Effects of Running Speed, Fatigue, and Bracing on Motor Control of Chronically Unstable Ankles

Courtney Ann Webster

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Maury A. Nussbaum, Chair
Michael L. Madigan
Michael J. Agnew
Hampton C. Gabler
Stefan Duma

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ABSTRACT

Ankle sprains are among the most common injuries for participants in running and jumping sports. Following an initial sprain injury, many (30-40%) will develop *chronic ankle instability* (CAI), characterized by a perception of instability and repeated sprain injuries. Quasi-static test methods indicate poor postural stability and joint position sense (JPS) as associated motor control deficits. Little research, though, has investigated ankle motor control under dynamic (simulated sport) or fatigue conditions. To better understand factors contributing to the increased sprain rate in adults with CAI, three studies were completed investigating the roles of running speed, fatigue, and ankle bracing on motor control in adults with CAI.

First, two groups with and without ankle instability performed dynamic athletic maneuvers at each of two running speeds. Joint kinematics and kinetics were measured to identify differences in motor control strategies. Participants also completed two quasi-static tests (JPS and single leg drop landings). The level of correspondence between quasi-static and dynamic test methods was of particular interest. A second study compared fatigue development and fatigue adaptations when executing single leg drop landings. Strength loss and ratings of perceived exertion measured fatigue development, and joint kinematics, kinetics, and muscle activation quantified drop landing performance. A final study examined whether ankle braces, a common treatment for ankle sprains, retained their effectiveness when an athlete was fatigued. JPS and ankle stiffness were measured before and after a fatigue protocol while using each of three brace conditions.

Overall, results indicated that adults with CAI exhibit distinct adaptations to changes in speed and to fatigue that may increase their risk for ankle reinjury. Specific changes, however, depended on the particular activity being performed. Single leg drop landing kinematics may be a good representation of kinematics during dynamic athletic performance. Neither test brace improved JPS following fatigue, but each may be effective in providing mechanical stiffness compared to an unbraced condition. The effectiveness of a particular test brace, however, may be gender-specific. Future work should focus on identifying the benefits of different braces under broader conditions to help inform brace selection.

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ABBREVIATIONS

AC	Aircast brace condition
ASIS	Anterior superior iliac spine
AW	Ankle wrap brace condition
CAI	Chronic ankle instability
CAIT	Cumberland Ankle Instability Tool
COP	Center of pressure
EMG	Electromyography
FAI	Functional ankle instability
GS	Gastrocnemius
JPS	Joint position sense
LMF	Localized muscle fatigue
MAI	Mechanical ankle instability
MVIC	Maximal voluntary isometric contraction
NB	No brace condition
PB	Peroneus brevis
PL	Peroneus longus
ROM	Range of motion
RPE	Rating of perceived exertion
TA	Tibialis anterior

Chapter 1

Background and Motivation

1.1 Prevalence of Ankle Sprain Injuries and Chronic Ankle Instability

The term *ankle sprain* describes an injury in which the ankle is suddenly and forcibly twisted (rotated), rolled (inversion/eversion) or dorsiflexed resulting in torn or damaged ligaments of the ankle. Recent studies indicate ankle sprain incidence rates of 2 - 6 injuries per 1000 person-years (Bridgman et al., 2003; Waterman et al., 2010) translating to approximately 600,000 to 2 million sprains per year in the US (Soboroff et al., 1984; Waterman et al., 2010). Nearly 50% of all ankle sprains occur during athletic activity (Waterman et al., 2010), and sprains are among the most common injuries to athletes in running and jumping sports (Garrick, 1977; Yeung et al., 1994; Hootman et al., 2007; Waterman, 2010).

The majority (~ 85%) of all ankle sprains are inversion injuries, and roughly 30-40% of people who sustain an initial sprain injury will develop chronic ankle instability (CAI: Garrick, 1977; Mack, 1982; Schaap et al., 1989; Peters et al., 1991). Freeman (1965) was the first to characterize this condition as a history of repeated ankle sprains to the same ankle and anecdotal reports of the ankle "giving way" during activity. These repeated sprains are thought to be a result of either mechanical or functional instability resulting from trauma to the lateral ligaments of the ankle during an initial sprain injury. Mechanical ankle instability (MAI) describes stretch, laxity, or a tear in the tissues or ligaments responsible for stabilization of the ankle (Hertel et al., 2002; Tropp, 2002; Bonnel et al., 2010). This diagnosis can be made through a physical examination of the ankle range of motion (ROM), using the anterior drawer and talar tilt tests or using stress radiography. Functional ankle instability (FAI), in contrast, describes proprioceptive deficits resulting from damage to receptors in the ligaments and joint capsule (Hertel et al., 2002; Tropp, 2002; Bonnel et al., 2010). Whether these two conditions can occur independently is debatable (Brown et al., 2008; Hubbard et al., 2007). Identification of CAI for research typically involves collecting ankle sprain history, confirming a self-reported perception of instability, and using survey instruments such as the Cumberland Ankle Instability Tool (CAIT), which has shown good reliability in identifying adults with ankle instability (Delahunt et al., 2010).

1.2 Costs of Repetitive Ankle Sprain Injuries

Recurrent ankle sprains are not only painful, but they can also incur substantial costs for medical treatment, rehabilitation, and supplies for sprain management. Athletic tape and ankle braces are common methods used for sprain management, and which help to limit ankle motion during recovery which may help prevent future injuries (Eils et al., 2002). The cost of taping the ankles of a single collegiate athlete over the course of one football season is roughly $400 (Rovere et al., 1988), and the cost of taping for the entire team is about $16,000 (Burks et al., 1991). The annual cost of treating these injuries varies by country but may range from to $26 million to $2 billion (Soboroff et al., 1984; ACC, 1996; Hupperets et al., 2009).

Perhaps more damaging to an athlete than the financial burden, however, is time lost from physical activity and the psychological effects of the injury. For example, ankle injuries account for 20 – 25% of time-loss injuries in running and jumping sports (Mack, 1982; ACC, 1996). A study of professional soccer players revealed that, over two seasons, ankle sprains resulted in players missing more than 12,000 days of play and over 2,000 matches (Woods et al., 2003). Time away from activity interferes with physical training and causes decrements in skilled performance, while the fear of re-injury and sense of decreased motor abilities can make the return to sports frustrating and stressful (Hardy, 1992; Vela and Denegar, 2010). For those with ankle instability, this time loss may be experienced after each sprain incident, and are often of a greater severity than the initial injury (Hawkins et al., 2001). Additionally, repetitive sprains may lead to the development of lateral posttraumatic ankle osteoarthritis and articular degeneration later in life (Harrington, 1979; Valderrabano et al., 2006).

1.3 Motor Control Deficits and Chronic Ankle Instability

An ankle is most stable when in the so-called “close-packed position”, defined as weight bearing in full dorsiflexion with a flexed knee (Alter, 1996). The mechanics of an ankle sprain typically involve excessive plantarflexion or ankle inversion at the instant of ground contact (Wright et al., 2000; Konradsen et al., 2000), coupled with an impact force creating an inversion moment about the ankle (Konradsen et al., 2000). Thus, ankle orientation and ground reaction forces substantially affect whether a particular ground contact event will result in a sprain injury. Avoiding ankle sprains requires recognition of improper ankle positioning and the rapid development of adequate muscle force to resist the inversion motion (Robbins & Waked, 1998). To understand why ground contact events more often lead to ankle sprains for adults with CAI, researchers have focused on identifying specific motor control deficits that may hinder the ability to detect or resist ankle inversion.

