THE ASSOCIATION BETWEEN THE CORE AND ANTERIOR CRUCIATE LIGAMENT INJURY RISK FACTORS

DISSERTATION

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Abstract

Non-contact anterior cruciate ligament (ACL) injury is a common serious injury among field and court sport athletes. Aspects of the core and trunk have been associated with ACL injury incidence, though the connection between the two is not well understood. The purpose of this project is to better understand this relationship through a series of experiments measuring aspects of the core as well as surrogates for ACL injury.

Specific studies in this project addressed various aspects of the relationship in unique ways. An intervention program lacking trunk-specific exercises resulted in increased peak external knee abduction moment (pEKAbM) during cutting (p=0.012) and decreased lateral trunk control (p=0.029) while an intervention program including these exercises did not significantly alter either (lateral trunk control, p=0.178; pEKAbM, p=0.116). Another study found lateral trunk lean during cutting had a significant, positive association with pEKAbM, even when controlling for speed, gender, and cutting angle (p=0.021). Finally, electromyography analysis of select core muscles revealed increased co-contraction of the L5 extensors had a significant, positive association with pEKAbM. This same study also found that an increase in the percent difference of L5 extensor activation was associated with a decrease in trunk control during an isolated task (0.014).

Combined, this project supports the hypothesis that control and movement of the core are associated with the loading conditions at the knee that have previously been linked to
ACL injury risk. Future work should use these results to develop and appropriately test intervention programs for their ability to alter lateral trunk control, pEKAbM, and, ultimately, ACL injury risk.
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1 Introduction

1.1 The Problem – ACL Rupture

The anterior cruciate ligament (ACL) is one of four major ligaments that help provide stability for the human knee. The ACL originates within the notch of the distal femur and terminates in the tibial plateau (Figure 1.1). The primary function of the ACL is to resist anterior (forward) translation and internal rotation of the tibia with respect to the femur. ACL rupture is one of the most common knee injuries for field and court sport athletes (i.e. basketball, American football, soccer, etc). A study of the most recent National Survey of Ambulatory Surgery, a comprehensive survey of hospital-based and free-standing surgery centers conducted by the National Center for Health Statistics (NCHS) at the Centers for Disease Control and Prevention (CDC), found that nearly 127,446 (95% CI; 95,124 to 159,768) ACL reconstructions were performed in 2006 (Kim et al., 2011). At an average estimated cost of $30,000-$40,000 for the surgery and subsequent related follow-up (Pearldiver, 2010a,b), annual direct medical costs for ACL reconstruction in the United States reaches into the billions.
The consequences of ACL injury extend beyond the medical costs of initial treatment. With rehabilitation for return to sport taking between 6 and 12 months, most athletes lose the remainder of their season. Injured athletes are at an increased risk of developing osteoarthritis, which could result in future complications and potential medical costs later in life (Ruiz et al., 2002). Additionally, lowered academic performance and loss of scholarships have been associated with ACL injuries in collegiate athletes (Freedman et al., 1998).

Approximately 70% of ACL ruptures in sport are thought to be non-contact in nature, meaning the injury is not a result of a direct blow by either a player or object (Griffin et al., 2000). These non-contact injuries are often a result of sudden changes in direction or
pivoting (Boden et al., 2000; McNair, 1990). Injury events usually result in a medial collapse (toward the midline) of the knee (Boden et al., 2000; Boden et al., 2009).

1.2 Core Stability

Core stability has recently become a popular topic in the popular media as well as in scientific literature (Kibler et al., 2006; Schuler, 2011). The popularity of core stability stems from what seems to be a widely-held hypothesis that the core plays an important role in athletic function and that improvements in core stability will result in improved athletic performance, reduced injury risk, or both. Inconsistent definitions and broad applications of the term core stability make synthesizing the current state of the literature difficult. To provide a frame of reference for the discussion contained here, the terms “core,” “trunk,” and “core stability,” will first be defined.

1.2.1 Core

The core region in some literature consists of all the muscles that cross the lumbar and inferior thoracic spine as well as all those that cross the hip – colloquially referred to as “nipples to knees.” Because many of the muscles that cross the hip joint act as prime movers of the lower extremity, this definition may be too broad. The muscles that cross the lumbar spine act primarily to move or hold the torso, which is quite different from the power generated by the gluteal muscles during jumping and landing. For this reason, this project will separate these regions and define the core as the region of the body bounded by the pelvis and diaphragm which includes the muscles of the abdomen and lower back (Figure 1.2).
1.2.2 Trunk

The trunk is an extension of the core that includes the thorax (chest and upper back) and this project will define the trunk as *the region of the body bounded by the pelvis and the clavicles* (Figure 1.2). Because the muscles of the core are largely responsible for the positioning and control of the trunk, and it is this positioning and control that this project
will focus on, the additional muscle content of this region is not of particular importance
for this discussion.

1.2.3 Core stability
According to the Merriam-Webster Dictionary, “stability” can be defined as “the
property of a body that causes it when disturbed from a condition of equilibrium or
steady motion to develop forces or moments that restore the original condition”
(Merriam-Webster.com, 2012). It follows, then, that “core stability” is the ability of the
core “to maintain or resume relative position of the trunk after a perturbation [or
disturbance]” (Zazulak et al., 2007b). This disturbance can either be external (i.e. another
player or object on the playing field) or from movement of the extremities. The muscles
of the core must continually react to the varying loading conditions and body positions to
maintain stability, making it a dynamic quality rather than an intrinsic property (Willson
et al., 2005). The realization of stability during athletic maneuvers requires appropriate
timing and intensity of core muscle activation in addition to muscular strength and
endurance. All of these attributes are essential in efficiently maintaining core stability.

1.3 Why the core and the ACL?
The muscles of the core are largely responsible for the positioning and control of the
trunk over the lower extremities. Since the trunk and upper body represent more than half
a person’s body weight (Dempster and Gaughran, 1967), poor trunk control and core
stability could place this mass in a position that results in adverse loading of the lower
extremity, including the knee, potentially leading to injury. Some speculate that better
core stability may be one of the factors associated with improved athletic performance
because the core provides “proximal stability for distal mobility” (Kibler et al., 2006). A
study of muscle activations revealed that muscles of the core activate prior to muscles responsible for moving the lower extremities (Hodges and Richardson, 1997). This core-first muscle activation pattern is thought to increase efficiency and effectiveness of the limb movement by providing a stable base for the movement to occur from. A recent study of professional baseball pitchers demonstrated that better control of the lumbopelvic region was associated with significantly more innings pitched as well as significantly fewer walks plus hits per inning pitched over one season (Chaudhari et al., 2011). This study supports the theory of better control leading to more efficient and effective movement of the extremities. While this study associates the core with upper extremity movement, it demonstrates that the core provides a stable base for better control of the extremities, suggesting that a similar association between the core and lower extremity may exist. Better control of the lower extremities may lead to decreased adverse loading of the knee, thereby reducing ACL injury risk.

1.4 Previous research linking core and ACL
Some evidence does exist suggesting that the core is linked to ACL injury risk. Three prospective studies have shown an association between ACL injury and measures of the core. Several intervention studies have been presented, although conclusions relating the core to ACL injury are difficult to draw. Finally, cross-sectional studies using surrogate measures for ACL injury provide some information linking the core to ACL injury risk, but since surrogates for injury are used definite conclusions are unattainable.

1.4.1 Prospective evidence
Prospective studies are a good way to investigate potential screening tools that could be used to categorize athletes by injury risk potential. The idea of a screening test is that
it relates observable pre-injury measures, in some way, to the mechanism of the injury, though this link does not have to be as straightforward as one may initially think. For instance, a biological condition could exist that influences both ligament properties and eye color. If this were true, it might be reasonable to screen for ACL injury risk using eye color, since eye color is easy to observe, even though eye color itself is not a component of the ACL injury mechanism.

With this in mind, investigators aim to find measures that are related in some way to the injury mechanism but that can be used on a large scale for screening purposes. Leetun et al. (2004), to my knowledge, was the first to attempt a prospective study for ACL injury risk using measures of the core. For their investigation they employed the prone plank and side bridge core endurance tests with intercollegiate athletes. These tests are simple to administer, but unfortunately no significant association between core muscle endurance and future injury status was found.

In contrast to Leetun et al.’s approach to using easy-to-administer tests, two prospective studies at Yale by Zazulak et al. used two different custom-made devices to test core stability as it is defined previously in this chapter (core stability: ability of the core “to maintain or resume relative position of the trunk after a perturbation [or disturbance]”) (Zazulak et al., 2007a,b). For their studies, Zazulak et al. recruited 140 female and 137 male varsity athletes who had no prior history of knee injury. A knee injury was defined as any ligament, meniscal, or patellofemoral injury diagnosed by a university sports medicine physician.

In the first study, trunk proprioception, or the ability of an individual to return the trunk to a starting position after being rotated in the transverse plane, was characterized
(Zazulak et al., 2007b). In the second study, trunk control, or the ability of the core to stop the movement of the trunk after a sudden, unexpected perturbation, was characterized (Zazulak et al., 2007a). Injury data was collected for three years following each of the tests.

Females who later experienced knee and knee ligament/meniscus injuries had significantly greater repositioning errors than uninjured females (Zazulak et al., 2007b). No significant differences were found between ACL-injured and uninjured females, though the ACL-injured sample size (n=4) was too small to draw definitive conclusions. Further, ligament-injured female athletes demonstrated significantly greater maximum lateral displacement after the sudden, unexpected perturbation than uninjured female athletes (Zazulak et al., 2007b). When results were collapsed across gender, knee-, knee ligament-, and ACL-injured athletes demonstrated significantly greater maximum lateral displacement after the sudden, unexpected perturbation than uninjured athletes.

Results from these two studies suggest a role for the core in knee injuries because it is reasonable to associate trunk proprioception and control to misalignment of the trunk over the lower extremity, resulting in increased adverse loading of the knee. It remains unknown whether this is indeed the mechanism for connecting these core measure to knee injuries, however, since performance in a controlled laboratory testing device may or may not transfer to how an athlete responds to the demands of play for his/her sport.

While the devices used in these studies are not suitable for large-scale screening, information gathered from their use can still be quite valuable. Long-term prospective studies are difficult, expensive, and, as the description implies, take a long time. Since performance in these devices is associated with lower extremity injury during prospective
studies, they may have potential as surrogates for ACL injury risk during short-term intervention or cross-sectional studies. Short-term intervention and cross-sectional studies usually have the benefit of being easier, cheaper, and quicker than longitudinal ones. While they do not measure injury directly, they are useful in evaluating interventions and identifying parameters that might be of particular interest. This information can then be fed back into longitudinal studies with more confidence that the intervention or parameters being evaluated will produce the results desired.

1.4.2 Intervention studies

Components of core stability training have been successfully used in injury prevention intervention programs that resulted in reduced ACL injury incidence (Emery et al., 2007; Gilchrist et al., 2008; Hewett et al., 1999; Mandelbaum et al., 2005; Myklebust et al., 2003; Petersen et al., 2005; Soligard et al., 2008). It is impractical to determine from these studies if the core stability component is necessary or if it is merely an adjunct component. Because the inclusion of components was not done on a systematic basis where components would be added/removed on an individual basis, it is not possible to determine each exercise’s contribution to injury incidence or functional change. Further, it is unknown whether core training in isolation would be effective or if core training needs to be a part of a comprehensive injury-prevention program in order to be effective.

1.4.3 Cross-sectional studies

Because cross-sectional studies do not track injuries, surrogate measures must be used to determine the relative contribution of tested parameters on ACL injury risk. The surrogate measure for ACL injury risk most often used, and with the most evidence, is
external knee abduction moment (EKAbM). An EKAbM occurs when the force generated between the ground and the lower limb (ground reaction force; GRF) has a moment arm lateral (on the outside) of the knee and acts to push the knee medially (toward the midline of the body). An increase in the magnitude of the GRF or the lateral moment arm will increase EKAbM. All body segment orientations, velocities, and orientations affect both the magnitude and orientation of the GRF, though some have more of an effect (leg segments) than others (arms and head). The leg segments influence EKAbM more than the arms or head due to their (legs’) increased mass contributing more to the composition of the GRF and their direct influence on the moment arm. Additionally, the mass and size of the person (more specifically the mass and size of body segments) will affect the magnitude of the GRF. Many scientists normalize EKAbM by body mass and/or height to account for these differences between subjects and allow comparisons between subjects to be based on movement pattern differences rather than body size (Andriacchi et al., 2005; Cortes et al., 2011; Mclean et al., 2007).

Increases in EKAbM have been associated with ACL injuries in a prospective study of female adolescent athletes (Hewett et al., 2005). Increases in EKAbM have also been associated with increased strain (elongation) of the ACL in cadaveric (Fleming et al., 2001; Markolf et al., 1995) and computer simulation studies (Shin et al., 2009; Shin et al., 2011).

One cross-sectional study investigated arm position and the effect that it had on peak EKAbM (Chaudhari et al., 2005). For this study marker-based motion capture and inverse dynamic techniques were used to calculate pEKAbM while subjects performed a 90° side-step cut with varying arm positions. The authors found that when the arms were
constrained closer to the body with either a football or lacrosse stick pEKA\text{bM} was significantly higher than when the arms were not constrained. This suggests that subjects alter their pEKA\text{bM} by extending their arms away from their trunk. While this is not a measure of the core directly, it does provide evidence that positioning of parts of the body above the waist does alter knee loading so it is possible that positioning of the trunk may also influence knee loading.

1.4.4 Limitations

While the above studies provide some evidence that the trunk influences ACL injury risk, they all lack information regarding how the core controls the trunk during the activities studied. To get a better sense of how the core controls the trunk during these activities core muscle activation must be assessed during them. Once the core’s trunk control strategy is better understood, more efficient intervention programs can be developed to target and alter improper control strategies.

1.5 Statement of Purpose

The current body of literature on the core and the ACL is limited. A theoretical evaluation of anatomy and movement during sport suggests a relationship between the two is likely (Section 1.3). In addition, some experimental evidence suggests the core may be associated with ACL injury risk (Section 1.4), but more needs to be done in this area to better understand their relationship (Figure 1.3). The fundamental purpose of this project is to develop the knowledge base connecting the core and the ACL. With a better understanding of how the core influences ACL injury risk factors, better, more effective ACL injury prevention programs can be developed and implemented – ultimately leading to fewer athletes experiencing this devastating injury.
Figure 1.3. Previous studies suggest a connection between trunk control, trunk position, EKAbM and ACL injury (black arrows). Little is known about how the trunk influences ACL injury, however. Does the trunk influence EKAbM (gray arrow), thereby influencing ACL injury? What are muscle activation strategies employed by the core that affect trunk control (green arrow), trunk positioning (green arrow), EKAbM (red arrow)?

Several studies were undertaken to evaluate whether trunk control, changes in trunk control, and dynamic trunk positioning are associated with pEKAbM, a surrogate for ACL injury risk. Further, the trunk control strategy employed by the core muscles was evaluated for its effect on trunk control, dynamic trunk positioning, and pEKAbM.

