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Effects of Increased Shoe Collar Height and Limb Dominance on Landing Knee Biomechanics in Collegiate Volleyball Players

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EFFECTS OF INCREASED SHOE COLLAR HEIGHT AND LIMB DOMINANCE ON LANDING KNEE BIOMECHANICS IN COLLEGIATE VOLLEYBALL PLAYERS

by

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ABSTRACT

Athletic footwear with higher collar heights are worn to restrict ankle motion. Reduced ankle dorsiflexion has been associated with increased frontal plane knee motion. Volleyball players wear mid-cut shoes (MC) that have an increased collar height rising slightly superior to the talocrural joint and malleoli. The purpose of this study was to determine the effect of MC and limb dominance on knee landing mechanics. It was hypothesized that participants would land with greater initial contact (IC) and peak frontal joint plane angles and moments and smaller IC and peak sagittal plane joint angles and moments at the knee in MC and dominant limb compared to low-top shoes (LT) and the non-dominant limb. We also hypothesized that vertical and posterior directed ground reaction forces (GRF) would be greater while wearing MC, yet the same between limbs. Seventeen female collegiate volleyball players performed unilateral forward drop landings on each limb in MC and LT styles of a volleyball shoe (Crazyflight, Adidas AG, Herzogenaurach, Germany). A two-way repeated measures ANOVA revealed significant interactions for internal peak knee flexion moment and peak lateral GRF. There was a shoe effect for peak dorsiflexion angle, peak plantarflexion moment, and peak medial GRF. A limb effect was found for peak knee abduction moment. The MC shoes did not increase knee frontal plane loading, but greater knee abduction moments were observed in the dominant limb. This suggests improving landing mechanics on the dominant limb may reduce knee injury risk while playing volleyball, regardless of LT or MC.

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DEDICATION

To my dad, Jason. I do not think I would be done with this without you letting me sit at one end of the table while you sat the other writing your reports and for always only being a phone call away if I ever needed to vent. Thank you for always pushing me to be better and do better as both a volleyball player and a person. Sorry for the gray hairs.

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LIST OF ABBREVIATIONS

ACL	Anterior Cruciate Ligament
ANOVA	Analysis of Variance
GRF	Ground Reaction Force
DVJ	Drop Vertical Jump
LT	Low-top
MC	Mid-cut
3D	Three-dimensional
N	Newtons
J	Joules
m	meters
Nm	Newton-meters
kg	Kilograms
ms	milliseconds
NCAA	National Collegiate Athletic Association
BW	Bodyweight
BM	Body Mass
NDL	Non-Dominant Low-Top
DL	Dominant Low-Top
NDM	Non-Dominant Mid-Cut
DM	Dominant Mid-Cut

CHAPTER I - INTRODUCTION

Background

In the United States, volleyball has the third largest participation rate in high-school female sports and has reached a record high number of athletes playing at the collegiate level (Chandran et al., 2021; Tillman et al., 2004a, 2004b). Volleyball has been characterized as an explosive jumping sport that relies on the speed and power of the lower extremity (Bisseling et al., 2008). There are a variety of tasks in every position that requires the ability to jump and land both bilaterally and unilaterally (Lobietti et al., 2010; Tillman et al., 2004a, 2004b). Previous research has found that during one five-set match, one volleyball player takes approximately 96 jumps and depending on the position, volleyball players may take between 32.3 to 76.8 jumps per hour or 300 to 450 jumps per training session (Charlton et al., 2017; Lima et al., 2019; Lobietti et al., 2010). In an analysis of volleyball players playing in two matches, it was observed that over 40% of landings were performed on a single leg (Tillman et al., 2004b). These landings place the lower extremity in an asymmetric position which may negatively impact the landing mechanics between dominant and non-dominant legs (Kalata et al., 2020).

Sixty-three percent of musculoskeletal injuries sustained in volleyball occur during the jumping and landing tasks of spiking and blocking, and over half of the injuries reported are non-contact or overuse injuries (Chandran et al., 2021; Xu et al., 2020). In female's collegiate volleyball, knee injuries account for the largest proportion of injuries reported, followed by ankle injuries (Chandran et al., 2021; Ferretti et al., 1992). The most common knee injury reported was a rupture to the anterior cruciate ligament (ACL) while lateral ankle sprains were the most common ankle injury

(Chandran et al., 2021; Ferretti et al., 1992). Volleyball players often wear ankle bracing, taping, and high-top shoes to restrict ankle motion to prevent ankle sprains (Frey et al., 2010; Janssen et al., 2017; Pedowitz et al., 2008). Although these devices may reduce the incidence of ankle sprains, research has established that the foot and knee are linked during landing and the manipulation of ankle mechanics changes the knee biomechanics (Frey et al., 2010; Pedowitz et al., 2008).

Limiting ankle range of motion in both the sagittal and frontal planes induces changes to knee biomechanics during landing tasks (Hagins et al., 2007; Ihmels et al., 2020; Mason-Mackay et al., 2016; Mündermann et al., 2003; Valenzuela et al., 2016; Vanwanseele et al., 2014; Venesky et al., 2006). Some studies have found that restricting ankle frontal plane range of motion increased the internal rotation moments at the knee (Mündermann et al., 2003; Valenzuela et al., 2016; Venesky et al., 2006). Other research has found that individuals with limited sagittal plane range of motion at the ankle have demonstrated greater frontal plane knee excursion, knee abduction angles, knee abduction moments, vertical and posterior ground reaction forces (GRF), knee anterior shear force, and reduced knee flexion angles (Fong et al., 2011; Hoch et al., 2015; Schroeder et al., 2021). While wearing ankle prophylactics that reduce ankle range of motion, changes to the knee biomechanics have been observed such as increased knee external rotation moment, decreased knee flexion angles, greater knee abduction angles, and greater knee internal rotation moments (Hagins et al., 2007; Ihmels et al., 2020; Malliaras et al., 2006; Mason-Mackay et al., 2016; Vanwanseele et al., 2014; Venesky et al., 2006).

Wearing shoes with an increased collar height, known as mid-cut (MC) or high-top shoes, is another strategy of reducing ankle sprains. Vanwanseele et al. (2014)

compared high-top shoes to a standard low-cut netball shoe during single leg landing in netball players (Vanwanseele et al., 2014). They reported that the high-top shoe did not restrict any motion at the ankle but did promote an increase in the internal plantar flexion moment and internal peak knee internal rotation moment (Vanwanseele et al., 2014). In contrast, other research has found that high-top shoes limit ankle frontal and sagittal plane motion during cutting and landing tasks (Brizuela et al., 1997; Daack and Senchina, 2014; Lam et al., 2015; Li et al., 2013; Tang et al., 2020; Yang et al., 2017). Reductions in total ankle range of motion in the sagittal and frontal planes have also been demonstrated while wearing a high-top shoe in both football and basketball (Daack and Senchina, 2014; Lam et al., 2015; Li et al., 2013; Tang et al., 2020). In the studies previously mentioned, it has been demonstrated that the basketball and football high-tops alter ankle mechanics. However, these studies did not report knee mechanics so the extent of how a high-top shoe impacts knee joint mechanics remains unclear.

With athletes exhibiting reduced ranges of motion at the ankle while wearing ankle prophylactics and reductions of ankle motion being associated to ACL injury mechanisms, there is a potential than an athlete may increase their risk of ACL injury by wearing a device designed to protect themselves from an ankle injury. The ACL functions to resist anterior tibial translation on the femur and assists in limiting knee internal rotation (Butler et al., 1980; Duthon et al., 2006; Markolf et al., 1976; Takeda et al., 1994). Medial knee collapse, caused by small knee flexion, greater knee abduction, and tibial rotation is also a common mechanism of ACL injuries. Video analyses have found that a majority of ACL injuries occur within the first 100 milliseconds (ms) of initial contact with the ground when landing or cutting (Koga et al., 2010; Krosshaug et al., 2007). Female athletes sustain non-contact ACL injuries at a rate two-to-eightfold greater than males (Arendt and Dick, 1995; Boden et al., 2010). Proposed explanations for the increased risk to these athletes are females have exhibited larger knee abduction moments, larger knee abduction angles, and smaller knee flexion angles when landing accompanied by intrinsic factors of a smaller ACL and a general joint laxity (Beynnon and Shultz, 2008; Chandrashekar et al., 2006; Ruedl et al., 2012).

Landing tasks have been implemented as a screening tool to identify athletes at risk of injury. The implementation of the drop vertical jump (DVJ) to be utilized as an ACL screening tool has been called into question with contradictory results between studies. Hewett et al. (2005) has associated the DVJ to ACL injury risk with increased knee abduction moments and ground reaction forces. The DVJ is limited by its bilateral nature when in sports, ACL injuries are most likely to occur in more dynamic tasks and in unilateral landings (Cruz et al., 2013; Peebles et al., 2020). Contrary to Hewett et al. (2005), Kristianslund and Krosshaug (2013) found no relationship between knee abduction moments and the DVJ. A disadvantage of the DVJ is that the task requires minimal horizontal force. In analyzing videos of ACL injuries, athletes rupture their ACL when they have both horizontal and vertical momentum while landing or cutting (Boden et al., 2000; Peebles et al., 2020). Therefore, a landing task that incorporates both horizontal and vertical components of momentum has been utilized to better represent the dynamic tasks of sports and improve the assessment of ACL injury risk (Decker et al., 2003; Fong et al., 2011; Padua et al., 2015; Schroeder et al., 2021). When comparing a stop jump to a DVJ, Peebles et al. (2020) found larger internal knee adduction moments during landing when both horizontal and vertical motion was accounted for in the task. A similar task, the forward vertical jump, has also been validated in combination with the Landing Error Scoring System to determine ACL injury risk in athletes (Padua et al., 2015). In single-leg landing tasks, Munro et al. (2017) reported moderate correlations ($r = 0.52$) of task to knee abduction angles and moments. This suggests that unilateral tasks may be a more useful screening tool for individuals participating in sports.

Side-to-side knee abduction moment asymmetry is a risk factor for ACL injuries in female athletes which makes the analysis of limb differences between unilateral landings import to assess (Hewett et al., 2005). Females have demonstrated significant asymmetries between their dominant and non-dominant limbs in peak knee abduction angles, GRF, and hip adduction angles (Edwards et al., 2012; Ford et al., 2003; Pappas and Carpes, 2012; Wang and Fu, 2019). In volleyball specific research, Tillman, Criss, et al. (2004) found greater muscular activation, a higher peak vertical and lateral GRF, and rate of loading when volleyball players landed on a single limb compared to landing on both limbs. Sinsurin et al. (2017) observed that volleyball players landed in an externally rotated knee position on the nondominant limb but in an internally rotated position on the dominant limb during a single leg landing task. At peak vertical GRF, greater knee internal rotation, less internal knee adduction moment, and a valgus alignment were observed when landing on the dominant limb compared to the nondominant limb (Sinsurin et al., 2017). In terms of kinetic parameters, Sinsurin et al. (2017) reported no significant differences in peak vertical GRF between limbs. The lack of GRF differences between limbs is in support of previous literature. In assessing the landing biomechanics, Ford et al., (2003), Van der Harst et al. (2007) and Wang and Fu (2019) observed no significant differences between peak GRF between limbs in a DVJ, unilateral landing, and single-leg distance hop respectively.

Footwear companies develop different high-top shoes designed specifically for various sports (Nigg et al., 1999). Minimal research has investigated the effects of a high-top shoe at the knee in basketball, football, and netball. In volleyball, a popular high-top shoe is a MC volleyball shoe where the shoe collar stops level at the malleoli (Cronin, 2001). One study that measured the ankle range of motion using a goniometer in volleyball shoes found that the ASICS Gel Airier MC volleyball shoe had a smaller dorsiflexion range of motion compared to the ASICS low-cut volleyball shoe (Cronin, 2001). This study addressed the impact MC volleyball shoes had on ankle range of motion, but it did not address the knee biomechanics. Reductions in dorsiflexion have been associated to greater frontal plane kinematics and kinetics at the knee (Malloy et al., 2015; Mauntel et al., 2013). Therefore, it is important understand the effect of MC shoes worn by volleyball players on biomechanical risk factors for ACL injury.

Purpose of the Study

The purpose of this study was to determine the effects of MC shoes on knee joint biomechanics during a unilateral drop landing task in female collegiate volleyball players. In addition, we aimed to determine the influence of leg dominance on knee joint biomechanics during landing while wearing MC shoes. Primary variables of interest include initial contact and peak joint angles and moments in the sagittal and frontal planes at the knee as well as peak GRF. Secondary variables of interest include the initial contact and peak joint angles and moments in the sagittal and frontal planes of the ankle joint.

Research Hypotheses

We hypothesized that an interaction would be present between shoe and limb conditions. Specifically, we hypothesized that the dominant limb in the MC shoe would have larger frontal plane knee joint angles and moments and smaller sagittal plane knee joint angles and moments than the non-dominant limb in the LT shoes. It was hypothesized that participants would exhibit smaller initial contact and peak knee flexion angles and peak knee joint moments in the sagittal plane yet greater initial contact and peak knee abduction angles and peak knee joint moments in the frontal plane while wearing the MC shoes compared to LT shoes. We also hypothesized that vertical and posterior directed ground reaction forces would be greater while wearing MC shoes. Further, it was hypothesized that smaller joint angles and moments in the sagittal plane and greater joint angles and moments in the frontal plane of the knee when landing would be observed in the dominant limb compared to the non-dominant limb but with the ground reaction forces remaining the same between limbs.

CHAPTER II – LITERATURE REVIEW

Introduction

The anterior cruciate ligament (ACL) is one of the most studied structures in the musculoskeletal system (Dienst et al., 2002). The ACL is one of the major stabilizers of knee motion which functions to resist anterior tibial translation on the femur and assist in limiting tibial internal rotation (Dienst et al., 2002; Hollis et al., 1991).

ACL injuries are the most common sports-related injury (Beynnon et al., 2014; Slater et al., 2019). Surgery and rehabilitation are predicted to cost $17,000 per ACL injury and cost the United States roughly $1 billion annually (DuPrey et al., 2016; Hewett et al., 2005). An ACL injury is a serious sports injury because it often requires ligament reconstruction with a rehabilitation process that spans from 6 months to a year (Nagano et al., 2009).

ACL injuries are commonly observed in sports like basketball, volleyball, soccer, handball, and netball. In volleyball specifically, knee injuries account for the most reported, with ACL injuries accounting for a majority of the injures (Chandran et al., 2021; Ferretti et al., 1992). ACL injuries have been reported to occur within the first 100 ms of movement during landing or cutting tasks. During landing and cutting tasks, the common mechanisms for ACL injuries includes a medial knee collapse with internal or external tibial rotation when the knee is close to extension (Hewett et al., 2005; Ireland, 2002, 1999; Koga et al., 2010; Krosshaug et al., 2007). Female athletes have a 2-8 x greater risk of sustaining an ACL injury compared to males (Arendt and Dick, 1995; Boden et al., 2010). A larger Q-angle, smaller intercondylar notch, joint laxity, and hormonal factors are some proposed explanations for the increased risk factors associated to females (Beynnon and Shultz, 2008; Ruedl et al., 2011). Compared to males, females generally have less knee flexion, increased abduction angles, and greater vertical and posterior ground reaction forces (GRF) compared to males.

Recent literature has discussed the role of the foot placement and dorsiflexion range of motion in implications for ACL risks. However, the normal relationship between the foot and the knee in sports is altered when the athlete wears ankle prophylactics. Ankle prophylactics are commonly worn to reduce motion at the ankle and prevent ankle sprains, but the effects of wearing ankle prophylactics at the other joints of the lower extremity are less understood (Venesky et al., 2006). Research has presented conflicting evidence on the effectiveness of ankle prophylactics on preventing ankle sprains, but it is well established that prophylactics like taping, bracing, and high-top shoes alters ankle mechanics.

During weightbearing movement, the foot and knee motions are linked as in the motion and force at the ankle influence the motion and force at the knee and vice versa (Venesky et al., 2006). Changing ankle mechanics affect the biomechanics of the knee. Manipulation of the ankle can induce changes to the knee that pose greater risks of sustaining an ACL injury. Limited research has presented the impact of high-top shoes on the knee mechanics in sports (Vanwanseele et al., 2014). With the large number of athletes that wear high-top shoes, it is important to assess the impact of wearing high-top shoes on the ankle and knee biomechanics, especially in a population that is already susceptible to ACL injuries and that plays a sport characterized by jumping and pivoting.

The purpose of this literature review will be to discuss the anatomy and function of the ACL, the ACL risk factors and injury mechanisms, the impact of ankle prophylactics, the relationship between the biomechanics of the ankle and the knee, the effect of ankle prophylactics on the ankle and knee biomechanics, and the current literature assessing the impact of high-top shoes specifically.

Overview of the ACL

Anatomy of the ACL

The ACL is a thick band of connective tissue composed of multiple collagen-based fascicles organized in a helical wave pattern (Dienst et al., 2002; Duthon et al., 2006). The matrix of the ACL is made up of collagen, glycosaminoglicans, glycol-conjugates, and elastic components (Duthon et al., 2006). The ACL is located within the synovial membrane of the tibiofemoral joint with its ligaments intra-articularly but remaining extra-synovially throughout its course (Dienst et al., 2002; Duthon et al., 2006). The irregular shaped ACL spans from the femur to the tibia with a mean length of 32 mm and a mean width of 11 mm (Amis and Dawkins, 1991; Dienst et al., 2002; Duthon et al., 2006). The femoral attachment of the ACL is on the posterior part of the lateral femoral condyle (Dienst et al., 2002). The femoral origin occupies on average 66% of the superior aspect of the intercondylar notch and ranges 11 to 24 mm in diameter and is titled 26° ± 6° forward from the vertical (Dienst et al., 2002; Duthon et al., 2006). At the femoral origin, the ACL is the narrowest and fans out to the tibial attachment. The tibial attachment of the ACL is located anteriorly and laterally to the medial intercondylar tubercle (Dienst et al., 2002; Markolf et al., 1976; Meyer and Haut, 2005; Zavatsky and O'Connor, 1992). The tibial attachment site has a mean width of 11 mm and a mean length in the anterior-posterior direction of 17 mm. Previous research has generally accepted the division of the ACL into two bundles: an anteromedial bundle and a posterolateral bundle (Duthon et al., 2006). The anteromedial bundle is the smaller of the two bundles that originates on the posterior and proximal aspect of the femoral attachment and inserts on to the anteromedial aspect of the tibial attachment. The larger posterolateral bundle originates at the distal aspect of the of the femoral attachment and insert at the posterolateral aspect of the tibial attachment (Colombet et al., 2006; Dienst et al., 2002).

The human knee has been geometrically modelled as a four-bar linkage to control the movement of the femur relative to the tibia (Zavatsky and O'Connor, 1992). This model predicts the rolling and sliding of the joint and can be utilized to calculate ligament shaped and fiber lengths. This kinematic model allows us to define the geometry of the knee in the sagittal plane in a passive manner to understand the function of ligaments in the knee without the presence of an external load (Zavatsky and O'Connor, 1992). However, due to the complexity of the knee, the kinematic linkage of the tibia with respect to the femur has been implemented to evaluate the 6 degrees of freedom in anatomical motion (Hollis et al., 1991). The anatomy of the ACL is important because its structure determines its function and influences its susceptibility to injury (Duthon et al., 2006). The wave-like matrix of the ACL fascicles serves the purposes previously termed as "crimp" and "recruitment" (Duthon et al., 2006). Crimp refers to the sinusoidal pattern in the matrix while recruitment refers to the increase tissue stiffness and an increase in fibrils becoming load bearing. Crimp and recruitment are mechanisms to control tension, absorb shock, and support large loads to allow the ACL to optimally function and provide stability to protect the tibiofemoral joint (Duthon et al., 2006).

Function of the ACL

In determining ACL function, past research that utilized unloaded cadaver studies have broken down the different functions of the anteromedial bundle and posterolateral bundle (Dienst et al., 2002; Kawaguchi et al., 2015; Li et al., 2004; Takeda et al., 1994). With passive flexion, the anteromedial bundle tightens while the posterolateral bundle is relaxed. The opposite of this is true when the knee is extended (Dienst et al., 2002; Li et al., 2004). Hollis et al. (1991) found that the anteromedial bundle length increased from 34.4 ± 1.0 mm to 38.0 ± 0.8 mm and the posterolateral bundle length decreased from 22.5 ± 1.2 mm to 15.4 ± 1.2 mm when the knee was passively flexed from 0 to 90°. Smaller changes in length occurred in the anteromedial bundle than the posterolateral bundle, this finding concurs with the in vivo study by Li et al. (2004). In a quasi-static lunge, the anteromedial bundle did not significantly change in length from full extension to flexion while the length of the posterolateral bundle significantly decreased past 60° of flexion (Li et al., 2004). Increasing the range of motion at the knee, Amis and Dawkins (1991) found that all the fiber bundles of the ACL continued to increase in length during the final 30° of extension. In the transverse plane, internal rotation lengthened the fibers more than external rotation (Amis and Dawkins, 1991). These studies demonstrated the increased role of the posterolateral bundle at low flexion angles and full extension. It is important to note that this research was conducted with passive tension in unloaded conditions.

Due to the weight-bearing activities of daily life, and especially in a sport setting that requires an increased weight bearing load, it is important to understand the overall function of the ACL. The primary function of the ACL is to resist anterior tibial translation (Butler et al., 1980; Dienst et al., 2002; Duthon et al., 2006; Markolf et al., 1976). Butler et al. (1980) analyzed the ACL in its entirety instead of divided into bundles in a controlled, cadaver study without tibial rotation. Butler et al. (1980) found that cutting the ACL greatly reduced the restraining force during an anterior drawer test. The ACL accounted for 85.1 ± 1.9% of the total restraining force at 90° of knee flexion and 87.2 ± 1.6 % at 30° of flexion (Butler et al., 1980).

The ACL has a secondary function of resisting tibial internal rotation (Dienst et al., 2002; Duthon et al., 2006; Markolf et al., 1976; Takeda et al., 1994). In assessing the roles of the cruciate and collateral ligaments of the knee, the removal of the ACL resulted in a decrease in stiffness while laxity increased with the greatest increase occurring near full extension for anterior stability (Markolf et al., 1976). When the medial collateral ligament and the ACL were sectioned together, the torsional stability changed with an increase in laxity with the largest increases observed at near full extension (Markolf et al., 1976). This research highlights that the ACL plays a role in both anterior and torsional stability of the knee as well as provides evidence that ACL's have an increased risk of injury in positions near or at full extension compared to 90° of flexion (Lipke et al., 1981; Markolf et al., 1976).

Loading of the ACL

ACL injuries occur when there are high levels of stress, high rates of strain, or a combination of both (Nordin, 2020). To understand the loads the ACL endures, femur-ACL-tibia complex (FATC) is often subjected to tensile testing (Takeda et al., 1994). Stress is the force per unit area and strain is the ratio of change in length versus the original length (Takeda et al., 1994). In a cadaver study, Meyer and Haut (2005) found that at flexion angles of 60°, 90°, and 120° ruptured the ACL at a combined peak load of 5.1 ± 2.1 kN. At 90° of flexion, the most force was required to rupture the knee at 6.0 ± 3.2 kN compared to 4.9 ± 1.5 at 60° and 4.4 ±1.0 kN at 120° (Meyer and Haut, 2005). In flexion angles less than 90°, less force was necessary to rupture the ACL, therefore making the ACL more susceptible to injury. Similar loading patterns were observed by Beynnon et al. (1992). After the implantation of the Hall Effect Strain Transducer (HEST), the shear loading of knee flexion was greater at 30° compared to 90° of flexion which was demonstrated with a 3.7% strain value of the anteromedial bundle at 30° and a 1.8% strain value at 90° (Beynnon et al., 1992). Beynnon et al. (1992) concluded that the ACL is strained between full extension and 48° during an active range of motion.

The loading of the ACL is influenced by age, maturation, knee flexion angles, and the direction of the applied load (Dienst et al., 2002). When investigating the impact of age, it has been established that younger ACLs can withstand greater amounts of force compared to older knees (Noyes and Grood, 1976; Woo et al., 1991). Woo et al. (1991) demonstrated the influence of age on the load ability of the ACL. Woo et al. (1991) found that the tensile load and linear stiffness of the ACL in both the anatomical orientation and tibial orientation was greater in the younger groups compared to the older groups. In anatomical orientation, the cadaveric knees in the range of 22 to 35 years old, had a linear stiffness of 242 ±16 N/mm and an ultimate load of 2160 ± 157 N. In the age range of 40 to 50, the linear stiffness was 220 ± 24 N/m and the ultimate load was 1503 ± 83 N. The older knees of 60- to 97-year-olds had a stiffness of 180 ± 25 N/mm and an ultimate load of 658 ± 129 (Woo et al., 1991). Through their research, Woo and colleagues (1991) established an ACL injury threshold of 2000 N. Noyes and Grood (1976) found younger individuals ruptured their ACL at a maximum force 2.4 times greater than the older group that ranged from 48 to 86 years old.

In addition to age, sex-based differences alter the load which an ACL can withstand. Chandrashekar et al. (2006) found a 14.49% difference in stress at failure and a 22.49% difference in the modulus of elasticity between females and males. Females demonstrated less resistance during the straining on the ACL and failed at lower rates which potentially explains why females sustain more ACL injuries than their male counterparts. Additionally, the length and size of the ACL was smaller and shorter in the females compared to males (Chandrashekar et al., 2006).

