University of Tennessee, Knoxville

TRACE: Tennessee Research and Creative Exchange

Masters Theses Graduate School

5-2017

Effects of a Raised Surface on Lower Extremity Kinematics, Kinetics, and Muscle Activation During a Sidecut in Recreational Female Softball Players

Lauren Elizabeth Schroeder
University of Tennessee, Knoxville, lschroe1@vols.utk.edu

Follow this and additional works at: https://trace.tennessee.edu/utk_gradthes

Part of the Biomechanics Commons, Exercise Science Commons, and the Sports Sciences Commons

Recommended Citation

Schroeder, Lauren Elizabeth, "Effects of a Raised Surface on Lower Extremity Kinematics, Kinetics, and Muscle Activation During a Sidecut in Recreational Female Softball Players. " Master's Thesis, University of Tennessee, 2017.
https://trace.tennessee.edu/utk_gradthes/4778

This Thesis is brought to you for free and open access by the Graduate School at TRACE: Tennessee Research and Creative Exchange. It has been accepted for inclusion in Masters Theses by an authorized administrator of TRACE: Tennessee Research and Creative Exchange. For more information, please contact trace@utk.edu.

www.manaraa.com

To the Graduate Council:

I am submitting herewith a thesis written by Lauren Elizabeth Schroeder entitled "Effects of a Raised Surface on Lower Extremity Kinematics, Kinetics, and Muscle Activation During a Sidecut in Recreational Female Softball Players." I have examined the final electronic copy of this thesis for form and content and recommend that it be accepted in partial fulfillment of the requirements for the degree of Master of Science, with a major in Kinesiology.

Joshua T. Weinhandl, Major Professor

We have read this thesis and recommend its acceptance:

Songning Zhang, John G. Orme

Accepted for the Council:
Dixie L. Thompson

Vice Provost and Dean of the Graduate School

(Original signatures are on file with official student records.)

www.manaraa.com

Effects of a Raised Surface on Lower Extremity Kinematics, Kinetics, and Muscle Activation During a Sidecut in Recreational Female Softball Players

A Thesis Presented for the
Master of Science
Degree
The University of Tennessee, Knoxville

Lauren Elizabeth Schroeder
May 2017

www.manaraa.com

Copyright © 2017 by Lauren Schroeder

All Rights Reserved.

المنارة للاستشارات

www.manaraa.com

ACKNOWLEDGEMENT

I would like to thank my advisor, Dr. Joshua Weinhandl, for the guidance and knowledge he has instilled in me throughout this process. I have been very grateful for his willingness to work with me on this novel study. It has truly been an honor working and learning from him these past two years. I look forward to continuing my education and learning with him while working towards my PhD. I also want to thank my committee members, Dr. Songning Zhang and Dr. John Orme, who have both provided useful constructive criticism and guidance along the way, making me not only a better writer, but also a better researcher. I want to thank Kevin Valenzuela for all his help in data collection and processing. He has expanded my knowledge of Biomechanics throughout the past two years, and I am truly grateful for it. I also want to thank my lab colleagues who have encouraged me and offered constructive criticism and advice during this process. Finally, I would like to thank my family and boyfriend, Bryce Dami, for all their love and encouragement throughout this entire process.

المنارة للاستشارات

www.manaraa.com

ABSTRACT

Noncontact anterior cruciate ligament (ACL) injury is a common sports-related injury. “High-risk” dynamic movements, such as a sidecut, have been associated with increasing the risk of noncontact ACL injury. Certain biomechanical abnormalities, specifically at the hip and knee, and neuromuscular abnormalities, such as unbalanced quadriceps-to-hamstrings activation ratios and certain activation patterns prior to initial contact and after initial contact, have also been associated with an increased likelihood of noncontact ACL injuries occurring. Approximately 78% of all NCAA Division I softball game-day injuries are classified as noncontact where there is no direct contact to a player. Internal derangement of the knee accounted for 221 game day injuries, and 31% of these injuries were noncontact ACL injuries. The base runner was at the greatest risk of injury, with 28.8% of athletes base running at the time of injury. Additionally, 9% of base runners, or 187 athletes, were injured while contacting the base. The purpose of this study was to determine the effects of a raised surface on lower extremity kinematics, kinetics, and muscle activation patterns during a sidecut, simulating rounding first base. Participants completed two base conditions – with a base present (WB) and no base (NB) present with a controlled entrance and exit speed. Results indicated the only biomechanical difference between base conditions was greater peak knee adduction moments in the NB condition compared to the WB condition. These findings suggest that the body may be in a better position when a raised surface is present during a sidecut and decrease the risk of noncontact ACL injury. Therefore, examining movement patterns at the ankle may provide a better explanation for noncontact ACL injuries that occur during this time. Regarding muscle activation, there was significantly greater quadriceps activation post-contact compared to pre-contact. Significantly greater quadriceps activation creates a large anterior shear force on the ACL, increasing risk of injury.

www.manaraa.com

TABLE OF CONTENTS

المنارة للاستشارات
www.manaraa.com

المنارة للاستشارات
www.manaraa.com

LIST OF TABLES

www.manaraa.com

LIST OF FIGURES

www.manaraa.com

LIST OF ABBREVIATIONS

ACL	anterior cruciate ligament
BF	biceps femoris muscle
CCI	co-contraction index
EMG	electromyography
GRF	ground reaction force
Hz	Hertz
kg	kilogram
LEFS	Lower Extremity Functional Scale
MH	medial hamstrings muscles (semimembranosus and semitendinosus)
m	meter
m/s	meters per second
ms	millisecond
MVIC	maximum voluntary isometric contraction
N	Newton
NB	no base sidecut condition
Nm	Newton-meter
VL	vastus lateralis muscle
VM	vastus medialis muscle
WB	with base sidecut condition
yrs	years

www.manaraa.com

CHAPTER 1:

INTRODUCTION

BACKGROUND

The anterior cruciate ligament (ACL) is one of the most important knee ligaments. It prevents both excessive anterior tibial translation in relation to the femur, as well as frontal and transverse planar tibial rotation about the femur [1, 2]. Unfortunately, almost half of ligamentous knee injuries are isolated to the ACL [3]. There are approximately 80,000 to 250,000 ACL injuries that occur in the United States each year due to ligament failure from excessive loading [4].

There are two classifications for ACL injuries: contact and noncontact. Both types of injury can result in a complete tear of the ACL. It has been indicated that both classifications of ACL injuries increase the risk for developing early knee osteoarthritis, and within 10-20 years after the injury, as many as 50% of individuals will demonstrate radiographic arthritis, significant pain, and functional limitations [5]. The mechanism behind contact ACL injuries can be easily identified. However, the exact mechanism in a noncontact ACL injury is more difficult to identify, which is problematic. Understanding the mechanisms behind these two classifications of ACL injuries creates the potential to reduce these types of injury.

Contact ACL injuries are easily identified because they involve direct player-to-player contact, and the mechanism for the injury is known [6]. Alarmingly, noncontact injuries, which involve no direct contact, account for approximately 70% of ACL injuries [7]. While there are numerous risk factors associated with noncontact ACL injuries, the exact mechanism of how these risk factors interact and result in an injury is poorly understood [3, 8, 9]. This lack of

www.manaraa.com

understanding regarding how noncontact ACL injury risk factors interact with one another is therefore a cause for concern in attempting to reduce the number of noncontact ACL injuries.

Females are at an increased risk of rupturing the ACL, especially in sporting events, and softball is no exception. In NCAA Division I, the total number of softball teams and number of female athletes have both increased over the years, with 1003 total teams and 19,628 participants by the 2014-2015 season [10]. An increased number of participants results in a potential increase in the number of injuries sustained in softball. The body can be broken up into five body parts that are injured in softball, and the lower extremity has the highest injury rate [11, 12]. It has been found that 8.7% of all game day injuries resulted in internal derangement of the knee, equating to 220.719 knee injuries. Thirty-one percent of the 220.719 knee injuries (n = 68.42) were classified as noncontact ACL injuries. The base runner has also been found to be at the highest risk of injury, with 187.31 game injuries occurring while rounding the base [12]. The percentage of noncontact ACL injuries is small compared to other types of injuries that occur to the body, but an alarming 88% of noncontact ACL injuries required 10 or more days of time loss [11]. Noncontact ACL injuries are not as prevalent in softball as in other sports, but still result in a long recovery period and may result in a loss of pre-injury skill level once the athlete returns to play. Understanding the mechanism of what is occurring while players are rounding the base can potentially reduce noncontact ACL injuries in the sport of softball.

ACL injuries result from ligament failure. These injuries occur when the load placed on the ligament exceeds the maximum failure load the ligament can withstand. Maximum failure loads for ACLs on young cadavers have been found to range between 1730 N and 2160 N [13, 14]. Loading in just one plane does not stress the ligament enough for it to fail. Dynamic movements that involve a combination of sagittal, frontal, and transverse loading generate a

المنارة للاستشارات
www.manaraa.com

large force on the ACL that can rupture the ligament [1, 15]. Any type of interactions between large anterior shear forces and abnormal frontal and transverse plane knee moments increases the chance of injury and is considered a risk factor for noncontact ACL injury [9, 15-17]. Specifically, anterior tibial fore plus internal tibial torque near extension, and anterior tibial force plus a valgus moment at more than 10° of flexion produced the greatest loading combinations for high ACL forces [1].

Noncontact ACL injuries occur during movements that involve sudden acceleration or deceleration and changes of direction, such as a sidecut, which involves planting on a fixed and then cutting in a different direction [3, 18-20]. Most noncontact ACL injuries occur during the first 40-100 ms of the plant leg coming into contact with the ground, indicating this as the time period where lower extremity abnormalities are more than likely present [21-23]. These abnormalities are more than likely caused by either biomechanical factors, neuromuscular factors, or a combination of the two.

Some biomechanical factors that are common in noncontact ACL injuries at initial contact are planting with the knee close to full extension (between 0° and 45° knee flexion), maximum internal knee rotation, and increased knee abduction angle, all of which tighten the ligament and increase the risk of rupture [15, 18, 20, 24-27]. There appears to be a causal link between initial contact knee abduction angle and the ensuing load experienced at the knee. Abnormal frontal plane alignment especially has the potential to create a vicious cycle of harmful ACL loading by creating unfavorable knee abduction angles at initial contact, which in turn increases the internal knee adduction moments, and therefore increases the risk for ACL injury [28]. Decreased flexion angles and increased abduction angles at the hip can also increase

المنارة للاستشارات
www.manaraa.com

the risk of ACL injury because of the greater load placed on the passive joint restraints to stabilize the knee [16, 29].

Certain neuromuscular factors also play a role in increasing the risk of noncontact ACL injury. Two muscle groups are responsible for maintaining the stability of the knee, the quadriceps and the hamstrings [24, 30]. During a sidecut, there is high activation of the quadriceps. Significantly greater quadriceps activation compared to hamstrings activation has been shown to increase the strain on the ACL between full extension and 45° of flexion [31]. This activation results in less knee flexion and a greater anterior pull on the tibia, causing a larger anterior tibial shear force on the ACL [2, 32]. The hamstrings act to counteract the quadriceps and reduce the strain placed on the ACL, but their function is dependent on the knee flexion angle [3, 24]. Between 30-90° of knee flexion, hamstring activity during simultaneous quadriceps activity has been shown to significantly reduced the strain on the ACL. However, from 0-30°of knee flexion, did not significantly reduce the strain on the ACL [31]. The level of hamstring activation determines how much the activation of the quadriceps is counterbalanced [24]. Because there is normally submaximal hamstring activation, the posteriorly directed shear force is not able to reduce the anterior shear force generated by the quadriceps, which therefore increases the loads placed on the ACL [33]. Examining co-activation levels of the quadriceps and hamstrings and how they compare during the first 100 ms of meeting the ground can aid in identifying imbalances between muscle groups. These imbalances can indicate if an individual generates large anterior shear forces, which place greater loads at the ACL. Hamstring activation levels need to be enough to counteract the quadriceps activation levels in order to reduce these loads.

المنارة للاستشارات
www.manaraa.com

There is a significant gender disparity in noncontact ACL injuries. Females are 2-8 times more likely to sustain this type of ACL injury [8, 34, 35]. These gender differences in ACL injury are also present softball and baseball [11, 36]. The question that arises is why are females at a greater risk than males for this type of injury? The two factors most associated with this gender disparity are biomechanical and neuromuscular. Females tend to land with decreased sagittal plane movement at the knee and hip, which leads to an increased anterior shear force on the ACL [27, 32]. McLean et al. [8] explained that differences in the sagittal plane alone do not explain the gender disparity, but pairing these differences with frontal plane gender differences may help explain the discrepancies in ACL injury between males and females. While females land more upright with less knee flexion, they also land with a more medially collapsed knee, which creates a much greater internal adduction moment about the knee [32, 35]. The combination of these two abnormal planar movements may help explain the injury rate differences between genders.

Females demonstrate greater quadriceps activity compared to hamstring activity during a sidecut, which unfortunately creates the debilitating anterior shear force at the knee [17, 20, 27, 32, 37]. The magnitude of hamstring activation is also decreased in females, which can result in decreased hamstring strength because the muscle group is not able to fire efficiently. Decreased hamstring strength has been shown to increase the peak load placed on the ACL [33]. The posterior shear force created by the hamstrings is not adequately scaled to the anterior shear force created by the quadriceps, resulting in an increased strain on the ACL. Because females have greater quadriceps activation levels compared to hamstring activation levels, their quadriceps-hamstring ratio is much greater than their male counterparts [17]. Again, the increased activation of the quadriceps increases the anterior shear force experienced at the knee,

المنارة للاستشارات
www.manaraa.com

increasing the strain on the ACL. Pre-activation levels also differ between genders. Females have a higher pre-activation of the quadriceps and lower pre-activation of the hamstrings compared to males [17, 34]. These pre-activation levels may help explain why there is a greater reliance on the quadriceps than the hamstrings in females at initial contact.

STATEMENT OF THE PROBLEM

Previous research has identified a gender discrepancy in noncontact ACL injuries while performing a sidecut. However, previous studies seemed to have focused on sports where female noncontact ACL injuries are more common, such as soccer, resulting in little research on why this type of injury occurs in other sports where it is not so prevalent (i.e. softball). While the number of noncontact ACL injuries is not as high in this sport, the amount of time loss is substantial. Therefore, analyzing the lower extremity while rounding a base will help aid in understanding why ACL injuries occur in female softball players.

Therefore, the purpose of this study was to determine the effects of including a raised surface on the kinematics, kinetics, and electromyography of the lower extremity while performing a sidecut, simulating rounding first base.

RESEARCH HYPOTHESES

Due to the lack of softball-specific research involving rounding a base, this study was an exploratory investigation to see if there were any differences at the hip and knee between the two base conditions.

Based on preliminary data, it was hypothesized that the with base (WB) condition would increase known ACL injury risk factors. Regarding EMG data, it was hypothesized that there would be greater vastus lateralis (VL) activity compared to vastus medialis (VM) activity, both pre-contact and post-contact, regardless of base condition. It was also hypothesized that there

المنارة للاستشارات
www.manaraa.com

would be greater hamstring activation pre-contact, but greater quadriceps activation post-contact, again regardless of the base condition.

INDEPENDENT VARIABLE

- Base condition – sidecut with no base (NB), sidecut with base (WB)

DEPENDENT VARIABLES

- Kinematic variables:
  - Sagittal plane joint angles:
    - Initial contact hip flexion angle
    - Initial contact knee flexion angle
  - Frontal plane joint angles:
    - Initial contact hip adduction angle
    - Initial contact knee abduction angle
    - Peak knee abduction angle
- Kinetic variables:
  - Sagittal plane joint moments:
    - Peak internal hip extensor moment
    - Peak internal knee extensor moment
  - Frontal plane joint moments:
    - Peak internal hip abduction moment
    - Peak internal knee adduction moment
- EMG variables:
  - Pre-contact vastus medialis-vastus lateralis co-contraction index
  - Post-contact vastus medialis-vastus lateralis co-contraction index

المنارة للاستشارات
www.manaraa.com

- Pre-contact quadriceps-hamstring co-contraction index
- Post-contact quadriceps-hamstring co-contraction index

LIMITATIONS

- Participant's entrance speed was controlled. Entrance speeds may vary in real game situations, which could affect results of the study.
- Standard lab shoes were worn during the sidecut instead of the typical cleats worn while playing. Therefore, any differences found could not be completely applicable to real game play.
- The participants completed each sidecut on a gym floor instead of a dirt field. Again, any differences found in the lab were therefore not completely applicable to real game play.

DELIMITATIONS

- The participants' age range was from 18-25 years old.
- Participants must have had a minimum of two years high school softball experience to ensure they were familiar with how to properly round the base.
- Participants must have been recreationally active at least three days per week, for a minimum of 30 minutes during each session. One session must have included dynamic movements such as running and cutting.
- Participants who have ever experienced lower extremity injuries that required surgery, ever suffered an ACL injury, or suffered a lower extremity injury in the past six months were excluded.
- Any participant who experienced knee pain on the day of data collection was excluded.

المنارة للاستشارات
www.manaraa.com

ASSUMPTIONS

- The twelve-camera infrared motion capture system (Vicon Motion Analysis, Inc., Centennial, CO, USA) and one force platform (BP600600, Advanced Mechanical Technology, Inc., Watertown, MA, USA) were accurately calibrated for each participant throughout the study.
- Participants were truthful while filling out the Lower Extremity Functional Scale.
- Participants were truthful with their lower extremity injury history and weekly activity levels.

