ENERGETIC ANALYSIS OF LANDING: A NOVEL APPROACH TO UNDERSTANDING ANTERIOR CRUCIATE LIGAMENT INJURIES

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A dissertation submitted to the faculty of the University of North Carolina at Chapel Hill in partial fulfillment of the requirements for the degree of Doctor of Philosophy in the Department of Interdisciplinary Human Movement Science (School of Medicine).

Chapel Hill
2011

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ABSTRACT

MARC FABIAN NORCROSS: Energetic Analysis of Landing: A Novel Approach To Understanding Anterior Cruciate Ligament Injuries
(Under the direction of J. Troy Blackburn)

Energetic analysis of landing combines kinematic and kinetic parameters across the landing period that have traditionally been evaluated independently and at discrete time points. This coupling of the kinematics and kinetics of multiple joints provides a more comprehensive description of the complex multi-segmental mechanics that occur during landing and in proposed anterior cruciate ligament (ACL) injury mechanisms. The purpose of this investigation was to utilize this form of analysis to 1) elucidate new knowledge regarding biomechanical factors that contribute to sagittal plane energy absorption (EA) patterns that are associated with high risk landing biomechanics related to ACL injury; 2) explore relationships between frontal and sagittal plane EA, and ACL-related landing biomechanics; and 3) clarify previous research regarding potential sex differences in lower extremity EA strategies. 82 volunteer subjects (41 males, 41 females; age = 20.1 ± 2.4 years; height = 1.74 ± 0.10 m; mass = 70.3 ± 16.1 kg) were included in this research study. Subjects had peak isometric strength measured prior to completing double leg jump landing and drop landing tasks during which biomechanics and were assessed. It was found that greater sagittal and frontal plane EA during the 100 ms after ground contact were indicative of biomechanical profiles that likely result in greater ACL loading due to sagittal and frontal plane mechanisms, respectively. However, there is no association between the magnitudes of sagittal and frontal plane EA during landing. Additionally, no sex differences in EA strategy were identified after controlling for initial joint kinematics indicating that landing posture, not sex, influences EA strategy. Finally, the combination of multi-factorial

biomechanical parameters is predictive of EA at the hip and ankle, but not at the knee and suggests that interventions aimed at reducing total lower extremity EA and thereby potentially decreasing knee joint loading during landing must facilitate changes across the entire kinetic chain. The results of this investigation provide significant information for understanding the way in which multi-joint lower extremity movement patterns during landing, quantified using EA analyses, affects ACL loading, and provides much-needed evidence for specific biomechanical factors that should be targeted in ACL injury prevention programs.

ACKNOWLEDGEMENTS

While I have made every attempt to be all-inclusive, I apologize in advance for any oversights and wish to thank everyone for their support and encouragement along my path to this point.

To my committee- Drs. Lewek, Padua, Shultz, and Weinhold: I thank you for your generous contributions of time and wisdom that have added tremendously to this project. My hope when formulating my committee was that it would be tougher and fairer than any reviewer that I may come across, and that I would be challenged to develop an idea and message that was both meaningful and scientifically sound. Thanks to all of you, I believe that together we have achieved those goals-

To my family- Mom, Dad, Ryan, Alisha, Dan, Karen, and Tom: Thank you for always believing in me and instilling in me the confidence to attempt new endeavors no matter how crazy they may seem. Your unwavering support behind every success and failure my entire life has not gone unnoticed. I love you.

To Troy: I can not tell you how grateful I am that you were willing to take a chance and invite me to work with you during such a pivotal point in your career. Your knowledge and expertise coupled with your compassion and broader view that life outside of the academic walls is important has and will continue to be truly appreciated. I could not have asked for a better mentor or friend through this process. Thank you.

To Kate: While you may have arrived at the end of this journey, you provided me with much needed perspective at exactly the right time.

To Kendal: I can’t believe how much you have grown over the past four years. We brought you to North Carolina as a baby, but you leave here a beautiful and independent little girl. Thank you for always being willing to help with my "project" and being a source of motivation as well as unconditional love.

To Emily: I may never be able to fully articulate just how thankful and grateful I am to have you in my life. How you ever signed off on a plan to quit our jobs and move across the country with a one year old is beyond me. However, it was just one of many sacrifices that you have made for our family, and for me. I can never thank you enough, but will continue to spend the rest of my life attempting to do so. Thank you for being my best friend and the most wonderful wife in the world. I love you!

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CHAPTER ONE

INTRODUCTION

1.1 The Consequences of Anterior Cruciate Ligament Injury

Anterior cruciate ligament (ACL) injuries are both debilitating and extremely costly to the American health care system. Each year, an estimated 250,000 ACL injuries occur in the United States,[14] resulting in annual surgical costs of more than $2 billion.[93] This figure does not account for costs associated with initial management and rehabilitation of these injuries, nor the treatment of long-term sequelae such as knee osteoarthritis (OA). Knee OA is three times more likely to develop in individuals who suffer knee joint injuries,[50] and has been documented radiographically in 40-50% of patients within 14 years following ACL injury irrespective of the treatment chosen.[91, 157] The additional non-surgical costs, coupled with the concomitant decline in patients' quality of life, imply that the true economic and social impact of ACL injury has been grossly underestimated. Moreover, despite more than 9,500 scholarly publications over the past 50 years dedicated to ACL injury, the exact causes of injury and specific factors to be targeted to effectively prevent its occurrence remain unknown. Therefore, continued research utilizing more comprehensive methods of biomechanical analysis are necessary to both advance our understanding and improve prevention efforts related to this traumatic injury.

1.2 The Role of Energetic Biomechanical Analyses in ACL Injury Research

Non-contact mechanisms account for 70-80% of all ACL injuries;[4, 52] occurring most commonly in dynamic activities involving rapid deceleration, cutting, and landing.[1, 138] During landing, impact with the ground induces hip, knee, and ankle flexion (dorsiflexion) motions of the lower extremity. Internal hip, knee, and ankle extension (plantarflexion) moments are produced via eccentric muscle contractions in response to this impact in an effort to control joint motion and absorb kinetic energy from the whole body system.[34] This energy absorption (EA) by the lower extremity musculature at individual joints can be calculated using energetic analyses in which kinematic (joint angular velocity) and kinetic (net joint moment) data are combined to quantify the energy flow at each joint that is responsible for producing the observed movement (Figure 1).[158] While typical, more common biomechanical analyses used in ACL injury research identify kinematic and kinetic parameters independently and at discrete time points, energetic analyses combine these data across the landing period. Further, the individual contributions of each joint to the total energy absorption of the lower extremity may be calculated and offers insight into the coordinated actions of the hip, knee, and ankle.[12, 77, 105] This coupling of the kinematics and kinetics of multiple joints provides a more comprehensive description of the complex multi-segmental mechanics that occur during landing and in proposed ACL-injury mechanisms.[70]

Though limited in scope, previous work suggested that greater EA by the neuromuscular system reduces loading of passive tissues (e.g. the ACL)[34] with greater total lower extremity EA in the sagittal plane associated with smaller vertical ground reaction forces (vGRF) and greater knee flexion displacements during landing.[146, 170] Additionally, the neuromuscular system increases both sagittal and frontal plane lower extremity EA in response to greater mechanical demands (e.g. increasing landing height).[104, 163, 164, 170] As a result, greater total lower extremity EA has been suggested to reduce the risk of ACL and other soft tissue injuries.[23, 36, 63, 117, 146] Recently, it has been reported that rather than just magnitude, the timing of sagittal plane EA is important in modifying the relationship between EA and landing biomechanics that are associated with ACL injury. Specifically, greater sagittal plane EA during the100 ms immediately following ground contact, and lesser EA during the time from 100 ms after ground contact to the minimum vertical position of the whole body center of mass are associated with higher risk landing biomechanics.[119] This suggests that limiting the magnitude of sagittal plane EA during the 100 ms after ground contact, which is the interval when peak ACL strain and injury likely occurs,[18, 76, 161] may be beneficial with respect to reducing ACL injury risk. However, total lower extremity EA is derived from the coordinated neuromechanical characteristics of the hip, knee, and ankle, and it is unclear which of these factors predispose individuals to large versus small total EA. As such, the identification of factors such as strength, muscle activation, and components of landing technique such as initial landing posture and active movement during landings would help to explain why strategies with greater total EA are utilized; and how these EA strategies might be modified.

It is also well-known that females display a two-to-eight times greater risk of ACL injury compared with males.[37, 52] Accordingly, a great deal of research has focused on identifying neuromechanical differences between sexes as a potential means to discover the underlying mechanism for non-contact ACL injury.[20, 28, 46, 72, 85, 132, 135, 142, 144, 146] To date, however, only two of these investigations have utilized more comprehensive energetic analyses, and both identified sex differences in EA.[28, 146] From these results, Decker et al.[28] proposed that use of a sex-specific EA strategy necessitated the adoption of an erect landing posture by females; a posture that has been implicated as contributing to their greater ACL injury risk.[15, 53, 67] However, as landing posture was not controlled for in these investigations, it is unclear if the observed sex difference in EA strategy is attributable to kinematic differences (i.e. landing posture) between sexes that are driven by other sex-related factors such as strength, or to differences in feed-forward neuromuscular control that occur as a result of sex-specific EA strategies. This distinction is critical to ascertain as these two scenarios would require drastically different intervention techniques in order to most effectively alter EA strategies with hopes of reducing the risk of non-contact ACL injury.

Finally, previous research also indicates that greater frontal plane loading at the knee contributes to increased ACL strain *in vitro*[97] and that peak external knee valgus moment during landing is a significant predictor of future non-contact ACL injury.[63] As a result, greater reliance on the frontal plane for EA during landing might be associated with an increased ACL injury risk. Additionally, there is evidence that greater co-contraction of the quadriceps and hamstrings can assist in reducing frontal plane knee loading.[89, 90] This suggests that there may be an inter-planar EA relationship whereby greater sagittal plane EA may reduce EA in the frontal plane. However, currently there is no research explicitly evaluating the relationships between ACL-related landing biomechanics, frontal plane EA, and sagittal plane EA.

It is evident that energetic analysis of landing holds great potential for increasing our understanding of multiple facets of non-contact ACL injury. However, the application of this technique to ACL injury research thus far has been nominal. As a result, the focus of this dissertation was to utilize energetic analyses in a three-pronged approach to: 1) elucidate new knowledge regarding modifiable biomechanical factors that contribute to sagittal plane EA patterns that have been associated with high risk landing biomechanics related to ACL injury; 2) explore relationships between lower extremity EA in the frontal and sagittal planes, and ACL-related landing biomechanics; and 3) clarify previous research regarding potential sex differences in lower extremity EA strategies. It is proposed that this approach will expand the current body of knowledge with respect to ACL injury from a mechanistic perspective as well as provide much-needed rationale for current and future non-contact ACL injury prevention program design.

1.3 Part I: The Identification of Biomechanical Predictors of Sagittal Plane Lower Extremity Energy Absorption

1.3.1 Background

While the theoretical basis linking greater EA with lesser risk of injury via a reduction in peak impact forces is generally accepted; this result has typically been observed in studies which have artificially manipulated landing conditions. Devita et al.[34] and Zhang et al.[170] observed greater EA and lesser peak impact forces in "soft" landings compared to "stiff" landings when subjects were instructed to alter the magnitude of their knee flexion displacement during drop landings. However, there is limited evidence that directly relates greater EA during landing to lesser peak impact forces under naturally occurring landing conditions. Recent work suggests that in individuals performing double-leg jump landings using their natural/preferred landing style, it is not the overall magnitude, but rather the timing of EA during landing that influences peak impact forces.[121] Norcross et al.[121] compared EA during the initial impact (INI: 100 ms following initial ground contact [IGC]), terminal (TER: 100 ms after IGC to minimum vertical position of the whole body COM), and total (TOT: INI + TER) phases of a double-leg jump landing between groups displaying high and low peak vGRF. While there was not a significant group difference in the total sagittal plane EA during the TOT landing period, the high vGRF group exhibited significantly greater total EA and a greater percentage of total EA during INI phase of landing compared with the low vGRF group (Table 1).

In addition to peak vGRF, the timing and magnitude of sagittal plane EA during landing influences other ACL injury risk factors as well. Norcross et al.[119] demonstrated that a greater magnitude of total lower extremity EA during the INI phase of double-leg jump landings was associated with greater peak vGRF [$r = 0.442$, $P = 0.021$], as well as greater anterior tibial shear force (ATSF) [$r = 0.747$, $P < 0.001$] and internal hip extension moment (HEM) [$r = 0.422$, $P = 0.028$]. However, greater total EA during the TER phase of landing was associated with smaller peak vGRF [$r = -0.534$, $P = 0.004$] and HEM [$r = -0.413$, $P < 0.032$]. These biomechanical variables are of particular interest as they either differ prospectively between individuals who subsequently suffer an ACL injury and those who do not (vGRF and HEM)[63] or are intimated to directly contribute to ACL loading (ATSF).[20] Additionally, the temporal relationship between the magnitude of EA and biomechanical ACL injury risk factors is evident when examining EA of individual lower extremity joints.[119]

Specifically, greater peak vGRF and HEM are associated with greater INI hip EA, greater peak ATSF is associated with greater INI hip and ankle EA, and greater TER knee EA is associated with lesser peak vGRF and HEM.[119] The aforementioned results suggest that individuals who absorb greater magnitudes of energy at the hip, ankle, and in total across the three lower extremity joints during the INI phase of landing utilize a movement strategy that may increase their risk of sustaining an ACL injury, while greater total and knee EA during the TER phase produces a biomechanical profile consistent with lesser ACL injury risk (Table 2). Given that EA results from eccentric muscle contraction, this greater INI EA is indicative of increased force and moment requirements of the extensor muscles early in the landing phase. As the knee is most extended and the ACL most vulnerable to quadriceps loading during this time, the combination of greater muscle forces and a more extended knee likely contribute to greater ACL strain.

To our knowledge, Norcross et al.[119] were the first to directly link lower extremity EA with biomechanical variables suggested in the literature as ACL injury risk factors. It is important to highlight the fact that joint power, and therefore EA, is determined by the combination of joint moment and angular velocity. Therefore, EA can be influenced by any number of potentially alterable biomechanical factors such as strength, muscle activation, joint position at IGC, and joint angular displacement and velocity during landing. Therefore, the purpose of Part I was to identify modifiable biomechanical factors that significantly predict the magnitude of hip, knee, and ankle energy absorption during the initial impact phase of double leg jump landings. We chose to focus on this phase of the landing task due to the fact that high magnitude EA in this phase is associated with a biomechanical profile consistent with greater ACL loading and injury risk; and because peak ACL strain is attained within this interval.[18, 76, 161] Secondarily, the face validity of categorizing individuals as having a higher risk of ACL injury based on the magnitude of total EA during the INI phase of landing was evaluated by comparing biomechanical parameters associated with non-contact

ACL injury between groups exhibiting greater and lesser total EA during the INI phase of landing.

1.3.2 Significance

While previous work has associated EA and proposed risk factors for non-contact ACL injury, it is not currently known what underlying biomechanical factors are responsible for influencing joint angular velocities and joint moments, and the subsequent EA profile. Joint positions at impact, which define the available joint ranges of motion,[107] and joint displacements during landing can both affect joint angular velocities.[104, 163] Similarly, strength and muscle activation amplitudes (i.e. EMG) utilized in combination with joint kinematics may influence the net joint moments. As a result, this investigation sought to build upon our previous work by completing a comprehensive neuromechanical analysis (kinematic, kinetic, and electromyographic) in an attempt to identify specific and modifiable predictors of EA profiles that have been previously associated with high risk landing biomechanics related to ACL injury. By identifying modifiable biomechanical variables that predict lower extremity EA, we suggest that we will also isolate specific components of current ACL injury prevention programs that may be responsible for the reduction in injury incidence that has been observed with their implementation.[61] As current programs[59, 62, 96, 110, 133, 152] vary greatly with respect to the included components (strength, flexibility, neuromuscular training, balance, plyometrics, etc.), intensity, and duration (10-75 minutes); the identification of key biomechanical parameters that should be addressed would serve to streamline these prevention programs in hopes of increasing both their effectiveness and efficiency.

Additionally, while greater EA during the INI phase of landing is associated with a less desirable biomechanical ACL injury risk factor profile, it is not currently known whether total INI EA might be useful as a mechanism to identify individuals at greater risk of non-contact ACL injury. It is well-documented that females display a greater likelihood than males of suffering a non-contact ACL injury.[37, 52] However, current ACL injury prevention programs are commonly directed only toward females,[62, 96, 110, 133] most often due to their greater risk of injury, despite the fact that a greater absolute number of ACL injuries are suffered by males.[26, 93, 128] Therefore, it is expected that while more females will be identified as high-risk using this EA method, there will be some males identified as high-risk who would otherwise have been labeled as low-risk when using sex alone to categorize injury risk. It is hoped that more effective prospective identification of high-risk athletes using criteria other than simply sex will allow for prevention programs to be more applicable to all individuals with a heightened risk of injury.

1.3.3 Research Questions, Hypotheses, and Approach

RQ 1A: What is the respective relationship between the magnitude of sagittal plane hip EA during the initial impact phase (INI) of a double leg jump landing (criterion variable) and the following biomechanical (predictor) variables?

1. Peak isometric hip extension strength
2. Mean gluteus maximus EMG activation amplitude during the 100 ms centered around IGC
3. Sagittal plane hip joint position at IGC
4. Peak hip flexion angle during the loading phase

RH 1A: Greater peak hip extension strength and gluteus maximus activation, but lesser hip flexion at initial ground contact and peak hip flexion during landing will be significant predictors of sagittal plane hip INI EA.

RQ 1B: What is the respective relationship between the magnitude of sagittal plane knee EA during the initial impact phase (INI) of a double leg jump landing (criterion variable) and the following biomechanical (predictor) variables?

1. Peak isometric knee extension strength
2. Peak isometric knee flexion strength
3. Mean hamstring EMG activation amplitude during the 100 ms centered around IGC

4. Mean quadriceps EMG activation amplitude during the 100 ms centered around IGC
5. Sagittal plane knee joint position at IGC
6. Peak knee flexion angle during the loading phase

<u>RH 1B</u>: Greater peak knee extension and flexion strength and quadriceps and hamstrings activation, but lesser knee flexion at initial ground contact and peak knee flexion during landing will be significant predictors of sagittal plane knee INI EA.

<u>RQ 1C</u>: What is the respective relationship between the magnitude of sagittal plane ankle EA during the initial impact phase (INI) of a double leg jump landing (criterion variable) and the following biomechanical (predictor) variables?

1. Peak isometric ankle extension (plantarflexion) strength
2. Mean gastrocnemius EMG activation amplitude during the 100 ms centered around IGC
3. Sagittal plane ankle joint position at IGC
4. Peak ankle flexion angle during the loading phase

<u>RH 1C</u>: Greater peak ankle extension strength, gastrocnemius activation and ankle extension angle at initial ground contact, but lesser peak ankle flexion during landing will be significant predictors of sagittal plane ankle INI EA.

<u>Approach</u>: Three, separate stepwise multiple linear regression analyses were used to identify biomechanical factors that significantly predict sagittal plane hip, knee, and ankle EA during INI.

<u>RQ 2</u>: Are there significant differences between groups exhibiting higher (large total sagittal plane EA during INI), moderate (moderate total sagittal plane EA during INI) and lower-risk (small total sagittal plane EA during INI) landing biomechanics related to non-contact ACL injury in the following dependent variables during a double leg jump landing?

A. Peak vGRF
B. Peak pGRF

C. Peak anterior tibial shear force (ATSF)
D. Peak internal hip extension moment (HEM)
E. Peak internal knee extension moment (KEM)
F. Peak internal knee varus moment (KVM)
G. Frontal plane knee angle at IGC
H. Peak knee valgus angle
I. Sagittal plane knee angle at IGC
J. Peak knee flexion angle

RH 2: Compared with the low and moderate risk landing biomechanics groups, the high-risk landing biomechanics group (large total EA during INI) will demonstrate significantly:

A. Greater peak vGRF
B. Greater peak pGRF
C. Greater peak anterior tibial shear force (ATSF)
D. Greater peak internal hip extension moment (HEM)
E. Greater peak internal knee extension moment (KEM)
F. Greater peak internal knee varus moment (KVM)
G. Greater frontal plane knee angle at IGC
H. Greater peak knee valgus angle
I. Lesser sagittal plane knee flexion angle at IGC
J. Lesser peak knee flexion angle

Approach: Subjects were grouped in tertiles based on their total sagittal plane EA during the INI phase of the double-leg jump landing. This arrangement of the EA data created groups who exhibit higher, moderate, and lower risk (highest, middle, and lowest tertiles, respectively) landing biomechanics related to ACL injury based on previous work regarding the relationship between EA during the INI phase of landing and biomechanical ACL injury risk factors.[119] A quasi-experimental design (static group comparisons) was used to determine significant differences in the dependent variables between groups using one-way ANOVA.

RQ 3: Is there a significant association between sex and non-contact ACL landing biomechanics risk group assignment via total INI EA?

RH 3: There will be a significant association between the high-risk landing biomechanics group (large total EA during INI) and females.

Approach: ACL-related landing biomechanics risk group (highest and lowest total INI EA) and sex were used as categorical variables to evaluate whether there is a significant association between group and sex using a X^2 test of association.

1.4 Part II: The Relationship between Sagittal Plane Energy Absorption, Frontal Plane Energy Absorption, and ACL Injury Risk Factors

1.4.1 Background

Females demonstrate greater knee valgus angles during landing compared to males,[46, 63, 132, 142] and frontal plane knee loading has been shown both *in vivo* using biomechanical modeling[21, 102] and *in vitro*[97] to contribute to ACL loading and lower injury threshold. Consequently, knee valgus angle and moment have been noted as risk factors[53, 63] and significant predictors of non-contact ACL injury.[63] We recently demonstrated that the magnitude of sagittal plane EA during jump landings is associated with biomechanical risk factors for ACL injury.[119] However, no significant relationships were identified between sagittal plane EA during INI and frontal plane biomechanics.[119] Conversely, Pollard et al.[136] reported that individuals exhibiting greater combined hip and knee flexion during landing displayed significantly greater sagittal plane hip and knee EA during the total landing phase, and lesser peak knee valgus angle and average internal knee varus moment. Their results are important as individuals who displayed greater combined peak hip and knee flexion also displayed more desirable frontal plane biomechanics. This suggests that sagittal plane EA may influence frontal plane risk factors. However, there are two limitations to this work. First, Pollard et al.[136] evaluated EA only over the TOT landing period; thereby potentially obscuring a temporal relationship between the timing of sagittal plane EA and frontal plane biomechanics. Second, these authors did not quantify the magnitude of frontal plane EA that occurred in their investigation. As such, it is unknown if greater sagittal plane EA required the use of lesser frontal plane EA; and whether lesser frontal plane EA was associated with the lesser knee valgus angle and internal knee varus moment. These limitations indicate that further research is necessary to determine the precise role that sagittal and frontal plane EA have in influencing frontal plane biomechanics.

It has also been proposed that increasing frontal plane hip stiffness, or stability, during landing can reduce frontal plane knee loading and subsequent ACL loading[21] due to the fact that hip adduction angle is a significant predictor of knee valgus angle.[130] Therefore, greater EA in the frontal plane at the hip (i.e. greater eccentric resistance to hip adduction) might reduce knee valgus motion and ACL loading caused by this frontal plane mechanism. However, there is currently only one published report which has focused on frontal plane EA during landing. Yeow et al.[164] observed significantly greater frontal plane EA at the hip and knee compared with the ankle in natural-style double leg landings from heights of 0.30 and 0.60 m. Further, they observed a significant increase in total frontal plane EA at greater landing heights; an increase that was primarily driven by an increase in EA at the hip.[164] Their results suggest that frontal plane EA is augmented in response to greater mechanical demands, much like in the sagittal plane, and that there is a greater reliance on the hip for frontal plane EA with increasing mechanical demands during landing.[104, 170 164] However, the primary limitations of this work are that it remains unknown whether frontal plane EA is directly associated with risk factors for non-contact ACL injury, and whether any significant relationships exist between frontal and sagittal plane EA. Additionally, should an association between frontal plane EA and high risk landing biomechanics exist, it is unknown whether groups performing different amounts of frontal plane EA during landing demonstrate meaningful differences on these ACL-related biomechanical factors. Therefore, the purpose of Part II was to explore the relationships between lower extremity

EA in the frontal and sagittal planes, and frontal plane landing biomechanics related to ACL injury.

1.4.2 Significance

Although the relationship between sagittal plane EA and ACL-related landing biomechanics has been investigated previously, there is currently no evidence linking frontal plane EA and landing biomechanics associated with non-contact ACL injury. While greater frontal plane hip EA during landing might serve to decrease ACL loading, it is plausible that greater knee EA in the frontal plane may be detrimental. This is because greater frontal plane knee EA is the result of either increased frontal plane knee angular velocity and/or increased frontal plane knee moment which contribute to frontal plane knee loading. However, Lloyd and Buchanan[89, 90] have demonstrated that the quadriceps and hamstrings musculature can support varus-valgus loading of the knee during both isometric and dynamic tasks, primarily via co-contraction. These results indicate a potential inter-planar EA relationship whereby greater sagittal plane knee EA (eccentric contraction of the quadriceps) could provide greater frontal plane support. As a result, the magnitude of frontal plane EA and frontal plane knee loading during landing might be mediated by increasing EA in the sagittal plane.

By identifying relationships between frontal plane EA, sagittal plane EA, and frontal plane biomechanics, we will be able to achieve two goals; 1) we can determine how frontal plane EA at the hip, knee and ankle influences ACL injury risk factors, and 2) we can construct a more thorough description of the multi-dimensional nature of energy dissipation during landing.

1.4.3 Research Questions, Hypotheses, and Approach

<u>RQ 4A</u>: Are there significant associations between total lower extremity, hip, knee, and ankle EA in the frontal plane during the INI phase of a double leg jump landing task and the following criterion variables?

A. Frontal plane knee angle at IGC
B. Peak knee valgus angle during TOT
C. Peak internal knee varus moment during TOT
D. Peak vGRF during TOT
E. Peak pGRF during TOT
F. Peak hip adduction angle during TOT
G. Total sagittal plane INI EA
H. Sagittal plane hip INI EA
I. Sagittal plane knee INI EA
J. Sagittal plane ankle INI EA

RH 4A: Greater total lower extremity, hip, knee, and ankle EA in the frontal plane during the INI phase of a double leg jump landing will be significantly associated with:

A. Greater knee valgus angle at IGC
B. Greater peak knee valgus angle during TOT
C. Greater peak internal knee varus moment during TOT
D. Greater peak vGRF during TOT
E. Greater peak pGRF during TOT
F. Greater peak hip adduction angle during TOT
G. Greater sagittal plane total INI EA
H. Greater sagittal plane hip INI EA
I. Greater sagittal plane knee INI EA
J. Greater sagittal plane ankle INI EA

RQ 4B: Are there significant associations between total lower extremity, hip, knee, and ankle EA in the frontal plane during the TER phase of a double leg jump landing task and the following criterion variables?

A. Frontal plane knee angle at IGC
B. Peak knee valgus angle during TOT
C. Peak internal knee varus moment during TOT
D. Peak vGRF during TOT
E. Peak pGRF during TOT
F. Peak hip adduction angle during TOT
G. Total sagittal plane EA during TER
H. Sagittal plane hip EA during TER
I. Sagittal plane knee EA TER
J. Sagittal plane ankle EA TER

RH 4B: Greater total lower extremity, hip, knee, and ankle EA in the frontal plane during the TER phase of a double leg jump landing will be significantly associated with:

A. Lesser knee valgus angle at IGC
B. Lesser peak knee valgus angle during TOT
C. Lesser peak internal knee varus moment during TOT
D. Lesser peak VGRF during TOT
E. Lesser peak pGRF during TOT
F. Lesser peak hip adduction angle during TOT
G. Greater sagittal plane total TER EA
H. Greater sagittal plane hip TER EA
I. Greater sagittal plane knee TER EA
J. Greater sagittal plane ankle TER EA

Approach: Simple, bivariate Pearson correlation coefficients were calculated to assess the relationships between total, hip, knee, and ankle frontal plane EA during the INI and TER phases of double-leg jump landings and the criterion variables.

RQ 4C: Are there significant differences between groups exhibiting higher, moderate, and lower magnitudes of total frontal plane EA during INI in the following dependent variables related to ACL injury during a double leg jump landing?

A. Frontal plane knee angle at IGC
B. Peak knee valgus angle
C. Peak hip adduction angle
D. Peak vGRF
E. Peak pGRF
F. Peak internal knee varus moment (KVM)

RH 4C: Compared with the lower and moderate frontal plane EA groups, the highest frontal plane EA group will demonstrate significantly:

A. Greater frontal plane knee angle at IGC
B. Greater peak knee valgus angle
C. Greater peak hip adduction angle
D. Greater peak vGRF
E. Greater peak pGRF
F. Greater peak internal knee varus moment (KVM)

Approach: Subjects were grouped in tertiles based on their total frontal plane EA during the INI phase of the double-leg jump landing. This arrangement of the EA data created groups who exhibited higher, moderate, and lower magnitudes of frontal plane INI EA. A quasi-experimental design (static group comparisons) was used to determine significant differences in the dependent variables between groups using one-way ANOVA.

1.5 Part III: Derivations of the Sex Difference in Energy Absorption Strategy

1.5.1 Background

Females tend to make contact with the ground during landing in a more erect position with the knee joint positioned in less flexion compared to males.[28, 94, 166] During landing, the quadriceps acts eccentrically to control knee flexion and has the greatest potential for generating anterior tibial shear force and loading the ACL at knee flexion angles between 10-30°.[35, 52, 75] Further, the posterior tibial shear force component of the hamstrings muscles, which can protect against excessive ACL loading, decreases as the knee joint is moved to less flexed positions.[131] This combination of increased ACL loading secondary to quadriceps contraction and decreased ACL protection provided by the hamstrings when landing in a more erect position has been implicated as one possible factor for the observed sex difference in ACL injury risk. As a result, increasing knee flexion during landing through technique instruction has been adopted as a common component in ACL injury prevention programs,[62, 96, 110] though the underlying reason for the more erect landing position in females continues to remain unknown.

Lower extremity EA results from the coordinated action of the hip, knee, and ankle. Several investigators have reported that these joints all contribute to EA, and that total lower extremity EA equals the sum of the energy absorbed at these joints.[33, 34, 79, 119, 146, 170] Further, the individual joint contributions to total EA change with alterations in landing height or landing style.[34, 170] Therefore, there may be numerous individualized strategies capable of achieving the same total magnitude of EA. However, despite the potential for many different strategies, the current literature suggests that there are two primary EA strategies employed during landing, and that these strategies are sex-specific. While both sexes seem to rely on the knee as the primary contributor to total lower extremity energy dissipation (33-47%), males utilize a strategy that emphasizes greater secondary contribution from the hip (30-42%) compared to the ankle (14-30%).[28, 104, 170] Alternatively, females tend to display an EA strategy with greater secondary contribution from the ankle (35-37%) compared to the hip (18-25%).[28, 34]

To our knowledge, only three investigations have directly compared the EA strategies of males and females. Decker et al. [28] and Schmitz et al.[146] observed the previously described ankle dominant energy absorption strategy and a more erect landing position in females during double-leg and single-leg landings, respectively; while Schmitz and Shultz[147] reported a greater magnitude and relative contribution to total EA from the knee in females compared to males when performing drop jump landings. Decker et al.[28] postulated that the more erect landing posture in females was the result of the preferential use of an EA strategy in which the knee and ankle provide greater relative contributions to total EA than the hip. Consequently, this erect landing posture may contribute to a greater risk of ACL injury in females by placing the knee in a more extended position at impact during landing.[15, 53, 67] This theory is partially supported by Devita et al.,[34] who evaluated females during "soft" (knee flexion at initial contact ≈ 28°), and "stiff" (knee flexion at initial contact ≈ 21°) landings that were artificially produced by instructing participants to limit the amount of knee flexion displacement during landing. They observed that the relative joint contributions to total energy absorption remained similar in each condition for the knee (37% vs. 31%) and hip (25% vs. 20%). However, while the contribution of the ankle to total EA was less during the "soft" condition compared to the "stiff" condition (37% vs. 50%); the contribution of the ankle during both conditions was still greater than the ankle contributions that have been previously reported in males during similar landing tasks.[28, 104, 170] These results suggest that a knee/ankle dominant EA strategy may persist in females regardless of the knee joint position at impact.

More recent work using the self-selected landing style of males and females (as opposed to experimentally manipulated landing conditions) performing double leg jump landings conflicts with these previous reports and suggests that energy absorption strategies may not be sex-specific.[118] Norcross et al.[118] reported no significant differences between sexes in hip, knee, and ankle contributions to total EA, nor in joint positions at initial ground contact. These results are in contrast to the previously identified sex differences in EA noted by Decker et al.[28] and Schmitz et al.[146] However, it should be noted that the double leg jump landing task induces greater horizontal velocity than the drop landing tasks used by Schmitz and Decker, and that these different demands during landing contributed to the differing results. Further, in these two previous investigations, significant sex differences in landing kinematics were present that were not identified by Norcross et al.[118] It is possible that the females sampled in the Norcross et al.[118] investigation may have exhibited biomechanical parameters (i.e. strength, muscle activation, etc.) that were comparable to those usually observed in males and that these underlying factors resulted in their joint positions at contact being similar to those typically observed in males. Based upon this discrepancy, it is plausible that initial joint positioning during landing, which can affect joint angular velocities and joint moments and thus the subsequent joint power profile (and as hypothesized in Part I), may be responsible for influencing EA strategy, instead of sex.[28, 170] Therefore, the purpose of Part III is to clarify previous research regarding potential sex-specific differences in lower extremity EA strategy by evaluating the influence of sex and initial landing posture on lower extremity EA during 0.60 meter drop-landings under natural/preferred conditions and conditions in which the initial knee angle is constrained.

1.5.2 Significance

The distinction between whether sex or landing posture influences lower extremity EA during landing is important because the results would suggest two different underlying mechanisms for the more erect landing position of females depending upon which factor (sex or landing posture) influences EA. Should sex influence EA strategy in such a way that a knee/ankle joint dominant EA strategy persists in females regardless of landing posture; it would suggest that this EA strategy is pre-programmed using a different, sex-specific motor program than that found in males. Females may configure their lower extremities prior to impact into the more erect landing posture in order to maximize the total EA capability of this knee/ankle dominant strategy. It is important to reiterate that increases in either net joint moment or joint angular velocity increase EA at a joint during landing. Positioning the ankle and knee in more extended positions at impact may serve to increase the available ROM at these joints allowing for greater angular velocities and thus greater EA. This notion is partially supported by previous research that indicates that females exhibit significantly greater joint angular velocities compared to males during landing.[28, 166] Therefore, although the more erect landing posture of females may be modified through instruction as is presently done in ACL injury prevention programs;[62, 96, 110] this technique change may result in an overall decrease in impact attenuation by limiting the available ROM at these joints and subsequently reducing the joint angular velocities. This reduction in total EA could then lead to a greater transfer of energy and stress on passive structures such as ligament, cartilage, and bone.[23, 34, 81] Additionally, in more mechanically demanding tasks requiring both greater EA and presenting greater potential for injury, females may revert to the more erect landing posture to maximize the magnitude of energy that they can absorb using their sex-specific knee/ankle dominant strategy. This suggests that simple instruction to land with greater flexion as is the norm in current prevention programs may not be sufficient to cause a permanent alteration in landing posture that will persist during landings with greater mechanical demands.

Alternatively, should landing posture influence EA strategy, regardless of sex, such that a knee/hip dominant EA strategy is utilized when landing with a flexed posture and a knee/ankle dominant strategy is used when landing with an erect posture; this finding would suggest that sex-specific EA strategies are not pre-selected. Rather, irrespective of sex, the EA strategy observed during landing is the result of the initial joint positions at impact. In this case, it may be that the erect landing posture of females is not a result of a pre-determined sex-specific EA. Instead, the initial landing posture might be derived from the influence of other biomechanical factor(s) (i.e. strength, muscle activation, etc.). As a result of these two drastically different scenarios, it is imperative to clarify previous work with respect to potential sex-specific EA strategies and to determine which factor- sex or landing posture- is actually influencing the EA strategies that have been reported.

1.5.3 Research Questions, Hypotheses, and Approach

RQ 5: Are there significant differences between sexes in the following EA variables during the initial impact phase (INI) of 0.60 meter drop landings performed using a preferred initial landing posture?

A. Relative joint (hip, knee, and ankle) contributions to total lower extremity EA
B. Magnitude of EA at the hip, knee, and ankle
C. Total lower extremity EA

RH 5: There will be significant differences between sexes in EA during the initial impact phase (INI) of 0.60 meter drop landings performed using a preferred initial landing posture such that:

A. 1) Females will exhibit greater contribution to total lower extremity EA from the ankle compared to males.
   2) Females will exhibit lesser contribution to total lower extremity EA from the hip compared to males.
B. Females will exhibit greater magnitudes of EA at the ankle and knee, but lesser magnitude EA at the hip compared to males.
C. Females will exhibit greater total lower extremity EA compared to males.

Approach: Subjects performed drop-landings from a 0.60 m high box in their preferred initial landing posture. Two, separate 2 (sex) x 3 (joint) mixed model repeated measures ANOVAs were used to determine the influence of sex on the relative joint contributions to total EA and individual joint EA magnitudes, while an independent samples *t*-test evaluated sex differences in total lower extremity EA.

RQ 6: Are there significant differences between sexes in the following EA variables during the initial impact phase (INI) of 0.60 meter drop landings when controlling for initial landing posture?

A. Relative joint (hip, knee, and ankle) contributions to total lower extremity EA
B. Magnitude EA at the hip, knee, and ankle
C. Total lower extremity EA

RH 6: There will not be significant differences between sexes in any EA variable during the initial impact phase (INI) of 0.60 meter drop landings after controlling for initial landing posture.

Approach: Subjects used real-time biofeedback regarding their sagittal plane knee joint position while performing 0.60 m drop-landings from an overhead drop bar to achieve a standardized flexed knee posture at IGC. Two, separate 2 (sex) x 3 (joint) mixed model repeated measures ANOVAs determined the influence of sex on the individual joint EA magnitudes and joint contributions to total EA, while an independent samples *t*-test was used to evaluate sex differences in total lower extremity EA.

RQ 7: Are the following EA variables during the initial impact phase (INI) of 0.60 meter drop landings affected by changing initial landing posture (flexed vs. erect) and are these changes modified by sex?