The application of forces on the ACL from multiple planes increases the loading of the ACL (Fukuda et al., 2003; Markolf et al., 1995; Quatman et al., 2014; Shin et al., 2011). Previous cadaver research has shown that the combination of tibial internal rotation moment and valgus moment increase the forces in the ACL (Fukuda et al., 2003; Markolf et al., 1976). Cadaver studies identified types of loading patterns that generate high levels of ACL force. Markolf et al. (1995) recorded the ACL ligament force and angle of knee flexion as the knee moved from a flexed 90° position to hyperextension under different loading conditions. The loading conditions were anterior tibial force plus internal or external tibial torque, anterior tibial force plus varus or valgus moment, internal tibial torque plus varus or valgus moment, and external tibial torque plus varus or valgus moment. At all flexion angles, the application of anterior tibial force increased the ACL load. The application of internal torque increased the ACL load at all flexion angles and with the addition of valgus moment, the load was increased from 10° to 70° of knee flexion. The highest levels of ACL load were observed with a combination of anterior tibial force and internal tibial torque near extension at angles of flexion less than 20° and a combination of anterior tibial force and valgus moment greater than 10° of flexion (Markolf et al., 1995). The authors conclude that controlling the anterior tibial displacement and limiting the varus-valgus angulation could prevent the increased ACL forces (Markolf et al., 1995).

Expanding the cadaveric research, simulation studies found that larger ACL strains and injuries were induced by multiplanar loading (Quatman et al., 2014; Shin et al., 2011). Quatman et al. (2014) assessed the ACL strain after a jump using different simulated landings that manipulated the magnitude of anterior tibial shear force, knee abduction, and internal tibial rotation, at 25° of flexion. Consistent with previous research, the ACL strain was significantly greater in a multiplanar loading situation with anterior tibial shear force, knee abduction, and internal tibial rotation rather than in these conditions alone (Quatman et al., 2014). Using *in vivo* human loading data, Shin et al. (2011) found that the peak strain to the ACL was affected by the application of a combination of valgus moment and internal tibial rotation moment during a single-leg landing task. The combination of valgus moment and internal rotation moment increased the strain of the ACL to greater than 0.077. At maximum valgus moment and maximum internal tibial rotation moment, the peak strain was 0.105. Noting that ACL ruptures occur in strain ranges from 0.09 to 0.15, the loads observed in the simulated single-leg landing task were large enough to potentially rupture the ACL (Shin et al., 2011).

Muscles that cross the knee play a large role impacting the strains and forces that the ACL endures (Takeda et al., 1994). *In vitro* and *in vivo* studies have measured ACL strain during quadricep and hamstring contractions. In a cadaver study, More et al. (1993)

confirmed the role of the hamstrings to provide stability to the knee. The hamstrings play a role in the unloading of the ACL during knee flexion-extension. The knee kinematics changed when adding a hamstring load that simulated a squat exercise. More et al. (1993) recorded a reduction in the amount of anterior tibial translation force and internal tibial rotation during knee flexion when the hamstring load was added (More et al., 1993). Lloyd and Buchanan (2001) concluded that the general goal of neuromuscular control is to increase the varus-valgus support as the knee extends. The conclusion was made based on their findings that the co-contraction of the hamstring and quadriceps and the activation of the gracilis and tensor fascia latae increased when varus-valgus loads were increased at the knee (Lloyd and Buchanan, 2001).

Although the quadriceps and hamstrings are highlighted as the primary muscles that influence loading at the knee, the ankle plantar flexors play a role in the absorption of landing forces (Devita and Skelly, 1992; Schmitz et al., 2007). DeVita and Skelly (1992) quantified the muscular work done by the ankle, hip, and knee to absorb the kinetic energy in both a bilateral stiff and soft landing. Overall, the body absorbed 19% more energy in the soft landing compared to the stiff landing (Devita and Skelly, 1992). The ankle plantar flexors absorbed the most energy with a contribution of 37% of energy in the soft landing and 50% of energy in the stiff landing. During a single-leg landing task, Schmitz et al. (2007) observed that the ankle joint contributed 78.2% of total energy absorption in males and 88.3% for females. As this research highlights the energy absorption, it is important to note that females consistently have greater energy absorption at the ankle and knee compared to males (Decker et al., 2003; Schmitz et al., 2007). The proper functioning of the plantar flexors play an important role in dissipating the forces and supporting lower extremity stability, especially in females. In a video analysis of ACL injuries, the gastrocnemius-soleus complex does not have time to absorb the GRF. The inadequate time to absorb the GRF increases the impulse force travelling up the leg to place greater loads on the ACL (Boden et al., 2009). In static activation tasks, Lloyd and Buchanon (2001) observed a large valgus moment support from the medial gastrocnemius. This finding supports the role of the gastrocnemius in stabilizing the knee in tasks that demand large plantar-flexion moments. Additionally, Reed-Jones and Vallis (2008) and Klyne and colleagues (2012) observed prolonged total muscular activation of the gastrocnemius in people with ACL deficient knees. The prolonged gastrocnemius activity was proposed to increase knee stability by compensating for the increased joint laxity (Klyne et al., 2012; Reed-Jones and Vallis, 2008).

ACL Injuries

Prevalence of ACL Injuries

Knee injuries in sports result in time lost from competition. ACL ruptures are the most common sports-related injury that accounts for annual incidence of up to 200,000 instances a year (Beynnon et al., 2014; Slater et al., 2019). Incidences of ACL injuries commonly occur in athletes within the age range of 15 and 25 (Beynnon et al., 2014). Within this accepted age range of 15 to 25 years old, college athletes have a significantly higher injury risk than high school athletes, with female college athletes having an even higher injury risk than their male counterparts (Beynnon et al., 2014). Female athletes tear their ACL at a 2 to 8 times higher rate than males and in an analysis conducted by the National Collegiate Athletic Association (NCAA), it was estimated that 1 in 10 college female athletes will sustain a knee injury (Arendt and Dick, 1995; Joseph et al., 2008). It is estimated that there is an annual increase of 1.3% ACL injuries (Hootman et al., 2007). Female participation in NCAA athletic programs and the number of NCAA institutions supporting female varsity programs continues to grow which makes the susceptibility of a female college athlete sustaining an ACL injury important to address (Arendt and Dick, 1995).

ACL injuries can be classified as contact or non-contact injuries. Contact ACL injuries occur with bodily contact with another player while non-contact ACL injuries occur without bodily contact (Montgomery et al., 2018). Nearly three-quarters of ACL injuries are non-contact injuries (Boden et al., 2010; Landis et al., 2018). Landing from a jump with the knee close to extension, quickly stopping after running, one-step stop landing with the knee in hyperextension, and planting and cutting directions are common mechanism of non-contact ACL injuries (Arendt and Dick, 1995; Boden et al., 2000; Nagano et al., 2009).

Risk Factors of ACL Injuries

ACL injuries result from a combination of both intrinsic and extrinsic risk factors. Intrinsic factors are not modifiable factors while extrinsic factors can be changed (Ireland, 2002; Ruedl et al., 2011). Lower extremity alignment, joint laxity, femoral notch size, ACL size, hormones, skill level, age, and gender are intrinsic factors that influence the susceptibility of ACL injury risk (Beynnon and Shultz, 2008). The environment, equipment, shoes, motivation, and floor surface are extrinsic factors (Ireland, 2002; Ruedl et al., 2011).It is well accepted and documented that female are more likely to sustain an ACL injury than males (Arendt and Dick, 1995; Ford et al., 2003; Hewett et al., 2005; Ireland, 2002, 1999; Schilaty et al., 2018). There are a multitude of intrinsic factors that increase the susceptibility of females to sustaining an ACL injury. Generally, females have a larger Q-angle, smaller trochanteric breadth, and shorter femoral length compared to males (Ireland, 2002). ACL literature has consistently reported that smaller intercondylar notches house a smaller ACL and are at a greater risk of sustaining a ligamentous injury. Comparing males to females, females had a smaller size ACL, a narrower femur, and a narrower intercondylar notch (Ireland, 2002). Female knee injuries are attributed to greater peak knee abduction angles and moments (Hewett et al., 2005; Ihmels et al., 2020). The intrinsic biomechanical factors of a female are relevant when assessing injury risk. Research has presented that the biomechanical measures alone account for 80% of variance in the knee abduction moment when landing (Myer et al., 2011).

Mechanisms of ACL Injuries

Based on their cadaver research, Markolf et al. (1995) discouraged the cutting or twisting on a planted foot to avoid high ACL forces. However, in the nature of sport, this recommendation is not attainable. Greater knee abduction angles and internal adduction moments have been previously identified as the primary mechanism for non-contact ACL injuries. Due to the multiplanar nature of sports, the assessment of ACL injury mechanisms has extended to include all planes of motion with the of knee valgus moment combined with limited knee flexion and internal tibial torque to produce a greater ACL-loading conditions (Oh et al., 2012). Video analysis of both landing and cutting maneuvers concur that ACL injuries occur within the first 100 ms after initial contact with minimal knee flexion (Koga et al., 2010; Krosshaug et al., 2007). In attempting to visualize the mechanism of a non-contact ACL injury, Ireland (1999) described the “position of no return.” This position is characteristic of a loss of trunk-pelvis-hip control with hip adduction and internal rotation, knee abduction with limited knee flexion, and tibial external rotation on an externally rotated foot (Ireland, 1999). Ireland (1999) characteristically described the “valgus collapse.” The valgus collapse is the knee moving medially as a result of large valgus or internal-external rotation forces (Koga et al., 2010; Krosshaug et al., 2007). In sports, the mechanisms of ACL injuries often occur when landing or when cutting. During these two tasks, the specific characteristics will be further assessed.

Landing Mechanisms and ACL Injuries

Landing is a critical component of sports like volleyball, basketball, and soccer. Landing from a jump has been identified as one of the primary non-contact mechanisms for an ACL injury in sports like basketball and volleyball (Arendt and Dick, 1995; Ferretti et al., 1992; Takahashi et al., 2019). During landing tasks, stiff landings, medial knee collapse, increased knee abduction angles, increased knee adduction moments, increased knee tibial internal rotation, increased vertical GRF, and decreased knee flexion are common mechanisms for ACL injuries (Decker et al., 2003; Hewett et al., 2005; Torry et al., 2011). The mechanism for ACL injuries occurs in both unilateral and bilateral landing. However, ACL injuries are more prevalent in unilateral tasks during sport and are linked to greater lower extremity demand like increased abduction angles and GRF (Munro et al., 2017; Nagano et al., 2009).

Regardless of task, there are sex-based differences in landing mechanics that place females at a larger risk for ACL injury. In assessing 205 female athletes involved in high-risk sports, Hewett and colleagues (2005) distinguished different lower extremity positions that increased the risk of ACL injuries. In the athletes that sustained an ACL injury, the knee abduction angle was 8.4° greater at initial contact and 7.6° greater at maximum compared to the non-injured knees. The maximum knee flexion at landing was 10.5° less in the injured knees compared to the uninjured. The injured knees compared to the uninjured knees had a 6.4 times larger external knee abduction moment and 20% greater vertical GRF (Hewett et al., 2005). Hewett et al. (2005) demonstrated that female athletes sustained ACL injuries under increased dynamic valgus and abduction loads. During a simulated landing task, Schilaty et al. (2018) observed a greater abduction torques at the knee in females compared to males. Females had 10.3 Nm increased knee abduction torque at 66 ms post impact and 22.3 Nm increased knee abduction torque at 100 ms compared to males. The increased knee torque values are important because they fall within the time range of sustaining an ACL injury. Additionally, the maximum values of external knee abduction moment were 52% greater than males (Schilaty et al., 2018). In addition to frontal plane differences at the knee while landing, Decker et al. (2003) and Leppänen et al. (2016) discussed potential implications of sagittal plane mechanisms for ACL injuries. Studying 171 female basketball and floorball players, new ACL injuries were associated with peak knee flexion angles and peak vertical GRF (Leppänen et al., 2017). A decreased peak knee flexion angle and increased vertical GRF, which are characteristics of a stiff landing, during a vertical drop jump were associated with an increased load at the knee and risk of ACL injury in the floorball and basketball players (Leppänen et al., 2017). Decker et al. (2003) found that females landed in a more erect posture with greater knee extension and ankle plantar flexion angles at initial ground contact with greater peak angular velocities compared to males. The erect posture of females when landing can be detrimental when landing unbalanced or in a state of fatigue because the quadriceps can pull the tibial anteriorly to strain the ACL (Decker et al., 2003; Pandy and Shelburne, 1997). Similar gender disparities were observed when assessing the difference in landing mechanics between males and females in a stop jump task (Peebles et al., 2020). Peebles et al. (2020) recorded a smaller peak knee flexion angle, a larger peak knee abduction angle, a larger frontal plane knee range of motion, and a larger peak knee internal adduction moment in females compared to males. These observations are consistent with past literature, however, Peebles et al. (2020) noted that the stop jump task was more representative test of sustaining an ACL injury because the stop jump task induced larger knee adduction moments and frontal plane knee loading than a drop-vertical jump.

It has been proposed that the positioning of the foot impacts the biomechanical factors associated with ACL injury (Kovács et al., 1999; Tran et al., 2016). The ankle angles influence the position of the trunk when landing that can place the lower extremity into a provocative position for injury (Sheehan et al., 2012). In assessing 1-legged landings in twenty athletes, Sheehan et al. (2012) found that athletes that sustained an ACL injury landed flat-footed with the center of mass 38 cm posterior to their base of support and smaller trunk angles in the sagittal plane while the uninjured athletes landed on their forefoot with a neutral trunk position. Sheehan et al. (2012) illustrated the impact the lower extremity positioning has on ACL injury risk when landing. Previous literature has linked forefoot contact to activities such as sprinting, bouncing, and spontaneous landing while heel-toe foot placement is associated to walking and running (Kovács et al., 1999). Kovács et al. (1999) demonstrated that the type of foot positioning while landing influenced the kinematics and kinetics of the lower extremity. With a heel-toe landing, the lower extremity landed in a stiffer posture with large vertical GRF and a smaller ankle joint angular displacement. With the heel-toe landing, -15.7 ± 5.4 J of work was done which only accounts for 23% of the work done in the forefoot contact in absorbing energy. In the flexion phase of landing, the largest torques came from the hip (37.3 ± 4.6% of total torque) and knee (44.8 ± 3.2% of total torque) with the heel-toe landing while the largest torques came from the ankle (37.3 ± 3.3% of total torque) and knee (37.3 ± 2.8% of total torque) in the forefoot landing (Kovács et al., 1999).

Tran et al. (2016) also identified the impact of different landing positions on the biomechanics of the lower extremity and its implication to ACL injury risk. Participants performed a bilateral landing activity and landed in a neutral position, a toe-in position, and a toe-out position (Tran et al., 2016). The neutral position was 0°, the toe-in position was 30° of internal rotation, and the toe-out position was 30° of external rotation. The toe-in landing was associated with increased hip adduction angles, knee abduction angles, and knee internal rotation angles at initial contact and peak compared to the neutral landing. Peak ankle dorsiflexion angle increased with the toe-in landing compared to the neutral landing. Decreases in the toe-in landing compared to the neutral landing were observed at peak hip flexion angle, peak knee flexion angle, peak knee external rotation angle, and peak and initial contact foot pronation angle. In the toe-out landing, increased hip abduction at peak and initial contact, increased foot pronation angle at initial contact, decreased knee abduction angle at peak and initial contact, decreased peak knee external rotation angle at initial contact, and decreased peak ankle dorsiflexion were found compared to the neutral landing position (Tran et al., 2016). This study is significant because when landing in the toe-in position, the risk factors for ACL injury are increased. Improper landing mechanics places the athletes into poor lower extremity body posture that is associated to ACL injury.

Cutting Mechanisms and ACL Injuries

Cutting mechanisms are another high-risk movement that can result in ACL injury. Similar to landing mechanics, ACL injuries while cutting occur with high loads of abduction forces at the knee while leaning on one foot (Boden et al., 2000; Nagano et al., 2009; Olsen et al., 2004). In sidestep cutting tasks, Mclean et al. (2004) found that the ACL cannot be injured based on the sagittal plane knee mechanics in an isolated setting. Increases in the external knee anterior force, abduction moment, and internal rotation moment were observed during the sidestepping task (McLean et al., 2004). However, the peak anterior drawer force never surpassed the 2,000 N threshold of ACL injury. Based on the research conducted by McLean et al. (2004), sagittal plane knee mechanics alone cannot rupture the ACL, so a combination of loading mechanisms are needed to produce an ACL injury in this task. Valgus loads and internal rotation loads impact are mechanisms observed that summate the forces of the knee to sustain an ACL injury. In a video analysis of ACL injuries in female netball players, the most common injury mechanism was the plant-and -cut movement with the 1-legged jump shot landing as the next most common mechanism (Olsen et al., 2004). Out of the twelve plant-and-cut ACL injuries, 4 of the injures were bilateral while 8 of the injures occurred during a single-leg push off. During the plant-and-cut task, ACL injuries occurred with a large valgus force and external or internal rotation of the knee at near full extension (Olsen et al., 2004). In comparing the ACL injury mechanisms between a single-leg landing task and plant-and-cut movement, the external tibial rotation at foot contact and peak internal tibial rotation were greater in the cutting task than the unilateral landing task while the peaks of knee abduction were similar across both tasks (Nagano et al., 2009). The similar abduction peaks but different rotational movements indicate that greater knee abduction angles influence injury risk in both types of ACL injury mechanisms but during the cutting task, the additive internal tibial rotation further risks sustaining an ACL injury (Nagano et al., 2009). In comparing a sidestep cut to a bilateral landing task, the kinematics and kinetics between the two tasks greatly differed (Kristianslund and Krosshaug, 2013). Kristianslund and Krosshaug (2013) observed lower knee flexion angles, a greater knee valgus angle, and a greater internal rotation angle at initial contact and maximum value when handball athletes perform a sidestep cut compared to a bilateral drop jump. Knee joint moments in all three planes were higher in the sidestep cut with the external knee abduction moment being 6 times higher than the drop jump (Kristianslund and Krosshaug, 2013). Kristianslund and Krosshaug (2013) also illustrated the increased demand of unilateral tasks compared to a bilateral task through the increased knee kinetics and kinematics during the cut compared to the drop jump.

ACL Screening Assessments

To identify athletes with poor landing mechanics at risk of sustaining an ACL injury, visual assessments have been implemented (Kristianslund and Krosshaug, 2013). Numerous studies have utilized the drop vertical jump (DVJ) to identify the biomechanical risk factored for ACL injury (Cruz et al., 2013; Decker et al., 2003; Ford et al., 2003; Hewett et al., 2005). The DVJ maneuver is a bilateral task consisting of dropping off a box, landing, and immediately performing a maximum vertical jump (Ford et al., 2003). Hewett et al. (2005) utilized the DVJ to predict the risk of ACL injury in 205 female athletes by assessing the joint angles and joint moments during the landing task. Hewett et al. (2005) demonstrated the association between the DVJ and ACL injury risk with knee abduction moments, knee abduction angles, and GRF. In the athletes that sustained an ACL rupture, their knee abduction angle was 8.4 degrees greater at initial contact, their peak knee abduction moment was 2.5 times greater, and their vertical GRF was 20% greater compared to the uninsured athletes (Hewett et al., 2005). The bilateral loading and uniplanar nature of the DVJ does not reflect an actual injury situation which limits the validity for predicting injury risk (Kristianslund and Krosshaug, 2013). Kristianslund and Krosshaug (2013)assessed the correlation between the kinematics and kinetics of the knee between a DVJ and a sports sidestep maneuver. The knee joint moments and knee joint angles were significantly different between DVJ and sidestep cuts (Kristianslund and Krosshaug, 2013). Additionally, poor correlations were established between the knee-valgus moments and DVJ and high knee abduction moments were not observed in the athletes that sustained an ACL injury (Kristianslund and Krosshaug, 2013).

Tasks that incorporate both vertical and horizontal movement patterns have been utilized in lab settings to better reflect sports movement (Cruz et al., 2013; Fong et al., 2011; Padua et al., 2015; Peebles et al., 2020; Schroeder et al., 2021). Forward vertical jumps and stop jumps have been implemented to represent sport-specific movements (Cruz et al., 2013; Padua et al., 2015; Peebles et al., 2020). To validate the use of these tasks in contrast to the DVJ, research studies have compared the lower extremity kinematics and kinetics when performing a DVJ to the forward vertical jump and stop jump (Cruz et al., 2013; Jones et al., 2014; Munro et al., 2017; Peebles et al., 2020). The stop jump reflects real-life sport movement as it is comprised of participants running towards and jumping onto a target (Peebles et al., 2020). In the stop jump, Peebles et al. (2020) found that the posterior GRF and peak internal knee adduction moment were larger than in the DVJ. Their findings agree with previous literature that compared a forward vertical jump to the DVJ (Cruz et al., 2013). Cruz et al. (2013) compared the lower extremity biomechanics of a drop landing, a DVJ, and a forward vertical jump. Like Peebles et al. (2020), Cruz et al. (2013) found that the external peak knee abduction moment, peak knee flexion moment, and peak hip adduction moment were significantly larger when performing the forward vertical jump compared to the DVJ. Additionally, the peak anterior tibial shear force was greater during the forward vertical jump compared the DVJ (Cruz et al., 2013). Based on the increased trunk and hip flexion angles and greater peal knee abduction and flexion moments, Cruz et al. (2013) concluded that the forward vertical jump was the most biomechanically demanding task which better simulates sports tasks. The forward vertical jump has been validated as a clinical assessment of landing biomechanics (Padua et al., 2015). In combination with the Landing Error Scoring System, the forward vertical jump identified biomechanical risk factors of movement quality with levels of 86% sensitivity and 64% specificity (Padua et al., 2015). Padua et al. (2015) found that athletes who sustained ACL injuries had higher Landing Error Scoring System scores than uninjured athletes when performing the forward vertical jump.

In another effort to simulate ACL injury mechanisms, unilateral landing tasks have been utilized. In comparing a single-legged landing to a DVJ, Munro et al., (2017) found greater frontal plane projection angles when performing a single-legged landing compared to a DVJ. These findings by the research by Munro et al. (2017) illustrate the increased kinematic demand between bilateral tasks and unilateral tasks. A moderate correlation between the knee-valgus angles and moments between the single-legged landing and the DVJ were established. Harty et al. (2011) also recorded strong relationships in frontal plane knee angles and moderate-to-good relationships in knee moments between a DVJ and a single-leg landing task. Considering previous research that has validated the DVJ to assess ACL biomechanical risk factors and the moderate-to-strong correlations established between kinematics and kinetics of landing between the DVJ and a unilateral task, the implication of the single-legged landing is a better screening tool for sports participants (Harty et al., 2011; Munro et al., 2017). The research conducted by Jones et al. (2014) further supports the inclusion of a unilateral screening task to identify ACL injury mechanics. Instead of comparing a single-legged landing to a DVJ, Jones et al. (2014) compared the knee biomechanics when performing a pivoting and a cutting maneuver. The peak knee abduction angles, knee abduction range of motion, and knee internal-external rotation range of motion between the single-legged landing and cut were strongly related and the knee abduction moments were moderately correlated (Jones et al., 2014). Peak knee abduction angles, knee abduction range of motion, and knee internal-external rotation range of motion were significantly correlated between the single-legged landing and a pivoting task (Jones et al., 2014). The results presented by Jones et al. (2014) show a relationship between landing and horizontal change of direction tasks which supports the use of unilateral landing tasks to identify at risk athletes. Athletes sustain ACL injuries in both cutting and landing tasks and with the correlations established between landing and cutting, these risk mechanisms can be identified through the use of a unilateral landing task.

To achieve both a horizontal motion and unilateral task, previous research has utilized a unilateral forward landing task that are comprised of participants jumping forward off of one foot from a 30 cm box onto a force plate 40% of the participants height away from the force plate on the opposite foot (Fong et al., 2011; Pappas and Carpes, 2012; Schroeder et al., 2021). Research that has utilized this task has been limited to only collect data on the dominant limb (Fong et al., 2011; Schroeder et al., 2021). A potential mechanism for ACL injuries is the neuromuscular asymmetry of leg dominance (Edwards et al., 2012; Ford et al., 2003). Leg dominance is a muscular strength and movement pattern asymmetry between limbs in which one limb has greater dynamic control and can more effectively absorb the high impact forces of sports (Edwards et al., 2012; Ford et al., 2003). Leg dominance asymmetries impair sports performance and have been proposed as a theory to explain the increased susceptibility of female athletes sustaining an ACL injury compared to males (Bishop et al., 2018; Ford et al., 2003; Pappas and Carpes, 2012). Ford et al. (2003) found that female basketball players landed with significantly larger maximum knee valgus angle on the dominant limb (27.6 ± 2.2º) compared to the non-dominant limb (12.5 ± 2.8º). This side-to-side difference in maximum knee valgus angles was not found in the male athletes. The female athletes landed with greater knee valgus range of motion and exhibited greater maximum knee valgus angles compared to the males (Ford et al., 2003). Pappas and Carpes (2012) further confirmed greater asymmetries in females compared to males. Pappas and Carpes (2012) investigated the effect of both gender and task on lower extremity limb asymmetries. During a forward landing task, Pappas and Carpes (2012) found that females had greater knee valgus and ankle abduction asymmetries than males. Greater knee valgus asymmetries and great hip adduction asymmetries were exhibited when performing a forward land compared to a drop land (Pappas and Carpes, 2012). Pappas and Carpes (2012) suggested that the higher knee valgus asymmetries potentially contribute to the increased risk of female athletes and that a forward land, which incorporates both horizontal and vertical motion, is a better modality of a test to identify asymmetries. Female athletes sustain ACL injuries when large asymmetries in leg weight distribution occur (Olsen et al., 2004; Pappas and Carpes, 2012). In assessing the knee biomechanics of female collegiate athletes using a DVJ, Morishige et al. (2019) found that the peak knee abduction angle was larger on the dominant limb compared to non-dominant limb and the knee flexion angle at initial contact when landing was smaller in the non-dominant limb. Morishige et al., (2019) concluded found that the dominant limb was at a 30.4% increased risk for a non-contact ACL injury while the non-dominant limb was at a 21.7% increased risk for injury in female collegiate athletes. Previous research has found leg-to-leg differences in knee load with side-to-side knee abduction moments 6.4 times greater in the injured limb compared to the uninjured limb which presents side-to-side knee valgus moment asymmetry as a risk factor for ACL injuries (Hewett et al., 2005).