OPERATIONAL DEFINITIONS

- Noncontact ACL injury was defined as one where there was no person-to-person contact to an individual. Coming in contact with a raised surface (i.e. base) was classified as a noncontact ACL injury.
- Joint moments reported were defined as the internal joint moments, which are torques that act to resist external torques. For example, an internal knee adduction moment would act to resist knee abduction torque created by external loads.
- Initial contact was defined as the instant where vertical ground reaction force is higher than 10 Newtons.
- The time frame of interest was between 50 ms before initial contact and 100 ms after initial contact [21-23]. It has been reported that pre-contact EMG activation levels in females can place the ACL at a greater risk of injury. It has also been suggested that the time when a noncontact ACL injury occurs is within the first 100 ms of foot contact.
- The convention used for joint kinematics and kinetics followed the right-hand rule.
    - Hip: flexion (+)/extension (-), adduction (+)/abduction (-)

المنارة للاستشارات
www.manaraa.com

- Knee: flexion (-)/extension (+), adduction (+)/abduction (-)

www.manaraa.com

CHAPTER 2:

REVIEW OF THE LITERATURE

INTRODUCTION

The purpose of this study is to examine how introducing a raised surface (i.e. a base) effects lower extremity kinematics, kinetics, and muscle activation patterns of recreationally active female softball players performing a 90° sidecut (simulating rounding first base). This chapter reviewed current literature discussing anterior cruciate ligament (ACL) anatomy, ACL planar loading mechanisms, biomechanical factors associated with noncontact ACL injuries, neuromuscular factors associated with noncontact ACL injuries, the increased noncontact ACL injury rate in females, the biomechanical and neuromuscular gender differences in noncontact ACL injuries, and the rate of ACL injury in female softball athletes.

ACL ANATOMY AND LOADING

ACL ANATOMY

The ACL is one of the most important knee ligaments and has become an increasing interest in numerous studies [1]. Of all the ligamentous knee injuries, almost half are isolated to the ACL [3]. The primary role of the ACL is to prevent excessive anterior tibial translation in relation to the femur, as well as frontal and transverse plane tibial rotation about the femur. It has been found that the ACL provides 87% of the total restraining force to anterior tibial translation when the knee is flexed at 30°, and provides 85% of the restraining force at 90° of knee flexion [38]. On average, the ACL is 38 mm long and 11 mm wide, originates on the medial aspect of the lateral femoral condyle, and inserts on the anterior intercondylar area of the tibia. These attachment sites aid in the function of the ligament during any sort of movement involving the knee. The ACL length-tension relationship is primarily controlled by the attachment site located

www.manaraa.com

on the femur, the combination of motions applied to the ACL (flexion, extension, rotation, etc.), ACL fiber length when the knee is at rest, and the attachment site locations at the tibia. There has been much debate about how the ligament itself is divided. One classification separates the ACL fibers into the anteromedial bundle and posterolateral bundle. The anteromedial fibers, anterior to the center of the ligament, lengthen while the knee is being flexed. The posterolateral fibers, posterior to the center of the ligament, lengthen while the knee is extending. While the two bundles seem to work separately in the sagittal plane, when motions are couples together, such as a combination of anterior translation and internal tibial rotation, the fibers in both bundles work together to resist any tibial displacement [2]. Any lengthening of the ACL increases tension, and if this load becomes too great, the ligament has the potential to fail and tear.

ACL PLANAR LOADING MECHANISMS

There are approximately 80,000 to 250,000 ACL tears that occur in the United States each year [4]. Noyes et al. [13] reported a failure load in young cadaver ligaments (16-26 yrs.) of 1730 N, and Woo et al. [14] reported a maximum failure load in young cadaver ligaments (22-35 yrs.) of 2160 N. McLean et al. [8] found that single planar loading mechanisms alone do not generate enough force at the ACL for it to rupture. Cadaver studies have shown that dynamic movements with a combination of sagittal, frontal, and transverse loading can generate a much larger force that can rupture the ACL [1]. Interestingly, Weinhandl et al. [9] found, through musculoskeletal modeling, that loading in the sagittal plane during a sidecut contributed to 62% of the total load placed on the ACL. This was due to a large anterior shear force from the patellar tendon and an increased shear tibiofemoral contact force. However, they also found that the frontal and transverse planes contributed the remaining 38% of the total load on the ligament (26% in the frontal plane and 12% in the transverse plane), indicating a combined loading on the

المنارة للاستشارات
www.manaraa.com

ACL. It has also been found that internal knee adduction moments of 125-210 Nm in the frontal plane and knee internal rotation moments between 35-80 Nm in the transverse plane can potentially damage the ligament [39].

Applying loads individually may not create enough force at the knee for the ligament to rupture, but combining loading conditions can generate greater forces at the ligament [1, 15]. Because the ACL specifically limits internal tibial rotation and anterior translation, the combination of these knee movements have the greatest potential to lead to the failure load of the ACL [2]. In a cadaver study conducted by Markolf et al. [1], two specific loading combinations were identified that place the highest forces on the ACL and create the greater risks of injury. These include anterior tibial force plus internal tibial torque, near knee extension, and anterior tibial force plus an internal knee adduction moment, at more than 10° of flexion [1]. If the knee experiences any type of movement where there is an interaction or combination of loading from more than one plane, specifically an interaction of sagittal plane shear forces with frontal and transverse plane knee moments, it is considered a risk factor and has the potential to rupture the ACL [9, 15-17].

The ACL is one of the most important ligaments in the knee. It is a small ligament in the knee responsible for preventing excessive tibial translation and rotation relative to the femur during dynamic movements. The ACL is lengthened during these movements, which causes increased tension on the ligament. Specific loading combinations at the knee, including anterior tibial force and an internal tibial torque with the knee near extension, have the most potential to cause ligament failure. Thus, understanding the basic anatomy and loading of the ACL can aid in understanding ACL injury mechanisms.

المنارة للاستشارات
www.manaraa.com

ACL INJURY OVERVIEW

Simply put, an ACL injury occurs when the applied load exceeds the load that the ACL can withstand [40]. ACL ruptures can be classified as either contact injuries or noncontact injuries. Both are equally devastating to the individual and can increase the risk for developing early onset knee osteoarthritis, but the difference between the two is knowing the exact mechanism which caused the ligament to rupture. Contact ACL injuries occur when there is player-to-player direct contact to the knee, and the mechanisms behind these types of ACL injuries are evident when looking at a clinical history of the injury [6]. Unfortunately, approximately 70% of ACL injuries involve no direct player contact, and therefore classified as noncontact. However, while the cause of contact ACL injuries is known, the exact mechanisms involved in noncontact ACL injuries are poorly understood [2, 3, 6, 8, 9]. This lack of understanding is cause for concern due to the high number of noncontact ACL injuries. Numerous biomechanical studies have identified specific risk factors that have potential to lead to ligament failure. Knowing how these risk factors contribute to a noncontact ACL injury creates the possibility to reduce the number of noncontact ACL injuries targeted intervention strategies.

BIOMECHANICAL FACTORS OF NONCONTACT ACL INJURY

Numerous studies have reported that the majority of noncontact ACL ruptures occur during dynamic movements that involve sudden acceleration or deceleration and changes of direction, such as planting and cutting (sidecut) on a fixed foot, and jump landings, specifically landings with straight knees or a single-leg landing [3, 8, 16, 18-20, 23, 28]. Most noncontact ACL injuries have been reported to occur during a range of the first 40 to 100 ms of the plant leg coming into contact with the flat ground, which implies that there must be some abnormalities

المنارة للاستشارات

www.manaraa.com

that occur at the knee during initial contact [21-23]. These abnormal biomechanics, specifically during a cutting movement, include: an anterior shear force, caused by large quadriceps contractions that occur with low knee flexion angles and a lack of hamstrings muscle activation, axial compression loads, hyperextension, medial collapse at the knee joint, internal tibial rotation, or a combination of the previously mentioned risk factors, such as a hyperextended knee with internal tibial rotation, or an extended knee with adduction loading [2, 25, 26, 41].

The sidecut is one of these high-risk dynamic movements that can potentially rupture the ACL. The sidecut was proposed to have three distinct phase, which include the preliminary deceleration phase, the plant and cut phase, and the take-off [6]. During this first phase, the entire body must decelerate, or slow down, in the sagittal plane to prepare for the cut. This phase is also where knee flexion occurs so the body can compensate for the increased load being placed on the joint. During the plant and cut phase, the body redirects itself by pivoting about the planted foot, thus creating a torque at the knee [19]. And finally in the take-off phase, increased knee flexion angles enable the body to generate sufficient knee extension torques to propel the body forward and complete the cutting movement [6]. Also, in order to do perform this type of movement, the body must create a large braking force in the anterior/posterior ground reaction force (GRF) component. The size of the cut angle and the entrance speed into the cut determine the magnitude of this GRF, which occurs during the first 25% of deceleration phase [18, 42]. The magnitude of this braking force indicates how much force is being loaded onto the lower extremity, and a greater cut angle, such as a 90° cut angle, would require a much greater force. However, this braking force may increase the risk of the ACL rupturing [43]. Understanding what is occurring at specific segments and joints is therefore essential in understanding how this deceleration component is related to noncontact ACL injuries [19].

www.manaraa.com

In an athletic task, such as the sidecut, the individual plants the foot and then twists toward the contralateral side with the knee relatively extended. Numerous studies have shown that at initial contact, if the knee is near full extension (between 0° and 45° knee flexion), or is at maximum internal rotation, along with increased adduction, the ACL will tighten and can be at risk for rupturing [15, 18, 20, 25-27, 32]. Having the knee undergo smaller flexion angles, close to full extension, suggests that there is greater load absorption in the sagittal plane to accommodate the large braking force and peak knee extensor moments, which contribute to the increased anterior tibial shear force on the ACL [19]. Weinhandl et al. [9] also found larger anterior shear forces in the sagittal plane, which were supplied by the patellar tendon. It should seem evident then that increasing the knee flexion angle would reduce this strain on the ligament, and has indeed been found to reduce the resultant strain [15]. However, while having a more extended knee does increase the load placed on the ACL, through musculoskeletal modeling McLean et al. [8] found that peak anterior drawer force was never positive, implying that the load placed on the ACL solely in the sagittal plane does not have the potential to rupture the ACL. They also introduced random perturbations to their model, which did produced increases in the peak anterior drawer during the sidecut, but the forces remained well below the 2000 N injury threshold determined by Woo et al. [14]. This indicates that a combination of loading mechanisms from all three planes must be linked to ACL injury.

Sagittal plane loads alone do not increase the risk of an ACL injury, but combining the sagittal plane with loads in the frontal and transverse planes may be more likely to contribute to injury. In the frontal plane, having the knee collapsing medially creates an unfavorable internal knee adduction moment, which increases the risk of ACL injury by tightening the ligament. Not surprising, McLean et al. [28] found that peak internal knee adduction moment during a sidecut

المنارة للاستشارات
www.manaraa.com

was dependent on the initial contact knee abduction angle. A greater knee abduction angle at initial contact increases the abduction alignment at the knee. This abduction alignment will then increase the internal knee adduction moment and loading at the knee throughout the entire weight bearing portion of the movement. It is obvious that reducing the initial knee abduction angle in the frontal plane will reduce this vicious cycle of ACL loading.

In the transverse plane, when the foot is firmly planted on a playing field and the body is cutting to a new direction, the knee twists during the sidecut towards the new direction. This twisting is what can cause the ACL to tear from the femur rotating about the tibia, due to the foot being fixed to the playing surface. Senter and Hame [3] reported that tibial torsion is the basic mechanism of noncontact ACL injury. Therefore, transverse plane tibial torque has become a greater focus of research in identifying the mechanisms, and reducing the rate, of noncontact ACL injuries in athletes. Cross and colleagues [25] also pointed out that the internal tibial rotation phase during the sidecut is the key in reducing injuries, and that if the ability to control internal tibial rotation is lost, the ACL will be placed under maximum tension and susceptible to rupturing. Finding a way to reduce tibial rotation and knee abduction through technique modifications will more than likely reduce how much strain is placed on the ACL and reduce the number of injuries [15].

While it is obvious that the knee is the joint where ACL injuries occur, it has become more common to examine the proximal hip joint and its influences on ACL strain. At initial contact of the cutting maneuver, there is a relatively low hip flexion angle, increased hip abduction or adduction, and increased hip internal rotation [15, 16, 19, 20, 26, 28]. When this small hip flexion angle is paired with a small knee flexion angle, the "passive joint restraints", or

المنارة للاستشارات
www.manaraa.com

the ligaments of the knee, have a much greater load placed on them in order to stabilize and counteract these abnormal knee motions [16].

Like the knee, changes in the sagittal plane, even at the hip, do not effect ACL injury risk as much as changes in the frontal and transverse planes. Havens and Sigward [19] reported that frontal and transverse plane hip biomechanics increased the risk of ACL injury because of the increased redirection angle. The internal rotation angle of the hip and greater hip adduction or abduction at initial contact are related to the internal knee adduction moment during a sidecut [16, 44]. Havens and Sigward [26] found that an initial contact hip internal rotation angle explained 25% of the variance in peak internal knee adduction moment. In their systematic review of 25 female lower extremity kinematics studies, Fox and colleagues [16] found that in most studies, hip adduction greater than 5° could be considered an abnormal hip position. In the same study, Fox et al. [16] also found peak hip internal rotation angles ranging from -2.53° to 24.81°. These "undesirable hip" positions "may result in relative knee positioning that leads to injury" [20].

This knee positioning, especially in the frontal and transverse planes, can become detrimental to the ACL by increasing the internal knee adduction moments experienced at the knee. Numerous studies have found that internal knee adduction moments during a sidecut are more sensitive to hip internal rotation, and that hip internal rotation is a significant predictor of peak internal knee adduction moments [23, 26, 28]. More specifically, Havens and Sigward [26] found that, at initial contact of a 90° cut, the smaller the internal rotation angle at the hip, the larger the peak internal knee adduction moment. This finding was in contrast to another study that found that increasing the angle of hip internal rotation caused an increase in the peak internal knee adduction position [16]. Most studies examine cut angles no greater than 60°, so it is

www.manaraa.com

possible that the relationship between hip internal rotation and abduction knee loading is dependent on the angle of the cut [26]. Performing a 90° cut also requires a greater redirection angle, which creates the possibility of putting the hip in an abnormal and unfavorable position by increasing the amount of rotation. This increased transverse movement at the hip may also increase ACL injury risk by compromising the surrounding musculature's ability to efficiently support the increased knee abduction loads [19, 28]. Either way, it appears that any lack of control or abnormal alignment at the hip has the potential to increase the internal knee adduction moment [44].

It is important to identify the biomechanical risk factors associated with noncontact ACL injuries, but it is also important to link the entire body's posture to knee loading during a sidecut. This link can be used to modify sidecut techniques and potentially reduce ACL injuries. One way to modify the posture of the entire body is by modifying where the initial foot contact is located. Dempsey and colleagues [15] analyzed videos of ACL injuries and found that athletes that suffered a noncontact ACL injury during a sidecut all displayed similar body postures. This body posture specifically showed that the foot was located further away from the midline of the body. They modified the sidecut technique by bringing the plant foot closer to midline of the body with a more upright torso and found that the peak internal knee adduction moment during the weight acceptance phase was reduced by 36%.

Kristianslund et al. [23] examined the same technique factors of a sidecut and included the width of the cut. They found that this factor, along with the initial knee abduction angle, produced the greatest effect on the peak internal knee adduction moment during the contact phase by increasing the moment arm. They also examined the influence of force production on peak internal knee adduction moment, assuming that an increased force production would result

المنارة للاستشارات
www.manaraa.com

in a higher peak internal knee adduction moment. However, they found a greater relationship between internal knee adduction moment and alignment than force magnitude, indicating that changing the alignment of the lower extremity, regardless of the force being produced, has the potential to reduce this internal knee adduction moment. The angles at which these modifications were implemented were smaller (45° and 30°, respectively), but it can be assumed that with any cut angle, bringing the foot closer to the midline of the body can potentially reduce this moment arm.

NEUROMUSCULAR FACTORS ASSOCIATED WITH ACL INJURY

Electromyography (EMG) records the electrical activity of muscles and provides a way to quantify the magnitude of muscle activation. It has become the common way in examining the activation patterns of muscle groups and individual muscles during different types of movement. Neuromuscular control of muscles around the knee, including both pre-activation and reactive muscular control, is vital for joint stability and indicates just how stiff the joint is during athletic tasks that can potentially lead to an ACL injury. It is believed that there is a high probability of ACL injuries occurring within the first 40-100 milliseconds after initial contact through video analysis [21-23].

Specifically, the co-activation of the knee extensors, the quadriceps, and the knee flexors, the hamstrings, are responsible for maintaining knee stability, and limiting tibial translation [24]. These two muscle groups can influence the ability to stabilize the knee in the frontal and transverse planes [30]. Besier and colleagues [30] reported two generalized neural strategies for how the body adjusts to counter any load that is applied to the knee during a sidecut. The first strategy, known as "selected activation", involves increasing the activation of particular muscles in order to counter external loads placed on the joint. The second strategy, generalized co-

www.manaraa.com

contraction, occurs when the quadriceps and hamstrings co-activate without any specific muscles. Both strategies work to stabilize the knee during frontal and transverse planar movements, but higher muscle activation patterns or lack of active neuromuscular control can potentially increase the risk of rupturing the ACL [33, 43].

An ACL injury can occur when the muscular force produced to counteract the tibia rotating is overcome, usually from high knee extension force production and low knee flexor force production. During a sidecut, there is high quadriceps muscle activation just before foot strike that frequently exceeds maximum isometric contraction [24]. When the quadriceps are contracted near full knee extension, the ACL becomes strained from the tibia being pulled anteriorly by the quadriceps muscles [2, 24]. Malinzak et al. [32] and Renström et al. [31] both found that ACL strain increases with higher quadriceps activation and knee flexion between 0-45°. This high activation results in reduced knee flexion and an increased patella-tendon-shift angle, which draws the tibia forward in relation to the femur. In examining the sidecut, Colby et al. [24] found knee flexion angles at foot strike to be 22°, which is in the range linked with greatest ACL strain, because of the increased quadriceps activation. During the deceleration of the sidecut, the knee flexes at a high speed while the quadriceps are also lengthening at high speeds. This combination of decreased knee flexion and muscle activation will most likely produce large knee forces which place it in a vulnerable position [24]. While a large load is placed on the ACL due to quadriceps activation near full extension, McLean et al. [8] explained why musculature forces in the sagittal plane alone cannot injure the ACL. With the knee in this position, the muscle fibers of the quadriceps are shortened in a way that significantly reduces their maximum force production, so the shear loads produced are unable to rupture the ligament.