A. Relative hip contribution to total lower extremity EA
B. Relative knee contribution to total lower extremity EA
C. Relative ankle contribution to total lower extremity EA
D. Magnitude of hip EA
E. Magnitude of knee EA
F. Magnitude of ankle EA
G. Total lower extremity EA

RH 7: Initial landing posture, but not sex, will significantly influence the following EA variables during the INI phase of landing such that compared with the erect condition all subjects in the flexed condition will exhibit:

A. Greater relative hip contribution to total lower extremity EA
B. No difference in the relative knee contribution to total lower extremity EA
C. Lesser relative ankle contribution to total lower extremity EA
D. Greater magnitude of hip EA
E. No difference in the magnitude of knee EA
F. Lesser magnitude of ankle EA
G. Lesser total lower extremity EA

Approach: Subjects used real-time biofeedback regarding their sagittal plane knee joint position while performing 0.60 m drop-landings from an overhead drop bar to achieve standardized flexed and erect knee postures at IGC. Seven, separate 2 (sex) x 2 (landing posture) mixed model repeated measures ANOVAs were used to determine the influence of sex and/or landing posture on the relative joint contributions to total EA, individual joint EA magnitudes, and the total magnitude of lower extremity EA.

1.6 Operational Definitions

Initial ground contact (IGC): The beginning of the total landing period was defined as the instant when the vertical component of the ground reaction force vector exceeds 10 Newtons.

Initial impact phase of landing (INI): The 100 ms immediately following initial ground contact (IGC)[27, 28]

Terminal phase of landing (TER): The period from 100 ms after IGC to the minimum vertical position of the entire body COM.[78]

Total landing period (TOT): The combined INI and TER phases of landing comprised of the period from IGC to the minimum vertical position of the entire body COM.

Dominant limb: The limb used to kick a ball for maximal distance.

Double leg jump landing: Subjects stood atop a 0.30 m tall box positioned 50% of their height behind a force plate. They then jumped forward and down toward the plate and landed with their dominant foot positioned in the center of the force plate and their non-dominant foot next to the force plate before immediately jumping up for maximum height.

Double leg drop landing: Subjects fell vertically from a height of 0.60 m and landed with their dominant foot positioned in the center of the force plate and their non-dominant foot next to the force plate before performing a terminal landing.

1.7 Assumptions

The following assumptions were made for this dissertation project:

1. Participants performed all testing protocols to the best of their ability and with maximum effort.

2. Participants were honest regarding their prior history with respect to the inclusion/exclusion criteria.

3. The biomechanical data collected during these experiments was reliable and valid for all participants.

4. Participants were not enrolled in a non-contact ACL injury prevention program at the time of testing.

1.8 Delimitations

The following delimitations were made for this dissertation project.

1. All participants were between the ages of 18-30 at the time of testing.

2. All kinematic and kinetic data was sampled using the same motion analysis system and force plate.

3. All strength data was collected using the same handheld dynamometer.

4. All participants had no history of ACL injury, lower extremity surgery, neurological disorder, or lower extremity injury within the 6 months preceding data collection that restricted activity for more than 3 days.

5. All participants were physically active as defined by participation in at least 30 minutes of activity a minimum of three days per week.

CHAPTER TWO

LITERATURE REVIEW

The purpose of the following review of the literature is to provide a contextual background for the previously proposed research questions, hypotheses, and significance statements. As such, this review concentrates on six primary topics: 1) ACL injury epidemiology with specific emphasis on the incidence, associated costs, and sex difference in injury risk; 2) previous research on sagittal and frontal plane biomechanical factors related to ACL injury; 3) the underlying theory and historical approaches to energetic analysis with particular attention to both the advantages and limitations of these different approaches; 4) the application of energetic analyses in investigating landing biomechanics; 5) a systematic review of previous investigations utilizing energetic analyses specific to ACL injury research; and 6) an explanation of how this investigation addresses specific gaps in the current body of knowledge and may substantially contribute to our understanding of non-contact ACL injury.

2.1 ACL Injury Epidemiology

2.1.1 ACL Injury Incidence

For non-contact ACL injuries in the United States, precise epidemiologic estimates are not known.[99] This lack of knowledge is attributed to a paucity of studies that capture both the annual number and incidence of ACL injury in an entire population;[99] and has led to the recommendation for a national ACL injury registry.[53] However, until these methods of injury surveillance are implemented, hospital surveys and a foreign ACL injury registry currently provide the best estimate of ACL injury incidence in the general population.

Using hospital survey data, the Centers for Disease Control and Prevention reported that over 100,000 knee cruciate ligament reconstructions were performed in 1996.[128] During 2006, 7,507 cruciate ligament reconstructions were performed in New York state alone; constituting a 21.5% increase in the number of reconstructions just a decade earlier.[93] Although posterior cruciate ligament reconstructions are included in both of these data sets, it may be fair to interpret these cases as ACL reconstructions for two reasons: 1) ACL injuries are overwhelmingly more common than PCL injuries;[110] and 2) Lyman et al.[93] identified that 99.3% of the total cruciate ligament reconstructions performed in New York State in 2006 were ACL reconstructions.

It is important to note that as not all ACL injuries are surgically reconstructed, the actual number of ACL injuries sustained is greater than the number of surgical reconstructions performed. Therefore, Boden et al.[14] have estimated that there may be as many as 250,000 ACL injuries in the United States annually. Utilizing health care survey information from 2003, Marshall et al.[99] estimated the annual number of cruciate ligament injuries in the United States to be about 200,000. While these figures do not represent a significantly high incidence of ACL injury in the general United States population; it has been suggested that as many as 1 out of every 90 patient visits to a physician for an unintentional injury is the result of a cruciate ligament injury.[99]

With respect to ACL injury incidence abroad, de Loës et al.[26] analyzed the injury data of 370,000 Swiss athletes, representing up to two-thirds of the 14-20 year old Swiss population, over a period of seven years. The overall incidence of cruciate ligament injury in this population was only 0.0059 per 100,000 athlete-hours of exposure; but, compared to the average expenditure for all knee injuries, the treatment of cruciate ligament injuries cost approximately 250% more.[26] Though the population incidence appears low, Marshall et al.[99] concluded that from the American health care system; non-contact ACL injury is a significant drain of both resources and money.

2.1.2 The Costs Associated with ACL Injury

As with incidence, there is no precise estimate of the total monetary cost of ACL injuries to the United States health care system. Due in part to the uncertainty surrounding the number of ACL injuries that occur annually, it is difficult to accurately quantify the total financial impact. With an average price in 1999 of $17,000 for both surgical reconstruction and rehabilitation,[62] Myer et al.[108] projected an annual cost of $650 million for these services in female secondary school and collegiate athletes alone. Boden et al.[14] used the same average price per injury to estimate the annual health care burden attributable to the reconstruction and rehabilitation of all ACL injuries in the United States to be about $1.5 billion. However, evidence indicates that the cost associated with only the surgical procedure (i.e., surgeon, anesthesiologist, operating room time, etc.) currently averages about $20,000.[16] Therefore, even if the number of ACL reconstructions performed annually in the United States has not increased since 1996, surgical reconstruction costs now exceed $2 billion each year. Compared to the combined cost for both surgical reconstruction and post-operative rehabilitation in 1999; the American health care system now spends $500 million more annually on ACL reconstructive surgeries alone. Further, it is not known how many additional millions of dollars are currently being spent for the post-surgical rehabilitation and medical imaging exams that these patients require.

In addition to the short-term costs previously addressed, there is a substantial long-term financial impact associated with ACL-injury; most notably from the treatment of knee osteoarthritis (OA). ACL-injured patients are at greater risk of developing knee OA[25, 42, 44, 50, 91, 92, 141, 153, 157] with approximately 50% of patients displaying OA within 10 years following injury.[109] Myklebust et al.[109] have proposed that within 20 years of injury, nearly all ACL-injured patients will develop knee OA. Further, the prevalence of knee OA is the same regardless of whether patients choose conservative management or surgical reconstruction.[43, 91, 111, 157] While the increased risk of OA may be partially attributable to the fact that up to 75% of ACL-injured individuals concomitantly injure their meniscus;[39] compared with isolated meniscus tears, patients suffering isolated ACL sprains display evidence of knee OA at a younger age.[141] This suggests that, regardless of any associated injuries, rupture of the ACL should be viewed as the impetus for the early onset of knee OA. Unfortunately, with the greatest number of ACL injuries occurring in individuals 16-18 years of age,[99, 150] early-onset knee OA may end up proving costlier over time to treat than the initial ACL injury.

2.1.3 Sex Difference in ACL Injury Risk

In addition to illuminating the substantial financial and societal implications, ACL related epidemiology has also provided a theoretical model for much of the research investigating both the mechanism and prevention of non-contact ACL injuries. While males suffer a greater absolute number of injuries,[26, 93, 128] females have a significantly greater risk of ACL injury in sports such as soccer,[1, 4, 11, 26, 54, 57, 62] basketball,[1, 4, 26, 54, 57, 106, 126], and handball.[26, 112, 113] As a result, many ACL-injury research studies have compared males and females across a variety of biomechanical factors with the idea that identified differences between the sexes might be important factors related to this injury. Though not without limitations, this experimental approach, in combination with basic science investigations, has successfully elucidated a number of biomechanical factors that are now accepted as probable contributors to non-contact ACL injury.

2.1.4 Summary of ACL Epidemiology

The precise incidence of non-contact ACL injury in the United States is unknown due to the lack of a national registry that would allow for study of the general population. Hospital surveys and foreign injury data suggest the incidence in the general population to be relatively low. Despite this low incidence, evidence indicates that the treatment of non-contact ACL injury demands considerable resources and money. While an unknown amount is spent on rehabilitation and other short-term needs like medical imaging; the costs associated with surgical reconstruction only are likely greater than $2 billion annually. In the long-term, ACL-injured patients are extremely likely to develop early-onset knee osteoarthritis. In combination, these short- and long-term costs constitute a tremendous financial burden. It has also been noted that although males suffer more injuries annually; females have a significantly greater risk of injury in selected sports. As a result, sex comparisons of landing mechanics have successfully been used to identify specific biomechanical factors related to non-contact ACL injury.

2.2 Biomechanical Factors Related to ACL Injury

Biomechanical, or neuromuscular, factors are one of four categories that have been used to classify risk factors for non-contact ACL injury.[53] While other intrinsic (anatomical and hormonal) and extrinsic (environmental) risk factor categories are important in understanding ACL injury mechanisms, we have chosen to focus on biomechanical factors as they are modifiable and thus may be targeted in injury prevention efforts. Further, although it is apparent that the mechanism of ACL injury is likely multi-planar (sagittal, frontal, and transverse),[2] this investigation will concentrate on the sagittal and frontal planes for two primary reasons; 1) the majority of total EA during landing occurs in these planes,[163, 164] and 2) there is preliminary evidence suggesting potential links between EA and ACL injury-related biomechanics for both the sagittal and frontal planes.[119, 136]

2.2.1 ACL Loading Mechanics

Prior to discussing specific biomechanical factors related to non-contact ACL injury, it is first necessary to describe the mechanical loading conditions at the knee that result in increased ACL strain, and ultimately ACL rupture. Using cadaveric knees, Berns et al.[8] reported that isolated anterior shear force significantly increased strain in the anteromedial bundle of the ACL; while isolated varus and valgus moments did not increase ACL strain. However, compared with isolated anterior force, the combination of anterior shear force and valgus moment resulted in significantly greater ACL strain.[8] Markolf et al.[97] also reported that isolated anterior tibial force application was the most direct mechanism of ACL loading. At full knee extension, the measured ACL forces were 150% of the applied anterior force; but, flexing the knee reduced the resultant ACL load.[97] However, at knee flexion angles greater than 10°, an applied valgus moment in combination with anterior shear force resulted in significantly greater ACL loading than that produced by anterior shear force alone.[97] Though considerable disagreement continues to persist about whether sagittal [88, 165] or frontal plane[102] loading is most responsible for ACL injury; the general consensus of the scientific community is that a combined ACL loading pattern is the most likely cause of excessive ACL loading.[53] Therefore, sagittal and frontal plane biomechanical features of human movement that contribute to greater ACL loading are considered deleterious.

2.2.2 Sagittal Plane Biomechanics

1. Movement Patterns

An overwhelming feature of biomechanical sex comparisons during landing is that compared with males, females contact the ground in a more erect posture with the hip and knee positioned in lesser flexion.[28, 68, 85, 94, 134, 135, 144] This more erect landing posture has been identified as a risk factor for ACL injury and is theorized to contribute to greater ACL loading in two ways; 1) by increasing ACL strain resulting from quadriceps muscle contraction, and 2) by increasing the peak impact forces.

In order to arrest the downward velocity of the whole body center of mass during landing, the lower extremity joints (hip, knee, and ankle) must resist external flexion moments caused by impact forces with internally generated extension moments.[34, 66] At the knee, the internal extension moment is generated by quadriceps contraction. However, quadriceps contraction is the primary contributor to anterior tibial shear force[165] with the sagittal plane position of the knee joint modifying the magnitude of the resultant ACL strain. *In vitro*[5, 35, 38] and *in vivo*[9, 10] experiments have demonstrated that quadriceps contraction between 0 and 30° of knee flexion significantly strains the ACL. Further, DeMorat et al.[29]

induced ACL injury in 6 out of 11 cadaver specimens positioned in 20° of knee flexion by applying an isolated 4,500 N quadriceps force.

Greater ACL strain due to a standardized quadriceps contraction at lesser knee flexion angles has been attributed to changes in the relative angles of both the patellar tendon and ACL with respect to the tibia.[165] Nunley et al.[123] reported that decreasing knee flexion angle causes an increase in the angle between the patella tendon and the tibial shaft; and results in a greater proportion of the quadriceps force being directed anteriorly relative to the tibia. Decreasing knee flexion angle also causes an increase in the elevation angle of the ACL,[60, 86, 143] defined as the angle between the longitudinal axis of the ACL and the tibial plateau.[86] This increase in the ACL elevation angle results in the ACL being oriented more vertically, a greater proportion of ACL loading being shear in nature as opposed to tensile, and a greater ACL strain with a given anterior shear force.[165] Compared with a more flexed knee under the same quadriceps loading conditions; a greater proportion of the quadriceps force is directed anteriorly, and this anterior tibial shear force produces greater ACL strain when the knee is less flexed.

With respect to peak impact forces during landing, several investigators have reported that more erect landing positions result in greater peak impact forces;[13, 34, 117, 127, 144] and that greater peak impact forces may cause greater internal moment demands, knee loading, and ACL injury risk.[28] Further, Hewett et al.[63] found that compared to uninjured females, females suffering ACL injuries prospectively exhibited significantly greater peak impact forces during jump-landings. However, while these results indicate that the impact forces of females should consistently be greater than males due to a more erect landing posture, the existing literature comparing the sexes is equivocal; a result of the fact that impact forces during landing are mediated by both joint motion and multi-joint coordination strategies. Schmitz et al[146] and Salci et al[144] reported greater peak vertical ground reaction forces in females, and lesser hip and knee joint flexion displacements[146] and peak flexion angles[144] during landing. Conversely, McNair and Prapavessis,[103] and Decker et al.[28] did not observe sex differences in peak vertical ground reaction forces. Decker's results are especially noteworthy as females landed in a significantly more erect posture compared to males. However, these females exhibited greater knee and ankle joint displacements; and greater hip, knee, and ankle joint angular velocities during landing: the result of which was a similar peak impact force as males, but achieved using a more erect landing posture. Similarly, Yu et al.[166] found peak impact and anterior tibial shear forces to be significantly associated with hip and knee joint angular velocities, but not joint position at ground contact.

Collectively, these results imply that peak impact forces and ACL loading cannot be adequately explained by evaluating either a single joint or kinematic parameter. Rather, the combination of; 1) initial joint position, 2) joint motion during landing, and 3) the coordinated activity of multiple joints (hip, knee, and ankle) affects the magnitude of the peak impact force during landing. Additionally, impact forces and sagittal plane knee position must be considered together when evaluating the magnitude of ACL loading and injury risk during landing tasks.

2. Muscle Activation

In addition to joint kinematics, the activation patterns of muscles acting in the sagittal plane have been identified as contributors to ACL loading and injury. Compared to males, females have demonstrated greater quadriceps and lesser hamstring activation amplitudes during numerous types of athletic movements.[19, 56, 82, 95, 114, 154, 168, 169] Further, while the hamstrings protect against ACL loading by producing posterior tibial shear force;[87, 98, 131, 140, 160] the quadriceps induces ACL strain by generating anterior tibial shear force.[30, 87, 161] Therefore, although muscle activation and force are not synonymous,[84] many investigators have concluded that the use of a quadriceps dominant activation pattern results in greater quadriceps forces during landing; thus increasing non-contact ACL injury risk.[22, 53]

While there have been many studies on thigh muscle activation patterns, there has been limited research on sagittal plane hip and ankle muscle activation amplitudes. Zazulak et al.[168] found lesser gluteus maximus activation in females during the pre-contact phase of landing. They concluded that lesser gluteus maximus activation may reduce females' ability to dynamically control the hip and pelvis; however, lower extremity kinematics were not measured in this investigation. Hollman et al.[65] reported a significant association between greater gluteus maximus activation and lesser knee valgus angle during a single-leg step-down, and postulated that the inadequate activation resulted in insufficient hip stabilization that contributed to increased knee valgus angle. Unfortunately, there are currently no investigations that have replicated the findings of Hollman et al.[65] in a more dynamic and challenging task; thereby leaving the idea of a relationship between sagittal plane hip muscle activation and knee kinematics primarily theoretical.[71, 116] With respect to sagittal plane ankle muscle activation, Landry et al.[83] reported greater medial and lateral gastrocnemius activation amplitudes during a side-cut task in females compared to males. Although they postulated that the greater gastrocnemius activation could help to dynamically stabilize the knee; the authors also conceded that greater gastrocnemius activation may be deleterious as gastrocnemius forces have been shown in both computer models[125] and *in vivo*[45] to load the ACL.

2.2.3 Frontal Plane Biomechanics

As described previously, valgus loading of the knee in combination with anterior tibial shear force strains the ACL. Greater knee valgus motion has been demonstrated in females compared to males;[46, 47, 74, 132, 142] and knee valgus angle (at initial contact and peak) and external knee valgus moment were found to be significant prospective predictors of non-contact ACL injury.[63] Accordingly, limiting frontal plane knee valgus motion and moments has been advocated to decrease ACL injury risk.[53] While the knee lacks musculature to effectively produce frontal plane knee motion directly, Lloyd and Buchanan[89, 90] have demonstrated that increased co-contraction of the quadriceps and hamstrings muscles contributes to reducing valgus-varus moments at the knee and indicates that sagittal plane mechanisms can contribute to reducing frontal plane knee loading. Additionally, hip adduction angle has been shown to be a significant predictor of knee valgus angle in the closed kinematic chain.[129] Therefore, effective control of frontal plane hip adduction by eccentric action of the hip abductors has been theorized to limit frontal plane hip motion.[48, 70, 71] However, investigations of hip abductor activation amplitudes during landing are equivocal. Hart et al.,[58] but not Zazulak et al.,[168] identified lesser gluteus medius activation in females compared to males with neither of these studies concomitantly reporting hip or knee kinematics or kinetics. Russell et al. [142] also failed to identify a sex difference in gluteus medius activation amplitude during jump landings; despite a significant sex difference in peak frontal plane knee position. As a result, the relationship between hip abductor activation amplitude and knee valgus is currently unclear.

Despite limited evidence to the contrary,[102] it is important to emphasize that pure frontal plane loading does not appear sufficient to cause ACL injury. Yu and Garrett[165] suggest that the magnitude of knee valgus moment reported to be predictive of ACL injury only results in ACL strain when combined with a significant anterior shear force; as the ACL does not substantially contribute to resisting valgus loads while the medial collateral ligament is intact.[7, 100, 101] Further, Fayad et al.[40] evaluated magnetic resonance images and reported no evidence of even grade I medial collateral ligament damage in 55% of confirmed ACL tears. Therefore, while frontal plane loading can certainly contribute to increased ACL strain; non-contact ACL injuries are very much associated with sagittal plane loading.

2.2.4 Summary of Biomechanical Factors Related to ACL Injury

Non-contact ACL injuries during active movement tasks are the result of excessive ACL strain caused by frontal and sagittal plane knee joint forces and moments generated by the individuals themselves.[165] Most often, these tasks are high velocity landings, cuts, and pivots that require the joints of the lower extremity to decelerate the whole body center of mass through coordinated flexion of the ankle, knee, and hip joints.[66] The amount of ACL strain is a function of both sagittal knee joint position and the magnitude of the impact force that must be attenuated by the body during landing.

Erect landing postures are proposed to present greater ACL injury risk as they are associated with greater impact forces, and less favorable patellar tendon-tibial shaft and ACL elevation angles; thereby increasing ACL strain. However, impact forces during landing may be mediated by; 1) initial joint position, 2) joint motion during the landing, and 3) the coordinated activity of these joints (hip, knee, and ankle). Finally, knee valgus angle and moment have been identified as predictors of non-contact ACL injury, though a combination of frontal and sagittal plane loading results in the greatest ACL strain.

A quadriceps dominant activation pattern in females is also thought to contribute to greater ACL loading. However, it should be noted that the magnitude of force produced by muscular contraction is not only determined by the muscle's activation amplitude; but also its maximum force capacity and muscle dynamics (i.e. length-tension and contraction velocity).[22] Considering that compared to females, males have been shown to have greater maximal quadriceps and hamstring strength;[55, 69, 85] a primary limitation with respect to the research on quadriceps dominance is that thigh muscle activation amplitudes and maximal strength are generally not considered together when evaluating landing biomechanics. This is important to note as the resultant force output of a muscle during a task is not determined solely by either the activation amplitude or maximum strength of the muscle; but rather by the relative utilization (i.e. degree of activation) of the underlying maximal strength capacity of that muscle.

With respect to movement patterns and non-contact ACL injury, the primary limitation of previous investigations is that complex multi-joint movement tasks are often evaluated using only a few individual pieces of kinematic and/or kinetic data. In an attempt to gain a more complete picture of complex landing mechanics, it may be more beneficial to combine this kinematic and kinetic information using more comprehensive biomechanical techniques like energetic analysis.

2.3 Approaches to Energetic Analysis

Historically, energetic analysis of human movements has typically involved calculations of external and/or internal work. External work is defined as the mechanical work done on either the body's center of mass or an external load, while internal work is the work done on the body segments resulting in motion of the segments.[145, 158, 167] In analyzing landing biomechanics, internal work is generally calculated for two reasons. First, the foot is generally in contact with a rigid surface, is assumed to not deform, and there is no displacement of the point of force application. Therefore, although there may be large ground reaction forces; these forces do not do external work on the body, but rather work on the segments is done by internal forces.[167] Second, the interest is usually in determining the mechanical work required to produce the observed segment motions. Through the years, there have been several different approaches used to quantify internal work. These approaches and their advantages and limitations have been categorized by Winter[158] and are summarized in the following sections.

2.3.1 Energy Increases in Segments

One of the earliest approaches, pioneered by Fenn,[41] was to calculate the potential and kinetic energies of individual segments during the time period of interest in order to determine the increase in the total energy of each segment. The individual segment energy increases were then summated in an effort to approximate the total internal work. However, this technique neglects two important facts: 1) energy may be transferred between segments passively by the joint forces acting through the joint centers resulting in a redistribution of mechanical energy; and 2) mechanical energy within a segment may transform between potential, linear kinetic, and rotational kinetic energies without changing the overall mechanical energy of the segement.[3, 158, 167] As such, this approach greatly overestimates the actual magnitude of internal work performed.[158]

2.3.2. Center of Mass Approach

An improvement on the increase in segmental energy approach was used by Cavagna and Margaria.[17] In this approach, kinetic and potential energies of the whole body center of mass are calculated during the time period of interest using force plate data. The change in the calculated whole body center of mass energy is then reported to represent the internal work performed. While this approach accounts for the segmental energy transfers and transformations ignored previously, it does not account for the mechanical work required for the "simultaneous increases and decreases in oppositely moving segments"; thereby underestimating the magnitude of total internal work.[158]

2.3.3. Sum of Segment Energies: Fraction Approach

Ralston and Lukin[139] and Winter et al.[159] proposed a technique that accounts for intersegment energy transfers and intrasegment energy transformations like the center of mass approach, but also accounts for the potential for simultaneous increases and decreases in individual segment energy. In this technique, the potential (E_P), translational kinetic (E_{TK}), and rotational kinetic (E_{RK}) energies of each individual segment are summated to calculate the total energy (E_S) of each segment at a given instant in time;

$E_S = E_P + E_{TK} + E_{RK}$.[158, 167]

This summation of the fractions of energy within the segment addresses the potential for energy transformation within a segment that was neglected in previous approaches. Then, the E_S of the individual segments are summated to calculate the total body energy (E_B) which accounts for the energy transfers between segments Finally, total body internal work is calculated by summating the changes in E_B during the time period of interest.[158] However, while an improvement from earlier techniques, the fractions approach has two primary limitations: 1) the calculation underestimates the true negative and positive work done by the muscular system as it fails to adequately account for concomitant energy generation and absorption at different joints;[158] and 2) while this approach quantifies the work done on the segments, the fractions approach does not provide insight into the sources of the internal work.[3, 145, 158, 167]

2.3.4. Joint Power and Work: Source Approach

Currently, the most commonly used approach for energetic analysis of human motion is the joint power and work, or source, approach. In this approach, standard inverse dynamics techniques are used to calculate net internal joint moments (M).[145, 158, 167] The net joint moments are then multiplied by instantaneous joint angular velocities (ω) to derive net joint power (P) curves;

$$P = M \times \omega.$$

Net joint work is then calculated by time integrating the joint power curves. When a net joint moment acts in the same direction as the joint angular velocity, a positive joint power is produced representing concentric muscle contraction.[104, 158, 167] Conversely, an eccentric contraction is indicated by a net joint moment and joint angular velocity acting in opposite directions. As a result, positive joint work is indicative of energy generation via concentric contraction, whereas negative joint work indicates energy absorption from the segments by the muscle-tendon unit via eccentric contraction.[104, 158, 167]

While the source approach to energetic analysis is an improvement over previous approaches, it also has two primary limitations that should be addressed. First, due to the use of net joint moments, the source approach does not quantify the actual "individual muscle contributions to mechanical work".[145] As a result, the total mechanical work of the muscles acting at a joint is underestimated when co-contraction exists and antagonist and agonist muscles act simultaneously to absorb and generate energy.[158] The solution to this error is to accurately quantify individual muscle forces and muscle velocities in order to calculate the individual muscle works and powers.[158] However, currently the techniques to acquire these variables are wanting.

The second limitation of the source approach is that biarticular muscles can act to absorb energy at one joint and generate energy at another joint; a fact known as intercompensation.[3, 167] For instance, immediately following IGC there can be a period of time in which the net joint power at the hip is positive, while the net joint power at the knee is negative.[34] During this time, the rectus femoris can act to generate energy at the hip to accelerate the pelvis anteriorly relative to the thigh, while simultaneously absorbing energy at the knee. As a result of this, researchers have developed methods that allow for net joint power calculations to be adjusted so that energy generated and absorbed at different joints by biarticular muscles may be accounted for and removed from the final joint work calculations.[73] However, recent investigations have indicated in both cycling[115] and normal walking[145] that joint work calculations neglecting intercompensation more closely approximate actual muscle-tendon work than joint work calculations with intercompensation. It is proposed that this underestimation when accounting for intercompensation is due to the fact that all of the negative work at one joint is assumed to be canceled out by positive work at another, or vice versa, by a biarticular muscle; thereby discounting the joint work done by uniarticular muscles.[145] Therefore, neglecting intercompensation of energy by biarticular muscles when using the source approach allows for more accurate estimation of the actual joint work.

2.3.5 Summary of Approaches to Energetic Analysis

There have been numerous approaches used over the years for calculating the internal work of the body to affect segmental motion. While early approaches such as those using energy increases in segments and the whole body center of mass were straightforward; these approaches greatly underestimated total internal work by either not accounting for energy transformations within segments and energy transfers between segments, or neglecting the mechanical work required for the simultaneous increase and decrease in energy of oppositely moving segments. The fractions approach is a marked improvement over these pervious methods, but quantifies the magnitude of work done on the segments rather than the sources of the mechanical energy. As a result, the sources approach for calculating internal work is the most advantageous. While it cannot quantify individual muscle contributions to net joint work secondary to inadequacies in current methods for estimating individual muscle forces and velocities, the source approach can fairly accurately quantify the net work at a joint and indicates whether that joint is generating or absorbing energy from the segments. Though intercompensation between joints by biarticular muscles does occur, previous work suggests that neglecting this effect during energetic calculations results in more representative estimates of actual musculotendinous work.[73, 145] Based upon these findings, this investigation will calculate net internal work using the source approach without accounting for joint intercompensation by biarticular muscles.

2.4 Energetics of Landing

During landing, the kinetic energy of the body is dissipated passively in structures such as bone, ligament and articular cartilage; and actively via eccentric muscle action in a process known as energy absorption or shock attenuation.[23, 81, 107, 124] It is proposed that energy absorption using eccentric muscle action is more significant than by using passive mechanisms;[107] and that greater energy dissipation using active mechanisms may reduce the loading of passive structures and decrease injury risk.[32, 34, 155, 156] Accordingly, previous investigations have been devoted to investigating two central components of energy absorption: 1) the relationships between the magnitude and timing of energy absorption and impact forces during landing; and 2) the biomechanical parameters that may influence energy absorption. The following sections discuss these two facets of landing energetics.

2.4.1 Relationships between the Magnitude and Timing of Energy Absorption and Impact Forces during Landing

The majority of investigations into the relationships between energy absorption magnitude and impact forces during landing have generally utilized sagittal plane analyses using two primary paradigms; either by changing landing height[104, 162-164, 170] or by artificially altering landing technique.[34, 148, 170] From greater landing heights, thereby inducing greater impact velocities, there are increases in sagittal plane energy absorption across all joints--the hip, knee, and ankle.[104, 170] This suggests that as the demands of a task increase, the muscular system responds accordingly in an attempt to attenuate impact forces through increases in negative mechanical joint work.[104, 170] Though peak impact forces increase as landing height increases, the rationale is that these peak impact forces are substantially less than what would be experienced had an increase in the magnitude of energy absorbed by the musculotendinous system not occurred.

The notion of greater magnitude of energy absorption being associated with lesser peak impact forces is also evident in the second paradigm—alteration of landing technique. Typically, investigators have manipulated landings to either be “soft” or “stiff” by instructing participants to augment or limit lower extremity displacements, respectively.[34, 148, 170] Using this paradigm, it has been observed that “soft” landings display greater lower extremity energy absorption and lesser peak impact forces, while “stiff” landings display lesser energy absorption and greater peak impact forces.[34, 148, 170] Therefore, the combined results of investigations using these two paradigms suggest that greater total lower extremity energy absorption is beneficial and reduces peak impact forces.

Unfortunately, a limitation of these previous investigations is that the relationship between energy absorption and peak impact forces has been evaluated using extreme landing techniques that are not typically utilized in everyday human movements. To address this limitation, Norcross et al.[121] compared lower extremity energy absorption between participants who demonstrated significantly different peak impact forces when landing using their preferred technique. It was found that there was no difference in the total energy absorbed during the time from initial ground contact until the minimum vertical position of the whole body center of mass. However, the group exhibiting significantly greater peak vertical ground reaction forces during landing absorbed greater energy during the first 100 ms following ground contact but lesser energy during the terminal phase; suggesting that a temporal relationship exists between energy absorption and peak impact forces.[121]

It is clear from these results that peak impact forces are attenuated by means of eccentric muscle action resulting in greater magnitude of lower extremity energy absorption. Although the relationship between greater energy absorption and lesser peak impact forces is readily apparent in landings using extremely "stiff" or "soft" techniques; this relationship is not evident when evaluating smaller ranges of peak impact forces and energy absorption as seen across subjects using more "normal" landing mechanics. In these situations, it appears that the timing of energy absorption is more influential than the magnitude of energy absorption with respect to peak impact forces. However, it is currently unknown exactly how multiple biomechanical parameters combine to influence both the magnitude and timing of energy absorption during landing.

2.4.2 Biomechanical Parameters and the Influence on Energy Absorption

The primary advantage of energetic analysis is that it combines kinetic (net joint moment) and kinematic (joint angular velocity) information to generate a more comprehensive representation of human movement. However, as a result, the magnitude of energy absorption may be influenced by any number of parameters; and the interaction of these different parameters has not been fully evaluated.

Lafortune et al.[80, 81] reported that increasing knee flexion angle at initial ground contact serves to decrease leg stiffness during running and results in decreased peak impact forces that are potentially indicative of greater energy absorption. However, Mizarahi and Susak[107] have argued that while increasing flexion of the joints at initial contact may decrease leg stiffness; this would effectively limit the available joint ranges of motion during landing and reduce the ability of muscles spanning these joints to absorb energy. This idea is supported by Zhang et al.[170] who observed increased hip, knee, and ankle angular displacements and energy absorption during soft compared to stiff landings. Similarly, McNitt-Gray[104] reported increases in peak hip, knee and ankle angular velocities and energy absorption as impact velocity was increased; and Yeow et al.[163] demonstrated greater knee angular velocity and energy absorption as landing height increased. Collectively, these results imply that modification of initial joint angles at ground contact and/or the magnitude of joint displacements during landing may influence joint angular velocities and the magnitude of lower extremity energy absorption.

In addition to kinematics, energy absorption is also influenced by the magnitude of the net joint moment acting in the opposite direction of the joint angular velocity. Further, the magnitude of this net joint moment is predominantly determined by active muscle forces. As a result, Schmitz and Shultz[147] evaluated the contribution of knee flexor and extensor strength on energy absorption under the premise that greater maximal strength might influence the underlying joint moment production capacity; and thus energy absorption. They found in females that maximum isometric knee extension strength accounted for only 11% of the variance in knee energy absorption, while neither maximum isometric knee flexion nor extension strength were predictive of energy absorption in males.[147] However, a limitation of this investigation is that electromyography was not utilized to quantify the activation magnitudes of the quadriceps and hamstrings. Therefore, it is plausible that maximal muscle strength in combination with voluntary muscle activation may influence joint energy absorption in two ways: 1) by altering the magnitude of the force generated by contraction and thus the active joint moment; or 2) by increasing the joint rotational stiffness,[51] or the resistance to angular motion, that could thereby reduce the joint angular displacement and/or angular velocity.

2.4.3 Summary of Energetics of Landing

The preceding sections have illustrated the influence that both the magnitude and timing of energy absorption have on peak impact forces. Further, it has been shown that energy absorption may potentially be influenced by kinematic, kinetic, strength, and muscle activation parameters- and that these parameters do not function independently. Specifically, the potential influence of initial joint angles at ground contact, joint displacements during landing, maximum strength, and utilization of that strength have been presented; with the evaluation of these same biomechanical parameters in isolation and at discrete time points proposed earlier to be a major limitation of previous research on non-contact ACL injury. However, though energetic analysis of human movement can address this limitation, its use to date specifically with respect to ACL injury has been minimal.

2.5 Energetic Analysis in ACL Injury Research

2.5.1 Sagittal Plane Landing Biomechanics

The first use of energetic analysis of landings specific to non-contact ACL injury was by Decker et al.[27] who compared healthy subjects to ACL reconstructed (ACLr) subjects during 60 cm drop landings. While no difference in peak vertical impact force was identified between groups, the joint coordination strategy utilized to dissipate energy during the 100 ms following impact was dissimilar. ACLr subjects absorbed significantly lesser energy at the hip; but significantly greater energy at the ankle compared to healthy subjects. However, the magnitude of knee energy absorption for both groups was the same. Additionally, the mechanisms underlying the variations in hip and ankle joint energy absorption were also different.

Compared to healthy subjects, the ACLr group demonstrated lesser net internal hip extension moment, but similar hip angular velocity; and similar net internal ankle plantarflexor moment, but greater ankle angular velocity.[27] Further, the ACLr group landed with the hip and ankle significantly more extended. These results reinforce the notion that alterations in either kinematic or kinetic parameters may influence joint energy absorption; and also highlight an important point related to lower extremity landing performance and energy absorption.

Energy absorption during landing is reliant on the coordinated action of the ankle, knee, and hip joints with the effects of this coordinated action reflected in the impact forces. As a result, the relative contributions of each joint to the total energy absorption can change, but the resultant effect may remain the same. This suggests the need to not look at the lower extremity joints in isolation; but rather, the specific contributions of each joint, and the total lower extremity energy absorption should be evaluated.

To this end, Schmitz et al.[146]and Decker et al.[28] compared the energy absorption strategies of males and females during single and double-leg drop landings, respectively. Schmitz et al.[146] reported greater peak vertical ground reaction force, lesser total lower extremity energy absorption, and greater relative contribution of the ankle to total energy absorption in females compared to males. Decker et al.[28] also observed a greater relative contribution of the ankle to total energy absorption in females as well as a lesser relative contribution from the hip; however, no difference in peak impact force was detected between sexes. These differences in relative joint contributions to the total energy absorption imply that feed-forward energy absorption strategies during landings may be sex-specific; with females employing an ankle/knee dominant strategy and males using a hip/knee dominant strategy. However, the initial landing kinematics differed between sexes in this investigation with females displaying a significantly more erect posture than males.[28] Decker et al[28] proposed that this more erect posture in females was the result of the ankle/knee dominant strategy. However, this hypothesis lacks confirmation and it has previously been demonstrated that initial joint positions may influence the magnitude of energy absorbed at different joints.[27, 81, 107] Further, Norcross et al.[118] reported no sex differences in joint contributions to energy absorption when initial landing kinematics were similar between sexes. Therefore, a major gap currently present in the literature is whether sex-specific feed-forward energy absorption strategies exist; or whether the results of Decker et al.[28] are driven by differences in landing posture with alternative biomechanical factors responsible for determining the landing posture utilized.

Additionally, despite numerous suggestions, there is still a lack of substantive evidence that directly connects energy absorption to biomechanical factors related to non-contact ACL injury. To date, only one investigation has demonstrated that this relationship exists. Norcross et al.[119] evaluated the relationships between sagittal plane total lower extremity, hip, knee, and ankle energy absorption; and seven biomechanical factors related to non-contact ACL injury. It was reported that not only the magnitude, but also the timing of energy absorption was important with respect to ACL related biomechanical factors. Specifically, greater energy absorption during the 100 ms following ground contact, encompassing the time of peak ACL strain (40-80 ms) measured *in vivo*,[10, 18] was associated with greater peak vertical ground reaction force, anterior tibial shear force, and hip extension moment; biomechanical factors that are generally considered unfavorable with respect to non-contact ACL injury.[20, 63] However, many of these relationships are not evident when evaluating energy absorption over the entire landing period (initial ground contact to minimum position of the whole body center of mass) suggesting a temporal aspect of energy absorption; and indicating that using this initial landing period is of paramount importance in future work.

2.5.2 Frontal Plane Landing Biomechanics

Finally, Norcross et al.[119] did not identify significant relationships between sagittal plane energy absorption and the two frontal plane ACL-injury risk factors—knee valgus angle and knee valgus moment. However, Pollard et al.[136] recently reported that a group of females displaying greater combined hip and knee flexion displacement; and greater sagittal plane hip and knee energy absorption over the entire landing period had lesser peak knee valgus angle and average internal knee varus moment than a low flexion group. They proposed that reduced sagittal plane energy dissipation resulted in greater use of frontal plane mechanisms for energy absoprtion.[136] This investigation is the first to suggest that there may be an interaction between sagittal and frontal plane energy absorption, and that this interaction might affect frontal plane loading and thus ACL strain. However, as frontal plane energy absorption was not quantified, any potential relationships remain speculative. Further, it is not currently known whether the magnitude of frontal plane energy absorption, reported to be less than 10% of sagittal plane energy absorption,[163, 164] is large enough to significantly affect frontal plane biomechanics.

