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Biomechanical Adaptations While Performing Bilateral Drop Landings With a Unilateral Ankle Tape Application

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BIOMECHANICAL ADAPTATIONS WHILE PERFORMING BILATERAL DROP LANDINGS WITH A UNILATERAL ANKLE TAPE APPLICATION

by

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ABSTRACT

BIOMECHANICAL ADAPTATIONS WHILE PERFORMING BILATERAL DROP LANDINGS WITH A UNILATERAL ANKLE TAPE APPLICATION

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Ankle sprains are the most common injury in sport and exercise performance, which makes the utilization of ankle taping a common procedure to both prevent potential sprain, as well as protect against reinjury. However, unilateral ankle taping may have unintended consequences on the mechanics of the ankle and other joints of both legs. The aim of this dissertation was to determine the effects of ankle taping on lower body kinetics and kinematics, stiffness, and coordination during a bilateral landing task.

Twelve female participants completed a total of 90 drop landings across two visits, randomized from landing platforms of 30, 45, and 60 cm while the dominant ankle was taped, and not taped on separate visits, before and after exercise. The three aims of this study are to address traditional landing biomechanics, lower-body, limb, and joint stiffnesses, and coordination and coordination variability.

Taping the dominant ankle reduced plantarflexion at contact and increased loading rate; however, the only effects of tape on joint moments were a decreased dominant knee peak flexion moment and an increased non-dominant ankle peak dorsiflexion moment. Exercise mitigated the effect of the tape on the non-dominant ankle but increased the peak hip flexion moment of the non-dominant leg, whether the ankle was taped or not.

Lower-body stiffness was not changed with platform height or the addition of unilateral ankle taping, suggesting a preferred lower-body stiffness for landing. Although, as height increased dominant limb stiffness decreased while non-dominant limb stiffness increased, potentially increasing risk for injury from the asymmetry. Taping the dominant ankle resulted in increased ipsilateral ankle and hip stiffnesses, but not in the knee, which might be a protective behavior for the knee, but increases the risk for injury in the ankle and hip.

There were minimal changes to kinetic and kinematic interlimb coordination of the legs from the unilateral ankle tape application, suggesting that coordination while landing is robust. The non-dominant hip was involved in all significant differences, suggesting an important role while landing.

These investigations highlight complex compensatory effects that unilateral ankle taping, exercise, and different heights has on landing mechanics of both legs.

This dissertation is dedicated to all my family and friends who have been there for me along the way; and my gratitude expressed to those below is in no particular order. My girlfriend Erin has been there for me every day throughout this journey. Without her support throughout this process, I would not have been able to get this far. For all your years of love and support, I am eternally grateful.

I would also like to express my appreciation to my family, who were always available to provide support and homemade meals that would always lift my spirits. My mother, Deanna, would take time out of her day to meet me for lunch and offer me a refuge from the demands of daily life. Throughout my educational journey from WVU to UTA, and finally to ODU and beyond, my father, Danny, was always there to visit for sporting events, and offer moving assistance which provided memories I will never forget. I would also like to thank my siblings, Katelyn and Dylan, whose academic achievements provided additional motivation to continue the pursuit of my own.

Lastly, I want to dedicate this work to my grandfather, Linwood Regan, who sadly passed away prior to the completion of this dissertation. He would always ask how I was doing, how things were going, and provide encouragement. Of the many lessons I learned from my Gramps, what will always stick with me was that once you set out your goals, you follow through with them. You give it your best effort and don't cut corners, because even if others won't notice, you know the value of your work from start to finish.

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NOMENCLATURE

ACL	Anterior Cruciate Ligament of the Knee
ASIS	Anterior Superior Iliac Spine of the Pelvis
ATFL	Anterior Talofibular Ligament of the Ankle
BW	Body Weight, kg
COM	Center of Mass
CRP	Continuous Relative Phase, degrees
DOF	Degrees of Freedom
DRP	Discrete Relative Phase
EMG	Electromyography
GRF	Ground Reaction Force, N
Hz	Hertz – Measure of Frequency Per Second
k	Stiffness
kN	KiloNewtons, kN
MVIC	Maximal Voluntary Isometric Contraction
N	Newton, N
Nm	Newton·meter, Nm
PSIS	Posterior Superior Iliac Spine of the Pelvis
ROM	Range of Motion, degrees
RPE	Rated Perceived Exertion, (No Units)

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CHAPTER 1

INTRODUCTION

1.1: Purpose

Ankle sprains are the most common injury to occur during exercise or sport participation (D. T.-P. Fong, Chan, Mok, Yung, & Chan, 2009; Hootman, Dick, & Agel, 2007). To mitigate further injury or re-injury, the ankle is often taped, or a brace is worn (Kaminski et al., 2013). Both forms of prophylaxis have been found to reduce the risk of re-injury to the ankle (Dizon & Reyes, 2010). However, change at one joint can have a concomitant effect at another joint, but the effect of taping an ankle on other joints in the kinetic chain are inconclusive (Niu, Feng, Wang, Jiang, & Zhang, 2016). Furthermore, prophylactic taping or bracing an ankle could also have an impact on the joints of the contralateral leg, but this has received minimal consideration in the literature (Dewar, Arnold, Wang, Drew, & Abboud, 2019). As ankle sprains commonly occur during sport and exercise participation, and a common preventative and treatment method is taping of the ankle, this dissertation seeks to understand the biomechanical changes which occur as a result of taping an ankle while performing landing maneuvers.

While engaging in exercise or sport participation, ankle sprains are most often sustained during movements which result in the foot in a plantarflexed and inverted position such as performing a lateral cutting maneuver or landing from a jump (D. T.-P. Fong, Chan, et al., 2009). In a lateral ankle sprain, which is the most common type of ankle sprain, the damage to the ligaments of the lateral ankle joint often occurs during the initial loading after early contact, as the foot is in a plantarflexed and inverted position thus exposing the lateral ligaments of the ankle joint to higher forces (Konradsen, Voigt, & Hojsgaard, 1997). Drop landings, which consist of stepping off an elevated platform onto the ground, are commonly used to better understand the influence of ankle sprains as they are a consistent and standardized method which is easily repeatable between participants and visits (Edwards, Steele, & McGhee, 2010). Particularly when a constraint is applied unilaterally and direct comparisons across limbs is of primary importance, a bilateral drop landing has an advantage compared to cutting tasks where directionality is a major factor, as the force attenuation of the drop landing can be more equitably distributed to provide a better understanding of how each limb responds to performing the same task at the same time.

Bilateral landings are often performed by individuals participating in exercise and sporting activities. Because ankle sprains can occur during both single-leg and bilateral landings, a bilateral landing task, especially with unilateral taping can provide beneficial details about how the tape effects both legs simultaneously. Investigations that use ankle supports often apply the support to both ankles; or, when using a single-leg landing, only apply the support to the test ankle. However, many athletes elect to only tape one ankle and little is known about how a unilateral ankle support impacts both legs during bilateral landings. The body of research regarding effects on bilateral outcomes after a unilateral limb modification has not been researched with depth, and questions remain about the impact on other joints outside of the joint which undergoes an external modification. Therefore, it is important to properly identify the key drivers of these bilateral effects and to quantify those findings.

The effect of taping an ankle on bilateral landings and potential injury risk can be quantified through biomechanical measures of ground reaction force (GRF), joint kinematics and joint kinetics. Force plates provide information about the force exerted upon the ground and therefore on the body, which when combined with three-dimensional motion capture providing kinematic information can be used to compute joint kinetic measures. Maximum values during ground contact indicate the extremes of GRF, motion of a joint, or joint moment that the individual experiences. In addition to these standard biomechanical variables, the lower-body or each leg can be considered to act as a spring during the impact, which can be quantified as stiffness (= force/displacement) while joint stiffness acts as a quasi-stiffness measure (= moment/angular displacement) due to the indirect relationship between joint position and joint moment. Measures of stiffness of the body and joints are informative about how an athlete adapts for impact with different surfaces or changing task constraints, which could be altered by taping an ankle (Ferris & Farley, 1997; Serpell, Ball, Scarvell, & Smith, 2012; SA Williams & BL Riemann, 2009). Continuous coordination measures during a landing could provide insight into whether taping one ankle alters coordination between joints of the same leg or between joints of the two legs.

Because ankle taping is primarily utilized when performing physical activity, it is important to understand how a bout of exercise alters the tape itself, as well as its effects on lower limb biomechanics. Exercise has been shown to decrease the restrictive properties of ankle taping after as little as ten minutes of exercise; however, the residual effectiveness measured as the protective ability to resist inversion sprains is not clear, as some studies report adequate residual protection, while others note little remaining protective capabilities (Best, Mauch, Böhle, Huth, & Brüggemann, 2014; Ricard, Sherwood, Schulthies, & Knight, 2000; Tamura et al., 2017).

Appropriate control of the legs during landing is critical to ensuring the musculoskeletal system does not incur damage during high impact landings. Stiffness describes the body's resistance to experiencing an external load (R. J. Butler, Crowell, & Davis, 2003). With increasing force attenuation demands especially as driven by increasing landing platform height, the body must adapt through several mechanical changes to mitigate the risk of musculoskeletal damage. Too much stiffness places the body at a higher risk for bony injury, while too little stiffness poses a higher risk of soft tissue damage (R. J. Butler et al., 2003). Control of lower body stiffness is regulated both passively and actively through ligaments, cartilage, tendon, bone, while musculature is critical to mitigating the external forces incurred during landing (R. J. Butler et al., 2003; Latash & Zatsiorsky, 1993; Rapoport, Mizrahi, Kimmel, Verbitsky, & Isakov, 2003). Stiffness can also be regulated through external factors such as joint tape, braces, prosthetics, or footwear (Hobara et al., 2013; Takahashi, Gross, van Werkhoven, Piazza, & Sawicki, 2016). However, the influence of taping one ankle joint on joint stiffness of other joints of the same leg or contralateral leg, as well as overall leg or body stiffness is unknown. Therefore, it is necessary to examine stiffness changes in a bilateral landing task due to a taped ankle, and to consider how increased drop height interacts with taping.

Although athletic maneuvers include numerous actions such as running, jumping, landing, etc., the combinations of these movements can change the demands of the body to attenuate the forces of these movements. Studies that assess stiffness often utilize continuous or repetitive tasks such as walking, hopping, or running as opposed to a discrete landing task, which can have differing demands based on the task itself, as well as performing secondary movements (Hobara et al., 2013; Hobara, Hashizume, & Kobayashi, 2017; Jin & Hahn, 2018; Krupenevich, Clark, Sawicki, & Franz, 2020; Sami Kuitunen, Komi, & Kyröläinen, 2002; S Kuitunen, Ogiso, & Komi, 2011; Mason-Mackay, Whatman, Reid, & Lorimer, 2016). Depending on the task performed, each joint in the leg has been shown to act differently when modulating stiffness to suit the task parameters (Jin & Hahn, 2018; Mason-Mackay, Whatman, Reid, et al., 2016). As such, the body of literature which focuses solely on the compressive resistance of landing without a secondary motor task is limited and requires further investigation. This is an important distinction as the additional demands of future controlled movements can alter the required stiffness to successfully perform the task.

As ankle taping reduces available dorsiflexion range of motion (ROM) during landing among other kinematic and kinetic properties of the lower body, it may be expected to alter coordinative relationships with other joints of the same and other leg while dissipating the forces of landing because of constrained movement at the ankle from taping. Bilateral landings require coordination between the joints of the leg as well as across limbs to distribute the forces of landing equitably to avoid overloading a limb or joint. With deviations to these coordinative patterns, there is a potential increased risk for injury as joints in the leg may have differing relationships while performing landings (Doherty, Bleakley, Hertel, Caulfield, Ryan, Sweeney, et al., 2016; Hughes, 2020). External ankle support acts as a constraint on the available degrees of freedom which potentially limits the variability of the system to adapt to perturbations (Herb, Chinn, & Hertel, 2016; Jagodinsky, 2016; Jagodinsky, Wilburn, Moore, Fox, & Weimar, 2020). As such, the coordinative relationships of the taped ankle to the other joints in the lower limb necessitate further investigation. Continuous relative phase (CRP) has been used as a measure to reveal coordination patterns and variability between joints or segments throughout a motion, which can provide more insight than discrete relative phase (DRP) which focuses on particular time points. CRP traditionally utilizes kinematic data of position and velocity to quantify coordination in joint motion; however, with a high impact action like landing, coordination of the joint moments might be a better representation of the control of the movement. Therefore, a novel technique involving assessing CRP with kinetic joint moment data combined with kinematic positional data will be used here. As previously noted, the body of unilateral external ankle support is limited, and so is knowledge regarding the effect of ankle support on landing coordination. Taken together, there is a need for further investigation of unilateral external ankle support on bilateral coordination of inter and intralimb kinetic and kinematic dynamics.

These series of investigations provide a comprehensive kinetic and kinematic analysis of the effects of unilateral ankle taping on performance of a bilateral drop landing, which has been lacking in the literature. The primary concepts to be explored are mechanics, stiffness, and kinematic and kinetic joint coordination. The specific aims of these studies seek to explore how ankle tape influences the ankle, knee, and hip joints in each limb independently as well as their coordination together. Further, what these studies aim to achieve is providing a novel insight into bilateral landing assessment to encourage considerations of the complex interactions in the kinetic chain which occur when using prophylactic ankle tape during physical activity.

1.2: Research Aims

The overall aim of these investigations is to better understand the effects of taping a single ankle while performing a bilateral landing task. While ankle taping is common in exercise and athletic participation, little is known about the biomechanical changes that occur to both limbs when only one limb is taped. Therefore, this study seeks to provide a holistic view of the biomechanics of bilateral landing through traditional kinetic and kinematic measures, joint and lower-body stiffness, and interlimb and intralimb coordination while only one limb receives an external mechanical restraint in the form of prophylactic ankle taping.

1.3: Aim One

The first aim of this dissertation is to understand if taping one ankle while performing a bilateral landing increases biomechanical risk factors for injury, and whether any of those effects are influenced from a bout of exercise. To determine the sources of these effects, peak values of ground reaction forces, 3D joint kinematics, and 3D joint kinetics will be assessed. More specifically, by performing a bilateral drop landing using motion capture and two force plates, where each foot contacts each respective force plate, the limbs can be assessed independently to determine kinetic and kinematic changes stemming from the application of ankle tape. Additionally, the effects of exercise will be a focus to understand the influence on biomechanics independently, as well as its influence on the properties of ankle tape. The aim of this study will further aid in understanding the kinetic and kinematic changes in the joints of the lower limbs while one ankle is taped. The data analysis presented in Study One will focus on the peak values of multiple kinetic and kinematic variables of the ankle, knee, and hip in three dimensions when landing from one height before and after an exercise protocol with and without the application of unilateral ankle tape to the dominant limb.

1.4: Aim Two

Applying ankle tape is expected to increase joint stiffness with the intent to mitigate injury risk to a healthy ankle, as well as to one which has previously been sprained. However, stiffness of the joints, as well as lower-body are adapted to internal and external factors including task demands, surface conditions, height, among others (R. J. Butler et al., 2003; Rapoport et al., 2003). Therefore, the aim of the second study is to determine how taping one ankle alters joint and lower-body stiffness when performing a bilateral landing task from differing heights. The specific focus of this investigation is to gain an understanding of ankle, knee, and hip joint, leg, and lower-body stiffnesses in the sagittal plane during the landing phase calculated as the difference between initial ground contact until the center of mass's (COM) vertical velocity becomes zero. Three landing platform heights serve to assess changes in stiffness with increasing loads both with and without a unilateral ankle tape application. The data analysis presented in Study Two will utilize the total ROM of the COM, and joints, from initial contact to maximum displacement, and the peak GRF and joint moments.

1.5: Aim Three

Bilateral landings require coordination between the joints of both limbs to ensure an equitable distribution of forces between the joints of the two legs to prevent any joint from being overloaded and at an increased risk for injury. Aim three seeks to determine whether interlimb or intralimb kinematic or kinetic coordination during a bilateral drop landing task is altered with the unilateral application of ankle tape to the dominant ankle. Alongside coordination, the variability of that coordination will also be assessed to provide information about how consistently landings are performed. Rather than utilizing solely the peak values of landing, the coordination assessment will consider the entire time series of the landing process utilizing CRP. To determine the role of a unilateral ankle taping on lower body coordination dynamics, intralimb comparisons will consist of ankle-knee, ankle-hip, and knee-hip, while interlimb comparisons will consist of the ankle-ankle, knee-knee, and hip-hip joints. These analyses will provide detailed information about the coordination and the variability of that coordination of the leg joint pairs, both in the same limb, as well as across limbs. All coordination pairs presented in Study Three will utilize both traditional means of calculating CRP, consisting of joint position and velocity, in addition to a novel methodology utilizing joint position and moment to calculate kinetic CRP.

CHAPTER 2
LITERATURE REVIEW

2.1: Landing

High impact landings are ever present in sport and exercise participation and place the body under great stress. Landing can be described as an impact of the feet with the ground and can be as minimal as the forces seen during the heel contact during walking, to the high forces sustained while landing from dramatic heights while performing complex athletic maneuvers such as gymnastic skills. The forces exerted upon the body while landing have a wide range depending on the type of activity. Ranging from just over one times the weight of their body during typical walking; while running, on average approximately 2.5 body weights (BW) on the limb making ground contact; up to landing from a jump can place loads in excess of five to ten BW, and even reportedly up to 14.4 times body weight in single-leg landing in gymnastics skills (Firminger et al., 2018; Keller et al., 1996; P. J. McNair & Prapavessis, 1999; McNitt-Gray, 1991; Panzer, Wood, Bates, & Mason, 1988). Not only are landings discrete and short term actions, but in sports where jumping is a key facet of participation, such as basketball and volleyball, participants can perform approximately 76 jumps in volleyball and approximately 60 jumps in a single basketball game (Detanico, Piucco, Reis, Mello, & Santos, 2009; R. Lima, Palao, & Clemente, 2019). Furthermore, the additional volume of practices, exercise, as well as number of games, compounded over a lifetime leads to a substantially large volume of landings performed. This effect of repeated loading stands to place the body at a higher risk for both acute as well as chronic lower body injury.

To minimize both the potential risk for acute and chronic musculoskeletal damage, especially to the ankle and knee, requires the utilization of proper landing technique (Decker, Torry, Wyland, Sterett, & Richard Steadman, 2003; Murphy, Connolly, & Beynnon, 2003). All joints in the lower body work together in a coordinated movement pattern to mitigate the high loads (Yeow, Lee, & Goh, 2011a). While landing, the impact absorption travels upward in a distal to proximal force transmission and using both legs in a bilateral fashion allows the forces to be spread between the two limbs, thus reducing the overall impact load on each limb (Lees, 1981). While some asymmetries in landing techniques can be observed with the naked eye, three-dimensional motion capture analysis and time synchronized force platforms allow for a full analysis of landing mechanics, providing detailed insight into the underlying kinetics and kinematics which facilitates our understanding of the complex physics of landing.

Landing safely requires coordinated muscular control which prepares the joints in the leg to absorb the instantaneous ground contact (Santello & McDonagh, 1998). Though the degree of preparatory activity may vary, as dependent upon factors such as height, visual input, and, environmental factors, pre-activated musculature remains a key component of landing preparation to facilitate a safe initial contact phase. Interestingly, previous modeling showed that muscular activation of ankle, knee, and hip musculature was unnecessary until the performer was landing from a height of 1.05 m or higher (Yeadon, King, Forrester, Caldwell, & Pain, 2010). The greater COM displacement compared to experimental data from a 1.8 m drop landing representing a failed landing, suggested that below the height of 1.05 m, even without muscular pre-activation, a person could land without losing balance and falling over (Yeadon et al., 2010). However, despite modeling suggesting that landings below 1.05 m can be performed without muscular pre-activation, muscular activation is important in preparing the body for landing, especially when important visual information has been removed (Santello, McDonagh, & Challis, 2001). Although it is possible to land without falling or sustaining serious injury without either vision or pre-activation, clearly those physiological mechanisms provide benefits to the performer and particularly pre-activation is an inherent trait when performing a landing. Even prior to an unexpected landing, preparatory muscular activation is noted after the fall is initiated, suggesting a necessary co-activation prior to impact (Melvill Jones & Watt, 1971).

As a component of safe landing, the body COM needs to be effectively slowed to zero in a slow and controlled fashion, effectively spreading the impact over a longer timeframe with the objective of reducing the peak impact force and loading rate, or how quickly that peak force is reached (P. J. McNair, Prapavessis, & Callender, 2000). Proper landing technique consists of making initial ground contact with the toe while resisting the downward force as the heel travels downward to make ground contact (P. J. McNair et al., 2000; McNitt-Gray, 1991). With a traditional toe-heel landing, a ground reaction force (GRF) profile with two peaks will occur representing the force peaks at both toe (F1) and heel contact (F2) (McNitt-Gray, 1991;

Riemann, Schmitz, Gale, & McCaw, 2002). Furthermore, the initial contact led by the toe allows for the intrinsic musculature in the foot, as well as the plantarflexor muscles to begin to resist the initial strain of landing prior to the heel contacting the ground, protecting the bony structure of the rearfoot. Importantly, the two GRF peaks allow for a richer analysis whereby time to those peaks provides insight to the rate at which the musculoskeletal system is loaded, including changes due to a previous injury. In a meta-analysis, it was shown that individuals who have sustained a stress fracture exhibit higher loading rates when compared to those who have not sustained a previous stress fracture, highlighting the importance of a more slow and controlled landing (Zadpoor & Nikooyan, 2011).

While performers might exhibit a preferred method of landing that models a consistent pattern unique to them, there have been multiple reports that instruction and feedback can be effectively provided which facilitate the reduction of peak ground reaction forces, as well as improving knee kinematics during landing (Khuu, Musalem, & Beach, 2015; Laughlin et al., 2011; P. J. McNair et al., 2000; Milner, Fairbrother, Srivatsan, & Zhang, 2012; Oñate et al., 2005; Onate, Guskiewicz, & Sullivan, 2001). What is important about these findings is that mechanics of landing can be altered relatively quickly, and poor mechanics can be improved with input from sports science professionals. Even more specifically, it can be highlighted that while certain intrinsic risk factors for common injuries, such as anterior cruciate ligament (ACL) injuries cannot be altered, informative instruction can mitigate other risk factors for ACL injury to promote a beneficial landing pattern. When instructed to land with a more flexible leg during single-leg landing, participants were observed to land with greater knee flexion, and achieved more functional ROM during the landing, resulting in an 11% decrease in peak ACL force loading (Laughlin et al., 2011). As such, the instruction of proper landing technique to exercisers and athletes can potentially reduce the amount of lower body injuries sustained due to the high forces incurred while landing.

Landing is a ubiquitous task in many forms of exercise and sport performance and is a common cause of non-contact injury to the lower limbs. Landing places high stress on the lower body to dissipate the forces of landing, and proper landing technique can act to reduce the experienced GRF and loading rate. As exercisers and athletes train, performance gains would be expected which can increase jump height, and accordingly place higher loads on the body when landing from these increased heights, which must be appropriately managed to prevent injury. Due to the high volume of landings a person will typically incur throughout a lifetime of sport and exercise activities, effective landing technique is critical to reduce the risk for acute and chronic injury. With appropriate education and practice, landing technique becomes more efficient, increasing positive adaptations of landing mechanics.

2.1.1: Landing Height

Landing places the body under high force loads, and the level of the experienced loads is dependent on the person's weight and the height of the landing due to the acceleration of gravity, and with increases in height, higher ground reaction forces will be incurred (McNitt-Gray, 1991; Yeow, Lee, & Goh, 2009a, 2009b, 2010; S.-N. Zhang, Bates, & Dufek, 2000). That increase in landing force creates a higher demand in the activation of musculature to control the joints during landing and mitigate the increased forces. While training, exercisers and athletes seek to increase functional strength in the lower body which aims to provide activity specific performance gains through increased power output in the legs. These training gains often facilitate an increased peak jump height, which is beneficial in sports where jump height is a key facet of performance, such as the high jump in track, basketball, and volleyball, among others.

With increased jumping potential, an increased demand on the lower body musculature will be incurred upon landing. The increased force demands upon landing from increased height present a number of challenges to the musculoskeletal system to overcome, one of which is a reduction in movement variability, as a reduction in variability can increase the risk for overuse injury (Howe, North, Waldron, & Bampouras, 2018; Nordin & Dufek, 2017, 2019). Furthermore, a decrease in variability of time to peak GRF, and loading rate with increasing drop heights has been noted (Howe et al., 2018). Critically, with an increase in peak height, the need for increased joint flexion is enhanced to facilitate a decreased COM rate of descent to zero velocity, thus spreading out the contact force over a longer period. This strategy has been reported in healthy females when performing drop jumps from increasing platform heights of 30, 40, and 50 cm, where increases in maximum knee flexion and hip flexion, as well as an increase in increased sagittal plane moments was observed for both joints (Dickin, Johann, Wang, & Popp, 2015).

Larger muscles in the leg have the ability to dissipate higher force loads, which has been noted in the frontal plane, the hip was the driving component in dissipating the forces of landing when increasing drop height from 30 to 60 cm in males (Yeow et al., 2009a). Thus, it would be expected that these muscles would play a larger role in dissipating larger forces associated with landing from higher heights. Through regression relationships of landing height with ground reaction forces, and knee kinetics and kinematics, the rate of increasing GRF from increased heights outpaced the increase in peak knee flexion potentially increasing injury risks when landing from larger heights (Yeow et al., 2009b). This is of critical importance as the role of force attenuation must not rely solely on a knee focused landing strategy. Additionally, in males, the knee has been shown to effectively mitigate the increasing landing heights when performing a bilateral landing through increased functional ROM, including increased flexion at contact and peak GRF (Yeow et al., 2010). Whereas when landing from moderate heights, the effects stemming from reduced dorsiflexion are less pronounced compared to landing from higher heights, potentially suggesting that the initial contact with the ground as controlled by the plantarflexors play an important role in primary force dissipation (Howe, Bampouras, North, & Waldron, 2019). As the landing height increases, an increased reliance on larger musculature of the knee and hip utilization has been noted as an increase in work performed (S.-N. Zhang et al., 2000).

As individuals improve their strength and jumping performance, the height which is attained by jumping is increased, and so too are the forces incurred while landing. Therefore, the importance of proper landing form is highlighted so that the risk for lower body injury is reduced. As height increases, force attenuation from the larger muscles in the leg are important, and the relationship between quadricep and hamstring co-activation is important to reduce risk for ACL injury, as it provides appropriate joint stiffness to attenuate the forces of landing (Ford, Myer, Schmitt, Uhl, & Hewett, 2011; Granata, Padua, & Wilson, 2002; Granata, Wilson, & Padua, 2002). The knee plays a critical role in aiding to dissipate the high forces of landing due to the increased strength of the musculature that crosses the joint, and its large

ROM in the sagittal plane, where a substantial amount of force dissipation occurs. With these factors, it is important to consider which landing protocol methodology will allow safe performance of the task while contributing relevant data.

2.1.2: Bilateral Drop Landings

Traditionally, a bilateral landing is the preferred method of landing compared to a unilateral landing, particularly from higher heights, as it provides a larger base of support, as well as sharing the load of the impact across the two limbs, reducing the effect of the peak force on each limb. While participating in sport and exercise, both single and bilateral landings are likely to be performed, thus the choice of landing protocol should be task dependent. There are a handful of methods to assess bilateral landings including step-off style drop landings, hang-drop landings, as well as participant-initiated jump landings (Afifi & Hinrichs, 2012; R. J. Butler, Willson, Fowler, & Queen, 2013; Harry, Freedman Silvernail, Mercer, & Dufek, 2018; McNitt-Gray, 1991). There are benefits and drawbacks to each methodology, and the choice of which method to utilize can depend on the research question which will ultimately elicit the best representative data set. While step-off style landing paradigms can elicit more rapid time to peak, and higher magnitude initial contact forces and increased risk for overuse injuries when compared to a jump style landing, the use of step-off style landings can highlight potentially detrimental effects which might not be observed during jump style landings (Afifi & Hinrichs, 2012; Harry et al., 2018). Assessing landing performance in a laboratory environment is often accomplished through the utilization of drop landings from a platform to standardize landing conditions, ease of setup and utilization, and to facilitate repeatability across participants.

With the large number of degrees of freedom in the musculoskeletal system, there exists an inherent potential for asymmetry when performing any physical task, which can appear in both in kinetics and kinematics. By utilizing a bilateral drop landing task, that symmetry can be observed, as well as offering standardized height at which the COM travels from the platform to initial contact, as opposed to potential height asymmetries when performing a landing following a jump. In this sense, the standardization of drop landing height could reduce the natural feeling of a landing across participants, especially when the height of the platform exceeds maximal jump height. Because drop landings require stepping off a platform, there will be some time needed for the trailing leg to catch up to the leading leg to facilitate a symmetrical landing; however, even during hang drop landings where both limbs are suspended directly above the force platforms, asymmetries have been observed (Collings, Gorman, Stuelcken, Mellifont, & Sayers, 2019; Schot, Bates, & Dufek, 1994). This would suggest that some degree of inherent asymmetry is naturally present, and not necessarily a task constraint, which is not surprising considering factors such as limb dominance, previous history of injury, or strength differences, among others. Ultimately, the purpose of utilizing drop landings is to provide robust, repeatable, and valuable information about the adaptability and performance of the musculoskeletal system as they mitigate higher stress demands while performing the landing.

Researchers have utilized both single-leg as well as bilateral landings to gain insight into lower limb kinetics and kinematics. While results from both types of landings provide important information into landing mechanics, it is important to note that there are inherent differences which impact how joints in the lower limb respond to landing, particularly regarding the ability to share the landing between limbs symmetrically (Heebner et al., 2017; McPherson, Dowling, Tubbs, & Paci, 2016; Yeow, Lee, & Goh, 2011b). Also, the consideration of sex-based differences in landing is important, as landing in a single-leg stance can exacerbate mechanics which predispose individuals to additional risk for injury, especially in females (Schmitz, Kulas, Perrin, Riemann, & Shultz, 2007). During bilateral landing, the dominant limb has been shown to have higher peak GRF and a faster time to peak when compared to the non-dominant limb highlighting a potential reliance on the dominant limb to take on a more involved role in the landing process (Britto, Franco, Pappas, & Carpes, 2015). This effect potentially stems from improved strength and coordination in the dominant limb with a preference to rely upon that limb when compared to the non-dominant limb (Edwards, Steele, Cook, Purdam, & McGhee, 2012; Ford, Myer, & Hewett, 2003).

Despite a potential effect of increased reliance on the dominant limb while performing a bilateral landing, the non-dominant limb still plays an important role in reducing the load that the dominant limb would otherwise experience in a unilateral landing. Beyond solely the additional load experienced by a limb during a unilateral landing, there are some mechanical changes which potentially increase the risk for injury. Since absorbing the impact of landing takes place predominantly in the sagittal plane, reductions in sagittal plane ROM can increase peak forces as well as rate of force development which both place additional demand on the joints in the leg, particularly in the knee (Decker et al., 2003; Yeow et al., 2010). While performing a drop landing utilizing both unilateral and bilateral landings, participants utilized less sagittal plane knee flexion and increased knee valgus while performing a single-leg landing, highlighting the beneficial influence of bilateral landings (Pappas, Hagins, Sheikhzadeh, Nordin, & Rose, 2007). Findings of increased joint ROM during a bilateral landing compared to a unilateral landing have been consistently reported (McPherson et al., 2016; Weinhandl, Joshi, & O'Connor, 2010). Further, single-leg landings exacerbate the deleterious effect of additional frontal plane motion at the knee, especially in females (Russell, Palmieri, Zinder, & Ingersoll, 2006).

Due to differences between unilateral and bilateral landing protocols, it is critical to evaluate the methodology used when drawing conclusions regarding changes in biomechanics. When performing both unilateral and bilateral drop jumps utilizing male athletes, sagittal plane mechanics between limbs when landing unilaterally were not different (McPherson et al., 2016). Similar results were seen in ACL loading of females between the limbs, suggesting a similar risk of injury across limbs (Mokhtarzadeh et al., 2017). During bilateral landings small asymmetries were noted in the knee and hip at impact, in addition to larger knee and hip excursion during the landing, although based on the small differences, their clinical significance is uncertain (McPherson et al., 2016). Similar asymmetries were noted in a drop jump task where females were noted to be more asymmetrical at landing as well as at peak force coinciding with a higher difference in peak force symmetry, with the dominant limb being greater (Gu et al., 2021). Although drop jump studies are frequently utilized, differences have been observed compared to drop landings force and leg displacement, which could be important factors when looking into coordination (Hackney, Clay, & James, 2016). Similarly, while performing a bilateral drop landing, the rate of force development between limbs has been shown to be symmetrical, with small differences in the dominant leg between multiple landings highlighting that largely symmetry is maintained though with subtle differences, likely attributable to internal characteristics such as limb dominance (Bates, Ford, Myer, & Hewett, 2013).

There are differences in landing mechanics between landing bilaterally and landing unilaterally, as well as the type of landing paradigm utilized. Step off drop landings allow ease of repeatability and more efficient data collection which makes them a beneficial task when acquiring numerous trials. A bilateral landing allows the forces of landing to be shared by both limbs; however, the role of the joints in the leg appear to play a different role in dissipating the force between unilateral and bilateral landings (Weinhandl et al., 2010). While landing from elevated heights, the knee is more suited to appropriately dissipate the high forces in a bilateral landing compared to a single-leg landing (Yeow et al., 2010). The utilization of both limbs during landing facilitates a lesser reliance on detrimental frontal plane motion, which is a key factor of ACL injury, and is already at a higher rate in females, highlighting the importance of appropriate landing technique (Kernozek, Torry, Van Hoof, Cowley, & Tanner, 2005; Russell et al., 2006). The distinction between the landing methodology utilized is important to make appropriate comparisons with previous findings; although findings may be similar, they are not necessarily equitable between differing methodologies, especially when considering the differences in risk factors for injuries in a female population. Ultimately, the use of a bilateral landing allows for kinetic and kinematic analyses of each limb independently, as well as how they function together during landing. This information is critically important when investigating differences between limbs, especially when limb asymmetries are present stemming from either intrinsic or extrinsic factors.

2.1.3: Gender Differences in Landing Mechanics

Although males and females perform largely the same gross motor skills while participating in exercise and sport, the kinetics and kinematics of these complex movements often differ and those differences can result in an increase for potential injury, notably with damage to the ACL, as well as ankle sprain. One of the most reported differences lies with the mechanics of the knee upon landing, where females tend to land with more knee valgus, or where the knee is closer to the midline than the point of contact of the foot with the ground, which places the ACL at a higher risk for damage.

Many of the differences in landing mechanics between males and females occur in the frontal plane, which is often a primary source of injury, particularly to the knee (Kernozek et al., 2005; Pappas & Carpes, 2012). Further, females have been shown to have an increased reliance on more frontal plane knee motion while landing (Carson & Ford, 2011; Cronstrom, Creaby, Nae, & Ageberg, 2016a, 2016b; Gehring, Melnyk, & Gollhofer, 2009; Jacobs, Uhl, Mattacola, Shapiro, & Rayens, 2007; Jenkins, Williams, Williams, Hefner, & Welch, 2017; Kernozek et al., 2005; S. G. McLean et al., 2007; Pappas & Carpes, 2012; Pappas, Hagins, et al., 2007; Pappas, Sheikhzadeh, Hagins, & Nordin, 2007; Russell et al., 2006). Increased knee valgus while landing is one of the primary causes of ACL damage, as the orientation of the knee increases strain on the ACL (Hewett et al., 2005). While individual improper movements can contribute to an increased risk for damage to the ACL; with multiple detrimental movements across the joints and in multiple planes, there is a further increased risk for injury (Jacobs et al., 2007).

Outside of varied mechanics at the knee, when females land from an elevated platform, they tend to land in a more erect posture, and utilize a force attenuation strategy involving a larger ROM at the ankle and hip with more force dissipation from the ankle and knee (Decker et al., 2003; Kernozek et al., 2005). Throughout the ankle ROM, the angular velocity through dorsiflexion was also higher in females, which could indicate less active control and potentially increase the risk for injury (Decker et al., 2003). Similarly, valgus movement velocity has been noted to be higher in females when performing a single-leg landing (Jenkins et al., 2017). The strategy of utilizing larger ankle ROM to attenuate a landing impact is suggested to attempt to reduce further loading up the kinetic chain, especially at the knee.

Regarding kinetics, females have been noted to land with higher peak GRF when compared to males (Pappas, Hagins, et al., 2007; Pappas, Sheikhzadeh, et al., 2007). Despite landing from the same height, a higher peak GRF upon landing places a greater load on the lower body which requires additional attenuation to prevent injury. Further, increased vertical GRF, has the potential to increase joint moments, which necessitates additional eccentric muscular activation to offset the additional joint torque. As such, females have been shown to have higher internal rotation and abduction moments at the knee while performing drop landings (S. G. McLean et al., 2007). While peak GRF was not reported in the study, with increased joint moments, there is potential for peak GRF values to also be increased.

While some gender-based differences in mechanics are benign, differences such as additional frontal plane knee motion, landing more upright, and with higher peak GRF are detrimental and can lead to an increased injury risk, particularly to the ACL. For this reason, it is important to not only understand the underlying sex-based differences while performing a landing task, but to also understand their role in changing the risk for injury. Furthermore, it is critical to understand the differences between landing mechanics which are intrinsic and non-modifiable, or if there are modifiable factors which can be altered with appropriate training to facilitate positive landing mechanics to mitigate injury risk.

2.2: Lower Extremity Injury

Injury is an unfortunate, yet common occurrence in sport and exercise, and of those, injuries to the lower body are the most common, especially considering that many forms of exercise and sport participation rely heavily on the lower body for successful task performance, including higher intensity activities like running and jumping (Hootman et al., 2007; Murphy et al., 2003). As such, lower body injuries present an issue with performance, and successful rehabilitation is critical to regain losses due to a sustained injury. Within the scope of the investigations conducted in this dissertation, two injuries are of interest, those being ankle sprains and damage to the ACL. The rationale for focusing on these injuries lies with the fact that ankle sprains are the most sustained injury to exercisers, while damage to the ACL is arguably the most debilitating musculoskeletal injury sustained during exercise and sport performance (D. T.-P. Fong, Chan, et al., 2009; Hewett, Ford, Hoogenboom, & Myer, 2010; Hootman et al., 2007).

Although there exist a number of factors which can influence the occurrence of ankle sprains, because they are the most common injury sustained while performing exercise and sport activities, it is important to understand the overarching effects of an ankle sprain (D. T.-P. Fong, Hong, Chan, Yung, & Chan, 2007; Hootman et al., 2007; Waterman, Owens, Davey, Zacchilli, & Belmont, 2010). Damage to the ACL is not as prevalent as damage to the ankle;

however, the short- and long-term effects of sustaining an injury to the ACL are highly debilitating to the individual (Kiapour & Murray, 2014). Furthermore, injury to the ACL is much more common in females than their male counterparts, and there is also some evidence to suggest that females also sustain ankle sprains at a higher rate than males as well (Arendt & Dick, 1995; Doherty et al., 2014; LaBella, Hennrikus, & Hewett, 2014; Montalvo et al., 2019). Therefore, these differences in injury occurrence and potential sources of these disparities will be explored.

2.2.1: Sex Disparities in Injury Rates

When participating in sport and exercise activities, there will always exist some degree of risk for sustaining musculoskeletal injuries, and the degree of potential injury risk depends on many factors. Particularly, as sex is a non-modifiable factor, its influence on injury risk is of importance especially when seeking to understand when females are noted to be at a higher risk of certain injuries when compared to their male counterparts. Additionally, the degree of injury dictates the amount of time removed from sport, and potentially daily life activities as well.

While the disparity in ACL damage between males and females is quite clear, the differences in ankle sprain have shown mixed reporting. A potential cause of the mixed reporting is that ankle sprains are more common across all ages and activity level, while most all ACL damage occurs while performing high intensity activity, and is typically incurred in those under 30 years of age, therefore study population is important to address (Sanders et al., 2016; Serpell, Scarvell, Ball, & Smith, 2012; Waterman et al., 2010). Females have been noted to have nearly twofold incidence of ankle sprain when compared to their male counterparts when examining epidemiological research studies (Doherty et al., 2014). While conversely, no sex-based difference in incidence of ankle sprain from reported visits to emergency departments in the United States has also been reported (Waterman et al., 2010). It is possible that the mixed findings can potentially be attributed to differences in search criteria, type of study design, study population, or underreporting; particularly if those who sustain an ankle sprain do not seek medical care. However, with the potential for higher incidence of ankle sprains in a female population, combined with the well-cited increased risk of ACL damage, it is important to understand not only the influence ankle sprains themselves have on other risks for injury, but also how the treatment and rehabilitation methods, particularly taping and bracing influence those potential risks for injury.

