The Effect of Soleus Fatigue During Sidestep Cutting Maneuvers: Implications for the ACL

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THE EFFECT OF SOLEUS FATIGUE DURING SIDESTEP CUTTING MANEUVERS: IMPLICATIONS FOR THE ACL

A thesis submitted in partial fulfillment of the requirements for the degree of Master of Science

By

MICHAEL WILLIAM CIESA
B.S., University of Dayton, 2013

2018
Wright State University
WRIGHT STATE UNIVERSITY
GRADUATE SCHOOL

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I HERBY RECOMMEND THAT THE THESIS PREPARED UNDER MY SUPERVISION BY Michael William Ciesa ENTITLED The Effect of Soleus Fatigue During Sidestep Cutting Maneuvers: Implications for the ACL BE ACCEPTED IN PARTIAL FULLFILLMENT OF THE REQUIREMENTS FOR THE DEGREE OF Master of Science.

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ABSTRACT

Ciesa, Michael William. M.S. Department of Neuroscience, Cell Biology and Physiology, Wright State University, 2018. The Effect of Soleus Fatigue During Sidestep Cutting Maneuvers: Implications for the ACL.

The soleus muscle is a monoarticular plantarflexor composed slow-twitch fatigue-resistant muscle fibers. Through its attachment to the proximal tibia, contraction of the soleus muscle in a closed kinetic chain (when the foot is planted) produces a posterior pulling force on the posterior proximal tibia. The anterior cruciate ligament (ACL) is responsible for preventing anterior displacement of the tibia relative to the femur. Through the production of a posterior pulling force on the tibia, soleus muscle contraction in a closed kinetic chain could help reduce strain on the ACL. Fatigue is a neuromuscular phenomenon that can alter biomechanical strategies during athletics and can increase an individual’s risk of being injured. The purpose of this study is to analyze the effects soleus muscle fatigue has on knee biomechanics during the sidestep cutting maneuvers.

To analyze the effects of soleus muscle fatigue, thirteen female subjects underwent a submaximal fatigue protocol targeting the soleus muscle. Biomechanical data during sidestep cutting tasks were gathered for pre-/post-soleus fatigue conditions. Additionally, dominant/nondominant limb differences were secondarily analyzed. Results showed a significant increase in peak knee extension moment in the post-soleus fatigue
condition during the sidestep cutting task. We concluded that fatigue or excessive stretch of the soleus muscle puts the ACL at an increased risk for injury during sidestep cutting maneuvers. These findings provide evidence that the triceps surae muscles have an influence on knee biomechanics during dynamic exercise. Further investigation of these factors can help with the development of ACL injury prevention and rehabilitation strategies.
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I would like to thank my thesis adviser, Dr. Andrew Froehle, for all his guidance and instruction throughout this project. He taught me much of what I know about the field of biomechanics. When I began this project, I had little to zero experience in gait analysis; Dr. Froehle taught me how to develop and test a hypothesis in a gait lab. He was always willing to help explain any details that were unclear to me. Without his guidance, instruction, and patience I would not have been able to complete this project.

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<table>
<thead>
<tr>
<th>Abbreviation</th>
<th>Definition</th>
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<tbody>
<tr>
<td>Abd</td>
<td>Abduction</td>
</tr>
<tr>
<td>ACL</td>
<td>Anterior cruciate ligament</td>
</tr>
<tr>
<td>Add</td>
<td>Adduction</td>
</tr>
<tr>
<td>ATT</td>
<td>Anterior tibial translation</td>
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<tr>
<td>Cm</td>
<td>Centimeters</td>
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<td>CL ES</td>
<td>Common language effect size</td>
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<td>EMG</td>
<td>Electromyography</td>
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<td>Ext</td>
<td>External</td>
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<td>Fem</td>
<td>Femoral</td>
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<td>IC</td>
<td>Initial contact</td>
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<td>Int</td>
<td>Internal</td>
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<td>Milliseconds</td>
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<td>N</td>
<td>Newton</td>
</tr>
<tr>
<td>NCAAA</td>
<td>National Collegiate Athletic Association</td>
</tr>
<tr>
<td>Nm</td>
<td>Newton meter</td>
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<tr>
<td>PCL</td>
<td>Posterior cruciate ligament</td>
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<tr>
<td>Pg</td>
<td>Picogram</td>
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<tr>
<td>ROM</td>
<td>Range of motion</td>
</tr>
<tr>
<td>Rot</td>
<td>Rotation</td>
</tr>
<tr>
<td>Tib</td>
<td>Tibial</td>
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<tr>
<td>vGRF</td>
<td>Vertical ground reaction force</td>
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I. INTRODUCTION

Anterior cruciate ligament (ACL) injuries are relatively common throughout high school and collegiate sports. ACL injuries can take professional, amateur and recreational athletes off the playing field for extended periods of time. The financial burden incurred from ACL injuries for many individuals can prove overwhelming. Not only are the surgical and rehabilitation costs costly but the resultant loss of mobility can greatly affect professional, amateur and recreational athlete’s ability to make a living. The mean lifetime cost to society for ACL reconstruction is approximately $38,121 per patient while the collective lifetime burden is estimated to cost approximately $7.6 billion per year (Mather III et al., 2013). Treating ACL injury through rehabilitation is even more costly with a lifetime cost to society of approximately $88,538 per patient; collectively, accruing a lifetime burden of $17.7 billion annually (Mather III et al., 2013). Furthermore, the potential annual cost attributed to the resultant development of long-term osteoarthritis is in the $2 billion to $4 billion range (Mather III et al., 2013). The economic impacts of ACL injuries on individuals and society illustrate why prevention should be a priority. In doing so, ACL injury-related morbidity and healthcare costs can be reduced.

The incidence of ACL tears varies between genders, with high school and collegiate female athletes more than twice as likely to tear their ACL than their male counterparts (Stanley et al., 2016). This difference in injury rate between genders
continues when comparing between contact and noncontact ACL injuries (Agel et al., 2005; Stanley et al., 2016). A review of the National Collegiate Athletic Association (NCAA) Injury Surveillance System database showed that female soccer and basketball athletes tore their ACL through a noncontact mechanism 3.3 to 4.5 times more often than their male counterparts (Agel et al., 2005). Explanations for these statistics are still under investigation. However, differences in sex hormone levels, knee anatomy, or biomechanical strategies may, in part, explain some of these variances.

The large size and location of the knee during dynamic maneuvers makes it more susceptible to influences from a wide variety of factors. For example, contraction of the quadriceps muscle group can place strain (change in length as a result of stress) on the ACL by pulling anteriorly on the tibia (Li et al., 1999; Renstrom et al., 1986; Crowley Community College et al., 2001). In this way, the quadriceps are considered an antagonist to the ACL. A large majority of the current literature focuses on the large muscle groups that cross the knee (such as the quadriceps and hamstrings) as they have more obvious influences on the movement of the knee. Possible protective effects of the hamstring muscles have been the central focus of much of the research. Due to its location on the posterior thigh and attachment to the posterior tibia, the hamstring muscles are thought to reduce stress on the ACL by pulling posteriorly on the tibia (Li et al., 1999; Renstrom et al., 1986; Simonsen et al., 2000). Cadaveric and computerized modeling studies have confirmed this relationship; however, studies have also shown that at angles of greater extension (<15°) the protective effect of the hamstrings on the ACL is diminished (Renstrom et al., 1986; Simonsen et al., 2000). This would suggest that the ACL is at greater risk of injury in these more extended positions.
While it has been shown that the ACL is injured more often while the knee is in an extended position, the ACL does not injure every time the knee performs dynamic maneuvers in an extended position (Cochrane et al., 2006; Koga et al., 2010). It is likely that other less studied muscles help to provide knee stability and protect the ACL while the knee is in more extended positions. The soleus muscle in the calf is one of these muscles that has been suggested to have a protective effect on the ACL (Elias et al., 2003; Mokhtarzadeh et al., 2014). With a proximal attachment to the posterior tibia, the soleus may reduce stress on the ACL during dynamic exercise by pulling posteriorly on the tibia, limiting anterior displacement of the tibia on the ACL (Elias et al., 2003). The current literature has only demonstrated this relationship through cadaveric and computer modeling studies ((Elias et al., 2003; Mokhtarzadeh et al., 2014). However, the soleus has been shown to be highly active during many dynamic in vivo exercise studies (Mokhtarzadeh et al., 2014; Morgan et al., 2014; Simonsen et al., 2000). This study will investigate the influence of the soleus on knee stability by observing the effects fatigue of the soleus will have knee biomechanics during a dynamic exercise seen commonly with ACL injuries, sidestep cutting. In doing so we hope to gain a greater understanding of the biomechanical factors that influence the knee during exercise so that more effective ACL injury prevention and rehabilitation strategies can be developed.

**Knee Anatomy**

The knee is a large multi-articulation joint in the body consisting of the medial and lateral tibiofemoral, and patellofemoral joints (Moore et al., 2009). Functionally, the tibiofemoral joint is a modified hinge joint that allows for flexion and extension of the knee. However, when the knee is in a flexed position internal and external rotation can
occur (Moore et al., 2009). The main extracapsular ligaments of the knee include the medial and lateral collateral ligaments which aid to resist valgus and varus stress, respectively. The intracapsular posterior cruciate ligament (PCL) attaches the lateral aspect of the medial femoral condyle to the posterior aspect of the tibial intercondylar eminence of the tibia. The PCL aids to resist posterior translation of the tibia relative to the femur in the sagittal plane. Working in a contrasting role, the anterior cruciate ligament attaches to the medial aspect of the lateral femoral condyle to the anterior aspect of the tibial intercondylar eminence and has the opposite function of the PCL. An additional joint, the proximal tibiofibular joint, is in close proximity to the knee. However, its joint capsule is separate from the knee and tends to move as a rigid unit, rotating about the knee with the remainder of the leg (Moore et al., 2009).

The primary muscles involved in the movement of the knee are the anterior thigh located quadriceps (rectus femoris, vastus medialis, vastus intermedius, and vastus lateralis), the posterior thigh located hamstrings (biceps femoris, semitendinous, and semimembranous) and the gastrocnemius in the calf. The distal portions of the vastus medialis, intermedius and lateralis join and continue with the rectus femoris tendon (patella tendon) to form the quadriceps tendon which attaches to the base of the patella and then indirectly to the tibial tuberosity via the patella ligament. The vastus medialis and vastus lateralis also attach independently to the tibia via the aponeuroses known as the medial and lateral retinacula. On the posterior thigh, the semitendinosus and semimembranous travel medially and attach to the superior aspect of the medial tibia and posterior aspect of the medial tibial condyle, respectively. Traveling laterally the biceps femoris attaches to the lateral side of the head of the femur (Moore et al., 2009). The
gastrocnemius is two-headed with proximal attachments at the lateral aspect of the lateral condyle and the superior aspect of the medial condyle. The gastrocnemius will then continue distally where it joins with the soleus muscle to form the calcaneal tendon (Achilles tendon) that attaches to the calcaneus bone (Moore et al., 2009). Deep to the gastrocnemius is the soleus muscle. The soleus spans the posterior aspects of the tibia and fibula and as a plantarflexor that cross the knee, it has no direct action on the knee. However, the soleus has major implications for the biomechanics of the knee when the leg is part of a closed kinetic chain (Brunner et al., 2013; Moore et al., 2009). The function of the soleus in a closed kinetic chain will be the primary focus of this study.

**Biomechanics of Common Movements and Exercise**

The various muscles of the lower extremity have their own characteristic contribution to movement. The firing of different muscle can vary depending on the movement being performed. The way the lower extremity muscles act on joints can have large influences on the ligaments of that joint. By understanding the contributions of the soleus muscle during exercise investigators can have a greater understanding of the influence it has on joints, such as the knee. The following sections will outline the basic biomechanics of commonly performed exercises, the role of the soleus on those biomechanics and the influences it may have on the ACL.

**Standing and Posture**

The muscles of the thigh have a reduced role compared to the muscles in the calf during standing and the maintenance of posture (Moore et al., 2009). Through its proximal attachments on the leg, the soleus muscle pulls posteriorly on the bones of the leg generating a dorsiflexion moment that keeps the leg erect while standing (Brunner et
al., 2013; Moore et al., 2009). As the line of gravity passes anteriorly to the leg the soleus is considered an antigravity because it works against the effects of gravity to maintain posture (Moore et al., 2009). Containing a large density of the proprioceptive muscle spindle receptors within its muscle belly the soleus works to maintain and correct posture and balance (Edgerton et al., 1975; Purves et al., 2012). As a point of reference, the gastrocnemius has a muscle spindle density of 0.40 spindles/gram whereas the soleus has 0.94 spindles/gram (Banks, 2006). Unlike the gastrocnemius, the soleus is continuously active throughout standing and becomes more active the farther away the center of gravity moves from the ankles (Moore et al., 2009; Giulio et al., 2009). During times of static movement, the ACL is in virtually no risk of injury as sagittal and frontal plane movement is relatively minimal if at all present.