Existing efforts to study CAI can be divided into two categories. The first is “quasi-static”, which is used here to describe studies employing either non-weight bearing or stationary protocols to evaluate ankle control. For example, several studies have reported greater postural sway among adults with ankle instability (Cornwall & Murrell, 1991; Goldie et al., 1994; Docherty et al., 2006; Brown et al., 2007 & 2010). One study even reported significant differences in postural stability control between the two ankles of the same participant, in which one ankle was frequently sprained and the other was uninjured (Gribble et al., 2004). Those with CAI also exhibit poorer joint position and movement sense (Ryan, 1994; Lentell et al., 1995; Jerosch & Bischof, 1996; Refshauge et al., 2000; Hartsell, 2000; Willems et al., 2002). Other efforts to quantify motor deficits with CAI have included measuring muscle reaction times to inversion perturbations (Konradsen & Ravn, 1991; Lofvenberg et al., 1995; Ebig et al., 1997; Vaes et al., 2002; Eechaute et al., 2009) and comparing ankle strength measures between injured and uninjured ankles (Birmingham et al., 1997; McKnight & Armstrong, 1997; Wilkerson et al., 1997; Konradsen et al., 1998; Kaminski et al., 1999 & 2003; Munn et al., 2003; Lin et al., 2009). Both approaches, though, have produced mixed results. The second category of research on CAI can be referred to as “functional evaluation”, where the primary focus is to determine how ankle instability or its treatment (e.g. ankle taping or bracing) affects athletic performance. Examples of such functional evaluations include shuttle runs, single-leg hopping, or the Star Excursion Balance Test (Gribble et al., 2004; Docherty et al., 2005; Monaghan et al., 2006; Hosseinimehr et al., 2010). Generally, measures of interest are performance times, jump distances, and reach distances, respectively.

While these quasi-static and functional evaluation techniques have revealed important differences between stable and unstable ankles, the degree to which they represent dynamic ankle control is not clear. Despite the frequency with which ankle sprains occur during sport, relatively few studies have quantified ankle control for adults with CAI under dynamic athletic conditions (i.e., those conditions causing the majority of ankle sprains). Drop and jump landings have been used to quantify specific differences in kinetics, kinematics, and muscle activation during rapid weight bearing and subsequent ankle stabilization (Riemann et al., 2002; Delahunt et al., 2009). Additionally, gait studies have revealed that participants with CAI demonstrate greater ankle inversion surrounding the instant of heel strike (Monaghan et al., 2006) and different gait termination strategies (Wikstrom et al., 2010). During athletic activity, however, there is an added component of running speed that likely influences how a task is executed, and kinetics and kinematics do change with speed (Novacheck, 1998). As mentioned, ankle joint positions/orientations (kinematics) and ground reaction forces (kinetics) play an important role in sprain injuries. Thus, quantifying ankle kinetic and kinematics during simulated athletic maneuvers at different running speeds is expected to help identify biomechanical differences in ankle stabilization related to CAI and improve our understanding of why those with CAI are so susceptible to repeated sprain injuries. Additionally, comparisons between dynamic ankle measures and measures of quasi-static or functional evaluation tests can aid in understanding how well these tests relate to ankle control during dynamic sport performance.

1.4 Fatigue and Motor Control Deficits with Chronic Ankle Instability

The majority of sports injuries are reported to occur during the latter half of a period of play or during the latter half of an athletic contest (Pinto et al., 1999; Gabbett, 2000). A study specific to ankle sprains, in professional soccer, found that nearly half of all sprains sustained over two seasons had occurred during the last third of each half (Woods et al., 2003). Although fatigue was not explicitly measured in these studies, it can be assumed that fatigue played an important role in these injuries.

A number of fatigue-induced changes in biomechanical and neuromuscular control have been identified among healthy ankles (Rozzi et al., 2000; Forestier et al., 2002; Gutierrez et al., 2007), but fatigue has not been well studied as a factor in recurrent sprain injuries for unstable ankles. Resistance to sprain is provided by a combination of reflexive and voluntary muscular force development and has some dependence on the direction and rate of an impact force (Konradsen et al., 2000). It follows, then, that conditions that influence joint positioning, ground reaction force characteristics, or reflex responses could affect sprain reinjury risk. Fatigue has been shown to influence each of these factors in healthy ankles.

In healthy ankles, both localized muscle fatigue (LMF) and whole-body fatigue lead to declines in passive and active joint position sense (Forestier et al., 2002; Mohammadi & Roozdar, 2010). This decline thought to occur due to a fatigue-induced increase in the threshold for muscle spindle firing (Rozzi et al., 2000). Considering that unstable ankles already demonstrate deficits in joint position sense (Jerosch & Bischof, 1996; Hartsell, 2000), the potential for fatigue to exacerbate sprain reinjury risk is apparent. Given its relevance to sprain occurrence, research is needed exploring the effects of fatigue on landing kinematics among adults with CAI.

Other fatigue-induced changes found in healthy ankles include decreased voluntary force generation. LMF of each of the primary ankle muscle groups (dorsiflexors, plantarflexors, invertors, and evertors) resulted in decreased voluntary torque generation capacity at the ankle and decreases in EMG amplitudes and median frequencies of the peroneus longus (PL), tibialis anterior (TA), and gastrocnemius (GS) muscles (Gutierrez et al., 2007). Thus, LMF at the ankle may leave a person more susceptible to ankle sprain due to diminished reactive ankle torque capacity. Further, reduced EMG amplitudes suggest that muscle pre-activation levels, which enable reactivity at ground contact (Konradsen et al., 2005), are also likely to be reduced. One study has also suggested that reflex amplitudes of PL and peroneus brevis (PB) – primary ankle evertors – also decrease following LMF (Jackson et al., 2009). Other studies, though, have reported no consistent change in reflex amplitude with LMF, but instead suggested that they may be gender-dependent (Wilson & Madigan, 2007).

As mentioned above, joint position sense is impaired in adults with CAI (Jerosch & Bischof, 1996; Hartsell, 2000). Therefore, fatigue is expected to result in further declines of motor control in this group. Using a functional reach test, whole-body fatigue caused a decrease in single-leg reach distances for both control and CAI groups with significant between-group differences for the lateral and anterolateral reach directions (Hosseinimehr et al., 2010). This indicated a loss of postural control for both groups following fatigue, but the CAI group exhibited greater losses than controls when challenged to reach in the lateral and anterolateral directions. Similar results were reported by Gribble et al. (2004), who found that LMF amplified the differences between the CAI ankles and their own uninjured contralateral ankles as well as between the CAI and control groups. Given that fatigue (whether whole-body or LMF) induces changes in neuromuscular and biomechanical control in stable ankles and that pre-existing deficiencies are present among adults with CAI, additional research on the contribution of fatigue to ankle sprains is needed. Such work could clarify whether fatigue poses a greater challenge to ankle stability for adults with CAI than those with healthy ankles.