Females have a clear increased risk for damage to the ACL when compared to males, although the exact degree of that increased risk is dependent on factors such as age, type of sport or exercise, level of participation, among others (Arendt & Dick, 1995; LaBella et al., 2014; Montalvo et al., 2019). The disparity in occurrence of ACL damage appears to be around puberty when females outpace the rate of ACL damage seen in males (LaBella et al., 2014). Damage to the ACL is tremendously debilitating to those who sustain this injury and has both immediate, as well as long lasting health as well as emotional and financial implications (Hewett et al., 2010). ACL damage does not heal well, and even with surgical intervention, the risk for reinjury is up to fifteen times greater (Paterno, Rauh, Schmitt, Ford, & Hewett, 2012). Furthermore, even after surgical repair of the ACL, less than two thirds of athletes will return to their previous level of performance (LaBella et al., 2014). Additionally, females are less likely than males to return to sport after ACL injury, even if they received an ACL reconstruction (LaBella et al., 2014).

Overall, there is evidence to support that females sustain ankle sprains at similar or higher rates when compared to males, and damage to the ACL at a higher rate than their male counterparts, with ACL injury being a substantially higher disparity. Regarding the ACL, the difference in sex-based injury rates appears to shift around puberty, when females overtake males. While ankle sprains are common across all ages and activities, ACL injury is typically only incurred during higher intensity activities, making it very relevant to exercisers and athletes to understand methods of prevention, and appropriate treatment to return to activity safely. While ankle sprains often heal well and normal sport performance and exercise can be resumed with proper treatment, ACL tears do not see the same degree of positive outcome. As such, it is important to understand sex-based differences to reduce the risk of injury, especially when some of these differences lead to a higher risk for injury in females.

2.2.2: Sex Specific Anatomical Risk Factors

Identification of differences in injury occurrence between sexes is an important first step towards understanding the root causes of those identified disparities. While external factors are potential contributors to differential injury risk between sexes, including sport or activity choice, contact sports, etc., the exact nature of those factors can be difficult to appropriately categorize and stratify across groups. Therefore, a focus on intrinsic factors, chiefly anatomical differences between sexes can provide a more succinct explanation of injury risk. Although some intrinsic factors such as strength, lean mass, fat mass, and flexibility among others, there are anatomical differences between males and females which are non-modifiable, and some of those differences predispose females to a higher risk for injury, more specifically to ACL damage (Collings et al., 2021).

Evidence for anatomical differences which pre-dispose females to more knee injuries is substantial; however, the volume of evidence for anatomical differences at the ankle is not as extensive. Females with larger calcaneal ROM and increased tibial varum were at a higher risk for ankle injury, which differed for males where increased talar tilt was a higher risk for ankle sprain (Beynnon, Renstrom, Alosa, Baumhauer, & Vacek, 2001). Not only does the ROM and bone orientation play a role, but the laxity within the joint appears to be a confounding issue, whereby there has been evidence suggesting that it is a risk factor for ankle sprain, and others suggesting it is of no influence (Beynnon et al., 2001; Hubbard & Hicks-Little, 2008; Murphy et al., 2003; Trevino & Lee, 2018). Passive resistance of the joint is achieved through the ligaments, and when musculature is either inactive, or sub-active, the strain placed on the ligament will be higher, as the musculature is not dampening the load. Though increased joint laxity is not strictly related to females, there is evidence to support that females have more ankle joint laxity when compared to males which shows a decrease in joint stiffness (Trevino & Lee, 2018).

At the knee, there are a handful of anatomic features which have been identified as sources of potential higher risk for sustaining an injury to the ACL. The anatomic structures of the distal femur, proximal tibia, and knee joint space; including tibial plateau slope (Cheung, Boguszewski, Joshi, Wang, & McAllister, 2015; McLean, Lucey, Rohrer, & Brandon, 2010), femoral notch width (Cheung et al., 2015; Ireland, 1999; Simon, Everhart, Nagaraja, & Chaudhari, 2010; Whitney et al., 2014) and ACL length, cross-sectional area, and volume (Chandrashekar, Slauterbeck, & Hashemi, 2005; Whitney et al., 2014) have been identified as potential sources of increased risk of ACL injury. Female ACLs have also been shown to fail at lower force thresholds when compared to their male counterparts during cadaveric testing (Chandrashekar, Mansouri, Slauterbeck, & Hashemi, 2006). The anatomy of the femoral notch located at the distal end of the femur also plays a unique role in how the ACL as well as the other cruciate ligaments are attached within the intracapsular space at the knee (Cheung et al., 2015). A narrower femoral notch has the potential to impinge the ACL as the knee goes through loading during flexion (Cheung et al., 2015; Ireland, Ballantyne, Little, & McClay, 2001; Simon et al., 2010; Whitney et al., 2014). Though data is inconclusive on whether females have comparatively smaller femoral notches when height matched to males, a smaller femoral notch has been suggested as a possible independent risk for ACL injury (Ireland, 1999; Whitney et al., 2014). Furthermore, the composition of the ACL as evidenced by MRI scans of the knee showed that decreased ACL volume and decreased femoral notch width were also independent predictors of ACL injury (Whitney et al., 2014). Females with a smaller ACL volume were at a higher risk for ACL injury when compared to females with a larger ACL volume, suggesting that a smaller ACL plays a significant role in injury risk (Whitney et al., 2014).

In addition to the ACL, the menisci provide the knee with stability, and their orientation along with the slope of the tibial plateau can be a risk factor for ACL injury (Cheung et al., 2015). Additionally, prior injury or degeneration of the menisci can contribute to a steeper posterior slope resulting in increased loading at the knee (Cheung et al., 2015). Patients who underwent a meniscectomy showed an increase in anterior and posterior tibial translation which are risk factors for ACL damage (Levy, Torzilli, & Warren, 1982). Further, the importance of the medial meniscus was shown to have a degree of anterior displacement restriction when the ACL was not functioning (Levy et al., 1982). These results further indicate the role the menisci play in restraining anterior tibial translation and sharing force loading at the knee.

Females have been observed to have an increased tibial posterior slope which is defined as the angle of the tibial plateau from the anterior to the posterior when viewed in the sagittal plane (Cheung et al., 2015). In conjunction with this, if the lateral slope of the tibia is greater than the medial slope it forces the tibia into an internal rotation during weight bearing putting additional strain on the ACL (Simon et al., 2010). Increased lateral posterior slope causes an anterior tibial translation, knee valgus, and internal tibial rotation, all of which put an additional strain on the ACL (Cheung et al., 2015).

Regarding the joints in the leg, there are sex-based differences in anatomical structure which can place females at a higher risk for injury when compared to males. There are a handful of sex-based anatomical differences at the proximal tibia, distal femur, and the joint space in between at the knee which place females at a higher risk for damage to the ACL when compared to males. There is less evidence for sex-based anatomical differences at the ankle which could increase for risk of ankle sprain; however, ankle sprains are still common injuries when participating in exercise and sport and are important to consider when investigating landing mechanics. Although there are unmodifiable anatomical characteristics which place females at a higher risk for lower extremity injuries, an understanding of these factors can facilitate training programs to promote movement strategies with the aim to mitigate these intrinsic risk factors.

2.2.3: Ankle Sprain

Ankle sprains are the most sustained musculoskeletal injury in exercise and sport participation (D. T.-P. Fong, Chan, et al., 2009; Hootman et al., 2007). Lateral ankle sprains are the most common type of sprain sustained, with the anterior talofibular ligament (ATFL) being the most often injured ligament in the lateral ankle complex followed by the calcaneofibular ligament, and are often the result of some combination of excessive inversion and plantarflexion (Doherty et al., 2014; D. T.-P. Fong et al., 2007; Kobayashi & Gamada, 2014; Safran, Benedetti, Bartolozzi, & Mandelbaum, 1999; Wright, Neptune, van den Bogert, & Nigg, 2000). The ATFL is highly important in providing support to the ankle complex as it acts to resist inversion while the foot is in a plantarflexed position (Bahr, Pena, Shine, Lew, & Engebretsen, 1998). Incidence of ankle sprain can vary based upon a number of factors including data reporting, anatomy, gender, age, sport or exercise participation among others (Beynnon, Murphy, & Alosa, 2002; Beynnon et al., 2001; Doherty et al., 2014; Leininger, Knox, &

Comstock, 2007; Swenson, Collins, Fields, & Comstock, 2013; Waterman et al., 2010). However, one of the most important determinants for incurring an ankle sprain is having sustained an ankle sprain previously (Beynnon, Vacek, Murphy, Alosa, & Paller, 2005; de Noronha, Franca, Haupenthal, & Nunes, 2013; Fousekis, Tsepis, & Vagenas, 2012; Raina & Nuhmani, 2014; Swenson et al., 2013; Swenson, Yard, Fields, & Dawn Comstock, 2009).

While many individuals will successfully heal after sustaining an ankle sprain and return to sport and exercise at or near their performance level pre-injury, there can still be lasting effects where residual laxity can be seen up to one year after the sprain (Hubbard & Hicks-Little, 2008). Interestingly, regarding laxity, the number of previous ankle sprains does not appear to have a relationship between lasting joint laxity six months after sustaining the most recent ankle sprain (Liu, Gustavsen, & Kaminski, 2013). While a single incident of sustaining a lateral ankle sprain can be treated and allow individuals to return to exercise or sport with minimal continual symptoms, subsequent ankle sprains can be problematic as the recurrent injury rate can be up to 74% (Kobayashi & Gamada, 2014). Ankles which have sustained a previous sprain have been shown to have more inversion rotation and lower anterior and inversion stiffness which are risk factors for sustaining an ankle sprain (Kovaleski et al., 2014). Further, previous ankle sprains have shown reduced variability in the knee and hip while performing single-leg jump landings potentially contributing to other lower body injury risk (Brown, Bowser, & Simpson, 2012). Individuals who have sustained multiple ankle sprains in the same ankle and develop chronic ankle instability (CAI) are often referred to as either 'non-copers' who identify one or more lasting deficits in their performance resulting from previous ankle sprains, or as 'copers' who do not identify many deficits with various aspects of performance (Doherty, Bleakley, Hertel, Caulfield, Ryan, & Delahunt, 2016). Individuals with CAI are an important group to address when post-ankle sprain investigations are concerned; however, when an investigation identifies CAI or other previous lower body injuries outside the scope of investigation, participation questionnaires which identify exclusionary criteria are important to control for potential confounding factors. The investigations performed as a part of this dissertation exclude history of lower body injuries which affect performance, and as such, CAI will not be discussed in depth.

Ankle sprains often occur in the early stages of weight acceptance, prior to any reactive activation of evertor muscles to resist the inversion load, placing a higher strain on the ligaments within the ankle (Konradsen et al., 1997). Support for this has been reported when a participant sustained an accidental inversion ankle sprain during a cutting maneuver as part of a research task, with a noted deviation in kinematics at 60 ms; and beginning at 110 ms the ankle showed a large increase in inversion and internal rotation angular velocities (D. T.-P. Fong, Hong, et al., 2009). Another study also recorded an accidental ankle sprain while a participant performed a cutting maneuver where the kinematics of the ankle were similar between the non-injury and injury trials until approximately 80 ms suggesting that shortly after contact is a critical determinant of whether an ankle sprain will occur (Kristianslund, Bahr, & Krosshaug, 2011). To actively compensate for the forces of landing, pre-contact muscle activation of complimentary musculature including the evertors, external rotators, and dorsiflexors must be able to supply a resistive torque in opposition of the external forces of the landing impact since stretch reflex mechanisms only somewhat contribute to post-landing muscular activation (Santello, 2005). Further, delayed reaction time has been observed in the peroneus muscle in those who have sustained a previous ankle sprain (Hoch & McKeon, 2014). While inversion is commonly noted in ankle sprains, internal rotation has also been reported as an injury mechanism, even when in dorsiflexion rather than the commonly reported plantarflexion, as internal rotation can also place a strain on the ATFL due to its orientation within the joint space (D. T.-P. Fong, Hong, et al., 2009; Golanó et al., 2010; Kobayashi & Gamada, 2014; Kristianslund et al., 2011).

During an investigation utilizing a stop-jump task involving stepping forward then performing a vertical jump to 50% maximal height, and another jump with maximal effort, an ankle sprain occurred in a male participant with CAI (Terada & Gribble, 2015). The participant completed two trials prior to incurring the sprain, which provided a unique insight into the mechanics of an ankle sprain and provided the ability to compare mechanics within the same participant. Prior to the injury, there was greater knee adduction and hip abduction, combined with less plantarflexion and knee flexion in the 200-millisecond period prior to the sprain (Terada & Gribble, 2015). Typical features of ankle sprain were exhibited including greater peak ankle inversion, knee adduction, and hip abduction combined with less knee flexion and hip flexion (Terada & Gribble, 2015). Compared to the successful trials, there was higher ankle energy dissipation in the sagittal plane, and less energy dissipation in the sagittal plane of the knee, as well as less ankle and knee energy dissipation in the frontal plane as well as the COM being higher and shifted to the non-injured side (Terada & Gribble, 2015).

During dynamic loading conditions, to prevent damage to ligaments, forces applied to them must remain below the failure point, through activation of the appropriate musculature. Passively, ligaments possess properties of a viscoelastic material, which gives them the ability to deform, and return to their original state provided they do not exceed approximately 8 – 10 percent of their original length where failure of the tissue will begin (Robi, Jakob, Matevz, & Matjaz, 2013; Woo et al., 2000). Through *in vitro* studies, researchers have been able to understand failure rates in tissue which can be applied to complex modeling techniques. A better understanding of the material properties and failure rate of ligaments facilitates how the musculature must function to offset the damaging external forces which can rupture ligaments. In a cadaveric study of eight ligaments in the ankle, it was noted the failure load of the anterior talofibular ligament was 297.1 N ± 80.3 (Funk, Hall, Crandall, & Pilkey, 2000). In a simulated ankle sprain addressing the individual contributions of eight ligaments in the ankle complex, the anterior talofibular ligament peaked at 20% strain while the ankle was inverted, plantar flexed, and internally rotated, highlighting the role of the foot position on the risk for damage to the ATFL (D. T.-P. Fong et al., 2011).

There is evidence for a potential elevated risk for ankle sprain in the dominant limb with one source coming from the additional load that athletes place on the dominant limb while performing sport specific tasks (Beynnon et al., 2002; Halabchi, Angoorani, Mirshahi, Shahi, & Mansournia, 2016). As noted previously, limb preference when performing these tasks can result in an increased utilization and reliance on that limb which can place that limb at an increased risk for injury (Edwards et al., 2012; Ford et al., 2003). In drop landings, electromyography (EMG) data revealed the non-dominant tibialis anterior muscle was more active than the dominant limb which provides additional stiffness at the ankle to resist the loading (Niu, Wang, He, Fan, & Zhao, 2011). The increased muscular activity in the non-dominant limb could highlight the role the non-dominant limb provides in supporting the performance of more sport-specific tasks, such as kicking a ball which is more likely to occur with the dominant leg, and this disparity in activity in the dominant limb could potentially lead to a reduced ability to protect the ankle during landing.

Ankle sprains remain as the most sustained injury in exercise and sport performance, and stem from several intrinsic and extrinsic factors. Though recovery and rehabilitation from sustaining an ankle sprain is often successful, one of the primary factors for sustaining an ankle sprain is a history of previous sprain(s). There is some evidence that the dominant limb is more likely to be sprained, and females have also been reported to incur higher rates of ankle sprain compared to males, which combined, highlight the importance of investigating movements which can result in both injuries. As ankle sprains are so common in sport and exercise performance, understanding and utilization of appropriate prevention and rehabilitation techniques are important in mitigating the negative short- and long-term physiological effects of ankle sprains.

2.2.4: Anterior Cruciate Ligament Injury

Though the incidence of ACL tears is lower than that of ankle sprains, when an ACL tear occurs, it is incredibly detrimental to the individual, and females are at a much higher risk for ACL damage when compared to their male counterparts. While there are many factors which can influence the incidence of injury, approximately 1 in 30 female athletes will sustain an ACL tear (Montalvo et al., 2019). Within the knee, the ACL serves an important role in rotational stabilization, and preventing anterior translation of the tibia relative to the femur. Damage to the ACL is a very serious injury which often results in reduction in functional movement deficits, lost time in sport and exercise performance, financial, and emotional strain, among other negative effects. It is important to understand common causes of ACL injury to provide education and training protocols which can assist in reducing the risk of sustaining an ACL injury.

The ACL is capable of sustaining high force loads in the stabilization of the knee, and the orientation of the applied force as well as the position of the knee joint contribute to the loading the ligament can withstand before sustaining damage (Woo, Hollis, Adams, Lyon, &

Takai, 1991). The strain threshold in the ACL has been reported to be around 15 – 19 % stretch beyond resting length before rupture noted in *in vitro* ligaments (D. L. Butler, Kay, & Stouffer, 1986; Taylor et al., 2011). Although the ACL can resist high strain loads, effective muscular control to moderate laxity is critical to provide support and minimize the force experienced by the ACL. Laxity within the knee has been shown to place the ACL under increased strain and at greater risk for damage during landing (Kiapour et al., 2014). This muscular control also provides dynamic stability to the knee during complex movements acting to protect the less compliant ligaments in the knee.

Knee stability during dynamic loading is maintained both passively by ligaments, as well as actively by musculature. The ACL is a significant supportive ligament in the knee, which is responsible for restraining nearly all anterior translation of the tibia relative to the femur and provides stability in the frontal and transverse planes (D. L. Butler, Noyes, & Grood, 1980; Levine et al., 2013; McLean, Mallett, & Arruda, 2015). A substantial portion of damage to the ACL occurs during non-contact situations which is an important consideration in the literature as contact-based injuries can be more difficult to quantify the specific nature of the injury (Boden, Dean, Feagin, & Garrett, 2000; Kiapour & Murray, 2014). Damage to the ACL is incredibly debilitating, and should athletes desire to return to sport, surgical intervention is almost a necessity, and even so, functionality at the same level as pre-injury is unlikely (McLean et al., 2015; Streich, Zimmermann, Bode, & Schmitt, 2011). Furthermore, adding to the overall detrimental effects of an ACL injury, females are less likely to return to sport participation when compared to males (LaBella et al., 2014).

Dynamic knee valgus is noted as one of four traditional features in ACL injury that have been observed in video evidence, with the others being a relatively straight contact leg, a substantial portion of weight on the affected leg, and a lateral trunk tilt (Hewett et al., 2010). Dynamic knee valgus is noted as a collapse of the knee, with the “distal femur toward and distal tibia away from the midline of the body” making it so that the knee is more towards the midline, relative to the point of contact of the foot (Hewett et al., 2005). Higher valgus motion at initial contact and during loading, along with higher GRF were noted in athletes who went on to sustain an ACL injury in the future (Hewett et al., 2005). Multiple factors have been identified in ACL injury risk along with knee valgus, including internal rotation of the thigh and knee with evidence that suggests knee valgus rotation is a primary factor in ACL strain (Fox, Bonacci, McLean, Spittle, & Saunders, 2014; Kiapour et al., 2015; Levine et al., 2013). Knee valgus is one of the most common factors and has been directly implicated in numerous mechanisms and patterns of injury to the ACL, which makes it a key mechanical component to address with proper training (Hashemi et al., 2011; Hewett et al., 2005; Ireland, 1999; Kanamori et al., 2002; Levine et al., 2013; Markolf et al., 1995; Olsen, Myklebust, Engebretsen, & Bahr, 2004; Quatman & Hewett, 2009; Shimokochi & Shultz, 2008).

Females sustain damage to the ACL at a rate of at least 1.7 to 6.2 times that of their male counterparts (Arendt & Dick, 1995; LaBella et al., 2014; Montalvo et al., 2019). The increased risk for ACL damage to females has been proposed to stem from several factors encompassing various anatomical differences as well as muscular control and landing strategies among others (Hewett et al., 2010; Huston, Greenfield, & Wojtys, 2000; Whitney et al., 2014). Until around puberty, the rate of ACL injury is similar between males and females, but around and following puberty, ACL injuries in females are higher in adolescence and adulthood; however, levels off at the professional level (Ford, Shapiro, Myer, Van Den Bogert, & Hewett, 2010; Holden, Boreham, & Delahunt, 2016; LaBella et al., 2014). A review of insurance claims over a five-year period of youth soccer players showed no higher incidence in number of ACL injuries between males and females suggesting that the onset of puberty is an important time point where sex-based differences in injury occurs (Shea, Pfeiffer, Wang, Curtin, & Apel, 2004). In late puberty females exhibited more frontal plane motion at the knee, noting that knee valgus is one of the primary contributing factors to ACL damage (Hewett, Myer, & Ford, 2004).

The anatomy of the ACL itself potentially plays a role in the increased rate of injury in females, where the mass of the ACL was shown to not increase with respect to height as it did in males which would serve to offset additional weight from the added height (Chandrashekar et al., 2005). Though, due to the smaller sample size of the study, body size was not able to be assessed as a separate factor, and potentially plays a role in these factors. However, there were still notable differences between males and females, as the mass of the ACL was shown to increase with height only in males, and there was an increase in ACL size with an increased notch width only in males (Chandrashekar et al., 2005). The ACL provides almost all of the total restraining force in the anterior direction, and if injured, does not have a highly successful capacity to fully heal, reducing the overall restraining force against external loading after injury (D. L. Butler et al., 1980; Cheung et al., 2015). In testing healthy male and female controls, as well as those with ACL deficiency showed that healthy females had significantly more single-leg stability than healthy males; however, after sustaining an ACL injury, males were significantly more stable on both the involved and uninvolved limb (Hewett, Paterno, & Myer, 2002). Following ACL reconstruction, males continued to have better total stability than females, with differences persisting to 12 months post-surgery which highlights the lasting deficits of sustaining an ACL injury in females (Hewett et al., 2002).

One of the factors of an increased risk of ACL damage, particularly in females can be described broadly as an asymmetry or imbalance which can stem from various sources including previous injury, strength and flexibility imbalances, ligament dominance, quadriceps dominance, leg dominance, and trunk dominance (Ford et al., 2003; Hewett et al., 2010; Hewett, Lindenfeld, Riccobene, & Noyes, 1999; Hewett, Stroupe, Nance, & Noyes, 1996; Knapik, Bauman, Jones, Harris, & Vaughan, 1991; Kuszewski, Gnat, Szlachta, Kaczynska, & Knapik, 2019). Limb dominance can play a role in increased risk for ACL damage, where females have been identified to utilize increased knee valgus in the dominant limb while landing compared to the non-dominant limb (Ford et al., 2003). While the dominant limb might be a higher risk for ACL damage due to a preference to utilize that limb more for demanding tasks, the non-dominant limb can be at risk due to factors such as less strength and control, highlighting an aspect of task specificity in the overall risk (Ford et al., 2003). Ligament dominance refers to inappropriate stabilizing muscular control in a joint, placing an additional load on the passive structures within the joint, while quadriceps dominance relates to a tendency to stiffen the knee largely with the quadriceps without as much hamstring utilization (Hewett et al., 2010). Ligament dominance is often pronounced at the knee, especially during landing, where individuals with less appropriate muscular control will typically show an increased valgus displacement, further straining the ACL (Ford et al., 2003; Hewett et al., 2010). The tendency for females to land in a more upright posture can contribute to the imbalance in muscular activation which further compounds the increased strain in the ACL (Blackburn & Padua, 2008, 2009; Decker et al., 2003; Durall et al., 2011; Hewett et al., 2010; Huston, Vibert, Ashton-Miller, & Wojtys, 2001; Kernozek et al., 2005).

When landing from 60 cm, participants reached peak ACL loading within the first 60 milliseconds, which is too early in the landing process to respond to counter the loading, so appropriate muscular control prior to the applied force is important to protect the knee (Kernozek & Ragan, 2008). In particular, when the knee is dampening more of the energy earlier in the landing combined with less influence of the ankle and hip, the risk for ACL injury is higher (Norcross, Blackburn, Goerger, & Padua, 2010). Considering the point in the landing phase when joints reach peak moment or peak joint displacement, it was noted that the knee valgus did not occur in the initial phase of landing when ACL injury risk is elevated (Pappas, Hagins, Sheikhzadeh, Nordin, & Rose, 2009). In a computational model, when performing a bilateral drop landing, the ACL was loaded only in the first 25% of the landing phase (Pflum, Shelburne, Torry, Decker, & Pandy, 2004). These findings are important as they address the importance of evaluating the kinetics and kinematics of landing throughout the entire loading phase.

Activation in the quadriceps and hamstrings are important protective factors to combat loading forces on the knee ligaments. In a study utilizing male participants, it was demonstrated that forcing the knee into abduction and adduction, could be mitigated with controlled activation of the musculature that crosses the knee (L.-Q. Zhang & Wang, 2001). This finding demonstrates the potential for reduction in joint laxity and knee abduction and adduction moments with appropriate muscular activation. Further, less effective muscular control is a predictor for both primary injury risk, as well as future reinjury risk (Levine et al., 2013). Improper co-activation of the hamstrings and quadriceps can potentially be a risk factor in ACL injury. In a study of elite handball athletes, athletes who showed reduced semitendinosus activity, and increased vastus lateralis activity during a cutting maneuver were at a higher risk for ACL injury (Zebis, Andersen, Bencke, Kjær, & Aagaard, 2009). Similarly, an unbalanced muscular activation between the medial hamstrings and quadriceps was shown to predict a higher knee abduction moment (Palmieri-Smith, McLean, Ashton-Miller, & Wojtys, 2009). To compensate for knee laxity, trained collegiate female athletes had a higher level of lateral hamstring activation which is a potential neuromuscular adaptation to protect the knee during landing (Rozzi, Lephart, Gear, & Fu, 1999). Appropriate neuromuscular control of the lower limbs is critical in mitigating injuries, and implementation of neuromuscular training has shown benefit improvement in mechanics to reduce improper movements that contribute to a higher ACL injury risk (Hewett et al., 1999; Hewett et al., 2005; Hewett et al., 1996; Noyes, Barber-Westin, Fleckenstein, Walsh, & West, 2005). After a six-week training program, adolescent female athletes were able to reduce peak landing forces and decreased both adduction and abduction knee moments (Hewett et al., 1996).

Damage to the ACL is tremendously debilitating and requires surgical intervention and physical therapy to return to a high level of physical activity and performance. Females have consistently been reported to sustain ACL damage at a higher rate than their male counterparts, and there are various anatomic and physiologic differences which contribute to this disparity. Appropriate neuromuscular control to regulate knee stiffness when landing is noted to be one of the most important mitigating factors to reduce the risk of ACL damage. Because appropriate neuromuscular control is a key factor in reducing the risk of damaging the ACL, training programs that focus on proper mechanics have been shown to reduce some of the key risk factors in ACL injury. As differences in ACL injury rates between males and females emerge around puberty, education and intervention of proper training could benefit female athletes to combat the primary risk factors for ACL damage.

2.3: Influence of Dorsiflexion Range of Motion on Landing

During landing, movement in the sagittal plane is the primary method for force absorption. As such, the ankle plays a critical role in sagittal plane mobility since the ankle is the first joint to begin force dissipation as the ground reaction force travels proximally up the kinetic chain. When assessing the motion of the ankle, it is important to address common measurement techniques. An arthrometer assesses joint laxity while the ankle is held in place and a force is applied to assess alterations in position, small electronic devices such as an inclinometer or an accelerometer can compare the position relative to gravity, or a goniometer which assesses the ROM of the ankle through a simple graduated protractor-like device to measure the angle of the shank relative to the foot where the ankle serves as the central axis (Bennell et al., 1998; Konor, Morton, Eckerson, & Grindstaff, 2012; Kovaleski, Hollis, Heitman, Gurchiek, & Pearsall, 2002). These forms of measurement provide important information about the motion at the ankle; however, it is important to understand the difference in what is measured with these techniques when observing the ankle joint. Arthrometry provides meaningful data about the ligamentous restraint properties, while goniometry provides information about ROM, typically highlighting musculotendon flexibility as well as the movement within the joint space. Additionally, the method of assessment can differ between non-weightbearing and weightbearing ROM assessment. During weightbearing ROM assessment, the participant is able to utilize additional motion with the talus which can provide a more realistic total functional ROM at the ankle (Dill, Begalle, Frank, Zinder, & Padua, 2014).

Across healthy participants, up to approximately 5-6 degrees difference in dorsiflexion ROM between limbs is noted as healthy, and weightbearing dorsiflexion ROM has been shown to be largely similar across limbs, suggesting no influence of limb dominance (Cosby & Hertel, 2011; Dowling, McPherson, & Paci, 2018; Hoch & McKeon, 2011; Howe, Bampouras, North, & Waldron, 2020; Konor et al., 2012; Rabin, Kozol, Spitzer, & Finestone, 2015). In one study which contrasts similar weightbearing dorsiflexion ROM, the non-dominant ankle had a larger ROM; however, it should also be noted that the mean difference was less than approximately seven degrees at which point a clinically relevant impairment would be noted (Hoch & McKeon, 2011; Rabin et al., 2015; Reid, Birmingham, & Alcock, 2007; Vicenzino, Branjerdporn, Teys, & Jordan, 2006). Ankle plantarflexion and dorsiflexion ranges of motion are influenced by several factors which could include joint laxity, muscular flexibility, previous injury, or external supportive devices. Because the triceps surae is a two joint muscle group, with control of plantarflexion and knee flexion, the flexibility of this muscle group is important for not only dorsiflexion ROM, but how that influences the behavior of the knee. Reduced dorsiflexion ROM has been shown to coincide with less knee flexion at contact and lower peak knee flexion (Dowling et al., 2018; C.-M. Fong, Blackburn, Norcross, McGrath, & Padua, 2011; Hoch, Farwell, Gaven, & Weinhandl, 2015; Howe et al., 2019; Malloy, Morgan, Meinerz, Geiser, & Kipp, 2015). Similarly, reduced knee flexion and increased frontal plane knee motion have also been presented when performing a squat, suggesting that these responses are common adaptations when dorsiflexion ROM is limited (Macrum, Bell, Boling, Lewek, & Padua, 2012; Mauntel et al., 2013).

The ankle plays a significant role in absorbing forces during landing (Devita & Skelly, 1992). Ankle sagittal plane ROM plays an important role in landing since most of the force is dissipated through movement in the sagittal plane. Within the joints of the legs, larger functional ROM adds mobility to utilize the sagittal plane motion to slow down the body COM over a larger ROM and over a longer period to reduce not only the peak force sustained, but the rate of force development. Furthermore, movements in the lower limb have complex interrelated relationships, and changes in one joint's mobility can require compensatory behavior at other joints. Therefore, it is important to understand the effects of reduced dorsiflexion flexibility on the kinetics and kinematics of other joints during dynamic loading conditions.

With decreased sagittal plane ankle ROM, an increase in motion in the frontal and transverse planes has been noted as a potential compensation which can increase risk for injury in the ankle and knee (Bell-Jenje et al., 2016; Dill et al., 2014; Howe et al., 2019; Y. L. Lima et al., 2018; Malloy et al., 2015). Reduced ankle dorsiflexion has been shown to increase knee valgus, potentially as a compensatory mechanism to offset the loss of ROM, which highlights that some compensatory movements can increase the risk for damage to the ACL, and improvements to dorsiflexion flexibility should be included in exercise programs (Cronstrom et al., 2016b; C.-M. Fong et al., 2011; Hagins, Pappas, Kremenic, Orishimo, & Rundle, 2007; Howe et al., 2019; Y. L. Lima et al., 2018; Macrum et al., 2012; Sigward, Ota, & Powers, 2008). More specifically, females with less ankle dorsiflexion flexibility exhibited a greater knee abduction moment and peak knee abduction angle, as well as decreased knee flexion, which are increased risk factors for sustaining an ACL injury (Malloy et al., 2015).

Reduced ankle dorsiflexion has been shown to have mixed results on certain kinetic and kinematic variables (Mason-Mackay, Whatman, & Reid, 2017). It is unclear if a limited dorsiflexion flexibility influences peak ground reaction forces, with findings both rejecting and supporting this effect (C.-M. Fong et al., 2011; Howe et al., 2019; Malloy et al., 2015; Whitting, Steele, McGhee, & Munro, 2011). In a bilateral drop landing task, reduced dorsiflexion flexibility in males and females did not alter vertical ground reaction force, or loading rate (Howe et al., 2019). Despite reduced ankle dorsiflexion flexibility while assessed statically, it is unclear if that carries over into dynamic loading. Multiple studies have reported that dorsiflexion ROM did not affect peak dorsiflexion angles while landing, which could be a factor of the high loads placed on the joint while landing from height, or even that the full use of ROM is not needed to dissipate the force of landing (Dill et al., 2014; Dowling et al., 2018; C.-M. Fong et al., 2011; Malloy et al., 2015; Mason-Mackay et al., 2017; Whitting et al., 2011).

The ankle plays a critical role in beginning the force dissipation from landing where larger functional ROM in the sagittal plane assists to slow the descent of the COM. Dorsiflexion flexibility is influenced by flexibility of the triceps surae musculature which can be modified with stretching and training programs, as well as the passive restraint provided by ligaments in the ankle complex. With less ankle sagittal plane flexibility, less utilization of knee flexion often coincides, in addition to frontal plane motion at the knee, which is a primary risk factor for damage to the ACL. Whether reduced ROM is due to intrinsic factors like flexibility or injury, or extrinsic ones like external ankle support, it is important to understand potential changes in the kinetics and kinematics at the knee and hip during landing.

2.4: Ankle Tape

A common method of treatment and prevention of ankle sprain is the utilization of prophylactic ankle supports, which includes both tape and braces (Dizon & Reyes, 2010; Kaminski et al., 2013). The choice between utilizing taping or bracing can depend on many factors including personal preference, availability of a sports medicine professional, injury history, among others. While ankle braces are easily applied by the performer, ankle taping requires proper technique from a trained person, often an athletic trainer which is not always available for exercisers and athletes. When utilizing tape to prevent lateral ankle sprains, the tape is applied in a way to provide support against inversion, where increases in resistance to an inversion moment show the successful application of the ankle tape (Manfroy, Ashton-Miller, & Wojtys, 1997). A common method of applying ankle tape is incorporating the Gibney technique which utilizes a procedure of taping patterns starting at the medial aspect of the foot and leg and pulled to the lateral aspect to provide support which aids in inversion resistance (Callaghan, 1997; Perrin & McLeod, 2018).

Ankle tape is utilized to constrain movement primarily in the frontal plane (inversion/eversion) with the aim to provide mechanical support to reduce the risk of ankle sprain, and there are also restrictions on movement in the sagittal plane (dorsiflexion/plantarflexion) (Cordova, Ingersoll, & LeBlanc, 2000; Cordova, Ingersoll, & Palmieri, 2002; Kaminski et al., 2013; Miller, Needle, Swanik, Gustavsen, & Kaminski, 2012; Quackenbush, Barker, Stone Fury, & Behm, 2008; Romero-Morales et al., 2020). Ankle tape serves as a preventative measure to reduce the risk for ankle sprains as well as a treatment method while returning to exercise and sport performance after sustaining an ankle sprain. As there is a great deal of overlap between the utilization of ankle tape and braces within the literature, both will be discussed, though ankle tape remains the primary focus as it was the methodology chosen for this dissertation.

2.4.1: Effects of External Ankle Support on Landing Mechanics

Two of the most used forms of external ankle support are tape or lace-up style semi-structured braces. Many studies have investigated the differences between those types of external ankle supports, and there are benefits to each method, with evidence to suggest that both methods provides sufficient support to the ankle, especially in those who have previously sustained an ankle sprain (Dizon & Reyes, 2010). Utilizing external ankle support aims to reduce the risk of sustaining damage to the ankle, most commonly a lateral ankle sprain. Although the precise degree of mechanical changes can differ between ankle taping and bracing, a key component of both types of ankle support are to provide inversion restriction (Hall, Simon, & Docherty, 2016). While the restrictive effects of ankle tape on motion at the ankle during static testing procedures are present, those restrictive capabilities of ankle tape need to carry over into dynamic loading situations to provide protection against ankle sprains during performance. Studies have largely shown that static measures of reduced dorsiflexion ROM from the ankle support do also carry over to restricted dorsiflexion ROM in dynamic movement during landing after the application of ankle support (Cordova, Takahashi, Kress, Brucker, & Finch, 2010; Kuni, Mussler, Kalkum, Schmitt, & Wolf, 2016; McCaw & Cerullo, 1999; Simpson et al., 2013; S. A. Williams, Ng, Stephens, Klem, & Wild, 2018).

The effect of ankle support on peak ground reaction force (GRF) has shown mixed evidence including decreases, increases, or no change in the peak GRF while external ankle support was applied (Abián-Vicén et al., 2008; Cordova et al., 2010; DiStefano, Padua, Brown, & Guskiewicz, 2008; Hodgson, Tis, Cobb, & Higbie, 2005; Mason-Mackay, Whatman, Reid, et al., 2016; Megalaa, Hiller, Ferreira, Beckenkamp, & Pappas, 2022; Okamatsu, 2014; Riemann et al., 2002; Stoffel et al., 2010; West, Ng, & Campbell, 2014; Yi, Brunt, Kim, & Fiolkowski, 2003). Differences in these findings can potentially be explained through various methodologies, as confounding variables can play a role in influencing outcomes which can include varied landing heights, participant instruction, or participant experience, among others. Despite the disparity in previously presented findings, a general trend in the two force components of GRF after ankle support was noted in a meta-analysis, regarding the force at initial contact of the toe, and the second contact with the heel (Niu et al., 2016). This meta-analysis analyzed 15 studies which revealed overall during toe to heel landings, while the first impact peak (F1) was not different between ankle support and without, the second impact peak (F2) was higher in the ankle support condition, and additionally, the time from contact to F1 and contact to F2 were both reduced (Niu et al., 2016). However, more specifically in a systematic review of only ankle braces, five investigations which were shared with the studies presented by Niu and colleagues, noted no change in peak GRF, which included both F1 and F2 values, adding to the differences between methods of ankle support (Mason-Mackay, Whatman, & Reid, 2016; Niu et al., 2016). As such, it appears that changes to peak GRF after the addition of external ankle support are not only task-dependent, but the specific type of ankle support utilized can play a role in whether GRF is influenced by that ankle support.

Evidence for a reduction in time to peak appears to be slightly clearer, with a number of studies reporting that after external ankle support, the time to F1, defined as T1 and the time to F2, defined as T2 are reduced (Cordova et al., 2010; Niu et al., 2016; Riemann et al., 2002; Simpson et al., 2013; Yi et al., 2003). However, it is important to consider the relationship between time to peak and the peak force itself. The rate at which force is placed on the body plays an important role in the deceleration of the body and the risk of injury (Riemann et al., 2002). Loading rate is described as the ratio of peak force to the time to reach that force. Riemann and colleagues noted that during a bilateral landing while wearing either ankle tape or a brace, neither F1 or F2 were greater than the control condition; however, both ankle support conditions showed a faster time to both T1 and T2 (Riemann et al., 2002). While not directly assessed in the investigation, the decrease in time to peak GRF has the potential to significantly increase the loading rate, despite no change to the peak GRF.

As a primary role of ankle tape is to reduce the chance of incurring an ankle sprain, which largely necessitates resistance against motion in the frontal plane, it stands to reason that the motion of the foot itself would be influenced. Due to the anatomy of the foot, the complex structure of the bones, ligaments, and intrinsic musculature allow the midfoot to move in all planes. As such, since ankle taping covers a significant portion of the foot, it is likely that there are segmental changes at the level of the midfoot because of ankle taping. In participants with CAI, Kuni and colleagues discovered that ankle taping reduced frontal plane movement at the midfoot and rearfoot during single-leg drop landing (Kuni et al., 2016). Changes to the joint moment in the frontal plane have shown varied results after ankle support noting an increase in the inversion moment after ankle bracing, while other investigations noting no differences in inversion and dorsiflexion moments (Dewar et al., 2019; Maeda et al., 2019; Vanwanseele, Stuelcken, Greene, & Smith, 2014; S. A. Williams et al., 2018).

While ankle taping acts to provide mechanical support to the ankle itself, there is evidence of kinetic and kinematic changes at other joints in the kinetic chain which arise due to external ankle support (S. A. Williams et al., 2018). There have been mixed reports on how ankle support influences knee kinematics in the sagittal plane with support for no change, as well as reduced flexion while landing with ankle support (Agres, Chrysanthou, & Raffalt, 2019; Cordova et al., 2010; West et al., 2014; S. A. Williams et al., 2018). Increased knee flexion at contact, however, does appear to be more consistent across investigations (DiStefano et al., 2008; Simpson et al., 2013; Stoffel et al., 2010). Increased knee flexion at contact and decreased ROM without a change in peak flexion while braced were also reported during forward jump landing (DiStefano et al., 2008). Similarly, performing a bilateral landing with an ankle brace, showed the knee more flexed at contact, without a change to peak flexion moment (Simpson et al., 2013). Taping and bracing have also shown not to influence sagittal plane knee moments while landing and performing sport related rebounds and receiving a pass (Vanwanseele et al., 2014; S. A. Williams et al., 2018). In both taping and bracing, there does not appear to be an influence in the hip kinematics in the frontal and sagittal planes (Cordova et al., 2010; Mason-Mackay, Whatman, & Reid, 2016; Simpson et al., 2013).