Pappas and Carpes (2012) and Hewett et al. (2005) used double leg-landing to observe asymmetries. Kinematic and kinetic asymmetries in the lower extremity have also been identified using single-leg landing assessments (Aizawa et al., 2018; Wang and Fu, 2019). Wang and Fu (2019) assessed the kinematic, kinetic, and changes in center of pressure of the lower extremity to analyze the differences between the dominant and non-dominant limbs during single-leg landings. In the non-dominant limb, smaller hip and knee ranges of motion in the sagittal plane and larger medial-lateral center of pressure displacements were found compared to the dominant limb (Wang and Fu, 2019). The large medial-lateral center of pressure displacement indicates a lack of muscular control and proprioception in the non-dominant limb compared to the dominant limb (Wang and Fu, 2019). The peak GRF and time to peak GRF remained similar between the limbs (Wang and Fu, 2019). Based on their findings, Wang and Fu (2019) concluded that an increased risk would be placed on the non-dominant limb with a greater load being transmitted to the knee. In contrast, Aizawa et al. (2018) reported GRF differences between the dominant limb and non-dominant limbs during single-leg lateral jump landings. The peak medial GRF and the medial GRF at peak vertical ground reaction force were larger in the non-dominant limb compared to the dominant limb. The peak medial GRF was 4.3% of body weight greater and the medial GRF at peak vertical GRF was 3.9% of body weight greater for the non-dominant limb compared to the dominant limb (Aizawa et al., 2018). These findings suggest that the non-dominant limb is less efficient in absorbing the frontal plane impact forces and at higher risk of sustaining knee injuries.

Limb dominance potentially compromises both limbs with the weaker limb unable to manage forces while the stronger limb may experience even greater forces and knee loading to compensate for the weaker limb (Ford et al., 2003). Furthermore, the assessment of both limbs is warranted to determine the magnitude of sustaining an ACL injury.

Implementation of Ankle Technologies in Sport

In addition to ACL injuries, ankle sprains are a common injury in sports (Ihmels et al., 2020; Wiley and Nigg, 1996). Fifteen to thirty percent of all injuries in sports are ankle sprains with 85% of these ankle injuries occurring to the lateral ankle structure (Venesky et al., 2006). These injuries commonly occur when the foot is forced into plantar flexion and inversion (Ihmels et al., 2020; Wiley and Nigg, 1996). In attempt to limit ankle sprains in sports, over 40% of athletes report wearing ankle braces (Ihmels et al., 2020). Common preventative measures to avoid ankle sprains are taping, cloth wrapping, orthotic devices, high-top shoes, or a combination of methods (Thacker et al., 1999). The preventative measures are designed to restrict frontal plane motion of the subtalar joint and ankle plantar flexion (Cordova et al., 2000). The effect of preventive measures of ankle sprains is limited by inconsistent methodology and implementation (Cordova et al., 2000; Thacker et al., 1999).

Previous research has presented conflicting results on the effectiveness of taping and bracing (Thacker et al., 1999). Sitler et al. (1994) found reduced incidence of ankle sprains in intramural male basketball players wearing a semirigid orthosis compared to the players without ankle protection. In contrast, investigating the same semirigid orthosis but with male soccer players, Surve et al. (1994) found no difference in the incidence rate of the players wearing the semirigid orthosis compared to the control group. In a football study, wearing a foot orthosis was demonstrated to be more effective than ankle taping in reducing the amount of ankle sprains. Wearing a foot orthosis resulted in an ankle sprain rate of 2.6 sprains per 1000 participant games compared to a rate of 4.9 sprains per 1000 participant while wearing ankle tape (Rovere et al., 1988).

Barrett et al. (1993) found that the high-top shoes were ineffective of preventing ankle sprains in intramural college basketball because the ankle injury rate per minute played was $4.80\ x\ 10^{-4}$ in a high-top shoe compared to an injury rate of $4.06\ x\ 10^{-4}$ in a low-top (LT) shoe. However, with the high-top shoe in combination with another preventative strategy, the protective effect was enhanced (Barrett et al., 1993; Garrick, 1973). Barrett et al. (1993), although insignificant, observed a reduction in the ankle injury rate wearing the high-top shoe with the addition of an inflatable cuff. A reduction from 32.8 ankle sprains per 1000 participant-games to 14.7 ankle sprains per 1000 participant-games was reported by Garrick (1973) when wearing a high-top shoe combined with ankle taping. Although the extent of the effectiveness of these protective measures are unknown, ankle prophylactics alter ankle mechanics during sports tasks (Ihmels et al., 2020; Mason-Mackay et al., 2016; Schroeder and Weinhandl, 2019; Vanwanseele et al., 2014).

Changes in the ankle joint range of motion have been accompanied with the use of ankle prophylactics in sports (Ihmels et al., 2020; Robinson et al., 1986; Wiley and Nigg, 1996). Robinson et al. (1986) designed a high-top shoe with pockets positioned at the anterior and posterior to the medial and lateral malleoli. The pockets allowed the researchers to add stiffeners to the shoe to systematically mock three different conditions of ankle support. Ankle sagittal and frontal plane ranges of motion were recorded in all three stiffener conditions and compared to the control shoe without the stiffeners. Significant differences in inversion and eversion range of motion were observed between the control shoe and the highest stiffened shoe condition. The plantar flexion range of motion was significantly reduced in all 3 stiffened shoe conditions compared to the controlled shoe (Robinson et al., 1986). The researchers noted a general trend that as the stiffness of the shoe increased, the range of motion for inversion, eversion, and plantar flexion decreased. The dorsiflexion range of motion deviated from the generalized trend by having no change in range of motion. However, the researchers noted that the shoe upper deformed and slid on the leg which was an error not quantified and may potentially explain the lack of change observed in the dorsiflexion range of motion (Robinson et al., 1986).

Implementation of training programs have also been suggested as an alternative to ankle prophylactic use. Studies conducted with male, Swedish soccer players, the implementation of a prevention training program reduced the incidence rates of ankle sprains within their competition season (Ekstrand et al., 1983; Tropp et al., 1985). As observed in soccer, comparable results regarding training programs were observed in volleyball. With the implementation of a 2-hour training session composed of ankle disk and landing technique training, the incidence rate of ankle sprains reduced post intervention-training in 719 adult Norwegian volleyball players (Bahr et al., 1997). These training programs may act as a solution to avoid manipulating the foot placement with a brace and increase the strength of the ankle to reduce the number of ankle injuries.

Cordova et al. (2000) reported that dorsiflexion range of motion is not of clinical importance when discussing lateral ankle sprains because the common mechanism for this injury occurs when the foot is forced into inversion while the ankle is plantar flexed. However, these altered biomechanics at the ankle in both the sagittal plane and frontal plane have been linked to changes in knee biomechanics, including risk factors associated to ACL injuries. It is important to address the current preventative strategies for the common ankle sprain because their restriction of foot motion induces kinematic and kinetic changes at the proximal joints.

Relationship between the Ankle and the Knee

Sagittal Plane Motion of the Ankle and Knee Biomechanics

The relationship between the ankle and the knee biomechanics has been well established. Previous literature has suggested that the range of motion in the sagittal plane at the ankle manipulates kinematic and kinetic variables at the knee and may increase pain and injury at the knee joint (Hagins et al., 2007; Ihmels et al., 2020). The restriction of ankle motion in the sagittal plane has been linked to decreased knee flexion, increased knee valgus displacement, and increased peak knee internal rotation moment (Fong et al., 2011; Mündermann et al., 2003; Oh et al., 2012; Venesky et al., 2006).

Reducing the sagittal plane range of motion at the ankle also affects knee biomechanics in the frontal plane. In individuals that exhibit medial knee displacement, characterized by the midpoint of the patella moving medially to the great toe during a single-leg squat, passive ankle dorsiflexion range of motion was the only variable that was statistically different than individuals without medial knee displacement (Mauntel et al., 2013). Through assessing the passive ranges of motion at the ankle, hip, and knee while performing a single-leg squat, Mauntel et al. (2013) linked limited ankle dorsiflexion to contribute to a dynamic knee valgus position. In a drop land task, Sigward et al. (2008) observed similar effects of ankle range of motion on frontal plane movement at the knee. Sigward et al. (2008) aimed to explain the association between the ankle and hip strength and range of motion on the frontal plane knee excursion during a landing task. A significant negative correlation between ankle dorsiflexion range of motion and frontal plane knee excursion was established. The dorsiflexion range of motion accounted for 10.8% of variability of the frontal plane knee movement (Sigward et al., 2008). In an intervention that aimed to increase dorsiflexion range of motion, Schroeder et al. (2021) measured the knee landing biomechanics before and after the intervention. Compared to the baseline measurements, the dorsiflexion range of motion improved in the active and passive range of motion assessments, the star excursion balance test, the weight bearing lunge test and the right leg drop jump land task after the stretching and joint mobilization intervention. Accompanied with the increased dorsiflexion range of motion post intervention, the knee adduction angle and external knee abduction moment decreased post intervention (Schroeder et al., 2021). The relationship established between ankle dorsiflexion range of motion and the knee adduction angle and abduction moment presented in this study is significant because it demonstrates that full range of motion at the ankle allows for proper functioning at the knee to avoid placing the knee in strenuous loading positions. Similar to Schroeder et al. (2021) and Sigward et al. (2008), Malloy et al. (2015) found similar relationships when female collegiate soccer players performed a DVJ task. During the landing phase of the vertical drop jump task, the females with less ankle dorsiflexion flexibility had larger peak external abduction moments and larger peak abduction angles (Malloy et al., 2015). The negative relationship between dorsiflexion range of motion and knee abduction moments and knee abduction angles in collegiate athletes is important because this positioning of the lower extremity places larger loads on the knee in a population that is highly susceptible ACL injury.

In addition to changes to knee biomechanics in the frontal plane, limited ankle range of motion in the sagittal plane has been linked to altering the knee biomechanics in the sagittal plane (Fong et al., 2011; Hoch et al., 2015; Malloy et al., 2015; Venesky et al., 2006). As previously mentioned, after implementing a dorsiflexion range of motion intervention, Schroeder et al. (2021) recorded positive increases in dorsiflexion range of motion. Post intervention, knee flexion angles also increased during the active and passive range of motion assessments and during the landing of a drop jump task (Schroeder et al., 2021). Schroeder et al. (2021) demonstrated that greater sagittal plane movement in the ankle improved sagittal plane movement in the knee. The results presented by Schroeder et al. (2021) were in support of previous research conducted by Hoch et al. (2015). Hoch et al. (2015) also utilized the weight bearing lunge test and single-legged drop landing tasks to assess dorsiflexion range of motion, lower extremity joint angles both at initial contact and maximum angles, sagittal plane displacement and GRF. A positive-moderate correlation was established between the ankle dorsiflexion range of motion, knee and hip flexion angles, and sagittal plane displacement during the weight bearing lunge test (Hoch et al., 2015). During the drop land task, a positive moderate correlation was observed between the hip flexion, hip, and knee displacement but was positively strongly correlated to knee flexion at both initial contact and maximum angle. In this study, a moderate to strong relationship was established between the sagittal plane motions throughout the lower extremity. Another finding of clinical importance to this study was that the static dorsiflexion range of motion was moderately correlated to the dorsiflexion range of motion and ankle displacement during landing (Hoch et al., 2015). This relationship finding is beneficial because it justifies the used of the weight bearing lunge test as a clinical assessment as used by Schroeder et al. (2021) and supports the uses of static range of motion tests that are more accessible as a mechanism to identify individuals that may be at a predisposed risk of sustaining a knee injury due to their lack of sagittal plane motion in the ankle. In concurrence with Hoch et al. (2015) and Schroeder et al. (2021), Fong et al. (2011) found that clinical assessments of lower extremity kinematics were related to landing biomechanics. Passive ankle dorsiflexion range of motion was measured using a manual goniometer in both a knee-flexed and a knee-extended position. Fong et al. (2011) aimed to identify the relationship between the passive measurements of dorsiflexion range of motion and knee-flexion displacement during a bilateral landing task. The extended-knee dorsiflexion range of motion was significantly correlated to the sagittal plane knee displacement while landing. The participants with larger ranges of dorsiflexion generally were associated to have greater knee flexion. Greater amounts of knee flexion place reduced the loading associated to the ACL (Fong et al., 2011). As previously mentioned, landing with smaller flexion angles are characteristics of stiff landings that add increased strain the ACL. Based on the research by Schroeder et al. (2021), Hoch et al. (2015), and Fong et al. (2011), full range of motion in the sagittal plane at the ankle allows for greater amounts of knee flexion to avoid stiff landings that are potential causes of ACL injuries.

In the studies conducted by Schroeder et al. (2021), Hoch et al. (2015), and Fong et al. (2011), limited sagittal plane movement of the ankle was associated with changes to kinematic variables in both the frontal and sagittal planes at the knee that places the ACL under a strenuous load. These studies also collected kinetic data that also supports the claim that reduced sagittal plane motion of the ankle increases ACL injury risk by imposing greater GRF on the lower extremity. Schroeder et al. (2021) observed a reduction in anterior shear force during the drop landing task when ankle dorsiflexion range of motion improved. Both Hoch et al. (2015) and Fong et al. (2011) reported significant correlations between dorsiflexion range of motion and vertical GRF during a landing task. As the dorsiflexion range of motion increased, the vertical GRF decreased no matter if the task was unilateral or bilateral (Fong et al., 2011; Hoch et al., 2015). The greater range of motion in the sagittal plane at both the knee and the ankle aids in increasing the length of time the forces can be dissipated throughout the lower extremity which is the potential explanation of the decreases in anterior shear force and vertical GRF observed in these studies (Schroeder et al., 2021).

As previously addressed, smaller knee flexion, greater knee abduction, increased internal rotation, and greater vertical and posterior GRF, increase ACL injury risk. Without an external device like footwear or ankle bracing, altering the foot mechanics changed the mechanics of the knee that placed the lower extremity into positions that increase the ACL loading and injury risk. Therefore, it is important to understand how ankle prophylactics manipulate the knee mechanics during landing because it could potentially increase the risk of the athlete sustaining an ACL injury.

Frontal Plane Motion of the Ankle and Knee Biomechanics

Recent studies have investigated the impact of the frontal plane range of motion at the ankle on the biomechanics at the knee (Mündermann et al., 2003; Valenzuela et al., 2016; Venesky et al., 2006). In different drop landing conditions, Valenzuela et al. (2016) observed that when landing on a surface with 25° of inversion and 25° of plantarflexion, kinematic changes to knee mechanics were induced. Compared to landing on a flat surface, greater peak knee abduction angles and lower peak knee flexion angles were recorded in the combined inversion and plantarflexion landing (Valenzuela et al., 2016).

Mündermann et al. (2003) and Venesky et al. (2006) observed that restricting frontal plane movement at the ankle increased internal rotation moments at the knee. With limited movement at the ankle in the frontal plane, Venesky et al. (2006) recorded greater knee external rotation torques compared to an unrestricted ankle. When reducing the maximum foot eversion, Mündermann et al. (2003) observed that the ankle inversion moment decreased, the knee external rotation moment increased, the maximum abduction moment was delayed, and the vertical impact peak and loading rate increased. The reduced frontal plane range of motion at the ankle placed the foot in an inverted position that caused greater rotation forces at the knee (Mündermann et al., 2003; Venesky et al., 2006). Greater rotational forces increase the load on the ACL as it functions to limit internal rotation at the knee.

Effects of Ankle Prophylactics on Knee Biomechanics in Sports Tasks

Ankle Prophylactics during Cutting Tasks

Ankle braces are worn to prevent lateral ankle sprains however, it is important to understand how the ankle braces affect the proximal joints. If the impact of ankle braces increases the risk of injury at the knee or hip joint, a better solution may be sought out to minimize injury risks at all joints in the lower extremity kinetic chain. Research has demonstrated that ankle braces that restrict ankle plantar flexion and dorsiflexion range of motion by $\geq 3°$ induces hazardous changes to the knee biomechanics that elevate injury risk during sports tasks (Hagins et al., 2007; Malliaras et al., 2006; Wahlstedt and Rasmussen-Barr, 2015).

In addition to the research examining the foot and knee independent of potential footwear or preventative strategies at the ankle, research has been conducted to determine the relationship between ankle prophylactics and knee biomechanics (Ihmels et al., 2020; Mason-Mackay et al., 2016; Vanwanseele et al., 2014; Venesky et al., 2006). Although statistically non-significant, the incidence rate of knee injuries increased slightly when wearing an ankle brace compared to being unbraced in both basketball and football (McGuine et al., 2012).

With the relationship established between sagittal plane movement of the ankle altering knee biomechanics, Ihmels et al. (2020) sought to test an ankle prophylactic, the Ankle Roll Guard (ARG), that allowed the full ranges of plantar flexion and dorsiflexion. The participants performed a single leg cut in the ARG, in an ankle brace, in tape, and in a controlled condition without any ankle support. As expected, the ankle brace and tape restricted plantarflexion range of motion compared to the ARG and control. From previous literature an assumption could be made that since the ARG allows full ankle range of motion in the sagittal plane, changes in knee biomechanics should not occur. This was supported by the lack of differences observed between the ARG and the control condition in all the knee biomechanics variables recorded (Ihmels et al., 2020). In contrast, the braced condition decreased ankle plantar flexion and dorsiflexion range of motion by 5.4°. In the braced condition, the peak knee abduction angle and peak knee internal rotation moment increased compared to the controlled and ARG condition (Ihmels et al., 2020). In all conditions, the total knee reaction moment was not affected however, while wearing the ARG, the percentage of the total knee reaction moment in the sagittal plane increased (Ihmels et al., 2020). The findings by Ihmels et al. (2020) regarding the ARG are in support of previous literature investigating the impact of hinged ankle braces on knee biomechanics during a 45° cutting task (Schroeder and Weinhandl, 2019). Schroeder and Weinhandl (2019) found that a hinged ankle brace restricted ankle frontal plane movement but allowed full ankle sagittal range of motion. Additionally, compared to the unbraced condition, no significant differences were observed in the GRF, and the knee flexion, abduction, and internal rotation angles at peak and initial contact when wearing the hinged ankle brace (Schroeder and Weinhandl, 2019). The findings by Schroeder and Weinhandl (2019) and Ihmels et al. (2020) support that ankle prophylactics that allow full sagittal plane range of motion at the ankle do not alter the knee biomechanics that could increase knee injury risk.

Ankle Prophylactics during Landing Tasks

Like cutting tasks, research has demonstrated that bracing can manipulate the mechanics of the lower extremity to induce greater risks of ACL injuries in landing tasks. Mason-Mackay et al. (2016) investigated the effect of ankle bracing on sagittal plane excursion, vertical GRF, and joint stiffness during three different landing tasks in female netball players. Increases in the leg and joint stiffness and decreases in the sagittal plane excursion at the knee and ankle were observed in the braced condition compared to the unbraced condition (Mason-Mackay et al., 2016). No difference in the vertical GRF between the braced and unbraced condition was recorded. From their findings, Mason-Mackay et al. (2016) concluded that the increased leg stiffness and reduced excursion with the ankle braces predisposes netball players to increase risk of knee injuries. It was hypothesized that the athletes increased their knee abduction and pronation to compensate for the reduced sagittal plane motion recorded during landing. Additionally, Mason-Mackay et al. (2016) theorized that the lack of kinetic changes observed in their study was due to the athletes increasing their frontal plane excursion at the knee. In another landing study aimed to determine the effect of an ankle prophylactic on knee mechanics, Venesky et al. (2006) found a significantly greater ankle torques when landing. When twenty-four college students performed drop landings onto a 20° slanted board in a braced condition, the ankle eversion torque and knee external rotation torque was significantly greater compared to the ankle and knee torques in the unbraced condition (Venesky et al., 2006). Venesky and colleagues (2006) suggested that the increased knee external rotation torque was observed because of a proposed increase in internal rotation and stiffer landing induced by the brace which have been established to increase the risk of knee injury.

Shoe Collar Height

Like ankle taping and bracing, the development of high-top shoes in sport were designed to prevent ankle sprains (Barrett et al., 1993; Robinson et al., 1986). As previously addressed, conclusive research has yet to be presented that high-top shoes are more effective than LT shoes in preventing ankle sprains (Barrett et al., 1993; Lai et al., 2020). Both court and field sport athletes have adopted wearing high-top shoes while competing (Daack and Senchina, 2014). Previous research has investigated the role of high-top footwear on lower extremity kinematics and kinetics in various sports such as basketball, football, and netball. Ankle taping and spatting have been shown to demonstrate little effectiveness as physical activity increases, whereas a high-top shoe may be a preferred substitute because it maintains it shape over time (Cordova et al., 2000; Daack and Senchina, 2014; Meana et al., 2008). This reasoning of using high-top shoes as an ankle sprain prevention strategy is strengthened by the research conducted by Daack & Senchina (2014) in football cleats. In this research the subject’s ankle range of motion in the sagittal and frontal plane were measured in a LT, a mid-top, and a high-top football cleat, before and after completing football field tasks. Pre- to post-exercise, the high-top cleat maintained the same ankle range of motion without losing its restrictive effects (Daack and Senchina, 2014). In addition, a significant effect of cleat on dorsiflexion and inversion was found between the high-top cleat and the LT cleat and the high-top cleat and the mid-top cleat. The high-top cleat greatly reduced dorsiflexion and inversion ranges of motion (Daack and Senchina, 2014). Similar results were found when measuring the ankle range of motion in the sagittal plane while wearing high-top basketball shoes and LT basketball shoes (Yang et al., 2017). Twelve male collegiate basketball players performed a weight-bearing dorsiflexion maneuver that required them to flex their ankles while in a squat position until their heel came off the floor while wearing a high-top shoe and a LT shoe (Yang et al., 2017). In the high-top shoe, the peak dorsiflexion angle was significantly smaller compared to the peak angle in the LT shoe. Outside of sports research, Rowson et al. (2010) measured dorsiflexion range of motion allowed in a high-top shoe and LT shoe in a cadaver study using a pivoting foot mount. Rowson et al. (2010) reported a reduction of in peak dorsiflexion angle of 7.2% when wearing the high-top shoe compared to the LT shoe.

High-top shoes limit dorsiflexion range of motion during dynamic cutting and landing tasks (Brizuela et al., 1997; Lam et al., 2015; Li et al., 2013; Tang et al., 2020). In comparing high-top football cleats to LT cleats during an anterior single-leg jump landing and a lateral single-leg drop landing, the high-top cleats resulted in a significantly smaller dorsiflexion range of motion and a smaller total range of sagittal plane motion at the ankle (Tang et al., 2020). Additionally, during the lateral jump landing task, the total frontal plane range of motion and ankle inversion range of motion was much smaller in the high-top cleat compared to the LT cleat (Tang et al., 2020). In basketball, the ankle biomechanics during cutting tasks and jumping tasks were influenced by an increased shoe collar height (Lam et al., 2015; Li et al., 2013). Li et al. (2013) recorded ankle joint quasi-static movement range of motion in the sagittal plane, ankle dorsiflexion maximum torque, passive maximum ankle dorsiflexion power, and active maximum ankle dorsiflexion power during the push-off phase of a single-leg and double-leg jump in both a high-top and LT basketball shoe. The ankle range of motion in the sagittal plane, dorsiflexion maximum power, and torque were significantly smaller in the high-top shoe compared to the LT shoe (Li et al., 2013). In another basketball study testing a high-top and LT basketball shoe, Lam et al. (2015) found when wearing a high-top shoe, the basketball players had smaller sagittal plane ranges of motion at the ankle during a lateral cutting task. In respect to frontal plane motion at the ankle, Lam et al. (2015) reported similar results to the study research by Tang et al. (2020). Lam et al. (2015) found less ankle inversion and external rotation when wearing the high-top basketball shoe compared to the LT shoe during a cutting task. Both studies incorporated a lateral component into their testing procedure, a lateral drop landing conducted by Tang et al. (2020) and a lateral cut by Lam et al. (2015). These studies illustrate that high-top shoes affect the range of motion in the frontal plane and sagittal plane of the ankle which have been previously addressed as potential risks for ACL injuries. However, these articles share a commonality of not collecting kinematic and kinetic data at the knee and using only male participants. Vanwanseele et al. (2014) compared the impact of a high-top shoe compared to a standard netball shoe on the ankle and knee mechanics during a single-leg landing in 44 netball players. The high-top shoe did not restrict any ankle range of motion but increased the internal ankle plantar flexion moment and increased the peak knee internal rotation moment by 15%. While wearing the high-top shoe, the netball players landed more internally rotated which has been demonstrated to contribute the knee injuries (Vanwanseele et al., 2014).

Volleyball

The implementation and research surrounding high-top shoes has expanded from court sports, like basketball, to field sports, like football. However, limited research has been conducted in other sports like volleyball. Research in volleyball is warranted because high-top shoes are commonly worn, females make up a large population of the players, and volleyball requires jumping and cutting tasks that place the athletes at a larger risk of sustaining ligamentous knee injuries. Volleyball is a popular sport worldwide. In the United States alone, there are 400,000 high school girls that participate in volleyball, making it the third largest participation rate in female sports (Tillman et al., 2004a, 2004b). At the collegiate level, the NCAA has reported that volleyball has continued to grow with a record high of 17,780 student athletes playing across 1,069 teams in the 2018-2019 academic year (Chandran et al., 2021). As previously addressed, female athletes are at a predisposed risk of sustaining ligamentous knee injuries. In reviewing the injury data from the NCAA Injury Surveillance program over the seasons from 2014 to 2019, knee injuries, followed by ankle injuries accounted for the largest amount all injuries reported (Chandran et al., 2021). Most of the injuries at the knee joint are ACL injuries (Ferretti et al., 1992). For all reported injuries, 74.7% of the injuries resulted from non-contact or overuse mechanisms (Chandran et al., 2021).