1.5 Influence of Fatigue and Bracing on Sprain Recurrence

While fatigue is a factor inherent to many sports that may affect ankle stability for all athletes, those with unstable ankles may be additionally affected by the use of an ankle brace. Taping and bracing are the most common treatments for unstable ankles, and considerable research has explored their effects on walking, running, and landing mechanics (Mundermann et al., 2003; Riemann et al., 2002; Cordova et al., 2010; Delahunt et al., 2009; Simpson et al., 1999;

Spaulding et al., 2003), postural stability (Rougier et al., 2004; Baier & Hopf et al., 1998), inversion perturbation responses (Cordova et al., 2007; Eils et al., 2002; Vaes et al., 2002), joint position sense (Feuerbach et al., 1994; Hartsell et al., 2000), and functional athletic performance (Gross et al., 1997; Hals et al., 2000; MacKean et al., 1995; Verbrugge et al., 1996). Generally, ankle braces are thought to be effective by increasing ankle joint stiffness (Cordova et al., 2007; Zinder et al., 2009) and improving joint proprioception (Feuerbach et al., 1994; Heit et al., 1996; Hartsell, 2000). Improvements may vary, however, based on the specific brace type.

Recalling the effects of fatigue discussed above, both LMF and whole-body fatigue decrease joint position sense (Allen et al., 2010; Carpenter et al., 1998; Forestier et al., 2002, Mohammadi & Roozdar, 2010). Therefore, as an athlete exercises with ankle support, the effects of ankle bracing and fatigue on joint proprioception would appear be offset each other. On the other hand, effects on ankle joint stiffness are more difficult to predict. One study has suggested that fatigue increases antagonistic co-contraction (Gregory et al., 1998), while others, in contrast, indicate that fatigue reduces ankle joint stiffness due to temporary muscle damage impairing cross-bridge formation (Hakkinen, 1983; Kuitunen et al., 2002). With regard to ankle joint stiffness, it would appear that bracing and fatigue may have either complimentary or opposing effects.

1.6 Present Research Goals

The present research aims to contribute new evidence regarding the influences of running speed, LMF, and ankle bracing on ankle sprain occurrence among adults with CAI. These areas are addressed in a series of three experiments described in Chapters 2-4 of this document. The overall objectives of this research were to: 1) determine how running speed affects ankle stabilization during athletic performance and evaluate quasi-static measures of ankle control as predictors of sprain reinjury risk during dynamic activity; 2) determine if LMF occurs more quickly or is more detrimental to motor control for adults with CAI compared to healthy controls; and 3) determine the mechanisms by which different types of braces provide ankle support to an active athlete under fatigued and non-fatigued conditions.

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Chapter 2

Analysis of Athletic Maneuver Performance for Adults with and without Chronic Ankle Instability and the Relationship between Dynamic and Quasi-static Ankle Measures

2.1 Abstract

Ankle sprains are a common injury in sports and can lead to a condition of chronic ankle instability (CAI) that is characterized by repeated sprains to the affected ankle. Identifying the causes of these recurrent sprains has been a subject of interest in sports injury literature, and much of the existing research has evaluated ankle motor function using quasi-static (non-weight bearing or stationary) test methods including joint position sense (JPS) and single-leg drop landing tests. The first goal of this study was to determine if adults with ankle instability exhibit differences in kinematics and kinetics compared to controls during several dynamic activities; such differences may help to explain the propensity of the former for repeated ankle injury. Dynamic maneuvers were performed at each of two running speeds to also determine whether the two groups exhibited similar speed-dependent changes in kinematics and kinetics. Analysis focused on the instant of ground contact and landing phase. Overall, the results indicated that the two groups have different methods for adapting to changes in running speed. Specifically, adults with CAI adapted to changes in running speed by increasing knee flexion and/or exhibiting greater peak frontal plane joint angles and moments. By contrast, the control group showed changes in transverse plane joint angles and moments as speed increased. A greater reliance on frontal plane exertions for dynamic control may contribute to an increased sprain reinjury risk among adults with CAI. As a second goal of this study, select kinematic measures from dynamic maneuvers were correlated with kinematic measures from quasi-static ankle evaluations. JPS error measures were not significantly correlated with GC error measures during dynamic maneuvers. JPS error measures may be a means of identifying motor deficits, but the magnitude of the errors appear to be unrelated to joint positioning errors during dynamic activities. Measures of ankle-ground contact angles during drop landings, however, had moderate to large correlations with ground contact angles during several dynamic maneuvers, particularly cut step maneuvers. When evaluating differences in ankle control, drop landings may provide a simpler, more controlled method of assessing kinematic measures that still relate well to ankle positioning during dynamic athletic activity.

2.2 Introduction

Ankle sprains account for roughly half of all sport injuries (Waterman et al., 2010) and are the most common injury in running and jumping sports (Garrick, 1977; Yeung et al., 1994; Hootman et al., 2007; Waterman et al., 2010). Sprains are often the result of increased ankle plantarflexion, inversion, or internal rotation (Garrick, 1977; Wright et al., 2000; Konradsen et al., 2000) at the instant of ground contact, coupled with a sudden impact force creating an inversion moment about the ankle (Konradsen et al., 2000). Roughly 30-40% of those who sustain an initial sprain injury develop chronic ankle instability (CAI; Garrick, 1977; Mack, 1982; Schaap et al., 1989; Peters et al., 1991), which is characterized by repeated sprains and sensations of instability or anecdotal reports of the ankle "giving way" during activity (Freeman, 1965).

To examine the causes of recurrent sprains, prior research has focused on identifying motor control deficits associated with CAI. Common test methods include static and dynamic postural stability assessments (Tropp et al., 1984; Cornwall & Murrell, 1991; Goldie et al., 1994; Bernier et al., 1997; Mitchell et al., 2008), muscle reaction times to inversion perturbations (Konradsen & Ravn, 1991; Lofvenberg et al., 1995; Ebig et al., 1997; Vaes et al., 2002; Eechaute et al., 2009), strength measures (Birmingham et al., 1997; McKnight & Armstrong, 1997; Wilkerson et al., 1997; Konradsen et al., 1998; Kaminski et al., 1999 & 2003; Munn et al., 2003; Lin et al., 2009), and measures of joint position and movement sense (Jerosch & Bischof, 1996; Lentell et al., 1995; Bernier et al., 1998; Refshauge et al., 2000; Willems et al., 2002). Instability is often attributed to physical and neural damage to the lateral ligaments of the ankle, leading to decreased proprioception and increased mechanical laxity (Hertel, 2002; Tropp, 2002; Bonnel et al., 2010). Several limitations have been identified among those with CAI, including greater postural sway (Cornwall & Murrell, 1991; Goldie et al., 1994; Docherty et al., 2006; Brown et al., 2007 & 2010) and poorer joint position and movement sense (Ryan, 1994; Lentell et al., 1995; Jerosch & Bischof, 1996; Refshauge et al., 2000; Hartsell, 2000; Willems et al., 2002). Adults with CAI also demonstrate functional deficits, as evidenced by reduced reach distances (Gribble et al., 2004) or poorer performance during single-leg hopping tasks (Jerosch & Bischof, 1996; Docherty et al., 2005).

