THESIS

THE ROLE OF LANDING FOOT ORIENTATION ON LINEAR TRACTION IN STOP AND STOP-JUMP TASKS

Submitted by

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ABSTRACT

THE ROLE OF LANDING FOOT ORIENTATION ON LINEAR TRACTION IN STOP AND STOP-JUMP TASKS

Introduction: The incidence of lower extremity injury has been shown to be greater on artificial turf (AT) than on natural grass across a variety of sports. Injury risk and performance are influenced by the traction characteristics of the foot-surface interface shortly after initial foot contact. The foot's orientation relative to the ground upon landing potentially contributes to these traction characteristics. Although landing foot orientation has been shown to be predictive of lower extremity injury risk on hardcourt surfaces, it remains unclear if foot orientation influences landing ground reaction forces and traction on AT. This information could contribute to modifications in athlete technique, cleat design, and surface characteristics to optimize athlete performance and reduce injury risk. The primary purpose of this investigation was to examine how foot orientation upon landing on AT during stop and stop-jump tasks influences linear traction and foot loading characteristics. Secondary goals were to investigate differences in landing strategy between males and females and the effect of subsequent task demands between the two movements.

Methods: Twenty-nine collegiate club-level or higher athletes (15 females) accustomed to competing on AT participated. A third-generation AT was prepared over a foam shock pad to manufacturer specifications with a sand base and crumb rubber performance infill. Isolated panels were secured over two side-by-side force platforms. Subject kinematics were measured using optical capture with reflective markers. Subjects performed six acceptable trials of a stop

task and a stop-jump task. Each limb was analyzed separately from initial foot contact through the landing phase. The representative average trial of each subject was used to determine differences between the limbs and sexes within and between each movement. Individual trials were used to explore the relationships between the initial foot progression angle and traction. Due to the limited number of forefoot landings for the two analyzed movements, correlations were only performed on the initial foot progression angles ranging from rearfoot to flatfoot.

Results: This investigation is especially novel since most reported literature on foot orientation has been conducted on hardcourt surfaces, not on AT. We found that initial foot progression angle was strongly correlated with the horizontal displacement of the foot before the cleat fully engaged with the AT but had limited influence on early ground reaction forces. We found no differences in initial foot progression angle between sexes or between movements, although horizontal ground reaction forces were greater for males than females and greater for the stop task compared to the stop-jump task.

Conclusion: Landing foot orientations, ranging from rearfoot to flatfoot, contribute to the horizontal movement across AT. The relationship between horizontal foot movement on AT and injury risk needs to be further analyzed, specifically by examining the joint loading mechanics at the ankle, knee, and hip.

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DEDICATION

For Mom-Mom,

sometimes just following one’s nose does work out.

TABLE OF CONTENTS

LIST OF TABLES

LIST OF FIGURES

CHAPTER 1
INTRODUCTION

Installations of third-generation artificial turf (AT) continue to rise in the United States, despite evidence that the incidence of lower extremity injury risk is greater on AT than on natural grass (NG) (Dragoo & Braun, 2010; Lanzetti et al., 2017; Mack et al., 2019; Poulos et al., 2014; Williams et al., 2016). Athletes across a range of sports have expressed concerns that playing on AT increases their risk of non-contact injury (Poulos et al., 2014). A recent analysis of per play injury rate occurrence during National Football League games showed an increased risk of lower extremity injury of sixteen percent on AT compared with NG (Mack et al., 2019). A significant increase in overuse-type injuries, which are more dependent upon the surface than traumatic injuries are, has also been found for professional rugby players on AT (Lanzetti et al., 2017). In addition, both professional soccer and rugby players have reported greater muscle and joint soreness and longer recovery times following competitions played on AT compared with NG (Poulos et al., 2014; Williams et al., 2016).

Injury risk and performance are influenced by the traction characteristics of the foot-surface interface. High traction is required for maximum performance, particularly to accelerate and change direction quickly, however, too much traction is associated with injury risk (Balazs, 2015; Villwock et al., 2009). "Footlock" is the phenomenon when the foot becomes stuck in the playing surface, which can ultimately lead to injury (Bowers & Martin, 1975; Cawley et al., 2003). The increased braking and acceleration that many AT surfaces allow, compared with NG, contribute to footlock (Fujitaka et al., 2017). Low traction, or slipping, is also associated with injury risk; however, a degree of traction release is beneficial to minimize footlock and to decrease potential loads on the lower extremity (Cawley et al., 2003; Dixon et al., 1999). This poses a unique problem; AT surfaces must be designed so that athletes can achieve high enough traction to maximize performance in acceleration and changes in direction, but not so high as to increase risk of lower extremity injury.

During athletic movements, injury is most likely to occur shortly after foot strike during the landing phase rather than during the propulsive, or take-off, phase (Dixon et al., 1999; Frederick, 1986; Nigg et al., 1987). On hardcourt, the peak ground reaction force (GRF) typically occurs shortly after foot strike and is associated with high decelerations of the lower extremity as the foot contacts the surface (Chappell et al., 2002). Higher GRFs have been correlated with acute lower extremity injuries, such as fractures and ligamentous injury, as well as chronic injuries such as stress fractures, tendonitis, and damage to articular cartilage (Dixon et al., 1999; Frederick, 1986; Mack et al., 2019; Nigg et al., 1987; Zhang et al., 2000). The increased braking and deceleration that many AT surfaces allow compared with NG may contribute to higher GRFs and the higher incidence of injury on AT compared with NG during the landing phase (Balazs et al., 2015; Ferrandino et al., 2015; Fujitaka et al., 2017). Arguably, these initial GRFs on AT may be more similar to those achieved on hardcourt than on NG due to the surface characteristics of AT.

An athlete's interaction with a playing surface must be considered in addition to the material properties of the surface and their effects on performance and injury risk. On hardcourt surfaces, the orientation of the foot upon landing has been shown to be predictive of lower extremity injury risk (Donnelly et al., 2017; Kristianslund et al., 2014). Various foot orientations upon landing, also called initial foot progression angles, are shown in Figure 1.1. Athletes who landed with a flatfoot or a rearfoot pattern were more likely to suffer an anterior cruciate ligament (ACL) rupture compared with athletes who landed on the forefoot while performing similar game-day movements across a variety of sports on hardcourt (Boden et al., 2009, 2010). It is likely that similar landing patterns would also increase risk of lower extremity injury if performed on AT. When landing with a negative foot progression angle, a rearfoot landing, the lower leg and foot segments act as one rigid segment instead of like an accordion and absorbing impact forces (Boden et al., 2010). Because there is significantly less time for the triceps surae musculature to contract and absorb forces, the impulsive force to the knee is increased. Since a landing with a positive foot progression angle, a forefoot landing, gives the soft tissue components of the leg time to dampen the force, landing on the forefoot could be a critical factor in preventing knee injury (Boden et al., 2010). While a negative foot progression angle upon landing most likely places a larger mechanical demand on the knee joint compared with other foot orientations, it remains unclear if the athlete's landing foot orientation influences their peak GRF (Donnelly et al., 2017).

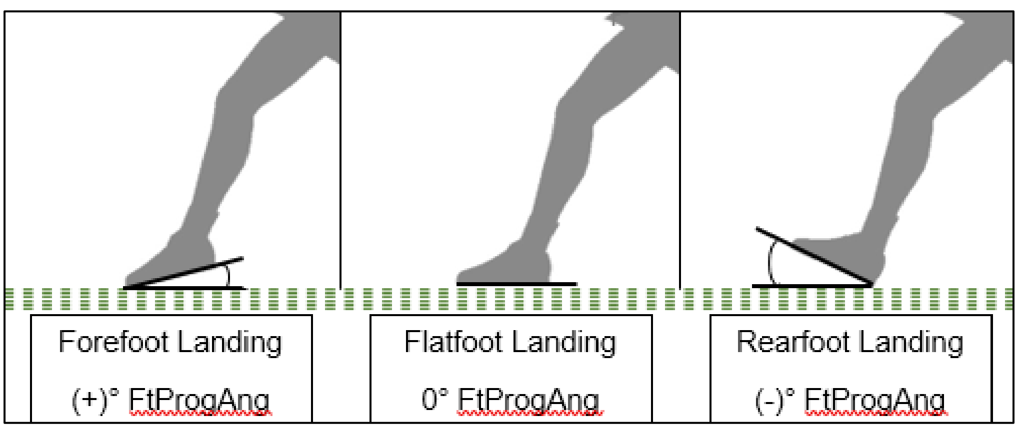

Figure 1.1: Foot Progression Angles. *Range of foot progression angles (FtProgAng) upon impact. A positive angle is indicative of a forefoot landing, zero degrees is a flatfoot landing, and a negative angle is indicative of a rearfoot landing.*

Furthermore, foot orientation upon landing potentially contributes to the traction characteristics during the landing phase. While studs on cleated footwear can be aligned to enhance penetration into the AT to optimize foot engagement, it remains unclear exactly how the foot orientation itself influences traction upon landing (Kirk et al., 2007). It is possible that a rearfoot landing may result in less initial traction because it will take longer for the entire cleat to achieve foot engagement fully compared with a more flatfooted landing which reaches full foot engagement more quickly.

Jumping and landing movements are integral to many sports and are commonly studied in laboratory settings (Zhang et al., 2000). Stop and stop-jump tasks are two such high-impact movements that have been associated with acute and overuse injuries (Zhang et al., 2000). Comparing these two movements to investigate linear traction is ideal because the stop task has been shown to produce greater peak horizontal GRFs than the stop-jump (Hass et al., 2003). Additionally, while Hass et al. (2003) did not report foot orientation upon landing, it appeared that the foot progression angles could be different between the stop and stop-jump tasks. The reported difference in the average ankle angle between the two movements was 12.9 degrees, although high variability in the stop task prevented this difference from being statistically significant (Hass et al., 2003). The preparation for the subsequent task demands of the stop-jump compared with the simpler stop task could be one influencing factor of the greater forces and the larger variability of the ankle angle upon landing for the stop task.

Landing mechanics also differ between sexes (Butler et al., 2013). Compared with males, during the landing phase of various stop-jump tasks, females exhibited greater knee extension, greater valgus moments, and greater tibial plateau anterior shear forces, putting the knee in a more compromised position and at greater risk for ACL injury (Chappell et al., 2002; Grund et al., 2007; Grund & Senner, 2006; Yin et al., 2015). Moreover, analytic videotape reviews of athletes who suffered ACL injury showed that female athletes were primarily injured during a simple deceleration maneuver, whereas male athletes were more frequently injured during more strenuous jumping tasks (Boden et al., 2009). If initial GRFs are different between these tasks, this would support the idea that ACL injury occurs at different GRFs between males and females. Differences between males and females are therefore expected between a simple stop task and the more strenuous stop-jump task. While many landing mechanics differ between males and females, it is unclear if these differences include foot orientation. As indicated above, it is plausible that differences in foot orientation upon landing influence traction and could contribute to injury risk.

Therefore, the primary purpose of this investigation was to examine how foot orientation upon landing on AT during stop and stop-jump tasks influences linear traction and foot loading characteristics. Secondary goals were to investigate differences in landing strategy between males and females and the effect of subsequent task demands between a stop task and a stop-jump task. We hypothesized that a flatfoot orientation upon landing will result in greater traction and loading forces compared with a rearfoot landing pattern. Additionally, we hypothesized that females will land with a more rearfoot orientation than males during stop and stop-jump tasks. Lastly, we hypothesized that the difference in subsequent physical demands between the stop-jump task compared with the stop task will result in greater rearfoot orientation, peak GRFs, and traction characteristics during the initial landing of the stop task. This information could contribute to modifications in athlete technique, cleat design, and surface characteristics to optimize athlete performance and reduce injury risk on AT.

1.1 Specific aims and hypotheses

<u>Aim 1:</u> To determine if the foot orientation upon landing during bilateral stop and stop-jump tasks influences peak GRFs and traction.

<u>Hypothesis 1</u>: A flatfoot orientation upon landing will result in greater traction and loading forces compared with an initial rearfoot landing.

Aim 2: To determine if males and females employ different foot orientations at foot contact during stop and stop-jump tasks.

Hypothesis 2: Females will land in a greater rearfoot orientation than males.

Aim 3: To determine if the subsequent demands of the stop-jump task compared with the stop task influence initial landing foot orientation, peak GRFs, and traction.

Hypothesis 3: The subsequent physical demands of the stop-jump compared with the stop task will result in greater rearfoot orientation, peak GRFs and traction characteristics for the stop task.

CHAPTER 2
REVIEW OF THE LITERATURE

There has been a significant rise in the installations of third-generation artificial turf (AT) surfaces in the United States despite evidence that natural grass (NG) may be superior to AT as it pertains to injury risk (Dragoo & Braun, 2010; Lanzetti et al., 2017; Mack et al., 2019; Poulos et al., 2014; Williams et al., 2016). As such, AT has been the source of considerable controversy among athletes, coaches, athletic trainers, and medical personnel at all levels of competition across a range of sports (Balazs, 2015; Soligard et al., 2012). Additionally, many athletes perceive that playing on AT increases their risk of non-contact injury more so than on NG (Poulos et al., 2014). Understanding how the athlete first interacts with the surface is essential to uncover how the injury risk to the athlete may be surface-dependent.

2.1 Artificial Turf Fields and Injury Risk:

In the United States, 12,500 AT fields are currently in use with approximately 1,400 new installations annually (Synthetic Turf Council, 2020). AT installments are chosen over NG in response to the increased demand for durable fields that can support high usage from multiple sports over many seasons, to provide more consistent field conditions compared with NG, and to have more favorable long-term cost profiles (Balazs et al., 2015; Sharma et al., 2016).

Current third-generation AT surfaces are comprised of a compacted sub-base layer, an optional shock pad layer, and the pile, or bladed carpet layer, as shown in Figure 2.1. Within the carpet layer, there is an optional stabilizing infill of sand followed by a performance infill, typically composed of recycled SBR rubber, also known as crumb rubber (Forrester & Tsui, 2014). The individual components of AT can be modified to affect the performance characteristics of the surface and injury risk (Alcántara et al., 2009; Severn et al., 2011).