Walking

The gait cycle consists of two phases swing phase and stance phase. The swing phase begins when the toe pushes off the ground, the leg moves forward, and the heel strikes the ground. When the heel strikes the ground the stance phase begins. During initial contact ankle dorsiflexor muscles eccentrically contract and hip extensors concentrically contract. These contractions flatten the foot and decelerate the forward swing of the thigh (Moore et al., 2009). The quadriceps muscles (largely the vasti muscles) then contract to begin extension of the knee (Brunner et al., 2013; Escamilla et al., 1998). However, as the knee approaches full extension the quadriceps become less effective. The soleus will then contract to compensate for the quadriceps inability to fully extend the knee (Brunner et al., 2013; Lenhart et al., 2014). Since a planted foot is part of a closed kinetic chain, the force generated through plantarflexion while the foot is flat
acts to pull posteriorly on the tibia and fibula extending the knee in a process known as plantarflexion-knee extension coupling (Brunner et al., 2013). As the opposite limb moves towards the stance phase, the pelvis moves forward, the ankle begins isotonic plantarflexion, and the swing phase begins as the toes push off the ground (Moore et al., 2009). This cycle will alternate between limbs throughout locomotion.

It is still possible for individuals to walk after sustaining ACL injury, indicating that it is not crucial that the ACL remain intact to perform a very low-intensity exercise. However, there are populations that compensate better without an ACL than others. ACL deficient individuals who do not compensate well for their injury experience decreased compressive forces and shear forces compared to control (Alkjaer et al., 2011). ACL deficient individuals also experience decreased knee flexion angles and moments as well as decreased knee adduction moments (Slater et al., 2017). ACL injury also results in diminished quadriceps muscles especially the vastus lateralis (Williams et al., 2003). Interestingly, hamstring activity is greater and soleus activity is diminished in ACL injured individuals during the mid-stance phase gait (Hurd et al., 2007). Mid-stance is when the soleus will begin to initiate knee extension (Lenhart et al., 2014). The diminished activity of the soleus during this stage in ACL-deficient individuals suggests an interesting relationship between the soleus and ACL that should be further investigated.

**Running**

Running biomechanics are very similar to that of walking except with much quicker stance phase (Moore et al., 2009). Running lacks the “double support” seen in walking and will often be associated with a very short period of no ground contact or
“float” (Moore et al., 2009; Kennedy, 2017). The faster the running speed the shorter the stance phase and the period of float the runner experiences (Kennedy, 2017). Prior to contacting the ground, muscles will activate in preparation of the eccentric contraction needed for absorption of ground reaction forces. This eccentric contraction allows the muscles to absorb energy like a spring that then releases that energy with the followed concentric contraction (Kennedy, 2017). The soleus is mildly activated during the preparatory phase however, its activity increases more than four times during the followed concentric contraction (Dolenec et al., 2015). This increased activation in the soleus indicates that it is integral in the generation of necessary propulsion forces during running.

While the ACL is more commonly injured during dynamic maneuvers such as sidestep and crossover cutting while running the ligament can undergo angles and moments that place stress on it (Cochrane et al., 2006). Sagittal plane knee angles do not typically exceed 50° flexion during the stance phase of running and are not significantly affected by running speed (Fukuchi et al., 2017). Peak knee extension moments can significantly increase with increased running speed but not exponentially. Fukuchi et al. (2017) found knee extension moment increases of 0.5 nm/kg when increasing speeds from 2.5 m/s to 4.5 m/s. While the increased extension moment can relate to a slightly increased risk of ACL rupture, sagittal factors alone are not likely to tear the ACL (Fukuchi et al., 2017; McLean et al., 2004). Frontal plane knee angles and moments remain relatively similar with increased running speed. Peak abduction moments will slightly increase from walking to running, however, increasing the running speed does not significantly increase the abduction moment (Fukuchi et al., 2017). The increases in
knee abduction angle and moment that have been found to result in ACL injury appear to be more from dynamic maneuvers such as sidestep and crossover cutting and not from the act of running alone (Cochrane et al., 2006; Hewett et al., 2005; Koga et al., 2010).

**Sidestep Cutting**

Sidestep cutting maneuvers are an action common to athletes and are performed across many different sports and athletic competitions. The action is often employed when an athlete attempts to change direction quickly, such as to pass by a defender in basketball or to perform a juke move in football. The sidestep cut maneuver can be divided into two phases corresponding to two peaks in vGRF (vertical force exerted by ground onto body), the deceleration phase and take-off phase (Simonsen et al., 2000). The quadriceps are active throughout the cutting maneuver but are more greatly activated during the deceleration phase (Simonsen et al., 2000; Xie et al., 2012). It is the anterior drawer of the quadriceps muscle that is hypothesized to create the anterior force vector of the shearing force that is thought to tear the ACL (Colby et al., 1997). During the deceleration phase, the hamstrings muscle shortens but are activated at no more than 37% of maximum. Throughout this phase, the soleus is highly active (Simonsen et al., 2000). As there is a generation of propulsive forces, the soleus is highly active during the take-off phase as well.

As the athlete begins the deceleration phase the knee is more extended. It will then flex and extend as it transitions to the take-off phase (Chan et al., 2009; McLean et al., 1998). This is rather consistent between genders (McLean et al., 1998). Differences begin to be seen when examining frontal plane motion. Females athletes have been shown to have significantly more abducted knee angles than men when performing
athletic maneuvers (McLean et al., 1998; Hewitt et al., 2005). The knee will reach peak adduction angles approximately halfway through the contact phase before transitioning to an abduction angle more dramatic than that seen at initial contact (Chan et al., 2009). While sagittal plane factors can contribute to ACL rupture, frontal plane factors have been shown to be more significant (see below). There are many factors that can alter knee kinematics during sidestep cutting maneuvers including ball handling, field position and movement anticipation (Chan et al., 2009; Houck et al., 2005; Shojaei et al., 2012).

Possible Mechanisms for Noncontact ACL Injury

A thirteen-year review of NCAA soccer and basketball athletes found that 67% of ACL injuries in female athletes were a result of a noncontact mechanism (Agel et al., 2005). These injuries typically occur when athletes perform deceleration, sidestep cutting, jump landing and pivoting maneuvers (Boden et al., 2009; Cochrane et al., 2006; Koga et al., 2010). In an analysis of thirty-four ACL injuries that occurred during Australian rules football, Cochrane et al. (2006) found that ACL injury occurred with the knee more extended and “giving way” most commonly in valgus and internal rotation directions during the deceleration phase of the athletic maneuver. It has been theorized that valgus loading (stress) applied to the knee increases tension on the ACL. When tension in the ACL is coupled with internal rotation of the tibia (relative to the femur) it can create a shearing force that may rupture the ACL (Kanamori et al., 2000; Markolf et al., 1995; Lloyd et al., 2001).

Using a model-based image-matching technique, Koga et al. (2010) reconstructed ten ACL injury situations in female athletes during handball and basketball matches. Consistent with the results of Cochrane et al. (2006), the majority of these cases occurred
in non-contact situations when the player was cutting. The most remarkable results occurred within the first 40ms from initial contact (IC). During this time, average flexion increased by 24°, knee abduction increased by 12° and the tibia internally rotated by 8°. Internal rotation of the tibia was followed by an abrupt external rotation of 17° supposedly when the ACL ruptured. From these results, Koga et al. (2010) hypothesized a mechanism wherein a lateral compressive force (valgal loading) coupled with an anterior pulling force (from quadriceps) results in posterior translation of the lateral femoral condyle, allowing the tibia to translate anteriorly and rotate internally resulting in rupture of the ACL.

Within the first 40ms of these ACL injury scenarios from the Koga et al. (2010) study, the average peak knee abduction angle was -11.3° and the average peak knee extension angle was 22.8°. Chan et al. (2009) found that female basketball players can experience average peak abduction angles of -7.39° and average peak extension angle of 34.54° within the first 20% of the stance phase while performing 45° sidestep cutting task. Xie et al. (2012) found these angles to be more dramatic with increasing cutting angle. While the 45° and 90° sidestep cutting task of Chan et al. (2009) and Xie et al. (2012) did not produce ACL injury, it does demonstrate that the sidestep cutting maneuver can place the ACL at higher risk of injury.

In addition to dramatic changes in frontal plane mechanics, Koga et al. (2010) hypothesized that a more extended knee resulting from increased quadriceps activity can increase anterior tibial translation (ATT). This increase in ATT places increased tension on the ACL, which if excessive and coupled with frontal place forces can lead to rupture of the ACL. While the magnitude to which sagittal plane mechanics can injure the ACL
is debatable, it is generally recognized that the ACL is at greater risk of injury while the knee is in a more extended position (DeMorat et al., 2004; McLean et al., 2004; Koga et al., 2010). Kim et al. (2015) reproduced three-dimensional models based on bone bruising seen in magnetic resonance images of knees with ACL injuries sustained from noncontact mechanisms. They used these models to predict the position of the femur and the tibia during the time of injury. Their findings suggested that the knee was in an extended position at the time of injury and that the tibia had undergone a significant degree of anterior translation and internal rotation. While some of the sagittal plane bruising may have occurred post-ACL rupture, these findings help corroborate previous research (Koga et al., 2010; DeMorat et al., 2004).

**Ankle Kinematics During ACL Injury**

In a closed kinetic chain, the mechanics of the ankle can have a striking impact on knee biomechanics. Depending on the speed and running style of the individual, initial contact will be made with forefoot, midfoot or heel during running (Kennedy, 2017). As the tibia moves forward, the dorsiflexion angle of the ankle will increase (Moore et al., 2009; Souza et al., 2015). As the pelvis moves forward, the knee extends and the ankle plantarflexes as the toes push off the ground (Moore et al., 2009). A foot striking the ground with an increased dorsiflexion angle will result in increased knee extension angle as well as increased hip extension (Souza et al., 2015). A more flexed knee allows more ready dissipation of energy absorbed at impact, therefore, running with a more extended tibia can lead to a greater risk of impact injuries (Kennedy, 2017; Souza et al., 2015).

Koga et al. (2017) used a model-based imaging matching technique to analyze ankle kinematics of ten ACL injuries in women’s handball and basketball. During injury,
they found all players landed with a heel strike (mean dorsiflexion of 2°) and became flatfooted with the ankle plantarflexion angle increasing by 12° (P = 0.096) within the first 20 ms. During the next 20 ms, the ankle dorsiflexion increased 12° with the foot remaining flat. From initial contact to 40 ms the ankle supination angle by 12° (7° at IC and 19° at 40 ms) and the ankle internally rotated by 13° (5° ext. rot. at IC to 8° int. rot. at 40 ms). Consistent with these results, Boden et al. (2009) also noted flat-footedness and decreased plantarflexion angle. These results demonstrate differences in ankle kinematics during noncontact ACL injuries suggesting the ankle kinematics may influence knee injuries.

**Risk Factors for ACL Injury**

**Anatomical**

Anatomical differences may contribute to the disparity of noncontact ACL injury prevalence between males and females. A four-year study of 859 West Point Cadets found a significant main effect of gender for a variety of variables including femoral notch width (measured from med surface of lat fem condyle to lat surface of med fem condyle), tibial intercondylar eminence width, femoral notch width/intercondylar eminence width index and generalized joint laxity (measured by ability to surpass normal ROM in at least three joints bilaterally) (Uhorchak et al., 2003). Other studies have shown a greater knee valgus resultant from greater pelvis width to be more prevalent in populations of women that have injured their ACL (Griffen et al., 2000). On average female cadets had slightly decreased fem notch/intercondylar eminence width index and generalized joint laxity nearly twice as great as their male counterparts (Uhorchak et al, 2003).
When comparing ACL injured versus non-injured populations Uhorchak et al. (2003) found a significant main effect of BMI, femoral notch width index (fem notch width/fem condylar width), notch/eminence width index (fem notch/intercondylar eminence) and generalized joint laxity. ACL injured populations had decreased notch width index and notch width/eminence width index. Additionally, injured populations had a mean generalized joint laxity 1.9 times greater than uninjured populations. Their results suggest a model where higher than average BMI and greater joint laxity combined with narrow femoral notch width was able to explain 75% of the non-contact ACL injuries in females.