While such studies have identified clear deficits in the CAI population, many of the test methods are "quasi-static" in nature. That is, participants were either seated (non-weight bearing) or standing in place. However, the relationship between quasi-static motor control deficits and sprain occurrence during athletic activity is not clear. Landing kinetics and kinematics (e.g.

ankle angle and vertical ground reaction force) are critical to determining whether a particular ground contact event will result in an ankle sprain (Wright et al., 2000; Konradsen et al., 2000). Research on stable ankles has revealed that lower body joint kinetics and kinematics are affected by running speed (Novacheck et al., 1998; Schache et al., 2011). Yet, only limited work has investigated unstable ankle control during walking or running or during simulated sporting tasks. These revealed that participants with CAI have greater ankle inversion near the instant of heel strike during walking (Monaghan et al., 2006), different gait termination strategies versus those with healthy ankles (Wikstrom et al., 2010), and less dorsiflexion during jogging (Drewes et al., 2009). Under more dynamic conditions, one study reported that unstable ankles experience greater vertical ground reaction force peak during "V"-style cutting maneuvers (Dayakidis & Boudolos, 2006). Some research groups have also used drop or jump landings to study ankle control under more realistic conditions (Riemann et al., 2002; Delahunt et al., 2009), and these landings involve rapid weight-bearing and balance control and are thus likely to be a good simulation of ground contact during sport. During athletic activity, however, there is an added component of speed that presumably influences the difficulty of ankle stabilization.

There were two primary goals of the present research. The first was to evaluate ankle and knee mechanics during simulated athletic maneuvers. We hypothesized that groups comprised of stable and unstable ankles would exhibit differences in joint kinematics/kinetics that may help explain recurrent sprain injuries within the unstable ankle group. An additional hypothesis was that landing mechanics would be dependent on running speed. We had participants perform common athletic maneuvers at each of two speeds. Joint kinetics and kinematics were compared within and between groups to determine whether speed-dependent changes were consistent across groups. The second goal was to assess the level of correspondence between common, quasi-static measures of motor control and measures obtained under more dynamic conditions. Quasi-static measures have already proven to be useful at identifying motor deficits in adults with CAI, and these types of measures are also more easily obtained. It is currently unknown, however, to what degree these quasi-static measures are representative of ankle control under dynamic conditions. Here, we determined the correlations between measures from quasi-static and dynamic ankle control. A finding of strong correlations between these would support the use of quasi-static assessments in quantifying motor deficits relevant to ankle injury occurring during dynamic athletic activities.

2.3 Methods

2.3.1 Participants and Overview of Experimental Design

Twenty-six participants (12 males, 14 females) aged 18 - 35 completed the study, and were assigned to either a control group (C) or an unstable ankle group (U) based on ankle sprain history and a score from the Cumberland Ankle Instability Tool (CAIT; Hiller et al., 2006). Both groups had six males and seven females. Those in the C group had a history of 0-1 ankle sprains and a CAIT score ≥ 28. Those in the U group had a history of at least two ankle inversion sprains to the same ankle, a CAIT score ≤ 25, and reported occasional sensations of ankle instability and several incidents of their ankle "giving way" during activity. Participants were matched, at the individual level, based on age (within 8 years), stature (within 13 cm), body mass (within 11.4 kg), and gender, and there were no significant differences in these between groups (Table 2.1). All participants were required to be active (exercising at least 2-3 times/week), and none reported any current or chronic musculoskeletal problems (other than ankle instability in the U group). Prior joint surgeries, current joint pain, neuropathies, muscle weakness of the lower limbs, and vestibular or balance disorders were exclusion criteria. In addition, participants were excluded if they were participating in any physical rehabilitation programs or regularly used ankle taping or braces. Verbal and written consent was provided by all participants, and all protocols were approved by the Virginia Tech Institutional Review Board.

Table 2.1. Mean (SD) participant characteristics for control and unstable ankle groups (n = 13 in each). *P* values are from unpaired *t* tests, and the symbol * indicates a significant ($p < 0.05$) difference between groups.

Measure	Control	Unstable	*p*
Age (yr)	23.8 (3.1)	22.4 (4.6)	0.38
Stature (cm)	173.0 (7.5)	174.6 (10.7)	0.67
Body Mass (kg)	72.3 (13.7)	72.5 (14.0)	0.97
CAIT Score	29.5 (0.8)	18.5 (3.7)	<0.0001*

Initially, a "test" leg was determined for each participant. For the U group, the test leg was the one with the lowest CAIT score. Leg dominance was also determined, by asking participants which leg they would use to kick a ball (Johnson & Johnson, 1993; Fernandes et al., 2000). Those in the C group were tested, in the experiment described below, using the leg with the same limb dominance (or non-dominance) as the test leg of their match in the U group. That is, if a U group participant was tested on their dominant leg, their matched pair was also tested using their dominant leg, and vice versa.

Three separate tests of ankle stability were completed by each participant, two of which were measures of quasi-static ankle stability (joint position sense and drop landing) and the third measured ankle stability during running and selected athletic maneuvers. Joint position sense was tested first, followed by the drop-landings, and running trials were completed last. This sequence was used for two primary reasons. First, it helped to mitigate any effects of fatigue on the initial, quasi-static, test measures. Second, both drop landings and running trials required surface markers for kinematic tracking, and completing these tests in adjacent order minimized errors due to marker movement or repositioning. To control for potential influences of foot wear, all procedures were performed using the same model of athletic shoes fitted to the nearest whole size.

2.3.2 Joint Position Sense (JPS)

A joint position sense (JPS) test was used to measure the accuracy of ankle repositioning. Testing was done using a commercial dynamometer (Biodex System 3, Biodex Medical Systems, Inc., Shirley, NY; Figure 2.1), and was done for two functional configurations, inversion/eversion (I/E) and plantarflexion/dorsiflexion (P/D), using the test leg and in a random order between participants. The dynamometer chair was reclined 20 degrees from the vertical, and participants were positioned with ~ 50 degrees of knee flexion. A foot plate was set such that ankle joint center-of-rotation was aligned with the dynamometer's axis of rotation. A support pad and Velcro™ strap were used to maintain the position of the thigh, and additional straps were used to hold the foot firmly to the foot plate. For P/D, a rigid counterweight was added to offset the weight of the foot and footplate, thereby ensuring that a similar effort level was used to complete both plantarflexion and dorsiflexion motions.

Figure 2.1. Experimental set up for testing I/E and P/D joint position sense (the former movement configuration is illustrated).

Range of motion (ROM) stops on the dynamometer were set to 20 degrees from the neutral position in both rotation directions, yielding a total ROM of 40 degrees. JPS trials were performed using a passive-active protocol. Initially, the participant's ankle was passively moved (at 2 degrees/sec) to a reference angle, then held there for ~ 10 seconds. Participants were then given active control with an isotonic load of 0.68 Nm. They were instructed to move their foot to the furthest ROM stop and then attempt to reposition their foot to the reference angle. Once they felt that the reference angle had been reached, they pressed a button to stop the dynamometer and indicate the end of the trial. Participants wore headphones and a blindfold to minimize audio and visual cues regarding their foot position. Prior to testing, participants completed extensive practice trials to acquaint them with these procedures. Five JPS trials were completed using each of four reference angles (±5 and ±15 degrees), and the order of the reference angles was randomized. For the P/D configuration, positive and negative reference angles respectively refer to dorsiflexion and plantarflexion displacements from the neutral position. For the I/E configuration, positive and negative reference angles refer to inversion and eversion angles, respectively. Dynamometer angles were sampled at 1024 Hz, and then low-pass filtered (2^{nd} order, bi-directional, Butterworth, cutoff frequency = 3 Hz).

