THE RELATIONSHIP BETWEEN DYNAMIC TRUNK STABILITY AND LOWER EXTREMITY AND TRUNK BIOMECHANICS DURING A CUTTING TASK

Barnett S Frank

A thesis submitted to the faculty of the University of North Carolina at Chapel Hill in partial fulfillment for the requirements of the degree of Master of Arts in the Department of Exercise and Sport Science (Athletic Training).

Chapel Hill
2010

Approved by:

Advisor: Dr. Darin Padua

Reader: Dr. Troy Blackburn

Reader: Mr. David Bell

Reader: Mr. Marc Norcross

ABSTRACT

The Correlation Between Dynamic Trunk Stability and Lower Extremity and Trunk Biomechanics During a Cutting Task

(Under the direction of: Dr. Darin A Padua)

Purpose: Examine the relationship between trunk neuromuscular control (NMC) during unstable-sitting (UST) and trunk and knee biomechanics during an athletic cutting task (ACT). **Design:** Descriptive Laboratory Study. **Methods:** Participants performed UST and ACT. Center of pressure (CoP) sway data were collected during UST. Peak triplanar trunk and knee biomechanics were collected during ACT. Pearson product-moment correlation coefficients were calculated among CoP data, and trunk and knee biomechanics. **Results:** Greater CoP sway area ($r = -0.548$, $p = 0.006$) and path ($r = -0.407$, $p = 0.032$) measures were associated with greater knee extension moment magnitudes. Transverse plane trunk angle was positively associated with both knee varus moment ($r = 0.371$, $p = 0.043$), and transverse plane trunk angle ($r = 0.587$, $p = 0.001$). **Conclusion:** There is evidence of a relationship between trunk NMC and ACL injury risk factors.

Key Words: ACL Injury, Trunk Stability, Core, ACL Injury Risk Factors

ACKNOWLEDGEMENTS

I would like to thank Dr. Darin Padua, Dr. Troy Blackburn, Dr. David R. Bell, Mr. Marc Norcross, and Benjamin Goerger, without whom this project would not have been possible. Thanks to their guidance, time, and dedication, this project became a reality. I would like to thank my peers in the master's of arts of athletic training program at UNC for their fun loving spirits over the past two years. I am grateful to have been able to work with such talented and understanding clinical supervisors, educators, and true friends in Mr. Alain Aguilar, Ms. Nicole Fava, and Ms. Carrie Rubertino-Shearer. I am specifically grateful to Mr. Alain Aguilar who has shown me the importance of tolerance, perseverance, and true devotion to progressing a profession.

I would also like to thank my undergraduate professors, and clinical educators, specifically Walter "Kip" Smith, who has taught me to be a true professional, and achieve excellence in all that I undertake. Thank you to Drs. Katie Grove and Joanne Klossner who were my navigators through my undergraduate career, and to Dr. John Schrader who opened my eyes to importance of "the big picture."

I am forever in-debted to Mr. Ron Hutchins who introduced me to the world of sports medicine. Thank you to Ron, who devoted much of his talent and resources to fostering a learning environment that brought to light a foundation to my current education and understanding.

Thank you to my parents for always believing in me, motivating me to succeed, and providing me with the resources to achieve all that I am capable of. You have been role models, friends, and mentors throughout my life and I thank you for all your time and patience throughout the years.

TABLE OF CONTENTS

LIST OF TABLES

LISTS OF FIGURES

LIST OF ABBREVIATIONS

- ACL - Anterior cruciate ligament
- ATSF - Anterior tibial shear force
- LCL - Lateral collateral ligament
- LE - Lower extremity
- LPHC - Lumbopelvic hip complex
- MCL - Medial collateral ligament
- NMC - Neuromuscular control
- PCL - Posterior cruciate ligament

CHAPTER I

INTRODUCTION

Anterior cruciate ligament (ACL) injury has become an increasingly prevalent and devastating injury for athletes. Seventy percent of all ACL injuries in the United States are non-contact in nature.[21, 22, 27, 67] There are destructive physical and psychological effects the ACL injured athlete experiences. ACL injury accounts for some of the greatest time lost in collegiate sports, a destructive psychological effect,[46] and the expenditure of financial resources that has been proposed to approach $2 billion nationally for surgical intervention, rehabilitation and prevention programs.[19]

In-vitro cadaveric research has indicated an increase in stress is experienced across the ACL during simulated jump landing and weight bearing conditions when the knee is in a less flexed position with concurrent external valgus moment application to the knee.[39, 64] Greater external knee valgus moment and knee valgus and tibial internal rotation angles, and lesser knee and hip flexion angles during landing have been theorized to place individuals at greater risk of non-contact ACL injury.[8, 27] Females exhibiting larger knee valgus angles and moments, and vertical ground reaction force during a jump-landing task demonstrate a greater prospective ACL injury risk than females presenting with lesser kinematic and kinetic values respectively.[27] Neuromuscular and biomechanical risk factors associated with non-contact ACL injury are of primary interest because they are considered modifiable.[21, 22, 27, 33, 49, 50, 57, 67]

The hip and knee joints of the lower extremity (LE) rely on the musculature originating from the lumbopelvic hip complex (LPHC). The rectus femoris of the quadriceps muscle group, and the hamstring muscles provide dynamic stability and control at the knee joint during athletic motions, such as jump-landing, and cutting tasks.[68, 71] Previous research has indicated differences in neuromuscular control (NMC) of the hip and knee musculature between males and females during athletic cutting tasks, with females displaying greater quadriceps activity and less hamstring activity compared to males, representing a difference in thigh muscle activity during cutting tasks.[35, 71] Specifically, the observed difference was observed during the initial phase of ground contact during an athletic task, the period when a majority of non-contact ACL injuries occur.[10, 22] The quadriceps musculature has been reported to exert an anterior tibial shear force (ATSF) across the knee joint when the knee articulation is in a more extended position, as it is during the initial phase of ground contact. Conversely, the hamstring musculature provides dynamic resistance to anterior translation of the tibia on the femur, supplementing the action of the ACL.[17, 65, 71] Force generation by the hamstring musculature during jump-landing tasks decreases strain on the ACL, thus decreasing the risk of injury.[65]

Goldfuss et al[17] reported a positive correlation between electromyographic activity of the quadriceps and hamstring musculature, and the knee joint's resistance to valgus loading, indicating a greater quadriceps/hamstring co-activation ratio increases the knee joint's ability to resist external valgus torque. An increase in resistance to external valgus torque decreases the amount of knee valgus posturing at the joint during weight bearing activity, possibly protecting the knee from valgus collapse, a risk factor of ACL injury.[17] Dynamic muscular restraint and force reduction across the knee joint may help prevent ACL injury during athletic tasks.[34, 36] In order for muscles to most effectively and efficiently provide a dynamically stable environment about the knee joint they must be at their optimal length, this association is defined as the length-tension relationship.[3] The foundation of this study proposes an unstable LPHC serving as the proximal attachment site for many of the dynamic muscular stabilizers of the knee, may alter the optimal and functional length-tension relationship of the muscles, not providing the necessary restraint against non-physiological, and potentially injurious accessory motion at the knee.

The LPHC is an anatomical region synonymous with the core or the trunk, and is the region of the body in which the center of mass (CoM) is located .[32] Adequate control of the region of the body's CoM by the trunk musculature is necessary for the production, transfer, and control of force and motion to distal segments of the kinetic chain.[32, 69, 70]

Deficits in trunk NMC have been shown to predict knee injury in Division I collegiate female athletes.[69] Trunk NMC was evaluated as a measure of trunk displacement after an unanticipated sudden release of a force. After the force release, trunk displacement was measured. Subjects attempted to maintain a static posture by resisting trunk sagittal plane flexion/extension and frontal plane lateral trunk flexion motion with isolated control from the trunk musculature. Subjects were required to sit in a device, immobilizing their lower extremities and pelvis, limiting the lower extremity's muscular influence of force exertion on the pelvis. The pelvis acted as a three-dimensional axis of rotation for the dynamic movement arm of the trunk and head. Zazulak et al[69] hypothesized the musculature of the trunk that was isolated during the task would act to resist motion of the upper body. Trunk muscle isolation was made possible through lower extremity immobilization in the apparatus. The cohort of collegiate female athletes was followed during a three-year period of active sport participation at the Division I collegiate level.[69] A positive relationship was observed between frontal plane lateral trunk motion and risk of knee ligament injury. Individuals exhibiting greater lateral trunk flexion during the sudden force release task injured their knee ligaments at a higher rate than those with a lesser degree of lateral trunk flexion. Greater trunk motion may result in greater motion or instability of the pelvic girdle, the origin of LE musculature that acts to dynamically stabilize the knee during athletic tasks. Zazulak et al's[69] findings indicate decreased NMC of the trunk musculature directly correlates with non-contact ACL injury, a LE pathology. Zazulak et al's[70] results validate the need for further evaluation of the relationship between NMC of the trunk dynamic stabilizers and lower extremity biomechanics during athletic tasks.

Cholewicki[12] examined the dynamic stability element of trunk NMC as subjects sat atop a rigid hemisphere on a force plate. The force plate measured center of pressure (CoP) location while subjects attempted to sit as still as possible for 20 seconds. Cholewicki[12] defined a large CoP travel to be representative of poor trunk stability. The subjects were unable to exert any external moments on their trunk using their extremities, and relied solely on trunk musculature to maintain stable sitting.

To date, no study has examined a measure of the stability element of trunk NMC and its relationship to proposed biomechanical risk factors associated with non-contact ACL injury. Past studies have focused on measures of trunk stability that may not directly correlate to CoM position, and were solely concerned with the position of the entire trunk[29, 68-70] No known methodology has incorporated a biomechanical analysis of a rapid change in direction, as in an athletic cutting task, an identified mechanism of non-contact ACL injury[10] and dynamic trunk stability measures. Therefore, the purpose of this study was to examine the relationship between the dynamic stability element of trunk NMC and LE biomechanical risk factors of ACL injury during a cutting task. The secondary purpose of the study was to examine the relationship between the dynamics stability element of trunk NMC through measures of CoP sway during unstable-sitting and trunk motion during a side-cutting task. We hypothesized decreased stability would be associated with greater trunk and knee motion in the frontal and transverse planes, and less motion in the sagittal plane, as well as smaller internal knee moments in the frontal and sagittal planes during the cutting task.

Independent Variables (predictor):

1. 95% CoP sway path area (square centimeters)
2. CoP Sway Path (centimeters)
3. CoP Sway Velocity (centimeters/second)
4. Peak frontal plane trunk angle during the first 50% of stance time of a cutting task
5. Peak sagittal plane trunk angle during the first 50% of stance time of a cutting task
6. Peak transverse plane trunk angle during the first 50% of stance time of a cutting task

Dependent Variables (criterion):

1. Peak frontal plane trunk angle during the first 50% of stance time of a cutting task

2. Peak sagittal plane trunk angle during the first 50% of stance time of a cutting task
3. Peak transverse plane trunk angle during the first 50% of stance time or a cutting task
4. Peak knee flexion angle during the first 50% of stance time of a cutting task
5. Peak internal knee extension moment during the first 50% of stance time of a cutting task
6. Peak knee valgus angle during the first 50% of stance time of a cutting task
7. Peak internal knee varus moment during the first 50% of stance time of a cutting task
8. Peak tibial internal rotation angle during the first 50% of the stance time of a cutting task

Research Questions:

1. Is there a significant relationship between CoP sway measures during unstable-sitting and peak triplanar trunk kinematics during a side-step cutting task?
2. Is there a significant relationship between CoP sway measures during unstable-sitting and peak triplanar knee kinematics during a side-step cutting task?
3. Is there a significant relationship between CoP sway measures during unstable-sitting and peak internal knee varus and extension moments during a side-step cutting task?
4. Is there a significant relationship between peak triplanar trunk kinematics and peak triplanar knee kinematics during a side-step cutting task?
5. Is there a significant relationship between peak triplanar trunk kinematics and peak internal knee varus and extension moments during a side-step cutting task?

Research Hypotheses:

1. We hypothesized a positive relationship between the measures of CoP sway and peak trunk kinematics in the frontal and transverse planes, and a negative relationship in the sagittal plane during the first 50% of the stance time during the side-step cutting task.
2. We hypothesized a positive relationship between the measures of CoP sway and peak knee kinematics in the frontal and transverse planes, and a negative relationship in the sagittal plane during the first 50% of the stance time during the side-step cutting task.
3. We hypothesized a positive relationship between the measures of CoP sway and peak internal knee and extension moments during the first 50% of the stance time of the side-step cutting task.
4. We hypothesized a positive relationship between peak triplanar trunk and knee kinematics during the first 50% of the stance time of the cutting task in the frontal and transverse planes, and a negative relationship in the sagittal plane.
5. We hypothesized a positive relationship between peak trunk kinematics in the frontal and transverse planes and a negative relationship in the sagittal plane and peak internal knee varus and extension moments during the first 50% of the stance time during the side-step cutting task.

Operational Definitions:

1. ACL Injury – Non- contact; forces applied to the knee at the time of injury resulting from the athlete's own movements and did not involve contact with another athlete or object, resulting in ACL tissue strain and macrofailure.
2. Cross-over Cut – Laterally cutting in the same direction as the subject's plant leg, as the non-planted leg anteriorly traverses the plant leg.
3. Cutting Maneuver – Locomotor task in which a subject is traveling anteriorly, primarily in the sagittal plane and changes forward locomotion into a bi-planar motion, including linear displacement in the frontal plane. The subject will use their dominant leg for both left and right directions. The subject will be required to perform a cross-over cut and a side-step cut.
4. Dominant Leg – Leg subjects defined they would use to kick a ball for maximum distance.
5. Dynamic Trunk Stability – Ability of the LPHC musculature to maintain a stable base of support, enabling the extremities to generate force extrinsically while the body is in a state of motion.
6. Dynamic Trunk Stability Task: Laboratory measure of CoP 95% elliptical sway area, sway path, and sway velocity over a force plate in which the subject will be unable to exert any external moments on the body via the extremities.
7. First 50% of Stance Time - Defined temporally as the first half of data points produced from initial ground contact during the cutting task to toe off.

8. Peak Internal Knee Extension Moment – Maximum muscular torque created about the knee joint in the sagittal plane causing the tibia to extend relative to the femur during the first 50% of the stance time of the side-step cutting task.

9. Peak Internal Knee Varus Moment – Maximum muscular torque created about the knee joint causing the tibia to adduct relative to the femur in the frontal plane during the first 50% of the stance time of the side-step cutting task.

10. Peak Internal Tibial Rotation Angle - Peak transverse plane knee angle defined by the vertex of the femur and tibia via the segment reference system in the transverse plane with an infero-superior longitudinal axis during the first 50% of the stance time of the side-step cutting task.

11. Peak Knee Flexion Angle – Peak sagittal plane knee angle defined by the vertex of the femur and tibia via the segment reference system in the sagittal plane with a mediolateral axis during the first 50% of the stance time of the side-step cutting task..

12. Peak Knee Valgus Angle – Peak frontal plane knee angle defined by the vertex of the femur and tibia via the segment reference system in the frontal plane with an anteroposterior axis during the first 50% of the stance time of the side-step cutting task..

13. Peak Trunk Flexion Angle – Peak sagittal plane angle of the trunk relative to the world-axis system about the Y-axis during the first 50% of the stance time of the side-step cutting task..

14. Peak Trunk Lateral Flexion Angle – Peak frontal plane angle of the trunk relative to the world-axis system about the X-axis during the first 50% of the stance time of the side-step cutting task.
15. Peak Trunk Rotation Angle - Peak transverse plane angle of the trunk relative to the world-axis system about the Z-axis during the first 50% of the stance time of the side-step cutting task.
16. Side-Step Cut – Laterally cutting in the opposite direction of the plant leg.
17. Vertical Ground Reaction Force – The purely vertical component of ground reaction force that exists parallel to the longitudinal axis of the world-axis system.

Delimitations

1. Subjects will include a wide array of individuals from the university population defined in the age range of 18 – 35 years old and physically active.
2. Exclusion criteria for subjects will limit subjects to have no previous history of ACL, LE, or lower back injury within the past 6 months.

Limitations

1. The researcher is limited in knowledge of the subjects' previous experience in balancing tasks.
2. The cutting task is limited to the laboratory setting and the Vicon system's capture volume.

3. The subjects' foot biomechanics may affect the outcome of the experiment, as pes-cavus/planus measurements will not be examined.

Assumptions

1. All measurements are reliable and valid.
2. The CoP's elliptical sway area, sway path, and sway velocity are valid measures of the stability element of trunk NMC.
3. All subjects will be capable of completing the cutting task.
4. Lateral cutting movements are a mechanism of ACL injury as a result of poor LE biomechanics in individuals.
5. Motion artifact is of no significance to reflective marker placement on subjects.

CHAPTER II

REVIEW OF THE LITERATURE

Introduction

ACL injury has become an increasingly prevalent devastating injury for the active population. The literature reveals a growing rate of ligament damage has occurred as a result from non-contact mechanisms.[21, 22] Some of the most common activities resulting in ACL injury include jumping, landing, lateral accelerations and decelerations, and sudden starting and stopping motions that are common in almost all athletic endeavors.[67, 70] Seventy percent of female ACL injuries are the result of a non-contact mechanism.[21, 22, 27, 67] The ACL injured athlete experiences a multitude of negative physical and psychological consequences.[46] ACL injury accounts for some of the greatest time lost in collegiate sports, with an average of six months for return to sport specific activity.[28] In 1999, Gottlob et al[19] proposed the expenditure of financial resources on surgical intervention, rehabilitation, and prevention programs to approach $2 billion nationally, with an average cost of nearly $12,000 per incidence. This literature review focuses on the functional anatomy of the knee joint, ACL injury epidemiology, the four risk factor categories of non-contact ACL injury: (1) anatomical, (2) environmental,(3) hormonal, and (4) biomechanics and the incorporation of trunk NMC and stability as a foundation for movement of the extremities. Specifically, this review will focus on element of biomechanics and neuromuscular control because these factors have been proposed to be modifiable,[22, 50, 68, 69] potentially preventing ACL injuries in the future. This review will concentrate on the relationship between trunk stability and the biomechanics of non-contact ACL injury.

Functional Knee Joint Anatomy

Tibiofemoral Articulation

The tibiofemoral joint (knee) is an articulation between the proximal tibia and the distal femur.[45] The distal articulating surface of the tibiofemoral joint is composed of the concave medial and lateral tibial plateaus, separated by the peaked inter-condyloid eminence,[30] serving anteriorly as an anterior cruciate ligament attachment site, and posteriorly as a posterior cruciate ligament attachment site.[30] The proximal articulating surface of the tibiofemoral joint is composed of the convex medial and lateral femoral condyles, separated by the concave intra-condylar fossa from which the two cruciate ligaments protrude.[45] The convex femoral condyles articulate with the shallow concave medial and lateral tibial plateaus. The articulating surface is deepened by the medial and lateral menisci of the knee, two fibrocartilagenous discs that function to provide force attenuation and increased surface area of the tibial plateaus as they contact the femoral condyles.[45] The bony anatomy of the knee joint is inherently unstable without adequate muscular (dynamic) and ligamentous (static) support[30]

Ligaments

The lateral collateral (LCL) and medial collateral (MCL) are the two collateral ligaments of the knee joint. The extra-capsular cord-like LCL extends proximally from the lateral femoral epicondyle posterior and superior to the insertion of the popliteus muscle, inserting distally on the lateral aspect of the fibular head.[45] The LCL is the primary static restraint against varus loading.[45] The MCL is divided into two portions; superficial and deep.[45] The deep portion arises from the medial femoral epicondyle, attaches into the joint capsule sending a slip into the medial meniscus. The extra-capsular, superficial portion of the MCL arises from the medial femoral epicondyle, inserting distally into medial tibial epicondyle, roughly 4-5 cm inferior to the joint line.[45] The superficial portion of the ligament is the primary medial stabilizer of the knee joint, resisting valgus loads.[45]

The two intra-articular ligaments of the knee joint are the anterior cruciate ligament (ACL) and posterior cruciate ligament (PCL). The ACL arises proximally from the posteromedial aspect of the lateral femoral condyle, within the intra-condylar fossa of the femur and inserts distally at the posterior aspect of the anterior inter-condyloid eminence of the tibia.[30, 45] The ACL provides static resistance against anterior tibial translation on the femur.[30] The PCL arises proximally from the anterior inferior lateral aspect of the medial femoral condyle and inserts distally at the posterior aspect of the inter-condyloid eminence on the tibia.[30] The PCL is the primary static restraint against posterior translation of the tibia on the femur. The PCL passes medially to the ACL as they cross in the tibiofemoral joint space.[30, 45] Together the ACL and PCL resist hyperextension, rotary instability, and varus/valgus loads across the knee joint.[30]

Musculature

The knee joint is dynamically supported in the sagittal, frontal, and transverse planes. The sagittal plane is supported against anterior translation by the hamstrings, arcuate complex, and iliotibial band.[30] The articulation is supported against posterior translation in the sagittal plane by the quadriceps and arcuate complex.[30] The frontal plane is stabilized medially by the pes anserine and vastus medialis oblique musculature, and laterally by the iliotibial band, biceps femoris and lateral gastrocnemius of the arcuate complex.[30] Transverse plane dynamic stabilization is contingent about four directions; anteromedial, anterolateral, posteromedial, and posterolateral.[30] Anteromedial stabilization is reinforced by the pes anserine and medial hamstring musculature.[30] Anterolateral stabilization is accomplished by the popliteus, lateral gastrocnemius, biceps femoris musculature of the arcuate complex, as well as the iliotibial band.[30] The medial hamstrings and vastus medialis oblique muscles provide posteromedial stabilization. The popliteus, lateral gastrocnemius, and biceps femoris muscles sustain posterolateral stabilization of the knee.[30]

Arthrokinematics

Traditionally the tibiofemoral joint is described as a modified hinge joint, with two degrees of freedom, flexion and extension about the mediolateral axis of the sagittal plane, and axial rotation about the longitudinal axis of the transverse plane.[44] Flexion is allowed 120°-160° depending on the position of the hip, as increased hip extension limits knee flexion.[44] Knee extension range varies from 0°-15° hyperextension.[44] The magnitude of axial rotation is dependent upon the sagittal plane position of the knee. In full extension the knee is in a close-packed position, limiting the value of axial rotation to almost 0°.[44] At 90° knee flexion the tibia is allowed roughly 40° lateral rotation and 30° medial rotation.[44].