The results of Uhorochak et al. (2003) identified many non-modifiable variables that may predispose the ACL to rupture. However, these anatomic differences cannot fully explain the prevalence and mechanism of noncontact ACL tears or how to prevent them. Investigation of modifiable variables such as muscular strength, flexibility, and neuromuscular joint control strategies can help give a complete understanding of the prevalence and mechanism of noncontact ACL injuries. This study will not investigate the impact of anatomical differences on ACL injury risk.

**Hormonal**

Through Western blot technique, androgen receptors have been found on the ACLs of young females (Lovering et al., 2005). Lovering et al. (2005) found that the concentration of testosterone positively correlates with the firmness of the ACL during ovulation and that the ratio of 17-β estradiol negatively correlates with ligament firmness during ovulation. Stijak et al. (2014) investigated the effects of androgen levels in ACL intact and injured females and found that on average testosterone levels were
approximately 23 pg/ml lower in ACL injured populations. Their results suggest decreased testosterone levels may be correlated with ACL injury. Collectively, the literature suggests that there may be varying androgen profiles between ACL-injured and non-injured females.

The age at which menarche begins in females has also been shown to correlate with increased knee abduction angles. Froehle et al. (2017) found that individuals an earlier age of menarche had a wider base of support (measured as bicristal breadth/step width) resulting in greater knee abduction angles during walking gait. However, Field (2016) did not find a correlation between age of menarche and dynamic knee abduction angles. There are interesting findings regarding the influence of individuals androgen history and makeup that warrants further investigation. Our study does not investigate the effects of sex hormones on ACL rupture.

**Biomechanical**

The main muscle groups that work on the tibia and influence the knee are the hamstrings, quadriceps, and triceps surae (soleus and gastrocnemius). The concentric contraction of these muscles results in pulling forces on the tibia, fibula or femur where they attach. The cumulative action of the quadriceps generates a superoanterior pulling force on the tibiofibular unit resulting in extension of the knee whereas, the cumulative contraction of the hamstrings generates superoposterior pulling force on the tibia resulting in flexion of the knee (Moore et al., 2009). Cadaveric studies have shown that simulated isometric and isotonic quadriceps contraction can increase *in situ* forces and strain on the ACL through anterior translation of the tibia relative to the femur. This is especially true at angles near full knee extension with the greatest *in situ* forces and strain
seen in the ACL at 15° of flexion (Renstrom et al., 1986; Li et al., 1999). When hamstring contraction is introduced strain in the ACL, produced by the anterior drawer of the quadriceps, is significantly reduced. However, the hamstrings become less effective from 0° to 15° of knee flexion (when they are eccentrically contracting) (Renstrom et al., 1986; Li et al., 1999). Simonsen et al. (2000) demonstrated that the hamstrings concentrically contracts during sidestep cutting maneuvers, however, researchers found that the hamstrings contracted at 27% to 37% of their maximum. The authors concluded that their ability of the hamstrings to reduce anterior load (stress) on the ACL was marginal at angles near full extension. The diminished ability of the hamstring muscles to reduce ATT at more extended positions could explain why more ACL ruptures occur with the knee in an extended position (Cochrane et al., 2006).

Neuromuscular Fatigue

There is not a single, simple, universal definition of fatigue. However, it could generally be accepted that fatigue is the reversible reduction in the maximal capacity of a muscle to generate force. Fatigue can be broadly categorized into central fatigue (failures in the brain, spinal cord and up to excitation site of motoneuron), peripheral fatigue because of failure at the neuromuscular junction, or peripheral fatigue due to muscle deficits (Williams et al., 2009). During repeated tetanic contractions, like that performed in this study, there are characteristic differences in the changes in force production between muscle types. Fast-twitch muscle fibers will experience an early, small reduction in force production due to impaired myofibril function. This is then followed later by a rapid reduction in force production due to decreased release of Ca^{2+} by the sarcoplasmic reticulum (SR). In slow-twitch muscle fibers, the initial reduction in force production is
not present and the late reduction in force production comes much later than it does in fast-twitch. This reduction in force production comes so much later that slow-twitch muscle fibers are referred to as fatigue-resistant. The fatigue resistance of slow-twitch muscle fibers comes from their ability to slow SR Ca\(^{+}\) pumping, reduce ATP consumption and employ aerobic metabolism (Williams et al., 2009). Nevertheless, after exhaustive measures, slow-twitch muscle fibers can still result in decreased force production due to fatigue (Kuchinad et al., 2004).

Fatigue in the muscles surrounding joints can result in a reduction of force production from a muscle and consequent alterations to knee biomechanics (Zebis et al., 2011; Smith et al., 2009; Gehring et al., 2009; Xia et al., 2017; Field, 2016). The manner in which the muscles act on the joint determines the patterns of kinematic and kinetic change that may occur. Smith et al. (2009) showed that fatigue had no gender bias effect on vGRF, frontal plane motion, and muscle activation during landing. Twenty-six volunteers (14 women and 12 men) performed maximum voluntary isometric contractions of the lower extremity muscles by pushing against an exercise bar with their knees bent at 60° flexion angles until fatigue was reached. The fatigued subjects were then instructed to jump from a 50cm platform onto a force plate, as kinematic and EMG data were collected and analyzed. Their results found no significant difference in pre- and post-fatigue EMG amplitudes change between men and women. However, fatigue did result in a significantly decreased peak vGRF in both males and females between the pre- and post-fatigued states. While fatigue did not result in significant changes in EMG amplitudes in both females and males, there were several individuals that exhibited increased EMG amplitudes after fatigue. There were twelve individuals with significant
changes in EMG amplitudes of the medial and lateral hamstring muscles with ten of those individuals having significant increases in EMG amplitude. Their results suggest a multifaceted response to global lower extremity fatigue between individuals.

Gehring et al. (2009) found contrasting results in that muscle activation patterns differed between men and women during vertical jump landing tasks. In their study, twenty-six physically active men and women jumped from a 52cm platform in pre- and post-fatigued conditions. Lower extremity fatigue was induced via a sub-maximal fatigue protocol involving isotonic contractions on a leg press machine. Fatigue resulted in changes in knee flexion angles that were more pronounced in males. Fatigue did not result in a change in muscle activation patterns within the respective genders. However, consistent with group analysis of Smith et al. (2009), fatigue did result in an overall reduction in muscle activation amplitude. Unlike Smith et al. (2009), Gehring et al. (2009) did find a significant reduction in medial and lateral hamstring activation. The increased EMG amplitudes seen with fatigue in the single subject analysis of Smith et al. (2009) further illustrates a variable response to fatigue between individuals likely a result of many possible factors including the type of contractions (isotonic vs isometric) performed in the fatigue protocol.

**Contributions of the Triceps Surae to ACL Stability**

The contributions of the quadriceps and hamstrings muscles to ACL loading have been heavily investigated; however, there is minimal literature on the contribution of the triceps surae muscles to ACL loading. As the hamstring muscles have a diminished ability to reduce ACL loading at angles closer to full extension it is especially important to understand the role the triceps surae may have in ACL loading. The gastrocnemius is a
large plantarflexor that contributes heavily to the magnitude of ankle torque in addition to flexion of the knee (Fleming et al., 2001). As the knee is flexed the ability of the gastrocnemius to generate plantarflexion torque is decreased because the muscle fibers are at a suboptimal length to length to also act on the ankle (Cresswell et al., 1995; Bojsen-Moller et al., 2004). At flexed positions, the soleus exerts greater magnitude over plantarflexion resulting in decreased overall plantarflexor torque (Arampatzis et al., 2006; Herbert-Losier et al., 2011; Price et al., 2003; Signorile et al., 2002). Being part of a kinetic chain that begins at the ankle, changes in the ankle kinematics and kinetics can influence knee dynamics and ACL strain through effects on the tibia (Koga et al., 2018; Sherbondy et al., 2003).

Cadaveric and computer modeling studies have demonstrated an antagonistic role of the gastrocnemius to the ACL (Elias et al., 2003; Fleming et al., 2001; Adouni et al., 2016). Elias et al. (2003) dissected cadaveric knees to expose the quadriceps, semimembranosus, biceps femoris tendons as well as the soleus and gastrocnemius. Cryoclamps and constant force springs were then used to apply Newton loads to the tendons. Independent gastrocnemius muscle loading tended to translate the tibia anteriorly as a result of a relative posterior motion of the femur. While independent soleus loading translated the proximal tibia posteriorly. Additionally, at 80° and 50° of flexion when both the gastrocnemius and soleus were loaded with equal force simultaneously, tibial translation was similar to that seen when the gastrocnemius was independently loaded. At flexion angles approaching 90° the soleus exerts greater influence over plantarflexion so to apply forces of equal magnitude at these angles is not
indicative of an *in vivo* scenario and an oversimplified model (Arampatzis et al., 2006; Herbert-Losier et al., 2011; Price et al., 2003; Signorile et al., 2002).

Using transcutaneous electrical muscle stimulation, Fleming et al. (2001) induced contractions of the gastrocnemius, quadriceps, and hamstrings in six subjects with healthy knee ligaments. They then measured the resultant ACL strain via a differential variable reluctance transducer. Stimulation of the gastrocnemius, with the knee at 15° flexion, resulted in increased ACL strain values approximately 3.5 times greater than in the relaxed state. Co-contraction of the gastrocnemius with the quadriceps produced ACL strain values more than 4.5 units greater than a relaxed state ACL. The increase in ACL strain produced through contraction of the gastrocnemius lead the investigators to conclude that the gastrocnemius is an antagonist to the ACL. Interestingly, the study also found at greater flexion angles (30° to 45°) the strain on the ACL produced by the gastrocnemius was significantly decreased compared to the more extended positions. This reduction in ACL strain could be a result of the decreased influence of the gastrocnemius to plantarflex the ankle at greater angles (Arampatzis et al., 2006; Herbert-Losier et al., 2011; Price et al., 2003; Signorile et al., 2002). The soleus may have possibly lead to even greater reductions in ACL strain however, the study did not look at the effects of soleus muscle stimulation.

Morgan et al. (2014) developed muscle-tendon actuated models of single-leg jump based off EMG and motion capture data from healthy male subjects. Their results showed increased estimated gastrocnemius and quadriceps muscle forces compared to the estimated hamstring muscle forces during the weight acceptance phase of landing. The estimated force from the hamstrings was not great enough to counterbalance the
estimated quadriceps force however, estimated gastrocnemius forces were elevated when ACL strain and ACL force was lower. The quadriceps muscles have been shown to increased strain on the ACL and cadaveric studies have shown that gastrocnemius contraction can increase this strain (Li et al., 1999; Renstrom et al., 1986; Fleming et al., 2001). The latter fact is counter to the results of Morgan et al. (2014) who proposed a mechanism where the gastrocnemius increases compressive forces in the knee during single-leg landing, limiting ATT and valgal loading.

Other studies have implicated the soleus muscle as having a key role in reducing ACL strain during single-leg jumps. Mokhtarzadeh et al. (2013) used EMG and motion capture data to develop scaled-generic models and estimate muscle forces occurring during single-leg landing jumps at 30cm and 60cm heights. Throughout the landing action, estimated muscle forces were greatest in the soleus followed sequentially by the quadriceps, hamstrings, and gastrocnemius. ACL strain values peaked around 28% to 32% of the landing phase. Estimated soleus muscle force peaked at approximately 49% to 70% of the landing phase, whereas, the estimated gastrocnemius and hamstring muscle forces peaked earlier in the landing phase (<51%). Based off these results Mokhtarzadeh et al. (2013) hypothesized that the soleus has a significant role in generating posterior forces as an attempt to decrease strain on the ACL. The early peak in the estimated gastrocnemius muscle force may be an attempt to generate the compressive forces suggested by Morgan et al. (2014). However, the magnitude of the estimated gastrocnemius muscle pulling force found by Mokhtarzadeh et al. (2013) was much less than the estimated forces found by Morgan et al. (2014). Morgan et al. (2014) suggested that this discrepancy may have been a result of the modeling technique used by their
study which may be more appropriate for dynamic simulations (Seth et al., 2011; Thelen et al., 2003; Thelen et al., 2006).