Absolute and true errors (Jerosch & Bischof, 1996; Willems et al., 2002) were obtained as dependent measures. Absolute error is the absolute value of the difference between the reposition and reference angles, and provides a measure of the offset between reposition and reference angles. True error is the signed magnitude of the difference between reposition and reference angles. A positive true error indicates a tendency to "overshoot" the reference position, meaning the replication angle was further from the neutral position than the reference angle. Conversely, a negative true error represents an “undershoot” error, in which the replication angle was closer to the neutral position than the reference angle. For each participant, the four best attempts at each reference angle (i.e., the least absolute error) were included in statistical analyses (Hartsell, 2000).

2.3.3 Single Leg Drop Landings

Participants completed single-leg drop landing trials using the test leg. To track lower-extremity kinematics, reflective surface markers were fixed bilaterally over several anatomical landmarks: anterior superior iliac spine (ASIS), greater trochanter, lateral and medial femoral condyles, lateral and medial malleoli, calcaneus, and bases of the first and fifth metatarsals. Rigid marker clusters were also attached to the thigh, shank, and foot segments of the test leg. In each trial, participants stood on a 30.5 cm high platform and were instructed to suspend their test leg over the floor, place their hands on their waist, drop themselves down from the platform onto their test leg, and maintain unipedal balance for 10 seconds (Figure 2.2). Both forward-facing and side-facing drop landings were completed; in the latter, the participant began turned 90 degrees on the box and dropped down to their side, again landing on the test leg. Practice was given for both configurations, and then five replications of each were completed, with forward-facing drops completed first. Participants landed on a force platform (Model #K20102, Bertec Corp., Columbus, OH), from ground reaction forces were sampled at 2500 Hz. A 6-camera system (Vicon MX 1.7.1, Vicon, Oxford, UK) was used to synchronously capture surface marker positions at 500 Hz.

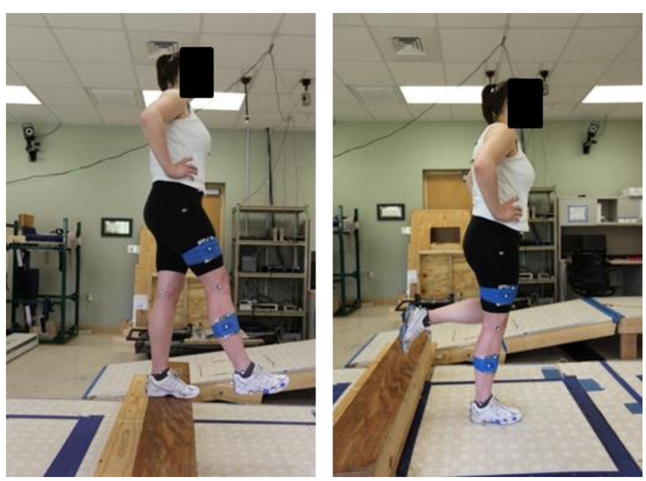

Figure 2.2. Forward drop landing. Participants dropped from a platform and landed on their “test” leg with their hands on their hips and contralateral knee flexed at ~ 90 deg.

Kinematic and kinetic data were low-pass filtered (Hz, 4th-order, bi-directional, Butterworth with cutoff of 20 and 50 Hz, respectively). The instant of ground contact (GC) was identified when the vertical component of the ground reaction force exceeded 10 N (Hreljac & Stergiou, 2000; Leitch et al., 2011). Dependent kinematic measures included sagittal, frontal, and transverse plane ankle and knee rotation angles at ground contact, and were calculated using the Euler angle approach (Hamill & Selbie, 2004). Ankle joint angles, moments, and power absorption were determined using 3D inverse dynamics (Winter 1990; Vaughan et al., 1999; Dumas et al., 2004). Subscripts are used to define the anatomical motion or plane to which the measures refer. Anatomical motions include flexion (FL), extension (EX), abduction (AB), adduction (AD), internal rotation (IR), external rotation (ER), dorsiflexion (DF), plantarflexion (PF), inversion (INV), and eversion (EV), and planes include sagittal (S), frontal (F), and transverse (T). Joint moments presented here refer to internal or reactive moments, and all reported kinetic values were normalized to individual body mass (Novacheck, 1998; Schache et al., 2011; Ishii et al., 2011). From the force platform, maximal mediolateral ground reaction force (ML GRF) was analyzed to consider the shear force acting about the inversion/eversion axis of the foot, and maximal lateral deviation of the center-of-pressure (COP_{LD}) were also derived. Note, ML forces and deviations were of interest, given their likely contribution to inversion ankle moments. COP_{LD} was determined as COP displacement relative to the foot midline as defined by the heel and toe markers (De Cock, 2008). Dependent measures analyzed were joint angles at GC and peak values of joint angles, moments, and power absorption during the landing phase, which was defined as the first 100 ms following ground contact (Decker et al., 2003). Dependent measures for the first three successful trials for each drop type were included in statistical analyses. A successful trial was defined as one in which the participant was able to execute the drop landing and maintain balance without touching down the other foot.

2.3.4 Dynamic Athletic Maneuvers

Participants completed a set of simulated athletic maneuvers while running on a 10-m walkway at each of two test speeds (2.5 m/s and 3.6 m/s). These speeds were chosen to be comparable to values used in other running experiments (Ferber et al., 2003; Drewes et al., 2009; Bischof et al., 2010; Whatman et al., 2011), and they represented two distinctly different levels of running effort that were attainable within lab space constraints. Running speed was controlled using a treadmill positioned beside the walkway. An elastic belt was stretched between the front drive axle of the treadmill and a pulley at the far end of the walkway. The treadmill motor was used to drive the belt at the target speed, and colored patches on the belt served as visual pacing cues.

Participants were given practice matching the belt speed, and the starting position was manipulated so that the test leg consistently landed on a force plate (Model #K20102, Bertec Corp., Columbus, OH) mounted centrally in the walkway. The specific athletic maneuvers were as follows:

- Running step (Run) – This involved a single footfall of the test leg on the force platform as the participant ran at the specified speed.
- Jump stop (JS) – The participant approached the force plate at the specified running speed, initiated a small jump forward starting ~1 m from the force platform, and landed bilaterally with only the test leg on the force plate.
- Cut step (Cut) – The participant ran down the walkway, stepped on the force platform with the test leg, and initiated a cut oriented 45 degrees from the original direction of motion. Tape was applied to the floor as a guide.
- Shuttle run (Shuttle) –The participant approached the force plate at the specified speed, stepped on the force platform with the test leg, then turned 180 degrees to run back toward the starting point.