Ultimately, ankle taping does indeed alter lower limb mechanics, although the precise extent that external ankle support affects the kinetics and kinematics of the lower limb is inconclusive, and is dependent on factors such as age, gender, type of activity, among others. Furthermore, the effect of a unilateral application of external ankle support on bilateral kinetics and kinematics has only recently been investigated, and only frontal plane variables were reported, thus this unique area requires further investigation. As more research seeks to uncover the effects of unilateral ankle taping and bracing on bilateral tasks, it is critical to remain aware of potential changes, both beneficial and detrimental to the legs independently. However, there are a handful of kinetic and kinematic measures which appear to be consistent across the literature. Pertaining specifically to drop landings, the time to peak force is decreased, typically represented as the time to the second of two peaks of ground contact when the heel makes contact, the externally supported ankle appears to approach the ground with less plantarflexion, and to offset the altered sagittal plane ankle motion, the knee acts to contact the ground in a more flexed position. With the changes in kinetics and kinematics that arise from taping or bracing an ankle, it is important to weigh the benefits to the potential drawbacks to determine the best course of action to benefit the musculoskeletal health of the performer.

2.4.2: Effect of Exercise on Ankle Tape

A multitude of investigations have sought to understand the effects of exercise on the changes in the restrictive properties of ankle tape. This is an important consideration, as losses in the beneficial restriction of a taped ankle can necessitate multiple applications throughout a training session or competitive event, or perhaps influence a performer's decision whether to use ankle tape. Mechanical restriction of ankle tape is one of the key benefits of taping the ankle, and that restriction has been shown to reduce after exercise as measured by losses in plantarflexion, dorsiflexion, inversion, and eversion restriction (Best et al., 2014; Cordova et al., 2000; Fumich, Ellison, Guerin, & Grace, 1981; Meana, Alegre, Elvira, & Aguado, 2008; Miller et al., 2012; Paris, Kokkaliaris, & Vardaxis, 1995; Purcell, Schuckman, Docherty, Schrader, & Poppy, 2009). After an acute bout of exercise, there are a handful of factors that have been noted to contribute to the loosening of ankle tape including numerous stretching movements, increased body heat, increased temperature in the tape itself, and perspiration contribute to the reduction in restrictive properties of ankle taping (Alt, Lohrer, & Gollhofer, 1999). However, depending on the duration, type of exercise, and the plane of motion assessed, the amount of restriction lost from the ankle taping varies, and can still provide beneficial support.

Losses from the initial restriction of ankle tape have been seen after as little as 10 minutes of exercise; however, this does not necessarily indicate that the mechanical support of the tape fails to provide protection from ankle sprains, or that the proprioceptive effects of taping are diminished, which is also a key feature of using ankle supports (Dizon & Reyes, 2010; Quackenbush et al., 2008). Following a 10-minute run, as well as additional exercises, ankle tape was still effective in resisting inversion during an unexpected inversion simulation (Ricard et al., 2000). After 20 minutes of an assorted exercise program, while there was an increase in ankle motion, there was still more restriction in the anterior displacement, and inversion/eversion rotation when compared to baseline measures as measured by an ankle arthrometer (Miller et al., 2012). During a 30-minute treadmill run assessed at multiple time points throughout the session, ankle tape was still effective in active inversion/eversion and plantarflexion restriction with relatively consistent active inversion/eversion ROM and maximum plantarflexion (Tamura et al., 2017). Plantarflexion ROM has been shown to reduce after the application of ankle tape, and exercise returned a larger ROM to both plantarflexion and dorsiflexion (Quackenbush et al., 2008). Regarding extended exercise including a variety of dynamic movements, after a 45-minute simulated soccer activity, as much as a 90% loss in the restrictive properties of ankle tape has been reported (Best et al., 2014).

Outside of the restrictive effects of tape, it is important to consider other areas of performance that might be influenced by exercise while the ankle is taped. After completing a

20-minute treadmill jog while wearing ankle tape or ankle braces, no differences in either peak forces F1 or F2 or the time to those forces were observed suggesting that the effects of external ankle support were not influenced by the exercise bout (Riemann et al., 2002). Although dorsiflexion and plantarflexion ROM increased after a ten minute treadmill run, there was no change in vertical jumping performance suggesting that taping does not reduce the capacity to achieve maximal vertical jumps (Quackenbush et al., 2008).

Despite the ROM restriction provided by ankle taping, repeated stretching, heat, and perspiration contribute to a reduction in mechanical restriction after exercise in as little as 10 minutes. One of the more important details regards the type of exercise performed, where running alone does not appear to influence the restrictive components of ankle tape, while more dynamic movements including lateral and direction changing maneuvers place more demands on the restrictive ability of the tape thus loosening the tape to a higher degree. While some investigations report that a significant ROM restriction remains after shorter bouts of exercise, with extended duration of exercise, restrictive properties of ankle tape have been shown to all but disappear. This is an important consideration when utilizing ankle tape in extended exercise and athletic performance, where re-application of the tape may be needed to provide the same level of support seen immediately after taping.

2.5: Stiffness

2.5.1: Definition and Calculation of Stiffness

Landing compresses the musculoskeletal system rapidly, creating high force loads at the joints which must be mediated to prevent injury. The human body has often been modeled as a linear mass-spring, highlighting the ability to compress and resist an external load (R. J. Butler et al., 2003; Serpell, Ball, et al., 2012). Compressive stiffness derives from Hooke's law, which states that the force (F) required to compress an object is related to its spring constant (k) and its resultant deformation (x), representing vertical stiffness (Equation 1) (R. J. Butler et al., 2003). Solving for the stiffness value rearranges the equation recommended for vertical (y) stiffness calculations by Serpell and colleagues (Equation 2) (Serpell, Ball, et al., 2012).

$$F = kx \tag{1}$$

$$K_{vert} = \frac{F_{\max}}{\Delta y} \tag{2}$$

Stiffness is maintained by a combination of anatomic features, including musculature, ligaments, cartilage, tendons, and bone (R. J. Butler et al., 2003; Latash & Zatsiorsky, 1993; Rapoport et al., 2003). However, the human body is not a perfect spring, where the deformation or displacement is not equivalent to the force applied as seen in Equation 1. Equation 2 is therefore a best approximation of the vertical quasi-stiffness in the body, considering the various elastic and non-elastic components within the leg. While the distinction in nomenclature is important to address, the information gained from assessing stiffness still provides meaningful information about how the body is resisting external forces when impacting the ground. The terms “vertical stiffness” and “leg stiffness” will be utilized throughout, with the understanding that they are not representative of a perfect spring, and are a measure of quasi-stiffness. Lower-body stiffness is representative of the combined effect of ankle, knee, and hip joint stiffnesses in both limbs combined. Lower-body stiffness can be determined from the vertical ground reaction force and the vertical displacement of the COM (Blickhan, 1989). Additionally, during movements which primarily take place in the vertical plane, leg stiffness is determined by the ground reaction force of that limb and the vertical compression of that leg (Serpell, Ball, et al., 2012). This particular methodology assumes an appropriate definition of the three-dimensional location of the femoral head as determined by three-dimensional motion capture (Serpell, Ball, et al., 2012).

In the literature, there is some discrepancy in reporting leg and vertical (lower-body) stiffness. While the two are similar, especially in a vertical task such as hopping or landing, lower-body stiffness is represented as the COM displacement, while leg stiffness is represented as the change in leg length, as measured by the vertical displacement of the femoral head (Serpell, Ball, et al., 2012). Often, studies state leg stiffness, when the calculation for vertical stiffness is utilized, thus it is important to understand the method used when comparing results across investigations (Serpell, Ball, et al., 2012). Because landing takes place primarily in the vertical direction, the vertical travel of the femoral head is an appropriate calculation with no need to assess the angle at which the leg contacts the ground as would be seen in gait. While supine, the body COM in females was located at approximately 56% of height, which would be near the pelvis (Virmavirta & Isolehto, 2014). During bilateral landing when using a lower body marker model and assigning the COM as the center of the pelvis, the total vertical travel of the femoral head and COM are likely to be similar. As joint stiffness utilizes the moment of the joint, it is dependent on the external forces applied, which makes it is important to normalize forces to body weight to ensure equitable comparisons (Farley, Glasheen, & McMahon, 1993).

While total vertical stiffness is represented as a linear mass-spring, joint stiffness is represented as a torsional spring whereby the force applied to the spring acts in a rotation rather than a linear plane (Farley & Morgenroth, 1999). Joint stiffness refers to the resistance to angular displacement and is calculated as the ratio peak joint moment (M_{max}) to the change in joint position ($\Delta\Theta$), maintained passively by ligaments, and actively in joint musculature (Equation 3) (Serpell, Ball, et al., 2012).

$$K_{Joint} = \frac{M_{\max}}{\Delta\Theta} \qquad 3$$

Vertical and joint stiffnesses are both measures of resistance to an external load, but they are calculated differently, where vertical stiffness is the ratio of the applied force over the change in COM; joint stiffness is a measure represented as the ratio of the joint moment to the change in joint angle. Joint stiffness has been noted to be a measure of quasi-stiffness, where there is not necessarily a direct relationship between a deformation and elastic energy store relative to the displacement of that joint (Latash & Zatsiorsky, 1993). While it is useful to assess the relationship between joint deformation and the joint moment, it is not a direct stiffness measure. However, the calculation of joint stiffness as noted in Equation 3 still provides meaningful information about the joint's resistance to change. Although additional components such as elastic and non-elastic forces can contribute to joint stiffness, moving forward, the term "joint stiffness" will be used, with the understanding that non-measured elastic and non-elastic components also contribute to joint stiffness (Latash & Zatsiorsky, 1993). Stiffness is a necessary quality to protect the body from damage due to impacts; however, the level of stiffness required can vary depending on the particular demands of a task (R. J. Butler et al., 2003). Too little stiffness can place additional stress on the ligaments, as the musculature is not appropriately regulating joint positional control to resist the load, whereas too much stiffness increases risk of bony injuries (R. J. Butler et al., 2003; Granata, Padua, et al., 2002; Williams III, Davis, Scholz, Hamill, & Buchanan, 2004).

2.5.2: Task Dependent Stiffness Modulation

Research in stiffness often utilizes tasks incorporating both the loading and unloading phases of impacts, such as hopping and running. Under these tasks, the legs undergo compression in the first half of ground contact, and subsequently unload during the second half. With activities like hopping, jumping, or running, there is a secondary task to perform the next motion rather than strictly the discrete absorptive compression phase. Furthermore, a critical point of consideration is the directionality of the action performed, where hopping and jumping are primarily single axis movements in the vertical direction, whereas running, while predominantly in the forward direction, has a vertical component of action as well. Due to this fact, when calculating stiffness during activities such as walking and running, a more complex model is needed whereby the angle of the leg as it contacts the ground can be accounted for to better understand the ground reaction force (Hobara et al., 2013; Lorimer, Keogh, & Hume, 2018).

Stiffness plays an important role in regulating task performance, and higher stiffness allows for rapid rebounding movements like hopping and running as the musculotendon unit is at a higher level of tension and ready to respond to activity demand, whereas lower stiffness can facilitate a secondary task with the aim of facilitating the stretch-shortening cycle, or the absorption of forces without a secondary task (R. J. Butler et al., 2003). Higher physical demands have been correlated with increasing stiffness across various tasks including hopping, running, and jumping (Brazier et al., 2014; Gunther & Blickhan, 2002; McMahon, Comfort, & Pearson, 2012). Furthermore, to perform a jump after landing requires the musculature to not only absorb the forces of landing, but to maintain a degree of activation to manage the transition of the stretch-shortening cycle from eccentric loading during force absorption to concentric contraction to propel the body upward. The effects of this additional task demand potentially reduces the ability to directly compare stiffness between drop landings and drop jumps.

During rhythmic and cyclic tasks, control of stiffness serves a dual purpose, where muscular control acts to both absorb the impact of landing, as well as to prepare for the following movement. During hopping tasks, leg stiffness has been shown to increase with increased frequency of hopping or height, highlighting the effect of a task demand on stiffness (Hobara et al., 2010; Hobara, Kanosue, & Suzuki, 2007; Hobara, Kobayashi, Yoshida, & Mochimaru, 2015; Padua, Arnold, Carcia, & Granata, 2005). Effectively, task demands of increased frequency or increased maximal height hopping necessitate a reduced time of ground contact, and decreased COM displacement, thus increasing leg stiffness. Even while performing the same task of a maximal countermovement jump, the addition of arm movement changes the stiffness values in the leg, highlighting the role that task demands play in modulating stiffness (Struzik, 2019). Interestingly, there does appear to be some similarity between tasks in terms of stiffness modulation, where knee stiffness while hopping at 1.5 Hz was similar to knee stiffness while performing a drop jump from 50 cm, which could potentially assist in more direct comparisons between stiffness metrics with varying tasks (Hobara et al., 2008).

However, while performing drop landings, the task prioritizes a slow and controlled descent of the body COM often through a larger ROM, whereby a more compliant spring mechanism facilitates the task parameters of absorbing the landing impact alone (Slater, Campbell, Smith, & Straker, 2015; I.-L. Wang, Wang, & Wang, 2015). Whereas when completing a drop jump, where the goal of the task is to reach a target height which is often maximal jump height, residual stiffness to control throughout the stretch-shorten cycle serves to prepare the legs to rebound would be preferred (S Kuitunen, Kyröläinen, Avela, & Komi, 2007). The demands of a landing task, therefore, demand a different approach to controlling lower body stiffness to absorb a drop landing without a secondary task of rebounding from the landing and are likely to be difficult to compare to those rebounding tasks more directly. Further, when considering differences in unilateral and bilateral landing mechanics, coinciding differences in stiffness modulation will also likely share similar differences to those in landing mechanics since stiffness is the combined measure of the forces exerted on the body and the coinciding position.

Stiffness is an important quality which both serves to aid performance as well as provide resistance to injury. While there is not a definite answer on what stiffness is ideal during a task, too much stiffness can increase risk to bony structures, and too little stiffness can increase risk to soft tissue. Vertical and joint stiffness have been shown to vary based on task demands, and as such, an ideal stiffness modulation is likely to meet the demands of the specific task. With the small body of literature related to solely the absorptive phase of landing, especially regarding landing from various heights, it is important to consider the differences in the task demands between a drop landing and a repetitive landing such as hopping or running that can influence stiffness in the leg.

2.5.3: Height of Landing and Stiffness

Increasing height places increased demand on the musculoskeletal system which must be attenuated appropriately to avoid injury. As such, these increased demands incur higher forces, higher loading rates, and alter joint kinematics, suggesting that alterations to lower body stiffness are bound to occur. Measures of stiffness are often investigated during repetitive loading and unloading conditions, such as hopping and running which consider the effects of stiffness as the leg spring compresses and rebounds to propel the COM in the intended direction of travel. In the case of running, an additional component involving the forward propulsive component along with a vertical one is important to consider.

Little evidence exists for measures of stiffness solely in the compressive stage of landing, such as performing a drop landing. In one of the only studies to utilize a bilateral drop landing task, leg stiffness was shown to decrease when landing from the higher heights of 60 and 80 cm when compared to 40 cm, but stiffness was consistent between the two higher heights of 60 and 80 cm (L.-I. Wang, 2009). Additionally, the knee showed an increase in stiffness from the 60 to the 80 cm platform highlighting the influence of height on leg and joint stiffnesses (L.-I. Wang, 2009). Another drop landing study found similar results where females performing a bilateral drop landing decreased leg stiffness when increasing landing height from 40 to 60 cm (I.-L. Wang et al., 2015). Interestingly, as males were also used in this study, leg stiffness was not significantly different between 40 and 60 cm, as reported in the previous investigation (I.-L. Wang et al., 2015; L.-I. Wang, 2009). It is worth noting that in the earlier study (L.-I. Wang, 2009) and the later study (I.-L. Wang et al., 2015) both methodologies utilized a bilateral landing on a solitary force plate which limits analysis to combined leg and joint stiffnesses. This distinction is especially relevant when the need to distinguish leg and joint stiffnesses independently is of paramount importance.

Little research has been completed assessing stiffness while performing a drop landing. Evidence supports that height appears to play a part in how stiffness is moderated in the legs to dampen the effect of the landing. With the available literature on landing stiffness, when landing from higher heights, a decrease in stiffness allows for the forces of the landing to be better dampened to reduce injury from the added height. Further research in this area is needed to better understand how height influences the absorptive component of landing, as well as how the joints in the leg are contributing to leg, and lower-body stiffness.

2.5.4: Individual Joint Contributions to Stiffness

When assessing leg stiffness and vertical stiffness, it is important to consider the contributions of the joints within the leg that make up the resistance to vertical displacement of the COM. Since vertical stiffness is a linear representation of the additive angular joint stiffnesses, it is important to understand how the joints of the leg contribute to vertical stiffness when performing various tasks (Lorimer et al., 2018). Often, stiffness is investigated utilizing repetitive tasks such as running or hopping, rather than discrete ones such as landing, making comparisons between results difficult. However, hopping can potentially provide the basis for the understanding of how stiffness in the leg would behave in drop landings since the plane of action is nearly entirely in the vertical direction for both tasks.

The degree to which each joint contributes to vertical stiffness appear to be task dependent, as reporting has suggested that each joint in the leg to be the primary contributor to stiffness regulation during hopping tasks, highlighting the task-dependency between single-leg or double-leg hopping, varied surface stiffnesses, hopping height, external perturbations, or even training (Arampatzis, Bruggemann, & Klapsing, 2001; Farley, Houdijk, Van Strien, & Louie, 1998; Farley & Morgenroth, 1999; Hobara et al., 2010; Hobara et al., 2008; Hobara et al., 2009; Lorimer et al., 2018; Maloney, Richards, Nixon, Harvey, & Fletcher, 2017). Task demands of hopping play an important role in the modulation of stiffness, as hopping at higher frequencies typically necessitates a decrease in vertical displacement of the COM, which results in a higher leg stiffness value when increasing frequency of hopping, though which joint contributes most to this increase is not consistently reported across investigations (Chang, Roiz, & Auyang, 2008; Hobara et al., 2017; Hobara et al., 2010; Padua et al., 2005). When comparing endurance trained athletes to power trained athletes while two-legged hopping at different frequencies, it was shown that power trained athletes had higher leg stiffness than endurance athletes at both frequencies (Hobara et al., 2008). Further, in the power trained group, ankle stiffness was higher than the endurance athletes in the faster condition, and knee stiffness was higher at the slower condition, which suggests that independent of the task, the type of training can modify joint preference in stiffness regulation.

Like leg and joint stiffness in hopping, while performing landings there is no clear evidence that a particular joint is the determining factor in stiffness modulation. During drop landing with a natural landing pattern, the primary goal is to land in a comfortable manner to reduce the feeling of the high impact forces from the elevated platform. Thus, an ideal preference for landing should center on extending the time of force generation and an increased ROM at the joints in the leg. Although landing with the aim of reducing the impact on the lower body is beneficial, there is no well-defined range for leg and joint stiffness which would indicate not too high of stiffness to increase risk of bony injuries, while not being too compliant to risk injury for soft tissue injury.

In a study comparing dancers to athletes, while performing a bilateral drop jump from 45 cm, dancers showed higher leg stiffness than athletes, yet both groups knee stiffness were not different (Ambegaonkar et al., 2011). This finding would suggest that the knee was not a primary factor in the higher leg stiffness, though it could not be determined which joint was a primary driver as ankle and hip stiffnesses were not calculated (Ambegaonkar et al., 2011). The ankle has been reported to be a primary factor in determining vertical stiffness during unilateral drop jumps in males as it accounted for most of the variance in stiffness measurement (Maloney et al., 2017). The ankle was also noted to be the leading determinant of alterations in leg stiffness in females performing a drop jump while wearing ankle braces (Mason-Mackay, Whatman, & Reid, 2016).

Like other aspects of landing mechanics, sex appears to play a role in how stiffness is regulated while landing. While performing a volleyball specific countermovement jump with a bilateral landing, Hughes and Watkins noted that females had lower normalized leg and knee stiffnesses when compared to males (Hughes & Watkins, 2008). During hopping, passive ankle stiffness and leg stiffness have also been shown to differ between males and females, where higher passive ankle stiffness was positively correlated to leg stiffness, which may predispose females to a greater risk of ACL injury from an increased reliance on the gastrocnemius (Hobara, Kato, Kobayashi, & Ogata, 2012). In a drop landing study, Wang and others noted that women utilized a knee dominant landing strategy whereby the knee stiffness decreased when landing from a higher height, suggesting a potential increased risk for injury due to decreased stability (I.-L. Wang et al., 2015). While performing a single-leg drop landing, male and female dancers were shown to exhibit a softer, more compliant landing when compared to athletes stemming from lower ankle and knee joint stiffness (Ward et al., 2019). This study also noted that the knee was the most variable joint in males, but no one joint was noted to be a primary factor in females (Ward et al., 2019).

Leg stiffness represents the combination of individual joint stiffnesses when performing a landing, and even if leg stiffness remains constant, the behavior at each joint might change under different conditions. Many investigations addressing stiffness utilize repetitive tasks such as walking or hopping, and there has been little focus on landing. Related to general biomechanics landing studies which do not directly measure stiffness, there is evidence that sex plays a role in how stiffness is regulated while performing landings. Further, differences between drop landing and drop jump designs are particularly important to distinguish as research highlights that joint and leg stiffnesses vary based on the task performed. There is not a consensus on which joint in the leg is most important in regulating stiffness while landing, even further highlighted by task demands such as varied height, secondary tasks like performing a maximal jump, or external supports, among others which can influence how the joints provide stiffness to dissipate the forces of landing.

2.5.5: External Ankle Support and Stiffness

Because ankle taping or bracing serve to add external support to the ankle joint, it stands to reason that there would be a noticeable change in the stiffness at the joint likely attributed to the accompanying reduction in ROM. While stiffness has been investigated after the application of ankle support in hopping, the available literature related to drop landing rather than running or hopping is limited and requires further study. As both hopping and drop landing occur almost entirely in the vertical direction, hopping studies investigating stiffness have the potential to provide meaningful similarities to landing from an elevated platform. In the only previous study to consider the effect of taping or bracing on joint stiffness in a drop landing task, applying ankle braces to both ankles resulted in increased ankle and leg stiffness with a negligible effect at the knee (Mason-Mackay, Whatman, Reid, et al., 2016). Furthermore, it is critical to understand the individual joint contributions to vertical and leg stiffness independently during bilateral landings, especially after the unilateral application of external ankle support.

In a two legged hopping task, the influence of both ankle tape and braces were assessed after a 10-minute treadmill jog on vertical leg stiffness, finding that neither taping or bracing altered vertical stiffness (SA Williams & BL Riemann, 2009). In this case vertical leg stiffness was measured more similarly to vertical stiffness since both legs were combined together for the stiffness calculations, and results suggested that either the exercise protocol reduced the restrictive properties of the ankle support, or that the joints of the lower limb compensated for the external ankle support (SA Williams & BL Riemann, 2009). A potential explanation for a lack of change in stiffness with ankle taping or bracing could also have been due to lesser impacts from hopping, as compared to landing from a higher height as seen in drop landings (SA Williams & BL Riemann, 2009).

Similarly, in an investigation utilizing single-leg hopping, there was no change in stiffness after external ankle support (Chang et al., 2008). During a single-legged hopping task, while leg stiffness increased with increased hopping frequency, the addition of external ankle support did not increase leg stiffness in the different hopping frequencies (Hobara et al., 2017). Like the conclusion of Williams and Riemann, it was proposed the idea that other joints in the kinetic chain compensated for the addition of ankle support. Evidence supports after external ankle support is applied during single- and double-leg hopping tasks that leg and vertical stiffness, do not change (Hobara et al., 2017; SA Williams & BL Riemann, 2009). With a clearer consensus that the addition of ankle support does not alter leg stiffness while hopping, and the minimal investigation that has been given to ankle supports on drop landing tasks, it cannot yet be concluded what effect would be expected when performing a bilateral drop landing task, particularly when only one limb receives ankle support.

2.6: Fatigue

In any activity involving physical activity, the onset of fatigue is often a possibility, and in investigations that utilize exercise, it is critical to understand if fatigue has occurred, even when fatigue is not a key focus of the investigation, to mitigate confounding variables. As such, there are many ways fatigue or exertion can be assessed both subjectively, with means such as questionnaires, or objectively through physiological assessments such as blood lactate or expired gas analysis; and choice in methodology is often driven by the predominant aim of the investigation (Borg, 1982; Brooks, Fahey, & White, 1996; N. Williams, 2017). As fatigue is not within the scope of the presented aims, the effects of fatigue on mechanics will only briefly be addressed.

Understanding fatigue in any research protocol is important because fatigue plays a role in altering various areas of lower body mechanics during landing, especially negatively affecting the knee (Borotikar, Newcomer, Koppes, & McLean, 2008; Chappell et al., 2005; Kernozek, Torry, & Iwasaki, 2008; Madigan & Pidcoe, 2003). After fatigue, females showed greater anterior shear joint moments and less knee flexion, both of which carry an increase in risk for damage to the ACL (Kernozek et al., 2008). Additionally, females have been shown to have an increase in peak knee valgus after fatigue, also a strong predictor for ACL injury (Dickin et al., 2015; Pappas, Sheikhzadeh, et al., 2007). The mechanical changes stemming from fatigue regarding increased knee valgus further detriments the muscular control of landing, increasing injury risk, when considering females already land with more frontal plane knee motion (Jacobs et al., 2007). Furthermore, females exhibited a number of ACL injury risk factors including larger knee abduction and internal rotation when fatigued (S. G. McLean et al., 2007). Another predictor of injury during landing is an increase in peak GRF, though changes to peak GRF have revealed conflicting results, shown to reduce after fatigue (Santamaria & Webster, 2010; M. P. Smith, Sizer, & Roger James, 2009), as well as increase after fatigue (Brazen, Todd, Ambegaonkar, Wunderlich, & Peterson, 2010; Pappas, Sheikhzadeh, et al., 2007).

During landing, appropriate muscular control at the knee to reduce strain on the ligaments in the knee is critically important, as under high loading, the musculature is more compliant, and better suited to dissipate the load without damage. Therefore, an increase in fatigue limits the ability of the musculature to provide appropriate muscular control to dissipate these forces. The degree of fatigue, as well as the amount of time engaged in activity are critical components in understanding increased injury risk. Potential evidence for a fatigue effect on risk for ankle sprain is supported by a recent meta-analysis whereby the occurrence of ankle sprain increased towards the end of a sporting event, or near the end of the first half in male athletes (de Noronha, Lay, Mcphee, Mnatzaganian, & Nunes, 2019). Similarly, in lengthy sport participation, anterior and posterior knee laxity has been shown to increase, which is a potential risk factor for ligamental damage (Steiner, Grana, Chillag, & Schelberg-Karnes, 1986).

Fatigue has been shown to alter lower body mechanics in various measures of kinetics and kinematics which can increase the risk of injury. Although the way fatigue affects landing mechanics might exhibit varied results based on the measures and protocol, there are indeed changes in mechanics which must be accounted for when exercise is a component of research design. As such, where exercise is utilized in a research protocol, it is important to mitigate the occurrence of fatigue, unless it is of direct interest in the investigation. Further, to ensure the viability of results, assessing fatigue during the investigation provides a safety net to rule out the confounding influence of exercise induced fatigue and bolster the efficacy of research findings.

2.7: Coordination

The human body is a combination of numerous biological systems which work independently, yet also work in a complex coordinative fashion to perform common everyday tasks. There are over 600 skeletal muscles which enable voluntary movement through over 100 degrees of freedom (Turvey, 1990). While performing a task, the regulation of these degrees of freedom into a coordinative pattern is of great importance to successful completion of a task (Fitch, Tuller, & Turvey, 1982). Initially when learning a skill, the body is often rather rigid, in the search to minimize the number of movements to focus centrally on the task (Turvey, Fitch, &

Tuller, 1982). Through the lifespan as skill increases in performing a task, degrees of freedom can be unlocked, where attentional resources can be centered on other external demands.

Coordinative patterns between muscles enable multiple structures to function together without the additional cognitive demand (Tuller, Turvey, & Fitch, 1982). When performing a motor task, even with a consistent endpoint, the path to get there will be variable in reaching that endpoint (Bernstein, 1967). Kelso suggests that groups of muscles work together in a synergistic fashion and behave like a non-linear oscillator (Kelso, Holt, Rubin, & Kugler, 1981). It has been observed that when hopping, the ankle, knee, and hip joints functioned in a highly coordinated pattern during the ground contact phase noting that the task demands of hopping successfully requires appropriate coordination of the joints in the leg (Rapoport et al., 2003). Ultimately, these properties result in the phasic and cyclic relationships of joints and segments over a time series. Because we have the capacity to adapt outside of linear constraints, we can adapt and reset to preferred movement patterns after perturbations. During gait, coupling is exhibited between the foot, shank, and thigh in a coordinative structure to dissipate external forces and continue forward motion. Similarly, during landing, coordinative patterns are exhibited by the joints in the lower body.

The ability to perform complex actions in a functional manner requires a high degree of coordination between the movements that comprise that skill. Specifically, in gross motor skills several muscle groups need to function together in a cohesive manner to complete the overarching task. While working together, not all muscle groups are working at the same time and in the same manner. As such, there exists an inherent variability as the involved muscles activate in a time and position dependent manner. Variability and complexity are important qualities in human movement, as they enable numerous ways to interact with a three-dimensional environment and adapt to perturbations. Complexity of movement patterns provides meaningful insight into healthy functioning (Lipsitz, 2002). Decreases in variability and loss of complexity in movements are indicative of a less adaptable system and can represent a less healthy biological system. Ankle taping acts as an external constraint at the ankle, and its effects have been shown to have various effects at not only the ankle, but other joints in the leg as well. Ankle taping, then, could be viewed as a type of perturbation that must be attenuated in some manner to limit any potential negative effects on mechanics.

While the means and standard deviations of biomechanical variables can provide insight into the trends of movement patterns, they lack the ability to uncover the small deviations that occur throughout a movement, and over a series of movements. Analyses of coordination, such as CRP allow for assessing movement dynamics throughout the duration of a movement, normalized so that each complete movement acts as an individual trial, thus allowing large volumes of data to be assessed. Though CRP is often applied to rhythmic movements such as walking, running, or hopping, movements which are repeated in a non-rhythmic pattern such as jumps or landings could benefit from this analytic method to compare performance over multiple trials of a discrete task.

2.7.1: Assessment of Coordination

CRP is a method to determine phasic relationships throughout a movement utilizing a time series of the kinematic properties of two joints (J. Hamill, Haddad, & McDermott, 2000). To produce a representation of CRP, a phase portrait is necessary to transform the position and velocity of a signal into a graphical representation of the signal's dynamics. Prior to calculating CRP, the joint or segment angles and velocities benefit from normalization to fit a range of -1 to 1 so as to center around the origin of the phase space (Fuchs, Jirsa, Haken, & Kelso, 1996; Lamb & Stockl, 2014). In a methodology suggested by Lamb and Stöckl in a review of CRP analyses, the joint angular displacement and angular velocities were normalized independently as to preserve the 0 value of velocity which represents a transitory component in a movement (Lamb & Stockl, 2014). Equation 4 demonstrates the normalization of the angular velocity, where Equation 5 demonstrates the normalization of angular displacement, where y(t) is the normalized time series (Lamb & Stockl, 2014).

$$f\big(y(t_i)\big) = \frac{y(t_i)}{\max\,(y(t))} \qquad 4$$

$$g\big(y(t_i)\big) = 2 * \left(\frac{y(t_i) - \min\big(y(t)\big)}{\max\big(y(t)\big) - \min\big(y(t)\big)}\right) - 1 \qquad 5$$

After normalization, the phase angle can then be calculated by taking the arctangent of the normalized velocity divided by the normalized position (Equation 6).

$$\Phi(t_i) = \arctan\left(\frac{f(y(t_i)}{g(y(t_i)}\right) \qquad 6$$

CRP takes the difference in phase angle at each time point from one joint to another, where Φ_{joint}(t) represents the phase angle of a joint which describes the phase relationship between joints (Equation 7).

$$CRP(t) = \Phi_{joint_1}(t) - \Phi_{joint_2}(t) \qquad 7$$

The full expansion of Equation 7 is further explained by Equation 8 (Lamb & Stockl, 2014).

$$CRP(t) = \arctan\left(\frac{f_1(y(t_i) * g_2(y(t_i) - f_2(y(t_i) * g_1(y(t_i)}{g_1(y(t_i) * g_2(y(t_i) + f_1(y(t_i) * f_2(y(t_i)}\right) \qquad 8$$

Phase plots provide the graphical representation of the signals of interest, whether kinematically with joint position and velocity, or kinetically with joint moment and joint position. In a cyclic task, the phase plot will typically resemble a repetitive pattern resembling somewhat of a circular or ovoid shape, whereas in a discrete task, the plot will have a clear beginning and end, as the movement is defined by a clear start and end position. Phase plots allow a visualization of how the signals of interest interact with one another, instead of through the time domain medium. Relative phase is the arithmetic difference between this phase angle for one joint and another joint at the same moment and can be continuously computed across the time series as CRP.

CRP enables comparisons of joint phases throughout the entirety of a movement, rather than the average throughout the series. As the role of CRP is to better evaluate the complexities of a movement throughout a time series, when analyzing the coordinative pairs, it is important to avoid boiling the results to a traditional mean and standard deviation. To assess coordination variability, the standard deviation at each point in the landing absorption phase between joints is calculated (Hamill, Palmer, & Van Emmerik, 2012; Hamill, van Emmerik, Heiderscheit, & Li, 1999). By taking the differences between the signals throughout the time series, it is possible to assess points in the landing where coordinative patterns differ. Because the signals have been percent normalized to the landing phase from initial contact to the lowest COM position when the COM velocity is equal to 0, variability at specific points in the landing phase can be highlighted. Ultimately, the benefit of utilizing CRP lies with the fact that it enables a better understanding of the complexities of movement throughout its entirety.

Assessment of coordination facilitates a deeper understanding about the qualities of a movement of not only joints independently, but how they interact with one another throughout the duration of the movement rather than boiling down a large amount of information into a single value. Traditionally, CRP is calculated using the kinematic components of movement: position and its first derivative, velocity which gives a representation of the qualities of the movement as they relate and interact together. This methodology works well to describe rhythmic movements such as gait and running, as the transitions between flexion and extension are less encumbered by highly shock-inducing maneuvers such as landing from a jump which might benefit from a kinetic approach to assessing coordination.

2.7.2: Considerations for Novel Methodologies of Coordination Assessment

Traditional methods of assessing coordination focus on the kinematic components of a movement, being position and its derivative, velocity. Hence traditionally, CRP has provided a measure of the kinematic coordination between joint or segment motions. This assumes that position and velocity are the appropriate descriptors of a joint or segment behavior, and the coordination between joints or segments. This makes sense when studying oscillating fingers, handheld pendulums, or other simple rhythmic movements. However, in tasks where a collision is significant (e.g., landings), other quantitative descriptors of the coordination may be more appropriate. Therefore, the question remains as to whether this method of coordination assessment is useful in all cases, and if other means of coordination calculation could better explain how joints are coordinated during more complex movements.

Because drop landing is a high velocity and high impact movement, the reliance of velocity as a factor of calculating phase has the potential to overestimate the phase angle calculation, as velocity is the numerator in the calculation of phase angle. Particularly, as the joints utilize differing ranges of motion while absorbing landing which further changes the relationship between joints when using velocity as a factor in comparing phases. Despite normalization of the signals, when the values of the joint velocity are so high, the slope of the line scaling factor as normalized from -1 to 1 is steeper and can under-represent the less extreme values of joint velocity. With a high-impact task such as a drop landing requiring control of the force dissipation, it is possible that the kinematics alone do not explain the complexities of the landing process and the kinetics might provide more meaningful information. By utilizing the joint moment instead of joint velocity, the values of joint moment are more constrained, and potentially allow for a more equitable comparison of coordination during landing using joint moment and joint position. Through the utilization of both kinetic and kinematic coordination calculations, it could be possible to uncover whether coordinating the motion or coordinating the force distribution takes precedence when landing.

Further, landings rely on large amounts of muscular contribution to slow the descent of the COM in a controlled manner. Thus, while performing a maneuver where equitable simultaneous force dissipation is paramount, coordination of forces experienced at the joints can provide insight to the control mechanics while landing, especially when a joint on one limb receives an external constraint. Beyond symmetrical kinematics, it is also worth noting the importance of symmetry of the kinetics during landing, to not overload one limb or joint and increase the risk for injury. As such, it becomes apparent that investigating the kinetic symmetry and coordination of the lower limbs would provide meaningful information about preferred force distribution while landing.

2.7.3: Coordination and External Ankle Support

Ankle taping acts to constrain motion at the ankle, ultimately restricting the available degrees of freedom during tasks. In terms of a protective effect against inversion ankle sprains, this effect is beneficial; however, with the reduced DOF comes the potential for a less variable movement pattern. When the musculoskeletal system experiences an external constraint, the ability to confront and respond to perturbations can be reduced, thus potentially increasing the risk for injury.

Bilateral ankle taping has been assessed during gait, utilizing a vector coding approach which allows the comparison of a proximal and distal segment angle to one another throughout a movement, noting that tape reduced variability in shank-rearfoot coupling, which suggests that taping acts as a constraint through its reduction in ROM (Herb et al., 2016). Utilizing a relative phase analysis during gait, the application of ankle braces have been shown to reduce variability in lower limb coupling patterns (Jagodinsky et al., 2020). In a more analogous task to drop landing due to the primarily vertical movement, during single-leg hopping, ankle bracing also reduced coupling variability in the foot-shank (Jagodinsky, 2016). Taken together, it appears that supporting the ankle externally limits the available motion at the ankle and makes tasks more consistent, which potentially limits the adaptability to external perturbations.

In a treadmill walking study, Alyami & Nessler limited participants' plantarflexion through a custom device placed on one ankle and calculated the phase coordination index, which assesses the phasic relationship between limbs, in a similar fashion to DRP (Alyami & Nessler, 2021; Plotnik, Giladi, & Hausdorff, 2007). While walking with the restrictive brace, asymmetry was larger than walking in the control group, suggesting that a unilateral restriction can influence a bilateral task like walking, which could also appear in other bilateral tasks, particularly when time synched coordination is important, as in a bilateral landing (Alyami & Nessler, 2021). The addition of ankle tape acts as an external constraint, whereby there is potential for a reduction in intertrial variability and possibly an increase in the need for other joints to modulate movement patterns to compensate, particularly when only one ankle is restricted. Furthermore, a better understanding of the coordination as well as the coordination variability throughout the joints in the limb will provide researchers with insight to the overarching effects of external ankle support.

2.7.4: Joint Coordination During Landing

When landing bilaterally, sharing the load equally between limbs maintains a more secure landing by keeping the force within the centralized base of support and sharing the load equally between limbs. During landing, as the joints of the leg go through their ROM to dissipate the forces from landing, it is important to consider their motion respective to the respective joint in the opposing limb, as well as the other joints in the same limb. Coordination describes the relationship of the joints throughout the entire landing process time-series, rather than a discrete time point such as the peak or mean of the movement, enabling a richer understanding of the complexities of the entire movement. Assessing the relationships between joints during landing utilizes the time-series data, and incorporates the non-linearity of biomechanical measures, rather than just utilizing the relative peaks of signals (Burgess-Limerick, Abernethy, & Neal, 1993).

Few previous studies have quantified coordinative patterns in intra-limb and inter-limb dynamics during landing. However, a recent investigation utilized DRP during single-leg landings without ankle support, showed females had greater variability in hip-knee and hip-ankle joint couplings regardless of whether they landed on the dominant or non-dominant limb, which could suggest a reliance on the hip to dissipate the force of the landing (Hughes, 2020). In contrast to bilateral landing, it was shown that in the dominant limb, greater mean relative phase variability in the knee-ankle coupling suggested the role of limb dominance when landing unilaterally, highlighting the importance of understanding coordinative relationships while landing bilaterally (Hughes, 2020). While the use of DRP allows for comparisons between joints, it is limited by utilizing discrete time points, rather than assessing coordination throughout a movement. Those findings could be further explored utilizing CRP to uncover differences in joint dynamics throughout the landing.

Little research has focused on the coordination of the joints in the leg throughout the duration of a landing task. When external ankle support has been utilized during continuous measurements, changes to the coupling and variability have been noted which highlights the potential for detrimental effects of force attenuation which could potentially increase the risk of joint injuries. Landing bilaterally is a complex movement to safely dissipate the forces of landing, and the dynamic behavior of the joints throughout that process can provide insight to how those forces are dissipated. As ankle support has been noted to change coupling and variability, in a bilateral landing task where coordination is an important component in landing safely, it is important to understand the influence a unilateral ankle taping could have on the lower limb coordination dynamics.