The results of Mokhtarzadeh et al. (2014) fall in line with cadaveric studies that suggest an agonistic role of the soleus to the ACL (Elias et al., 2003). However, as mentioned previously cadaveric studies can greatly oversimplify the interactions occurring at the knee and do not necessarily reflect the biomechanics occurring during dynamic exercise. While unable to produce many statistically significant results the findings of Elias et al. (2003) did provide evidence that while the soleus does not cross the knee, it can still affect knee biomechanics. Through the use of clinical and cadaveric techniques, Sherbondy et al. (2003) found 35% to 45% reductions in ATT with increased tension from the gastrocnemius and soleus muscles. To simulate gastrocnemius and soleus pull the ankle was from a plantarflexion (starting at 30° plantarflexion) to dorsiflexed position (ending at 10°) while the knee was in a 30° angle and a KT-1000 arthrometer measured tibial translation. After the clinical demonstration of the effects of dorsiflexion, the results were confirmed through cadaveric studies. The cadaveric gastrocnemius was then separated and cut from the soleus and the above procedure was repeated with no significant change from the clinical findings. These results suggest that the soleus is largely acting on the tibia at 30° flexion and that it can in reducing ATT in a closed kinetic chain. It also suggests that the gastrocnemius may not significantly increase ATT as previous cadaveric studies had suggested.

**Hypotheses**

There are many factors associated with non-contact ACL ruptures including anatomical, hormonal, biomechanical and neuromuscular fatigue. It is likely that multiple
factors working together injure the ACL in noncontact situations. The focus of this study is on the biomechanical alterations that occur with neuromuscular fatigue or reduced functional capacity of the soleus muscle during sidestep cutting task. There is relatively limited information regarding the role of the soleus during dynamic exercise, however, its function during gait and posture is well documented. Cadaveric studies have outlined a clear agonistic role of the soleus to the ACL. It is our aim to bridge the gap between cadaveric and in vivo studies by providing further illumination of the role of the soleus by reducing its functional capacity during dynamic exercise. To the best of our knowledge, this is the first study that has examined the kinematic and kinetic effects that occur after putting the soleus through a targeted submaximal fatigue protocol. Additionally, this study performed a secondary analysis of the impact of leg dominance during dynamic sidestep cutting tasks. There is conflicting literature regarding the impacts of leg dominance during dynamic exercise (Greska et al., 2017; Brophy et al., 2010; Ford et al., 2003). We aim to provide further clarification regarding the influence of leg dominance and ACL injury risk.

This study tested several hypotheses regarding the effects of fatigue on knee biomechanics. Firstly, we hypothesized that soleus fatigue will increase knee flexion angle and moment during sidestep cutting maneuvers as it will have a reduced capacity to pull posteriorly on the tibia and extend the knee. Secondly, we hypothesized that soleus fatigue will increase anterior translation of the tibia during sidestep cutting maneuvers due to its inability to pull posteriorly on the tibia. Thirdly, we hypothesized that there will be asymmetries between the dominant and nondominant limbs. Brophy et al. (2010) found that soccer players were more likely to injure their support leg during gameplay
and Ford et al. (2003) found greater knee valgus in the dominant limb during drop vertical jump. As the knee experiences large valgal loading during side cuts we suspect to find these during side cutting tasks as well.
II. MATERIALS AND METHODS

Subject Population

Thirteen healthy female subjects were recruited and screened for activity level and ACL injury history prior to undergoing testing. Subjects were eligible for this study if they participated in at least moderate weekly activity and had no previous history of ACL injury. All subjects played organized sports (i.e. basketball, volleyball, soccer, softball) from ages 8 to 18 years and continued to participate in organized or recreational sports or athletic activities thereafter. All subjects considered themselves to have higher or much higher activity levels than their peers in the general population. The organized sports and recreational activities that subjects participated in at the time of the study included soccer, basketball, volleyball, mixed martial arts, distance running, cycling, swimming, powerlifting and general cardio.

Exclusion criteria included (1) subjects with a history of anterior cruciate ligament injury, (2) subjects with a current injury or pain that may affect athletic performance, (3) subject age <18 or >28. Experimental procedures were provided to the subject for review and all questions regarding experimental procedure were answered prior to testing. Informed consent was obtained from all subjects. The study was approved by the Wright State University institutional review board.
Data Collection and Instrumentation

Anthropometric data was gathered in a standard fashion (Lohman et al., 1988) while the subjects wore non-loose athletic wear and socks. Weight (to nearest 0.1 kg; scale), height (to nearest 0.1 cm; stadiometer), and sitting height (to nearest 0.1 cm; stadiometer and chair) were measured. Subischial limb length was calculated as height minus sitting height. As part of the pre-testing questionnaire, each subject was asked to specify their dominant leg. Dominant leg was defined as the leg that would be used to kick a soccer ball as far as possible.

Biomechanical data were gathered in the Wright State University three-dimensional motion analysis lab at the Lifespan Health Research Center (LHRC). The motion analysis lab is complete with six high-speed Osprey cameras (Motion Analysis Corp., Santa Rosa, CA) synchronized with three forces plates in a 15-meter walkway. One of the three force plates (Kistler Type 9281B11, Kistler Instruments, Winterthur, Switzerland) was used for data gathering. Markers were placed on all subjects for all trials by the same investigator according to the Helen Hayes system (Kadaba et al., 1990). Twenty-five reflective markers were used including bilateral markers at the mid-acromion process, lateral humeral epicondyles, mid-wrist, anterior superior iliac spine, mid-thigh, medial femoral epicondyle, lateral femoral epicondyle, mid-tibial shaft, medial malleolus, lateral malleolus, heel, and head of the second metatarsal. Additionally, a unilateral midline sacral maker was used. All twenty-five markers were used during the static trial. The medial femoral epicondyle and medial malleoli markers were removed prior to the dynamic trials. Athletic tape was used to secure the bilateral second
metatarsal, heels and lateral malleoli markers. Trials with poor marker recognition were removed from the data set.

Cortex 6.0 software was used to collect biomechanical data which was then processed using MacGait 1.0 software (Motion Analysis Corp., Santa Rosa, CA). Reported moments are external moments and were calculated using inverse dynamics. All moments were normalized for body mass. Ground reaction forces are expressed as multiples of body weight. The following conventions are used for angle and moment signs: positive values denote flexion, adduction/valgus, and internal rotation; while negative values denote extension, abduction/varus, and external rotation.

**Experimental Procedure**

Subjects came to the gait analysis lab at the LHRC for two sessions. In the first session prior to data collection, subjects answered a questionnaire regarding their exercise and knee injury history. Control data (i.e. unfatigued) was gathered for sidestep cutting task during the first session (see below for detailed methods). After control data was collected, investigators assessed the maximum number of repetitions of seated calf raise with 50% body weight resistance that subjects could perform. Condition data (i.e. fatigued) was gathered during the second session approximately 1-2 weeks after the first session to allow adequate rest for the lower limb muscles after the seated calf raise exercise. During the second session, subjects first underwent the submaximal fatigue protocol prior to data collection. Due to the nature of the fatigue protocol, not all the markers could be placed on the subject prior to starting the fatigue protocol. Therefore, immediately after the fatigue protocol, the remainder of the necessary markers (bilateral thigh, bilateral anterior superior iliac spine, bilateral second metatarsal, and bilateral
shank) were quickly placed in the appropriate areas on the subject. The subject then performed the sidestep cutting maneuvers in the soleus fatigued state while kinematic and kinetic data were collected.

**Sidestep Cutting Task**

At the beginning of the control visit (first session) subjects were given a thorough explanation of the sidestep cutting maneuver. Subjects were asked to perform practice trials of the sidestep cutting task until they became comfortable with the maneuver. Cochrane et al. (2006) found that during sidestep cutting ACL injuries, individuals typically approached the cut and a medium jog. Therefore, subjects were instructed to approach the force plate at a medium to high-speed jog/run. Upon approaching the force plate the subject was instructed to plant with one foot on the force plate and make a sidestep cut at a 45° angle in the direction opposite of their plant foot (figure 1) (Chan et al., 2009; Sankey et al., 2015). Tape was used on the floor to mark the angle that the subject was to make their cut (figure 2). Additionally, a researcher with a blocking dummy pad was placed at the end of the run to help subjects aim as well as slow down, if needed. To ensure that the subject would land on the force plate in natural stride, the appropriate starting distance was determined during the practice trials.

The direction subjects cut at was predetermined prior to the trial. Each subject was asked to complete five sidestep cutting trials to the left and five sidestep cutting trials to the right. If a subject failed to make a clean, full-foot contact with the force plate they were asked to repeat the trial. The three best trials from each leg were processed and used for statistical analysis.
Submaximal Fatigue Protocol

To set a standard for the submaximal fatigue protocol, the maximum repetitions of isotonic plantarflexions with 50% body weight resistance subjects could perform was determined. This was determined during the first session after control data was collected. During the assessment, the subject sat on a height-adjustable examination table with their toes on and their heels hanging off the edge of a 31cm plyometric box (figure 4). The height of the table was adjusted so that the subject’s knees were placed in a >90° flexion angle. At knee flexion angles >90° the gastrocnemius activity is decreased, and the soleus becomes largely responsible for plantarflexion (Arampatzis et al., 2006; Herbert-Losier et al., 2011; Price et al., 2003; Signorile et al., 2002). A barbell wrapped in a cushioned towel (for comfort) was loaded with 50% of the subject’s body weight and then placed in line with the subject’s heels on the subject’s thighs (figure 3a). The subject was then instructed to loosely hold the barbell in place with her hands and raise the weight by pushing through the balls of their feet in a plantarflexion motion (figure 3b). Upon reaching peak isotonic plantarflexion, the subject was asked to pause for a researcher measured two-count. The subject would then return to the start position and repeat the motion until the participant could no longer complete the motion with proper form or the subject felt as if they could not perform any further repetitions. The number of repetitions performed during this first session were used as a standard for the submaximal fatigue protocol in the subsequent session.

At the beginning of the second session, subjects performed the submaximal fatigue protocol prior to performing the sidestep cutting trials. During this session, subjects performed the same plantarflexion exercise with 50% body weight at 70% of the maximum repetitions they performed in the first session.
Statistical Analysis

Several hypotheses related to dynamic maneuvers and soleus fatigue were tested. Motion and force data were processed and analyzed from initial contact (IC) with the force plate to toe off (TO) during the sidestep cutting maneuvers. Initial contact was defined as the frame in which a non-zero ground reaction force (vGRF) was first measured by the force plate. Toe off was defined as the frame in which a ground reaction force was no longer detected. Maximum, minimum and average sagittal and frontal plane knee angles and moments, and sagittal plane ankle angles and moments of the plant limb were analyzed.

The findings of Koga et al., (2010) suggest that ACL injury may occur within the first 40ms from initial contact. With a capture rate of 120 frames per second, it was calculated that five frames were approximately 41.67ms. As 40ms fell in between four and five frames after initial contact, 41.67ms (five frames after initial contact) was used. Therefore, sagittal and frontal plane knee angles and moments and sagittal plane ankle angles and moments from IC to 41.67ms after initial contact were also analyzed.

To analyze sagittal plane movement of the tibia, the kinematics of the shank marker were isolated. Average sagittal plane acceleration from IC to TO and from IC to 41.67ms after initial contact was analyzed. The acceleration of the shank marker at maximum knee flexion and maximum knee extension as well as the maximum marker acceleration during the contact phase were analyzed.