1.6 Outline of Upcoming Chapters
Chapter 2 describes a randomized controlled trial in which two interventions, both very similar except for the inclusion or exclusion of focused trunk training, were used as a tool to elicit changes in trunk control. Changes in pEKAbM were also measured to determine if changes in trunk control were associated with changes in an ACL injury risk factor. Chapter 3 describes a study designed to determine the associated between...
pEKAbM and positioning of the trunk during a dynamic, side-step cutting task. Chapter 4 describes a study in which trunk muscles activation during both a quasi-static trunk control task and a side-step cutting task was compared to known surrogates for ACL injury risk. Chapter 5 presents a comprehensive discussion of Chapters 2-4 and outlines future work for this area.
2 RCT of the Effects of a Trunk Stabilization Program on Trunk Control and Knee Loading

2.1 Abstract

Background: Many athletic maneuvers involve coordination of movement between the lower and upper extremities, suggesting that better core muscle utilization may lead to improved athletic performance and reduced injury risk. Purpose: To determine to what extent a training program with focused trunk stabilization exercises would improve measures of core performance, leg strength, agility, and dynamic knee loading, compared to a program incorporating only resistance training. Methods: 37 male subjects were randomly assigned to either a resistance training only (RT) or a resistance and trunk stabilization (TS) training program, each lasting 6 weeks. Core strength and endurance, trunk control, knee loading during an unanticipated cut, leg strength, and agility were collected pre- and post-training. Results: The TS group significantly improved only core
endurance when compared to the RT group (side bridge; p=0.050). The TS group improved lateral core strength (MVIC, cable on non-dominant side; 44.5±61.3N, p=0.037). Both groups increased leg strength (estimated 1RM during deadlift; TS: 55.1±46.5lbs, p=0.003; RT: 33.4±17.5lbs, p<0.001) and decreased sagittal plane trunk control (sudden force release (SFR) test; cable in front; TS: 2.54±3.68°, p=0.045; RT: 3.47±2.83°, p=0.004), but only the RT group decreased lateral trunk control (SFR test; cable on dominant side; 1.36±1.65°, p=0.029). The RT group improved standing broad jump (73.2±108.4mm, p=0.049), but also showed increased knee abduction moment during the unanticipated cut (1.503 fold increase [%bw*ht], p=0.012). **Conclusions:** Trunk stabilization exercises improved core endurance relative to resistance training, but not core strength, trunk control or knee loading. The RT program resulted in worsened lateral trunk control and knee abduction moment. Together these results suggest the potential importance of targeted trunk control training to address these known ACL injury risk factors.

### 2.2 Introduction

Core strength and stability are popular terms in both the scientific and popular media because it is thought that increases in one or both will improve athletic performance as well as aid in the treatment and prevention of injury (Kibler et al., 2006; Schuler, 2011). Hodges and Richardson (Hodges and Richardson, 1997) reported that abdominal muscle activity precedes activity of the prime mover of the limb, providing a stable base for limb movement, which Kibler et al. (2006) referred to as “proximal stability for distal mobility.” The idea of the core contributing significantly to athletic function has construct validity, as many athletic movements require energy to be transferred from the
legs to the arms, or vice versa, through the core to complete the task. From a biomechanical standpoint, it is also reasonable to hypothesize that the core could influence lower extremity injury risk by altering loading since the core is responsible for positioning about half (Dempster and Gaughran, 1967) of an athlete’s body mass over the lower extremity at risk. Malalignment of this mass could increase adverse loading of the joints and other structures in the leg(s), increasing injury risk. However, due at least in part to a lack of data and the broad application of the terms core strength and stability, there has been confusion and conflicting evidence in the scientific literature regarding the relationships between measures of the core, performance and injury. For the purposes of this study, the “core” is defined as the region of the body bound by the pelvis and diaphragm and includes the muscles of the abdomen and lower back, while the “trunk,” or “torso,” is defined as encompassing the abdomen and thorax. “Trunk control” will be considered an aspect of core stability as the core is largely responsible for the movement and control of the trunk.

Links between measures of core stability and strength and athletic performance have been investigated, with conflicting results. Mills et al. (Mills et al., 2005) found that females who completed targeted core stabilization training improved their supine trunk control (Sahrmann test), lower-extremity agility (T-test), and vertical jump height; while females who completed a treatment program consisting of crunches and other more conventional abdominal exercises also improved their supine trunk control. Another intervention study utilizing targeted core muscle exercises on a Swiss Ball also demonstrated improvements in supine trunk control, but found that this training program did not improve running economy (Stanton et al., 2004). Both studies investigated
correlations between Sahrmann test scores and athletic performance measures, but found no evidence of an association. This result is perhaps not surprising given the lack of sensitivity of the Sahrmann test to discriminate within this relatively homogeneous population and the small changes between pre- and post-testing for each study.

Other studies attempted to connect abdominal endurance to measures of performance with varying results. Cross-sectional studies have found correlations between core endurance measures and 20-m sprint, 40-m sprint, short shuttle, vertical jump, 1 repetition maximum power clean, bench press, T-run agility test and maximum number of reps of a single leg squat (Nesser et al., 2008; Okada et al., 2011). In contrast, Tse et al. (Tse et al., 2005) reported that a training program focused on core endurance in collegiate rowing athletes improved side flexion endurance, but did not improve vertical jump, broad jump, shuttle run, 40-m sprint, overheard medicine ball throw, or 2,000-m max rowing ergometer test.

In addition to the above-mentioned reports that investigated the connection between measurements of the core and performance, a smaller number of studies have investigated the connection between the core and lower extremity injury, though these studies have only produced indirect associations. In a prospective cohort study of hip and core muscle strength and endurance that they considered measures of “core stability,” Leetun et al. (Leetun et al., 2004) found hip external rotation strength to be a predictor of lower extremity injury. Zazulak et al. (Zazulak et al., 2007a,b) defined “trunk control” as “the body’s ability to maintain or resume an equilibrium position of the trunk after perturbation” and found that deficits in lateral trunk control and trunk proprioception in the transverse plane each predicted future risk for knee injury in collegiate athletes.
However, these two studies do not address whether the core is contributing directly to the dynamic loading environment of the lower extremities. An alternative hypothesis is that these measures of the core are coincidentally associated with lower extremity strength, endurance, or neuromuscular control, and that these lower extremity factors alone are contributing to the differences in injury patterns. Using video analysis of anterior cruciate ligament (ACL) injuries, Hewett et al. (Hewett et al., 2009) suggested that lateral trunk lean was an important factor in the ACL injury mechanism in women.

Given that associations in cross-sectional and observational cohort studies cannot separate between a coincidental connection and a true causal relationship between measures of the core and lower extremity injury, intervention studies comparing various training programs are critical to improving our understanding of the relationship between the core musculature and ACL injury risk. As described in a recent meta-analysis of ACL injury prevention programs (Hewett et al., 2006), several studies have found success with ACL injury prevention programs that include exercises directed to core strength and stability (Heidt et al., 2000; Hewett et al., 1999; Mandelbaum et al., 2005; Myklebust et al., 2003; Petersen et al., 2005). Myer et al. (Myer et al., 2005) demonstrated that an intervention program that incorporated core strength and balance-training components reduced knee loading patterns implicated in ACL injury. However, the conservative design of these programs (i.e. including any and all exercises that might be helpful) has so far precluded the determination of whether the core-directed exercises provide any meaningful contribution to reducing injury risk or are merely an extraneous addition to the intervention. More controlled comparisons of exercise interventions with and without trunk training would be most useful in evaluating the relative contribution of trunk
control training to lower extremity mechanics, thereby addressing potential mechanisms by which core training could alter risk for injury.

To the authors’ knowledge there have been no comprehensive studies to date investigating the effectiveness of a training program including focused trunk stabilization exercises on improving lower extremity biomechanics, athletic performance and trunk or core measures (control, endurance, and strength). The primary hypothesis of this study was to compare changes in peak external knee abduction and tibial internal rotation moments, which have been previously associated with ACL strain (Fleming et al., 2001; Markolf et al., 1995; Shin et al., 2009; Shin et al., 2011) and injury (Hewett et al., 2005), between a training program incorporating focused trunk stabilization exercises and a traditional resistance training program. We also hypothesized that a trunk stabilization program would improve core endurance and strength, trunk control, lower extremity biomechanics and athletic performance, relative to a traditional resistance training program. Furthermore, we hypothesized that pre- to post-changes in each of these variables would occur within each exercise program.

2.3 Methods
2.3.1 Design
After providing IRB-approved informed consent, subjects were randomly assigned to either the treatment group or the control group using a permuted block randomization scheme. Testing personnel were blinded to the group assignment during both testing sessions. Subjects were blinded to the details of the other training program.
2.3.2 Subjects

An a priori sample size estimate was obtained based on a desired improvement of one standard deviation (effect size of 1.0) in the treatment group over the control group in the two primary outcomes of peak knee abduction and tibial internal rotation moments. A Bonferroni correction was used to control type I error (each outcome tested at $\alpha = 0.025$, 1-sided). Allowing for >80% power, it was estimated that 18 subjects per group would be needed in the final analyses. This was adjusted to 20 subjects per group to account for a predicted 10% dropout.

Thirty seven male subjects were initially enrolled in the study. All subjects completed an on-line survey to assess eligibility for participation. Eligibility requirements included participation in organized full-contact American football at the high school level within the last 5 years, no known history of ACL injury or prior surgery, no history of other serious lower extremity or torso injury, and a current Marx Activity Rating Scale score $\geq 9$ (Marx et al., 2001). Written informed consent was obtained from all subjects prior to initiating their participation. Subjects were randomly assigned to a 6-week resistance training (RT) program [control group] or a 6-week trunk stabilization (TS) program [treatment group]. All subjects completed testing of agility, dynamic biomechanical loading, trunk control, and core strength and endurance less than 14 days before starting the training program. One subject aggravated a previous foot injury during initial testing and therefore was not assigned an intervention, leaving 36 subjects randomized to an intervention. In order to remain an active participant in the study, subjects were permitted to miss at most 4 total sessions, and no more than 3 consecutive sessions. A pro-rated
payment schedule was used to induce participation, leading to a maximum possible payment of $300 upon completing the study.

Fourteen subjects withdrew from the study after being allocated to an intervention program: one due to the recurrence of a prior shoulder injury, three because of illnesses resulting in more than a week of missed training, four due to changes in their schedules reducing attendance below the required minimum, and 6 due to reasons unknown to the investigators. Therefore, 22 subjects completed all requirements (Figure 2.1).
Figure 2.1 Flow diagram of subject participation throughout all aspects of the study.

2.3.3 Testing

Pre-testing was completed prior to training during a single session no more than 14 days before the start of the interventions. Post-testing was completed within 14 days following training. Testing personnel were blinded to the intervention group assignment for both testing sessions.
**Biomechanical loading.** A lower-body Point Cluster Technique marker set and functional hip joint center estimation was used as previously described to estimate lower extremity kinematics (Andriacchi et al., 1998; Camomilla et al., 2006). An 8-camera motion capture system (Vicon MX-F40 @ 300 Hz) was used for motion tracking during a 45° unanticipated run-to-cut maneuver, planting on the dominant leg. We chose to study unanticipated cutting as this more closely resembles the environment experienced during sport, and differences have previously been reported between anticipated and unanticipated cutting (Besier et al., 2001a). Subjects started on a pressure sensitive mat, placed at a distance that allowed them to start from rest and take three steps to get up to a self-selected jogging speed before planting for their cut. Just before the subjects planted for their cut, an arrow pointing in one of two directions would illuminate, instructing the subjects to either continue running straight or to perform the cut. The direction of the arrow was chosen at random. For a trial to be accepted, subjects had to plant their foot within an area (80cm wide x 60cm long) defined by two force plates [Bertec 4060-10] placed side-by-side, and their change of direction had to take them over top of a line drawn on the ground at a 45° angle to the approach path, originating at the center of the force plates. The time delay between when subjects left the pressure sensitive mat and when the arrow would illuminate could be adjusted for each subject to account for speed and reaction time differences. Dominant foot was determined as the foot the subject would prefer to use to kick a ball. Peak knee abduction moments and tibial internal rotation moments were estimated using custom BodyBuilder and Matlab scripts for 3 trials during weight acceptance phase and normalized by body weight and height. We chose peak knee abduction and tibial internal rotation moments as our primary outcomes.
because previous in vitro and simulation studies have identified both moments as biomechanical predictors of ACL strain (Fleming et al., 2001; Markolf et al., 1995; Shin et al., 2009; Shin et al., 2011). Furthermore, a prospective in vivo study associated high knee abduction moments with future ACL injury (Hewett et al., 2005).

**Trunk control and core strength.** For the purposes of this study, the trunk is defined as the region encompassing the abdomen and thorax, and the definition of trunk control proposed by Zazulak et al. (Zazulak et al., 2007a), “the body’s ability to maintain or resume an equilibrium position of the trunk after perturbation,” was adopted. A sudden force release test device modeled after the one previously developed (Cholewicki and Vanvliet Iv, 2002; Radebold et al., 2000; Zazulak et al., 2007a) was utilized in order to test trunk control and core strength. The pelvis was secured with the subject in a semi-kneeling position. From this position, an instrumented cable attached to an upper body harness at approximately the level of the subject’s 10th thoracic vertebra ran over a pulley to an electromagnet and load cell (Omega Engineering, Inc., DMD-46). Core strength was assessed by instructing subjects to pull maximally against the cable. A custom LabView user interface was used to display and record the maximum and instantaneous forces generated by the subject. Following collection of core strength data, trunk control was assessed by instructing the subject to pull isometrically against the cable with 30% of their measured maximum force for that direction. The tester then deactivated the electromagnet at a random time between 1 and 5 seconds after steady state was achieved, releasing the tension in the cable and forcing the subject to react in an attempt to maintain the same trunk position. A marker placed at the level of the 7th cervical vertebra and 4 markers placed at the corners of the sudden force release device were used to calculate
the maximum angular displacement of the trunk after release using custom MATLAB scripts. Subjects performed 3 sudden force release trials in each of forward, backward, non-dominant, and dominant directions, with 15 seconds of rest between trials. Non-dominant and dominant sides were identified based on which leg the subject would preferentially use to kick a soccer ball a maximum distance. Order of directions was standardized (backward, right, front, left) to minimize error in the pre-to-post comparison. Because the location of the upper body harness was adjacent to the thoracolumbar junction, the muscles engaged during these trunk control and core strength tasks were primary those of the core rather than the upper torso.