المنارة للاستشارات
www.manaraa.com

The hamstring muscles are also important in stabilizing the knee. They act as an antagonist to the quadriceps and reduce ACL strain by resisting mediolateral and anterior tibial translation forces, but their function is dependent on knee flexion angle [2, 3, 24]. ACL strain is significantly reduced when the hamstrings and quadriceps are simultaneously activation with the knee flexed 30-90º. Unfortunately, the same study found that from 0-30° of knee flexion, simultaneous activation of the hamstrings and quadriceps did not significantly reduce the strain on the ACL [31]. There is normally submaximal hamstring muscle activation at and after foot strike. This submaximal activation indicates decreased hamstring force production, which has been implicated as a potential mechanism for increased ACL injury risk. Submaximal hamstring activation, paired with increased quadriceps force production, can result in a significant tibial anterior draw, straining the ACL [24]. Weinhandl et al. [33] confirmed that ACL loading increased when there was decreased hamstring strength because of the reduced posteriorly directed shear force, indicating that greater strength and a greater force produced by the hamstrings can decrease anterior tibial draw by increasing knee flexion. In fact, Senter and Hame [3] found that knee flexion angles between 15 and 60° decreased ACL force due to the co-contraction from the hamstrings. Zebis et al. [45] also found that with neuromuscular training, hamstring activation increased while quadriceps activation remained the same, thus potentially reducing the load placed on the ACL and decreasing injury risk.

Because of the function of both the quadriceps and hamstrings and how they affect loading on the ACL, examining co-activation patterns of the quadriceps and hamstrings has become the common way in identifying muscular imbalances. The co-activation between these two muscle groups creates muscular "active stability" for the knee by acting as "active restraints" and providing muscular protection [41]. There is typically a 2:1 ratio of quadriceps strength to

المنارة للاستشارات
www.manaraa.com

hamstring strength (Q:H) in the general population. This ratio has been suggested to reveal the muscular stability around the knee and indicate whether or not there is an increased risk of injury, especially during dynamic movements [46]. This 2:1 Q:H ratio indicates a larger force exerted by the quadriceps compared to the hamstrings. This increased force produced from the quadriceps increases the ACL loading and risks ACL injury, but if the co-activation levels from the hamstrings are scaled enough to resist this anterior pull, the load is not too great on the ACL. However, if there is an even greater ratio between quadriceps and hamstrings activation levels, the anterior shear load will dramatically increase on the ACL. An imbalance between the quadriceps and hamstrings co-activation can result in "passive stability", where reliance is placed on the ligament and not the muscles [41]. To reduce the load on the ligament, hamstring activation levels must be enough to counteract the quadriceps activation levels. Identifying activation and strength imbalances between the two muscle groups may be critical in protecting the ACL from excessive loads.

Movements that involve a rapid deceleration and cutting increase the risk of rupturing the ACL. Examining both the biomechanical and neuromuscular factors associated with noncontact injuries are important in creating a way to possibly reduce this type of injury. The sagittal plane alone cannot produce enough force to rupture the ACL, but smaller knee flexion angles contribute to increasing the load on the ligament. A more extended knee also increases the tibial anterior draw, placing even more strain on the ACL. Movements in the frontal and transverse planes are more likely to contribute to noncontact ACL injuries. In any movement where the knee medially collapses, thus creating a large internal knee adduction moment, and internal rotation control is lost, a large strain is placed on the ligament. Abnormalities at the hip also influence ACL strain. More specifically, hip internal rotation is a significant predictor of knee

المنارة للاستشارات
www.manaraa.com

abduction. Initial contact body position can also affect how much strain is placed on the ligament. The closer the foot is to the midline of the body decreases the strain on the ACL, which in turn reduces the stress in the ACL, thus reducing the risk of a noncontact injury. Neuromuscular control also contributes to noncontact ACL injuries. A higher quadriceps force magnitude pulls the tibia forward and creates an increased anterior shear force. This can be counteracted by the hamstrings if their contractions are sufficient enough to counteract the large quadriceps force, creating a posterior shear force. However, decreased hamstring strength typically indicates that the anterior shear force will not be counteracted. Understanding these biomechanical and neuromuscular factors associated with noncontact ACL injuries create the possibility of reducing these types of injuries.

GENDER DIFFERENCES IN ACL INJURY

Injury rates have shown that females are 2-8 times more likely to sustain a noncontact ACL injury compared to males [47-50]. Gray et al. [47] was among the first to report gender differences in basketball players who suffered ACL injuries. It has been found that regardless of the sport, female noncontact ACL injury rates are significantly higher than male noncontact ACL injury rates [50]. The question that has become the basis of several studies is are there any lower extremity differences between males and females during athletic tasks that lead to this gender difference in injury rate? Ireland [51] explained that noncontact ACL injuries are multifactorial, and these factors can be categorized as intrinsic, extrinsic, or a combination of the two. She describes intrinsic factors as those that cannot be changed due to an individual's anatomy and physiology. These factors include alignment, hyperextension, physiologic rotatory laxity, ACL size, femoral notch size and shape, hormonal influences, and inherited skills. Extrinsic factors are those that can be changed and are controlled by the individual. These include the

المنارة للاستشارات
www.manaraa.com

strengthening of muscle groups, conditioning, shoes, and motivation. And finally, combined factors can potentially be changed and include proprioception, neuromuscular, order of firing, and acquired skills. There has been increased interest in studies that focus on the biomechanical and neuromuscular control factors that differ between males and females because they are the most likely to contribute to the increased injury rate in females and have the potential to be changed and corrected.

BIOMECHANICAL GENDER DIFFERENCES

The first factor that differs between genders is lower extremity biomechanics. The position that females tend to adopt is landing in a more upright position with decreased knee and hip flexion. This decreased sagittal plane movement is not able to protect the ACL and leads to an increased anterior shear force on the ligament. Malinzak et al. [32] found lower knee flexion angles in females, with an 8° difference between males and females that remained consistent during the entire sidecut. Wallace et al. [27] also found females to have greater extension angles compared to males (10.14° vs. 17.43°). Ireland [51] reported similar findings, with females showing knee flexion values of 24.6° compared to 29.8° in males. Interestingly, Wallace et al. [27] found that females reached their maximum knee flexion angles the same time as males, even though they exhibited greater knee extension angles. This makes it probable that females have an increased knee flexion velocity, which can possibly decrease the amount of control females have in the sagittal plane.

While McLean et al. [8] found sagittal plane movements do not solely place the ACL at an increased risk of injury, pairing sagittal plane differences with frontal plane gender differences may help explain the increased rate of injury in females. At the time of injury during a sidecut, females have smaller knee flexion angles paired with a more medially collapsed knee

www.manaraa.com

compared to their male counterparts [27, 32, 35]. Ireland [51] reported knee abduction angles at 11.1° in females, but only miniscule 1.9° angles in males. Sigward and Powers [35] found no significant kinematic differences between males and females, but they were the first to find females who experienced greater frontal plane knee moments compared to males in the early part of the sidecut. Eighty percent of their female subjects experienced a greater medially collapse of the knee, resulting in internal knee adduction moments up to two times greater compared to only 40% of males who demonstrated internal knee adduction moment [35]. This medial knee collapse also occurs faster in females than males, giving the body less time to adjust if it is in an undesirable position [37]. While studying the kinematics and kinetics of baseball and softball players, Wallace et al. [27] examined females displaying rotations in the transverse plane that places them at a greater risk for an ACL injury. This transverse plane rotation, paired with the greater extension moments found in females, reverts back to the loading characteristics that Markolf et al. [1] described to have the most potential for tearing the ACL.

Bencke and colleagues [44] observed in female handball players that peak internal knee adduction moments and external knee rotation coincided with peak internal hip external rotation moments and hip abduction moments, showing that the connection between knee and hip alignments to be of even greater importance in females. Females also have increased internal hip rotation compared to men, which has been shown to be a predictor of ACL injury [37]. Pollard et al. [52] found that female athletes exhibited all of the biomechanical risk factors at the hip that are associated with ACL injuries. Females had significantly greater hip internal rotation (7.7° vs. -1.0°), decreased hip flexion (49.3° vs. 54.0°), greater internal hip adductor moments (-1.69 Nm/kg vs. -0.87 Nm/kg), and decreased internal hip extensor moments (5.36 Nm/kg vs. 6.67 Nm/kg). These results indicate that females rely on the frontal and transverse planes more

المنارة للاستشارات
www.manaraa.com

heavily than their male counterparts. The increased internal rotation during dynamic activities can also potentially alter the lower extremity alignment, increasing ACL injury risk. The decreased extensor moments and decreased hip flexion angles in females indicate that the male subjects were better able to engage their hip extensors and control their movements in the sagittal plane during deceleration. It is clear that the motions in the frontal and transverse planes pose a much greater risk for females than males in the possibility of injuring the ACL.

NEUROMUSCULAR GENDER DIFFERENCES

The second factor where gender differences are present is lower extremity muscle activation patterns. Abnormal neuromuscular control in females, especially during dynamic athletic movements, is viewed as one of the most critical factors that can contribute to the increased injury rate in females [28, 51]. Luckily, this abnormal neuromuscular control is considered a combined factor, with both intrinsic and extrinsic components, meaning that it has the potential to change and improve [51]. These factors can be changed by improving the order of neuromuscular activation patterns so the body becomes more efficient at performing dynamic movements.

Muscle activation differences between genders are also present in the sagittal plane when examining the quadriceps and hamstrings activation levels. Females, due to the smaller flexor moments and the increased extensor moments of the knee, display increased quadriceps activity (191% MVIC) compared to men (151% MVIC) during the first 20% of the foot contact phase [35]. Malinzak et al. [32] also found similar results, with the normalized quadriceps activation in females being consistently above their male counterparts, with differences between 17% and 40%. This increased activation, specifically in the vastus lateralis (40% more in females than males during the loading phase), suggests that females exhibit a greater reliance on quadriceps

المنارة للاستشارات
www.manaraa.com

muscle activation to stabilize the knee and protect it from anterolateral subluxations [17, 20, 27, 32, 37]. As stated previously, this increased quadriceps activation unfortunately draws the tibia anteriorly, in relation to the femur, severely stressing the ACL when there is low knee flexion angles [24].

Unfortunately, while females have a greater magnitude of activation and strength in the quadriceps, they have a decreased magnitude of hamstring activation and strength during a sidecut [32, 37]. Weinhandl et al. [33] demonstrated that reduced hamstring strength in females increased the peak load placed on the ACL by 36%. They found a 44% increased load on the ACL in the sagittal plane, due to the anterior shear force of the tibiofemoral contact and posterior shear force of the hamstrings, and a 24% increase in the frontal plane ACL loading. Malinzak et al. [32] found that male hamstring activation levels were greater than 20% compared to their female counterparts. Interestingly, studies have found that hamstring activation levels were similar between males and females, yet females recruited more quadriceps and were unable to properly scale their hamstring activation to the same levels of the quadriceps [17]. Thus reduced hamstring strength in females creates an inability to scale hamstring force production levels to similar levels achieved by the quadriceps, which causes an imbalance between the two muscle groups [17]. This imbalance between the two muscle groups creates excessive anterior shear force and minimal posterior shear force, which significantly increases the risk of an ACL injury.

Gender differences are present in the quadriceps-hamstring ratio during the loading phase of the sidecut, which reveals how females are more quadriceps dominate than males. Having this quadriceps-dominate characteristic can reduce knee stability and stiffness. Ireland [51] reported stiffness values of males up to 473%, while females only had stiffness values up to 217%. The female’s body weight will not be able to be properly supported due to this laxity in the knee.

المنارة للاستشارات
www.manaraa.com

Hanson et al. [17] found a greater quadriceps-to-hamstrings ratio in female soccer players (1.26) compared to males (0.88), demonstrating that the female participants used more of their quadriceps than hamstrings during a sidecut. It is interesting to note that Sigward and Powers [35] found no differences in gender hamstring activation. So while there may be minimal differences in hamstring activation levels between genders, the increased quadriceps activation levels in females creates an excessive anterior shear force that cannot be counteracted by the hamstring, leaving the ACL in a more vulnerable position to injury [32]. This shows that the quadriceps activation level is the primary factor in causing such larger ratio differences between males and females. Either way, it is clear that an imbalance between hamstring and quadriceps strength in females increases the strain of the ACL.

Examining the pre-activation levels of the quadriceps and hamstrings and the gender differences of these levels has also become an interest in several studies. Examining these levels indicate how much force can potentially be generated during dynamic tasks. The proper amount of force to absorb the impact needs to be generated before the task is performed to prepare the body. Hanson et al. [17] found a higher pre-activation level, 31% more than males, of the vastus lateralis, indicating that this pre-activation is preparing the muscle for movement. It is scaled greater in females than males because females rely more heavily on the vastus lateralis during a sidecut. Bencke and Zebis [34] found significantly lower hamstring pre-activation in females compared to males, in both the semitendinosus (33 ± 12% vs. 46 ± 14%) and biceps femoris (30 ± 10% vs. 52 ± 22%). This lower pre-activation in females during initial contact indicates that less force will be produced by the hamstrings to pull the tibia posteriorly, and this decreased force means that the hamstrings will not be able to protect the ACL sufficiently. In the same study, the hamstring-to-quadriceps pre-activity ratio was also significantly lower in females than

المنارة للاستشارات
www.manaraa.com

males during a sidecut (0.52 ± 0.13 vs. 0.91 ± 0.42), again indicating a greater imbalance between hamstrings and quadriceps in females. Hanson et al. [17] found that females lacked hamstring activation in the preparatory phase compared to males, but reported the quadriceps-to-hamstring ratio instead of hamstring-to-quadriceps ratio. They found that males had a co-activation ratio of 0.81 ± 0.26, compared to 1.16 ± 0.74 for females, revealing another significant pre-activation muscular imbalance in females. This decreased preparatory hamstring activity and increased quadriceps activity indicates that the hamstring will not be able to generate enough force to pull the tibia posteriorly and protect the ACL from excessive loading. It is possible to improve hamstring pre-activity and decrease the risk of ACL injury. With neuromuscular training, Zebis et al. [45] saw a significant increasing in semitendinosus activation from 41±12% to 52±16%, 50 ms before initial contact. Semitendinosus activation 10 ms after initial contact also increased from 29±12% to 39±20%. These findings reveal an important neuromuscular adaptation that can potentially reduce the risk of noncontact ACL injury.

Both biomechanical and neuromuscular gender differences may explain why females are at an increased risk of noncontact ACL injuries. Females exhibit smaller knee flexion angles and greater knee adduction angles compared to males. These angles create knee moments that place a tremendous amount of strain on the ligament. Females also rely more heavily on their quadriceps and have decreased hamstring strength, both of which contribute to ACL injury because of the excessive anterior shear force that cannot be counteracted by the hamstrings. Females also do not pre-activate their hamstrings to the same level as males, indicating that the proper amount of force needed to absorb the impact of dynamic tasks cannot be generated. Increasing the knee flexion angles, increasing hamstring strength, and improving neuromuscular activation patterns

المنارة للاستشارات
www.manaraa.com

puts the knee, and therefore the ligament, in a more favorable position that can reduce the risk of a noncontact injury in females.

FEMALE SOFTBALL PLAYER ACL INJURY RATE

In the 2003-2004 academic year, there were a total of 912 NCAA softball varsity teams, with 16,079 athletes participating in the sport. By 2014-2015, the total number of teams has increased to 1003 with 19,628 participants [10]. While softball has grown in popularity throughout the years, it also means that the possible number of injuries sustained by female athletes has increased. There are five major body parts that are injured in softball, which include the head/neck, upper extremity, trunk/back, lower extremity, and other/system. While the upper extremity had a substantial percentage of injuries in games (33.1%) and practices (33.0%), the lower extremity sustains the greatest percentage of injuries in both games and practices (43.3% & 40.8%, respectively) [12]. An epidemiology study conducted by Hootman and colleagues [11], reviewed 15 different sport injury reports and found that in both games (53.8%) and practices (53.7%), the lower extremity had the highest injury rate. It is interesting to note that the gender differences seen in ACL injury rates are also present between softball and baseball, with softball players experiencing higher injury rates than baseball players [11, 36].

Marshall et al. [12] found a disturbing 78.2 % of all game day softball injuries (n = 2537 total injuries) were caused by noncontact means, such as running bases. It was also found that 8.7% of all game day injuries resulted in internal derangement of the knee, for a total of 221 knee injuries. Notably, this internal derangement of the knee resulted in the greatest amount of activity time loss. Thirty-one percent of the 221 injuries were classified as ACL injuries, equating to 69 total ACL injuries in this data set. The base runner was reported to be at the highest risk of injury, with 28.8% of athletes (n = 731) being in that position at the time of injury. Also, 187

www.manaraa.com

game injuries were sustained while rounding the base. While interpreting these injury rates, it should be understood that not every college that participates in Division I softball report injury rates to the NCAA Injury Surveillance System, and only 11.8% of schools sponsoring Division I softball participated in the data collection [12]. More accurate injury rates would be available if more schools reported their injuries.

Hootman et al. [11] also reported injury rates in different sports and found that 129 ACL injuries occurred in softball over a 15-year period, equal to 2.4% of all softball injuries. While this is a small percentage, they noted that 88% (n = 113.52) of these ACL injuries resulted in 10 or more days of time loss [12]. This amount of time removed from practice and game play can result in an increase amount of rehabilitation time to get back to pre-injury activity and skill level. It is interesting to point out that the gender discrepancy in noncontact ACL injuries is also present between softball and baseball players, with softball players sustaining a higher rate of noncontact ACL injuries compared to baseball players [11, 36]. In fact, a study analyzing sex differences in ACL injuries in collegiate sports found the highest ACL injury rate ratio to exist between softball and baseball (IRR = 6.61), indicating softball players were over six times more likely to sustain a noncontact ACL injury [36]. While the number of ACL injuries is lower in softball compared to other sports, such as football, basketball, and soccer, ACL injuries are still a major injury sustained by these athletes. Analyzing what is occurring at the knee while a player is rounding the base is therefore important to understand why this time of play results in a larger number of ACL ruptures.