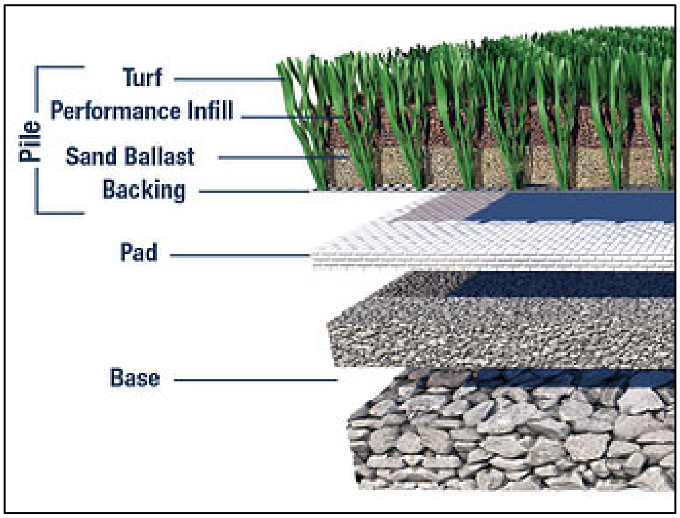

Figure 2.1 Components of the Artificial Turf Field System *(USGreentech, 2021). The foundational base layer is topped by an optional foam shock pad which is topped by the bladed carpet layer infilled with an optional stabilizing layer of sand and the performance infill.*

The injury risk on AT may in fact vary depending on the sport played. In addition, the multiple real-world confounding factors such as the temperature of the air and field, the field moisture level, the maintenance quality of the field, the age of the field, the athletes' footwear, and encounters with other players could all potentially affect the outcome of injury rates on AT.

A recent analysis of per play injury rate occurrence during National Football League games showed an increased risk of lower extremity injury of sixteen percent on AT compared with NG (Mack et al., 2019). Specifically, for football players, the anterior cruciate ligament (ACL) is especially at increased risk for injury when competing on AT (Hershman et al., 2012; Loughran et al., 2019). At the collegiate football level, Dragoo and Braun (2010) reported a 43% elevated risk in ACL injuries on AT than on NG. 40% of these ACL injuries occurred by a non-contact mechanism and primarily occurred on offensive passing and running plays (Dragoo & Braun, 2010).

A review of Major League Soccer injuries revealed that significantly more ankle injuries including sprains, fractures, and Achille's tendon injuries occur on AT than on NG (Calloway et al., 2019; Ekstrand et al., 2006). Other studies, however, have found no difference in injury rate between the two surfaces for male or female professional, collegiate, or youth soccer players (Aoki et al., 2010; Bianco et al., 2016; Bjørneboe et al., 2010; Ekstrand et al., 2011; Fuller et al., 2007; Kristenson et al., 2014; Lanzetti et al., 2017; Soligard et al., 2012; Steffen et al., 2007). Some authors have even reported a lower incidence and severity rate of injuries to the pelvis, hip, and knee on AT than on NG for both male and female collegiate soccer players (Balazs, 2015; Meyers, 2013, 2017).

Interestingly, both professional soccer and rugby players have reported greater muscle and joint soreness and longer recovery times following competitions on AT than on NG (Poulos et al., 2014; Williams et al., 2016). However, for elite rugby players, there is no clear difference in the incidence or severity of injuries between the two playing surfaces during practice or during gameday competition (Fuller et al., 2007; Williams et al., 2016). Lanzetti et al. (2017) argue that when traumatic and overuse injuries for rugby players are analyzed separately, then a significant increase in overuse-type injuries becomes apparent on AT than on NG. Traumatic rugby injuries are primarily through a player-to-player contact mechanism, whereas overuse injuries are more dependent upon the surface and are thus more relevant to a discussion about the role of playing surfaces on injury rate.

2.2 Surface Friction and Traction:

One of the surface characteristics that affects injury risk and performance is friction. Friction remains one of the most familiar, yet least understood facets of mechanics (Brungraber, 1976). According to Dowson's *History of Tribology* (1998), "surface interactions dictate or control the functioning of every device developed by man to enhance the quality of life through his inventiveness and the utilization of the resources of the physical world." In the late fourteenth century, Leonardo da Vinci first stated the classical laws of friction (Milburn & Barry, 1998).

Others have experimentally confirmed and adapted these laws over time. Classically, friction is defined as “a force of resistance acting on a body which prevents or retards slipping of the body relative to the surface with which it is in contact” (Hibbeler, 1998). This force always acts tangentially to the surface to oppose the possible motion of the body. The first classical law of friction is that the friction force is independent of the area of the interacting surfaces. The second law states that the friction force is independent of the sliding velocity. The third law maintains that the friction force is directly proportional to the normal load as shown by Equation 2.1 where F_S is the friction force associated with sliding, F_N is the normal force, and μ is the coefficient of friction.

$$F_S = \mu F_N \tag{2.1}$$

In 1835, it was proposed that friction could be described as static or dynamic (Milburn & Barry, 1998). Static friction is the maximum force needed before motion occurs of a body along a surface while dynamic friction is the force necessary to maintain motion of the body along the surface. Static and dynamic coefficients of friction are both less than 1.0, such that the friction force is less than the normal force. Classically, dynamic friction is slightly less than static friction (Dixon et al., 1999).

In addition to translational friction, rotational friction also exists. Translational, or linear, friction refers to the change in position of a body with no change in its orientation along a linear path, such as a foot sliding along the ground (Frederick, 1986). Rotational friction refers to a fixed body with a torsional force applied along an axis of rotation, such as a foot pivoting on the ground. For the purposes of this review, translational friction rather than rotational friction will be the primary focus.

The classical laws of friction were developed in consideration of dry, non-deformable homogenous surfaces. When different, non-homogenous surfaces interact, such as a studded cleat on AT, all the classical laws of friction are violated (Dixon et al., 1999). First, the friction force increases with an increase in contact area (Hibbeler, 1998). Second, the friction force is inversely dependent on the velocity (Hibbeler, 1998). Third, the friction force can exceed the normal force (Milburn & Barry, 1998; Nigg, 1990). Lastly, the dynamic friction force could be greater than the static friction force for a cleat moving across AT (Milburn & Barry, 1998).

This introduces the concept of traction. While friction and traction are both features which enable the athlete to perform without excessive slipping, 'friction' is a property of the surface and 'traction' is a characteristic of the interaction between the body and the surface. For example, a football player relies on the surface friction to initiate, stop, or change direction, and traction is related to the propulsive or braking force generated on the surface by the football player to execute these tasks (Milburn & Barry, 1998). Traction, similar to friction, is the resistance to movement between a body and the surface, however, the unique traits specific to non-homogenous surfaces must be considered when calculating traction forces.

2.3 Artificial Turf Fields and Traction:

Advances in AT design and development are focused on factors that may decrease the risk of lower extremity injury while still allowing for elevated levels of athletic performance. The traction characteristics of the foot-surface interaction could influence both the perceived and actual injury risk as well as the performance on AT. Several components of the AT system can be independently manipulated to affect traction including fiber type and density, type of infill, levels of compaction, and the use of a shock pad (Lozano-Berges et al., 2019).

Crumb rubber is the most used infill product for AT fields. Over time, the aging and wear of the crumb rubber creates a smaller particle size and a smoother shape which results in greater compaction of the infill. A decreased infill height significantly increases the risk of lower extremity injury by decreasing the friction properties of the field (Fujitaka et al., 2017; Meyers, 2017). Fujitaka et al. (2017) observed that as AT ages, the long pile of the carpet layer narrows, shortens, and flattens. Since it is less expensive to add more infill than to replace the bladed carpet layer, even when new infill product is added to the deteriorated pile, the studs of athletes' cleats cannot penetrate effectively into the surface, resulting in less traction on the AT (Fujitaka et al., 2017).

Even the shock pad and base layers under the carpet layer can impact the traction characteristics of the AT. The subbase is composed of asphalt, compacted gravel, or similar material and is designed to support and transmit the loads produced on the surface. Irregularities in this layer will affect the mechanical properties of the playing surface, such as the surface stability, which could result in increased injury (Sánchez-Sánchez et al., 2014). The purpose of the optional cushioning underlay, or shock pad, installed over the subbase is to maintain the properties of stiffness and shock absorption of the playing surface and to decrease impact forces. It also helps to mitigate the uneven effect of the infill compaction in high-use areas of the AT (Sánchez-Sánchez et al., 2014). By affecting impact ground reaction forces (GRFs), these layers under the AT indirectly affect some traction characteristics, such as the required coefficient of friction (RCOF). The RCOF is defined as the maximal ratio of the horizontal GRF to the vertical GRF. A higher RCOF indicates greater traction on the surface, such as the increased capability of stopping quickly.

Traction is important to consider in the design of AT for maximal performance and minimal injury risk. Surface studies have found that AT surfaces have higher levels of friction than NG (Driscoll et al., 2015). Higher surface friction could increase lower extremity injuries, such as knee and ankle sprains, which are especially dependent upon the foot-surface interface (Driscoll et al., 2015). However, high traction on AT is required for maximal performance, particularly to accelerate and change direction quickly. Too much traction, though, is associated with injury risk (Balazs et al., 2015; Villwock et al., 2009). “Footlock” is the phenomenon when the foot becomes stuck in the playing surface, which can ultimately lead to injury (Bowers & Martin, 1975; Cawley et al., 2003). The increased braking and acceleration that AT allows compared with NG may contribute to footlock (Fujitaka et al., 2017). Low traction, or slipping, is also associated with injury risk; however, a degree of traction release is beneficial to decrease potential loads on the lower extremity (Dixon et al., 1999; Dowling et al., 2010). This poses a unique problem; AT must be designed so that athletes can achieve enough traction to maximize performance, but not contribute to such a level of traction as to increase the risk of lower extremity injury.

2.4 Mechanical Testing of Traction:

The performance of the AT surface involves its ability to absorb and disperse impact forces in addition to providing appropriate traction. Mechanical testing can give a good indication if the surface is performing as expected and gives researchers a better understanding of how the surface responds to the applied forces. However, because human movement patterns are often unique to each athlete and can vary with each performance trial, caution should be exercised to not mistakenly generalize mechanical test results to infer how a human will interact with a surface (Bates, 1996). For example, when measured mechanically, cleats with bladed studs were shown to have comparatively low rotational traction, but when measured in the actual human-surface interaction, the bladed studs were associated with causing excessive stud fixation and contributing to lower extremity injury (Silva et al., 2017). In addition, when performing tasks on surfaces with different mechanical properties, athletes will adjust their movement patterns, further complicating the ability to compare the effect of the surface on athletes' performance (Creagh et al., 1998). Nevertheless, laboratory testing is the most reliable method to measure kinetic and kinematic data from athletes at the level of the surface interface, although advances are being made in wearable devices to collect *in vivo* data with potentially even less disruption to the athlete (Kirk et al., 2007).

One of the most common ways to assess the traction characteristics of AT is to use a mechanical traction test device (Cawley et al., 2003; Milburn & Barry, 1998). Devices have been designed to study both translational and rotational traction, but research conducted on rotational traction are far more common. Often a mechanical torque wrench is used to collect rotational traction data (Cawley et al., 2003; Livesay et al., 2006). The TrakTester is one of the more advanced devices as it applies realistic loadings through an artificial foot model along anatomical axes of the leg (Lehner et al., 2013). It was developed by Grund et al. (2007) to evaluate the factors at the shoe-surface interface influencing ACL injury during cutting movements. Devices designed to study translational traction on AT generally involve moving a shoe across the surface at a predefined speed (Cawley et al., 2003). The traction parameter for that shoe-surface interface is the maximum force achieved during the translation. The most studied movement is the anterior-posterior translation of the shoe forefoot to represent an athlete's push-off movement from rest to a sprint. While this yields information about the AT, it does not provide as much information about injury risk as it relates to linear traction. During athletic movements, the landing phase of a stride or a jump rather than the take-off phase, is most associated with injury risk as described in more detail later in this review. A linear traction device to represent an athlete's landing would therefore be more informative of traction characteristics related to the potential injury risk to the athlete.

2.5 Cleat-Surface Interface:

The studded cleat serves as a fundamental link between the athlete and the AT, and how the cleat interacts with the surface has been shown to influence performance and injury risk (Müller et al., 2010). Cleats are designed to provide optimal traction when the studs penetrate the surface completely so that the outsole of the shoe is in contact with the surface (Kirk et al., 2007). Pressure points are created when the vertical ground reaction force is transmitted only through the studs. Traction can be altered by modifying the stud configuration and the stud shape (Driscoll et al., 2015). As such, a wide variety of sport-specific cleat outsoles are available in an attempt to optimize performance and traction on the wide range of potential playing surfaces (Lehner et al., 2013).

Soccer players, regardless of the playing surface, experience higher rates of ankle sprains than any other injury, and the stud pattern design of their cleats can influence this risk (O'Connor & James, 2013). Butler et al. (2014) reported that male soccer players landed with greater ankle dorsiflexion when wearing cleats with bladed studs compared with rounded studs. Female soccer players wearing the bladed cleats landed with reduced knee flexion compared with the cleats with rounded studs (Butler et al., 2014). Both these landing patterns, increased ankle dorsiflexion angles and decreased knee flexion angles, have been shown on hardcourt surfaces to increase lower extremity injury risk (Boden et al., 2010). It is likely the same mechanism of injury that occurs on hardcourt occurs on AT, which suggests that wearing bladed cleats could contribute to higher injury rates (Butler et al., 2014).

2.6 The Landing Phase:

As alluded to above, during athletic movements, injury is most likely to occur shortly after foot strike during the landing phase rather than during the propulsive, or take-off, phase (Dixon et al., 1999; Frederick, 1986; Nigg et al., 1987). Compared with the take-off phase, greater stress is placed upon the ACL of the knee during the landing phase in stop-jump tasks on hardcourt (Boden et al., 2010; Chappell et al., 2002). The first peak horizontal and vertical GRFs typically occur shortly after foot strike and the peak GRF is associated with high decelerations of the lower extremity as the foot impacts the surface as shown in Figure 2.2 (Chappell et al., 2002). These high forces have been linked to acute lower extremity injuries, such as fractures and ligamentous injury, as well as chronic injuries, such as stress fractures, tendonitis, and damage to articular cartilage (Dixon et al., 1999; Frederick, 1986; Mack et al., 2019; Nigg et al., 1987; Zhang et al., 2000). Since AT allows for increased braking and dynamic deceleration compared with NG, this may contribute to greater horizontal and vertical GRFs on AT than on NG (Fujitaka et al., 2017). These GRFs on AT may be more similar to impact GRFs on hardcourt than on NG and may partially explain the higher incidence of injury on AT compared with NG during the landing phase (Balazs et al., 2015; Ferrandino et al., 2015; Fujitaka et al., 2017).