Despite a developing acceptance of the importance of energetic analysis in evaluating movement patterns that may be associated with increased injury risk; there is a paucity of research evaluating what biomechanical factors predict energetic profiles. Currently, there is only one investigation that has attempted to determine if energy absorption can be predicted using biomechanical factors.[147] This investigation measured maximum isometric knee flexion and extension strength and was able to explain about 11% of the variance in knee energy absorption. However, given the numerous biomechanical parameters that have been purported to influence energy absorption like strength, muscle activation, initial joint angle, and joint displacement; the limited amount of energy absorption explained in this investigation by strength alone is not surprising and signifies the need for a more comprehensive investigation measuring a greater number of these biomechanical factors across multiple joints.

2.6 Synthesis and Conclusions

This review of the literature makes clear that non-contact ACL injuries are a significant financial and social burden; not only in the short term, but in the long term as well.

Previous research has successfully identified mechanisms of ACL loading with combined sagittal and frontal plane loading identified as being most dangerous. While a number of sagittal and frontal plane biomechanical factors thought to contribute to non-contact ACL injury have been identified using sex comparisons over the years, the primary limitation of these investigations are that they usually evaluate individual biomechanical parameters; and at discrete time points. In contrast, energetic analysis of human movement combines kinematic and kinetic information over the movement period providing a more comprehensive quantification of human movement. Despite this obvious benefit, the application of this technique to ACL injury research has been minimal. As a result, there are three primary knowledge gaps that were addressed in this investigation: 1) the biomechanical factors that predict sagittal plane energy absorption profiles previously associated with ACL injury related landing biomechanics and the face validity of classifying ACL injury risk using energetic profiles; 2) the relationship between sagittal and frontal plane energy absorption and frontal plane landing biomechanics associated with ACL injury; and 3) the influences of sex and initial landing posture on sagittal plane energy absorption. It is suggested that this investigation will substantially contribute to the body of knowledge surrounding ACL injury by addressing these gaps in knowledge to identify modifiable biomechanical factors that are predictive of an energetic profile consistent with high-risk landing mechanics (Part I); provide evidence of an association between sagittal and frontal plane energy absorption and frontal plane risk factors for non-contact ACL injury (Part II); and demonstrate that sex differences in landing posture are not the result of sex-specific energy absorption strategies, but rather the energy absorption strategy observed is due to the landing posture utilized (Part III).

CHAPTER THREE

METHODS

3.1 Experimental Design

Subjects reported to the Neuromuscular Research Laboratory at The University of North Carolina at Chapel Hill for one testing session lasting approximately 1.5 hours. During this testing session, all subjects performed three separate assessments. First, subjects performed maximal voluntary isometric contractions (MVIC) of the hip extensors, knee extensors, knee flexors, and ankle plantarflexors during which electromyography (EMG) amplitudes and peak force production were measured. Then, subjects completed; 1) a double leg jump landing task, and 2) drop landings in preferred, erect, and flexed lower extremity joint postures during which lower extremity kinematics, kinetics, and EMG amplitudes were sampled. The order of these two landing tasks was counter-balanced to control for potential effects of fatigue. For all assessments, data was sampled from the dominant lower extremity, defined as the limb used to kick a ball for maximal distance.

Cross-sectional correlational and quasi-experimental designs were used to address the proposed research questions. Three, separate stepwise multiple linear regression analyses were used to identify biomechanical factors that significantly predict sagittal plane, hip, knee, and ankle EA during the initial impact phase of landing (INI) [100 ms immediately following initial ground contact (IGC)][27, 28] **(RQ 1)**. Subjects were then divided into tertiles (27 subjects in the highest and lowest tertiles, and 28 in the middle tertile) based on their total sagittal plane EA during the INI phase of landing. This arrangement of the EA data created groups who exhibit higher, moderate, and lower risk (highest, middle, and lowest tertiles, respectively) landing biomechanics related to ACL injury based on previous work regarding the relationship between EA during the INI phase of landing and biomechanical ACL injury risk factors.[119] A quasi-experimental design (static group comparisons) was used to evaluate differences in knee valgus angle at initial ground contact (IGC), and peak vGRF, pGRF, ATSF, HEM, KEM, knee flexion angle, knee valgus angle, and internal knee varus moment (KVM) during the total landing period (IGC to minimum vertical position of the entire body COM)[79, 170] of the double leg jump landing task between these groups using one-way ANOVA **(RQ 2)**. The highest and lowest EA groups based on total sagittal plane EA were used along with sex as categorical variables to evaluate whether there is a significant association between sex and EA group using a X^2 test of association **(RQ 3).** Simple bivariate Pearson correlation coefficients were calculated to assess the relationships between frontal and sagittal plane total lower extremity, hip, knee, and ankle EA during the INI and TER phase of landing and biomechanical ACL injury risk factors (peak knee valgus angle, hip adduction angle, internal knee varus moment and ATSF, knee valgus angle at IGC) during the double leg jump landing task **(RQ 4A and 4B)**. Subjects were also divided into tertiles (27 subjects in the highest and lowest tertiles, and 28 in the middle tertile) based on their total frontal plane EA during the INI phase of landing with this arrangement creating groups who exhibited higher, moderate, and lower magnitudes of total frontal plane INI EA during the double leg jump landing task. A quasi-experimental design (static group comparisons) was used to evaluate differences in knee valgus angle at initial ground contact (IGC), and peak knee valgus angle, hip adduction angle, vGRF, pGRF, and KVM during the total landing period between these groups using one-way ANOVA **(RQ 4C)**. Two, separate 2 (sex) x 3 (joint) mixed model repeated measures ANOVAs were used to determine the influences of sex and joint on the magnitudes of hip, knee, and ankle INI EA and the relative joint contributions to total INI EA during 0.60 m drop-landings in both preferred **(RQ 5)** and flexed **(RQ 6)** initial landing postures. Sex differences in total lower extremity EA during INI during the preferred **(RQ 5)** and flexed **(RQ 6)** initial landing postures were evaluated using independent samples *t*-tests. Finally, seven, separate 2 (sex) x 2 (landing posture) mixed model repeated measures ANOVAs were used to evaluate the influence of sex and initial landing posture on sagittal plane total lower extremity, hip, knee and ankle EA, and the relative joint contributions to total lower extremity EA during the initial impact phase of landing **(RQ 7)**.

3.2 Subjects

Eighty-two volunteer subjects (41 males, 41 females; age = 20.1 ± 2.4 years; height = 1.74 ± 0.10 m; mass = 70.3 ± 16.1 kg) were included in this research study and recruited from the general population of the University of North Carolina at Chapel Hill. *A priori* power analyses indicate that a sample size of 82 would be sufficient to achieve a statistical power of at least 0.80 with $\alpha = 0.05$ for each research question **(see Section 3.6)**. Subjects were recruited directly from classes, and via informational flyers and e-mail in accordance with the policies of the university's Institutional Review Board (IRB). Criteria for subject inclusion in the study were: 1) 18-30 years of age, 2) healthy and physically active as defined by participation in at least 30 minutes of physical activity a minimum of three times per week, 3) no history of ACL injury, lower extremity surgery, or neurological disorder, and 4) no lower extremity injury within the six months preceding data collection. Prior to participation, all subjects read and signed an IRB-approved informed consent form that described the procedures and risks associated with participation in the study. Following the informed consent process, subject height and mass were measured and used for model generation.

3.3 Instrumentation

3.3.1 Surface Electromyography (EMG) Preparation

This investigation measured EMG amplitudes of the vastus lateralis (VL), vastus medialis (VM), biceps femoris (BF), medial hamstrings (MH), gluteus maximus (GMax), gluteus medius (GMed), lateral gastrocnemius (LG), and medial gastrocnemius (MG) of the dominant leg during MVICs, the double leg jump landing task, and the drop landing task using surface EMG. Prior to data collection, preamplified/active bipolar surface EMG electrodes (DelSys Bagnoli-8, DelSys, Inc., Boston, MA: interelectrode distance = 10 mm; amplification factor = 1,000 – 10,000 (20 – 450 Hz); CMMR @ 60 Hz > 80 dB; input impedance > 10^{15} Ω//0.2pF) were placed over each muscle as described below parallel to the direction of action potential propagation.

For the VL, electrodes were placed 2/3 of the distance from the anterior superior iliac spine (ASIS) of the pelvis to the lateral side of the patella, while the VM electrodes were positioned 80% of the distance from the ASIS to the medial tibiofemoral joint line. The BF and MH electrodes were placed midway between the ischial tuberosity and the lateral and medial condyles of the tibia, respectively. The GMax electrodes were positioned 30% of the measured distance from the second sacral vertebra (S2) to the greater trochanter with the GMed electrodes placed 30% of the distance from the greater trochanter to the iliac crest. Finally, the LG electrodes were positioned 1/3 of the distance from the fibular head to the calcaneus, and the MG electrodes were placed over the most prominent portion of the muscle belly. These electrode placements are in accordance with the widely accepted SENIAM project (Surface Electromyography for the Non-Invasive Assessment of Muscles) guidelines.[149] A reference electrode was placed over the anteromedial aspect of the tibia distal to the tibial tuberosity.

To reduce impedance to the EMG signal and allow for proper electrode fixation, electrode sites were prepared by shaving any hair from the immediate vicinity of the desired electrode location with a hair clipper before lightly abrading the skin with an abrasive pad, and cleansing the skin with isopropyl alcohol. Electrode placement and minimal crosstalk were confirmed using an oscilloscope by performing manual muscle testing while palpating the muscle of interest.[64] All electrodes and wires were secured using pre-wrap and athletic tape.

3.3.2 Kinematic and Kinetic Analysis Preparation

Lower-extremity and trunk kinematics were assessed during the double leg jump landing and drop-landing tasks using an electromagnetic motion capture system (Motion Star, Ascension Technology Corp., Burlington, VT). Electromagnetic tracking sensors were positioned over the third metatarsal of the foot, anteromedial aspect of the shank, and lateral thigh of the dominant leg, as well as on the sacrum and C7 spinous process of the trunk. These sensors were placed over areas of minimal muscle mass, and secured with athletic tape to reduce motion artifact. Global and segment axis systems were established with the positive X axis designated as forward/anteriorly, the positive Y axis leftward/medially, and the positive Z axis upward/superiorly. The dominant lower extremity, pelvis, and thorax were modeled using the MotionMonitor motion analysis software (Innovative Sports Training, Chicago, IL) by digitizing the ankle, knee, and hip joint centers and the T12 spinous process. Ankle and knee joint centers were defined as the midpoint of the digitized medial and lateral malleoli and the medial and lateral femoral condyles, respectively. The hip joint center was predicted by digitizing external landmarks on the pelvis as described by Bell et al.[6] Ground reaction forces during the double leg jump landing and drop-landing tasks were measured using a non-conductive force plate (Bertec 1060-NC, Bertec Corp., Columbus, OH) with the axis system of the force plate aligned with the global axis system.

3.4 Task Protocols

3.4.1 Maximal Voluntary Isometric Contractions (MVICs) Assessment

Subjects performed three 5 second MVIC assessments for hip extension (GMax), hip abduction (GMed), knee flexion (BF and MH), knee extension (VL and VM), and ankle plantarflexion (LG and MG) in standardized testing positions[64] against gravity and a handheld dynamometer with one minute of rest between trials to minimize the risk of fatigue (Figure 2).[120] The testing order for the MVIC assessments was counter-balanced to eliminate the potential for an order effect. A 100 ms moving average was used to identify maximal EMG amplitude during the middle 3 seconds of the MVIC trials following data processing (see description below). The average amplitude across the three trials was used to normalize the EMG activation measured during the landing tasks and allow for comparison of EMG activation amplitudes between subjects. The reliability and precision of this normalization technique is reported is presented in Table 3 ($ICC_{2,1}$ = 0.88 - 0.97, SEM = 0.031 – 0.055 V). Additionally, peak isometric force was recorded during these trials using a handheld dynamometer (Chatillon CSD 300, Amtek, Inc., Largo, FL) and multiplied by segment length to calculate peak torque. The observed reliability and precision of the measured strength data is shown in Table 4 ($ICC_{2,1}$ = 0.93 – 0.96, SEM = 5.49 – 14.43 N*m).

3.4.2 Double Leg Jump Landing Task

Lower-extremity EMG, kinematics, and kinetics were assessed during 5 trials of a double leg jump landing task with at least 30 seconds of rest between trials. Subjects stood atop a 30 cm box placed a horizontal distance equal to 50% of their height behind a force plate. They then jumped forward off the box and performed a double-leg landing with only the dominant foot in contact with the force plate before jumping vertically for maximum height. GMax, GMed, BF, MH, VL, VM, LG, and MG EMG amplitudes, ground reaction forces, and three-dimensional ankle, knee, and hip joint kinematics were assessed simultaneously during the jump landing task. Subjects were provided with a minimum of 3 practice trials to familiarize themselves with the task.

3.4.3 Drop Landing Task

Subjects also completed drop-landings from a height of 0.60 m in three different landing posture conditions [preferred (P), flexed (F), and erect (E)] during which lower extremity kinematics, kinetics, and EMG were sampled. This approximate height has been used previously in a number of studies examining lower extremity energetics.[28, 34, 104, 170] All subjects completed trials in the P condition first so as to prevent potential contamination of their natural landing strategy caused by the artificial F and E landing conditions. Subjects were instructed to lead off of a 0.60 m tall box positioned directly behind the force plate with their dominant foot before rolling off of the box with their non-dominant foot making sure to not lower themselves toward the floor. They then performed a double leg terminal landing with their dominant foot positioned completely on the force plate and their non-dominant foot positioned on the floor next to the force plate.

Following completion of the P condition, subjects completed drop landings in the F and E conditions with the order of these conditions counterbalanced. Drop-landings in the F and E conditions required subjects to land with their knee flexion angle at initial contact within 35 ± 5° and 20 ± 5°, respectively. These target angles were chosen as they are similar to the mean knee flexion angles at initial contact exhibited by male (F) and female (E) subjects during a previous study that demonstrated a sex difference in energy absorption strategy during a 0.60 m drop-landing.[28]

For the F and E conditions, subjects hung from an adjustable, overhead drop bar attached to a wooden support frame positioned around a force plate that served as the landing target for the dominant foot of each subject (Figure 3A). In order to maintain a standardized drop-landing height of 0.60 m and placement of the dominant foot over the force plate following movement of the lower extremity segments to position the knee in the proper testing position, individualized adjustment of the vertical height and horizontal position of the overhead drop bar was performed for each subject in each landing posture condition. The adjustment magnitudes and positions for each subject were calculated as functions of the knee joint angle in each condition with segment lengths derived using subject height and anthropometric tables.[31] The calculation of the adjustment magnitudes and distances was performed using a custom-designed Excel spreadsheet (Microsoft Corp., Redmond, WA).

To facilitate positioning in the desired knee flexion position during F and E trials, subjects were provided with real-time visual and auditory biofeedback regarding their knee flexion angle using the Motion Monitor motion analysis system and a computer monitor. This real-time biofeedback consisted of a line graph with a superimposed target box representing the desired knee flexion position ± 5° of deviation (Figure 3B). As subjects altered their knee joint position, they saw a cursor on the screen move in real-time in response to the joint position alteration. When they successfully positioned the cursor within the target window, a beep was generated that provided auditory confirmation that they were positioned at the desired knee joint angles for the F and E conditions, respectively. Though they only received real-time feedback regarding the knee joint position of their dominant leg, subjects were instructed to move both legs in unison so that positioning of the non-dominant leg was as close to that of the dominant leg as possible. They were also instructed to flex their hips in order to bring their feet beneath their body while their knee is positioned within the target joint angle range in order to create a lower body configuration conducive for landing.

For each F and E trial, subjects hung from the overhead drop bar with their arms straight. They then used the biofeedback to assist them in positioning their knee in the desired joint position and flexed their hips as described previously. Upon successful joint positioning, subjects heard the aforementioned auditory signal indicating that they may let go of the overhead drop bar. Subjects attempted to maintain their knee position during the drop and perform a double leg terminal landing on the floor below. In order for a trial to be deemed successful, the entire dominant foot had to land on the force plate with the non-dominant foot placed on the platform next to the force plate, and the knee flexion angle at IGC must have been within the range specified for the F and E conditions, respectively. Knee flexion angle and vGRF were calculated and displayed using the Motion Monitor software immediately following each trial. All subjects completed a minimum of 3 practice trials and attempted to perform 5 successful testing trials in the P, F, and E conditions with a maximum of 8 testing trials allowed in each condition. Subjects were provided with at least 30 seconds of rest between trials and 2 minutes of rest between conditions to minimize the potential effects of fatigue.

3.5 Data Sampling, Processing, and Reduction

Kinematic and analog (electromyographic and kinetic) data were be sampled at 120 Hz and 1,200 Hz, respectively, using The Motion Monitor motion analysis software. Raw kinematic data were low-pass filtered using a fourth-order, zero-phase-lag Butterworth filter with a cutoff frequency of 10 Hz,[28] time synchronized with the analog data, and then re-sampled to 1,200 Hz. Joint angular positions were calculated based on a right hand convention using Euler angles in a Y (flexion/extension), X' (adduction/abduction), Z" (internal/external rotation) rotation sequence with motion defined about the hip as the thigh relative to the sacrum, about the knee as the shank relative to the thigh, and about the ankle as the foot relative to the shank. Instantaneous joint angular velocities were calculated as the 1st derivative of angular position.

Custom computer software (LabVIEW, National Instruments, Austin, TX) was used for processing and reduction of recorded electromyographic signals. EMG data was corrected for DC bias, bandpass filtered (20-350 Hz, zero-phase-lag, 4th order Butterworth), and smoothed using a root-mean-square (RMS) sliding window function with a time constant of 25 ms. A 100 ms moving average was used to identify maximal EMG amplitude during the middle 3 seconds of the MVIC trials, with the largest 100 ms period representing maximal activation. For jump landing trials, the mean EMG amplitude of the RMS smoothed waveform was calculated during the period from 50 ms before to 50 ms after initial ground contact.

Force plate data were low-pass filtered at 60 Hz (4th order zero-phase lag Butterworth)[79] and combined with kinematic and anthropometric[31] data to calculate the net internal force on the shank at the knee joint and net internal joint moments of force at the hip, knee, and ankle using an inverse dynamics solution[49] within The Motion Monitor software. ATSF, vGRF, and pGRF were designated as positive, while angular conventions assigned positive values for hip extension, hip adduction, knee flexion, knee varus, and ankle extension (plantarflexion).

Custom computer software (LabVIEW, National Instruments, Austin, TX) was used to generate sagittal and frontal plane hip, knee, and ankle joint power curves by multiplying joint angular velocities and net joint moments for each double leg jump landing and drop landing trial ($P = M \times \omega$). Negative mechanical joint work was calculated by integrating the negative portion of the joint power curves [28, 33, 104, 146] during; 1) the initial impact phase (INI) over the 100 ms immediately following initial ground contact (VGRF > 10 N),[27, 28] and 2) the terminal phase of the landing (TER) defined as the interval between 100ms after initial ground contact and COM Min [79] with these negative joint work values representing EA by the muscle-tendon unit.[104, 158] Total lower extremity EA was calculated by summing the negative joint work at each individual joint[33, 146, 170] with the relative contribution of the hip, knee, and ankle to total energy absorption calculated as the EA at the respective joint divided by the total lower extremity EA. All EA values were assigned as positive by convention to simplify their interpretation during data analysis.

Finally, the same custom software was used to identify peak values for vGRF, ATSF, knee flexion angle, internal HEM, internal KVM, knee valgus angle, hip adduction angle, and knee valgus angle at IGC during the TOT landing period of double leg jump landings. Mean values for all dependent variables were calculated across the five trials for each landing task/condition with only the dominant leg of each subject used for data analysis as in previous investigations.[28, 148, 170] vGRF and ATSF were normalized to subject body weight (x BW^{-1}), while all strength measures (peak torques), and internal hip extension, knee extension, and knee varus moments were normalized to the product of subject height and weight (x [BW*Ht]$^{-1}$). EMG activation amplitudes for each individual muscle were expressed as a percentage of MVIC (%MVIC). The calculated %MVIC values for the BF and MH, VL and VM, and LG and MG during the jump landing trials were then averaged to represent the activation amplitudes of the hamstring, quadriceps, and gastrocnemius groups, respectively.[151] Energy absorption variables were expressed as a percentage of the product of subject height and weight (% BW*Ht).[27, 28]

3.6 Statistical Analyses

All data was subjected to a series of consistency checks and screening procedures to insure that they meet inferential statistical assumptions. Statistical significance was established *a priori* as $\alpha \leq 0.05$, and all analyses were conducted using SPSS version 17.0 (SPSS, Inc., Chicago, IL). Power analysis for RQ 1 indicated that a sample of 82 subjects will provide an *a priori* statistical power greater than 0.85 to detect an $R^2 = 0.25$ with a maximum of 6 predictor variables at $\alpha = 0.05$.[137] The predictor variables were entered into the regression models using a step-wise technique with six representing the maximum number of predictor variables for a regression model in this investigation. The order of entry for the predictor variables into each regression equation was based on the highest individual bivariate correlation with each criterion variable. The specific predictor variables used for each regression analysis are shown in Table 4. For each separate regression analysis, all of the identified predictor variables were entered into the regression equation, but only those variables significantly increased the multiple R^2 value of the model remained included in order to construct the most parsimonious regression model. For RQ 2, a power analysis conducted using previously collected data in 27 subjects performing the same double leg jump landing task[119] indicated that a sample size of 82 (27 per group) would allow for *a priori* statistical power of 0.80 with $\alpha = 0.05$ to detect mean differences between groups with the one-way ANOVA model for dependent variables exhibiting medium-large effect sizes ($f >$ 0.35).[137] Table 5 outlines the statistical analyses used for each research question.

CHAPTER FOUR

RESULTS

4.1 Introduction

The complete results of the dissertation are presented below organized by research question. In total, 82 physically active volunteers (41 males, 41 females; age = 20.1 ± 2.4 years; height = 1.74 ± 0.10 m; mass = 70.3 ± 16.1 kg) were tested. However, for some research questions, the number of subjects included in the final analysis differed due to errors with subject performance, data collection, or data reduction. In these instances, descriptive data and an explanation for the use of a smaller subject sample are provided prior to the results.

4.2 Research Question One

Only 77 of the 82 total subjects (40 males, 37 females; Age = 20.1 ± 2.2 years; Height = 1.74 ± 0.10 m; Mass = 70.3 ± 16.1 kg) were included in the initial analysis for research question one due to a lack of medial hamstring EMG data in the first 5 subjects because of equipment error. Data from these 77 volunteers were subjected to a screening procedure to identify any potential outliers prior to performing the regression analyses. As a result of this procedure, 4 participants were excluded from the final regression analyses due to quadriceps and/or hamstrings EMG activation amplitudes that were deemed outliers. Outliers were defined as having a mean value greater than three times the inter-quartile range of values for all subjects. This screening procedure resulted in 73 participants (39 Males, 34 Females) being included in the final analyses. Descriptive statistics for the predictor and dependent variables are shown in Table 7.

4.2.1 Part A: Hip INI EA

The regression analysis for hip EA revealed that the linear combination of greater peak hip flexion (R^2 change = 11.2%, P = 0.004), lesser hip flexion at IGC (R^2 change = 12.1%, P = 0.001), and greater peak hip extension strength (R^2 change = 6.4%, P = 0.014) predicted greater hip EA (R = 0.545, R^2 = 0.297, Adjusted R^2 = 0.266, P < 0.001) with mean gluteus maximus EMG not explaining additional variance (P = 0.388). Table 8 presents the parameter estimates and standardized coefficients for this final regression model.

4.2.2 Part B: Knee INI EA

No biomechanical factors were identified that significantly predict knee EA using a stepwise multiple regression model. As a result, the parameter estimates presented in Table 9 are for a non-significant model in which the variables were forced in using an enter method.

4.2.3 Part C: Ankle INI EA

Table 10 indicates the parameter estimates and standardized coefficients for the regression model that significantly predicted ankle EA. The linear combination of greater ankle extension at IGC (R^2 change = 28.0%, P < 0.001), greater peak ankle flexion (R^2 change = 10.7%, P = 0.001), greater peak ankle extension strength (R^2 change = 6.7%, P = 0.005), and greater mean gastrocnemius EMG (R^2 change = 4.2%, P = 0.020) predicted greater ankle EA (R = 0.704, R^2 = 0.496, Adjusted R^2 = 0.466, P < 0.001).

4.3 Research Questions Two and Three

All 82 volunteer participants were included in the analyses for research questions 2 and 3. Table 11 displays descriptive statistics and frequency counts by sex for the three EA groups, while Table 12 reports the mean time following ground contact that the peak value for each dependent variable occurred. EA group assignment by tertile successfully created three groups with significantly different sagittal plane EA during INI ($F_{2,79}$ = 133.093, p < 0.001) (Table 11). With respect to the biomechanical variables related to ACL injury, we observed significant differences between groups for peak ATSF ($F_{2,79}$ = 4.767, p = 0.011), KEM ($F_{2,79}$ = 11.092, p < 0.001), and pGRF ($F_{2,79}$ = 10.582, p < 0.001) (Table 13). Post hoc testing revealed that that the High group landed with significantly greater peak KEM than both the Moderate group (p = 0.004) and the Low group (p < 0.001). However, no significant difference in KEM was detected between the Moderate and Low EA groups (p = 0.398) (Figure 4). The High group demonstrated significantly greater peak ATSF compared to the Low group (p = 0.009); though no significant differences were noted between the High and Moderate groups (p = 0.113) or the Moderate and Low groups (p = 0.557) (Figure 5). Peak pGRF was also greater in the High group compared to Moderate group (p = 0.001) and the Low group (p < 0.001), but the pGRF of the Moderate and Low groups were not significantly different (p = 0.843) (Figure 6). No EA group differences were noted for any other biomechanical variable of interest (p > 0.05) (Tables 13 and 14). There was also no significant association between sex and High vs. Low EA group assignment (X^2 = 1.20, p = 0.273) (Table 11). However, the EA groups did demonstrate significant differences for peak hip flexion ($F_{2,79}$ = 3.207, p = 0.046) and knee flexion ($F_{2,79}$ = 6.160, p = 0.003) angular velocities, but not for peak dorsiflexion angular velocity during this task.(Table 14). Specifically, the High group exhibited significantly greater hip flexion (p = 0.035) and knee flexion (p = 0.005) angular velocity than the Low group, and the Moderate group displayed significantly greater peak knee flexion angular velocity than the Low group (p = 0.019).

4.4 Research Question Four

All 82 subjects were included in the analyses for research question 4. Table 15 displays the means and standard deviations for frontal and sagittal plane EA during the INI and TER phase of landing. Means and standard deviations for the key biomechanical factors associated with non-contact ACL injury are presented in Table 16.

4.4.1 Parts A and B: Frontal Plane EA and Biomechanical Factor Relationships

Correlation coefficients between total, hip, knee, and ankle EA in the frontal plane during the INI and TER phases of landing; and the biomechanical factors related to ACL injury are shown in Tables 17 and 18, respectively. During the INI phase, significant relationships were identified between total frontal plane EA and frontal plane knee angle at IGC (r = -0.518, $p < 0.001$), peak knee valgus angle (r = -0.662, p < 0.001), peak hip adduction angle (r = 0.462, $p < 0.001$), peak pGRF (r = 0.225, $p = 0.042$), and peak KVM (r = 0.698, $p < 0.001$). Frontal plane knee EA during INI was also significantly associated with frontal plane knee angle at IGC (r = -0.589, $p < 0.001$), peak knee valgus angle (r = -0.732, p < 0.001), peak hip adduction angle (r = 0.462, $p < 0.001$), peak pGRF (r = 0.279, $p = 0.011$), and peak KVM (r = 0.717, $p < 0.001$). These results indicate that greater total and knee frontal plane INI EA is associated with greater knee valgus angle at IGC, peak knee valgus, peak hip adduction, peak pGRF, and peak KVM. Further, greater peak KVM was also related to greater frontal plane ankle INI EA (r = 0.260, $p = 0.018$). There were no other significant relationships between frontal plane ankle and hip INI EA and the biomechanical factors of interest.

During the TER phase of landing, greater total frontal plane EA was significantly associated with greater knee valgus angle at IGC (r = -0.233, $p = 0.035$), peak knee valgus angle (r = -0.457, $p < 0.001$), and peak KVM (r = 0.284, $p = 0.010$); while greater frontal plane knee EA was associated with greater knee valgus angle at IGC (r = -0.450, $p < 0.001$), peak knee valgus angle (r = -0.625 $p < 0.001$), peak hip adduction angle (r = 0.333, $p = 0.002$), and peak KVM (r = 0.446, $p < 0.001$). However, lesser peak hip adduction angle was significantly associated with greater hip TER EA (r = -0.338, $p = 0.002$); and lesser peak pGRF was significantly related to greater frontal plane hip (r = -0.268, $p = 0.015$) and ankle (r = -0.269, $p = 0.014$) TER EA. No other significant relationships were identified between frontal plane EA and the biomechanical factors related to ACL injury.

Table 19 displays the correlation coefficients between sagittal and frontal plane EA during the INI and TER phases of landing. Greater sagittal plane knee IN EA was associated with greater frontal plane hip INI EA (r = 0.301, p = 0.006); and greater sagittal plane ankle INI EA was associated with greater frontal plane ankle INI EA (r = 0.224, p = 0.043). No other significant relationships between frontal and sagittal plane EA during the INI phase were identified. During the TER phase, greater total sagittal plane EA was associated with greater frontal plane total (r = 0.287, p = 0.009), hip (r = 0.314, p = 0.004), and ankle (r = 0.225, p = 0.042) EA. Similarly, greater sagittal plane hip TER EA was significantly related to greater frontal plane total (r = 0.264, p = 0.017), hip (r = 0.287, p = 0.009), and ankle (r = 0.337, p = 0.002) TER EA. Finally, greater sagittal plane knee TER EA was significantly associated with greater total frontal plane (r = 0.244, p = 0.027) and hip (r = 0.270, p = 0.014) TER EA. There were no other significant relationships between sagittal and frontal plane TER EA.

4.4.2 Part C: Frontal Plane EA Group Comparisons

Subject allocation to tertiles based upon total frontal plane INI EA was successful in creating three groups demonstrating high, moderate, and low frontal plane INI EA ($F_{2,79}$ = 55.501, p < 0.001) (Table 20). One-way ANOVA detected significant EA group differences for frontal plane knee angle at IGC ($F_{2,79}$ = 5.782, p = 0.005), peak knee valgus angle ($F_{2,79}$ = 19.874, p < 0.001), peak hip adduction angle ($F_{2,79}$ = 4.529, p = 0.014), peak pGRF ($F_{2,79}$ = 4.030, p = 0.022), and peak KVM ($F_{2,79}$ = 17.883, p = 0.001), but no group differences for peak vGRF ($F_{2,79}$ = 0.444, p = 0.643) (Table 20). Post hoc testing revealed that that the High EA group landed with significantly greater knee valgus angle than the Low EA group (p = 0.003), and displayed significantly greater peak knee valgus angles during landing compared to both the Moderate EA (p < 0.001) and Low EA (p < 0.001) groups. The High EA group also demonstrated significantly greater peak hip adduction angle compared to the Low EA group (p = 0.015), greater peak pGRF compared to the Moderate EA group (p =

0.032), and greater peak KVM during landing than the Moderate EA (p = 0.001) and Low EA (p < 0.001) groups. Additionally, as with the sagittal plane, there were significant EA group differences in peak hip adduction ($F_{2,79}$ = 4.885, p = 0.010) and knee valgus ($F_{2,79}$ = 39.275, p < 0.001) angular velocities (Table 20). The High group exhibited significantly greater peak hip adduction velocity than the Low group (p = 0.007) and greater peak knee valgus angular velocity than both the Moderate (p < 0.001) and Low (p < 0.001) groups.

4.5 Research Questions Five, Six, and Seven

Only 80 of the 82 total subjects (40 females, 40 males) were included in the initial analysis for this arm of the investigation due to excessive noise in the force plate data secondary to a voltage overload in the extended range controller used with the electromagnetic capture system during at least one of the drop landing conditions. Of these 80 subjects, 27 participants (19 females and 8 males) were unable to successfully complete drop landings in both the F and E conditions as their mean knee flexion angle at ground contact for their 5 best trials did not meet the established criteria. Further, 3 males were excluded from performing drop landings from the bar due to concerns over the stability of the wooden frame to support their mass. As a result, 50 participants (21 females, Age = 20.2 ± 2.0 years; Height = 1.66 ± 0.06 m; Mass = 59.7 ± 8.9 kg: 29 males; Age = 21.3 ± 2.3 years; Height = 1.81 ± 0.06 m; Mass = 75.7 ± 6.8 kg) were included in the final analysis. Table 21 displays descriptive statistics for initial contact joint positions and peak flexion angles during the landings, while Table 22 reports EA magnitude and relative joint contributions to total EA stratified by sex.

During the preferred condition, there were no sex differences in hip (t_{48} = 0.726, p = 0.471) or knee (t_{48} = -0.002, p = 0.999) flexion angles at initial contact, but females demonstrated approximately 7.5° more ankle plantarflexion at contact compared to males (t_{48} = -2.409, p = 0.046). With respect to EA magnitude, significant main effects for sex ($F_{1,48}$ = 9.674, p = 0.003) and joint ($F_{2,96}$ = 45.145, p < 0.001) were identified, but there was not a sex x joint interaction ($F_{2,96}$ = 0.961, p = 0.359). Post-hoc testing revealed the knee absorbed a significantly greater magnitude of energy than the hip (p < 0.001) and the ankle (p < 0.001), and that the magnitude of EA at the ankle was significantly greater than the hip (p < 0.001) (Figure 7). A significant main effect of joint ($F_{2,96}$ = 42.702, p < 0.001) was noted for the contribution to total EA during P, but the main effect of sex ($F_{1,48}$ = 2.473, p = 0.122) and sex x joint interaction effect ($F_{2,96}$ = 0.177, p = 0.767) were not significant. The knee contribution to total EA was significantly greater than the ankle (p < 0.001) and hip (p < 0.001) contributions, while the ankle contribution to total EA was significantly greater than the hip contribution (p < 0.001) (Figure 7). Additionally, females performed greater total EA compared to males during this condition (t_{48} = 3.110, p = 0.003).

In the F condition, there were again no sex differences in hip (t_{48} = 0.426, p = 0.672) or knee (t_{48} = 0.574, p = 0.569) flexion angles at initial contact. However, compared to males, females demonstrated approximately 9.5° more ankle plantarflexion at contact (t_{48} = -2.511, p = 0.015). Similar to the P condition, significant main effects for sex ($F_{1,48}$ = 13.709, p = 0.001) and joint ($F_{2,96}$ = 19.600, p < 0.001), but no sex x joint interaction ($F_{2,96}$ = 0.036, p = 0.942) were identified for EA magnitude (Figure 8). The magnitude of ankle EA was significantly greater than the magnitude of hip EA (p = 0.001), and the magnitude of knee EA was greater than the magnitudes of ankle EA (p = 0.002) and hip EA (p < 0.001). A significant main effect of joint ($F_{2,96}$ = 21.233, p < 0.001), but no sex main effect ($F_{1,48}$ = 0.125, p = 0.725) or sex x joint interaction ($F_{2,96}$ = 0.410, p = 0.630) were identified for the joint contributions to EA (Figure 8). Compared to the hip (p < 0.001) and ankle (p = 0.001), the knee was the greatest contributor to total EA with the ankle contribution to total EA greater than the hip contribution (p = 0.001). Females also absorbed greater total energy than males in the F condition (t_{48} = 3.702, p = 0.001) (Figure 9).

Four of the seven 2 (Sex) x 2 (Posture) ANOVA models used to evaluate the influences of sex and landing posture on the dependent variables individually were significant. Significant main effects for posture were identified with the F condition exhibiting greater hip contribution to total EA ($F_{1,48} = 4.082$, $p = 0.049$), lesser ankle contribution to total EA ($F_{1,48} = 11.593$, $p < 0.001$), lesser magnitude of ankle EA ($F_{1,48} = 30.722$, $p < 0.001$), and lesser total EA ($F_{1,48} = 13.063$, $p = 0.001$) compared to the E condition (Figures 3, 5, and 6). Additionally, there was a main effect for sex ($F_{1,48} = 15.170$, $p < 0.001$) with females displaying greater total EA compared to males. No other significant main effects were noted and there were no significant sex x posture interaction effects identified for any outcome measure (Figures 9-12).

CHAPTER FIVE

DISCUSSION

5.1 Introduction

The purpose of this chapter is to provide brief synopses of the primary findings of this investigation as four manuscripts have been included (Appendices A-D) that provide greater detail and in-depth discussion of the results for all research questions except for 4B. For this question, a complete discussion is provided. Rather than addressing each research question individually, related research questions have been grouped and discussed within a single synopsis. Additionally, the synopses are presented in an order that allows for conclusions drawn from one aspect of this investigation to be built upon by subsequent aspects of the dissertation. Finally, the conclusion presents an overall synthesis of the dissertation and discusses how these results make a meaningful contribution to the existing knowledge about ACL injury related landing biomechanics.

5.2 Sagittal plane EA and landing biomechanics: Research Questions 2 and 3

Research questions 2 and 3 are addressed in Manuscript I (Appendix A). The primary finding of this aspect of the dissertation is that individuals absorbing a greater magnitude of energy in the sagittal plane during the INI phase of landing utilize a movement strategy that likely results in greater ACL loading. This is evidenced by the fact that the High EA group exhibited significantly greater peak KEM, ATSF, and pGRF compared to the Low EA group without differences in sagittal plane knee kinematics. The greater KEM and ATSF demonstrated by the High EA group agreed with the hypotheses and have been identified in previous research as contributors to ACL loading. However, there were no significant differences in IGC or peak knee flexion angles between the three EA groups (Table 13). Accordingly, the greater observed sagittal plane knee kinetics, in concert with the same knee kinematics, are most likely indicative of greater ACL loading in the High EA group due to sagittal plane mechanisms.