SUMMARY

Noncontact ACL injuries are one of the most detrimental injuries an athlete can sustain. Most require reconstructive surgery and a prolonged rehabilitation period before athletes can

www.manaraa.com

return to their sport. And even then, there is a chance that the athlete may never reach their pre-injury activity and skill level. Those who suffer an ACL injury also have an increased risk of developing knee osteoarthritis later in life [9, 15]. Understanding the anatomy of the ACL, what occurs to the ligament when a load is placed on it, and the different factors associated with ACL injuries are important in developing intervention programs to reduce the number of ACL ruptures athletes sustain. The increased rate of ACL injuries in females compared to males is also concerning, and identifying the underlining cause of this increased rate is vital in possibly reducing or eliminating this difference. Just like any other sport, female softball players are at risk of suffering an ACL injury, especially during base running. Understanding what is occurring at the knee while a player is coming in contact with the base is important to potentially reduce the number of ACL injuries in softball.

للاستشارات المنارة
www.manaraa.com

CHAPTER 3:

METHODS

The purpose of this study was to examine how introducing a raised surface effects lower extremity kinematics, kinetics, and muscle activation patterns of recreationally active female softball players performing a 90° sidecut. This chapter described the methods used to conduct the study.

PARTICIPANTS

Participants were recruited via fliers, word of mouth, and emails. If participants were recruited via word of mouth, an email, with the study flier attached, was sent to the individual to ensure that they were still interested and qualified for the study.

Ten recreationally active females (age: 21.5±1.96 yrs, height: 1.7±0.04 m, mass: 66.99±10.87 kg) participated in this study. G*Power 3.1.9.2 [53] was used to estimate the sample size. Mean peak internal knee adduction moments and their associated standard deviations from four previous studies were used to calculate effect size [26, 28, 35, 54]. Participants had a minimum of two years of high school softball experience to ensure they were familiar with proper base rounding technique. Recreationally active was defined as being physically active at least 3 days per week for a minimum of 30 minutes each session. One of these sessions had to include dynamic movements, such as running and cutting. Participants were excluded if they ever had a lower extremity injury that required surgery (e.g., ligament rupture, meniscus repair, bone fracture), had ever suffered an ACL injury, or had suffered a lower extremity injury in the past six months. Participants were also excluded if they were experiencing any lower extremity pain the day of data collection. Participants provided written consent before data collection commenced, which was approved by the University Institutional Review Board prior to testing.

www.manaraa.com

Participants filled out the Lower Extremity Functional Scale (LEFS) to determine if they had any difficulties performing different daily activities [55]. The minimal detectable change and minimal clinically important difference of LEFS was 9, which suggests that a change of greater than 9 scale points on the LEFS is a true change in lower extremity function [55]. Thus, any participant who scored lower than a 71 out of 80 on the LEFS was excluded from the study. During this time, the participant's demographic information (age, height, years' experience) was recorded. Weight was collected from the static trial.

All participants wore a pair of Spandex shorts, a generic short-sleeved t-shirt, and a pair of standard lab running shoes (Noveto, Adidas, USA). Tight Spandex shorts were to ensure there was minimal movement of the marker set, and standard lab shoes were to decrease any variability and to ensure that any differences observed were not due to shoe type.

INSTRUMENTATION

EQUIPMENT SETUP

A twelve-camera infrared motion capture system (200 Hz, Vicon Motion Analysis, Inc., Centennial, CO, USA) was used to collect marker coordinate data. Cameras were calibrated prior to data collection. Each camera collected a minimum of 6000 wand counts to ensure each marker was captured during the dynamic movement used for the study. The capture volume included the area involved in the sidecut (i.e. the immediate area surrounding the base, including the entrance and exit lanes). A 60x60 cm AMTI force platform (2000 Hz, BP600600, Advanced Mechanical Technology, Inc., Watertown, MA, USA) was used to measure ground reaction forces. Before data collection began, the force platform was checked to ensure there was no excess noise that would interfere with collection. Each force platform was zeroed prior to data collection and throughout the data collection to remove any residual noise.

المنارة للاستشارات
www.manaraa.com

ELECTROMYOGRAPHY

Wireless EMG sensors were placed on predetermined muscles of the right lower extremity. The Trigno™ Wireless EMG system and sensors (2000 Hz, Delsys, Inc., Natick, MA, USA) were used due to the high-speed, dynamic movement performed. Four sensors were used during data collection. Each sensor was placed 1 centimeter apart on the muscle belly of the vastus medialis (VM), vastus lateralis (VL), biceps femoris (BF), and medial hamstrings (MH), specifically the semitendinosus, following the muscle fiber direction. These muscles were chosen to represent the knee extensors and flexors. The exact locations of the sensors were identified according to the guidelines provided by Rainoldi et al. [56]. The site for the BF was located approximately 35% along the reference line starting from the ischial tuberosity to the lateral side of the popliteus cavity. MH was located approximately 36% along the reference line from the ischial tuberosity to the medial side of the popliteus cavity. VL was located approximately 94 mm along the reference line from the superior lateral side of the patella to the anterior superior iliac spine. The site for VM was located approximately 52 mm from the superior medial side of the patella along the reference line medially oriented at an angle of 50° with respect to the anterior superior iliac spine. Muscle activation data was collected simultaneously with marker coordinate and ground reaction force data, using the Nexus 2.5 (Vicon Motion Analysis, Inc., Centennial, CO, USA). The Trigno™ Control Utility program was used to guarantee the sensors were properly functioning throughout the data collection.

PARTICIPANT SETUP

Each participant came to the lab for one session to complete testing. Participants completed a five-minute warm-up jog, at a self-selected pace, on a treadmill located in the lab. They stretched the muscles of the lower extremities to their satisfaction. Maximum voluntary

المنارة للاستشارات
www.manaraa.com

isometric contraction (MVIC) testing was completed after EMG sensors were placed to allow comparison of the EMG activation levels between participants and muscles. EMG levels were also normalized to the MVIC readings obtained. An isokinetic dynamometer (Biodex Medical Systems, Inc., Shirley, New York, USA) was used to collect MVIC readings. Both quadriceps and hamstrings MVIC readings were collected with the right knee flexed at a 45° [17]. The participant performed maximum effort isometric contractions against the fixed lever arm, performing extension for the quadriceps and flexion for the hamstrings. Warm-up trials were performed to familiarize participants with the testing protocol. Once the participant was comfortable with the MVIC protocol, three MVIC trials were collected for each muscle group. There was a three-second countdown, followed by five seconds of maximal effort. During the three-second countdown, each second was associated with an increase in effort (i.e. second three relaxed, second two at 50% of maximal effort, second one at 75% of maximal effort) until maximal contraction was achieved. The middle three seconds from the five-second maximal effort trials were used to assure a steady-state activity level for each muscle [57]. From the three trials, the maximum MVIC value was used to normalize the EMG data collected during movement trials. Therefore, a percentage of the MVIC was used to interpret the EMG results.

To define segment coordinate systems, anatomical reflective markers were placed on the right leg of the participant. The pelvis was defined with anatomical markers at the left and right iliac crests, as well as the left and right greater trochanters. Anatomical markers were also placed at the medial and lateral femoral epicondyles, medial and lateral malleoli, and first and fifth metatarsal heads to define the femur, shank and foot, respectively. Tracking markers were placed on the right leg of the participant to track the participant's movement. Three semi-rigid thermoplastic shells, each having four reflective markers, were placed on Velcro-sensitive

المنارة للاستشارات
www.manaraa.com

neoprene wraps on the pelvis, thigh, and shank. One semi-rigid thermoplastic shell with four reflective markers was also secured to the lateral heel of the shoe.

One static trial was collected, which consisted of the participant standing still, with their arms crossed over their chest. Once the static trial was collected, the anatomical markers were removed, leaving only the tracking marker used for data collection. A dynamic range of motion (ROM) movement trial was then collected. During the dynamic ROM trial, the participant first extended their entire leg, from the hip, strait forwards and bring it back a standing position. This was followed by the leg being extended out to a forward 45° angle, out laterally from the body, out to a backwards 45° angle, and extended straight backwards. Then, the participant bent their right knee at a 90° angle and extended the leg forwards and back to the 90° angle three times. With the knee still bent at 90°, the participant plantar flexed and dorsiflexed the right foot three times each. Once the dynamic ROM trial was collected, data collection began.

TESTING PROTOCOL

The sidecut task simulated rounding first base, off the participant's right foot to the opposite side. They stayed within an angle of 60° to 90° relative to the original line of progression. The range of entrance angles was measured and marked off using a goniometer in relation to where the base was placed on the force platform. Entrance and exit lanes were taped on the ground to create a running lane for the participants and to assure that they stayed within the preferred angle. The running lane lead directly to the base. Two meters before the base, the participant began simulating "rounding the base" in order to achieve the proper angle of the cut and maintain the required speed. The exit lane followed the normal running path that would lead to second base in a game-like situation. Participants had to maintain their speed for two steps after leaving the base or plate before they began to slow down. Two base conditions were used

المنارة للاستشارات
www.manaraa.com

for the study (Figure 1). The first condition (WB) required participants to perform the sidecut on a base (Schutt Sports Jack Corbet MLB Hollywood Baseball Base, Schutt Sports, Litchfield, IL) placed in the center of the force platform. The second condition (NB) required participants to perform the sidecut on just the force platform. During the NB condition, there was an outline of the base taped on the force platform so the participants replicated the same cut pattern as when the base was present. Participants aimed for the bottom left corner ("inside corner") for the WB condition, and the bottom left corner of the base outline for the NB condition. During each trial, foot placement was closely monitored to accurately replicate how the sidecut is performed in a game (Figure 2a). If foot placement was improper (Figures 2b-d), the trial was redone. The testing procedures were the same for both conditions. Base conditions were counterbalanced between all participants.

Participants were instructed of sidecut task and were allowed to practice as many times as necessary until they were comfortable with the route. Participants started 7 meters away from the force platform so there was ample room to reach the required speed. Two pairs of timing gates (63501 IR, Lafayette Instrument Inc., IN, USA) were used to maintain the entrance and exit speeds of each trial. The timing gaits were placed 1 m apart from each other and were at, or close to, shoulder height. Participants were required to maintain an entrance speed of 4.0 m/s ± 0.25 m/s and an exit speed of 3.75 m/s ± 0.25 m/s. The first pair of timing gates were placed 2 meters before the participant began to deviate from their forward linear motion. Participants ran towards the force platform, began "rounding the base" two meters prior to contacting the force platform, performed the sidecut within the predetermined angle range, and continued running for 7 m. Each participant was required to maintain their speed for two steps after contacting the base. Five successful trials were collected for each condition. A successful trial consisted of the participant

المنارة للاستشارات
www.manaraa.com

staying within the running lane, maintaining the required entrance speed of 4.0 m/s ± 0.25 m/s, correctly contacting the inside corner of the base, staying within the exit lane, and maintaining the required exit speed of 3.75 m/s ± 0.25 m/s.

DATA PROCESSING AND ANALYSIS

Visual3D software suite (v5, C-Motion, Inc., Rockville, MD) was used to compute kinematic and kinetic data of the right lower extremity. Because non-contact ACL injuries normally occur shortly after initial contact, only the first 100 ms after initial contact (post-contact) was analyzed [23]. There have also been some indications that pre-activation levels may contribute to ACL loading levels [17, 34]. Therefore, 50 ms prior to initial contact (pre-contact) were also analyzed. Raw marker coordinate and GRF data from each trial were low-pass filtered using a fourth-order, zero lag Butterworth filter with a cutoff frequency of 20 Hz [23]. A four-segment (pelvis, right thigh, right shank, and right foot) skeletal kinematic model was created using the standing calibration trial. Three-dimensional ankle, knee, and hip angles were calculated using a joint coordinate system approach [58]. Hip joint centers were located according to Bennett et al. [59]. The knee joint center was the midpoint between the epicondyle markers [58], and the ankle joint center was the midpoint between the malleoli markers [60]. Internally applied, three-dimensional joint kinetics were calculated using a Newton-Euler approach [61], and projected to the joint coordinate system [62]. Body segment parameters were estimated from Dempster et al. [63]. Global maximum values for each dependent variable, from each trial, were extracted from Visual3D. All peak values occurred during the stance phase of the sidecut. A vertical ground reaction force threshold of 10 N indicated initial foot contact (IC).

All EMG data (MVIC trials and motion trials) were pre-amplified and band-pass filtered using a fourth-order, zero lag Butterworth filter with a high pass cutoff frequency of 10 Hz and a

المنارة للاستشارات
www.manaraa.com

low pass cutoff frequency of 350 Hz to remove any noise. The signal was then full-wave rectified and low pass filtered at 5 Hz [33]. The middle three seconds for each MVIC trial was used to assure a steady-state activity level for each muscle [57]. The maximum MVIC value for each muscle was identified from the three MVIC trials. EMG data for each muscle from each motion trial was then normalized to its respected peak MVIC value. For each trail, the medial/lateral co-contraction index of the quadriceps (VM:VL CCI) was then computed as the ratio of average VM-to-VL activation from the 50 ms pre-contact to the 100 ms post-contact.

$$Pre\text{-}VM{:}VL\ CCI = \frac{avg(VM_{50ms-HS})}{avg(VL_{50ms-HS})} \quad (1)$$

$$Post\text{-}VM{:}VL\ CCI = \frac{avg(VM_{HS-100ms\ post})}{avg(VL_{HS-100ms\ post})} \quad (2)$$

Likewise, quadriceps/hamstrings co-contraction index (Q:H CCI) was computed as the ratio of average quadriceps-to-hamstrings activation. Due to the possibility of high hamstring pre-contact activation, the Q:H CCI was separated into pre-contact Q:H CCI and post-contact Q:H CCI. Average quadriceps activation was calculated as the sum of average VM and average VL activation. While average hamstrings activation was calculated as the sum of average BF and average MH activation.

$$Q = (avgVM + avgVL) \quad (3)$$

$$H = (avgBF + avgMH) \quad (4)$$

$$Pre\text{-}contact\ Q{:}H\ CCI = \frac{(Q_{50ms\ pre-HS})}{(H_{50ms\ pre-HS})} \quad (5)$$

$$Post\text{-}contact\ Q{:}H\ CCI = \frac{(Q_{HS-100ms\ post})}{(H_{HS-100ms\ post})} \quad (6)$$

All CCIs were computed separately for each trial. The average CCI value was then computed over the five trials. Q:H CCI values of 1.0 indicated equal average activation levels between the quadriceps and hamstrings, and VM:VL CCI values of 1.0 indicated equal activation

المنارة للاستشارات
www.manaraa.com

between the vastus medialis and vastus lateralis. Q:H CCI values greater than 1.0 indicated increased quadriceps activation compared to hamstrings activation. Q:H CCI values less than 1.0 indicated greater hamstrings activation compared to quadriceps activation [17]. VM:VL CCI values greater than 1.0 indicated greater vastus medialis activation than vastus lateralis, while VM:VL CCI values less than 1.0 indicated greater vastus lateralis activation than vastus medialis activation.

STATISTICAL ANALYSIS

Due to the small sample size, normal distribution was assumed for all variables. Dependent *t*-tests were conducted for all kinematic and kinetic dependent variables between base conditions. An independent *t*-test was utilized to determine if an order effect was present between the two base conditions. Two 2x2 (condition x contact time) repeated measure ANOVAs were performed to examine the main effects of condition and contact time for EMG data, and to determine if any significant interactions were present between condition and contact time. The first ANOVA was performed for the VM:VL CCI, and the second was performed for the Q:H CCI. Abnormalities were present in one quadriceps sensor for two participants. Therefore, VM:VL CCIs could not be calculated for those two participants. Significance for all variables was $p < 0.05$, with a power of 0.80 [64]. All statistical analyses were performed through SPSS (v24.0, SPSS Inc., Chicago, IL, USA).

المنارة للاستشارات
www.manaraa.com

CHAPTER 4:

EFFECTS OF A RAISED SURFACE ON LOWER EXTREMITY KINEMATICS, KINETICS, AND MUSCLE ACTIVATION DURING A SIDECUT IN RECREATIONAL FEMALE SOFTBALL PLAYERS

المنارة للاستشارات
www.manaraa.com

ABSTRACT

Noncontact anterior cruciate ligament (ACL) injury is a common sports-related injury. “High-risk” dynamic movements, such as a sidecut, have been associated with increasing the risk of noncontact ACL injury. Certain biomechanical abnormalities, specifically at the hip and knee, and neuromuscular abnormalities, such as unbalanced quadriceps-to-hamstrings activation ratios and certain activation patterns prior to initial contact and after initial contact, have also been associated with an increased likelihood of noncontact ACL injuries occurring. Approximately 78% of all NCAA Division I softball game-day injuries are classified as noncontact injury where there is no direct contact to a player. Internal derangement of the knee accounted for 221 game day injuries, and 31% of these injuries were noncontact ACL injuries. The base runner was at the greatest risk of injury, with 28.8% of athletes base running at the time of injury. Additionally, 9% of base runners, or 187 athletes, were injured while contacting the base. The purpose of this study was to determine the effects of a raised surface on lower extremity kinematics, kinetics, and muscle activation patterns during a sidecut, simulating rounding first base. Participants completed two base conditions – with a base present (WB) and no base (NB) present with a controlled entrance and exit speed. Results indicated the only biomechanical difference between base conditions was greater peak knee adduction moments in the NB condition compared to the WB condition. These findings suggest that the body may be in a better position when a raised surface is present during a sidecut and decrease the risk of noncontact ACL injury. Therefore, examining movement patterns at the ankle may provide a better explanation for noncontact ACL injuries that occur during this time. Regarding muscle activation, there was significantly greater quadriceps activation post-contact compared to pre-contact. Significantly greater quadriceps activation creates a large anterior shear force on the ACL, increasing risk of injury.

www.manaraa.com

INTRODUCTION

The anterior cruciate ligament (ACL) is one of the most important knee ligaments, preventing excessive anterior tibial translation, as well as frontal and transverse plane tibial rotation [1]. Unfortunately, there are approximately 80,000 to 250,000 ACL tears in the United States each year [4]. ACL injuries are classified as either a contact injury, occurring when there is direct player-to-player contact to the knee, or a noncontact injury, occurring when there is no direct contact to the knee. Approximately 70% of ACL injuries are noncontact [7] and occur during "high risk" dynamic movements that involve sudden deceleration and changes of direction, such as a sidecut [18]. These types of movements have the potential to create abnormal loading on the ACL. Simply put, an ACL injury occurs when the applied load exceeds the load that the ACL can withstand [40].