The second GRF peak corresponds with the take-off phase of a stride or a jump. Since this is generally a more controlled application of force than during the landing phase, the take-off phase has not been related to overuse injuries (Dixon et al., 1999; Frederick, 1986).

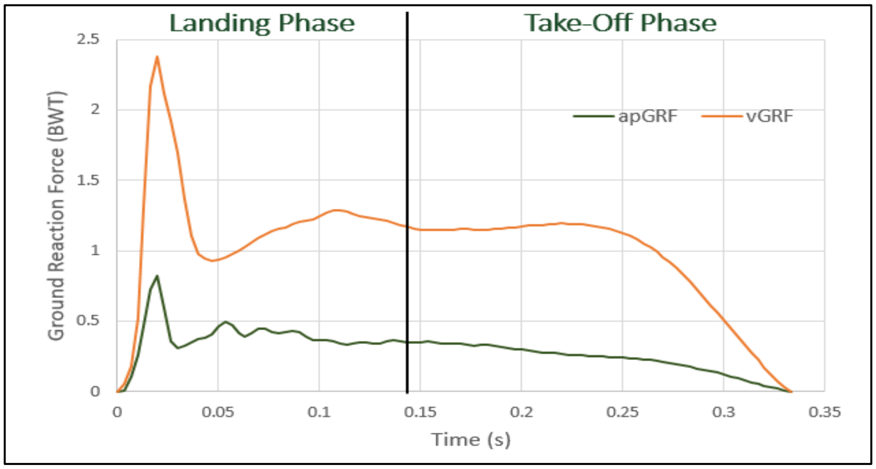

Figure 2.2 Horizontal and Vertical GRFs during Landing and Take-off. *GRFs normalized to bodyweight for males and females during the landing and take-off phases of a stop-jump task on AT.*

2.7 Foot Orientation and Traction

An athlete's interaction with a playing surface must be considered in addition to the material properties of the surface and their effects on performance and injury risk. The orientation of the foot and the foot's velocity immediately before impact with the surface have been shown to predict injury risk on hardcourt surfaces (Donnelly et al., 2017; Ferrandino et al., 2015; Kirk et al., 2007; Kristianslund et al., 2014). While little is reported on the foot orientation upon impact, it is far more common to report other lower extremity kinematic data, understanding the foot orientation upon landing could reveal critical information about injury risk and performance.

Donnelly et al. (2017) showed that a hardcourt rearfoot landing compared with a forefoot landing places a larger mechanical demand on the knee joint, which is associated with increased ACL injury risk, however, a forefoot landing may place a larger mechanical demand on the ankle. Reviewed video footage of athletes who suffered an ACL rupture during various court sports were found to have landed flatfooted or with the rearfoot compared with the athletes who landed on the forefoot who did not suffer injury while performing similar movements (Boden et al., 2009). Reaching a flatfoot position sooner upon landing gives less time for the triceps surae musculature to contract and absorb forces which increases the impulsive force to the knee (Boden et al., 2010). Conversely, a forefoot landing gives the soft tissue components of the leg time to dampen the force and could be a critical aspect in preventing ACL injury. Investigating the foot orientation when the athlete is at the highest risk for injury, during the landing phase, may provide information to potentially train athletes to land safely.

Ferrandinoa et al. (2015) concluded that one of the elements that may influence linear traction is the angle with which the studs initially contact the AT. Indeed, Kirk et al. (2007) found that the studs, especially rounded studs, could be better aligned to enhance penetration into the AT to optimize traction properties. Most cleats are designed with the studs aligned normal to the sole. Assuming the sole impacts the ground parallel to the surface, this is the most efficient design for stud penetration and optimal traction (Driscoll et al., 2015; Kirk et al., 2007). However, if the foot is oriented so that the cleat interacts with the surface at an angle, such as with either a forefoot or a rearfoot landing, an obliquely angled stud design would be more advantageous (Kirk et al., 2007). These findings support that the foot orientation upon landing does play a role in the traction characteristics of regularly performed athletic tasks.

Jumping and landing movements are integral to many sports and are commonly studied in laboratory settings (Zhang et al., 2000). Stop and stop-jump tasks are two such high-impact movements with strong horizontal components that have been associated with acute and overuse injuries (Zhang et al., 2000). Non-contact ACL injury risk has been associated with greater GRFs, increased proximal tibia anterior shear force, and greater knee flexion moment during the initial landing in stop-jump tasks (Sell et al., 2007; Yu et al., 2006). Stop-jump tasks are more commonly performed in research settings than stop tasks, but the initial GRFs have been shown to be different between the two movements (Hass et al., 2003). The stop has been shown to produce greater peak horizontal GRFs than the stop-jump (Hass et al., 2003). Additionally, while Hass et al. (2003) did not report landing foot orientation, it appears that the foot progression angles could be different between the stop and stop-jump tasks. The reported difference in the average ankle angles between the two movements was 12.9 degrees, although the high variability in the stop task prevented this difference from being statistically significant (Hass et al., 2003). It is possible that the preparation for the subsequent physical demands of the stop-jump compared with the simpler stop task could be one influencing factor of the greater forces and the larger variability of the ankle angle upon landing with the stop task.

2.8 Sex Differences During the Landing Phase:

Landing mechanics also differ between sexes (Butler et al., 2013). It has long been shown that females sustain more non-contact ACL injuries during sport than males. In an evaluation across basketball, ice hockey, lacrosse, soccer, and baseball/softball, the incidence rate for female athletes sustaining ACL injuries is more than two times the rate for male athletes at both the high school and the collegiate levels (Stanley et al., 2016). The National Collegiate Athletic Association reported the incidence rate is three to four times higher for collegiate female basketball and soccer players than for their male counterparts (Arendt et al., 1999). One reason for this disparity between males and females could be attributed to altered landing mechanics. Compared with males, for various stop-jump tasks, females landed with greater knee extension, greater valgus moments, and greater tibial plateau anterior shear forces, putting the knee in a more compromised position and at greater risk for ACL injury (Chappell et al., 2002; Grund et al., 2007; Grund & Senner, 2006; Yin et al., 2015). These landing differences between sexes was found in a study of youth recreational soccer players by the age of twelve and the differences were even more apparent by the age of sixteen (Yu et al., 2005). In addition, during a drop landing task, the dynamic knee valgus peak was found during the deceleration phase of landing in females, but in the acceleration phase in males, suggesting that females collapse more rapidly into knee valgus compared with males, putting the ACL at increased risk for injury (Joseph et al., 2011). Furthermore, during the preparation for the landing, females had increased hamstring activation before landing and decreased activation after landing compared with males, which also increases the risk of injury to the ACL (Chappell et al., 2007). In addition, analytic videotape reviews of athletes who suffered ACL injury found that female athletes were primarily injured during a simple deceleration maneuver, whereas male athletes were more commonly injured during more strenuous jumping tasks (Boden et al., 2009). If initial GRFs are different between these tasks, this would support the idea that ACL injury occurs at different GRFs between males and females. Differences between males and females are therefore expected between a simple stop task and the more strenuous stop-jump task. While many landing mechanics differ between males and females, it is unclear if these differences include foot orientation. As indicated above, it is plausible that differences in foot orientation upon landing influences traction and could contribute to injury risk.

2.9 Summary of the Literature:

In summary, the use of artificial turf fields is becoming increasingly more prevalent in the United States and while injury rates on artificial turf compared with natural grass are not clear, athletes perceive there to be a greater risk on artificial turf. One surface-dependent performance characteristic that is also associated with injury risk is traction. There are several mechanical methods to determine rotational and translational traction, however, to determine biomechanical factors dependent upon variable human movement it is best to obtain human subject data in the laboratory setting. During athletic movement, injury is most likely to occur during the landing phase rather than the take-off phase of a stride or jump. Flatfoot or rearfoot landings result in greater ground reaction forces and potentially increase the risk for lower extremity injury regardless of the surface. Little is known about whether traction is influenced by the orientation of the foot upon impact. This information could be used to help modify athlete technique, shoe design, and artificial turf surface characteristics for optimal athlete performance with reduced risk for injury.

CHAPTER 3
METHODS

3.1 Subjects:

Thirty-two healthy collegiate club level or higher athletes (16 females) between 18-30 years of age volunteered for the investigation. Subjects were competitive in their sport during the previous year, used their cleats at least twice per week for the last month, were pain/soreness/injury free at the time of data collection, and did not wear ankle or knee orthopedic braces during exercise. A minimum of two weeks full participation in athletic activities without restriction was required and subjects had to be free of any condition or medication affecting balance. Subjects were familiar with artificial turf (AT) prior to participating in this study. Before collecting any data, the testing procedures were explained to the subjects, eligibility was confirmed by reviewing the subjects' completed health-history questionnaire, and university-approved, written informed consent was obtained. Subjects were monetarily compensated for their time at the conclusion of their second visit.

3.1.1 Prior injuries

Prior injuries were not exclusionary if the subject met all other criteria listed above. Based on the health history questionnaire, one male and one female subject had a history of a right anterior-cruciate ligament (ACL) tear at least three years prior, two female subjects had a history of a fractured right foot/ankle seven years prior, and one female had a history of a fractured left hip nine years prior to data collection.

3.2 Artificial Turf:

The artificial turf used in this investigation was a 5cm monofilament, slit-film blend with 1.25cm spacing (Shaw Sports Turf, Calhoun, GA) secured with screws and carpet tape to a

2.54cm thick expandable foam shock pad (PowerBase YSR, Brock USA). The crumb rubber (CR) performance layer infill was added over a silica sand base, consistent with manufacturer recommendations at a ratio of 9.0/17.1 kg/m^2 CR/sand. The styrene-butadiene CR was new ambient ground fine rubber granulate originally sourced from car/truck tires (Genova USA, Inc., Beaumont, TX). Artificial turf panels were made to cover a 1.2x3.7m performance area and additional panels surrounded this area for safety [*Figure 3.1*]. All turf, shock pad, and infill material were new at the start of the study. Isolated turf/shock pad panels were secured with carpet tape over two isolated 40x60cm force platforms (Bertec Corp., Columbus, OH) which were flush to the surrounding concrete floor with a 3mm gap around all edges. New panels over the force platforms were introduced after every 10 subjects. All turf was conditioned by athlete-use prior to collecting research data. In-fill depth was measured over the force platforms before each subject, leveling and adding infill material as necessary to maintain 19mm blade reveal.

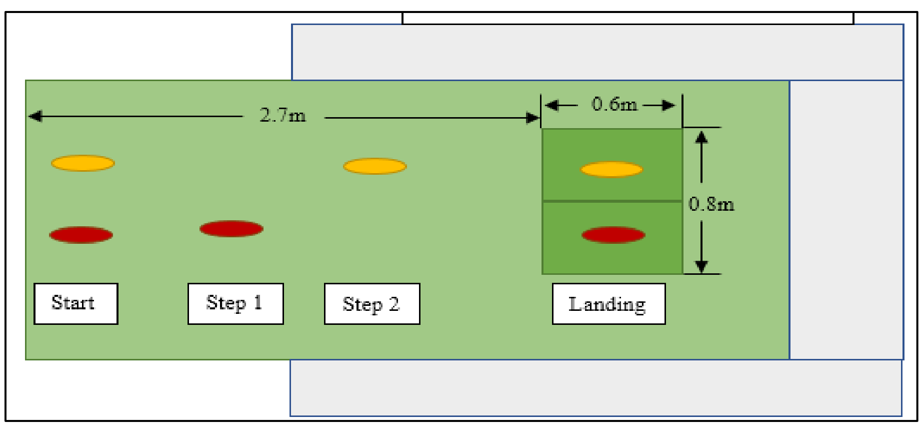

Figure 3.1: Artificial Turf Performance Area. *The AT infilled with CR performance area (light green) was surrounded by additional AT for safety (gray). 2.7m of run-up space was available for the required two step lead-in starting with step 1 performed with the right foot (red) and step 2 with the left foot (yellow) then landing bilaterally with each foot near the center of each of the two isolated 40x60cm force platforms (dark green).*

3.3 Data Collection:

3.3.1 Visit one

The experimental data collection occurred over two visits. Subjects were encouraged to abstain from strenuous exercise in the 24 hours prior to each visit. The first visit was designed for familiarization, measurement of required anthropometrics, and the performance of a functional movement assessment. After completing the functional movement assessment in court shoes, subjects donned their cleats and practiced the more dynamic movements to be assessed on their second visit.

3.3.2 Visit two

The second visit was completed within ten days of the first visit. Subjects wore appropriate tightfitting, spandex-like clothing and a full-body reflective 14mm diameter marker set for use in Plug-in-Gait (Vicon, Centennial, CO) was applied with toupee tape. Medial knee and ankle markers were removed after capturing the subjects' static 'motorcycle' pose and constructing their wireframe model. Subjects then warmed-up on a stationary bike. After donning their cleats, additional specific practice was given before each of the different dynamic movement trials. A minimum of one successful practice was required before collecting data. In order, the subjects performed six acceptable trials of a bilateral drop landing, reverse pivot, vertical hop, Two-Step Stop (TSS), Two-Step Stop Jump (TSSJ), and a unilateral 90-degree cut. Only the TSS and TSSJ were assessed here.

3.3.3 Comparison in-fill

In addition to the standard CR infill, a second wood-based infill product was also assessed during this investigation. All trials were performed on one preparation of the artificial turf before taking a ten-minute break and completing the trials on the second turf preparation.

Turf type was presented in a random order, alternating between subjects. Only the TSS and TSSJ performed on the CR infill were assessed here.

3.3.4 Stop and stop-jump tasks

Both movements started with a required two-step lead-in which began with the right foot [*Figure 3.1*]. The maximum distance a subject's heels could start from the edge of the force platforms was 2.7m. For the TSS, the two-step lead-in to gain maximal controlled horizontal speed preceded a bilateral landing on the force platforms [*Figure 3.2*]. Subjects were instructed to stop as quickly as possible and to hold the athletic-ready position for at least one second prior to returning to a relaxed standing position. The TSSJ began with the same two-step lead-in to gain maximal controlled horizontal speed before the bilateral landing then immediately after making ground contact, subjects jumped vertically with maximal effort [*Figure 3.3*]. Subjects could use their arms if desired and were required to have a controlled landing on the force platforms.