The tasks of volleyball place athletes in at risk situations. In a single match alone, a volleyball player takes approximately 96 jumps and 60% of injuries that occur in volleyball occur when landing without contact with another player (Ferretti et al., 1992; Lobietti et al., 2010; Takahashi et al., 2019). Spiking, blocking, serving, and setting a volleyball all require jumping and landing tasks. Breaking these tasks down even further, some landing tasks are unilateral while others are bilateral. In quantifying the jumps of a collegiate player during a match, Tillman et al. (2004) reported that over 40% of the landings were unilateral in nature. When performing a jump float serve and setting, athletes tend to land on one foot while when spiking, spike serving, and blocking, volleyball players tend to land on two feet in a controlled environment (Lobietti et al., 2010). Xu et al. (2021) investigated the kinematic and kinetic impact of a volleyball player landing on a single leg compared to both legs following a volleyball spike. At the ankle, the plantarflexion angle, plantarflexion moment, ankle joint power, and dorsiflexion angular velocity were much greater in the single-leg landing than the double leg landing. At the knee and hip, the flexion angles and flexion angular velocities were smaller in the single-leg landing than the double leg. The single-leg landing had a greater knee and hip extension moments and knee and hip joint powers Greater peak vertical GRF and loading rates were observed in the single-leg landing compared to the double leg landing (Xu et al., 2020). The angles and moments recorded by Xu et al. (2021) all fall within the 50 ms of landing which places these biomechanical changes in the time frame of sustaining an ACL injury. As previously addressed, lower knee flexion angles and greater vertical GRF contribute to increased ACL loading. The findings by Xu et al. (2021) illustrates that volleyball players landing on a single limb are at risk of sustaining an ACL injury. In addition to landing on one limb, the subsequent task after landing can place volleyball players at an increased risk of sustaining an ACL injury. With the continuous nature of volleyball, the game does not often allow for an athlete to land and stabilize. In the situations where a volleyball player must land and run compared to a land and stick mechanism, higher knee abduction moments and larger abduction angles have been reported (Zahradnik et al., 2014). Besides volleyball tasks, differences in between male and female volleyball player have been reported. Compared to male volleyball players, female volleyball players land with less hip and knee flexion and greater vertical GRF, resulting in a stiffer landing (Salci et al., 2004).

As previously addressed, leg dominance and asymmetries may impair sports performance and are risk factors for knee injuries (Bishop et al., 2018; Kozinc and Šarabon, 2020). Volleyball players have been reported to have side-to-side difference in the lower extremity biomechanics that predisposes these athletes to injury (Kalata et al., 2020; Sinsurin et al., 2017). Based on the different positions of volleyball players, different asymmetries have been reported in female athletes (Castanharo et al., 2011). Castanharo et al. (2011) identified the asymmetries present in a setter, middle hitter, right-side hitter, and an outside hitter during a countermovement jump and drop landings. In the setting position, the hip-knee ratio for the knee joint moment was smaller in the right limb compared to the left limb, indicating an increased load to the knee (Castanharo et al., 2011). The middle hitter had larger peak vertical GRF, hip flexion joint moments, knee extension joint moments, and a longer rate of force development in the left leg compared to the right leg (Castanharo et al., 2011). The right-side hitter had a longer rate of force development, greater peak vertical GRF, a larger hip flexion joint moment, and a smaller knee extension joint moment in the left limb compared to the right limb (Castanharo et al., 2011). The outside hitter exhibited asymmetries in the knee joint moments and frontal plane hip range of motion. Hip range of motion in the frontal plane of the left limb was larger than in the right limb. The left limb of the outside hitter had a greater internal knee adductor moment and greater knee abduction compared to the right limb (Castanharo et al., 2011). Though limited by a small sample of volleyball players, Castanharo et al. (2011) demonstrated that asymmetries in the lower limb occur in all volleyball players with variation in inter-limb differences based on their position.

Kalata et al. (2020) classified the sport of volleyball as an asymmetrical sport comprised of alternating cyclic, balanced unilateral loading and acyclic, unbalanced loading movements. Within these movement patterns, volleyball players have developed different landing strategies between dominant and non-dominant limb (Sinsurin et al., 2017). In unilateral landings, Sinsurin et al. (2017) observed that the dominant limbs initially landed in a more internally rotated position and at peak vertical GRF, the dominant limb had greater knee internal rotation, less internal knee varus moment, and a greater knee valgus position than the non-dominant limb. No differences were found in the peak vertical GRF but the time to peak vertical GRF in the non-dominant limb was longer compared to the dominant limb (Sinsurin et al., 2017). This study implicates that the dominant limb of volleyball players may be at a greater risk of knee injury compared to the non-dominant limb. Thirty five percent of unilateral landings in volleyball land on the left foot while ten percent of unilateral landings occur on the right-foot (Tillman et al., 2004b). All of the volleyball players in the study by Sinsurin et al. (2017) were right-leg and right-hand dominant. Sinsurin et al. (2017) contributed a greater landing frequency on the left foot to right-handed players shifting their mass to the left from lateral trunk flexion while hitting the ball. It was reasoned that the dominant limb was at a greater risk of knee injury in this study because the non-dominant limb had adapted to utilize an efficient strategy of attenuating high-impact forces while landing from the repeated impact during volleyball (Sinsurin et al., 2017).

A factor that may contribute to different landing strategies between limbs is muscular strength. In comparison to athletes that do triathlons, play tennis, play soccer, and do sports aerobics, Kalata et al. (2020) found that volleyball players had the greatest level of asymmetries in the isokinetic strength of the knee extensors and the knee flexors. Wilkosz et al. (2021) measured the peak torque of the hamstrings and quadriceps in both limbs of elite male volleyball players during a concentric isokinetic test and a static test. During the isokinetic test, at angular velocities of 180º/s and 300º/s, the knee extensor strength was greater in the right limb compared to the left limb with larger values of peak torque (Wilkosz et al., 2021). The peak torque of the knee flexors was much smaller than the knee extensors but between limbs, the right leg reached greater peak torques during the isokinetic test (Wilkosz et al., 2021). As observed in the isokinetic task, the peak torques for both the knee extensor and flexors during the static task were greater in the right leg with the peak knee extensor torques exceeding the strength of the knee flexors (Wilkosz et al., 2021). From their findings, Wilkosz et al. (2021) attributed a low strength ratio and knee joint instability in volleyball players to the greater strength of the knee extensors in the right limb compared to the knee flexors in the same leg and the knee extensors in the left leg.

In previous volleyball research that sought to quantify what musculature and joint factors were associated with patellar tendon injury, ankle dorsiflexion range of motion was the only factor significantly associated with this common knee injury (Malliaras et al., 2006). Fifty percent of volleyball players sustain a patellar tendon injury (Malliaras et al., 2006). In testing 113 male and volleyball players, jump height, plantar flexor strength, year of playing, and hours playing per week were not significantly correlated to patellar tendinopathy (Malliaras et al., 2006). The relationship between dorsiflexion range of motion and patellar tendon injuries established in volleyball players in conjunction with research illustrating that high-top shoe limit dorsiflexion range of motion in sports yields that it is important to address the effects of high-top shoes at the ankle and knee joint in volleyball players. This research will aid in the development of an understanding of the effect of the shoes and the potential implications of wearing this footwear technology.

Conclusion

Non-contact ACL injuries are highly prevalent in sports with female athletes at a more susceptible risk than male athletes. In attempt to limit the ankle sprains, athletes wear ankle braces, ankle tape, and high-top shoes and it has been established that these preventative measures impact the mechanics of the ankle. During landing tasks, foot position and ankle mechanics impact the biomechanics of the knee in both the sagittal and frontal planes. Previous research has found that ankle braces that limit sagittal and frontal plane motion at the ankle increase the frontal plane movement and reduce sagittal plane movement at the knee which are associated to ACL injury mechanisms. Wearing high-top shoes is one method that has been used to reduce ankle sprain risk, however, limited research has been conducted on the effects of high-top shoes as it pertains to both ankle and knee joint biomechanics. To advance current research, analyzing the effect of high-top shoes on both the ankle and knee would better assess the role of high-top shoe's implication for potential musculature injuries. The prevalence of high-top shoes in sports, the greater ACL injury risk of females, and the task demands of volleyball emphasize the importance in understanding the impact of these shoes on the lower extremity biomechanics to aid in the prevention of ACL injuries.

CHAPTER III - METHODS

Participants

A convenience sample of eighteen NCAA Division I female collegiate volleyball players were recruited via word of mouth to participate in a single lab visit. Informed consent was provided by all participants and the study procedures were approved by the University of Southern Mississippi Institutional Review Board.

Inclusion Criteria

The inclusion criteria for the study consisted of female participants that were members of a collegiate volleyball team, and under no current practice limitations set by the team physician.

Exclusion Criteria

Participants were excluded from the study if the participants were not members of a collegiate volleyball program, were under playing restrictions set by the team physician, had any current or previous cardiovascular, metabolic, or neurological diseases, were pregnant, had a history of ACL injury or other knee injuries, or had any current or past lower extremity musculoskeletal injury in the past 6 months.

Pre-test Procedures

Each participant was provided verbal and electronic instruction via email for pre-testing procedures. Participants were required to refrain from any vigorous activity and consuming alcohol 24 hours prior to the testing session. To ensure the participants readiness, the pre-test instructions were confirmed via a questionnaire at the start of the session. Any participant that did not adhere to the pre-test instructions was rescheduled for another session.

Experimental Design

This study utilized a within and between subject randomized crossover design. An a priori power analysis for a two-way repeated-measures analysis of variance (ANOVA) indicated a sample size of 17 to achieve an alpha 0.05 and a beta of 0.80 (Faul et al., 2007). Thus, we recruited 18 participants to participate in the study conducted in the biomechanics laboratory. In the visit, the participants performed 20 unilateral forward drop landing tasks, 5 landings on each leg, in 2 shoe conditions. The participants all wore t-shirts and shorts to perform the landing tasks.

Participants performed forward unilateral drop landing tasks in two shoe collar height conditions. All participants wore the Adidas Crazyflight volleyball shoe for the low-top (LT) condition and the Adidas Crazyflight mid volleyball shoe for the mid-cut (MC) condition. Both shoes have identical characteristics of a lightweight stretch mesh upper, a thermoplastic polyurethane yarn reinforced midfoot, a molded sock liner, a boost midsole, and a Top Grip rubber outsole (Adidas AG, Herzogenaurach, Germany). The difference between the shoe conditions is the fit around the ankle. The shoe collar on the MC volleyball shoes rises above the malleoli while the regular, LT cut does not cross the ankle. For a standard US Women’s Adidas size 9 shoe, the MC had a collar height of 11 cm and the LT had a collar height of 9 cm.

Instrumentation

Lower extremity three-dimensional (3D) marker coordinate data was collected at 240 Hz using a 6-camera Qualisys motion capture system. Ground reaction force (GRF) data was measured synchronously at 1200 Hz using one AMTI in-ground force plate. Single anatomic reflective anatomic markers were placed bilaterally on the acromion process, iliac crest, greater trochanter, medial and lateral femoral epicondyles, medial and lateral malleoli, distal end of the second toe, and the first and fifth metatarsal heads. Four rigid thermoplastic segmental tracking clusters were placed on the trunk and pelvis, and bilaterally on the thighs, shanks, and heel.

Experimental Procedures

Anthropometric Measurements

Participant height was measured using a stadiometer and weight was measured while the participant stood still on the force plate. Leg dominance was determined by asking the participant what leg they would use to kick a soccer ball (Avedesian et al., 2019; Mercado-Palomino et al., 2021). The volleyball position was recorded to assess which foot the participant took off from and landed on the most during play. Participant demographics are listed in Table 1.

Experimental Assessments

Prior to performing the forward drop landing tasks, participants performed a 5-minute warm-up of running at a self-selected pace on a treadmill. The participants were also given a familiarization period to practice the unilateral landings prior to data collection. After the warmup and anthropometric measurements were recorded, the reflective anatomic and segmental tracking cluster markers were placed on the participant. With all the markers on, the participants were instructed to stand on the force plate with their legs shoulder width apart and their arms crossed and away from their body (Bennett et al., 2018). The participants were instructed to remain still and look straight ahead while a static trial of three seconds was recorded. After the static trial, the anatomic markers were removed while the clusters remained on to collect the data during landing.

Forward Drop Landing. The order of the landing limb was randomized first, followed by the shoe condition. The forward drop landing task was used to simulate a landing task performed in the sport of volleyball. The participants completed the task from a 30 cm box placed 40% of the height of the participant away from the leading edge of the force plate (Fong et al., 2011). While completing the task, participants jumped forward off one foot and landed on the opposite foot followed by maintaining balance for three seconds. The kinematics and kinetics were recorded during the forward drop landing task. The first five successful trials on each leading foot were used for the data analysis. Trials were repeated if the participants landed off the force plate, did not maintain balance after landing, or jumped vertically from the box. Between trials, participants were given 30 seconds of rest and between shoe conditions, the participants were given 3 minutes of rest.

Data Processing

Kinematic Data

Raw marker coordinate data was gap filled and then exported out of Qualisys Track Manager to Visual3D to be further processed. The 3D marker coordinate data was low-pass filtered using a fourth-order, zero lag, recursive Butterworth filter with a cutoff frequency of 10 Hz (Movahed et al., 2019; Xu et al., 2020). Angular computations were completed using a Cardan rotational sequence (X-Y-Z). Ankle dorsiflexion and inversion, knee extension and adduction, and hip flexion and adduction angles and moments were indicated by positive values. From the static trial, a model of the pelvis, right and left thigs, right and left shanks, and the right and left foot was created. The hip joint center was placed 25% of the distance from the ipsilateral to contralateral greater trochanter markers (Weinhandl and O'Connor, 2010). The knee joint centers were defined as the midpoint between the medial and lateral femoral epicondyle markers (Grood and Suntay, 1983). The ankle joint center was defined as the midpoint between the medial and lateral malleoli markers (Wu et al., 2002). Lower extremity joint angles were computed in the joint coordinate system (Grood and Suntay, 1983).

Ground Reaction Forces

GRF data was low-pass filtered using a fourth-order, zero lag, recursive Butterworth filter with a cutoff frequency of 10 Hz (Kristianslund et al., 2012). All GRF data was normalized to body weight before statistical analysis. Vertical GRF data was used to identify initial contact and determine the initiation of the landing phase (Nagano et al., 2009).

Kinetic Data

Internal joint moments were calculated using a Newton-Euler inverse dynamics approach and projected into the joint coordinate system (Kristianslund et al., 2014). Joint moments were normalized to body mass.

Variable Identification

From the static trial, a 6-degree freedom kinematic and kinetic model of the lower extremity and trunk was created for each participant using Visual 3D software (Version 6, C-Motion, Inc., Germantown, MD, USA). The static trial was referenced as the participant's neutral position and the kinematic variables were related back to this orientation (Jones et al., 2014). The variables of interest were all identified in the landing phase of the forward drop landing through Visual3D. The landing phase was defined as initial foot contact to peak knee flexion (Garcia et al., 2022; Schroeder et al., 2021; Xu et al., 2020). Initial contact was defined as the first time point vertical GRF reached the threshold of 10 N. The initial contact angles and joint moments were identified at the kinetic event of initial contact landing on the force platform. Within the landing phase, peak angles, peak GRF, and peak joint moments were identified in Visual3D. Peak joint moments have been correlated to initial contact angles at the ankle, knee, and hip (McLean et al., 2005). Total range of motion was defined as final angular position of the joint minus the initial angular position of the joint. All variables were confirmed via manual visual inspection.

Statistical Analysis

Descriptive statistics, including means and standard deviations, were calculated for all dependent variables. A two-way [Limb (Dominant vs. Non-Dominant) × Shoe (MC vs LT)] repeated-measures ANOVA was performed to evaluate the impact of the shoe conditions and limb dominance on the knee and ankle biomechanical variables. Shoe condition was treated as the within-subject factor while limb dominance was treated as the between-subject factor. If a significant main effect was found, post hoc pairwise t-tests with a Bonferroni correction were performed to identify the location of statistical significance. Statistical significance was set at $\alpha = 0.05$. Partial eta squared (η_p^2) was calculated as the effect size of the repeated measures ANOVA. Small, medium, and large effect sizes corresponded to the values of 0.0099, 0.0588, and 0.1379, respectively (Norouzian and Plonsky, 2018; Richardson, 2011). All statistical analyses were performed using SPSS software (version 27, SPSS, Chicago, IL).

CHAPTER IV – RESULTS

Out of the 18 participants recruited, 1 participant was excluded from the study due to a previous ACL rupture. A two-way repeated measures ANOVA did not reveal any statistically significant interactions between shoe and limb for any knee kinematic variables (Table 2). There was a significant main effect of shoe for peak dorsiflexion angle ($p = 0.038$), indicating greater peak dorsiflexion with low-top (LT) shoes (Table 2).

A significant interaction between shoe and limb was observed for internal peak knee flexion moment ($F(1, 17) = 4.620$, $p = 0.047$, Table 3). However, the post-hoc analysis did not reveal any statistically significant comparisons. A significant main effect of limb was found for internal peak knee abduction moment ($p = 0.014$), indicating greater peak knee abduction moment at the dominant limb. A main effect of shoe was observed at the ankle, with greater internal peak plantarflexion moment in the LT shoes ($p = 0.019$).

A significant interaction was found for peak lateral GRF ($F(1, 17) = 8.083$, $p = 0.012$, Table 4). A post-hoc analysis revealed greater peak lateral GRF in the LT shoe compared to the MC for the ND limb ($p = 0.003$). A significant main effect of shoe was found for peak medial GRF ($p = 0.017$), suggesting peak medially oriented GRF was greater in MC shoes.

Table 1 Participant Demographics presented as mean ± sd.

Age (yrs)	20.12 ± 1.32
Weight (kg)	72.29 ± 9.52
Height (cm)	175.53 ± 6.37
Shoe Size (US)	10.18 ± 1.26
Right Foot Dominant	16
Left Foot Dominant	1

Table 2 Kinematic variables of the knee and ankle joints. Both peak angles and angles at initial contact (IC) are presented as mean ± sd for the non-dominant low-top shoe (NDL), non-dominant mid-cut shoe (NDM), dominant low-top shoe (DL), and dominant mid-cut shoe (DM). Interaction and main effect results are presented as p-value (η^2). **Bold** represent statistical significance.

	NDL	NDM	DL	DM	Shoe	Limb	Interaction
Ankle							
Plantarflexion Angle	-34.9 ± 4.71	-34.1 ± 3.22	-32.7 ± 6.70	-33.5 ± 3.44	1.00 (0.000)	0.140 (0.131)	0.236 (0.087)
Dorsiflexion Angle	7.7 ± 5.41	6.1 ± 4.85	9.1 ± 4.44	7.2 ± 4.39	**0.038** (0.242)	0.178 (0.110)	0.745 (0.007)
Inversion Angle	0.1 ± 5.31	0.9 ± 4.92	1.3 ± 5.28	2.3 ± 4.81	0.148 (0.126)	0.231 (0.088)	0.868 (0.002)
Eversion Angle	-9.8 ± 2.67	-8.8 ± 3.49	-10.5 ± 3.52	-11.0 ± 2.54	0.658 (0.013)	0.120 (0.145)	0.156 (0.121)
Sagittal Plane IC	-35.1 ± 4.69	-34.2 ± 3.28	-30.2 ± 12.99	-33.5 ± 3.44	0.474 (0.033)	0.081 (0.178)	0.156 (0.122)
Frontal Plane IC	-0.5 ± 5.73	0.4 ± 5.22	1.2 ± 5.35	2.2 ± 4.93	0.139 (0.131)	0.148 (0.126)	0.901 (0.001)
Knee							
Flexion Angle	-57.2 ± 8.23	-57.6 ± 8.87	-59.4 ± 7.65	-59.5 ± 7.69	0.848 (0.002)	0.227 (0.090)	0.832 (0.003)
Extension Angle	-14.3 ± 5.35	-14.1 ± 5.27	-14.3 ± 3.95	-14.2 ± 4.79	0.703 (0.009)	0.991 (0.000)	0.843 (0.003)
Abduction Angle	-0.4 ± 3.21	-1.0 ± 3.18	-1.9 ± 4.02	-1.2 ± 2.82	0.913 (0.001)	0.330 (0.059)	0.083 (0.176)
Adduction Angle	4.9 ± 3.02	4.7 ± 2.93	4.8 ± 3.27	5.5 ± 3.25	0.682 (0.011)	0.626 (0.015)	0.243 (0.084)
Sagittal Plane IC	-14.3 ± 5.32	-14.0 ± 5.25	-14.3 ± 3.95	-14.2 ± 4.64	0.709 (0.009)	0.928 (0.001)	0.672 (0.011)
Frontal Plane IC	2.1 ± 1.83	2.0 ± 2.33	1.8 ± 2.51	2.0 ± 2.48	0.821 (0.003)	0.767 (0.006)	0.622 (0.016)

Table 3 Kinetic variables of the knee and ankle joints. Internal peak joint moments are presented as mean ± sd in % body mass (BM) for the non-dominant low-top shoe (NDL), non-dominant mid-cut shoe (NDM), dominant low-top shoe (DL), and dominant mid-cut shoe (DM). Interaction and main effect results are presented as p-value (η2). Bold represent statistical significance.

	NDL	NDM	DL	DM	Shoe	Limb	Interaction
Ankle							
Dorsiflexion Moment	-0.0 ± 0.02	-0.0 ± 0.06	0.0± 0.17	-0.2 ± 0.05	0.395 (0.046)	0.142 (0.130)	0.138 (0.132)
Plantarflexion Moment	-1.8 ± 0.34	-1.6 ± 0.24	-1.7 ± 0.45	-1.6 ± 0.34	**0.019** (0.298)	0.689 (0.010)	0.079 (0.180)
Eversion Moment	-0.2 ± 0.12	-.2 ± 0.13	-0.1 ± 0.05	-0.1 ± 0.09	0.359(0.053)	0.121 (0.144)	0.473 (0.033)
Inversion Moment	0.1 ± 0.12	0.1 ± 0.13	0.2 ± 0.07	0.1 ± 0.09	0.333 (0.059)	0.150 (0.125)	0.469 (0.033)
Knee							
Flexion Moment	-0.2 ± 0.09	-0.1 ± 0.15	-0.1 ± 0.12	-0.2 ± 0.10	0.190 (0.105)	0.520 (0.026)	**0.047** (0.224)
Extension Moment	3.1 ± 0.53	3.1 ± 0.46	3.2 ± 0.47	3.2 ± 0.47	0.277 (0.073)	0.544 (0.023)	0.759 (0.006)
Adduction Moment	0.1 ± 0.05	0.1 ± 0.06	0.0 ± 0.06	0.1 ± 0.10	0.459 (0.035)	0.592 (0.018)	0.442 (0.037)
Abduction Moment	-0.9 ± 0.24	-0.9 ± 0.28	-1.0 ± 0.25	-1.1 ± 0.30	0.280 (0.072)	**0.014** (0.323)	0.068 (0.193)

Table 4 Three-dimensional ground reaction forces presented as mean ± sd in % bodyweight (BW) for the non-dominant low-top shoe (NDL), non-dominant mid-cut shoe (NDM), dominant low-top shoe (DL), and dominant mid-cut shoe (DM). Interaction and main effect results are presented as p-value (η^2). **Bold** represent statistical significance.

	NDL	NDM	DL	DM	Shoe	Limb	Interaction
Mediolateral							
Max	0.010 ± 0.02$^{\chi}$	0.000 ± 0.02	0.007 ± 0.03	0.017 ± 0.03	0.968 (0.000)	0.309 (0.064)	**0.012** (0.336)
Min	-0.123 ± 0.03	-0.140 ± 0.04	-0.122 ± 0.03	-0.122 ± 0.03	**0.017** (0.309)	0.204 (0.099)	0.196 (0.102)
Anteroposterior							
Max	-0.082 ± 0.01	-0.092 ± 0.02	-0.077 ± 0.03	-0.082 ± 0.02	0.083 (0.176)	0.097 (0.163)	0.625 (0.015)
Min	-0.693 ± 0.06	-0.704 ± 0.05	-0.672 ± 0.05	-0.678 ± 0.06	0.345 (0.056)	0.126 (0.140)	0.795 (0.004)
Vertical							
Max	2.844 ± 0.36	2.805 ± 0.32	2.769 ± 0.38	2.765 ± 0.37	0.563 (0.021)	0.155 (0.122)	0.456 (0.035)

$^{\chi =}$ Statistically significant difference from the NDM

Table 5 Duration of landing phase in all conditions. The duration of the landing phase is presented as mean ± sd in milliseconds (ms) for the non-dominant low-top shoe (NDL), non-dominant mid-cut shoe (NDM), dominant low-top shoe (DL), dominant mid-cut shoe (DM), the non-dominant limb, the dominant limb, the MC, the LT, and overall.