Certain biomechanical risk factors have been identified that are associated with noncontact ACL injuries. These include increased anterior tibial draw with a knee close to full extension (0 to 45° of knee flexion), and excessive internally applied knee adduction and internal rotation moments [1]. Research has also indicated abnormalities at the hip, such as decreased hip flexion angles, during a sidecut can increase the risk of ACL injury [52]. It has been suggested that this decreased hip flexion is due to the inability to decelerate in the sagittal plane, thus resulting in an increased reliance on movements in the frontal and transverse planes [52]. Most noncontact ACL injuries occur during a range of the first 40 milliseconds (ms) to 100 ms after initial contact, indicating that these biomechanical abnormalities occur during this time range [21, 23]. Some research has also found that differences in pre-activation (occurring before initial contact) can influence whether or not the proper amount of force needed to stabilize the knee is generated [17]. Females, specifically, display lower hamstring pre-activity, indicating a different

المنارة للاستشارات
www.manaraa.com

neuromuscular strategy that can result in a predisposed risk of noncontact ACL injuries occurring [34].

Injury rates have shown that females are 2-8 times more likely to sustain a noncontact ACL injury compared to males [47-50]. Females exhibit decreased hip and knee flexion, which can lead to increased anterior shear force on the ACL [32]. Increased knee abduction angles and internal knee adduction moments have also been seen in females, which can increase ACL injury risk [35, 51, 65]. Females exhibit increased quadriceps activity during sidecutting, suggesting a greater reliance on the quadriceps to stabilize the knee [17, 32]. Interestingly, when comparing lateral and medial quadriceps activations, females tend to display greater lateral quadriceps activation during sidecutting [66]. Increased quadriceps activity draws the tibia anteriorly, putting severe stress on the ACL when there is minimal knee flexion [24, 31]. Furthermore, disproportional recruitment of the lateral quadriceps influences tibial anterior shear force and potentially leads to loss of control in the frontal plane, which in turn can increase ACL injury risk [67]. The hamstrings act to pull the tibia posteriorly, which reduces the stress placed on the ACL [31]. Unfortunately, females exhibit decreased hamstring activation during dynamic movements, suggesting females may be unable to adequately reduce the stress placed on the ACL during cutting movements [32]. If there is an imbalanced ratio between quadriceps activation and hamstrings activation, an imbalance of anterior shear force to posterior shear force is created, which significantly increases the risk of ACL injury [17].

Fast-pitch softball has increased with popularity, with 19,628 NCAA Division I female athletes participating in the 2014-15 season [10]. Increased popularity in the sport means the probability of injury increases. It has been found that 78.2% (n = 2537) of all game day softball injuries were caused by noncontact means, such as running bases [12]. Internal derangement of

المنارة للاستشارات
www.manaraa.com

the knee accounted for 221 of all game day injuries, and 31% of these were classified as noncontact ACL injuries [12]. The base runner has been found to be at the highest risk of injury, with 28.8% of athletes base running at the time of injury, and 9% of injuries, or 187 injuries, were sustained while contacting the base [12]. While the number of ACL injuries is smaller compared to sports such as football or soccer, the injury still results in the most amount of game-play time lost [12]. Therefore, analyzing the lower extremity while rounding a base can potentially aid in understanding why ACL injuries occur in female softball players.

To date, no study has analyzed the effects of a raised surface on noncontact ACL injury risk factors in softball players. Therefore, the purpose of this study was to determine the effects of a raised surface (i.e. base) on the kinematics, kinetics, and muscle activation of the lower extremity while performing a sidecut in recreational softball players. Firstly, it was hypothesized that including a raised surface would increase known ACL injury risk variables. Secondly, it was hypothesized there would be greater vastus lateralis activation compared to vastus medialis activation, regardless of base condition. Lastly, it was hypothesized there would be greater hamstring activation pre-contact, but greater quadriceps activation post-contact, regardless of base condition.

METHODS

PARTICIPANTS

Ten recreationally active females (age: 21.5±2.07 yrs, height: 1.70±0.04 m, mass: 66.99±11.46 kg) participated in the study (Table 1). Inclusion criteria were between 18 to 25 years old, a minimum of two years high school softball experience to ensure they were familiar with the proper base running technique, and being recreationally active. Recreationally active was defined as being physically active at least 3 sessions per week for a minimum of 30 minutes

المنارة للاستشارات
www.manaraa.com

each session. One session had to include dynamic movements, such as running and cutting. Participants were excluded if they ever had a lower extremity injury that required surgery (e.g., ligament rupture, meniscus repair, bone fracture), had ever suffered an ACL injury, or had suffered a lower extremity injury in the past six months. Participants were also excluded if they were experiencing any lower extremity pain the day of data collection. Participants were also excluded if they scored lower than 71 on the Lower Extremity Functional Scale (LEFS).

An *a priori* power analysis, using mean peak knee adduction moment results from previous research [26, 28, 35, 54], indicated a sample size of 10 were needed for an *alpha* of 0.05 and a *beta* of 0.80. Participants provided written consent before data collection commenced, which was approved by the University Institutional Review Board prior to testing. Participants filled out the LEFS to determine if they had any difficulties performing different daily tasks [55]. The minimal detectable change and minimal clinically important difference are both 9 scale points, indicating a true change in lower extremity function.

EXPERIMENTAL PROCEDURES

A twelve-camera motion capture system (200 Hz, Vicon Motion Analysis, Inc., Centennial, CO, USA) was used to collect three-dimensional (3D) marker coordinate data of the right leg. One AMTI force platform (2000 Hz, BP600600, Advanced Mechanical Technology, Inc., Watertown, MA, USA) was used to measure ground reaction force (GRF) and identify initial contact and toe-off. The Trigno™ Wireless EMG system and four sensors (2000 Hz, Delsys, Inc., Natick, MA, USA) were used to measure muscle activation of the right leg during data collection. Prior to placement, each participants' skin was shaved and cleaned with alcohol, Sensors were placed on the muscle belly of the vastus medialis (VM), vastus lateralis (VL), biceps femoris (BF), and the semitendinosus (MH) with an interelectrode distance of 1 cm. The

المنارة للاستشارات
www.manaraa.com

exact locations were identified according to the guidelines provided by Rainoldi et al. [56]. Before data collection commenced, an isokinetic dynamometer (Biodex Medical Systems, Inc., Shirley, New York, USA) was used to collect maximum voluntary isometric contraction (MVIC) data for both the quadriceps and the hamstrings were collected with the right knee flexed at 45° [68]. Each MVIC trial began with a three-second ramp-up contraction, where participants began increasing the force, until they reached 100% of their maximal effort [69]. Maximal contractions were held for five seconds, and three trials were collected for each muscle group. These MVIC values were then used to normalize EMG levels and compare activation levels between participants.

Following MVIC testing, ten anatomical reflective markers were placed unilaterally on the 1st and 5th metatarsals, lateral and medial malleoli, and lateral and medial femoral epicondyles of the right leg. To define the pelvis, anatomical markers were placed on the left and right iliac crests, as well as the left and right greater trochanters. Three semi-rigid thermoplastic shells, each with four reflective markers, were placed on the shank, thigh, and pelvis. One semi-rigid thermoplastic shell with four reflective markers was secured to the lateral heel of the participant's shoe.

Two test conditions were utilized in the study (Figure 1). The first condition (WB) required participants to perform the sidecut on a raised surface (i.e. base, Schutt Sports Jack Corbet MLB Hollywood Baseball Base, Schutt Sports, Litchfield, IL, USA) placed in the center of the force platform. The second condition (NB) required participants to perform the same sidecut pattern, but on a flat surface. An outline of the base was taped on the force platform so participants could replicate the same cut pattern. Cutting angles were required to be between 60° 90° from the original direction of motion, with this range marked with tape on the floor. Two

المنارة للاستشارات
www.manaraa.com

pairs of timing gates (63501 IR, Lafayette Instrument Inc., IN, USA), placed 1 m apart at shoulder height, were used to maintain an entrance speed of 4.0 m/s ± 0.25 m/s and an exit speed of 3.75 m/s ± 0.25 m/s. The first pair of timing gates was placed 2 meters before the participant met the force plate, when the participant began deviating from their forward linear motion, and the second pair was placed immediately after contact with the force platform. Participants were required to maintain their speed for two full strides after completing the sidecut. A trial was considered successful if the participant maintained the entrance speed, contacted the "inside corner" of the base, or base outline with only the forefoot, and maintained the exit speed after completing the movement. Five successful trials were collected for each base condition.

DATA ANALYSIS

Reconstructed 3D marker coordinates and force data were used to compute kinematic and kinetic data of the right leg (Visual3D, v5, C-Motion, Inc., Rockville, MD, USA). Coordinate and GRF data were both filtered using a fourth-order, low-pass Butterworth filter with a cutoff frequency of 20 Hz [70]. Three-dimensional ankle, knee, and hip angles were calculated using a joint coordinate system [58]. Hip joint centers were located according to Bennett et al. [59]. The knee joint center was the midpoint between the epicondyle markers [58], and the ankle joint center was the midpoint between the malleoli markers [60]. Internally applied, three-dimensional joint kinetics were calculated using a Newton-Euler approach [61], and projected to the joint coordinate system [62]. All kinetic data were normalized to body mass. Body segment masses were estimated from Dempster et al. [63] and segment moment of inertias were estimated from Hanavan [71]. All peak values occurred during the stance phase of the sidecut. A vertical ground reaction force threshold of 10 N indicated both initial foot contact (IC) and toe-off. For the

المنارة للاستشارات
www.manaraa.com

purposes of this study, only a contact time range of 50 ms prior to initial contact (pre-contact) and 100 ms after initial contact (post-contact) were utilized for all analyses [21, 23].

Raw EMG data for MVIC and motion trials were pre-amplified and filtered with a band-pass Butterworth filter (10-350 Hz). The signal was then full wave rectified and low pass filtered at 5 Hz to create a linear envelope. The middle three seconds for each MVIC trial was used to assure a steady state activation and to identify the maximum MVIC value for each muscle for normalization purposes. EMG collected from each trial were expressed as a percentage of the EMG obtained during MVIC (%MVIC). Pre-contact (1) and post-contact (2) medial-to-lateral co-contraction indices of the quadriceps (VM:VL CCI). The VM:VL CCI was computed as the ratio of average VM activation to average VL activation.

$$Pre\text{-}contact\ VM{:}VL\ CCI = \frac{avg(VM_{50ms-HS})}{avg(VL_{50ms-HS})} \tag{1}$$

$$Post\text{-}contact\ VM{:}VL\ CCI = \frac{avg(VM_{HS-100ms\ post})}{avg(VL_{HS-100ms\ post})} \tag{2}$$

Likewise, pre-contact (3) and post-contact (4) quadriceps-to-hamstrings co-contraction indices (Q:H CCI) were computed for each trial. Q:H CCI was computed as the ratio of average quadriceps activation to average hamstring activation. Average quadriceps activation was calculated as the sum of average VM and average VL activation (5). While average hamstrings activation was calculated as the sum of average BF and average MH activation (6).

$$Pre\text{-}contact\ Q{:}H\ CCI = \frac{(Q_{50ms\ pre-HS})}{(H_{50ms\ pre-HS})} \tag{3}$$

$$Post\text{-}contact\ Q{:}H\ CCI = \frac{(Q_{HS-100ms\ post})}{(H_{HS-100ms\ post})} \tag{4}$$

$$Q = (avgVM + avgVL) \tag{5}$$

$$H = (avgBF + avgMH) \tag{6}$$

المنارة للاستشارات
www.manaraa.com

The dependent variables evaluated in this study included: initial contact hip and knee joint kinematics in the sagittal and frontal planes, peak knee abduction angle, peak hip and knee joint kinetics (internal moments) in the sagittal and frontal planes, and pre-contact and post-contact VM:VL CCI and Q:H CCI. All dependent variables were evaluated from 50 ms pre-contact to 100 ms post-contact. For each participant, all dependent variables represented the mean of the five trials collected.

STATISTICAL ANALYSIS

Paired-samples *t*-tests were conducted for all kinematic and kinetic dependent variables between base conditions. An independent *t*-test between those who completed the NB condition first and those who completed the WB condition first was utilized to determine if an order effect was present between the two base conditions. Two 2x2 (condition x contact time) repeated measure ANOVAs were performed to examine the main effects of condition and contact time for EMG data, and to determine if any significant interactions were present between condition and contact time. The first ANOVA was performed for the VM:VL CCI, and the second was performed for the Q:H CCI. Abnormalities were present in one quadriceps sensor for two participants. Therefore, VM:VL CCIs could not be calculated for those two participants. It was predicted that the WB condition would increase ACL injury risk factors. However, a non – directional hypothesis was tested because a result in either direction would be important. Significance for all variables was $p \leq 0.05$ [64]. All statistical analyses were performed through SPSS (v24.0, SPSS Inc., Chicago, IL, USA).

www.manaraa.com

RESULTS

KINEMATICS AND KINETICS

Dependent *t*-tests determined there were no significant differences between NB and WB conditions for sagittal (p = 0.746) and frontal (p = 0.779) initial contact knee and sagittal (p = 0.859) and frontal (p = 0.705) initial contact hip kinematics. There was also no significant difference in peak knee abduction angle (p = 0.917) between the two conditions (Table 3).

Dependent *t*-tests revealed a significant difference between base conditions for peak knee adduction moment (p = 0.036). Participants demonstrated larger peak knee adduction moments in the NB condition compared to the WB condition (0.51±0.46 vs. 0.31±0.29 Nm/kg, Table 4). No significant differences were found for peak knee extensor moment (p = 0.991). There were no significant differences in sagittal (p = 0.663) and frontal (p = 0.102) peak hip moments.

There was a significant order effect for the WB peak hip abduction moment (p = 0.022) and the NB peak knee adduction moment (p = 0.049, Table 7). This order effect was due to one participant's data. Once the participant's data for WB hip abduction moment and NB knee adduction moment was removed, there was no significant order effect. It should be noted that while no longer significant (p = 0.075), the mean difference remained quite large (M = 0.375).

ELECTROMYOGRAPHY

There were no main effects in VM:VL CCIs between base condition or contact time. There was a significant main effect involving contact time regarding Q:H CCI, indicating significantly greater quadriceps activation post-contact than pre-contact ($F_{1,7}$ = 18.565, p = 0.003, Table 6).

www.manaraa.com

DISCUSSION

The aim of this study was to determine effects of a raised surface on lower extremity biomechanics and muscle activation patterns during a sidecut in female softball players. Participants demonstrated significantly greater peak knee adduction moments when performing the sidecut on the flat surface compared to the raised surface. Additionally, there were no differences between VM and VL activation, both pre-contact and post-contact. However, there was a significant difference pre- and post-contact regarding Q:H CCI, with significantly greater quadriceps activation post-contact than pre-contact. It is important to note that this is the first study to examine these two specific conditions for females, rather than comparing the same condition between genders.

With respect to kinetics, our findings are similar to those previously reported for female knee adduction moments during sidecut movements [35, 54, 72]. Our hypothesis of the WB condition increasing ACL injury risk factors was not supported. It was believed that including a raised surface during a "high-risk" dynamic movement, such as the sidecut, would cause the distal end of the tibia to become more abducted, which would then work up the kinetic chain to result in a greater knee adduction moment to resist the increased abduction angle. This was not the case. It is interesting to note that while the initial contact knee abduction angles were similar between the NB and WB conditions (-9.52±4.29° vs. -9.60±3.91°, respectively, Table 3), a greater adduction moment was seen for the NB condition. It has been found that a large peak knee adduction moment is highly correlated with a large initial contact knee abduction angle in sidecutting, suggesting that increased knee abduction angles result in increased knee adduction moments [28]. If this were the case in the current study, the peak knee adduction moments would be the relatively the same for both conditions due to the initial contact knee abduction angles

www.manaraa.com

being relatively similar. A smaller peak knee adduction moment in the WB condition suggests that contacting a raised surface may provide a more optimal surface to perform and complete the sidecut, causing the GRF to potentially align more closely with the knee joint compared to cutting on a flat surface.

Hanson et al. [17] and Myer et al. [66] found that females exhibit greater vastus lateralis activity compared to males after initial contact, suggesting a different activation pattern that may result in harmful ACL loading when females perform a sidecut. Increased activation of the lateral quadriceps increases lateral joint compression, as well as causing the tibia to become more abducted [66]. This abnormal neuromuscular control potentially causes an increased internal knee adduction moment, thus increasing the risk of ACL injury. While not significant, it should be noted that females in this study exhibited slightly greater vastus medialis activity compared to vastus lateralis activity (Table 5). Therefore, the results of the current study suggest that increased ACL injury risk in female athletes performing a sidecut may not be due to unbalanced quadriceps activation pre- or post-contact.

Previous research has found that females exhibit large quadriceps activation magnitudes during the stance phase of dynamic movements, which has been shown to directly load the ACL and increase injury risk [17, 32]. Participants in the current study also demonstrated large quadriceps activation post-contact in both base conditions (Table 6). Increased quadriceps activation should not be surprising, as the quadriceps stabilize the knee and prevent excessive flexion. The peak knee extension moments, regardless of condition, were quite large (Table 4), also suggesting large contributions from the quadriceps. Increased quadriceps contraction has been shown to increase the load on the ACL and increase the risk of ACL injury, unless hamstring activation is sufficient to counteract the anterior tibial draw caused by the quadriceps

المنارة للاستشارات
www.manaraa.com

[31]. It should be noted that our post-contact Q:H CCI values for both NB (3.72±1.41) and WB (3.19±1.87) (Table 6) conditions are much greater than that found by Hanson et al. (1.55±0.63) [17]. Differences in cutting angles between the present study and Hanson et al. (60-90° vs 60°, respectively) may suggest different neuromuscular patterns are required to complete the sidecut.