Figure 3.2: Two-Step Stop Task. *Two-step lead-in starting with the right foot and landing bilaterally in the athletic-ready position.*

Figure 3.3: Two-Step Stop Jump Task. *Two-step lead-in starting with the right foot, landing bilaterally, then immediately jumping vertically.*

3.3.5 Ground reaction forces

Ground reaction forces (GRFs) were sampled at 1200Hz simultaneously with optical motion capture at 300Hz (10 camera, Vicon Nexus). Raw marker coordinates were gap filled and Woltring low-pass filtered (MSE 10mm) in Vicon. Inverse dynamics was performed using Vicon's Plug-in-Gait for the calculation of joint kinematics and kinetics. Custom Matlab code (MathWorks, Natick, MA) was used for any additional calculations and the extraction of variables for all movements. Exported GRFs and joint kinetics were 4th-order recursively Butterworth low-pass filtered at 100Hz. Velocities were computed using First Central Differences. All forces were normalized to bodyweight.

3.4 Analysis:

Each foot was analyzed separately. Initial contact of each foot was detected with a vertical GRF (vGRF) threshold of 10N. The landing phase for both the TSS and TSSJ was defined from initial foot contact until maximum knee flexion.

Foot orientation was explored using the Plug-in-Gait Foot Progression Angle (FtProgAng) in the sagittal plane. The FtProgAng was normalized to zero degrees when the foot was flat on the ground at the end of the landing phase. An initial negative FtProgAng was indicative of a rearfoot orientation upon ground contact and an initial positive FtProgAng was indicative of a forefoot orientation upon ground contact. A flatfoot orientation had an initial FtProgAng of zero degrees. Variables to assess traction included the time to Full Foot Engagement (FFE) with the turf after initial contact. FFE was defined as the first instance when both the FtProgAng was within five degrees of flatfoot and the Foot Center of Mass (COM) Vertical Velocity (FtCOMvVel) was greater than -0.1m/s after initial contact [*Figure 3.4*]. The Anterior-Posterior Foot COM displacement prior to FFE and after FFE was quantified, as well as the peak vGRF and anterior-posterior GRF (apGRF) within the first 100ms of landing, the rate of force development within the first 100ms, the average vGRF and apGRF in the first 50ms and 100ms, and the maximal Required Coefficient of Friction (RCOF) during the landing phase. RCOF is the maximal ratio of apGRF to vGRF.

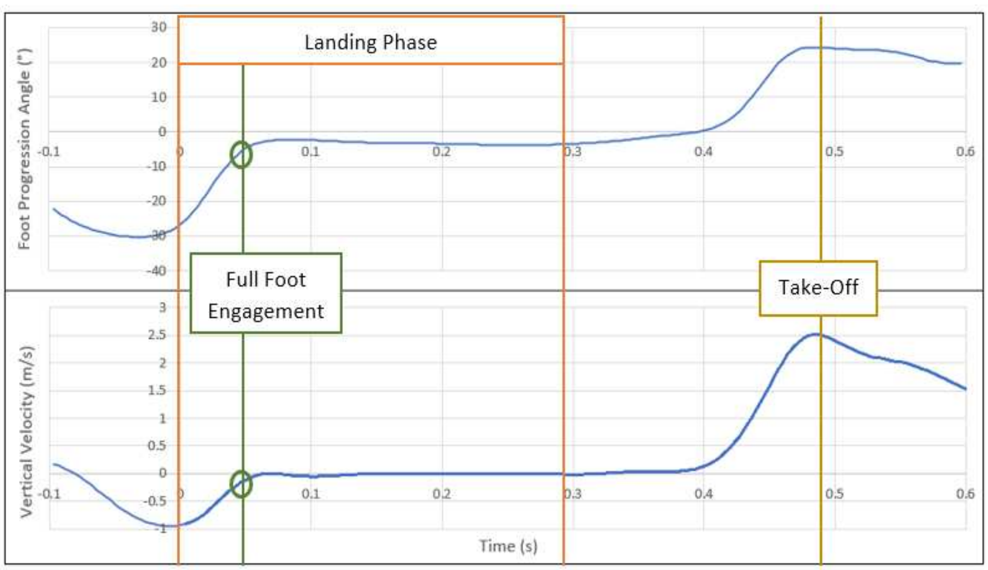

Figure 3.4: Initial Foot Progression Angle and Foot Center of Mass Vertical Velocity TSSJ Exemplar Plot. *Full Foot Engagement occurs when the Foot Progression Angle is within five degrees of flatfoot orientation and when the Foot COM Vertical Velocity is greater than -0.1m/s.*

3.4.1 Five trial selection

Each subject's best five trials for each movement were used in this assessment, as determined by outlier analysis of variables associated with the right leg at initial impact [*Appendix Table 1*]. The process started by using each subject's first five trials, if an outlier was identified in one of the fourteen selection variables it was replaced with the subject's sixth trial. Variables associated with the right foot's initial contact were selected because the right foot typically made impact slightly before the left. Variables assessed included: sagittal plane angle and angular velocity of the hip, knee, ankle, and foot, anterior-posterior and vertical foot COM velocities, and anterior-posterior and vertical whole-body COM positions and velocities. These variables were assessed for outliers both as individual trials and after a representative average trial was made from subjects' best five trials.

3.4.2 Statistics

The representative average trial from each subject was used to determine differences between sex and limb within and between each movement with a 4x2 repeated measures ANOVA. Bonferroni post-hoc tests were used to assess significant main effects for movement and limb. Independent t-tests were used to explore differences between sex. There were no extreme outliers. Normality was assessed with the Kolmogorov-Smirnov Test. Differences found with the ANOVAs for any variable that was not normally distributed were then verified with Friedman's two-way analysis of variance by ranks non-parametric test followed by Bonferroni post-hoc tests to assess significant main effects for movement and limb. Differences between sex for not normally distributed variables were verified with independent samples median test. Nine of the assessed variables had at least one non-parametric movement/limb. The ANOVA results were verified as described for six of these variables. Three of the variables had minor discrepancies between their ANOVA result and the non-parametric tests and these non-parametric findings are noted in the Results chapter.

3.4.3 Correlations with initial foot progression angle

The relationships between FtProgAng at initial ground contact and traction were explored with Pearson's Correlation (r) using individual trials. Each limb within each movement was assessed separately within the whole group as well as within each sex. Moderate correlations are defined as having r-values between 0.5 and 0.7, moderately strong correlations have r-values between 0.7 and 0.8, and strong correlations have r-values between 0.8 and 0.9. Correlation r-values greater than 0.9 are good for predicting (Vincent, 1999). All statistics were performed in IBM SPSS v26 (Armonk, NY) with $\alpha=0.05$.

CHAPTER 4
RESULTS

4.1 Subject Characteristics:

Two subjects were excluded from data analysis. One subject was injured between visits and not able to complete the second visit. Another subject's data was deemed unusable due to a marker misplacement at the heel during data collection. Two subjects only had four acceptable trials for the two-step stop (TSS); they were included in the analysis using only the four available trials for this movement. All included subjects had five trials for the two-step jump (TSSJ).

Twenty-nine athletes (15 females) ranging in age from 18 to 26 years were included in the analysis *[Table 4.1*]. Average age was not different between females and males (*p=0.051*), although males were on average taller (*p<0.001*) and of greater mass than females (*p<0.001*). All athletes were competitive in their ground-based sport during the previous year. Cleats worn during the study were the subject's own and ranged in stud shape. Based on preferred kicking leg, most subjects were right-leg dominant.

Table 4.1: Subject Characteristics. *Mean and (standard deviation) for age, height, and mass. (Number of subjects) listed for sport, stud shape, and leg dominance. $^{\Delta}p<0.001$ between sex.*

	Whole Group	**Female**	**Male**
n	29	15	14
Age (yrs)	20.8 (1.9)	20.1 (1.5)	21.5 (2.1)
Height (cm)$^{\Delta}$	174.4 (9.6)	167.6 (6.1)	181.6 (7.0)
Mass (kg)$^{\Delta}$	71.3 (11.1)	64.5 (8.4)	78.7 (8.8)
Sport	Soccer (9) Lacrosse (7) Ultimate Frisbee (7) Rugby (6)	Soccer (4) Lacrosse (3) Ultimate Frisbee (4) Rugby (4)	Soccer (5) Lacrosse (4) Ultimate Frisbee (3) Rugby (2)
Stud Shape	Blade (8), Round (6), Blade+Round (10), Other (5)	Blade (4), Round (2), Blade+Round (7), Other (2)	Blade (4), Round (4), Blade+Round (3), Other (3)
Leg Dominance	Right (25), Left (3), No Preference (1)	Right (13), Left (1), No Preference (1)	Right (12), Left (2)

4.2 Landing Characteristics:

4.2.1 Foot contact time difference between limbs

With the required two-step approach, only one subject contacted the ground with their left foot first based on their five-trial average. As a whole group, the right foot impacted the force plate 0.02 (±0.01) seconds prior to the left foot for both the TSS and the TSSJ and from the 2x2 ANOVA there was no difference between males and females (*main effect for movement* $p=0.579$*, main effect for sex* $p=0.422$).

4.2.2 Whole-body center of mass position and velocity at foot impact

The 4x2 ANOVAs of whole-body center of mass (COM) position and velocity at initial foot contact with the ground comparing the two movements (TSS and TSSJ), the two limbs (right and left), and the two sexes (male and female) revealed that there were differences between the movements, limbs, and sexes. The whole-body COM anterior-posterior position upon landing (iCOMap) was further behind the force platform for the TSS compared with the TSSJ, was further behind when the right foot made contact compared with the left for both movements, and the iCOMap was further behind when males contacted the force plate compared with females (*main effect for movement and limb* $p<0.001$*, main effect for sex* $p<0.001$) [*Table 4.2*]. The whole-body COM anterior-posterior velocity upon landing (iCOMapVel) was not different between movements but was faster when the right foot made contact compared with the left for both movements, and males had a faster iCOMapVel than females (*main effect for movement and limb* $p<0.001$*, main effect for sex* $p=0.001$) [*Table 4.2*]. The whole-body COM vertical position upon initial foot contact with the ground (iCOMv) was significantly higher for the TSSJ compared with the TSS when the right foot contacted the force platform, the iCOMv was higher when the right foot contacted compared with the left for both movements, and it was higher for males compared with females (*main effect for movement and limb p<0.001, main effect for sex p=0.030*) [*Table 4.2*].

Table 4.2: Whole-Body Center of Mass (COM) Characteristics Upon Initial (i) Ground Contact. *Mean and (standard deviation) for when each foot made ground contact. ap = anterior-posterior, v = vertical, Vel = velocity. Units: positions in m, velocities in m/s. ●main effect for movement and limb p<0.05, ∆main effect for sex p<0.05, ■Friedman's main effect for movement and limb p<0.05, ◊independent samples median test for sex p<0.05 (see text for further detail).*

	Whole Group		Female		Male	
Right	**TSS**	**TSSJ**	**TSS**	**TSSJ**	**TSS**	**TSSJ**
iCOMap●∆	-0.42	-0.39	-0.39	-0.34	-0.45	-0.44
	(0.05)	(0.06)	(0.05)	(0.04)	(0.04)	(0.05)
iCOMapVel●∆	2.25	2.24	2.13	2.07	2.37	2.42
	(0.24)	(0.27)	(0.21)	(0.19)	(0.20)	(0.23)
iCOMv●∆	0.81	0.83	0.80	0.81	0.83	0.85
	(0.04)	(0.04)	(0.05)	(0.04)	(0.02)	(0.03)
iCOMvVel	-1.38	-1.42	-1.33	-1.43	-1.43	-1.40
	(0.21)	(0.19)	(0.24)	(0.20)	(0.16)	(0.18)
Left	**TSS**	**TSSJ**	**TSS**	**TSSJ**	**TSS**	**TSSJ**
iCOMap●∆	-0.38	-0.34	-0.36	-0.31	-0.40	-0.37
	(0.04)	(0.05)	(0.04)	(0.04)	(0.02)	(0.03)
iCOMapVel●∆	2.11	2.12	2.00	1.98	2.23	2.28
	(0.23)	(0.26)	(0.20)	(0.20)	(0.21)	(0.23)
iCOMv●∆	0.78	0.79	0.77	0.78	0.80	0.81
	(0.04)	(0.04)	(0.05)	(0.04)	(0.02)	(0.03)
iCOMvVel■	-1.33	-1.41	-1.30	-1.45	-1.37	-1.37
	(0.20)	(0.20)	(0.24)	(0.18)	(0.15)	(0.21)

The whole-body COM vertical velocity upon landing (iCOMvVel) was not normally distributed across both movements for each limb. The iCOMvVel was faster for the TSSJ than for the TSS for the left limb only and there were no differences between the limbs or between males and females (*Friedman's main effect between movement and limb p=0.029, independent samples median test for sex p ≥ 0.066*) [*Table 4.2*].

4.2.3 Foot center of mass velocity at impact

The 4x2 ANOVAs of foot COM velocity at initial foot contact with the ground showed that there were differences between the movements, limbs, and sexes. Upon impact, the foot COM anterior-posterior velocity upon landing (iFtCOMapVel) was faster for the TSS compared with the TSSJ for both limbs, the left foot contacted the force platform faster than the right foot for the TSS, and the iFtCOMapVel was faster for males compared with females (*main effect for movement and limb p<0.001, main effect for sex p=0.001*) [*Table 4.3*]. The vertical velocity of the foot COM upon impact (iFtCOMvVel) was faster for the TSSJ than the TSS when the right foot made ground contact, the right foot contacted the force platform faster than the left foot for both movements and the iFtCOMvVel was faster when males contacted the ground compared with females (*main effect for movement and limb p<0.001, main effect for sex p=0.001*) [*Table 4.3*].