While the High EA group displayed significantly greater peak pGRF compared to both the Moderate and Low groups (Figure 6), there were no significant differences between groups for peak vGRF (Table 12). This result is in contrast to a previous exploratory investigation in which there was a significant association between peak vGRF and total sagittal plane EA; though, only 19.5% of the variance in vGRF was explained by sagittal plane EA.[119] There is also limited evidence to suggest that the posterior component of the GRF is just as, if not more, important than the vertical component in explaining knee joint loading. Yu et al.[166] reported significant associations between both peak pGRF and vGRF; and ATSF and KEM. However, they found that peak pGRF occurred at the same time as peak ATSF and KEM, and explained 72% and 74% of the variance in these same variables compared to only 26% and 32% of the variance, respectively, for vGRF.[166] Collectively, these results imply that increases in either vGRF or pGRF likely result in greater ACL loading during landing. As such, the greater peak pGRF exhibited by the High EA group further supports the notion that a movement strategy involving greater lower extremity EA during INI increases resultant ACL loading due to sagittal plane mechanisms. However, there were no sagittal plane EA group differences noted for knee valgus angle at IGC, peak knee valgus angle, or peak internal KVM indicating that the magnitude of sagittal plane INI EA does not influence frontal plane knee loading.

Finally, it is apparent that quantification of total sagittal plane EA to infer non-contact ACL injury risk is unfounded as there was not a significant association between EA group assignment (High vs. Low) and sex. Given the overwhelming evidence indicating the greater risk of ACL injury in females,[37, 52] it would be expected that there would be a greater proportion of females assigned to the High EA group if this measure was indeed indicative of injury risk. However, this result also indicates that males and females have an equal likelihood of utilizing a landing strategy (High EA) that results in greater ACL loading due to sagittal plane mechanisms.

5.3 Frontal plane EA, landing biomechanics, and inter-planar EA relationships: Research Question 4

The principal findings of the frontal plane EA analyses are that greater frontal plane EA during the INI phase of landing is associated with a less favorable frontal plane biomechanical profile that likely contributes to greater ACL loading. Additionally, there is not a significant inter-planar EA relationship such that greater sagittal plane EA mitigates the magnitude of frontal plane EA in the 100 ms immediately following ground contact during double-leg jump landings. Research questions 4A and 4C are addressed in Manuscript II (Appendix B), while research question 4B is discussed in detail in this section.

The associations between frontal plane EA and the biomechanical factors of interest generally agreed with our hypotheses. As expected, greater frontal plane total and knee INI EA were significantly associated with less desirable values for all biomechanical factors except for peak vGRF (Table 16). However, similar associations between frontal plane hip and ankle INI EA and the biomechanical factors of interest were not observed. At these joints, only greater ankle INI EA was correlated with greater peak KVM, and the strength of this association was relatively weak (r = 0.260).

The results related to frontal plane EA during the TER phase of landing were less clear, however. We observed a similar pattern in that greater frontal plane total and knee EA were associated with greater frontal plane knee angle at IGC, peak knee valgus angle, and peak KVM. Additionally, greater frontal plane knee TER EA was associated with greater peak hip adduction angle (Table 17). However, it is interesting to note that in each of these cases the strength of the association between the biomechanical factors and TER

EA measure is weaker than for the INI EA measure. There are also three instances in which the relationship between frontal plane TER EA and frontal plane biomechanics related to ACL injury would be considered favorable: 1) greater hip EA and lesser peak hip adduction, 2) greater hip EA and lesser peak pGRF, and 3) greater ankle EA and lesser peak pGRF. It has been proposed that increasing frontal plane hip stiffness, or stability, during landing might reduce frontal plane knee loading and subsequent ACL loading[21] due to the fact that hip adduction angle is a significant predictor of knee valgus angle.[130] While we did observe that greater frontal plane hip TER EA was associated with lesser peak hip adduction, we failed to identify a relationship between hip EA and frontal plane knee biomechanics. Therefore, while we suggest that greater total frontal plane and knee EA during the TER phase are related to unfavorable frontal plane biomechanics like in the INI phase; the influence that greater hip and ankle TER EA may have on frontal plane biomechanics is still unclear and should be investigated in future studies.

The results of the frontal plane INI EA group comparisons generally agreed with the hypotheses. Greater knee valgus angle at IGC, peak knee valgus and hip adduction angles, peak KVM, and peak pGRF were identified in the High EA group compared to the Low EA group with only peak vGRF not differing between the EA groups (Table 19). However, the lack of group differences in peak vGRF was not surprising given our previous sagittal plane analyses[122] and the fact that investigations comparing peak vGRF between sexes (i.e. higher and lower ACL injury risk) are generally equivocal.[28, 103, 144, 146] While the investigation is clearly limited in drawing any conclusions regarding injury outcome, it is apparent that the High frontal plane EA group displayed frontal plane knee biomechanics that are sufficiently different than the Low frontal plane EA group to potentially result in greater frontal plane knee loading. Hewett et al.[63] reported that females who went on to suffer a non-contact ACL injury demonstrated 8.4° more knee valgus angle at IGC, 7.6° greater peak knee valgus angle, and about 2.5 times more frontal plane knee moment than uninjured females. By comparison, the High EA group displayed 6.6° more knee valgus angle at IGC, 14.4° greater peak knee valgus angle, and about 2.1 times greater frontal plane knee moment compared to the Low EA group. To evaluate this notion, a secondary analysis was performed to determine whether there was a significant association between sex and total frontal plane INI EA group assignment (High vs. Low) given the increased risk of non-contact ACL injury in females.[52] A significant association between sex and frontal plane INI EA group (X^2 = 4.909, p = 0.027) was identified with females being 3.6 times more likely to be in the High INI EA group. Accordingly, it is likely that landing strategies with greater total frontal plane INI EA are likely to cause greater ACL loading due to frontal plane mechanisms.

Finally, the lack of a consistent association between frontal and sagittal plane EA was unexpected and differed from the hypotheses as it was anticipated that greater sagittal plane EA would mitigate the magnitude of frontal plane EA required during landing (Table 18). In contrast, during the TER phase of landing, greater total, hip, and knee sagittal plane EA were associated with greater total and hip frontal plane TER EA. Further, greater total and hip sagittal plane TER EA were related to greater frontal plane ankle TER EA. These results are interesting in that frontal plane hip and ankle TER EA were the only frontal plane EA measures to be associated with more favorable frontal plane landing biomechanics (Table 17); and greater sagittal plane TER EA is indicative of a more favorable sagittal plane biomechanical profile. However, it should be noted that the relatively weak strength of these inter-planar relationships and aforementioned lack of influence of frontal plane hip EA on frontal plane knee biomechanics marginalize the value of these findings. Apart from relatively weak associations between sagittal plane knee and frontal plane hip (r = 0.301) and sagittal and frontal plane ankle (r = 0.224) EA, there were no significant relationships identified between the magnitudes of sagittal and frontal plane EA during INI indicating that the magnitude of INI EA in the sagittal plane does not necessarily influence the magnitude of frontal plane INI EA.

5.4 Derivations of sex-specific EA strategies: Research Questions 5, 6, and 7

Research questions 5-7 are addressed in Manuscript III (Appendix C). The primary findings of this part of the investigation are: 1) the lack of sex-specific EA strategies during drop landings when the initial landing postures of males and females are similar; and 2) that altering landing posture (i.e. knee flexion angle at ground contact) influences both EA magnitude and the relative joint contributions to total EA; but that sex does not modify these changes.

Contrary to the hypotheses, no significant differences in initial hip or knee flexion angles were identified between males and females when performing drop landings using a preferred landing strategy. However, females made contact with the ground in approximately 7.5° more plantarflexion than males which is consistent with previous research.[28] When using this similar preferred initial landing posture, there was not a sex difference in the relative joint contributions to total EA (i.e. sex-specific EA strategies), as all subjects exhibited the greatest contribution to total EA from the knee, a secondary contribution from the ankle, and a tertiary contribution from the hip.

There were also no sex-specific joint EA patterns detected during landings in the F condition; during which subjects were manipulated to land with the same flexed lower extremity configuration. In these landings, there were no sex differences in hip and knee flexion angles at initial contact; but females again exhibited greater ankle plantarflexion at contact (approximately 9.5°) compared to males. As with the P condition, we observed no differences between males and females in the relative joint contributions to total EA (Knee>Hip>Ankle), and sex did not modify the relative joint contributions to EA (Figure 8). Collectively, the results from the P and F conditions provide strong evidence that sex-specific feed-forward EA strategies do not exist as there were not different EA strategies in males and females when performing landings using both preferred and constrained landing postures.

The results comparing the F and E postures further indicate that altering initial landing posture significantly influences both the magnitude and relative contributions of select joints to total EA, but that these changes are not modified by sex (Figures 10-12). Compared to the E condition, drop landings in the F condition resulted in significantly greater hip and lesser ankle contributions to total EA for both males and females.

5.5 Biomechanical predictors of sagittal plane EA: Research Question 1

The primary finding of this aspect of the dissertation is that the combination of multi-factorial biomechanical factors is predictive of EA at the hip and ankle, but not at the knee with these results addressed in Manuscript IV (Appendix D). This suggests that interventions aimed at reducing total lower extremity EA and thereby potentially decreasing knee joint loading during landing must facilitate changes across the entire kinetic chain.

The greatest contributor to total EA during the double leg jump landing task was the knee, accounting for 65% of the total energy absorbed during the initial 100 ms following ground contact (Figure 13). Interestingly, however, the knee was the only joint for which EA could not be predicted using kinematic, strength, and muscle activation factors (Table 9). At the hip, we found that greater EA was predicted by greater peak hip flexion, lesser hip flexion at IGC, and greater peak isometric hip extension strength (Table 8). Similar to the hip, greater ankle EA was predicted by greater ankle extension (plantarflexion) at IGC, peak ankle flexion, peak isometric extension strength, and mean gastrocnemius EMG (Table 10).

The finding related to greater ankle extension at IGC is particularly interesting as more extended ankle and knee positions along with greater ankle EA has been noted previously in females;[28] a population with a greater risk of ACL injury.[52] Additionally, Blackburn and Padua[12] have noted lower extremity joint coupling in both males and females whereby flexion or extension displacements at one joint facilitates this same greater angular displacements at the other joints. Therefore, a secondary analysis was performed to determine whether this kinematic coupling across joints of the lower extremity was present with respect to joint positions at IGC. We found that this within limb coordination was present as lesser hip flexion angle at IGC was associated with lesser knee flexion angle at IGC (r = -0.398, $P < 0.001$); and greater ankle extension angle at IGC was associated with more extended knee (r = -0.584, $P < 0.001$) and hip (r = 0.349, $P = 0.002$) positions at IGC. We propose that the more extended lower extremity posture is the result of a "reaching" strategy in which the knee and hip are concomitantly positioned in greater extension as the ankle is extended to "reach" toward the ground during landing. This suggests that a more comprehensive strategy aimed at instructing participants to increase flexion angles at IGC in all three lower extremity joints may be needed to effectively facilitate greater knee flexion angles at IGC.

Based upon these results, it is recommended that increasing hip and ankle flexion angles, and reducing gastrocnemius activation at IGC may reduce the magnitude of lower extremity EA during the 100 ms after ground contact. Interventions aimed at incorporating verbal and visual feedback along with technique instruction may be most effective in altering initial hip and ankle kinematics as previous investigators have successfully increased knee flexion angle at IGC using these techniques.[24] Additionally, it is suggested that by increasing ankle flexion there will be an associated increase in knee and hip flexion due to kinematic coupling of the lower extremity. However, it is currently unknown whether an intervention to reduce gastrocnemius activation at IGC is viable.

5.6 Conclusion

The overarching idea of this dissertation was to utilize a more comprehensive type of biomechanical analysis to expand the current body of knowledge with respect to ACL injury. This goal was achieved by: 1) elucidating new knowledge regarding modifiable biomechanical factors that contribute to sagittal plane EA patterns that have been associated with high risk landing biomechanics related to ACL injury; 2) exploring relationships between lower extremity EA in the frontal and sagittal planes, and ACL-related landing biomechanics; and 3) clarifying previous research regarding potential sex differences in lower extremity EA strategies.

The results of this investigation provide significant information for understanding the way in which multi-joint lower extremity movement patterns during landing, quantified using EA analyses, affects ACL loading. It was demonstrated that greater sagittal plane EA in the 100 ms following ground contact is indicative of a biomechanical landing profile with greater peak internal knee extension moment, anterior tibial shear force, and pGRF that likely results in increased ACL loading due to sagittal plane mechanisms. Further, no association between sex and sagittal plane INI EA group was identified, signifying that there is an equal likelihood for males and females to land using this deleterious sagittal plane strategy. It was also demonstrated that greater frontal plane INI EA was indicative of frontal plane landing biomechanics that likely increase ACL loading due to purely frontal plane mechanisms, and that females were 3.6 times more likely than males to exhibit higher frontal plane INI EA during landing. However, there was not a significant relationship between the magnitudes of sagittal and frontal plane EA during the INI phase of landing, indicating that these values are independent of one another. Given these findings, it is suggested that individuals who absorb a higher magnitude of energy in both the sagittal and frontal planes immediately following ground contact would be at the highest risk of non-contact ACL injury, as they would experience greater combined sagittal and frontal plane ACL loading. Further, the increased risk of ACL injury noted in females may be due to the fact that females are significantly more likely than males to land with higher frontal plane INI EA, but just as likely to land with high sagittal plane INI EA, which would increase their likelihood of being subjected to greater combined sagittal and frontal plane ACL loading.

In addition, this investigation was the first to predict the magnitude of sagittal plane hip and ankle EA that occurs during the time interval in which ACL rupture likely takes place using biomechanical factors that are modifiable. These results are important in that specific biomechanical factors were identified that may be targeted in an effort to reduce the magnitude of sagittal plane INI EA and potentially reduce ACL loading. Based upon these findings, it is suggested that increasing hip and ankle flexion, and decreasing gastrocnemius activation at IGC should be addressed via technique training to modify high risk sagittal plane landing biomechanics. Further, it was elucidated that lesser ankle extension at IGC may facilitate greater knee and hip flexion angles at impact through a kinematic coupling of the joints of the lower extremity.

Finally, it was demonstrated that initial landing posture, rather than sex, influences both the magnitude of sagittal plane EA during landing as well as the relative joint contributions to total sagittal plane EA. Compared to an erect landing posture, males and females both demonstrate lesser ankle and total EA, lesser ankle contribution to total EA, and greater hip contribution to total EA when landing in a flexed posture. Further, after controlling for initial landing kinematics, no sex differences in EA strategy were identified indicating that sex-specific feed-forward EA strategies do not exist. Therefore, the more erect landing posture of females that has been reported in the literature is most likely influenced by another sex-related factor such as strength with future research necessary to elucidate this factor(s).

As a whole, the knowledge gleaned from this investigation adds to the current body of ACL injury literature by making a meaningful contribution with respect to landing biomechanics related to ACL injury as well as providing much-needed evidence for specific biomechanical factors that should be targeted in ACL injury prevention programs.

Table 1. Comparison of peak vertical ground reaction force (vGRF), and sagittal plane energy absorption during double leg jump landings using a preferred technique. (Adapted from Norcross et al.[121])

	vGRF	INI EA	TER EA	TOT EA	% EA during INI
High vGRF	3.25 ± 0.33	14.26 ± 3.04	2.83 ± 0.98	17.09 ± 3.02	83.07 ± 6.04
Low vGRF	1.97 ± 0.28	11.24 ± 2.18	4.66 ± 2.47	15.90 ± 3.24	71.82 ± 12.01
P	< 0.001	0.020	0.051	0.407	0.020

Table 2. Significant Pearson bivariate correlations between sagittal plane energy absorption and biomechanical factors related to ACL injury. (Adapted from Norcross et al.[119])

	Total INI EA	Total TER EA	Hip INI EA	Knee TER EA	Ankle INI EA
vGRF	$r = 0.442$ $P = 0.021$	$r = -0.534$ $P = 0.004$	$r = 0.771$ $P < 0.001$	$r = -0.571$ $P = 0.002$	
ATSF	$r = 0.747$ $P < 0.001$		$r = 0.479$ $P = 0.011$		$r = 0.529$ $P = 0.005$
HEM	$r = 0.422$ $P = 0.028$	$r = -0.413$ $P = 0.032$	$r = 0.807$ $P < 0.001$	$r = -0.486$ $P = 0.010$	

Table 3. Intraclass correlation coefficients (ICC 2,1) and standard errors of measurement (SEM) of mean voltage during MVICs measured using surface electromyography

Muscle	**ICC (2,1)**	**SEM (V)**
Gluteus Maximus	0.97	0.037
Vastus Lateralis	0.92	0.055
Vastus Medialis	0.89	0.054
Biceps Femoris	0.91	0.043
Medial Hamstrings	0.89	0.039
Lateral Gastrocnemius	0.94	0.037
Medial Gastrocnemius	0.88	0.031

Table 4. Intraclass correlation coefficients (ICC 2,1)and standard errors of measurement (SEM), and SEM expressed as a percentage of mean isometric strength during MVICs measured using handheld dynamometry

Joint Motion	ICC (2,1)	SEM (N*m)	SEM as % of Mean
Hip Extension	0.95	9.69	8.02
Knee Extension	0.95	14.43	8.33
Knee Flexion	0.96	5.49	7.31
Ankle Extension	0.93	5.80	5.83

Table 5. Specific predictor variables for the multiple stepwise regression analyses

Dependent Variable	Predictor Variables
Hip EA	Hip Position at IGC Peak Hip Flexion Peak Hip Extension Strength Mean Gluteus Maximus EMG
Knee EA	Knee Position at IGC Peak Knee Flexion Peak Knee Extension Strength Peak Knee Flexion Strength Mean Quadriceps EMG Mean Hamstrings EMG
Ankle EA	Ankle Position at IGC Peak Ankle Flexion Peak Ankle Extension Strength Mean Gastrocnemius EMG

Table 6. Statistical analysis plan by research question

RQ	Description	Data Source	Method
1	To identify predictors of EA during INI	<u>Criterion Variables:</u> A. Hip EA during INI B. Knee EA during INI C. Ankle EA during INI <u>Predictor Variables:</u> A. Hip extension strength, mean GMax activation amplitude, sagittal plane hip position at IGC, and peak hip flexion B. Knee extension and flexion strength, mean quadriceps and hamstring activation amplitude, knee position at IGC, and peak knee flexion C. Ankle extension strength, mean gastrocnemius activation amplitude, ankle position at IGC, and peak ankle flexion	<u>Stepwise Multiple Linear Regression</u> - A stepwise selection method was used for model generation with collinearity assessed using the variance inflation factor (VIF) - Overall R^2, and unstandardized and standardized β-coefficients and their associated *P* values will be reported
2	To determine if biomechanical factors related to ACL injury differ between groups exhibiting high, moderate, and low magnitudes of sagittal plane INI EA	<u>Dependent Variables:</u> Peak: - vGRF - pGRF - ATSF - internal HEM - knee flexion angle - knee valgus angle -internal KVM Knee valgus angle at IGC <u>Independent Variable:</u> Group (defined by Total sagittal plane INI EA tertiles)	<u>One-way ANOVA</u> - Static comparison across groups for each dependent variable -Pairwise comparisons were conducted using Tukey's HSD following significant ANOVA models

3	To determine if sex and higher-risk landing biomechanics related to ACL injury group are significantly associated	<u>Categorical Variables:</u> Sex ACL related landing biomechanics group (High and Low) based upon total EA during INI	<u>X^2 test of association</u>
4A and 4B	To assess the relationships between frontal plane EA, sagittal plane EA, and biomechanical factors related to ACL injury	<u>Predictor Variables:</u> Total, hip, knee, and ankle frontal plane EA during INI and TER phases <u>Criterion Variables:</u> Peak: - vGRF - pGRF - knee valgus angle - internal KVM - hip adduction angle Knee valgus angle at IGC Total, hip, knee, and ankle sagittal plane EA during INI and TER	<u>Pearson correlation</u> - Separate bivariate correlations

4C	To determine if biomechanical factors related to ACL injury differ between groups exhibiting high, moderate, and low magnitudes of frontal plane INI EA	<u>Dependent Variables:</u> Peak: - vGRF - pGRF - knee valgus angle - hip adduction angle -internal KVM Knee valgus angle at IGC <u>Independent Variable:</u> Group (defined by Total frontal plane INI EA tertiles)	<u>One-way ANOVA</u> - Static comparison across groups for each dependent variable - Pairwise comparisons were conducted *post hoc* using Tukey's HSD following significant ANOVA models
5	To evaluate sex differences in sagittal plane EA during landings using preferred initial landing postures	<u>Dependent Variables</u> A. - Relative contributions of the hip, knee, and ankle to total EA -Hip, knee, and ankle EA B. Total lower extremity EA	A. <u>ANOVA</u> -Two, separate 2 (Sex) x 3 (Joint) repeated-measures ANOVAs Planned pairwise comparisons were conducted post hoc using a Bonferroni correction for multiple *t*-tests following significant ANOVA models B. <u>Independent samples *t*-test</u>

6	To evaluate sex differences in sagittal plane EA when controlling for initial landing posture (Flexed Posture)	<u>Dependent Variables</u> A. - Relative contributions of the hip, knee, and ankle to total EA -Hip, knee, and ankle EA B. Total lower extremity EA	A. <u>ANOVA</u> -Two, separate 2 (Sex) x 3 (Joint) repeated-measures ANOVAs Planned pairwise comparisons were conducted post hoc using a Bonferroni correction for multiple *t*-tests following significant ANOVA models B. <u>Independent samples *t*-test</u>
7	To evaluate the influences of landing posture and sex on sagittal plane EA	<u>Dependent Variables</u> - Relative contributions of the hip, knee, and ankle to total EA - Hip, knee, and ankle EA -Total lower extremity EA <u>Independent Variables</u> - Sex - Landing Posture (Flexed and Erect)	<u>ANOVA</u> Seven, separate 2 (Sex) x 2 (Landing Posture) repeated-measures ANOVAs Planned pairwise comparisons were conducted post hoc using a Bonferroni correction for multiple *t*-tests following significant ANOVA models

Table 7. Descriptive statistics for initial hip, knee, and ankle EA and biomechanical predictor variables

Variable	Mean	SD
Hip EA (%BW*Ht)	2.35	1.35
Knee EA (%BW*Ht)	8.95	2.65
Ankle EA (%BW*Ht)	2.47	1.59
Hip Position at IGC (°)	-34.4	10.3
Peak Hip Flexion (°)	-67.9	16.3
Peak Hip Extension Strength (x[BW*Ht]$^{-1}$)	0.101	0.023
Mean Gluteus Maximus EMG (%MVIC)	37.3	27.9
Knee Position at IGC (°)	23.3	8.47
Peak Knee Flexion (°)	92.3	15
Peak Knee Extension Strength (x[BW*Ht]$^{-1}$)	0.145	0.03
Peak Knee Flexion Strength (x[BW*Ht]$^{-1}$)	0.063	0.014
Mean Quadriceps EMG (%MVIC)	120.6	53.6
Mean Hamstrings EMG (%MVIC)	33.4	18.8
Ankle Position at IGC (°)	36.8	18.1
Peak Ankle Flexion (°)	-15.7	7.4
Peak Ankle Extension Strength (x[BW*Ht]$^{-1}$)	0.087	0.018
Mean Gastrocnemius EMG (%MVIC)	111.4	68.8

Table 8. Regression coefficients from the final model when predicting INI hip EA from the biomechanical variables (Significant coefficient if $P < 0.05$)

Variable	Parameter Estimate	SE	Standardized Coefficient	t Value	P Value
Hip EA					
Intercept	-0.402	0.881		-0.457	0.649
Peak Hip Flexion	-0.043	0.010	-0.518	-4.478	<0.001
Hip Position at IGC	0.049	0.015	0.377	3.247	0.002
Peak Hip Extension Strength	15.211	6.053	0.254	2.513	0.014

Table 9. Regression coefficients from the final model when predicting INI knee EA from the biomechanical variables (Significant coefficient if $P < 0.05$)

Variable	Parameter Estimate	SE	Standardized Coefficient	*t* Value	*P* Value
Knee EA					
Intercept	7.999	2.661		3.006	0.004
Mean Quadriceps EMG	-0.009	0.007	-0.181	-1.318	0.192
Peak Knee Extension Strength	-16.190	16.670	-0.181	-0.971	0.335
Peak Knee Flexion Strength	11.518	35.885	0.060	0.321	0.749
Mean Hamstrings EMG	0.018	0.019	0.126	0.960	0.341
Knee Position at IGC	0.036	0.042	0.114	0.846	0.401
Peak Knee Flexion	0.024	0.024	0.137	0.999	0.322

Table 10. Regression coefficients from the final model when predicting INI ankle EA from the biomechanical variables (Significant coefficient if *P* < 0.05)

Variable	Parameter Estimate	SE	Standardized Coefficient	*t* Value	*P* Value
Ankle EA					
Intercept	-3.124	0.817		-3.822	<0.001
Ankle Position at IGC	0.049	0.008	0.557	5.804	<0.001
Peak Dorsiflexion	-0.065	0.020	-0.306	-3.317	0.001
Peak Ankle Extension Strength	25.418	7.677	0.291	3.311	0.001
Mean Gastrocnemius EMG	0.005	0.002	0.216	2.385	0.02

Table 11. Sagittal plane EA descriptives and frequency counts by sex

	Energy Absorption Group		
	High	Moderate	Low
Mean ± SD (%BW*Ht)	16.99 ± 1.85*†	13.37 ± 0.78*	10.50 ± 1.57
95% CI	(16.25, 17.72)	(13.07, 13.68)	(9.88, 11.12)
Males	13	11	17
Females	14	17	10
Total	27	28	27

Significantly different from Low EA (*) and Moderate EA (†) groups, $P < 0.05$

Table 12. Time instance following initial ground contact for the occurrence of peak biomechanical variables during the double leg jump landing task

Variable	Mean (ms)	SD
vGRF	32.7	8.2
pGRF	22.0	13.4
ATSF	110.7	67.4
HEM	41.1	26.1
KEM	57.7	34.2
KVM	52.0	37.3
Knee Valgus	108.0	45.8
Hip Adduction	112.0	63.3

Table 13. Sagittal plane EA group comparisons for kinetic variables (High EA group significantly different from Low (*) and Moderate (†) EA groups, $P < 0.05$)

Variable	EA Group	Mean ± SD	95% CI	$F_{(2,79)}$	P-value
vGRF (xBW^{-1})	High	2.94 ± 0.66	(2.68, 3.21)	0.102	0.903
	Moderate	2.86 ± 0.89	(2.51, 3.21)		
	Low	2.94 ± 0.82	(2.62, 3.26)		
pGRF (xBW-1)	High*†	0.96 ± 0.27	(0.86, 1.07)	10.582	< 0.001
	Moderate	0.74 ± 0.20	(0.67, 0.82)		
	Low	0.71 ± 0.18	(0.64, 0.78)		
ATSF (xBW-1)	High*	1.01 ± 0.14	(0.96, 1.07)	4.767	0.011
	Moderate	0.92 ± 0.19	(0.85, 1.00)		
	Low	0.87 ± 0.17	(0.81, 0.94)		
HEM (x[BW*Ht]-1)	High	0.29 ± 0.13	(0.24, 0.35)	0.580	0.562
	Moderate	0.28 ± 0.13	(0.23, 0.34)		
	Low	0.32 ± 0.13	(0.27, 0.37)		
KEM (x[BW*Ht]-1)	High*†	0.21 ± 0.05	(0.19, 0.23)	11.092	< 0.001
	Moderate	0.17 ± 0.05	(0.16, 0.19)		
	Low	0.16 ± 0.03	(0.15, 0.17)		
KVM (x[BW*Ht]-1)	High	0.08 ± 0.05	(0.06, 0.11)	0.027	0.973
	Moderate	0.08 ± 0.03	(0.07, 0.10)		
	Low	0.09 ± 0.05	(0.07, 0.11)		

Significant at $P < 0.05$

Table 14. Sagittal plane EA group comparisons for kinematic variables (Low EA group significantly different from High (*) and Moderate (†) EA groups, *P* < 0.05)

Variable	EA Group	Mean ± SD	95% CI	$F_{(2,79)}$	P-value
Sagittal plane knee angle at IGC (°)	High	22.73 ± 6.96	(19.98, 25.49)	0.015	0.471
	Moderate	23.11 ± 8.92	(19.65, 26.57)		
	Low	23.03 ± 9.60	(19.23, 26.83)		
Frontal plane knee angle at IGC (°)	High	-7.73 ± 8.17	(-8.85, -1.85)	0.760	0.985
	Moderate	-7.34 ± 5.53	(-9.49, -5.20)		
	Low	-6.81 ± 7.60	(-10.96, -4.50)		
Peak knee flexion angle (°)	High	93.82 ± 14.16	(73.91, 99.42)	1.143	0.324
	Moderate	91.15 ± 14.75	(71.72, 96.87)		
	Low	87.74 ± 15.47	(61.96, 93.86)		
Peak knee valgus angle (°)	High	-14.37 ± 11.15	(-18.78, -9.96)	1.310	0.276
	Moderate	-18.12 ± 8.86	(-21.56, -14.69)		
	Low	-18.57 ± 11.35	(-23.06, -14.08)		
Peak hip flexion velocity (°/s)	High*	-346.03 ± 74.14	(-375.36, -316.70)	3.207	0.046
	Moderate	-314.41 ± 103.62	(-354.59, -274.23)		
	Low	-284.56 ± 86.71	(-318.87, -250.26)		
Peak knee flexion velocity (°/s)	High*	699.15 ± 77.95	(668.32, 729.99)	6.160	0.003
	Moderate†	686.35 ± 103.81	(646.09, 726.60)		
	Low	618.24 ± 89.46	(582.85, 653.63)		
Peak ankle dorsiflexion velocity (°/s)	High	-662.74 ± 146.74	(-720.80, -604.69)	2.400	0.097
	Moderate	-606.47 ± 288.37	(-718.29, -494.65)		
	Low	-514.80 ± 287.75	(-628.63, -400.97)		

Significant at *P* < 0.05

Table 15. Sagittal and frontal plane INI and TER EA descriptives (Mean ± SD)

	Energy Absorption (%BW*Ht)	
	Sagittal Plane	Frontal Plane
INI Total	13.62 ± 3.02	1.53 ± 1.24
INI Hip	2.26 ± 1.34	0.20 ± 0.26
INI Knee	8.98 ± 2.69	1.05 ± 1.08
INI Ankle	2.37 ± 1.64	0.28 ± 0.32
TER Total	3.72 ± 2.29	0.45 ± 0.37
TER Hip	0.83 ± 1.02	0.15 ± 0.21
TER Knee	2.19 ± 1.29	0.26 ± 0.29
TER Ankle	0.70 ± 0.40	0.03 ± 0.05

Table 16. Frontal plane biomechanical descriptives during the double leg jump landing task

	Frontal knee angle at IGC(°)	Peak knee valgus (°)	Peak hip adduction (°)	Peak vGRF (x BW^{-1})	Peak pGRF (x BW^{-1})	Peak KVM (x $[BW*Ht]^{-1}$)
Mean	-6.81	-17.04	2.96	2.91	0.81	0.085
SD	7.60	10.54	7.39	0.79	0.24	0.046

Table 17. Simple bivariate correlations between frontal plane INI EA and frontal plane biomechanics during the double leg jump landing task (Significant at $P < 0.05$)

	Frontal Plane EA			
Biomechanical Variables	INI Total	INI Hip	INI Knee	INI Ankle
FPK angle at IGC	r = -0.518 p < 0.001	r = -0.048 p = 0.665	r = -0.589 p < 0.001	r = 0.013 p = 0.905
Peak knee valgus angle	r = -0.662 p < 0.001	r = -0.036 p = 0.750	r = -0.732 p < 0.001	r = -0.073 p = 0.515
Peak hip adduction angle	r = 0.462 p < 0.001	r = -0.040 p = 0.724	r = 0.462 p < 0.001	r = 0.155 p = 0.165
Peak vGRF	r = 0.144 p = 0.197	r = 0.046 p = 0.680	r = 0.139 p = 0.211	r = 0.051 p = 0.651
Peak pGRF	r = 0.225 p = 0.042	r = -0.071 p = 0.529	r = 0.279 p = 0.011	r = -0.007 p = 0.949
Peak KVM	r = 0.698 p < 0.001	r = 0.037 p = 0.741	r = 0.717 p < 0.001	r = 0.260 p = 0.018

Table 18. Simple bivariate correlations between frontal plane TER EA and frontal plane biomechanics during the double leg jump landing task (Significant at *P* < 0.05)

	Frontal Plane EA			
Biomechanical Variables	TER Total	TER Hip	TER Knee	TER Ankle
FPK angle at IGC	r = -0.233 p = 0.035	r = 0.190 p = 0.087	r = -0.450 p < 0.001	r = 0.058 p = 0.606
Peak knee valgus angle	r = -0.457 p < 0.001	r = 0.022 p = 0.847	r = -0.625 p < 0.001	r = 0.133 p = 0.233
Peak hip adduction angle	r = 0.041 p = 0.715	r = -0.338 p = 0.002	r = 0.333 p = 0.002	r = -0.153 p = 0.170
Peak vGRF	r = -0.144 p = 0.142	r = -0.188 p = 0.091	r = -0.040 p = 0.720	r = -0.160 p = 0.152
Peak pGRF	r = -0.159 p = 0.154	r = -0.268 p = 0.015	r = 0.046 p = 0.684	r = -0.269 p = 0.014
Peak KVM	r = 0.284 p = 0.010	r = -0.083 p = 0.461	r = 0.446 p < 0.001	r = -0.111 p = 0.319

Table 19. Simple bivariate correlations between frontal and sagittal plane EA during the INI and TER phases of the double leg jump landing task (Significant at $P < 0.05$)

	Frontal Plane EA			
Sagittal Plane EA	INI Total	INI Hip	INI Knee	INI Ankle
INI Total	r = -0.015 p = 0.890	r = 0.139 p = 0.212	r = -0.054 p = 0.628	r = 0.010 p = 0.928
INI Hip	r = -0.095 p = 0.398	r = -0.117 p = 0.296	r = -0.096 p = 0.391	r = 0.050 p = 0.653
INI Knee	r = 0.002 p = 0.987	r = 0.301 p = 0.006	r = 0.025 p = 0.823	r = -0.151 p = 0.175
INI Ankle	r = 0.046 p = 0.683	r = -0.141 p = 0.208	r = 0.019 p = 0.862	r = 0.224 p = 0.043
	TER Total	TER Hip	TER Knee	TER Ankle
TER Total	r = 0.287 p = 0.009	r = 0.314 p = 0.004	r = 0.091 p = 0.414	r = 0.225 p = 0.042
TER Hip	r = 0.264 p = 0.017	r = 0.287 p = 0.009	r = 0.063 p = 0.575	r = 0.337 p = 0.002
TER Knee	r = 0.244 p = 0.027	r = 0.270 p = 0.014	r = 0.090 p = 0.423	r = 0.115 p = 0.303
TER Ankle	r = 0.178 p = 0.110	r = 0.193 p = 0.083	r = 0.073 p = 0.515	r = 0.058 p = 0.607

Table 20. Frontal plane EA group comparisons for frontal plane landing biomechanics (High EA group significantly different from Low (*) and Moderate (†) EA groups, P < 0.05)

Variable	EA Group	Mean ± SD	95% CI	$F_{(2,79)}$	P-value
Total frontal plane EA INI (%BW*Ht)	High*†	2.81 ± 1.37	(2.27, 3.35)	55.501	< 0.001
	Moderate	1.23 ± 0.24	(1.14, 1.32)		
	Low	0.55 ± 0.21	(0.47, 0.63)		
Frontal plane knee angle at IGC (°)	High*	-10.34 ± 7.81	(-13.43, -7.25)	5.782	0.005
	Moderate	-6.38 ± 7.69	(-9.36, -3.40)		
	Low	-3.73 ± 5.89	(-6.06, -1.40)		
Peak knee valgus angle (°)	High*†	-25.41 ± 8.66	(-28.83, -21.98)	19.874	< 0.001
	Moderate	-14.75 ± 10.31	(-18.75, -10.75)		
	Low	-11.04 ± 6.69	(-13.68, -8.39)		
Peak hip adduction angle (°)	High*	6.25 ± 7.74	(3.19, 9.32)	4.529	0.014
	Moderate	1.90 ± 6.88	(-0.77, 4.57)		
	Low	0.76 ± 6.59	(-1.85, 3.37)		
vGRF (xBW^{-1})	High	2.97 ± 0.67	(2.71, 3.24)	0.444	0.643
	Moderate	2.96 ± 0.95	(2.60, 3.34)		
	Low	2.80 ± 0.73	(2.51, 3.08)		
pGRF (xBW^{-1})	High†	0.91 ± 0.27	(0.80, 1.02)	4.030	0.022
	Moderate	0.75 ± 0.23	(0.66, 0.84)		
	Low	0.76 ± 0.20	(0.68, 0.84)		
KVM ($x[BW*Ht]^{-1}$)	High*†	0.119 ± 0.047	(0.101, 0.139)	17.883	< 0.001
	Moderate	0.079 ± 0.036	(0.065, 0.093)		
	Low	0.058 ± 0.031	(0.046, 0.070)		
Peak hip adduction velocity (°/s)	High*	84.35 ± 54.50	(62.79, 105.91)	4.885	0.010
	Moderate	63.14 ± 46.20	(45.22, 81.05)		
	Low	46.83 ± 27.53	(35.94, 57.72)		
Peak knee valgus velocity (°/s)	High*†	-247.62 ± 77.45	(-278.26, -216.98)	39.275	< 0.001
	Moderate	-123.95 ± 74.07	(-152.67, -95.23)		
	Low	-85.59 ± 57.56	(-108.35, -62.82)		
Peak ankle eversion velocity (°/s)	High	-182.05 ± 119.77	(-255.77, -155.15)	0.255	0.776
	Moderate	-193.56 ± 114.62	(-238.01, -149.12)		
	Low	-205.46 ± 127.18	(-229.43, -134.67)		

Significant at *P* < 0.05

Table 21. Joint position at initial ground contact and peak joint flexion during the preferred, flexed, and erect drop landing conditions [Mean (SD)]

		Position at IGC (°)		
		Hip	Knee	Ankle
Preferred	Male	12.90 (10.78)	16.23 (8.72)	40.64 (13.76)
	Female	14.85 (6.92)	16.24 (8.98)	*48.07 (10.88)
	Overall	13.72 (9.32)	16.24 (8.74)	43.76 (13.05)
Flexed	Male	26.89 (11.16)	34.00 (2.04)	31.00 (13.71)
	Female	28.15 (8.97)	33.66 (2.24)	*40.16 (11.25)
	Overall	†27.41 (10.22)	†33.86 (2.11)	†34.85 (13.41)
Erect	Male	15.00 (15.99)	19.95 (2.00)	38.18 (11.46)
	Female	14.37 (7.53)	19.39 (2.12)	*47.59 (9.25)
	Overall	14.74 (13.01)	19.72 (2.05)	42.13 (11.49)

		Peak Joint Flexion (°)		
		Hip	Knee	Ankle
Preferred	Male	46.70 (27.55)	77.42 (24.96)	-14.35 (10.49)
	Female	52.25 (18.42)	80.22 (14.46)	-11.44 (12.37)
	Overall	49.03 (24.08)	78.60 (21.05)	-13.13 (11.29)
Flexed	Male	60.84 (24.85)	89.80 (15.27)	-18.89 (8.88)
	Female	70.38 (14.55)	95.59 (10.79)	-17.19 (9.16)
	Overall	†64.85 (21.49)	†92.23 (13.75)	†-18.17 (8.95)
Erect	Male	44.76 (27.87)	75.61 (13.78)	-16.74 (7.98)
	Female	52.18 (20.80)	80.80 (14.82)	-12.97 (9.27)
	Overall	47.88 (25.18)	77.79 (14.31)	-15.16 (8.67)

Significantly less flexed (*) than males and more flexed (†) than Erect and Preferred conditions, $P < 0.05$

Table 22. Descriptive statistics for energy absorption magnitude and joint contribution to total energy absorption during the preferred, flexed, and erect drop landing conditions stratified by sex [Mean (SD)]

		Energy Absorption (%BW*Ht)			
		Hip	Knee	Ankle	Total
Preferred	Male	3.12 (1.94)	8.07 (3.58)	4.50 (2.25)	15.68 (4.14)
	Female	3.60 (1.76)	10.24 (3.96)	5.47 (3.25)	*19.32 (3.99)
Flexed	Male	2.70 (2.18)	6.14 (2.48)	4.01 (2.10)	12.85 (3.06)
	Female	3.52 (1.98)	7.21 (3.56)	5.11 (2.61)	*15.84 (2.44)
Erect	Male	2.81 (2.10)	6.33 (2.59)	4.52 (2.20)	13.66 (3.24)
	Female	3.21 (2.14)	7.66 (4.46)	6.08 (3.12)	*16.95 (2.90)

		Contribution to Total EA (%)		
		Hip	Knee	Ankle
Preferred	Male	19.49 (9.98)	50.88 (17.74)	29.63 (14.86)
	Female	19.12 (10.81)	53.22 (19.14)	27.66 (13.61)
Flexed	Male	19.48 (13.27)	49.25 (19.23)	31.27 (16.91)
	Female	22.67 (13.20)	45.25 (19.31)	32.08 (14.08)
Erect	Male	19.34 (12.13)	47.36 (17.93)	33.28 (15.70)
	Female	19.62 (13.89)	44.12 (21.45)	36.26 (16.95)

Figure 1. The quantification of net energy flow

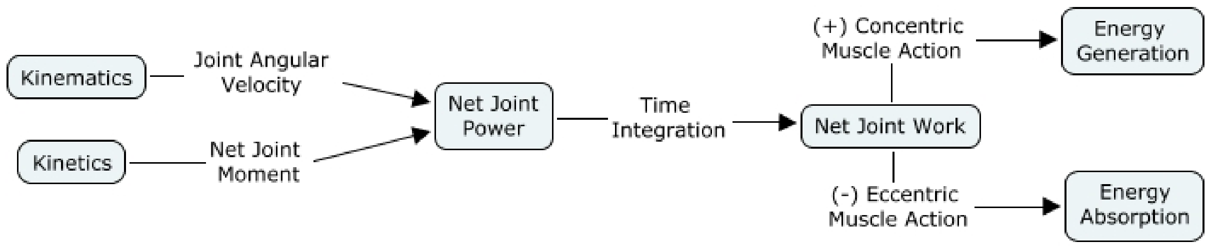


Figure 2. Maximal voluntary isometric contraction testing positions

A. Hip Extension

B. Knee Extension

C. Knee Flexion

D. Ankle Extension (Plantarflexion)

Figure 3. Adjustable drop bar used to complete 0.60 m double leg drop landings (A) and depiction of biofeedback display with knee flexion angle target window and cursor indicating instantaneous knee flexion angle (B).