All statistical tests were performed using SAS 9.3 (SAS Inc., Cary, NC, USA) with the significance level set at $\alpha = 0.05$. The statistical analysis techniques that were used include linear mixed model analysis, paired t-tests, Pearson’s correlation, and effect size analysis. To investigate the effects of soleus fatigue data from both dominant and
non-dominant limbs were pooled and analyzed. To investigate for inter-limb asymmetries data from both pre- and post-fatigue conditions were pooled and analyzed. As the sample size was small and statistical power may have been limited, effect size analysis was run, in accordance with previous methods (Lakens, 2013), for all variables with p-values <0.10.
Figure 1. Schematic of experimental setup
**Figure 1**

Diagram illustrating experimental setup for sidestep cutting trials. Six high-speed osprey cameras surrounding the 15-meter walkway with a Kistler Type 9281B11 force plate. Red lines indicate the taped pathway subjects followed during data collection. Tape was laid at a 45° angle.
Figure 2. Photograph of experimental set up

Force plate where subjects made sidestep cut

Starting Point

Running path
Figure 2

Photograph of experimental setup. Subjects ran between thick black lines. Red tape designates the 45° cutting angle. Colored lines designate potential starting positions. Starting position was adjusted, as appropriate, for each subject. Kistler force plate is labeled.
Figure 3. Diagram of submaximal fatigue protocol

A) Barbell loaded with 50% body weight
   31cm plyometric box

B) Height-adjustable bench
Diagram illustrating submaximal fatigue protocol. (A) Starting and ending positions for the submaximal fatigue protocol. Subjects sat on a height adjustable bench with their toes on the edge and heels hanging off a 31cm plyometric box. Bench height was adjusted so that knees were placed in a 90° angle. A barbell loaded with 50% of the subject’s body weight was placed on subject’s thighs while subject loosely held the bar in position. (B) Subjects were instructed to raise the barbell by pushing off the 31cm plyometric box with their toes, isotonically plantarflexing ankle. At the peak of plantarflexion, subject was instructed to hold the position for an investigator measured two-count. Subjects then returned to the starting position.
**Figure 4.** Picture of submaximal fatigue protocol setup
Figure 4

Picture showing submaximal fatigue protocol setup. Subjects sat on the height adjustable bench. Bench height was adjusted so that knees were placed in a 90° angle. Subject’s toes were placed on a 31cm plyometric box while heels hung off the edge. The distance the box was placed from the bench was adjusted so that tibia was perpendicular to the ground.
III. RESULTS

Subjects were an average age of 23.5 ± 2.8 years with an average height of 166 ± 6.8 cm, an average weight of 62.8 ± 7.4 kg, and average leg length of 74.2 ± 3.9 cm. In the pre-testing questionnaire, all subjects indicated that had higher or much higher activity levels than their peers. During the control session (1st sessions), the maximum repetitions of seated calf raise (with knees at 90°) performed by subjects on average was 163.4 ± 78.4 repetitions (table 1). There was an average weight of 31.39 ± 3.6 kg loaded on the barbell for the submaximal fatigue protocol for the first and second sessions. During the submaximal fatigue protocol in the second session, subjects were instructed to perform on average 115 ± 55.1 repetitions (70% of maximum repetitions performed during the first session) of seated calf raises prior to testing.

Interactions between Plant Limb and Soleus Fatigue Condition

A mixed models analysis showed no significant interaction between plant limb and soleus fatigue condition. There was no interaction between limb and soleus fatigue for sagittal plane knee angles (maximum flexion angle P = 0.460, minimum flexion angle P = 0.572, average flexion angle 0.491) or sagittal plane knee moments (peak flexion moment P = 0.958, peak extension moment P = 0.869, average sagittal plane moment 0.965). Frontal plane knee angles (maximum add angle P = 0.937, maximum abd angle P = 0.418, average frontal plane angle P = 0.497) and frontal plane knee moments (peak add moment P = 0.958, peak abd moment P = 0.869, average frontal plane moment (P =
showed no interaction between limb dominance and soleus fatigue. Average tibia marker acceleration ($P = 0.522$), maximum acceleration ($P = 0.842$), acceleration at maximum extension ($P = 0.950$), and acceleration at maximum flexion ($P = 0.346$) showed no interaction between limb dominance and soleus fatigue condition. Lastly, there was no interaction between limb dominance and soleus fatigue in ankle sagittal plane angles (maximum plantarflexion angle $P = 0.480$, maximum dorsiflexion angle $P = 0.963$, average sagittal plane angle $0.943$) and sagittal plane ankle moments (peak plantarflexion moment $P = 0.735$, peak dorsiflexion moment $P = 0.792$, average sagittal plane ankle moment $P = 0.938$). The results of the mixed model analysis demonstrate that the manner in which both the dominant and non-dominant limbs under pre- and post-fatigued conditions changed were consistent.

**Effects of Soleus Fatigue**

**Kinematics and Kinetics throughout Entire Contact Period**

Throughout the entire contact period data was only significant or near significance for sagittal plane angles and moments. In the soleus fatigued condition, the knee was more extended throughout all data points. The peak flexion angle ($P = 0.055$) and the peak extension angle ($P = 0.076$) were approximately $3^\circ$ more extended after fatigue (table 2). The average angle ($P = 0.045$) throughout the contact period was also $3^\circ$ more extended. Effect size analysis showed a medium effect of soleus fatigue on maximum flexion and extension angles (table 7). Maximum knee extension moment (table 3) was significantly greater (pre = -0.69 Nm Kg$^{-1}$; post = -0.88) in the soleus fatigued state ($P = 0.035$). Effect size analysis showed a medium effect (HG = 0.42; CL ES = 0.62) of soleus fatigue on extension moment.
Tibial Marker Acceleration

A change in the direction of sagittal plane motion occurs at peak knee extension and flexion angles. As the quadriceps is the main knee extensor at greater flexion angles, this muscle group will be responsible for initiating knee extension after the knee reaches maximum knee flexion (Brunner et al., 2013). To obtain an understanding of quadriceps activity (without EMG capability) we measured the acceleration of the tibia marker at maximum knee flexion as this is the likely point of maximum quadriceps contraction. Because of its anterior attachment to the tibia greater movement of the tibia likely relates to greater quadriceps activity. Our results showed a significant increase in the magnitude of tibial marker linear acceleration in the anterior direction (a negative value indicates lab’s forward direction or direction of subject’s travel) at maximum knee flexion angle in the soleus fatigued state (P = 0.035). At maximum knee flexion angle in the post-fatigue condition, the tibial marker was approximately 35.9 m/s² greater than the pre-fatigued condition (table 6). However, the standard deviation in the post-fatigue condition was quite large (approximately 115.8 m/s²). Soleus fatigue had no significant effect on maximum shank acceleration, average shank acceleration or shank acceleration at maximum knee extension.

Kinematics and Kinetics through First 41.67ms after Initial Contact

Within the first 41.67ms after initial contact soleus fatigue had no significant effect on sagittal and frontal plane angles or moments at the knee or ankle. Soleus fatigue had a medium effect (HG = 0.35; CL ES = 0.62) on knee adduction angle which was approximately 1.31° greater in the pre-fatigued state (P = 0.095). Average ankle sagittal plane moment tended toward increased plantarflexion (pre-fatigue = 0.34 ± 0.39 Nm Kg⁻¹
post-fatigue = 0.22 ± 0.34 Nm Kg$^{-1}$) however, while not significant ($P = 0.059$), soleus fatigue had a medium effect ($HG = 0.32; CL ES = 0.64$) on sagittal plane moment.

**Sagittal Plane Ankle Kinematics and Kinetics**

There were no significant effects of soleus fatigue on sagittal plane angles or moments at the ankle. Peak plantarflexion angle ($P = 0.599$) and peak dorsiflexion angle both decreased by less than a degree ($P = 0.472$) in the post-fatigue condition (table 2). Peak plantarflexion moment ($P = 0.892$) and dorsiflexion moment ($P = 0.293$) had very minimal to no change in the post-fatigue condition.

**Dominant Leg versus Nondominant Leg: All Conditions**

*Kinematics and Kinetics throughout Entire Contact Period*

The dominant limb experienced significantly greater peak adduction angle ($2.14^\circ$ greater) and significantly greater peak abduction moment ($0.33$ Nm Kg$^{-1}$) than the nondominant limb (table 2, table 3). Average frontal plane moment trended significantly toward greater abduction moment ($P = 0.002$) in the dominant limb. Effect size analysis (table 8) demonstrated a large effect of limb on the maximum adduction angle ($HG = 0.52; CL ES 0.68$), peak adduction moment ($HG = 0.95; CL ES = 0.75$), peak abduction moment ($HG = 0.70; CL ES = 0.73$) and average sagittal plane moment ($HG = 1.31; CL ES = 0.81$).

With regards to the sagittal plane, peak flexion moment was significantly greater (i.e. more flexed) in the dominant limb (table 3). There was found to be a medium effect of limb on the maximum flexion moment ($HG = 0.57; CL ES = 0.71$; table 8) and average flexion moment ($HG = 0.52; CL ES = 0.71$; table 8).
Kinematics and Kinetics through the First 41.67ms After Initial Contact

The knee frontal plane moment trended significantly (P = <0.001) toward abduction in the dominant limb (table 5). There was found to be a large effect (HG = 1.29; CL ES = 0.81) of limb on frontal plane moment within the first 41.67ms. Knee sagittal plane angle within the first 41.67ms was found to be statistically significant (P = 0.049) towards extension (dominant = 29.96 ± 8.33°; nondominant = 31.10 ± 8.44°) nonetheless, an effect analysis showed that limb had a small effect (HG = 0.13; CL ES = 0.56) on knee flexion angle. Ankle kinematics were not significant within the first 41.67ms (table 4). While ankle sagittal plane moment trended significantly (P = 0.037) toward greater plantarflexion in the dominant limb (dominant = 0.21 ± 0.37 Nm Kg⁻¹; nondominant = 0.35 ± 0.35 Nm Kg⁻¹; table 5).

Tibial Marker Acceleration

Average acceleration of the tibial marker in the sagittal plane was significantly greater (P = 0.046) in the dominant limb (dominant = -4.8 ± 3.1 m/s²; nondominant = -3.2 ± 3.6 m/s²). Effect analysis showed that limb dominance had a medium effect (HG = 0.47; CL ES = 0.66) on average tibial marker acceleration.

Sagittal Plane Ankle Kinematics and Kinetics

Sagittal plane ankle moments were statistically significant with a medium effect (table 3, table 8) at multiple points during the sidestep maneuver. The maximum plantarflexion moment was 0.72 Nm Kg⁻¹ greater in the dominant limb than nondominant limb (P = 0.028). Maximum dorsiflexion moment was 0.10 Nm Kg⁻¹ greater in the nondominant limb (P = 0.047). Average sagittal plane moment trended significantly (P =
0.025) toward plantarflexion in the dominant limb (dominant = 0.83 ± 0.22 Nm Kg⁻¹; nondominant = 0.93 ± 0.21 Nm Kg⁻¹).

**Table 1. Repetitions performed during submaximal fatigue protocol**

<table>
<thead>
<tr>
<th>Repetitions Performed at 50% Body Weight During Submaximal Fatigue Protocol</th>
<th>1st Session</th>
<th>2nd Session</th>
</tr>
</thead>
<tbody>
<tr>
<td>Average Performed</td>
<td>163.4 ± 78.4</td>
<td>115 ± 55.1</td>
</tr>
<tr>
<td>Maximum Performed</td>
<td>282</td>
<td>197</td>
</tr>
<tr>
<td>Minimum Performed</td>
<td>52</td>
<td>36</td>
</tr>
</tbody>
</table>
Table 1

Average repetitions of seated calf raise performed for all subjects. The maximum number of repetitions performed by a subject and the minimum number of repetitions performed by a subject.
Table 2. Peak joint angles during sidestep cutting task

<table>
<thead>
<tr>
<th>Joint-Action</th>
<th>Pre-soleus fatigue</th>
<th>Post-soleus fatigue</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Knee-Flexion</td>
<td>54.62 ± 10.10</td>
<td>51.70 ± 7.63</td>
<td>0.055</td>
</tr>
<tr>
<td>Knee-Extension</td>
<td>19.00 ± 8.79</td>
<td>16.54 ± 6.69</td>
<td>0.076</td>
</tr>
<tr>
<td>Knee-Adduction</td>
<td>0.64 ± 0.49</td>
<td>0.77 ± 0.44</td>
<td>0.221</td>
</tr>
<tr>
<td>Knee-Abduction</td>
<td>-0.63 ± 0.47</td>
<td>-0.74 ± 0.50</td>
<td>0.296</td>
</tr>
<tr>
<td>Ankle-Plantarflexion</td>
<td>-34.55 ± 9.00</td>
<td>-34.07 ± 8.37</td>
<td>0.599</td>
</tr>
<tr>
<td>Ankle-Dorsiflexion</td>
<td>7.10 ± 6.92</td>
<td>6.45 ± 6.42</td>
<td>0.472</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Joint-Action</th>
<th>Dominant limb</th>
<th>Non-dominant limb</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Knee-Flexion</td>
<td>53.48 ± 9.14</td>
<td>52.84 ± 8.99</td>
<td>0.942</td>
</tr>
<tr>
<td>Knee-Extension</td>
<td>17.66 ± 8.35</td>
<td>17.89 ± 7.45</td>
<td>0.920</td>
</tr>
<tr>
<td>Knee-Adduction</td>
<td>6.55 ± 4.31</td>
<td>4.42 ± 3.60</td>
<td>0.014*</td>
</tr>
<tr>
<td>Knee-Abduction</td>
<td>-0.38 ± 2.99</td>
<td>-1.55 ± 4.30</td>
<td>0.676</td>
</tr>
<tr>
<td>Ankle-Plantarflexion</td>
<td>-34.77 ± 8.87</td>
<td>-33.84 ± 8.49</td>
<td>0.508</td>
</tr>
<tr>
<td>Ankle-Dorsiflexion</td>
<td>6.61 ± 6.61</td>
<td>6.94 ± 6.75</td>
<td>0.584</td>
</tr>
</tbody>
</table>