Table 4.3: Foot Center of Mass (COM) and Foot Progression Angle (FtProgAng) Characteristics Upon Initial (i) Ground Contact. *Mean and (standard deviation) for when each foot made ground contact. ap = anterior-posterior, v = vertical, Vel = velocity. Units: positions in m, linear velocities in m/s, angular velocities in °/s. ●main effect for movement and limb p<0.05, Δmain effect for sex p<0.05 (see text for further detail).*

	Whole Group		Female		Male	
Right	**TSS**	**TSSJ**	**TSS**	**TSSJ**	**TSS**	**TSSJ**
iFtCOMapVel●Δ	1.86 (0.53)	1.48 (0.46)	1.67 (0.50)	1.24 (0.34)	2.07 (0.49)	1.74 (0.44)
iFtCOMvVel●Δ	-1.22 (0.26)	-1.15 (0.22)	-1.06 (0.18)	-1.04 (0.16)	-1.39 (0.24)	-1.27 (0.22)
iFtProgAng●	-20.69 (11.43)	-16.69 (15.00)	-18.09 (12.97)	-11.97 (14.14)	-23.49 (9.16)	-21.74 (14.70)
iFtProgAngVel●Δ	378.78 (250.50)	232.58 (247.63)	276.31 (235.90)	136.64 (198.05)	488.56 (223.90)	335.38 (260.45)
Left	**TSS**	**TSSJ**	**TSS**	**TSSJ**	**TSS**	**TSSJ**
iFtCOMapVel●Δ	2.05 (0.59)	1.54 (0.51)	1.74 (0.51)	1.28 (0.48)	2.37 (0.49)	1.82 (0.39)
iFtCOMvVel●Δ	-0.97 (0.27)	-0.94 (0.22)	-0.82 (0.17)	-0.90 (0.17)	-1.13 (0.26)	-0.98 (0.26)
iFtProgAng●	-13.28 (7.27)	-10.64 (9.44)	-10.34 (6.08)	-9.16 (9.01)	-16.42 (7.32)	-12.24 (9.95)
iFtProgAngVel●Δ	246.04 (200.66)	135.76 (225.34)	153.42 (140.11)	92.34 (205.82)	345.27 (212.32)	182.28 (243.41)

4.2.4 Foot progression angle and angular velocity at impact

The 4x2 ANOVAs of foot progression angle and angular velocity upon impact with the ground revealed that there were differences between the movements, limbs, and sexes for the angular velocity, but only a limb difference for the foot progression angle. The foot progression angle upon impact (iFtProgAng) was not different between the two movements nor was there a difference between the sexes, but the right foot was oriented for a more rearfoot landing (a more negative FtProgAng) upon impact than the left foot for both movements (*main effect for movement and limb p<0.001, main effect for sex p=0.060*) [*Table 4.3*] [*Figure 4.1*]. The foot progression angular velocity at impact (iFtProgAngVel) was faster for the TSS compared with the TSSJ for both limbs, the right foot was rotating faster than the left for the TSS at impact, and the iFtProgAngVel was faster when males made ground contact compared with females (*main effect for movement and limb p<0.001, main effect for sex p=0.008*) [*Table 4.3*].

4.2.5 Figure explanation

Figures 4.1 through 4.11 are formatted as box-and-whisker plots of the subjects' representative average trials displaying the complete range of results and quartiles for the right (R) and left (L) limbs for the two movements (TSS and TSSJ). Mean is represented by an X. Individual dots represent outliers, there were no extreme outliers. ★*p<0.05.*

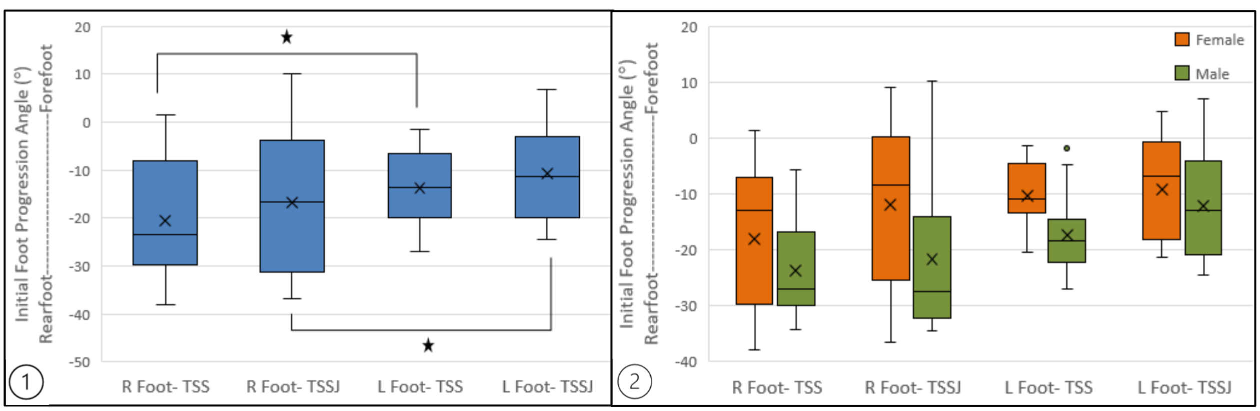

Figure 4.1: Initial Foot Progression Angle Upon Ground Contact. *The range of foot orientations in degrees from rearfoot to forefoot for each limb and for each movement for the whole group of subjects (1) and separated by sex (2).*

4.3 Full Foot Engagement:

Full foot engagement (FFE), when the cleat is engaged completely with the AT, occurred for the whole group on average 0.06 (±0.02) seconds after the right foot made initial ground contact and 0.04 (±0.01) seconds after the left foot made initial ground contact with no differences between movements or sex (*main effect for movement and limb p<0.001, main effect for sex p=0.509*) [*Figure 4.2*]. The foot COM anterior-posterior displacement before FFE (preFFEFtCOMap) was significantly greater for the TSS compared with the TSSJ and the right foot had greater preFFEFtCOMap than the left foot for both movements, but there were no differences between sexes (*main effect for movement and limb p<0.001, main effect for sex p=0.056*) [*Figure 4.3*]. The foot COM anterior-posterior displacement after FFE until the end of the landing phase (postFFEFtCOMap) was significantly greater for the TSSJ than for the TSS, males had greater postFFEFtCOMap than females, but there were no differences between the limbs (*main effect for movement and limb p<0.001, main effect for sex p=0.001*) [*Figure 4.4*].

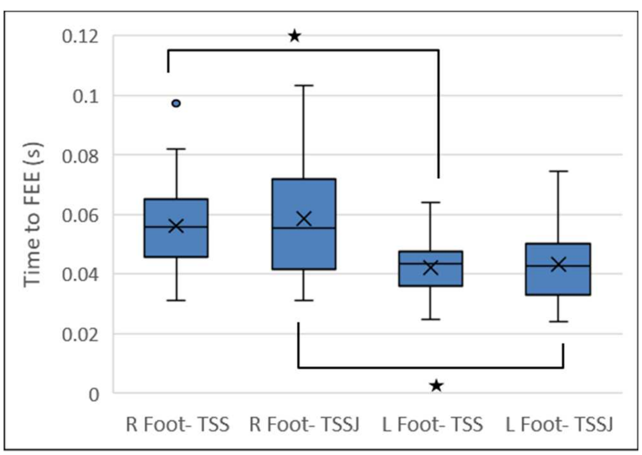

Figure 4.2: Time to Full Foot Engagement (FFE). *Time in seconds from initial foot contact for each limb and for each movement for the whole group of subjects.*

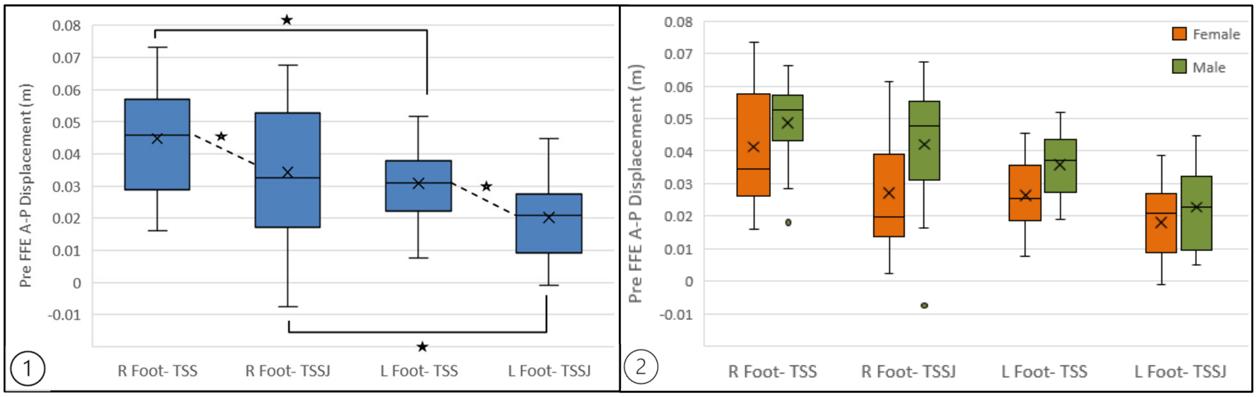

Figure 4.3: Pre-Full Foot Engagement (FFE) Anterior-Posterior (A-P) Foot COM Displacement. *Displacement (m) from initial foot contact to FFE for each limb and for each movement for the whole group of subjects (1) and separated by sex (2).*

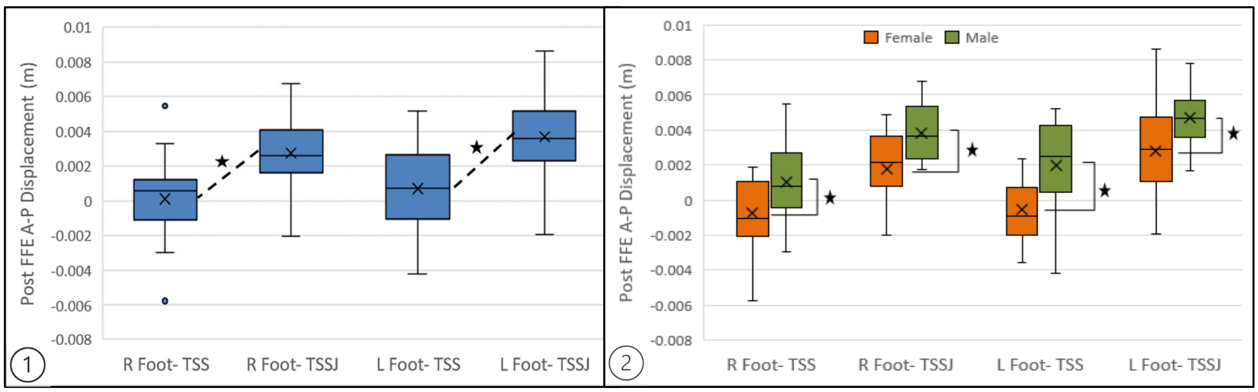

Figure 4.4: Post-Full Foot Engagement (FFE) Anterior-Posterior (A-P) Foot COM Displacement. *Displacement (m) from FEE to the Foot COM position at the end of the landing phase for each limb and for each movement for the whole group of subjects (1) and separated by sex (2).*

4.4 Characteristics of Landing Ground Reaction Forces:

4.4.1 Ground reaction forces upon impact

The 4x2 ANOVAs of the ground reaction force (GRF) normalized to bodyweight to compare the two movements (TSS and TSSJ), the two limbs (right and left), and the two sexes (male and female) revealed that there were differences between the movements and the limbs primarily for the horizontal forces but not for the vertical forces [*Figures 4.5-4.8*] The peak anterior-posterior GRF within the first 100ms upon impact (PapGRF100) was greater for the TSS than for the TSSJ for both limbs, was greater for the right foot compared with the left, and was greater for males compared with females (*main effect for movement and limb p<0.001, main effect for sex p=0.009*) [*Figure 4.5*] [*Table 4.4*]. The time when the PapGRF100 occurred (PapGRF100time) was later for the right limb than the left for both movements but was not different between the movements or the sexes (*main effect for movement and limb p<0.001, main effect for sex p=0.711*) [*Table 4.4*]. The average anterior-posterior GRF within the first 100ms upon impact (AapGRF100) was greater for the TSS than for the TSSJ for the right limb only, was greater for the right foot compared with the left for both movements, and there were no differences between the sexes (*main effect for movement and limb p<0.001, main effect for sex p=0.290*) [*Table 4.4*].

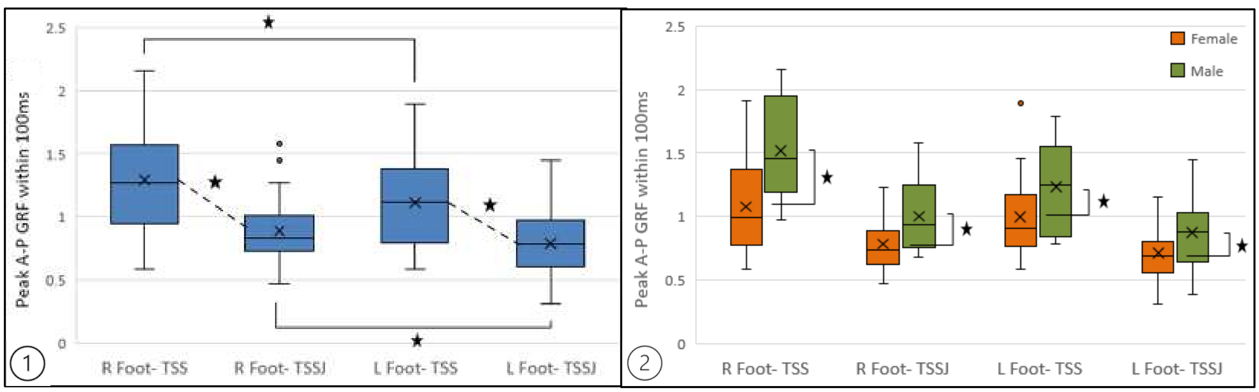

Figure 4.5:Peak Anterior-Posterior (A-P) GRF within 100ms of Initial Foot Contact. *Maximal GRF (BWT) for each limb and for each movement for the whole group of subjects (1) and separated by sex (2).*

The AapGRF within the first 50ms (AapGRF50) was not normally distributed across both movements for each limb. The AapGRF50 was greater for the TSS than for the TSSJ for both limbs but was not different between limbs or between males and females (*Friedman's main effect between movement and limb p<0.001, independent samples median test for sex p ≥ 0.466)* [*Figure 4.6*] [*Table 4.4*].