A.

B.

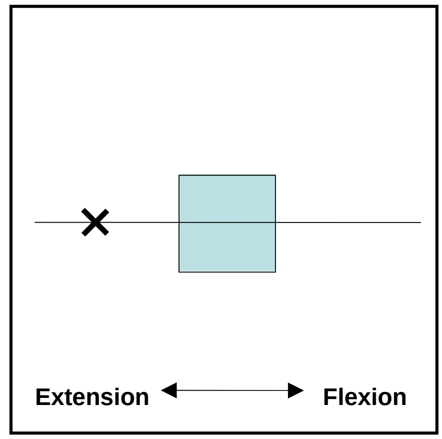
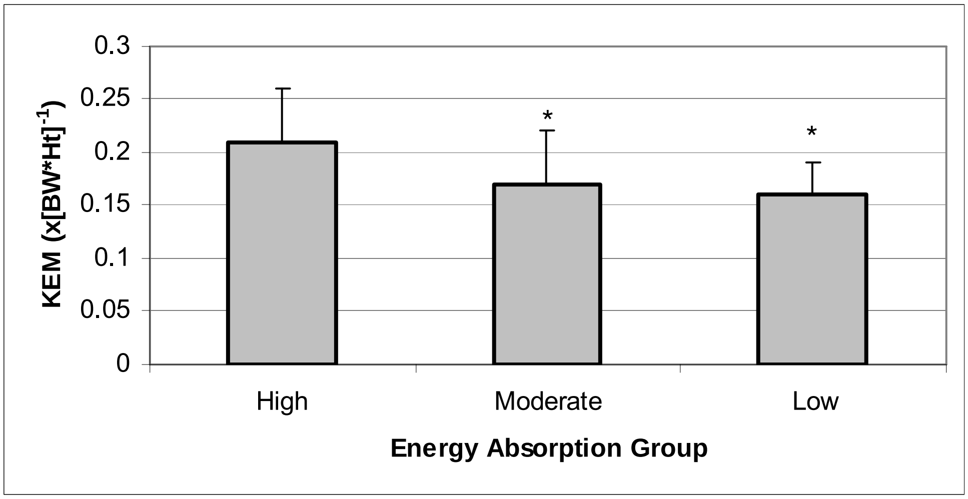
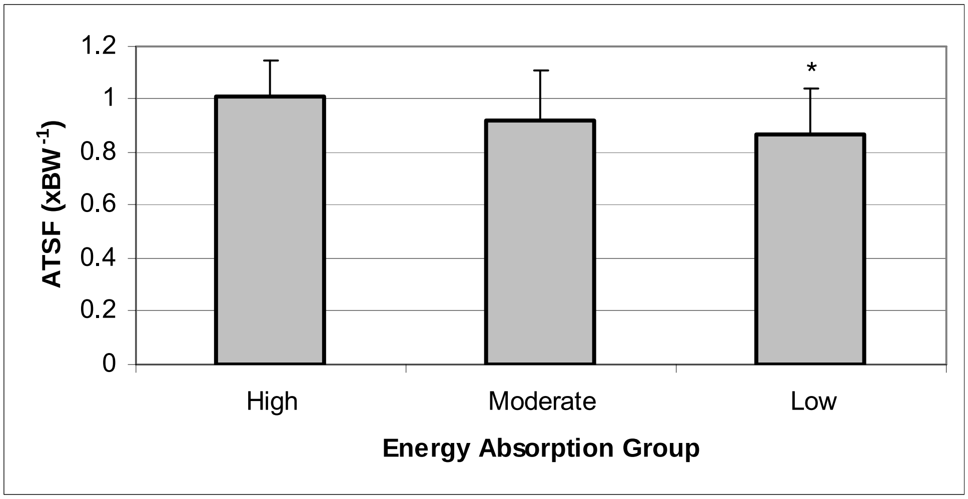


Figure 4. Sagittal plane EA group comparison of peak internal knee extension moment (KEM) during the double leg jump landing task



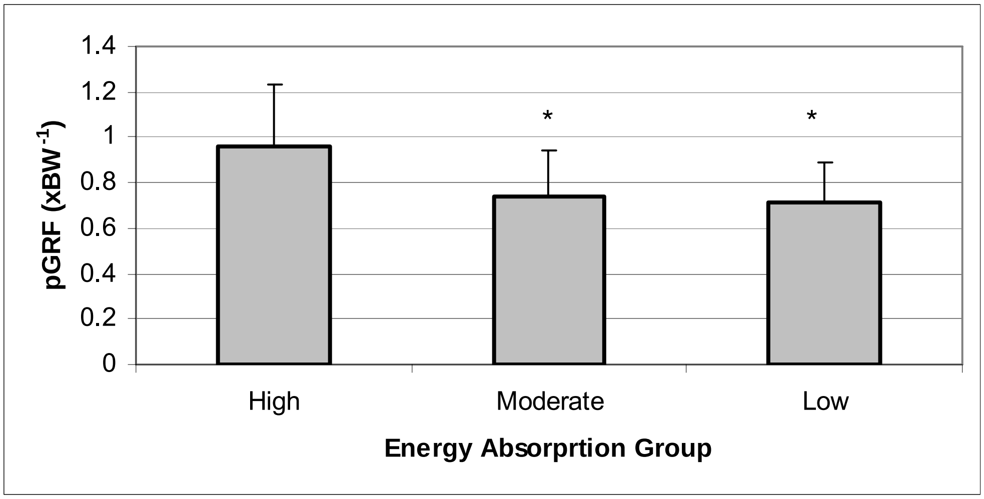
*Significantly different from High EA group (*P* < 0.05)

Figure 5. Sagittal plane EA group comparison of peak anterior tibial shear force (ATSF) during the double leg jump landing task



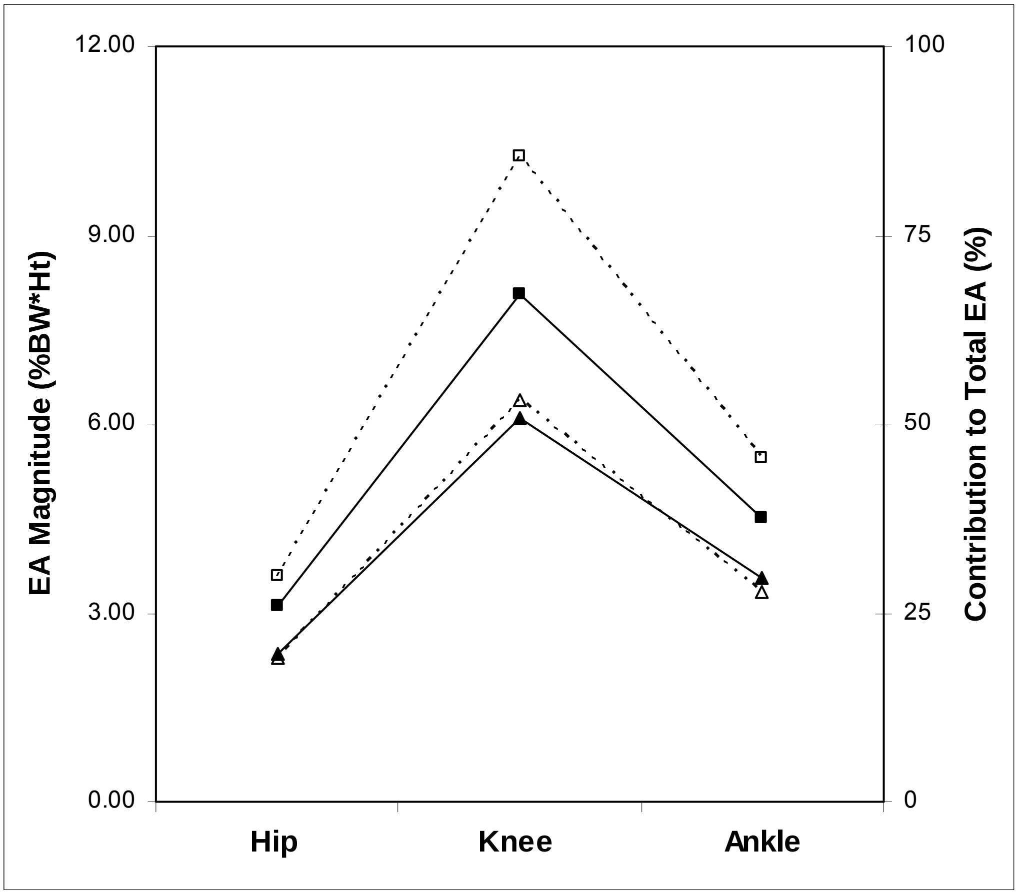
*Significantly different from High EA group (*P* < 0.05)

Figure 6. Sagittal plane EA group comparison of peak posterior ground reaction force (pGRF) during the double leg jump landing task



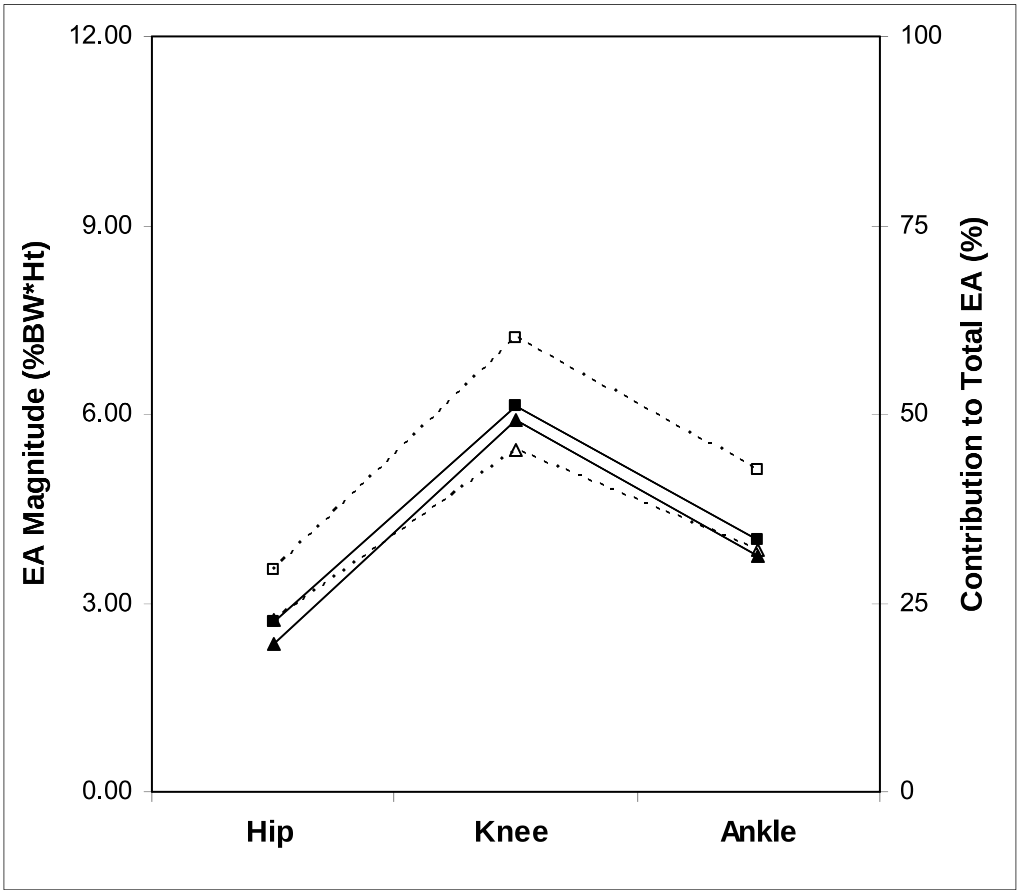
*Significantly different from High EA group (*P* < 0.05)

Figure 7. Joint energy absorption (EA) magnitude (squares) and relative joint contributions to total EA (triangles) for males (solid) and females (dashed) during the preferred landing condition



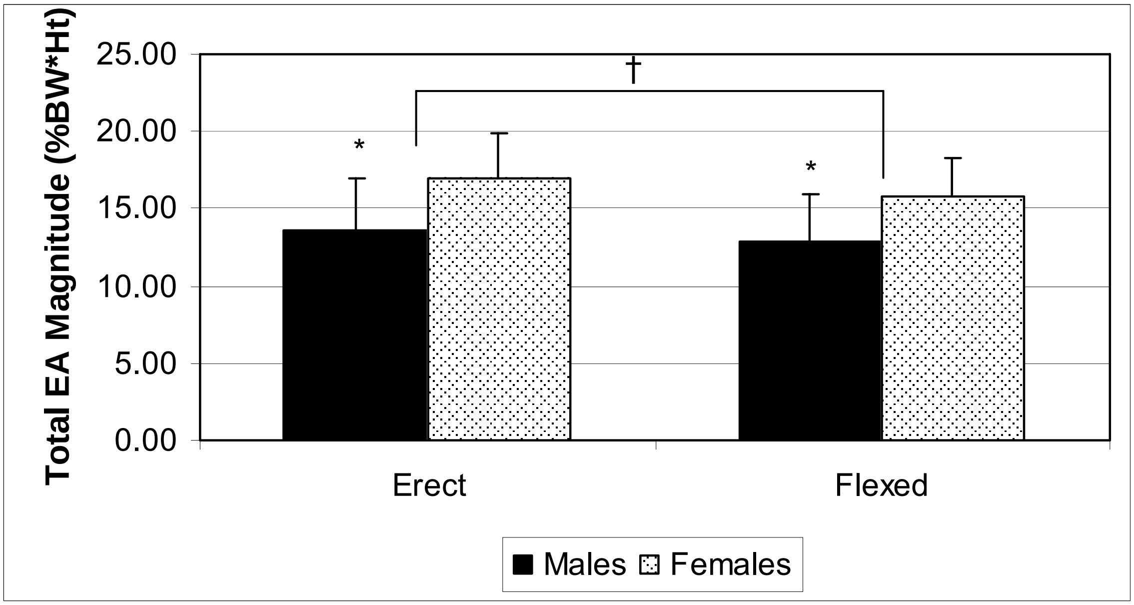
Magnitude: Main effects for sex (P = 0.003) and joint (P < 0.001) with Knee EA > Ankle EA > Hip EA (P < 0.05)
Contribution: Main effect for joint (P < 0.001) with Knee > Ankle > Hip (P < 0.05)

Figure 8. Joint energy absorption (EA) magnitude (squares) and relative joint contributions to total EA (triangles) for males (solid) and females (dashed) during the flexed landing condition



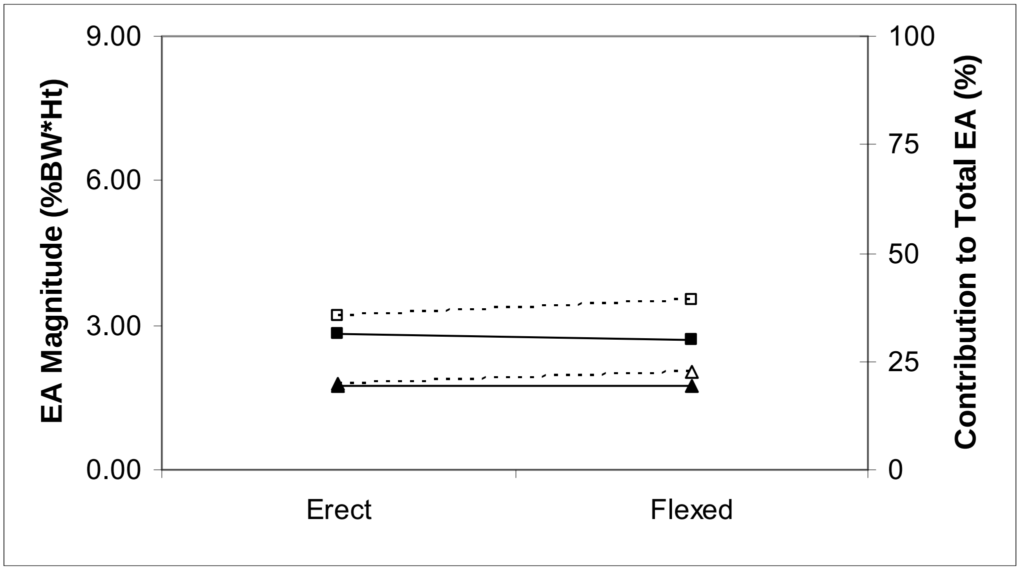
Magnitude: Main effects for sex (*P*= 0.001) and joint (*p* < 0.001) with Knee EA > Ankle EA > Hip EA (*P* < 0.05)
Contribution: Main effect for joint (*P* < 0.001) with Knee > Ankle > Hip (*P* < 0.05)

Figure 9. Total energy absorption magnitude during the Preferred, Erect and Flexed landing conditions



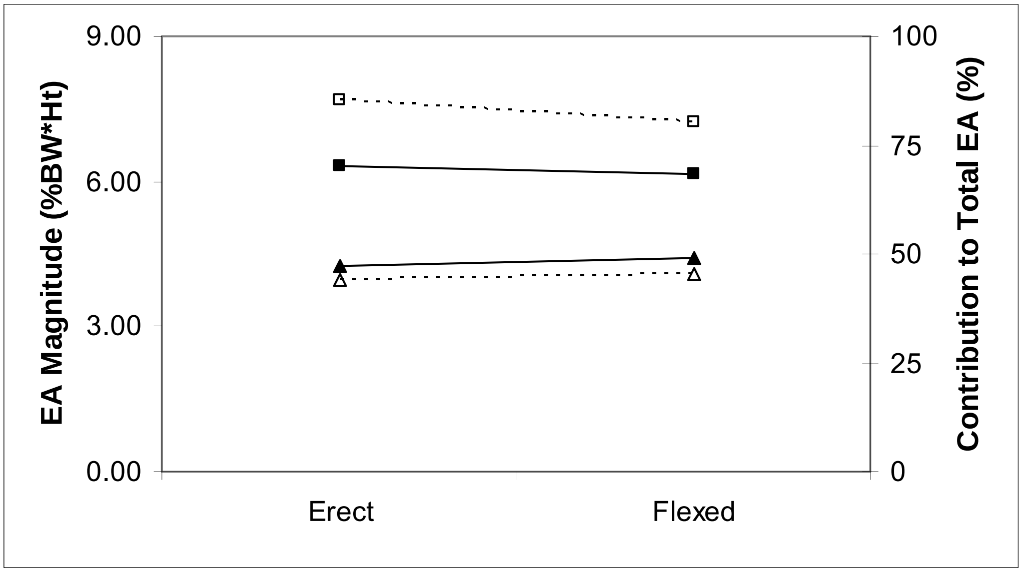
***** = significant difference between Males and Females ($P < 0.05$); † = Erect significantly greater than Flexed ($P < 0.05$)

Figure 10. Hip energy absorption (EA) magnitude (squares) and hip contribution to total EA (triangles) for males (solid) and females (dashed) during the Erect and Flexed landing conditions



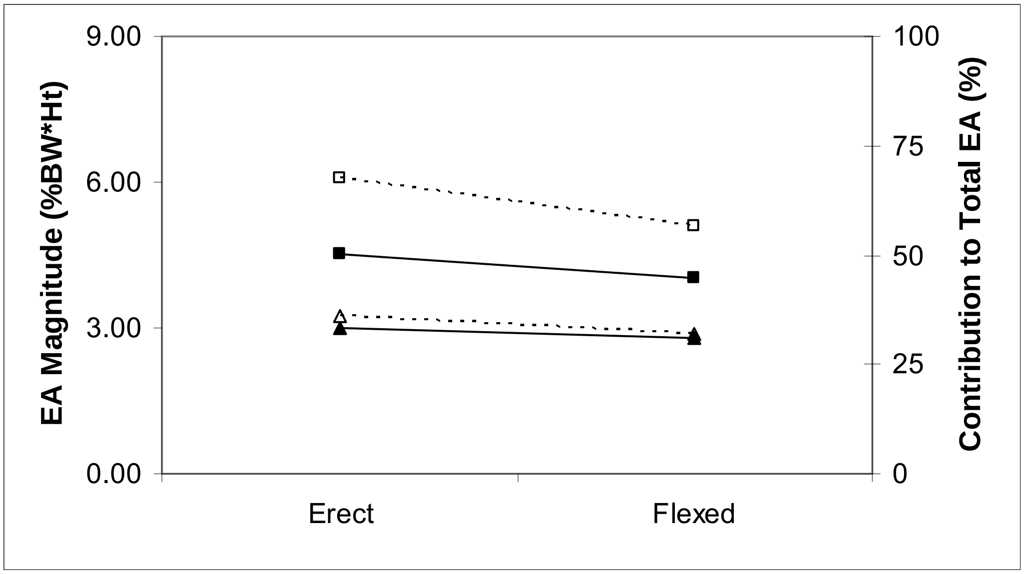
Contribution: Main effect for posture with Flexed > Erect (*P* < 0.05)

Figure 11. Knee energy absorption (EA) magnitude (squares) and knee contribution to total EA (triangles) for males (solid) and females (dashed) during the Erect and Flexed landing conditions



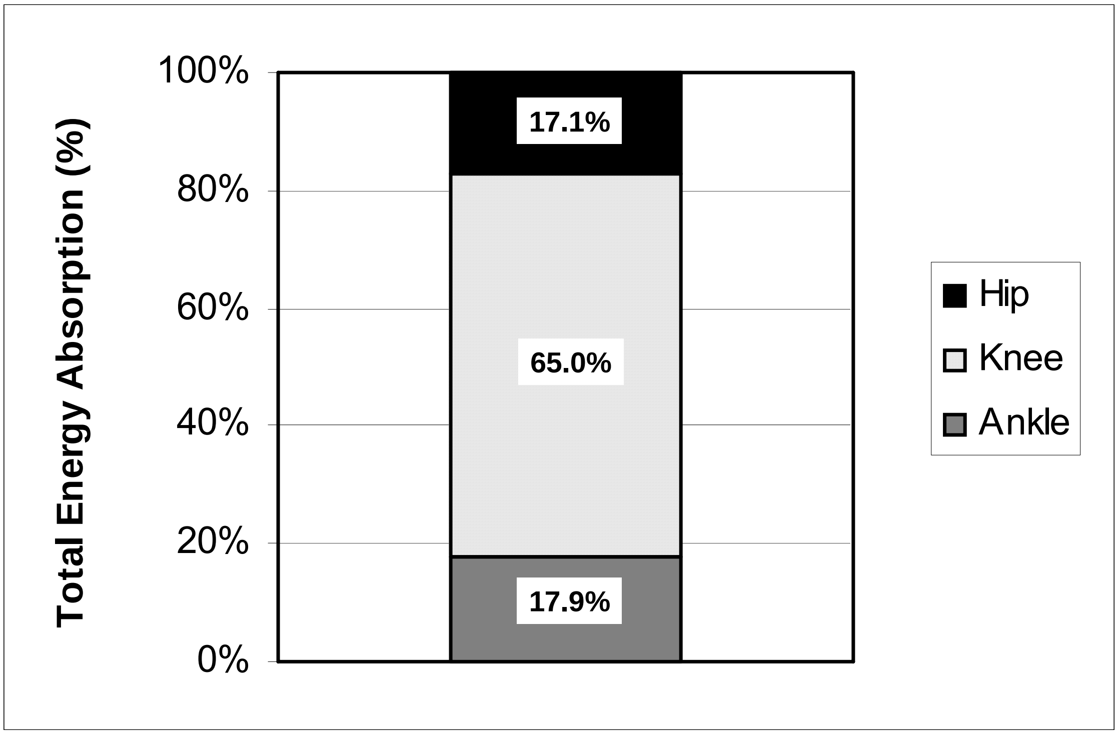
No significant main or interaction effects (*P* > 0.05)

Figure 12. Ankle energy absorption (EA) magnitude (squares) and ankle contribution to total EA (triangles) for males (solid) and females (dashed) during the Erect and Flexed landing conditions



Magnitude: Main effect for posture with Flexed < Erect ($P < 0.001$)
Contribution: Main effect posture with Flexed < Erect ($P < 0.001$)

Figure 13. Mean relative joint contributions to total energy absorption during the 100 ms immediately following ground contact



APPENDIX ONE: MANUSCRIPT I

Manuscript I

Group differences in lower extremity energy absorption and landing biomechanics.

Part I: Sagittal plane analyses.

ABSTRACT

Context: Eccentric muscle actions of the hip, knee, and ankle extensors absorb kinetic energy from the system during landing. Greater total lower extremity energy absorption (EA) in the sagittal plane during the initial impact phase (INI: 100 ms immediately following ground contact) of landing has been associated with landing biomechanics that are considered high-risk for anterior cruciate ligament (ACL) injury. However, it is unknown whether meaningful differences in ACL-related landing biomechanics are present in groups exhibiting high, moderate, and low magnitudes of EA during landing, and whether quantification of EA might be a mechanism to better identify individuals at higher risk of ACL injury.

Objective: To compare landing biomechanics between high, moderate and low EA groups, and determine whether there is an association between sex and EA group.

Design: Descriptive laboratory study.

Setting: Research laboratory.

Patients or Other Participants: Eighty-two healthy, physically active volunteers.

Intervention(s): Landing biomechanics were assessed using an electromagnetic capture system and force plate during double leg jump landings.

Main Outcome Measure(s): Total sagittal plane lower extremity EA was used to group participants into high, moderate, and low EA tertiles. Sagittal and frontal plane knee angles at ground contact, and peak vertical and posterior ground reaction forces, anterior tibial shear force, internal hip extension, knee extension, and knee varus moments; and, knee flexion and knee valgus angles were identified during the landing task. One-way ANOVA was used to compare EA groups across these variables.

Results: The High EA group exhibited greater peak knee extension moment than both the Moderate ($P < 0.05$) and Low EA ($P < 0.05$) groups, and greater anterior tibial shear force than the Low EA group ($P < 0.05$). Peak posterior ground reaction force was significantly greater in the High group compared to the Moderate ($P < 0.05$) Low groups ($P < 0.05$). No other significant group differences were noted. There was not a significant association between sex and High vs. Low EA group assignment ($P = 0.273$).

Conclusions: Greater sagittal plane INI EA is likely indicative of greater ACL loading due to sagittal plane mechanisms. However, there is no evidence that sagittal plane EA influences frontal plane biomechanics. Further, there is no association between sex and EA group assignment suggesting that quantification of sagittal plane INI EA to infer ACL injury risk is not supported.

INTRODUCTION

Non-contact mechanisms account for 70-80% of all anterior cruciate ligament (ACL) injuries;[2, 20] occurring most commonly in dynamic activities involving rapid deceleration, cutting, and landing.[1, 39] During landing, internal hip, knee, and ankle extension (plantarflexion) moments must be produced via eccentric muscle contractions to both control joint motion and absorb the kinetic energy of the system.[15] This energy absorption (EA) by the lower extremity musculature can be calculated using energetic analyses in which kinematic (joint angular velocity) and kinetic (net joint moment) data are combined to quantify the energy flow at each joint that is responsible for producing the observed movement.[43]

While conventional biomechanical analyses used in ACL injury research identify kinematic and kinetic parameters independently and at discrete time points, energetic analyses quantify these data across the landing period and combine the individual contributions of the hip, knee, and ankle to the total lower extremity energy absorption in order to provide insight into the coordinated actions of these joints.[7, 27, 34] This coupling of the kinematics and kinetics of multiple joints provides a more comprehensive description of the complex multi-segmental mechanics that occur during landing and in proposed ACL-injury mechanisms.[25]

Previous work has suggested that greater EA by the neuromuscular system over the entire landing period during drop landings reduces the loading of passive tissues such as the ACL;[15] with greater total lower extremity EA in the sagittal plane associated with smaller vertical ground reaction forces (vGRF) and greater knee flexion displacements during landing.[42, 47] However, these results have typically been observed in studies which have artificially manipulated landing conditions. Devita et al.[15] and Zhang et al.[47] observed greater EA and lesser peak impact forces in "soft" landings compared to "stiff" landings when subjects were instructed to alter the magnitude of their knee flexion displacement during drop landings. To date, there is limited evidence that has directly evaluated the influence of sagittal plane EA during natural landing conditions on peak impact forces and other biomechanical factors specifically related to non-contact ACL injury.

Recently, Norcross et al.[35] reported the first direct associations between EA and biomechanical factors related to non-contact ACL injury in individuals using their preferred landing style. This exploratory analysis identified that it is not just the magnitude, but also the timing of EA during landing which influences these biomechanical factors. Specifically, greater total lower extremity EA in the sagittal plane during the initial impact phase of landing (INI: 100 ms immediately following initial ground contact [IGC]) was associated with greater peak vGRF, anterior tibial shear force (ATSF), and internal hip extension moment; factors generally considered to be unfavorable with respect to ACL injury risk.[9, 23] However, greater total EA during the terminal phase of landing (TER: 100 ms after IGC to the minimum vertical position of the whole body center of mass) was associated with lesser peak values of these same biomechanical factors.[35] As a result, it was suggested that EA during landing may serve to quantify movement strategies that could result in greater ACL injury risk.[35] Though promising in its preliminary results, this investigation has two principal limitations. First, while significant relationships between lower extremity EA and key ACL-related biomechanical factors were identified, it is unknown whether groups performing different amounts of sagittal plane EA during landing demonstrate meaningful differences on these ACL-related biomechanical factors. Second, although quantification of sagittal plane EA appears to accurately synthesize an overall sagittal plane biomechanical landing profile; it is not clear whether quantification of sagittal plane EA might be useful as a mechanism to identify individuals at greater risk of non-contact ACL injury. It is well-documented that females display a greater likelihood than males of suffering a non-contact ACL injury,[17, 20] despite the fact that a greater absolute number of ACL injuries are suffered by males.[10, 30, 37]

As a result, sagittal plane EA could potentially serve as a more effective means of prospectively identifying high-risk athletes using a criteria other than simply sex.

The purpose of this study was to address these limitations by: 1) determining whether there were significant differences between high, moderate, and low sagittal plane EA groups on various biomechanical factors that are associated with non-contact ACL injury; and 2) to evaluate the face validity of using sagittal plane EA during INI to identify ACL injury risk by determining whether there is a significant association between sex and sagittal plane EA group assignment. We hypothesized that individuals in the high EA group would display significantly less favorable values across all biomechanical variables compared to the moderate and low EA groups, and that there would be a significant association between the high EA group and females.

METHODS

Participants

Eighty-two volunteers (41 males, 41 females; age = 20.1 ± 2.4 years; height = 1.74 ± 0.10 m; mass = 70.3 ± 16.1 kg) participated in this study after reading and signing an Institutional Review Board approved consent form. All subjects were physically active (participating in at least 30 minutes of physical activity three times per week), and generally healthy with no history of ACL injury, neurological disorder, lower extremity surgery, or lower extremity injury within the six months preceding data collection.

Subject Preparation and Experimental Procedures

The height and mass of each subject were recorded prior to data collection and used for biomechanical model generation and normalization of the dependent variables. Lower-extremity and trunk kinematics were assessed using an electromagnetic motion capture system (Motion Star, Ascension Technology Corp., Burlington, VT, USA). Six degree of freedom electromagnetic tracking sensors were positioned over the third metatarsal of the foot, anteromedial aspect of the shank, and lateral thigh of the dominant leg, defined as the leg used to kick a ball for maximum distance; as well as on the sacrum and C7 spinous process of the trunk. These sensors were placed over areas of minimal muscle mass, and secured with pre-wrap and athletic tape to reduce motion artifact. Global and segment axis systems were established with the positive X axis designated as forward/anteriorly, the positive Y axis leftward/medially, and the positive Z axis upward/superiorly. A segment-linkage model of the dominant lower extremity, pelvis, and thorax was created using the MotionMonitor motion analysis software (Innovative Sports Training, Inc., Chicago, IL, USA) by digitizing the ankle, knee, and hip joint centers and the T12 spinous process. Ankle and knee joint centers were defined as the midpoint of the digitized medial and lateral malleoli and the medial and lateral femoral condyles, respectively. The hip joint center was predicted using external landmarks on the pelvis as described by Bell et al.[4]

Double leg jump landings were performed by having subjects stand atop a 30 cm high box that was set a distance equal to 50% of the subjects' height away from the edge of a nonconductive force plate (Type 4060-NC, Bertec Corporation, Columbus, OH, USA) whose axis system was aligned with the global axis system. Subjects were instructed to jump down and forward toward the force plate, contact the ground with both feet at the same time with their dominant foot near the center of the force plate and their non-dominant foot positioned next to the force plate, and then immediately jump up for maximum height using both legs. Subjects performed 3 practice trials and 5 successful testing trials with 30 seconds of rest between trials to minimize the potential effects of fatigue. Trials were deemed successful if subjects jumped from the box and landed with both feet at the same time, completely contacted the force plate with only the dominant foot, and performed the landing task and subsequent maximum jump in a fluid motion.

Data Sampling and Reduction

Kinematic and kinetic data were sampled at 120 and 1,200 Hz, respectively, using the MotionMonitor motion analysis software. Raw kinematic data were low-pass filtered using a fourth-order, zero-phase-lag Butterworth filter with a cutoff frequency of 10 Hz,[12] time-synchronized with the kinetic data, and re-sampled at 1,200 Hz. Joint angular positions were calculated based on a right hand convention using Euler angles in a YX'Z'' rotation sequence, and instantaneous joint angular velocities were calculated as the 1st derivative of angular position. Motion was defined about the hip as the thigh relative to the sacrum, about the knee as the shank relative to the thigh, and about the ankle as the foot relative to the shank. Kinetic data were low-pass filtered at 60 Hz (4th order zero-phase lag Butterworth) [28] and combined with kinematic and anthropometric data to calculate the net internal joint moments of force at the hip, knee, and ankle, and the net internal force on the shank at the knee joint using an inverse dynamics solution.[19]

Custom computer software (LabVIEW, National Instruments Corporation, Austin, TX, USA) was used to multiply sagittal plane joint angular velocities and net joint moments in order to generate hip, knee, and ankle joint power curves for each landing trial ($P = M \times \omega$). The negative portion of the joint power curves were then integrated to calculate negative mechanical joint work[12, 14, 33, 42] during the INI phase of landing (the 100 ms following IGC [VGRF > 10 N]).[11, 12] Finally, total negative lower extremity joint work was calculated by summing the negative joint works calculated at the hip, knee, and ankle.[14, 42, 47] This value then represents the total sagittal plane lower extremity EA, as negative joint work is indicative of EA by the muscle-tendon unit.[33, 43] The same custom software was used to identify sagittal and frontal plane knee angles at IGC, and peak values for: vGRF; posterior ground reaction force (pGRF); ATSF; internal hip extension (HEM), knee extension (KEM), and knee varus (KVM) moments; and, knee flexion and knee valgus angles during the total landing phase (IGC to the minimum vertical position of the whole body center of mass).[28, 47] Ground reaction and segmental forces were normalized to subject body weight (x BW^{-1}), net joint moments normalized to the product of subject height and weight (x $[BW*Ht]^{-1}$), and energy absorption expressed as a percentage of the product of subject height and weight

(% BW*Ht). All dependent variables were averaged across the five jump landing trials of each subject prior to statistical analysis.

Statistical Analysis

Total EA data were arranged into tertiles to create three distinct EA groups: High, Moderate, and Low, respectively. Static comparisons across EA groups for each biomechanical factor were made using ten separate one-way ANOVA models. For significant ANOVA models, post-hoc testing to identify group differences on these dependent variables was performed using Tukey's HSD. A 2 x 2 contingency table was constructed using sex and EA group (H and L) as categorical variables and a Pearson X^2 test of association was used to determine whether a significant association existed between sex and EA group assignment. All analyses were conducted using commercially available software (SPSS 17.0, SPSS Inc., Chicago, IL, USA) with statistical significance established a priori as $\alpha \leq 0.05$.