Current developments in the literature suggest up to 6 degrees of freedom of motion within the knee joint; flexion and extension with mediolateral translation about a mediolateral axis, varus and valgus angulation with anteroposterior translation about an anteroposterior axis, and internal and external rotation with superoinferior translation about a longitudinal axis.[18, 44] The current description of tibiofemoral arthrokinematics lends to the foundation of triplanar movement of the human body.

To account for the current description of the six degrees of freedom of joint motion, a thorough understanding of accessory joint motion must be understood. During knee motion in the sagittal plane, a combination of roll, glide, and spin occurs at the articulating surfaces.[44] As the knee moves into extension in a closed kinetic chain, the femur glides posteriorly, rolls anteriorly, and subsequently medially rotates during the terminal 30° of extension on the tibial plateaus. The medial femoral rotation is described as the screw home mechanism, locking the knee in extension. During knee extension in an open kinetic chain, the tibia glides anteriorly, rolls anteriorly on the femoral condyles, and subsequently rotates laterally during the terminal 30° of extension. During closed kinetic chain flexion, the medial and lateral femoral condyles glide anteriorly, roll posteriorly, and externally rotate on the medial and lateral tibial plateaus during the first 30° of motion to unlock the knee joint. During open kinetic chain knee flexion, the tibial plateaus glide and roll posteriorly, and internally rotate on the femur during the first 30° of motion to unlock the knee joint.[44]

The accessory joint motion is produced by what Muller[47] describes as a four-bar linkage system. The four bars of the linkage system are formed by the (1) ACL, (2) PCL, (3) anteroposterior line connecting the femoral attachments of the ACL and PCL, (4) anteroposterior line connecting the tibial attachments of the ACL and PCL. In a physiological knee, the cruciate ligaments are elastic, and retain a constant length during sagittal plane knee motion. The linkage system controls conventional accessory motion about the knee joint specific to roll and glide. During closed chain extension of the knee, the forward rolling femoral condyles place stress on the PCL, in turn pulling the femur posteriorly. During closed chain flexion of the knee, the femoral condyles roll posteriorly, increasing stress on the ACL, in turn pulling the femur anteriorly. Injury to the cruciate ligaments significantly disrupts the physiologic action of the four bar linkage system, altering knee joint arthrokinematics. The altered joint mechanics predispose individuals for joint pathologies described by disrupted articular tissue, such as meniscal tears, cartilaginous degeneration, and the potential for inflammatory conditions such as osteoarthritis.[22, 26, 41, 44]

ACL Injury Definition

ACL injury is defined as a mechanical failure of the of the fibrous connective tissue viscoelastic ligature of the knee. The ACL is a viscoelastic structure. Linear strain placed on the ligament is resisted by the ligament itself, as an increase in strain force is directly proportional to the ligament's internal resistance force against the external strain. Ligament injury can be described by a graphical model comparing the effects of external tissue load (stress) on tissue deformation (strain). During the physiological resting state of the knee, a normal tension is held in the ACL as described by the cross-bar linkage system.[47, 48] This resting load does not cause any deformation of ligamentous tissue. As load increases past a physiological value, tissue deformation ensues, linearly described as an elastic change in tissue length. As the load increases, the elastic deformation increases (length increase) proportionately to a point. The critical value point is the load value that deforms the tissue past its elastic range. As the tissue is deformed past its elastic range, it can no longer return to its resting length, and is plastically extended. Past the critical load point, the fibrous connective tissue fails, resulting in micro-tears within the connective tissue substance of the ligament. As the load continues to increase, macrofailure of the tissue occurs, and macroscopic tissue deformation results.

Tissue rupture can present as full-thickness or partial thickness.[58] In a full thickness tear of the tissue, complete integrity of the ACL structure is lost, and presents radiographically and arthroscopically as a disrupted structure.[58] A full thickness tear of the ACL results in gross instability of the knee joint. In a partial-thickness tear, the ligament is deformed but shares a longitudinal link between the two ends.[58] The partial-thickness pathology is a result of plastic tissue deformation and irreversible tissue damage similar to the full-thickness tear, and presents with instability as the tissue has been deformed past its physiological state and is lax within the joint. The approximation between the two ends may still provide restraint against clinically applied loads, but translation of the tibia relative to the femur may be evident.

Clinically, a physical examination of the knee joint after suspected ACL injury does not provide adequate information regarding extent of injury.[58] Extent of ligament damage can be estimated with magnetic resonance imaging (MRI) technique and confirmed with direct observation via arthroscopy.[58]

Non-contact ACL Injury Definition

Marshall[22] et al defines the incidence of a non-contact ACL injury as "forces applied to the knee at the time of injury resulting from the athlete's own movements and did not involve contact with another athlete or object." The focus of the remainder of this review will be on the non-contact type of ACL injury.

Epidemiology

ACL injury incidence has been reported as high as 200,000 cases annually[4] within the united states, with 70% of these injuries non-contact in nature.[22] This results in approximately 140,000 non-contact ACL injuries annually. Individuals sustaining non-contact injuries who opt for surgical repair of the ruptured ligament face a financial burden of roughly $12,000, totaling $2 billion nationally.[19]

The literature has suggested ACL injury has long-standing physical effects, including an increased risk of joint degeneration, such as osteoarthritis, which may lead to decreased levels of physical activity later in life. Individuals experiencing levels of joint degeneration that interfere with quality of life may require knee joint replacement. Louboutin et al[41] followed a cohort of ACL injured elite athletes for 35 years, 42% of the subjects required total joint replacement. At 20 years follow-up the reported risk of developing symptomatic osteoarthritis was decreased in the surgical group of the study, indicating the need for surgical intervention to improve long-term quality of life.[16]

Griffin et al[22] established four risk factor categories specific to non-contact ACL injury including, (1) Environmental, (2) Anatomical, (3) Hormonal, and (4) Biomechanical and Neuromuscular Control.

Environmental factors include prophylactic knee bracing and the shoe-surface interface. The literature is inconclusive on the role of prophylactic knee bracing and its effects on ACL injury incidence.[22] The primary setting for knee brace research is in the realm of contact sports and does not have a great deal of direct application to the prevention of non-contact ACL injury. Najibi et al[51] proposed a decreased injury risk in contact sports, and a potential increased injury risk in non-contact sports. The primary benefit of a prophylactic knee brace may be increased proprioception and kinesthesia that is provided by the skin-brace sensory interface, but research in this area is inconclusive and limited.[21] The shoe-surface interface is of concern to the non-contact injury mechanism. An increased coefficient of friction between the shoe and the playing surface increases the risk of non-contact ACL injury. Long studded cleats and a dry playing surface increase the coefficient of friction between the shoe and playing surface. A firmly planted LE has less degrees of freedom then a freely moving LE, translating to greater motion at the LE joints which may be required to attenuate additional forces that a freely moving distal extremity may be able to manage when its motion is less restricted.

Anatomical

Individuals present a great degree of anatomical variability. To date no literature has found any single static anatomical factor that places an individual at a greater risk of injury.[22] Specific lower extremity factors that have been suggested to increase an individual's risk of ACL injury are increased femoral anteversion, increased quadriceps angle (Q angle), excessive tibial torsion, increased foot pronation, and decreased femoral notch size.[21, 30, 59] The proposed anatomical risk factors are found more commonly in females, but no conclusive evidence has shown these anatomical differences to contribute to the increased incidence in ACL injury in females compared to their male counterparts.[21, 22]

The two most researched factors have been an increased quadriceps angle and decreased femoral notch size. Quadriceps angle in females has been shown to be significantly greater than in their male counterparts.[25] Shambaugh et al[59] studied 45 recreational athletes finding a larger quadriceps angle correlated to an increased risk of knee injury. However, other studies have shown no conclusive evidence that quadriceps angle increases an individual's risk of ACL injury.[20] Quadriceps angle likely changes during dynamic activity specific to athletic tasks,[21] thus it is difficult to conclude an increased static quadriceps angle predisposes an individual to non-contact ACL injury.

In 1993 Souryal et al[61] studied a cohort of 905 high school athletes, finding a stenotic intercondylar notch was a significant predictor of ACL injury, whereas Lombardo et al[40] in found intercondylar notch size to have no significant influence on ACL injury in 615 professional basketball players. Review finds inconclusive evidence of increased quadriceps angle and intercondylar notch stenosis as predictors of ACL injury. There is no conclusive evidence that specific anatomical factors predispose individuals to non-contact ACL injury.[21, 22, 30, 40]

Hormonal

Estrogen receptor sites have been identified on the female ACL.[38] One major difference between the female and male sex is that of the presence of the menstrual cycle, and the levels of hormones present in the body. The menstrual cycle in the female introduces the fluctuation of hormone levels within an average 28-day period. ACL laxity has been observed to change in response to the levels of reproductive hormones.[24, 60, 62] Additionally, the risk of sustaining a non-contact ACL injury is not equal across the menstrual cycle.[4, 26, 66] Arendt et al's(E. A. Arendt, MD, un-published data, 1999) study of 108 ACL injured NCAA female basketball players revealed that athletes tended to experience injury just before or after the onset of menses. However, this area is not without controversy with several other studies indicating that ACL injury risk is equal across the menstrual cycle. Further research is warranted in this area to produce conclusive findings regarding the significant effects of female sex hormones on knee injury risk in the athlete.[21, 22]

Biomechanical & Neuromuscular Control Factors

Biomechanical and neuromuscular control factors of ACL injury are of specific concern because they have been proposed to be modifiable.[22, 50, 68, 69]

Kinematics

Lower extremity kinematics during dynamic tasks has been investigated, the literature has proposed increased knee valgus angle,[15, 64, 68-70],decreased knee flexion angle,[7, 8] increased hip adduction angle,[27, 50] and increased tibial internal rotation[2, 27] to be risk factors of ACL injury during athletic participation.

Knee Flexion Angle

Previous research has shown low knee flexion angles influence ACL loading.[10, 27, 37] In a prospective study, Hewett et al[27] observed a 10.5° decrease in knee flexion angle between injured and un-injured female athletes. Lin et al[37] showed smaller knee flexion angles in simulated ACL injury trials of stop-jump tasks. Boden et al[10] observed lesser degrees of knee flexion in 27 video-taped instances of non-contact ACL injuries in athletes. In 2008 Blackburn and Padua et al[8] showed that knee flexion angle can be increased during a drop-landing task when individuals were instructed to forward flex their trunk during landing. Blackburn and Padua et al[8] proposed the sine product of the insertion angle of the quadriceps/patellar tendon derives the resultant anterior shear force contribution from the quadriceps on the tibia. As the insertion angle increases (knee flexion angle decreases), the resultant quadriceps force exerts an anterior shear force along the anteroposterior axis that is greater than that of a decreased tendon insertion angle cosine product. Tendon insertion angle has been proposed to be inversely proportional to knee flexion angle, thus a decreased knee flexion angle results in an increased anteroposterior axis force of the quadriceps on the tibia.[7, 8] The increase in anteroposterior axis force likely attributes to increased anterior tibial shear force. Anterior tibial shear force directly loads the ACL.[21, 22]

Knee Valgus & Hip Adduction Angle

Increased knee valgus posturing during a dynamic task has been shown to be a biomechanical risk factor of ACL injury.[15, 64, 68-70] Cadaveric studies have shown a greater degree of stress is placed on the ACL during closed chain loading in the knee valgus positioning versus closed chain loading in the neutral frontal plane position[43, 64]. When the knee is dynamically loaded into a flexed and valgus position, the ACL experiences significantly greater strain compared to the knee that is loaded in the flexed position only.[43, 64] This valgus loaded position may increase the risk of ACL injury due to the increased degree of stress that is placed on the ACL when absorbing force from a dynamic motion, such as landing or cutting in athletics.[26, 27, 50]

Hip adduction angle and knee valgus angle have been shown to be positively related.[27] Knee valgus may increase as result of poor femoral proximal control. Theoretically, this is caused by hip abductor muscle weakness or adductor tightness/overactivity, as the individual is unable to control the hip during a dynamic task, preventing the femur from collapsing into adduction. Further research is required on specific modifiers of knee valgus.[22, 27]

Tibial Internal Rotation Angle

Increased tibial internal rotation during athletic tasks has been theorized to increase the risk of knee injury with common involvement of the ACL.[21, 22, 27, 43, 67] In a review article, Yu et al[67] indicates transverse plane rotation of the tibia alone does not cause isolated ACL injury. Commonly, ACL injury may result from a combination of internal tibial rotation with a host of biomechanical risk factors present during the injury mechanism. Yu et al[67] theorizes increased internal tibial rotation leads to injury of additional soft tissue structures of the knee when the posturing occurs during injury.

The ACL arises proximally from the posteromedial aspect of the lateral femoral condyle within the intra-condylar fossa of the femur and inserts distally at the posterior aspect of the anterior inter-condyloid eminence of the tibia.[30, 45] When the femur externally rotates on a fixed tibia, such as would occur in the closed kinetic chain, the ACL experiences an increase in stress.[21, 22, 30, 67] The open kinetic chain equivalent of femoral external rotation is tibial internal rotation. A direct longitudinal stress is placed on the ACL during open chain tibial internal rotation or closed chain femoral external rotation.[44]

Kinetics

Kinetics are the forces that cause motion, in the case of the knee, these forces are translated across the joint, as the joint experiences these forces, the inert structures of the knee are loaded.[30] During athletic tasks, the knee experiences forces from the external environment and the internal force producing musculature. This review focuses on the external influence from vertical ground reaction force and knee valgus moment,[21, 22, 26, 27] and the internal force caused by knee extension moment caused by the quadriceps, resulting in anterior tibial shear force.[21, 22, 30]

External Knee Valgus Moment

Knee valgus angle results from a knee valgus moment about the anteroposterior axis of the frontal plane, it is similar to internal knee varus moment. As the external knee valgus moment increases, knee valgus angle will increase.[27] Increased knee valgus moment results in an increased risk of non-contact ACL injury,[21, 22, 27, 50, 70] as a consequence of increased strain in the ACL during dynamic tasks caused by increased knee valgus angle.[15, 21, 22, 30, 57]

Vertical Ground Reaction Force

Increased vertical ground reaction force results from landing stiff legged and in a position of shallow knee flexion.[8, 21, 22, 30] As the force travels superiorly through the lower extremity, it crosses the knee joint where the ACL is exposed to the force. An increased ground reaction force results in an increased load on the ACL during athletic tasks as the knee extension musculature must counter a greater resultant external flexion force[9, 21, 22, 30] The increased load on the ACL results in greater strain within the ligament, increasing the risk of injury.[30]

Knee Extension Moment

To absorb ground reaction force adequately, the lower extremity acts as a rotational spring at the knee. As the vertical ground reaction force is applied to the body, the knee flexes to dissipate this force.[8, 30] To prevent the knee from collapsing into full flexion, the quadriceps musculature contracts, exerting an internal extension torque about the mediolateral axis of knee as the vertical ground reaction force causes an external flexion torque.[8, 21, 22, 64] The extension torque caused by the quadriceps introduces strain to the ACL in the anterior direction, especially when the knee is near or approaches full extension.[8, 14, 36, 64] This anteriorly directed strain is defined as internal ATSF. ATSF pulls the tibia forward relative to the femur, the exact motion the ACL restrains.[30, 63] When this force is high and un-checked by the hamstring musculature, a greater degree of stress is placed across the ACL, deformation past the elastic range results, and ligament injury ensues.[13, 30]

Kinematic & Kinetics Summary

Increased static knee valgus alone has not been shown to be a predictor of ACL injury.[22, 27] Increased dynamic knee valgus angle however, has been shown to increase the risk of ACL injury, as it has been observed to occur concomitantly with increased hip adduction, decreased knee flexion angle, increased tibial internal rotation angle, increased knee valgus moment, and increased vertical ground reaction force.[10, 22, 27, 43, 64, 67] As these biomechanical risk factors occur together, they place the individual at a greater risk of non-contact ACL injury[10, 22, 27, 43, 64, 67].

The focus of this study will concentrate on the triplanar motion of the knee during a cutting task, and its relationship to NMC of the trunk. The trunk houses the body's center of mass CoM,[11, 12, 27, 32, 50, 68-70] control of Com during a task is paramount for permitting functional and safe athletic function. During a cutting task, the CoM is required to rapidly change its direction of travel. Adequate control of the trunk manipulates the CoM as it exerts force about the knee due to gravity and additional acceleration forces.[11, 27, 32]

Trunk Stability

Kibler et al[32] defines trunk stability as the "ability to control position and motion of the trunk over the pelvis, thereby allowing optimum production, transfer, and control of force and motion to the terminal (anatomical) segment in integrated athletic, kinetic chain activities." Trunk stability is of utmost importance in understanding and preventing ACL injury, as it is a modifiable element.[11, 49, 50] Previous literature has concentrated on the effects of training or modifying the action of the lower kinetic chain and ACL injury prevention.[21, 22] The study will focus on the relationship between NMC of the trunk and its relationship to LE movement patterns during a dynamic task. Previous research by Blackburn and Padua et al[8] has highlighted a significant relationship between trunk and knee motion in the sagittal plane. Subjects were instructed to land with a forward trunk positioning, that concomitantly increased knee flexion and hip flexion range of motion during a drop-landing task. Increased hip and knee flexion during athletic tasks, specifically landing, has been indicated to decrease the risk of ACL injury, placing less stress on the ACL during activity.[14, 15, 21, 22, 64]

Using a cohort of collegiate athletes, Zazulak et al[69] introduced a direct relationship between a laboratory measure of trunk NMC and ACL injury. Subjects were placed in a force release apparatus where all extremity influence was removed, subjects relied solely on their trunk musculature to resist motion after sudden force release. A positive relationship was observed between trunk motion (variable of trunk NMC) and non-contact ACL injury incidence in female athletes only.[69] Zazulak et al[69] found increased frontal plane trunk motion to be the strongest predictor of ACL injury.

A significant relationship has also been observed between trunk proprioception and ACL injury.[70] Subjects were placed in a rotational seat apparatus that required subjects to actively and passively reposition themselves into previously defined and experienced angular positions. In the passive trials, a stepper motor repositioned the subjects, subjects were then required to signal with a hand trigger the point they believed themselves to be located at the initial angular position. In the active trials, subjects repositioned themselves with their trunk musculature to the initial angular position. The cohort was followed throughout their collegiate careers. Zazulak et al[70] found a significant relationship between deficits in active trunk repositioning and knee injury incidence in the female athletes.