	Landing Phase
NDL	0.181 ± 0.05
NDM	0.208 ± 0.05
DL	0.192 ± 0.04
DM	0.201 ± 0.03
Non-Dominant Limb	0.194 ± 0.05
Dominant Limb	0.197 ± 0.04
MC	0.204 ± 0.04
LT	0.186 ± 0.05
Overall	0.195 ± 0.05

CHAPTER V – DISCUSSION

This study aimed to determine the effects of mid-cut (MC) shoes on knee joint landing biomechanics in NCAA Division I collegiate volleyball players. For the interaction between shoe and limb, we hypothesized that the dominant limb in the MC shoe would have larger frontal plane knee joint angles and moments and smaller sagittal plane knee joint angles and moments than the non-dominant limb in the low-top shoes (LT). It was hypothesized that while wearing the MC shoe, the participants would land with larger initial contact and peak angles and moments in the frontal plane and with smaller initial contact and peak angles and peak knee joint moments in the sagittal plane. It was also hypothesized that ground reaction force (GRF) would be directed more vertical and posterior while wearing the MC shoes. This study also addressed the impact of limb dominance on knee injury risk in volleyball players. It was hypothesized that the dominant limb would have greater joint angles and moments in the frontal plane and smaller joint angles and moments in the sagittal plane, putting the limb at greater risk of injury compared to the non-dominant limb.

Shoe Collar Height

Our hypotheses regarding knee kinematics and kinetics while wearing the MC shoes were not supported. No statistically significant differences were found for initial contact and peak joint angles or peak knee joint moments at the knee between shoe conditions (Figures 1, 2). Previous literature has associated reduced passive ankle dorsiflexion range of motion to increased frontal plane knee angles and moments (Hagins et al., 2007; Malloy et al., 2015; Sigward et al., 2008). We found the MC shoe reduced the peak dorsiflexion angle (Figure 3) and reduced the internal peak plantarflexion moment while landing compared to the LT (Figure 4). With our findings at the ankle, we anticipated changes in knee mechanics. Although changes at the ankle were observed, our results suggest that the MC shoes did not induce changes to landing mechanics that increased frontal plane knee joint angles and moments. Vanwanseele et al. (2014) investigated knee landing mechanics while wearing different shoe collar heights in netball players and reported no statistically significant differences in the peak sagittal and frontal plane knee moments when landing in the high top shoes compared to a standard, low-cut netball shoe. They also reported an increased knee joint loading in the form of increased peak knee internal rotation moment in high top shoes. The high top shoe condition evaluated by Vanwanseele et al. (2014) was a basketball shoe with a collar height rising above the level of the talocrural joint. Whereas in this study, we used a MC shoe that has a collar height rising superior to the level of talocrural joint and to the lateral malleolus. The smaller peak dorsiflexion angle in the MC is similar to findings of reduced sagittal plane ankle range of motion in basketball and football high tops (Fu et al., 2014; Lam et al., 2015; Rowson et al., 2010; Tang et al., 2020). The structural design differences between a MC and a high top may be large enough to change how each type of shoe impacts knee mechanics. Although the MC shoe reduced sagittal plane motion, the comparatively lower collar height (relative to high-top shoes) may not be sufficient to elicit comparable changes in ankle or knee mechanics as high top shoes. The effect of increased collar height of the MC volleyball shoe in this study was not different enough from the LT volleyball shoe to impact knee mechanics while landing. Therefore, when choosing types of footwear, players can likely wear MC and LT volleyball shoes without significantly affecting landing mechanics at the knee joint.

The reductions in ankle sagittal plane kinematics and kinetics observed while wearing the MC did not impact knee joint mechanics, which could be contributed to the dynamic nature of the task performed, and collar height of the shoe worn. The MC shoes reduced peak dorsiflexion motion which was similar to the reductions in high top shoes found by Li et al. (2013) and Tang et al. (2020). However, Li et al. (2013) reported a 5° reduction in dorsiflexion range of motion and Tang et al. (2020) reported a 2.1° reduction in dorsiflexion range of motion in the high top shoes whereas an average of 1.25° reduction of peak dorsiflexion angle was found while wearing the MC shoes in the current study. Li et al. (2013) and Tang et al. (2020) measured dorsiflexion range of motion passively in contrast to the measurement of dorsiflexion angles during a dynamic landing task in the current study. It is apparent that the MC shoes did not inhibit sagittal plane motion at the ankle to the same extent as high top shoes, which further supports our claim that the high tops and MC shoe collar heights are sufficiently different to affect lower extremity joints in different ways. No differences in ankle range of motion in both the sagittal and frontal planes between the high top and LT shoe during unilateral landings were reported by Vanwanseele et al. (2014). This suggests a passive assessment of dorsiflexion range of motion may not best represent the dorsiflexion range of motion required for a dynamic task. Although Vanwanseele et al. (2014) reported an increase in knee internal rotation moment while wearing the high top shoes, the authors contributed these findings potentially to the unfamiliarity of netball players in high top shoes. Netballers do not commonly wear high top shoes when playing and the researchers noted that their participants had never played in high top shoes. The lack of kinematic changes at the ankle does not support why changes in loading were observed at the knee in the netball players. Volleyball players in the current study are familiar with both types of shoes which may explain why knee landing mechanics did not change across shoe collar heights. After polling the athletes on their shoe preference, 13 of the 17 participants preferred to play in the MC. The small, yet significant, differences in ankle kinematics while wearing the MC compared to the LT were not enough to increase frontal plane knee motion that has been associated to ACL injury risk, which suggests this style of MC shoe may not contribute significantly increasing internal knee adduction moments in volleyball players.

Other possible explanations for the lack of statistically significant differences in knee mechanics were the task that was performed, and the population studied. In the previous literature that demonstrated relationships between frontal plane knee excursion and ankle sagittal plane ROM, the tasks and participant demographics were different from the current study. Hagins et al. (2007) utilized landing on a tilted surface from a 40 cm box with professional dancers, Malloy et al. (2015) used a drop vertical jump with female collegiate soccer players, and Sigward et al. (2008) used a vertical drop jump from a 40 cm box with high school soccer players. In each of the previous studies, participants performed their landing task from a height slightly greater than in the current study, did not have a horizontal component to the task, had a subsequent jump after the landing phase, and performed bilateral landings. Additionally, although each of these studies utilized athletes, volleyball players are exposed to a large number of landings from the frequent maximal vertical jumps throughout practice and matches. Soccer players and dancers may not undergo vertical jumping and landing to the same extent as volleyball players. The volleyball players in the current study may have adapted to have more efficient landing mechanics from their repeated exposures to landing while playing. Fong et al. (2011) similarly required participants to jump off a 30 cm box placed 40% of the participants height away from the force plate and land on both feet. Consistent with our results, Fong et al. (2011) reported no association between dorsiflexion range of motion and knee-valgus displacement during landing. The 30 cm box height may not have been large enough to observe changes in landing strategies at the knee.

Our hypotheses regarding GRFs while landing were not supported. It was anticipated that the vertical and posterior GRF would increase while landing in the MC shoe compared to the LT shoe. Rather, we found that participants landed with greater medially directed GRF while landing in the MC shoe while vertical and posterior GRF remained similar between shoes (Figure 5). These results are consistent with Vanwanseele et al. (2014) who observed comparable peak vertical GRF between a high top and a LT shoe. However, previous research has found moderate correlations between passive ankle dorsiflexion range of motion in a LT shoe and vertical GRF (Fong et al., 2011; Hoch et al., 2015). For posterior GRF, a moderate correlation between passive ankle dorsiflexion range of motion was reported by Fong et al. (2011) and Hagins et al. (2007) reported an increase of 10.2% of the participants body weight in posterior ground reaction force when landing on the tilted surface that reduced ankle sagittal plane ROM. These studies did not report any relationship between ankle dorsiflexion range of motion and mediolateral GRF. Fong et al. (2011) and Hagins et al. (2007) also did not report any joint angles or moments that may have impacted their GRF results. Potential GRF differences presented here could be attributed to different landing styles. At initial contact, participants in the current study landed within the range of 30-36° of plantarflexion and 14° of knee flexion while Hoch et al. (2015) reported 26.24° of plantar flexion and 13.21° of knee flexion at initial contact. Greater sagittal plane excursion during the landing phase attenuates forces throughout the lower extremity (Fong et al., 2011; Hoch et al., 2015). The larger initial contact angles in the sagittal plane at both the knee and ankle in this study may have contributed to vertical and posterior GRF remaining similar across shoe conditions. Frontal plane GRF is related to the magnitude of inversion and eversion moments (Sacco et al., 2004; Simpson and Jiang, 1999). Although larger medial GRF were observed while wearing the MC shoe, no significant differences occurred in the frontal plane ankle angles or moments between shoe conditions. The magnitude of the mediolateral GRF between shoe collar heights does not seem to be large enough to affect the frontal plane ankle kinetics. A function of wearing shoes with an increased shoe collar height in sports are to limit frontal plane motion of the ankle to reduce the risk of a lateral ankle sprain injury (Barrett et al., 1993; Fu et al., 2014). Based on the results of this study, if coaches and athletic training staff are choosing footwear solely on ankle support, a more restrictive shoe is warranted than the MC volleyball shoe which did not support greater ankle stability by reducing excessive frontal plane ankle motion. Between the MC and LT shoes in this study, no initial contact joint angles at the knee or ankle were observed suggesting that volleyball players used similar landing strategies in both types of shoes and the MC were not restrictive enough to alter initial contact that may have increased GRF.

Limb

An interaction between shoe and limb was found for maximum mediolateral GRF. While wearing the MC in the non-dominant limb, the mediolateral GRF was directed more medial than wearing the LT in the non-dominant limb. However, though not statistically significant, the GRF was directed more medial in the LT than the MC in the dominant limb (Figure 6). Depending on the limb, the direction of the GRF differed between shoe conditions. This suggests that force-production strategies between dominant and non-dominant limb may play a greater role in GRF production than shoe collar height. Previous research has reported that the dominant limb exhibited greater peak vertical GRF and greater medially directed GRF when landing compared to the non-dominant limb (Seymore et al., 2019). Accompanied with a greater medially directed GRF in the dominant limb, Seymore et al. (2019) reported reductions in knee abduction ROM, greater knee flexion ROM, and smaller external peak knee abduction moments. Medially directed GRF aids in reduction of frontal plane knee mechanics that reduces frontal plane angles and moments at the knee (Seymore et al., 2019; Ueno et al., 2020). In the current study, the non-dominant limb had greater medially directed GRF and smaller internal peak knee abduction moments compared to landing on the dominant limb. In both limbs, the participants landed with knee adduction angles accompanied with internal knee abduction moments. Although external knee abduction moments have been associated to ACL injuries, greater internal knee abduction moments have been correlated with osteoarthritis (p =0.003, Baliunas et al., 2002). Additionally, internal knee abduction moments has been associated as a surrogate measure mediolateral force distribution that may alter muscle forces, joint kinematics, and limb alignment (Kutzner et al., 2013). The participants in this study may have landed more balanced on their non-dominant limb while greater amounts of internal knee abduction moment required when landing on the dominant limb to balance and improve the dynamic stability at the knee (Schipplein and

Andriacchi, 1991). Based on the GRF and knee abduction moments, it can be hypothesized that in this study, the volleyball players may have been more susceptible to risk factors associated to knee injuries on their dominant limb which is in support of our hypothesis but in opposition to the knee at risk presented by Seymore et al (2019). Seymore et al. (2019) assessed twelve healthy females during a bilateral drop task from a 30 cm box whereas the current study evaluated female collegiate volleyball players.

This discrepancy between limb mechanics drove another aim of this study; to analyze the effect of limb dominance and identify which limb was more susceptible to greater frontal plane knee mechanics when landing in volleyball players. Leg dominance has been identified as a risk factor ACL injury but conflicting research has presented different limbs at risk (Ford et al., 2003; Myer et al., 2011; Ruedl et al., 2012). The different task demands, experience level, and age across sports highlights the importance to assess limb dominance with each athlete within a sport to best represent their potential risks. Recreational female soccer players and recreational female skiers suffer more ACL injuries on their non-dominant limb (Brophy et al., 2010; Ruedl et al., 2012). However, in high-school aged female basketball players the dominant limb has been shown to have greater knee abduction angles, predisposing this limb to greater knee loading (Ford et al., 2003).

In volleyball specifically, some players have been identified to have muscular imbalance asymmetries and biomechanical differences (Kalata et al., 2020; Sinsurin et al., 2017). Volleyball players, compared to other sports such as tennis and soccer, were found to have greater asymmetries between limbs (Kalata et al., 2020). In this study, we found that the dominant limb exhibited greater internal knee abduction moments and laterally directed GRF which we anticipated due to the jumping and landing tasks in inherent in volleyball (Figure 5, Figure 7). In right-handed female volleyball players, Lobietti et al. (2010) assessed all skills over 6 professional female volleyball matches and found a greater percentage of the females landed on their left foot than the right foot. Additionally, Tillman et al. (2004) reported that when collegiate volleyball players landed unilaterally from a spike jump, 78% of the players landed on their left limb compared to 21% on their right limb. Right handed players typically land on their left limb while hitting a ball due to left lateral trunk flexion and a left shift of center of mass (Sinsurin et al., 2017). Sinsurin et al. (2017) observed that volleyball players landed on their dominant limb with greater knee abduction angles and smaller internal knee adduction moments compared to landing on their non-dominant limb. With the high frequency of right-handed volleyball players landing on their left limb, muscular adaptations were anticipated to improve landing strategy by volleyball players on the left limb. In right footed dominant participants, Niu et al. (2011) found that the non-dominant ankle adopted a better protective strategy when landing while the dominant ankle was at greater injury risk while landing. Eighty-eight percent of the participants in this study (n = 15) were right-handed and 94% (n = 16) were right-limb dominant. It is probable that these athletes adapted to the repeated amounts of landing on their non-dominant limb. Potential adaptions made to the non-dominant limb supports our findings as to why we observed smaller frontal plane knee moments in non-dominant limb compared to the dominant limb. With the larger internal knee abduction moments found in the dominant limb, the volleyball players in this study potentially adapted to the greater frequency of landing on their non-dominant limb and may be at a greater risk of knee injuries on their dominant limb. In contrast to other sports, volleyball players may be susceptible to greater risk of knee injury on their dominant limb, which could be attributed to the task demands of volleyball as previously addressed, or the of the playing level of the athlete. Different sport-specific demands and intensities of play impact kinematic and kinetic performance. A comparison of female recreational athletes to a female collegiate athletes found that ACL injury risk factors like peak knee abduction angles, external peak knee abduction moments, and peak knee flexion, affected dominant and non-dominant limbs differently between groups (Morishige et al., 2019). For recreational athletes, peak and initial contact knee abduction angles and knee internal rotation at initial contact were significantly larger in the non-dominant limb whereas the dominant limb of the collegiate athletes had a larger peak knee abduction angles and smaller knee flexion angles at initial contact (Morishige et al., 2019). The findings by Morishige et al. (2019) combined with Sinsurin et al. (2017), emphasize the need to both address sport and level specific factors when assessing limb asymmetries and training programs to reduce knee injury risk. In the current study, the participants were collegiate female athletes that participate in a sport that requires a large amount of jumping and landing. Augmented feedback, knee ligament injury prevention program, and landing training have been implemented as successful ways to improve female volleyball players landing mechanics (Noyes et al., 2011; Parsons and Alexander, 2012). By identifying that the dominant limb may be more susceptible to injury, volleyball and strength training sessions athletes participate in can be utilized to improve landing techniques to reduce the internal peak knee abduction moment observed in the dominant limb.

Sagittal and frontal plane angles at the knee and ankle during landing are presented in Figure 1. Although the main kinematic variables of interest in this study were initial contact and peak angles, we believe it is relevant to address the frontal plane angles at the knee at the end of the landing phase. The dominant limb had frontal plane knee angles trending to knee adduction while the non-dominant limb trended towards knee abduction approaching peak knee flexion (i.e., the end of the landing phase). The increasing trend in knee adduction angles in the dominant limb could potentially explain increased peak knee abduction moment in the dominant limb observed during landing (Figure 2). Further, a larger deviation in the magnitude of the frontal plane knee angles is observed between shoe conditions for the dominant limb compared to the non-dominant limb. Within the dominant limb, the MC shoe seemed to promote greater knee adduction angles than the LT shoe. As the body decelerates from landing, the increasing trend of increasing knee adduction angels could be attributed to the control of the trunk. The ability to control to trunk during the landing is important, as large ipsilateral trunk lean has been linked to knee frontal plane angles and moments (Cannon et al., 2021). The participants may have adopted different landing strategies with more ipsilateral trunk lean than a neutral trunk position in the dominant limb than the non-dominant limb that increased the knee adduction angles as the participants reached the end of landing. Additionally, the reduction of peak dorsiflexion angle in the MC shoe may have contributed to the greater difference in knee adduction angles between the MC and LT in the dominant limb. Qualitatively, in Figure 1 the reduction of peak dorsiflexion angle occurs at a similar time to peak knee flexion. At the end of the landing phase, when greater knee adduction angles in the MC shoe are observed, smaller sagittal plane motion in the MC shoe may have caused a change in landing strategy. This change potentially occurred to counteract the lost ankle motion whereas the LT with less restriction, may have allowed the participants to adopt a more neutral landing strategy. The greater knee adduction angles in the dominant limb as the knee approaches peak knee flexion may increase the frontal plane loading at the knee, and the greater dispersion in the trends between shoes and limbs may suggest that an analysis of the entirety of the landing phase may be warranted for injury implication risk between shoes and limbs.

The duration of the landing phase of the participants in the current study is reported in table 5. On average, the landing phase was 195 ms. Sport specific ACL ruptures have been identified to occur within the first 100 ms of landing (Koga et al., 2010; Krosshaug et al., 2007). A 200 ms time window has been utilized to demonstrate both the relationship of the ACL-hamstring reflex arc (Tsuda et al., 2001) and the relationship between anterior tibial acceleration, tibial slope, and ACL strain during simulated landing (McLean et al., 2011). Though the 200 ms window has been used to assess factors associated to ACL injury risk, caution is warranted when assessing the significant findings of this research to ACL injury risk due to some of the peak angles and moments potentially occurring after the 100 ms window ACL injuries.

Limitations

Results of this study need to be considered within in the context of several limitations, including using a forward drop land task, players from the same team, and the shoes. The forward unilateral landing task was utilized in this study because it has been validated as a clinical assessment of landing biomechanics (Padua et al., 2015), better represents the multiplanar nature of sports (Cruz et al., 2013), and have been used to identify limb asymmetries (Pappas and Carpes, 2012). Although the unilateral nature of the forward drop landing task is representative of single-leg landings in volleyball, this task does not replicate the multidirectional three-step sequence volleyball players use to jump and land. Mercado-Palomino et al. (2021) found that during block jump landings, volleyball players had different movement strategies between the dominant and non-dominant limbs depending on the direction of the blocking task and depending on which limb was the leading limb and which limb was the trailing limb. In our analysis, player position was not a consideration. It may be beneficial for future research to assess volleyball players by their position and their main movement patterns within that position to better assess limb dominance. The shoes themselves may have been a limitation to this study. Lam et al. (2019) reported wear time did not change ankle biomechanics in a basketball shoe but wear time was associated to greater GRF. The participants each brought in their own shoes which may have been worn different amounts of time. By result, the shoes could have potentially altered landing. Additionally, only sagittal plane and frontal plane variables were considered in this study. Transverse plane variables such as larger internal rotation moments and internal rotation angles, have been show to increase knee joint loading and increase ACL injury risk which is which is a factor to be considered when investigating knee joint loading in future studies (Shin et al., 2011).

Conclusion

The MC shoe limited dorsiflexion range of motion and decreased plantarflexion moments at the ankle, but these effects did not have a great enough magnitude to increase frontal plane kinematics and kinetics at the knee in volleyball players when landing. Wearing shoes with increased collar heights, like MC shoes, have been implemented into sports to reduce the occurrence of ankle sprains due to excessive inversion (Fu et al., 2014). However, wearing the MC volleyball shoes compared to the LTs in the current study did not reduce frontal plane motion of the ankle. Therefore, volleyball coaches and athletic training staffs can utilize this knowledge when making the best decision of footwear to wear to improve performance and reduce injury risk. From analyzing the inter-limb landing mechanics in this study, collegiate volleyball players demonstrated greater internal knee abduction moments when landing on their dominant limb potentially making this limb at greater risk of knee injury. Training regiments should focus on improving knee joint landing mechanics in the dominant limb of volleyball players to reduce injury risk and loading of the knee while playing.

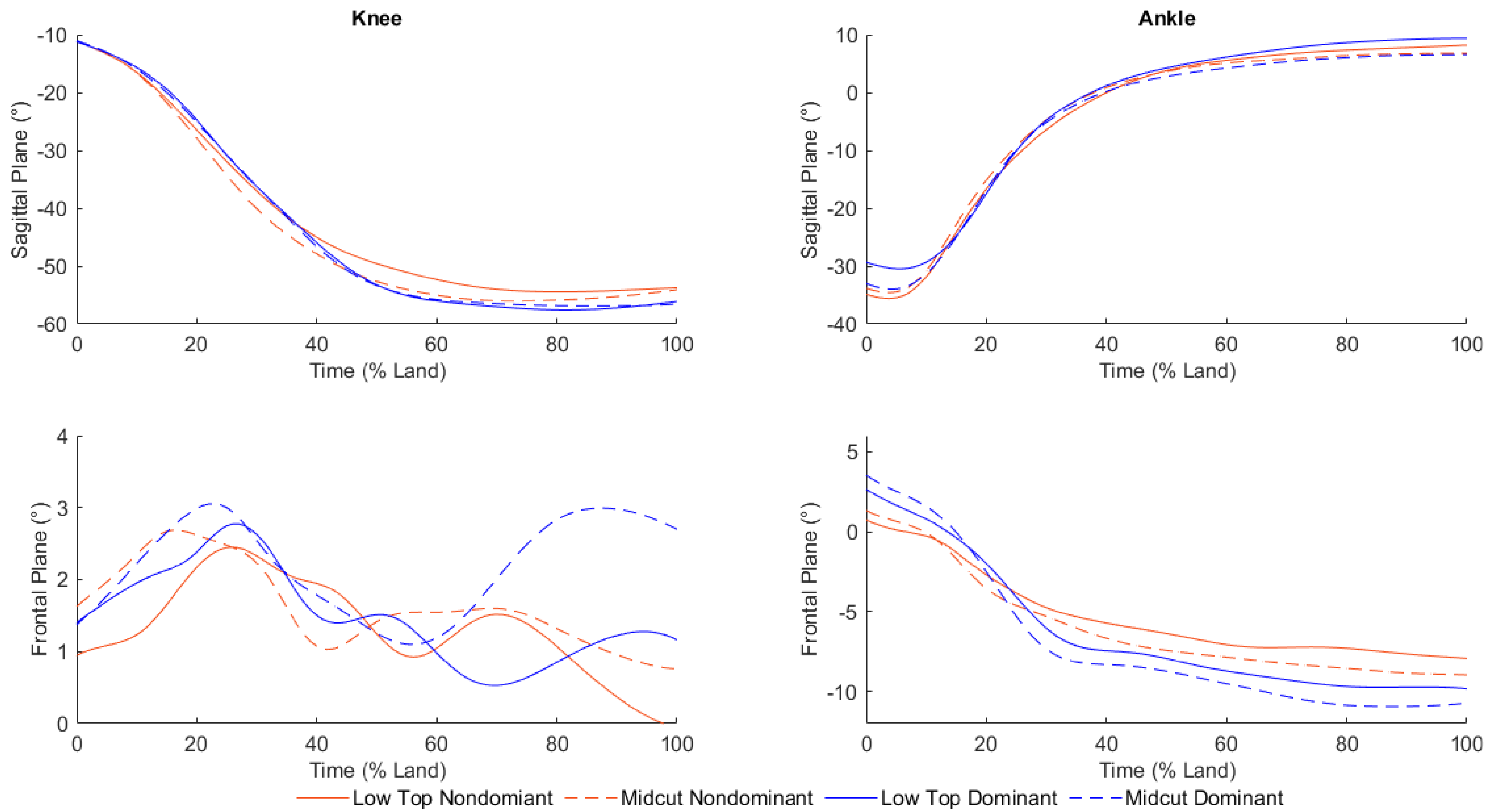

Figure 1. Knee and ankle joint angles in the sagittal and frontal plane during landing.