RESULTS

Table 1 displays descriptive statistics and frequency counts by sex for the three EA groups. EA group assignment by tertile successfully created three groups with significantly different sagittal plane EA during INI ($F_{2,79} = 133.093$, $p < 0.001$) (Table 1). With respect to the biomechanical variables related to ACL injury, we observed significant differences between groups for peak ATSF ($F_{2,79} = 4.767$, $p = 0.011$), KEM ($F_{2,79} = 11.092$, $p < 0.001$), and pGRF ($F_{2,79} = 10.582$, $p < 0.001$) (Table 2). Post hoc testing revealed that that the High group landed with significantly greater peak KEM than both the Moderate group ($p = 0.004$) and the Low group ($p < 0.001$). However, no significant difference in KEM was detected between the Moderate and Low EA groups ($p = 0.158$) (Figure 1). The High group demonstrated significantly greater peak ATSF compared to the Low group ($p = 0.009$); though no significant differences were noted between the High and Moderate groups ($p = 0.113$) or the Moderate and Low groups ($p = 0.557$) (Figure 2). Peak pGRF was also greater in the High group compared to Moderate group (p = 0.001) and the Low group (p < 0.001), but the pGRF of the Moderate and Low groups were not significantly different (p = 0.843) (Figure 3). No EA group differences were noted for any other biomechanical variable of interest (p > 0.05) (Tables 2 and 3). There was also no significant association between sex and High vs. Low EA group assignment (X^2 = 1.20, p = 0.273) (Table 1).

DISCUSSION

The primary finding of this investigation is that individuals absorbing a greater magnitude of energy in the sagittal plane during the INI phase of landing utilize a movement strategy that likely results in greater ACL loading. This is evidenced by the fact that the High EA group exhibits significantly greater peak KEM, ATSF, and pGRF compared to the Low EA group without differences in sagittal plane knee kinematics.

The greater KEM and ATSF demonstrated by the High EA group agreed with our hypotheses and have been identified in previous research as contributors to ACL loading. During landing, the lower extremity joints must resist rapid flexion induced by impact forces with internally generated extension moments.[15, 24] At the knee, the internal extension moment is generated by quadriceps contraction which has been identified as the primary contributor to anterior tibial shear force.[45] *In vitro*[3, 16, 18] and *in vivo*[5, 6] experiments have demonstrated that quadriceps contraction between 0 and 30° of knee flexion, and the ensuing anterior tibial shear force, significantly strains the ACL. Further, DeMorat et al.[13] successfully induced ACL injury in 6 out of 11 cadaver specimens with the application of simply an isolated quadriceps force. As a result, our findings indicate that movement strategies with greater sagittal plane EA during the 100 ms immediately following ground contact result in greater KEM and ATSF; and thus greater quadriceps forces that can potentially induce greater ACL loading.

The resultant strain on the ACL due to a standardized quadriceps contraction may be influenced by the sagittal plane position of the knee. Nunley et al.[36] reported that the angle between the patella tendon and the tibial shaft decreases as the knee progresses in to flexion, resulting in a smaller proportion of the quadriceps force being directed anteriorly relative to the tibia. The elevation angle of the ACL,[22, 29, 40] defined as the angle between the longitudinal axis of the ACL and the tibial plateau,[29] also decreases with knee flexion, resulting in the ACL being oriented less vertically; a smaller proportion of ACL loading being shear in nature as opposed to tensile; and a smaller ACL strain with a given anterior shear force.[45] Therefore, under the same quadriceps loading conditions, positioning the knee in greater flexion would result in lesser ACL strain. Accordingly, it is plausible that the High EA group exhibited greater KEM and ATSF, but in a more flexed knee position, thereby mediating the effects of the greater quadriceps force and experiencing resultant ACL loading that was comparable to the other groups. However, there were no significant differences in IGC or peak knee flexion angles between the three EA groups (Table 3). Accordingly, we feel that the greater observed sagittal plane knee kinetics, in concert with the same knee kinematics, are indicative of greater ACL loading in the High EA group.

The results of the present investigation were also surprising with respect to peak impact forces during landing. While the High EA group displayed significantly greater peak pGRF compared to both the Moderate and Low groups (Figure 3), there were no significant differences between groups for peak vGRF (Table 2). This result is in contrast to our previous exploratory investigation in which there was a significant association between peak vGRF and total sagittal plane EA; though, only 19.5% of the variance in vGRF was explained by sagittal plane EA.[35] It is known that both the posterior and vertical components of the GRF can induce a flexion moment relative to the knee that must be resisted by quadriceps contraction and increase ACL loading.[45] As such, increased vGRF or pGRF may affect knee joint loading. In a prospective investigation, Hewett et al.[23] found that ACL-injured females displayed peak vGRF that were 20% greater than uninjured controls. However, it is difficult to accurately compare the magnitudes of the vGRF in the present study to this investigation as the authors did not normalize their measured GRF to account for subject mass. Additionally, the existing literature comparing the sexes (i.e. higher and lower ACL injury risk) on vGRF is equivocal. Schmitz et al[42] and Salci et al[41] reported greater peak vertical ground reaction forces in females, while McNair and Prapavessis[32] and Decker et al.[12] did not observe sex differences in peak vGRF during landing. There is also limited evidence to suggest that the posterior component of the GRF is just as, if not more, important than the vertical component in explaining knee joint loading. Yu et al.[46] reported significant associations between both peak pGRF and vGRF; and ATSF and KEM. However, they found that peak pGRF occurred at the same time as peak ATSF and KEM; and explained 72% and 74% of the variance in these same variables compared to only 26% and 32% of the variance, respectively, for vGRF.[46] Collectively, these results imply that increases in either vGRF or pGRF likely result in greater ACL loading during landing. As such, the greater peak pGRF exhibited by the High EA group, even without a concomitant group difference in peak vGRF, lends further support to the notion that a movement strategy involving greater lower extremity EA during INI increases resultant ACL loading due to sagittal plane mechanisms.

A lack of EA group differences in peak HEM (Table 2) was unexpected given our previous investigation in which there was a significant relationship between total EA and peak HEM.[35] However, as with peak vGRF, the strength of the relationship observed previously was relatively weak with only 18% of the variance in peak HEM explained by total sagittal plane EA. Additionally, the current investigation utilized a sample size three times greater than our previous study, thereby decreasing the influence of more extreme values that may have driven the significance of our previous result. Given these discrepancies, we feel that further investigation of the relationship between total EA and peak HEM is warranted.

The results of this investigation confirmed our exploratory findings that indicated a lack of relationship between total sagittal plane EA and frontal plane knee kinematics and kinetics.[35] There were no group differences noted for knee valgus angle at IGC, peak knee valgus angle, or peak internal KVM. These frontal plane variables are important as knee valgus angle (at initial contact and peak) and peak external knee valgus moment were found to be significant prospective predictors of non-contact ACL injury.[23] Additionally, at knee flexion angles greater than 10°, an externally applied valgus moment in combination with anterior shear force results in significantly greater ACL loading than that produced by anterior shear force alone.[31] Accordingly, limiting frontal plane knee valgus motion and moments has been advocated to decrease ACL injury risk.[21] Pollard et al.[38] reported that individuals exhibiting greater combined peak hip and knee flexion during landing displayed significantly greater sagittal plane hip and knee EA, and lesser peak knee valgus angle and average internal knee varus moment. These authors speculated that greater use of sagittal plane EA may have reduced the magnitude of EA in the frontal plane and thereby influenced frontal plane knee biomechanics. It is important to note that Pollard et al.[38] calculated sagittal plane EA from IGC to peak knee flexion compared to the 100 ms following IGC used in the current study. We chose to focus on this INI phase for two reasons: 1) our previous results identified a temporal relationship between EA and high-risk landing biomechanics in which greater INI EA and lesser TER EA were considered unfavorable; and 2) peak ACL strain and injury likely occur during this period.[8, 26, 44] The failure of the High EA group to exhibit a less favorable frontal biomechanical profile compared to the other groups suggests that the magnitude of sagittal plane EA during INI does not influence frontal plane biomechanics and the associated ACL loading caused by frontal plane mechanisms. We suggest that future investigations should more closely examine inter-planar EA relationships as well as the direct influence of frontal plane EA on frontal plane biomechanics.

Finally, it is apparent from this investigation that quantification of total sagittal plane EA to infer non-contact ACL injury risk is unfounded. Contrary to our hypothesis, there was not a significant association between EA group assignment (High vs. Low) and sex. Given the overwhelming evidence indicating the greater risk of ACL injury in females,[17, 20] it would be expected that there would be a greater proportion of females assigned to the High EA group if this measure was indeed indicative of injury risk. However, this result also indicates that males and females have an equal likelihood of utilizing a landing strategy (High EA) that results in greater ACL loading due to sagittal plane mechanisms. As such, we propose that there are likely associations between sex and frontal and/or transverse plane landing biomechanics that lead to the increased risk of ACL injury in females.

CONCLUSIONS

The results of this study provide significant information for understanding the way in which EA during landing affects ACL loading. Landing with greater sagittal plane EA during the 100 ms immediately following ground contact results in sagittal plane knee kinetics and impact forces that likely increase ACL loading due to sagittal plane mechanisms. However, there is no association between sex and sagittal plane INI EA during landing indicating that the magnitude of sagittal plane EA during landing is not modified by sex. Additionally, sagittal plane INI EA does not appear to influence frontal plane knee biomechanics. Future research should determine what biomechanical factors are predictive of sagittal plane EA and whether sagittal plane EA may be modified via an intervention program to decrease ACL loading attributable to this plane. Further, the relationships between frontal plane EA; and frontal plane biomechanics and sagittal plane EA should be more closely investigated.

Acknowledgements

This study was supported by the NATA Research & Education Foundation Doctoral Grant Program.

Table 1. Sagittal plane EA group descriptives and frequency counts by sex

	Energy Absorption Group		
	High	Moderate	Low
Mean ± SD (%BW*Ht)	16.99 ± 1.85*†	13.37 ± 0.78*	10.50 ± 1.57
95% CI	(16.25, 17.72)	(13.07, 13.68)	(9.88, 11.12)
Males	13	11	17
Females	14	17	10
Total	27	28	27

Significantly different from Low EA (*) and Moderate EA (†) groups, $P < 0.05$

Table 2. Sagittal plane EA group comparisons for kinetic variables (High EA group significantly different from Low (*) and Moderate (†) EA groups, P < 0.05)

Variable	EA Group	Mean ± SD	95% CI	$F_{(2,79)}$	P-value
vGRF (xBW^{-1})	High	2.94 ± 0.66	(2.68, 3.21)	0.102	0.903
	Moderate	2.86 ± 0.89	(2.51, 3.21)		
	Low	2.94 ± 0.82	(2.62, 3.26)		
pGRF (xBW-1)	High*†	0.96 ± 0.27	(0.86, 1.07)	10.582	< 0.001
	Moderate	0.74 ± 0.20	(0.67, 0.82)		
	Low	0.71 ± 0.18	(0.64, 0.78)		
ATSF (xBW-1)	High*	1.01 ± 0.14	(0.96, 1.07)	4.767	0.011
	Moderate	0.92 ± 0.19	(0.85, 1.00)		
	Low	0.87 ± 0.17	(0.81, 0.94)		
HEM (x[BW*Ht]-1)	High	0.29 ± 0.13	(0.24, 0.35)	0.580	0.562
	Moderate	0.28 ± 0.13	(0.23, 0.34)		
	Low	0.32 ± 0.13	(0.27, 0.37)		
KEM (x[BW*Ht]-1)	High*†	0.21 ± 0.05	(0.19, 0.23)	11.092	< 0.001
	Moderate	0.17 ± 0.05	(0.16, 0.19)		
	Low	0.16 ± 0.03	(0.15, 0.17)		
KVM (x[BW*Ht]-1)	High	0.08 ± 0.05	(0.06, 0.11)	0.027	0.973
	Moderate	0.08 ± 0.03	(0.07, 0.10)		
	Low	0.09 ± 0.05	(0.07, 0.11)		

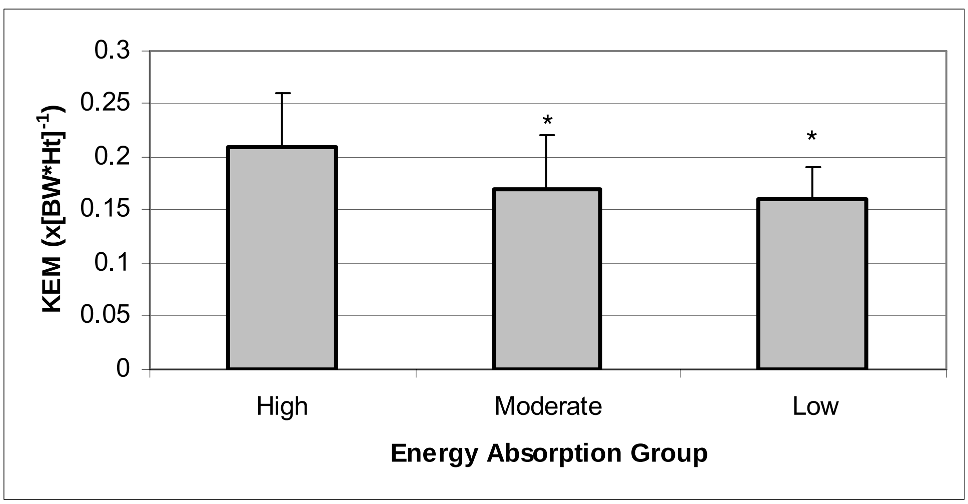
Significant at $P < 0.05$

Table 3. Sagittal plane EA group comparisons for kinematic variables

Variable	EA Group	Mean ± SD	95% CI	$F_{(2,79)}$	P-value
Sagittal plane knee angle at IGC (°)	High	22.73 ± 6.96	(19.98, 25.49)	0.015	0.471
	Moderate	23.11 ± 8.92	(19.65, 26.57)		
	Low	23.03 ± 9.60	(19.23, 26.83)		
Frontal plane knee angle at IGC (°)	High	-7.73 ± 8.17	(-8.85, -1.85)	0.760	0.985
	Moderate	-7.34 ± 5.53	(-9.49, -5.20)		
	Low	-6.81 ± 7.60	(-10.96, -4.50)		
Peak knee flexion angle (°)	High	93.82 ± 14.16	(73.91, 99.42)	1.143	0.324
	Moderate	91.15 ± 14.75	(71.72, 96.87)		
	Low	87.74 ± 15.47	(61.96, 93.86)		
Peak knee valgus angle (°)	High	-14.37 ± 11.15	(-18.78, -9.96)	1.310	0.276
	Moderate	-18.12 ± 8.86	(-21.56, -14.69)		
	Low	-18.57 ± 11.35	(-23.06, -14.08)		

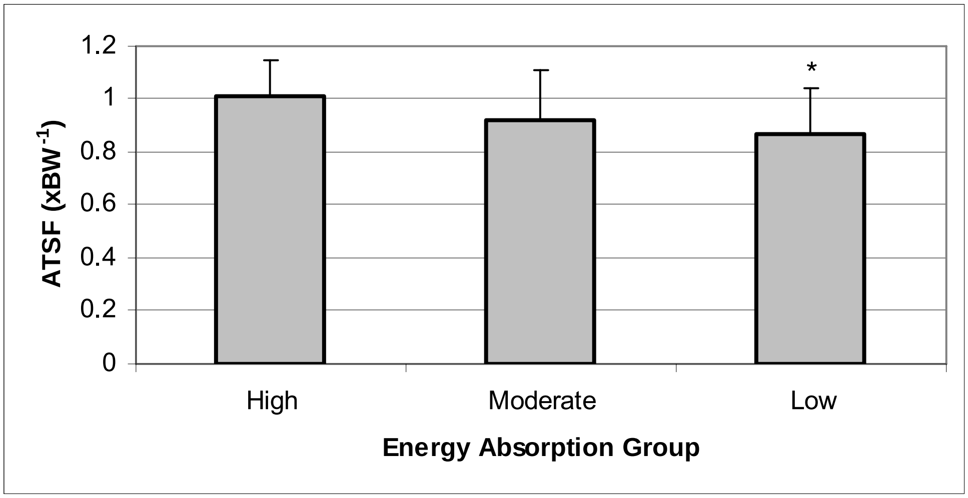
Significant at $P < 0.05$

Figure 1. Sagittal plane EA group comparison of peak internal knee extension moment (KEM) during the double leg jump landing task



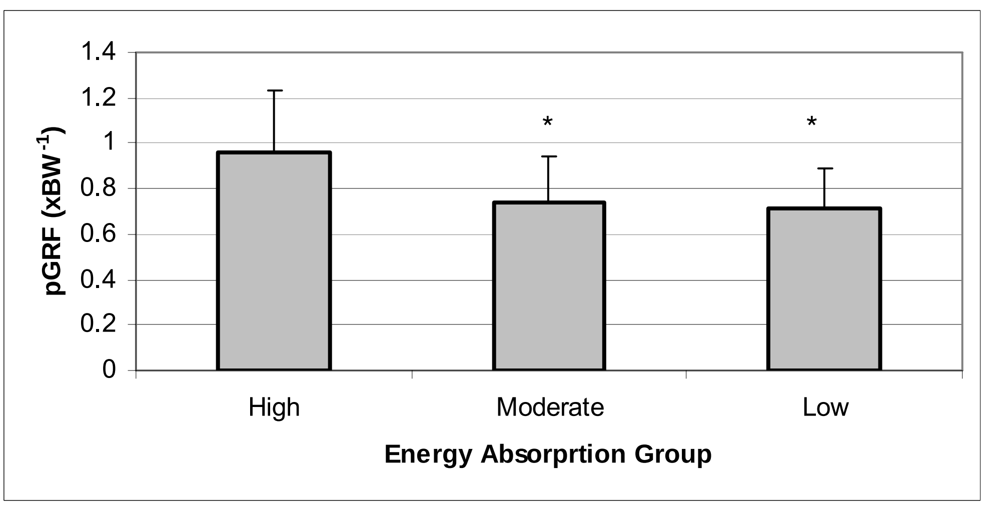
*Significantly different from High EA group ($P < 0.05$)

Figure 2. Sagittal plane EA group comparison of peak anterior tibial shear force (ATSF) during the double leg jump landing task



*Significantly different from High EA group (*P* < 0.05)

Figure 3. Sagittal plane EA group comparison of peak posterior ground reaction force (pGRF) during the double leg jump landing task



*Significantly different from High EA group (*P* < 0.05)

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APPENDIX TWO: MANUSCRIPT II

Manuscript II

Lower extremity energy absorption and landing biomechanics. Part II: Frontal plane analyses and inter-planar energy absorption relationships.

ABSTRACT

Context: Greater sagittal plane energy absorption (EA) by the lower extremity musculature during the initial impact phase (INI: 100 ms immediately following ground contact) of landing is consistent with a biomechanical profile that increases anterior cruciate ligament (ACL) strain due to sagittal plane mechanisms. While sagittal plane EA does not influence frontal plane biomechanics that are associated with ACL injury, it is unknown whether frontal plane EA is related to frontal plane landing biomechanics that may increase the risk of ACL injury, or if there is a relationship between the magnitudes of sagittal and frontal plane EA.

Objective: To: 1) evaluate relationships between frontal plane EA and frontal plane landing biomechanics, 2) compare landing biomechanics between high, moderate and low frontal plane EA groups, and 3) evaluate the relationships between frontal and sagittal plane EA during landing.

Design: Descriptive laboratory study.

Setting: Research laboratory.

Patients or Other Participants: Eighty-two healthy, physically active volunteers.

Intervention(s): Landing biomechanics were assessed using an electromagnetic motion capture system and force plate during double leg jump landings.

Main Outcome Measure(s): Frontal and sagittal plane total, hip, knee, and ankle INI EA. Total frontal plane INI EA was used to group participants into high, moderate, and low EA tertiles. Frontal plane knee angle at ground contact, and peak vertical and posterior ground reaction forces, internal knee varus moment, and knee valgus and hip adduction angles were identified during the landing task. Simple bivariate correlations were used to evaluate the relationships between frontal plane EA, and the biomechanical factors and sagittal plane EA. Biomechanical variables were compared across EA group using one-way ANOVA.

Results: Greater total and knee frontal plane INI EA were associated with greater knee valgus angle at ground contact, and greater peak knee valgus angle, hip adduction angle, posterior ground reaction force and knee varus moment, while greater frontal plane ankle INI EA was related to greater peak knee varus moment ($R^2 = 0.051 – 0.536$, $P < 0.05$). The High EA group exhibited greater knee valgus angle at IGC ($P = 0.001$) and greater peak hip adduction angle ($P = 0.007$) compared to the Low EA group. The High EA group also displayed greater peak knee valgus angle during landing compared to the Moderate EA group ($p < 0.001$) and Low EA group ($p < 0.001$) as well as greater peak knee varus moment than the Moderate EA ($p = 0.001$) and Low EA ($p < 0.001$) groups. The majority of frontal and sagittal plane EA relationships were not significant with only greater sagittal knee EA being associated with greater frontal hip EA ($r = 0.301$, $p = 0.006$), and greater sagittal ankle EA being associated with greater frontal ankle EA ($r = 0.224$, $p = 0.043$).

Conclusions: Greater frontal plane INI EA is associated with a less favorable frontal plane biomechanical landing profile that likely results in greater ACL loading due to frontal plane mechanisms. Additionally, the magnitudes of sagittal and frontal plane EA during landing are independent. Individuals absorbing large magnitudes of energy in both planes immediately following landing may have an increased risk of ACL injury.

INTRODUCTION

Females display a two-to-eight times greater risk of non-contact anterior cruciate ligament (ACL) injury compared with males.[1, 2] Accordingly, a great deal of research has focused on identifying neuromechanical differences between sexes to discover the underlying mechanism for non-contact ACL injury.[3-7] Greater knee valgus angle at initial ground contact (IGC) and peak knee valgus angle during landing have been identified in females compared to males.[4, 6, 8, 9] Further, frontal plane knee loading has been shown both *in vivo* using biomechanical modeling[10, 11] and *in vitro*[12] to contribute to greater ACL loading. Consequently, knee valgus angle and frontal plane knee moment have been identified as predictors of non-contact ACL injury risk, and limiting these frontal plane biomechanical factors has been advocated to decrease ACL injury risk.[9, 13]

We demonstrated in Part I of this investigation that greater sagittal plane lower extremity energy absorption (EA) during the initial impact phase (INI: 100 ms following IGC) of double leg jump landings resulted in a sagittal plane biomechanical profile that likely contributes to greater ACL loading.[14] Specifically, greater peak posterior ground reaction force (pGRF), internal knee extension moment (KEM), and anterior tibial shear force (ATSF) were observed in the highest EA group compared to the lowest EA group.[14] As a result, it was proposed that landing with greater sagittal plane EA during INI results in greater ACL loading due to sagittal plane mechanisms. However, no EA group differences were identified for frontal plane knee kinematics or kinetics, indicating that the magnitude of sagittal plane EA may not directly influence frontal plane landing biomechanics thought to contribute to ACL loading. Therefore, it is important to expand this energetic analysis beyond the sagittal plane to evaluate whether frontal plane EA influences ACL loading attributable to frontal plane mechanisms.

To our knowledge, there is currently only one published report which has directly evaluated frontal plane EA during landing. Using a double leg drop landing task, Yeow et al.[15] observed greater frontal plane EA at the hip and knee compared with the ankle and an increase in frontal plane EA in response to increased landing height. However, this investigation did not explicitly evaluate the relationships between frontal plane EA and frontal plane biomechanics that have been associated with ACL injury.[15] Lloyd and Buchanan[16, 17] have demonstrated that the quadriceps and hamstrings musculature can support varus-valgus loading of the knee during both isometric and dynamic tasks, primarily via co-contraction. These results indicate a potential inter-planar EA relationship whereby greater sagittal plane knee EA (eccentric contraction of the quadriceps) could enhance frontal plane support. As a result, the magnitude of frontal plane EA and frontal plane knee loading during landing might be mediated by EA in the sagittal plane.

This notion is supported by Pollard et al.[18] who reported that females exhibiting greater combined peak hip and knee flexion during double leg drop landings displayed greater sagittal plane EA but lesser peak knee valgus angle and average internal knee varus moment. These authors postulated that the greater sagittal plane EA in the high flexion group necessitated lesser frontal plane knee EA, thus reducing ACL loading due to frontal plane mechanisms. However, there are two primary limitations to this study. First, Pollard et al.[18] calculated EA from IGC to peak knee flexion. Our previous work indicates that greater EA during the INI period is unfavorable in terms of ACL loading, while greater EA later in the landing phase is more desirable.[19] Therefore, the time interval used by Pollard et al. could obscure the temporal relationship between sagittal plane EA and biomechanical ACL injury risk factors. Further, peak ACL loading[20, 21] and injury[22] occur within the first 100 ms of landing, indicating that evaluating EA during this INI phase may be more applicable to ACL injury risk. Second, the magnitude of frontal plane EA was not actually calculated by Pollard et al.,[18] leaving any potential inter-planar EA relationships purely speculative.

The current investigation sought to expand upon our sagittal plane EA analyses by 1) determining if frontal plane EA is associated with frontal plane biomechanics related to ACL injury; 2) comparing the magnitudes of these biomechanical variables between groups displaying high, moderate, and low frontal plane EA; and 3) evaluating the relationships between EA in the sagittal and frontal planes. We hypothesized that greater frontal plane EA would be associated with a less favorable biomechanical profile, and that the high frontal plane EA group would display the least desirable biomechanical values compared to the moderate and low frontal plane EA groups. Additionally, we hypothesized that greater sagittal plane EA would be associated with lesser frontal plane EA.

METHODS

Participants

Eighty-two physically active (participating in at least 30 minutes of physical activity 3 times per week) volunteers (41 males, 41 females; age = 20.1 ± 2.4 years; height = 1.74 ± 0.10 m; mass = 70.3 ± 16.1 kg) participated in this study after reading and signing an Institutional Review Board approved consent form. All subjects were generally healthy with no history of ACL injury, neurological disorder, lower extremity surgery, or lower extremity injury within the six months prior to data collection.

Subject Preparation and Experimental Procedures

Prior to data collection, the height and mass of each subject were recorded and used for generation of the biomechanical model and normalization of the dependent variables. Lower-extremity and trunk kinematics were assessed using an electromagnetic motion capture system (MotionStar, Ascension Technology Corp., Burlington, VT, USA). Electromagnetic tracking sensors were positioned over the third metatarsal, anteromedial aspect of the shank, and lateral thigh of the dominant leg (the leg used to kick a ball for maximum distance), and the sacrum and C7 spinous process. Sensors were placed over areas of minimal muscle mass, and secured with pre-wrap and athletic tape to reduce motion artifact. Global and segment axis systems were established with the positive X axis designated as forward/anteriorly, the positive Y axis leftward/medially, and the positive Z axis upward/superiorly. A segment-linkage model of the dominant lower extremity, pelvis, and thorax was created using the MotionMonitor motion analysis software (Innovative Sports Training, Inc., Chicago, IL, USA) by digitizing the ankle, knee, and hip joint centers and the T12 spinous process. The midpoints of the digitized medial and lateral malleoli and the medial and lateral femoral condyles defined the ankle and knee joint centers, respectively. The hip joint center was predicted using external landmarks on the pelvis as described by Bell et al.[23]

Subjects were instructed to stand atop a 30 cm high box that was set a distance equal to 50% of the subjects' height away from the edge of a nonconductive force plate (Type 4060-NC, Bertec Corporation, Columbus, OH, USA) whose axis system was aligned with the global axis system. They then performed double leg jump landings by jumping down and forward toward the force plate, contacting the ground with both feet at the same time with their dominant foot near the center of the force plate and their non-dominant foot positioned next to the force plate, and then immediately jumping up for maximum height using both legs. Subjects performed 3 practice trials and 5 successful testing trials with 30 seconds of rest between trials to minimize the potential effects of fatigue. Trials in which the subject jumped from the box and landed with both feet at the same time, completely contacted the force plate with only the dominant foot, and performed the landing task and subsequent maximum jump in a fluid motion were deemed successful.

Data Sampling and Reduction

Kinematic and kinetic data were sampled at 120 and 1,200 Hz, respectively, using the MotionMonitor motion analysis software. Raw kinematic data were low-pass filtered at 10 Hz (4th order zero-phase lag Butterworth),[24] time-synchronized with the kinetic data, and re-sampled at 1,200 Hz. Joint angular positions were calculated using Euler angles based on a right hand convention in a YX'Z'' rotation sequence. Instantaneous joint angular velocities were calculated as the 1st derivative of angular position. Motion was defined about the ankle as the foot relative to the shank, about the knee as the shank relative to the thigh, and about the hip as the thigh relative to the sacrum. Ground reaction forces were low-pass filtered using a fourth-order, zero-phase-lag Butterworth filter with a cutoff frequency of 60 Hz,[25] and combined with kinematic and anthropometric data to calculate the net internal joint moments of force at the ankle, knee, and hip, and the net internal force on the shank at the knee joint using an inverse dynamics solution.[26]

Custom computer software (LabVIEW, National Instruments Corporation, Austin, TX, USA) was used to calculate frontal and sagittal plane hip, knee, and ankle joint power curves as the product of angular velocity and net joint moment ($P = M \times \omega$). The negative portions of joint power curves were then integrated to calculate negative mechanical joint work[24, 27-29] during the INI[24, 30] phase of landing. Next, total sagittal and frontal plane joint work was calculated by summing the negative joint work at each individual joint during this time interval.[28, 29, 31] Negative joint work values represent energy absorption by the muscle-tendon unit, [27, 32] and all EA values were assigned to be positive by convention. The same custom software was also used to identify frontal plane knee angle at IGC, and peak values for knee valgus and hip adduction angles, vertical ground reaction force (vGRF), pGRF, and internal knee varus (KVM) moment during the interval from IGC to the minimum vertical position of the whole body center of mass.[25, 31] GRFs were normalized to subject body weight (x BW^{-1}), KVM normalized to the product of subject height and weight (x $[BW*Ht]^{-1}$), and EA expressed as a percentage of the product of subject height and weight (% BW*Ht). All dependent variables were averaged across the five jump landing trials of each subject prior to statistical analysis.

Statistical Analysis

Simple, bivariate Pearson correlation coefficients were used to assess the relationships between total, hip, knee, and ankle frontal plane INI EA during double-leg jump landings and the criterion biomechanical variables. The magnitude of total frontal plane EA during INI was then used to create three distinct frontal plane EA groups: High, Moderate, and Low, respectively. Static comparisons across EA groups for each biomechanical factor were made using separate one-way ANOVA models. Pairwise comparisons with a Bonferroni correction for multiple independent *t*-tests were employed for significant ANOVA models to identify specific group differences on these dependent variables. Finally, simple, bivariate Pearson correlation coefficients were used to assess the relationships between frontal and sagittal plane EA during the INI landing phase. All analyses were conducted using commercially available software (SPSS 17.0, SPSS Inc., Chicago, IL, USA) with statistical significance established a priori as $\alpha \leq 0.05$.

RESULTS

Table 1 displays the means and standard deviations for frontal and sagittal plane EA during the INI phase of landing. Means and standard deviations for the key biomechanical factors associated with non-contact ACL injury are presented in **Table 2**.

Frontal Plane EA and Biomechanical Factor Relationships

Correlation coefficients between total, hip, knee, and ankle EA in the frontal plane during the INI phase of landing, and the biomechanical factors related to ACL injury are shown in **Table 3**. Significant relationships were identified between total frontal plane EA and frontal plane knee angle at IGC ($r = -0.518$, $p < 0.001$), peak knee valgus angle ($r = -0.662$, $p < 0.001$), peak hip adduction angle ($r = 0.462$, $p < 0.001$), peak pGRF ($r = 0.225$, $p = 0.042$), and peak KVM ($r = 0.698$, $p < 0.001$). Frontal plane knee EA during INI was also significantly associated with frontal plane knee angle at IGC ($r = -0.589$, $p < 0.001$), peak knee valgus angle ($r = -0.732$, $p < 0.001$), peak hip adduction angle ($r = 0.462$, $p < 0.001$), peak pGRF ($r = 0.279$, $p = 0.011$), and peak KVM ($r = 0.717$, $p < 0.001$). These results indicate that greater total and knee frontal plane INI EA are associated with greater knee valgus angle at IGC, peak knee valgus, peak hip adduction, peak pGRF, and peak KVM. Further, greater peak KVM was also related to greater frontal plane ankle INI EA ($r = 0.260$, $p = 0.018$). There were no other significant relationships between frontal plane ankle and hip INI EA and the biomechanical factors of interest.

Frontal Plane EA Group Comparisons

Subject allocation to tertiles based upon total frontal plane INI EA was successful in creating three groups demonstrating high, moderate, and low frontal plane EA ($F_{2,79} = 55.501$, $p < 0.001$) (**Table 4**). One-way ANOVA detected significant EA group differences for frontal plane knee angle at IGC ($F_{2,79} = 5.782$, $p = 0.005$), peak knee valgus angle ($F_{2,79} = 19.874$, $p < 0.001$), peak hip adduction angle ($F_{2,79} = 4.529$, $p = 0.014$), peak pGRF ($F_{2,79} = 4.030$, $p = 0.022$), and peak KVM ($F_{2,79} = 17.883$, $p = 0.001$), but no group differences for peak vGRF ($F_{2,79} = 0.444$, $p = 0.643$) (**Table 4**). Post hoc testing revealed that that the High EA group landed with significantly greater knee valgus angle at IGC than the Low EA group ($p = 0.001$), and displayed significantly greater peak knee valgus angles during landing compared to both the Moderate EA ($p < 0.001$) and Low EA ($p < 0.001$) groups. The High EA group also demonstrated significantly greater peak hip adduction angle compared to the Low EA group ($p = 0.007$), and greater peak KVM during landing than the Moderate EA ($p = 0.001$) and Low EA ($p < 0.001$) groups.

Inter-planar EA Relationships

Table 5 displays the correlation coefficients between sagittal and frontal plane EA during the INI phase of landing. Greater sagittal plane knee INI EA was associated with greater frontal plane hip INI EA ($r = 0.301$, $p = 0.006$); and greater sagittal plane ankle INI EA was associated with greater frontal plane ankle INI EA ($r = 0.224$, $p = 0.043$). No other significant relationships between frontal and sagittal plane EA during the INI phase were identified.

DISCUSSION

The principal findings of Part II of this investigation are that greater frontal plane EA during the INI phase of landing is associated with a less favorable frontal plane biomechanical profile that likely contributes to greater ACL loading. Additionally, there is not a significant inter-planar EA relationship such that greater sagittal plane EA mitigates the magnitude of frontal plane EA in the 100 ms immediately following ground contact during double-leg jump landings.

The associations between frontal plane EA and the biomechanical factors of interest generally agreed with our hypotheses. As expected, greater frontal plane total and knee INI EA were significantly associated with less desirable values for all biomechanical factors except for peak vGRF (Table 3). However, similar associations between frontal plane hip and ankle INI EA and the biomechanical factors of interest were not observed. At these joints, only greater ankle INI EA was correlated with greater peak KVM, and the strength of this association was relatively weak (r = 0.260). These results indicate that greater total EA in the frontal plane during INI, which is primarily achieved at the knee (70% of the total frontal plane INI EA), is indicative of a frontal plane biomechanical landing profile consistent with greater knee valgus angles (peak and at IGC) and frontal plane knee moment; factors that are important as they have been shown prospectively to be associated with future ACL injury.[9]

The results of the frontal plane INI EA group comparisons also agreed with our hypotheses. Greater knee valgus angle at IGC, peak knee valgus and hip adduction angles, peak KVM, and peak pGRF were identified in the High EA group compared to the Low EA group with only peak vGRF not differing between the EA groups (Table 4). However, the lack of group differences in peak vGRF was not surprising given our previous sagittal plane analyses[33] and the fact that investigations comparing peak vGRF between sexes (i.e. higher and lower ACL injury risk) are generally equivocal.[24, 28, 34, 35] With respect to frontal plane knee biomechanics, the mean differences between EA groups also appear to be consequential when compared with previous investigations. Hewett et al.[9] reported that females who went on to suffer a non-contact ACL injury demonstrated 8.4° more knee valgus angle at IGC, 7.6° greater peak knee valgus angle, and about 2.5 times more frontal plane knee moment than uninjured females. By comparison, the High EA group displayed 6.6° more knee valgus angle at IGC, 14.4° greater peak knee valgus angle, and about 2.1 times greater frontal plane knee moment compared to the Low EA group. While the current investigation is clearly limited in drawing any conclusions regarding injury outcome, it is apparent that the High EA group displayed frontal plane knee biomechanics that are sufficiently different than the Low EA group to potentially result in greater frontal plane knee loading. Accordingly, we propose that landing strategies with greater total frontal plane INI EA are likely to cause greater ACL loading due to frontal plane mechanisms.

Despite the greater risk of non-contact ACL injury in females,[2] we did not identify a relationship between sex and sagittal plane INI EA group in Part I of this investigation.[33] Therefore, we performed a secondary analysis to determine whether there was a significant association between sex and total frontal plane EA group assignment (High vs. Low). In contrast to the sagittal plane, we identified a significant association between sex and frontal plane INI EA group ($X^2 = 4.909$, $p = 0.027$) with females being 3.6 times more likely to be in the High INI EA group. While males and females have an equal likelihood of landing with greater sagittal plane EA and subsequently greater ACL loading due to sagittal plane mechanisms, females are more likely to absorb greater energy in the frontal plane during INI and load the ACL via frontal plane mechanisms. Additionally, as combined, multi-planar knee loading has been shown to result in greater ACL strain than pure sagittal or frontal plane loading,[12] we suggest that this increased likelihood of greater frontal plane INI EA coupled with a similar chance of landing with greater sagittal plane INI EA in females may contribute to their increased risk of ACL injury. To further evaluate this notion, we identified

19 subjects who were assigned to either the High sagittal and High frontal plane (2 males, 8 females) or Low sagittal and Low frontal plane (6 males, 3 females) EA groups. While we were unable to identify a significant association between sex and High-High vs. Low-Low group allocation due to small cell frequencies ($X^2 = 4.232$, Exact $p = 0.070$), we did determine that females were 8.00 times more likely to be in the High-High group than males ($p = 0.050$). Additionally, when compared across key ACL-related biomechanical variables, the High-High group demonstrated 7.5° greater knee valgus at IGC, 11.3° more peak knee valgus angle, 71% more peak pGRF, 46% more peak KEM, and 115% more peak KVM ($p < 0.05$) than the Low-Low group. We suggest that identification of individuals who perform greater magnitudes of INI EA in both the sagittal and frontal planes during landing may be a means to accurately discriminate individuals who display high-risk landing biomechanics in multiple planes.