**Core endurance.** Core endurance was assessed at pre and post testing by recording participant performance in side plank, prone plank, and trunk flexor exercises using adapted protocols from McGill et al. (Mcgill et al., 1999). Each test was performed once since these tests have been shown to have a reliability >0.97 (Mcgill et al., 1999). The side plank test was performed on a hard rubber mat. The feet were placed one on top of the other, right upper arm was perpendicular to the ground with the elbow resting on the mat, and the left arm was placed over the chest, with the left hand on the right shoulder. Time was stopped when any other part of the subject’s body touched the floor. The side plank test is “suggested to optimally challenge quadratus lumborum and the muscles of the anterolateral [core] wall” (Evans et al., 2007). For the prone plank, the subject lay prone off the edge of an exam table with the ASIS at the edge and heels under a metal frame attached to the floor. The test was started when the subject assumed a horizontal position, with arms crossed over the chest. A small wooden block with 3 bubble levels indicating ±10° and horizontal was placed between the shoulder blades. Time stopped
when the subject drifted 10° either above or below horizontal. The prone plank test, also
referred to as the Sorensen Test, provides a “global measure of back extension endurance
capacity” (Moreau et al., 2001). The flexor endurance test was performed on a lightly
padded exam table. Feet were placed flat on the table, knees and hips were at
approximately 90° of flexion. The subject started with their back against a wooden wedge
(76cm long) 60° from horizontal. Time began when the wedge was moved 10cm back
from the subject and ended when the subject’s back touched the wedge. Subjects were
not permitted to lean forward from their initial position during the test. The flexor
endurance test targets the muscles of the abdominal wall (Evans et al., 2007).

**Leg strength.** While maximizing participant leg strength was not a primary goal of
the interventions, it was assessed at pre- and post-testing to determine its relative
contribution to changes in biomechanical loading and athletic performance. Participant
leg strength was estimated using an equation \(1-{\text{RM}}=100\times{\text{rep}}\text{ wt}/(48.8+53.8\times e^{-0.075\times{\text{reps}}})\),
previously evaluated by LeSuer, et al. (Lesuer et al., 1997), involving variables of weight
used (rep wt) and repetitions (reps) completed during a deadlift exercise. The
repetition number and weight were obtained from the last set (third or fourth depending
on group assignment) of deadlift completed during the first and last training session. In
an effort to maximize participant safety, a deadlift exercise was chosen in favor of a
back-loaded squatting exercise. In order to assess strength in a functional movement, the
deadlift was chosen preferentially over a machine-oriented leg press.

**Athletic performance.** The 3-cone test, 20-yard short shuttle test, and standing broad
jump, all standardized components of the National Football League (NFL) Combine,
were completed by each subject at pre and post testing. All tests were administered as
described by the NFL (NFL, 2008). The 3-cone and 20-yard short shuttle tests both assess
agility, while the broad jump is intended as a measure of strength. All 3 tests require
whole-body coordination and stability to be performed optimally. After hearing the
instructions and seeing a demonstration, each subject was allowed two practice trials.
Each test was then performed once by each subject. Times for the 3-cone and 20-yard
short shuttle were determined from video-recordings. Broad jump distance was
determined from a measuring tape secured to the floor next to the subject. Each athlete
performed the 3 tests in the same order (3-cone test, 20-yard shuttle test, then broad
jump) to eliminate variability in performance based on the order of completion.

2.3.4 Training

All subjects were asked to attend three supervised 1-hour training sessions per week
for a total of six weeks. In order to minimize confounding variables, subjects were
instructed to make no adjustments to their level of cardiovascular activity performed
outside of the training sessions. Subjects self-reported outside training activity over the
preceding days at each supervised session.

Sessions for both groups began with ten minutes of dynamic movement warm-up
consisting of an inverted hamstring stretch with arm reach (10 per side), walking lunges
(10 per side), the elbow-to-instep stretch (10 per side), arm circles (10 large, 10 small),
and crossing arm swings (10 over/ 10 under). The RT group then completed 4 sets of
each resistance training exercise. In contrast, after the warm-up the TS group performed
15 minutes of trunk stabilization exercises. After the trunk stabilization exercises, the TS
subjects then completed the resistance training portion of the session but did only 3 sets
for each exercise.


**Trunk stabilization.** Subjects were first instructed how to maintain appropriate trunk posture during the exercises using the neutral position of the spine (Mcgill, 1998). Eight exercises which targeted the major muscles of the torso were selected based on a review of the literature (Table 2.1) (Akuthota and Nadler, 2004; Mcgill, 1998, 2009; Nadler et al., 2002). Each exercise was progressed through increased levels or “phases” of difficulty over the 6 week period. The eight trunk stabilization exercises were prone planks (6 phases), side-planks (5 phases), lunges (forward, reverse, and side; difficulty increased by holding heavier dumbbells), supine bridge (2 phases), hip abduction (2 phases), quadruped exercises (2 phases), sagittal abdominal curls (2 phases) and diagonal abdominal curls (2 phases). To fit within the 15-minute time limit and provide variety from day to day, only half of the exercises were performed at each session. Prone and side planks, lunges, and sagittal abdominal curls were performed at one session and the bridging, hip abduction, quadruped and diagonal abdominal curl exercises were performed at the next.
Table 2.1 List of exercises performed during training. Workouts #1 and #2 were performed on alternating days. The RT group performed 4 sets of each standard workout exercise. The TS group started with 4 exercises from the trunk stabilization program before completing 3 sets of each standard workout exercise. The first 4 and last 4 listed in the trunk stabilization program were grouped together and performed on alternating days. Subscripts indicate the number of phases for each exercise; † indicates difficulty was increased by holding heavier dumbbells.

<table>
<thead>
<tr>
<th>Workout #1</th>
<th>Workout #2</th>
<th>Trunk Stabilization</th>
</tr>
</thead>
<tbody>
<tr>
<td>bench press</td>
<td>front squat/back squat</td>
<td>prone planks⁶</td>
</tr>
<tr>
<td>deadlift</td>
<td>bent-over row</td>
<td>side planks⁵</td>
</tr>
<tr>
<td>pull-ups</td>
<td>Romanian deadlift</td>
<td>front, back, and side</td>
</tr>
<tr>
<td>double-leg hamstring</td>
<td>incline chest press</td>
<td>sagittal abdominal curls²</td>
</tr>
<tr>
<td>overhead press</td>
<td>step-ups</td>
<td>diagonal abdominal curls³</td>
</tr>
<tr>
<td>lat pulldown</td>
<td>single leg hamstring curls</td>
<td>hip abduction²</td>
</tr>
<tr>
<td>bicep curls</td>
<td>lying triceps extensions</td>
<td>quadruped exercises²</td>
</tr>
<tr>
<td></td>
<td></td>
<td>supine bridge²</td>
</tr>
</tbody>
</table>

**Resistance training.** The resistance training portion of the study was designed to focus on strengthening all major muscle groups, consistent with basic athletic training preparation. For subject safety, certain dynamic lifting movements commonly utilized in athletic training were excluded from use in this study. Two different pre-designated workouts were performed at alternating training sessions for the duration of the study (Table 2.1). Exercises for workout #1 included bench press, deadlift, pull-ups, double-leg hamstring curls, overhead press, lat pulldown, and bicep curls. Workout #2 consisted of front squat/back squat, bent-over row, Romanian deadlift, incline chest press, step-ups, single leg hamstring curls, and supine triceps extensions. Subjects were instructed to perform each lift using the greatest resistance that would allow completion of all required
repetitions for the day. As the program progressed, the desired number of repetitions was decreased from 10 to 8 to 6.

2.3.5 Data analysis

Paired two-tailed t-tests were used to evaluate change within a training group before and after the intervention for the strength and agility outcomes. In order to test for a differential effect between the training groups, a two-sample two-tailed t-test was used to compare the individual participant-level differences from baseline to post-intervention between the groups. A mixed effects modeling approach was used to examine the primary outcomes of knee abduction moment and tibia internal rotation moment. Both outcomes were log-transformed to satisfy the model assumptions of normality and homoscedasticity. Training group, assessment time point (baseline or post-intervention), cut angle, speed and the interaction between group and time point were included as fixed effects. A random intercept and a random effect for the comparison of interest, the interaction of training group and assessment time point, were included by participant. The level of significance for the two primary comparisons was set at 0.025, per the sample size estimation. All other analyses were done at \( \alpha = 0.05 \). Due to equipment problems during testing, pre-test biomechanical loading during the cut was not available for one TS subject.
2.4 Results

Body mass at pre-test (Table 2.2) did not differ significantly (p>0.05) between groups or between pre- and post-testing. A summary of all results, including between and within group analyses, can be found in Table 2.2.

Biomechanical loading. No differences were observed in the pre-post changes across the randomized groups for either of the primary outcomes (p=0.63 for knee abduction moment, p=0.15 for tibial internal rotation moment). However, the RT group showed increases in knee loading during the unanticipated cut. The RT significantly increased peak knee abduction moment (1.503 fold increase [%bw*ht], p=0.012). The RT group did not significantly change peak tibial internal rotation moment (1.124 fold increase [%bw*ht], p=0.617). The TS group did not significantly change peak knee abduction moment (1.349 fold increase [%bw*ht], p=0.116) or tibial internal rotation moment (0.651 fold increase [%bw*ht], p=0.110).

Core strength. Only the TS group increased core strength (Figure 2.2). The TS group significantly (p<0.05) increased lateral core strength when pulling toward the dominant side (cable attached to the non-dominant side; 44.5±61.3N, p=0.037). The TS group also showed a trend (p≤0.1) of increased core strength when flexing forward (cable attached to the back; 34.6±60.5N, p=0.087). Nonetheless, significant core strength differences were not observed between training groups in any direction (p=0.160-0.407).
Table 2.2 Summary results for all tests performed. All data are presented as mean ± SD except for knee moments, which are presented as geometric mean estimates (95% CI) at the mean speed and cut angle based on a mixed effects regression model. *p≤0.05. **p≤0.1.

<table>
<thead>
<tr>
<th>Demographics</th>
<th>Intervention Group</th>
<th>Pre-Test</th>
<th>Post-Test</th>
<th>Change (Δ=post-pre)</th>
<th>Effect Size</th>
<th>Δ p-value</th>
<th>ΔTS:ΔRT p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Age [yrs]</td>
<td>TS</td>
<td>20.5±1.2</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td></td>
<td>RT</td>
<td>20.3±1.5</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Height [m]</td>
<td>TS</td>
<td>1.82±0.06</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td></td>
<td>RT</td>
<td>1.81±0.05</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Mass [kg]</td>
<td>TS</td>
<td>82.8±7.6</td>
<td>83.1±6.8</td>
<td>0.3±1.5</td>
<td>-</td>
<td>0.622</td>
<td>0.547</td>
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<tr>
<td></td>
<td>RT</td>
<td>82.9±5.4</td>
<td>83.7±5.5</td>
<td>0.8±1.5</td>
<td>-</td>
<td>0.119</td>
<td></td>
</tr>
<tr>
<td>Peak Moments [%bw*ht]</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Knee Abduction</td>
<td>TS</td>
<td>2.15 (1.38-3.34)</td>
<td>2.9 (1.77-4.75)</td>
<td>34.9%</td>
<td>-</td>
<td>0.116</td>
<td>0.634</td>
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<tr>
<td></td>
<td>RT</td>
<td>3.03 (2.10-4.37)</td>
<td>4.56 (3.00-6.90)</td>
<td>50.3%</td>
<td>-</td>
<td>0.012*</td>
<td></td>
</tr>
<tr>
<td>Tibial Internal Rotation</td>
<td>TS</td>
<td>0.72 (0.46-1.14)</td>
<td>0.47 (0.22-0.99)</td>
<td>-34.9%</td>
<td>-</td>
<td>0.110</td>
<td>0.154</td>
</tr>
<tr>
<td>Core Strength [N]</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Back</td>
<td>TS</td>
<td>581.1±81.6</td>
<td>615.7±96.1</td>
<td>34.6±60.5</td>
<td>0.57</td>
<td>0.087**</td>
<td>0.407</td>
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<tr>
<td></td>
<td>RT</td>
<td>557.6±107.3</td>
<td>563.4±96.5</td>
<td>5.8±95.1</td>
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<tr>
<td>Dominant</td>
<td>TS</td>
<td>466.7±61.5</td>
<td>509.0±99.5</td>
<td>42.2±191.3</td>
<td>0.46</td>
<td>0.156</td>
<td>0.225</td>
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<tr>
<td></td>
<td>RT</td>
<td>495.6±92.7</td>
<td>492.0±119.9</td>
<td>-3.6±79.9</td>
<td>-0.05</td>
<td>0.885</td>
<td></td>
</tr>
<tr>
<td>Front</td>
<td>TS</td>
<td>702.1±214.6</td>
<td>745.6±192.6</td>
<td>43.5±100.1</td>
<td>0.43</td>
<td>0.180</td>
<td>0.160</td>
</tr>
<tr>
<td></td>
<td>RT</td>
<td>830.7±170.6</td>
<td>802.2±162.3</td>
<td>-28.5±129.1</td>
<td>-0.22</td>
<td>0.481</td>
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<td>Non-Dominant</td>
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<td>467.6±92.4</td>
<td>512.1±105.2</td>
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<tr>
<td>Side Bridge</td>
<td>TS</td>
<td>8.9±1.7</td>
<td>10.1±2.2</td>
<td>1.2±3.0</td>
<td>0.40</td>
<td>0.224</td>
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<td>8.8±2.3</td>
<td>10.2±3.8</td>
<td>1.4±2.8</td>
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<td>0.146</td>
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<td>Prone Plank</td>
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<td>5.0±1.8</td>
<td>5.6±1.7</td>
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<td>0.178</td>
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</tr>
<tr>
<td></td>
<td>RT</td>
<td>4.9±1.4</td>
<td>6.3±2.3</td>
<td>1.4±1.6</td>
<td>0.88</td>
<td>0.029*</td>
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<td>Flexor Endurance</td>
<td>TS</td>
<td>8.5±3.1</td>
<td>11.1±5.0</td>
<td>2.5±3.7</td>
<td>0.68</td>
<td>0.045*</td>
<td>0.529</td>
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<tr>
<td></td>
<td>RT</td>
<td>8.9±3.5</td>
<td>12.3±5.3</td>
<td>3.5±2.8</td>
<td>1.25</td>
<td>0.004*</td>
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<td>Leg Strength</td>
<td>DL 1RM [N]</td>
<td>887.6±204.8</td>
<td>1,133.5±354.5</td>
<td>245.9±207.1</td>
<td>1.18</td>
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<td>RT</td>
<td>789.3±72.4</td>
<td>938.3±85.8</td>
<td>149.0±77.8</td>
<td>1.91</td>
<td>&lt;0.001*</td>
<td></td>
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<tr>
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<td>3-Cone Drill [sec]</td>
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<td>0.01±0.28</td>
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<td>8.28±0.40</td>
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<td>-0.40</td>
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<tr>
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<td>4.95±0.10</td>
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<td>-0.53</td>
<td>0.108</td>
<td>0.851</td>
</tr>
<tr>
<td></td>
<td>RT</td>
<td>5.13±0.23</td>
<td>5.01±0.29</td>
<td>-0.12±0.21</td>
<td>-0.57</td>
<td>0.093**</td>
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<tr>
<td>Broad Jump [mm]</td>
<td>TS</td>
<td>2,397.0±220.6</td>
<td>2,422.4±218.8</td>
<td>25.4±71.3</td>
<td>0.36</td>
<td>0.265</td>
<td>0.236</td>
</tr>
<tr>
<td></td>
<td>RT</td>
<td>2,129.1±210.3</td>
<td>2,202.3±204.5</td>
<td>73.2±108.4</td>
<td>0.68</td>
<td>0.049*</td>
<td></td>
</tr>
</tbody>
</table>

32
Trunk control. Both groups increased the peak angle after release during the SFR test, i.e. worsened their trunk control (Figure 2.2). The TS group significantly increased peak trunk angle after release when the cable was attached in the front (2.54±3.68°, p=0.045) while the RT group significantly increased peak angle after release when the cable was attached to the dominant side (1.36±1.65°, p=0.029) and in front (3.47±2.83°, p=0.004). Trunk control changes were not significantly different between groups in any direction (p=0.332-0.850).
Core endurance. Only the TS group showed a trend toward increased core endurance (Figure 2.3). Core endurance times increased for both the prone plank (17.5±26.9s, p=0.056) and flexor endurance (85.1±152.9s, p=0.095) tests. Only the side bridge test showed a significant difference between the two groups with the TS group increasing endurance times while the RT group decreased times (ΔTS: 6.8±15.6s, p=0.187; ΔRT: -7.1±15.4s, p=0.157; ΔTS-ΔRT: 13.9s, p=0.050, effect size=0.825).