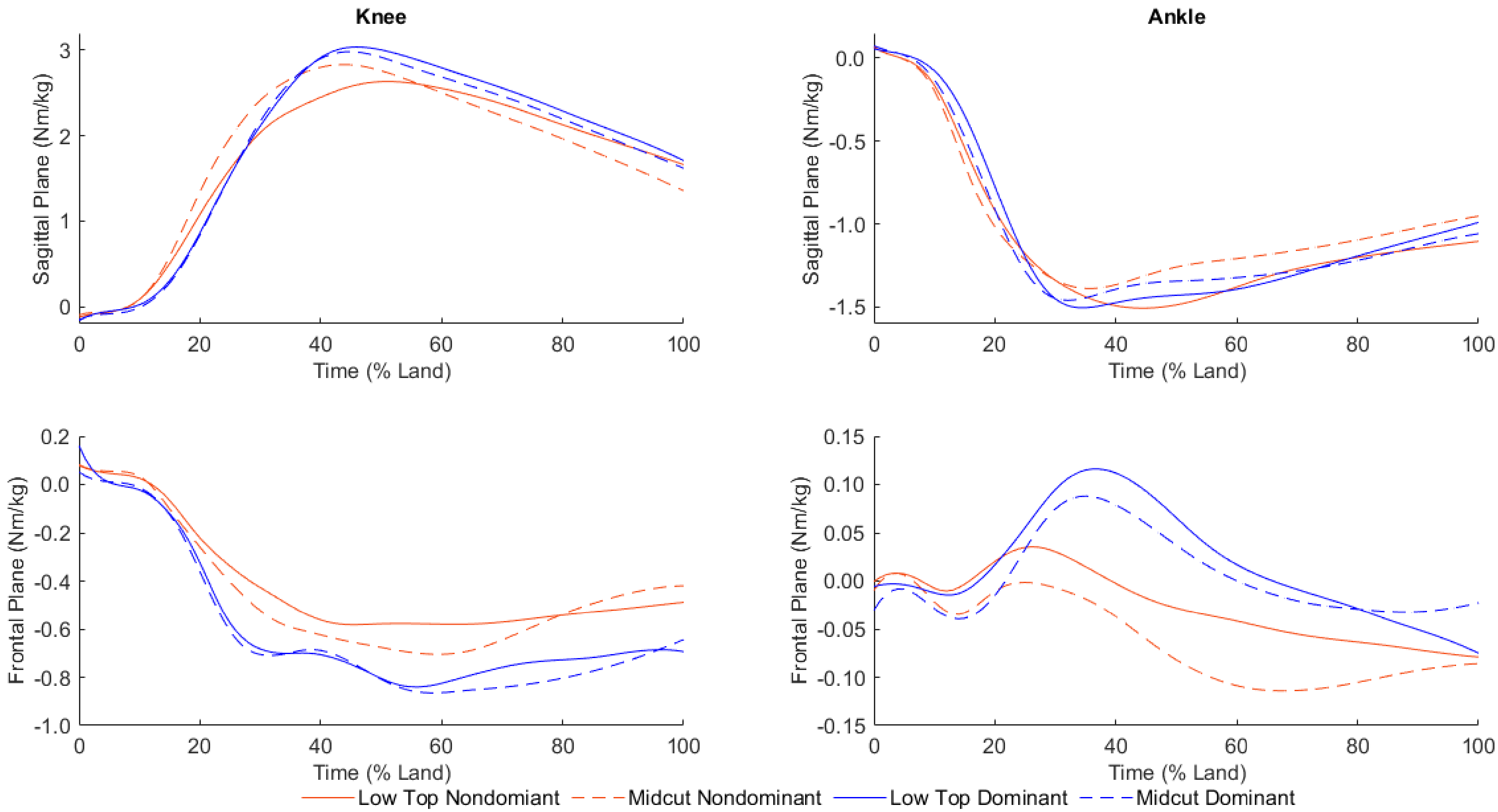

Figure 2. Knee and ankle joint moments in the sagittal and frontal plane during landing.

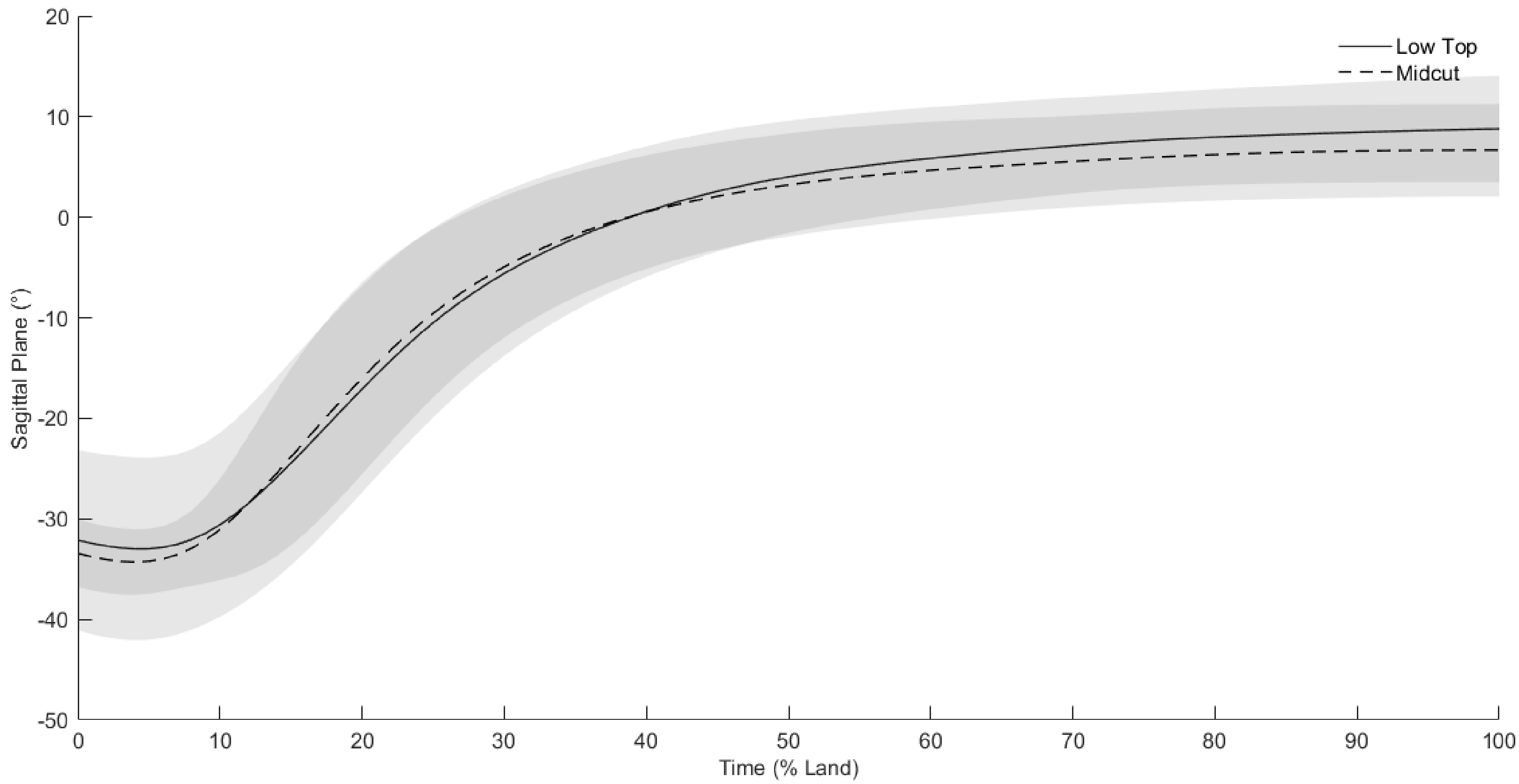

Figure 3. Sagittal plane ankle joint angles between shoes during landing.

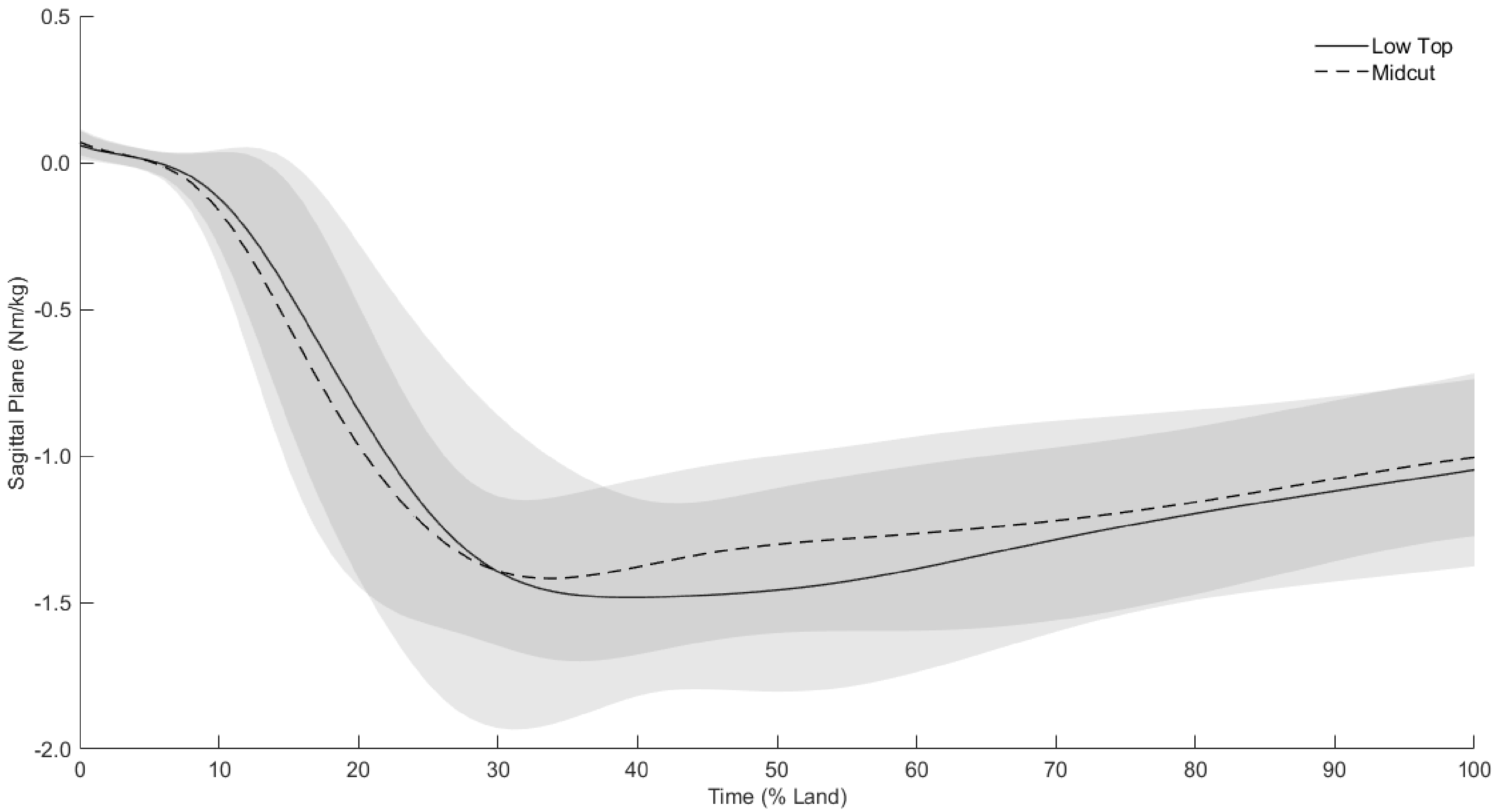

Figure 4. Sagittal plane ankle joint moment between shoes during landing.

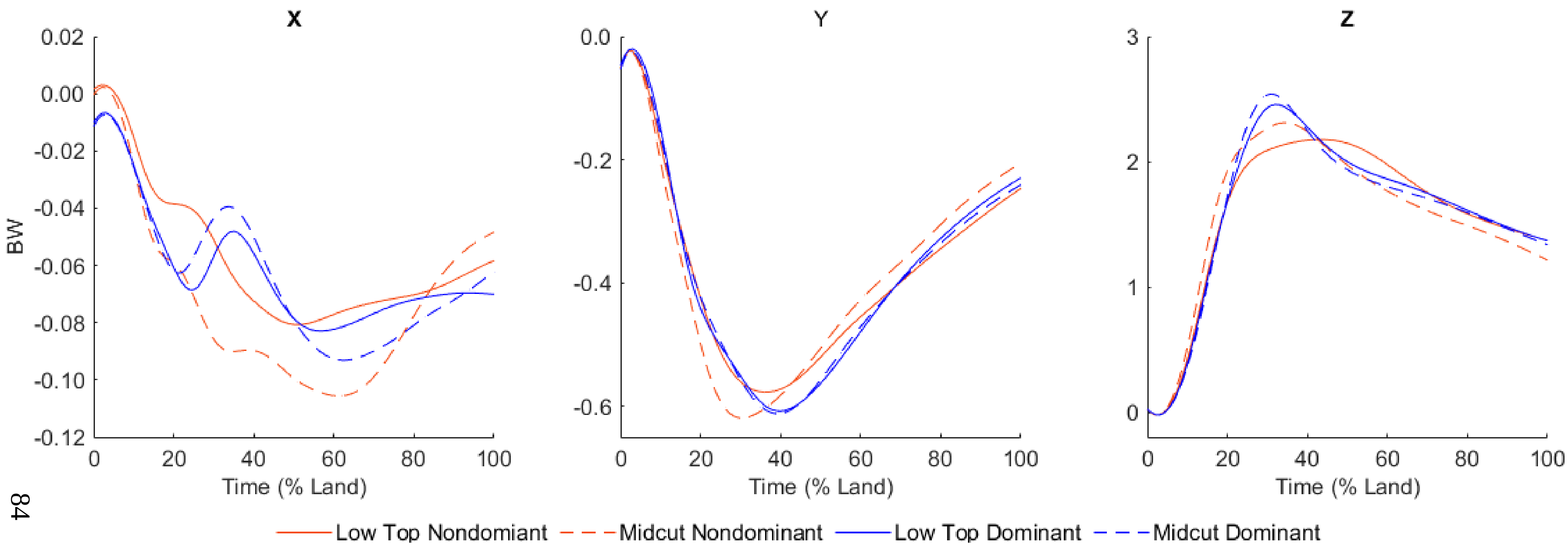

Figure 5. Mediolateral, anterior-posterior, and vertical ground reaction forces during landing.

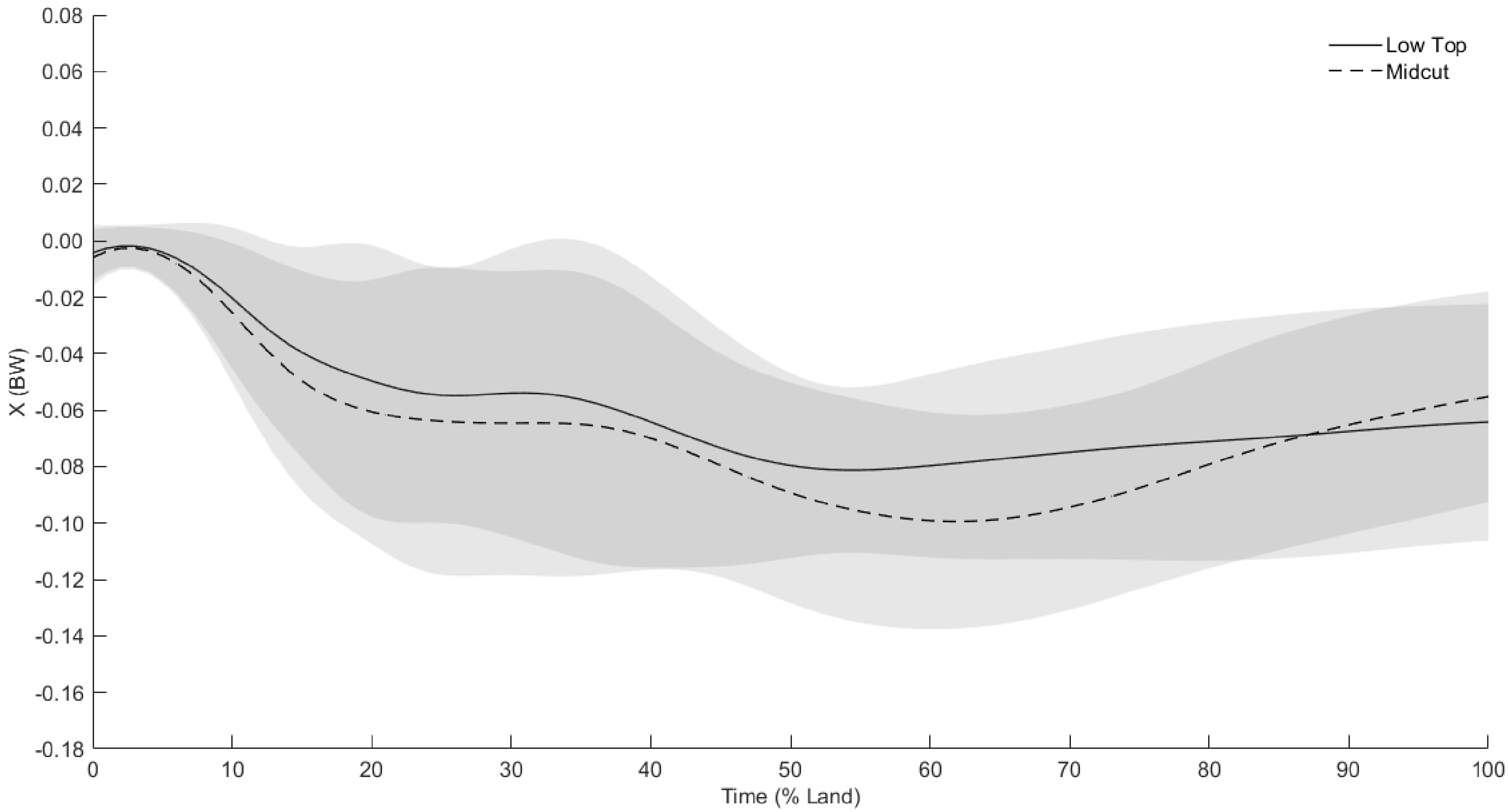

Figure 6. Mediolateral ground reaction force between shoes during landing.

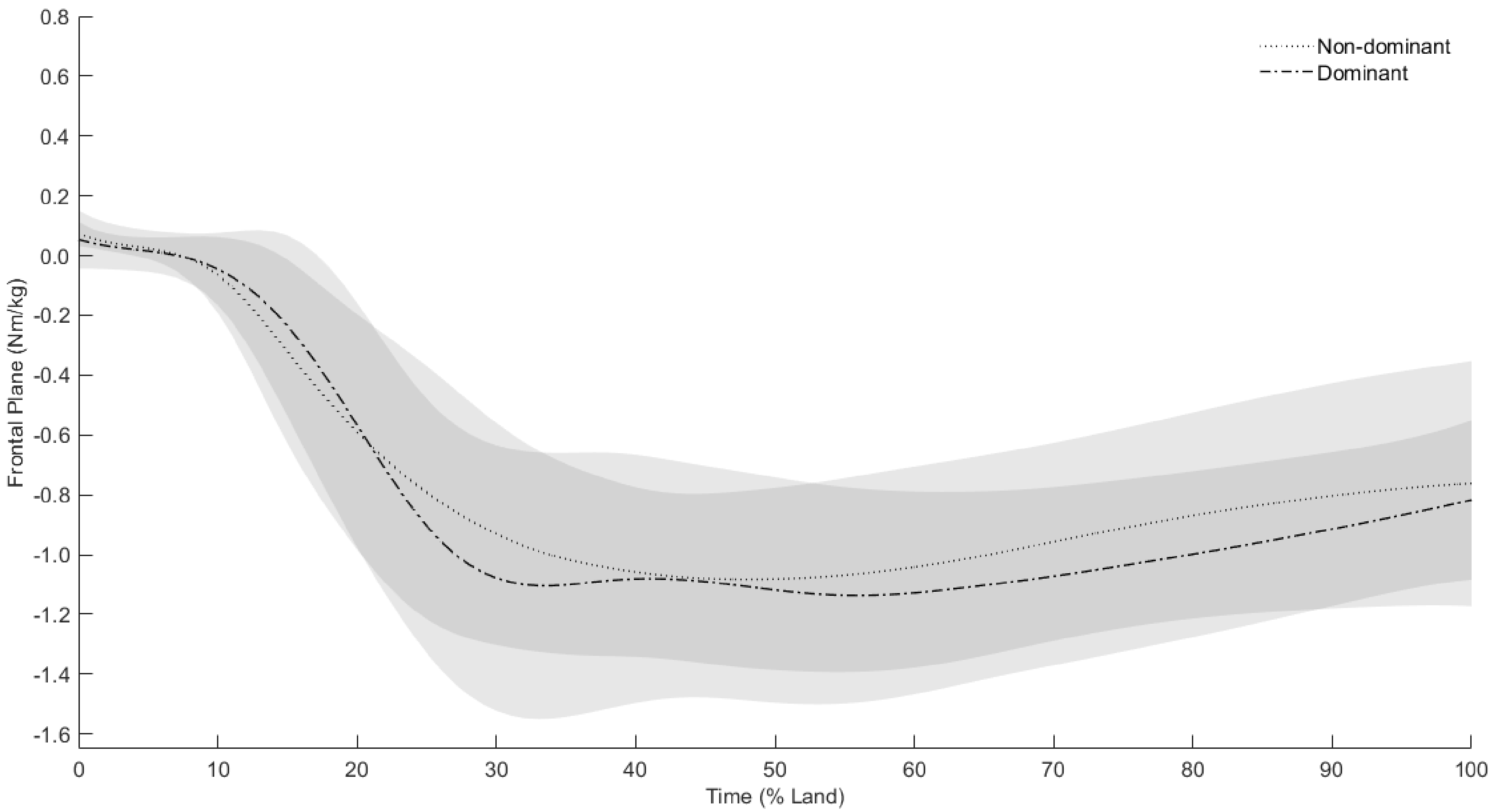

Figure 7. Frontal-plane knee moment between limbs during landing.

APPENDIX A –Subject-Specific Variables

Table A1. Peak knee flexion angles for each subject during landing (degrees). All joint angles are presented as mean for the non-dominant low-top shoe (NDL), non-dominant mid-cut shoe (NDM), dominant low-top shoe (DL), and dominant mid-cut shoe (DM). The mean and standard deviation (S.D.) are presented across all participants.

Subject	NDL	DL	NDM	DM
S1	-56.1	-55.8	-59.4	-63.4
S2	-62.9	-71.2	-66.9	-75.4
S3	-58.8	-65.7	-57.3	-58.6
S4	-62.6	-54.3	-61.6	-54
S5	-69.2	-60.4	-74.1	-65.3
S6	-55.7	-65.2	-56	-69
S7	-65.5	-61.4	-65.8	-53.1
S8	-55	-53.2	-61.8	-67.9
S9	-43.5	-49	-43	-50.1
S10	-59	-54.6	-52.4	-58.1
S11	-57.9	-66.8	-60.4	-62.5
S12	-64.1	-70.1	-62.2	-68.2
S13	-51	-70.4	-50.3	-50.2
S14	-57.7	-54.1	-53.6	-55
S15	-48.7	-50	-53.5	-55.7
S16	-60.4	-58.8	-62.3	-54.1
S17	-37.5	-49.3	-36.8	-50.4
Mean	-56.8	-59.4	-57.5	-59.5
S.D.	8.0	7.6	8.9	7.7

Table A2. Peak knee extension angle for each subject during landing (degrees). All joint angles are presented as mean for the non-dominant low-top shoe (NDL), non-dominant mid-cut shoe (NDM), dominant low-top shoe (DL), and dominant mid-cut shoe (DM). The mean and standard deviation (S.D.) are presented across all participants.

Subject	NDL	DL	NDM	DM
S1	-14.8	-15.3	-14.4	-14.8
S2	-12.8	-21.7	-15.9	-23.6
S3	-8.1	-17.8	-9.8	-14.4
S4	-19.2	-16.3	-17.1	-16.5
S5	-27.9	-19.5	-27.0	-18.4
S6	-16.7	-19.3	-16.9	-18.6
S7	-15.6	-9.0	-14.1	-5.8
S8	-16.8	-14.8	-16.4	-20.8
S9	-11.9	-9.1	-10.3	-11.8
S10	-14.8	-12.1	-15.9	-13.3
S11	-11.7	-11.1	-10.9	-8.5
S12	-13.4	-14.4	-14.3	-18.8
S13	-12.6	-15.8	-9.5	-11.4
S14	-12.5	-12.0	-9.1	-11.0
S15	-14.4	-13.3	-15.6	-11.5
S16	-18.2	-13.4	-18.2	-13.1
S17	-1.8	-7.5	-2.4	-8.3
Mean	-14.3	-14.3	-14.0	-14.2
S.D.	5.3	4.0	5.3	4.8

Table A3. Peak knee abduction angle for each subject during landing (degrees). All joint angles are presented as mean for the non-dominant low-top shoe (NDL), non-dominant mid-cut shoe (NDM), dominant low-top shoe (DL), and dominant mid-cut shoe (DM). The mean and standard deviation (S.D.) are presented across all participants.

Subject	NDL	DL	NDM	DM
S1	1.9	-0.2	-1.2	-2.0
S2	-0.3	-9.0	-2.3	-6.8
S3	-1.2	-11.4	-2.1	-5.4
S4	-0.3	-3.4	-1.1	-2.3
S5	-0.4	-1.0	0.0	3.4
S6	2.7	-0.6	0.8	-2.2
S7	-2.4	-0.1	-2.4	-1.9
S8	-6.7	-6.0	-3.5	-5.3
S9	-1.0	1.6	-4	0.9
S10	4.6	-1.7	4.1	0.4
S11	2.2	0.5	1.0	0.8
S12	-6.1	-1.7	-8.6	-0.6
S13	0.4	-5.7	-1.2	-1.8
S14	1.5	3.4	2.7	2.7
S15	4.3	2.9	4.1	2.1
S16	-0.9	-0.7	-4.0	-0.9
S17	-4.4	0.3	0.1	-1.0
Mean	-0.4	-1.9	-1.0	-1.2
S.D.	3.2	4.0	3.2	2.8

Table A4. Peak knee adduction angle for each subject during landing (degrees). All joint angles are presented as mean for the non-dominant low-top shoe (NDL), non-dominant mid-cut shoe (NDM), dominant low-top shoe (DL), and dominant mid-cut shoe (DM). The mean and standard deviation (S.D.) are presented across all participants.

Subject	NDL	DL	NDM	DM
S1	6.6	5.0	5.3	5.5
S2	4.9	0.0	3.9	1.8
S3	6.0	-2.0	3.4	2.1
S4	4.5	0.8	3.2	1.3
S5	4.9	5.3	9.2	11.7
S6	4.7	5.9	6.2	8.5
S7	5.5	8.5	4.0	4.5
S8	0.7	0.8	0.4	1.7
S9	0.4	2.9	-0.7	4.7
S10	11.5	4.0	8.8	4.6
S11	8.6	7.2	7.5	7.9
S12	4.1	7.4	5.3	8.7
S13	3.0	8.6	1.7	5.7
S14	6.1	6.3	6.6	9.4
S15	8.6	8.3	8.7	9.4
S16	3.4	8.0	2.4	3.5
S17	-0.1	5.1	3.5	1.9
Mean	4.9	4.8	4.7	5.5
S.D.	3.0	3.3	2.9	3.2

Table A5. Sagittal plane knee initial contact angle for each subject during landing (degrees). All joint angles are presented as mean for the non-dominant low-top shoe (NDL), non-dominant mid-cut shoe (NDM), dominant low-top shoe (DL), and dominant mid-cut shoe (DM). The mean and standard deviation (S.D.) are presented across all participants.