Before dynamic movements, such as a sidecut, lower extremity muscles must be activated prior to ground contact in order to build up the force required for impact [34]. Specifically, the hamstrings must be pre-activated so they can produce enough force to counteract the anterior tibial pull due to increased quadriceps activation. Hamstring activation has been found to significantly decrease loading on the ACL, thus decreasing ACL injury risk [31]. Our findings do show increased hamstring activation compared to quadriceps activation prior to initial contact for both NB and WB conditions (0.82±0.32, 0.84±0.53, respectively, Table 6). Unfortunately, these CCI values suggest that pre-contact hamstring activation may not be sufficient to produce a force that can compensate the large quadriceps activation produced post-contact, thus reducing the potential for ACL protection. These results emphasize the importance of neuromuscular training, which has been found to improve hamstring activation both prior to initial contact and after initial contact and decrease ACL loading [45].

There are certain limitations that should be noted with the current study. Firstly, participants were recreational athletes who were not required to follow the same training protocol before data collection and had only a minimum of two years of high school softball experience. Thus, skill levels between participants were more than likely not the same. Secondly, the softball movement used in this study was performed on a gym floor and in a standard pair of lab shoes, rather than on a dirt field in cleats. Therefore, these findings are not completely applicable to game-day situations. The combination of rounding the base on a dirt field in cleats

المنارة للاستشارات
www.manaraa.com

may result in movement patterns and loading that could possible increase the risk of an ACL injury. Further investigation is needed to assess how cleats and dirt affect ACL-causing risk factors. Finally, foot placement, as well as entrance and exit speeds, were also controlled to minimize possible variation due to speed differences or foot placement variations. In a game-like situation, the entrance speed may be much faster, decreasing the amount of control the athlete has over foot placement and increasing the chance for injury.

CONCLUSIONS

Our findings suggest that introducing a raised surface, with a controlled foot pattern and speed, may reduce the risk of ACL injury by decreasing the knee adduction moment. As there were no differences seen at the hip joint between both conditions, examining movement patterns at the ankle may provide a better explanation for why ACL injuries occur while rounding the base. Examining abnormal foot strikes may also provide a more in depth explanation for why ACL injuries in softball players while rounding a base. Quadriceps activation was significantly greater post-contact, and hamstring pre-activation may not be sufficient to counteract the anterior pull caused by quadriceps activation post-contact. Neuromuscular training to increase hamstring activation pre-contact could potentially decrease the load applied to the ACL and decrease the risk of injury while rounding a base.

المنارة للاستشارات
www.manaraa.com

REFERENCES

www.manaraa.com

[1] Markolf, K. L., Burchfield, D. M., Shapiro, M. M., Shepard, M. F., Finerman, G. A., and Slauterbeck, J. L., 1995, "Combined knee loading states that generate high anterior cruciate ligament forces," J Orthop Res, 13(6), pp. 930-935.
[2] Noyes, F. R., and Barber-Westin, S. D., 2012, "The ACL: Anatomy, Biomechanics, Mechanisms of Injury, and the Gender Disparity," ACL Injuries in the Female Athlete: Causes, Impacts, and Conditioning Programs, R. F. Noyes, and S. Barber-Westin, eds., Springer Berlin Heidelberg, Berlin, Heidelberg, pp. 3-24.
[3] Senter, C., and Hame, S. L., 2006, "Biomechanical analysis of tibial torque and knee flexion angle: implications for understanding knee injury," Sports Med, 36(8), pp. 635-641.
[4] Griffin, L. Y., Albohm, M. J., Arendt, E. A., Bahr, R., Beynnon, B. D., Demaio, M., Dick, R. W., Engebretsen, L., Garrett, W. E., Jr., Hannafin, J. A., Hewett, T. E., Huston, L. J., Ireland, M. L., Johnson, R. J., Lephart, S., Mandelbaum, B. R., Mann, B. J., Marks, P. H., Marshall, S. W., Myklebust, G., Noyes, F. R., Powers, C., Shields, C., Jr., Shultz, S. J., Silvers, H., Slauterbeck, J., Taylor, D. C., Teitz, C. C., Wojtys, E. M., and Yu, B., 2006, "Understanding and preventing noncontact anterior cruciate ligament injuries: a review of the Hunt Valley II meeting, January 2005," Am J Sports Med, 34(9), pp. 1512-1532.
[5] Lohmander, L. S., Englund, P. M., Dahl, L. L., and Roos, E. M., 2007, "The long-term consequence of anterior cruciate ligament and meniscus injuries: osteoarthritis," Am J Sports Med, 35(10), pp. 1756-1769.
[6] McLean, S. G., Myers, P. T., Neal, R. J., and Walters, M. R., 1998, "A quantitative analysis of knee joint kinematics during the sidestep cutting maneuver. Implications for non-contact anterior cruciate ligament injury," Bull Hosp Jt Dis, 57(1), pp. 30-38.
[7] Boden, B. P., Dean, G. S., Feagin, J. A., Jr., and Garrett, W. E., Jr., 2000, "Mechanisms of anterior cruciate ligament injury," Orthopedics, 23(6), pp. 573-578.
[8] McLean, S. G., Huang, X., Su, A., and Van Den Bogert, A. J., 2004, "Sagittal plane biomechanics cannot injure the ACL during sidestep cutting," Clin Biomech (Bristol, Avon), 19(8), pp. 828-838.
[9] Weinhandl, J. T., Earl-Boehm, J. E., Ebersole, K. T., Huddleston, W. E., Armstrong, B. S., and O'Connor, K. M., 2013, "Anticipatory effects on anterior cruciate ligament loading during sidestep cutting," Clin Biomech (Bristol, Avon), 28(6), pp. 655-663.
[10] "*1981/82-2014/15 NCAA Sports Sponsorship and Participation Rates Report*," National Collegiate Athletic Association, Indianapolis, IN, October 2015.
[11] Hootman, J. M., Dick, R., and Agel, J., 2007, "Epidemiology of collegiate injuries for 15 sports: summary and recommendations for injury prevention initiatives," J Athl Train, 42(2), pp. 311-319.
[12] Marshall, S. W., Hamstra-Wright, K. L., Dick, R., Grove, K. A., and Agel, J., 2007, "Descriptive epidemiology of collegiate women's softball injuries: National Collegiate Athletic Association Injury Surveillance System, 1988-1989 through 2003-2004," J Athl Train, 42(2), pp. 286-294.
[13] Noyes, F. R., and Grood, E. S., 1976, "The strength of the anterior cruciate ligament in humans and Rhesus monkeys," J Bone Joint Surg Am, 58(8), pp. 1074-1082.
[14] Woo, S. L., Hollis, J. M., Adams, D. J., Lyon, R. M., and Takai, S., 1991, "Tensile properties of the human femur-anterior cruciate ligament-tibia complex. The effects of specimen age and orientation," Am J Sports Med, 19(3), pp. 217-225.

المنارة للاستشارات
www.manaraa.com

[15] Dempsey, A. R., Lloyd, D. G., Elliott, B. C., Steele, J. R., and Munro, B. J., 2009, "Changing sidestep cutting technique reduces knee valgus loading," Am J Sports Med, 37(11), pp. 2194-2200.
[16] Fox, A. S., Bonacci, J., McLean, S. G., Spittle, M., and Saunders, N., 2014, "What is normal? Female lower limb kinematic profiles during athletic tasks used to examine anterior cruciate ligament injury risk: a systematic review," Sports Med, 44(6), pp. 815-832.
[17] Hanson, A. M., Padua, D. A., Troy Blackburn, J., Prentice, W. E., and Hirth, C. J., 2008, "Muscle activation during side-step cutting maneuvers in male and female soccer athletes," J Athl Train, 43(2), pp. 133-143.
[18] Besier, T. F., Lloyd, D. G., Cochrane, J. L., and Ackland, T. R., 2001, "External loading of the knee joint during running and cutting maneuvers," Med Sci Sports Exerc, 33(7), pp. 1168-1175.
[19] Havens, K. L., and Sigward, S. M., 2015, "Joint and segmental mechanics differ between cutting maneuvers in skilled athletes," Gait Posture, 41(1), pp. 33-38.
[20] Hurd, W. J., Chmielewski, T. L., Axe, M. J., Davis, I., and Snyder-Mackler, L., 2004, "Differences in normal and perturbed walking kinematics between male and female athletes," Clin Biomech (Bristol, Avon), 19(5), pp. 465-472.
[21] Koga, H., Nakamae, A., Shima, Y., Iwasa, J., Myklebust, G., Engebretsen, L., Bahr, R., and Krosshaug, T., 2010, "Mechanisms for noncontact anterior cruciate ligament injuries: knee joint kinematics in 10 injury situations from female team handball and basketball," Am J Sports Med, 38(11), pp. 2218-2225.
[22] Krosshaug, T., Nakamae, A., Boden, B. P., Engebretsen, L., Smith, G., Slauterbeck, J. R., Hewett, T. E., and Bahr, R., 2007, "Mechanisms of anterior cruciate ligament injury in basketball: video analysis of 39 cases," Am J Sports Med, 35(3), pp. 359-367.
[23] Kristianslund, E., Faul, O., Bahr, R., Myklebust, G., and Krosshaug, T., 2014, "Sidestep cutting technique and knee abduction loading: implications for ACL prevention exercises," Br J Sports Med, 48(9), pp. 779-783.
[24] Colby, S., Francisco, A., Yu, B., Kirkendall, D., Finch, M., and Garrett, W., Jr., 2000, "Electromyographic and kinematic analysis of cutting maneuvers. Implications for anterior cruciate ligament injury," Am J Sports Med, 28(2), pp. 234-240.
[25] Cross, M. J., Gibbs, N. J., and Bryant, G. J., 1989, "An analysis of the sidestep cutting manoeuvre," Am J Sports Med, 17(3), pp. 363-366.
[26] Havens, K. L., and Sigward, S. M., 2015, "Cutting mechanics: relation to performance and anterior cruciate ligament injury risk," Med Sci Sports Exerc, 47(4), pp. 818-824.
[27] Wallace, B. J., Kernozek, T. W., and Bothwell, E. C., 2007, "Lower extremity kinematics and kinetics of Division III collegiate baseball and softball players while performing a modified pro-agility task," J Sports Med Phys Fitness, 47(4), pp. 377-384.
[28] McLean, S. G., Huang, X., and van den Bogert, A. J., 2005, "Association between lower extremity posture at contact and peak knee valgus moment during sidestepping: implications for ACL injury," Clin Biomech (Bristol, Avon), 20(8), pp. 863-870.
[29] McLean, S. G., Walker, K. B., and van den Bogert, A. J., 2005, "Effect of gender on lower extremity kinematics during rapid direction changes: an integrated analysis of three sports movements," J Sci Med Sport, 8(4), pp. 411-422.
[30] Besier, T. F., Lloyd, D. G., and Ackland, T. R., 2003, "Muscle activation strategies at the knee during running and cutting maneuvers," Med Sci Sports Exerc, 35(1), pp. 119-127.

المنارة للاستشارات
www.manaraa.com

[31] Renstrom, P., Arms, S. W., Stanwyck, T. S., Johnson, R. J., and Pope, M. H., 1986, "Strain within the anterior cruciate ligament during hamstring and quadriceps activity," Am J Sports Med, 14(1), pp. 83-87.
[32] Malinzak, R. A., Colby, S. M., Kirkendall, D. T., Yu, B., and Garrett, W. E., 2001, "A comparison of knee joint motion patterns between men and women in selected athletic tasks," Clin Biomech (Bristol, Avon), 16(5), pp. 438-445.
[33] Weinhandl, J. T., Earl-Boehm, J. E., Ebersole, K. T., Huddleston, W. E., Armstrong, B. S., and O'Connor, K. M., 2014, "Reduced hamstring strength increases anterior cruciate ligament loading during anticipated sidestep cutting," Clin Biomech (Bristol, Avon), 29(7), pp. 752-759.
[34] Bencke, J., and Zebis, M. K., 2011, "The influence of gender on neuromuscular pre-activity during side-cutting," J Electromyogr Kinesiol, 21(2), pp. 371-375.
[35] Sigward, S. M., and Powers, C. M., 2006, "The influence of gender on knee kinematics, kinetics and muscle activation patterns during side-step cutting," Clin Biomech (Bristol, Avon), 21(1), pp. 41-48.
[36] Stanley, L. E., Kerr, Z. Y., Dompier, T. P., and Padua, D. A., 2016, "Sex Differences in the Incidence of Anterior Cruciate Ligament, Medial Collateral Ligament, and Meniscal Injuries in Collegiate and High School Sports: 2009-2010 Through 2013-2014," Am J Sports Med, 44(6), pp. 1565-1572.
[37] Ebben, W. P., 2012, "Analysis of Male and Female Athletes' Muscle Activation Patterns During Running, Cutting, and Jumping," ACL Injuries in the Female Athlete: Causes, Impacts, and Conditioning Programs, R. F. Noyes, and S. Barber-Westin, eds., Springer Berlin Heidelberg, Berlin, Heidelberg, pp. 149-168.
[38] Butler, D. L., Noyes, F. R., and Grood, E. S., 1980, "Ligamentous restraints to anterior-posterior drawer in the human knee. A biomechanical study," J Bone Joint Surg Am, 62(2), pp. 259-270.
[39] Seering, W. P., Piziali, R. L., Nagel, D. A., and Schurman, D. J., 1980, "The function of the primary ligaments of the knee in varus-valgus and axial rotation," J Biomech, 13(9), pp. 785-794.
[40] Slauterbeck, J. R., Hickox, J. R., Beynnon, B., and Hardy, D. M., 2006, "Anterior cruciate ligament biology and its relationship to injury forces," Orthop Clin North Am, 37(4), pp. 585-591.
[41] Hashemi, J., Breighner, R., Chandrashekar, N., Hardy, D. M., Chaudhari, A. M., Shultz, S. J., Slauterbeck, J. R., and Beynnon, B. D., 2011, "Hip extension, knee flexion paradox: a new mechanism for non-contact ACL injury," J Biomech, 44(4), pp. 577-585.
[42] Vanrenterghem, J., Venables, E., Pataky, T., and Robinson, M. A., 2012, "The effect of running speed on knee mechanical loading in females during side cutting," J Biomech, 45(14), pp. 2444-2449.
[43] Houck, J., 2003, "Muscle activation patterns of selected lower extremity muscles during stepping and cutting tasks," J Electromyogr Kinesiol, 13(6), pp. 545-554.
[44] Bencke, J., Curtis, D., Krogshede, C., Jensen, L. K., Bandholm, T., and Zebis, M. K., 2013, "Biomechanical evaluation of the side-cutting manoeuvre associated with ACL injury in young female handball players," Knee Surg Sports Traumatol Arthrosc, 21(8), pp. 1876-1881.
[45] Zebis, M. K., Bencke, J., Andersen, L. L., Dossing, S., Alkjaer, T., Magnusson, S. P., Kjaer, M., and Aagaard, P., 2008, "The effects of neuromuscular training on knee joint motor control during sidecutting in female elite soccer and handball players," Clin J Sport Med, 18(4), pp. 329-337.

المنارة للاستشارات
www.manaraa.com

[46] Aagaard, P., Simonsen, E. B., Trolle, M., Bangsbo, J., and Klausen, K., 1995, "Isokinetic hamstring/quadriceps strength ratio: influence from joint angular velocity, gravity correction and contraction mode," Acta Physiol Scand, 154(4), pp. 421-427.
[47] Gray, J., Taunton, J. E., McKenzie, D. C., Clement, D. B., McConkey, J. P., and Davidson, R. G., 1985, "A survey of injuries to the anterior cruciate ligament of the knee in female basketball players," Int J Sports Med, 6(6), pp. 314-316.
[48] Messina, D. F., Farney, W. C., and DeLee, J. C., 1999, "The incidence of injury in Texas high school basketball. A prospective study among male and female athletes," Am J Sports Med, 27(3), pp. 294-299.
[49] Arendt, E., and Dick, R., 1995, "Knee injury patterns among men and women in collegiate basketball and soccer. NCAA data and review of literature," Am J Sports Med, 23(6), pp. 694-701.
[50] Agel, J., Arendt, E. A., and Bershadsky, B., 2005, "Anterior cruciate ligament injury in national collegiate athletic association basketball and soccer: a 13-year review," Am J Sports Med, 33(4), pp. 524-530.
[51] Ireland, M. L., 2002, "The female ACL: why is it more prone to injury?," Orthop Clin North Am, 33(4), pp. 637-651.
[52] Pollard, C. D., Sigward, S. M., and Powers, C. M., 2007, "Gender differences in hip joint kinematics and kinetics during side-step cutting maneuver," Clin J Sport Med, 17(1), pp. 38-42.
[53] Faul, F., Erdfelder, E., Lang, A. G., and Buchner, A., 2007, "G*Power 3: a flexible statistical power analysis program for the social, behavioral, and biomedical sciences," Behav Res Methods, 39(2), pp. 175-191.
[54] Sigward, S., and Powers, C. M., 2006, "The influence of experience on knee mechanics during side-step cutting in females," Clin Biomech (Bristol, Avon), 21(7), pp. 740-747.
[55] Binkley, J. M., Stratford, P. W., Lott, S. A., and Riddle, D. L., 1999, "The Lower Extremity Functional Scale (LEFS): scale development, measurement properties, and clinical application. North American Orthopaedic Rehabilitation Research Network," Phys Ther, 79(4), pp. 371-383.
[56] Rainoldi, A., Melchiorri, G., and Caruso, I., 2004, "A method for positioning electrodes during surface EMG recordings in lower limb muscles," J Neurosci Methods, 134(1), pp. 37-43.
[57] Blackburn, J. T., Riemann, B. L., Padua, D. A., and Guskiewicz, K. M., 2004, "Sex comparison of extensibility, passive, and active stiffness of the knee flexors," Clin Biomech (Bristol, Avon), 19(1), pp. 36-43.
[58] Grood, E. S., and Suntay, W. J., 1983, "A joint coordinate system for the clinical description of three-dimensional motions: application to the knee," J Biomech Eng, 105(2), pp. 136-144.
[59] Bennett, H. J., Shen, G., Weinhandl, J. T., and Zhang, S., 2016, "Validation of the greater trochanter method with radiographic measurements of frontal plane hip joint centers and knee mechanical axis angles and two other hip joint center methods," J Biomech, 49(13), pp. 3047-3051.
[60] Wu, G., Siegler, S., Allard, P., Kirtley, C., Leardini, A., Rosenbaum, D., Whittle, M., D'Lima, D. D., Cristofolini, L., Witte, H., Schmid, O., Stokes, I., Standardization, and Terminology Committee of the International Society of, B., 2002, "ISB recommendation on definitions of joint coordinate system of various joints for the reporting of human joint motion--part I: ankle, hip, and spine. International Society of Biomechanics," J Biomech, 35(4), pp. 543-548.