Cholewicki et al[12] investigated a dynamic measure of postural control of the trunk using an unstable-sitting apparatus. The apparatus removed all external influence from the extremities on the trunk, having subjects seated with their legs in contact with a seat device that remained connected to a hemisphere, upon which subject's flank rested atop. Subject's upper extremities were held across their chests. The unstable-sitting apparatus was composed of a hemisphere atop a force plate that measured the system's CoP excursion. Cholewicki defined postural control of the trunk in this study as the "ability to maintain a stable sitting position without interference from the extremities, relying solely on the lumbopelvic hip musculature to maintain equilibrium with measure of the CoP path."[12] Cholewicki defined greater values of excursion measures to represent poor trunk stability and control, while smaller excursion values represented satisfactory control and stability of the trunk.[12]

In the current study, we implemented the methodology of Cholewicki and the unstable-sitting apparatus to measure trunk stability. Zazulak's et al's[69, 70] methods are representative of reaction, with the ability to resist motion after a sudden force release, and proprioception with rotational trunk angle repositioning. Cholewicki's[12] methodology introduces an element of necessity to maintain stability of the LPHC that is the variable of interest in the current study.

Trunk Stability and the Lower Extremity

Trunk muscle activity and CoM position has been indicated to precede lower extremity function during athletic tasks.[5, 27, 29, 32, 33, 68, 69] As we draw on the lower extremity system during an athletic task, and focus the center of the system about the knee joint, we are lead to concentrate above and below the joint. Below the joint the ankle and foot contact the ground closing the kinetic chain. Above the knee joint is the hip, and globally the LPHC, which houses the trunk musculature. The literature has established that the core region of the body is the area in which the body's CoM most commonly exists;[68-70] the trunk encompasses the core region. Establishing the knee joint as a three-dimensional fulcrum or center point of the lower extremity system, an inverted double pendulum model[42, 52] of the system can be established.

Once the shank is fixed to the ground via the foot-ground interface, degrees of freedom of movement about this interface are limited to the three tri-axial movement points of the hip, knee, and ankle.[21, 22, 30, 52] Above the knee fulcrum at the femur-LPHC interface, the degrees of freedom in movement are still un-limited.[32] The CoM sits atop the femur moment arm, free to move in all planes of motion. The dynamic moment contributes to an increased demand across the lower extremity at the knee and then terminally at the ankle. The contractile mechanisms attempting to resist instability at the knee must account for instability at the LPHC as well, as the knee and hip joint share common musculature, such as the rectus femoris, medial and lateral hamstring, and pes anserine group. This rationale leads to the focus of this study, to examine the relationship between LPHC NMC and lower extremity knee biomechanics during an inherently unstable task of cutting.

Rationale

The relationship between dynamic control of the trunk and lower extremity biomechanics is of interest because NMC of the trunk is thought to be modifiable.[11, 21, 22, 50, 68] If NMC of the trunk is in fact modifiable, and the current study establishes a significant relationship between NMC of the trunk and lower extremity biomechanics, NMC training of the trunk may decrease the presence of poor lower extremity biomechanics, potentially decreasing the risk of ACL injury.[11]

Until the current study, no known research has related the multidimensional measure of the stability element of trunk NMC, as in Cholewicki et al's[12] measure to the lower extremity biomechanics of a known dynamic mechanism of non-contact ACL injury, an athletic cutting maneuver. Past studies have focused on the static measures of NMC and lower extremity biomechanics.[27, 33, 69, 70] No known methodology incorporates an athletic task in which proposed biomechanical risk factors can be measured and their relationship to a laboratory measure of trunk stability. The relationship between trunk NMC and LE biomechanics during an inherently unstable task is of importance due to the ramifications of an unstable LPHC and its potential to influence mechanics of the knee during athletic tasks.

The pelvic girdle is a proximal attachment site for many muscles that act on the knee. It is hypothesized that a stable bony origin for the bi-articulate musculature will decrease the presence of extraneous forces and motion across the more distal knee joint, as the muscles originating from the stable pelvis will be capable of exerting adequate stabilizing forces across the knee joint that may otherwise be disrupted or decreased due to excessive LPHC motion.

The focus of this study is to investigate a measure of trunk stability[12] and the prevalence of biomechanical risk factors of ACL injury during a cutting task.

CHAPTER III

METHODS

Experimental Design

We used simple correlation analyses to evaluate 1) relationships between trunk NMC and trunk motion during a cutting task and 2) relationships between trunk motion and knee kinematics and kinetics during a cutting task. Specifically the predictor variables of CoP elliptical sway area, sway path, and sway velocity and the criterion variables of peak frontal, sagittal, and transverse plane trunk angles; knee flexion, valgus, and tibial internal rotation angles; and internal knee extension and varus moments during the first 50% of the stance phase of the side-step cutting task were used to evaluate these relationships. In a secondary analysis we examined relationships between peak frontal, sagittal, and transverse plane trunk angles (predictor variables) and peak knee flexion, valgus, and tibial internal rotation angles, and internal knee extension and varus moments (criterion variables) during the first 50% of the stance phase of the side-step cutting task

Subjects

Thirty subjects from the university setting volunteered for this study (age range 18-35). Subjects were eligible for participation if they were recreationally active, participating in physical activity for at least twenty minutes at least three times per week.

Subjects were excluded from the study if they could not perform the tasks, had a history of ACL injury, had lower extremity surgery in the past year, had a low back injury or pain at the time of the study, had previously participated in an ACL injury prevention program, or had suffered any known lower extremity injury within the past six months prior to testing.[8] An injury was defined as a traumatic event or presence of symptoms that restricted activity for more than three days. Prior to data collection, subjects read and signed an informed consent form (see appendix A) approved by the institutional review board.

Instrumentation

3-Dimensional Motion Capture System

An infrared optical motion capture system (Vicon MX Camera (7), Vicon Motion Systems, Los Angeles, California) was used to collect kinematic data at a sampling rate of 150 Hz as subjects performed the cutting task. The measurements were recorded by Vicon Nexus 1.4.116 motion capture software (Vicon Motion Systems, Los Angeles, California). The capture volume was established centered about a force plate (Bertec Corporation, Columbus, OH) which was used to record ground reaction force during the cutting task at 1500 Hz. Force plate data were collected to derive kinetic variables via inverse dynamics, as well as to provide timestamp points throughout ground contact during the cutting task.

World and segmental axis systems were established by a right hand three-dimensional Cartesian coordinate system. The positive X-axis was designated forward/anterior direction, the positive Y-axis leftward (world)/medially (right

LE)/laterally (left LE), and the positive Z-axis upward/superiorly[8].The motion capture system was calibrated prior to each data collection session. All kinematic and kinetic data collected during the cutting task trials was imported into The Motion Monitor version 8 software system (Innovative Sports Training, Inc., Chicago, IL) for analysis and subject modeling.

Joint angles were calculated using an Euler angle sequence with the distal segment relative to the proximal segment, with the primary axis of rotation about the Y-axis (flex/ext), secondary axis of rotation about the X-axis (add/abd), and tertiary rotation about the Z-axis (internal/external rotation). Sagittal plane knee flexion (+)/extension (-) and frontal plane knee valgus (-)/varus (+) - (right LE), knee valgus (+)/varus (-) - (left LE), transverse plane tibial internal rotation (+)/external rotation (-) – (right LE), internal rotation (-)/external rotation (+) – (left LE) were analyzed. Sagittal plane trunk flexion (+)/extension (-), frontal plane trunk leftward flexion (-)/rightward lateral flexion (+), and transverse plane rightward rotation (-)/leftward rotation (+) were analyzed. Kinematic data were filtered using a low-pass Butterworth digital filter at a cutoff frequency of 15 Hz.

Subjects wore non-reflective spandex shorts and shirts during motion analysis. A set of 25 static/calibration reflective tracking markers were placed on the subject for the static modeling trial. After the static trial, 4 medial markers were removed and the remaining 21 reflective markers were used for motion analysis of the cutting task. The dynamic trial tracking marker set included left/right acromion process, L4-L5 joint space, right/left anterior superior iliac spine, right/left greater trochanter, right/left anterior thigh, right/left lateral epicondyle, right/left anterior shank, right/left lateral malleolus, right/left calcaneous, right/left 1st metatarsal head, and right/left 5th metatarsal head. The four medially placed static/calibration markers include; right/left medial epicondyle, right/left medial malleolus. Reflective markers were secured to the skin and black spandex with double-sided carpet tape.

A static trial was collected with the subject facing the positive X-axis of the world coordinate system. Joint centers of the hip, knee, ankle, and the trunk at the L4-L5 level were established with the described marker set. The hip joint centers were calculated using the Bell method.[6] The knee joint center was calculated as the midpoint between the medial and lateral knee joint line markers. The ankle joint center was calculated as the midpoint between the lateral and medial malleolus. The trunk axis of rotation was defined as the L4-L5 joint interspace marker permitting rotation of the thorax segment in all three planes. Medial markers were removed after the static trial was conducted, and a subject template was established from the static trial as medial markers and joint centers were defined virtually during the experimental trials. A segment reference system was established on the subject via the static trial digitization process. Relative joint angles were based on segment positions of the subject illustrated by the reflective marker positions, defined as the distal segment relative to the proximal segment. Inverse dynamics procedures were used to calculate the kinetic variables of internal knee varus extension moments during the cutting task trials.

Unstable-Sitting Apparatus

A plyometric box with dimensions 61 cm x 61 cm x 45.72 cm with a conductive force plate (Bertec Corporation, Columbus, OH) supported a wooden seat atop a 30 cm polycarbonate resin hemisphere. The wooden seat included an adjustable foot rest to ensure the subjects' hips and knees were positioned at 90º of flexion, with their ankle joints at neutral (0º).[12] The plyometric box was fitted with handrails, allowing subjects to steady themselves when initially sitting on the unstable seat. A computer running Motion Monitor version 8 software system (Innovative Sports Training, Inc., Chicago, IL) sampled data from the conductive force plate (Bertec Corporation, Columbus, OH), specifically vertical ground reaction forces and moments about the X-, Y-, and Z-axes were sampled at 100 Hz using a low-pass 20 Hz Butterworth filter. Kinetic data were used to calculate CoP location during data reduction.[12, 53-56]

Procedures

Subjects reported to the Sports Medicine Research Laboratory (SMRL) for a single testing session. Subjects completed a health history and physical activity readiness questionnaire (PAR-Q) (see appendix D) to ensure compliance with the study's inclusion criteria, and to record demographic data. The demographic data obtained from the questionnaire included information regarding subject age, level of activity (ie. collegiate athlete, club sport athlete, intramural athlete, recreational athlete, physically active individual (fitness), sedentary individual), previous athletic experience, and previous injury history. The subject's dominant leg was defined as the leg they would use to kick a soccer ball for maximum distance. This leg was used as the test limb.

Subjects were fitted with non-reflective black spandex clothing. The experimenter provided explanation and demonstration of the unstable-sitting and cutting tasks. Prior to each task, the subject warmed-up on a stationary bike for 5 minutes at a level of 11 RPE and completed a static lower-extremity stretching program. Order of task assignment remained constant, with subjects performing the cutting task prior to the unstable-sitting task. The cutting task was completed prior to the unstable sitting task to decrease the likelihood of introducing the potential effects of trunk muscle fatigue.

Cutting Task

The task required subjects to stand in the capture volume at 50% of their height from the leading edge of the force plate and double-leg jump over a short hurdle (17.0 cm) placed 25% the subject's body height from the leading edge of the force plate. Subjects landed on their dominant foot, contacting only the force plate, cutting either left or right at a 60º angle from the center of the trailing edge of the force plate. The order of cutting direction was predetermined and counterbalanced by the experimenter. Subjects cut off their dominant leg. A cut in the ipsilateral direction of the dominant leg required the subject to cross-cut their dominant leg with their non-dominant leg. A cut in the contralateral direction required the subject to side-step cut with their dominant leg (side-step cut). A successful trial included the test extremity's foot landing on the force plate to signal initial ground contact as the beginning of the stance time (vertical ground reaction force > 10 N) and continued until toe-off (vertical ground reaction < 10 N). All variables were examined during the first 50% of the stance phase.

The cutting task was re-explained to the subject before the trials. The subject was allowed to practice the task a maximum of three times in each direction, alternating between side-step and cross-over cuts. After the practice trials were completed, the subject performed 10 trials of the task, alternating cutting direction, with an equal number of cuts in each direction, left (5) and right (5). The subject was given a 30 second rest period between each trial.

Unstable-sitting Task

The task required the subject to maintain stability on the sitting apparatus without aid. Initially, the subject maintained stability with the handgrips while the researcher helped to position the subject such that his/her hips and knees were at 90º of flexion, and the ankles were at neutral (0º flexion). The subject was instructed to keep the medial surfaces of the knee joints "touching" throughout each trial. The subject was told to continue holding onto the handgrips until the researcher cued them to "let go" to begin the stability task, at which point the subject crossed his/her arms across the chest. Once the subject established equilibrium, they verbally cued the researcher when they were "ready." The researcher began data collection of CoP location once he received the "ready" cue from the subject. Kinetic data were sampled for a one-minute trial. If the subject could not complete the one-minute trial without touching down or having to stabilize him/herself with the handrails, the trial was discarded. One-minute of rest was given between each trial, and five trials were recorded. The unstable-sitting task was re-explained to the subject before the trials. The subjects were allowed to practice the unstable-sitting task a maximum of three times.

Data Reduction

All data reduction was completed using customized Matlab software (v 7.0, The Mathworks, Inc., Natick, MA). Kinematic and kinetic data of the side-step cutting task were reduced after data collection. The mean values for biomechanical variables for each subject were calculated and used in data analysis.

CoP elliptical sway area, sway path, and sway velocity were calculated from CoP location data. CoP sway area was calculated as elliptical area in square centimeters encompassing 95% of the CoP location points during the one-minute trial. CoP sway path was calculated as the total displacement path in centimeters of the CoP during the one-minute trial. CoP sway velocity was calculated as CoP displacement divided by time (cm/s). The mean CoP sway area, sway path, and sway velocity for each subject was calculated from the three middle trials (2, 3, 4) if all 5 trials had no episodes of touching down. If there was an episode of touching down in the middle three trials, trial 1 or 5 replaced one of the middle trials. If two middle trials had episodes of touching down, trials 1 and 5 replaced two of the middle trials.

Peak triplanar trunk and lower extremity kinematics and internal knee extension and varus moments were calculated during the first 50% of the stance phase of the side-step cutting task. Internal knee moments were standardized to the product of the subject's weight (N) and height (m),. Average peak angles and moments were calculated across the 5 side-step cutting trials.

Statistical Analysis

Statistical significance was set a-priori at $p < 0.05$. 24 bivariate Pearson r correlations were run to examine the relationships between CoP sway measures (3) during unstable-sitting and the trunk (3) and lower extremity mechanics (5) during the side-step cutting task. 15 bivariate Pearson r correlations were run to examine the relationships between triplanar trunk kinematics (3) and lower extremity mechanics (5) during the cutting task. SPSS version 16.0 (SPSS INC, Chicago, Ill) was used for all statistical calculations.

Research Design Rubric

Question	**Description**	**Data Source**	**Method**
1	Is there a significant relationship unstable-sitting CoP sway measures and triplanar trunk kinematics during a side-step cutting task?	*Predictors:* - CoP sway area - CoP sway path - CoP sway velocity *Criterion:* - Peak sagittal plane trunk angle - Peak frontal plane trunk angle - Peak transverse plane trunk angle	Pearson r correlations (9 total)
2	Is there a significant relationship between CoP sway measures during unstable-sitting and peak triplanar knee kinematics during a side-step cutting task?	*Predictors:* - CoP sway area - CoP sway path - CoP sway velocity *Criterion:* - Peak sagittal plane knee angle - Peak frontal plane knee angle - Peak transverse plane knee angle	Pearson r correlations (9 total)

3	Is there a significant relationship between CoP sway measures during unstable-sitting and peak internal knee varus and extension moments during a side-step cutting task?	*<u>Predictors:</u>* - CoP sway area - CoP sway path - CoP sway velocity *<u>Criterion:</u>* - Peak internal knee extension moment - Peak internal knee varus moment	Pearson r correlations (6 total)
4	Is there a significant relationship between peak triplanar trunk kinematics and peak triplanar knee kinematics during a side-step cutting task?	*<u>Predictors:</u>* - Peak sagittal plane trunk angle - Peak frontal plane trunk angle - Peak transverse plane trunk angle *<u>Criterion:</u>* - Peak sagittal plane knee angle - Peak frontal plane knee angle - Peak transverse plane knee angle	Pearson r correlations (9 total)
5.	Is there a significant relationship between peak triplanar trunk kinematics and peak internal knee varus and extension moments during a side-step cutting task?	*<u>Predictors:</u>* - Peak sagittal plane trunk angle - Peak frontal plane trunk angle - Peak transverse plane trunk angle *<u>Criterion:</u>* - Peak internal knee extension moment - Peak internal knee varus moment	Pearson r correlations (6 total)

CHAPTER IV

RESULTS

Thirty subjects (age = 20.5 ± 2.3 years, height = 173.78 ± 9.28 centimeters, mass = 67.50 ± 11.39 kilograms), 15 males (age = 20.9 ± 2.7 years, height = 178.06 ± 8.23 centimeters, mass = 75.08 ± 8.94 kilograms), and 15 females (age = 20.1 ± 1.9 years, height = 196.49 ± 8.44 centimeters, mass = 59.92 ± 8.11 kilograms) were tested. Three female and three male subjects' CoP elliptical sway areas, two female subjects' CoP sway paths, and three female subjects' sway velocities were not used in the statistical analysis due to equipment error. Means and standard deviations for CoP elliptical sway area, sway path, and sway velocity (Table 1.), tri-planar trunk kinematics during the side-step cutting task (Table 2.), tri-planar knee kinematics (Table 3.), external knee valgus moment and internal knee flexion moment (Table 4.) are presented.

Center of Pressure and Trunk Kinematics Correlations

Correlation coefficients were computed among the three predictor variables of CoP elliptical sway area, sway path, sway velocity, and the three criterion variables of peak triplanar trunk angles during the first 50% of stance phase of the side-step cutting task. The results of the correlational analyses presented in Table 5 show that two of the nine correlations were statistically significant. Significant positive relationships between the predictor variables of CoP sway path and velocity and the criterion variable of peak frontal plane trunk angle were observed. A moderate, positive, linear relationship (r(26) = 0.401, p = 0.034) was observed between CoP sway path and peak frontal plane trunk angle during the first 50% of the stance time of the side-step cutting task. This relationship suggests that as CoP sway path increases, peak sagittal plane trunk angle during the first 50% of the stance time increases. A moderate, positive, linear relationship (r(25) = 0.448, p = 0.019) was observed between CoP sway velocity and peak frontal plane trunk angle during the first 50% of the stance time during the side-step cutting task. This relationship suggests that as CoP sway velocity increases, peak frontal plane trunk angle during the first 50% of the stance time increases.

Center of Pressure and Knee Kinematics Correlations

Correlation coefficients were computed among the three predictor variables of CoP elliptical sway area, sway path, sway velocity, and the three criterion variables of peak triplanar knee angles during the first 50% of the stance time during the side-step cutting task. The results of the correlational analyses presented in Table 6 show that none of the nine correlations were statistically significant at the $p < 0.05$ level.

Center of Pressure and Knee Kinetics Correlations

Correlation coefficients were computed among the three predictor variables of CoP elliptical sway area, sway path, sway velocity, and the two criterion variables of peak internal knee varus moment and peak internal knee extension moment during the first 50% of the stance time during the side-step cutting task. The results of the correlational analyses presented in Table 7 show that two of the six correlations were statistically significant. Significant positive relationships between the predictor variables of CoP elliptical sway area and sway path, and the criterion variable of peak internal knee extension moment were observed. A strong, negative, linear relationship ($r(22) = -0.548$, $p = 0.006$) was observed between CoP elliptical sway area and peak internal knee extension moment during the first 50% of the stance time of the side-step cutting task. By directional convention a more negative internal knee extension value indicates a greater internal knee extension moment magnitude. This relationship suggests that as CoP elliptical sway area increases, peak internal knee extension moment magnitude during the first 50% of the stance time increases. A moderate, negative, linear relationship ($r(26) = -0.407$, $p = 0.006$) was observed between CoP sway path and peak internal knee extension moment during the first 50% of the stance time of the side-step cutting task. This relationship suggests that as CoP sway path increases, peak internal knee extension moment magnitude during the first 50% of the stance time increases.