The order of running speeds was randomized for each participant. Practice was given prior to data collection for each maneuver, to ensure that the participant understood and could safely perform each maneuver at each speed. Subsequently, five trials of each of the athletic maneuvers were completed. Running step trials were completed first out of convenience, and the order of the remaining maneuvers was randomized. Kinematics and ground reaction forces were recorded and filtered as described previously, and dependent measures were the same as for the drop landing tests (i.e., joint angles at GC and peak values of joint angles, moments and power absorption during the landing phase). The first three successful trials for each dynamic activity type were used for statistical analyses. Here, a trial was deemed successful if the participant matched the pace of the treadmill timing belt (via visual inspection), the test leg landed entirely on the force plate during maneuver execution, and no contact was made between the force plate and the contralateral leg. Proper foot placement was confirmed by observing each motion capture file to verify that the reflective markers of the foot landed within the bounds of the force plate and coincided with a single, continuous vertical GRF impulse while in contact with the platform.

2.3.5 Statistical Analyses

All statistical analyses were performed using JMP 10.0 software (SAS Institute Inc., Cary NC), using the restricted maximum likelihood (REML) approach, and with significance concluded when $p < 0.05$. Summary statistics are presented as means (SD). Separate mixed-factor analyses of variance (ANOVA) models were used for each protocol, all of which included group assignment as a factor. For JPS analysis, reference angle was included as an additional factor, and separate analyses were completed for each configuration (P/D and I/E). Drop landing analyses were also completed separately for each drop landing type (forward and side). Analyses for the dynamic protocols were completed separately for each of the four types of maneuvers (run, jump stop, cut step, shuttle run) using ANOVA models that had running speed as an additional factor. Where relevant, post-hoc paired comparisons were done using Tukey's Honestly Significant Difference (HSD) test and interaction effects were explored using simple effects analyses.

Correspondence between JPS error values and ankle GC angles during dynamic athletic maneuvers were assessed using bivariate coefficients of correlation (ρ). For these, mean ankle position was determined for each participant during the dynamic trials, and an absolute mean difference score was computed describing the participant's mean position deviation across trials. These mean difference scores were then compared with mean JPS error measures. Correlations (absolute value) were qualitatively interpreted (Cohen, 1988; Hopkins et al., 2009) as trivial (0.0-0.1), small (0.1-0.3), moderate (0.3-0.5), large (0.5-0.7), very large (0.7-0.9), or extremely large (0.9-1.0). A similar approach was used to compare measures between drop landings and the dynamic tasks. Bivariate correlations were also obtained between kinetic and kinematic measures from dynamic maneuvers and drop landings. For simplicity, only correlations between measures about the same anatomical axis are presented.

2.4 Results

Complete summary statistics and statistical results for each of the analyses are provided in Appendix A. Given the focus of this work, the presentation of results is limited primarily to main and interactive effects of group. Main effects of running speed are also noted, in cases where it affected landing mechanics that are related to sprain occurrence (specifically, frontal plane angles, moments, and power absorption).

2.4.1 Joint Position Sense and Single Leg Drop Landings

While ANOVA analyses revealed a significant main effect of reference angle, no significant main or interactive group effects were found for either absolute or true JPS error measures in either configuration ($p > 0.25$).

No group effects were significant during forward drop landings. In an effect that approached significance ($p = 0.089$), the U group had greater frontal plane ankle power absorption (8.3(11.1) W/kg) than the C group (3.2(2.6) W/kg). A significant group difference during side landings was found only for transverse plane power absorption at the knee ($F_{(1,21)} = 7.18$, $p = 0.01$), and which was 0.7(0.6) W/kg for the C group versus 0.2(0.3) W/kg for the U group. Also approaching significance were measures of transverse plane GC angle ($p = 0.09$) and peak ER angle ($p = 0.10$) during the landing phase of side drop landings. The C group had greater ankle IR (4.4 deg, ~68%) at GC than the U group, and in the landing phase the U group exhibited a small degree of ER whereas the C group did not. Instead, the C group retained a degree of ankle IR throughout landing phase only reducing its magnitude to about 1.2(3.8) deg.

2.4.2 Dynamic Athletic Maneuvers

<u>Run</u>

There was a significant group x speed interaction effect ($F_{(1,125)} = 4.24$, $p = 0.04$) on knee FL angle at GC. Specifically, at the faster running speed, knee FL at GC increased by 6.4 deg (~62%) in the U group, but increased by only 4.2° (~44%) in the C group. The U group also showed a small but significant increase ($F_{(1,125)} = 8.91$, $p = 0.003$) in peak knee FL angle (2.7°, ~6%) during the landing phase as speed increased. Group x speed interactions were also present for sagittal plane ankle joint moments ($F_{(1,128)} = 6.44$, $p = 0.01$). At the slower speed, the C group exhibited a peak ankle DF moment of 2.4 (1.8) Nm/kg which, although small in magnitude, was over twice as large as that of the U group whose peak reached 1.0 (1.5) Nm/kg. At the faster speed, group differences were smaller and non-significant, and the C and U groups had peak DF moments of 1.8 (2.0) Nm/kg and 1.4 (1.6) Nm/kg, respectively. There was also a significant interactive effect of group x speed on ML GRF ($F_{(1,127)} = 5.19$, $p = 0.02$). ML GRF increased for both groups as speed increased, but the change was larger (0.19 N/kg, ~22%) and significant for the U group.

Jump Stop

The group x speed interaction effect was significant ($F_{(1,122)} = 6.50$, $p = 0.01$) for transverse plane ankle angle at GC. As speed increased, the C group showed a 60% (2.1°) reduction in ankle IR at GC while the U group remained unchanged. During landing, the groups exhibited opposite trends with regard to peak knee FL angle ($F_{(1,118)} = 22.80$, $p < 0.0001$; Figure 2.3). In response to the faster running speed, the C group reduced peak knee FL by about 3.2° (~5%). Conversely, the U group increased their peak knee FL during landing by roughly the same magnitude (2.8°, ~5%). The U group was also found to have twice the knee P_T ($F_{(1,125)} = 5.06$, $p = 0.0001$) at the faster running speed (3.8 (3.6) W/kg) compared to the slower speed (1.8 (1.8) W/kg) while the C group showed a smaller, non-significant increase (0.6 W/kg, ~27%) in knee $P_{T.}$

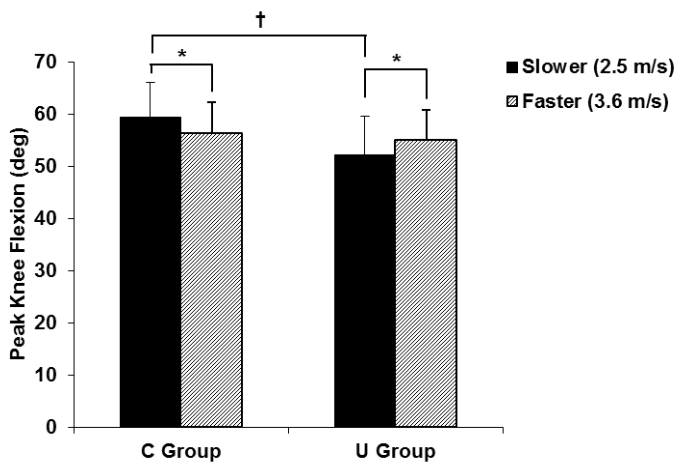

Figure 2.3. Group and speed effects on peak knee flexion angle during the landing phase of the jump stop maneuver. The symbols † and * indicate significant differences between and within groups, respectively ($p < 0.05$).