2.8: Summary

Lower body injuries are an unfortunate yet common occurrence during both exercise and sport performance. Females have been consistently reported to sustain damage to the ACL at a higher rate than their male counterparts, and there is some evidence which shows they also sustain ankle sprains at a higher rate. There are several anatomic and physiological differences which account for this disparity, and while training can moderate some of the negative aspects, some of these sex-based differences are non-modifiable. A thorough understanding of the mechanics that predispose individuals to certain injuries enables the design of intervention programs to reduce injury risk. Because of the prevalence of ankle sprains, external ankle support is often used to both prevent ankle sprains, as well as support the ankle after sustaining a sprain. Drop landings have been commonly utilized to assess kinetics and kinematics because they are like the landings often performed in exercise and sport performance and are easily repeatable within the lab. Through the above presented literature, internal factors such as sex, age, and experience; along with external factors such as ankle taping, landing height, and exercise all play a role in altering lower body landing mechanics; however, the complex interweaving relationships have yet to be fully explored. Moreover, the body of literature specifically related to stiffness and coordination during landing tasks has not yet been investigated with great depth and requires more study.

External ankle support in the form of taping and bracing has been shown to alter lower body kinetics and kinematics in several ways. However, it is still unclear how the unilateral application of ankle support influences kinetics and kinematics while performing a bilateral landing. Therefore, the overarching aims of these investigations presented in this dissertation are to provide a more comprehensive view of the influence of unilateral ankle taping on the kinetics, kinematics, and time dependent components of bilateral drop landings. More specifically, this series of investigations sets out to fill gaps in the research which have not yet been addressed or require further investigation, particularly in the stiffness and coordination literature. Through addressing the biomechanics, joint and total stiffnesses, as well as coordinative patterns during landing, a more complete understanding of the influence of unilateral ankle tape when performing a bilateral landing can be attained.

CHAPTER 3
STUDY METHODOLOGY

3.1: Participants and Study Design

Each aim will be addressed through one investigative data set which consists of two visits to the laboratory in a repeated measures design. Prior to recruitment or testing of participants, Old Dominion University Institutional Review Board approval [984816-8] was attained and was maintained throughout the duration of all participant data collection (Appendix A1), and participants were provided information about the procedure and provided informed consent (Appendix A2). To participate in the investigation, inclusion criteria needed to be met. Those criteria consisted of being at least 18 years of age at the start of the investigation, no history of ACL injury, no history of injury to the lower limb that continues to influence daily function, and physically active for at least 3-4 days per week with the ability to run for at least 15 minutes without stopping (Appendix A3). Participants were excluded if they did not meet all the inclusion criteria. Based on a power analysis (G*Power v 3.1.9.7), twelve participants were determined to be the minimum required (Faul, Erdfelder, Lang, & Buchner, 2007). Thirteen college and young adult aged female participants were recruited for participation; however, due to incomplete kinetic data for one participant, data analysis was completed with the remaining twelve participants (age 24.0 ± 3.6 years, height 1.64 ± 0.06 m, weight 58.7 ± 4.4 kg). All participants were recruited from the campus of Old Dominion University through convenience sampling. Participants came to the laboratory for testing on two separate visits separated by a minimum of twenty-four hours. In a counterbalanced order, participants would complete the control/no tape landing set on one day and the experimental/tape landing set on the other day. In the ankle tape condition, ankle tape was applied by a certified athletic trainer utilizing the Gibney closed basketweave with heel lock and figure-eight which consisted of two anchors, three stirrups, three horseshoes, two figure eights, two heel locks, and then closed (Perrin & McLeod, 2018). For the ankle tape procedure, no pre-wrap or adhesive spray were utilized; however, heel and lace pads were applied for participant comfort, and Johnson & Johnson cloth athletic tape was used for all participants.

3.2: Data Collection

Upon the first visit to the laboratory, the participants were provided the details of the investigation and provided informed consent prior to any data collection. At the start of each data collection, participants were outfitted with four Trigno surface electromyography (EMG) sensors (Delsys Inc., Natick, MA, USA) placed on the tibialis anterior, gastrocnemius medialis, rectus femoris, and biceps femoris of the dominant limb. To ensure a proper adherence to the skin with minimal impedance, the area of skin was prepared by the removal of hair should any be present, followed by a topical abrasion utilizing an abrasive and conductive gel and sterilization with an alcohol wipe (Konrad, 2005). The dominant limb was defined as the leg which the participants would use to kick a ball for maximum distance. To normalize the EMG signal, participants completed a maximal voluntary isometric contraction (MVIC) against manual resistance for each muscle in an open kinetic chain. To test the MVIC for each of the muscles, the below procedures were followed. The tibialis anterior was tested with the participant in a seated position with their legs extended on a trainers table and asked to pull their toes towards their head. To test the medial gastrocnemius, participants laid prone on the trainer table with their foot hanging off the end and asked to point their toes away from them. The rectus femoris was assessed with the participant seated with their knee at the edge of the table and asked to kick forward. The biceps femoris was tested with the participants seated with their knee at the edge of the table and asked to pull their lower leg to the back. Following the MVIC testing, participants were then outfitted with a retroreflective marker array on the lower body for three-dimensional motion capture data collection. Although EMG data was collected as a part of the protocol, the data was not utilized within the scope of the aims presented here.

A full three-dimensional kinematic analysis using an 8-camera array (Vicon, Oxford, UK) was paired with kinetic data from two six degrees-of-freedom force plates (AMTI, USA), whereby one force plate was located under each limb. Kinematic data was sampled at 120 Hz, and synced with kinetic data at 960 Hz, and EMG at 1,920 Hz. After completion of a power-spectral-density analysis, kinematic data was low pass filtered with a cutoff frequency at 8 Hz using a fourth order zero-lag Butterworth filter. Ground reaction force data were low pass filtered at a frequency of 12 Hz using a fourth order zero-lag Butterworth filter. Kinematic data were processed in Visual3D (v4.93, C-Motion, Germantown, MD USA) and additional mathematical calculations were completed utilizing a custom script in MATLAB (The MathWorks Inc., Natick, MA USA). Ground reaction forces were normalized to participant body weight.

A rigid body, linked segment lower body model based upon anthropometric measurements reported by de Leva was designed (de Leva, 1996a). Markers were placed on the anterior and posterior superior iliac spine in addition to the iliac crest to define the pelvis. The thigh segment was defined through markers placed on the greater trochanter, and medial and lateral femoral condyles. Shank segments were defined by markers placed at the tibial tubercle, and medial and lateral malleoli. The foot was represented as a single segment using the calcaneus, head of the first metatarsal, base, and head of the fifth metatarsal, and a marker on the distal hallux was utilized for positional displacement calculations. Cluster arrays were placed on the shank and thigh segments for tracking during trials. The hip joint center was defined utilizing the de Leva measurements using the greater trochanter markers and the iliac crest markers by placing the hip joint center proximal to the greater trochanter, as defined by the segmental length hip joint center, then subtracted by the scaled difference between the greater trochanter and iliac spine (de Leva, 1996a). The knee and ankle joint centers were defined as the midpoint of the medial and lateral femoral epicondyles and malleoli, respectively. Prior to data collection, participants were recorded in a static position to determine specific anatomical definitions followed by a functional calibration to facilitate motion analysis tracking and labeling. A picture of the marker set utilized for the model can be found in Appendix A4.

3.3: Task Procedures

As a part of the research design, participants jogged for fifteen minutes at a self-selected pace with zero degrees of incline on a pressure instrumented treadmill (Zebris Medical GmbH, Germany). The landing protocol consisted of a step off drop landing from a custom-built platform (Appendix A5), where participants placed the distal aspect of the non-dominant toe at the edge of the platform and hung their dominant limb over the force platforms. During the landing, participants were instructed to place their hands on their shoulders and maintain them in that position while landing to reduce the occlusion of the hip markers. Participants were instructed to adopt a natural landing pattern that felt comfortable and could practice landing with one foot on each force plate until they felt comfortable doing so. A successful trial was defined as landing with each foot landing wholly on an independent force plate and returning to an upright posture without losing balance. In the event participants had an unsuccessful landing, additional landings were performed until a successful landing was achieved.

On each visit to the laboratory, participants completed five baseline trials at each height (30, 45, and 60 cm) in a randomized order to assess for any differences between visits to the lab. This was followed by five more landings from each height after either receiving ankle tape or the control (no tape), followed by the last set of five landings from each height after completing the fifteen-minute run exercise protocol. This comprised fifteen landings from each height totaling forty-five landings per visit to the laboratory, totaling ninety landings between both days. Each visit took approximately 75 minutes, totaling 150 minutes between the two visits to the lab. Following the exercise protocol, participants were asked to assess their rate of perceived exertion (RPE), and were allowed to rest and were reassessed on their RPE prior to completing the last grouping of landings (Borg, 1982). The RPE chart can be found in Appendix A6.

Prior to starting a grouping of landings, each participant was measured with a manual goniometer in a weight-bearing lunge on their dominant ankle, adapting the method presented in Konor et al. where participants were able to place their hands on the wall for support; however, they were not tasked with placing their knee on the wall (Konor et al., 2012). The baseline measurement allowed for assessment of any differences between visits to the lab, while immediately after tape served to assess the restrictive effect of the ankle tape immediately after application, and after exercise to assess any loss in restrictive motion because of the exercise protocol. While the non-dominant ankle ROM was not assessed, previous research has shown weightbearing dorsiflexion ROM to be similar between limbs in healthy participants (Dowling et al., 2018; Konor et al., 2012). A visual representation of the protocol can be seen in Figure 1.

3.4: Dependent Variables

Manual goniometer dorsiflexion flexibility is expressed as the angle between the fifth metatarsal head and center line of the fibula with the lateral malleolus serving as the fulcrum, similar to previous methodology and adjusted to accommodate the available handheld manual goniometer (Konor et al., 2012). In the motion capture trials, the ankle angle is defined as the angle of the foot relative to the shank, where standing upright defines the angle as zero degrees (neutral), where plantar- and dorsiflexion will be reported values away from neutral. Thus, the angles reported in the manual goniometer measurement are not directly comparable with the motion capture data. However, the manual goniometer measurement only serves to assess weight-bearing dorsiflexion flexibility prior to the landing trials, while the motion capture data provides all data for the active landing trials. Frontal plane kinetics and kinematics are reported as inverted values in the left limb to match the reporting in the right limb, as those values follow the right-hand rule.

For Study One, traditional inverse dynamics variables were the focus of the analysis, including vertical GRF, loading rate, loading time, joint kinematics at ground contact, and maximum values of joint kinetics and kinematics in the frontal and sagittal planes. Analysis of these variables will provide a view of traditional biomechanical measures to uncover the influence of unilateral ankle taping while performing a landing both before and after a bout of exercise. In Study Two, assessment of stiffness was the focus where the variables of interest included vertical and joint stiffnesses, in addition to the COM and joint ROM which one of the components defining the stiffness. Analysis of these variables will provide information about how the joints and the body changes stiffnesses with the application of unilateral ankle tape when landing from varied heights. Lastly, for Study Three, metrics of CRP and coordination variability were the focus of the investigation. The variables of interest consisted of kinetic and kinematic CRP and their associated variability which will provide information about how the ankle, knee, and hip are coordinated and how that coordination varies within the same leg, as well as the matched joints in both legs.

3.5: Statistical Analyses

Dependent variables in Studies One and Two were analyzed with repeated measures ANOVA's utilizing SPSS (IBM SPSS Version 24, SPSS Inc., Chicago, IL, USA). Significance was set to an $\alpha \leq 0.05$, and any comparisons which violated assumptions of sphericity were corrected using the Greenhouse-Geiser correction. Effect sizes were calculated using partial eta squared (η_p^2) where 0.01 represents a small effect, 0.06 represents a medium effect, and 0.14 represents a large effect (Cohen, 1988). Sidak post-hoc tests will be used to further reveal significant interactions. Both kinematic and kinetic coordination data was statistically analyzed in MATLAB utilizing open source code for statistical parametric mapping (SPM) (v M.0.4.7) to determine points in the landing that were significantly different between taped and un-taped conditions (Pataky, 2010, 2022; Penny, Friston, Ashburner, Kiebel, & Nichols, 2011). Further descriptions of methodology and statistics will be discussed in each respective study.

3.6: Protocol Organization

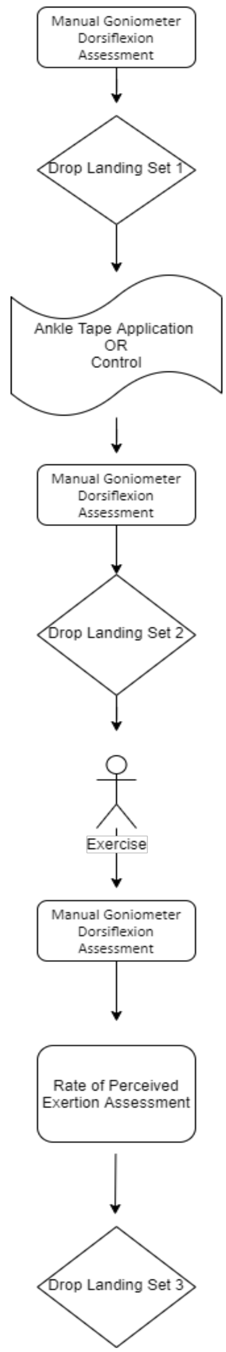

Figure 1. Study procedure repeated on two days. Order of taping and control days was counterbalanced.

CHAPTER 4

STUDY ONE – BIOMECHANICS OF BILATERAL DROP LANDING WITH UNILATERAL ANKLE TAPE

4.1: Introduction

Ankle sprains are the most common injury in exercise and sport participation and are typically incurred after a cutting maneuver or a high impact landing (Doherty et al., 2014; Roos et al., 2017). Among individuals who have sustained an ankle sprain, residual ligament laxity is common, and often in the treatment and prevention of further ankle sprains the use of prophylactic ankle supports, including tape or braces, is recommended (Denegar, Hertel, & Fonseca, 2002; Dizon & Reyes, 2010; Kaminski et al., 2013). An important aspect of ankle taping or bracing is the resultant restriction of ankle motion in both the frontal and sagittal planes, reducing inversion/eversion and dorsiflexion ranges of motion to limit the excessive motion which causes ligamentous damage (Cordova et al., 2000; Miller et al., 2012). Despite the benefits of movement restriction in the ankle to prevent ankle sprains, there is potential for alterations to kinetic measures, including peak GRF and loading rate, which could increase injury risk elsewhere in the limb (Abián-Vicén et al., 2008; Cordova et al., 2010; DiStefano et al., 2008; Hodgson et al., 2005; Mason-Mackay, Whatman, Reid, et al., 2016; Megalaa et al., 2022; Okamatsu, 2014; Riemann et al., 2002; Stoffel et al., 2010; West et al., 2014; Yi et al., 2003).

Knee kinematics can also change due to ankle taping. Increased knee flexion at contact has been reported previously, although differences in total ROM during landing is not clear (Cordova et al., 2010; DiStefano et al., 2008; S. A. Williams et al., 2018). Of particular concern is the potential for an increase in dynamic knee valgus while landing (Howe et al., 2019; Y. L. Lima et al., 2018; Malloy et al., 2015; Sigward et al., 2008). Large knee valgus has been implicated as a mechanism in ACL injury in a number of studies (Hashemi et al., 2011; Hewett et al., 2005; Ireland, 1999; Kanamori et al., 2002; Markolf et al., 1995; Quatman & Hewett, 2009), suggesting that an ankle prophylactic could have deleterious effects on other joints during landings. Women may be especially susceptible to such concerns as females have been shown to utilize a greater ROM at the ankle when compared to males, presenting the possibility of further deficits to safe landing mechanics while an ankle is taped (Decker et al., 2003; Kernozek et al., 2005; S. G. McLean et al., 2007). There does not appear to be a clear influence of ankle taping or bracing on hip kinematics (Cordova et al., 2010; Mason-Mackay, Whatman, & Reid, 2016; Simpson et al., 2013). While researchers have begun to consider the effects taping or bracing an ankle has on other joints of the ipsilateral leg, little consideration has been made for the impact on the joints of the contralateral leg.

Often, athletes only tape one ankle; however, little is known about how a unilateral ankle support impacts both legs during bilateral landings, as of the writing of this dissertation, only one study has investigated this paradigm (Dewar et al., 2019). To understand the biomechanical effects of ankle taping of one or both ankles, many studies have investigated single leg and bilateral landings, but it is still not clear if there are adaptations across both limbs when only one ankle is taped. Investigations that utilize ankle supports often apply the support to both ankles; or, when utilizing a single leg landing, only apply the support to the test ankle. The one study which has investigated the effect of bracing one ankle when performing a bilateral landing, found that the effects of ankle bracing were localized to only the braced ankle, which saw a reduction in inversion angle and an increase in inversion moment with no changes to the unbraced ankle (Dewar et al., 2019). Unfortunately, the study did not measure sagittal plane mechanics of the ankle, nor any mechanics of the knees and hips, and the sample was predominantly male. Therefore, the effects of unilateral ankle taping on the kinetics and kinematics of the proximal joints of the kinetic chain during a bilateral landing requires further investigation. Also, given that women have a higher incidence of ankle and knee injuries, ankle taping could negatively influence landing mechanics stemming from a preferential utilization of greater ankle sagittal plane ROM when compared to men (Arendt & Dick, 1995; Decker et al., 2003; Doherty et al., 2014; Kernozek et al., 2005; Montalvo et al., 2019).

After ankle taping or putting on a brace, athletes' warm-up and begin playing or training. While this changes the temperature and elastic properties of the muscles, tendons, and ligaments, the mechanical properties of the tape or brace can also change. Exercise has been shown to decrease the restrictive properties of ankle taping after as little as ten minutes of exercise; however, the protective ability to resist inversion sprains is not clear, as some studies report adequate residual protection, while others note little remaining protective capabilities (Best et al., 2014; Meana et al., 2008; Purcell et al., 2009; Ricard et al., 2000;

Tamura et al., 2017). As such, it is important to better understand the level of residual effectiveness of ankle taping after an exercise protocol with the aim to uncover any subsequent effects on the ankle and other joints of the taped and un-taped legs.

Therefore, the aims of this study were to understand the influence that unilateral ankle taping has on the frontal and sagittal plane kinetics and kinematics of not only the ankle, but of the knee and hip as well. Additionally, this study seeks to understand if exercise has a mitigating effect on the mechanical restriction of ankle taping. The current investigation focused on understanding the biomechanics of landing from 60 cm, under tape and no tape conditions and both before and after exercise. The biomechanical variables of interest were peak vertical ground reaction force, time to peak force, loading rate, and lower limb joint motion and moments in the sagittal and frontal planes.

As previous investigations have found an increased loading rate from a faster time to peak force, it was hypothesized that ankle taping will increase the loading rate of the taped limb, as well as the loading rate for both legs (Cordova et al., 2010; Niu et al., 2016; Riemann et al., 2002; Simpson et al., 2013; Yi et al., 2003). While the overall reporting on sagittal plane mechanics at the knee is not clear, as the ankle tape provides sagittal plane restriction to the ankle, participants are expected to adopt a strategy of increasing knee angle at contact, while peak flexion remains constant, as has been presented previously (Cordova et al., 2010; DiStefano et al., 2008; S. A. Williams et al., 2018). Like the results of Dewar and colleagues, it is expected that no changes in frontal plane ankle mechanics will be observed in the un-taped ankle. Because reduced ankle dorsiflexion flexibility has been shown to increase frontal plane knee movement, it is hypothesized that during a bilateral landing, frontal plane knee movement will be increased in the dominant knee after taping the dominant ankle. Exercise is expected to mitigate the restrictive effects of the tape which will then carry over to the dynamic mechanics when landing, revealing an increased dorsiflexion ROM during landing.

4.2: Methodology

After attaining local institutional review board approval, twelve female participants (age: 24 ± 3.6 years, height: 1.64 ± 6.29 m, weight: 58.68 ± 4.42 kg) were recruited to participate in this investigation. Previous injury to the ACL, or any lower body injury which limits current physical activity were exclusionary criteria. Additionally, participants were physically active as defined by currently exercising 3-4 times per week and able to run for at least 15 minutes without stopping, in addition to at least six months free of any lower body injury. All participants were provided the same model shoe in the appropriate size to standardize footwear.

Participants visited the laboratory on two separate days to complete the control and experimental conditions in a counterbalanced order, with a minimum of 24 hours rest in between laboratory visits. Under the experimental condition, ankle tape was applied by a certified athletic trainer to the dominant ankle, defined as the leg which the participant would use to kick a ball for greatest distance. The ankle tape was applied using the Gibney closed basketweave with heel lock and figure-eight, which consisted of two anchors, three stirrups, three horseshoes, two figure eights, and two heel locks (Perrin & McLeod, 2018). To standardize landing, participants were asked to keep their arms crossed, with hands on each shoulder and to land in a comfortable, natural pattern. Participants completed a drop landing by stepping off a 60 cm high box, leading with their dominant leg, and landing bilaterally with one foot on each force plate. Two six-degrees-of-freedom force plates (AMTI, Watertown, MA, USA) sampled at 960 Hz. Participants were outfitted with 44 retro-reflective markers in a rigid linked segment lower body model using de Leva's segmental anthropometric definitions (de Leva, 1996a). Markers were placed on the anterior and posterior superior iliac spine in addition to the iliac crest to define the pelvis. The thigh segment was defined through markers placed on the greater trochanter, and medial and lateral femoral condyles. Shank segments were defined by markers placed at the tibial tubercle, and medial and lateral malleoli. The foot was represented as a single segment using the calcaneus, head of the first metatarsal, base and head of the fifth metatarsal, and a marker on the distal hallux was utilized for positional displacement calculations. Cluster arrays were placed on the shank and thigh segments for tracking during trials. Three-dimensional motion capture was assessed utilizing an eight-camera array sampled at 120 Hz (Vicon, Oxford, UK).

Upon each visit to the laboratory, participants completed baseline trials to account for differences between days. Each day's collection included the completion of five baseline landings, followed by five landings after the application of ankle tape or the control (no tape), and another five landings after a 15-minute self-selected pace forward run on a treadmill. Participants were allowed to rest until they were at an RPE value of 9 (very light) or lower. A successful trial involved participants landing with each foot fully on its respective force plate and maintaining balance while returning to a static upright posture. Any unsuccessful trials were repeated until the required number of successful landings were attained. For this investigation, only landings from the 60 cm platform were utilized for the data analysis. Prior to each set of landings, dorsiflexion ROM of the dominant ankle was measured by manual handheld goniometer with the participant in a weight-bearing lunge position.

Both kinetics and kinematics have been shown to change after fatigue (Kernozek et al., 2008; Madigan & Pidcoe, 2003; Scott G McLean et al., 2007; Pappas, Sheikhzadeh, et al., 2007), thus, to mitigate this potential effect, participants rated their level of perceived exertion (Borg, 1982), and were allowed time to rest before completing the last group of landings.

After completion of a power spectral density analysis, kinematic data was low pass filtered with a cutoff frequency at 8 Hz, while kinetic data were low pass filtered with a cutoff frequency at 12 Hz using a fourth order zero-lag Butterworth filter in Visual3D (v4.93, C-Motion, Germantown, MD USA). Additional mathematical calculations were completed in MATLAB (The MathWorks Inc., Natick, MA). Weight-bearing dorsiflexion measurements are expressed as the angle between the fifth metatarsal head and center line of the fibula with the lateral malleolus serving as the fulcrum, similar to previous methodology and adjusted to accommodate the available handheld manual goniometer (Konor et al., 2012). In reporting the degrees of dorsiflexion flexibility while performing the weightbearing lunge, a larger number represents the foot in a neutral position when standing upright, meaning less dorsiflexion flexibility. In the motion capture trials, the ankle angle is defined as the angle of the foot relative to the shank, where standing upright defines the angle as zero degrees. As such, both plantarflexion and dorsiflexion angles are represented as positive values, as those terms provide the directionality of the motion. Prior to analysis, all variables were assessed for statistically significant differences between each day’s baseline measures. Variables of interest included peak GRF, loading rate, kinematics of the ankle, knee, and hip at initial contact, and peak values of frontal and sagittal kinetics and kinematics of the ankle, knee, and hip. Paired t-tests were used to compare baseline values of each dependent variable between the two testing days. To assess the effects of tape and exercise, 2 (condition) by 2 (exercise) repeated measures ANOVAs were run on each dependent variable using SPSS software (IBM SPSS Version 24, SPSS Inc., Chicago, IL), and effect sizes are represented by η_p^2, where 0.01 represents a small effect, 0.06 represents a medium effect, and 0.14 represents a large effect (Cohen, 1988). For any violations of sphericity, Greenhouse-Geiser corrections were used.

4.3: Results

The preliminary baseline analyses revealed that none of the variables assessed were different between visits to the laboratory, suggesting that any changes observed were a result of the experimental protocol, rather than the extraneous effects stemming from which day participants visited the lab. Ankle tape significantly restricted weight-bearing dorsiflexion in the dominant ankle from baseline measurement to immediately after taping showing that the ankle tape restricted ankle motion as expected, $F(1,11) = 6.36$, $p < 0.01$, $\eta_p^2 = 0.37$ (Table 1). After exercise, weight-bearing dorsiflexion increased in the dominant ankle while taped, coming closer to the pre-tape baseline measurement. There was a significant interaction between tape and exercise in manual goniometer assessment of dorsiflexion flexibility of the dominant ankle, $F(1,11) = 6.89$, $p = .02$, $\eta_p^2 = 0.39$. (Table 1), where dorsiflexion flexibility remained constant in the control condition both pre and post-exercise, while in the taped condition, exercise increased dorsiflexion flexibility from pre-exercise to after exercise (Table 1). However, there were no overall independent effects of tape, nor exercise alone, on dominant ankle dorsiflexion flexibility.

There were no differences in perceived exertion between the day where ankle tape was applied and the control day (ankle tape = 7.8 ± 2.7; no ankle tape = 7.7 ± 2.1).

No interaction was noted between tape and exercise for total peak GRF, nor was the effect of tape or exercise alone. Total loading rate showed no significant interaction between exercise and tape, nor was the effect of exercise alone significant; however, the effect of tape alone increased loading rate, $F(1,11) = 6.27$, $p = .03$, $\eta_p^2 = 0.36$ (Table 2). This was explained by a significant decrease in the time to peak force, $F(1,11) = 14.70$, $p < 0.01$, $\eta_p^2 = 0.57$ (Table 2). In the dominant limb, peak GRF did not reveal significant results in the interaction between exercise and tape, nor was the effect of tape or exercise alone significant. There was no significant interaction between exercise and tape in the dominant limb loading rate, nor was the effect of exercise alone. Dominant limb loading rate increased in the ankle tape condition, $F(1,11) = 5.54$, $p = .04$, $\eta_p^2 = 0.33$ (Table 3). This effect was driven by a decreased loading time under the tape condition, $F(1,11) = 10.68$, $p < 0.01$, $\eta_p^2 = 0.49$ (Table 3).

There was no significant interaction of time and tape in the non-dominant peak GRF, nor was the effect of tape, or exercise alone. In the non-dominant limb, there was no significant interaction between tape and exercise, nor were the effects of tape, or exercise alone significant in altering loading rate.

After exercise, ankle plantarflexion increased in the taped ankle, although exercise had no effect on ankle plantarflexion of the un-taped ankle. These findings are supported by a significant interaction between tape and exercise on dominant ankle plantarflexion angle at impact, $F(1,11) = 8.61$, $p = .01$, $\eta_p^2 = 0.44$ (Table 3). Before exercise, tape resulted in reduced dominant ankle plantarflexion at ground contact (Table 3). In contrast with plantarflexion of the dominant ankle at ground contact, peak dorsiflexion was not significantly influenced by tape, exercise, or an interaction of tape and exercise. For the non-dominant ankle, peak dorsiflexion decreased with exercise, $F(1,11) = 17.67$, $p < 0.01$, $\eta_p^2 = 0.62$ (Table 4). Peak dorsiflexion was not significantly affected by tape, or a combination of tape and exercise. There was no significant interaction between tape and exercise on the non-dominant ankle plantarflexion angle at contact, nor were the effects of tape, or exercise alone significant.

The peak flexion moment of the dominant ankle was not significantly impacted by taping, exercise, or an interaction of taping and exercise. Similarly, no significant effects were observed for the peak frontal moment of the dominant ankle. There was no significant interaction between tape and exercise, nor was there any significant effect of tape application or exercise alone on dominant ankle peak frontal moment. Turning to the non-dominant ankle, the peak dorsiflexion moment was significantly impacted by an interaction of tape and exercise, $F(1,11) = 5.72$, $p = .04$, $\eta_p^2 = 0.34$, but not from tape or exercise alone (Table 4). While the peak dorsiflexion moment remained constant before and after exercise in the no tape condition, the ankle taping procedure resulted in increased peak dorsiflexion moment that decreased after exercise. Similar to the dominant ankle, there were no significant effects on peak frontal plane moment of the non-dominant ankle due to main effects of tape, exercise, or an interaction of the two.

There was no interaction between ankle tape and exercise, nor was the effect of exercise significant on dominant knee peak flexion moment. Dominant knee peak flexion moment was decreased under the ankle tape condition, $F(1,11) = 13.98$, $p < 0.01$, $\eta_p^2 = 0.56$ (Table 3). There was no interaction between tape and exercise in the non-dominant knee peak flexion moment, nor were the independent effects of tape or exercise.

There was no interaction of tape and timing on dominant knee valgus moment, nor were the individual effect of taping, or exercise. Dominant knee peak valgus showed no interaction between tape and exercise, nor was the effect of tape, or exercise alone. The non-dominant knee valgus moment was not significantly changed in the exercise and tape interaction, taping or exercise alone. Additionally, in the non-dominant knee, there was no significant interaction between exercise and tape on peak valgus, nor were individual effects of taping, or exercise.

Dominant hip flexion moment showed no significant interaction between exercise and tape, nor was the effect of tape, or exercise alone. There was no significant interaction between tape and exercise in the dominant hip peak frontal plane moment, nor were independent effects of tape or exercise significant.

Non-dominant hip flexion moment revealed no significant interaction between exercise and tape, nor a significant effect of tape alone; however, exercise alone increased non-dominant hip flexion moment, $F(1,11) = 7.01$, $p = .02$, $\eta_p^2 = 0.39$ (Table 4). Abduction/adduction moment in the non-dominant hip was not impacted by the interaction between tape and exercise, nor were independent effects of tape, or exercise alone. All significant effects reported above were large effect sizes.

	Control	Tape
Pre-Exercise	63.2° ± 11.2	67.9° ± 8.5
Post-Exercise	64.6° ± 10.1	64.2° ± 8.8

Table 1. Manual goniometer measurements.

	Control		Tape	
	Pre-Exercise	Post-Exercise	Pre-Exercise	Post-Exercise
Combined GRF (BW)	4.85 ± 0.75	4.92 ± 0.65	4.89 ± 0.61	4.89 ± 0.58
Combined Loading Rate (BW/s) †	98.2 ± 24.9	100.7 ± 23.3	108.3 ± 24.7	104.1 ± 23.4
Combined Loading Time (ms) †	51.1 ± 6.7	50.5 ± 6.8	47.0 ± 7.5	48.9 ± 6.2

Table 2. GRF and COM measures for combined right and left legs. Means and standard deviations are provided, where † represents a significant tape effect. No effect of exercise was significant.

Dominant Limb				
	Control		Tape	
	Pre-Exercise	Post-Exercise	Pre-Exercise	Post-Exercise
Manual Goniometer Flexibility (Degrees) ‡	63.2 ± 11.2	64.6 ± 10.1	67.9 ± 8.5	64.2 ± 8.8
Peak GRF (BW)	2.62 ± 0.36	2.58 ± 0.34	2.61 ± 0.27	2.62 ± 0.30
Loading Rate (BW/s) †	53.9 ± 13.0	53.9 ± 12.4	60.4 ± 13.2	57.5 ± 11.1
Loading Time (ms) †	50.1 ± 6.2	49.6 ± 6.6	45.1 ± 7.9	47.1 ± 5.8
Plantarflexion at Contact (Degrees) ‡ †	26.2 ± 6.9	24.9 ± 5.6	19.9 ± 8.2	23.0 ± 5.4
Peak Dorsiflexion (Degrees)	28.4 ± 6.2	29.3 ± 9.8	28.6 ± 6.9	26.6 ± 6.3
Peak Dorsiflexion Moment (Nm/kg)	2.1 ± 0.3	2.0 ± 0.3	2.0 ± 0.3	2.0 ± 0.2
Peak Ankle Inversion Moment (Nm/kg)	-0.1 ± 0.2	-0.1 ± 0.2	-0.1 ± 0.2	-0.1 ± 0.1
Knee Flexion at Impact (Degrees)	23.2 ± 7.2	22.8 ± 6.4	23.9 ± 7.3	22.8 ± 5.3
Peak Knee Flexion (Degrees)	96.6 ± 12.3	95.6 ± 12.8	96.3 ± 9.6	93.8 ± 10.8
Peak Knee Flexion Moment (Nm/kg) †	2.7 ± 0.3	2.8 ± 0.4	2.5 ± 0.2	2.5 ± 0.3
Peak Knee Valgus (Degrees)	11.7 ± 4.4	12.6 ± 4.5	11.1 ± 5.5	11.5 ± 6.5
Peak Knee Valgus Moment (Nm/kg)	0.6 ± 0.2	0.7 ± 0.5	0.6 ± 0.2	0.6 ± 0.3
Hip Flexion at Impact (Degrees) ‡	31.5 ± 6.1	29.3 ± 8.1	29.7 ± 6.1	31.7 ± 5.7
Peak Hip Flexion (Degrees)	50.8 ± 9.9	52.2 ± 9.7	52.2 ± 7.9	52.9 ± 9.3
Peak Hip Flexion Moment (Nm/kg)	2.1 ± 0.6	2.2 ± 1.0	2.4 ± 0.8	2.1 ± 0.6
Peak Hip Abduction Moment (Nm/kg)	0.2 ± 0.4	0.6 ± 1.2	0.3 ± 0.5	0.3 ± 0.5

Table 3. Dominant limb dependent variables. Means and standard deviations are provided, where † denotes a significant tape effect, ᵋ denotes an exercise effect, and ‡ denotes a significant interaction.

Non-Dominant Limb				
	Control		Tape	
	Pre-Exercise	Post-Exercise	Pre-Exercise	Post-Exercise
Peak GRF (BW)	2.33 ± 0.41	2.41 ± 0.36	2.40 ± 0.39	2.37 ± 0.32
Loading Rate (BW/s)	49.4 ± 16.2	52.0 ± 14.7	51.3 ± 14.6	52.6 ± 13.9
Loading Time (ms)	50.6 ± 10.5	48.7 ± 9.0	49.1 ± 8.1	47.1 ± 7.4
Plantarflexion at Contact (Degrees)	23.4 ± 4.5	22.6 ± 3.3	21.9 ± 5.0	21.0 ± 5.7
Peak Dorsiflexion (Degrees) $^{\varepsilon}$	27.5 ± 7.7	26.6 ± 6.9	28.3 ± 8.0	26.4 ± 7.4
Peak Dorsiflexion Moment (Nm/kg) ‡	1.8 ± 0.3	1.8 ± 0.2	1.9 ± 0.3	1.8 ± 0.2
Peak Ankle Inversion Moment (Nm/kg)	0.2 ± 0.1	0.2 ± 0.1	0.2 ± 0.1	0.2 ± 0.1
Knee Flexion at Impact (Degrees)	27.1 ± 6.2	26.8 ± 7.3	26.4 ± 5.5	25.7 ± 5.2
Peak Knee Flexion (Degrees)	97.5 ± 11.1	97.5 ± 11.3	97.3 ± 10.1	94.8 ± 10.2
Peak Knee Flexion Moment (Nm/kg)	2.8 ± 0.8	2.7 ± 0.4	2.6 ± 0.5	2.5 ± 0.4
Peak Knee Valgus (Degrees)	12.4 ± 8.1	11.7 ± 8.4	12. ± 8.8	11.8 ± 8.5
Peak Knee Valgus Moment (Nm/kg)	0.6 ± 0.2	0.6 ± 0.2	0.5 ± 0.2	0.5 ± 0.2
Hip Flexion at Impact (Degrees)	34.6 ± 5.4	34.2 ± 10.3	33.9 ± 5.9	35.8 ± 4.1
Peak Hip Flexion (Degrees)	48.9 ± 11.5	51.1 ± 10.4	49.3 ± 10.0	49.6 ± 9.1
Peak Hip Flexion Moment (Nm/kg) $^{\varepsilon}$	2.3 ± 2.2	2.5 ± 2.0	1.6 ± 0.6	1.9 ± 0.6
Peak Hip Abduction Moment (Nm/kg)	-0.3 ± 2.9	-0.1 ± 2.4	0.4 ± 0.3	0.5 ± 0.3

Table 4. Non-dominant limb dependent variables. Means and standard deviations are provided, where † denotes a significant tape effect, $^{\varepsilon}$ denotes an exercise effect, and ‡ denotes a significant interaction.

4.4: Discussion

The primary aims of this investigation were to understand the influence of unilateral ankle tape on bilateral kinetics and kinematics of all three joints in the leg in both the frontal and sagittal planes while performing bilateral drop landings before and after exercise. The ankle tape procedure effectively constrained weight-bearing dorsiflexion and exercise mitigated the constrained ROM as anticipated. Despite ankle taping reducing ROM passively, only plantarflexion angle at contact was reduced, with no change in peak dorsiflexion, which is an important aspect of ankle taping, as excessive plantarflexion at ground contact is a typical feature of a lateral ankle sprain, along with excessive inversion. This finding has previously been reported, although Simpson and others noted that peak dorsiflexion was also reduced while wearing an ankle brace, and Cordova and others reported reduced dorsiflexion after taping (Cordova et al., 2000; Simpson et al., 2013). It is possible that the high force attenuation demands when landing from 60 cm exceed the restrictive properties of ankle taping in the sagittal plane, thus negating the reduction seen in the weight-bearing lunge flexibility. It is useful to note that 60 cm is significantly higher than many exercisers and even elite athletes and particularly college-aged females, who made up a significant portion of the current investigation, will be able to raise their COM during a jump, which is around 43-50 cm (Lockie et al., 2016; Sands et al., 2019; Smith et al., 2007; Wisloff, Castagna, Helgerud, Jones, & Hoff, 2004). However, previous investigations have also utilized heights of 60 cm and higher (Mokhtarzadeh et al., 2017; Niu, Wang, Jiang, & Zhang, 2018; I.-L. Wang et al., 2015). In addition, understanding the effects of increased height remain important, as other activities can involve landings from similar heights or higher, such as gymnastics or extreme sports such as skateboarding, among others.

Exercise had no significant effects on GRF, loading rate, or the kinetics and kinematics in the dominant leg. This was somewhat surprising as the observed decrease in dorsiflexion ROM immediately after ankle tape did not appear to play a role in altering any kinetic or kinematic variables because of the restricted motion at the ankle; however, this could have been mediated by the ROM restriction presented here, or the demands of landing from the height of the landing. When considering the participants' self-reported RPE values indicating extremely light to very light exertion, the lack of significant differences between the pre- and post-exercise would indicate that the participants were indeed not influenced by fatigue. Exercise was an important component in altering the non-dominant limb, as peak ankle dorsiflexion moment was increased while the dominant ankle was taped, followed by a mitigating effect of exercise which brought the dorsiflexion moment equal to the control condition. Even though the participants in this investigation were allowed to rest and complete the last set of landings when they did not feel fatigued, a 15-minute bout of exercise was enough to change biomechanical parameters. This could potentially increase the risk for injury, specifically as a decrease in dorsiflexion while landing has the potential to increase the loading rate when the ankle does not go through the same ROM. It appears that after exercise, a lesser reliance on increased motion at the non-dominant ankle to slow the COM transmits up the leg resulting in an increase in hip flexion moment after exercise, potentially exhibiting an increased reliance on hip musculature to resist the forces of landing. Ankle taping did not play a role in altering perceived levels of exertion after completing the exercise protocol. While this was not a specific aim of this study, it was important to note that ankle tape did not complicate the effect of exercise on perceptions of fatigue.

Ankle tape increased the loading rate in the taped limb as well as the combined loading rate, but did not increase any of the peak ground reaction forces. The changes in loading rate have been previously reported as an effect of ankle taping, as the time to peak force was reduced (Niu et al., 2016). Changes in peak ground reaction force after ankle support has shown mixed results, and the results of this investigation would support no change in peak GRF (Abián-Vicén et al., 2008; Cordova et al., 2010; DiStefano et al., 2008; Niu et al., 2016). Even when considering only one ankle was taped, the peak GRF for the taped limb was not different after taping, further supporting the evidence for no change to peak GRF after ankle tape. This result demonstrates that while the peak force is not increased, time is reduced, resulting in the potential for increased injury risk as the body is loaded more rapidly with less time to slow the COM.