Figure 2.3 Changes (mean ± SD) in core endurance performance. A positive change indicates an increase from pre- to post-testing. The bracket for the side bridge test indicates a two-sample test. *p ≤0.05, **p ≤0.1.

Leg strength. Both groups significantly increased deadlift 1RM (TS: 55.1±46.5lbs, p=0.003; RT: 33.4±17.5lbs, p<0.001; Figure 2.4) although these increases were not significantly different from each other (p=0.162).
**Athletic performance.** Only the RT group experienced significant changes in athletic performance tests but these changes were not significantly different from the TS group (p=0.236-0.851). The RT group significantly increased broad jump distance (73.2±108.4mm, p=0.049; Figure 2.4) and showed a trend toward a decrease in short shuttle time (-0.1±0.2s, p=0.093, Table 2.2).

![Figure 2.4 Changes (mean ± SD) in estimated 1RM deadlift and maximum broad jump distance. A positive change indicates an increase from pre- to post-testing. *p≤0.05, **p≤0.1.](image)

2.5 Discussion

Significant between-group effects of training regimen were not observed for the primary outcomes, most likely due to the large variability between subjects and a lower-than desired sample size. Nevertheless, secondary within group analyses suggest that a traditional whole body resistance training program, like the one used by the RT group
herein, is associated with an increase in knee abduction moment during a 45°
unanticipated cut, even when accounting for approach speed and cut angle as covariates.
Increases in knee abduction moment may suggest an increased risk of ACL rupture as it
has been associated with an increase in ACL strain (Fleming et al., 2001; Shin et al.,
2009). In contrast, a significant increase in knee abduction moment was not seen in the
athletes who completed the TS program, suggesting that the trunk stabilization exercises
may limit any potential negative effects of the resistance training.

Neither training group demonstrated improved trunk control as measured by the SFR
test. The RT group actually showed a decrease in lateral control of the trunk while the TS
appeared to maintain lateral trunk control through the training period. The RT group also
exhibited an increase in peak knee abduction moment during the cutting task suggesting a
potential connection between lateral control of the trunk and knee loading during cutting.
This result is consistent with both Zazulak et al. (Zazulak et al., 2007a), who reported that
lateral trunk control was a predictor of knee injury risk, and Hewett et al. (Hewett et al.,
2009), who used observations of non-contact ACL injury events to conclude that lateral
trunk position may be an important factor in the ACL injury mechanism.

The findings that neither the RT nor the TS group demonstrated an improvement in
trunk control suggests that improvements in this measure cannot be attained with static
trunk stabilization exercises or general whole body resistance training; and that a more
task specific training program focusing on neuromuscular control of the trunk and lower
extremity may be necessary. Pedersen et al. (Pedersen et al., 2009) reported that
recreational soccer training of novice athletes, including repeated rapid perturbations of
the upper and lower body, significantly decreased both reaction time and distance moved
after sudden trunk loading compared to running and negative control groups. Even though their activity-level matched control group trained by running, an exercise thought to incorporate significant core muscle activity, only the group that routinely trained with the repeated rapid trunk loading experienced in soccer improved in sudden trunk loading response. Our findings complement this work, suggesting that an intermediate intervention directed towards trunk stabilization but without repeated rapid trunk loading was not successful in inducing the desired changes in sudden trunk loading response.

The TS group showed significant improvement in core strength and a trend toward improved endurance, while the RT group did not show improvement in either measure. This result is perhaps not surprising because the TS program incorporated side bridge and prone plank stabilization as well as sagittal plane and diagonal abdominal curls. Tse et al. (Tse et al., 2005) noted similar findings with a significant increase in side bridge endurance after a core endurance training program with no change in the control group. Interestingly, in the current study, only side bridge endurance change was significantly different between groups, although the lack of significant differences between training programs may be due to the small sample sizes and large variance observed between subjects for the other core strength and endurance tests. Anecdotally, the extremely large variances noted, especially in flexor endurance, appeared to the examiners to be at least partially attributable to differences in mindset between subjects. Many subjects reached volitional exhaustion before reaching the outward appearance of physical exhaustion that was reached in other strength and endurance tests, which calls into question the utility of these measures in this particular population for assessing athletes’ capabilities.
When considered altogether, the results of this study support the concept of specificity of training (Wilmore et al., 2008). Both groups improved in the deadlift 1RM, an exercise that each group included in their training, but only the TS group, which had focused trunk exercises incorporated into their training, improved in measures of core endurance and strength. An apparent negative effect was observed in the RT group for lateral trunk control and lower extremity loading during cutting, supporting previous literature reporting a link between lateral trunk control and lower extremity injury (Hewett et al., 2009; Zazulak et al., 2007a). In contrast, the TS program appeared to help subjects maintain, not improve, lateral trunk control and lower extremity loading during cutting. While the final sample size in this study was not large enough to prove this null hypothesis, the trend from these results nevertheless suggests that focused trunk stabilization may be able to help maintain levels of trunk control and lower extremity loading but may not be enough to result in statistical improvements in either.

The results of this study should be considered in light of its limitations. Each of the individual pre- and post-tests has its own limitations, and we direct the reader to the referenced articles for discussions of these tests (Andriacchi et al., 1998; Besier et al., 2001a; Besier et al., 2001b; Camomilla et al., 2006; Cholewicki and Vanvliet Iv, 2002; Evans et al., 2007; Lesuer et al., 1997; Mcgill et al., 1999; Moreau et al., 2001; Radebold et al., 2000; Zazulak et al., 2007a). One of the primary limitations of the intervention design in this study was that the trunk stabilization exercises performed at each training session may not have provided a sufficient stimulus to elicit the desired improvements in trunk control from pre- to post-testing in this population of healthy, athletic men who recently competed at the high school level in American football. However, informal
feedback from the subjects acknowledging that they found the exercises challenging suggests that the program was of appropriate difficulty. In addition, subjects did not exceed the most difficult progressions in the TS exercises, further supporting our conclusion that the TS exercises created an appropriate challenge for the athletes. The results of our study do support those of others suggesting that trunk specific training will improve aspects of core endurance but not performance (Tse et al., 2005) and that core endurance may not be correlated with performance measures (Nesser et al., 2008; Okada et al., 2011).

A second limitation of the intervention design in this study was that 18 sessions over a 6-week period may not be enough to elicit changes in trunk control or lower extremity moments in healthy, athletic men. While the athletes progressed through phases of the TS program, they did not reach the most difficult levels suggesting that more improvements may have been possible with a longer training period. A 6-week training program was chosen because other interventions of this duration have been shown to improve landing techniques, increase vertical jump height, reduce ACL injury risk, and improve scores on the Sahrmann test (Hewett et al., 1999; Hewett et al., 1996; Myer et al., 2005; Stanton et al., 2004). However, the soccer study performed by Pedersen et al. (Pedersen et al., 2009) included 16 weeks of training between the pre- and post-tests, indicating that a longer training program may be a consideration for future studies.

Finally, sample size is another limitation of our study. As stated previously, 18 subjects were needed per training group to obtain appropriate power to prove the null hypotheses (that training had no effect within each group). Because of the high number of dropouts only 11 subjects remained per group for the statistical analysis. However, the
statistically significant differences observed are nevertheless valid, as type I error is less than 5% for these conclusions. While we made every effort to recruit and retain the appropriate number of subjects, retention of the university student population in the summer cohort was hindered by subjects withdrawing to travel away from campus, to accept summer employment, or for other unreported reasons, while recruitment and initial enrollment of the autumn cohort were challenged by scheduling conflicts with classes and training facility availability.

2.6 Conclusions
A 6-week training program incorporating trunk stabilization exercises significantly improved only trunk endurance, not lower extremity biomechanics, trunk control, core strength, athletic performance or leg strength, when compared to a 6-week training program including just resistance training. However, a training program including only resistance training, without trunk stabilization exercises, was associated with an increase in knee loading during unanticipated cutting. In contrast, a training program with a trunk stabilization component appeared to help athletes maintain, but not reduce, knee loading. A similar result was seen for lateral trunk control as the traditional resistance training group showed a decrease in lateral trunk control while the group that incorporated trunk stabilization did not show a significant trunk control change. Both programs improved leg strength while only the trunk stabilization program was able to improve core strength and endurance. These results, when considered with previous work, suggest that the inclusion of trunk stabilization in dynamic whole-body activities and of sports-specific perturbation training for both the trunk and the lower extremities may be necessary to elicit desired improvements in trunk control and lower extremity biomechanics.
2.7 Acknowledgements
The authors would like to thank the NFL Charities Foundation and OSU Sports
Medicine Initiative Grants Program for funding this study.
Knee moments during run-to-cut maneuvers are associated with lateral trunk positioning

3.1 Abstract
Non-contact anterior cruciate ligament (ACL) injuries account for approximately 70% of ACL ruptures and often occur during a sudden change in direction or pivot. Decreased neuromuscular control of the trunk in a controlled perturbation task has previously been associated with ACL injury incidence, while knee abduction moments and tibial internal rotation moments have been associated with ACL strain and ACL injury incidence. In this study, the association between movement of the trunk during a run-to-cut maneuver and loading of the knee during the same activity was investigated. External knee moments and trunk angles were quantified during a run-to-cut maneuver for 29 individuals. The trunk angles examined were outside tilt (frontal plane angle of the torso from vertical), angle between the ground reaction force (GRF) and the torso in the plane containing the GRF and shoulders (torso-GRF_shoulders); and angle between
GRF and torso in the plane containing the GRF and pelvis (torso-GRF_pelvis). Significant positive associations were found between torso angles and peak knee abduction moments (outside tilt, p=0.002; and torso-GRF_shoulders, p=0.036) while a significant negative association was found between peak tibial internal rotation moment and outside tilt (p=0.021). Because the peaks of these moments occur at different times and minimal axial rotation moment is observed at peak knee abduction moment (-0.29±0.46 %BW*ht), the positive association between peak knee abduction moment and torso lean suggests that increasing torso lean may increase ACL load and risk of injury.

3.2 Introduction

Anterior cruciate ligament (ACL) rupture is one of the most common knee injuries for athletes competing in field and court sports. Hewett et al. reported that more than 2,200 ACL ruptures are expected to occur in female American National Collegiate Athletic Association (NCAA) athletes each year and that a conservative estimate for orthopaedic care for this group alone would amount to over $37 million (Hewett et al., 1999). When one considers that these injuries occur to males and females participating in sport at all levels from junior to elite to adult recreational, the total direct medical costs reach into the billions (Pearldiver, 2010a,b). The consequences of ACL rupture also go beyond the medical costs of initial treatment. Injured athletes also face the potential loss of the remainder of the season, long-term disability, increased risk of osteoarthritis, elevated pain levels during sport, and, for collegiate athletes, lowered academic performance and possible loss of a scholarship (Freedman et al., 1998; Ruiz et al., 2002).

Approximately 70% of ACL ruptures are non-contact in nature, meaning that they occur without a direct blow to the leg by an object or opposing player, and most of these
occur during sudden changes in direction and pivoting (Boden et al., 2000; Griffin et al., 2000; McNair et al., 1990). Previous studies have shown that both knee abduction moments and tibial internal rotation moments load the ACL and that these moments have an interaction effect at physiologic load levels, creating strains approaching the reported range of ACL rupture (Fleming et al., 2001; Kanamori et al., 2002; Shin et al., 2009; Shin et al., 2011). One prospective cohort study has shown that peak knee abduction moment during a drop-jump activity could predict ACL rupture with 73% specificity and 78% sensitivity in a population of 205 female athletes, further implicating increased peak knee abduction moments as a factor in ACL injury risk (Hewett et al., 2005).

A prospective study by Zazulak et al. found that deficits in lateral trunk control, measured using an isolated trunk perturbation test, were correlated with ACL injury incidence (Zazulak et al., 2007a). During this test the subject’s pelvis was secured to isolate the torso and upper extremity. The subject then isometrically pulled against a cable until the cable was randomly released. Those subjects with increased angular deviation of the trunk after the release were observed to have a higher relative risk of ACL injury than those that had better neuromuscular control of the trunk. While this study suggested that a relationship does exist between trunk control and ACL injury risk, it did not address the underlying cause of this observed relationship. One potential explanation for this result is that control of the trunk in the isolated trunk perturbation test is a good indication of dynamic trunk control during certain cutting and pivoting tasks and that the resulting trunk position directly influences the dynamic loading of the knee. However, an alternative explanation is that this relationship is merely coincidental—athletes who are in better overall physical condition may display better trunk control, but
they may be less apt to sustain an ACL injury due to lower-extremity factors that are also associated with being in better overall physical condition.

The goal of this study was to determine if control of the trunk during a run-to-cut maneuver was correlated with the loading of the knee during the same activity. It was hypothesized that increased torso motion away from the cutting direction would have a significant, positive association with both knee abduction moments and tibial internal rotation moments.

3.3 Methods

3.3.1 Subjects

Thirty healthy subjects participated in this study after providing IRB approved consent. Subjects had no prior history of lower extremity surgery or serious injury nor did they have a previous history of open abdominal surgery. All subjects were pain free at the time of testing and were fit enough to comfortably jog for more than 10 minutes. Excess torso marker occlusion prevented the calculation of torso angles for one female subject. The remaining 29 subjects (Table 3.1) were included in the analyses presented.
Table 3.1 Subject demographic data split by gender (15 males, 14 females).