Cut Step

Several significant speed and group x speed interaction effects were found on measures during cut step maneuvers. Overall, an increase in running speed resulted in a 2.5-degree (~52%) increase ($F_{(1,122)} = 8.32$, $p = 0.005$) in ankle INV angle at GC. Peak ankle INV and ER angles during landing also increased significantly ($F_{(1,121)} = 19.68$, $p < 0.0001$; $F_{(1,121)} = 10.71$, $p = 0.001$, respectively) with running speed, by means of 3° (~22%) and 2.7° (~56%), respectively. Although the group x speed interaction effect on ankle INV only approached significance ($p = 0.08$), the relative increase with speed was greater for the C group compared to the U group (2.9°, ~91% vs. 2°, ~32%). However, the magnitude of ankle INV was still about 2° greater for the U group at either speed.

During the landing phase, several significant group x speed interactions were found. The U group had a significantly greater peak knee AB angle (Figure 2.4; $F_{(1,117)} = 7.60$, $p = 0.01$) by 2.5°(~24%) at the faster speed compared to the slower speed, whereas peak knee AB angle for the C group was unchanged across speed conditions. As speed increased, the C group had an increase in peak knee ER angle (3.2°, 26%; $F_{(1,117)} = 4.01$, $p = 0.047$), whereas the change in peak knee ER angle for the U group was not significant.

Regarding kinetic measures, significant group x speed interaction effects were found for measures of peak ankle EV moment ($F_{(1,121)} = 5.40$, $p = 0.02$) and peak knee AB ($F_{(1,117)} = 6.71$, $p = 0.01$) and IR ($F_{(1,117)} = 9.74$, $p = 0.002$) moments (Figure 2.5). At the slower speed, the C group exhibited a peak ankle EV stabilization moment three times the magnitude of the U group, at 1.2 (1.4) Nm/kg and 0.39 (0.76) Nm/kg, respectively. As speed increased, however, peak ankle EV for the C group remained nearly constant while for the U group it increased three-fold to a level that matched the magnitude of the C group. A similar group x speed interaction was observed for peak knee AB moment. At the faster running speed, the U group exhibited a significantly greater peak knee AB moment (0.88 Nm/kg, ~275%) than at the slower running speed, with a magnitude comparable to that of the C group. Conversely, the C group increased peak knee IR moment (0.6 Nm/kg, ~55%; $F_{(1,117)} = 9.74$, $p = 0.002$) as speed increased, and, at the faster speed, the two groups exhibited a similar peak IR moment of 1.7 Nm/kg.

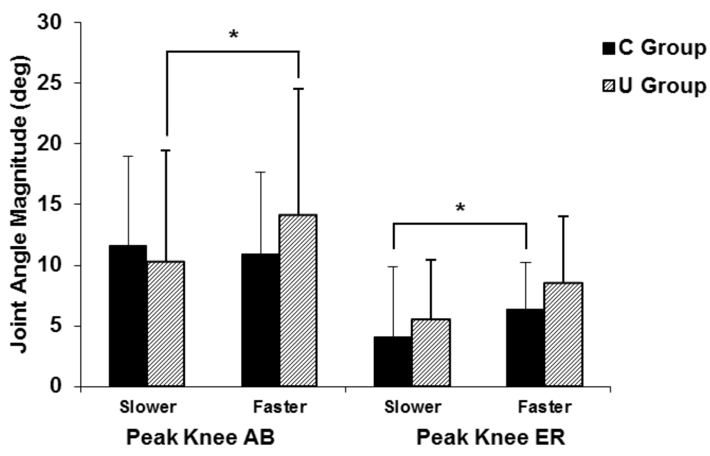

Figure 2.4. Peak joint angles during landing phase of the cut step maneuver. The symbol * indicates significant speed effects ($p < 0.05$).

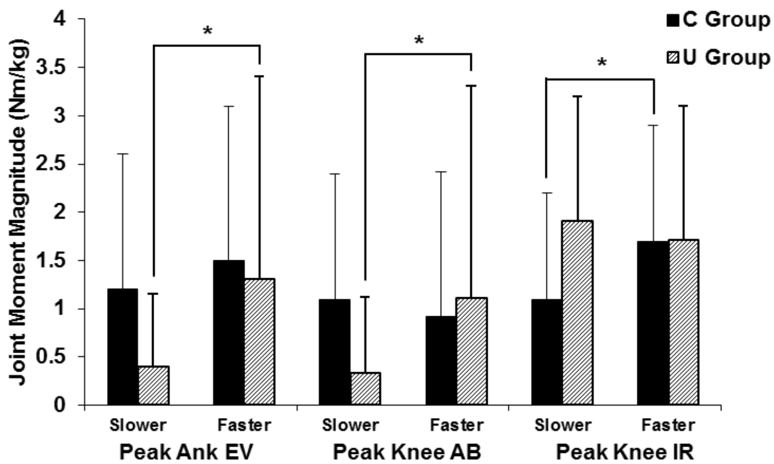

Figure 2.5. Peak joint moments during landing phase of the cut step maneuver. The symbol * indicates speed effects ($p < 0.05$).

Shuttle Run

No main or interactive effects of group were found during Shuttle run trials, though there were main effects of running speed on several measures. Most notably, with an increase in speed peak knee ER increased ($F_{(1,117)} = 32.27$, $p < 0.0001$) by 3.6° (~15%) and peak ankle EV and knee AB moments each increased by about 21% (0.64 Nm/kg, $F_{(1,125)} = 15.77$, $p = 0.0001$ and 0.52 Nm/kg, $F_{(1,117)} = 5.15$, $p = 0.03$, respectively). Also, peak ankle and knee P_F increased with speed by 5.9 (~55%) and 1.6 (~77%) W/kg, respectively ($F_{(1,125)} = 12.82$, $p = 0.005$; $F_{(1,118)} = 19.20$, $p < 0.0001$).

Interestingly, all dynamic trial types (Run, JS, Cut) with the exception of the Shuttle run showed an increase in ML GRF as running speed increased. Run trial results were discussed previously. Mean ML GRF increased with speed by 0.4 N/kg (~17%) for JS trials ($F_{(1,121)} = 6.62$, $p = 0.01$) and 0.6 N/kg (~9%) for Cut trials ($F_{(1,121)} = 4.67$, $p = 0.03$).

2.4.3 Correlations

Tables of complete bivariate correlation results are provided in the Appendix (Tables A11-A13).

Joint Position Sense and Dynamic Maneuvers

JPS error scores had mostly trivial to small correlations with absolute GC angle position error from the ankle during dynamic maneuvers. Here, we present correlations between JPS measures and slower-speed dynamic maneuvers first and then for the faster-speed maneuvers. Absolute errors from the small DF reference angle (+5 deg) had a significant, positive correlation ($\rho = 0.44$, $p = 0.02$) with the absolute plantar/dorsiflexion (PF/DF) GC error during JS maneuvers performed at the slower running speed. Positive correlations that approached significance were also observed between PF/DF GC error during slower-speed JS trials and the absolute and true JPS errors from the large dorsiflexion reference angle (+15 deg; $\rho = 0.33$, $p = 0.099$; $\rho = 0.34$, $p = 0.093$, respectively). Slower-speed run trials produced correlations between PF/DF GC errors and absolute JPS error from the large PF reference angle (-15 deg) that also approached significance ($\rho = 0.34$, $p = 0.09$). The only correlation between I/E JPS errors and slower-speed dynamic maneuvers that was significant, or approached significance, was found for the Shuttle run. True JPS error associated with the large EV reference angle (-15 deg) had a negative correlation with inversion/eversion (INV/EV) GC positioning error during the Shuttle run that approached significance ($\rho = -0.33$, $p = 0.096$).