Ankle taping did not change the dorsiflexion or inversion moment in taped ankle. Following the application of ankle support, there have been mixed findings regarding changes in joint moment. Dewar and colleagues noted an increase in the inversion moment after ankle bracing, while other investigations found no differences in inversion and dorsiflexion moments (Dewar et al., 2019; Maeda et al., 2019; Vanwanseele et al., 2014; S. A. Williams et al., 2018). Further, Sato and colleagues observed that for drop landing on an inclined platform, ankle taping reduced the inversion moment by a mechanism of a reduction of the mediolateral moment arm which was posited to be a result of the mechanical restriction of the ankle taping procedure (Sato, Nunome, Hopper, & Ikegami, 2019). In the present investigation, only a drop landing was utilized, and could have been the reason behind no significant difference in the inversion moment since the landing movement, with an assumedly healthy landing technique, occurs almost entirely in the vertical direction. Whereas, perhaps in a task which also takes place in the mediolateral direction such as cutting, taping could influence the inversion moment.

In the dominant knee, the decrease in moment while ankle tape is applied could potentially act as a compensatory adaptation, although there was no change in peak flexion angle. While the reduction in moment could mitigate the impact, it could potentially suggest a less stable joint, which might increase the overall risk for injury to that knee. Interestingly, the reduction in knee moment after ankle support has not been previously reported, with studies reporting no change during drop landings (Maeda et al., 2019; Vanwanseele et al., 2014; S. A. Williams et al., 2018). However, Mason-Mackay and colleagues suggest that while performing a drop jump with ankle bracing, no change in stiffness combined with reduced knee flexion would result in a knee flexion moment decrease, although they did not measure this directly (Mason-Mackay, Whatman, Reid, et al., 2016). The only change noted in the non-dominant limb because of taping the dominant ankle was a reduction in ankle dorsiflexion moment. Although Dewar and colleagues did not address sagittal plane variables, they found no differences in the inversion moment of the un-braced ankle (Dewar et al., 2019). One of the more interesting findings was that only ankle dorsiflexion moment was significantly different in the non-dominant limb because taping the dominant ankle. This could be an important landing adaptation since there was little compensation in the non-dominant limb from taping the dominant ankle. This increase in the ankle flexion moment suggests that the non-dominant ankle might play more of a role in attenuating the forces from a bilateral landing when the dominant ankle has the constraint of ankle tape.

A potential limitation was that only the dominant ankle was measured with the manual goniometer meaning that any change in dorsiflexion flexibility in the non-dominant ankle after exercise was not able to be addressed. It would be of interest for future investigations to understand the influence of varied heights on peak dorsiflexion or total ROM in a taped ankle to determine if at lower heights, additional ROM is not needed, and the taping procedure does not obstruct typical performance. Additionally, it would be interesting to assess the differences that varied heights might have on landing mechanics after an exercise protocol, even without a factor of fatigue.

4.5: Conclusion

Taping the dominant ankle has significant effects on sagittal plane mechanics during bilateral drop landings. The taped ankle has reduced plantarflexion at contact and the loading rate is increased. This presents an interesting relationship, as the reduction in plantarflexion at contact is a beneficial aspect of ankle taping, whereas increased loading rate places that limb under higher stress. However, with a reduced peak knee flexion moment and no significant change in the ankle and hip moments, there is no clear indication if any joint in the dominant limb might be at increased risk for injury. But interestingly, taping the dominant ankle significantly increases peak dorsiflexion moment of the non-dominant ankle, which could be a compensatory mechanism to take on an additional role in dissipating the force of landing, but over time may place too much additional stress on that ankle. Exercise mitigates the effect of the tape on the non-dominant ankle but increases the peak hip flexion moment of the non-dominant leg, whether the ankle is taped or not. These findings highlight complex compensatory effects that taping an ankle or exercise can have on the mechanics of a landing.

CHAPTER 5

STUDY TWO – STIFFNESS OF THE LOWER BODY AFTER UNILATERAL TAPE APPLICATION WHILE PERFORMING DROP LANDINGS FROM VARIED HEIGHTS

5.1: Introduction

Ankle sprains are one of the most commonly sustained injuries across all age groups and skill levels in exercise and sport participation, and remain as the leading source of lower body injury (Doherty et al., 2014; Fernandez, Yard, & Comstock, 2007; D. T.-P. Fong et al., 2007; Roos et al., 2017). High-impact landings are a common factor when sustaining a non-contact ankle sprain, as it puts great strain on the soft tissues in the foot-ankle complex. Ankle taping is a common preventative and rehabilitative measure to provide rigidity and external mechanical support to the ligaments within the ankle, ultimately constraining the ROM in the frontal and sagittal planes (Cordova, Dorrough, Kious, Ingersoll, & Merrick, 2007; Cordova et al., 2000; Purcell et al., 2009; Shapiro, Kabo, Mitchell, Loren, & Tsenter, 1994). In the role of providing support to the ankle, taping is effective in providing mechanical support to reduce inversion by reducing inversion ROM by up to 50 %, and reducing dorsiflexion ROM by 20 % (Alt et al., 1999; Best et al., 2014; Eils, Imberge, Völker, & Rosenbaum, 2007).

Landing compresses the musculoskeletal system rapidly which creates high loads at the joints that must be mediated to prevent injury. Proper landing technique consists of making initial ground contact with the ball of the foot while resisting the load as the heel travels downward to make ground contact (P. J. McNair et al., 2000; McNitt-Gray, 1991). Utilization of a larger overall joint ROM at the ankle, knee, and hip can decelerate the COM in a controlled manner over a longer period, which can reduce stiffness (P. J. McNair et al., 2000). Females have been suggested to utilize a greater ROM at the ankle and hip during single and double-leg landings when compared to males (Decker et al., 2003; Kernozek et al., 2008; Kernozek et al., 2005; Scott G McLean et al., 2007; Murphy et al., 2003; Schmitz et al., 2007). These findings suggest that with the restriction of dorsiflexion ROM and increased external mechanical support from ankle taping there would be changes to landing mechanics particularly in females.

The human body has often been modeled as a spring, as it can resist external load while it compresses (R. J. Butler et al., 2003; Serpell, Ball, et al., 2012). In assessing compressive stiffness, Hooke's law states that a force (F) required to compress an object is related to its spring constant (k) and its resultant deformation (x), (Equation 1) (R. J. Butler et al., 2003). To solve for vertical stiffness, the equation can be rearranged to (Equation 2) (Serpell, Ball, et al., 2012).

$$F = kx \tag{1}$$

$$K_{vert} = \frac{F_{\max}}{\Delta y} \tag{2}$$

Stiffness is maintained by a combination of anatomic features, including musculature, ligaments, and tendons (Latash & Zatsiorsky, 1993). Stiffness is a necessary quality to protect the body during impacts; however, the level of stiffness needed is task dependent and either too much or too little stiffness can be a precursor to injury (R. J. Butler et al., 2003). Additionally, some degree of stiffness is necessary for musculature to actively regulate joint position to resist external loading (R. J. Butler et al., 2003). While vertical stiffness can be viewed as simple linear compression taking the quotient of GRF and COM displacement, joint stiffness is noted as quasi-stiffness, and is modeled as a torsional spring where the stiffness is operating rotationally, utilizing the quotient of joint moment with angular displacement (Farley et al., 1998; Serpell, Ball, et al., 2012). Vertical stiffness describes how the lower body is compressing while dissipating the forces of landing, while joint stiffness can give some insight to how the joints contribute to vertical stiffness. Joint stiffness refers to the resistance to angular displacement, calculated as the peak joint moment (M_{max}) over the change in joint position ($\Delta\theta$), and is maintained both passively by ligaments and actively by musculature at the joint (Equation 3).

$$K_{Joint} = \frac{M_{\max}}{\Delta \Theta} \tag{3}$$

As ROM is a critical component of joint stiffness calculations, external movement restriction such as that provided by ankle tape is likely to play a major role in influencing joint stiffness.

Muscular activation has been shown to play a critical role in regulating joint stiffness to support the knee during landing (Kim & Hong, 2011; McNair, Wood, & Marshall, 1992). Landing strategies more reliant on ligamentous restraint rather than appropriate muscular activation place the individual at a higher risk for ligament damage as the strain can exceed the mechanical resistance capabilities of the tissue. Improper muscular control is often noted as a key feature of damage to the ACL, and similarly, appropriate muscular control of the ankle when landing can reduce the risk for ankle sprains as injury often occurs in the initial weight acceptance prior to reactive control (Kernozek & Ragan, 2008; Konradsen et al., 1997; Santello, 2005). During hopping, Granata and colleagues noted that females utilized less leg stiffness than males despite completing the same task, suggesting the importance of leg stiffness in providing biomechanical stability to reduce risk of knee injury (Granata, Padua, et al., 2002). Quadricep and hamstring musculature play an integral role in modulating lower body stiffness, as both muscle groups contain bi-articular muscles responsible for both knee and hip control. Though it is unclear how much or how little stiffness is most beneficial for a task, females have been shown to have decreased joint stiffness compared to males (Quatman, Ford, Myer, Paterno, & Hewett, 2008). As such, it is important to better understand how joint stiffness contributes to vertical stiffness, as there remains some speculation on how joint stiffness influences vertical stiffness, especially under differing task constraints (Farley et al., 1998; Farley & Morgenroth, 1999; Hobara et al., 2010; Hobara et al., 2009; Serpell, Ball, et al., 2012).

Williams and Riemann (S Williams & BL Riemann, 2009) found no difference in vertical leg stiffness after taping during a hopping task, suggesting the existence of potential compensatory measures; however, without examining joint stiffness, the origin of this mechanism remains unclear. Furthermore, stiffness was assessed after an exercise protocol, where the loosening of ankle tape was a likely contributing factor to the results. Utilizing ankle bracing during single leg hopping at different frequencies, Hobara and colleagues (Hobara et al., 2017) also found no change in vertical stiffness after wearing an ankle brace, suggesting that other factors must be altered to maintain a consistent vertical stiffness. While increasing frequency in hopping can provide a more challenging task, increasing the height of landing could also represent a more challenging task to dissipate the higher forces of landing. Evidence on stiffness while landing from various heights is limited; however, the results of those investigations highlighted changes in leg and joint stiffnesses with changing heights (I.-L. Wang et al., 2015; L.-I. Wang, 2009).

While the external restriction of movement caused by ankle taping offers support to the ankle, it is unclear exactly how the other joints of the ipsilateral and contralateral limbs modulate stiffness to mitigate high impact landing forces. These previous studies investigated vertical stiffness, and not the stiffness of contributing joints which could provide insight into the control of vertical stiffness. There has been one study which investigated the effect of taping or bracing on joint stiffness while performing a drop landing task, where braces were applied to both ankles which resulted in increased ankle and leg stiffness with a negligible effect at the knee (Mason-Mackay, Whatman, Reid, et al., 2016). The effect of taping one ankle on the stiffness of the contralateral leg has not been investigated.

Multiple stiffness calculations were reviewed by Serpell and colleagues (Serpell, Ball, et al., 2012) highlighting the difference in methodologies when calculating vertical, leg, and joint stiffnesses. General recommendations for stiffness calculations were suggested based on the volume of utilization in the literature, and those methodologies were utilized here. Vertical stiffness will be defined as the ratio of change in force to the change in COM position (Farley, Blickhan, Saito, & Taylor, 1991; Farley & Gonzalez, 1996; Hobara et al., 2008; Morin, Dalleau, Kyröläinen, Jeannin, & Belli, 2005; Serpell, Ball, et al., 2012). Joint stiffness is calculated as the ratio of joint moment to joint angular change (Farley & Morgenroth, 1999).

The aim of this investigation was to determine the effect of unilateral ankle taping on vertical and joint stiffnesses during a bilateral landing task. Stiffness has often been investigated during gait and hopping tasks, but landing stiffness has not received much attention. Further, the effect of varied heights on vertical and joint stiffnesses has received little investigation and necessitates further study. As such, it is hypothesized that with increasing height, vertical stiffness will decrease and the knee will be a primary driver of the changes in vertical stiffness. Through the utilization of a landing protocol, the scope of stiffness evaluated here lies with the compressive resistance alone, rather than the effect of compression and energy return. Due to the increased external mechanical support with ankle taping, it is hypothesized that joint stiffness at the ankle will increase. Furthermore, it is hypothesized that the unilateral application of ankle tape will cause changes in stiffness in other joints in both legs serving to compensate for the additional task constraint.

5.2: Methodology

After attaining local institutional review board approval, twelve female participants age (24 ± 3.6 years), height (1.64 ± 6.29 m), body mass (58.68 ± 4.42 kg) were recruited to participate in this investigation. Participants were excluded if they had previous ACL injury or other injury to the lower limbs within the last six months, or lasting injury that affects daily function. Additionally, participants were free of any neuromuscular disorder that would affect task performance. Participants were outfitted with 25-mm retro-reflective anatomic markers placed at the iliac crest, greater trochanter, medial and lateral femoral condyles, tibial tubercle, medial and lateral malleolus, and calcaneus. While markers located at the anterior and posterior superior iliac spine tracked the pelvis, and markers placed on the head of the first and fifth metatarsal, base of the fifth metatarsal, and distal hallux served as tracking markers for the foot. Additionally, marker clusters were placed on the lateral aspects of the thighs and shanks. Foot, shank, thigh, and pelvis segments were created in a rigid linked segment model using de Leva's anthropometric definitions (de Leva, 1996a). Data was collected using an eight-camera array (Vicon, Oxford, UK) sampled at 120 Hz in synchronization with two six-degrees-of-freedom force plates (AMTI, Watertown, MA, USA) sampled at 960 Hz.

Participants visited the laboratory on two separate days with a minimum of twenty-four hours rest, completing the landing protocol under in both the control or experimental condition in a repeated measures design. Testing order was counterbalanced, and baseline measurements were taken both days after arrival in the lab, and before the application of tape to assess for differences in baseline measures between sessions. In the experimental condition, ankle tape was applied in the same method by a certified athletic trainer to the dominant ankle, defined as the leg the participant would use to kick a ball the farthest. All participants were provided standardized athletic shoes in their preferred size to wear during the landing trials. Participants were tasked with stepping off a platform at 30, 45, and 60 cm in height in a randomized order, leading with their dominant leg, landing bilaterally with one foot per force plate. Furthermore, participants were asked to keep their arms crossed, with hands on their shoulders, and upon landing to stand upright and maintain a neutral posture which ensured balance was maintained. Fifteen landings were performed from each height on each laboratory visit for a total of thirty landings; however, only the first ten were utilized for this analysis, as the last five were performed after an exercise protocol and excluded to minimize the confounding influence on joint stiffness.

After completion of a power-spectral-density analysis, kinematic data was low pass filtered with a cutoff frequency at 8 Hz using a fourth order zero-lag Butterworth filter. Ground reaction force data were low pass filtered at a frequency of 12 Hz using a fourth order zero-lag Butterworth filter. As body weight influences stiffness values, ground reaction force data was normalized to participant body weight prior to stiffness calculations (Farley et al., 1993).

Vertical stiffness was calculated as the ratio of the peak vertical ground reaction force to the total COM displacement (Equation 2) (Farley & Gonzalez, 1996; Farley & Morgenroth, 1999; Hobara et al., 2010). COM was defined as the three-dimensional vertical coordinate center of the pelvis from the markers placed on the anterior and posterior iliac spines.

$$K_{Vert} = \frac{F\max}{\Delta y} \quad (2)$$

Leg stiffness was calculated in the same manner as Equation 2, by using the ratio of the peak ground reaction force of the individual limb to the vertical displacement of the greater trochanter of that limb.

Individual peak joint stiffness was calculated as the ratio of peak joint moment to joint ROM from contact to maximal flexion in the sagittal plane (Farley & Morgenroth, 1999; Hobara et al., 2013; Hobara et al., 2010) (Equation 3).

$$K_{joint} = \frac{M_{\max}}{\Delta \Theta} \quad (3)$$

All dependent variables were assessed with a 2 (condition) x 3 (height) repeated measures ANOVA utilizing IBM SPSS Version 23 (IBM, Chicago, IL, USA) with significance set to 0.05. Similarly, the effect of day on baseline variables was assessed with a 2 (day) x 3 height) repeated measures ANOVA. Effect sizes were calculated with η_p^2, where a value of 0.01 represents a small effect, 0.06 is a medium effect, and 0.14 is a large effect, as well as Sidak post-hoc tests to further explore results (Cohen, 1988).

5.3: Results

Initial comparisons were completed to test for statistically significant differences between visits to the lab, to determine a possibility of an order effect. No significant differences between days were noted, and as such, experimental condition order was not considered as a dependent variable.

There were no significant effects of condition, height, or interactions on peak total vertical stiffness (Table 5). There was no significant interaction between condition and height, nor was the effect of tape significant in altering dominant limb stiffness. However, height decreased dominant leg stiffness $F(2,22) = 6.14$, $p < 0.01$, $\eta_p^2 = 0.36$ (Table 6). Pairwise comparisons revealed a significant difference between only the 30 cm and 60 cm heights, $p < 0.01$. Height significantly increased non-dominant leg stiffness, $F(2,22) = 22.09$, $p < 0.01$, $\eta_p^2 = 0.67$ (Table 7). Pairwise comparisons revealed leg stiffness was significantly lower at 30 cm when compared to 60 cm, $p < 0.01$, and that 45 cm was less than 60 cm, $p < 0.01$.

The dominant ankle stiffness showed no significant interaction between height and condition; however, taping increased stiffness, F(1,11) = 14.06, $p < 0.01$, $\eta_p^2 = 0.56$ (Table 6). Drop height also increased dominant ankle stiffness, $F(2,22) = 30.31$, $p < 0.01$, $\eta_p^2 = 0.73$ (Table 6). Post-hoc analyses showed relationships between all heights $p < 0.01$. The increase in stiffness at the ankle while taped was caused by a significant reduction in ROM, $F(1,11) = 57.16$, $p < 0.01$, $\eta_p^2 = 0.84$ (Table 6), with no significant change in joint moment. Dominant ankle peak moment did not exhibit a significant interaction of height and tape, nor was tape alone significant. Height significantly increased peak ankle moment $F(1.32,14.56) = 53.30$, $p < 0.01$, $\eta_p^2 = 0.83$, with Sidak post-hoc tests revealing significant relationships between all heights, $p < 0.01$.

The non-dominant ankle stiffness showed no significant interaction between height and condition, nor any significant effect of ankle tape. Height significantly increased stiffness in the non-dominant ankle, $F(2,22) = 12.15$, $p < 0.01$, $\eta_p^2 = 0.52$ (Table 7). Pairwise comparisons revealed significant differences between 30 cm and 60 cm, $p < 0.01$, as well as from 45 cm and 60 cm, $p = 0.02$. Height also increased non-dominant ankle moment, $F(1.30,14.31) = 119.95$ $p < 0.01$, $\eta_p^2 = 0.92$ (Table 7). Sidak post-hoc tests revealing significant relationships between all heights, $p < 0.01$.

In the non-dominant knee, there was no significant interaction between height and condition, nor was the application of tape or height. Height significantly increased the peak non-dominant knee moment, $F(2,22) = 47.05$, $p < 0.01$, $\eta_p^2 = 0.81$ (Table 7). Sidak post-hoc revealed significant differences between all heights, $p < 0.01$.

There was no significant interaction of height and tape, nor was the effect of height significant on dominant hip stiffness. Ankle tape had a significant effect on increasing dominant hip stiffness, $F(1,11) = 7.30$, $p = 0.02$, $\eta_p^2 = 0.40$ (Table 6). The hip also experienced both a significant increase from height in joint moment, $F(1,11) = 37.48$, $p < 0.01$, $\eta_p^2 = 0.77$ (Table 6), and ROM (44.07 ± 10.57 to 46.20 ± 9.98), $F(1,11) = 6.78$, $p = 0.03$, $\eta_p^2 = 0.38$ (Table 6).

There was no significant interaction between height and condition for the non-dominant hip stiffness, nor was tape significant. Height significantly decreased non-dominant hip stiffness $F(2,22) = 4.32$, $p = 0.03$, $\eta_p^2 = 0.28$ (Table 7). There were no significant relationships revealed by a Sidak post-hoc test. Taping the dominant ankle had no significant effect on any stiffness measure at the ankle, knee, or hip of the non-dominant side. All significant effects reported above were large effect sizes.

	Control			Tape		
	30 cm	45 cm	60 cm	30 cm	45 cm	60 cm
Stiffness (kN/kg/m)	0.12 ± 0.04	0.13 ± 0.04	0.13 ± 0.03	0.13 ± 0.03	0.13 ± 0.04	0.13 ± 0.03
Peak GRF (BW) Ƕ	3.35 ± 0.57	4.07 ± 0.77	4.85 ± 0.75	3.35 ± 0.47	4.06 ± 0.64	4.89 ± 0.61
COM displacement (m) ‡ Ƕ	0.29 ± 0.06	0.34 ± 0.08	0.38 ± 0.08	0.27 ± 0.05	0.33 ± 0.06	0.38 ± 0.06

Table 5. Lower-body limb dependent variables. Means and standard deviations are provided, where † represents a significant tape effect, Ƕ represents a significant effect of height, and ‡ represents a significant interaction.

	Dominant Limb					
	Control			Tape		
	30 cm	45 cm	60 cm	30 cm	45 cm	60 cm
Leg Stiffness (kN/kg/m) Ƕ	0.074 ± 0.02	0.074 ± 0.03	0.070 ± 0.02	0.080 ± 0.02	0.074 ± 0.02	0.070 ± 0.01
Ankle Stiffness (kN/kg/deg) † Ƕ	0.028 ± 0.004	0.033 ± 0.008	0.036 ± 0.007	0.032 ± 0.005	0.037 ± 0.007	0.041 ± 0.008
Knee Stiffness (kN/kg/deg)	0.039 ± 0.009	0.042 ± 0.016	0.040 ± 0.011	0.038 ± 0.008	0.039 ± 0.008	0.038 ± 0.009
Hip Stiffness (kN/kg/deg) †	0.048 ± 0.03	0.043 ± 0.02	0.037 ± 0.01	0.049 ± 0.02	0.054 ± 0.03	0.046 ± 0.02
Peak GRF (BW)	2.03 ± 0.30	2.33 ± 0.36	2.62 ± 0.36	2.09 ± 0.25	2.36 ± 0.29	2.61 ± 0.27
Leg Displacement (m) ‡ Ƕ	0.29 ± 0.06	0.33 ± 0.07	0.37 ± 0.07	0.26 ± 0.04	0.32 ± 0.06	0.36 ± 0.06
Peak Ankle Moment (Nm/kg) Ƕ	1.46 ± 0.18	1.84 ± 0.37	2.03 ± 0.37	1.42 ± 0.21	1.71 ± 0.24	1.98 ± 0.31
Ankle ROM (degrees) Ƕ †	52.9 ± 4.1	55.9 ± 3.3	56.8 ± 2.9	44.4 ± 4.5	46.7 ± 5.7	49.5 ± 6.5
Peak Knee Moment (Nm/kg)	2.2 ± 0.6	2.5 ± 1.0	2.4 ± 0.4	2.1 ± 0.2	2.4 ± 0.5	2.4 ± 0.3
Knee ROM (degrees) Ƕ †	80.6 ± 11.1	87.1 ± 12.0	92.2 ± 9.9	81.0 ± 10.4	90.4 ± 9.1	96.3 ± 9.6
Peak Hip Moment (Nm/kg) Ƕ	1.9 ± 1.3	2.2 ± 1.3	2.3 ± 0.5	1.8 ± 0.4	2.5 ± 1.3	2.6 ± 0.5
Hip ROM (degrees) Ƕ †	37.1 ± 10.6	44.5 ± 12.4	50.1 ± 9.9	38.8 ± 11.2	47.2 ± 10.1	52.2 ± 7.9

Table 6. Dominant limb dependent variables. Means and standard deviations are provided, where † represents a significant tape effect, Ƕ represents a significant effect of height, and ‡ represents a significant interaction.

	Non-Dominant Limb					
	Control			Tape		
	30 cm	45 cm	60 cm	30 cm	45 cm	60 cm
Leg Stiffness (kN/kg/m) Ƕ	.058 ± .021	.059 ± .017	.066 ± .017	.058 ± .017	.060 ± .018	.065 ± .016
Ankle Stiffness (kN/kg/deg) Ƕ	0.029 ± 0.007	0.032 ± 0.01	0.035 ± 0.008	0.032 ± 0.009	0.032 ± 0.008	0.035 ± 0.009
Knee Stiffness (kN/kg/deg)	0.046 ± 0.034	0.048 ± 0.05	0.036 ± 0.023	0.046 ± 0.030	0.038 ± 0.026	0.037 ± 0.025
Hip Stiffness (kN/kg/deg) Ƕ	0.050 ± 0.029	0.037 ± 0.014	0.036 ± 0.014	0.051 ± 0.024	0.044 ± 0.020	0.034 ± 0.011
Peak GRF (BW)	1.46 ± 0.34	1.90 ± 0.41	2.33 ± 0.41	1.53 ± 0.27	1.91 ± 0.35	2.40 ± 0.39
Leg Displacement (m) ‡ Ƕ	0.29 ± 0.06	0.34 ± 0.08	0.38 ± 0.08	0.27 ± 0.05	0.33 ± 0.06	0.38 ± 0.06
Peak Ankle Moment (Nm/kg) Ƕ	1.09 ± 0.18	1.43 ± 0.18	1.76 ± 0.28	1.06 ± 0.16	1.37 ± 0.21	1.78 ± 0.24
Ankle ROM (degrees) Ƕ	37.3 ± 8.1	45.3 ± 7.8	49.5 ± 6.4	33.7 ± 8.0	42.2 ± 8.1	49.9 ± 6.9
Peak Knee Moment (Nm/kg) ‡ Ƕ	1.8 ± 0.5	2.0 ± 0.5	2.3 ± 0.6	1.8 ± 0.5	2.0 ± 0.6	2.5 ± 0.7
Knee ROM (degrees) Ƕ †	82.0 ± 9.8	88.1 ± 12.1	93.8 ± 10.2	82.4 ± 10.4	90.8 ± 9.7	97.3 ± 10.1
Peak Hip Moment (Nm/kg) Ƕ	1.3 ± 0.4	1.5 ± 0.5	2.0 ± 0.6	1.4 ± 0.4	1.7 ± 0.5	2.0 ± 0.6
Hip ROM (degrees) Ƕ	29.8 ± 12.1	40.5 ± 12.6	47.6 ± 12.3	30.1 ± 11.8	40.8 ± 11.7	49.3 ± 10.0

Table 7. Non-dominant limb dependent variables. Means and standard deviations are provided, where † represents a significant tape effect, Ƕ represents a significant effect of height, and ‡ represents a significant interaction.

5.4: Discussion

A key aim of this investigation was to understand the influence of unilateral ankle tape applied to the dominant limb and how it impacted the dominant and non-dominant limbs independently regarding the resultant vertical, and joint stiffnesses. To assess the individual contribution of each limb, a bilateral landing protocol was designed so that each limb was measured independently using two force plates. Taping an ankle resulted in increased joint stiffness in both the ankle and hip of the ipsilateral leg for women performing a drop landing. Previous research has indicated that females utilize greater ankle dorsiflexion to mitigate the forces of landing when compared to males (Decker et al., 2003; Kernozek et al., 2005). Because ankle taping reduces dorsiflexion ROM, and ROM is a key determinant of stiffness calculations, it is logical that stiffness would be increased at the ankle, especially when females prioritize the utilization of ankle ROM to attenuate landing forces. In professional Australian Rules football players, those who went on to sustain a hamstring injury noted greater leg stiffness values than those players who did not sustain a hamstring injury (Watsford et al., 2010). As noted by Mason-Mackay, after applying ankle braces, there were increases in ankle and leg stiffnesses with a negligible effect on the knee, while the hip was not reported (Mason-Mackay, Whatman, Reid, et al., 2016). As such, an increase in joint or leg stiffnesses because of ankle taping could potentially place the individual at higher risk for bony damage.

The results of this investigation suggest that taping the ankle appears to only alter joint stiffnesses of the leg that was taped. Specifically, taping the dominant ankle significantly increased ankle and hip stiffness of that leg but did not impact stiffness of the knee. Interestingly, despite increased ankle and hip stiffnesses, the overall vertical leg stiffness did not change which could have been mitigated by small, non-significant changes in the knee. This raises an interesting point, as Mason-Mackay noted a small increase in leg stiffness, they reported an unclear or negligible stiffness increase at the knee after ankle bracing (Mason-Mackay, Whatman, Reid, et al., 2016). While the dominant hip stiffness was increased because of tape, there was no change in peak moment, but an increase in ROM. Based on how joint stiffness is calculated, these findings of ROM and moment seem counterintuitive; however, the small effect size in ROM combined with the fact that both the 45 and 60 cm platforms produced higher moments while taped, but not the 30 cm platform, could explain the overall increase in hip stiffness as a result of ankle tape. Following the application of external ankle support, no change to leg and vertical stiffness has been reported during both single-leg and two-legged hopping tasks after applying external ankle support, suggesting evidence for joint compensation strategies (Chang et al., 2008; Hobara et al., 2017; SA Williams & BL Riemann, 2009). Overall lower-body stiffness in the current investigation was shown to be invariant to changes from tape, further providing evidence of some form of compensation strategies.

Regarding the influence of unilateral ankle support, in the only previous study to utilize a bilateral landing after unilateral ankle support, Dewar reported a decrease in inversion movement and an increase in inversion moment, but only in the braced ankle (Dewar et al., 2019). However, only frontal plane kinetics and kinematics were reported. The current study confirms that taping one ankle did not significantly affect stiffness of the joints in the other non-taped leg. Evidence that this was likely due to the taping and not due to leg dominance comes from a study which found that bracing both ankles significantly increases stiffness of both ankles (Mason-Mackay, Whatman, Reid, et al., 2016). However, the previous study also reported increased leg stiffness in both legs after bracing which was not observed in the current study (Mason-Mackay, Whatman, Reid, et al., 2016). The previously reported difference in leg stiffness after bracing showed a small effect size, and when compared to the current study, could potentially be attributed to differences in the choice of ankle support. While the current investigation noted an increase in hip stiffness, the effect of ankle support on the hip has not yet been reported; thus, it is difficult to draw a comparison to previous literature (Mason-Mackay, Whatman, Reid, et al., 2016). The fact that taping the dominant ankle in this investigation was localized to changes in stiffness of the dominant limb is interesting, as it suggests ankle taping only influences stiffness in the joints of the limb which received ankle tape and is a novel addition to the literature.

Drop height influences the stiffness of both legs, but surprisingly, in opposite directions. Dominant leg stiffness decreased with increasing height, while the non-dominant leg stiffness increased with increasing height. Across all heights, the non-dominant limb had lower peak GRF than the dominant limb which could enable the non-dominant limb to take on additional loading to distribute the load more equally between limbs. The opposing changes in stiffness in each leg effectively balanced out total stiffness which was uninfluenced by drop height. Furthermore, as the platform height increased, the leg stiffness values became closer to one another, suggesting a more even force distribution strategy with increased height. This is a new finding, as previous research assessing stiffness utilizing different platform heights did not consider the effects of limbs independently (I.-L. Wang et al., 2015; L.-I. Wang, 2009). Altering the height of the drop significantly altered the ankle and hip in the non-dominant limb, where there was an increase in the non-dominant ankle stiffness, and a decrease in the non-dominant hip stiffness. An overall increase in leg stiffness with drop height suggests that the ankle plays a larger role in modulating stiffness with height. Overall, leg and joint stiffnesses varied significantly with drop height but showed opposing effects between the two legs. Interestingly, the present findings of total and leg stiffnesses somewhat conflict with data reported by Wang and colleagues, who noted that specifically in females, lower-body stiffness decreased when performing a drop landing with increasing heights of 40 to 60 cm (I.-L. Wang et al., 2015).

While it does not appear that this investigation revealed an intralimb compensation mechanism regarding ankle taping, there does appear to be an interlimb compensatory behavior in terms of varied landing heights, where the dominant limb shows a decreasing stiffness with height, while the non-dominant limb shows an increasing stiffness with height which maintained a consistent lower-body stiffness. Interestingly, lower body and joint stiffnesses have been reported to increase with increasing performance and task demands (Brazier et al., 2014). However, the exact influence of the driving force between joint, leg, and lower-body stiffnesses has been debated (Brazier et al., 2014; Serpell, Ball, et al., 2012).

During tasks such as hopping or running, increased lower-body stiffness from a reduction in COM displacement serves to minimize ground contact time enabling the performance of more rapid movements (Hobara et al., 2010; Hobara et al., 2007; Hobara et al., 2015). Whereas, in single-leg drop jumps from heights above 30 cm, stiffness has been shown to reduce, which serves an important role to reduce the strain on the body, although it resulted in the loss of jump height (L.-I. Wang & Peng, 2014). Additionally, contrary to rhythmic or cyclic movements, drop landings are task oriented to reduce loading forces without a secondary performance task, potentially limiting the generalizability of results of previous investigations which utilize rhythmic or cyclic tasks. Further evidence for a task-centered difference is provided by findings where bilateral drop jumps from increasing heights resulted in attaining similar jump height while maintaining similar leg stiffness values, whereas performing bilateral drop landings of increasing heights, leg stiffness decreased (I.-L. Wang et al., 2015; L.-I. Wang & Peng, 2014). When taken together, it becomes apparent that joint, leg, and lower-body stiffness are modifiable based upon task parameters. Furthermore, Mason-Mackay and colleagues assessed stiffness while performing both drop landings and drop jumps (Mason-Mackay, Whatman, Reid, et al., 2016). While the researchers did not compare the results between the two tasks, left leg stiffness was higher in the drop jump compared to the drop landing, and left knee stiffness was lower in the drop jump compared to the drop landing, suggesting task-dependent stiffness modulation occurred (Mason-Mackay, Whatman, Reid, et al., 2016).

A potential area of future research stemmed from an interesting finding that peak GRF and peak joint moments almost always occurred prior to peak COM displacement of joint flexion. This phenomenon has been expressed in previous research; however, it typically occurred in hopping (Hobara et al., 2010). Previous researchers have typically allowed for some difference between maximal GRF and minimal COM position by calculating stiffness using the COM position at the time of peak GRF, despite the two values not occurring at the same time point (Hobara et al., 2010; Hobara et al., 2015). Horita and colleagues have previously addressed that pre-landing stiffness could potentially influence the initial contact stiffness measures (Horita, Komi, Nicol, & Kyröläinen, 1996; Horita, Komi, Nicol, & Kyröläinen, 2002). Differences in timing relationships could also be a task-dependent factor, as this investigation used only the compressive phase of landing with no secondary task such as a jump. It is possible that the demand of performing a secondary task immediately after ground contact, specifically with the goal to minimize ground contact time could bring the peak GRF and minimal COM position closer together.

It is worth addressing a potential limiting factor for vertical leg displacement calculations, which was that projected landmarks denoting left and right leg femoral head centers were based off the pelvis COM. To use the greater trochanter for leg stiffness assessment, a virtual marker was projected to the hip joint center. While the greater trochanter marker was utilized as a calibration only marker and not for active tracking, it would have served as the landmark to track the change in leg length, therefore, it was posited that this methodology would provide a more consistent representation of leg length specifically in the vertical direction. This effect led to nearly identical vertical displacements between both legs, although it is not unrealistic that while performing drop landings the right and left legs would be within two centimeters of each other in vertical displacement.

5.5: Conclusion

Ultimately, it appears that applying ankle tape unilaterally only influences the taped side, through increases in ankle and hip stiffnesses, although leg stiffness remains constant. This would support previous research which has noted that females utilize an ankle and hip focused landing strategy (Decker et al., 2003; Kernozek et al., 2005). There were small, but non-significant changes at the knee which mitigated a significant change to the leg stiffness. As the dominant ankle and hip increased stiffness after ankle tape, but not the knee, this could potentially indicate a strategy where the stiffness in the knee is maintained to reduce the potential for injury. As there are not well-defined stiffness values while performing various tasks, it is unclear if an increase in stiffness at the hip would influence injury risk. Particularly as the musculature supporting the hip is larger, and increasing stiffness may not influence the injury risk at the hip as much as changes to stiffness in the ankle due to the smaller musculature supporting the ankle. This finding expands the body of stiffness literature, as the effects of unilateral ankle support on stiffness during bilateral drop landings have not been previously reported. Lower-body stiffness in the current investigation was shown to be invariant to changes from either tape or height of landing platform, further providing evidence of some form of compensation strategies. The height from which landing is initiated also plays a unique role in altering both the dominant and non-dominant limbs with opposing effects exhibiting a decrease in stiffness in the dominant limb, and an increase in stiffness in the non-dominant limb, effectively neutralizing any changes in combined lower-body vertical stiffness. Furthermore, this investigation provides novel findings, as previous stiffness research utilizing drop landings of varied heights has not reported the influence of individual limbs. This investigation is the first to assess lower-body, leg, and joint stiffnesses while performing a drop landing in one comprehensive analysis.

CHAPTER 6

STUDY THREE – KINEMATIC AND KINETIC JOINT COORDINATION IN THE LOWER BODY WHILE PERFORMING A DROP LANDING

6.1: Introduction

To land safely from a drop, humans dissipate forces through the ankles, knees, and hips. This requires coordination between the joints within a leg (intralimb) and between the legs (interlimb) to avoid overloading a single joint. Studies One and Two of this dissertation found that taping an ankle has varied effects on the kinematics and kinetics of joints of the same leg but not the other leg, during bilateral drop landings. Additionally, exercise and drop height had differential effects on dominant and non-dominant legs. Together, these findings suggest that coordination between joints within the same leg and between joints of different legs are influenced by changes to environmental constraints (taped ankle), organismic constraints (warm-up from exercise), and task constraints (different drop heights) (Newell, 1986). The focus of this study is to investigate how taping one ankle joint affects intralimb and interlimb coordination during a double-leg drop landing.

Joint coordination during a landing task has received only modest investigation and the impact of joint supports has not been assessed. In one study, intralimb coordination for a double-leg landing from a 60 cm platform was examined in men using nonlinear regression techniques (Yeow et al., 2011a). The relationship between ankle and knee joint angle was best fit by a natural logarithm, while the relationship between the hip and knee was best fit by an exponential function. Additionally, relative joint angles differed before and after peak ground reaction force, suggesting a change in coordination during the landing. While the relationship between joint angles can be informative, coordination can also be studied by assessing higher-order relationships like the relative phase. DRP has been used to quantify coordination during single-leg landings in a study on sex-based differences (Hughes, 2020). This measure is based on the relative time of occurrence of certain events during a landing, such as the difference in time between peak flexion of two joints relative to the overall landing time from ground contact to zero velocity. Using standard deviation of DRP, women were found to have greater variability in hip-knee and hip-ankle joint couplings than men (Hughes, 2020). Knee-ankle coupling of the dominant limb was also found to be more variable than the non-dominant limb for both men and women (Hughes, 2020). In a similar study, but of bilateral landings, women were observed to land with greater asymmetry between the legs (Hughes & Watkins, 2008). Intralimb motion has shown dependent relationships in landing mechanics, and although the mathematical relationships about the knee were logarithmic and exponential with the ankle and the hip, respectively, their modeled behavior suggests that the joints in the leg are coordinated during the absorption of landing forces (Yeow et al., 2011a). Therefore, greater asymmetry and variability in joint coordination during landings potentially increases the strain on passive structures in the joints and may be a factor in the higher prevalence of lower-limb injuries, such as ACL tears, that are observed in women. Given increased injuries are likely to result in greater use of joint supports, it is especially important to study intralimb and interlimb coordination during landings and the effects of taping or bracing a joint in a female population.

A limitation of the previous research on landing coordination in women is the use of DRP. While DRP can reveal aspects of coordination in different tasks, it only indicates coordination relative to the timing of particular elements of the movement and not continuously throughout the movement. Indeed, as mentioned earlier, intralimb joint coordination changes have been found in landings of men (Yeow et al., 2011a). A richer understanding of how the joints are coupled within and between legs throughout a landing task can be achieved by quantifying coordination at each moment in time by CRP.