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<td></td>
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<tr>
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<tr>
<td>F</td>
<td>25.5±6.2</td>
<td>20-42</td>
</tr>
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</tr>
<tr>
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<tr>
<td>F</td>
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<tr>
<td>F</td>
<td>60.7±6.0</td>
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3.3.2 Unanticipated run-to-cut maneuver

Subjects started on a pressure sensitive mat then ran 3 steps at a self-selected pace before planting their dominant foot within a target area defined by a force plate 40cm wide by 60cm long [Bertec 4060-10; Bertec Corp; Columbus, Ohio] embedded in the floor to record the ground reaction forces acting upon the foot during stance. During the approach, before the subjects planted for their cut, an arrow pointing either straight ahead or away from the planting foot, instructing the subjects to either continue running straight or to perform a side-step cut. The direction of the arrow (straight or cut) was chosen at random. A successful cut was one in which the plant foot was completely within the 40cm wide by 60cm long target area and the subject’s change of direction took him/her over top of a line drawn on the floor at a 45° angle to the approach direction, originating at the middle of the force plate and progressing away from the plant foot side. The distance between the pressure mat and the plant foot target as well as the time between when the subject left the pressure mat and when an arrow would be illuminated were adjusted for each subject to account for stride length, speed, and reaction time differences between subjects. Dominant foot was established as the foot the subject would prefer to
use to kick a ball. Figure 3.1 shows the laboratory set-up for the unanticipated cutting maneuver.

Figure 3.1 Unancipated run-to-cut set-up. Subject started on the pressure sensitive mat, took three steps at a self-selected jog pace before planting their fourth step within the force plate target area. A direction indicator arrow would illuminate before the plant indicating whether to perform a side-step cut or continue straight. Right and left foot plants are possible with this set-up.

3.3.3 Data collection
Marker data (Figure 3.2) was collected at 300 Hz using 8 Vicon MX-F40 cameras [Vicon; Oxford, UK] and filtered using Vicon Nexus’ Woltring filter with generalized
cross-validation (GCV) (Woltring, 1986). Ground reaction forces were sampled at 300Hz from a Bertec 4060-10 force plate [Bertec Corp; Columbus, Ohio].

### 3.3.4 External knee moments

The point-cluster technique (Andriacchi et al., 1998) (Figure 3.2), as well as custom Vicon BodyBuilder and MATLAB [MathWorks, Inc., Natick, MA] scripts, were used to calculate kinematics and kinetics of the lower extremities during the cut. Peak externally-applied knee abduction and tibial internal rotation moments (pKabM and pTIRM, respectively) were estimated during the weight acceptance phase of the cut since the majority of non-contact ACL injuries occur during this time (Boden et al., 2000). Weight acceptance was defined as the interval between initial contact and peak knee flexion. Knee moments were normalized by body weight and height [%BW*ht].

### 3.3.5 Torso angles

Upper body markers based on the Plug-In Gait marker set (Gutierrez et al., 2003) (Figure 3.2) were used to determine three torso angles (Figure 3.3). Because previous studies have implicated lateral trunk position and control as ACL injury risk factors (Hewett et al., 2009; Zazulak et al., 2007a), we examined frontal-plane torso angles. As the primary torso variable of interest, torso outside tilt was defined as the angle the torso deviated from vertical, with positive being a torso lean away from the direction of cutting (toward the plant foot side). Peak outside tilt was found during the weight acceptance phase of the cut (foot strike to peak knee flexion). Angles between the torso midline and the ground reaction force (GRF) were also calculated in two different planes to explore whether these different formulations of torso angle would result in different relationships to knee loading variables. A GRF_pelvis plane was defined by the normal to the cross
product of a vector from the midpoint of the left anterior and posterior superior iliac spines to those of the right and the GRF vector (Figure 3.3A, Plane zoa). A GRF_shoulders plane was defined by the normal to the cross product of a vector from the left to right shoulder marker and the GRF vector (Figure 3.3B, Plane zob). The torso midline was initially defined as the vector from the midpoint of the lower torso markers (sternum and T10) to the midpoint of the upper torso markers (clavicle and C7), and then projected into one of the two planes for angle calculations. The torso-GRF_pelvis angle was defined as the angle between the GRF and the projection of the torso midline into the GRF_pelvis plane. The torso-GRF_shoulders angle was defined as the angle between the GRF and the projection of the torso midline into the GRF_shoulders plane. Maximum torso-GRF angles were found during weight acceptance. Angles between the torso and GRF were chosen as the GRF is the decelerating force during the weight acceptance phase of the maneuver, when ACL injury occurs. Due to the relative rotation between the shoulders and the pelvis during a cutting task we chose to project the angle of the torso onto two frontal planes, one defined using the shoulders and the other using the pelvis, to examine the relationship between lateral trunk angles and knee moments.
Figure 3.2 Anterior (A) and posterior (B) views of the lower body Point-Cluster marker set combined with the torso portion of the Plug-In Gait marker set.
Figure 3.3 Posterior (A, B) and top views (C) of a subject performing the unanticipated run-to-cut maneuver, planting on their dominant limb, and changing directions to the left. Both the plane containing the GRF and the midpoints of the right and left pelvic markers (A; yellow and orange axes, Plane zoa) and the plane containing the GRF and the shoulder markers (B; red and orange axes, Plane zob) are depicted. The orange vector (\( \mathbf{z_0} \)) is common to both planes and is parallel to the GRF (green). The torso-GRF angles are the angles between \( \mathbf{z_0} \) and the spine projected into the respective plane (zoa: torso-GRF_pelvis, zob: torso_GRF_shoulders). Note the lag in upper torso rotation seen in (C) (angle aob). Torso lean away from vertical can also be observed in (A) and (B).

3.3.6 Statistics
The association between knee moments and torso angles was evaluated using linear mixed models where approach speed, cutting angle and gender were considered as covariates. A compound symmetry covariance structure was chosen to account for the correlation between the knee moments from the three trials for each subject. Peak knee abduction moment and peak tibial internal rotation moment were evaluated separately against torso outside tilt (primary comparisons) as well as the other two torso angles leading to a total of 6 models (2 moments x 3 torso angles). Sensitivity studies with other
covariance structures and data transformations were conducted and confirmed the conclusions are the same regardless of the covariance structure or data transformation. Reported p-values were not adjusted for multiple comparisons.

### 3.4 Results

Typical knee moment and trunk angle curves are presented in Figure 3.4. Average statistics and ranges for parameters used in the models are presented in Table 3.2. The results of the mixed effects models are summarized in Table 3.3. Significant positive associations were found between peak knee abduction moment (pKAbM) and outside tilt angle (p=0.002) and torso-GRF_shoulders angle (p=0.036). Peak tibial internal rotation moment (pTIRM) was found to have a significant negative association with outside tilt angle (p=0.021) and a trend towards a negative association with torso-GRF_pelvis angle (p=0.0642). There was no evidence for a significant association between the other knee moment - torso angle combinations. The association between knee moments and approach speed varied between significant, marginally significant, and not significant in the models (p=0.043-0.167) but was always positive. Cut angle showed a significant positive association in all models (p=0.009-0.035). Gender was not a significant factor in any of the models (p=0.377-0.982).
Figure 3.4 Typical externally-applied knee abduction (A) and tibial internal rotation (B) moment curves as well as trunk angles (outside tilt (C), torso-GRF_shoulders (D), and torso-GRF_pelvis (E) angles) from midswing to midswing of the planting foot. The shaded regions represent weight acceptance, defined as the interval from initial contact (A-C) or when the vertical ground reaction force (vGRF) first exceeded 10% of a subject’s body weight (>10%BW) (D, E) to peak knee flexion. Peak values during weight acceptance were used for subsequent analysis. Vertical dashed lines mark the maximum position of the curve in the region of interest.
Table 3.2 Means, standard deviations, and ranges for variables of interest split by gender (15 males, 14 females).

<table>
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<tr>
<th>Variable</th>
<th>mean±stdev</th>
<th>range</th>
</tr>
</thead>
<tbody>
<tr>
<td>pKAbM [%BW*ht]</td>
<td></td>
<td></td>
</tr>
<tr>
<td>M</td>
<td>3.09±2.30</td>
<td>-0.30-9.58</td>
</tr>
<tr>
<td>F</td>
<td>3.01±2.66</td>
<td>0.37-11.23</td>
</tr>
<tr>
<td>pTIRM [%BW*ht]</td>
<td></td>
<td></td>
</tr>
<tr>
<td>M</td>
<td>0.93±0.68</td>
<td>-0.01-2.85</td>
</tr>
<tr>
<td>F</td>
<td>0.87±0.50</td>
<td>0.09-1.89</td>
</tr>
<tr>
<td>outside tilt [deg]</td>
<td></td>
<td></td>
</tr>
<tr>
<td>M</td>
<td>10.3±10.1</td>
<td>-6.0-34.7</td>
</tr>
<tr>
<td>F</td>
<td>6.4±3.3</td>
<td>-0.57-13.6</td>
</tr>
<tr>
<td>torso-GRF_shoulders angle [deg]</td>
<td></td>
<td></td>
</tr>
<tr>
<td>M</td>
<td>25.3±4.9</td>
<td>13.0-33.4</td>
</tr>
<tr>
<td>F</td>
<td>24.8±4.2</td>
<td>15.3-34.5</td>
</tr>
<tr>
<td>torso-GRF_pelvis angle [deg]</td>
<td></td>
<td></td>
</tr>
<tr>
<td>M</td>
<td>33.4±9.4</td>
<td>15.7-57.5</td>
</tr>
<tr>
<td>F</td>
<td>31.1±7.7</td>
<td>17.2-49.5</td>
</tr>
<tr>
<td>Approach Speed [m/s]</td>
<td></td>
<td></td>
</tr>
<tr>
<td>M</td>
<td>2.85±0.31</td>
<td>2.29-3.68</td>
</tr>
<tr>
<td>F</td>
<td>2.66±0.27</td>
<td>2.23-3.10</td>
</tr>
<tr>
<td>Cut Angle [deg]</td>
<td></td>
<td></td>
</tr>
<tr>
<td>M</td>
<td>44.3±5.5</td>
<td>34.3-63.0</td>
</tr>
<tr>
<td>F</td>
<td>45.6±4.2</td>
<td>35.0-53.0</td>
</tr>
</tbody>
</table>
Table 3.3 Slope [%BW*ht per degree (°)] and p-values for interactions between normalized knee moments [%BW*ht] and torso angles [°] during a 45° unanticipated run-to-cut maneuver. Speed, cut angle, and gender were considered as covariates in the mixed effects model. Significant (p<0.05) values bolded and denoted with †. Marginally significant values (p<0.1) denoted with ‡.

<table>
<thead>
<tr>
<th>Trunk Angle</th>
<th>Slope</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Peak knee abduction moment</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>outside tilt</td>
<td>0.13†</td>
<td>0.002</td>
</tr>
<tr>
<td>torso-GRF_pelvis</td>
<td>0.05</td>
<td>0.181</td>
</tr>
<tr>
<td>torso-GRF_shoulders</td>
<td>0.12†</td>
<td>0.036</td>
</tr>
<tr>
<td><strong>Peak tibial internal rotation moment</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>outside tilt</td>
<td>-0.03†</td>
<td>0.021</td>
</tr>
<tr>
<td>torso-GRF_pelvis</td>
<td>-0.02‡</td>
<td>0.064</td>
</tr>
<tr>
<td>torso-GRF_shoulders</td>
<td>-0.02</td>
<td>0.272</td>
</tr>
</tbody>
</table>

3.5 Discussion
The results of this study supported our hypothesis that knee abduction moments would have a positive association with torso motion away from the cutting direction, but did not support our hypothesis that tibial internal rotation moments would also have a positive association with torso motion away from the cutting direction. Significant positive associations between pKAbM and outside tilt and torso-GRF_shoulders angles suggest that as the torso moves away from the cutting direction peak knee abduction moment increases, potentially increasing ACL load and risk of ACL injury (Fleming et al., 2001; Hewett et al., 2005; Kanamori et al., 2002; Shin et al., 2009; Shin et al., 2011). Negative associations between pTIRM and outside tilt and torso-GRF_pelvis angles suggest that as the torso moves away from the cutting direction peak tibial internal...
rotation moment decreases, potentially decreasing risk of ACL injury (Kanamori et al., 2002; Shin et al., 2011).

The significant positive associations between pKAbM and torso angles indicate that an increased lean of the torso away from the cutting direction increases peak knee abduction moments. This effect is expected, because deviation of the torso from vertical causes the GRF to pass more lateral of the knee, thereby increasing its moment arm and the resulting knee abduction moment. Significant negative associations between pTIRM and torso angles indicate that an increased lean of the torso away from the cutting direction decreases the peak tibial internal rotation moment. This effect of an increased lean should act to pull the GRF and center of pressure laterally, decreasing the moment arm that would create a tibial internal rotation moment.

Even though the negative association between torso angles and pTIRM moments might suggest a protective effect of torso lean on ACL injury risk, the positive association between pKAbM and torso lean is of most importance because the peaks of the two moments do not occur at the same time (e.g. Figure 3.4). Tibial internal rotation moments are roughly neutral at pKAbM (average TIRM @ pKAbM, -0.29±0.46%) suggesting the loading pattern experienced by the ACL is minimally affected by the tibial internal rotation moment. Therefore, an increase in torso lean may increase ACL load and risk of injury, despite the reduction in pTIRM at another time point during weight acceptance.

One limitation of this study is that it does not provide insight into the causes for the increased torso lean observed in some subjects. One possible explanation is a lack of control of the trunk. In this scenario the pelvis and lower extremities progress toward the
cutting direction, but the trunk musculature does not activate in such a way to keep the torso in line with the lower extremity. Without this proactive strategy, the torso could lag behind the pelvis, resulting in torso lean away from the cutting direction. Another explanation is that the subject is utilizing an active control strategy to lean away from the cutting direction. It could be that the subject is trying to counterbalance the lower body which is moving in the direction of the cut by moving the torso in the opposite direction. Subjects could also be employing this active control strategy in an effort to maximize power generation similar to the way that in a countermovement jump subjects bend at the hip, knee and ankle before extending in the jumping motion. Future studies should examine the root cause of the increased trunk angles using electromyography or other means to quantify trunk muscle activation.

Another limitation of the study was that it was only powered to examine the relationship between the primary torso variable of interest, torso outside tilt, vs. each peak knee moment. Torso-GRF_pelvis and Torso-GRF_shoulders angles were included as secondary measures of torso angle in this analysis. As this study was the first to examine these variables in a biomechanical laboratory study, no correction for multiple comparisons was applied. Therefore, the results from this initial study should be regarded as preliminary until confirmed by future studies.

### 3.6 Conclusions

This study observed an association between frontal-plane trunk lean and knee loading, providing evidence of a direct mechanical connection between dynamic trunk control and known risk factors for ACL injury. Further work is necessary to fully characterize the mechanical relationship between movement and control of the trunk and
loading on the knee, but in the future it may be possible to design exercise programs that
target both the trunk and lower extremities to reduce ACL injury risk.