*Indicates statistically significant p-value (p ≤ 0.05).
Table 2

Average peak extension, flexion, dorsiflexion and plantarflexion angles and p-values during the sidestep cutting task. The top section represents the results of pre-fatigued and post-fatigued conditions. The bottom section represents the results for all conditions between dominant and nondominant legs.
Table 3. Peak joint moments during sidestep cutting task

<table>
<thead>
<tr>
<th>Joint-Action</th>
<th>Pre-soleus fatigue</th>
<th>Post-soleus fatigue</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Knee-Flexion</td>
<td>2.05 ± 0.46</td>
<td>2.08 ± 0.53</td>
<td>0.864</td>
</tr>
<tr>
<td>Knee-Extension</td>
<td>-0.69 ± 0.48</td>
<td>-0.88 ± 0.40</td>
<td>0.035*</td>
</tr>
<tr>
<td>Knee-Adduction</td>
<td>0.64 ± 0.49</td>
<td>0.77 ± 0.44</td>
<td>0.317</td>
</tr>
<tr>
<td>Knee-Abduction</td>
<td>-0.63 ± 0.47</td>
<td>-0.74 ± 0.50</td>
<td>0.296</td>
</tr>
<tr>
<td>Ankle-Plantarflexion</td>
<td>-0.13 ± 0.23</td>
<td>-0.13 ± 0.21</td>
<td>0.599</td>
</tr>
<tr>
<td>Ankle-Dorsiflexion</td>
<td>1.64 ± 0.30</td>
<td>1.59 ± 0.24</td>
<td>0.892</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Joint-Action</th>
<th>Dominant limb</th>
<th>Non-dominant limb</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Knee-Flexion</td>
<td>2.21 ± 0.49</td>
<td>1.93 ± 0.46</td>
<td>0.014*</td>
</tr>
<tr>
<td>Knee-Extension</td>
<td>-0.72 ± 0.41</td>
<td>-0.86 ± 0.48</td>
<td>0.224</td>
</tr>
<tr>
<td>Knee-Adduction</td>
<td>0.50 ± 0.35</td>
<td>0.91 ± 0.49</td>
<td>0.006*</td>
</tr>
<tr>
<td>Knee-Abduction</td>
<td>-0.85 ± 0.42</td>
<td>-0.52 ± 0.49</td>
<td>&lt;0.001*</td>
</tr>
<tr>
<td>Ankle-Plantarflexion</td>
<td>-0.17 ± 0.23</td>
<td>-0.09 ± 0.19</td>
<td>0.028*</td>
</tr>
<tr>
<td>Ankle-Dorsiflexion</td>
<td>1.57 ± 0.23</td>
<td>1.67 ± 0.30</td>
<td>0.047*</td>
</tr>
</tbody>
</table>

*Indicates statistically significant p-value (p ≤ 0.05)
Table 3
Average peak extension, flexion, dorsiflexion and plantarflexion moments and p-values during the sidestep cutting task. Top section represents the results of pre-fatigued and post-fatigued conditions. Bottom section represents the results between dominant and nondominant legs.
Figure 5. Box and whiskers plot of peak extensions moments
**Figure 5**

Effects of fatigue on knee extension moment in nondominant (top) and dominant (bottom) limbs. Empty bars: pre-fatigued, filled bars: post-fatigue. Data displayed as maximum, minimum, interquartile range, and median. X indicates mean.
Table 4. Joint angles through first 41.67ms of Sidestep Cutting Task

Joint Angles (°) through first 41.67ms of Sidestep Cutting Task

<table>
<thead>
<tr>
<th>Joint-Plane</th>
<th>Pre-soleus fatigue</th>
<th>Post-fatigued</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Knee-Sagittal</td>
<td>32.09 ± 9.66</td>
<td>28.98 ± 6.55</td>
<td>0.431</td>
</tr>
<tr>
<td>Knee-Frontal</td>
<td>2.98 ± 3.71</td>
<td>1.67 ± 3.60</td>
<td>0.095</td>
</tr>
<tr>
<td>Ankle-Sagittal</td>
<td>-13.94 ± 9.28</td>
<td>-15.98 ± 9.91</td>
<td>0.127</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Joint-Plane</th>
<th>Dominant Limb</th>
<th>Non-dominant Limb</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Knee-Sagittal</td>
<td>29.96 ± 8.33</td>
<td>31.10 ± 8.44</td>
<td>0.049*</td>
</tr>
<tr>
<td>Knee-Frontal</td>
<td>2.78 ± 3.03</td>
<td>1.87 ± 4.24</td>
<td>0.223</td>
</tr>
<tr>
<td>Ankle-Sagittal</td>
<td>-15.87 ± 8.63</td>
<td>-14.05 ± 10.50</td>
<td>0.129</td>
</tr>
</tbody>
</table>

Note: negative values indicate extension and abduction; positive values indicate flexion and adduction

*Indicates statistically significant p-value (p ≤ 0.05)
**Table 4**

Mean averages of sagittal and frontal plane angle through the first 41.67ms of sidestep cutting task. P-values are also displayed. Top section represents the results between pre-fatigued and post-fatigued conditions. Bottom section represents the results between dominant and nondominant legs.
Table 5. Joint moments through first 41.67ms of sidestep cutting task

<table>
<thead>
<tr>
<th>Joint-Plane</th>
<th>Pre-soleus fatigue</th>
<th>Post-soleus fatigue</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Knee-Sagittal</td>
<td>0.33 ± 0.45</td>
<td>0.18 ± 0.31</td>
<td>0.142</td>
</tr>
<tr>
<td>Knee-Frontal</td>
<td>-0.03 ± 0.36</td>
<td>-0.08 ± 0.31</td>
<td>0.520</td>
</tr>
<tr>
<td>Ankle-Sagittal</td>
<td>0.34 ± 0.39</td>
<td>0.22 ± 0.34</td>
<td>0.059</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Joint-Plane</th>
<th>Dominant limb</th>
<th>Non-dominant limb</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Knee-Sagittal</td>
<td>0.32 ± 0.41</td>
<td>0.19 ± 0.37</td>
<td>0.079*</td>
</tr>
<tr>
<td>Knee-Frontal</td>
<td>-0.25 ± 0.27</td>
<td>0.13 ± 0.30</td>
<td>&lt;0.001*</td>
</tr>
<tr>
<td>Ankle-Sagittal</td>
<td>0.21 ± 0.37</td>
<td>0.35 ± 0.36</td>
<td>0.037*</td>
</tr>
</tbody>
</table>

Note: negative values indicate motion toward extension and abduction; positive values indicate motion toward flexion and adduction

*Indicates statistically significant p-value (p ≤ 0.05)
Table 5
Mean averages of sagittal and frontal plane moments through the first 41.67ms of sidestep cutting task. P-values are also displayed. Top section represents the results between pre-fatigued and post-fatigued conditions. Bottom section represents the results of dominant and nondominant legs.
Table 6. Tibial marker acceleration at peak joint angles

<table>
<thead>
<tr>
<th>Tibial Marker Acceleration During Sidestep Cutting Exercise</th>
<th>Sagittal Acceleration (m/s²)</th>
<th>Pre-soleus fatigue</th>
<th>Post-soleus fatigue</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>At Maximum Knee Flexion Angle</td>
<td>-25.97 ± 40.33</td>
<td>-61.92 ± 115.78</td>
<td>0.035*</td>
<td></td>
</tr>
<tr>
<td>At Maximum Knee Extension Angle</td>
<td>8.01 ± 36.61</td>
<td>5.59 ± 43.51</td>
<td>0.82</td>
<td></td>
</tr>
</tbody>
</table>

*Indicates statistically significant p-value (p ≤ 0.05)
Table 6
Tibia marker linear acceleration in the sagittal plane at maximum knee flexion and knee extension. Subjects were moving in the lab’s negative direction. A more negative acceleration indicates a higher acceleration in the direction of travel (i.e. anterior direction). Values are reported as m/s\(^2\). *Denotes statistically significant p-value (p<0.05).
Table 7. Pre-/Post-Fatigue Effect Size Analysis

<table>
<thead>
<tr>
<th>Variable with p-value &lt;0.10</th>
<th>p-value</th>
<th>Hedge’s g</th>
<th>Common Language Effect Size</th>
</tr>
</thead>
<tbody>
<tr>
<td>Average Flexion Angle</td>
<td>0.45*</td>
<td>0.32</td>
<td>0.64</td>
</tr>
<tr>
<td>Peak Flexion Angle</td>
<td>0.055</td>
<td>0.32</td>
<td>0.63</td>
</tr>
<tr>
<td>Peak Extension Angle</td>
<td>0.076</td>
<td>0.31</td>
<td>0.63</td>
</tr>
<tr>
<td>Average Flexion Moment</td>
<td>0.067</td>
<td>0.27</td>
<td>0.63</td>
</tr>
<tr>
<td>Peak Extension Moment</td>
<td>0.035*</td>
<td>0.42</td>
<td>0.62</td>
</tr>
<tr>
<td>Sagittal Tibia Marker Acceleration at Max Knee Flexion</td>
<td>0.035*</td>
<td>0.45</td>
<td>0.61</td>
</tr>
<tr>
<td>Average Plantarflexion Moment through 1st 41.67ms</td>
<td>0.82</td>
<td>0.32</td>
<td>0.64</td>
</tr>
</tbody>
</table>

Note: small effect: HG<0.3; medium effect: 0.3< HG<0.8; large effect: HG>0.8
*Indicates statistically significant p-value (p ≤ 0.05)
Table 7
Results of effect size analysis for the pre-fatigue and post-fatigue conditions during sidestep cutting task. Hedge’s g values less than 0.3 are considered a small effect, Hedge’s g between 0.3 and 0.8 are considered a medium effect, and Hedge’s g greater than 0.8 are considered a large effect. Common language effect size provides a rough percentage of how likely two variables affect each other. *Denotes a statistically significant p-value (p<0.05).
<table>
<thead>
<tr>
<th>Variable with P-value &lt;0.10</th>
<th>p-value</th>
<th>Hedge’s g</th>
<th>Common Language Effect Size</th>
</tr>
</thead>
<tbody>
<tr>
<td>Peak Knee Adduction Angle</td>
<td>0.014*</td>
<td>0.57</td>
<td>0.71</td>
</tr>
<tr>
<td>Peak Knee Flexion Moment</td>
<td>0.014*</td>
<td>0.52</td>
<td>0.68</td>
</tr>
<tr>
<td>Peak Knee Adduction Moment</td>
<td>0.006*</td>
<td>0.95</td>
<td>0.75</td>
</tr>
<tr>
<td>Peak Knee Abduction Moment</td>
<td>&lt;0.001*</td>
<td>0.70</td>
<td>0.73</td>
</tr>
<tr>
<td>Peak Ankle Plantarflexion Moment</td>
<td>0.028*</td>
<td>0.37</td>
<td>0.64</td>
</tr>
<tr>
<td>Peak Ankle Dorsiflexion Moment</td>
<td>0.047*</td>
<td>0.37</td>
<td>0.64</td>
</tr>
<tr>
<td>Average Knee-Sagittal Plane Moment</td>
<td>0.028*</td>
<td>0.52</td>
<td>0.71</td>
</tr>
<tr>
<td>Average Knee-Frontal Plane Moment</td>
<td>0.002*</td>
<td>1.31</td>
<td>0.81</td>
</tr>
<tr>
<td>Average Ankle-Sagittal Plane Moment</td>
<td>0.025*</td>
<td>0.41</td>
<td>0.66</td>
</tr>
<tr>
<td>Avg Knee-Sagittal Plane Angle 41.67ms after IC</td>
<td>0.049*</td>
<td>0.13</td>
<td>0.56</td>
</tr>
<tr>
<td>Avg Knee-Sagittal Plane Moment through 1st 41.67ms</td>
<td>0.079</td>
<td>0.32</td>
<td>0.66</td>
</tr>
<tr>
<td>Avg Knee-Frontal Plane Moment through 1st 41.67ms</td>
<td>&lt;0.001*</td>
<td>1.29</td>
<td>0.81</td>
</tr>
<tr>
<td>Avg Ankle-Sagittal Plane Moment through 1st 41.67ms</td>
<td>0.037*</td>
<td>0.41</td>
<td>0.66</td>
</tr>
<tr>
<td>Average Tibia Marker Acceleration</td>
<td>0.046*</td>
<td>0.47</td>
<td>0.66</td>
</tr>
</tbody>
</table>