Trunk Kinematics and Knee Kinematics During the Side-Step Cutting Task Correlations

Correlation coefficients were computed among the three predictor variables of peak triplanar trunk angles during the first 50% of the stance time, and the three criterion variables of peak triplanar knee angles during the first 50% of the stance time during the side-step cutting task. The results of the correlational analyses presented in Table 8 show that one of the nine correlations was statistically significant. A strong, positive, linear relationship ($r(28) = 0.586$, $p = 0.001$) was observed between the predictor variable peak transverse plane trunk angle and the criterion variable peak tibial internal rotation angle during the first 50% of the stance time of the side-step cutting task. This relationship suggests that as peak transverse plane trunk angle during the first 50% of the stance time increases, peak tibial internal rotation angle during the first 50% of the stance time increases.

Trunk Kinematics and Knee Kinetics Correlations

Correlation coefficients were computed among the three predictor variables of peak triplanar trunk kinematics during the first 50% of the stance time, and the two criterion variables of peak internal knee extension moment and peak internal knee varus moment during the first 50% of the stance time during the side-step cutting task. The results of the correlational analyses presented in Table 9 show that one of the nine correlations was statistically significant. A significant, positive, relationship between the predictor variable of peak transverse plane trunk angle and peak internal knee varus moment was observed. A moderate, positive, linear relationship ($r(28) = 0.371$, $p = 0.043$) was observed between peak transverse plane trunk angle and peak internal knee varus moment during the first 50% of the stance time of the side-step cutting task. This relationship suggests that as peak transverse plane trunk angle during the first 50% of the stance time increases, peak internal knee varus moment during the first 50% of the stance time increases.

CHAPTER V

DISCUSSION

The primary purpose of this study was to examine the relationships between measures of trunk stability during an unstable-sitting task, and trunk and knee biomechanics during a side-step cutting task. A secondary purpose was to examine the relationships between trunk and knee biomechanics during the same task. The results indicate that lesser trunk stability is associated with greater peak frontal plane trunk angle and peak internal knee extension moment during the initial 50% of stance time. Further, greater peak trunk rotation angle away from the stance leg during the first 50% of stance time during side-step cutting is associated with greater peak internal knee varus moment and peak tibial internal rotation angle during the first 50% of stance time during side-step cutting. To our knowledge, our findings are the first to demonstrate significant relationships between NMC of the trunk, trunk mechanics, and knee mechanics during an athletic task.

The primary finding of our study is the negative correlation between trunk stability and knee extension moment. Specifically, greater CoP elliptical sway area and sway path during an unstable-sitting task is associated with greater peak knee extension moment during side-step cutting. Previous research has demonstrated that an increase in trunk flexion angle decreases vertical ground reaction force, increases knee flexion angle, and decreases knee extension moment during the loading time in drop-landing tasks.[9] Landing in a more erect posture places the body's CoM mass in a more posterior position relative to the center of rotation of the knee joint, compared to landing with an increased trunk flexion angle. The increase in CoM posterior displacement increases the moment arm of the body's CoM mass relative to the knee joint. The greater moment arm results in a greater flexion moment about the knee, thus the quadriceps musculature is required to exert a greater magnitude internal extension moment about the knee.[9] Combined with an increase in knee extension moment and a decrease in knee flexion angle, there is a potential to increase ACL loading via ATSF.[9, 13, 39] Theoretically, increased CoP elliptical sway area and sway path indicate decreased trunk control, thus, increasing knee extension moment.

The rationale for the link between decreased trunk control and increased knee extension moment is supported by previous findings of the body's CoM influence on lower extremity biomechanics.[8, 9] The ability of subjects to actively increase their trunk flexion angle during the loading time with provided augmented feedback indicates trunk NMC specific to CoM position during the loading time of drop-landing tasks is modifiable in the sagittal plane. An inability to control the body's CoM during athletic tasks may manifest in a more erect posture. However, a relationship between trunk stability during unstable sitting and sagittal plane trunk angle during the cutting task was not observed, a relationship between trunk stability and internal knee extension moment did manifest. It is important to note no relationship between sagittal plane trunk angle or knee extension moment was observed during the cutting task as indicated in previous research.[8, 9] It is possible the lack of an association between sagittal plane trunk angle and knee extension moment is specific to our methodology; incorporating a more inclusive triplanar acceleration in contrast to previous methodology,[8, 9] which implemented a sagittal plane dominant acceleration. Although no direct relationship between sagittal plane trunk kinematics during cutting and trunk stability during unstable sitting, or sagittal plane trunk kinematics during cutting and internal knee extension moment was presented, it is important to note an association between trunk stability during unstable sitting and internal knee extension moment was observed. Our study's findings direct future research to examine the effects of the trunk stability element of trunk NMC on trunk position and internal knee extension moment during tasks that have been proposed to be mechanisms of ACL injury. Future research methodology would require interventions that are specifically oriented toward improving trunk stability to determine the effect of the stability component of trunk NMC on trunk position and internal knee extension moment.

Another important finding of this study is the association between trunk stability and frontal plane trunk angle during the side-step cutting task. Greater magnitudes of CoP sway area and sway velocity are associated with greater lateral trunk flexion angles towards the stance leg. In theory, increased lateral trunk flexion is also indicative of decreased trunk control. It is interesting to note that in a prospective study of Division I collegiate athletes, increased lateral trunk flexion after sudden force release was a significant predictor of subsequent knee injury in females.[69, 70] While the methodology between our studies differs, it is apparent that the same kinematic pattern of increased lateral trunk flexion is observed during a cutting task in individuals demonstrating lesser trunk stability, and during a dynamic perturbation in individuals who subsequently suffer a knee injury. This suggests that there are important links between trunk NMC, lower extremity mechanics, and injury. However, further research is needed to elucidate the specific mechanisms underlying these observations.

The next significant correlation of our study was the positive relationship between peak transverse plane trunk angle and knee varus moment during side-step cutting. This indicates that greater amounts of trunk rotation (away from the stance leg) are associated with greater knee varus moments. Internal knee varus moment is similar to external knee valgus moment, which is a prospective risk factor for non-contact ACL injury[27]. Our findings agree with Houck et al,[29] who observed a positive relationship between frontal plane peak trunk angle and hip abduction moment during side-step cutting. Houck et al's findings differ from our study in that they are specific to the frontal plane. We found relationships between two planes, trunk motion in the transverse plane, and knee kinetics in the frontal plane.

Increased frontal plane trunk angle drives hip abduction moment, a frontal plane torque proposed to act as a stabilizing force for the lower extremity.[29] The abduction moment prevents excessive hip adduction angle from occurring as a result of increased lateral CoM positioning at the trunk (lateral flexion).[29] Previous research is in agreement with our study, presenting an association between trunk motion and lower extremity kinetics, yet previous findings have only indicated intra-planar relationships,[23, 29] versus the our study's findings, implicating a relationship between trunk kinematics in the transverse plane, and knee kinetics in the frontal plane. Although our findings differ from Houck et al's,[29] previous in-vitro cadaveric research has implicated external knee valgus moment to increase ACL strain during a simulated jump-landing task and combined external loading conditions of the knee.[43, 64] The findings of our study demonstrate that increased trunk rotation away from the stance leg increases internal knee varus moment, an identified risk factor of ACL injury.[27] Our study provides a link between transverse plane trunk motion and knee injury risk.[27]

It is possible that the variation between our results regarding inter-planar relationships and previous findings specific to intra-planar relationships may be a function of task specificity. Previous literature has examined slower unanticipated cutting tasks during walking gait,[29] versus anticipated, higher speed cutting tasks as in our methodology. It is possible that an increase in transverse plane trunk motion occurs during higher speed anticipated cutting tasks versus unanticipated changes of direction during walking. Our study provides evidence of a positive relationship between trunk rotation angle and knee varus moment, a risk factor of ACL injury.[1, 22, 27]

The final significant relationship we observed was the positive correlation between trunk rotation and tibial rotation (relative to the femur) during the side-step cutting task. In a cadaveric study, Markolf et al[43] demonstrated that addition of internal tibial torque induced the greatest amount of ACL loading when combined with an anterior tibial shear force, compared to knee valgus/anterior tibial force application at various knee flexion angles. Greater amounts of trunk rotation (away from the stance leg, towards the direction of the cut) were associated with greater amounts of tibial internal rotation, a proposed risk factor for ACL injury.[1, 31, 43] This result suggests that transverse plane moments causing rotation superiorly at the trunk may be transferred through the lower extremity kinetic chain, manifesting inferiorly at the knee in tibial rotation. This notion is supported by previous research which indicates trunk motion precedes all other body segment motion during locomotor tasks requiring a change in direction of travel.[52]

It is proposed that the trunk motion precedes lower extremity motion in order to accelerate the body's CoM in the new direction of motion.[29, 52] Furthermore, Imwalle et al[31] also observed this concept of segment kinematic symmetry, in which there was a positive association between transverse plane knee and hip internal rotation in female soccer players. While the finding of our study is limited in that it is focused on the relationship between transverse plane trunk and knee kinematics without evaluating the kinematics of the thigh, it is clear the intra-planar segmental relationship noted by Imwalle et al seems to persist in non-adjacent segments. As such, our study provides preliminary evidence for a link between transverse plane trunk motion and a proposed risk factor for ACL injury; tibial internal rotation.

Combined loading across all three cardinal planes has been demonstrated to impart the greatest load on the ACL.[43] The results of our study indicate decreased trunk stability during unstable sitting and increased transverse plane motion of the trunk during an athletic cutting task are associated with lower extremity biomechanical risk factors for ACL injury in all three planes. In the sagittal plane we observed greater internal knee extension moment to be associated with less trunk stability during unstable sitting. In the frontal plane we observed greater internal knee varus moment to be associated with greater transverse plane trunk angle away from the stance leg. In the transverse plane we observed greater internal tibial rotation angle to be associated with greater transverse plane trunk angle away from the stance leg. Our results support previous findings of Zazulak et al,[69] identifying decreased trunk NMC to be a risk factor of ACL injury. Our study links trunk NMC to lower extremity biomechanical risk factors of ACL injury in all three planes. The results of our study direct future research to establish an understanding of the effects of NMC of the trunk on lower extremity biomechanical risk factors for ACL injury.

Contrary to our hypotheses, we were not able to demonstrate any significant relationships between measures of trunk stability and knee kinematics. It is possible that the magnitude of any potential relationships between our trunk stability measures and knee biomechanics during a side-step cutting task are either too small to be detected in the current analysis, or that the task itself was not demanding enough dynamically to bring about biomechanical associations. While we did observe that decreased trunk stability was associated with greater internal knee extension moment demand, it may be that within the cutting task the participants demonstrating lesser trunk stability were still capable of meeting the biomechanical demands of the side-step cutting task, and therefore did not display knee mechanics that were appreciably different than those with greater trunk stability. This suggests that future research should examine more demanding tasks as well as multiple elements of trunk NMC, as previous research has identified that deficits in reactive and proprioceptive elements of trunk NMC are predictive of knee injury risk.[69, 70] A model encompassing proprioceptive, reactive, and stability components of trunk NMC may permit future research to more comprehensively associate trunk NMC with lower extremity biomechanics during athletic tasks that are related to knee injury.

Limitations

This study has four primary limitations that should be addressed. First, as a correlational design can only establish relationships between variables and not determine cause and effect, the results of this study are limited in that they are preliminary, exploratory evidence that should be used as rationale for future research investigations. Second, subjects with a history of balance training and performance of cutting tasks may exhibit improved performance over those who have not had previous exposure to the tasks. As the inclusion criteria for the study included participation in physical activity for at least 20 minutes a day at least 3 times per week, the activity level of our participants ranged from this minimum to Division I collegiate varsity athletes. It is not possible to discern the effects, if any, of physical activity participation level on the variables of interest. Thus, the study's findings can only be generalized to this specific population in regards to physical activity participation level. Third, the average age of the subjects in this investigation was 20.5 ± 2.3 years. While this demographic is representative of college-aged young-adults, our findings should not be generalized to older or younger populations. Finally, we assumed that the CoP measures were representative of the body's ability to stabilize the trunk without influence from the extremities or head. Although participants were instructed to maintain a standardized test position (arms across chest, looking straight ahead) minor movement of the extremities and head may have had the ability to influence the CoP results.

Conclusions

1. Decreased trunk stability is associated with increased peak knee extension moment, a risk factor for ACL injury. Therefore, training to increase trunk stability may result in decreased peak knee extension moment during athletic tasks, and a potential reduction in ACL injury risk.

2. Decreased trunk stability is associated with increased lateral trunk flexion toward the stance leg during a side-step cutting task. Increased lateral trunk flexion towards the stance leg during cutting tasks has been associated with ACL injury risk. Trunk stability training may decrease lateral trunk flexion towards the stance leg during athletic tasks, possibly decreasing the presence of lower extremity mechanics associated with ACL injury, and ACL injury risk.
3. Increased trunk rotation away from the stance leg is associated with increased knee varus moment and increased tibial internal rotation, identified risk factors of ACL injury. Future ACL injury prevention strategies decreasing trunk rotation during side-step cutting tasks may decrease knee biomechanics associated with ACL injury.
4. No significant relationships were observed between trunk stability and knee kinematics. The stability element of trunk NMC may not be directly associated with knee kinematics during a cutting task. Future research should implement comprehensive measures of trunk NMC.

Table 1. Center of pressure means during unstable sitting task, values are in cm^2, cm, and cm/s respectively and standard deviations

Ellipse Sway Area	Sway Path	Sway Velocity
50.29 ± 18.50	81.02 ± 14.44	1.34 ± 0.23

Table 2. Peak trunk kinematics means during side-step cutting, values are in degrees and standard deviations

Sagittal Plane Trunk Angle	Frontal Plane Trunk Angle	Transverse Plan Trunk Angle
36.80 ± 9.45	6.80 ± 2.82	6.92 ± 8.24

Table 3. Peak knee kinematics means during side-step cutting task, values are in degrees and standard deviations

Sagittal Plane Knee Angle	Frontal Plane Knee Angle	Transverse Plane Knee Angle
46.94 ± 4.72	6.75 ± 4.96	0.767 ± 7.38

Table 4. Peak internal knee kinetics (moments) means during side-step cutting task, mean values are quotient of subjects' absolute moments divided by the product of the subjects' masses (N) and heights (m)

Knee Varus Moment	Knee Extension Moment
0.01 ± 0.01	-0.12 ± 0.03

Table 5. Pearson Correlation Coefficients for Unstable Sitting Center of Pressure and Peak Trunk Kinematics During a Side-Step Cutting Task

	Ellipse Area		Sway Path		Sway Velocity	
	r	*p*	r	*p*	r	*p*
Sagittal Plane Trunk Angle	0.026	0.903	0.094	0.635	0.079	0.694
Frontal Plane Trunk Angle	0.207	0.332	0.401*	0.034	0.448*	0.019
Transverse Trunk Angle	0.240	0.258	0.180	0.359	0.301	0.127

*correlation is significant at the 0.05 level (two-tailed)

Table 6. Pearson Correlation Coefficients for Unstable Sitting Center of Pressure Data and Peak Knee Kinematics During a Side-Step Cutting Task

	Ellipse Area		Sway Path		Sway Velocity	
	r	p	r	p	r	p
Sagittal Plane Knee Angle	0.399	0.053	0.120	0.543	0.119	0.555
Frontal Plane Knee Angle	-0.208	0.330	0.012	0.950	0.028	0.891
Transverse Plane Knee Angle	0.346	0.098	0.278	0.153	0.362	0.064

Table 7. Pearson Correlation Coefficients for Unstable Sitting Center of Pressure Data and Peak Internal Knee Kinetics (moments) During a Side-Step Cutting Task

	Ellipse Area		Sway Path		Sway Velocity	
	r	p	r	p	r	p
Knee Varus Moment	-0.137	0.523	0.034	0.863	0.054	0.789
Knee Extension Moment	-0.548**	0.006	-0.407*	0.032	-0.363	0.063

*correlation is significant at the 0.05 level (two-tailed)

**correlation is significant at the 0.01 level (two-tailed)

Table 8. Pearson Correlation Coefficients for Peak Trunk and Knee Kinematics During a Side-Step Cutting Task

	Sagittal Plane Trunk Angle		Frontal Plane Trunk Angle		Transverse Plane Trunk Angle	
	r	p	r	p	r	p
Sagittal Plane Knee Angle	0.282	0.131	0.143	0.452	0.233	0.215
Frontal Plane Knee Angle	-0.139	0.460	0.218	0.247	-0.088	0.642
Transverse Plane Knee Angle	0.212	0.260	0.295	0.113	0.587**	0.001

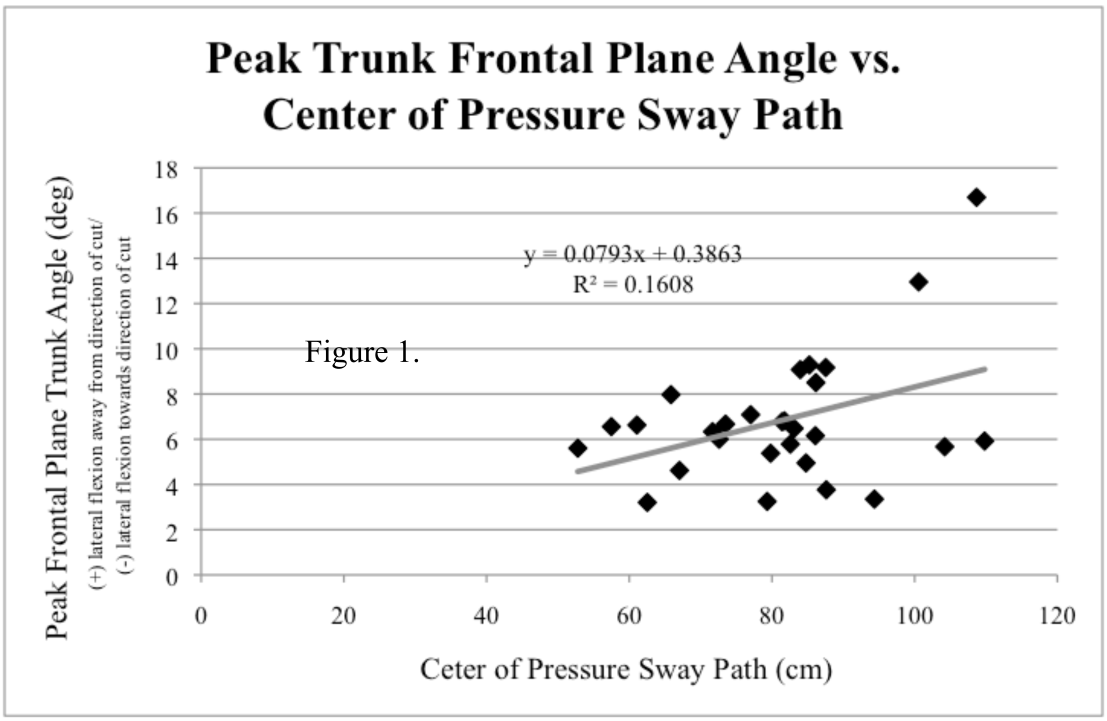
**correlation is significant at the 0.01 level (two-tailed)

Table 9. Pearson Correlation Coefficients for Peak Trunk Kinematics and Peak Internal Knee Kinetics (moments) During a Side-Step Cutting Task

	Sagittal Plane Trunk Angle		Frontal Plane Trunk Angle		Transverse Plane Trunk Angle	
	r	p	r	p	r	p
Knee Varus Moment	0.094	0.620	0.268	0.152	0.371*	0.043
Knee Extension Moment	-0.020	0.918	-0.065	0.734	-0.100	0.600

*correlation is significant at the 0.05 level (two-tailed)

Figure 1.



Peak Trunk Frontal Plane Angle vs.
Center of Pressure Sway Path
Peak Frontal Plane Trunk Angle (deg)
(+) lateral flexion away from direction of cut/
(-) lateral flexion towards direction of cut
y = 0.0793x + 0.3863
R² = 0.1608
Figure 1.
0
2
4
6
8
10
12
14
16
18
0
20
40
60
80
100
120
Ceter of Pressure Sway Path (cm)

Figure 2.