Finally, the lack of a consistent association between frontal and sagittal plane EA was unexpected and differed from our hypotheses, as we anticipated that greater sagittal plane EA would mitigate the magnitude of frontal plane EA required during landing (Table 5). Apart from relatively weak associations between sagittal plane knee and frontal plane hip (r = 0.301) and sagittal and frontal plane ankle (r = 0.224) EA, there were no significant relationships identified between the magnitudes of sagittal and frontal plane EA during INI. This is in stark contrast to both our expectations and previous research that has postulated that greater sagittal plane EA would limit frontal plane EA and thus frontal plane knee loading.[18] Though it appears counterintuitive that the magnitude of sagittal and frontal plane EA are independent, it is important to note that there is not a fixed magnitude of energy that is absorbed by all individuals during such a limited portion (100 ms) of landing. Even though the total energy of the system during these landings is relatively standardized, in addition to energy being absorbed via eccentric contraction, energy may be transformed into translational and rotational kinetic energy, as well as potential energy in each segment of the system (body) and in each plane.[32, 36] As a result, the magnitude of energy needed to be absorbed via eccentric muscle contraction during this time period is variable and dependent upon the motion of the individual segments. Therefore, the magnitude of INI EA in the sagittal plane does not necessarily influence the magnitude of frontal plane INI EA resulting in the lack of association noted in the current investigation.

CONCLUSION

Despite the fact that ACL strain is greater under a combination of anterior shear force and frontal plane knee moment compared with the isolated application of these components,[12, 37] considerable disagreement continues to persist about whether sagittal [38, 39] or frontal plane[10] loading is most responsible for ACL injury. In Part I of this investigation, we demonstrated that greater INI EA in the sagittal plane was indicative of a biomechanical landing profile with greater peak internal knee extension moment, anterior tibial shear force, and pGRF that likely results in greater ACL loading due to sagittal plane mechanisms. Further, no association was identified between sex and sagittal plane INI EA group signifying that there is an equal likelihood for males and females to land using this deleterious sagittal plane strategy. In Part II of this study, we reported that greater frontal plane INI EA was indicative of frontal plane landing biomechanics that likely increase ACL loading due to purely frontal plane mechanisms; and that females were 3.6 times more likely than males to exhibit higher frontal plane INI EA during landing. Additionally, we found that there was not a significant relationship between the magnitudes of sagittal and frontal plane EA during the INI phase of landing indicating that these values are independent of one another. Given these findings, we hypothesize that individuals who absorb a higher magnitude of energy in both the sagittal and frontal planes immediately following ground contact would be at the highest risk of non-contact ACL injury, as they would experience greater combined sagittal and frontal plane ACL loading. However, future prospective investigation is necessary to test this hypothesis. Further, we speculate that the increased risk of ACL injury noted in females may be due to the fact that females are significantly more likely than males to land with higher frontal plane INI EA; but just as likely to land with high sagittal plane INI EA which would make them more likely to be subjected to greater combined sagittal and frontal plane ACL loading. As such, we suggest that identifying biomechanical factors contributing to greater sagittal and frontal plane INI EA in future studies is paramount and might assist in the design of more efficacious ACL injury prevention programs. Given that the magnitude of EA during landing is influenced by factors that affect either joint moments or joint angular velocities,[27, 40, 41] we suggest that changing modifiable parameters like muscular strength, muscle activation, initial contact joint positions, and the magnitude of joint motion during landing may successfully alter INI EA and potentially reduce ACL injury risk.

Acknowledgements

This study was supported by the NATA Research & Education Foundation Doctoral Grant Program.

Table 1. Sagittal and frontal plane INI EA descriptives (Mean ± SD)

Energy Absorption (%BW*Ht)		
	Sagittal Plane	Frontal Plane
Total	13.62 ± 3.02	1.53 ± 1.24
Hip	2.26 ± 1.34	0.20 ± 0.26
Knee	8.98 ± 2.69	1.05 ± 1.08
Ankle	2.37 ± 1.64	0.28 ± 0.32

Table 2. Frontal plane biomechanical descriptives during the double leg jump landing task

	Frontal knee angle at IGC (°)	Peak knee valgus (°)	Peak hip adduction (°)	Peak vGRF (x BW^{-1})	Peak pGRF (x BW^{-1})	Peak KVM (x $[BW*Ht]^{-1}$)
Mean	-6.81	-17.04	2.96	2.91	0.81	0.085
SD	7.60	10.54	7.39	0.79	0.24	0.046

Table 3. Simple bivariate correlations between frontal plane INI EA and frontal plane biomechanics during the double leg jump landing task (Significant at $P < 0.05$)

Biomechanical Variables	Frontal Plane EA INI Total	INI Hip	INI Knee	INI Ankle
FPK angle at IGC	r = -0.518 p < 0.001	r = -0.048 p = 0.665	r = -0.589 p < 0.001	r = 0.013 p = 0.905
Peak knee valgus angle	r = -0.662 p < 0.001	r = -0.036 p = 0.750	r = -0.732 p < 0.001	r = -0.073 p = 0.515
Peak hip adduction angle	r = 0.462 p < 0.001	r = -0.040 p = 0.724	r = 0.462 p < 0.001	r = 0.155 p = 0.165
Peak vGRF	r = 0.144 p = 0.197	r = 0.046 p = 0.680	r = 0.139 p = 0.211	r = 0.051 p = 0.651
Peak pGRF*	r = 0.225 p = 0.042	r = -0.071 p = 0.529	r = 0.279 p = 0.011	r = -0.007 p = 0.949
Peak KVM	r = 0.698 p < 0.001	r = 0.037 p = 0.741	r = 0.717 p < 0.001	r = 0.260 p = 0.018

Table 4. Frontal plane EA group comparisons for frontal plane landing biomechanics (High EA group significantly different from Low (*) and Moderate (†) EA groups, P < 0.05)

Variable	EA Group	Mean ± SD	95% CI	$F_{(2,79)}$	P-value
Total frontal plane EA INI (%BW*Ht)	High*†	2.81 ± 1.37	(2.27, 3.35)	55.501	< 0.001
	Moderate	1.23 ± 0.24	(1.14, 1.32)		
	Low	0.55 ± 0.21	(0.47, 0.63)		
Frontal plane knee angle at IGC (°)	High*	-10.34 ± 7.81	(-13.43, -7.25)	5.782	0.005
	Moderate	-6.38 ± 7.69	(-9.36, -3.40)		
	Low	-3.73 ± 5.89	(-6.06, -1.40)		
Peak knee valgus angle (°)	High*†	-25.41 ± 8.66	(-28.83, -21.98)	19.874	< 0.001
	Moderate	-14.75 ± 10.31	(-18.75, -10.75)		
	Low	-11.04 ± 6.69	(-13.68, -8.39)		
Peak hip adduction angle (°)	High*	6.25 ± 7.74	(3.19, 9.32)	4.529	0.014
	Moderate	1.90 ± 6.88	(-0.77, 4.57)		
	Low	0.76 ± 6.59	(-1.85, 3.37)		
vGRF (xBW^{-1})	High	2.97 ± 0.67	(2.71, 3.24)	0.444	0.643
	Moderate	2.96 ± 0.95	(2.60, 3.34)		
	Low	2.80 ± 0.73	(2.51, 3.08)		
pGRF (xBW^{-1})	High	0.91 ± 0.27	(0.80, 1.02)	4.030	0.022
	Moderate	0.75 ± 0.23	(0.66, 0.84)		
	Low	0.76 ± 0.20	(0.68, 0.84)		
KVM ($x[BW*Ht]^{-1}$)	High*†	0.119 ± 0.047	(0.101, 0.139)	17.883	< 0.001
	Moderate	0.079 ± 0.036	(0.065, 0.093)		
	Low	0.058 ± 0.031	(0.046, 0.070)		

Table 5. Simple bivariate correlations between frontal and sagittal plane EA during the INI and TER phases of the double leg jump landing task (Significant at P < 0.05)

	Frontal Plane INI EA			
Sagittal Plane INI EA	INI Total	INI Hip	INI Knee	INI Ankle
Total	r = -0.015 p = 0.890	r = 0.139 p = 0.212	r = -0.054 p = 0.628	r = 0.010 p = 0.928
Hip	r = -0.095 p = 0.398	r = -0.117 p = 0.296	r = -0.096 p = 0.391	r = 0.050 p = 0.653
Knee	r = 0.002 p = 0.987	r = 0.301 p = 0.006	r = 0.025 p = 0.823	r = -0.151 p = 0.175
Ankle	r = 0.046 p = 0.683	r = -0.141 p = 0.208	r = 0.019 p = 0.862	r = 0.224 p = 0.043

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APPENDIX THREE: MANUSCRIPT III

Manuscript III

The influence of sex and landing posture on lower extremity energy absorption during drop landings

Abstract

Background: Females are at a greater risk of anterior cruciate ligament (ACL) injury compared to males with a more erect landing posture suggested to contribute to this greater risk of injury. It has been suggested that the more erect landing posture of females is the result of a sex-specific sagittal plane energy absorption (EA) strategy where females utilize a greater contribution to total EA from the ankle, and absorb a greater magnitude of energy during the 100 ms following ground contact which has been associated with high risk landing biomechanics related to ACL injury. However, as this sex-specific strategy has only been shown when initial landing postures differ between sexes, it is unknown whether sex or landing posture is responsible influencing EA strategy.

Methods: Total, hip, knee and ankle energy absorption were measured in 50 individuals (29 males, 21 females) performing 60-cm terminal drop landings under three conditions: preferred, flexed, and erect during which landing postures were controlled.

Findings: Sex differences in EA strategy were not identified when males and females landed with similar lower extremity postures. The magnitudes of EA during landing as well as the relative joint contributions to total EA were influenced by landing posture, but not sex. All subjects demonstrated lesser ankle and total EA, lesser ankle contribution to total EA, and greater hip contribution to total EA when landing in a flexed vs. the preferred and erect postures.

Interpretation: The more erect landing posture of females that has been reported in the literature is most likely influenced by other sex-related factors such as strength and is not the result of a sex-specific EA strategy.

Keywords: Energy Absorption; ACL; Anterior cruciate ligament; Landing; Sex differences

1. Introduction

Compared to males, females have a significantly greater risk of anterior cruciate ligament (ACL) injury (Agel et al., 2005; Arendt et al., 1999; de Loës et al., 2000; Myklebust et al., 1997) and tend to exhibit a more erect posture during landing with the knee joint positioned in less flexion at ground contact (Decker et al., 2003; Malinzak et al., 2001; Yu et al., 2006). During landing, the quadriceps acts eccentrically to control knee flexion and has the greatest potential for generating anterior tibial shear force and loading the ACL at knee flexion angles between 10-30° (Draganich and Vahey, 1990; Griffin et al., 2000; Kirkendall and Garrett, 2000). Further, the posterior tibial shear force component of the hamstrings muscles, which can protect against excessive ACL loading, decreases as the knee joint is moved to less flexed positions (Pandy and Shelburne, 1997). This combination of increased ACL loading secondary to quadriceps contraction and decreased ACL protection provided by the hamstrings when landing in a more erect position has been implicated as one possible factor for the observed sex difference in ACL injury risk. As a result, increasing knee flexion during landing through technique instruction has been adopted as a common component in ACL injury prevention programs (Hewett et al., 1999; Mandelbaum et al., 2005; Myklebust et al., 2003). However, the underlying reasons for the more erect landing position of females continue to remain unknown.

Decker et al. (2003) postulated that sex differences in landing postures are the result of sex-specific sagittal plane energy absorption (EA) strategies in which males and females preferentially utilize either the hip or ankle, respectively, in conjunction with the knee as the primary joints with which to absorb energy. They proposed that the erect landing posture of females is caused by this female specific ankle and knee joint dominant EA strategy, with the erect landing posture serving to maximize the energy absorbed by females during landing (Decker et al., 2003). However, landing posture may also influence joint contributions to EA by affecting joint angular velocities, joint moments, and the subsequent joint power profile (Zhang et al., 2000). Sex differences in joint contributions to EA have only been reported when accompanied by differences in initial landing posture (Decker et al., 2003; Norcross et al., 2010). Therefore, it is unknown whether sex-specific EA strategies truly exist and are responsible for determining the initial joint configurations used during landing. It is possible that instead of a sex-specific EA strategy, other sex-related biomechanical factors, such as strength, are responsible for the observed sex differences in landing posture; and that landing posture alone influences the relative joint contributions to EA. Clinically, this distinction is significant as it determines whether simple biomechanical factors, or a more complex feed-forward EA strategy, would need to be changed in order to facilitate a safer, more flexed landing posture in females.

Given this issue, the purpose of this investigation was to clarify previous research regarding the existence of sex-specific EA strategies by evaluating the influences of sex and landing posture on EA strategy during drop landings in preferred, flexed, and erect landing postures. We hypothesized that compared to males; females would demonstrate a more erect landing posture and an ankle/knee dominant EA strategy in the preferred condition; but that no sex differences in EA would be identified after controlling for landing posture (i.e. during flexed and erect conditions). Further, we hypothesized that the magnitude of EA and the relative joint contributions to total sagittal plane EA would be significantly influenced by landing posture, but that sex would not modify these effects.

2. Methods

2.1 Subjects

Eighty physically active (40 females, 40 males) volunteers participated in this investigation after reading and signing an Institutional Review Board approved consent form. All subjects were recreationally active (participating in at least 30 minutes of physical activity at least three times per week); and generally healthy with no history of ACL injury, neurological disorder, lower extremity surgery, or lower extremity injury within the six months prior to data collection.

2.2 Subject and equipment preparation

The height and mass of each subject were recorded prior to data collection, and used for generation of the biomechanical model and normalization of the dependent variables. An electromagnetic motion capture system (MotionStar, Ascension Technology Corp., Burlington, VT, USA) and five 6 degree of freedom electromagnetic tracking sensors were used to assess lower-extremity and trunk kinematics. Sensors were positioned over the third metatarsal, anteromedial shank, and lateral thigh of the dominant leg (defined as the leg used to kick a ball for maximum distance), as well as the sacrum and C7 spinous process. In addition to being placed over areas of minimal muscle mass, the sensors were secured with pre-wrap and athletic tape in order to reduce motion artifact. Global and segmental axis systems were established using a right-hand coordinate system with the positive X axis designated as forward/anteriorly, the positive Y axis leftward/medially, and the positive Z axis upward/superiorly. The MotionMonitor motion analysis software (Innovative Sports Training, Inc., Chicago, IL, USA) was used to create a segment-linkage model of the dominant lower extremity, pelvis, and thorax by digitizing the ankle, knee, and hip joint centers and the T12 spinous process. Ankle and knee joint centers were defined as the midpoints of the digitized medial and lateral malleoli, and the medial and lateral femoral condyles, respectively. The hip joint center was predicted using external landmarks on the pelvis as described by Bell et al. (1989). Finally, a nonconductive force plate (Type 4060-NC, Bertec Corporation, Columbus, OH, USA), whose axis system was aligned with the global axis system, was used to measure reaction forces and moments during the drop landing trials.

2.3 Experimental procedures

Following experimental set-up, subjects completed double leg drop-landings from a height of 0.60 m in three different landing postures: preferred (P), flexed (F), and erect (E). All subjects completed the P condition first to prevent contamination of their preferred landing strategy caused by the artificial F and E landing conditions. For the P condition, subjects stood atop a 0.60 m tall box positioned directly behind the force plate before reaching out with their dominant foot to position it over the force plate (Supplementary Figure A). They were instructed to roll forward off of the box using their non-dominant foot without jumping or lowering themselves in order to initiate a drop; and then to perform a double leg terminal landing with their dominant foot positioned completely on the force plate and their non-dominant foot positioned on the floor next to the force plate. Subjects were given no other instructions regarding landing technique or performance.

Following completion of the P condition, subjects completed drop landings in F and E postures in a counterbalanced order. Drop landings in the F and E conditions required subjects to position their knee in 35 ± 5° and 20 ± 5° of flexion, respectively. These target angles were chosen as they are similar to the mean knee flexion angles at initial contact exhibited by male (F) and female (E) subjects in a previous study that demonstrated sex differences in EA strategy during 0.60 m drop-landings (Decker et al., 2003). Subjects hung from an overhead bar attached to a wooden support frame positioned around the force plate (Supplementary Figure B and C). To maintain a standardized drop height of 0.60 m for all conditions, the height of the overhead bar was adjusted as a function of subject height and the knee and hip joint angles in each condition.

While subjects hung from the bar, they were provided with biofeedback regarding their knee flexion angle using the Motion Monitor motion analysis system and a computer monitor to facilitate landing with F and E postures. As subjects changed their knee flexion angle, they saw a cursor on the screen move in real-time reflecting their joint position. An auditory signal was also triggered when subjects successfully positioned the cursor within the target window. Though participants only received feedback regarding knee joint position of their dominant leg, they were instructed to move both legs in unison. Following proper knee joint positioning, subjects let go of the bar to initiate the drop and were instructed to maintain their body position until the instant of ground contact before completing the terminal landing. Knee flexion angle and vertical ground reaction force were calculated and displayed immediately following each trial using the Motion Monitor software, and used to determine whether the drop landing trial was successful as defined by the knee flexion angle at ground contact being within the specified range for the F and E conditions.

All subjects completed a minimum of 3 practice trials and up to 8 testing trials in the P, F, and E conditions in hopes of capturing 5 successful trials for each condition. Subjects were provided with at least 30 seconds of rest between trials and 2 minutes of rest between conditions to minimize the potential effects of fatigue.

2.3 Data sampling and reduction

Kinematic and kinetic data were sampled at 120 Hz and 1,200 Hz, respectively, using The Motion Monitor motion analysis software. Raw kinematic data were low-pass filtered using a fourth-order, zero-phase-lag Butterworth filter with a cutoff frequency of 10 Hz (Decker et al., 2003), time synchronized with the kinetic data, and then re-sampled to 1,200 Hz. Joint angular positions were calculated based on a right hand convention using Euler angles in a Y (flexion/extension), X' (adduction/abduction), Z" (internal/external rotation) rotation sequence with motion defined about the hip as the thigh relative to the pelvis, about the knee as the shank relative to the thigh, and about the ankle as the foot relative to the shank. Instantaneous joint angular velocities were calculated as the 1st derivative of angular position. Kinetic data were low-pass filtered at 60 Hz (4th order zero-phase lag Butterworth) (Kulas et al., 2006) and combined with kinematic and anthropometric (Dempster et al., 1959) data to calculate the net internal joint moments of force at the hip, knee, and ankle using an inverse dynamics solution (Gagnon and Gagnon, 1992) within The Motion Monitor software.

Custom computer software (LabVIEW, National Instruments, Austin, TX, USA) was used to generate sagittal plane hip, knee, and ankle joint power curves by multiplying joint angular velocities and net joint moments for each drop landing trial ($P = M \times \omega$). Negative mechanical joint work, representing EA by the muscle-tendon unit (McNitt-Gray, 1993; Winter, 2005), was calculated by integrating the negative portion of these joint power curves (Decker et al., 2003; McNitt-Gray, 1993; Schmitz et al., 2007; DeVita et al., 2008) during the 100 ms immediately following initial ground contact (VGRF > 10 N) (Decker et al., 2003; Decker et al., 2002). Total lower extremity EA was calculated by summing the EA at each individual joint (Schmitz et al., 2007; DeVita et al., 2008; Zhang et al., 2000) with the relative contribution of the hip, knee, and ankle to total energy absorption calculated as the EA at the respective joint divided by the total lower extremity EA. Mean EA values were calculated across the five trials for each landing posture and expressed as a percentage of the product of subject height and weight (% BW*Ht) (Norcross et al., 2010). All EA values as well as hip flexion, knee flexion, and ankle plantarflexion were assigned to be positive by convention to simplify their interpretation during data analysis.

2.4 Statistical analyses

Of the 80 subjects tested, 27 participants (19 females and 8 males) were unable to successfully complete drop landings in both the F and E conditions as their mean knee flexion angle at ground contact for their 5 best trials did not meet the established criteria. Further, 3 males were excluded from performing drops landings from the bar due to concerns over the stability of the wooden frame to support their mass. As a result, 50 participants (21 females, Age = 20.2 ± 2.0 years; Height = 1.66 ± 0.06 m; Mass = 59.7 ± 8.9 kg: 29 males; Age = 21.3 ± 2.3 years; Height = 1.81 ± 0.06 m; Mass = 75.7 ± 6.8 kg) were included in the final analysis.

Separate 2 (Sex) x 3 (Joint) repeated measures ANOVAs were used to evaluate sex differences in the magnitude of sagittal plane EA and relative joint contributions to total EA during drop landings in the P condition. Additionally, separate 2 (Sex) x 3 (Joint) repeated measures ANOVAs were used to evaluate whether sex differences in the magnitude of sagittal plane EA and relative joint contributions to total EA still existed after controlling for initial landing kinematics (F condition). Independent samples *t*-tests were used to test for significant sex differences in total sagittal plane EA; and hip, knee, and ankle joint angles at initial ground contact during these same conditions. To evaluate the influences of sex and landing posture on the individual EA magnitudes (total, hip, knee, and ankle) and relative joint contributions (hip, knee, and ankle) to total EA, seven separate 2 (Sex) x 2 (Posture: Flexed vs. Erect) repeated measured ANOVAs were used with planned pairwise comparisons conducted post hoc using a Bonferroni correction for multiple *t*-tests following significant ANOVA models. We chose to compare the F and E conditions in these analyses as pilot testing indicated that the vertical and horizontal velocity of the whole body center of mass at impact in F and E landings were not different in contrast with a slightly greater horizontal velocity at impact in the P landing. All analyses were conducted using commercially available software (SPSS 17.0, SPSS Inc., Chicago, IL, USA) with statistical significance established a priori as $\alpha \leq 0.05$.

3. Results

Table 1 displays descriptive statistics for initial contact joint positions, EA magnitude, and relative joint contributions to total EA stratified by sex. During the preferred condition, there were no sex differences in hip ($t_{48} = 0.726$, $p = 0.471$) or knee ($t_{48} = -0.002$, $p = 0.999$) flexion angles at initial contact, but females demonstrated approximately 7.5° more ankle plantarflexion at contact compared to males ($t_{48} = -2.409$, $p = 0.046$). With respect to EA magnitude, significant main effects for sex ($F_{1,48} = 9.674$, $p = 0.003$) and joint ($F_{2,96} = 45.145$, $p < 0.001$) were identified, but there was not a sex x joint interaction ($F_{2,96} = 0.961$, *p*

= 0.359). Post-hoc testing revealed the knee absorbed a significantly greater magnitude of energy than the hip ($p < 0.001$) and the ankle ($p < 0.001$), and that the magnitude of EA at the ankle was significantly greater than the hip ($p < 0.001$) (Figure 1). A significant main effect of joint ($F_{2,96} = 42.702$, $p < 0.001$) was noted for the contribution to total EA during P, but the main effect of sex ($F_{1,48} = 2.473$, $p = 0.122$) and sex x joint interaction effect ($F_{2,96} = 0.177$, $p = 0.767$) were not significant. The knee contribution to total EA was significantly greater than the ankle ($p < 0.001$) and hip ($p < 0.001$) contributions, while the ankle contribution to total EA was significantly greater than the hip contribution ($p < 0.001$) (Figure 1). Additionally, females performed greater total EA compared to males during this condition ($t_{48} = 3.110$, $p = 0.003$).

In the F condition, there were again no sex differences in hip ($t_{48} = 0.426$, $p = 0.672$) or knee ($t_{48} = 0.574$, $p = 0.569$) flexion angles at initial contact. However, compared to males, females demonstrated approximately 9.5° more ankle plantarflexion at contact ($t_{48} = -2.511$, $p = 0.015$). Similar to the P condition, significant main effects for sex ($F_{1,48} = 13.709$, $p = 0.001$) and joint ($F_{2,96} = 19.600$, $p < 0.001$), but no sex x joint interaction ($F_{2,96} = 0.036$, $p = 0.942$) were identified for EA magnitude (Figure 2). The magnitude of ankle EA was significantly greater than the magnitude of hip EA ($p = 0.001$), and the magnitude of knee EA was greater than the magnitudes of ankle EA ($p = 0.002$) and hip EA ($p < 0.001$). A significant main effect of joint ($F_{2,96} = 21.233$, $p < 0.001$), but no sex main effect ($F_{1,48} = 0.125$, $p = 0.725$) or sex x joint interaction ($F_{2,96} = 0.410$, $p = 0.630$) were identified for the joint contributions to EA (Figure 2). Compared to the hip ($p < 0.001$) and ankle ($p = 0.001$), the knee was the greatest contributor to total EA with the ankle contribution to total EA greater than the hip contribution ($p = 0.001$). Females also absorbed greater total energy than males in the F condition ($t_{48} = 3.702$, $p = 0.001$) (Figure 6).

Four of the seven 2 (Sex) x 2 (Posture) ANOVA models used to evaluate the influences of sex and landing posture on the dependent variables individually were significant. Significant main effects for posture were identified with the F condition exhibiting greater hip contribution to total EA ($F_{1,48}$ = 4.082, p = 0.049), lesser ankle contribution to total EA ($F_{1,48}$ = 11.593, p < 0.001), lesser magnitude of ankle EA ($F_{1,48}$ = 30.722, p < 0.001), and lesser total EA ($F_{1,48}$ = 13.063, p = 0.001) compared to the E condition (Figures 3, 5, and 6). Additionally, there was a main effect for sex ($F_{1,48}$ = 15.170, p < 0.001) with females absorbing greater total EA compared to males. No other significant main effects were noted and there were no significant sex x posture interaction effects identified for any outcome measure (Figures 3-6).

4. Discussion

The primary findings of this investigation are: 1) sex-specific EA strategies during drop landings are not present when the initial landing postures of males and females are similar; and 2) altering landing posture (i.e. knee flexion angle at ground contact) influences both EA magnitude and the relative joint contributions to total EA, but sex does not modify these changes.

Contrary to our hypothesis, we did not observe significant differences in initial hip or knee flexion angles between males and females when performing drop landings using a preferred landing strategy. However, females made contact with the ground in approximately 7.5° more plantarflexion than males which is consistent with previous research (Decker et al., 2003). When using this similar preferred initial landing posture, we did not identify a sex difference in the relative joint contributions to total EA (i.e. sex-specific EA strategies), as all subjects exhibited the greatest contribution to total EA from the knee, a secondary contribution from the ankle, and a tertiary contribution from the hip (Figure 1). Due to the discrepancy between these findings and the results of previous research (Decker et al., 2003), we postulated that excluding individuals who were unable to successfully complete drop landings in all three conditions may have potentially confounded our results by biasing our subject sample. Therefore, we ran secondary analyses on the initial contact angles and relative joint contributions to total EA during the P condition using all 80 subjects in order to confirm our findings. As with the initial analysis, there were no sex differences in initial contact hip and knee flexion angles, but about 7° more ankle plantarflexion was observed at initial contact in females compared to males. Additionally, this analysis identified the same significant main effect of joint ($F_{2,156}$ = 97.974, *p* < 0.001), but no main effect for sex ($F_{1,78}$ = 0.276, *p* = 0.601) or sex x joint interaction effect ($F_{2,156}$ = 2.415, *p* = 0.109). Based upon these identical findings, we are confidant that the results we obtained with our initial analysis are valid, and that these results clearly indicate a lack of sex difference in EA strategy during landings when males and females exhibit similar initial contact postures. However, we are unsure of the reason for the lack of sex differences in initial contact kinematics in this investigation. We specifically chose to utilize the 60-cm drop landing task in order to replicate the experiment performed by Decker et al. (2003) in hopes that we would observe sex differences in landing kinematics during the preferred condition. It is possible that our larger subject sample (50 vs. 17) and/or the younger average age of our subjects (21 vs.27 years of age) contributed to the differing results of the studies using the same task.

As with the preferred condition, we were also unable to detect sex-specific joint EA patterns during landings in the F condition; during which we manipulated males and females to land with the same flexed lower extremity configuration. In these landings, there were no sex differences in hip and knee flexion angles at initial contact, but females again exhibited greater ankle plantarflexion at contact (approximately 9.5°) compared to males. As with the P condition, we observed no differences between males and females in the relative joint contributions to total EA (Knee>Hip>Ankle), and sex did not modify the relative joint contributions to EA (Figure 2). Collectively, we feel that the results from the P and F conditions provide strong evidence that sex-specific feed-forward EA strategies do not exist, as we were unable to detect different EA strategies in males and females when performing landings using both preferred and constrained landing postures.

Our results comparing the F and E postures further indicate that altering initial landing posture significantly influences both the magnitude and relative contributions of select joints to total EA, but that these changes are not modified by sex (Figures 3-5). Compared to the E condition, drop landings in the F condition resulted in significantly greater hip and lesser ankle contributions to total EA for both males and females. However, the mean increase in hip contribution was just 1.5%, while the decrease in ankle contribution was about 3%. These values are in sharp contrast to the 10-20% differences in hip and ankle contributions to total EA that have been reported between sexes (Decker et al., 2003), and following changes in landing height and technique (Zhang et al., 2000). However, we suggest that the limited magnitudes of the observed change in joint contribution to total EA in the current investigation, if evaluated independently, mask the influence that landing posture has on the actual eccentric work performed at each joint during landing.

Compared to the E condition, F landings demonstrated the same magnitude of hip EA, but significantly lesser ankle and total EA. There was also a trend for lesser magnitude of knee EA during the F condition ($F_{1,48} = 3.783$, $p = 0.085$) (Figure 4). Therefore, despite the similar relative joint contributions to total EA in the F and E conditions, there was a greater magnitude of total EA during the 100 ms immediately following ground contact when using a more erect posture (Figure 6). This greater magnitude of EA during the 100 ms after ground contact may be clinically relevant, as recent work from our laboratory indicates that greater total sagittal plane EA during this time interval in individuals performing double leg jump landings likely increases ACL loading due to sagittal plane mechanisms (Norcross et al., In Preparation). However, given the inherent differences between the drop landing and jump landing tasks, generalizing these findings to the current results is speculative. We suggest that future research should directly evaluate the relationships between greater total EA during drop landings and ACL injury related landing biomechanics.

The primary limitation of this investigation is the potential that landings in the F and E conditions were not representative of an individual's true landing performance due to the artificial manner in which we induced the desired landing posture. As a result, we specifically chose not to compare the F and E landing conditions directly to the P condition, but instead opted to only compare these constrained landing conditions to each other. Further, in all three landing conditions, subjects in the current investigation demonstrated mean ankle plantarflexion angles at contact that were 10-40° greater than the initial plantarflexion angles exhibited in the investigation by Decker et al. (2003) (Males = 11.3°; Females = 21.3°). While we cannot rule out that the less plantar flexed position at impact may contribute to the discrepancies noted with our results, we do not believe this to be the case, as females in that study demonstrated the same joint absorption strategy (Hip<Ankle<Knee) as subjects in the current investigation and landed with similar knee and hip flexion positions.

5. Conclusions

Initial landing posture, rather than sex, influences both the magnitude of EA during landing as well as the relative joint contributions to total EA. Compared to an erect landing posture, subjects demonstrated lesser ankle and total EA, lesser ankle contribution to total EA, and greater hip contribution to total EA when landing in a flexed posture, irrespective of sex. Further, we were unable to identify sex differences in EA strategy when males and females landed with similar lower extremity postures, indicating that sex-specific EA strategies likely do not exist. As a result, the more erect landing posture of females that has been reported in the literature is most likely influenced by another sex-related factor such as strength. Future research is necessary to elucidate this factor(s) in order to most effectively elicit greater flexion during landing and potentially reduce ACL injury risk.

Table 1. Joint position at initial ground contact, energy absorption magnitude, and joint contribution to total energy absorption during the preferred, flexed, and erect drop landing conditions stratified by sex [Mean (SD)]

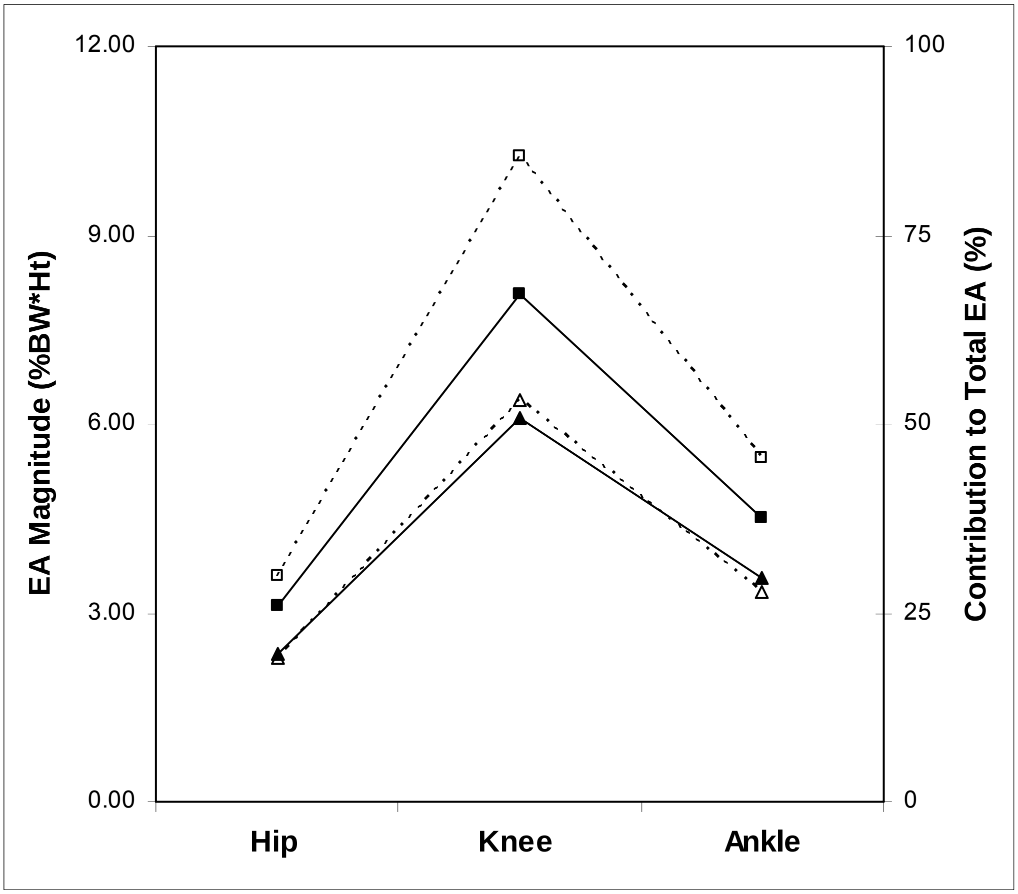
		Position at IGC (°)		
		Hip	Knee	Ankle
Preferred	Male	12.90 (10.78)	16.23 (8.72)	40.64 (13.76)
	Female	14.85 (6.92)	16.24 (8.98)	*48.07 (10.88)
Flexed	Male	26.89 (11.16)	34.00 (2.04)	31.00 (13.71)
	Female	28.15 (8.97)	33.66 (2.24)	*40.16 (11.25)
Erect	Male	15.00 (15.99)	19.95 (2.00)	38.18 (11.46)
	Female	14.37 (7.53)	19.39 (2.12)	*47.59 (9.25)

		Energy Absorption (%BW*Ht)			
		Hip	Knee	Ankle	Total
Preferred	Male	3.12 (1.94)	8.07 (3.58)	4.50 (2.25)	15.68 (4.14)
	Female	3.60 (1.76)	10.24 (3.96)	5.47 (3.25)	*19.32 (3.99)
Flexed	Male	2.70 (2.18)	6.14 (2.48)	4.01 (2.10)	12.85 (3.06)
	Female	3.52 (1.98)	7.21 (3.56)	5.11 (2.61)	*15.84 (2.44)
Erect	Male	2.81 (2.10)	6.33 (2.59)	4.52 (2.20)	13.66 (3.24)
	Female	3.21 (2.14)	7.66 (4.46)	6.08 (3.12)	*16.95 (2.90)

		Contribution to Total EA (%)		
		Hip	Knee	Ankle
Preferred	Male	19.49 (9.98)	50.88 (17.74)	29.63 (14.86)
	Female	19.12 (10.81)	53.22 (19.14)	27.66 (13.61)
Flexed	Male	19.48 (13.27)	49.25 (19.23)	31.27 (16.91)
	Female	22.67 (13.20)	45.25 (19.31)	32.08 (14.08)
Erect	Male	19.34 (12.13)	47.36 (17.93)	33.28 (15.70)
	Female	19.62 (13.89)	44.12 (21.45)	36.26 (16.95)

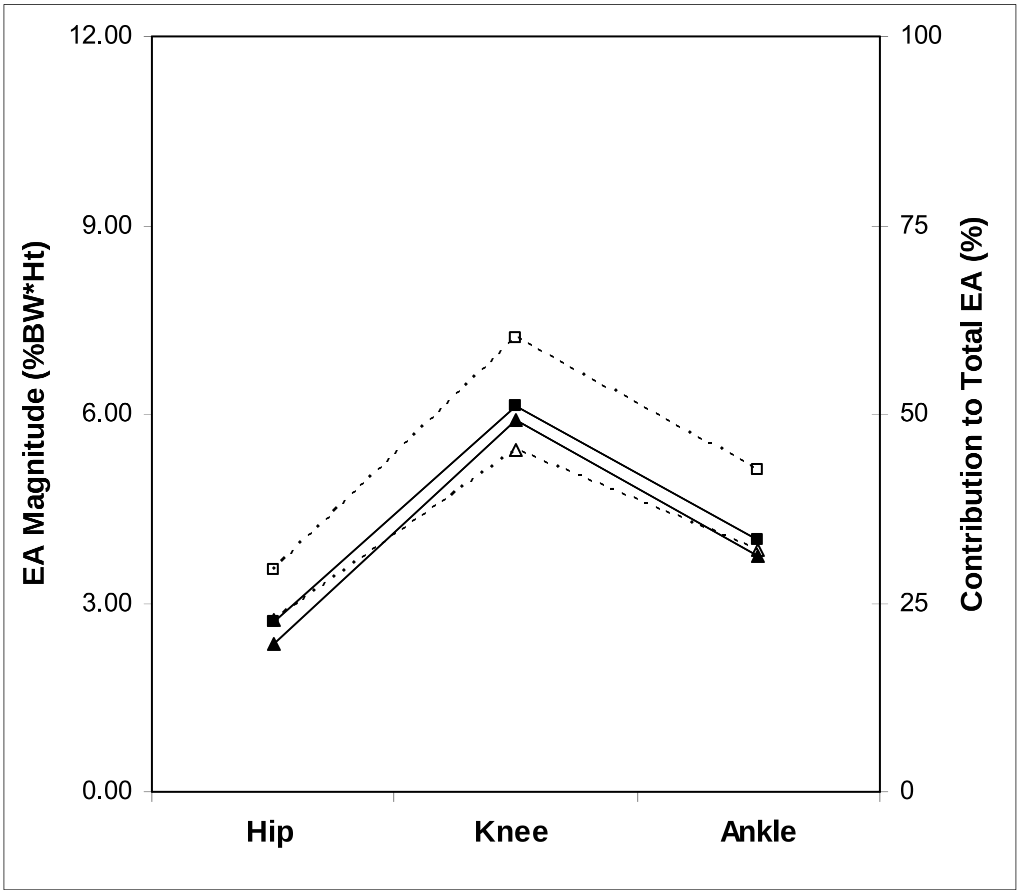
*Significantly greater than males, $P < 0.05$

Figure 1. Joint energy absorption (EA) magnitude (squares) and relative joint contributions to total EA (triangles) for males (solid) and females (dashed) during the Preferred landing condition.