Note: small effect: HG<0.3; medium effect: 0.3< HG<0.8; large effect: HG>0.8
*Indicates statistically significant p-value (p ≤ 0.05)
Table 8
Results of effect size analysis for the dominant and nondominant legs during sidestep cutting task. Hedge’s g values less than 0.3 are considered a small effect, Hedge’s g between 0.3 and 0.8 are considered a medium effect, and Hedge’s g greater than 0.8 are considered a large effect. Common language effect size provides a rough percentage of how likely two variables affect each other. *Denotes a statistically significant p-value (p<0.05).
IV. DISCUSSION

The purpose of this study was to investigate knee biomechanics during sidestep cutting maneuvers after undergoing a submaximal fatigue protocol targeting the soleus muscle. There is relatively limited information regarding the role of the soleus muscle during dynamic exercise, however, its function during gait and posture is well documented. Cadaveric studies have outlined an agonistic role of the soleus to the ACL (Elias et al., 2003; Sherbondy et al., 2003). Model-based studies have suggested that the soleus muscle produces posterior forces on the proximal tibia, therefore, reducing strain on the ACL (Mokhtazadeh et al., 2013). Muscular fatigue has been shown to affect knee biomechanics and studies that evaluate the effects of fatigue can provide a greater understanding of the function of a muscle during dynamic exercise (Gehring et al., 2009; Smith et al., 2009). There were multiple hypotheses tested and the results of our study support a few. Our first hypothesis that soleus fatigue increased flexion angle and moment during sidestep cutting task was not supported. Instead, knee extension angle and moment both increased. Our second hypothesis that soleus fatigue increases movement of the tibia during sidestep cutting task was supported by our results. Our third hypothesis that limb asymmetries would be present during sidestep cutting maneuvers was supported.
Effects of Fatigue on Knee Kinematics and Kinetics During Sidestep Cutting Task

Our results show that the knee is in a more extended position with a greater knee extension moment after soleus targeted submaximal fatigue protocol. The post-fatigue condition resulted in a 3° increase in average knee extension angle and 3° and a 0.19 Nm Kg\(^{-1}\) increase in peak knee extension moment. This increase in knee extension angle and moment may be at increased ACL injury risk as the ACL has been shown to be injured while the knee is in more extended positions (Cochrane et al., 2006; Kim et al., 2015; Koga et al., 2010). Important to note is that while sagittal factors are implicated as key components in the ACL tearing mechanism, frontal and transverse plane kinematics and kinetics have been shown to be more impactful (Cochrane et al., 2006; Koga et al., 2010; McLean et al., 2004). The following sections will provide background information and present possible hypotheses to explain our results.

Synergistic Relationship of Soleus and Quadriceps

To understand our findings, it is important to recognize the synergistic relationship observed in several studies between the soleus and quadriceps muscles (Dyer et al., 2011; Iles et al., 2000; Maupas et al., 2017; Suzuki et al., 2014). The results of Dyer et al. (2011) found increased coactivation and reduced inhibition of the soleus and quadriceps in the paretic leg of stroke victims during static knee extension activity. Dyer et al. (2011) hypothesized that a spinal pathway is involved in coactivation of knee and ankle extensors. Further evidence of a spinal pathway can be seen when studying those suffering from cerebral palsy. Individuals suffering from cerebral palsy have characteristic changes in gait consisting of abnormal plantarflexion-knee extension coupling as a result of central nervous system abnormalities (Sangeux et al., 2015).
This linkage between the soleus and quadriceps can also be seen in individuals unaffected by central nervous system deficits. Suzuki et al. (2014) found evidence of selective activation of the soleus when knee extensor and plantarflexion activity are coupled. When the quadriceps and plantarflexor muscles underwent simultaneous isometric contraction in a fully extended knee, EMG activity in the rectus femoris and soleus muscles were increased while medial gastrocnemius activity was depressed. The authors suggested that the soleus acted to compensate for the loss of medial gastrocnemius plantarflexion activity. A later, similar study found a linear relationship between quadriceps and soleus activation and concluded that this relationship was due to muscle synergies rather than joint torques (Suzuki et al., 2017).

**Quadriceps Compensation for Soleus Fatigue Hypothesis**

Fatigue is loosely defined as a decreased ability of the muscle to generate the maximum capacity of force (Williams et al., 2009). Theoretically, fatigue of a muscle should decrease the pulling force and consequent joint moment (or torque) that a muscle is able to generate (Williams et al., 2009). We hypothesized that fatigue of the soleus would result in increased flexion angle and moment. As fatigue should diminish the ability of the soleus to generate force for the secondary extension action it performs on the knee during gait. However, our results showed the opposite suggesting that other muscles may be compensating for the lack of soleus output.

Assuming our submaximal fatigue protocol induced soleus fatigue, decreasing its ability to produce force, the increased extension moment could be a result of compensation by the synergistic quadriceps muscle group. Being a larger muscle group the quadriceps muscles would exert less refined motor control and could explain why
there is an overcompensated extension moment (Purves et al., 2012). In a closed kinetic chain exercise, the monoarticular vasti muscles are approximately 50% more active than the rectus femoris (Escamilla et al., 1998). The overcompensated extension moment may be from increased activation of the vasti muscles. Increased activity of the rectus femoris is not likely as this would most likely result in increased hip flexion. Without EMG data it is not possible to confirm that the soleus was fatigued and that that fatigue resulted in increased quadriceps muscle activity. Nonetheless, increased contribution from the quadriceps would explain the increased sagittal plane acceleration of the tibial marker at maximum knee flexion (P = 0.035). This is the angle the quadriceps would generate the greatest pulling force as it moves to extend the knee. Therefore, greater activation of the quadriceps could explain the increased tibia movement.

Stutzig et al. (2015) were able to show synergistic muscle compensation as a response to fatigue. Specifically, they showed that the gastrocnemius compensated for decreased soleus plantarflexion force after electrically stimulated soleus fatigue was induced. They found similar decreases in plantarflexion torque (with the knee at 80° flexion) after fatigue of the soleus, medial gastrocnemius, and lateral gastrocnemius (performed in different sessions). However, as the soleus is largely responsible for plantarflexion at such a large flexion angle it is entirely likely that the gastrocnemius was not compensating for the soleus but that the soleus was not fatigued during the plantarflexion exercises. Due to its composition, it is very difficult to fatigue the soleus. Our study also lacked the ability to assess fatigue level via EMG technique. Therefore, other possible hypotheses must also be considered when explaining the results of our study.
Primed Soleus-Stretch Reflex Hypothesis

A component found in a large quantity in the soleus are muscle spindles (Banks, 2006). Muscle spindles are a mechanoreceptor involved in the stretch reflex response. The stretch reflex response could be another possible explanation for the increased extension moment seen in our results. While performing the isotonic plantarflexion exercise during the submaximal fatigue protocol, the soleus would have been repeatedly stretched activating muscle spindles in the muscle belly. Upon activation, the intrafusal fibers within the muscle spindle would have activated mechanically gated ion channels sending action potentials to Ia afferent fibers of the tibial nerve. These action potentials would then have traveled to the spinal cord where they synapse onto α motor neurons. The α motor neurons would then send action potentials to the soleus and its synergistic muscles increased their activation and contraction (Purves et al., 2012; Suzuki et al., 2014). Due to the previously demonstrated coactivation properties, it is likely that the quadriceps muscles would be a synergistic muscle experiencing increased activation through this pathway (Dyer et al., 2011; Iles et al., 2000; Maupas et al., 2017; Suzuki et al., 2014).

While increased quadriceps activation cannot be confirmed without EMG data, increased activation and contraction from the quadriceps would explain the increased sagittal acceleration of the tibia marker seen at maximum knee flexion. The stretch reflex would also account for the decreased dorsiflexion moment seen within the first 41.67ms as increased soleus activity from activation through the soleus stretch reflex pathway would lead to greater plantarflexion activity. If our fatigue protocol was not able to reduce the force capacity (fatigue) the soleus could exert on the proximal tibia, increased
activity of the soleus and its synergistic muscles through the stretch reflex could explain our results.

Soleus Fatigue and ACL Rupture Risk

Greater knee extension angles and increased anterior tibial translation have been found to be associated with noncontact ACL injuries (Koga et al., 2010; Cochrane et al., 2005). After undergoing our submaximal fatigue protocol, subjects displayed increased knee extension angles and moments (table 2, table 3). While only speculative without EMG data, increased extension angles and moments may be a result of increased quadriceps activity. Sharing a common insertion point on the tibia, increased activity in any of the quadriceps muscles could result in increased pull on the proximal tibia resulting in increased anterior tibial translation at the knee and greater strain on the ACL (DeMorat et al., 2004; Li et al., 1999; Renstrom et al., 1986).

As increased anterior drawer from the quadriceps can strain the ACL (DeMorat et al., 2004) factors that increase quadriceps dominance may indirectly increase strain on the ACL. Multiplanar factors are more likely to tear the ACL than sagittal factors alone (Cochrane et al., 2006; Koga et al., 2010; McLean et al., 2004). Koga et al. (2010) hypothesized that valgal loading coupled with the anterior drawer from quadriceps contraction causes the lateral femoral condyle to shift posteriorly resulting in anterior translation of the tibia. This is then followed by internal rotation of the tibia and consequent rupture of the ACL. Female athletes already utilize the quadriceps more often than males during athletic activity (Hanson et al., 2008; Huston et al., 1996; Hewett et al., 1996). Whether due to soleus fatigue or increased soleus stretch reflex activity, increased quadriceps anterior drawer can lead to greater ACL strain. The increased sagittal strain
coupled with maneuvers that place valgus shear stress on the ACL may increase an individual’s risk of rupturing the ACL, especially females.

Due to its high composition of fatigue-resistant muscle fibers, it is unlikely that soleus fatigue will occur during a normal course of an athletic competition. Interestingly, there was a wide range (52 to 282 repetitions) in the number of plantarflexion repetitions subjects were able to perform (table 1). Those subjects that performed more repetitions indicated a high involvement in endurance sports such as running. Therefore, athletes with insufficient endurance training may be at higher risk for soleus fatigue and consequently higher risk for ACL rupture.

Factors that impede the ability of the soleus to function properly in knee extension, such as injury or decreased functional capacity, may result in quadriceps compensation that could place the ACL at greater risk for rupture. Farrag et al. (2016) found that habitual high-heel use significantly decreased plantarflexion torque when the knee was in the soleus dominated flexed position. There was no difference found when the gastrocnemius dominated plantarflexion in the extended knee position. They concluded that habitual high-heel use diminishes that functionality of the soleus. Due to the spinal reflex pathway between the soleus and quadriceps, it is conceivable that increased quadriceps recruitment could result from decreased soleus contribution during knee extension. Further studies should investigate the role soleus strength has in the knee extension coupling phase of gait and other, high-intensity motions.
**Challenges and Limitations of Fatigue Protocol**

The soleus is composed of 70-90% slow-twitch muscle fibers with an approximate innervation ratio of 180 muscle fibers per motor neuron making it highly resistant to fatigue. In contrast, the gastrocnemius has a more even distribution of slow-twitch to fast-twitch muscle fibers with an innervation ratio of 1000-2000 muscle fibers per motor neuron (Edgerton et al., 1975; Johnson et al., 1973; Purves et al., 2012). The smaller fiber per motor neuron ratio allows the soleus to participate in more refined, sustained movements like maintenance of posture (Purves et al., 2012). The composition of the soleus makes it very difficult to fatigue which is evident through the relatively large number of repetitions (163 ± 78) performed by our subjects (table 1) during the submaximal fatigue protocol.