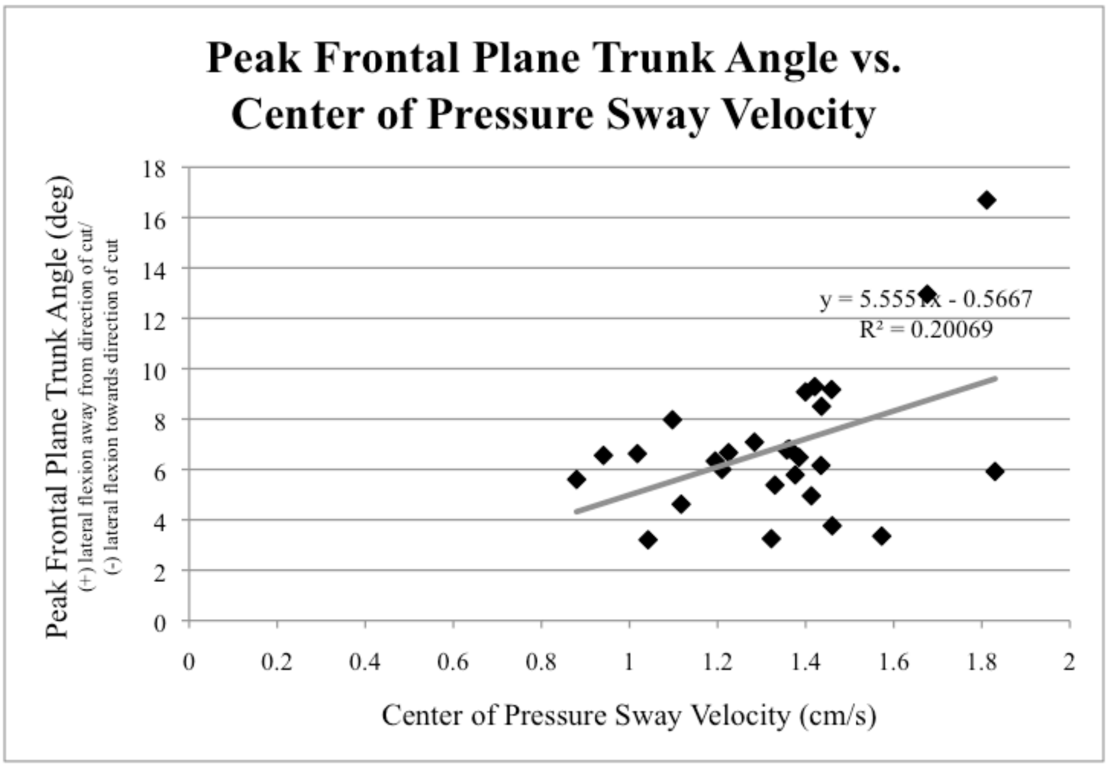
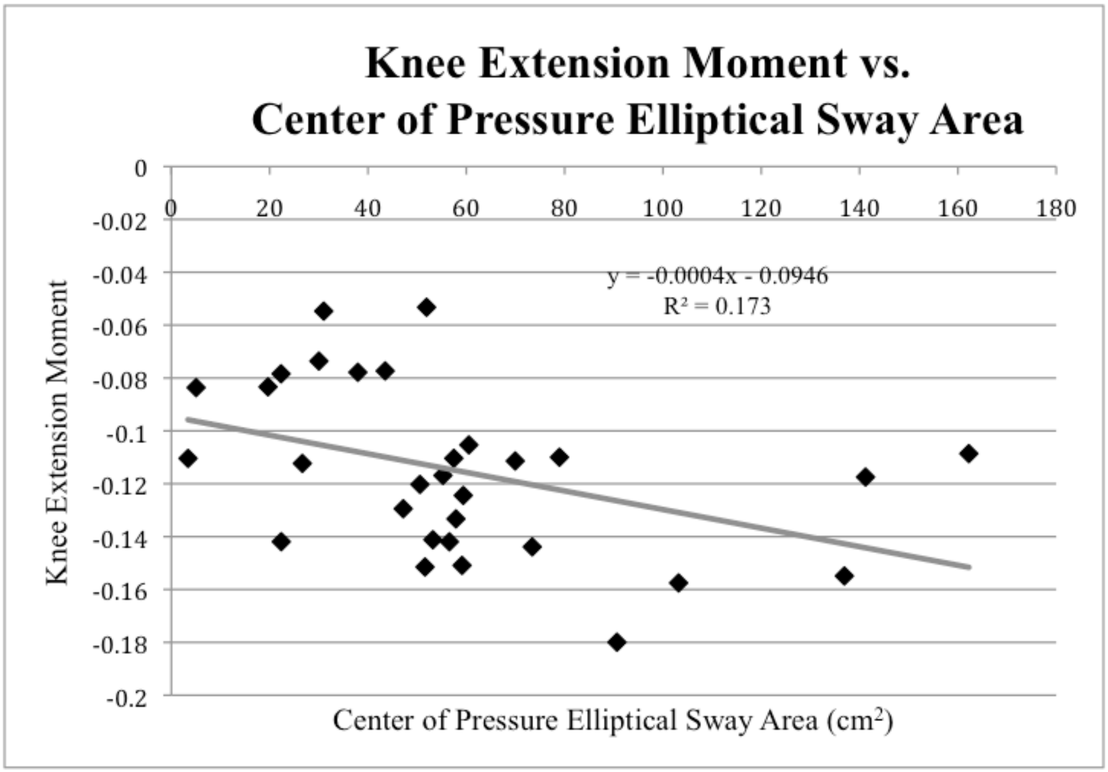

Peak Frontal Plane Trunk Angle vs. Center of Pressure Sway Velocity
Peak Frontal Plane Trunk Angle (deg)
(+) lateral flexion away from direction of cut/
(-) lateral flexion towards direction of cut
y = 5.5551x - 0.5667
R² = 0.20069
Center of Pressure Sway Velocity (cm/s)
0
2
4
6
8
10
12
14
16
18
0
0.2
0.4
0.6
0.8
1
1.2
1.4
1.6
1.8
2

Figure 3.



Knee Extension Moment vs.
Center of Pressure Elliptical Sway Area
Knee Extension Moment
Center of Pressure Elliptical Sway Area (cm2)
y = -0.0004x - 0.0946
R² = 0.173
0
-0.02
-0.04
-0.06
-0.08
-0.1
-0.12
-0.14
-0.16
-0.18
-0.2
0
20
40
60
80
100
120
140
160
180

Figure 4.

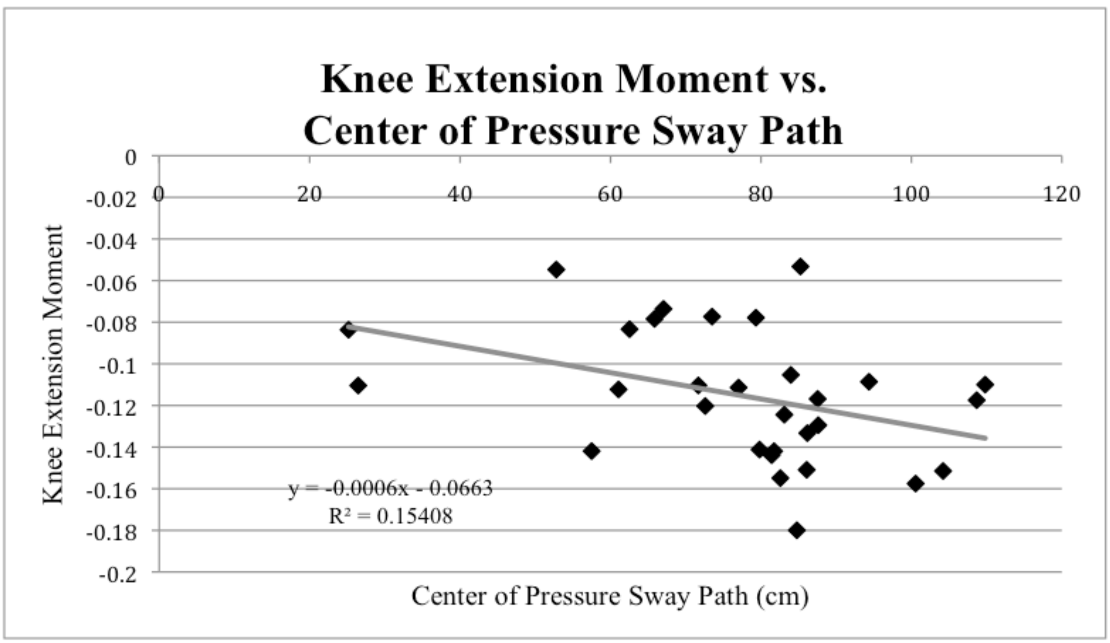
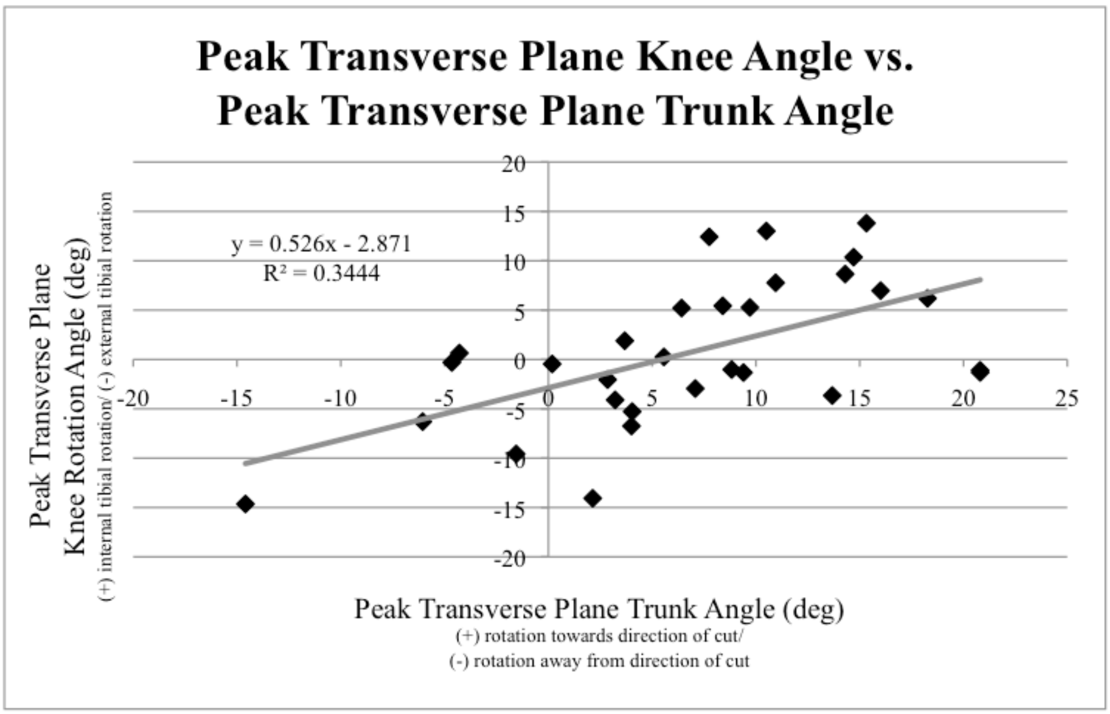
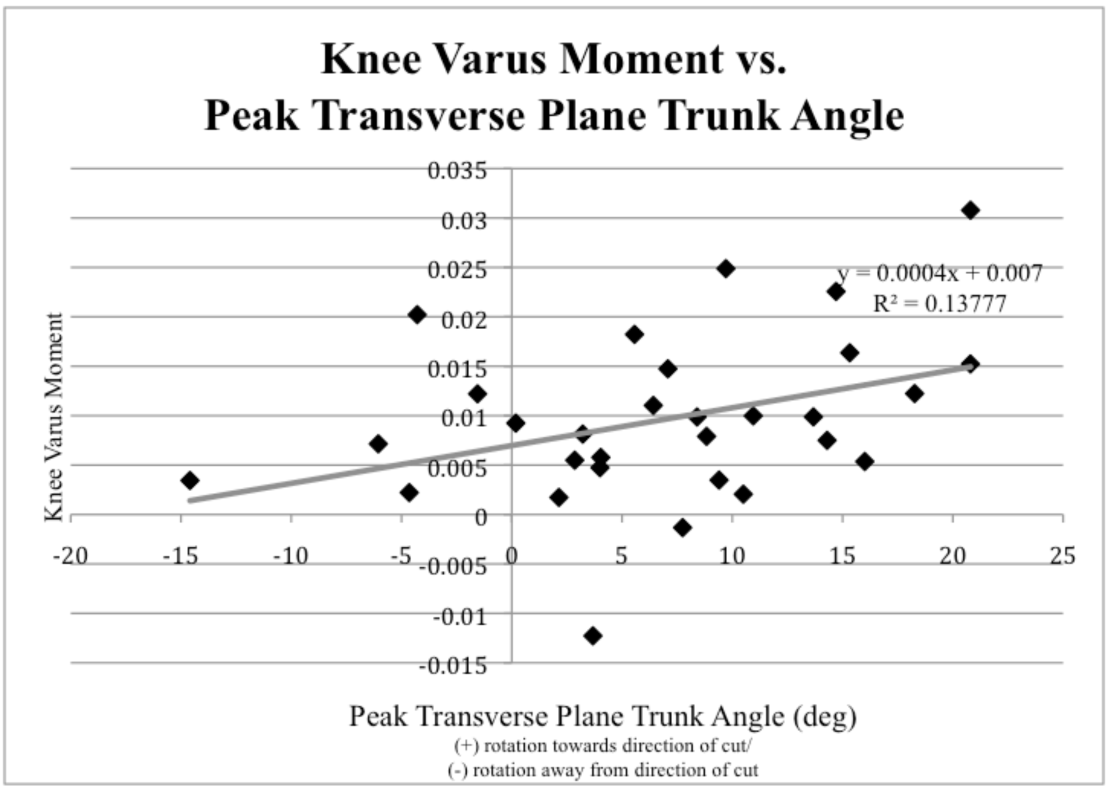

Knee Extension Moment vs.
Center of Pressure Sway Path
Knee Extension Moment
0
-0.02
-0.04
-0.06
-0.08
-0.1
-0.12
-0.14
-0.16
-0.18
-0.2
0
20
40
60
80
100
120
y = -0.0006x - 0.0663
R² = 0.15408
Center of Pressure Sway Path (cm)

Figure 5.



Peak Transverse Plane Knee Angle vs.
Peak Transverse Plane Trunk Angle
y = 0.526x - 2.871
R² = 0.3444
Peak Transverse Plane
Knee Rotation Angle (deg)
(+) internal tibial rotation/ (-) external tibial rotation
Peak Transverse Plane Trunk Angle (deg)
(+) rotation towards direction of cut/
(-) rotation away from direction of cut
20
15
10
5
0
-5
-10
-15
-20
-20
-15
-10
-5
0
5
10
15
20
25

Figure 6.



Knee Varus Moment vs.
Peak Transverse Plane Trunk Angle
Knee Varus Moment
0.035
0.03
0.025
0.02
0.015
0.01
0.005
0
-0.005
-0.01
-0.015
-20
-15
-10
-5
0
5
10
15
20
25
y = 0.0004x + 0.007
R² = 0.13777
Peak Transverse Plane Trunk Angle (deg)
(+) rotation towards direction of cut/
(-) rotation away from direction of cut

APPENDIX A

University of North Carolina-Chapel Hill
Consent to Participate in a Research Study
Adult Subjects

IRB Study # __09-2204__

Consent Form Version Date: 7/10/2009

Title of Study: The correlation between dynamic trunk stability, clinical trunk stability measures, and lower extremity and trunk biomechanics during a cutting task
Principal Investigators: Barnett S. Frank BS, ATC & Taryn M. Nicoletta BS, ATC
UNC-Chapel Hill Department: Exercise and Sport Science
UNC-Chapel Hill Phone number: (919)-962-7187
Email Address: *bsfrank@email.unc.edu & tmnicole@email.unc.edu*
Co-Investigators: Darin A. Padua, PhD, ATC, J. Troy Blackburn, PhD, ATC, Steven M. Zinder, PhD, ATC David R. Bell MEd, ATC, Benjamin M. Goerger, MS, ATC, & Marc F. Norcross, MA, ATC
Faculty Advisor: Darin A. Padua PhD, ATC
Study Contact telephone number: (203)-512-4235 – Barnett Frank &
(603)-662-7585 – Taryn Nicoletta
Study Contact email: bsfrank@email.unc.edu & tmnicole@email.unc.edu

---

<u>What are some general things you should know about research studies?</u>
You are being asked to take part in a research study. To join the study is voluntary. You may refuse to join, or you may withdraw your consent to be in the study, for any reason.

Research studies are designed to obtain new knowledge that may help other people in the future. You may not receive any direct benefit from being in the research study. There also may be risks to being in research studies.

Deciding not to be in the study or leaving the study before it is done will not affect your relationship with the researcher, your health care provider, or the University of North Carolina-Chapel Hill. If you are a patient with an illness, you do not have to be in the research study in order to receive health care.

Details about this study are discussed below. It is important that you understand this information so that you can make an informed choice about being in this research study. You will be given a copy of this consent form. You should ask the researchers named above, or staff members who may assist them, any questions you have about this study at any time.

What is the purpose of this study?

Lower extremity and low back injuries are a common and disabling occurrence during athletic participation. Those unable to adequately control the motion of their trunk during athletic tasks may be at a greater risk for injury.

The first purpose of this study is to examine if a relationship between previously defined risk factors of knee injury are more prevalent in individuals with poor trunk stability, strength, and control.

The second purpose of this study is to examine if there is a relationship between a laboratory measure of trunk stability and 3 clinical measures of trunk stability, strength, and control.

You are being asked to be in the study because you are a healthy individual aged 18-35. You are currently free of any lower back and lower extremity injury, have no history of anterior cruciate ligament (ACL) injury or surgery, and no history of lower back or abdominal wall surgery.

Are there any reasons you should not be in this study?

You should not be in this study if you have (or have had) any of the following conditions:

- Any known lower extremity or abdominal wall injury within six months prior to testing that may affect your ability to perform the testing protocol. An injury in this case is defined as the presence of injury symptoms that kept you out of physical activity for more than three days
- Any prior history to the anterior cruciate ligament of either knee, or known deficiency of the anterior cruciate ligament in either knee
- A history of surgery in either lower extremity in the past year, or any history of knee surgery
- Any known medical condition that would prevent you from participating in physical activity
- Any neuromuscular condition that prevents the normal functioning of the musculature in the lower extremity, lower back, or abdominal wall
- Current signs and symptoms of low back pain or pathology
- Previous history of abdominal wall surgery

How many people will take part in this study?

If you decide to be in this study, you will be one of approximately 60 people in this research study.

How long will your part in this study last?

Your participation in this study will last approximately 90 minutes. You will only be tested once.

What will happen if you take part in the study?

You will come to the Sports Medicine Research Laboratory in the basement of Fetzer Gym for a single testing session. You will need to wear athletic shoes, 1 pair of socks, shorts falling above the knee joint, and a t-shirt (and a sports bra for women) so that you can comfortably perform the tests during this study. When you first come into the lab, you will be asked to complete a short survey about your exercise history and basic health history. You will also sign this consent form if you decide to participate in the study. You will then have your height and weight measured. You will perform a self-paced 5-minute warm-up on a stationary bicycle, followed by some time to stretch your legs and body.

When you have finished warming-up you will perform a series of tests on your lower body and trunk.

1. <u>Movement Assessment</u>: You will be asked to put on a pair of spandex shorts and a tank top. Reflective markers will be placed on your shoulders, hips, back, thighs, knees, lower legs, ankles, heels, and feet. Someone of the same sex will be present for your comfort during the placement of these markers. After the markers have been placed on your body, you will be asked to stand in the center of the room on two plates while 7 infrared video cameras record the position of the reflective markers and 2 video cameras record your position within the movement area. The infrared video cameras do not capture images; only the markers are visible on the computer screen, the 2 video cameras will record your whole body motion (your face will be recognizable, however, the video recordings of your movement assessment will be kept in the Sports Medicine Research Laboratory under lock and key, in which the investigators are the only individuals who have access to recordings). Once this standing trial has been completed, you will be asked to perform a sidestep cutting task. You will start be standing on a line that is set 50% of your body height from the front edge of a metal plate. A short hurdle (17cm, or 6 inches) will be placed halfway between you and the metal plate. Prior to each testing trial the investigator will instruct you to cut in either a right or a left direction. When the investigator says "go," you will jump over the hurdle. You will land on your dominant foot (the foot you would use to kick a soccer ball with) on one of the plates, and cut to the side previously indicated by the investigator. You will have 3-5 practice trials to make sure you feel comfortable performing the side-step cut. After the practice trials, you will perform several repetitions in both directions (until you have performed at total of 10 trials; 5 cuts to your non-dominant, and 5 cuts to your dominant side), with 30 seconds of rest between each repetition.