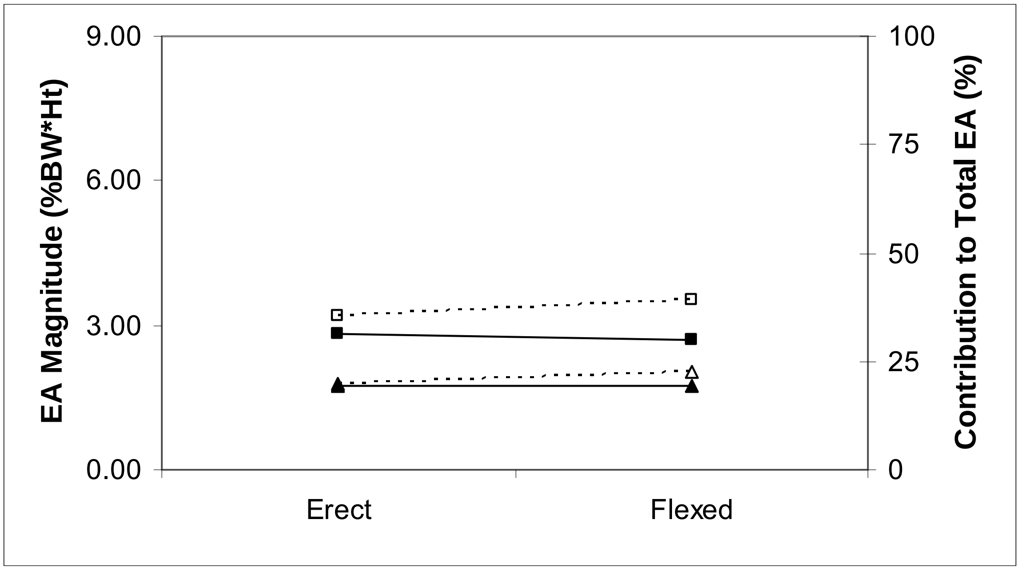
Magnitude: Main effects for sex ($p = 0.003$) and joint ($p < 0.001$) with Knee EA > Ankle EA > Hip EA ($p < 0.05$)
Contribution: Main effect for joint ($p < 0.001$) with Knee > Ankle > Hip ($p < 0.05$)

Figure 2. Joint energy absorption (EA) magnitude (squares) and relative joint contributions to total EA (triangles) for males (solid) and females (dashed) during the Flexed landing condition.



Magnitude: Main effects for sex ($p = 0.001$) and joint ($p < 0.001$) with Knee EA > Ankle EA > Hip EA ($p < 0.05$)
Contribution: Main effect for joint ($p < 0.001$) with Knee > Ankle > Hip ($p < 0.05$)

Figure 3. Hip energy absorption (EA) magnitude (squares) and hip contribution to total EA (triangles) for males (solid) and females (dashed) during the Erect and Flexed landing conditions.



Contribution: Main effect for posture with Flexed > Erect (p = 0.049)

Figure 4. Knee energy absorption (EA) magnitude (squares) and hip contribution to total EA (triangles) for males (solid) and females (dashed) during the Erect and Flexed landing conditions.

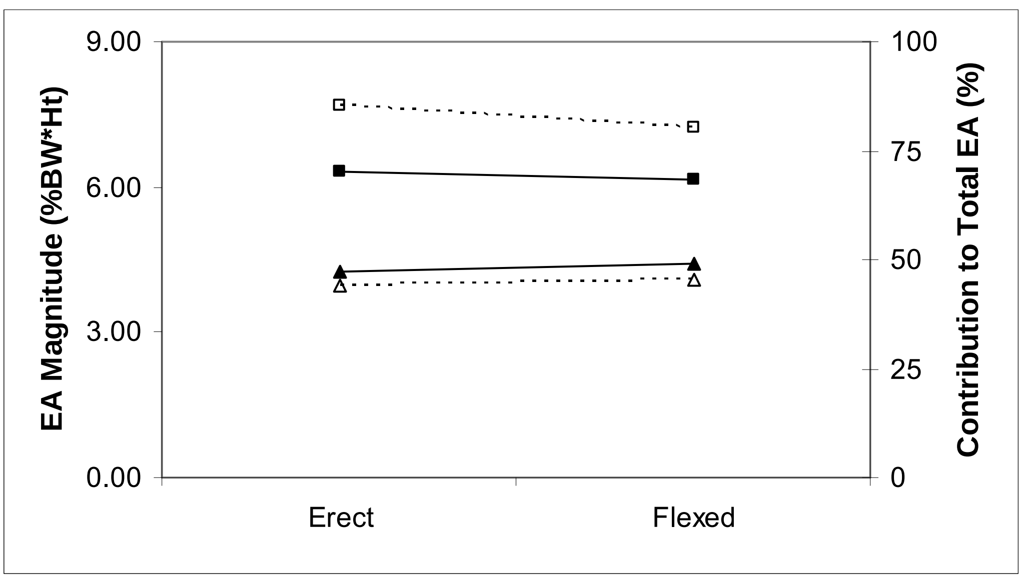
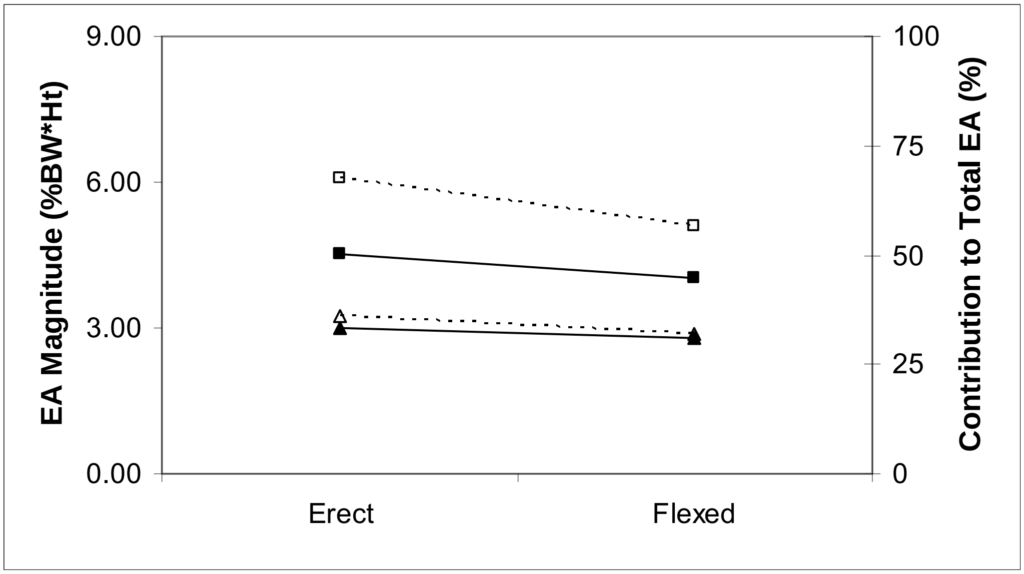
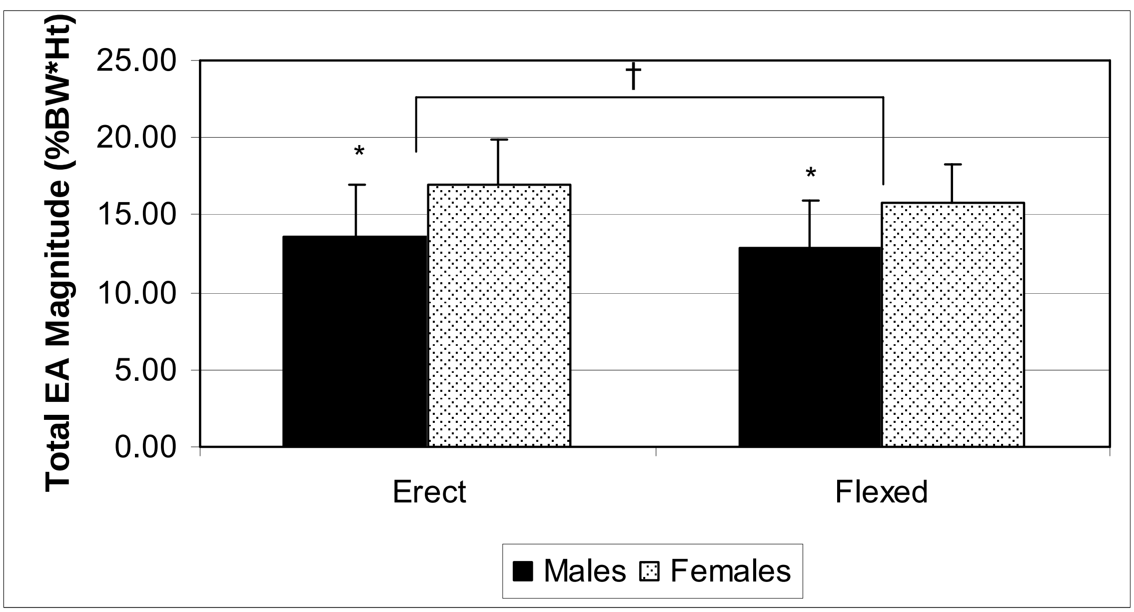


Figure 5. Ankle energy absorption (EA) magnitude (squares) and hip contribution to total EA (triangles) for males (solid) and females (dashed) during the Erect and Flexed landing conditions.



Magnitude: Main effect for posture with Flexed < Erect ($p < 0.001$)
Contribution: Main effect for posture with Flexed < Erect ($p < 0.001$)

Figure 6. Total energy absorption (EA) magnitude during the Preferred, Erect and Flexed landing conditions.



* = significant difference between Males and Females ($p < 0.05$); † = Erect significantly greater than Flexed ($p < 0.05$)

Supplementary Figure. Subject positioning during the preferred (A), flexed (B), and erect (C) drop landing conditions

A.

B.

C.

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APPENDIX FOUR: MANUSCRIPT IV

Manuscript IV

Biomechanical predictors of sagittal plane energy absorption during landing

ABSTRACT

Background: Greater sagittal plane energy absorption (EA) in the 100 ms immediately following ground contact during landing likely results in greater anterior cruciate ligament (ACL) loading and is the result of the combined EA of the hip, knee, and ankle. Further, greater total EA has been associated with landing biomechanics that are purported to increase ACL injury risk. However, it is currently unknown what modifiable biomechanical factors are predictive of lower extremity EA and should be addressed in ACL injury prevention programs.

Purpose: To identify modifiable biomechanical predictors of hip, knee, and ankle EA.

Study Design: Descriptive laboratory study.

Methods: Seventy-seven volunteers (40 males, 37 females; Age = 20.1 ± 2.2 years; Height = 1.74 ± 0.10 m; Mass = 70.3 ± 16.1 kg) had peak strength and muscle activation measured during maximal voluntary isometric hip extension, knee extension, knee flexion, and ankle plantarflexion contractions. Subjects then performed double-leg jump landings during which lower extremity biomechanics were assessed (kinematics and kinetics, and electromyography). Multiple, stepwise regression analyses were used to predict hip, knee, and ankle EA during the 100 ms following ground contact from the measured biomechanical factors.

Results: Greater peak hip flexion, lesser hip flexion at ground contact, and greater peak hip extension strength significantly predicted greater hip EA ($R = 0.545$, $R^2 = 0.297$, Adjusted $R^2 = 0.266$, $P < 0.001$). No biomechanical factors were identified that significantly predicted knee EA ($P > 0.05$). Greater ankle extension (plantarflexion) at ground contact, peak ankle flexion angle, peak ankle extension strength, and mean gastrocnemius EMG significantly predicted greater ankle EA (R = 0.704, R^2 = 0.496, Adjusted R^2 = 0.466, P < 0.001).

Conclusion: Increasing hip flexion, decreasing gastrocnemius activation, and increasing ankle flexion at initial ground contact may cause a decrease in total lower extremity EA, lesser knee joint loading, and potentially reduce ACL injury risk.

Key words: energy absorption, anterior cruciate ligament, landing biomechanics

INTRODUCTION

While precise epidemiological estimates are not known, it has been suggested that as many as 250,000 anterior cruciate ligament (ACL) injuries occur in the United States annually with an estimated health care cost of $1.5 billion.[1] These high costs coupled with the increased risk of long-term sequelae such as knee osteoarthritis [2-4] have been the impetus in the development and implementation of intervention programs aimed at reducing the risk of non-contact ACL injury. However, current ACL-injury prevention programs vary greatly with respect to the included components (i.e. strength, flexibility, etc.), intensity, and duration; as well as their success in both altering high risk landing biomechanics and reducing ACL-injury incidence.[5-10]

There is increasing evidence that peak ACL strain and injury likely occur within the first 100 ms following ground contact during dynamic tasks such as landing.[11-13] Accordingly, mitigating high risk landing biomechanics during this time period that contribute to greater ACL loading may be an effective way to reduce ACL injury risk. Recent work investigating landing biomechanics using energetic analyses in which the kinematic and kinetic parameters of motion are combined[14] has demonstrated a biomechanical profile consistent with greater ACL loading in individuals absorbing a greater magnitude of energy during the initial 100 ms following ground contact.[15] Specifically, greater EA during this time interval is associated with greater peak knee extension moment, anterior tibial shear force, and posterior ground reaction force.[15] This suggests that reducing the magnitude of total sagittal plane EA during this time period could result in lesser knee joint loading. However, it is currently unknown what, if any, modifiable biomechanical factors are predictive of sagittal plane EA magnitude.

Total lower extremity EA is calculated as the sum of EA at the hip, knee, and ankle[16-18] and is indicative of eccentric muscle actions during movement. Further, individual joint EAs are derived by integrating the negative portion of the joint power curves, with joint power calculated as the product of the resultant joint moments and joint angular velocities ($P = M \times \omega$).[14] Therefore, biomechanical factors that affect either joint angular velocities or joint moments likely influence the magnitude of EA at a specific joint and in total during landing. Joint positions at ground contact, which define the available joint ranges of motion,[19] and joint displacements during landing can both affect joint angular velocities.[20, 21] Similarly, muscular strength and activation influence net joint moments. Modifying these factors may reduce total EA, thus improving high risk landing biomechanics. However, the influences of these factors on lower extremity EA have yet to be examined. Therefore, the purpose of this investigation was to identify modifiable biomechanical predictors of the magnitude of lower extremity joint EA during the 100 ms following ground contact using a comprehensive neuromechanical analysis (kinematics, kinetics, and electromyography).

METHODS

Participants

Seventy-seven volunteers (40 males, 37 females; Age = 20.1 ± 2.2 years; Height = 1.74 ± 0.10 m; Mass = 70.3 ± 16.1 kg) participated in this study after reading and signing an Institutional Review Board approved consent form. All subjects were physically active (participating in at least 30 minutes of physical activity three times per week), and generally healthy with no history of ACL injury, neurological disorder, lower extremity surgery, or lower extremity injury within the six months preceding data collection. Following the informed consent process, subject height and mass were recorded and used for biomechanical model generation and normalization of the dependent variables. All testing was performed on the dominant leg defined as the leg that would be used to kick a ball for maximum distance.

Subject Preparation and Experimental Procedures

Surface Electromyography Preparation and MVIC Assessment

Preamplified/active bipolar surface EMG electrodes (DelSys Bagnoli-8, DelSys, Inc., Boston, MA: interelectrode distance = 10 mm; amplification factor = 1,000 – 10,000 (20 –

450 Hz); CMMR @ 60 Hz > 80 dB; input impedance > 10^{15} Ω//0.2pF) were placed parallel to the direction of action potential propagation over the vastus lateralis (VL), vastus medialis (VM), biceps femoris long head (BF), medial hamstrings (MH), gluteus maximus (GMax), lateral gastrocnemius (LG), and medial gastrocnemius (MG) in standardized positions as described below and in accordance with the widely accepted SENIAM project (Surface Electromyography for the Non-Invasive Assessment of Muscles) guidelines.[22] The VL electrode was placed 2/3 of the distance from the anterior superior iliac spine (ASIS) of the pelvis to the lateral side of the patella, while the VM electrode was positioned 80% of the distance from the ASIS to the medial tibiofemoral joint line. The BF and MH electrodes were placed midway between the ischial tuberosity and the lateral and medial condyles of the tibia, respectively. The GMax electrode was positioned 30% of the distance from the second sacral vertebra (S2) to the greater trochanter. Finally, the LG electrode was positioned 1/3 of the distance from the fibular head to the calcaneus, and the MG electrode placed over the most prominent portion of the muscle belly. A reference electrode was also placed over the anteromedial aspect of the tibia distal to the tibial tuberosity. All electrodes and wires were then secured using pre-wrap and athletic tape. To reduce impedance to the EMG signal and allow for proper electrode fixation, electrode sites were prepared by shaving any hair from the immediate vicinity of the desired electrode location before abrading the area and cleansing the skin with isopropyl alcohol. Proper electrode placement and minimal crosstalk were confirmed using an oscilloscope prior to testing.

Following EMG electrode placement, subjects performed three 5 second maximal voluntary isometric contraction (MVIC) trials for hip extension (GMax), knee flexion (BF and MH), knee extension (VL and VM), and ankle extension (plantarflexion) (LG and MG) in standardized testing positions[23] against a handheld dynamometer (Chatillon CSD 300, Amtek, Inc., Largo, FL) (Figure 1). For each trial, peak isometric force was recorded and the EMG signals were sampled using the MotionMonitor motion analysis software (Innovative

Sports Training, Inc., Chicago, IL, USA). The testing order for the MVIC assessments was randomized without replacement and one minute of rest was provided to subjects between trials to minimize the risk of fatigue.

Kinematic and Kinetic Analysis Preparation and Jump Landing Assessment

Lower-extremity and trunk kinematics were assessed using an electromagnetic motion capture system (Motion Star, Ascension Technology Corp., Burlington, VT, USA). Six degree of freedom electromagnetic tracking sensors were positioned over the third metatarsal, anteromedial shank, and lateral thigh, as well as on the sacrum and C7 spinous process. These sensors were placed over areas of minimal muscle mass, and secured with pre-wrap and athletic tape to reduce motion artifact. Global and segment axis systems were established with the positive X axis designated as anteriorly/forward, the positive Y axis medially/leftward, and the positive Z axis superiorly/upward. A segment-linkage model of the dominant lower extremity, pelvis, and thorax was created using the MotionMonitor by digitizing the T12 spinous process and hip, knee, and ankle joint centers. The hip joint center was predicted using digitized landmarks on the pelvis as described by Bell et al.[24] Ankle and knee joint centers were defined as the midpoints of the digitized medial and lateral malleoli and the medial and lateral femoral condyles, respectively.

Subjects performed double leg jump landings by standing atop a 30 cm high box that was set a distance equal to 50% of their height away from the edge of a nonconductive force plate (Type 4060-NC, Bertec Corporation, Columbus, OH, USA) whose axis system was aligned with the global axis system. Subjects were instructed to jump down and forward toward the force plate, contact the ground with both feet at the same time with their dominant foot near the center of the force plate and their non-dominant foot positioned next to the force plate, and then immediately jump up for maximum height using both legs. All subjects performed 3 practice trials and 5 successful testing trials with 30 seconds of rest between trials to minimize the potential effects of fatigue. Trials were deemed successful if subjects jumped from the box and landed with both feet at the same time, completely contacted the force plate with only the dominant foot, and performed the landing task and subsequent maximum jump in a fluid motion.

Data Sampling, Processing, and Reduction

Kinematic and analog (electromyographic and kinetic) data were sampled at 120 Hz and 1,200 Hz, respectively, using The MotionMonitor motion analysis software. Raw kinematic data were low-pass filtered using a fourth-order, zero-phase-lag Butterworth filter with a cutoff frequency of 10 Hz,[25] time synchronized with the analog data, and then re-sampled to 1,200 Hz. Joint angular positions were calculated based on a right hand convention using Euler angles in a Y (flexion/extension), X' (adduction/abduction), Z" (internal/external rotation) rotation sequence with motion defined about the hip as the thigh relative to the pelvis, about the knee as the shank relative to the thigh, and about the ankle as the foot relative to the shank. Instantaneous joint angular velocities were calculated as the 1st derivative of angular position.

Custom computer software (LabVIEW, National Instruments, Austin, TX) was used for processing and reducing the recorded EMG signals. EMG data were corrected for DC bias, bandpass filtered (20-350 Hz, zero-phase-lag, 4th order Butterworth), and smoothed using a root-mean-square (RMS) sliding window function with a time constant of 25 ms. A 100 ms moving average was used to identify maximal EMG amplitude during the middle 3s of the MVIC trials, with the largest 100 ms period representing maximal activation. For jump landing trials, the mean EMG amplitude of the RMS smoothed waveform was calculated during the period from 50 ms before to 50 ms after initial ground contact (VGRF > 10 N). EMG activation amplitudes for each individual muscle during the jump landing trials were expressed as a percentage of the mean of the activation amplitudes measured during the three MVIC trials (%MVIC). The calculated %MVIC values for the BF and MH, VL and VM, and LG and MG during the jump landing trials were then averaged and represented the activation amplitudes of the hamstring, quadriceps, and gastrocnemius muscle groups, respectively.[26]

Force plate data were low-pass filtered at 60 Hz (4^{th} order zero-phase lag Butterworth)[27] and combined with kinematic and anthropometric[28] data to calculate the net internal joint moments of force at the hip, knee, and ankle using an inverse dynamics solution[29] within The MotionMonitor software. Custom computer software (LabVIEW, National Instruments, Austin, TX) was then used to generate sagittal plane hip, knee, and ankle joint power curves by multiplying joint angular velocities and net joint moments for each jump landing trial ($P = M \times \omega$). Negative mechanical joint work was calculated by integrating the negative portion of the joint power curves[16, 17, 20, 25] during the 100 ms immediately following initial ground contact.[25, 30] Total lower extremity EA was calculated by summing the negative joint work at each individual joint.[16-18] All EA values were assigned as positive by convention and expressed as a percentage of the product of subject height and weight (% BW*Ht).[25, 30]

Finally, the same custom software was used to identify sagittal plane hip, knee, and ankle joint angles at IGC, the peak values of these same angles during the time from IGC to the minimum vertical position of the whole body center of mass, and the lengths of the model generated thigh, shank, and foot segments. The peak isometric force recorded during the MVIC trials was then multiplied by segment length to calculate peak torques (strength) which were normalized to the product of subject height and weight ($\times [BW*Ht]^{-1}$). Prior to statistical analysis, all dependent variables were averaged across the five jump landing or three MVIC trials, respectively.

Statistical Analysis

Data were subjected to a screening procedure to identify any potential outliers prior to further statistical analysis. As a result of this procedure, four participants were excluded from the final analysis due to quadriceps and/or hamstrings EMG activation amplitudes that were deemed outliers. Outliers were defined as having a mean value greater than three times the inter-quartile range of values for all subjects. This screening procedure resulted in 73 participants (39 Males, 34 Females) being included in the final analyses.

Three separate, stepwise multiple linear regression models were used to predict the magnitudes of hip, knee, and ankle EA from the measured biomechanical variables (Table 1). The order of entry for the predictor variables into each regression equation was based on the highest individual bivariate correlation with each criterion variable. The probability of F for entry and removal from the models were set at 0.10 and 0.20, respectively, in order to construct the most parsimonious regression models. All analyses were conducted using commercially available software (SPSS 17.0, SPSS Inc., Chicago, IL, USA) with statistical significance established a priori as $\alpha \leq 0.05$.

RESULTS

Descriptive statistics for the predictor and dependent variables are shown in Table 2. The regression analysis for hip EA revealed that the linear combination of greater peak hip flexion (R^2 change = 11.2%, $P = 0.004$), lesser hip flexion at IGC (R^2 change = 12.1%, $P = 0.001$), and greater peak hip extension strength (R^2 change = 6.4%, $P = 0.014$) predicted greater hip EA ($R = 0.545$, $R^2 = 0.297$, Adjusted $R^2 = 0.266$, $P < 0.001$) with mean gluteus maximus EMG not explaining additional variance ($P = 0.388$). Table 3 presents the parameter estimates and standardized coefficients for this final regression model. No biomechanical factors were identified that significantly predict knee EA using a stepwise multiple regression model. As a result, the parameter estimates presented in Table 4 are for a non-significant model in which the variables were forced in using an enter method. Finally, Table 5 indicates the parameter estimates and standardized coefficients for the regression model that significantly predicted ankle EA. The linear combination of greater ankle extension at IGC (R^2 change = 28.0%, $P < 0.001$), greater peak ankle flexion (R^2 change = 10.7%, $P = 0.001$), greater peak ankle extension strength (R^2 change = 6.7%, $P =$

0.005), and greater mean gastrocnemius EMG (R^2 change = 4.2%, P = 0.020) predicted greater ankle EA (R = 0.704, R^2 = 0.496, Adjusted R^2 = 0.466, P < 0.001).

DISCUSSION

The primary findings of this investigation are that the combination of multi-factorial biomechanical factors is predictive of EA at the hip and ankle, but not at the knee. This suggests that interventions aimed at reducing total lower extremity EA and thereby potentially decreasing knee joint loading during landing must facilitate changes across the entire kinetic chain.

The greatest contributor to total EA during the double leg jump landing task was the knee, accounting for 65% of the total energy absorbed during the initial 100 ms following ground contact (Figure 2). Interestingly, however, the knee was the only joint for which EA could not be predicted using kinematic, strength, and muscle activation factors. Recently, Schmitz and Shultz[31] reported a sex-specific relationship between strength and EA during 45 cm drop jump landings with greater knee extensor strength predicting greater knee EA (R^2 = 0.11) in females only. Additionally, females absorbed 69% more energy at the knee compared to males during this investigation.[31] As a result of these findings, we performed a secondary analysis in which we included sex as an additional predictor variable, but were again unable to significantly predict knee EA. We suggest that inherent differences between the two tasks (i.e. predominantly vertical vs. combined vertical and horizontal velocity) resulted in different knee contributions to total EA (20% vs. 65%), and may potentially explain the differing results.

At the hip, we found that greater EA was predicted by greater peak hip flexion, lesser hip flexion at IGC, and greater peak isometric hip extension strength (Table 2). However, we do not suggest that all of these biomechanical factors should be altered in an attempt to reduce the total lower extremity EA during landing and potentially affect knee joint loading. As EA represents eccentric muscle action, it is not surprising that greater strength was related to greater EA. In fact, this strength-EA relationship is in agreement with the previously discussed findings of Schmitz and Shultz,[31] as well as Sandler and Rabinovitch[32] who demonstrated increased energy absorption secondary to increased muscular strength when investigating falls using a biomechanical model. Further, Zhang et al.[18] and Yeow et al.[33] have demonstrated through the manipulation of landing height that greater impact velocities result in a greater magnitude of energy that must be absorbed. As a result, a reduction in strength may limit the ability to meet the demands of common athletic tasks that are much more challenging than the double leg landing utilized in this study. With respect to peak hip flexion, Pollard et al.[34] demonstrated that individuals exhibiting greater combined peak knee and hip flexions during landing absorbed greater energy at the hip and suggested that limiting sagittal plane motion during landing may place individuals at greater risk for ACL injury. Therefore, of the significant biomechanical predictors of hip EA, we suggest that ACL injury interventions should facilitate increases in hip flexion angle at IGC in an effort to reduce total lower extremity EA during landing and potentially reduce knee joint loading.

Similar to the hip, greater ankle EA was predicted by greater ankle extension at IGC, peak ankle flexion, peak isometric ankle extension strength, and mean gastrocnemius EMG (Table 5). The relationship between peak strength and EA identified at the hip and by Schmitz and Shultz[31] is again evident, but at the ankle the addition of greater gastrocnemius activation during the 100 ms centered around IGC increases the magnitude of ankle EA. We propose that this combination of greater peak ankle extension strength coupled with greater gastrocnemius activation likely results in greater apparent musculotendinous stiffness of the ankle extensors. Therefore, greater work must be done on the muscle-tendon unit to push the ankle into flexion during landing, thereby contributing to greater EA. As a result, reducing gastrocnemius activation amplitude at IGC may decrease the magnitude of ankle EA

In addition to strength and activation of the ankle extensors, we also identified greater peak ankle flexion and greater ankle extension angle at IGC as significant predictors of greater ankle EA during the initial 100 ms of landing. The finding related to greater ankle extension at IGC is particularly interesting as more extended ankle and knee positions along with greater ankle EA have been noted previously in females;[25] a population with a greater risk of ACL injury.[35] Additionally, Blackburn and Padua[36] noted lower extremity joint coupling in both males and females whereby flexion or extension displacements at one joint facilitates this same greater angular displacements at the other joints. Therefore, we performed a secondary analysis to determine whether this kinematic coupling across joints of the lower extremity was present with respect to joint positions at IGC. We found that this within limb coordination was present as lesser hip flexion angle at IGC was associated with lesser knee flexion angle at IGC ($r = -0.398$, $P < 0.001$); and greater ankle extension angle at IGC was associated with more extended knee ($r = -0.584$, $P < 0.001$) and hip ($r = 0.349$, $P = 0.002$) positions at IGC. We propose that the more extended lower extremity posture is the result of a “reaching” strategy in which the knee and hip are concomitantly positioned in greater extension as the ankle is extended to “reach” toward the ground during landing. This suggests that a more comprehensive strategy aimed at instructing participants to increase flexion angles at IGC in all three lower extremity joints may be needed to effectively facilitate greater knee flexion angles at IGC. Simply focusing on isolated knee joint position without concern for the ankle may lead to more extended knee and hip positions at ground contact persisting if an individual continues to utilize relatively greater ankle extension at IGC. Further, as an erect posture has been shown to result in greater total EA,[25, 37] this more erect posture might also result in greater ACL loading. However, future research is needed to more closely examine this notion.

Based upon our results, we recommend that increasing hip and ankle flexion angles, and reducing gastrocnemius activation at IGC may reduce the magnitude of lower extremity

EA during the 100 ms after ground contact. We suggest that an intervention aimed at incorporating verbal and visual feedback along with technique instruction may be most effective in altering initial hip and ankle kinematics as previous investigators have successfully increased knee flexion angle at IGC using these techniques.[38] Additionally, we suspect that by increasing ankle flexion there will be an associated increase in knee and hip flexion due to kinematic coupling of the lower extremity. With respect to reducing gastrocnemius activity, we are currently unsure of how to best facilitate this change. Cowling et al.[38] reported that simple instructions to activate the hamstrings earlier during landing was not effective at eliciting a change in hamstring activity and suggested that more specialized training might be needed. However, it is unknown if a more intensive intervention would be necessary to cause a reduction in the activation of the gastrocnemius.

Limitations

As with any study evaluating landing biomechanics, care should be taken when generalizing these findings to tasks other than the double leg jump landing task utilized. Task differences in landing height, type of landing (terminal vs. countermovement), and direction of the center of mass velocity at impact can dramatically influence both the magnitudes and relative joint contributions to total EA. We also acknowledge that as EA is indicative of eccentric muscle action, we are limited in that we assessed peak strength during isometric, rather than eccentric contractions. This choice to test peak strength using isometric contractions against a handheld dynamometer was made as we wanted to include predictors in our model that were not only modifiable, but also more easily measured in a clinical setting. Additionally, while our previous work indicates that greater total EA is associated with a biomechanical profile likely to increase ACL loading,[15] we do not know if greater total EA is directly related to the risk of ACL injury. Future research should evaluate if the proposed recommendations to reduce EA actually do so, and whether those changes result in a more favorable landing biomechanics and decreased ACL injury risk.

Summary

This investigation was the first to predict the magnitude of hip and ankle EA that occurs during the time interval in which ACL rupture likely takes place using biomechanical factors that are modifiable. Based upon our findings, we suggest that increasing hip flexion, decreasing gastrocnemius activation, and increasing ankle flexion at IGC may cause a decrease in total lower extremity EA and reduce knee joint loading. Further, we propose that utilization of greater ankle flexion at IGC may facilitate greater knee and hip flexion angles at impact through a kinematic coupling of the joints of the lower extremity and that these changes might be achieved using verbal feedback and technique instruction as has been done in previous investigations.

Table 1. Specific predictor variables for the multiple stepwise regression analyses

Dependent Variable	Predictor Variables
Hip EA	Hip Position at IGC Peak Hip Flexion Peak Hip Extension Strength Mean Gluteus Maximus EMG
Knee EA	Knee Position at IGC Peak Knee Flexion Peak Knee Extension Strength Peak Knee Flexion Strength Mean Quadriceps EMG Mean Hamstrings EMG
Ankle EA	Ankle Position at IGC Peak Ankle Flexion Peak Ankle Extension Strength Mean Gastrocnemius EMG

Table 2. Descriptive statistics for initial hip, knee, and ankle EA and biomechanical predictor variables

Variable	Mean	SD
Hip EA (%BW*Ht)	2.35	1.35
Knee EA (%BW*Ht)	8.95	2.65
Ankle EA (%BW*Ht)	2.47	1.59
Hip Position at IGC (°)	-34.4	10.3
Peak Hip Flexion (°)	-67.9	16.3
Peak Hip Extension Strength (x$[BW*Ht]^{-1}$)	0.101	0.023
Mean Gluteus Maximus EMG (%MVIC)	37.3	27.9
Knee Position at IGC (°)	23.3	8.47
Peak Knee Flexion (°)	92.3	15
Peak Knee Extension Strength (x$[BW*Ht]^{-1}$)	0.145	0.03
Peak Knee Flexion Strength (x$[BW*Ht]^{-1}$)	0.063	0.014
Mean Quadriceps EMG (%MVIC)	120.6	53.6
Mean Hamstrings EMG (%MVIC)	33.4	18.8
Ankle Position at IGC (°)	36.8	18.1
Peak Ankle Flexion (°)	-15.7	7.4
Peak Ankle Extension Strength (x$[BW*Ht]^{-1}$)	0.087	0.018
Mean Gastrocnemius EMG (%MVIC)	111.4	68.8

Table 3. Regression coefficients from the final model when predicting INI hip EA from the biomechanical variables (Significant coefficient if $P < 0.05$)

Variable	Parameter Estimate	SE	Standardized Coefficient	*t* Value	*P* Value
Hip EA					
Intercept	-0.402	0.881		-0.457	0.649
Peak Hip Flexion	-0.043	0.010	-0.518	-4.478	<0.001
Hip Position at IGC	0.049	0.015	0.377	3.247	0.002
Peak Hip Extension Strength	15.211	6.053	0.254	2.513	0.014

Table 4. Regression coefficients from the final model when predicting INI knee EA from the biomechanical variables (Significant coefficient if $P < 0.05$)

Variable	Parameter Estimate	SE	Standardized Coefficient	*t* Value	*P* Value
Knee EA					
Intercept	7.999	2.661		3.006	0.004
Mean Quadriceps EMG	-0.009	0.007	-0.181	-1.318	0.192
Peak Knee Extension Strength	-16.190	16.670	-0.181	-0.971	0.335
Peak Knee Flexion Strength	11.518	35.885	0.060	0.321	0.749
Mean Hamstrings EMG	0.018	0.019	0.126	0.960	0.341
Knee Position at IGC	0.036	0.042	0.114	0.846	0.401
Peak Knee Flexion	0.024	0.024	0.137	0.999	0.322

Table 5. Regression coefficients from the final model when predicting INI ankle EA from the biomechanical variables (Significant coefficient if $P < 0.05$)

Variable	Parameter Estimate	SE	Standardized Coefficient	*t* Value	*P* Value
Ankle EA					
Intercept	-3.124	0.817		-3.822	<0.001
Ankle Position at IGC	0.049	0.008	0.557	5.804	<0.001
Peak Dorsiflexion	-0.065	0.020	-0.306	-3.317	0.001
Peak Ankle Extension Strength	25.418	7.677	0.291	3.311	0.001
Mean Gastrocnemius EMG	0.005	0.002	0.216	2.385	0.02

Figure 1. Maximal voluntary isometric contraction testing positions

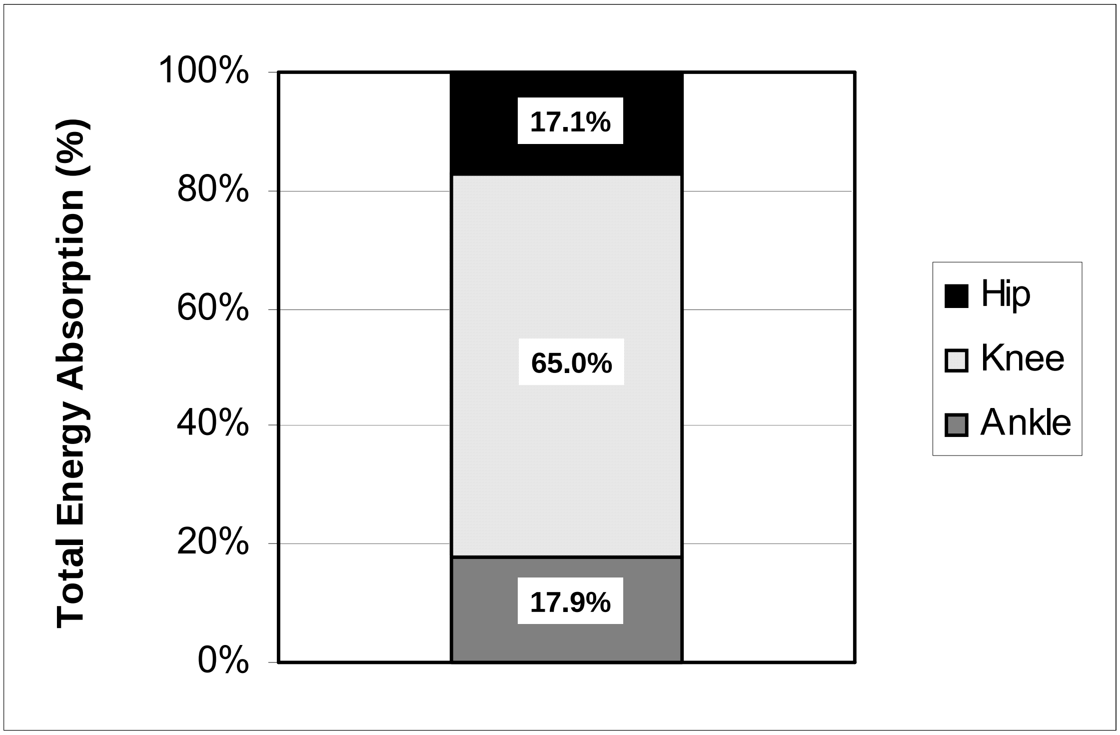
A. Hip Extension

B. Knee Extension

C. Knee Flexion

D. Ankle Extension (Plantarflexion)

Figure 2. Mean relative joint contributions to total energy absorption during the 100 ms immediately following ground contact.



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