Our methods were unable to ascertain if the soleus muscle underwent isolated fatigue. Through its common distal insertion, the soleus is intimately linked to the gastrocnemius. Both of these muscles engage in ankle plantarflexion, however, the magnitude of influence the soleus has over plantarflexion can be adjusted through knee flexion (Arampatzis et al., 2006; Herbert-Losier et al., 2011; Price et al., 2003; Signorile et al., 2002). EMG techniques measure the electrical activity that initiates muscle activity. EMG studies have shown that with increasing knee flexion angle the soleus gains increasing influence over plantarflexion activity (Arampatzis et al., 2006; Herbert-Losier et al., 2011; Price et al., 2003; Signorile et al., 2002). However, EMG studies do not demonstrate selective isolation of the soleus with increasing knee flexion (Herbert-Losier et al., 2011; Price et al., 2003; Signorile et al., 2002).
Magnetic resonance imaging (MRI) measures the signal changes resulting from shifts in water brought about through metabolic changes during contraction (Fisher et al., 1991; Fleckenstein et al., 1989; Price et al., 1998). MRI measurements could, therefore, provide a greater indication of muscular work occurring during exercise than EMG. An MRI/EMG comparison study by Price et al. (2003) examined leg muscle activity during plantarflexion at various knee flexion angles. With the knee at 90° flexion, their results showed significantly increased T2 levels (relating to increased metabolic activity) in the soleus compared to the nearly nonexistent T2 levels in the medial and lateral gastrocnemius. Nonetheless, their results still showed EMG activity in the medial and lateral gastrocnemius at 90° knee flexion. The presence of electrical activity without T2 level change suggests the muscle underwent neurological recruitment but did not engage in metabolic activity great enough to induce T2 level change (Price et al., 2003). The authors explained that at 90° knee flexion perhaps the medial and lateral gastrocnemius were passively moved to a position on the length-tension curve that did not allow for cross-bridge cycling thus resulting in decreased metabolic output. They concluded that at 90° knee flexion the soleus must dominate plantarflexion force production.

While we cannot definitively say that our submaximal fatigue protocol fatigued the soleus in isolation, we are confident that our protocol preferentially targeted the soleus. EMG and MRI technique would have been able to confirm with greater confidence that our protocol fatigued the slow-twitch soleus muscle, however, those techniques were not available for this study. Our submaximal fatigue protocol required subjects to perform weighted plantarflexions with knees flexed until they could no longer perform the movement with good technique or tapped out due to self-perceived fatigue.
The point at which subjects determined they were fatigued may have been due to
sensation caused by extensive stretching (Alters, 1996). Either scenario likely involves
decreased the muscular performance of the soleus muscle and would therefore still
provide insight into the effects of decreased soleus performance or functional capacity.

**Leg Dominance During Sidestep Cutting Task**

After analysis of the pre- and post-fatigue conditions, results from both conditions
were pooled and differences between the dominant and nondominant legs were analyzed.
Sagittal plane knee kinematics and kinetics were relatively similar in both limbs. The
dominant limb experienced greater peak and average flexion moments, however,
increased flexion angles or moments have not been shown to increase ACL rupture risk
(Hewitt et al., 2005; Koga et al., 2010). There were significant differences in sagittal
plane ankle kinetics with the dominant limb experiencing a decreased peak dorsiflexion
moment and greater peak plantarflexion moment (table 3). Decreased dorsiflexion
flexibility has been shown to result in decreased knee flexion angles during a drop
vertical jump task (Malloy et al, 2015). Our results did not show a similar relationship.
The dominant ankle experienced minimally decreased dorsiflexion moment
(approximately 0.10) that resulted in increased knee flexion moment (table 3). A video
analysis by Boden et al. (2009) showed significantly less plantarflexion ankle angles at
initial contact were characteristic of ACL injuries. Our results did not show any
significant difference in ankle kinematics (table 2). While sagittal plane kinematics and
kinetics can increase the risk of ACL rupture, sagittal factors alone are not likely to tear
the ACL (DeMorat et al., 2004; Kim et al., 2001; Koga et al., 2010; McLean et al., 2004).
Frontal and transverse plane kinematics and kinetics have been shown to be greater predictors for noncontact ACL injuries than sagittal plane factors alone (Cochrane et al., 2005; Hewett et al., 2005; Koga et al., 2010). The results of Hewitt et al. (2005) indicated that female athletes who employ landing strategies involving increased abduction angles and moments are at significantly greater risk for ACL rupture. This is consistent with other studies describing valgal loading as an integral component in the mechanism producing noncontact ACL tears (Cochrane et al., 2005; Hewett et al., 2005; Koga et al., 2010). The greater abduction moment in the dominant limb seen in our results suggests the dominant limb is at greater risk for ACL rupture.

There is conflicting data on the influence of leg dominance on noncontact ACL rupture risk. Greska et al. (2017) found no difference between legs when female athletes performed sidestep cutting tasks. Their results are contrary to those of Brophy et al. (2010) who performed a retrospective analysis and found that female soccer athletes were more likely to injure their nondominant leg than their dominant leg. Our results showing significantly increased peak abduction moments (table 3) in the dominant limb. Ford et al. (2003) that found increased peak abduction angles in female basketball athletes performing drop vertical jump tasks. Furthermore, Ford et al. (2003) found the dominant limb to be significantly more abducted. Additionally, our common language effect size analysis (table 8) showed an 81% chance that the dominant knee would experience greater abduction moments than the nondominant limb during the sidestep cutting task.

Contrary to previous studies, our results support the hypothesis of asymmetries during sidestep cutting maneuvers. The results of our study suggest the dominant limb is at greater risk due to its tendency to undergo greater abduction angles and moments.
(Hewitt et al., 2005; Ford et al., 2003). Differences in findings are most likely due to
differences in data gathering techniques. Greska et al. (2017) had subjects undergo
unanticipated 45° sidestep cuts whereas our study had subjects perform anticipated 45°
sidestep cuts. Moreover, Brophy et al. (2010) retrospectively analyzed injuries that had
occurred during gameplay. It is seemingly apparent that there are many factors that
influence dominant versus nondominant knee kinematics and kinetics. These include but
are not limited to athletic task, cognitive recognition, and environment. Recognition of
limb asymmetries during athletic tasks can help in the development of prevention
measures, further studies should continue to analyze limb asymmetries during exercise.
While differences in the degree of limb asymmetries between male and females have
been found (Brophy et al., 2010; Ford et al., 2003), it is especially important for female
population studies to include analysis of limb asymmetries.

**Kinematics and Kinetics within First 41.67ms**

Model-based image-matching analysis of noncontact ACL injuries suggests that
the during injury the ACL ruptures within the first 40ms of the athletic maneuver (Koga
et al., 2010). Our study found no significant differences in the sagittal plane motion
between the pre- and post-soleus fatigue conditions within the first 41.67ms after initial
contact. Average dorsiflexion moment decreased after fatigue and although not
significant it began to approach significance (P = 0.059). While not significant, knee
extension moment within the first 41.67ms was increased after soleus fatigue (table 5).
This is consistent with the plantarflexion-knee extension mechanism (Brunner et al.,
2013). Again, while not significant (P = 0.431) knee extension angle was increased
(32.09° pre-fatigue, 28.98° post-fatigue) within the first 41.67ms after soleus fatigue and
fell into the flexion angle range (11° to 30°) seen with ACL injuries in the study by Koga et al. (2010). As described above, increased extension angle and moments can place the knee in positions that compromise ACL stability (DeMorat et al., 2004; Kim et al., 2001; Koga et al., 2010). Increased extension angles and moments may be a result of increased quadriceps activation which can increase anterior drawer of the proximal tibia. Decreased dorsiflexion moment may be a result of unintended fatigue of ankle dorsiflexors however, the increased knee extension angles and moments seen in our study makes it more likely that there were increased soleus and quadriceps activation through plantarflexion-knee extension coupling pathway. Additionally, decreased dorsiflexion moment within the first 41.67ms provides light evidence that soleus fatigue (defined as a decreased ability to produce force) did not occur after our submaximal fatigue protocol. If soleus fatigue did occur, it resulted in increased activation of plantarflexors and/or quadriceps most likely via the spinal pathway seen with abnormal plantarflexion-knee extension coactivation (Dyer et al., 2011). Further studies should include EMG data from the soleus, gastrocnemius, and quadriceps to investigate the pattern of activation in response to increased plantarflexor use and/or fatigue.

Leg dominance had significant effects on knee flexion angle and knee abduction moment. The dominant limb was slightly more extended (approximately 2°) in the dominant limb. Ankle dorsiflexion moment was decreased within the first 41.67ms in the dominant limb however, this did not result in a greater knee extension moment. The lack of increased knee extension moment with increased plantarflexion moment provides further light-evidence that the increased knee extension seen in the pre- and post-fatigue conditions were a result of increased soleus-quadriceps coactivation.
Consistent with the results of Koga et al. (2010) knee abduction was significantly increased within the first 41.67ms in the dominant limb. Effect analysis showed a large effect of leg dominance on abduction moment within the first 41.67ms. As greater abduction moments have been shown to increase the risk of ACL rupture (Hewett et al., 2005; Koga et al., 2010), these results suggest that the dominant limb is at greater risk for ACL rupture. Future studies should continue to investigate the effect of leg dominance on knee kinematics and kinetics.

**Study Limitations**

There are several limitations that should be considered when interpreting the results of this study. The foremost consideration is the submaximal soleus fatigue protocol employed. For reasons outlined above (page 59) isolated soleus muscle fatigue is difficult to achieve. Based on previous studies we are confident that we preferentially targeted the soleus, however, it is entirely possible that our protocol did not fatigue the soleus but thoroughly stretched it. Additionally, our fatigue protocol relied on subject-reported, subjective perception of fatigue. It is likely that there was variation in perceived levels of fatigue between subjects. EMG data was not available to be used in this study but would help provide a more objective measurement of fatigue. Furthermore, it is possible the effects of our submaximal fatigue may not have endured throughout the entirety of the sidestep cutting task. We attempted to reduce these effects by limiting the time between submaximal fatigue protocol and data collection.

The exercise history for our sample population included subjects with variable exercise history. Neuromuscular control can vary between high-level athletes and the general population (Viitasalo et al., 1998). We attempted to recruit a subject population
representative of recreational and high-level athletes by including subjects with and without athletic experience beyond recreation. Pre- and post-fatigue data were collected during different appointments, it is possible that there were differences in marker placement between visits. To diminish the likelihood of marker placement differences, the same investigator positioned markers for each visit. It is highly unlikely that systematic differences in marker placement between visits explain the results of the study.

**Future Studies**

The results of our study provide evidence to warrant further investigation of the triceps surae muscles on knee biomechanics. To effectively rehabilitate the ACL and limit chances of re-injury, the many variables that impact ACL rupture should be recognized and properly weighed. The complex interactions of the gastrocnemius and soleus and their influence on posture, balance, and joint loading should be further investigated. Our study shows a clear effect of soleus targeted exercise on knee dynamics. Further studies should treat the gastrocnemius and soleus as two different muscle with possible contrasting influences during exercise. Through neural connections to the quadriceps muscle group, the soleus may impact the stability of the ACL in dynamic exercise. Future studies should assess muscle activation patterns and activation levels during exercise when the soleus is in a diminished state. Greater understanding of the neural pathway between the soleus and quadriceps muscle could help lead to better rehabilitation measures for those suffering from neurological disorders that affect limb musculature.
Conclusions

The results of our study do not support our first hypothesis that soleus fatigue increased flexion angle during sidestep cutting exercise. Our second hypothesis that soleus fatigue increases movement of the tibia during sidestep cutting tasks, as well as, our third hypothesis that limb asymmetries are present during sidestep cutting tasks was supported by our results. Based on our results the following conclusions can be made:

- **An isotonic submaximal fatigue protocol targeting the soleus results in greater knee extension angles and moments during sidestep cutting tasks.**
  While speculative at this point, this may be due to a synergistic relationship between the soleus and quadriceps muscles.

- **An isotonic submaximal fatigue protocol targeting the soleus increases movement of the tibia at maximum knee flexion during sidestep cutting task.**
  This is the angles at which quadriceps activity is greatest (as it initiates a change in direction of the limb). Therefore, the increased anterior translation of the proximal tibia may be due to increased quadriceps activity.

- **Isotonic submaximal fatigue protocol targeting the soleus has no significant effect on frontal plane kinematics and kinetics during sidestep cutting tasks.**
  The soleus attaches to and pulls posteriorly on the proximal tibia while in a closed kinetic chain. Our study provided no evidence that the soleus has a medial or lateral pull on the proximal tibia.

- **A relatively high-degree of resisted plantarflexion repetitions is required to reach levels of perceived soleus muscle fatigue.** This is evidenced by the large number of average repetitions (163.4 ± 78.4) performed by our subjects.
• An isotonic submaximal fatigue protocol targeting the soleus does not significantly alter ankle kinematics and kinetics during sidestep cutting tasks. In a closed kinetic chain, concentric contraction of the soleus results in a posterior pulling on the proximal tibia. A similar response is seen during the maintenance of posture while standing.

• The dominant limb undergoes greater abduction moments than the nondominant limb during sidestep cutting tasks. There is conflicting literature regarding limb asymmetries during sidestep cutting tasks. Our study shows with large degree of confidence that the dominant limb experiences increased knee abduction moments. The reasoning for this phenomenon is currently not known.
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