2. <u>Unstable Sitting (Clinical Trunk Stability and Control) Assessment</u>: You will be asked to sit atop the unstable sitting apparatus chair which is composed of a 30 cm plastic hemisphere and a footplate to rest your feet. The unstable sitting chair rests upon a forceplate that measures the location of the sitting apparatus throughout the testing period. The forceplate is embedded in a wooden box approximately one meter above the ground. The wooden box has handrails that are available for you to grasp to stabilize yourself when sitting in the device in between testing trials. Once you are sitting atop the apparatus your trunk and thorax will be secured in the apparatus with seatbelt-like safety harnesses. While the experimenter secures your upper body into the device, you are to grab onto the handrails on your left and right.

Once you are secure within the unstable sitting device you will undergo the testing procedure and are allowed 3 practice trials and 5 recorded trials. During each trial you will be asked to let go of the hand rails and cross your arms across your chest, placing each hand on the opposite shoulder, and to stabilize yourself and the unstable sitting apparatus. Once you feel that you are stable in the seat, you will inform the experimenter with the verbal cue "ready." Once you cue the experimenter, a 60-second trial recording the location of the sitting apparatus will begin. Once the 60-second trial has elapsed, the experimenter will verbally cue you with the word "done" and you may grasp the handrails on the sides of the wooden box. You will be given one-minute of rest between each of the 5 trials. Once five successful trials have been recorded the safety belts will be unstrapped and you will exit the apparatus.

3. Clinical Measures of Trunk Stability, Control, and Strength:
   You will then complete a series of 3 clinical trunk stability tests.

   The following tests will be administered in a randomized order following the abdominal hollowing task. These tests will be videotaped.

   - Human Arrow Test: The human arrow is an isometric (no motion) contraction performed on the floor. You will be instructed to push up off of the floor on to your elbows and toes with your shoulders and elbows bent to 90 degrees. You will be asked to maintain a straight posture, keeping the back flat with mild lordosis, the legs straight, and the shoulders retracted. You will hold this position until failure occurs, which includes increasing lumbar lordosis, increasing hip flexion, touching down with the knees or placing the hands flat to the floor. The trial will then be over. You will perform this task once.

   - Side Plank Test: The side plank test will be performed on the floor. You will be instructed to lay on your dominant side and to push up off the floor. You will be asked to hold this position with the legs extended, and the top foot placed in front of the lower foot for support. You will support yourself on one elbow and the feet while lifting your hips off the floor. The uninvolved arm is held across the chest with the hand placed on the opposite shoulder. You will hold this position until failure occurs, which includes increasing hip adduction or abduction more than 10 degrees from neutral, increasing hip flexion more than 10 degrees from neutral, or increasing lumbar lordosis. The trial will then be over. You will perform this task once.

   - Seated Ball Test: You will be asked to maintain the same upright posture as you did during the unstable sitting task while seated on a stability ball. The hips and knees will be bent to approximately 90 degrees. Arms will be crossed across the chest and you will be instructed to lift your feet off of the ground. You will be asked to maintain proper posture, including mild lumbar lordosis, neutral shoulders, head, and neck. You will maintain this position for 60 seconds and your errors will be counted. Errors for this test include touching one or both feet to the floor, uncrossing the arms, increasing lumbar lordosis or decreasing hip flexion more that 30 degrees (i.e. rolling back and forth on the ball). You will be allotted one practice trial for up to 30 s and then will be given one minute of rest. You will then perform 5, 60 second trials with one minute of rest between each trial.

What are the possible risks or discomforts involved with being in this study?
Participation in this study may involve the following risks:

Common: You may experience muscle soreness in your abdominal and lower back muscles one to two days after testing. This soreness may be a result of abnormal stress to your trunk musculature as the clinical measures of trunk stability and control target the muscles of the abdomen and the lower back. This soreness should be no greater than that as result of an abdominal and lower back exercise routine.

Uncommon: There is a minor risk of lower extremity joint injury that may be incurred during the cutting task. The task itself is inherently athletic in nature and requires a rapid change in forward motion. As the subject, you will be required to travel in a straight-line forward motion and then be cued to "cut" in a leftward or rightward direction. The quick change in direction may result in a minor loss of balance and a moment of instability in which you are at minor risk for injuring joint structures of the lower extremity. The risk of joint structure injury is no greater than that during participation in athletic tasks or physical activity in which rapid changes in direction of travel are required.

If at any point during the testing you feel abnormal or extreme discomfort, please inform the experimenter and the testing session will be stopped immediately.

In addition, there may be uncommon or previously unknown risks that might occur. You are encouraged to report any problems to the researchers.

What if we learn about new findings or information during the study?
You will be given any new information gained during the course of the study that might affect your willingness to continue your participation.

How will your privacy be protected?
No subjects will be identified in any report or publication about this study. Although every effort will be made to keep research records private, there may be times when

federal or state law requires the disclosure of such records, including personal information. This is very unlikely, but if disclosure is ever required, UNC-Chapel Hill will take steps allowable by law to protect the privacy of personal information. In some cases, your information in this research study could be reviewed by representatives of the University, research sponsors, or government agencies for purposes such as quality control or safety.

You will be assigned a subject ID number. All of the data collected during the study will use the subject ID for identification purposes. All data will be kept in a locked filing cabinet in a locked room in the Sports Medicine Research Laboratory for the duration of the study. The subject ID number 'key' (that is the only document that links your name to the data) will be securely stored in a locked file cabinet in a separate location from the data. Any data stored on the computer will be identified only by subject number and protected by password, which only the primary investigator and any co-investigators directly involved in data collection and reduction for the study will know. All paper documents will be destroyed once the data is in the computer.

What will happen if you are injured by this research?
All research involves a chance that something bad might happen to you. This may include the risk of personal injury. In spite of all safety measures, you might develop a reaction or injury from being in this study. If such problems occur, the researchers will help you get medical care, but any costs for the medical care will be billed to you and/or your insurance company. The University of North Carolina at Chapel Hill has not set aside funds to pay you for any such reactions or injuries, or for the related medical care. However, by signing this form, you do not give up any of your legal rights.

What if you want to stop before your part in the study is complete?
You are free to cease testing at any time during the physical testing. You can withdraw from this study at any time, without penalty. The investigators also have the right to stop your participation at any time. This could be because you have had an unexpected reaction, or have failed to follow instructions, or because the entire study has been stopped.

Will you receive anything for being in this study?
You will not receive any monetary compensation for this study.

Will it cost you anything to be in this study?
It will not cost you anything to be in this study. You are only responsible for your transportation to and from the testing site in Fetzer Gymnasium on the campus of the University of North Carolina at Chapel Hill. If you need a parking permit, a temporary permit will be given to you by the primary investigators.

What if you are a UNC student?
You may choose not to be in the study or to stop being in the study before it is over at any time. This will not affect your class standing or grades at UNC-Chapel Hill. You will not be offered or receive any special consideration if you take part in this research.

What if you are a UNC employee?
Taking part in this research is not a part of your University duties, and refusing will not affect your job. You will not be offered or receive any special job-related consideration if you take part in this research.

Who is sponsoring this study?
There is no sponsoring affiliation with this study.

What if you have questions about this study?
You have the right to ask, and have answered, any questions you may have about this research. If you have questions, or if a research-related injury occurs, you should contact the researchers listed on the first page of this form.

What if you have questions about your rights as a research subject?
All research on human volunteers is reviewed by a committee that works to protect your rights and welfare. If you have questions or concerns about your rights as a research subject you may contact, anonymously if you wish, the Institutional Review Board at 919-966-3113 or by email to IRB_subjects@unc.edu.

- -

Title of Study: The correlation between dynamic trunk stability, clinical trunk stability measures, and lower extremity and trunk biomechanics during a cutting task

Principal Investigators: Barnett S. Frank BS, ATC & Taryn M. Nicoletta BS, ATC

Subject's Agreement:
I have read the information provided above. I have asked all the questions I have at this time. I voluntarily agree to participate in this research study.

______________________________________ __________________
Signature of Research Subject Date

____
Printed Name of Research Subject

______________________________________ __________________
Signature of Person Obtaining Consent Date

____
Printed Name of Person Obtaining Consent

APPENDIX B

Movement Assessment Study

Department of Exercise and Sport Science

Research subjects needed to participate in a study analyzing core strength and knee injury.

Participants should be:

- 18 – 35 years of age
- In good physical health

If you participate in this study you will:

- Participate in one testing session lasting approximately 45 minutes
- Complete two tests measuring core strength and stability
- Complete one test analyzing 3D motion during an athletic cutting task

You should not participate in this study if:

- You have a prior history of anterior cruciate ligament (ACL) injury
- You have a history of lower back surgery
- You have had a lower extremity or low-back injury within the past 6 months

Contact Barnett Frank or Taryn Nicoletta if you are interested or for more information.

Barnett:
(203)-512-4235 or bsfrank@email.unc.edu

Taryn:
(603)-662-7585 or tmnicole@email.unc.edu

Barnett Frank
(203)-512-4235
bsfrank@email.unc.edu
Taryn Nicoletta
(603)-662-7585
tmnicole@email.unc.ed

Barnett Frank
(203)-512-4235
bsfrank@email.unc.edu
Taryn Nicoletta
(603)-662-7585
tmnicole@email.unc.ed

Barnett Frank
(203)-512-4235
bsfrank@email.unc.edu
Taryn Nicoletta
(603)-662-7585
tmnicole@email.unc.ed

Barnett Frank
(203)-512-4235
bsfrank@email.unc.edu
Taryn Nicoletta
(603)-662-7585
tmnicole@email.unc.ed

Barnett Frank
(203)-512-4235
bsfrank@email.unc.edu
Taryn Nicoletta
(603)-662-7585
tmnicole@email.unc.ed

Barnett Frank
(203)-512-4235
bsfrank@email.unc.edu
Taryn Nicoletta
(603)-662-7585
tmnicole@email.unc.ed

Barnett Frank
(203)-512-4235
bsfrank@email.unc.edu
Taryn Nicoletta
(603)-662-7585
tmnicole@email.unc.ed

APPENDIX C

Subject: Movement Assessment Study

Volunteers needed to participate in a research study analyzing core strength and it's relationship to knee injury. This study is sponsored by The Department of Exercise and Sport Science.

Participants should be:

- 18 – 35 years of age
- In good physical health

If you participate in this study you will:

- Participate in one testing session lasting approximately 45 minutes
- Complete two tests measuring core strength and stability
- Complete one test analyzing 3D motion during an athletic cutting task

You should not participate in this study if:

- You have a prior history of anterior cruciate ligament (ACL) injury
- You have a history of lower back surgery
- You have had a lower extremity or low-back injury within the past 6 months

Please contact Barnett Frank and/or Taryn Nicoletta if you are interested in volunteering, or would like more information about the study.

Barnett Frank
Department of Exercise & Sport Science

bsfrank@email.unc.edu
(203)-512-4235

Taryn Nicoletta
Department of Exercise & Sport Science
tmnicole@email.unc.edu
(603)-662-7585

This study had been approved by the IRB #: 09-2204

APPENDIX D

HEALTH AND ACTIVITY QUESTIONNAIRE

The correlation between dynamic trunk stability, clinical trunk stability measures, and lower extremity and trunk biomechanics during an unanticipated cutting task

PART 1: Demographics

1. What is your age?	________	
2. What is your gender?	Male	Female
3. If you were going to kick a ball for maximum distance, which leg would you use?	RIGHT	LEFT

PART 2: Health History

4. Are you currently in good health?	YES	NO
5. Have you ever been diagnosed with any cardiac condition (such as tachycardia, bradycardia, fibrillation, heart murmur, etc.)?	YES	NO
5a. If YES, please explain:		
6. Have you ever been diagnosed with any neurologic condition (such as brain injury, spinal cord injury, Parkinson's disease, multiple sclerosis, epilepsy, etc.) that affects the way your muscles work?	YES	NO
6a. If YES, please explain:		
7. Have you ever had asthma?	YES	NO
7a. If YES, do you still take medications?	YES	NO
8. Do you have diabetes?	YES	NO
9. Have you ever been diagnosed with high blood pressure (greater than 140/90)?	YES	NO
10. Have you ever experienced heat stroke or heat exhaustion?	YES	NO
10a. If YES, when?		
11. Have you ever needed hospitalization for a non-surgical reason?	YES	NO
11a. If YES, why were you hospitalized?		
11b. Is YES, when were you hospitalized?		
12. Have you ever had surgery?	YES	NO
12a. If YES, when did you have surgery?		
12b. If YES, what surgical procedure was performed?		

Physical Activity Readiness Questionnaire (PAR-Q)

(Representative of the American College of Sports Medicine standards)

YES	NO	
___	___	1. Has a doctor ever said that you have a heart condition and recommended only medically supervised activity?
___	___	2. Do you have chest pain brought on by physical activity?
___	___	3. Have you developed chest pain in the past month?
___	___	4. Have you on one or more occasions lost consciousness or fallen over as a result of dizziness?
___	___	5. Do you have a bone or joint problem that could be aggravated by the proposed physical activity?
___	___	6. Has a doctor ever recommended medication for your blood pressure or a heart condition?
___	___	7. Are you aware, through your own experience or a doctor's advice, of any other physical reason that would prohibit you from exercising without medical supervision?

If you answer "yes" to any of these questions, call your personal physician or healthcare provider before increasing your physical activity.

APPENDIX E

SUB ID:			HT:	
DATE:			WT:	
Unstable Sitting				

Trial #	***Comments***
1	
2	
3	
4	
5	

Cutting Task				

Trial #	***Comments***
1 ______	
2 ______	
3 ______	
4 ______	
5 ______	
6 ______	
7 ______	
8 ______	
9 ______	
10 ______	

APPENDIX F

The Correlation Between Dynamic Trunk Stability and Lower Extremity and Trunk Biomechanics During a Cutting Task

Background: Trunk position and postural control influence lower extremity biomechanics during athletic tasks. However, the relationships between trunk stability, trunk motion, and lower extremity biomechanics during a cutting task have not been examined. **Hypothesis/Purpose:** To examine the relationship between a laboratory measure of trunk neuromuscular control (NMC) and trunk and knee biomechanics during an athletic cutting task. We hypothesized that decreased NMC would be associated with a biomechanical profile during the cutting task consistent with greater anterior cruciate ligament (ACL) loading and injury risk. **Design:** Descriptive Laboratory Study. **Methods:** 30 healthy, recreationally active individuals performed an unstable-sitting task and an athletic cutting task. Center of pressure (CoP) sway data were collected during the unstable-sitting task. Peak triplanar trunk and knee angles, and peak internal knee extension and varus moments were calculated during the first 50% of the stance phase during the cutting task. Pearson product-moment correlation coefficients were calculated among the CoP sway data, trunk and knee kinematics, and knee kinetics. **Results:** Greater CoP sway area ($r = -0.548$, $p = 0.006$) and CoP sway path ($r = -0.407$, $p = 0.032$) were associated with greater knee extension moment (negative angular convention). Greater transverse plane trunk angle was associated with greater knee varus moment (r =

0.371, p = 0.043) and Greater transverse plane trunk angle was associated with greater transverse plane trunk angle (r = 0.587, p = 0.001). No other correlations between trunk stability, trunk motion, and knee biomechanics were observed. **Conclusion:** The results of this study support a link between trunk stability and lower extremity biomechanical risk factors for ACL injury. Improved neuromuscular control (NMC) of the trunk may decrease the presence of lower extremity risk factors for ACL injury.

Key Words: ACL Injury, Trunk Stability, Core, ACL Injury Risk Factors

INTRODUCTION

Mechanisms of non-contact ACL injury have been observed during athletic activities requiring rapid decelerations and changes in direction, such as drop-landing and cutting maneuvers.[1, 5, 11] In-vitro cadaveric studies have indicated that combined external knee valgus and internal rotation forces applied at shallow flexion angles increase ACL strain compared to strain values when these forces are applied independently.[14, 24] Athletic motions are multi-planar in nature and have the ability impart forces on the ACL in multiple planes. When loads are not met with adequate resistance from the body's stabilizing musculature, the body may assume dangerous posturing, placing excessive strain on the anatomy, thus tissue injury may result.[1, 25]

Decreased trunk flexion during athletic tasks increases forces associated with ACL loading[4]. Individuals with decreased trunk NMC have been observed to injure their ACL's at a greater rate than those with greater trunk control.[26, 27] Furthermore, individuals who perform athletic cutting tasks with less trunk flexion and greater lateral trunk flexion over the stance limb display biomechanical profiles associated with greater ACL loading and a greater risk of ACL injury.[11] The trunk is the segment of the body in which the body's center of mass (CoM) most commonly exists.[11, 15, 18] Poor control of the body's CoM over the lower extremity may place excessive loads over the lower extremity due to the combined accelerations required during rapid changes in direction of travel and gravity. The specific relationship between NMC of the trunk and its influence on ACL injury risk factors is not known at this time. The purpose our study was to examine potential relationships between a laboratory measure of trunk NMC during unstable-sitting, and proposed trunk and knee biomechanical risk factors of ACL injury during an athletic cutting task, a described mechanism of ACL injury.

MATERIALS AND METHODS

We used a correlational design to investigate relationships between trunk NMC and biomechanical factors associated with non-contact ACL injury during a cutting task. We used simple correlation analyses to evaluate 1) relationships between trunk NMC and trunk motion during a cutting task and 2) relationships between trunk motion and knee kinematics and kinetics during a side-step cutting task.

Subjects

Thirty subjects from the university setting volunteered for this study (age range 18-35). Subjects were eligible for participation if they were recreationally active, participating in physical activity for at least twenty minutes at least three times per week. Subjects were excluded from the study if they could not perform the tasks, had a history of ACL injury, had lower extremity surgery in the past year, had a low back injury or pain at the time of the study, had previously participated in an ACL injury prevention program, or had suffered any known lower extremity injury within the past six months prior to testing.[3] An injury was defined as a traumatic event or presence of symptoms that restricted activity for more than three days. Prior to data collection, subjects read and signed an informed consent form approved by the institutional review board.

Procedures

Subjects reported to the laboratory for a single testing session that involved two separate assessments: 1) trunk NMC on an unstable-sitting surface and 2) a side-step cutting maneuver. The cutting task was always performed first to avoid any influence that fatigue of the trunk musculature following the unstable-sitting task may have exerted on trunk and lower extremity and biomechanics. Subjects completed a health history and physical activity readiness questionnaire (PAR-Q) to ensure compliance with the study's inclusion criteria, and to record demographic data (age, level of activity, previous athletic experience, and previous injury history). All data were collected from the dominant leg, defined as the leg used to kick a soccer ball for maximum distance. Subjects were fitted with non-reflective black spandex clothing and warmed-up on a stationary bike for 5 minutes at a level of 11 RPE and completed a static lower-extremity stretching program.

A set of 25 reflective markers was then placed on the left/right acromion process, L4-L5 joint space, right/left anterior superior iliac spine, right/left greater trochanter, right/left anterior thigh, right/left lateral epicondyle, right/left anterior shank, right/left lateral malleolus, right/left calcaneous, right/left 1st metatarsal head, and right/left 5th metatarsal head using double-sided tape. A seven-camera infrared motion capture system

(Vicon MX, Vicon Motion Systems, Los Angeles, California) interfaced with a force plate (Bertec Corporation, Columbus, OH) was used to sample to the 3-dimensional positions of these markers at 150 Hz and ground reaction forces at 1,500 Hz using Vicon Nexus 1.4.116 motion capture software. World and segmental axis systems were established by a right hand three-dimensional Cartesian coordinate system. The positive X-axis was designated forward/anterior direction, the positive Y-axis leftward/medially, and the positive Z-axis upward/superiorly [3].

A static trial was collected with the subject facing the positive X-axis of the world coordinate system. Hip joint centers were calculated using the Bell method.[2] Knee and ankle joint centers were calculated as the midpoint between the medial and lateral femoral epicondyles and malleoli, respectively. The trunk center was defined by the L4-L5 joint interspace marker. Markers were then removed from the medial epicondyles and malleoli after the static trial.

Cutting Task

The cutting task was explained and demonstrated by the principal investigator, subjects were allowed to practice the task a maximum of three times. After the practice trials were completed, subjects performed 10 trials of the task, alternating cutting direction, with an equal number of cuts in each direction, left (5) and right (5). The subject was given a 45 second rest period between each trial.

The cutting task began with subjects standing in the capture volume a distance of 50% body height from the force plate. They then performed a double-leg jump over a 17cm hurdle, landed with the dominant foot on the force plate, and cut to the contralateral side at an angle of 610 degrees.