3.7 Acknowledgements

The authors gratefully acknowledge financial support from the OSU Sports Medicine
Initiative Grants Program and the NFL Charities Foundation.
The influence of core muscle activation on trunk control and dynamic knee loading

4.1 Abstract

Background: Anterior cruciate ligament (ACL) rupture is a common serious injury experienced by field and court sport athletes. Trunk control and peak external knee abduction moment (pEKAbM) have been associated with ACL injury and ACL strain, respectively. Trunk positioning, in the form of peak outside tilt during cutting, has been associated with pEKAbM. To address ACL injury risk through altered trunk control, core muscle activation during cutting and an isolated trunk control task must be better understood. Purpose: To investigate whether preactivation of the core muscles will influence variables associated with ACL injury or injury risk; namely trunk control, peak outside tilt of the trunk during cutting, and peak external knee abduction moment (pEKAbM) during cutting. Methods: Forty-six subjects completed unanticipated cutting and trunk control assessment tasks. Average co-contraction indices (avgCCI) and percent differences (avg%DIFF) were calculated for internal obliques (IO), external obliques (EO), and L5 extensors (L5) just prior to planting for a cut and before release during the trunk control task. pEKAbM and peak outside tilt of the trunk were also calculated for the cut while peak lateral angular displacement was found during the trunk control task.
**Results:** A significant negative association was found between pEKAbM and avgCCI_IO (p=0.048) while a significant positive association was found between pEKAbM and avgCCI_L5 (p=0.005). A significant positive association was found between peak angular trunk displacement during the trunk control task and avg%DIFF_L5 (p=0.014). **Conclusions:** Increased co-contraction of the L5 extensors, stiffening the spine, may contribute to increased risk of ACL injury risk by elevating adverse loading at the knee. A reduced side-to-side activation difference of the L5 extensors may improve trunk control. Future work is necessary to determine if intervention programs aimed at reducing ACL injury rates should consider addressing core muscle activation and trunk positioning.

**4.2 Introduction**

Anterior cruciate ligament (ACL) injury is a common, serious injury among field and court sport athletes. Consequences of ACL rupture can include missing the rest of the season, decreased academic performance for collegiate athletes (Freedman et al., 1998), persistent pain even following extensive rehab, and increased risk of developing osteoarthritis earlier in life (Ruiz et al., 2002). Because an estimated 70% of ACL injuries are non-contact in nature (Griffin et al., 2000), often the result of sudden changes in direction or pivoting (Boden et al., 2000; Mcnair et al., 1990), it is possible that many of them may be preventable. For the current investigation, peak external knee abduction moment (pEKAbM) will be used as a surrogate for ACL injury risk because increased pEKAbM has been implicated in ACL injury risk (Hewett et al., 2005) and ACL strain (Fleming et al., 2001; Markolf et al., 1995; Shin et al., 2009; Shin et al., 2011).
Core stability is a fashionable topic in the scientific literature and in the popular press because many think that core stability has implications for injury prevention, athletic performance and rehabilitation (Kibler et al., 2006; Schuler, 2011). Hodges and Richardson demonstrated that muscles of the abdomen activate prior to the activation of the prime movers of the limb (Hodges and Richardson, 1997). The hypothesis is that this core muscle activation creates a stable base for the movement of the limb, which Kibler et al. referred to as “proximal stability for distal mobility” (Kibler et al., 2006). However, due at least in part to a lack of data and the broad application of the term core stability, there has been confusion and conflicting evidence in the scientific literature regarding the relationships between measures of the core, performance and injury. For the purposes of this study, the core will be defined as \textit{the region of the body bounded by the pelvis and diaphragm which includes the muscles of the abdomen and lower back}. The trunk, a combination of the core and the thorax, will be defined as \textit{the region of the body bounded by the pelvis and the clavicles}.

The core is responsible for the positioning of more than half the body’s mass over the lower extremity when erect. Misplacement of this mass could lead to undesirable loading of the lower extremities. In fact, previous research has shown that increases in lateral trunk lean are associated with increases in adverse loading at the knee (increased pEKA\textsubscript{bM}) during a cutting task (Jamison et al., 2012b; Chapter 3). Furthermore, another study found that adverse loading of the knee during a cutting task increased when subject’s arms were constrained by either a football or lacrosse stick (Chaudhari et al., 2005). While this last study did not measure trunk position, it does suggest that the position of the body over the lower extremity can influence lower extremity loading.
In a prospective study, trunk control, defined in the study as “the body’s ability to maintain or resume an equilibrium position of the trunk after perturbation,” was assessed using a sudden force release device and then injury history was tracked for 3 years (Zazulak et al., 2007a). The athletes that had poorer trunk control were more likely to sustain an ACL tear during the follow-up period than those whose trunk control was better, further supporting the hypothesis that control of the trunk may be associated with lower extremity injury risk.

A recent intervention study also provides evidence that the core may be related to ACL injury risk factors. In this randomized control trial, subjects were allocated to either a resistance training only program or a trunk stabilization program, with the main difference between the two programs being the inclusion or exclusion of 15 minutes of trunk stabilization exercises (Jamison et al., 2012a; Chapter 2). The authors found that following the resistance training only program a significant worsening of lateral trunk control and knee loading (pEKAbM) while cutting were observed. In contrast, there was no evidence that the trunk stabilization program altered either of these outcomes. More work is needed in this area, but this evidence points again to an association between the core and lower extremity loading and potential injury risk.

This cross-sectional, prospective, and intervention based evidence links lateral control and position of the trunk with ACL injury or pEKAbM, a surrogate for ACL injury. For this reason, lateral trunk control, as defined by Zazulak et al. (2007), and lateral trunk positioning, as measured by Jamison et al.(2012b; Chapter 3) will be examined in the present study.
While some evidence exists associating the core with lower extremity loading and injury, the root cause still has yet to be investigated. The way in which the musculature of the core acts to control the trunk is an important piece of the puzzle as this information could help guide the development of efficient and effective intervention programs that improve trunk control and reduce injury risk.

The preparatory, or anticipatory, response of the core muscles may be of most importance when preventing ACL injury. Previous reports estimate that ACL rupture occurs within the first 40ms after initial contact of the foot with the ground (Koga et al., 2010; Shin et al., 2007) which is faster than reported muscle reaction times during a sudden loading task (Skotte et al., 2004). Because the body does not have enough time to react after a perturbation (whether due to landing or other external load), the state of the system at the time of the perturbation (i.e. preactivation of the muscles) may play a significant role on the response observed. Furthermore, because both the right and left sides of the core act in concert to support and control the trunk, the relative activations of the muscles on both sides of the body should be considered.

Preactivation of the core muscles can act to stiffen the spinal cord, possibly reducing motion of the trunk. This activation strategy might present itself as an increase in co-contraction, i.e. an increase in the activation of both muscles being considered. Preactivation of the core muscles can also act to pull the trunk laterally. This activation strategy might present itself as an increased difference in the activations of the muscles being considered. If the core muscles on the right side of the core are activated more than muscles on the left, the trunk may be pulled toward the right side.
The purpose of the present study is to investigate whether preactivation of the core muscles will influence variables associated with ACL injury or injury risk; namely trunk control, trunk positioning during cutting, and pEKBaM during cutting. Specifically, we hypothesized that increased co-contraction of the core muscles would lead to decreased trunk motion for both a trunk control assessment task and while cutting. We also hypothesized that a decrease in the difference in side-to-side core muscle activation would lead to decreased trunk motion during both tasks. Because lateral trunk position has a positive association with pEKBaM (Chapter 3; Jamison et al., 2012b), we hypothesized that decreased pEKBaM would be associated with increased co-contraction and decreased side-to-side differences in the activation of the core muscles during a cutting task.

4.3 Methods

4.3.1 Subjects

46 subjects (23M, 23F; height=1.75±0.98m; mass=70.9±12.5kg; age=22.9±4.2yrs) participated in this study after providing IRB approved consent. Subjects had no prior history of lower extremity or abdominal surgery. Subjects had no history of serious lower extremity injury (ACL, ligament, tendon, muscle, or meniscus tear) or abdominal hernia. Subjects with a BMI greater than 30 were excluded. Subjects had to have been active in athletic participation at least 3 times a week for the previous 3 months before testing and have a Tegner activity score greater than 3 (Tegner and Lysholm, 1985). Requiring a Tegner score of greater than 3 ensured that, at minimum, subjects cycled, cross-country skied, or ran on even ground at least twice a week.
4.3.2 Unanticipated run-to-cut maneuver

An 8-camera passive marker motion capture system (MX-F40; Vicon; Oxford, UK) was used to track marker positions during a 45° unanticipated run-to-cut maneuver performed on the dominant foot. Unanticipated cutting was observed because it more closely matches the environment experienced during sport, when players have to react to changing game situations. Furthermore, differences have been reported between anticipated and unanticipated cutting (Besier et al., 2001a). For the cut, subjects started on a pressure sensitive mat, took three steps at a self-selected pace, then planted their dominant foot within an area (80cm wide x 60cm long) defined by two force plates (Bertec 4060-10; Bertec Corp; Columbus, OH) placed side-by-side (Figure 4.1). The distance between the starting mat and the force plate was customized so that each subject could maintain a natural stride and strike the force plate with the desired foot. At a set delay after leaving the start mat, one of two arrows, chosen at random, would illuminate instructing the subject to either continue to run straight ahead or to perform the side-step cut. A successful cut was one in which the subject planted his/her dominant foot within the area of the force plates and his/her new direction lay over a line drawn on the ground at a 45° angle to the approach path, originating at the center of the force plates. The time delay between the subject leaving the mat and arrow illumination was adjusted for each subject to account for speed and reaction time differences. The dominant foot was the one which the subject would use to maximally kick a ball. Four successful cuts were recorded and used for analysis.
Figure 4.1 Unanticipated run-to-cut maneuver set-up. Subjects started on the pressure sensitive start mat, took three steps at a self-selected jog pace, then planted their dominant foot within the target area. A direction indicator would light up during the approach indicating whether to continue to run straight or to perform the side-step cut. Direction was chosen at random. Both right and left foot plants are possible with this set-up.

Subjects wore standard compression shorts (Figure 4.2) provided by the lab requiring some thigh and pelvis makers to be placed on the shorts instead of the skin. Subjects performed a series of squats, lunges and other activities before marker placement to get the shorts to “settle” and reduce movement during testing.
Figure 4.2 Anterior and posterior views of the lower body Point-Cluster marker set combined with the torso portion of the Plug-In Gait marker set. EMG electrodes for external obliques and L5 extensors (internal oblique electrodes hidden by shorts). Transmitters attached to electrodes were taped to keep them secured.

Marker data was captured at 300Hz and filtered using Vicon Nexus’ Woltring filter with generalized cross-validation (GCV) (Woltring, 1986). Ground reaction forces were sampled at 300Hz to match the marker capture rate.
**External knee abduction moment.** During the unanticipated cut, a lower-body Point Cluster Technique marker set (Figure 4.2) was used in conjunction with a functional hip joint center estimation to estimate lower extremity kinematics as previous described (Andriacchi et al., 1998; Camomilla et al., 2006). External knee abduction moments were estimated using custom Vicon BodyBuilder and MATLAB (MathWorks, Inc., Natick, MA) scripts and normalized to body weight and height. The peak external knee abduction moment (pEKAbM) was found during the weight acceptance phase of the cut (initial contact and peak knee flexion) for 4 trials.

**Outside Tilt.** During the unanticipated cut, the upper body Plug-In Gait marker set (Gutierrez et al., 2003) (Figure 4.2) was used to determine the peak frontal plane torso angle. Outside tilt was defined as the angle between the torso and vertical, with positive being a torso lean away from the cutting direction (or toward the plant foot side). Peak outside tilt was found during the weight acceptance phase of the cut (foot strike to peak knee flexion). This is the same procedure used by Jamison et al. (2012b, Chapter 3) when associating frontal plane trunk angles and pEKAbM.

### 4.3.3 Lateral Trunk Control

For the purposes of this study, trunk control was defined as “the body’s ability to maintain or resume an equilibrium position of the trunk after perturbation” (Zazulak et al., 2007a) and was assessed using a sudden force release device (SFR; Figure 4.3) modeled after one previously developed to assess trunk control (Cholewicki and Vanvliet Iv, 2002; Radebold et al., 2000; Radebold et al., 2001; Zazulak et al., 2007a). For the test, the subject is in a semi-kneeling position with the pelvis secured. An instrumented cable is then attached laterally at approximately the 10th thoracic vertebra to an upper
body harness on the non-dominant side. The cable runs over a pulley and is attached to an electromagnet and load cell in series (DMD-46; Omega Engineering, Inc; Stamford, CT). To assess trunk control, subjects isometrically pulled against the cable with a previously prescribed force (108N-males; 72N-females) (Zazulak et al., 2007a). A custom LabView program displayed the target and real time force output to help the subject achieve the desired force. At a random time interval (between 1 and 5 seconds) after the subject had stabilized at the desired force output, the electromagnet was deactivated, releasing the tension in the cable, and forcing the subject to react in an attempt to remain vertical or regain vertical equilibrium. Three trials were collected and used for analysis.

Figure 4.3 Apparatus for obtaining core muscle maximum voluntary isometric contractions (MVIC) and testing lateral trunk control. Loads can be applied to resist flexion (A), extension (B), or lateral bending (C) to both sides. For MVICs, the subject pulls maximally against the cable while the magnet remains engaged. When testing lateral trunk control, the subject pulls isometrically against the cable, attached to either the left or right side, to maintain a desired force. The magnet is disengaged, releasing the tension in the cable, after a random duration of time. Retro-reflective markers on the subject and the apparatus are used to measure the subject’s trunk displacement in response to the force release. (A) reproduced from Radebold et al., 2001.
A marker placed at the level of the 7th cervical vertebra and 4 markers placed on the testing device were recorded using the same passive marker motion capture system used for cutting, though the capture rate was reduced to 150Hz. Custom MATLAB scripts were used to calculate the maximum lateral angular displacement of the trunk after release. Three trials were collected when the cable was attached to the non-dominant side of the subject because the subject needs to use their non-dominant side core muscles to react to the sudden release of force in this test as well as to react to a 45° side-step cut when planting with their dominant foot.

Figure 4.4 Sudden force release (A) and cutting (B) for a right-foot dominant subject. When the cable is attached to the subject’s non-dominant side (left side) during the SFR task, the subject must react to and pull his trunk left (yellow arrow) after the cable release to stop motion and return to vertical. When the subject plants with his dominant foot (right foot) to perform a cut, he must also pull his trunk left (yellow arrow) to complete the maneuver.
### 4.3.4 Electromyography

Trunk muscle activation was quantified using wireless surface electromyography (EMG) (Noraxon USA, Inc; Scottsdale, AZ) during both the trunk control and unanticipated cut tests. Surface electrodes (Vermed, Inc; Bellows Falls, VT) were placed bilaterally over the internal obliques (IO), external obliques (EO), and L5 extensors (L5) according to McGill et al. (1996) (Figure 4.5). All electrodes were pre-gelled Ag/AgCl and were oriented parallel to muscle fibers. Electrode locations were shaved then lightly abraded and cleaned with alcohol pads. At IO and L5 muscles, two single electrode discs (A10005; 38.1mm diameter) were used with the end opposite the pull-tab trimmed to allow a closer interelectrode distance (30.5mm) (Figure 4.5, right). Dual electrodes were used for the EO muscles (A10011; 41.275x82.55mm) with an interelectrode distance of 42mm (Figure 4.5, left).