www.manaraa.com

[61] Bresler, B., and Frankel, J., 1950, "The forces and moments in the leg during level walking.," Trans. Am. Soc. Mech. Engrs., 72, pp. 27-36.
[62] Kristianslund, E., Krosshaug, T., Mok, K. M., McLean, S., and van den Bogert, A. J., 2014, "Expressing the joint moments of drop jumps and sidestep cutting in different reference frames--does it matter?," J Biomech, 47(1), pp. 193-199.
[63] Dempster, W. T., Gabel, W. C., and Felts, W. J., 1959, "The anthropometry of the manual work space for the seated subject," Am J Phys Anthropol, 17, pp. 289-317.
[64] Cohen, J., 1992, "A power primer," Psychol Bull, 112(1), pp. 155-159.
[65] Sigward, S. M., and Powers, C. M., 2007, "Loading characteristics of females exhibiting excessive valgus moments during cutting," Clin Biomech (Bristol, Avon), 22(7), pp. 827-833.
[66] Myer, G. D., Ford, K. R., and Hewett, T. E., 2005, "The effects of gender on quadriceps muscle activation strategies during a maneuver that mimics a high ACL injury risk position," J Electromyogr Kinesiol, 15(2), pp. 181-189.
[67] Sell, T. C., Ferris, C. M., Abt, J. P., Tsai, Y. S., Myers, J. B., Fu, F. H., and Lephart, S. M., 2007, "Predictors of proximal tibia anterior shear force during a vertical stop-jump," J Orthop Res, 25(12), pp. 1589-1597.
[68] LeVeau B., A. G., 1992, "Interpretation of the electromyographic signal. ," Selected Topics in Surface Eletromyography for Use in the Occupational Setting: Expert Perspectives., U. S. D. o. H. a. H. Services, ed.Atlanta, GA, pp. 70-95.
[69] Konrad, P., 2005, The ABC of EMG: A Practical Introduction to Kinesiological Eletromyography, Noraxon USA, Scottsdale, AZ.
[70] Kristianslund, E., Krosshaug, T., and van den Bogert, A. J., 2012, "Effect of low pass filtering on joint moments from inverse dynamics: implications for injury prevention," J Biomech, 45(4), pp. 666-671.
[71] Hanavan, E. P., Jr., 1964, "A Mathematical Model of the Human Body. Amrl-Tr-64-102," AMRL TR, pp. 1-149.
[72] McLean, S. G., Lipfert, S. W., and van den Bogert, A. J., 2004, "Effect of gender and defensive opponent on the biomechanics of sidestep cutting," Med Sci Sports Exerc, 36(6), pp. 1008-1016.

www.manaraa.com

APPENDICES

المنارة للاستشارات
www.manaraa.com

APPENDIX A. PARTICIPANT DEMOGRAPHICS

TABLE 1. Participant Demographics: mean ± STD.

Height (m)	Weight (kg)	Age (yrs)	Total Playing Experience (yrs)	High School/College Playing Experience (yrs)
1.70±0.04	66.99±11.46	21.5±2.07	12±3.92	4.3±1.35

المنارة للاستشارات
www.manaraa.com

TABLE 2. Individual Participant Demographics.

Subject	Height (m)	Weight (kg)	Age (yrs)	LEFS	Total Playing Experience (yrs)	High School/College Playing Experience (yrs)
1	1.72	67.67	25	79	17	8
2	1.68	60.89	21	80	8	4
3	1.67	62.91	21	80	8	3
4	1.715	62.51	21	80	15	4
5	1.725	62.18	22	80	13	4
6	1.645	68.83	21	80	13	5
7	1.715	61.43	25	76	17	4
8	1.78	96.71	19	78	8	4
9	1.705	55.01	19	80	15	4
10	1.68	71.79	21	77	6	3

www.manaraa.com

APPENDIX B. CHAPTER 4 TABLES

TABLE 3. Kinematic dependent *t*-test results (degrees): mean ± STD

Variable	NB	WB	*p*-value	Effect Size (*d*)
IC Hip Flexion Angle	38.46±7.61	38.29±5.91	0.859	0.03
IC Hip Adduction Angle	-9.14±8.95	-8.72±7.18	0.705	0.05
IC Knee Flexion Angle	-27.11±6.62	-27.48±4.61	0.746	0.07
IC Knee Abduction Angle	-1.41±5.77	-1.72±4.74	0.779	0.06
Peak Knee Abduction Angle	-9.52±4.29	-9.60±3.91	0.917	0.02

IC: initial contact

المنارة للاستشارات
www.manaraa.com

TABLE 4. Kinetic dependent *t*-test results (Nm/kg): mean ± STD

Variable	NB	WB	*p*-value	Effect Size (*d*)
Peak Hip Extension Moment	-2.15±0.43	-2.09±0.44	0.663	0.15
Peak Hip Abduction Moment	-1.27±0.41	-1.44±0.32	0.102	0.49
Peak Knee Extension Moment	2.74±0.50	2.74±0.59	0.991	0.00
Peak Knee Adduction Moment*	0.51±0.46	0.31±0.29	**0.036**	0.55

*****significant difference

للاستشارات المنارة

www.manaraa.com

TABLE 5. VM:VL CCI: mean ± STD.

				Cond.	Time	Cond*Time
		NB	**WB**	*p*	*p*	*p*
VM:VL CCI	**Pre**	1.21±0.80	1.09±0.48	0.391	0.133	0.289
	Post	1.31±0.41	1.29±0.52			

www.manaraa.com

TABLE 6. Q:H CCI: mean ± STD.

				Cond.	Time	Cond*Time
		NB	**WB**	*p*	*p*	*p*
Q:H CCI	**Pre**	0.82±0.32	0.84±0.53	0.919	**0.003**	0.710
	Post	3.72±1.41[a]	3.19±1.87[a]			

[a]: Significantly different from pre-contact

www.manaraa.com

TABLE 7. Independent *t*-test results: test for condition order effect.

	Mean Difference	*p*-value
NB IC Hip Flexion Angle	0.92	0.862
WB IC Hip Flexion Angle	-1.12	0.783
NB IC Hip Adduction Angle	-3.19	0.603
WB IC Hip Adduction Angle	-4.26	0.380
NB Peak Hip Extension Moment	0.23	0.436
WB Peak Hip Extension Moment	0.04	0.902
NB Peak Hip Abduction Moment	0.38	0.148
WB Peak Hip Abduction Moment	0.44	**0.022**
NB IC Knee Flexion Angle	3.44	0.444
WB IC Knee Flexion Angle	2.22	0.479
NB IC Knee Abduction Angle	-0.58	0.885
WB IC Knee Abduction Angle	1.63	0.616
NB Peak Knee Abduction Angle	-1.48	0.614
WB Peak Knee Abduction Angle	0.16	0.953
NB Peak Knee Extension Moment	0.42	0.207
WB Peak Knee Extension Moment	0.29	0.465
NB Peak Knee Adduction Moment	0.55	**0.049**
WB Peak Knee Adduction Moment	0.24	0.203+

www.manaraa.com

APPENDIX C. CHAPTER 4 FIGURES

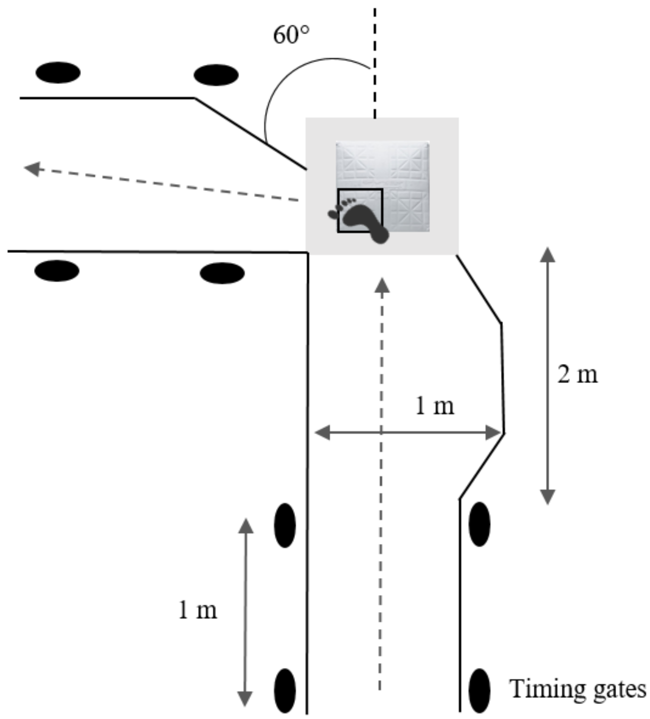

FIGURE 1. Experimental set-up. Tape was used to provide a range of 60 to 90° from the original direction of motion for the participant to perform the sidecut. Participants contacted the force plate, or base depending on condition, with their right foot and changed direction to the left. An outline of the base was provided, as well as an outline of where foot contact needed to occur, to ensure the same motion was used.

www.manaraa.com

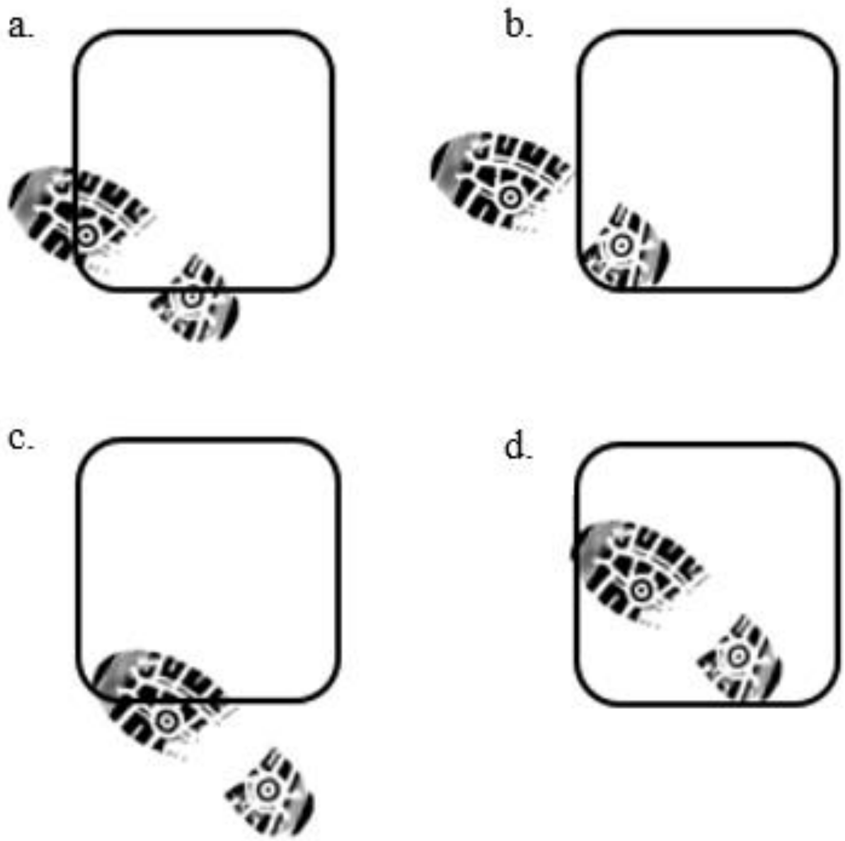

FIGURE 2. Foot placement. Foot placement was controlled during each trial. Figure 2a represents proper foot placement. Figures 2b-d represent improper foot placements. Trials were redone if foot placement was improper.

www.manaraa.com

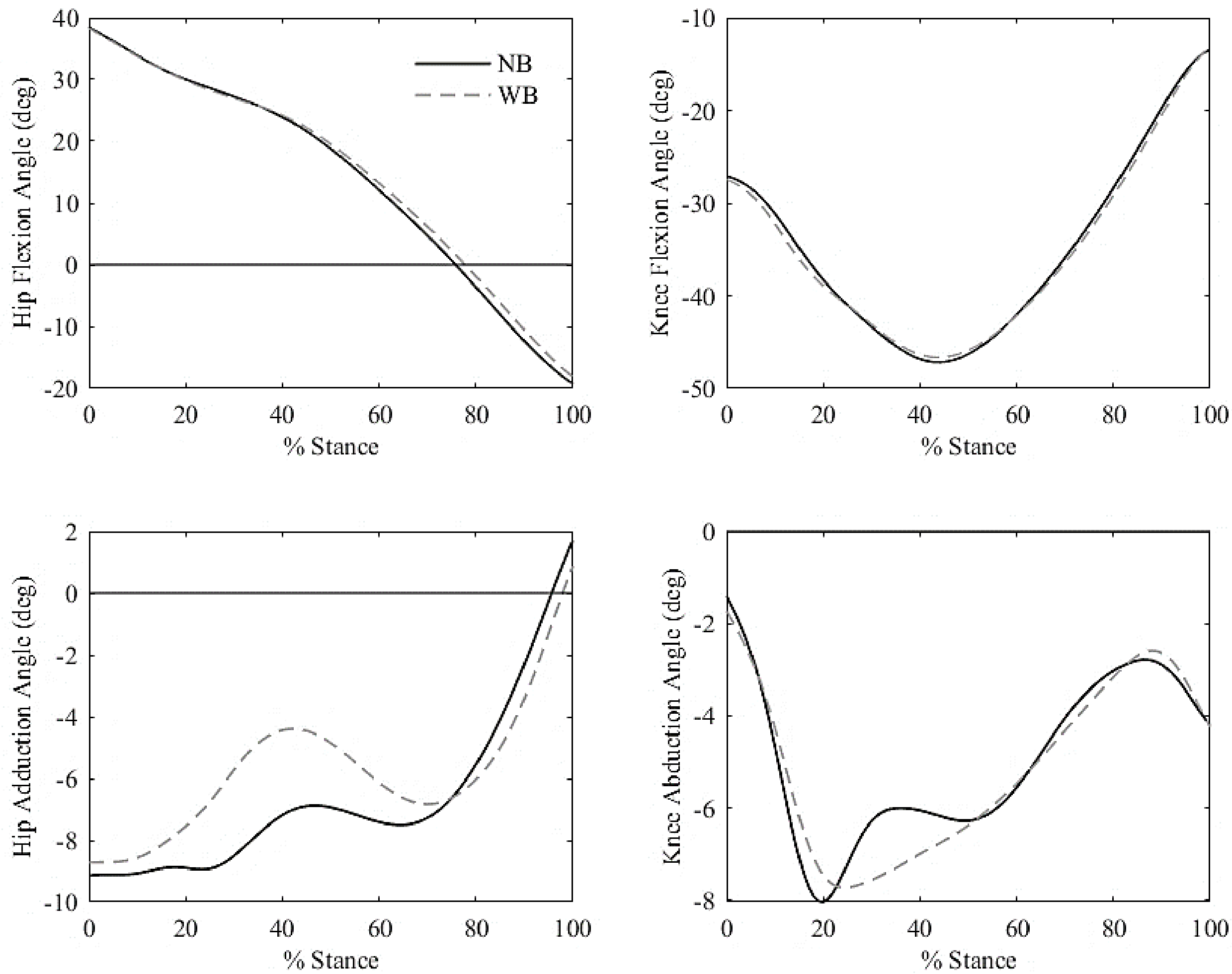

FIGURE 3. Kinematic ensemble curves. Comparison of hip and knee joint kinematics between the two base running conditions in the sagittal (upper) and frontal (lower) planes. No significant differences were observed between conditions.

المنارة للاستشارات
www.manaraa.com

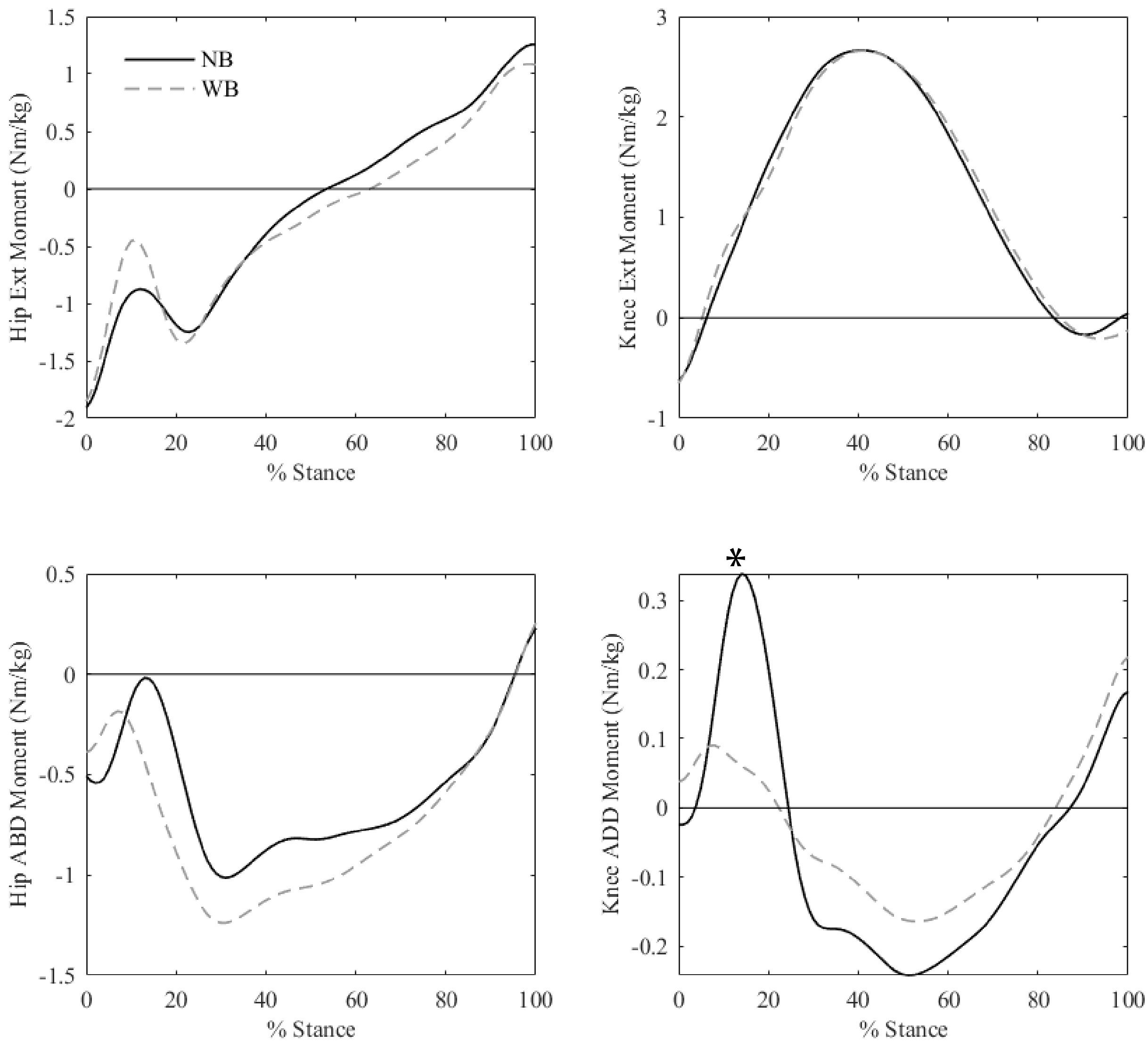

FIGURE 4. Kinetic ensemble curves. Comparison of hip and knee joint kinetics between the two base running conditions in the sagittal (upper) and frontal (lower) planes. * indicates a significant difference between NB and WB conditions ($p < 0.05$).

www.manaraa.com

APPENDIX D. INFORMED CONSENT

Informed Consent

The Effect of a Raised Surface on Lower Extremity Kinematics, Kinetics, and Muscle Activation During a 90° Sidecut in Female Recreational Softball Players

INTRODUCTION
Participants are invited to participate in a research study conducted in the University of Tennessee Biomechanics Lab (HPER 136). The purpose of this study is to determine the effects of including a raised surface on hip, knee and ankle movements and muscle activations will performing a 90° sidecut, simulating rounding first base.