Typically, CRP calculations begin with a phase plot for an individual joint motion to represent the relationship between angular velocity and angular position, where angular velocity and angular position have first been normalized by subtracting the mean to center the data around zero and dividing by the velocity and position range, respectively, to give values between -1 and +1 (Equations 4 and 5). In a methodology suggested by Lamb and Stöckl in a review of CRP analyses, the joint angular displacement and angular velocities were normalized independently as to preserve the 0 value of velocity which represents a transitory component in a movement (Lamb & Stockl, 2014). Equation 4 demonstrates the normalization of the angular velocity, where Equation 5 demonstrates the normalization of angular displacement, where y(t) is the normalized time series (Lamb & Stockl, 2014).

$$f\big(y(t_i)\big) = \frac{y(t_i)}{\max\,(y(t))} \quad 4$$

$$g\big(y(t_i)\big) = 2 * \left(\frac{y(t_i) - \min\big(y(t)\big)}{\max\big(y(t)\big) - \min\big(y(t)\big)}\right) - 1 \quad 5$$

While a phase plot for a discrete task such as a landing will not show a full limit cycle, the plot will still reveal if a systematic pattern occurs, or all areas of phase space are visited. The phase angle (α) at each moment in time (t_i) can be computed between a vector from a position by velocity coordinate pair (A) to the origin and the positive position axis, according to Equation 9:

$$\alpha_{kinematic}(t_i) \; = \arctan\left(\frac{velocity_{joint}(t_i)}{angle_{joint}(t_i)}\right) \quad 9$$

This process of computing the phase angle time series is repeated for other joints. CRP of joint kinematics ($\phi_{kinematic}$) can then be computed as the arithmetic difference between the phase angle of one joint and the phase angle of a second joint at the same moment in time, and is repeated across the time series according to Equation 10:

$$\varphi_{kinematic}(t_i) = \alpha_{joint1}(t_i) - \alpha_{joint2}(t_i) \quad 10$$

Plotting CRP for joint pairs during a landing task will indicate how the joints are coordinated together and how that changes during different phases of the landing. By including both landings with and without a taped ankle on a CRP plot, differences in coordination arising from the ankle tape can be identified.

Traditionally, CRP has provided a measure of the kinematic coordination between joint or segment motions. This assumes that position and velocity are the appropriate descriptors of a joint or segment behavior, and the coordination between joints or segments. This makes sense when studying oscillating fingers, handheld pendulums, or other simple rhythmic movements, where the kinematics are assumed to be the defining characteristics of the behavior (Amazeen, Amazeen, & Turvey, 1998). However, in tasks where a collision is significant (e.g., landings), other quantitative descriptors of the coordination may be more appropriate. During a landing, the lower limb joints perform negative work to bring the COM to a controlled stop. This negative work is achieved through the moment and angular displacement of each joint. The ratio of this joint moment to angular displacement is the joint stiffness. In Study Two of this dissertation, the overall joint stiffness across the landing was computed for each joint. Taping was found to increase overall joint stiffness of the ankle and hip of the taped side, but not the contralateral leg; hence, we may expect that coordination of stiffness between joints changes with ankle taping. Phase plots of joint moment against angular position could be informative about how joint stiffness changes in each joint during a landing. The continuous phase angle for joint kinetics for each joint can then be calculated according to Equation 11:

$$\alpha_{kinetic}(t_i) = \arctan\left(\frac{moment_{joint}(t_i)}{angle_{joint}(t_i)}\right) \qquad 11$$

To determine if taping one ankle effects coordination between the dynamics of the joints, CRP of joint dynamics ($\varphi_{kinetic}$) can be computed between pairs of joints as per Equation 12:

$$\varphi_{kinetic}(t_i) = \alpha_{joint1}(t_i) - \alpha_{joint2}(t_i) \qquad 12$$

Plots of CRP between joint pairs can provide information about the phase relationship between the stiffness of two joints throughout the landing, where CRP can indicate if the joints maintain a consistent in-phase or phase lagged relationship (flat slope close or far from 0 degrees, respectively) or a steep slope shows rapid changes in phase in one joint compared with another. Variability of CRP between trials can inform about whether the pattern is consistent or not across landings. Therefore, quantifying CRP of kinematic and kinetic joint coordination could be informative about coordination during a landing with and without ankle taping.

No previous research has examined the effect of ankle support on coordination during a drop landing, however, the effect of ankle support on intralimb coordination has been studied in rhythmic locomotion tasks (Herb et al., 2016; Jagodinsky, 2016; Jagodinsky et al., 2020). Coordination variability in gait after taping has been assessed using a vector coding approach. Tape led to reduced variability in shank-rearfoot coupling, which supports the idea that taping acts as a constraint through reducing ROM and thereby limiting available degrees of freedom (Herb et al., 2016). Using a CRP analysis of gait, foot-shank coordination was shown to be more in-phase during initial foot contact while braced (Jagodinsky et al., 2020). As in the early stance phase, weight is just being accepted by the foot, hence this finding may apply to the early phase of a drop landing. In a task more analogous to a drop landing, ankle bracing was found to reduce coupling variability between the foot-shank segments during single leg hopping (Jagodinsky, 2016). Hence, taping or bracing an ankle appears to result in more in-phase coordination and reduced coordination variability of the constrained segments during running or hopping. In the matched interlimb pairs, more in-phase coordination would be preferential, as it allows for a more symmetrical landing profile across both limbs while performing a bilateral task. How an ankle support impacts coordination with other joints of the same leg and between the legs is unknown.

The aim of the current study was to determine how taping an ankle influences intralimb and interlimb coordination during a double-leg drop landing. Coordination was quantified throughout the landing using CRP of joint kinematics (angular position and velocity) as well as joint kinetics (angular position and moment). As women are more susceptible to joint injuries than men, and sex differences have been observed in landing kinematics and coordination, this study focused on young, physically active women. To mitigate injury risk, the joints within and between the legs were expected to be closely coordinated to reduce overloading one joint. Based on the findings of Yeow and colleagues, the ankle was expected to phase lead the knee and hip early in the landing, and then phase lag during the latter part of the landing (Yeow et al., 2011a). Coordination between matching joints of the legs was expected to be approximately in-phase, in an effort to load the legs symmetrically (Bates et al., 2013). Small deviations from in-phase coordination could arise from limb dominance or the task of stepping off a 60 cm platform.

The predictions for taping an ankle are based on previous findings for intralimb kinematic coordination during rhythmic tasks, as this has not been assessed for landings. During a landing, taping an ankle reduces the ankle's mobility (ROM). This constraint of one ankle is expected to alter its kinematic and kinetic coordination with other joints that are not being constrained. Specifically, taping an ankle is predicted to increase kinematic coordination asymmetry from in-phase for the ankle-knee and ankle-hip joint pairs of the same leg, influencing the role the hip plays while landing. Kinematic coordination asymmetry will also increase between the ankles of the two legs, particularly in the early stage of the landing, as Study One revealed a significant difference in plantarflexion angle at initial contact. Similarly, kinematic joint coordination variability is predicted to increase for intralimb ankle-knee and ankle-hip joint coupling of the taped leg, as well as interlimb coordination of the two ankles, when the ankle is taped. While joint kinetic coordination has not been previously investigated, we expect that taping an ankle will have similar effects on joint kinetic coordination as kinematic coordination. Taping an ankle is predicted to increase asymmetry of kinetic joint coordination for the ankle-knee and ankle-hip couplings of the taped leg, as well as the interlimb coordination between the ankles. These same joint combinations are also expected to increase in kinetic joint coordination when one ankle is taped. These changes in joint kinetic coordination are expected to reveal that taping an ankle alters the dynamics of the legs which could place the joints at a risk for injury as the joints undergo altered kinetic responses. This study will contribute to our understanding of joint coordination during a double-leg landing, as well as how coordination changes when a joint is constrained.

6.2: Methodology

This study utilized the procedure noted in the Study Methodology (Chapter 3), where twelve healthy, recreationally active women without a history of lower body injury which affects daily living activities, were recruited to participate in this investigation. Three-dimensional motion capture recorded kinematic data for ankle, knee, and hip motion of both legs, while two force plates recorded kinetic data from each independent limb. Participants were tasked to perform a drop landing from 60 cm, leading with their dominant limb, but landing with both feet at approximately the same time. The landing procedure was completed on two days in a repeated measures design, where upon arrival to the lab, a baseline landing set was completed (5 landings), followed by the control (no tape)/experimental (tape) condition (5 landings) in a counter-balanced order. In the ankle tape condition, ankle tape was applied only to the dominant ankle by a certified athletic trainer utilizing the Gibney closed basketweave with heel lock and figure-eight, which consisted of two anchors, three stirrups, three horseshoes, two figure eights, two heel locks, and then closed (Perrin & McLeod, 2018). Five trials were performed at each assessment point during the investigation, totaling twenty landings between the two visits.

To compute CRP, first all sagittal plane joint positions, velocities, and moments of the ankle, knee, and hip, were normalized (Equations 4 and 5), then the phase angle of each joint was calculated, for both kinetic and kinematic calculations (Equations 9 and 11). Phase and CRP are established for the drop landings from the period of initial contact until the center-of-mass (COM) velocity was equal to 0 and were normalized to 101 data points. Kinematic and kinetic CRP were calculated as shown in Equations 10 and 12, respectively.

A total of nine coordination pairs, including three intralimb comparisons in each limb (ankle-knee, ankle-hip, and knee-hip), and three interlimb comparisons (ankle-ankle, knee-knee, and hip-hip). Each coordination pair was plotted across the normalized landing phase to reveal their relative coordination throughout the landing cycle. The five baseline trials captured each day immediately upon arriving in the lab served to establish if any differences occurred between visits to the lab. When assessing CRP outputs, values range from -180 to 180, where -180 and 180 both represent a perfect anti-phase relationship, and a value of 0 represents a perfect phasic relationship (Lamb & Stockl, 2014). Between conditions, a positive value represents the distal joint in the joint pair leading the proximal joint, and for intralimb comparisons, a positive value represents the dominant limb joint leading the non-dominant joint (Lamb & Stockl, 2014). This was repeated for each joint pair to consider the nine intra and interlimb coordination combinations. To assess coordination variability, the standard deviation at each time point in the landing phase across the trials for each individual for a given condition was computed.

Both kinematic and kinetic coordination and their associated coordination variability data was statistically analyzed in MATLAB (v. R2020b) utilizing open source code for statistical parametric mapping (SPM) (v M.0.4.7) to determine points in the landing that were significantly different between taped and un-taped conditions (Pataky, 2010, 2022; Penny et al., 2011). SPM is a technique which uses continuous comparisons to look for regional differences in a data set which are above a critical value, based on probability distributions according to random field theory (Adler, 2010; Pataky, 2010). Paired two-tailed t-tests were used in SPM, which computed a critical t value based on an alpha = 0.05 for determining when coordination or coordination variability differed significantly between the taped and un-taped conditions (Pataky, 2010, 2022). This critical SPM(t) value varies based on the smoothness of the data set (Pataky, 2010).

6.3: Results

Baseline data were compared between visits to the lab to account for any differences between session days. The order of the tape or no-tape condition day was counterbalanced between the 12 participants. Of the 18 coordination pairs analyzed with the paired two-tailed t-test SPM, there were 4 at baseline which revealed significant differences between visits to the lab. Kinematic ankle-knee coordination in both legs showed that the ankle trailed the knee significantly more on day one than day two from 60-74% of the landing in the dominant limb and 68-82% of the landing in the non-dominant limb. Kinematic knee-knee coordination also differed significantly between baselines, with the dominant knee trailing the non-dominant knee more during 76-87% of the landing at the first visit. For kinetic coordination, only the dominant leg knee-hip coordination differed significantly between days, and this occurred very briefly at 55% of the landing. None of the coordination pairings found to differ significantly at baseline, differed significantly between the taped and un-taped conditions.

The SPM paired two-tailed t-test were run on the two conditions to assess for differences between the tape and no-tape conditions. In the CRP plots, in interlimb pairs, a positive value indicates that the dominant side joint leads the non-dominant side; and in intralimb pairs, a positive value indicates that the distal joint leads the proximal joint. Regarding kinematic coordination, interlimb hip-hip coordination showed a significant difference between 86-95% (Figure 2) of the landing, with the dominant hip trailing more behind the non-dominant hip for the taped compared with the non-taped condition. Throughout the landing, mean kinematic hip-hip CRP was largely in phase, with coordination variability ranging approximately 20 degrees (Figure 4). Similarly, the knee-knee and ankle-ankle pairs were largely in phase throughout the landing (Figure 2) with a mean coordination variability less than 20 degrees (Figure 4). There were no significant kinematic coordination differences in intralimb coordination for joints of the dominant limb. However, the non-dominant ankle-hip coordination varied significantly between conditions at 84-95% of the landing (Figure 3). The ankle trailed the hip more under the tape condition (Figure 3) and the knee trailed the hip more in the tape condition at 82-96% of the landing (Figure 3). In both the non-dominant ankle-hip and knee-hip kinematic CRP pairs, the coordination variability was also very small at the times when there were significant differences (Figure 5)

The SPM analysis of kinetic coordination revealed only brief significant differences between conditions. The taped condition differed significantly from the non-taped condition for interlimb kinetic hip-hip coordination at 40% of the landing with the non-dominant trailing the dominant hip while the dominant ankle was taped (Figure 6); and for non-dominant ankle-hip coordination at 15% of the landing where the ankle lead the hip when the dominant ankle was taped (Figure 7). However, neither effect was sustained, suggesting that the noted difference may not be representative of a meaningful and persistent change. Mean kinetic interlimb pairs were all largely in phase throughout the landing (Figure 6) and the coordination variability was predominantly under approximately 25 degrees, with a peak at 40 degrees in the hip-hip pair (Figure 8). Interlimb kinetic CRP pairs had a larger range of coordination (Figure 7) with much larger coordination variability up to 80 degrees out of phase (Figure 9). However, the kinetic ankle-knee CRP started largely out of phase, but after approximately 30% of the landing, they trended to near perfectly in phase (Figure 7), and the coordination variability was also small from that point until the end of the landing (Figure 9).

Across all interlimb and intralimb pairs of kinetic and kinematic CRP, no significant differences were found (Figures 4, 5, 8, and 9). All results of SPM paired two-tailed t-tests can be found in the Appendix A7.

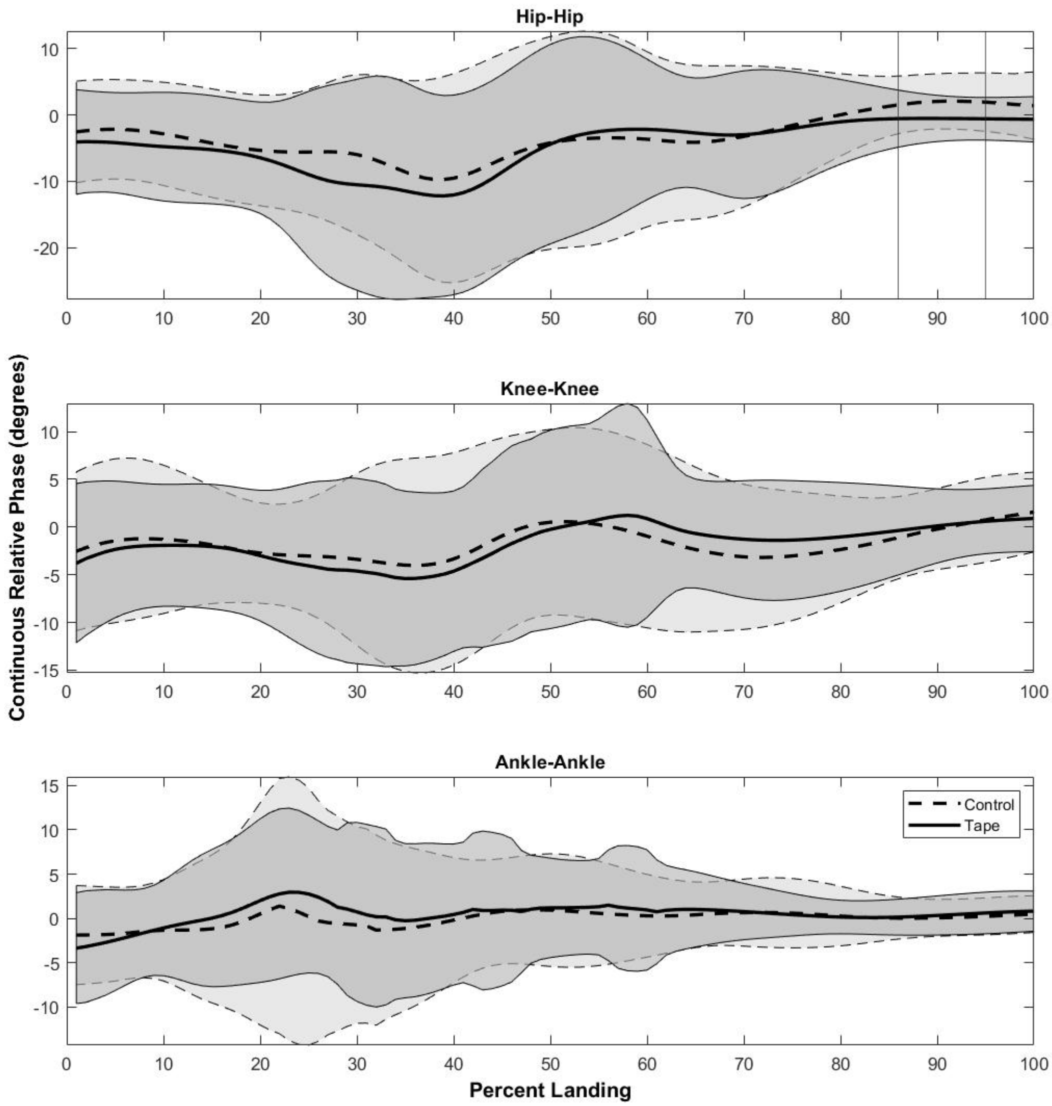

Figure 2. Comparison of the taped and un-taped kinematic CRP of the interlimb joint pairs in degrees (M ± SD) from initial contact to landing completion, where the vertical lines represent the region of the landing where there were significant differences between conditions.

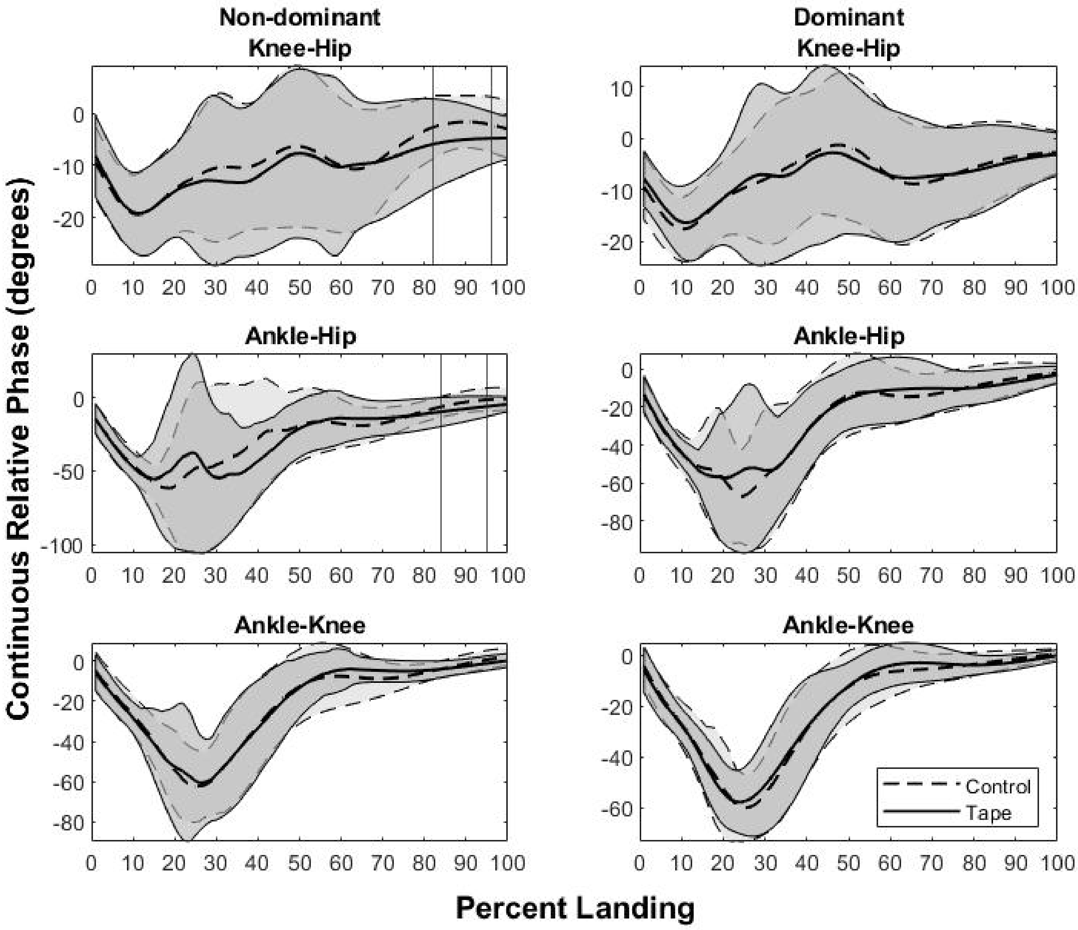

Figure 3. Comparison of the taped and un-taped kinematic CRP of the non-dominant (left) and dominant (right) intralimb joint pairs. Displayed in degrees (M ± SD) from initial contact to landing completion, where the vertical lines represent the region of the landing where there were significant differences between conditions.

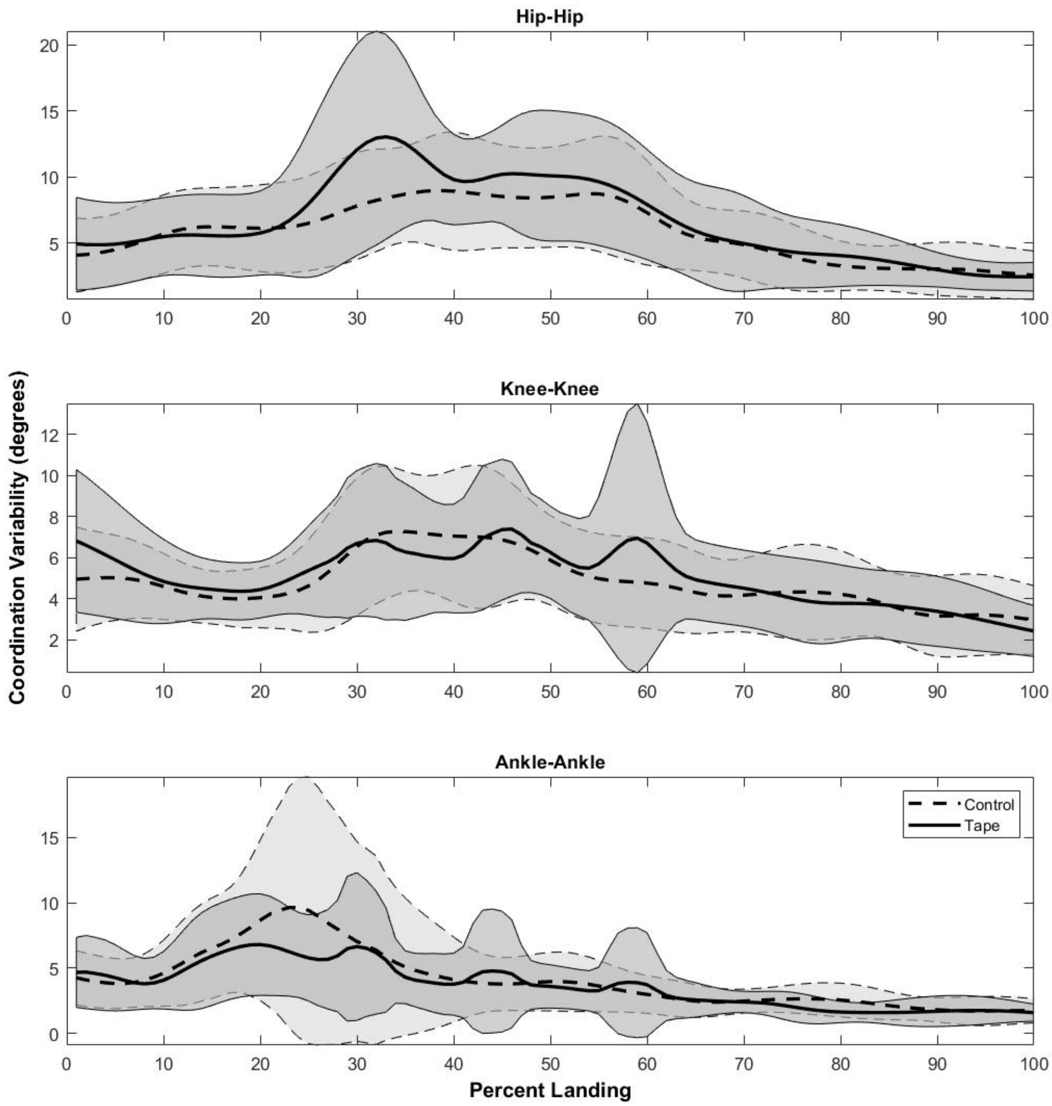

Figure 4. Comparison of the taped and un-taped kinematic interlimb coordination variability between interlimb joint pairs. Displayed in degrees (M ± SD) from initial contact to landing completion.

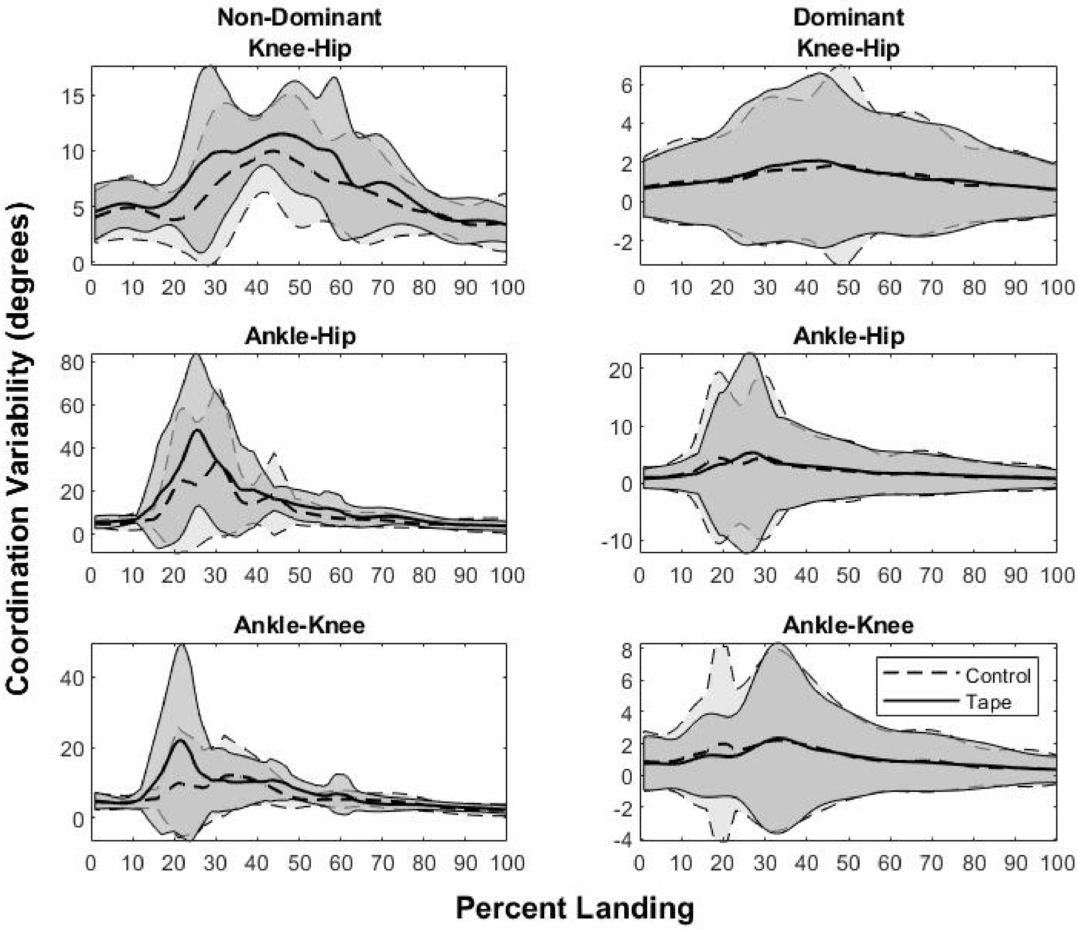

Figure 5. Comparison of the taped and un-taped kinematic intralimb coordination variability of the non-dominant (left) and dominant (right) intralimb joint pairs. Displayed in degrees (M ± SD) from initial contact to landing completion.

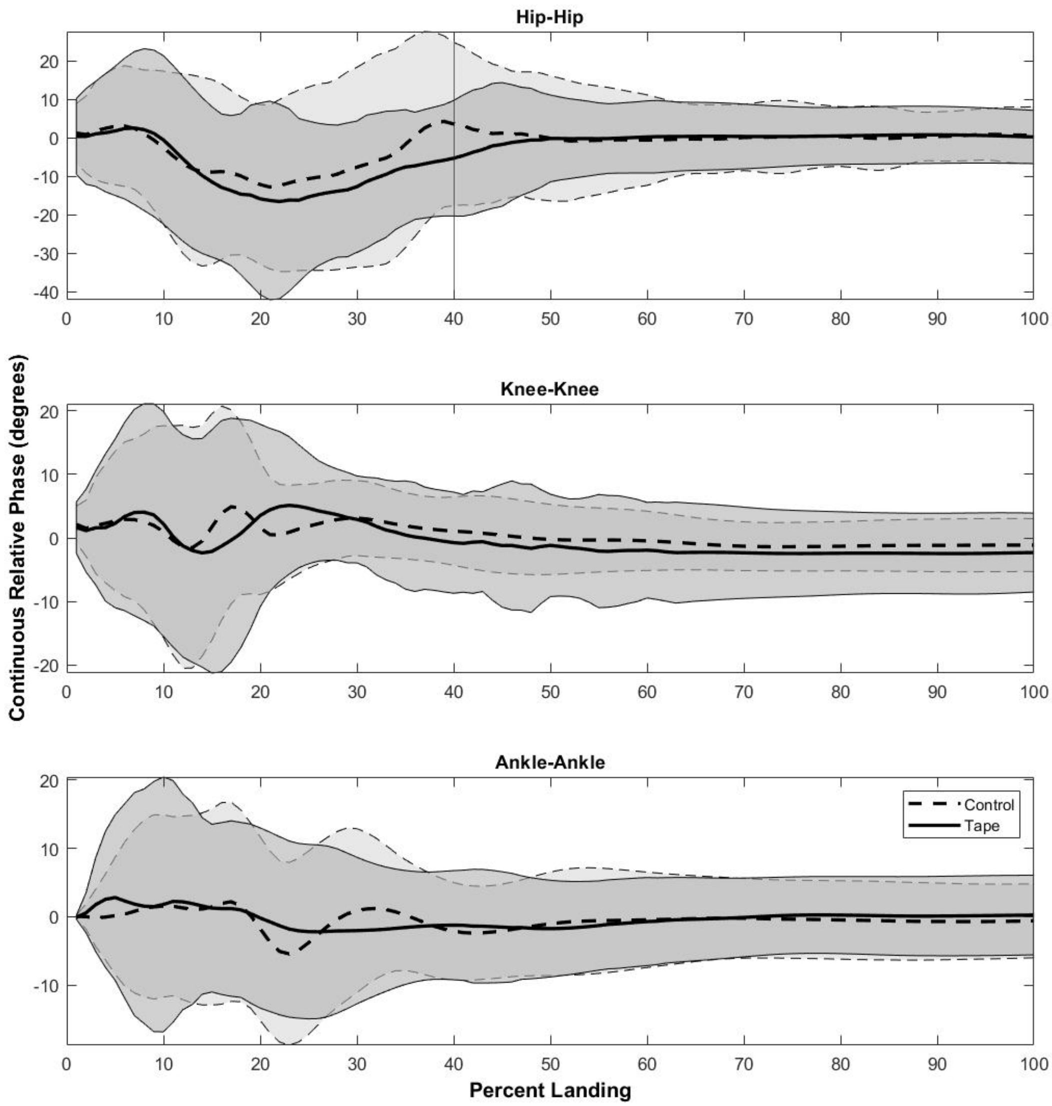

Figure 6. Comparison of the taped and un-taped kinetic CRP of the interlimb joint pairs. Displayed in degrees (M ± SD) from initial contact to landing completion, where the vertical line represents the time point in the landing where there was a significant difference between conditions.

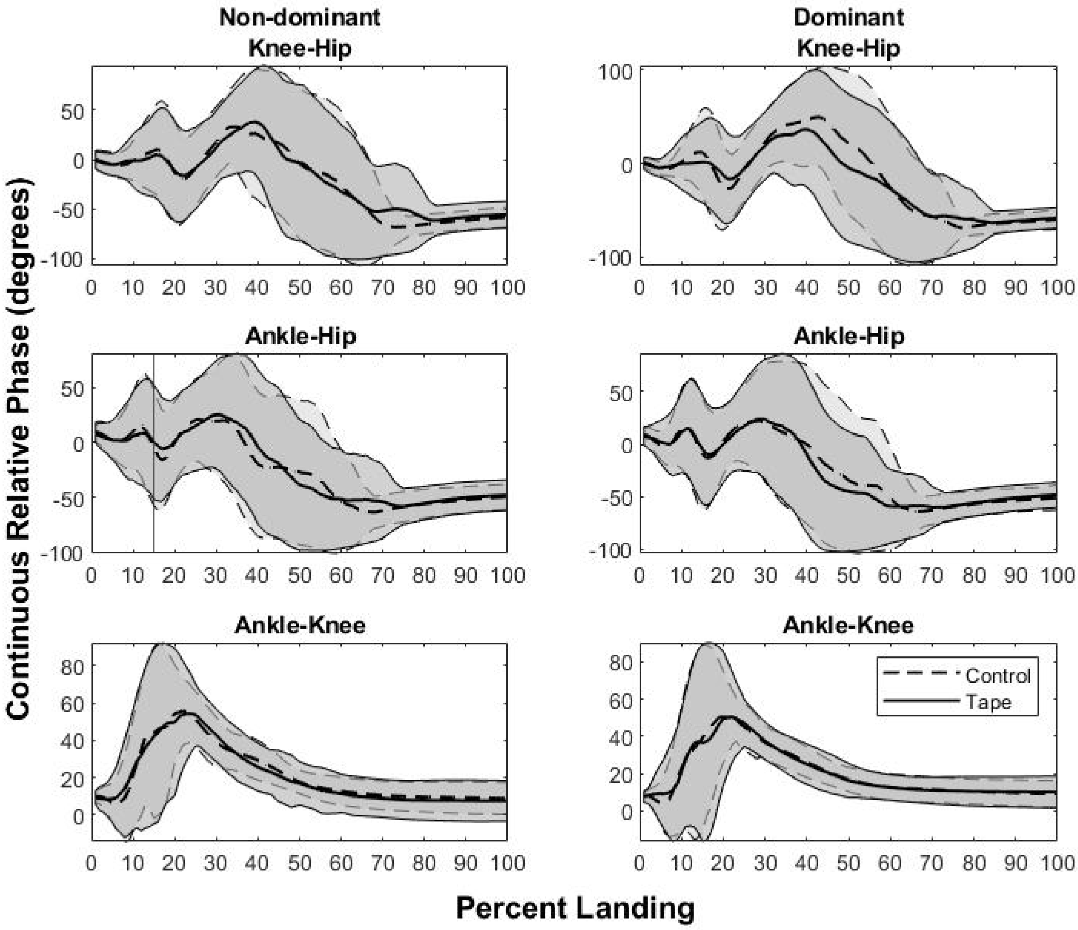

Figure 7. Comparison of the taped and un-taped kinetic CRP of the non-dominant (left) and dominant (right) intralimb joint pairs. Displayed in degrees (M ± SD) from initial contact to landing completion, where the vertical line represents the time point in the landing where there was a significant difference between conditions.

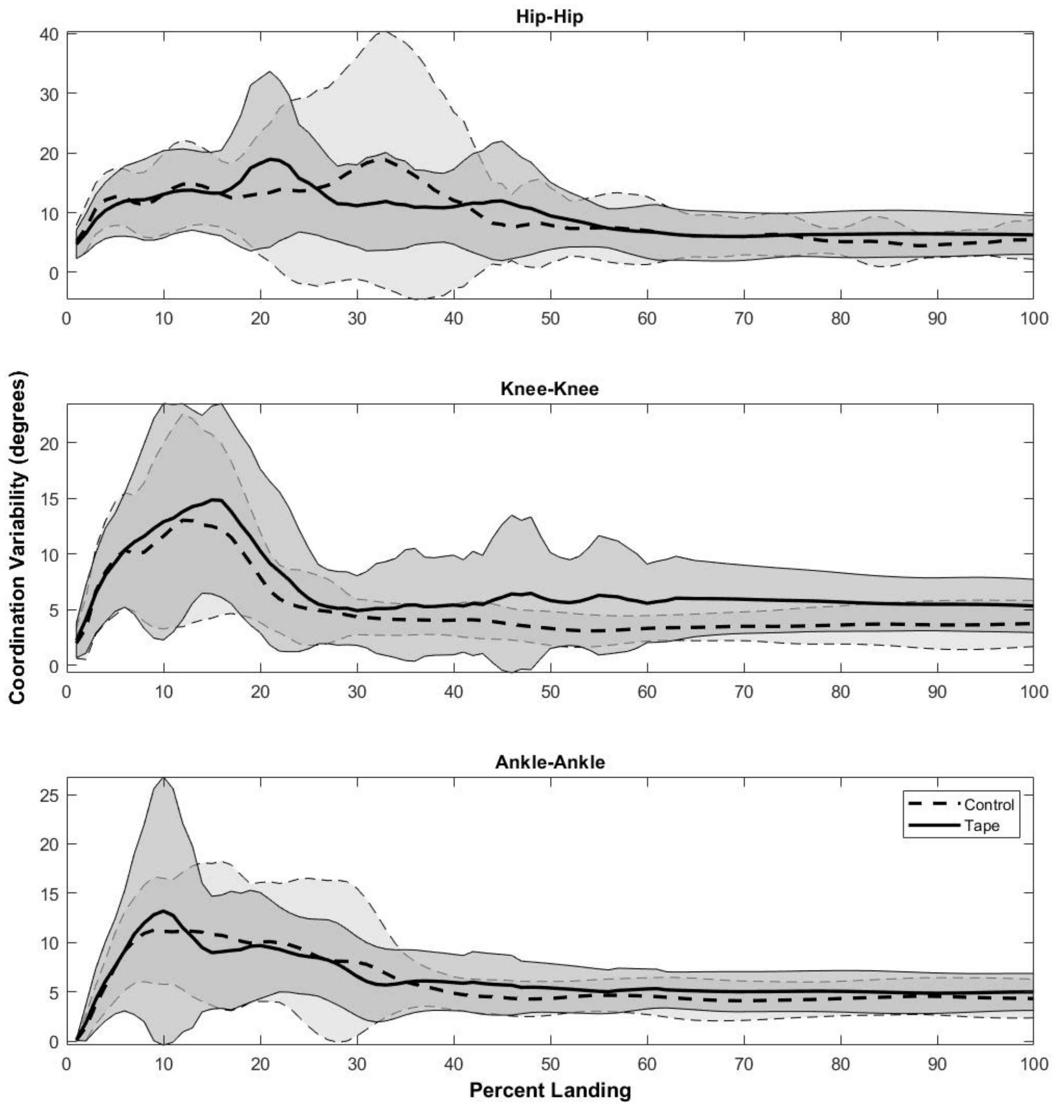

Figure 8. Comparison of the taped and un-taped kinetic interlimb coordination variability between interlimb joint pairs. Displayed in degrees (M ± SD) from initial contact to landing completion.