![Figure 4.5 Electrodes, wires, and transmitters for external obliques and L5 extensors (internal oblique electrodes hidden by shorts). Transmitters attached to electrodes were taped to keep them secured.](image)

Raw EMG data (recorded at 1500 Hz) were high-pass filtered at 10Hz with a zero-lag 4th order Butterworth filter to remove motion artifact. Filtered EMG data were then full wave rectified and smoothed using an RMS filter with a 20ms window. The RMS data
was then normalized by the maximum voluntary isometric contraction (MVIC) for each muscle to obtain continuous EMG curves of %MVIC.

MVIC trials were performed for trunk flexion, extension, and lateral bending to each side while the subject was in the SFR testing apparatus. Subjects pulled maximimally and isometrically against the cable, with the magnet engaged the entire time. MVIC trials were also filtered at 10Hz, rectified and smoothed using an RMS filter with a 20ms window. A 500ms running average of the RMS smoothed data was calculated for each muscle during each of the 4 MVIC trials. The highest 500ms running average, in any direction, was used for normalization.

As mentioned previously in the introduction, the anticipatory or preparatory response of the core before a perturbation may be the most important due to the lack of time to respond between the perturbation and the injury incident. EMG data is inherently noisy, even after filtering and smoothing, so the average activation over a short time period before the perturbation event (hereafter denoted as a “preactivation interval”) was used for analysis. For the unanticipated cut, the perturbation of interest is the initial contact of the plant foot with the ground. In order to standardize the preactivation interval for analysis, the flight phase (interval between contralateral toe-off to initial contact for the plant foot; 88.1±57.1ms) was used. For the trunk control task (SFR), the perturbation event is the cable release. Because the trunk control task is much more controlled, with all subjects pulling isometrically against the cable for at least one second prior to release, a standard preactivation interval length of 100ms preceding release was used. A 100ms time interval was used because it was similar to the duration of the preactivation interval for the cutting task (88.1±57.1ms).
Careful quality control measures were implemented to ensure only data of the highest quality was used in the analyses. This was accomplished by manual inspection of each EMG averaging section used for all muscles and all activities. EMG of questionable quality (inappropriate magnitude suggesting signal was due to movement of electrodes, missing data due to loss of connection between wireless transmitters and receiver during testing, etc.) was marked as invalid and withheld from subsequent analysis. 234 of the possible 1104 (46 subjects x 6 muscles x 4 trials) EMG sections were removed for the cutting task while only 4 of the possible 828 (46 subjects x 6 muscles x 3 trials) EMG sections were removed for the trunk control task.

For the remaining EMG sections, average co-contraction indices (avgCCI; Equation 4-1) as well as average percent differences (avg%DIFF; Equation 4-2) were calculated for each of the three muscle pairs (external obliques, internal obliques, and L5 extensors).

\[
avgCCI = \frac{1}{n} \sum_{i=1}^{n} \left( \frac{lower\ EMG_i}{higher\ EMG_i} \times (lower\ EMG_i + higher\ EMG_i) \right)
\]
Equation 4.1

\[
avg\%DIFF = \frac{1}{n} \sum_{i=1}^{n} \left( \frac{higher\ EMG_i - lower\ EMG_i}{(higher\ EMG_i + lower\ EMG_i)/2} \right) \times 100\%
\]
\[
where\ the\ interval\ of\ interest\ is\ a\ total\ of\ n\ frames
\]
Equation 4.2

Both the co-contraction index and percent difference evaluate the relative, simultaneous activation of antagonistic muscles. As discussed previously, lateral trunk motion is of greatest interest for the ACL injury mechanism. For the core, muscles of the same name on opposite sides of the body will act to laterally pull the trunk in different directions, making them antagonists for lateral trunk flexion.
The calculation of the co-contraction index, based off previously published work (Lewek et al., 2005), incorporates the relative activation of antagonistic muscles with their combined magnitudes to evaluate the level of co-contraction. Percent difference incorporates the difference in simultaneous activation of antagonistic muscles expressed relative to the average activation of the two muscles to evaluate how much more one muscle is activated compared to the other.

While avgCCI and avg%DIFF may look similar, they are quantifying different things. A similar magnitude increase in both activations of the muscles would result in an increase in avgCCI but a decrease in avg%DIFF \[ \uparrow\text{lowerEMG} + \downarrow\text{higherEMG} = \uparrow\text{avgCCI} \& \downarrow\text{avg%DIFF} \]. This indicates that avgCCI is a good indication of when both muscles increase or decrease activation together. An increase in co-contraction of core muscles is likely to lead to stiffening of the abdomen and support of the trunk. If the lower activated muscle decreased its contraction while the higher activated muscle increased its contraction, but with a similar magnitude of change, avgCCI would decrease while avg%DIFF would increase \[ \downarrow\text{lowerEMG} + \uparrow\text{higherEMG} = \downarrow\text{avgCCI} \& \uparrow\text{avg%DIFF} \]. This indicates that changes in avg%DIFF are a good indication of when the activations of the muscles being considered change in opposite directions. An increased percent difference means an increase in the potential for motion in the direction of the muscle with elevated activation while a small percent difference would indicate that both muscles are activated approximately the same resulting in little-to-no potential for movement.
4.3.5 Statistics

The association between dependent and independent variables was evaluated using linear mixed effects models. Linear mixed effects models include a fixed effect for all variables used except subject, which is a random effect. The use of a random subject effect allows examination of associations within an individual rather than only intersubject relationships. Therefore, the significant relationships reveal how modifying independent variables can be expected to result in changes in dependent variables for individual subjects.

A compound symmetry covariance structure was chosen to account for the correlation between the dependent variables of the multiple trials. Each dependent variable was evaluated separately. Two sets of independent variables were evaluated for all models. The first was the set of 3 avgCCI and the second was the set of 3 avg%DIFF. All avgCCI and avg%DIFF (except avg%DIFF_L5) were log transformed to meet normality assumptions of the model. A significance level of $\alpha=0.05$ was set a priori. No correction for multiple comparisons was applied.

Unanticipated run-to-cut maneuver. For unanticipated cutting, both pEKA$\text{bM}$ and peak outside tilt were evaluated as dependent variables with approach speed, cut angle, and gender added as covariates. pEKA$\text{bM}$ was log transformed to meet normality assumptions of the model.

Trunk control. For trunk control, peak angular displacement after release was evaluated as the dependent variable with no covariates added. Peak angular displacement after release was log transformed to meet normality assumptions of the model.
4.4 Results

Average statistics and ranges for parameters used in the models are presented in Table 4.1. The results of the mixed effects models can be found in Table 4.2.

4.4.1 Unanticipated run-to-cut maneuver

Peak outside tilt. There was no evidence for a significant association between peak outside tilt and avgCCI or avg%DIFF for any muscle group. The associations between peak outside tilt and approach speed (p=0.071, 0.075; for avgCCI and avg%DIFF, respectively), cut angle (p=0.140, 0.142), and gender (p=0.180, 0.149) were not significant.

External knee abduction moment. Two significant associations were found between peak external knee abduction moment (pEKAbM) and avgCCI. A significant negative association was found between pEKAbM and avgCCI_IO (p=0.048) while a significant positive association was found between pEKAbM and avgCCI_L5 (p=0.005). No significant associations were found between any of the three avg%DIFF. The associations between pEKAbM and approach speed (p=0.024, 0.011; for avgCCI and avg%DIFF, respectively) were significant while associations between pEKAbM and cut angle (p=0.082, 0.180) and gender (p=0.196, 0.332) were not significant.

4.4.2 Lateral Trunk Control

There was no evidence for a significant association between peak angular displacement after release and any of the avgCCI. A significant positive association was found between peak angular displacement after release and avg%DIFF_L5 (p=0.014).
Table 4.1 Means, standard deviations, and ranges for variables of interest (dependent, independent, and covariates). Numbers in parentheses are negative.

<table>
<thead>
<tr>
<th>Variable</th>
<th>mean±stdev</th>
<th>range</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cutting</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Peak outside tilt [°]</td>
<td>8.56±5.4</td>
<td>(4.0)-23.9</td>
</tr>
<tr>
<td>pEKAbM [%BW*ht]</td>
<td>8.51±6.7</td>
<td>0.0-37.3</td>
</tr>
<tr>
<td>avgCCI_EO [%MVIC]</td>
<td>26.7±15.1</td>
<td>4.3-76.5</td>
</tr>
<tr>
<td>avgCCI_IO [%MVIC]</td>
<td>30.4±20.5</td>
<td>7.5-135.6</td>
</tr>
<tr>
<td>avgCCI_L5 [%MVIC]</td>
<td>28.819.2</td>
<td>3.4-121.1</td>
</tr>
<tr>
<td>avg%DIFF _EO [%]</td>
<td>63.0±27.6</td>
<td>18.3-153.8</td>
</tr>
<tr>
<td>avg%DIFF _IO [%]</td>
<td>76.0±34.2</td>
<td>6.7-168.3</td>
</tr>
<tr>
<td>avg%DIFF _L5 [%]</td>
<td>61.9±25.6</td>
<td>10.8-139.3</td>
</tr>
<tr>
<td>Approach speed [m/s]</td>
<td>3.61±0.4</td>
<td>2.65-4.46</td>
</tr>
<tr>
<td>Cut angle [deg]</td>
<td>45.3±6.5</td>
<td>24.2-63.1</td>
</tr>
<tr>
<td>Trunk Control</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Peak angular displacement [°]</td>
<td>2.73±1.05</td>
<td>1.05-7.65</td>
</tr>
<tr>
<td>avgCCI_EO [%MVIC]</td>
<td>2.91±2.32</td>
<td>0.45-18.0</td>
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<tr>
<td>avgCCI_IO [%MVIC]</td>
<td>3.60±2.68</td>
<td>0.19-19.0</td>
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<tr>
<td>avgCCI_L5 [%MVIC]</td>
<td>3.12±2.44</td>
<td>0.49-16.2</td>
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<tr>
<td>avg%DIFF _IO [%]</td>
<td>69.5±32.7</td>
<td>19.9-171.0</td>
</tr>
<tr>
<td>avg%DIFF _L5 [%]</td>
<td>78.0±35.5</td>
<td>12.5-168.0</td>
</tr>
</tbody>
</table>
Table 4.2 Slope and p-values for interactions between dependent variables (peak outside tilt of the torso during cutting, natural log transformed pEKA_bM, and peak lateral angular displacement after release in the SFR) and independent variables (average co-contraction index [avgCCI] and average percent difference [avg%DIFF]). All independent variables natural log transformed except avg%DIFF_L5. Analyses for cutting variables included approach speed, cut angle, and gender added as covariates. Significant (p<0.05) associations bolded and denoted with †.

<table>
<thead>
<tr>
<th>Independent Variable</th>
<th>Slope</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Peak outside tilt [°]</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Ln(avgCCI_EO)</td>
<td>-0.8043</td>
<td>0.499</td>
</tr>
<tr>
<td>Ln(avgCCI_IO)</td>
<td>-0.014</td>
<td>0.918</td>
</tr>
<tr>
<td>Ln(avgCCI_L5)</td>
<td>1.296</td>
<td>0.218</td>
</tr>
<tr>
<td>Ln(avg%DIFF_EO)</td>
<td>0.972</td>
<td>0.419</td>
</tr>
<tr>
<td>Ln(avg%DIFF_IO)</td>
<td>0.514</td>
<td>0.644</td>
</tr>
<tr>
<td>avg%DIFF_L5</td>
<td>-0.014</td>
<td>0.447</td>
</tr>
<tr>
<td>Ln(Peak external knee abduction moment [%BW*ht])</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Ln(avgCCI_EO)</td>
<td>0.185</td>
<td>0.424</td>
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<tr>
<td>Ln(avgCCI_IO)</td>
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<td>0.048</td>
</tr>
<tr>
<td>Ln(avgCCI_L5)</td>
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<td>0.005</td>
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<tr>
<td>Ln(avg%DIFF_EO)</td>
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<tr>
<td>Ln(avg%DIFF_IO)</td>
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<tr>
<td>avg%DIFF_L5</td>
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<td>0.260</td>
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<tr>
<td>Ln(Peak angular displacement after release [°])</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Ln(avgCCI_EO)</td>
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<td>0.827</td>
</tr>
<tr>
<td>Ln(avgCCI_IO)</td>
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<td>0.941</td>
</tr>
<tr>
<td>Ln(avgCCI_L5)</td>
<td>0.048</td>
<td>0.451</td>
</tr>
<tr>
<td>Ln(avg%DIFF_EO)</td>
<td>-0.016</td>
<td>0.807</td>
</tr>
<tr>
<td>Ln(avg%DIFF_IO)</td>
<td>-0.011</td>
<td>0.843</td>
</tr>
<tr>
<td>avg%DIFF_L5</td>
<td>0.002 †</td>
<td>0.014</td>
</tr>
</tbody>
</table>
4.5 Discussion

The results of this study did not support our hypothesis that increased co-contraction indices and decreased percent differences for core muscles would be associated with decreased peak trunk angles during an unanticipated cutting task. Two significant associations were found between pEKAbM and core muscle preactivation during the cutting task; one contradicting and the other confirming our hypothesis. First, contradicting our hypothesis, the significant positive association between avgCCI_L5 and pEKAbM suggests that as co-contraction of the L5 extensors increases prior to initial contact of a cut so too does the peak knee abduction moment, potentially increasing ACL strain and risk of ACL injury (Fleming et al., 2001; Hewett et al., 2005; Kanamori et al., 2002; Shin et al., 2009; Shin et al., 2011). Prior to initial contact of a cut pEKAbM decreases, potentially reducing ACL strain and risk of ACL injury. The results of this study did support our hypothesis that decreases in avg%DIFF would be associated with decreases in trunk displacement, though this relationship was only seen for the L5 extensors during the trunk control test and not for any of the other core muscle pairs or for unanticipated cutting.

The lack of evidence for a significant association between co-contraction indices or percent differences of core muscles before initial contact for cutting is likely due to the complex movement associated with cutting and the complex architecture of the trunk musculature. Even though the period analyzed was pre-contact, the subject has already been given the signal to cut and is already preparing for that cut. This preparation involves all three planes of motion, and all the core muscles analyzed influence motion in the frontal plane and at least one other plane. In addition to frontal plane motion, the L5
extensors contribute to sagittal plane trunk motion. Internal and external obliques each contribute to both sagittal and transverse plane motion (Figure 4.6). Because of the complex nature of cutting and the complex anatomy and function of the core musculature, a more complex analysis may be necessary to determine how the muscles of the core influence the motion of the trunk during cutting.

![Figure 4.6 Representation of obliques showing lines of action. The more superficial left external oblique is not pictured in order to show the left internal oblique which lies beneath it. The blue arrows indicate action lines of the obliques. A contraction of the right external oblique and the left internal oblique would act to pull the right shoulder forward (sagittal plane) and across the body (transverse plane).](image)

The significant positive association between the co-contraction index of the L5 extensors and pEKAbM indicates that as the L5 extensors increase co-contraction, potentially increasing the stiffness the spine before impact with the ground, peak knee
abduction moment just after impact increases. If the spine is stiff from increased co-contraction of the L5 extensors, the kinetic energy of the upper body will not be absorbed, but transferred, through the core. This transfer of energy, rather than absorption, may be manifested in higher knee abduction moments. The association between pEKAbM (dependent variable) and peak outside tilt, while considering the covariates of speed, cut angle and gender, was also evaluated. The significant positive association between pEKAbM and peak outside tilt (p<0.0001) matches that previously published by Jamison et al. (2012b; Chapter 3). This positive association between knee abduction moment and lateral trunk position, along with the positive association between co-contraction of the L5 extensors and peak knee abduction moments, suggests that while the trunk is in a disadvantageous position as it relates to the loading of the knee, the lack of energy absorption by the core may put the knee in further danger.