ELIGIBILITY
To participate in this study, you must be between the ages of 18 and 25 and be currently recreationally active. We define recreationally active as being physically active at least 3 days per week for a minimum of 30 minutes each session, and one session must include dynamic movements (i.e. running and cutting). You must have a minimum of 2 years high school softball experience. You must NOT have: undergone surgery for a lower extremity injury (e.g., ligament rupture, meniscus repair, bone fracture), have had an anterior cruciate ligament (ACL) injury, or suffered a lower extremity injury in the past six months.

INFORMATION ABOUT PARTICIPANTS' INVOLVEMENT IN THE STUDY
You will come into the Biomechanics Lab for one session, which will last approximately an hour and a half. You will also complete the Lower Extremity Functional Scale to ensure that you are qualified to participate. A score of 71/80 or lower on the scale will exclude you from the study. You will change into Spandex shorts and a generic t-shirt, which will be provided. Height and weight will be taken, followed by a 5-minute warmup jog and stretching of the legs. Wireless electromyography (EMG) electrodes will be placed on your right quadriceps and hamstrings. Maximum contraction trials will be completed, where you will maximally contract your quadriceps, and then hamstrings, for five seconds. This information will be used to normalize the EMG data collected during the motion trials.
Reflective markers will then be placed on your right leg, which will be used to track movement for each condition. For the first condition, you will perform a sidecut that will be simulating rounding first base with a base present. In the second condition, you will perform the same route, but no base will be present. You will be allowed to practice the sidecut motion until you are comfortable with the route and the speed at which the sidecut needs to be completed in. Five successful trials for each condition will then be completed. The session will conclude once these ten total trials are collected.

RISKS
Because a dynamic sidecut movement is being performed, there is a possibility of lower extremity injury. The participant will be required to warmup with a 5-minute jog and stretching of the lower extremity to ensure their muscles are ready for this dynamic movement. The speed at which this movement will be performed will be slower than a game-like situation, ensuring

www.manaraa.com

that the participant will have control of their body throughout the sidecut. Practice runs will be before data collection begins until the participant feels comfortable with the route they are to take.

Skin irritation/rash or mild burn where electrodes are placed due to skin preparation, or adhesive may occur. Although the likelihood of these risks are low, they are still known risks associated with EMG.

A loss of confidentiality is also a possible risk related to your participation. Such a disclosure might link you to your data or your association with the study. Even when safeguards are used to minimize the likelihood of an unintentional disclosure, such a disclosure is still a risk associated with the study.

BENEFITS

There will be no direct benefits to the participant. The data collected from the participant may help provide a better understanding of what mechanisms are involved in ACL injuries that occur in softball players. The data collected may also provide athletes and coaches a better insight into how to improve base running to reduce the risk of ACL injuries occurring. A better understanding of neuromuscular characteristics during this type of movement may also lead to training interventions that may help reduce the prevalence of ACL injuries in softball players.

CONFIDENTIALITY

The information collected in this study will be kept confidential. Each participant will be identified by a given number. Data will be stored securely, both in a password-protected computer desktop and in a locked drawer in the Biomechanics lab. Information will be available only to persons conducting the study unless participants specifically give permission in writing to do otherwise. No reference will be made in oral or written reports which could link participants to the study.

CONTACT INFORMATION

If you have questions at any time about the study or the procedures, (or you experience adverse effects as a result of participating in this study,) you may contact the researcher, Lauren Schroeder, at lschroe1@vol.utk.edu or (865) 974-2091 (office number), or Dr. Joshua Weinhandl, PhD at jweinhan@utk.edu. If you have questions about your rights as a participant, you may contact the University of Tennessee IRB Compliance Officer at utkirb@utk.edu or (865) 974-7697.

PARTICIPATION

Your participation in this study is voluntary; you may decline to participate without penalty. If you decide to participate, you may withdraw from the study at any time without penalty and without loss of benefits to which you are otherwise entitled. Your participation may be ended by the investigators without regard to your consent, such as if you no longer maintain compliance with the study procedures.

المنارة للاستشارات
www.manaraa.com

CONSENT

I have read the above information. I have received a copy of this form. I agree to participate in this study.

Participant's Name (printed) ____

Participant's Signature ______________________________

المنارة للاستشارات

www.manaraa.com

APPENDIX E. LOWER EXTREMITY FUNCTIONAL SCALE

Lower Extremity Functional Scale

We are interested in knowing whether or not you are having any difficulty at all with the activities listed below. Please provide an honest answer for each activity.

KEY

0 - Extreme difficulty or unable to perform activity
1 - Quite a bit of difficulty
2 - Moderate difficulty
3 - A little bit of difficulty
4 - No difficulty

Today, ***do you*** or ***would you*** have any difficulty at all with:	Extreme 0	Quite a bit 1	Moderate 2	Minimal 3	None 4
1. Any of your usual work, housework or school activities	☐	☐	☐	☐	☐
2. Your usual hobbies, recreational or sporting activities	☐	☐	☐	☐	☐
3. Getting into or out of the bath	☐	☐	☐	☐	☐
4. Walking between rooms	☐	☐	☐	☐	☐
5. Putting on your shoes or socks	☐	☐	☐	☐	☐
6. Squatting	☐	☐	☐	☐	☐
7. Lifting an object, like a bag of groceries from the floor	☐	☐	☐	☐	☐
8. Performing light activities around your home	☐	☐	☐	☐	☐
9. Performing heavy activities around your home	☐	☐	☐	☐	☐
10. Getting into or out of a car	☐	☐	☐	☐	☐
11. Walking 2 blocks	☐	☐	☐	☐	☐
12. Walking a mile	☐	☐	☐	☐	☐
13. Going up or down 10 stairs (about 1 flight)	☐	☐	☐	☐	☐
14. Standing for 1 hour	☐	☐	☐	☐	☐
15. Sitting for 1 hour	☐	☐	☐	☐	☐
16. Running on even ground	☐	☐	☐	☐	☐
17. Running on uneven ground	☐	☐	☐	☐	☐
18. Making sharp turns while running fast	☐	☐	☐	☐	☐
19. Hopping	☐	☐	☐	☐	☐
20. Rolling over in bed	☐	☐	☐	☐	☐

www.manaraa.com

APPENDIX F. IRB APPROVAL LETTER

THE UNIVERSITY OF TENNESSEE
KNOXVILLE

December 12, 2016
Lauren Schroeder,
UTK - Kinesiology Recreation & Sport Studies

Re: UTK IRB-16-03387-XP
Study Title: The effect of a raised surface on lower extremity kinematics, kinetics, and muscle activation during a 90 degree sidecut in female recreational softball players

Dear Lauren Schroeder:

The UTK Institutional Review Board (IRB) reviewed your application for the above referenced project. It determined that your application is eligible for expedited review under 45 CFR 46.110(b)(1),categories (4), (6) and (7). The IRB has reviewed these materials and determined that they do comply with proper consideration for the rights and welfare of human subjects and the regulatory requirements for the protection of human subjects.

Therefore, this letter constitutes full approval by the IRB of your application (version1.1) as submitted, including Informed Consent (v2.1), Recruitment Flyer (v1.2), Recruitment Email (v1.5), and the LEFS Questionnaire (v1.0). The listed documents have been dated and stamped IRB approved. Approval of this study will be valid from December 12, 2016 to December 11, 2017.

In the event that subjects are to be recruited using solicitation materials, such as brochures, posters, web-based advertisements, etc., these materials must receive prior approval of the IRB. Any revisions in the approved application must also be submitted to and approved by the IRB prior to implementation. In addition, you are responsible for reporting any unanticipated serious adverse events or other problems involving risks to subjects or others in the manner required by the local IRB policy.

Finally, re-approval of your project is required by the IRB in accord with the conditions specified above. You may not continue the research study beyond the time or other limits specified unless you obtain prior written approval of the IRB. Sincerely,

Colleen P. Gilrane

Colleen P. Gilrane, Ph.D.
Chair

Institutional Review Board | Office of Research & Engagement
1534 White Avenue Knoxville, TN 37996-1529
865-974-7697 865-974-7400 fax irb.utk.edu

BIG ORANGE. BIG IDEAS.
Flagship Campus of the University of Tennessee System

المنارة للاستشارات
www.manaraa.com

APPENDIX G. INDIVIDUAL RESULTS FOR SELECT VARIABLES

TABLE 8. Individual Initial Contact Hip Kinematics: mean ± STD

Subject	IC Hip Flexion Angle (deg)		IC Hip Adduction Angle (deg)	
	NB	WB	NB	WB
1	39.999±2.488	41.140±3.136	-3.756±3.270	-6.575±4.288
2	29.591±1.176	27.316±1.495	-9.126±1.783	-14.109±3.056
3	32.873±8.344	36.236±1.910	-24.299±3.697	-19.980±1.808
4	35.170±3.385	35.453±2.818	-14.738±4.971	-12.941±5.281
5	46.405±2.591	42.180±1.109	0.392±3.448	-4.034±2.106
6	43.373±3.130	40.768±4.939	-1.277±2.586	0.230±4.013
7	37.510±2.516	37.314±3.441	-1.124±3.133	-0.470±3.434
8	36.254±2.814	38.784±1.207	-7.292±2.060	-6.167±1.595
9	29.796±4.089	33.906±3.234	-23.449±4.405	-18.384±2.282
10	53.629±2.042	49.774±2.573	-6.773±5.947	-4.771±5.660

NB: No base
WB: With base

المنارة للاستشارات
www.manaraa.com

TABLE 9. Individual Initial Contact and Peak Knee Kinematics: mean ± STD

Subject	IC Knee Flexion Angle (deg)		IC Knee Abduction Angle (deg)		Peak Knee Abduction Angle (deg)	
	NB	WB	NB	WB	NB	WB
1	-24.356±4.533	-25.719±8.055	-0.726±1.302	-1.296±0.630	-7.388±0.517	-6.942±1.250
2	-30.772±1.157	-32.467±2.103	-12.735±1.770	-8.708±1.255	-19.606±1.231	-17.574±0.466
3	-34.087±4.116	-29.622±4.653	6.040±3.331	4.051±1.460	-5.666±1.902	-7.201±1.108
4	-26.840±7.329	-26.464±5.734	-0.250±1.962	-0.333±0.739	-9.838±3.515	-6.825±2.631
5	-26.177±2.992	-27.429±1.147	5.154±2.429	1.166±1.140	-8.710±2.032	-9.814±2.371
6	-35.647±3.049	-32.430±6.515	-6.785±0.669	-7.388±1.107	-13.083±2.073	-14.162±1.661
7	-27.560±3.894	-25.691±4.768	0.874±2.248	-6.281±3.203	-6.970±0.806	-11.843±3.496
8	-22.521±2.273	-29.478±2.707	-5.003±0.500	-1.789±1.416	-10.773±1.263	-8.236±0.593
9	-12.399±2.683	-16.387±2.786	2.995±1.826	5.758±0.873	-4.870±1.444	-4.630±1.366
10	-30.786±3.360	-29.091±3.948	-3.661±2.093	-2.425±1.988	-8.277±1.613	-8.745±1.457

المنارة للاستشارات
www.manaraa.com

TABLE 10. Individual Peak Hip Kinetics: mean ± STD

Subject	Peak Hip Extension Moment (Nm/kg)		Peak Hip Abduction Moment (Nm/kg)	
	NB	WB	NB	WB
1	-2.903±8.983	-3.089±40.922	-1.959±14.398	-2.030±43.069
2	-2.090±6.713	-1.977±23.495	-1.069±13.368	-0.885±9.487
3	-1.944±15.225	-1.842±2.280	-0.753±12.849	-1.277±18.153
4	-1.897±23.048	-1.638±35.018	-0.856±9.772	-1.171±5.870
5	-1.698±26.542	-2.131±17.533	-1.357±14.838	-1.290±7.269
6	-1.817±7.155	-2.121±24.626	-1.318±8.897	-1.666±12.706
7	-2.053±0.578	-1.473±0.324	-1.900±0.220	-1.586±0.216
8	-2.594±0.162	-2.007±0.172	-1.400±0.180	-1.737±0.097
9	-1.752±0.062	-2.333±0.526	-0.901±0.242	-1.423±0.158
10	-2.728±0.236	-2.264±0.342	-1.229±0.114	-1.350±0155

المنارة للاستشارات
www.manaraa.com

TABLE 11. Individual Peak Knee Kinetics: mean ± STD

Subject	Peak Knee Extension Moment (Nm/kg)		Peak Knee Adduction Moment (Nm/kg)	
	NB	WB	NB	WB
1	2.293±0.084	1.852±0.553	0.345±0.190	0.291±0.080
2	2.675±0.078	2.861±0.282	1.497±0.196	1.095±0.192
3	2.316±0.305	3.139±0.311	0.168±0.158	0.154±0.049
4	3.450±0.661	3.535±0.256	0.846±0.167	0.234±0.166
5	2.652±0.176	2.267±0.310	0.149±0.038	0.147±0.068
6	3.461±0.185	3.199±0.254	0.518±0.088	0.166±0.080
7	2.362±0.397	2.763±0.608	-0.077±0.119	0.123±0.162
8	2.205±0.074	2.007±0.105	0.193±0.199	0.193±0.058
9	3.385±0.258	3.378±0.118	0.794±0.271	0.422±0.100
10	2.567±0.224	2.379±0.261	0.627±0.393	0.242±0.174

www.manaraa.com

TABLE 12. Individual Pre-Contact and Post-Contact VM:VL CCI: mean ± STD

Subject	Pre-Contact VM:VL CCI (%MVIC)		Post-Contact VM:VL CCI (%MVIC)	
	NB	WB	NB	WB
1	-	-	-	-
2	0.829±0.275	0.775±0.247	1.562±.0.144	0.862±0.197
3	0.589±0.150	1.080±0.283	1.419±0.465	2.171±0.182
4	0.774±0.080	0.918±0.145	0.661±0.206	0.917±0.114
5	2.170±0.467	2.211±0.527	1.748±0.217	1.788±0.334
6	2.768±1.424	1.236±0.522	8.218±6.418	1.139±0.366
7	0.834±0.208	1.009±0.184	0.828±0.128	1.026±0.150
8	-	-	-	-
9	0.851±0.159	0.749±0.193	1.383±0.238	1.396±0.253
10	1.528±0.531	1.913±0.353	1.710±0.628	1.285±0.457

VM: vastus medialis
VL: vastus laterlis

www.manaraa.com

TABLE 13. Individual Pre-Contact and Post-Contact Q:H CCI: mean ± STD

Subject	Pre-Contact Q:H CCI (%MVIC)		Post-Contact Q:H CCI (%MVIC)	
	NB	WB	NB	WB
1	1.358±0.320	1.698±0.655	6.142±1.366	6.586±1.797
2	1.379±0.202	1.587±0.138	5.836±1.027	6.174±1.232
3	0.648±0.108	0.419±0.288	3.490±1.021	2.530±0.200
4	0.987±0.133	1.462±0.369	3.877±1.301	2.896±0.551
5	0.569±0.274	0.776±0.239	2.861±0.572	4.167±1.027
6	0.506±0.106	0.520±0.177	3.949±2.011	1.395±0.378
7	0.645±0.104	0.386±0.074	2.084±0.341	1.499±0.274
8	0.609±0.333	0.401±0.095	4.163±2.143	2.867±0.601
9	0.668±0.227	0.506±0.073	2.042±0.378	1.885±0.207
10	0.839±0191	0.663±0.210	2.735±0.554	1.935±0.245

Q: quadriceps
H: hamstrings

المنارة للاستشارات
www.manaraa.com

VITA

Lauren Schroeder was born in Greenfield, IN in 1992 to Aubrey and Carol Schroeder. She is the oldest of two children and grew up in Greenfield, IN. Lauren graduated from New Palestine High School in 2010. She graduated from Ball State University in 2014 with a degree in Exercise Science. After graduation, she completed a M.S. in Kinesiology with an emphasis in Biomechanics at the University of Tennessee, Knoxville in Spring 2017. She will begin working on her PhD in Kinesiology, with a specialization in Biomechanics, at the University of Tennessee, Knoxville in Fall 2017.

www.manaraa.com