Subject	NDL	DL	NDM	DM
S1	-14.8	-15.3	-14.4	-14.8
S2	-12.8	-21.7	-15.9	-23.6
S3	-8.1	-17.8	-9.8	-14.4
S4	-19.2	-16.3	-17.1	-16.5
S5	-27.9	-19.5	-27	-18.4
S6	-16.7	-19.3	-16.9	-18.6
S7	-15.6	-9.0	-14.1	-7.2
S8	-16.8	-14.8	-16.4	-20.8
S9	-11.9	-9.1	-10.3	-11.8
S10	-14.8	-12.1	-15.9	-13.3
S11	-11.7	-11.1	-10.9	-8.5
S12	-13.4	-14.4	-14.3	-18.8
S13	-12.6	-15.8	-9.5	-11.4
S14	-12.5	-12.0	-9.1	-11
S15	-14.4	-13.3	-15.6	-11.5
S16	-18.2	-13.4	-18.2	-13.1
S17	-1.8	-7.5	-2.4	-8.3
Mean	-14.3	-14.3	-14.0	-14.2
S.D.	5.3	4.0	5.3	4.6

Table A6. Frontal plane knee initial contact angle for each subject during landing (degrees). All joint angles are presented as mean for the non-dominant low-top shoe (NDL), non-dominant mid-cut shoe (NDM), dominant low-top shoe (DL), and dominant mid-cut shoe (DM). The mean and standard deviation (S.D.) are presented across all participants.

Subject	NDL	DL	NDM	DM
S1	3.9	3.3	2.5	4.0
S2	1.4	-0.4	2.0	0.6
S3	2.9	-2.2	2.2	1.6
S4	-0.2	-0.6	0.4	-1.3
S5	1.7	3.4	2.5	3.7
S6	2.9	4.2	4.5	3.1
S7	1.7	1.0	-0.2	-0.4
S8	0.7	0.8	-1.1	0.7
S9	-0.2	1.8	-1.5	1.0
S10	4.7	3.7	4.5	4.1
S11	2.3	5.1	1.0	2.9
S12	3.0	4.2	5.1	7.7
S13	0.6	-2.5	-0.3	-1.6
S14	3.1	3.6	3.4	2.7
S15	6.1	5.5	6.9	5.1
S16	1.2	-0.7	0.7	0.3
S17	-0.7	1.1	1.5	-0.1
Mean	2.1	1.8	2.0	2.0
S.D.	1.8	2.5	2.3	2.5

Table A7. Peak plantarflexion angle for each subject during landing (degrees). All joint angles are presented as mean for the non-dominant low-top shoe (NDL), non-dominant mid-cut shoe (NDM), dominant low-top shoe (DL), and dominant mid-cut shoe (DM). The mean and standard deviation (S.D.) are presented across all participants.

Subject	NDL	DL	NDM	DM
S1	-39.6	-37.8	-36.8	-36.4
S2	-36.2	-31.2	-34.8	-27.1
S3	-31.8	-26.1	-31.6	-30.5
S4	-34.9	-39.1	-34.9	-35.3
S5	-36.2	-33.9	-31.9	-30.9
S6	-31.1	-31.4	-29.9	-28.6
S7	-29.9	-23.7	-32.8	-34.1
S8	-27.2	-33.0	-31.3	-33.0
S9	-37.4	-40.5	-36.0	-38.5
S10	-38.8	-40.5	-35.6	-37.1
S11	-36.7	-30.9	-35.4	-32.1
S12	-35.0	-31.0	-30.0	-30.9
S13	-32.1	-14.0	-32.4	-32.3
S14	-33.2	-33.7	-38.5	-36.0
S15	-31.7	-35.3	-31.4	-31.7
S16	-37.1	-37	-35.6	-37.7
S17	-47.9	-36.8	-42.4	-38.2
Mean	-35.1	-32.7	-34.2	-33.5
S.D.	4.7	6.7	3.3	3.4

Table A8. Peak dorsiflexion angle for each subject during landing (degrees). All joint angles are presented as mean for the non-dominant low-top shoe (NDL), non-dominant mid-cut shoe (NDM), dominant low-top shoe (DL), and dominant mid-cut shoe (DM). The mean and standard deviation (S.D.) are presented across all participants.

Subject	NDL	DL	NDM	DM
S1	5.7	3.1	9.1	9.0
S2	12.6	12.5	12.8	14.1
S3	11.5	9.5	6.0	5.5
S4	5.2	2.8	2.9	0.6
S5	5.7	4.7	8.4	4.3
S6	3.4	10.7	4.4	10.6
S7	12.9	13.8	8.7	6.4
S8	12.0	8.9	8.5	12.1
S9	-1.2	7.1	-2.9	0.8
S10	10.4	8.5	3.8	9.8
S11	2.4	14.4	3.4	11.6
S12	11.4	17.9	13.5	9.7
S13	1.9	11.1	2.3	0.7
S14	15.3	13.2	5.1	9.3
S15	5.7	6.0	8.5	9.4
S16	12.0	8.4	12.6	7.5
S17	-1.1	2.6	-3.5	0.5
Mean	7.4	9.1	6.1	7.2
S.D.	5.2	4.4	4.9	4.4

Table A9. Peak eversion angle for each subject during landing (degrees). All joint angles are presented as mean for the non-dominant low-top shoe (NDL), non-dominant mid-cut shoe (NDM), dominant low-top shoe (DL), and dominant mid-cut shoe (DM). The mean and standard deviation (S.D.) are presented across all participants.

Subject	NDL	DL	NDM	DM
S1	-9.3	-9.9	-8.4	-11.9
S2	-6.8	-6.0	-3.8	-6.9
S3	-5.9	-6.6	-6.7	-6.4
S4	-9.5	-9.1	-11.7	-10.9
S5	-12.6	-8.5	-15.7	-10.8
S6	-8.6	-12.5	-4.9	-12.2
S7	-9.4	-16.0	-10.1	-10.3
S8	-5.8	-9.1	-7.5	-14.4
S9	-8.0	-12.6	-5.8	-14.4
S10	-14.8	-6.8	-10.5	-8.8
S11	-13.7	-13.8	-13.0	-10.6
S12	-7.1	-8.0	-3.2	-10.4
S13	-10.9	-17.5	-6.2	-15.9
S14	-12.5	-7.1	-8.7	-10.5
S15	-10.6	-8.7	-11.9	-13.1
S16	-11.6	-15.1	-12.1	-11.3
S17	-9.2	-11.6	-10.1	-8.8
Mean	-9.8	-10.5	-8.8	-11.0
S.D.	2.7	3.5	3.5	2.5

Table A10. Peak inversion angle for each subject during landing (degrees). All joint angles are presented as mean for the non-dominant low-top shoe (NDL), non-dominant mid-cut shoe (NDM), dominant low-top shoe (DL), and dominant mid-cut shoe (DM). The mean and standard deviation (S.D.) are presented across all participants.

Subject	NDL	DL	NDM	DM
S1	2.9	2.3	2.0	2.9
S2	6.3	7.6	3.9	8.6
S3	9.4	10.6	8.7	10.5
S4	0.5	7.6	-1.4	3.2
S5	-1.0	3.7	-4.5	5.6
S6	-3.9	-3.7	1.4	3.6
S7	0.1	0.3	1.6	4.0
S8	7.4	-1.4	4.3	4.1
S9	-1.7	-4.4	6.1	-5.0
S10	3.9	3.6	6.8	2.6
S11	-1.9	-4.2	2.6	0.2
S12	6.9	8.9	5.4	8.4
S13	-3.8	-6.5	-3.1	-3.0
S14	-8.9	3.1	-5.3	-6.5
S15	-3.9	1.0	-6.3	3.6
S16	-4.4	-5.4	1.1	-4.0
S17	-6.8	-1.4	-7.9	0.7
Mean	0.1	1.3	0.9	2.3
S.D.	5.3	5.3	4.9	4.8

Table A11. Sagittal plane ankle initial contact angle for each subject during landing (degrees). All joint angles are presented as mean for the non-dominant low-top shoe (NDL), non-dominant mid-cut shoe (NDM), dominant low-top shoe (DL), and dominant mid-cut shoe (DM). The mean and standard deviation (S.D.) are presented across all participants.

Subject	NDL	DL	NDM	DM
S1	-39.6	-37.8	-36.8	-36.4
S2	-36.2	-31.2	-34.8	-27.1
S3	-31.8	-22.5	-31.6	-30.5
S4	-34.9	-39.1	-34.9	-35.3
S5	-36.2	-33.9	-31.9	-30.9
S6	-31.1	-31.4	-29.9	-28.6
S7	-29.9	-9.5	-32.8	-33.2
S8	-27.2	-33.0	-31.3	-33.0
S9	-37.4	-40.5	-36.0	-38.5
S10	-38.8	-40.5	-35.6	-37.1
S11	-36.7	-30.9	-35.4	-32.1
S12	-35.0	-31.0	-30.0	-30.9
S13	-32.1	11.1	-32.4	-32.3
S14	-33.2	-33.7	-38.5	-36.0
S15	-31.7	-35.3	-31.4	-31.7
S16	-37.1	-37.0	-35.6	-37.7
S17	-47.9	-36.8	-42.4	-38.2
Mean	-35.1	-30.2	-34.2	-33.5
S.D.	4.7	13.0	3.3	3.4

Table A12. Frontal plane ankle initial contact angle for each subject during landing (degrees). All joint angles are presented as mean for the non-dominant low-top shoe (NDL), non-dominant mid-cut shoe (NDM), dominant low-top shoe (DL), and dominant mid-cut shoe (DM). The mean and standard deviation (S.D.) are presented across all participants.

Subject	NDL	DL	NDM	DM
S1	2.9	2.3	2.0	2.9
S2	5.1	7.5	0.9	8.6
S3	9.4	10.6	8.7	10.5
S4	0.4	7.3	-1.4	3.2
S5	-1.2	3.7	-4.5	5.6
S6	-3.9	-3.7	0.8	3.6
S7	-0.4	0.3	1.5	3.4
S8	6.1	-1.6	4.3	4.1
S9	-2.0	-5.5	6.1	-5.0
S10	3.9	3.6	6.8	2.6
S11	-1.9	-4.2	2.6	0.2
S12	6.9	8.9	5.2	8.4
S13	-3.8	-6.6	-4.0	-3.0
S14	-12.3	3.1	-8.5	-7.6
S15	-4.7	0.9	-6.4	3.6
S16	-4.6	-5.4	1.1	-4.0
S17	-8.5	-1.4	-8.2	0.7
Mean	-0.5	1.2	0.4	2.2
S.D.	5.7	5.4	5.2	4.9

Table A13. Peak knee flexion moment for each subject during landing (Nm/kg). All joint moments are presented as mean for the non-dominant low-top shoe (NDL), non-dominant mid-cut shoe (NDM), dominant low-top shoe (DL), and dominant mid-cut shoe (DM). The mean and standard deviation (S.D.) are presented across all participants.

Subject	NDL	DL	NDM	DM
S1	0.08	-0.05	0.07	-0.07
S2	-0.08	-0.10	-0.05	-0.11
S3	-0.13	-0.06	-0.16	-0.15
S4	-0.08	-0.18	-0.03	-0.22
S5	0.01	-0.12	-0.04	-0.09
S6	-0.05	-0.15	-0.10	-0.18
S7	-0.05	-0.07	-0.02	-0.27
S8	-0.01	0.01	-0.03	-0.06
S9	-0.1	-0.23	-0.11	-0.33
S10	-0.09	-0.26	-0.01	-0.26
S11	-0.07	-0.27	-0.08	-0.20
S12	-0.06	-0.18	-0.10	-0.22
S13	-0.13	0.14	-0.17	-0.34
S14	-0.08	-0.22	-0.03	-0.33
S15	0.06	-0.04	0.01	-0.11
S16	-0.05	-0.04	-0.05	-0.04
S17	-0.14	-0.35	-0.21	-0.35
Mean	-0.06	-0.13	-0.07	-0.20
S.D.	0.06	0.12	0.07	0.11

Table A14. Peak knee extension moment for each subject during landing (Nm/kg). All joint moments are presented as mean for the non-dominant low-top shoe (NDL), non-dominant mid-cut shoe (NDM), dominant low-top shoe (DL), and dominant mid-cut shoe (DM). The mean and standard deviation (S.D.) are presented across all participants.

Subject	NDL	DL	NDM	DM
S1	2.90	3.68	2.94	3.41
S2	3.56	3.70	3.73	3.58
S3	3.18	2.77	3.06	3.18
S4	3.30	2.38	3.59	2.26
S5	3.64	3.17	3.75	3.13
S6	3.11	3.52	3.18	3.26
S7	3.21	2.89	2.95	2.77
S8	3.67	3.66	3.31	3.85
S9	2.02	3.03	2.43	3.02
S10	2.82	3.37	2.96	3.33
S11	3.68	3.64	3.53	3.55
S12	3.06	3.31	3.41	3.55
S13	3.21	2.86	3.08	3.12
S14	2.91	3.27	2.85	3.45
S15	2.62	2.88	2.84	2.91
S16	3.94	3.46	3.80	3.58
S17	2.18	2.10	2.17	2.08
Mean	3.12	3.16	3.15	3.18
S.D.	0.52	0.47	0.46	0.47

Table A15. Peak knee abduction moment for each subject during landing (Nm/kg). All joint moments are presented as mean for the non-dominant low-top shoe (NDL), non-dominant mid-cut shoe (NDM), dominant low-top shoe (DL), and dominant mid-cut shoe (DM). The mean and standard deviation (S.D.) are presented across all participants.

Subject	NDL	DL	NDM	DM
S1	-0.74	-1.23	-0.77	-1.13
S2	-0.74	-0.98	-0.65	-1.03
S3	-0.82	-0.58	-0.75	-0.75
S4	-0.69	-0.81	-0.60	-0.90
S5	-0.9	-1.02	-1.03	-1.33
S6	-0.88	-0.71	-0.95	-1.07
S7	-0.88	-1.02	-0.77	-0.92
S8	-0.88	-0.78	-0.98	-0.84
S9	-0.76	-0.98	-0.37	-1.11
S10	-1.33	-0.87	-1.32	-1.09
S11	-1.17	-1.08	-1.00	-1.28
S12	-0.48	-0.89	-0.41	-0.77
S13	-0.94	-1.48	-1.05	-1.54
S14	-1.45	-1.33	-1.48	-1.82
S15	-0.73	-0.97	-0.86	-0.96
S16	-0.95	-1.35	-0.81	-1.17
S17	-0.66	-0.74	-0.84	-0.61
Mean	-0.88	-0.99	-0.86	-1.08
S.D.	0.24	0.25	0.28	0.30

Table A16. Peak knee adduction moment for each subject during landing (Nm/kg). All joint moments are presented as mean for the non-dominant low-top shoe (NDL), non-dominant mid-cut shoe (NDM), dominant low-top shoe (DL), and dominant mid-cut shoe (DM). The mean and standard deviation (S.D.) are presented across all participants.

Subject	NDL	DL	NDM	DM
S1	0.15	0.00	0.13	-0.02
S2	0.07	-0.11	0.04	-0.12
S3	0.14	-0.11	0.10	-0.15
S4	0.04	0.08	0.07	0.07
S5	0.12	0.04	0.12	0.04
S6	0.00	0.15	-0.01	0.15
S7	0.04	0.09	0.01	0.14
S8	0.03	-0.02	0.03	-0.03
S9	0.19	0.05	0.15	0.12
S10	0.06	0.08	0.05	0.07
S11	0.08	0.17	0.14	0.11
S12	0.04	0.16	0.13	0.16
S13	0.05	0.11	0.10	0.06
S14	0.00	-0.01	0.03	0.18
S15	0.05	0.01	-0.03	0.04
S16	0.08	0.04	0.05	0.05
S17	0.07	0.11	0.06	0.18
Mean	0.07	0.05	0.07	0.06
S.D.	0.05	0.08	0.06	0.10

Table A17. Peak plantarflexion moment for each subject during landing (Nm/kg). All joint moments are presented as mean for the non-dominant low-top shoe (NDL), non-dominant mid-cut shoe (NDM), dominant low-top shoe (DL), and dominant mid-cut shoe (DM). The mean and standard deviation (S.D.) are presented across all participants.

Subject	NDL	DL	NDM	DM
S1	-2.36	-2.02	-1.98	-1.75
S2	-1.58	-1.04	-1.54	-1.07
S3	-1.12	-1.18	-1.19	-1.44
S4	-1.65	-1.68	-1.52	-1.49
S5	-1.98	-1.84	-1.83	-1.64
S6	-2.22	-1.87	-1.53	-1.43
S7	-1.52	-1.32	-1.43	-1.48
S8	-1.84	-2.31	-1.66	-2.05
S9	-2.22	-2.69	-1.85	-2.2
S10	-2.13	-2.18	-1.84	-1.97
S11	-1.69	-1.61	-1.48	-1.31
S12	-1.55	-1.41	-1.54	-1.42
S13	-1.51	-1.14	-1.64	-1.53
S14	-1.68	-1.46	-1.72	-1.61
S15	-1.41	-1.42	-1.36	-1.05
S16	-2.09	-1.96	-1.87	-2.2
S17	-1.51	-1.58	-1.02	-1.45
Mean	-1.77	-1.69	-1.59	-1.59
S.D.	0.34	0.45	0.25	0.34

Table A18. Peak dorsiflexion moment for each subject during landing (Nm/kg). All joint moments are presented as mean for the non-dominant low-top shoe (NDL), non-dominant mid-cut shoe (NDM), dominant low-top shoe (DL), and dominant mid-cut shoe (DM). The mean and standard deviation (S.D.) are presented across all participants.

Subject	NDL	DL	NDM	DM
S1	-0.02	0.05	-0.01	0.05
S2	-0.03	0.08	-0.03	0.12
S3	-0.02	0.08	-0.03	0.05
S4	-0.04	-0.02	-0.03	-0.02
S5	-0.03	-0.04	-0.05	-0.01
S6	-0.05	-0.05	-0.03	-0.04
S7	-0.03	0.15	-0.02	-0.04
S8	-0.03	-0.01	-0.03	-0.01
S9	-0.07	-0.02	-0.05	-0.07
S10	-0.04	-0.04	-0.04	-0.03
S11	-0.05	-0.07	-0.04	-0.04
S12	-0.02	-0.01	-0.05	-0.03
S13	-0.04	0.65	-0.05	-0.02
S14	-0.04	-0.04	-0.04	-0.08
S15	-0.02	-0.03	-0.02	-0.03
S16	-0.01	-0.02	-0.01	-0.02
S17	-0.06	-0.07	-0.07	-0.08
Mean	-0.04	0.03	-0.04	-0.02
S.D.	0.02	0.17	0.02	0.05

Table A19. Peak eversion moment for each subject during landing (Nm/kg). All joint moments are presented as mean for the non-dominant low-top shoe (NDL), non-dominant mid-cut shoe (NDM), dominant low-top shoe (DL), and dominant mid-cut shoe (DM). The mean and standard deviation (S.D.) are presented across all participants.

Subject	NDL	DL	NDM	DM
S1	-0.08	-0.08	-0.09	-0.13
S2	-0.15	-0.12	-0.15	-0.13
S3	-0.13	-0.12	-0.08	-0.11
S4	-0.05	-0.14	-0.05	-0.13
S5	-0.16	-0.09	-0.11	-0.06
S6	-0.08	-0.01	-0.08	0.00
S7	-0.13	-0.13	-0.27	-0.27
S8	-0.27	-0.04	-0.26	-0.01
S9	-0.09	-0.13	0.01	-0.05
S10	-0.16	-0.03	-0.24	-0.04
S11	0.01	-0.02	0.00	-0.01
S12	-0.04	-0.10	-0.13	-0.13
S13	-0.09	-0.15	-0.14	-0.14
S14	-0.48	-0.05	-0.48	-0.15
S15	-0.33	-0.04	-0.30	-0.03
S16	-0.06	-0.18	-0.05	-0.33
S17	-0.27	-0.01	-0.26	0.01
Mean	-0.15	-0.08	-0.16	-0.10
S.D.	0.12	0.05	0.13	0.09

Table A20. Peak inversion moment for each subject during landing (Nm/kg). All joint moments are presented as mean for the non-dominant low-top shoe (NDL), non-dominant mid-cut shoe (NDM), dominant low-top shoe (DL), and dominant mid-cut shoe (DM). The mean and standard deviation (S.D.) are presented across all participants.

Subject	NDL	DL	NDM	DM
S1	0.10	0.23	0.06	0.13
S2	0.15	0.11	0.07	0.09
S3	0.15	0.17	0.28	0.19
S4	0.14	0.13	0.18	0.11
S5	0.12	0.17	0.13	0.19
S6	0.00	0.14	0.02	0.18
S7	0.00	0.06	-0.02	0.00
S8	0.00	0.31	0.02	0.31
S9	0.11	0.05	0.31	0.13
S10	0.04	0.14	0.00	0.08
S11	0.41	0.26	0.41	0.26
S12	0.37	0.17	0.13	0.18
S13	0.06	0.13	0.01	0.04
S14	-0.01	0.17	-0.01	0.03
S15	0.00	0.16	-0.02	0.19
S16	0.14	0.06	0.09	0.00
S17	-0.01	0.22	-0.02	0.18
Mean	0.10	0.16	0.10	0.13
S.D.	0.12	0.07	0.13	0.09

Table A21. Maximum mediolateral ground reaction force for each subject during landing (Nm/kg). All ground reaction forces are presented as mean for the non-dominant low-top shoe (NDL), non-dominant mid-cut shoe (NDM), dominant low-top shoe (DL), and dominant mid-cut shoe (DM). The mean and standard deviation (S.D.) are presented across all participants.

Subject	NDL	DL	NDM	DM
S1	0.03	0.07	0.02	0.06
S2	0.01	-0.04	0.01	-0.05
S3	-0.02	0.00	-0.02	0.01
S4	-0.01	-0.01	-0.01	0.00
S5	0.05	0.01	0.06	0.08
S6	0.02	0.02	-0.01	0.06
S7	-0.01	0.01	-0.01	0.00
S8	0.01	0.02	0.00	0.00
S9	0.01	-0.01	-0.01	-0.01
S10	0.04	0.00	0.01	-0.01
S11	0.02	0.07	0.00	0.05
S12	-0.03	-0.02	-0.03	0.01
S13	-0.01	0.00	0.00	0.04
S14	0.03	-0.02	0.01	0.04
S15	0.01	-0.01	0.00	-0.02
S16	0.02	0.01	0.00	0.03
S17	0.02	0.01	0.01	0.00
Mean	0.01	0.01	0.00	0.02
S.D.	0.02	0.03	0.02	0.03

Table A22. Minimum mediolateral ground reaction force for each subject during landing (BW). All ground reaction forces are presented as mean for the non-dominant low-top shoe (NDL), non-dominant mid-cut shoe (NDM), dominant low-top shoe (DL), and dominant mid-cut shoe (DM). The mean and standard deviation (S.D.) are presented across all participants.

Subject	NDL	DL	NDM	DM
S1	-0.08	-0.08	-0.11	-0.08
S2	-0.14	-0.16	-0.12	-0.15
S3	-0.12	-0.12	-0.12	-0.12
S4	-0.12	-0.11	-0.14	-0.11
S5	-0.08	-0.08	-0.09	-0.07
S6	-0.07	-0.07	-0.11	-0.08
S7	-0.12	-0.10	-0.14	-0.12
S8	-0.18	-0.09	-0.20	-0.11
S9	-0.12	-0.16	-0.17	-0.18
S10	-0.13	-0.13	-0.13	-0.14
S11	-0.17	-0.12	-0.17	-0.15
S12	-0.17	-0.13	-0.15	-0.14
S13	-0.12	-0.20	-0.2	-0.12
S14	-0.14	-0.15	-0.19	-0.12
S15	-0.13	-0.14	-0.14	-0.13
S16	-0.09	-0.12	-0.08	-0.14
S17	-0.12	-0.1	-0.10	-0.11
Mean	-0.12	-0.12	-0.14	-0.12
S.D.	0.03	0.03	0.04	0.03

Table A23. Maximum anterior-posterior ground reaction force for each subject during landing (BW). All ground reaction forces are presented as mean for the non-dominant low-top shoe (NDL), non-dominant mid-cut shoe (NDM), dominant low-top shoe (DL), and dominant mid-cut shoe (DM). The mean and standard deviation (S.D.) are presented across all participants.

Subject	NDL	DL	NDM	DM
S1	-0.09	-0.12	-0.10	-0.09
S2	-0.09	-0.10	-0.10	-0.10
S3	-0.07	-0.05	-0.08	-0.08
S4	-0.07	-0.07	-0.09	-0.07
S5	-0.07	-0.07	-0.08	-0.05
S6	-0.08	-0.06	-0.07	-0.07
S7	-0.08	-0.05	-0.09	-0.09
S8	-0.10	-0.09	-0.08	-0.08
S9	-0.07	-0.10	-0.09	-0.11
S10	-0.10	-0.11	-0.12	-0.10
S11	-0.09	-0.10	-0.08	-0.10
S12	-0.08	-0.08	-0.10	-0.08
S13	-0.08	0.03	-0.08	-0.07
S14	-0.07	-0.10	-0.10	-0.08
S15	-0.09	-0.07	-0.07	-0.06
S16	-0.08	-0.09	-0.09	-0.08
S17	-0.09	-0.07	-0.09	-0.07
Mean	-0.08	-0.08	-0.09	-0.08
S.D.	0.01	0.04	0.01	0.02

Table A24. Minimum anterior-posterior ground reaction force for each subject during landing (BW). All ground reaction forces are presented as mean for the non-dominant low-top shoe (NDL), non-dominant mid-cut shoe (NDM), dominant low-top shoe (DL), and dominant mid-cut shoe (DM). The mean and standard deviation (S.D.) are presented across all participants.