Table 4.4: GRF Normalized to Bodyweight within 100ms or 50ms of Initial Foot Contact. *Mean and (standard deviation) for when each foot made ground contact. P = peak, A = average, ap = anterior-posterior, v = vertical. Units: forces normalized to bodyweight, time in s. ●main effect for movement and limb $p<0.05$, ∆main effect for sex $p<0.05$, ■Friedman's main effect for movement and limb $p<0.05$, ◊independent samples median test for sex $p<0.05$ (see text for further detail).*

Right	**Whole Group**		**Female**		**Male**	
	TSS	**TSSJ**	**TSS**	**TSSJ**	**TSS**	**TSSJ**
PapGRF100$^{\bullet\Delta}$	1.30	0.89	1.08	0.78	1.53	1.00
	(0.45)	(0.27)	(0.37)	(0.22)	(0.42)	(0.28)
PapGRF100time$^{\bullet}$	0.04	0.04	0.04	0.04	0.03	0.05
	(0.01)	(0.02)	(0.01)	(0.02)	(0.01)	(0.02)
AapGRF100$^{\bullet}$	0.61	0.52	0.58	0.49	0.65	0.55
	(0.13)	(0.13)	(0.12)	(0.14)	(0.14)	(0.13)
AapGRF50$^{\blacksquare}$	0.59	0.45	0.54	0.44	0.64	0.46
	(0.17)	(0.15)	(0.17)	(0.17)	(0.16)	(0.14)
PvGRF100	2.23	2.01	2.02	1.98	2.45	2.05
	(0.57)	(0.59)	(0.53)	(0.66)	(0.53)	(0.53)
PvGRF100time$^{\bullet}$	0.03	0.04	0.03	0.05	0.03	0.04
	(0.01)	(0.02)	(0.01)	(0.03)	(0.01)	(0.02)
AvGRF100$^{\bullet}$	1.09	1.19	1.04	1.17	1.14	1.21
	(0.18)	(0.28)	(0.16)	(0.30)	(0.19)	(0.26)
AvGRF50$^{\bullet}$	1.22	1.10	1.13	1.06	1.31	1.14
	(0.25)	(0.32)	(0.24)	(0.35)	(0.24)	(0.28)
Left	**Whole Group**		**Female**		**Male**	
	TSS	**TSSJ**	**TSS**	**TSSJ**	**TSS**	**TSSJ**
PapGRF100$^{\bullet\Delta}$	1.11	0.79	1.00	0.71	1.24	0.88
	(0.36)	(0.28)	(0.34)	(0.23)	(0.35)	(0.30)
PapGRF100time$^{\bullet}$	0.03	0.03	0.03	0.03	0.03	0.04
	(0.01)	(0.02)	(0.01)	(0.02)	(0.01)	(0.02)
AapGRF100$^{\bullet}$	0.50	0.44	0.49	0.42	0.51	0.46
	(0.11)	(0.15)	(0.11)	(0.16)	(0.11)	(0.15)
AapGRF50$^{\blacksquare}$	0.49	0.38	0.48	0.37	0.50	0.38
	(0.12)	(0.15)	(0.13)	(0.15)	(0.12)	(0.15)
PvGRF100	2.08	1.90	1.94	1.81	2.24	1.99
	(0.53)	(0.63)	(0.52)	(0.56)	(0.51)	(0.71)
PvGRF100time$^{\bullet}$	0.03	0.04	0.03	0.04	0.02	0.04
	(0.01)	(0.02)	(0.01)	(0.02)	(0.00)	(0.02)
AvGRF100$^{\bullet}$	0.92	1.01	0.92	1.02	0.92	1.00
	(0.13)	(0.32)	(0.14)	(0.34)	(0.13)	(0.32)
AvGRF50$^{\bullet}$	1.03	1.01	1.01	1.02	1.05	1.00
	(0.20)	(0.32)	(0.21)	(0.34)	(0.18)	(0.32)

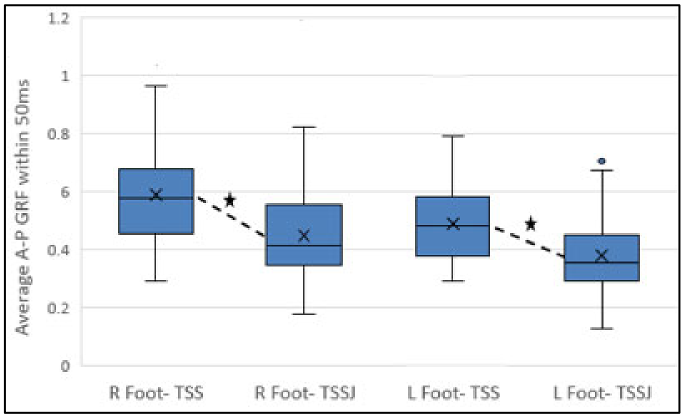

Figure 4.6: Average Anterior-Posterior (A-P) GRF within 50ms of Initial Foot Contact. *Average GRF (BWT) for each limb and for each movement for the whole group of subjects.*

The peak vertical GRF within the first 100ms upon impact (PvGRF100) was not different between the movements, limbs, or sexes (*main effect for movement and limb p=0.003, main effect for sex p=0.210*) [*Figure 4.7*] [*Table 4.4*]. The time when the PvGRF100 occurred (PvGRF100time) was later for the TSSJ compared with the TSS for both limbs but was not different between limbs or sexes (*main effect for movement and limb p<0.001, main effect for sex p=0.288*) [*Table 4.4*]. The average vertical GRF within the first 100ms upon impact (AvGRF100) was greater for the right foot than the left for both movements but there were no differences between the movements or the sexes (*main effect for movement and limb p<0.001, main effect for sex p=0.706*) [*Table 4.4*]. The AvGRF within the first 50ms (AvGRF50) was greater for the right limb than the left for the TSS only and there were no differences between the movements or sexes (*main effect for movement and limb p<0.001, main effect for sex p=0.447*) [*Figure 4.8*] [*Table 4.4*].

4.4.2 Rate of ground reaction force development

The 4x2 ANOVAs of the rate of force development from initial ground contact to peak GRF normalized to bodyweight within the first 100ms to compare the two movements (TSS and TSSJ), the two limbs (right and left), and the two sexes (male and female) revealed that there were differences between the movements but not the limbs for both the apGRF and the vGRF. The maximum rate of apGRF development within the first 100ms after impact (MaxRateapGRF) was greater for the TSS compared with the TSSJ for both limbs and was greater for males than females but was not different between limbs (*main effect for movement and limb p<0.001, main effect for sex p=0.005*) [*Figure 4.9*] [*Table 4.5*]. The time when MaxRateapGRF occurred (MaxRateapGRFtime) was not different between movements or limbs but occurred later for males than females (*main effect for movement and limb p=0.112, main effect for sex p=0.016*) [*Table 4.5*].

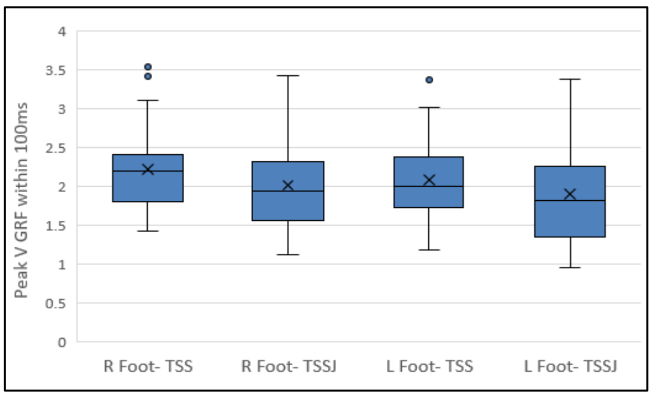

Figure 4.7: Peak Vertical (V) GRF within 100ms of Initial Foot Contact. *Maximal GRF (BWT) for each limb and for each movement for the whole group of subjects.*

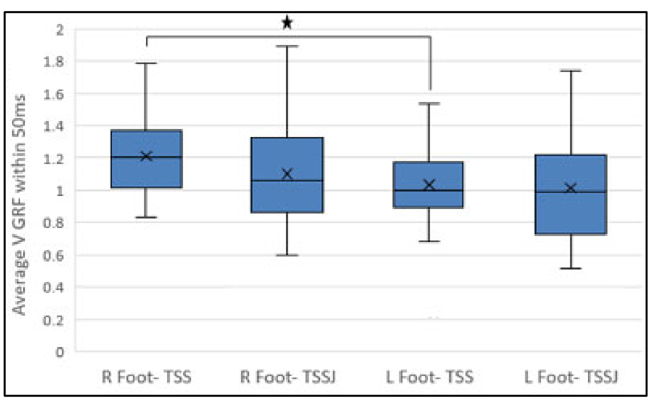

Figure 4.8: Average Vertical (V) GRF within 50ms of Initial Foot Contact. *Average GRF (BWT) for each limb and for each movement for the whole group of subjects.*

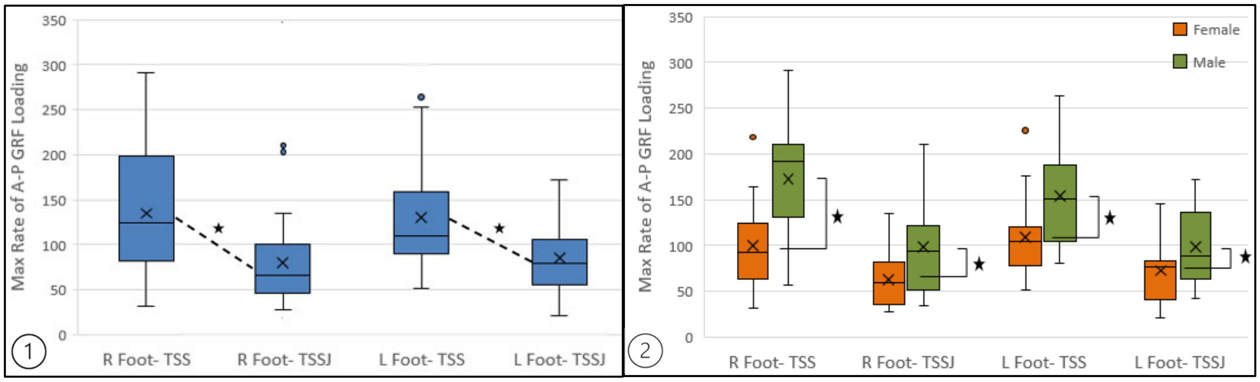

Figure 4.9: Rate of Anterior-Posterior (A-P) GRF Loading within 100ms of Foot Contact. *Maximal loading rate (s^{-1}) from initial foot contact to peak apGRF (BWT) for each limb and for each movement for the whole group of subjects (1) and separated by sex (2).*

Table 4.5: Maximum Rate of GRF Loading Normalized to Bodyweight within 100ms of Initial Foot Contact. *Mean and (standard deviation) for when each foot made ground contact. ap = anterior-posterior, v = vertical. Units: rate in s^{-1}, time in s. ●main effect for movement and limb $p<0.05$, Δmain effect for sex $p<0.05$, ■Friedman's main effect for movement and limb $p<0.05$, ◊independent samples median test for sex $p<0.05$ (see text for further detail).*

	Whole Group		Female		Male	
Right	**TSS**	**TSSJ**	**TSS**	**TSSJ**	**TSS**	**TSSJ**
MaxRateapGRF●Δ	136.1	80.2	99.5	63.0	175.4	98.7
	(67.8)	(46.7)	(48.4)	(30.7)	(64.7)	(54.5)
MaxRateapGRFtimeΔ	0.022	0.020	0.020	0.016	0.024	0.024
	(0.007)	(0.009)	(0.006)	(0.005)	(0.008)	(0.010)
MaxRatevGRF■◊	221.5	165.7	186.7	157.8	258.9	174.2
	(77.9)	(74.7)	(72.0)	(89.1)	(67.7)	(57.5)
MaxRatevGRFtime	0.018	0.018	0.018	0.018	0.018	0.018
	(0.005)	(0.006)	(0.005)	(0.006)	(0.005)	(0.007)
Left	**TSS**	**TSSJ**	**TSS**	**TSSJ**	**TSS**	**TSSJ**
MaxRateapGRF●Δ	130.4	85.1	109.3	72.8	153.1	98.3
	(54.8)	(39.4)	(44.6)	(32.5)	(57.1)	(42.9)
MaxRateapGRFtimeΔ	0.020	0.018	0.018	0.016	0.022	0.020
	(0.007)	(0.008)	(0.007)	(0.007)	(0.007)	(0.010)
MaxRatevGRF■	215.2	167.2	185.2	144.6	247.3	191.5
	(82.3)	(90.6)	(73.1)	(68.6)	(81.6)	(106.7)
MaxRatevGRFtime	0.017	0.016	0.016	0.017	0.017	0.016
	(0.004)	(0.008)	(0.004)	(0.009)	(0.003)	(0.007)

The maximum rate of vGRF development within the first 100ms after impact (MaxRatevGRF) was not normally distributed across both movements for each limb. The MaxRatevGRF was greater for the TSS compared with the TSSJ for both limbs, was not different between limbs, and was greater for males than females for the TSS for the right limb only (*Friedman's main effect between movement and limb p<0.001, independent samples median test for sex for R-TSS p=0.027, all else p ≥ 0.143)* [*Table 4.5*].

The 4x2 ANOVA of the time when MaxRatevGRF occurred (MaxRatevGRFtime) was not different between movements, limbs, or sexes (*main effect for movement and limb p=0.684, main effect for sex p=0.967*) [*Table 4.5*].

4.5 Required Coefficient of Friction:

The 4x2 ANOVAs of the required coefficient of friction within the first 100ms of initial foot impact (RCOF) revealed that the RCOF was greater for the TSS than the TSSJ for both limbs, was greater for the right limb compared with the left for the TSSJ and was greater for males than females (*main effect for movement and limb p<0.001, main effect for sex p<0.001*) [*Figure 4.10*]. The time when RCOF occurred (RCOFtime) was later for the TSS than the TSSJ for the left limb only and there were no other differences between limbs or sex (*main effect for movement and limb p=0.002, main effect for sex p=0.760*) [*Figure 4.11*].