Unstable-Sitting Task

A plyometric box with dimensions 61 cm x 61 cm x 45.72 cm with a conductive force plate (Bertec Corporation, Columbus, OH) supported a wooden seat atop a 30 cm polycarbonate resin hemisphere. The wooden seat included an adjustable foot rest to ensure the subjects' hips and knees were positioned at 90º of flexion, with their ankle joints at neutral (0º).[6] The plyometric box was fitted with handrails, allowing subjects to steady themselves when initially sitting on the unstable seat. The conductive force plate (Bertec Corporation, Columbus, OH) sampled vertical ground reaction force and moments about the X-, Y-, and Z-axes at 100 Hz, with a low-pass 20 Hz Butterworth filter. Kinetic data was used to calculate CoP location during data reduction.[6, 20-23]

The task required the subject to maintain stability on the sitting apparatus without aid. The subject maintained stability with the handgrips. While the subject maintained stability with assistance from the handgrips, the researcher helped to position the subject such that their hips and knees were at 90º of flexion, and their ankles were at neutral, or 0º flexion. The subject was instructed to keep the medial surfaces of their knee joints "touching" throughout each trial. The subject was told to continue holding onto the handgrips until the researcher cued them to "let go" to begin the stability task. The subject was instructed to cross their arms across their chest. Once the subject established equilibrium, they verbally cued the researcher when they were "ready." The researcher began data collection of CoP location once they received the "ready" cue from the subject. Kinetic data was sampled for a one-minute trial. If the subject could not complete the one-minute trial without touching down or having to stabilize themselves with the handrails, the trial was discarded. One-minute of rest was given between each trial, and five trials were recorded. The unstable-sitting task was re-explained to the subject before the trials. The subjects were allowed to practice the unstable-sitting task a maximum of three times.

Data Reduction and Analysis

All kinematic and kinetic data were imported into The Motion Monitor version 8 software system (Innovative Sports Training, Inc., Chicago, IL) for data reduction. Joint angles were calculated as motion of the distal segment relative to the proximal segment using a Y(sagittal), X(frontal), Z(transverse) Euler angle rotation sequence. Kinematic and kinetic data were filtered using a low-pass Butterworth digital filter, at a cutoff frequency of 15 Hz. Inverse dynamics procedures were used to calculate internal knee varus and extension moments during the cutting task.

Data reduction was completed using customized Matlab software (v 7.0, The Mathworks, Inc., Natick, MA). Peak sagittal, frontal, and transverse plane knee and trunk angles were calculated during the initial 50% of the stance phase during the cutting task. The stance phase was defnied as the interval between initial ground contact (vertical ground reaction > 10N) and toe-off (vertical ground reaction < 10N).

CoP elliptical sway area, sway path, and sway velocity were calculated from CoP location data. CoP sway area was calculated as elliptical area in square centimeters encompassing 95% of the CoP location points during the one-minute trial. CoP sway path was calculated as the total displacement path in centimeters of the CoP during the one-minute trial. CoP sway velocity was calculated as CoP displacement divided by time (cm/s). The mean CoP sway area, sway path, and sway velocity for each subject was calculated from the three middle trials (2, 3, 4) if all 5 trials had no episodes of touching down. If there was an episode of touching down in the middle three trials, trial 1 or 5 replaced one of the middle trials. If two middle trials had episodes of touching down, trials 1 and 5 replaced two of the middle trials.

Triplanar trunk and lower extremity kinematics, and internal knee extension and varus moments were calculated as peak angles and moments observed during the first 50% of the stance time during the side-step cutting task. Internal knee moments were calculated as the quotient of each subject's absolute moments, divided by the product of the subject's mass (N) and height (m), calculated moments were unitless, and were represented as directional magnitudes. Average peak angles and moments were calculated across the 5 side-step cutting trials.

Statistical Analysis

Statistical significance was set a-priori at $p < 0.05$. 24 bivariate Pearson r correlations were run to examine the relationships between CoP sway measures (3) during unstable-sitting and the trunk (3) and lower extremity mechanics (5) during the side-step cutting task. 15 bivariate Pearson r correlations were run to examine the relationships between triplanar trunk kinematics (3) and lower extremity mechanics (5) during the cutting task. SPSS version 16.0 (SPSS INC, Chicago, Ill) was used for all statistical calculations.

RESULTS

Thirty subjects (age = 20.5 ± 2.3 years, height = 173.78 ± 9.28 centimeters, mass = 67.50 ± 11.39 kilograms), 15 males (age = 20.9 ± 2.7 years, height = 178.06 ± 8.23 centimeters, mass = 75.08 ± 8.94 kilograms), and 15 females (age = 20.1 ± 1.9 years, height = 196.49 ± 8.44 centimeters, mass = 59.92 ± 8.11 kilograms) were tested. Three female and three male subjects' center of pressure elliptical sway areas, and two female subjects' center of pressure sway paths, and three females subjects' sway velocities were not used in the statistical analysis due to equipment error. Means and standard deviations for CoP elliptical sway area, sway path, and sway velocity are presented in Table 1. Means and standard deviations for tri-planar trunk kinematics during the side-step cutting task are presented in Table 2. Means and standard deviations for tri-planar knee kinematics are presented in Table 3. Means and standard deviations for external knee valgus moment and internal knee flexion moment are presented in Table 4.

Center of Pressure and Trunk Kinematics Correlations

Correlation coefficients were computed among the three predictor variables of CoP elliptical sway area, sway path, sway velocity, and the three criterion variables of peak triplanar trunk angles during the first 50% of stance time during the side-step cutting task. The results of the correlational analyses presented in Table 5 show that two of the nine correlations were statistically significant, and were greater than or equal to 0.35. Significant positive relationships between the predictor variables of CoP sway path and velocity and the criterion variable of peak frontal plane trunk angle were observed. A moderate, positive, linear relationship ($r(26) = 0.401$, $p = 0.034$) was observed between

CoP sway path and peak frontal plane trunk angle during the first 50% of the stance time of the side-step cutting task. This relationship suggests that as CoP sway path increases, peak sagittal plane trunk angle during the first 50% of the stance time increases. A moderate, positive, linear relationship ($r(25) = 0.448$, $p = 0.019$) was observed between CoP sway velocity and peak frontal plane trunk angle during the first 50% of the stance time during the side-step cutting task. This relationship suggests that as CoP sway velocity increases, peak sagittal plane trunk angle during the first 50% of the stance time increases.

Center of Pressure and Knee Kinematics Correlations

Correlation coefficients were computed among the three predictor variables of CoP elliptical sway area, sway path, sway velocity, and the three criterion variables of peak triplanar knee angles during the first 50% of the stance time during the side-step cutting task. The results of the correlational analyses presented in Table 6 show that none of the nine correlations were statistically significant at the $p < 0.05$ level, and were not greater than or equal to 0.35.

Center of Pressure and Knee Kinetics Correlations

Correlation coefficients were computed among the three predictor variables of CoP elliptical sway area, sway path, sway velocity, and the two criterion variables of peak internal knee varus moment and peak internal knee extension moment during the first 50% of the stance time during the side-step cutting task. The results of the correlational analyses presented in Table 7 show that two of the six correlations were statistically significant. Significant positive relationships between the predictor variables of CoP elliptical sway area and sway path, and the criterion variable of peak internal knee extension moment were observed. A strong, negative, linear relationship ($r(22) = -0.548$, $p = 0.006$) was observed between CoP elliptical sway area and peak internal knee extension moment during the first 50% of the stance time of the side-step cutting task. This relationship suggests that as CoP elliptical sway area increases, peak internal knee extension moment magnitude during the first 50% of the stance time increases. A moderate, negative, linear relationship ($r(26) = -0.407$, $p = 0.006$) was observed between CoP sway path and peak internal knee extension moment during the first 50% of the stance time of the side-step cutting task. This relationship suggests that as CoP sway path increases, peak internal knee extension moment magnitude during the first 50% of the stance time increases.

Trunk Kinematics and Knee Kinematics During the Side-Step Cutting Task Correlations

Correlation coefficients were computed among the three predictor variables of peak triplanar trunk angles during the first 50% of the stance time, and the three criterion variables of peak triplanar knee angles during the first 50% of the stance time during the side-step cutting task. The results of the correlational analyses presented in Table 8 show that one of the nine correlations was statistically significant, and greater than or equal to 0.35. A significant positive relationship between the predictor variable of peak transverse plane trunk angle and peak tibial internal rotation angle was observed. A strong, positive, linear relationship ($r(28) = 0.586$, $p = 0.001$) was observed between peak transverse plane trunk angle and peak tibial internal rotation angle during the first 50% of the stance time of the side-step cutting task. This relationship suggests that as peak transverse plane trunk angle during the first 50% of the stance time increases, peak tibial internal rotation angle during the first 50% of the stance time increases.

Trunk Kinematics and Knee Kinetics Correlations

Correlation coefficients were computed among the three predictor variables of peak triplanar trunk kinematics during the first 50% of the stance time, and the two criterion variables of peak internal knee extension moment and peak internal knee varus moment during the first 50% of the stance time during the side-step cutting task. The results of the correlational analyses presented in Table 9 show that one of the nine correlations was statistically significant, and greater than or equal to 0.35. A significant, positive, relationship between the predictor variable of peak transverse plane trunk angle and peak internal knee varus moment was observed. A moderate, positive, linear relationship ($r(28) = 0.371$, $p = 0.043$) was observed between peak transverse plane trunk angle and peak internal knee varus moment during the first 50% of the stance time of the side-step cutting task. This relationship suggests that as peak transverse plane trunk angle during the first 50% of the stance time increases, peak internal knee varus moment during the first 50% of the stance time increases.

DISCUSSION

The primary purpose of this study was to examine the relationships between measures of trunk stability during an unstable-sitting task, and trunk and knee biomechanics during a side-step cutting task. A secondary purpose was to examine the relationships between trunk and knee biomechanics during the same task. The results indicate that lesser trunk stability is associated with greater peak frontal plane trunk angle and peak knee extension moment during the initial 50% of stance time. Further, greater peak trunk rotation angle away from the stance leg during the first 50% of stance time during side-step cutting is associated with greater peak knee varus moment and peak tibial internal rotation angle during the first 50% of stance time during side-step cutting. To our knowledge, our findings are the first to demonstrate significant relationships between NMC of the trunk, trunk mechanics, and knee mechanics during an athletic task.

The primary finding of our study is the negative correlation between trunk stability and knee extension moment. Specifically, greater CoP elliptical sway area and sway path during an unstable-sitting task is associated with greater peak knee extension moment during side-step cutting. Previous research has demonstrated that an increase in trunk flexion angle decreases vertical ground reaction force, increases knee flexion angle, and decreases knee extension moment during the loading time in drop-landing tasks.[4] Landing in a more erect posture places the body's CoM mass in a more posterior position relative to the center of rotation of the knee joint, compared to landing with an increased trunk flexion angle. The increase in CoM posterior displacement increases the moment arm of the body's CoM mass relative to the knee joint. The greater moment arm results in a greater flexion moment about the knee, thus the quadriceps musculature is required to exert a greater magnitude internal extension moment about the knee.[4] Combined with an increase in knee extension moment and a decrease in knee flexion angle, there is a potential to increase ACL loading via ATSF.[4, 7, 16] Theoretically, increased CoP elliptical sway area and sway path indicate decreased trunk control, thus, increasing knee extension moment.

The rationale for the link between decreased trunk control and increased knee extension moment is supported by previous findings of the body's CoM influence on lower extremity biomechanics.[3, 4] The ability of subjects to actively increase their trunk flexion angle during the loading time with provided augmented feedback, indicates trunk NMC specific to CoM position during the loading time of drop-landing tasks is modifiable in the sagittal plane. An inability to control the body's CoM during athletic tasks may manifest in individuals who position themselves in a more erect posture. Our study's findings direct future research to examine the effects of the trunk stability element of trunk NMC on trunk position and internal knee extension moment during tasks that have been proposed to be mechanisms of ACL injury. Future research methodology would require interventions that are specific to trunk stability to determine the effect of the stability component of trunk NMC on trunk position and internal knee extension moment.

Another important finding of this study is the association between trunk stability and frontal plane trunk angle during the side-step cutting task. Greater magnitudes of CoP sway area and sway velocity are associated with greater lateral trunk flexion angles towards the stance leg. In theory, increased lateral trunk flexion is also indicative of decreased trunk control. It is interesting to note that in a prospective study of Division I collegiate athletes, increased lateral trunk flexion after sudden force release was a significant predictor of subsequent knee injury in females.[26, 27] While the methodology between our studies differs, it is apparent that the same kinematic pattern of increased lateral trunk flexion is observed during a cutting task in individuals demonstrating lesser trunk stability, and during a dynamic perturbation in individuals who subsequently suffer a knee injury. This suggests that there are important links between trunk NMC, lower extremity mechanics, and injury. However, further research is needed to elucidate the specific mechanisms underlying these observations.

The next significant correlation of our study was the positive relationship between peak transverse plane trunk angle and knee varus moment during side-step cutting. This indicates that greater amounts of trunk rotation (away from the stance leg) are associated with greater knee varus moments. Internal knee varus moment is similar to external knee valgus moment, which is a prospective risk factor for non-contact ACL injury[10]. Our findings agree with Houck et al,[12] who observed a positive relationship between frontal plane peak trunk angle and hip abduction moment during side-step cutting. Houck et al's findings differ from our study in that they are specific to the frontal plane. We found relationships between two planes, trunk motion in the transverse plane, and knee kinetics in the frontal plane.

Increased frontal plane trunk angle drives hip abduction moment, a frontal plane torque proposed to act as a stabilizing force for the lower extremity.[12] The abduction moment prevents excessive hip adduction angle from occurring as a result of increased lateral CoM positioning at the trunk (lateral flexion).[12] Previous research is in agreement with our study, presenting an association between trunk motion and lower extremity kinetics, yet previous findings have only indicated intra-planar relationships,[9, 12] versus the our study's findings, implicating a relationship between trunk kinematics in the transverse plane, and knee kinetics in the frontal plane. Although our findings differ from

Houck et al's,[12] previous in-vitro cadaveric research has implicated external knee valgus moment to increase ACL strain during a simulated jump-landing task and combined external loading conditions of the knee.[17, 24] The findings of our study demonstrate increased trunk rotation away from the stance leg increases internal knee varus moment, an identified risk factor of ACL injury.[10] Our study provides a direct link between transverse plane trunk motion and knee injury risk.[10]

It is possible the variation between our results regarding inter-planar relationships and previous findings specific to intra-planar relationships may be a function of task specificity. Previous literature has examined slower unanticipated cutting tasks during walking gait,[12] versus anticipated, higher speed cutting tasks as in our methodology. It is possible an increase in transverse plane trunk motion occurs during higher speed anticipated cutting tasks versus unanticipated changes of direction during walking. Our study provides evidence of a positive relationship between trunk rotation angle and knee varus moment, a risk factor of ACL injury.[1, 8, 10]

The final significant relationship we observed was the positive correlation between trunk rotation and tibial rotation during the side-step cutting task. In a cadaveric study, Markolf et al[17] demonstrated addition of internal tibial torque induced the greatest amount of ACL loading when combined with an anterior tibial force, compared to knee valgus/anterior tibial force application at various knee flexion angles. Greater amounts of trunk rotation (away from the stance leg, towards the direction of the cut) were associated with greater amounts of tibial internal rotation, a proposed risk factor of ACL injury.[1, 13, 17] This result suggests transverse plane moments causing rotation superiorly at the trunk may be transferred through the lower extremity kinetic chain, manifest inferiorly at the knee in tibial rotation. This notion is supported by previous research which indicates trunk motion precedes all other body segment motion during locomotor tasks requiring a change in direction of travel.[19] It is proposed that the trunk motion precedes lower extremity motion in order to accelerate the body's CoM in the new direction of travel.[12, 19] Furthermore, Imwalle et al[13] also observed this concept of segment kinematic symmetry, in which there was a positive association between transverse plane knee and hip internal rotation in female soccer players. While the finding of our study is limited in that it is focused on the relationship between transverse plane trunk and knee kinematics without evaluating the kinematics of the thigh, it is clear the intra-planar segmental relationship noted by Imwalle et al seems to persist in non-adjacent segments. As such, our study provides preliminary evidence for a direct link between transverse plane trunk motion and a proposed risk factor of ACL injury; tibial internal rotation.

Combined loading across all three cardinal planes has been demonstrated to impart the greatest load on the ACL.[17] The results of our study indicate decreased trunk stability during unstable sitting, and increased transverse plane motion of the trunk during an athletic cutting task are associated with lower extremity biomechanical risk factors of ACL injury in all three planes. In the sagittal plane we observed increases in internal knee extension moment to be associated with decreases in trunk stability during unstable sitting. In the frontal plane we observed increases in internal knee varus moment to be associated with increases in transverse plane trunk angle away from the stance leg. In the transverse plane we observed increases in internal tibial rotation angle to be associated with increases in transverse plane trunk angle away from the stance leg. Our results support previous findings of Zazulak et al,[26] identifying decreased trunk NMC to be a risk factor of ACL injury. Our study directly links trunk NMC to lower extremity biomechanical risk factors of ACL injury in all three planes. The results of our study direct future research to establish an understanding of the effects of NMC of the trunk on lower extremity biomechanical risk factors of ACL injury.

Contrary to our hypotheses, we were not able to demonstrate any significant relationships between measures of trunk stability and knee kinematics. It is possible that the magnitude of any potential relationships between our trunk stability measures and knee biomechanics during a side-step cutting task are either too small to be detected in the current analysis, or that the task itself was not demanding enough dynamically to bring about biomechanical changes. While we did observe decreased trunk stability was associated with greater internal knee extension moment demand, it may be that in this task the participants with lesser stability met those demands and therefore did not display knee mechanics that were appreciably different than those with greater trunk stability. This suggests that future research should examine more demanding tasks as well as multiple elements of trunk NMC, as previous research has identified that deficits in reactive and proprioceptive elements of trunk NMC are predictive of knee injury.[26, 27] A model encompassing proprioceptive, reactive, and stability components of trunk NMC may permit future research to more comprehensively associate trunk NMC with lower extremity biomechanics during athletic tasks that are related to knee injury.

Limitations

This current study has four primary limitations that should be addressed. First, as a correlational design can only establish relationships between variables and not determine cause and effect, the results of this study are limited in that they are preliminary, exploratory evidence that should be used as rationale for future research investigations. Second, subjects with a history of balance training and performance of cutting tasks may exhibit improved performance over those who have not had previous exposure to the tasks. As the inclusion criteria for the study was participation in physical activity for at least 20 minutes a day at least 3 times per week, the activity level of our participants ranged from this minimum to Division I collegiate varsity athletes. It is not possible to discern the effects, if any, of physical activity participation level on the variables of interest. Thus, the study's findings can only be generalized to this specific population in regards to physical activity participation level. Third, the average age of the subjects in this investigation was 20.5 ± 2.3 years. While this demographic is representative of college-aged young-adults, our findings should not be generalized to older or younger populations. Finally, we assumed that the CoP measures were representative of the body's ability to stabilize the trunk without influence from the extremities or head. Although participants were instructed to maintain a standardized test position (arms across chest, looking straight ahead) minor movement of the extremities and head may have had the ability to influence the CoP results.

Conclusions

5. Decreased trunk stability is associated with increased peak knee extension moment, a risk factor for ACL injury. Therefore, training to increase trunk stability may result in decreased peak knee extension moment during athletic tasks, and a potential reduction in ACL injury risk.

6. Decreased trunk stability is associated with increased lateral trunk flexion toward the stance leg during a side-step cutting task. Increased lateral trunk flexion towards the stance leg during cutting tasks has been associated with ACL injury risk. Trunk stability training may decrease lateral trunk flexion towards the stance leg during athletic tasks, possibly decreasing the presence of lower extremity mechanics associated with ACL injury, and ACL injury risk.
7. Increased trunk rotation away from the stance leg is associated with increased knee varus moment and increased tibial internal rotation, identified risk factors of ACL injury. Future ACL injury prevention strategies decreasing trunk rotation during side-step cutting tasks may decrease knee biomechanics associated with ACL injury.
8. No significant relationships were observed between trunk stability and knee kinematics. The stability element of trunk NMC may not be directly associated with knee kinematics during a cutting task. Future research should implement comprehensive measures of trunk NMC.

REFERENCES FOR APPENDIX F

1. Alentorn-Geli E, Myer GD, Silvers HJ, et al. Prevention of non-contact anterior cruciate ligament injuries in soccer players. Part 1: Mechanisms of injury and underlying risk factors. *Knee Surg Sports Traumatol Arthrosc.* Jul 2009;17(7):705-729.

2. Bell AL, Pedersen DR, Brand RA. A comparison of the accuracy of several hip center location prediction methods. *J Biomech.* 1990;23(6):617-621.