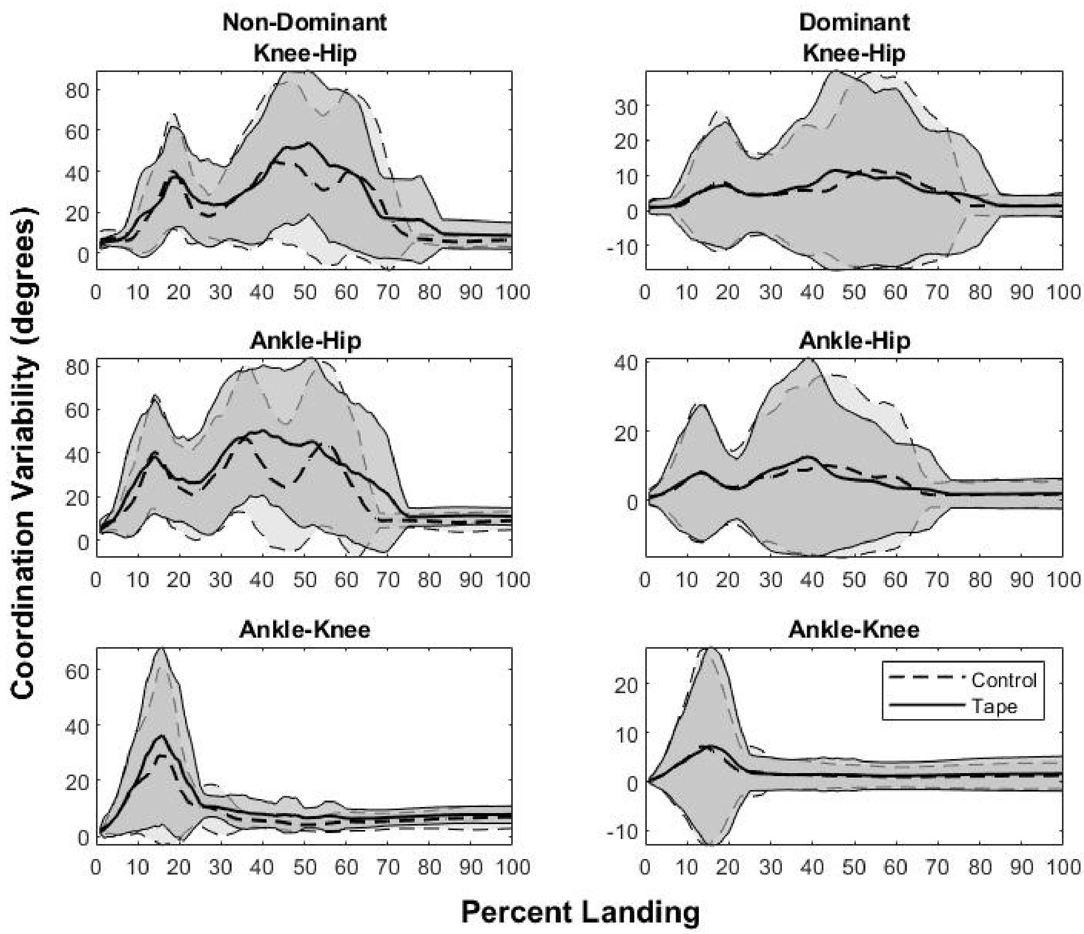

Figure 9. Comparison of the taped and un-taped kinetic intralimb coordination variability of the non-dominant (left) and dominant (right) intralimb joint pairs. Displayed in degrees (M ± SD) from initial contact to landing completion.

6.4: Discussion

The aim of this investigation was to understand the influence of a unilateral ankle taping on the kinematic and kinetic coordination of the ankle, knee, and hip while performing a bilateral drop landing. The primary hypothesis was that the taped ankle would result in changes from in-phase to more out-of-phase kinetic and kinematic coordination of the ankle-ankle, ankle-knee, and ankle-hip pairs. It was also hypothesized that the interlimb kinetic and kinematic coordination variability would increase while one ankle was taped. In the CRP plots, positive values mean the distal joint leads the proximal for the intralimb pairs, and a positive value means the dominant leads the non-dominant side for the interlimb pairs. Differences noted in the baseline trials between visit one and visit two suggest the potential for a learning effect, but the significant results between days were not seen in the same pairings during the control and experimental trials.

6.4.1: Interlimb Kinematic CRP

There were minimal significant effects across kinematic and kinetic CRP pairs, and no significant differences were observed in any kinetic or kinematic coordination variability due to the unilateral application of ankle tape while performing a drop landing. Kinematic interlimb coordination pairs were largely in phase throughout the landing and showed a shrinking variability towards the end of the landing. While performing a bilateral task, the matched joints across the limbs moved more in phase with better coordinative relationships (Kelso et al., 1981). The only significant difference in interlimb kinematic coordination occurred in the hip-hip pairing at the end of the landing when the variability was reduced relative to the rest of the landing. This contrasts with the initial hypothesis that the ankle was likely to be the source of differences in coordination, as the more proximal joints in the leg would be less affected as the forces traveled up the kinetic chain. The taped ankle was found to have differing effects from the un-taped ankle in Studies One and Two, particularly at the early stage of landing, which was expected to play a role in affecting kinematic coordination. The non-dominant hip was different from the dominant hip at the end of the landing where in the no-tape trials, the dominant hip led the non-dominant hip, but in the tape trials, the non-dominant hip led the dominant hip. Differences involving the non-dominant hip will also be discussed in the intralimb pairings. All differences in the kinematic CRP involved the non-dominant hip as one of the joints in the pairings. The other interlimb kinematic CRP pairs, both ankle-ankle and knee-knee were both mostly in phase throughout the landing and showed shrinking variability towards the end of the landing.

6.4.2: Interlimb Kinetic CRP

Despite two significant effects in the kinetic CRP data, those only showed significant differences at a single time point in the landing and may not be representative of a meaningful, and lasting effect. During landings, coordination of the kinetics of interlimb pairs remained approximately in-phase (close to 0 degrees), largely uninfluenced by whether an ankle was taped or not. This pattern could be seen as more ideal since the relationship between force distribution and respective joint position which comprises the kinetic CRP, remains more equal across the limbs. During the landing task, kinetic symmetry would be beneficial to prioritize when performing a bilateral landing to complete the task in the safest manner, despite the unilateral constraint from ankle tape. A more equal distribution of forces during the landing process would allow for each limb to share the load and prevent one limb from being overloaded which could increase the risk of injury. As a demand of the landing task from 60 cm, it is possible that the unilateral ankle tape did not provide enough external resistance to meaningfully change the kinetic CRP profile of the lower limbs, thus maintaining no significant sustained differences in kinetic relative phase. This may serve as a beneficial effect, as the dynamic relationships between the interlimb and intralimb joint pairs were not changed as a result of the ankle tape.

Considering coordination variability of interlimb kinetic CRP pairings, it is evident that there was a large amount of variability across both the tape and no tape conditions, particularly compared to the kinematic CRP. The larger kinetic CRP variability in both conditions could be a result of the high force attenuation demands of the task. Utilization of kinetic values for calculation of CRP is a novel technique, and as far as the author is aware, this is the first study to utilize CRP calculated with joint moment and joint angle during a bilateral drop landing task. It is possible that performing a landing successfully does not necessitate a tightly controlled and less variable force attenuation, while the overall motion of the landing is more consistent. Low variability can potentially increase the risk for overuse injury, and too much variability can represent an uncontrolled movement pattern. In this investigation, the application of ankle tape did not change the variability in any of the kinematic or kinetic interlimb and intralimb pairs, which would suggest that the preferred variability seen in the participants is maintained, despite unilateral ankle taping.

6.4.3: Intralimb Kinematic CRP

Intralimb non-dominant ankle-hip, and the non-dominant knee-hip kinematic CRP pairs were significantly different between the tape and no-tape conditions. Although the dominant and non-dominant limbs were not directly compared, there was much larger between-subject variability in the non-dominant limb. It is possible that the non-dominant limb does not exhibit the same level of control as the dominant limb, though the overall phasic behavior between the limbs appears similar. No differences were found in the dominant intralimb CRP pairs. One of the most interesting findings was that while performing a bilateral drop landing with the dominant ankle taped, it was the non-dominant hip which appeared to be the joint most affected by kinematic CRP changes. The non-dominant hip was a primary factor in the changes observed in kinematic CRP pairings, both with interlimb hip-hip, as well as intralimb ankle-hip and knee-hip pairs where it is possible that additional hip movement strategy could serve to offset potential overloading of the other joints.

Differences noted in pairings with the hip as a predominant factor in the landing have been previously identified during a single-leg landing in females, where hip utilization was more variable in the ankle-hip and knee-hip pairs which was suggested to be an attempt to reduce loading (Hughes, 2020). Utilizing DRP during a single-leg landing task revealed coordination variability differences in hip-knee and hip-ankle pairs, where this effect was also seen in the current study using kinematic CRP. This highlights that specifically the non-dominant hip may play a significant role in bilateral landings and to compensate for the restriction from the ankle tape on the dominant ankle. In Study Two, it was shown that in the tape condition, the dominant hip used additional ROM throughout the landing, and there was no change to the non-dominant hip ROM, which could be a source of the deviation in the kinematic CRP in the intralimb ankle-hip and knee-hip pairs, as well as in the interlimb hip-hip pair. A potential source of the differences in the kinematic CRP could stem from ROM involving the non-dominant hip as the participants were in the lowest point of the landing where less flexibility at the hip could alter the relative positioning. Across all intralimb kinematic pairs the proximal joint tended to phase lead the distal joint which was most apparent in the ankle-hip and ankle-knee pairings.

6.4.4: Intralimb Kinetic CRP

As noted in Study One, no differences in peak GRF were noted between the tape and no tape conditions; however, there was a significant increase in the loading rate in the dominant limb, but not the dominant limb in the taped condition. Considering that kinetic CRP is calculated outside of strictly the time domain utilizing both the joint moment as well as the position, the effect of different loading rates between limbs in the initial stage of landing did not influence the overall kinetic coordination between the limbs. In the intralimb kinetic CRP pairs, only the non-dominant ankle-hip had a significant difference; however, it was only for one percent of the landing and not sustained, which might suggest that the difference observed is not indicative of a substantial change to the overall coordination between the ankle and hip. While there appeared to be a trend across all kinematic coordination pairs for the proximal joint to lead the distal joint, a similar pattern was not observed across the kinetic coordination pairs. The ankle-knee kinetic CRP pairs in both limbs showed the ankle phase leading the knee with the peak phasic difference occurring around 25% of the landing. The knee-hip and ankle hip kinetic CRP pairs were mostly in phase until approximately 40% of the landing before there was a trend for the proximal joint to phase lead the distal joint which remained until the completion of the landing. Both the knee and hip have large musculature that contributes to landing safely, and when looking at the kinetic coordination, it appeared that the more proximal joint took on a larger role in the second half of the landing.

Like the kinematic CRP, the kinetic CRP of the ankle-knee pairs in both limbs became more tightly coupled and in phase after about 50% of the landing until the end. It is likely that the large forces sustained early in the landing contribute to the larger variability, and after the peak GRF, the latter stage of landing can become more tightly coupled. All intralimb kinetic CRP pairs became consistent and less variable for about the last 15% of the landing, even if they were not in phase, suggesting that in the final part of the landing, the joints are more coupled and consistent in their relationship. Overall, all intralimb kinetic CRP pairs had a wider CRP range than the kinematic CRP pairs which could suggest that a variable force attenuation strategy still allows for a successful landing.

6.4.5 Intralimb Kinematic and Kinetic CRP

One interesting finding regarding the intralimb kinematic and kinetic CRP was the behavior of the ankle-knee. In both dominant and non-dominant kinematic ankle-knee CRP pairs, the knee phase led the ankle in the first 50% of the landing with the peak occurring approximately at 25%, while in the kinetic ankle-knee CRP pairs, the ankle phase led the knee in the first 50% of the landing, with the same peak occurring at approximately 25%. This would suggest that during the first 50%, and even more specifically with the peak occurring at 25% of the landing, that the knee is doing more of the kinematic changes through the larger ROM, while the ankle is taking on a larger kinetic load to dissipate the forces of landing. A similar pattern, though less dramatic, was seen in the dominant and non-dominant ankle-hip pairings. In the kinematic CRP pairs, the hip phase led the ankle, while the ankle phase led the hip in the kinetic CRP pairs. With both the ankle-hip and ankle-knee pairs, it could be interpreted that the ankles play an important role in mitigating the forces of landing, particularly in the early stage of landing, especially considering the ankle is the first joint to undergo force attenuation as it is most distal in the kinetic chain. The ankle has been previously reported to be the joint where significant forces are dissipated (Devita & Skelly, 1992). Contrary to the initial hypothesis, the differences which did occur in CRP appeared in the latter stages of the landing, rather than at the beginning. This was a surprising finding, as some of the kinetic and kinematic differences that were noted in Studies One and Two were related to differences at the initial stage of landing in the dominant ankle and hip angles at contact and the loading rate and loading time in the dominant limb. As CRP calculations incorporate joint velocity with joint position for kinematic CRP, and joint moment with joint position for kinetic, the combination of variables and transforming data into the phase space could negate the significant findings from kinetic or kinematic variables alone.

In the non-dominant intralimb pairs, the distal joint trailed the proximal joint in the tape condition for both the ankle-hip and knee-hip pairs. This was an interesting finding, as Study One highlighted that both the dominant ankle and hip flexion were reduced at the initial contact, rather than at the end of the landing with no change to the peak ROM, yet the differences in CRP appeared with the ankle and hip in the non-dominant limb at the ending stage of the landing, in addition to the knee-hip differences at the end of the landing. The CRP of the ankle-knee pairs in both limbs (Figures 6.4.2 and 6.4.6) are similar to previously reported joint flexion ratios throughout a bilateral landing in males, finding that intralimb coordination is nonlinear (Yeow et al., 2011a). The CRP in this study highlights that coordination is variable and complex while performing a landing, suggesting that an exponential or logarithmic relationship as found in Yeow, et al. may be too simplistic (Yeow et al., 2011a). Further, with the addition of kinetic CRP analysis, the present investigation uncovered the differing relationships between kinetic and kinematic CRP while performing a bilateral landing. Particularly in the ankle-knee pairing in both limbs, the ankle led the knee in its kinetic phase, but trailed the knee in its kinematic phase, which presents a more robust view of how the ankle and knee interact while performing a landing.

6.4.6: Ankle Taping on CRP

Across all inter- and intralimb kinetic and kinematic CRP pairs, there was little influence from taping one ankle. The lack of significant findings resulting from taping one ankle could be viewed that kinematic coordination is robust with moderate variability and kinetic coordination having larger variability when landing; therefore, changes from ankle tape are not greater than the normal variability. As Study One highlighted the protective mechanism from tape where there was reduced plantarflexion at contact, ankle taping still appears to be providing one of the key benefits to preventing ankle sprain. At the same time, ankle taping is not drastically altering the coordinative relationships both in the interlimb as well as the intralimb pairs, especially in the initial stage of landing where ankle sprains are more likely to occur. In both kinetic and kinematic CRP, the variability shrinks at the end and the significant differences from ankle tape are seen when the variability is smaller. Considering that the differences occurred in the non-dominant hip, it is possible that attentional focus played a role as the non-dominant hip is the farthest joint from the ankle which received tape; therefore, increasing attention to the taped ankle and away from the farthest joint from the taped ankle.

6.4.7: Coordination Variability

Applying ankle tape to the dominant ankle did not result in any significant differences across kinematic and kinetic CRP variability, which is likely a result of the large variability of the CRP values across all pairs. Because variability was high across the trials and participants, whether ankle tape was applied or not did not significantly influence the size of the observed variability. It is possible that performing a bilateral landing as completed in this investigation does not necessitate an approach with minimal variability as all participants performed the landings in a safe manner. A possible source of the larger variability could stem from the unfamiliar task demands of landing from a platform 60 cm above the ground, especially when performing in a controlled laboratory experiment. Additionally, although instruction and feedback were provided to the participants, how they placed their non-dominant foot over the edge of the platform to initiate the landing could vary, which could influence how the dominant leg 'caught up' to the non-dominant leg in air to enable a bilateral landing.

Finding no significant differences with coordination variability was interesting because ankle bracing has been reported to decrease variability in gait (Jagodinsky et al., 2020). Although the task demands are different, the restricted motion stemming from the ankle tape could have acted to maintain a more coupled effect therefore reducing the variability in the landing specifically in the ankle-ankle interlimb pair. It could be possible that increasing the number of trials or participants in the current investigation might influence the coordination variability as the kinematic interlimb ankle-ankle pair appeared to have less variability while taped, but there was not a significant difference.

6.5: Conclusion

Most often, CRP has been utilized to describe repetitive, rhythmic movements such as walking and running (Lamb & Stockl, 2014). Discrete movements such as squatting and jumping have also been assessed; however, the landing process still has not received much attention (Hu, Kobayashi, Zhou, & Lam, 2021). Interlimb kinetic and kinematic coordination was largely in phase throughout the landing, and was not influenced by taping the dominant ankle, highlighting that coordination while landing is robust. These results highlight that there is likely a demand to not only land with a symmetrical movement in the joints, but a symmetrical force distribution across both limbs, even when the dominant ankle is taped. This investigation also highlighted that the non-dominant hip was most influenced by the application of ankle tape to the dominant ankle, as all significant differences in CRP involved the non-dominant hip. This effect could potentially stem from attentional focus since the non-dominant hip is the farthest removed from the taped ankle.

Importantly, this investigation highlights the benefit of utilizing both kinematic and kinetic CRP to describe the varied effects in both interlimb and intralimb coordination. One of the primary examples of this effect was in the ankle-knee kinetic and kinematic CRP pairs where the joints showed inverse phasic relationships depending on whether kinetic or kinematic CRP was assessed while performing a landing, which was irrespective of ankle taping. Overall, the results presented here give a better understanding of how bilateral landings are performed and the role that the joints take on to slow the COM to a controlled stop and land safely.

CHAPTER 7
SUMMARY

7.1: Dissertation Aims

The overall goal of this dissertation was to investigate the effects of unilateral ankle taping on the performance of a bilateral drop landing using a multifaceted analytical approach. The first aim sought to understand bilateral differences at the ankle, knee, and hip joints using traditional kinematic and kinetic variables before and after a bout of exercise. Aim two sought to understand the effects of unilateral ankle taping on lower-body, leg, and individual joint stiffnesses when landing from three different heights. Lastly, the third aim sought to understand how the joints in one limb, as well as the matched joints across limbs were coordinated while performing a bilateral landing with and without the application of unilateral ankle tape.

7.2: Study One

Taping the dominant ankle had several effects across joints in both lower limbs during bilateral drop landings. These findings were explored in general kinetic and kinematic measures, vertical and joint stiffnesses, and coordination. Ankle taping restricted weightbearing dorsiflexion flexibility; however, peak dorsiflexion while landing was not reduced, rather only the plantarflexion angle at contact was reduced. No changes were noted in total or individual limb peak GRF after taping the dominant ankle; however, the total loading rate was increased which was driven by an increase in the dominant limb loading rate alone. The increase in loading rate alone, or combined with the imbalance of increased loading rate in the taped limb but not in the non-taped limb could potentially increase the risk for injury. After a unilateral application of ankle tape to the dominant limb, nearly all effects noted because of ankle tape were localized to the ankle and knee of the limb that received tape, where only the peak ankle flexion moment was changed in the non-dominant limb. With the increase in the dorsiflexion moment in the non-taped ankle, the additional loading incurred after taping the dominant ankle could increase the risk for injury in that ankle. However, after a bout of exercise, the dorsiflexion moment returned to baseline values highlighting that the potential injury risk may only be present shortly after ankle tape is applied to the other ankle, and some loosening of the tape on the taped ankle might enable less reliance on the non-taped ankle during the landing.

Exercise did increase the plantarflexion angle at contact in the taped ankle without changing the peak dorsiflexion, although the plantarflexion angle at contact was still restricted compared to the control condition. This highlights that the tape was still providing some protective benefits at ground contact after exercise, and there was not a change in the baseline dorsiflexion motion while landing. Interestingly, the effects of exercise alone predominantly affected the non-dominant limb whereby the ankle saw a reduction in peak dorsiflexion as well as a greater peak hip flexion moment. These changes could potentially increase the risk of injury as the ankle is still undergoing the same force attenuation but over a smaller ROM and that the hip is taking on an additional role in dissipating the forces of landing. These findings highlight complex compensatory effects that taping an ankle or exercise can have on the mechanics of a landing.

7.3: Study Two

Study Two investigated the role of ankle taping on lower-body, leg, and joint stiffness when landing from varied heights. Neither height nor ankle taping influenced total body stiffness. Though the peak GRF increased with height, the additional use of COM displacement through functional ROM in the leg offset the additional force demands of landing to maintain a constant level of stiffness. This finding suggests that while performing a landing where the goal is to land safely, participants make use of additional ROM to maintain a constant lower-body stiffness. This lower-body stiffness was uninfluenced by the height of the platform, but it did influence each leg's stiffness, although in opposing directions where the dominant limb stiffness decreased with increasing height, while the non-dominant limb increased stiffness with increasing height. This imbalance could potentially increase the risk for injury as the non-dominant limb is taking on a greater role in mitigating the forces of landing. Taping the dominant ankle increased stiffness in the ankle and hip of that leg without increasing overall leg stiffness. With the increase in the dominant ankle and hip stiffness, but no change to the knee indicate that the loading is prioritized to be dissipated with the ankle and hip rather than the knee, which could potentially act as a protective measure to reduce the risk of injury to the knee, although this strategy could increase the risk for injury at the ankle and hip.

7.4: Study Three

When assessing the entire landing process using time-series CRP of both kinetic and kinematic data, more detailed information about the landing was uncovered. Largely, the interlimb kinetic and kinematic CRP joint pairs were unchanged with the application of unilateral ankle tape. The interlimb kinetic and kinematic joint pairs moved almost completely in phase with each other throughout the landing showing that unilateral ankle taping on the dominant limb did not change how the joints in that limb moved relative to the non-dominant limb. This was an interesting finding, particularly at the ankle as Studies One and Two highlighted both kinetic and kinematic changes in the taped ankle. Although interlimb kinetic and kinematic CRP were largely in phase, there was larger variability in the kinetic CRP when compared to kinematic CRP which could indicate that the way the forces of landing are dissipated is more variable, whereas the motion of the landing is more constrained. Further, towards the end of the landing, the coordination variability in both kinetic and kinematic interlimb CRP shrank suggesting that the latter stages of landing are more consistent, irrespective of ankle taping.

In all coordination pairs where there were significant differences because of taping the dominant ankle, the non-dominant hip was influenced. In the interlimb hip-hip pair, the non-dominant hip phase led the dominant hip when the dominant ankle was taped, but phase lagged the dominant hip in the no-tape trials. While the dominant ankle was taped, the non-dominant hip phase led both the non-dominant ankle and knee toward the end of the landing when the variability was reduced. With the non-dominant hip involved in all significant differences, it is possible that there was an attentional effect since the non-dominant hip was farthest removed from the taped ankle, and less focus was at that joint which could influence its relationship between the ankle and knee in the same leg, as well as the other hip. Also, as the differences appeared towards the end of the landing, it is possible that flexibility played a role as the joints would be most flexed while in a squatting position.

The ankle-knee pairing in both limbs was interesting because of the differing behavior between kinetic and kinematic CRP. In the ankle-knee kinematic CRP, the knee phase led the ankle for the first half of the landing, with the peak difference occurring at about 25% of the landing, while in the kinetic CRP, the ankle phase led the knee for the first half of the landing with the peak difference occurring at about 25% of the landing. This was an interesting finding because it highlighted that the knee was doing most of the movement during the first half of the landing, but the ankle was taking on a larger role in the force dissipation. This highlights the benefit of assessing both kinetic and kinematic CRP, as it paints a more complete picture of how the joints are coordinated. While this inverse behavior between kinetic and kinematic intralimb CRP pairs was seen in the ankle-knee pairing, it was not as apparent in the ankle-hip and knee-hip pairs.

In the kinetic CRP results, only two pairings showed any significant results, and they were each only for one data point and not sustained in the landing. In all kinetic and kinematic CRP pairs, no significant differences were noted for coordination variability suggesting that the ankle taping procedure did not influence the naturally occurring variability while performing a drop landing. In both kinetic and kinematic intralimb CRP, the non-dominant limb was much more variable than the dominant limb. This could highlight that the non-dominant limb is less controlled while landing when compared to the dominant limb and could potentially increase the risk of injury in that limb.

7.5: Clinical Implications

Taping an ankle remains a common practice in exercise and sport performance, with the aim to reduce the risk of sustaining an ankle sprain, or re-injury if an ankle sprain previously occurred. Study One highlighted that when initial contact occurred, the taped ankle was less plantarflexed, which is a key factor in reducing the risk of sustaining a lateral ankle sprain. Importantly, the results of this investigation did not show an increase in either knee valgus motion or moment, where increased valgus kinetics and kinematics are a risk factor for sustaining an ACL injury (Hashemi et al., 2011; Hewett et al., 2010; Hewett et al., 2005; Ireland, 1999; Kanamori et al., 2002; Levine et al., 2013; Markolf et al., 1995; Olsen et al., 2004; Quatman & Hewett, 2009; Shimokochi & Shultz, 2008). There is some evidence which would suggest that applying ankle support can negatively influence knee mechanics (Dizon & Reyes, 2010; Mason-Mackay, Whatman, & Reid, 2016). However, in the present study, outside of a change to the dominant knee moment, which may or may not be detrimental, no other negative changes were noted in the knees. Loading rate of the taped leg was seen to increase after ankle taping which places more stress on that limb, even without an increase in the peak GRF (Niu et al., 2016). Study Two highlighted that taping an ankle increased stiffness in that ankle and hip, but not the knee, with no changes to the non-taped limb. The additional stiffness from the tape may aid in protecting the ankle but pass along additional work to the hip in the taped limb. Study Three highlighted that taping one ankle did not cause many changes in either kinetic or kinematic inter- and intralimb pairs; and no observed changes occurred in the early stages of landing where ankle sprains are more likely to occur.

Taping one ankle provided effective support in positioning the foot at ground contact with less plantarflexion which reduces one of the risk factors of sustaining a lateral ankle sprain in that ankle. While at the same time, no negative effects were noted in the non-taped ankle. When looking at stiffnesses, the taped ankle, along with the intralimb hip increased; however, as there are not well-defined stiffness values for safe landings, it is unclear if the changes noted to stiffness are potentially detrimental. Research has indicated that too much or too little stiffness can present issues, so provided that the baseline stiffness was enough to protect the musculature and ligaments, and the increase in stiffness from taping is not so stiff that the risk for bony injuries is increased. Therefore, it is difficult to make a determination on whether or not taping was beneficial or detrimental (R. J. Butler et al., 2003; Granata, Padua, et al., 2002; Williams III et al., 2004). A key point to note is that without changes in the non-taped limb, the existence of asymmetrical stiffnesses across limbs could potentially increase the risk for injury resulting from the imbalance, as leg dominance can play a role in ACL injury risk (Ford et al., 2003). However, Study Three highlighted that taping one ankle did not influence either kinetic or kinematic coordination across the interlimb ankle-ankle pair which would suggest that the relationship between the ankles is not altered with tape and that the preferred landing strategy can still be maintained.

When deciding on whether to tape one ankle or both ankles, there are some key points which should be addressed. While applying tape unilaterally does create differences between the two limbs through changes observed in the taped ankle, it raises the question if it would be advisable to tape both ankles if one ankle is taped. Each study presented here highlights unique aspects of landing, and not one study alone can provide a clear answer to this question. The present investigation did not exclude individuals with previous ankle sprain if that ankle sprain occurred more than 12 months prior to participation, or if it was perceived to not influence daily living activities; which is an important consideration as both tape and braces have been shown to be effective in preventing additional ankle sprains to offset mechanical and functional losses after an initial ankle sprain (Dizon & Reyes, 2010). Even with differences noted in general biomechanical variables, along with joint stiffnesses, the way in which the joints are coordinated does not have many changes, which suggests that unilateral taping could still be a beneficial clinical decision. On the other hand, with the asymmetries seen from the joint stiffnesses, long-term use of unilateral ankle tape may further exacerbate these asymmetries and the performers may adopt other compensatory behaviors which could increase other injury risks.

Across the three investigations presented here, there were a variety of kinetic and kinematic changes brought on by unilateral ankle tape, with some positive protective mechanisms, some negative, and others with no change at all. Clearly the question of whether to tape only one ankle, or tape both, is a complex one with many factors at play. The findings from this dissertation lean more to suggest that continuing to tape one ankle is still a worthwhile practice. However, the clinical implications from the results presented here would best be discussed between the performers and sports medicine professionals, especially considering things like previous injury history, or exercise and sport specific tasks, where there may be a higher risk for ankle sprain. Ultimately, further investigation would be needed before a more definite clinical recommendation could be made.

7.6: Summary

Together, these studies highlight that while taping one ankle and performing a bilateral landing does not drastically alter lower body mechanics, there are some changes which could potentially increase the risk for injury because of the unilateral constraint. Results from these investigations showed that when applying a unilateral constraint through ankle taping, and performing a bilateral task, differences are seen not just at the taped ankle, but also in the knee and hip of that leg, as well as in the joints in the other leg. Through the implementation of multiple analytical techniques, the influence of unilaterally applying ankle tape and performing a bilateral landing can be more thoroughly understood as it influences various aspects of kinetics and kinematics.

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APPENDIX

Old Dominion University Institutional Review Board, Norfolk, VA Approval – [984816-8] The Effect of Ankle Taping on Drop Landing. Board Reference #'s - 16-219; 17-235; 18-238; 19-013. Initial Approval Date: 11/17/2016.

A1: INSTITUTIONAL REVIEW BOARD APPLICATION

OLD DOMINION UNIVERSITY
HUMAN SUBJECT RESEARCH REVIEW APPLICATION FORM

Responsible Project Investigator (RPI)		
Responsible Project Investigator: The RPI must be a member of ODU faculty or staff who will serve as the project supervisor and be held accountable for all aspects of the project. Students cannot be listed as RPIs.		
First Name: Daniel	**Middle Initial: M**	**Last Name: Russell**
Telephone: 757-683-6016	**Fax Number:**	**E-mail:dmrussel@odu.edu**
Office Address: Innovation Research Park Building II, Suite 200, 4211 Monarch Way		
City: Norfolk	**State: VA**	**Zip: 23508**
Department: School of Physical Therapy and Athletic Training	**College: Health Sciences**	
Complete Title of Research Project: The Effect of Ankle Taping on Drop Landings		**Code Name (one word): Drop-Landings**
Investigators If more investigators exist than lines provide, please attach a separate list.		
Investigator(s): Individuals who are directly responsible for any of the following: the project's design, implementation, consent process, data collection, and/or data analysis.		
First Name: Eric	**Middle Initial: D**	**Last Name: Jenkins**
Telephone: 757-345-1964	**Fax Number:**	**Email: ejenk002@odu.edu**
Office Address: Innovation Research Park Building II, Suite 200, 4211 Monarch Way		
City: Norfolk	**State: VA**	**Zip: 23508**
Department: School of Physical Therapy and Athletic Training	**College: Health Sciences**	
Affiliation: __Faculty __Staff	**X** Graduate Student __Other________________	__ Undergraduate Student
First Name: Cortney	**Middle Initial:**	**Last Name: Armitano**
Telephone: 757-683-6002	**Fax Number:**	**Email: carmi002@odu.edu**
Office Address: Innovation Research Park Building II, Suite 200, 4211 Monarch Way		
City: Norfolk	**State: VA**	**Zip: 23508**
Department: School of Physical Therapy and Athletic Training	**College: Health Sciences**	
Affiliation: __Faculty __Staff	**X** Graduate Student __Other________________	__ Undergraduate Student
List all information for additional investigators on attachment and check here: __		
Type of Research		

1. This study is being conducted as part of (check all that apply):
__Faculty Research
X Doctoral Dissertation
__Masters Thesis
__Non-Thesis Graduate Student Research
__Honors or Individual Problems Project
__Other______________________

Funding

2. How is the research project funded?
X Research is **not funded (go to 3)**
__Research is **funded (go to 2a)**
__**Funding decision is pending** (funding decision has not been made) (**go to 2a)**

2a. What is the type of funding source? (Check all that apply)
__Federal Grant or Contract
Agency Proposal Number____
Grant Start Date (MM/DD/YY) ___________________ Grant End Date (MM/DD/YY) ___________________
__State or Municipal Grant or Contract
__Private Foundation
__Corporate contract
__Other (specify):____

2b. Who is the point of contact at the funding source?
Name:
Mailing Address:

Telephone: **Email:**

Research Dates

3a. Date you wish to start research (MM/DD/YY): 11/20/2016
3b. Date you plan to end research (MM/DD/YY): 11/19/2017 (End date for data collection and analysis)
Note: Protocols are approved for a maximum of 1 year. If a proposed project is intended to last beyond the approval period, continuing review and reapproval are necessary.

Research Location

4. Where will the experiment be conducted? (Check all that apply)
X On Campus (Building and Room Number)
Innovation Research Park Building II, Suite 200, 4211 Monarch Way

__ Off-Campus (Street Address)

Human Subjects Review

5. Has this project been reviewed by any other committee (university, governmental, private sector) for the protection of human research subjects?

__Yes

X No (**If no, go to 6)**

5a. If yes, is ODU conducting the "primary" review?

__Yes

__No (**If no, go to 5b)**

5b. Who is conducting the primary review?

Study Purpose

6. Describe the rationale for the research project.

Previous research investigations have looked into the influence of ankle taping and bracing on the mechanics of the lower limbs; however, none have provided an all-encompassing research design that will provide numerous findings on kinetics and kinematics of the lower body. Through the use of electromyography to record muscular activation, 3D motion capture and forceplates to record impact forces, the combination of that data will provide numerous data points that will add to the current body of research and progress the body of knowledge on ankle tape with regard to lower body kinetics and kinematics.

Subjects

7. What will be the maximum number of subjects in the study? 20

7a. Indicate the approximate number of: Males – 0 Females – 20

7b. What is the age of subjects? (Check all that apply)

__Children (1-17 years old) **X** Adults (18-65 years old)

__Elderly (64-years and older)

7c. Will students be enrolled in the study? (Check all that apply)

X Undergraduate students dept* Kinesiology and Rehabilitation **X** Advanced students dept Kinesiology and Rehabilitation

*If students are under 18 years old, parental consent must be obtained

7d. Provide rationale for the choice of subjects. Enumerate any additional defining characteristics, including age, of the subject population. (e.g., symptomatology, history, socio-economic status).

Females aged 18-35 will be the only gender studied. This is based on the large body of research on ACL injuries on female athletes, and the influence of ankle stability is a factor in the risk for ACL tears. Age of participants is set to include a large population of healthy adult women with no physical limitations that would impede performance of this study, or pose a risk to their health. The researchers are looking at a number of variables, and the addition of males to observe a gender effect would double the time needed to complete the study and increase the number of participants. By limiting it to only females, the variance is reduced, thus, giving way for stronger statistical significance.

Vulnerable Subjects

8. Are research subjects being used whose ability to give informed voluntary consent may be in question? (**e.g., children, persons with AIDS, mentally disabled, psychiatric patients, prisoners.)**

__Yes (**If yes, explain the procedures to be employed to enroll them and to ensure their protection).**

X No

8b. What type of vulnerable subjects are being enrolled? (**check all that apply)**

__Critically Ill Patients
__Prisoners
__Pregnant Women
__Other____________________
__Mentally Disabled or Cognitively Impaired Individuals
__Physically Handicapped
__Children

Recruitment

9. How will participants be recruited? (**Please submit a copy of the sign-up sheet, newspaper advertisement, or any other protocol or procedure which will be used to recruit subjects.)**

__Internet
__Newspaper/radio/television advertising
X Posters/brochures/letters
X Other – Simple selection of participants enrolled in the department through word of mouth

Comments:

Inclusion and Exclusion Criteria

10. Are subjects equitably chosen for participation in the study? (**no one group is excluded without justification)**

__Yes

X No (**If no, specify criteria and justify in detail below.)**

Females will be the only gender studied. This is based on the large body of research on ACL injuries on female athletes, and the influence of ankle stability is a factor in the risk for ACL tears. The researchers are looking at a number of variables, and the addition of males to observe a gender effect would double the time needed to complete the study and increase the number of participants. By limiting it to only females, the variance is reduced, thus, giving way for stronger statistical significance.

10a. Does the study require special evaluation and screening of potential subjects to determine their appropriateness for inclusion in the study?

X Yes (**If yes, briefly elaborate on the screening process and attach the screening questionnaire.)**

__No

Participants will be asked if they are of legal consent age (18), as well as history of injury to the lower limbs that would risk their personal safety through participation in addition to providing a skewed data set. Additionally, they will be asked about their current exercise habits and involvement of running, as running is a part of the protocol, moderate fitness will be necessary to prevent fatigue and injury.

Experimental Procedures

11. Describe the experimental procedures that will be followed. (Include a succinct, but comprehensive statement of the methodology relating to the human subjects. You are encouraged to include a discussion of statistical procedures used to determine the sample size.)
Participants will be asked to wear form fitting elastic clothing to facilitate the application of reflective markers placed on anatomical landmarks that will be utilized by a 3D motion capture camera system. A lower body 3D reflective marker set will be applied that will allow the researchers to establish joint centers and centers of mass, which will aid in kinetic and kinematic analyses. Additionally, participants will have surface EMG placed on four muscles (gastrocnemius, tibialis anterior, rectus femoris and the biceps femoris) that will aid in giving insight to muscular activation. In order to remove any impedance, the participant will have a 3x3" patch of hair shaved, lightly exfoliated and cleaned with alcohol. Participants will then perform a drop jump landing onto an AMTI forceplate from three different heights of 30cm, 45cm and 60cm, based on previous research studies' protocols. Participants will perform 5 baseline drop landings from each height, 5 immediately after taping at each height and 5 after the exercise bout at each height, for a total of 45 drop landings. Participants will be allowed to employ their own landing technique with no coaching or training provided by the researchers. All participants will complete the control condition with no tape applied, then receive ankle tape applied to one ankle by an ATC and complete a 20 minute exercise bout consisting of a treadmill run and lateral shuffling, followed by another series of drop jump landings. Participants will be given a pair of laboratory shoes to standardize across all participants. Functional ROM data will be gathered at the start, immediately after taping as well as after the exercise bout. Data will be collected using Vicon Nexus 3D motion capture software that will also analyze forceplate data. Kinetic and kinematic data will be analyzed in Visual3D, and EMG data will be analyzed in a custom MatLab program. Statistics will be run through SAS utilizing an ANOVA and post-hoc tests to determine further significant impact. The sample size was derived from numerous other studies that implemented a similar protocol.

11a. Will any aversive or painful procedures be employed (e.g., shock, the threat of shock or punishment, experimentally induced stress?)
__Yes (**If yes, specify and justify in detail below.)**
X No

11b. Will the deliberate deception of research participants be involved as part of the experimental procedure?
__Yes (**If yes, explain the nature of the deception, why it is necessary, any possible risks that may result from the deception, and the nature of the debriefing with specific reference to the deception.)**
X No

Attach copies of the following items:
(**listed in 11 above)** Research Protocol(s)
___Questionnaire
___Copies of any instructions or debriefings given
___If the research is part of a research proposal submitted for federal, state or external funding, submit a copy of the FULL proposal

Compensation

12. How much time will be required of each subject?
Two visits consisting of approximately 75 minutes, for a total time commitment of 150 minutes.

12a. Will research subjects receive course credit for participating in the study?
__Yes (**If yes, please explain in comments section.)**
X No
Comments:

12b. Are there any other forms of compensation that may be used? (e.g. Money)
__Yes (**If yes, please explain in comments section.)**
X No
Comments:

12c. Are there any penalties for subjects who do not show up for a research session?
__Yes (**If yes, please explain in comments section.)**
X No
Comments:

Informed Consent

13. Do you intend to obtain informed consent from subjects?
X Yes (**please answer question 13a)**
__No (**please complete Appendix F: Request for Waiver of Consent Form)**

13a. Describe the procedures that will be used to obtain Informed Consent and attach the Informed Consent Document (follow the guidelines for preparation of the University Informed Consent Form).
Note: Subjects MUST be given a description of the procedures and rationale for the study to the extent possible. The benefits and ANY risks associated with participating in the study MUST be enumerated. The subjects MUST be informed of their right to terminate the experiment at any time. If there is no risk associated with the study and participants' signature on the informed consent sheet is the only identifying information about the name of the subject, then the subjects' signature may not be necessary.

Informed consent will be obtained from each participant before any study related procedure are performed, and any possible risks are explained to their complete understanding, where any questions or concerns will be addressed fully. Participants will be instructed on how their personal information will be kept anonymous and secure. The participants will be instructed to read over the attached Informed Consent form, and will be asked if they have any further questions or concerns. Additionally, they will be told that their participation is voluntary and they may remove themselves from the study at any point without penalty.

Risks

14. What are potential risks of the research? (Check all that apply)
X physical harm
__psychological harm
__Release of confidential information
__Other________________________________

14a. Describe any potential risks to subjects for the activities proposed and describe the steps that will be taken to minimize the risks. Include any risks to the subject's physical well being, privacy, dignity, emotions, employability, and criminal and legal status. A detailed, comparative statement of the risk (harm or likelihood) must also be described in the consent form.
As with any study involving exercise, the risk of injury is present; however, steps will be taken to ensure that risk is minimal. Participants will be fully instructed on the details and requirements of the study, including running, lateral shuffling and landing from heights up to 60cm. Should they not feel comfortable with these requirements, they will be allowed to forfeit participation without penalty. Participants will be required to run, shuffle, and jump; should any of the test parameters be in question, the participants will be provided with a demonstration to aid in minimizing potential risk. It is possible to have minor skin irritation at the site of electrode placement, where the skin is shaved and lightly exfoliated, as well as a possible reaction to the adhesive tape used. Risk of injury would include: fatigue, soreness, strains, sprains, as well as any others that would accompany typical exercise. Should any injury occur, there is a certified athletic trainer available for immediate injury appraisal, treatment, and referral, should that be necessary.

Please attach the following (if you have developed them)
The script by the experimenter to disclose potential harm and likelihood (risk) prior to the subject's choice to participate.
Noted in Informed Consent Document

Benefits

15. Assess the potential benefits that may accrue to the individual subject as well as to others as a result of the proposed study. Do the potential benefits justify the possible risks involved? Although you may mention general benefits to society, such speculative benefits should not be presented to a subject as a direct benefit for informed consent.
No direct benefits will be gained by the participant for their completion.