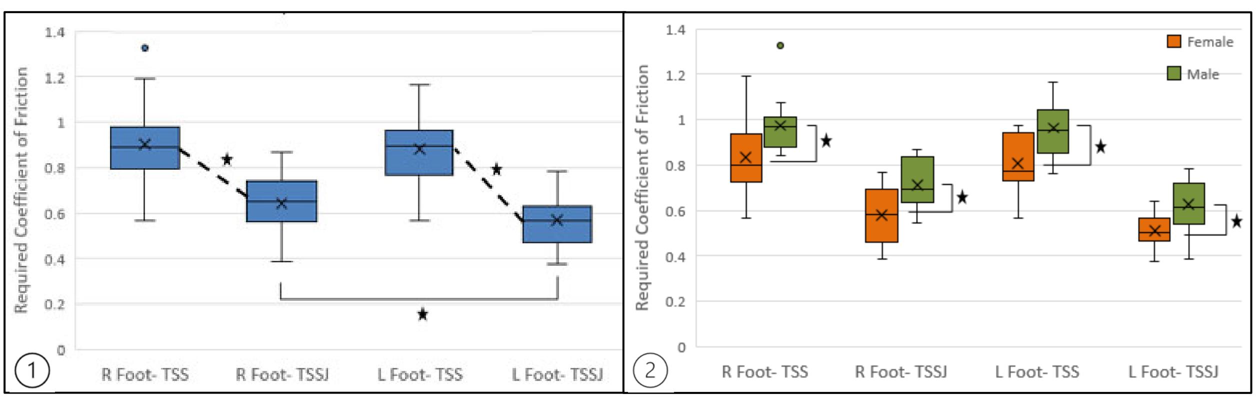

Figure 4.10: Required Coefficient of Friction within 100ms of Foot Contact. *The maximal ratio of vGRF to apGRF for each limb and movement for whole group of subjects (1) and separated by sex (2).*

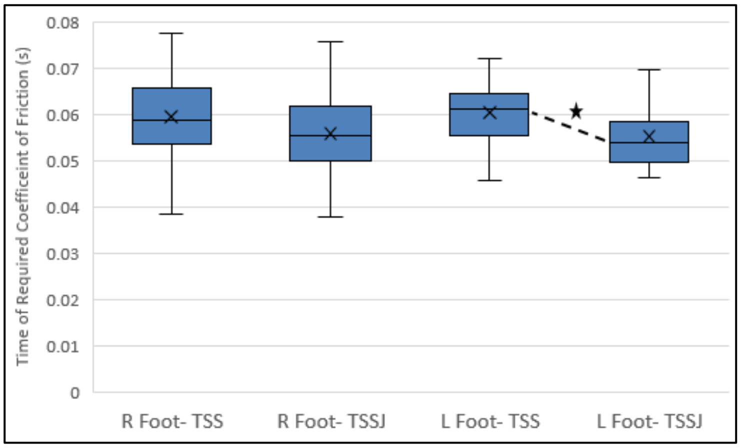

Figure 4.11: Time of Maximal Required Coefficient of Friction*. Time (s) within first 100ms of foot contact to occurrence of maximal RCOF for each limb and movement for the whole group of subjects.*

4.6 Correlations with Initial Foot Progression Angle Upon Impact:

The relationships between iFtProgAng and traction were explored with Pearson's Correlation (r) using individual trials to assess the whole group as well as within each sex [*Appendix Table 2*]. The small number of trials with forefoot initial contact were removed from this portion of the analysis. Moderate correlations, r-values between 0.5 and 0.7, with iFtProgAng were seen for the TSS upon right foot impact with the PapGRF100time and the iFtCOMvVel for the whole group, females, and males. For the TSSJ upon right foot impact, moderate correlations with iFtProgAng were found with the time to FFE for the whole group, females, and males. Moderate correlations were also found with the iCOMap and iFtCOMapVel for females only for both the TSS and the TSSJ.

Upon left foot impact for the TSS, moderate correlations with iFtProgAng were observed with the PapGRF100time for the whole group, females, and males, with the time to FFE for females and males, with the iFtCOMapVel for the whole group, and with the iFtCOMvVel for the whole group and males. For the TSSJ upon left foot impact, moderate correlations with iFtProgAng were seen with the iCOMap and iFtCOMapVel for the whole group and females, with the time to FFE for the whole group and males, and with the iCOMvVel and postFFEFtCOMap for males only.

Moderately strong correlations, defined as r-values between 0.7 and 0.8, and strong correlations, r-values greater than 0.8, for both the right and left limbs upon impact, for both movements, across the whole group, females, and males were found with the preFFEFtCOMap [*Figure 4.12*].

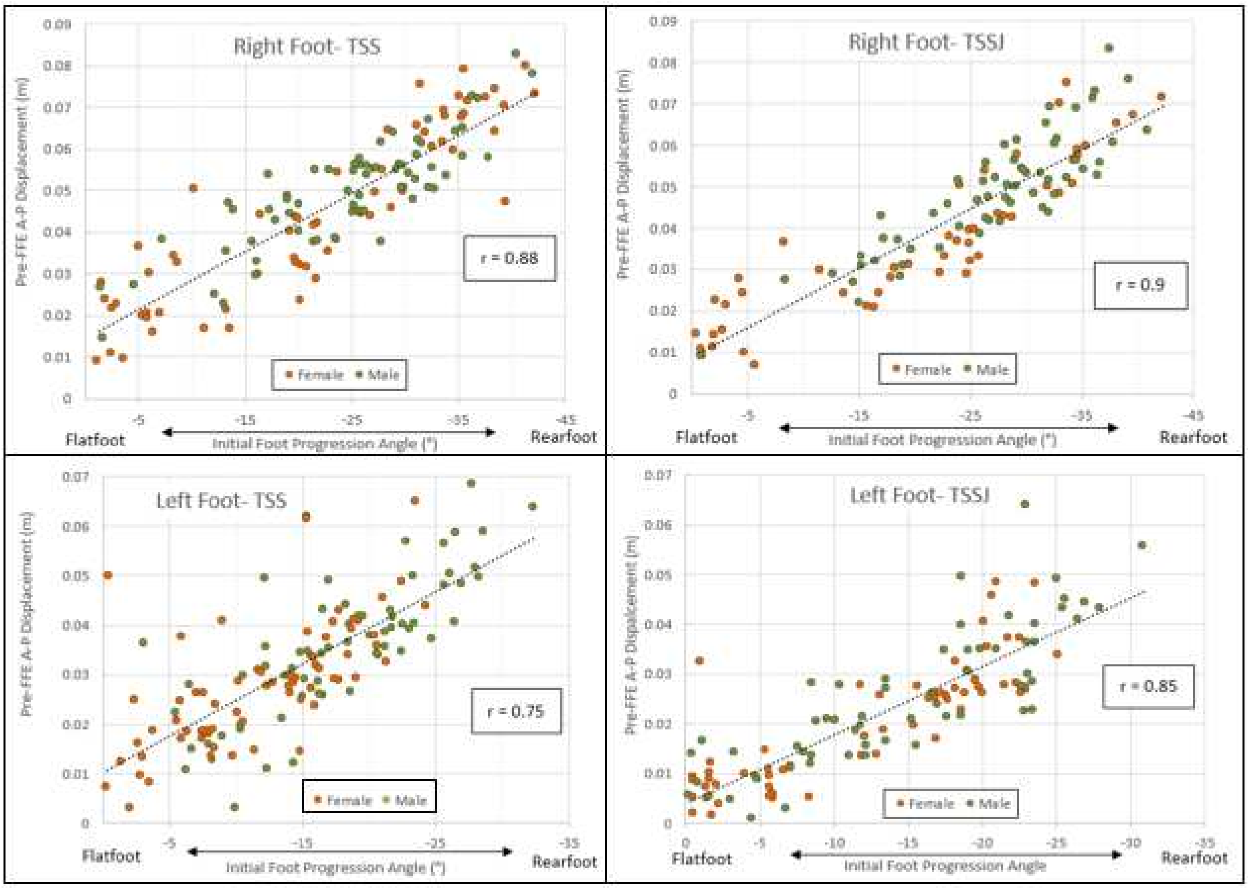

***Figure 4.12**: **Pre-Full Foot Engagement (FFE) Anterior-Posterior (A-P) Displacement Correlated with Initial Foot Progression Angle.** Moderately strong to strong correlations between preFFEFtCOMap and iFtProgAng for each limb, for each movement.*

CHAPTER 5
DISCUSSION

5.1 Study Aims:

The primary goal of this investigation was to examine how the foot progression angle upon landing on artificial turf (AT) during stop and stop-jump tasks influences traction and foot loading characteristics. Secondary goals were to investigate differences in landing strategy between males and females and the effect of subsequent task demands between the two movements. Overall, our findings suggest that landing foot orientation contributes to some linear traction characteristics on AT but does not differ between males and females and does not influence landing phase ground reaction forces (GRFs).

5.1.1 Aim 1: Differences in foot orientation upon landing

Our first hypothesis, that a flatfoot orientation upon landing will result in greater peak GRFs and traction compared with an initial rearfoot landing, is partially supported by our findings. The foot progression angle influences some traction characteristics, but not GRFs.

For both movements, the right foot was oriented for a greater rearfoot landing, a more negative foot progression angle, than the left foot. When oriented in a more rearfoot landing pattern at impact, the foot took a longer time to reach full foot engagement (FFE) and had greater horizontal displacement before FFE than when the foot had a more flatfoot orientation at impact. There were moderately strong (r-value greater than .7) to strong (r-value greater than .8) correlations between each limb's foot progression angle at impact and the foot's pre-FFE horizontal displacement. For the right foot of the two-step jump, this correlation value was high enough to be predictive. The correlation was weak (r-value less than 0.5) between foot orientation at impact and the post FFE displacement.

Assessing the horizontal displacement of the cleat across the AT surface is critical to understanding the balance between a surface that provides for high enough traction to maximize athlete performance, but not so high as to contribute to footlock (Cawley et al., 2003). The current study's findings suggest that foot orientation highly contributes to the displacement of the foot until the cleat can fully engage with the AT. Likely, the studs on the heel of the cleat provide little resistance as the foot moves through the AT in a rearfoot landing. This displacement occurs within 0.06 seconds after initial foot contact; so quickly, that there is limited ability in this period to decelerate the whole body prior to FFE. However, it remains unclear to what extent this linear movement of the foot before FFE contributes to performance or injury risk. Further analysis of joint kinematics and kinetics is needed to investigate whether the slightly greater time to FFE is enough to alter joint loading mechanics.

It is important to note that the differences in GRFs during the landing phase between the limbs may be attributed more to the differences in timing of impact, rather than the landing foot orientation, because the right foot impacted the surface slightly before the left. There was a weak correlation for the whole group of subjects between foot orientation at impact and horizontal or vertical GRFs. While it does not appear that foot orientation at impact is associated with GRFs, it is possible that whole-body COM velocity, foot COM velocity, or foot progression angular velocity at impact may be more strongly correlated with initial GRFs. Achieving higher velocities before impact would result in greater landing forces as discussed below between male and female peak horizontal GRFs at impact. In addition, the whole-body COM positions were different between the two limbs. Both the horizontal and vertical positions were further away from the force platform when the right foot made contact compared with the left so position may also be more strongly correlated with GRF at impact.

5.1.2 Aim 2: Differences between males and females

Our findings did not support our second hypothesis, that females land with a greater rearfoot orientation than males. We found no differences in the foot progression angle at impact between sexes. However, it should be noted that while all subjects met the inclusion criteria to participate, males and females were not matched, allowing for potential group differences to occur.

The noted differences in whole-body position at impact between males and females, that males were further behind and higher above the foot at ground contact, can be attributed to the taller heights and longer body segments of the male subjects. In addition, males were generally moving faster than females at initial ground contact. While males had greater velocity at impact, the horizontal displacement was not different before FFE. After FFE, however, males did have greater displacement after FFE compared with females, likely attributed to the different absolute horizontal GRFs between sexes. The bodyweight normalized average horizontal GRFs were not different between males and females, however, the peak horizontal GRF, the rate of horizontal force loading, and the RCOF were greater for males. Together, this suggests that traction needs are higher in males, leading to the greater horizontal displacement after FFE.

Boden et al. (2009) found through the analytic review of video-recorded anterior-cruciate ligament (ACL) injuries that female athletes were more commonly injured during simple deceleration maneuvers, such as stop tasks, whereas male athletes were primarily injured during more strenuous jumping tasks. However, we did not find differences between sexes for the primarily rearfoot landing orientations in the two movements that would suggest this finding. While the landing mechanics have been shown to be different between males and females across various stop-jump tasks, from the current study, it does not appear that these differences include landing foot orientation (Butler et al., 2013; Chappell et al., 2002; Grund et al., 2007; Grund & Senner, 2006; Yin et al., 2015). There were also weak correlations between foot orientation and horizontal or vertical GRFs. Accordingly, foot orientation at landing does not explain the difference in peak horizontal GRF between sexes. Again, the greater whole-body and foot COM velocities at impact, as achieved by males, is likely more highly associated with achieving greater horizontal GRFs than landing foot orientation.

5.1.3 Aim 3: Differences between stop and stop-jump tasks

Our third hypothesis, that differences in task demands between the two movements will result in greater initial landing rearfoot orientation, foot loading, and traction characteristics for the stop task compared with the stop-jump task is partially supported by our findings. At impact, the foot orientation was not different between the two movements, although the horizontal foot loading and traction forces were greater for the two-step stop (TSS) than for the two-step stop-jump (TSSJ).

While we did not find statistically significant differences in foot orientation at impact between the two movements, the stop task did trend towards a more rearfoot landing orientation than the TSSJ. Donnelly et al. (2017) showed that more rearfoot landing orientations place greater mechanical demand on the knee joint and could affect lower extremity injury. In the current study, there was a large amount of variation for both movements and a greater amount of variation for the TSSJ compared with the TSS. This high variability could have prevented the difference between the movements from being statistically significant. Hass et al. (2003) reported high ankle angle variability at ground contact for a stop task, but not for a stop-jump task. The variability within the stop task likely prevented their approximately thirteen-degree difference between the two movements from being statistically significant (Hass et al., 2003).

Although the subjects' whole-body COM velocity was not different between the movements at foot contact, compared with the TSSJ, the foot COM had a greater horizontal velocity and foot progression angular velocity for the TSS. Then, once the foot contacted the surface, the horizontal displacement before FFE was greater for the TSS suggesting that foot COM velocity influences this displacement in addition to the horizontal GRFs as discussed above. There were no differences in when FFE occurred between the two movements.

Importantly, the horizontal foot loading forces were different between the two movements, but the vertical forces were not different. Our findings of greater horizontal peak and average ground reaction forces for the stop task compared with the stop-jump task are in agreement with Hass et al. (2003). We also found greater RCOF for the TSS compared with the TSSJ and the rates of horizontal and vertical force loading were also greater for the TSS compared with the TSSJ. This suggests that there is a greater need for traction for the TSS to halt forward movement than for the TSSJ.

It seems likely the subsequent task demands of the stop-jump task compared with the stop task could influence the horizontal loading forces across the foot at impact. Because non-contact lower extremity injuries, such as ACL ruptures, have been associated with greater landing GRFs, caution is warranted when performing the "simple" stop task (Yu, 2006). While only a trend was shown here, and not a statistical difference between the two movements, if the greater GRFs associated with stop tasks are coupled with rearfoot orientation upon landing, even greater injury risk would be expected for this movement.