The significant negative association between the co-contraction index of the IO and pEKAbM indicates that if IO cocontraction increases, pEKAbM decreases. In addition to contributing to lateral trunk motion, the IO also act to flex and twist the trunk. If both IO were activated significantly, the trunk may flex forward. Flexing of the trunk forward before landing may help position the trunk to absorb the kinetic energy of the upper body during the landing, transferring less energy to the lower extremities, which could result in a lower pEKAbM. A recent study comparing 20 ACL-injury events to similar non-injury events using video analysis showed that injured athletes had significantly less trunk flexion than the athletes who were not injured (Sheehan et al., 2012). These results support ours since a decreased co-contraction of the IO just before foot contact, associated with increased pEKAbM just after foot contact, may result in decreased
forward trunk flexion at initial contact, which has previously been associated with ACL-injury events.

The significant positive association between percent difference of the L5 extensors and peak angular displacement of the trunk after release during the trunk control task indicates that as the activation difference between the two sides decreases, or they become closer to the same, the trunk moves less and more control of the trunk is exhibited after the perturbation. The difference in activations during the isometric pull just before release can be thought of as the potential for the system to move since the pair of muscles are acting to pull the trunk laterally in opposite directions while motion is constrained by the cable (Figure 4.7). After release, this potential has to be overcome in order to halt and reverse the movement. In this way, the smaller the difference in activations, and the less potential the system has, the easier it is for movement to be stopped and reversed after the potential is released (in the form of removing the constraining force in this example).
Figure 4.7 Simplification of L5 extensor activation and the spine during the sudden force release task. The left L5 extensor is activated more (greater color saturation) than the right (less color saturation) in order to produce the desired force in the cable (black bar), which is anchored. When the left L5 extensor increased activation and the right decreased activation, avg%DIFF increased. This increases the potential for the system to move left, once the restraint is released [\( \uparrow L + \downarrow R = \uparrow \text{avg%DIFF} = \uparrow \text{left movement potential} \)]. To stop motion to the left after release and return to vertical, the left and right L5 extensors must reverse relative activation (right must be activated more than left).

The similarities and differences between avgCCI and avg%DIFF should also be considered when evaluating these results. First, the relationship between avgCCI and avg%DIFF was explored using 6 mixed effects models (2 activities x 3 muscle pairs) identical to those used previously, including the addition of covariates for cutting. For both activities, avg%DIFF was used as the dependent variable while avgCCI was the independent variable. Again, all avgCCI and avg%DIFF (except avg%DIFF_L5) were log transformed to meet normality assumptions. Results indicate that avgCCI and
avg%DIFF were significantly (all p<0.0001) and negatively associated. This indicates that as the co-contraction index increases, the difference between the two muscles decreases, i.e. the activations become more similar. If the lower EMG and higher EMG become closer, the ratio for avgCCI (Equations 4.1) will increase, increasing avgCCI, while the numerator for avg%DIFF (Equation 4.2) will decrease, decreasing avg%DIFF. However, this does not give much insight into the activations of the individual muscles used in each characterization.

To further investigate the relationship between avgCCI and avg%DIFF, mixed effects models similar to those used previously, including the addition of covariates for cutting, were applied to examine the influences of the independent variables on relative muscle activation for each activity-muscle pair combination (2 quantifications of relative muscle activation (avgCCI and avg%DIFF) x 2 activities x 3 muscle pairs). Again all avgCCI and avg%DIFF (except avg%DIFF_L5) were log transformed to meet normality assumptions. Independent variables were the average activations (during the preactivation interval) of the individual muscles used to calculate the relative muscle activation (avgCCI or avg%DIFF). In all models, avg%DIFF had a significant (p=<0.0001 to 0.0035), positive association with one muscle of the pair and a significant (p=<0.0001 to 0.0033), negative association with the other. This result indicates that avg%DIFF best indicates, in our study, when one muscle of the pair is increasing in activation while the other muscle of the pair is decreasing in activation. For cutting, avgCCI had a significant (all p<0.0001), positive association with all muscles. For the sudden force release task, avgCCI had a significant (p<0.0001), positive association with 4 of the 6 possible muscles. The other two muscles did not have a significant association with their
respective avgCCI (p=0.5918 for non-cable side L5; p=0.2798 for cable side EO). This result indicates that in our study, avgCCI best indicates when muscles are increasing in activation.

The results of this study should be considered in light of its limitations. This study was not powered to evaluate all the statistical models used. As this was the first study to examine core muscle activation during a task as dynamic as cutting, no correction for multiple comparisons was applied. The results of this study may be best used as preliminary pilot data for designing future studies to confirm its observations.

Only three core muscle pairs were considered for this study. Other core muscles may influence trunk motion during cutting and the sudden force release task, but technical challenges make quantifying their activations impractical at this time. The multifidus, quadratus lumborum, and transverse abdominis lie deep to more superficial structures and require fine wire EMG or ultrasound to characterize their activations. The current fine wire EMG and ultrasound techniques are not suitable for use with an activity as dynamic as cutting. Further, a study comparing fine wire EMG measured activation of quadratus lumborum and transverse abdominis to surface EMG measured activation of lower lumbar extensors and internal obliques found that quadratus lumborum activation was predicted well by the lower lumbar extensor activation (RMS difference < 6% MVIC) while transverse abdominis activation as predicted well by internal oblique activation (RMS difference < 15% MVIC; McGill et al., 1996). Additionally, the rectus abdominis was not considered because its main function is to flex the trunk forward, while lateral trunk motion was considered in the current study. Future studies may want to include the rectus abdominis and forward trunk flexion when quantifying ACL injury risk, however.
Another limitation is the simplicity of the analysis used to evaluate core muscle activation during an unanticipated cutting task. A simple model was chosen to aid in interpretation of the results for communication to clinicians and to translate the results to interventions. However, both the anatomy and function of the core muscles, as well as the task, are complex in nature. The use of a simple model may have masked more complex relationships from being observed. Therefore more complex models and analyses may be necessary to fully characterize the role of the core muscles during cutting. Nevertheless, the models presented here are a good first step in evaluating this relationship.

4.6 Conclusions

This study observed a significant positive association between co-contraction of the L5 extensors and knee abduction moments suggesting that a stiff spine and trunk may not be advantageous for protecting the knee during an unanticipated cut. This study also observed a significant positive association between percent difference in side-to-side activation of the L5 extensors and peak angular displacement of the trunk after release during a trunk control task. This result suggests that a control strategy that calls for more activation of the antagonist muscles with less activation of the agonist muscles prior to release may lead to an improved ability to respond to a perturbation. Future work is necessary to confirm these results but, if confirmed, these results give some insight into what may be advantageous muscle activation strategies for cutting and responding to a perturbation. Additionally, exercises that aim to improve trunk muscle activation strategies in response to perturbation may be beneficial additions to ACL injury prevention programs.
4.7 Acknowledgements

The authors would like to thank Under Armour for funding this study.
5 Discussion and Future Work

5.1 Project Discussion

Through this work, a better understanding of how the core influences ACL injury risk was realized (Figure 5.1). First, the parallels between trunk control and adverse knee loading following interventions suggest that trunk control may influence ACL injury risk. Further, the lack of improvement in trunk control following a trunk stabilization program implies that trunk stabilization training may not be the most efficient and effect program for improving trunk control. The association between trunk position and knee loads during cutting indicates that trunk positioning may be part of the ACL injury mechanism. Lastly, a core muscle activation strategy that stiffens that spine and reduces forward trunk flexion may increase adverse loading of the knee, also contributing to the ACL injury mechanism.
Previous studies have connected lateral trunk control (Zazulak et al., 2007a), lateral trunk angle during dynamic activity (Hewett et al., 2009), and EKAbM (Hewett et al., 2005) with ACL injury (black arrows). Gray arrows indicate associations investigated in this project. A green “✓” indicates a significant association was found while a red “✗” indicates a significant association was not found. A green “✓” and a red “✗” together indicate a weak connection was found. A “▲” indicates changes in the two variables were investigated. The chapter detailing the study undertaken for each investigation is also specified (Ch#).

In Chapter 2, an intervention study where interventions were used as tools for altering lateral trunk control (associated with ACL injury; Zazulak et al., 2007a) while observing the subsequent changes in peak external knee abduction moment (pEKAbM), a surrogate for ACL injury risk. Lateral trunk control and pEKAbM followed the same trend relative to ACL injury risk for both intervention groups which suggests that lateral trunk control may be an important factor in the ACL injury mechanism. Further, since trunk control was not altered following a trunk stabilization program (Chapter 2) but did improve after a training program incorporating perturbations (Pedersen et al., 2009), perturbation
training is likely a key component in intervention programs which are aimed at improving trunk control.

In Chapter 3 a significant, positive association was found between lateral tilt of the trunk and pEKAbM during cutting, suggesting that as the trunk laterally tilts more from vertical, adverse knee loading increases, increasing ACL injury risk. This implies that the position of the trunk directly affects the loading environment of the knee and compliments previous a video analysis of ACL injury events (Hewett et al., 2009), further supporting the hypothesis that the trunk may be a factor in the ACL injury mechanism.

The study outlined in Chapter 4 determined how core muscles activate during both sudden force release (SFR) and cutting tasks, potentially influencing ACL injury risk. These activation strategies were then compared to trunk displacement during the SFR task and trunk position and pEKAbM during the cutting task. A significant, positive association was found between trunk displacement during the trunk control task and the average percent difference in left-to-right lower lumbar spine (L5) extensor activation. This suggests that a control strategy emphasizing increased antagonistic muscle activation of the lower lumbar spine extensors may be advantageous for improving trunk control in this well-defined task. For cutting, a significant, positive association was found between the L5 co-contraction index and pEKAbM while a significant negative association was found between the internal oblique co-contraction index and pEKAbM. An increase in L5 and a decrease in internal oblique co-contraction likely leads to an increase in stiffness and a decrease in forward flexion of the spine and trunk, reducing the trunk’s ability to absorb energy upon impact while transferring this energy to the lower extremities. This transfer, rather than absorption, of energy may cause elevated adverse
loading of the knee, potentially increasing ACL injury risk. Additionally, lateral trunk position during the cut was not significantly associated with any measure of core muscle activation investigated, though this is most likely due to the extremely complex nature of the cutting task as well as the anatomy and function of the core muscles.

5.2 Future Work

While the studies contained in this project begin to explore the relationship between the core and ACL injury risk, more should be done in this area to confirm these results and expand upon them (Figure 5.2). The most important thing that needs to be accomplished, as it pertains to the core and the ACL, is the development of an intervention program that alters trunk control and, as a result, alters surrogates for ACL injury risk like pEKAbM. As discussed in Sections 2.5 and 5.1, the most effective intervention programs will likely contain a trunk perturbation component.
Figure 5.2 Representation of how the work from this project fits into the ACL injury prevention landscape. “→” represents an association, not necessarily a causation, between variables. “↑” and “↓” represent an increase and decrease in variables, respectively.
While developing and testing interventions (with perturbation components) for altering trunk control, changes in knee loading should also be studied. Because a reduction in ACL injury risk is ultimately desired, any programs that improve adverse knee loading (regardless of changes in trunk control) may still be valuable in reducing ACL injury incidence. I hypothesize, however, that training programs that improve trunk control will also improve knee loading; establishing a causal relationship between the two. An ideal study will compare several different training programs; one which contains only trunk perturbation components, one which only incorporates lower extremity exercises (e.g. landing technique training (Hewett et al., 1999), strength training (Herman et al., 2009)) and one with a combination of the two. I hypothesize that the combination program will be most effective at altering ACL injury risk factors because it ought to improve both trunk control and lower extremity movement patterns, both of which should improve knee loading. However, testing training programs which focus only on the core and improving trunk control are still important. Showing that improvements in trunk control, on their own, can alter lower extremity loading justifies the inclusion of trunk focused exercises in ACL injury prevention programs and is an important step in proving that trunk control alters ACL injury risk.

In addition to determining if perturbation training is necessary and effective at improving ACL injury risk factors, training focused on the positioning of the trunk during cutting and other changes in direction may also be important (Chapter 3). Teaching athletes to lean into, not away from, their changes in direction may be a necessary component of effective and efficient trunk control training aimed at reducing ACL injury. Perturbation training may improve the core’s ability to react to sudden perturbations to
the trunk while teaching athletes to lean into their changes of direction may help improve their desired response.

It is not clear how long interventions of this nature need to last in order to be effective. A study by Hewett et al. (1999) showed that female athletes completing a 6-week jump-training program were less likely to sustain an ACL injury than untrained female athletes. In contrast, the soccer intervention performed by Pedersen et al. (2009), which improved trunk control, consisted of 16 weeks of training. It is possible, though, that the changes in trunk control occurred within the first 6 weeks of training and that the additional time was unnecessary. Long training programs (12 to 18 weeks) with intermediate testing sessions would be able to evaluate when changes are occurring and allow for a better estimate of how long an effective training program needs to be.

Once a training program has been found to be productive at improving trunk control and reducing adverse loading of the knee, prospective studies using the training program should be undertaken to evaluate its effectiveness at reducing actual ACL injuries. In the end, the goal is to reduce ACL injuries so a reduction in ACL injury incidence is the final measuring stick with which intervention programs should be measured.

5.3 Summary
The aim of this dissertation was to improve the knowledge base confirming or contradicting a connection between the core and the ACL. The studies undertaken to accomplish this aim build upon previous research as well as each other to accomplish this aim. The study detailed in Chapter 2 demonstrated that the trunk stabilization program used was not effective at improving trunk control or adverse knee loading. It did, however, demonstrate that the whole body resistance training program used was
associated with decreased trunk control and increased adverse loading at the knee. The fact that trunk control and adverse knee loading followed similar patterns for both training groups suggests that the trunk control and knee loading might be connected. Chapter 3 showed increased lateral lean of the trunk away from the cutting direction was associated with increased adverse loading of the knee, suggesting that dynamic trunk positioning may be an important factor in the ACL injury mechanism. Finally, Chapter 4 found that increased lumbar stiffness, a result of increased co-contraction of the lumbar extensors, was associated with increased adverse knee loading during cutting. An increasingly stiff spine and trunk may not be able to absorb the energy from the impact of the plant foot striking the ground, instead transferring this energy to the lower extremity, where it manifests itself as an increase in adverse knee loading. This study also found that increasing antagonist lumbar extensor activation prior to a sudden force release is advantageous for reducing trunk motion post-release. Overall, the results of these studies, and this project, represent valuable information about how the core may be influencing ACL injury risk. Having a better understanding of this connection should lead to better ACL injury prevention programs and, hopefully, fewer ACL injuries as a result of these improved prevention programs.
References