Protection of Anonymity

16. Describe in detail the procedures for protecting the anonymity (meaning that no one will ever be able to know the names) of the research subjects. If anonymity is impossible, then describe in detail the procedures for safeguarding data and confidential records. These procedures relate to how well you reduce the risk that a subject may be exposed or associated with the data.

Anonymity will be kept through stringent protective behaviors by the researchers. The only personal information will contain the participants' name and a coded number that will be used to ensure proper data collection, which will be kept separate from the research data in a secure locked file cabinet in the Center for Brain Research and Rehabilitation. Only anonymous data will be reported in the research findings. Once all data has been collected, the identifying code page will be destroyed.

Drugs or Devices

17. Will any drugs, devices, or chemical biological agents be used with the subjects?

__Yes (**If yes, please attach Appendix G: Drugs, Agents, and Devices Form)**

X No

Biological Materials

18. Will this research involve the collection, analysis, or banking of human biological materials (cells, tissues, fluids, DNA?)

__Yes (**If yes, please attach Appendix H: Biological Materials Form)**

X No

Training

19. Briefly explain the nature of the training and supervision of anyone who is involved in the actual data collection, research design, or in conducting the research. This information should be sufficient for the IRB to determine that the RPI and investigators possess the necessary skills or qualifications to conduct the study.

Daniel Russell, PhD is an assistant professor in the Kinesiology and Rehabilitation Sciences program and has conducted numerous human movement, utilizing similar equipment for more than 20 years of research investigations with previous IRB approval.

Eric Jenkins received his master's degree in exercise physiology from the University of Texas at Arlington where he participated in numerous research investigations currently in preparation for peer reviewed journal submission. He also has current CITI training accreditation.

Cortney Armitano is a certified athletic trainer, responsible for the application of the ankle tape and is available during testing, should any injury occur. She also has current CITI training accreditation.

Human Subjects and HIPAA Training

20. A. All investigators (including graduate students enrolled in Thesis and Dissertation projects involving human subjects) must document completion of the CITI Human Subject Protection course.
(Attach a copy of all CITI Human Subject Protection completion certificates.)
Date RPI completed Human Subject Protection training: 11/9/2015

B. RPI's who propose studies with patient populations must document HIPAA training by accessing the NIH booklet entitled "Protecting Personal Health Information in Research: Understanding the HIPAA Privacy Rule" at: http://privacyruleandresearch.nih.gov/pr_02.asp. and must submit an attachment to the review application stating that the material has been read and will be adhered to in the proposed research. The attachment must include the date the material was read, which must be within the 12 months prior to the application. (If you are submitting this attachment with your application the RPI must initial here:_________________

PLEASE NOTE:

- You may begin research when the University Institutional Review Board gives you final WRITTEN notice of its approval.
- You MUST inform the committee of ANY adverse event, changes in the method, personnel, funding, or procedure.
- At any time the committee reserves the right to re-review a research project, to request additional information, to monitor the research for compliance, to inspect the data and consent forms, to interview subjects that have participated in the research, and if necessary to terminate a research investigation.

Daniel M. Russell 11/7/2016

A2: INFORMED CONSENT DOCUMENT

INFORMED CONSENT DOCUMENT

OLD DOMINION UNIVERSITY

PROJECT TITLE: The Effect of Ankle Taping on Drop Landings

INTRODUCTION
The purposes of this form are to give you information that may affect your decision whether to say YES or NO to participation in this research, and to record the consent of those who say YES. This project, The Influence of External Bracing on Walking and Postural Control, will be conducted at Old Dominion University in the Center for Brain Research and Rehabilitation, which is on the second floor of the Innovation Research Park 2 Building on Monarch Way.

RESEARCHERS
The Responsible Project Investigator is Daniel Russell, PhD, Assistant Professor of Physical Therapy & Athletic Training, College of Health Sciences. Co-investigator Eric Jenkins, a graduate student in the Kinesiology & Rehabilitation, PhD.

DESCRIPTION OF RESEARCH STUDY
The purpose of this research is to better understand the effect of ankle tape on the mechanics (technique) on landings from an elevated platform at three different heights.
The expected duration of your participation will be approximately 75 minutes per visit for two separate visits, for a total of 150 minutes. This study will require nothing other than your time and participation.
The procedures for this study are as follows. You will be asked to change into laboratory clothing (spandex shorts and a spandex shirt) and fitted with reflective markers on anatomical landmarks (foot, ankle, knee, and hip) as well as electromyography (EMG) sensors that sit on the skin. In order to prep for the EMG, any hair must be removed from a 3x3" patch of skin at four locations (calf, quadriceps, hamstrings, and the front of lower leg) and then cleaned with an alcohol wipe. After this is completed, you will stand still and be recorded for the 3D camera system. Next you will perform the first series of 15 drop landings from the three different heights (5 each from 30, 45, and 60 cm) for a total of 45 landings. Following the first jumps, you will then receive or not receive ankle tape to one ankle, then perform the second series of landings. Immediately after those landings, you will perform the 20 minute exercise routine which consists of a comfortable paced jog and lateral shuffle maneuvers. Then the last series of drop landings will be conducted. You will then have the reflective markers and EMG removed. This will be completed for both the first and second visit.

If you say YES, then your participation will last for approximately 1 hour on two separate days at the Center for Brain Research and Rehabilitation, IRP2 Building, Monarch Way. Approximately 20 women will be participating in this study.

EXCLUSIONARY CRITERIA
To the best of your knowledge, you should not have any heart or breathing problems or muscle, bone or joint pain that would become worse through walking and moving. A moderate level of fitness is important for this investigation, as it involves an exercise protocol, so familiarity with this level of exercise will facilitate participation. If you had any problems that make it unsafe for you to walk or participate in any activity in this study please tell the researcher. Researchers reserve the right to screen participants if they deem it necessary, this would include any history of lower limb injury, especially any ankle injury within the last 12 months. Only females will be studied for this investigation.

RISKS AND BENEFITS
RISKS: As with any exercise program, there is a minimal risk of injury; the risks might include musculoskeletal injuries involved in running and jumping. In addition, fatigue is possible, since you will be

asked to jog and do light to moderate exercise for a short period. These risks and discomforts will be reduced by proper instruction of any movements you might be unfamiliar with, in addition to your self-selected pace for the jogging. You will also be allowed to rest after the exercise, should you desire. There is a possibility of temporary redness to the skin from the electrodes.

BENEFITS: There is no direct benefit to you for participating in this study. The main benefit is in helping researchers to understand the influence of ankle taping on drop landing mechanics.

COSTS AND PAYMENTS

The researchers are unable to give you any payment for participating in this study.

NEW INFORMATION

If the researchers find new information during this study that would reasonably change your decision about participating, then they will give it to you.

CONFIDENTIALITY

The researchers will take all reasonable steps to keep your information private. In their records, the researchers will identify you by number, and all of your information will be filed according to number. Your name and any identifying information will not be kept in the same file as your data. Images of you walking will not be used in a classroom for teaching or for research presentations unless you have specifically signed a "permission to use images" form giving the researchers permission to use the images in these ways. The results of this study may be used in reports, presentations, and publications; but the researcher will not identify you. Of course, your records may be subpoenaed by court order or inspected by government bodies with oversight authority.

WITHDRAWAL PRIVILEGE

It is OK for you to say NO. Even if you say YES now, you are free to say NO later, and walk away or withdraw from the study -- at any time. Your decision will not affect your relationship with Old Dominion University, or otherwise cause a loss of benefits to which you might otherwise be entitled. The researchers reserve the right to withdraw your participation in this study, at any time, if they observe potential problems with your continued participation.

COMPENSATION FOR ILLNESS AND INJURY

If you say YES, then your consent in this document does not waive any of your legal rights. However, in the event of injury arising from this study, neither Old Dominion University nor the researchers are able to give you any money, insurance coverage, free medical care, or any other compensation for such injury. In the event that you suffer injury as a result of participation in any research project, you may contact Dr. Daniel Russell the Responsible Project Investigator at 757-683-6016 or call Dr. Tancy Vandecar-Burdin, the current IRB chair, at 757 683 3802 at Old Dominion University, who will be glad to review the matter with you.

VOLUNTARY CONSENT

By signing this form, you are saying several things. You are saying that you have read this form or have had it read to you, that you are satisfied that you understand this form, the research study, and its risks and benefits. The researchers should have answered any questions you may have had about the research. If you have any questions later on, then the researchers should be able to answer them: Dr. Daniel Russell (Responsible Project Investigator) 757-683-6016

If at any time you feel pressured to participate, or if you have any questions about your rights or this form, then you should call Dr. Tancy Vandecar-Burdin, the current IRB chair, at 757 683 3802, or the Old Dominion University Office of Research, at 757 683 3460.

And importantly, by signing below, you are telling the researcher YES, that you agree to participate in this study. The researcher should give you a copy of this form for your records.

Subject's Printed Name & Signature	Date

INVESTIGATOR'S STATEMENT

I certify that I have explained to this subject the nature and purpose of this research, including benefits, risks, costs, and any experimental procedures. I have described the rights and protections afforded to human subjects and have done nothing to pressure, coerce, or falsely entice this subject into participating. I am aware of my obligations under state and federal laws, and promise compliance. I have answered the subject's questions and have encouraged him/her to ask additional questions at any time during the course of this study. I have witnessed the above signature(s) on this consent form.

Investigator's Printed Name & Signature	Date

A3: SCREENING QUESTIONNAIRE

Screening Questionnaire

1. Are you at least 18 years of age at the start of this study?

2. Have you ever sustained an injury to your lower limbs that continues to impact your daily life? If so, briefly explain.

3. Have you sustained an injury to your lower limbs in the past 12 months (sprained ankle, broken bone, torn ACL, etc.)?

4. Are you physically active at least 3-4 days per week (for this study, physically active includes activities in which you would run for at least 15 minutes per exercise session)?

A4: MODEL DEFINITION

The model was defined by two limbs consisting of three rigid segments (foot, shank, and thigh) with a single segment pelvis. The ankle joint was defined by the medial and lateral malleoli. The knee joint was defined by the medial and lateral femoral epicondyles. The hip joint center was defined using the de Leva method proximal to the greater trochanter and defined as the difference between the hip joint center-greater trochanter and the scaled greater trochanter-iliac spine (de Leva, 1996a, 1996b).

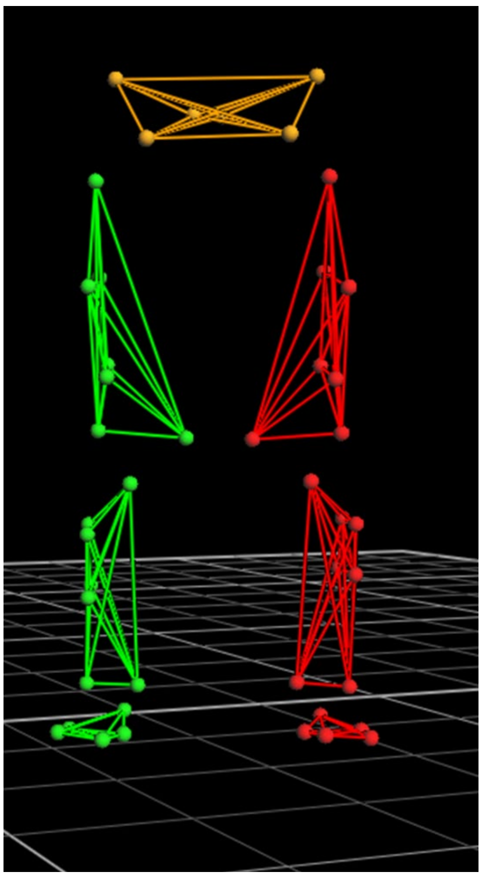

Figure 10. Lower body marker set for three-dimensional motion capture analysis based on segmental properties derived from De Leva (de Leva, 1996a).

A5: LANDING PLATFORM

Figure 11. Custom built landing platform with three heights of 30, 45, and 60 cm pictured with the two force plates used for kinetic analysis.

A6: RATED PERCEIVED EXERTION SCALE

Borg RPE	
Score	Level of exertion
6	No exertion at all
7	
7.5	Extremely light
8	
9	Very light
10	
11	Light
12	
13	Somewhat hard
14	
15	Hard (heavy)
16	
17	Very hard
18	
19	Extremely hard
20	Maximal exertion

Figure 12. Borg Rating of Perceived Exertion (RPE) Scale (Borg, 1982).

A7: SPM COORDINATION AND COORDINATION VARIABILITY FIGURES

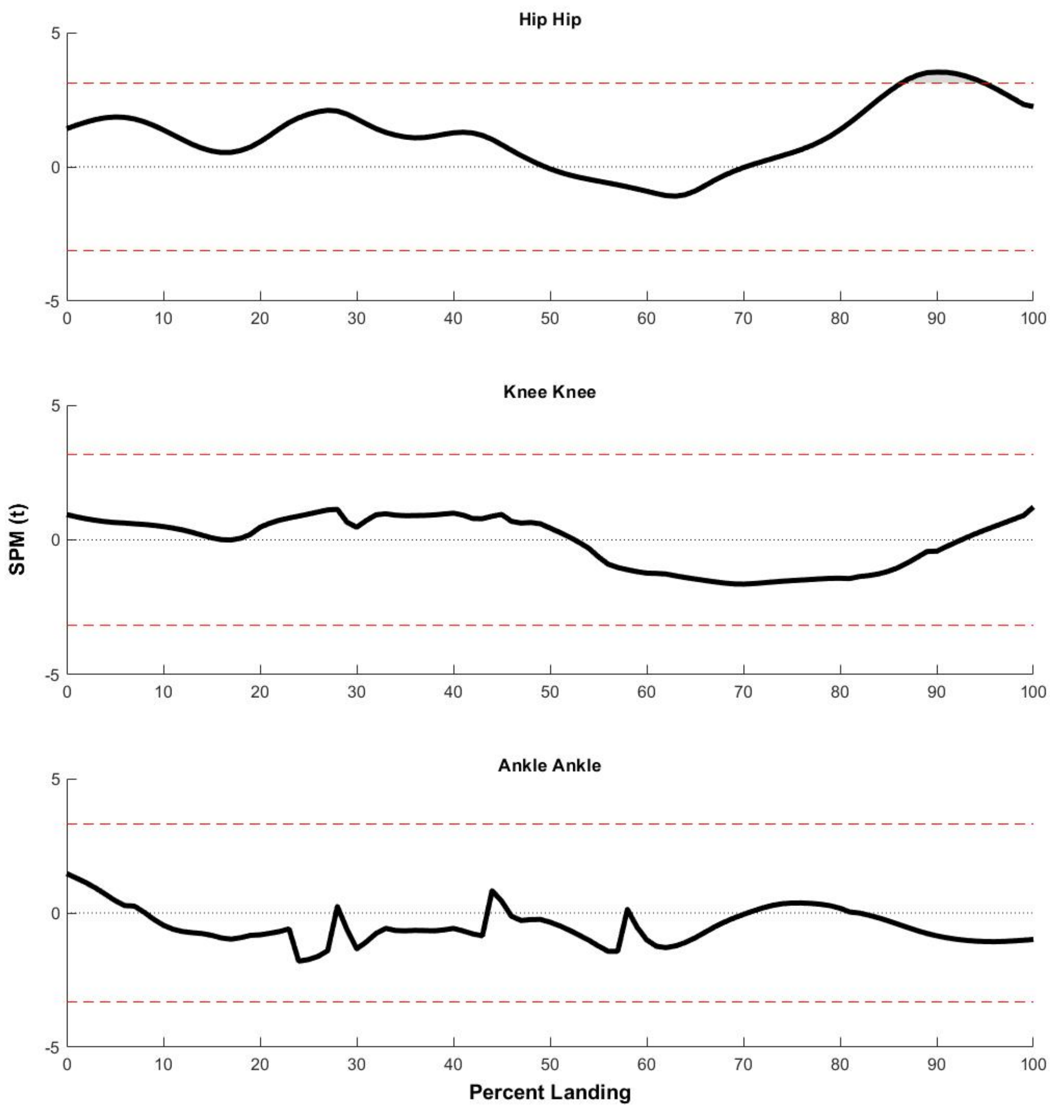

Figure 13. SPM of interlimb kinematic CRP.

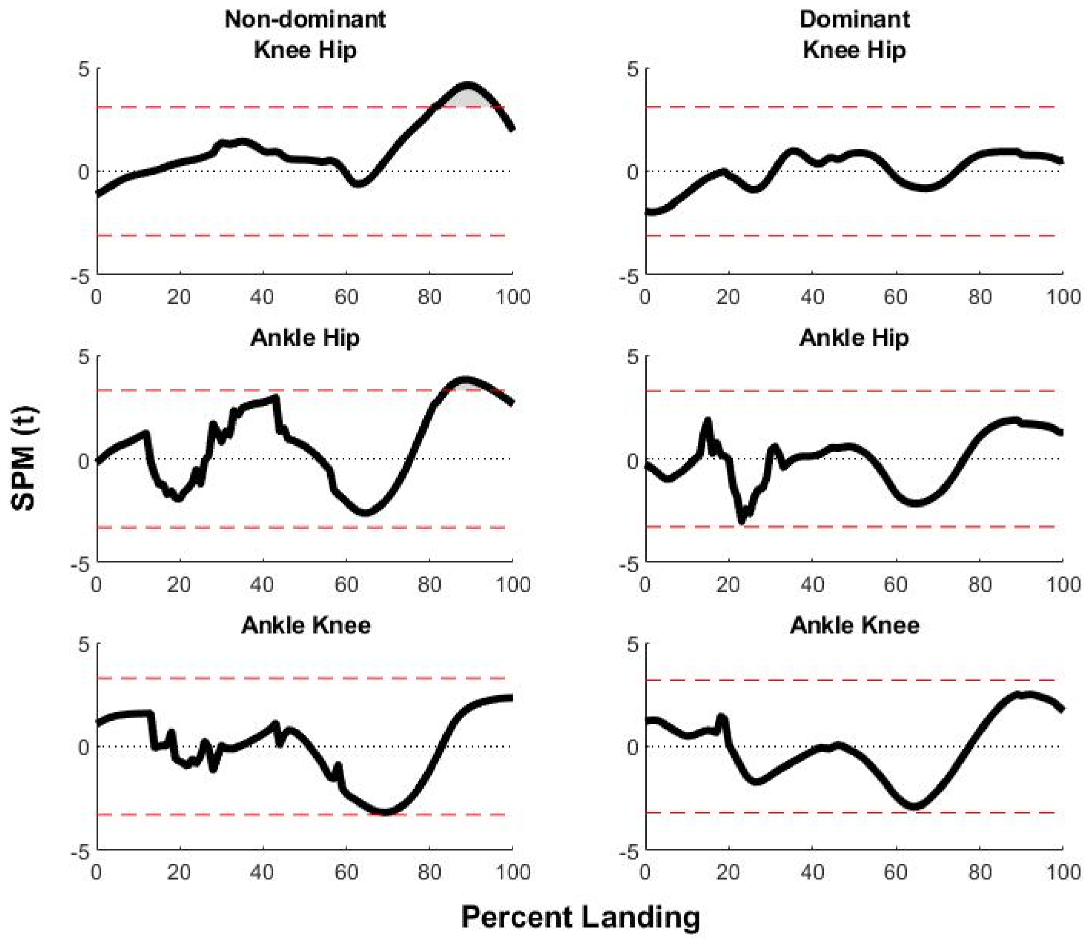

Figure 14. SPM of intralimb kinematic CRP.

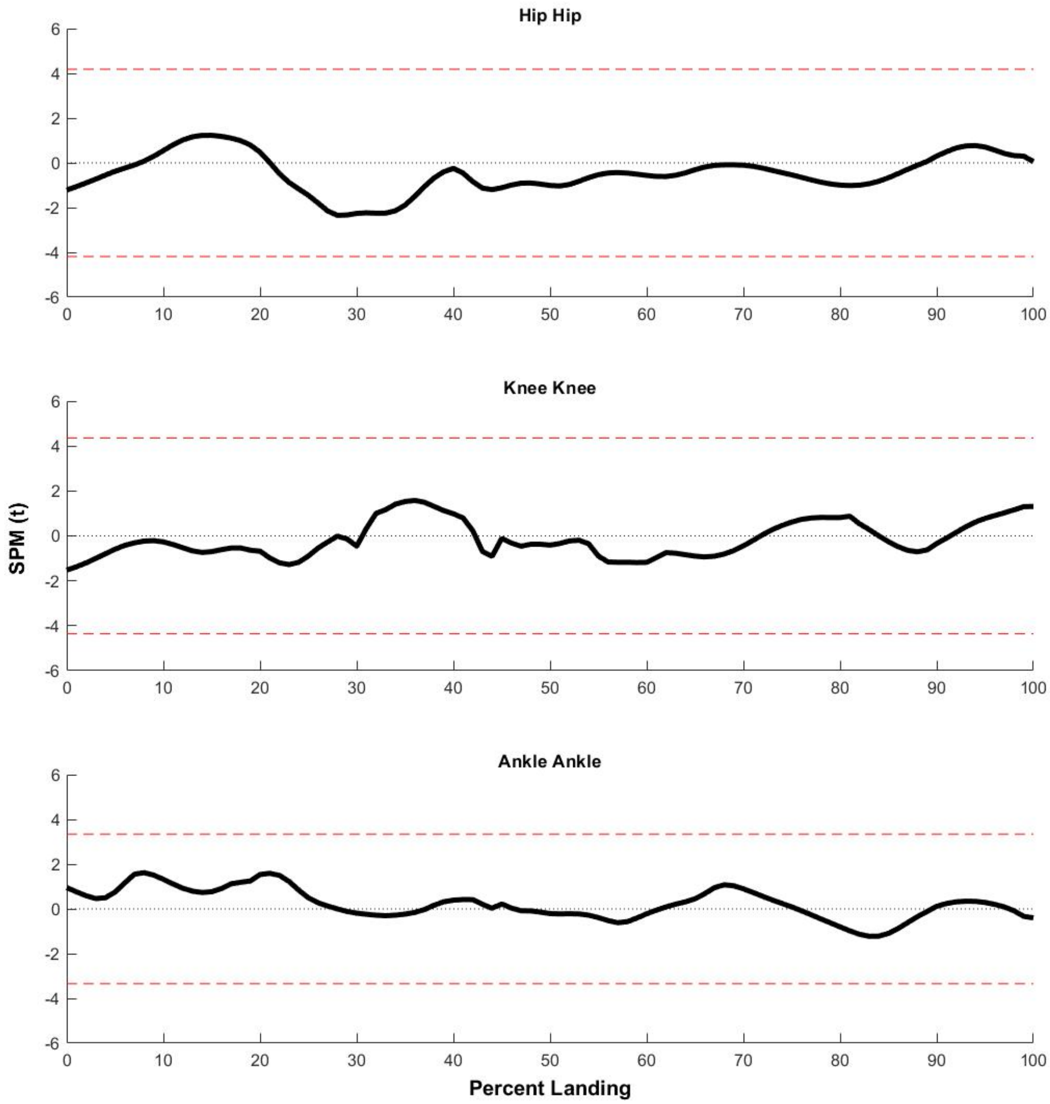

Figure 15. SPM of interlimb kinematic coordination variability.

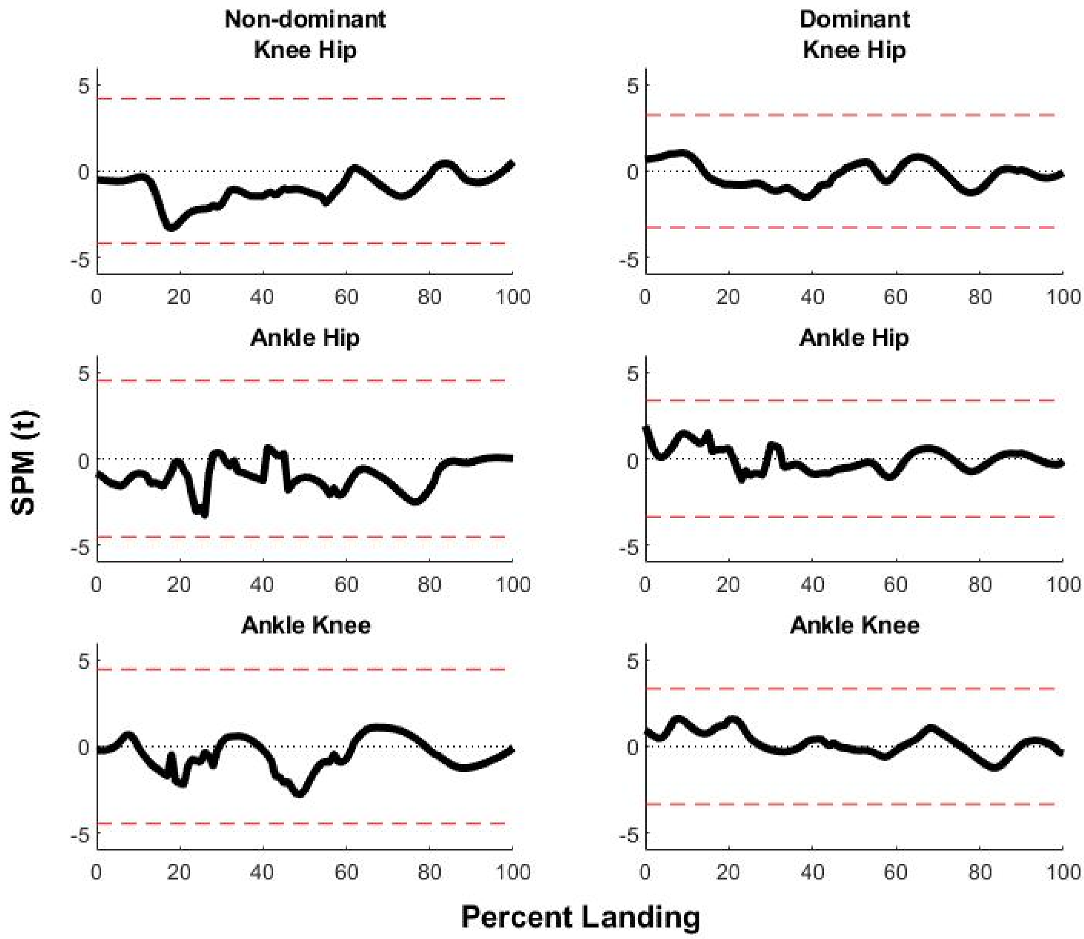

Figure 16. SPM of intralimb kinematic coordination variability.

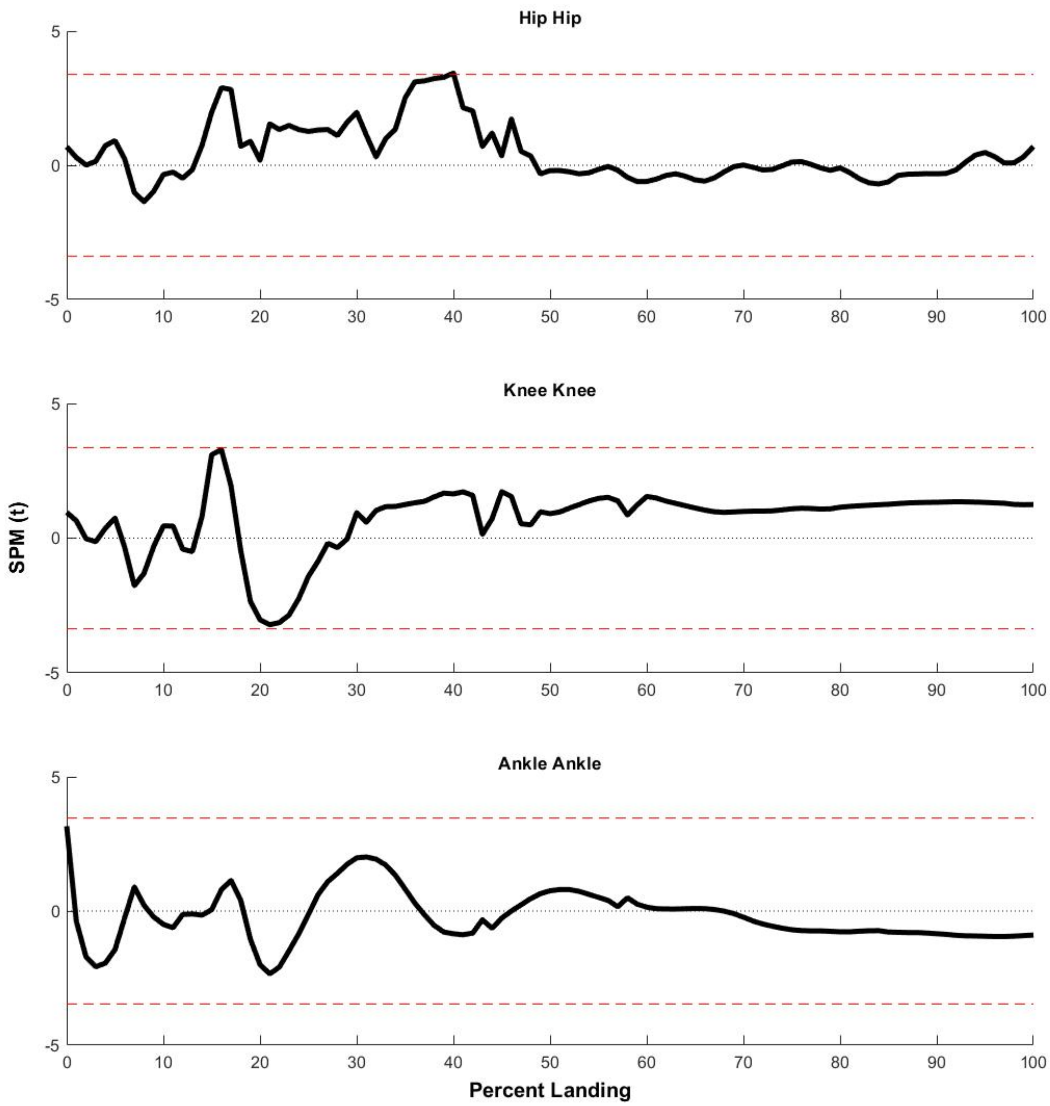

Figure 17. SPM of kinetic interlimb coordination.

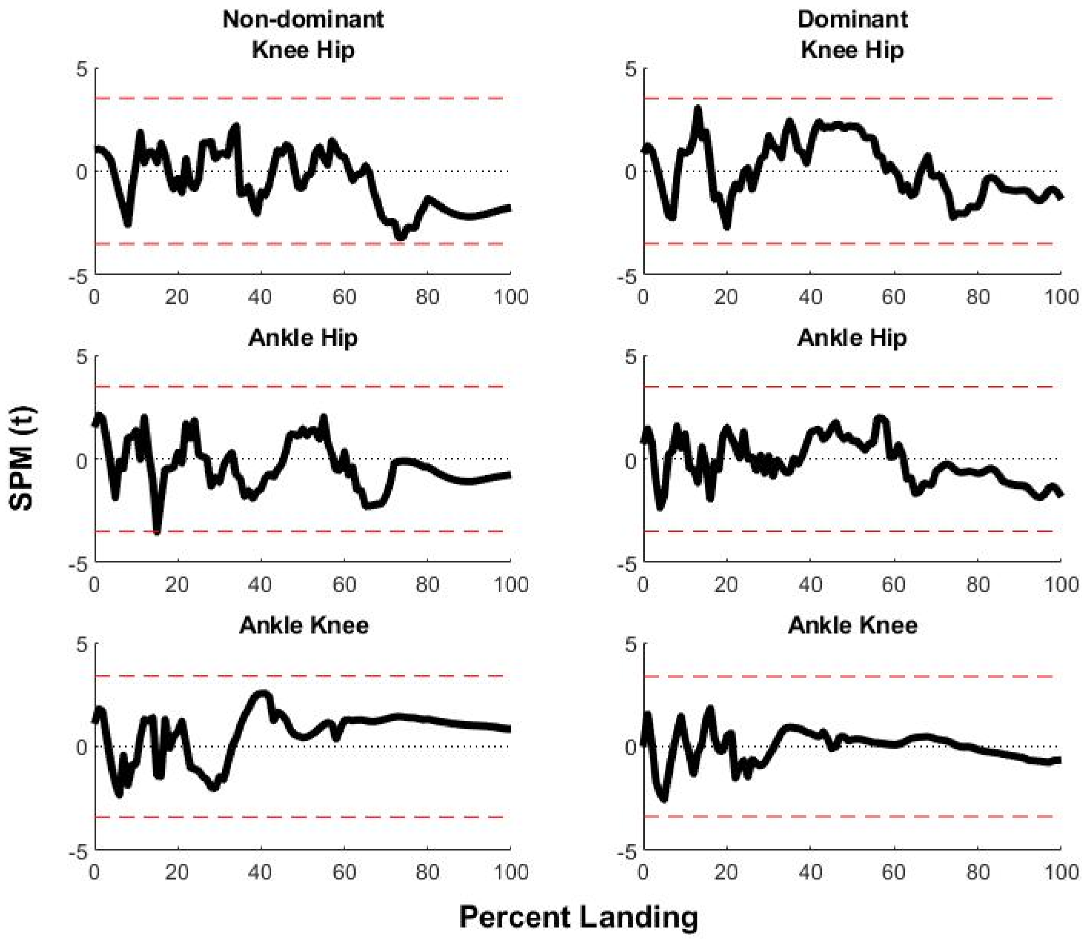

Figure 18. SPM of kinetic intralimb coordination.

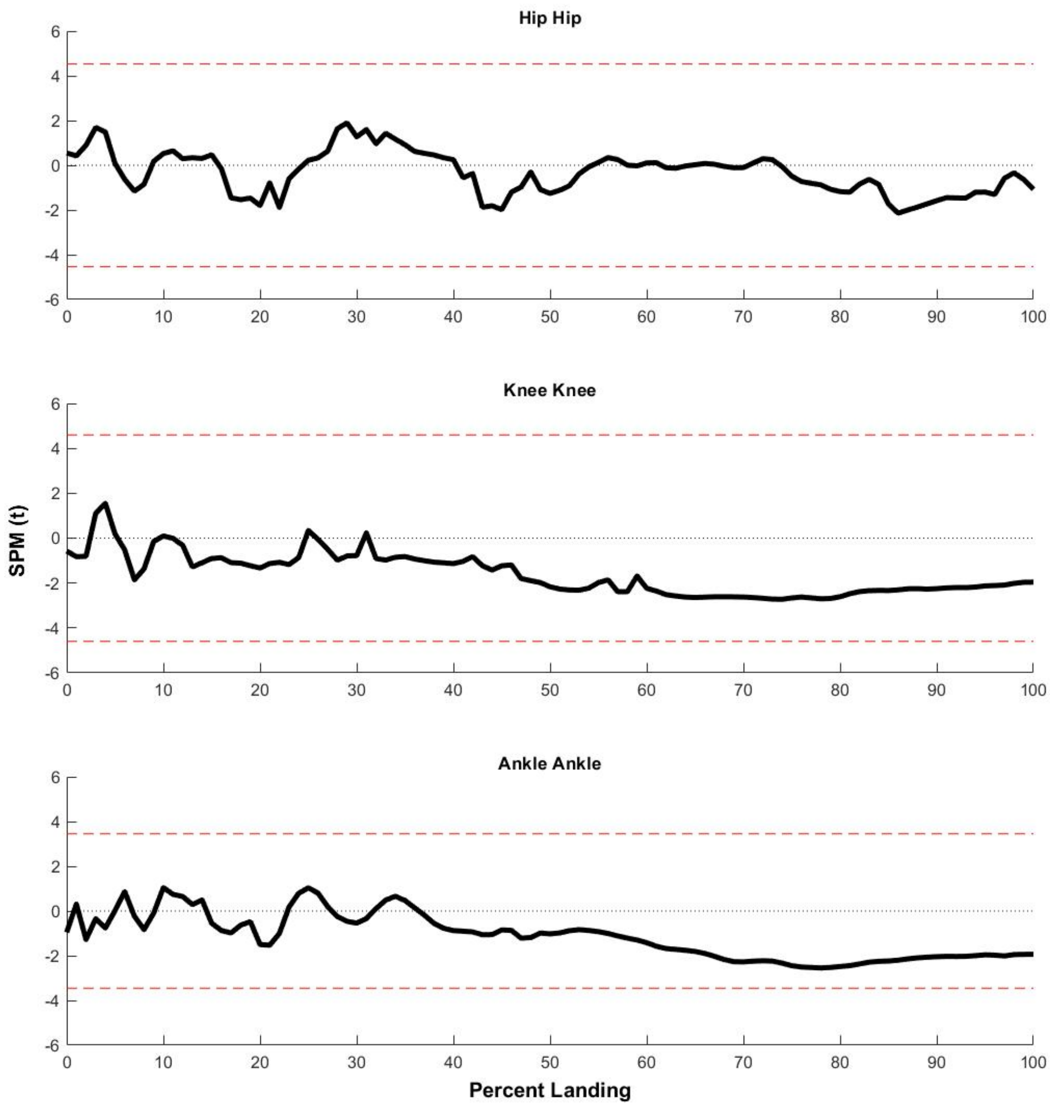

Figure 19. SPM of kinetic interlimb coordination variability.

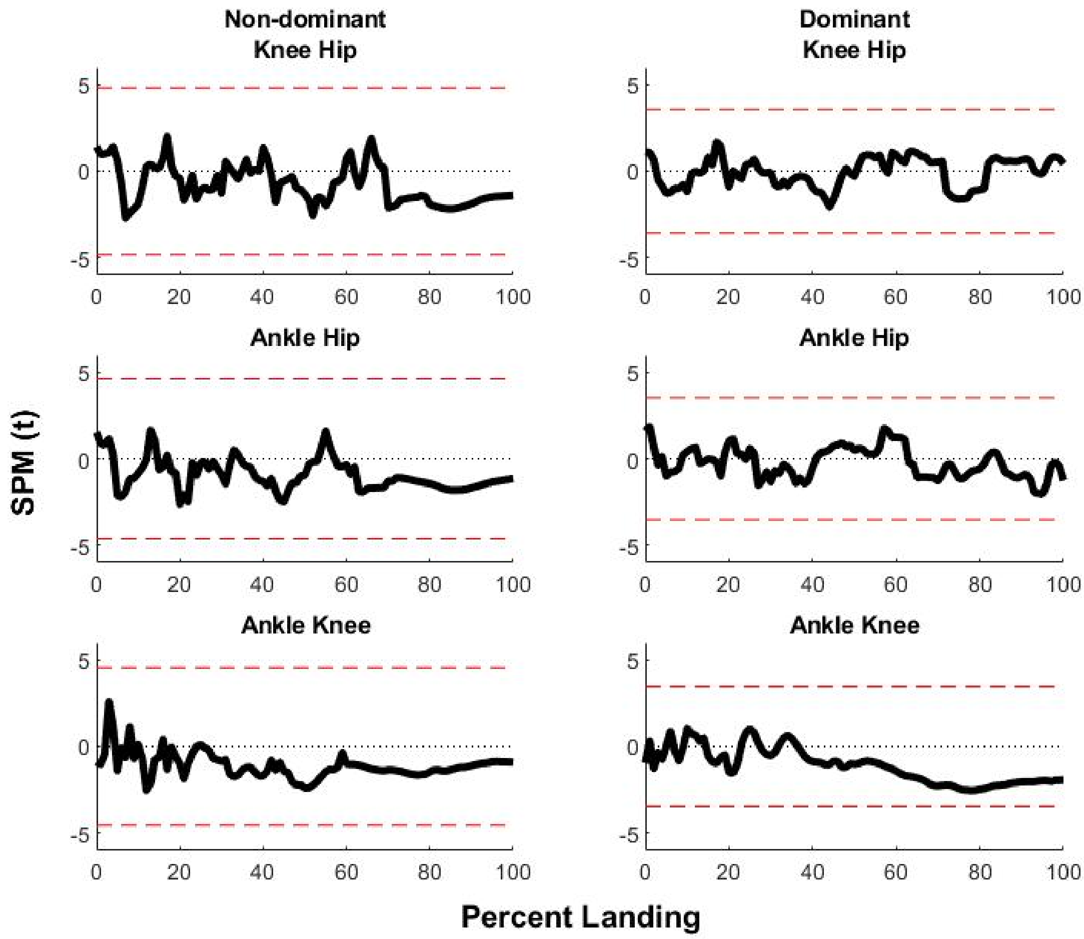

Figure 20. SPM of kinetic intralimb coordination variability.

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Biographical Sketch

Eric Jenkins earned his Bachelor of Science degree from West Virginia University in Sport and Exercise Psychology in 2013. Following his completion of a B.S., Eric attained a Master of Science at The University of Texas at Arlington in Exercise Physiology in 2016. Eric currently works for Motek | DIH Technology Inc. as a Clinical and Scientific Affairs Manager for the North American and Aisa Pacific regions. At the time of this dissertation submission, Eric has attained one peer-reviewed publication.

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