5.2 Limitations and Future Directions:

One of the limitations in the current study was the dimensions of the laboratory in relation to the in-ground force platforms. Athletes only had a maximum distance from the wall to the force platforms of 2.7 meters to perform two steps before their bilateral landing. This required the taller athletes to adapt the stride length of their first step more than the shorter subjects to fit in the required two steps. This space limitation could have influenced subjects' ability to gain maximal horizontal velocity. Future work should consider the physical constraints of the lab and possibly limit the required lead-in to a single step or a modified stride-jump technique. The two-step approach was utilized in the current study to have subjects gain maximal horizontal velocity before targeting the force platform. Limiting the lead-in to one step in future work may still allow athletes to gain enough forward momentum and to be more consistent from trial to trial. In the current study, we reported wide variability in the initial foot progression angle both within and between subjects, which could have prevented differences between males and females or between the movements from being statistically significant. However, the two-step approach did provide similar whole-body COM velocities between the TSS and TSSJ, allowing for an equitable comparison between the two movements at impact.

While the movements performed in the current study were familiar to our subjects, our subjects were not necessarily accustomed to performing these types of athletic movements while targeting a specific landing zone. Athletes may have modified their technique for the lab setting and this could have introduced unintended variability. Variability reduces the ability to detect statistical significance and should be limited. Even though the subjects' first visit to the laboratory was one to primarily gain familiarization with the lab setting and to practice the dynamic movements, it is conceivable that this still did not allow for enough practice to gain complete familiarization with the movements while targeting the force platforms before data collection on the second visit. Possibly decreasing the time between the first and second visit as scheduling allows would also assist subjects in the familiarization process.

To increase subject comfort level and to create more generalizable results, our subjects wore their own cleats they were accustomed to playing in for their sport. This led to a wide range in cleat type by sport, stud shape, and stud configuration across the twenty-nine subjects, as each pair of cleats was unique. The goal was that this would also be most realistic to what is seen in competition on AT. We therefore cannot make any conclusions on the influence of the cleat itself on traction characteristics. Because the stud design and configuration are known to influence traction in mechanical studies, future work should investigate the effect of stud shape and configuration on traction and control for cleat variations in human performance studies (Driscoll et al., 2015; Kirk et al., 2007). To better assess traction and injury risk in future mechanical studies, using a linear traction device to represent the landing phase would be more relevant because many of the linear traction studies involve the movement of the shoe to mimic take-off. In addition, our findings suggest that a linear traction device would not need to replicate various foot progression angles; a flatfoot orientation should suffice.

Another limitation in the current study was that athletes primarily landed with a rearfoot to flatfoot orientation for the two movements analyzed; the few forefoot landings were removed from the correlation analysis. As such, the range of possible foot orientations upon landing was limited without comparison to forefoot landings. Boden et al. (2010) found that athletes who reached a flatfoot position sooner after landing increased their lower extremity injury risk due to the limited time for the triceps surae musculature to contract and absorb forces, increasing the impulsive force to the knee. They concluded that a forefoot landing would be safest since this would give the soft tissue components of the leg time to dampen the force (Boden et. al., 2010). While increasing the time for the musculature to absorb GRFs upon impact until a flatfoot position is reached, injury risk also depends on whether the musculature is already contracted at the time of impact to compress and stabilize the joint (Morgan et al., 2014). In rearfoot to flatfoot orientations, the triceps surae musculature is not as active or producing as much force as it is for a forefoot landing (Morgan et al., 2014). Future work should incorporate movements with the full range of foot progression angles upon landing to investigate the role of foot orientation and foot loading characteristics. It is highly possible that a forefoot landing orientation could influence the loading characteristics to a greater degree than the more passive rearfoot landing primarily studied here.

Other future work should focus on athletic movements with unilateral, rather than bilateral landings. The bilateral landings assessed here are commonly used in current biomechanical tests to screen for high-risk landing patterns related to ACL injury (Cruz et al., 2013). However, single-leg landings result in more lower extremity injuries than the more stable double-leg landings (Koga et al., 2010; Taylor et al., 2016). Single-leg landings also result in greater GRFs than double-leg landings early in the landing phase (Taylor et al., 2016). Assessing the foot's orientation at impact for a unilateral landing in the sagittal plane could yield relevant insight into the role foot orientation plays in traction and injury risk. In addition to landing from a jump, pivoting and cutting movements are highly associated with lower extremity injury (Shimokochi & Shultz, 2008). Specific to ACL injury, incidence rates are highest in multidirectional sports with repetitive movements beyond the sagittal plane, such as lateral shuffling and cutting (Boden et al., 2009; Shimokochi & Shultz, 2008). Therefore, future work should investigate the medial-lateral and rotational traction characteristics as they pertain to the landing foot progression angle and injury risk.

5.3 Conclusion:

This investigation was especially novel as most reported literature on foot orientation has been conducted on hardcourt surfaces, not on AT. Overall, our findings suggest that rearfoot to flatfoot landing foot orientations contribute to some linear traction characteristics on AT, but do not differ between males and females. Additionally, there may not be an association between landing foot orientation and foot loading forces when the initial foot contact is by the rearfoot. The factor most strongly correlated with foot progression angle upon impact is the horizontal foot COM displacement before FFE. In addition to assessing forefoot impacts, cleat design, and unilateral tasks, future work is needed to determine if this horizontal movement across the surface contributes to injury risk by examining the joint loading mechanics at the ankle, knee, and hip. This information could further contribute to modifications in athlete technique, cleat design, and surface characteristics to optimize athlete performance and to reduce injury risk.

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APPENDIX

Appendix Table 1: Indicators for Trial Selection. *Mean (and standard deviation) for landing characteristics of initial (i) ground contact by the right foot used in the selection of the five trials for each subject. ap = anterior-posterior, v = vertical, Vel = velocity, AngVel = angular velocity. Units: positions in m, velocities in m/s, angular positions in °, angular velocities in °/s. (See text for further detail).*

Right Limb	Whole Group		Female		Male	
	TSS	TSSJ	TSS	TSSJ	TSS	TSSJ
iCOMap	-0.42	-0.39	-0.39	-0.34	-0.45	-0.44
	(0.05)	(0.06)	(0.05)	(0.04)	(0.04)	(0.05)
iCOMapVel	2.25	2.24	2.13	2.07	2.37	2.42
	(0.24)	(0.27)	(0.21)	(0.19)	(0.20)	(0.23)
iCOMv	0.81	0.83	0.80	0.81	0.83	0.85
	(0.04)	(0.04)	(0.05)	(0.04)	(0.02)	(0.03)
iCOMvVel	-1.38	-1.42	-1.33	-1.43	-1.43	-1.40
	(0.21)	(0.19)	(0.24)	(0.20)	(0.16)	(0.18)
iFtCOMapVel	1.86	1.48	1.67	1.24	2.07	1.74
	(0.53)	(0.46)	(0.50)	(0.34)	(0.49)	(0.44)
iFtCOMvVel	-1.22	-1.15	-1.06	-1.04	-1.39	-1.27
	(0.26)	(0.22)	(0.18)	(0.16)	(0.24)	(0.22)
iFtProgAng	-20.69	-16.69	-18.09	-11.97	-23.49	-21.74
	(11.43)	(15.00)	(12.97)	(14.14)	(9.16)	(14.70)
iFtProgAngVel	378.78	232.58	276.31	136.64	488.56	335.38
	(250.50)	(247.63)	(235.90)	(198.05)	(223.90)	(260.45)
iAnkleAng	-3.02	-2.33	-4.04	-5.65	-1.93	1.23
	(9.18)	(14.27)	(12.12)	(13.94)	(4.55)	(14.24)
iAnkleAngVel	-258	-59	-131	61	-394	-188
	(300)	(297)	(283)	(233)	(263)	(311)
iKneeAng	27.90	31.86	28.11	32.40	27.67	31.29
	(9.35)	(9.22)	(7.71)	(8.23)	(11.15)	(10.47)
iKneeAngVel	183	229	210	271	155	184
	(125)	(110)	(91)	(75)	(152)	(125)
iHipAng	52.29	56.58	51.16	57.09	53.49	56.03
	(10.40)	(8.81)	(13.69)	(10.65)	(5.30)	(6.67)
iHipAngVel	4	32	26	57	-19	5
	(77)	(73)	(75)	(53)	(72)	(82)

Appendix Table 2: Landing Characteristics Upon Impact Correlated with the Initial (i) Foot Progression Angle. *Correlation values (r) and significant values (p). Shaded r values indicate moderately strong (r >0.7) to strong (r >0.8) correlations. ap = anterior-posterior, v = vertical, Vel = velocity, FFE = full foot engagement, P = peak, A = average, RCOF = required coefficient of friction*

RIGHT FOOT	**TSS**						**TSSJ**					
	Whole Group		**Female**		**Male**		**Whole Group**		**Female**		**Male**	
	r	p	r	p	r	p	r	p	r	p	r	p
Bodyweight	0.12	0.18	0.03	0.80	0.09	0.49	0.30	0.00	0.18	0.21	0.11	0.40
iCOMap	0.47	0.00	0.50	0.00	0.42	0.00	0.48	0.00	0.52	0.00	0.35	0.01
iCOMapVel	0.20	0.02	0.19	0.13	0.08	0.50	0.13	0.19	0.13	0.37	0.04	0.74
iCOMv	0.06	0.50	0.20	0.12	0.04	0.76	0.17	0.08	0.01	0.93	0.08	0.57
iCOMvVel	0.50	0.00	0.52	0.00	0.53	0.00	0.38	0.00	0.42	0.00	0.18	0.16
iFtCOMapVel	0.36	0.00	0.51	0.00	0.07	0.59	0.47	0.00	0.52	0.00	0.21	0.10
iFtCOMvVel	0.50	0.00	0.52	0.00	0.53	0.00	0.38	0.00	0.42	0.00	0.18	0.16
Time to FFE	0.33	0.00	0.35	0.00	0.47	0.04	0.51	0.00	0.58	0.00	0.63	0.00
preFFEFtCOMap	0.88	0.00	0.90	0.00	0.85	0.00	0.90	0.00	0.90	0.00	0.88	0.00
postFFEFtCOMap	0.10	0.27	0.17	0.18	0.07	0.58	0.13	0.18	0.06	0.68	0.04	0.78
PapGRF100	0.09	0.28	0.05	0.67	0.01	0.92	0.05	0.60	0.37	0.01	0.06	0.62
PapGRF100time	0.61	0.00	0.65	0.00	0.64	0.00	0.36	0.00	0.45	0.00	0.24	0.06
AapGRF100	0.05	0.53	0.06	0.65	0.17	0.17	0.09	0.35	0.31	0.03	0.14	0.30
AapGRF50	0.15	0.09	0.22	0.07	0.21	0.10	0.23	0.02	0.42	0.00	0.20	0.12
PvGRF100	0.13	0.14	0.15	0.23	0.32	0.01	0.12	0.22	0.13	0.36	0.30	0.02
PvGRF100time	0.01	0.90	0.02	0.86	0.12	0.35	0.16	0.09	0.26	0.07	0.04	0.78
AvGRF100	0.10	0.24	0.17	0.18	0.05	0.70	0.03	0.76	0.04	0.79	0.06	0.64
AGRF50	0.06	0.50	0.05	0.68	0.07	0.60	0.04	0.71	0.05	0.75	0.13	0.33
MaxRateapGRF	0.07	0.42	0.07	0.58	0.05	0.69	0.08	0.41	0.30	0.04	0.19	0.16
MaxRatevGRF	0.03	0.75	0.02	0.87	0.15	0.23	0.00	1.00	0.04	0.78	0.23	0.08
MaxapRCOF	0.03	0.71	0.04	0.76	0.05	0.66	0.03	0.78	0.11	0.44	0.21	0.10
LEFT FOOT	**TSS**						**TSSJ**					
	Whole Group		**Female**		**Male**		**Whole Group**		**Female**		**Male**	
	r	p	r	p	r	p	r	p	r	p	r	p
Bodyweight	0.39	0.00	0.02	0.89	0.26	0.04	0.19	0.04	0.14	0.30	0.29	0.02
iCOMap	0.43	0.00	0.23	0.06	0.34	0.01	0.57	0.00	0.69	0.00	0.49	0.00
iCOMapVel	0.39	0.00	0.28	0.02	0.17	0.17	0.22	0.01	0.22	0.09	0.11	0.41
iCOMv	0.01	0.93	0.38	0.00	0.03	0.80	0.09	0.35	0.18	0.17	0.31	0.02
iCOMvVel	0.09	0.29	0.15	0.21	0.19	0.14	0.22	0.01	0.14	0.29	0.59	0.00
iFtCOMapVel	0.53	0.00	0.43	0.00	0.41	0.00	0.56	0.00	0.65	0.00	0.45	0.00
iFtCOMvVel	0.57	0.00	0.38	0.00	0.50	0.00	0.39	0.00	0.25	0.05	0.48	0.00
Time to FFE	0.42	0.00	0.55	0.00	0.60	0.00	0.51	0.00	0.48	0.00	0.63	0.00
preFFEFtCOMap	0.75	0.00	0.68	0.00	0.76	0.00	0.85	0.00	0.85	0.00	0.84	0.00
postFFEFtCOMap	0.31	0.00	0.28	0.02	0.11	0.40	0.32	0.00	0.07	0.57	0.52	0.00

PapGRF100	0.00	1.00	0.12	0.34	0.18	0.16	0.29	0.00	0.18	0.17	0.50	0.00
PapGRF100time	0.67	0.00	0.66	0.00	0.63	0.00	0.31	0.00	0.28	0.03	0.34	0.01
AapGRF100	0.07	0.43	0.08	0.53	0.19	0.12	0.23	0.01	0.12	0.38	0.41	0.00
AapGRF50	0.14	0.12	0.16	0.21	0.23	0.07	0.29	0.00	0.13	0.34	0.48	0.00
PvGRF100	0.05	0.61	0.07	0.57	0.31	0.01	0.24	0.01	0.16	0.22	0.40	0.00
PvGRF100time	0.35	0.00	0.32	0.01	0.48	0.00	0.10	0.26	0.11	0.42	0.37	0.00
AvGRF100	0.13	0.15	0.14	0.25	0.17	0.18	0.20	0.03	0.07	0.59	0.35	0.01
AGRF50	0.02	0.79	0.06	0.64	0.08	0.51	0.20	0.03	0.07	0.59	0.35	0.01
MaxRateapGRF	0.20	0.02	0.05	0.66	0.05	0.69	0.13	0.14	0.05	0.71	0.30	0.02
MaxRatevGRF	0.01	0.87	0.01	0.92	0.32	0.01	0.20	0.03	0.02	0.86	0.45	0.00
MaxapRCOF	0.21	0.01	0.07	0.55	0.04	0.77	0.03	0.78	0.21	0.11	0.25	0.05