3. Blackburn JT, Padua DA. Influence of trunk flexion on hip and knee joint kinematics during a controlled drop landing. *Clin Biomech (Bristol, Avon).* Mar 2008;23(3):313-319.

4. Blackburn JT, Padua DA. Sagittal-plane trunk position, landing forces, and quadriceps electromyographic activity. *J Athl Train.* Mar-Apr 2009;44(2):174-179.

5. Boden BP, Dean GS, Feagin JA, Jr., Garrett WE, Jr. Mechanisms of anterior cruciate ligament injury. *Orthopedics.* Jun 2000;23(6):573-578.

6. Cholewicki J, Polzhofer GK, Radebold A. Postural control of trunk during unstable sitting. *J Biomech.* Dec 2000;33(12):1733-1737.

7. DeMorat G, Weinhold P, Blackburn T, Chudik S, Garrett W. Aggressive quadriceps loading can induce noncontact anterior cruciate ligament injury. *Am J Sports Med.* Mar 2004;32(2):477-483.

8. Griffin LY, Albohm MJ, Arendt EA, et al. Understanding and preventing noncontact anterior cruciate ligament injuries: a review of the Hunt Valley II meeting, January 2005. *Am J Sports Med.* Sep 2006;34(9):1512-1532.

9. Gupta RT, Vankoski S, Novak RA, Dias LS. Trunk kinematics and the influence on valgus knee stress in persons with high sacral level myelomeningocele. *J Pediatr Orthop.* Jan-Feb 2005;25(1):89-94.

10. Hewett TE, Myer GD, Ford KR, et al. Biomechanical measures of neuromuscular control and valgus loading of the knee predict anterior cruciate ligament injury risk in female athletes: a prospective study. *Am J Sports Med.* Apr 2005;33(4):492-501.

11. Hewett TE, Torg JS, Boden BP. Video analysis of trunk and knee motion during non-contact anterior cruciate ligament injury in female athletes: lateral trunk and knee abduction motion are combined components of the injury mechanism. *Br J Sports Med.* Jun 2009;43(6):417-422.

12. Houck JR, Duncan A, De Haven KE. Comparison of frontal plane trunk kinematics and hip and knee moments during anticipated and unanticipated walking and side step cutting tasks. *Gait Posture.* Nov 2006;24(3):314-322.

13. Imwalle LE, Myer GD, Ford KR, Hewett TE. Relationship between hip and knee kinematics in athletic women during cutting maneuvers: a possible link to noncontact anterior cruciate ligament injury and prevention. *J Strength Cond Res.* Nov 2009;23(8):2223-2230.

14. Kanamori A, Woo SL, Ma CB, et al. The forces in the anterior cruciate ligament and knee kinematics during a simulated pivot shift test: A human cadaveric study using robotic technology. *Arthroscopy.* Sep 2000;16(6):633-639.

15. Kibler WB, Press J, Sciascia A. The role of core stability in athletic function. *Sports Med.* 2006;36(3):189-198.

16. Lo J, Muller O, Wunschel M, Bauer S, Wulker N. Forces in anterior cruciate ligament during simulated weight-bearing flexion with anterior and internal rotational tibial load. *J Biomech.* 2008;41(9):1855-1861.

17. Markolf KL, Burchfield DM, Shapiro MM, Shepard MF, Finerman GA, Slauterbeck JL. Combined knee loading states that generate high anterior cruciate ligament forces. *J Orthop Res.* Nov 1995;13(6):930-935.

18. Myer GD, Chu DA, Brent JL, Hewett TE. Trunk and hip control neuromuscular training for the prevention of knee joint injury. *Clin Sports Med.* Jul 2008;27(3):425-448, ix.

19. Patla AE, Adkin A, Ballard T. Online steering: coordination and control of body center of mass, head and body reorientation. *Exp Brain Res.* Dec 1999;129(4):629-634.

20. Radebold A, Cholewicki J, Polzhofer GK, Greene HS. Impaired postural control of the lumbar spine is associated with delayed muscle response times in patients with chronic idiopathic low back pain. *Spine.* Apr 1 2001;26(7):724-730.

21. Reeves NP, Cholewicki J, Lee AS, Mysliwiec LW. The effects of stochastic resonance stimulation on spine proprioception and postural control in chronic low back pain patients. *Spine.* Feb 15 2009;34(4):316-321.

22. Reeves NP, Cholewicki J, Narendra KS. Effects of reflex delays on postural control during unstable seated balance. *J Biomech.* Jan 19 2009;42(2):164-170.

23. Reeves NP, Everding VQ, Cholewicki J, Morrisette DC. The effects of trunk stiffness on postural control during unstable seated balance. *Exp Brain Res.* Oct 2006;174(4):694-700.

24. Withrow TJ, Huston LJ, Wojtys EM, Ashton-Miller JA. The effect of an impulsive knee valgus moment on in vitro relative ACL strain during a simulated jump landing. *Clin Biomech (Bristol, Avon).* Nov 2006;21(9):977-983.

25. Zazulak B, Cholewicki J, Reeves NP. Neuromuscular control of trunk stability: clinical implications for sports injury prevention. *J Am Acad Orthop Surg.* Sep 2008;16(9):497-505.

26. Zazulak BT, Hewett TE, Reeves NP, Goldberg B, Cholewicki J. Deficits in neuromuscular control of the trunk predict knee injury risk: a prospective biomechanical-epidemiologic study. *Am J Sports Med.* Jul 2007;35(7):1123-1130.

27. Zazulak BT, Hewett TE, Reeves NP, Goldberg B, Cholewicki J. The effects of core proprioception on knee injury: a prospective biomechanical-epidemiological study. *Am J Sports Med.* Mar 2007;35(3):368-373.

REFERENCES

1. Alentorn-Geli E, Myer GD, Silvers HJ, et al. Prevention of non-contact anterior cruciate ligament injuries in soccer players. Part 1: Mechanisms of injury and underlying risk factors. *Knee Surg Sports Traumatol Arthrosc.* Jul 2009;17(7):705-729.

2. Amis AA, Dawkins GP. Functional anatomy of the anterior cruciate ligament. Fibre bundle actions related to ligament replacements and injuries. *J Bone Joint Surg Br.* Mar 1991;73(2):260-267.

3. An KN. Muscle force and its role in joint dynamic stability. *Clin Orthop Relat Res.* 10/2002 2002(403suppl):S37.

4. Arendt E, Dick R. Knee injury patterns among men and women in collegiate basketball and soccer. NCAA data and review of literature. *Am J Sports Med.* Nov-Dec 1995;23(6):694-701.

5. Beales M, Flay N, McKinney R, et al. Mutations in the large subunit of U2AF disrupt pre-mRNA splicing, cell cycle progression and nuclear structure. *Yeast.* Aug 2000;16(11):1001-1013.

6. Bell AL, Pedersen DR, Brand RA. A comparison of the accuracy of several hip center location prediction methods. *J Biomech.* 1990;23(6):617-621.

7. Bennett DR, Blackburn JT, Boling MC, McGrath M, Walusz H, Padua DA. The relationship between anterior tibial shear force during a jump landing task and quadriceps and hamstring strength. *Clin Biomech (Bristol, Avon).* Nov 2008;23(9):1165-1171.

8. Blackburn JT, Padua DA. Influence of trunk flexion on hip and knee joint kinematics during a controlled drop landing. *Clin Biomech (Bristol, Avon).* Mar 2008;23(3):313-319.

9. Blackburn JT, Padua DA. Sagittal-plane trunk position, landing forces, and quadriceps electromyographic activity. *J Athl Train.* Mar-Apr 2009;44(2):174-179.

10. Boden BP, Dean GS, Feagin JA, Jr., Garrett WE, Jr. Mechanisms of anterior cruciate ligament injury. *Orthopedics.* Jun 2000;23(6):573-578.

11. Borghuis J, Hof AL, Lemmink KA. The importance of sensory-motor control in providing core stability: implications for measurement and training. *Sports Med.* 2008;38(11):893-916.

12. Cholewicki J, Polzhofer GK, Radebold A. Postural control of trunk during unstable sitting. *J Biomech.* Dec 2000;33(12):1733-1737.

13. DeMorat G, Weinhold P, Blackburn T, Chudik S, Garrett W. Aggressive quadriceps loading can induce noncontact anterior cruciate ligament injury. *Am J Sports Med.* Mar 2004;32(2):477-483.

14. Fleming BC, Beynnon BD, Renstrom PA, et al. The strain behavior of the anterior cruciate ligament during stair climbing: an in vivo study. *Arthroscopy.* Mar 1999;15(2):185-191.

15. Fleming BC, Renstrom PA, Beynnon BD, et al. The effect of weightbearing and external loading on anterior cruciate ligament strain. *J Biomech.* Feb 2001;34(2):163-170.

16. Giugliano DN, Solomon JL. ACL tears in female athletes. *Phys Med Rehabil Clin N Am.* Aug 2007;18(3):417-438, viii.

17. Goldfuss AJ, Morehouse CA, LeVeau BF. Effect of muscular tension on knee stability. *Med Sci Sports.* Winter 1973;5(4):267-271.

18. Goodfellow J, O'Connor J. The mechanics of the knee and prosthesis design. *J Bone Joint Surg Br.* Aug 1978;60-B(3):358-369.

19. Gottlob CA, Baker CL, Jr., Pellissier JM, Colvin L. Cost effectiveness of anterior cruciate ligament reconstruction in young adults. *Clin Orthop Relat Res.* Oct 1999(367):272-282.

20. Gray J, Taunton JE, McKenzie DC, Clement DB, McConkey JP, Davidson RG. A survey of injuries to the anterior cruciate ligament of the knee in female basketball players. *Int J Sports Med.* Dec 1985;6(6):314-316.

21. Griffin LY, Agel J, Albohm MJ, et al. Noncontact anterior cruciate ligament injuries: risk factors and prevention strategies. *J Am Acad Orthop Surg.* May-Jun 2000;8(3):141-150.

22. Griffin LY, Albohm MJ, Arendt EA, et al. Understanding and preventing noncontact anterior cruciate ligament injuries: a review of the Hunt Valley II meeting, January 2005. *Am J Sports Med.* Sep 2006;34(9):1512-1532.

23. Gupta RT, Vankoski S, Novak RA, Dias LS. Trunk kinematics and the influence on valgus knee stress in persons with high sacral level myelomeningocele. *J Pediatr Orthop.* Jan-Feb 2005;25(1):89-94.

24. Heitz NA, Eisenman PA, Beck CL, Walker JA. Hormonal Changes Throughout the Menstrual Cycle and Increased Anterior Cruciate Ligament Laxity in Females. *J Athl Train.* Apr 1999;34(2):144-149.

25. Herrington L, Nester C. Q-angle undervalued? The relationship between Q-angle and medio-lateral position of the patella. *Clin Biomech (Bristol, Avon).* Dec 2004;19(10):1070-1073.

26. Hewett TE. Predisposition to ACL injuries in female athletes versus male athletes. *Orthopedics.* Jan 2008;31(1):26-28.

27. Hewett TE, Myer GD, Ford KR, et al. Biomechanical measures of neuromuscular control and valgus loading of the knee predict anterior cruciate ligament injury risk in female athletes: a prospective study. *Am J Sports Med.* Apr 2005;33(4):492-501.

28. Hootman JM, Dick R, Agel J. Epidemiology of collegiate injuries for 15 sports: summary and recommendations for injury prevention initiatives. *J Athl Train.* Apr-Jun 2007;42(2):311-319.

29. Houck JR, Duncan A, De Haven KE. Comparison of frontal plane trunk kinematics and hip and knee moments during anticipated and unanticipated walking and side step cutting tasks. *Gait Posture.* Nov 2006;24(3):314-322.

30. Hughes G, Watkins J. A risk-factor model for anterior cruciate ligament injury. *Sports Med.* 2006;36(5):411-428.

31. Imwalle LE, Myer GD, Ford KR, Hewett TE. Relationship between hip and knee kinematics in athletic women during cutting maneuvers: a possible link to noncontact anterior cruciate ligament injury and prevention. *J Strength Cond Res.* Nov 2009;23(8):2223-2230.

32. Kibler WB, Press J, Sciascia A. The role of core stability in athletic function. *Sports Med.* 2006;36(3):189-198.

33. Kulas AS, Schmitz RJ, Shultz SJ, Henning JM, Perrin DH. Sex-specific abdominal activation strategies during landing. *J Athl Train.* Oct-Dec 2006;41(4):381-386.

34. Landry SC, McKean KA, Hubley-Kozey CL, Stanish WD, Deluzio KJ. Neuromuscular and lower limb biomechanical differences exist between male and female elite adolescent soccer players during an unanticipated run and crosscut maneuver. *Am J Sports Med.* Nov 2007;35(11):1901-1911.

35. Landry SC, McKean KA, Hubley-Kozey CL, Stanish WD, Deluzio KJ. Gender differences exist in neuromuscular control patterns during the pre-contact and early stance phase of an unanticipated side-cut and cross-cut maneuver in 15-18 years old adolescent soccer players. *J Electromyogr Kinesiol.* Oct 2009;19(5):e370-379.

36. Li G, Rudy TW, Sakane M, Kanamori A, Ma CB, Woo SL. The importance of quadriceps and hamstring muscle loading on knee kinematics and in-situ forces in the ACL. *J Biomech.* Apr 1999;32(4):395-400.

37. Lin CF, Gross M, Ji C, et al. A stochastic biomechanical model for risk and risk factors of non-contact anterior cruciate ligament injuries. *J Biomech.* Mar 11 2009;42(4):418-423.

38. Liu SH A-SR, Panossian V, et al. Primary immonulocalization of estrogen and progesteron target cells in the human anterior cruciate ligament. *Journal Orthop Res.* 1996;14:526-533.

39. Lo J, Muller O, Wunschel M, Bauer S, Wulker N. Forces in anterior cruciate ligament during simulated weight-bearing flexion with anterior and internal rotational tibial load. *J Biomech.* 2008;41(9):1855-1861.

40. Lombardo S, Sethi PM, Starkey C. Intercondylar notch stenosis is not a risk factor for anterior cruciate ligament tears in professional male basketball players: an 11-year prospective study. *Am J Sports Med.* Jan 2005;33(1):29-34.

41. Louboutin H, Debarge R, Richou J, et al. Osteoarthritis in patients with anterior cruciate ligament rupture: A review of risk factors. *Knee.* Dec 19 2008.

42. MacKinnon CD, Winter DA. Control of whole body balance in the frontal plane during human walking. *J Biomech.* Jun 1993;26(6):633-644.

43. Markolf KL, Burchfield DM, Shapiro MM, Shepard MF, Finerman GA, Slauterbeck JL. Combined knee loading states that generate high anterior cruciate ligament forces. *J Orthop Res.* Nov 1995;13(6):930-935.

44. McGinty G, Irrgang JJ, Pezzullo D. Biomechanical considerations for of the knee. *Clin Biomech (Bristol, Avon).* Mar 2000;15(3):160-166.

45. Moore K. *Clinically Oriented Anatomy*. Fifth ed. Baltimore: Lippincott Williams & Wilkins; 2006.

46. Morrey MA, Stuart MJ, Smith AM, Wiese-Bjornstal DM. A longitudinal examination of athletes' emotional and cognitive responses to anterior cruciate ligament injury. *Clin J Sport Med.* Apr 1999;9(2):63-69.

47. Muller W. *The knee: Form, Function, and Ligament Reconstruction*. New York: Springer; 1985.

48. Muller W. Form and function of the knee. Its relation to high performance and to sports. *Am J Sports Med.* 1996;24(6 Suppl):S104-106.

49. Myer GD, Brent JL, Ford KR, Hewett TE. A pilot study to determine the effect of trunk and hip focused neuromuscular training on hip and knee isokinetic strength. *Br J Sports Med.* Jul 2008;42(7):614-619.

50. Myer GD, Chu DA, Brent JL, Hewett TE. Trunk and hip control neuromuscular training for the prevention of knee joint injury. *Clin Sports Med.* Jul 2008;27(3):425-448, ix.

51. Najibi S, Albright JP. The use of knee braces, part 1: Prophylactic knee braces in contact sports. *Am J Sports Med.* Apr 2005;33(4):602-611.

52. Patla AE, Adkin A, Ballard T. Online steering: coordination and control of body center of mass, head and body reorientation. *Exp Brain Res.* Dec 1999;129(4):629-634.

53. Radebold A, Cholewicki J, Polzhofer GK, Greene HS. Impaired postural control of the lumbar spine is associated with delayed muscle response times in patients with chronic idiopathic low back pain. *Spine.* Apr 1 2001;26(7):724-730.

54. Reeves NP, Cholewicki J, Lee AS, Mysliwiec LW. The effects of stochastic resonance stimulation on spine proprioception and postural control in chronic low back pain patients. *Spine.* Feb 15 2009;34(4):316-321.

55. Reeves NP, Cholewicki J, Narendra KS. Effects of reflex delays on postural control during unstable seated balance. *J Biomech.* Jan 19 2009;42(2):164-170.

56. Reeves NP, Everding VQ, Cholewicki J, Morrisette DC. The effects of trunk stiffness on postural control during unstable seated balance. *Exp Brain Res.* Oct 2006;174(4):694-700.

57. Renstrom P, Ljungqvist A, Arendt E, et al. Non-contact ACL injuries in female athletes: an International Olympic Committee current concepts statement. *Br J Sports Med.* Jun 2008;42(6):394-412.

58. Rijke AM, Vierhout PA. Graded stress radiography in acute injury to the lateral ligaments of the ankle. *Acta Radiol.* Mar 1990;31(2):151-155.

59. Shambaugh JP, Klein A, Herbert JH. Structural measures as predictors of injury basketball players. *Med Sci Sports Exerc.* May 1991;23(5):522-527.

60. Shultz SJ, Kirk SE, Johnson ML, Sander TC, Perrin DH. Relationship between sex hormones and anterior knee laxity across the menstrual cycle. *Med Sci Sports Exerc.* Jul 2004;36(7):1165-1174.

61. Souryal TO, Freeman TR. Intercondylar notch size and anterior cruciate ligament injuries in athletes. A prospective study. *Am J Sports Med.* Jul-Aug 1993;21(4):535-539.

62. Van Lunen BL, Roberts J, Branch JD, Dowling EA. Association of Menstrual-Cycle Hormone Changes with Anterior Cruciate Ligament Laxity Measurements. *J Athl Train.* Dec 2003;38(4):298-303.

63. Wheeless III M, Clifford R. . Wheeless' Textbook of Orthopaedics. In: Wheeless III M, Clifford R. Nunley, II, MD, James A. Urbaniak, MD, James R., ed; 2008.

64. Withrow TJ, Huston LJ, Wojtys EM, Ashton-Miller JA. The effect of an impulsive knee valgus moment on in vitro relative ACL strain during a simulated jump landing. *Clin Biomech (Bristol, Avon).* Nov 2006;21(9):977-983.

65. Withrow TJ, Huston LJ, Wojtys EM, Ashton-Miller JA. Effect of varying hamstring tension on anterior cruciate ligament strain during in vitro impulsive knee flexion and compression loading. *J Bone Joint Surg Am.* Apr 2008;90(4):815-823.

66. Wojtys EM, Huston LJ, Boynton MD, Spindler KP, Lindenfeld TN. The effect of the menstrual cycle on anterior cruciate ligament injuries in women as determined by hormone levels. *Am J Sports Med.* Mar-Apr 2002;30(2):182-188.

67. Yu B, Garrett WE. Mechanisms of non-contact ACL injuries. *Br J Sports Med.* Aug 2007;41 Suppl 1:i47-51.

68. Zazulak B, Cholewicki J, Reeves NP. Neuromuscular control of trunk stability: clinical implications for sports injury prevention. *J Am Acad Orthop Surg.* Sep 2008;16(9):497-505.

69. Zazulak BT, Hewett TE, Reeves NP, Goldberg B, Cholewicki J. Deficits in neuromuscular control of the trunk predict knee injury risk: a prospective biomechanical-epidemiologic study. *Am J Sports Med.* Jul 2007;35(7):1123-1130.

70. Zazulak BT, Hewett TE, Reeves NP, Goldberg B, Cholewicki J. The effects of core proprioception on knee injury: a prospective biomechanical-epidemiological study. *Am J Sports Med.* Mar 2007;35(3):368-373.

71. Zazulak BT, Ponce PL, Straub SJ, Medvecky MJ, Avedisian L, Hewett TE. Gender comparison of hip muscle activity during single-leg landing. *J Orthop Sports Phys Ther.* May 2005;35(5):292-299.