Subject	NDL	DL	NDM	DM
S1	-0.59	-0.74	-0.59	-0.63
S2	-0.75	-0.74	-0.69	-0.70
S3	-0.75	-0.70	-0.73	-0.70
S4	-0.69	-0.57	-0.74	-0.60
S5	-0.75	-0.72	-0.76	-0.68
S6	-0.76	-0.68	-0.75	-0.63
S7	-0.70	-0.67	-0.68	-0.72
S8	-0.59	-0.69	-0.69	-0.58
S9	-0.71	-0.74	-0.77	-0.76
S10	-0.64	-0.68	-0.66	-0.65
S11	-0.71	-0.58	-0.74	-0.59
S12	-0.69	-0.70	-0.69	-0.68
S13	-0.79	-0.63	-0.77	-0.84
S14	-0.65	-0.63	-0.69	-0.66
S15	-0.68	-0.7	-0.68	-0.73
S16	-0.67	-0.65	-0.65	-0.69
S17	-0.66	-0.62	-0.69	-0.69
Mean	-0.69	-0.67	-0.70	-0.68
S.D.	0.06	0.05	0.05	0.06

Table A25. Maximum vertical ground reaction force for each subject during landing (BW). All ground reaction forces are presented as mean for the non-dominant low-top shoe (NDL), non-dominant mid-cut shoe (NDM), dominant low-top shoe (DL), and dominant mid-cut shoe (DM). The mean and standard deviation (S.D.) are presented across all participants.

Subject	NDL	DL	NDM	DM
S1	3.27	3.46	3.17	3.31
S2	2.70	2.60	2.64	2.46
S3	2.59	2.50	2.62	2.73
S4	2.40	2.45	2.82	2.34
S5	2.61	2.54	2.72	2.71
S6	2.97	2.54	2.67	2.29
S7	2.21	2.27	2.36	2.43
S8	3.20	3.29	2.94	3.08
S9	3.33	3.37	3.25	3.46
S10	2.93	2.93	2.89	2.88
S11	3.49	3.15	3.36	3.07
S12	2.66	2.53	2.69	2.53
S13	2.63	2.37	2.54	2.77
S14	3.11	3.08	3.27	3.22
S15	2.42	2.44	2.21	2.29
S16	3.00	2.96	2.92	3.00
S17	2.82	2.59	2.63	2.44
Mean	2.84	2.77	2.81	2.77
S.D.	0.36	0.38	0.32	0.37

APPENDIX B –IRB Approval Letter

Office *of*
Research Integrity

118 COLLEGE DRIVE #5116 • HATTIESBURG, MS | 601.266.6756 | WWW.USM.EDU/ORI

NOTICE OF INSTITUTIONAL REVIEW BOARD ACTION

The project below has been reviewed by The University of Southern Mississippi Institutional Review Board in accordance with Federal Drug Administration regulations (21 CFR 26, 111), Department of Health and Human Services regulations (45 CFR Part 46), and University Policy to ensure:

- The risks to subjects are minimized and reasonable in relation to the anticipated benefits.
- The selection of subjects is equitable.
- Informed consent is adequate and appropriately documented.
- Where appropriate, the research plan makes adequate provisions for monitoring the data collected to ensure the safety of the subjects.
- Where appropriate, there are adequate provisions to protect the privacy of subjects and to maintain the confidentiality of all data.
- Appropriate additional safeguards have been included to protect vulnerable subjects.
- Any unanticipated, serious, or continuing problems encountered involving risks to subjects must be reported immediately. Problems should be reported to ORI via the Incident submission on InfoEd IRB.
- The period of approval is twelve months. An application for renewal must be submitted for projects exceeding twelve months.

PROTOCOL NUMBER:	22-923
PROJECT TITLE:	Effects of Increased Shoe Collar Height and Limb Dominance on Landing Knee Biomechanics in Collegiate Volleyball Players
SCHOOL/PROGRAM	Kinesiology
RESEARCHERS:	PI: Lindsey Legg Investigators: Legg, Lindsey~Thorsen, Tanner~
IRB COMMITTEE ACTION:	Approved
CATEGORY:	Expedited Category
PERIOD OF APPROVAL:	15-Jun-2022 to 14-Jun-2023

Donald Sacco

Donald Sacco, Ph.D.
Institutional Review Board Chairperson

APPENDIX C –Participant Screening Form

Screening Form:

□ By checking this box, I certify that I am 18 years of age or older
□ Free of musculoskeletal injury for the past 6 months
□ Abstained from alcohol and vigorous activity in the past 12 hours
□ Completed the PAR-Q

Which foot would I use to kick a volleyball with?

- □ Right
- □ Left

What is my primary volleyball position?

- □ Setter
- □ Libero/DS
- □ Middle
- □ Outside
- □ Right Side
- □ Other

Signature of Participant: ____
Date: ______________

Signature of Researcher: ____
Date: ______________

S#__________

APPENDIX D –Informed Consent

ORI | Office of Research Integrity

INSTITUTIONAL REVIEW BOARD

STANDARD (SIGNED) INFORMED CONSENT

STANDARD (SIGNED) INFORMED CONSENT PROCEDURES
• ***Use of this template is <u>optional</u>.*** However, by federal regulations (45 CFR 46.116), all consent documentation must address each of the required elements listed below (purpose, procedures, duration, benefits, risks, alternative procedures, confidentiality, whom to contact in case of injury, and a statement that participation is voluntary). • Signed copies of the consent form should be provided to all participants. Last Edited May 18th, 2022

<table>
<tr><td colspan="3">Today's date:05/31/2022</td></tr>
<tr><th colspan="3">PROJECT INFORMATION</th></tr>
<tr><td colspan="3">Project Title: Effects of Increased Shoe Collar Height and Limb Dominance on Landing Knee Biomechanics in Collegiate Volleyball Players</td></tr>
<tr><td colspan="3">Protocol Number: 22-923</td></tr>
<tr><td>Principal Investigator: Lindsey Legg</td><td>Phone: (850)543-7686</td><td>Email: Lindsey.Legg@usm.edu</td></tr>
<tr><td>College: Education and Human Sciences</td><td colspan="2">School and Program: School of Kinesiology and Nutrition</td></tr>
<tr><th colspan="3">RESEARCH DESCRIPTION</th></tr>
<tr><td colspan="3">1. Purpose:
The purpose of this study is to determine the effects of midcut shoes on the knee joint biomechanics during a unilateral drop landing task in female collegiate volleyball players. In addition, we aimed to determine the influence of leg dominance on knee joint biomechanics during landing while wearing midcut shoes.

2. Description of Study:</td></tr>
</table>

I will be asked to complete 1 visit that will be less than 1.5 hours long. Twenty-four hours prior to the visit, I will be asked to sustain from any vigorous activity or alcohol. Upon arrival, I will complete the PAR-Q questionnaire. Once cleared, I will be familiarized to the informed consent document and provide my consent. After providing consent, verbal instructions will be given to ensure that I understand the tasks within the study. My height will be measured using a stadiometer and weight will be measured while standing on a force plate. After a warmup consisting of a 5-minute run at a self-selected pace on the treadmill and a familiarization period, I will have 8 electromyography electrodes, 22 anatomic markers, and 8 segmental markers, placed onto the muscles of interest and anatomical locations. The electrodes will be placed bilaterally on medial head of the gastrocnemius, vastus medialis, vastus lateralis, and biceps femoris. Electrode placement will occur according to the standards set for each muscle by the SENIAM project (Surface Electromyography for the Non-Invasive Assessment of Muscles). The anatomic markers were placed bilaterally on the acromion process, iliac crest, greater trochanter, medial and lateral femoral epicondyles, medial and lateral malleoli, distal end of the second toe, and the first and fifth metatarsal heads. The segmental markers were placed on the trunk and pelvis, and bilaterally on the thighs, shanks, and heel. I will stand still on the force plate for a static trial and then have the anatomic markers removed. Shoe condition and limb order was then randomized. The forward drop landing task is a unilateral task comprised of jumping forward off one foot and landing on my opposite foot followed by maintaining balance for three seconds. I will perform five successful trials of the forward drop landing task on both limbs in one shoe condition. Then, three minutes of rest will be given before I repeat the same procedures with the other shoe condition. Ground reaction force, reflective marker, and electromyography recordings will be collected during landing. The totally of the landing tasks will take no more than 15 minutes.

3. **Benefits:**
   I will likely receive no direct benefit from participating in this research.

4. **Risks:**
   Adverse events of this research study are listed with the following:

"Rare" – occur less than 1% or less than 1 in 100 people.

Injury risk is rare. The possible risk for injury would be due to landing wrong. The landing is occurring from jumping off a stationary box which has a small incidence of incorrect landing. The laboratory staff will stop any falling because of a mis-landing. Muscle strains, muscle tears, ankle sprains, and sudden movement-related injuries are other potential injury risks while engaging in the study.

I understand that I am not waiving any legal rights or releasing the institution or their agents from liability from negligence. I understand that in the event of injury resulting from the research procedures, The University of Southern Mississippi does not have the funds budgeted for compensation for 1) lost wages, 2) medical treatment, or 3) reimbursement for such injuries. The University will help, however, obtain medical attention which I may require while involved in the study by securing transportation to the nearest medical facility.

5. **Confidentiality:**
   Any information from this research study will be kept as confidential as possible. Information produced by this study will be stored in the researcher's file and de-identified by a subject number only. The documents connecting my name to specific information about myself will be kept in a separate, secure file cabinet outside of the lab. Information contained in my records may not be given to anyone unaffiliated with the study in a form that could identify me without my written consent, except as required by law. The results of the study may be published in a medical book or journal. However, my name will not be used in any publication without my permission.

6. **Alternative Procedures:**
   There are no alternative procedures.

7. **Participant's Assurance:**
   This project and this consent form have been reviewed by USM's Institutional Review Board, which ensures that research projects involving human subjects follow federal regulations. Any questions or concerns about rights as a research participant should be

directed to the Chair of the Institutional Review Board, The University of Southern Mississippi, 118 College Drive #5125, Hattiesburg, MS 39406-0001, 601-266-5997.

Any questions about this research project should be directed to the Principal Investigator using the contact information provided above.

CONSENT TO PARTICIPATE IN RESEARCH

Participant's Name: ____________________________

I hereby consent to participate in this research project. All research procedures and their purpose were explained to me, and I had the opportunity to ask questions about both the procedures and their purpose. I received information about all expected benefits, risks, inconveniences, or discomforts, and I had the opportunity to ask questions about them. I understand my participation in the project is completely voluntary and that I may withdraw from the project at any time without penalty, prejudice, or loss of benefits. I understand the extent to which my personal information will be kept confidential. As the research proceeds, I understand that any new information that emerges and that might be relevant to my willingness to continue my participation will be provided to me.

(Include the following information only if applicable. Otherwise delete this entire paragraph before submitting for IRB approval:) The University of Southern Mississippi has no mechanism to provide compensation for participants who may incur injuries as a result of participation in research projects. However, efforts will be made to make available the facilities and professional skills at the University. Participants may incur charges as a result of treatment related to research injuries. Information regarding treatment or the absence of treatment has been given above.

____

____

Research Participant

Person Explaining the Study

____

____

Date

Date

APPENDIX E –PAR-Q

2021 PAR-Q+

The Physical Activity Readiness Questionnaire for Everyone

The health benefits of regular physical activity are clear; more people should engage in physical activity every day of the week. Participating in physical activity is very safe for MOST people. This questionnaire will tell you whether it is necessary for you to seek further advice from your doctor OR a qualified exercise professional before becoming more physically active.

GENERAL HEALTH QUESTIONS

Please read the 7 questions below carefully and answer each one honestly: check YES or NO.	YES	NO
1) Has your doctor ever said that you have a heart condition ☐ **OR** high blood pressure ☐?	☐	☐
2) Do you feel pain in your chest at rest, during your daily activities of living, **OR** when you do physical activity?	☐	☐
3) Do you lose balance because of dizziness **OR** have you lost consciousness in the last 12 months? Please answer **NO** if your dizziness was associated with over-breathing (including during vigorous exercise).	☐	☐
4) Have you ever been diagnosed with another chronic medical condition (other than heart disease or high blood pressure)? **PLEASE LIST CONDITION(S) HERE:** ______	☐	☐
5) Are you currently taking prescribed medications for a chronic medical condition? **PLEASE LIST CONDITION(S) AND MEDICATIONS HERE:** ______	☐	☐
6) Do you currently have (or have had within the past 12 months) a bone, joint, or soft tissue (muscle, ligament, or tendon) problem that could be made worse by becoming more physically active? Please answer **NO** if you had a problem in the past, but it **does not limit your current ability** to be physically active. **PLEASE LIST CONDITION(S) HERE:** ______	☐	☐
7) Has your doctor ever said that you should only do medically supervised physical activity?	☐	☐

If you answered NO to all of the questions above, you are cleared for physical activity.
Please sign the PARTICIPANT DECLARATION. You do not need to complete Pages 2 and 3.

- Start becoming much more physically active – start slowly and build up gradually.
- Follow Global Physical Activity Guidelines for your age (https://www.who.int/publications/i/item/9789240015128).
- You may take part in a health and fitness appraisal.
- If you are over the age of 45 yr and NOT accustomed to regular vigorous to maximal effort exercise, consult a qualified exercise professional before engaging in this intensity of exercise.
- If you have any further questions, contact a qualified exercise professional.

PARTICIPANT DECLARATION
If you are less than the legal age required for consent or require the assent of a care provider, your parent, guardian or care provider must also sign this form.

I, the undersigned, have read, understood to my full satisfaction and completed this questionnaire. I acknowledge that this physical activity clearance is valid for a maximum of 12 months from the date it is completed and becomes invalid if my condition changes. I also acknowledge that the community/fitness center may retain a copy of this form for its records. In these instances, it will maintain the confidentiality of the same, complying with applicable law.

NAME ______________________ DATE ______________

SIGNATURE ______________________ WITNESS ______________

SIGNATURE OF PARENT/GUARDIAN/CARE PROVIDER ______________________

If you answered YES to one or more of the questions above, COMPLETE PAGES 2 AND 3.

Delay becoming more active if:

- You have a temporary illness such as a cold or fever; it is best to wait until you feel better.
- You are pregnant - talk to your health care practitioner, your physician, a qualified exercise professional, and/or complete the ePARmed-X+ at www.eparmeds.com before becoming more physically active.
- Your health changes - answer the questions on Pages 2 and 3 of this document and/or talk to your doctor or a qualified exercise professional before continuing with any physical activity program.

 1 / 4
01-11-2020

2021 PAR-Q+

FOLLOW-UP QUESTIONS ABOUT YOUR MEDICAL CONDITION(S)

1. Do you have Arthritis, Osteoporosis, or Back Problems?

If the above condition(s) is/are present, answer questions 1a-1c — If **NO** ☐ go to question 2

1a.	Do you have difficulty controlling your condition with medications or other physician-prescribed therapies? (Answer **NO** if you are not currently taking medications or other treatments)	YES ☐ NO ☐
1b.	Do you have joint problems causing pain, a recent fracture or fracture caused by osteoporosis or cancer, displaced vertebra (e.g., spondylolisthesis), and/or spondylolysis/pars defect (a crack in the bony ring on the back of the spinal column)?	YES ☐ NO ☐
1c.	Have you had steroid injections or taken steroid tablets regularly for more than 3 months?	YES ☐ NO ☐

2. Do you currently have Cancer of any kind?

If the above condition(s) is/are present, answer questions 2a-2b — If **NO** ☐ go to question 3

2a.	Does your cancer diagnosis include any of the following types: lung/bronchogenic, multiple myeloma (cancer of plasma cells), head, and/or neck?	YES ☐ NO ☐
2b.	Are you currently receiving cancer therapy (such as chemotheraphy or radiotherapy)?	YES ☐ NO ☐

3. Do you have a Heart or Cardiovascular Condition? This includes Coronary Artery Disease, Heart Failure, Diagnosed Abnormality of Heart Rhythm

If the above condition(s) is/are present, answer questions 3a-3d — If **NO** ☐ go to question 4

3a.	Do you have difficulty controlling your condition with medications or other physician-prescribed therapies? (Answer **NO** if you are not currently taking medications or other treatments)	YES ☐ NO ☐
3b.	Do you have an irregular heart beat that requires medical management? (e.g., atrial fibrillation, premature ventricular contraction)	YES ☐ NO ☐
3c.	Do you have chronic heart failure?	YES ☐ NO ☐
3d.	Do you have diagnosed coronary artery (cardiovascular) disease and have not participated in regular physical activity in the last 2 months?	YES ☐ NO ☐

4. Do you currently have High Blood Pressure?

If the above condition(s) is/are present, answer questions 4a-4b — If **NO** ☐ go to question 5

4a.	Do you have difficulty controlling your condition with medications or other physician-prescribed therapies? (Answer **NO** if you are not currently taking medications or other treatments)	YES ☐ NO ☐
4b.	Do you have a resting blood pressure equal to or greater than 160/90 mmHg with or without medication? (Answer **YES** if you do not know your resting blood pressure)	YES ☐ NO ☐

5. Do you have any Metabolic Conditions? This includes Type 1 Diabetes, Type 2 Diabetes, Pre-Diabetes

If the above condition(s) is/are present, answer questions 5a-5e — If **NO** ☐ go to question 6

5a.	Do you often have difficulty controlling your blood sugar levels with foods, medications, or other physician-prescribed therapies?	YES ☐ NO ☐
5b.	Do you often suffer from signs and symptoms of low blood sugar (hypoglycemia) following exercise and/or during activities of daily living? Signs of hypoglycemia may include shakiness, nervousness, unusual irritability, abnormal sweating, dizziness or light-headedness, mental confusion, difficulty speaking, weakness, or sleepiness.	YES ☐ NO ☐
5c.	Do you have any signs or symptoms of diabetes complications such as heart or vascular disease and/or complications affecting your eyes, kidneys, **OR** the sensation in your toes and feet?	YES ☐ NO ☐
5d.	Do you have other metabolic conditions (such as current pregnancy-related diabetes, chronic kidney disease, or liver problems)?	YES ☐ NO ☐
5e.	Are you planning to engage in what for you is unusually high (or vigorous) intensity exercise in the near future?	YES ☐ NO ☐

2021 PAR-Q+

6.	**Do you have any Mental Health Problems or Learning Difficulties?** This includes Alzheimer's, Dementia, Depression, Anxiety Disorder, Eating Disorder, Psychotic Disorder, Intellectual Disability, Down Syndrome	
	If the above condition(s) is/are present, answer questions 6a-6b If **NO** ☐ go to question 7	
6a.	Do you have difficulty controlling your condition with medications or other physician-prescribed therapies? (Answer **NO** if you are not currently taking medications or other treatments)	YES ☐ NO ☐
6b.	Do you have Down Syndrome **AND** back problems affecting nerves or muscles?	YES ☐ NO ☐
7.	**Do you have a Respiratory Disease?** *This includes Chronic Obstructive Pulmonary Disease, Asthma, Pulmonary High Blood Pressure*	
	If the above condition(s) is/are present, answer questions 7a-7d If **NO** ☐ go to question 8	
7a.	Do you have difficulty controlling your condition with medications or other physician-prescribed therapies? (Answer **NO** if you are not currently taking medications or other treatments)	YES ☐ NO ☐
7b.	Has your doctor ever said your blood oxygen level is low at rest or during exercise and/or that you require supplemental oxygen therapy?	YES ☐ NO ☐
7c.	If asthmatic, do you currently have symptoms of chest tightness, wheezing, laboured breathing, consistent cough (more than 2 days/week), or have you used your rescue medication more than twice in the last week?	YES ☐ NO ☐
7d.	Has your doctor ever said you have high blood pressure in the blood vessels of your lungs?	YES ☐ NO ☐
8.	**Do you have a Spinal Cord Injury?** *This includes Tetraplegia and Paraplegia*	
	If the above condition(s) is/are present, answer questions 8a-8c If **NO** ☐ go to question 9	
8a.	Do you have difficulty controlling your condition with medications or other physician-prescribed therapies? (Answer **NO** if you are not currently taking medications or other treatments)	YES ☐ NO ☐
8b.	Do you commonly exhibit low resting blood pressure significant enough to cause dizziness, light-headedness, and/or fainting?	YES ☐ NO ☐
8c.	Has your physician indicated that you exhibit sudden bouts of high blood pressure (known as Autonomic Dysreflexia)?	YES ☐ NO ☐
9.	**Have you had a Stroke?** *This includes Transient Ischemic Attack (TIA) or Cerebrovascular Event*	
	If the above condition(s) is/are present, answer questions 9a-9c If **NO** ☐ go to question 10	
9a.	Do you have difficulty controlling your condition with medications or other physician-prescribed therapies? (Answer **NO** if you are not currently taking medications or other treatments)	YES ☐ NO ☐
9b.	Do you have any impairment in walking or mobility?	YES ☐ NO ☐
9c.	Have you experienced a stroke or impairment in nerves or muscles in the past 6 months?	YES ☐ NO ☐
10.	**Do you have any other medical condition not listed above or do you have two or more medical conditions?**	
	If you have other medical conditions, answer questions 10a-10c If **NO** ☐ read the Page 4 recommendations	
10a.	Have you experienced a blackout, fainted, or lost consciousness as a result of a head injury within the last 12 months **OR** have you had a diagnosed concussion within the last 12 months?	YES ☐ NO ☐
10b.	Do you have a medical condition that is not listed (such as epilepsy, neurological conditions, kidney problems)?	YES ☐ NO ☐
10c.	Do you currently live with two or more medical conditions?	YES ☐ NO ☐

PLEASE LIST YOUR MEDICAL CONDITION(S) AND ANY RELATED MEDICATIONS HERE: ______________________

____

GO to Page 4 for recommendations about your current medical condition(s) and sign the PARTICIPANT DECLARATION.

01-11-2020

2021 PAR-Q+

If you answered NO to all of the FOLLOW-UP questions (pgs. 2-3) about your medical condition, you are ready to become more physically active - sign the PARTICIPANT DECLARATION below:

- It is advised that you consult a qualified exercise professional to help you develop a safe and effective physical activity plan to meet your health needs.
- You are encouraged to start slowly and build up gradually - 20 to 60 minutes of low to moderate intensity exercise, 3-5 days per week including aerobic and muscle strengthening exercises.
- As you progress, you should aim to accumulate 150 minutes or more of moderate intensity physical activity per week.
- If you are over the age of 45 yr and **NOT** accustomed to regular vigorous to maximal effort exercise, consult a qualified exercise professional before engaging in this intensity of exercise.

If you answered **YES** to **one or more of the follow-up questions** about your medical condition:
You should seek further information before becoming more physically active or engaging in a fitness appraisal. You should complete the specially designed online screening and exercise recommendations program - the **ePARmed-X+ at www.eparmedx.com** and/or visit a qualified exercise professional to work through the ePARmed-X+ and for further information.

Delay becoming more active if:

- You have a temporary illness such as a cold or fever; it is best to wait until you feel better.
- You are pregnant - talk to your health care practitioner, your physician, a qualified exercise professional, and/or complete the ePARmed-X+ **at www.eparmedx.com** before becoming more physically active.
- Your health changes - talk to your doctor or qualified exercise professional before continuing with any physical activity program.

- You are encouraged to photocopy the PAR-Q+. You must use the entire questionnaire and NO changes are permitted.
- The authors, the PAR-Q+ Collaboration, partner organizations, and their agents assume no liability for persons who undertake physical activity and/or make use of the PAR-Q+ or ePARmed-X+. If in doubt after completing the questionnaire, consult your doctor prior to physical activity.

PARTICIPANT DECLARATION

- All persons who have completed the PAR-Q+ please read and sign the declaration below.
- If you are less than the legal age required for consent or require the assent of a care provider, your parent, guardian or care provider must also sign this form.

I, the undersigned, have read, understood to my full satisfaction and completed this questionnaire. I acknowledge that this physical activity clearance is valid for a maximum of 12 months from the date it is completed and becomes invalid if my condition changes. I also acknowledge that the community/fitness center may retain a copy of this form for records. In these instances, it will maintain the confidentiality of the same, complying with applicable law.

NAME ______________________ DATE ______________________

SIGNATURE ______________________ WITNESS ______________________

SIGNATURE OF PARENT/GUARDIAN/CARE PROVIDER ______________________

For more information, please contact
www.eparmedx.com
Email: eparmedx@gmail.com

Citation for PAR-Q+
Warburton DER, Jamnik VK, Bredin SSD, and Gledhill N on behalf of the PAR-Q+ Collaboration. The Physical Activity Readiness Questionnaire for Everyone (PAR-Q+) and Electronic Physical Activity Readiness Medical Examination (ePARmed-X+). Health & Fitness Journal of Canada 4(2):3-23, 2011.

The PAR-Q+ was created using the evidence-based AGREE process (1) by the PAR-Q+ Collaboration chaired by Dr. Darren E. R. Warburton with Dr. Norman Gledhill, Dr. Veronica Jamnik, and Dr. Donald C. McKenzie (2). Production of this document has been made possible through financial contributions from the Public Health Agency of Canada and the BC Ministry of Health Services. The views expressed herein do not necessarily represent the views of the Public Health Agency of Canada or the BC Ministry of Health Services.

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