At the faster running speed, even fewer significant correlations were found between JPS errors and GC positioning errors. During the Cut trials, INV/EV GC errors were significantly and negatively correlated to true JPS errors for both large INV and EV reference angles ($\rho = -0.42$, $p = 0.04$; $\rho = -0.49$, $p = 0.03$, respectively). Two correlations also approached significance for JS trials performed at the faster running speed: INV/EV GC errors during JS trials were negatively correlated with both absolute JPS error at 15 deg of EV ($\rho = -0.35$, $p = 0.08$) and true JPS error at 5 deg of INV ($\rho = -0.35$, $p = 0.08$).

<u>Drop Landings and Dynamic Maneuvers</u>

Correlations between measures of joint power absorption during drop landings and dynamic maneuvers were small or trivial ($\rho < 0.30$). However, several significant moderate - large correlations were found between ankle GC angle measures during drop landings and dynamic maneuvers. In general, the highest correlations were seen for PF/DF and internal/external rotation (IR/ER) angles. At the slower running speed, ankle GC angles during Run trials had moderate, positive correlations ($\rho = 0.44$, $p = 0.03$; $\rho = 0.41$, $p = 0.04$, respectively) with INV/EV and IR/ER GC angles during forward drop landings. At the faster running speed, the correlation with INV/EV GC angles was no longer significant, but the correlation between IR/ER GC angles increased to $\rho = 0.53$ ($p = 0.005$). Also, PF/DF GC angles during forward drops and faster-speed Run trials had a moderate, positive correlation of $\rho = 0.47$ ($p = 0.02$). PF/DF and IR/ER GC angles from slower-speed Run trials were also significantly correlated with the corresponding GC angles during side drop landings ($\rho = 0.50$, $p = 0.01$; $\rho = 0.40$, $p = 0.048$, respectively), and these correlations increased at the faster running speed ($\rho = 0.66$, $p = 0.003$; $\rho = 0.67$, $p = 0.003$, respectively).

Measures of PF/DF GC angles during slower-speed JS trials had a large positive correlation (ρ = 0. 0.56, $p = 0.003$) with PF/DF GC angles from forward drop landings. IR/ER GC angles from slower-speed JS trials and forward drop landings were also moderately and positively correlated ($\rho = 0.40$, $p = 0.04$). Similar results were obtained for correlations between GC angles from slower-speed JS trials and side drop landings. Again, measures of PF/DF GC angles for these two actions had a large positive correlation of 0.59 ($p = 0.002$), and measures of IR/ER GC angles had a moderate positive correlation ($\rho = 0.42$, $p = 0.04$). As speed increased however, the correlations between JS GC angles and forward drop landing GC angles declined and were no longer significant. Faster-speed JS trials retained a significant positive correlation of PF/DF GC angles with PF/DF GC angles from side drop landings, but the correlation between IR/ER GC angles were lower and not statistically significant.

PF/DF and IR/ER GC angles of slower-speed Cut trials were significantly correlated with the corresponding GC angles from both forward ($\rho = 0.68$, $p = 0.0002$; $\rho = 0.42$, $p = 0.04$, respectively) and side ($\rho = 0.56$, $p = 0.004$ and $\rho = 0.57$, $p = 0.003$) drop landings. At the faster speed, GC angles for Cut trials were significantly and positively correlated with corresponding GC angles for forward ($0.43 < \rho < 0.58$) and side ($0.41 < \rho < 0.58$) drop landings along all three anatomical axes (PF/DF, INV/EV, IR/ER).

In slower-speed Shuttle runs, PF/DF GC angles were moderately and positively correlated ($\rho = 0.46$, $p = 0.02$) with PF/DF GC angles from forward drop landings. As running speed increased, the PF/DF correlation declined and was not significant, but a moderate and positive correlation ($\rho = 0.46$, $p = 0.02$) was observed between INV/EV GC angles during Shuttle and forward drop landings. GC angles from Shuttle run trials had more significant correlations with GC angles from side drop landings than with those from forward drop landings. PF/DF GC angles from slower-speed Shuttle trials and side drop landings had a large positive correlation ($\rho = 0.52$, $p = 0.007$), and IR/ER GC angles between these two actions were moderately and positively correlated ($\rho = 0.44$, $p = 0.03$). At the faster running speed, Shuttle run GC angles had moderate positive correlations ($0.42 < \rho < 0.47$) with side drop GC angles about all three anatomical axes.

2.5 Discussion

2.5.1 Joint Position Sense and Single Leg Drop Landings

Though greater JPS errors with ankle instability have been reported previously (Jerosch & Bischof, 1996; Boyle & Negus, 1998; Willems et al. 2002), another found no difference (Gross, 1987), and no significant group differences were found here. There was, however, a significant main effect of reference angle suggesting that true errors for DF reference angles were larger (~1.5 degrees) than errors for PF reference angles. A study by Forestier et al. (2002) also found significantly greater error for DF angles than PF angles following fatigue, but, generally, differences due to reference angles are more often reported based on the magnitude of the reference angle rather than the direction of the JPS test. That is, greater JPS errors are typically expected for larger magnitude reference angles (Glencross and Thornton, 1981; Goble, 2010). Specific differences between reference angles were not a primary focus here, but because differences were observed, bivariate correlations were computed using JPS errors for each reference angle level to determine if a given reference angle magnitude had a stronger correlation with GC angles during drop landing or dynamic maneuvers. The implications of these correlations are discussed in Section 4.3 below.

Drop landings and variations of this technique have been used as a means of studying the potential for non-contact sports injuries, particularly to the knee and ankle. Magnitudes of joint angles, moments, and power absorption reported here are similar to those reported by other studies (Decker et al., 2003; Yeow et al., 2009; Niu et al., 2011; Ali et al., 2012). Some group-level differences in drop landing kinetics and kinematics were evident. Forward drop landing measures suggested that the U group had greater frontal plane ankle power absorption than the C group. Note that joint powers are a product of joint moment and joint angular velocity. Greater ankle power absorption would therefore indicate an increase in joint moment, an increase in joint angular velocity, or both. As the frontal plane joint moments (INV, EV) were equivalent between groups, the greater ankle frontal plane power absorption for the U group was likely due to greater ankle angular velocity during landing. Greater joint velocities during landing are thought to indicate poorer restraint of joint motion and be an indicator of greater ankle injury risk (Niu et al., 2011).

Group differences were also found during side drop landings. The C group landed with greater ankle IR and remained in IR throughout the landing phase. Given that ankle IR is a mechanism of sprain injury (Garrick, 1977), landing with greater IR GC angle would presumably put the C group at greater risk of sprain injury. The C group appeared to control the drop landing maneuver through transverse plane power absorption at the knee. The U group, in contrast, appeared to do so by externally rotating the ankle. During drop landing, a differential use of ankle and knee joints was apparent between groups to sustain impact, with the U group exerting control at the ankle level and the C group relying more on the knee for control. Recent principal component analysis of ankle joint kinematics during a single-leg land-and-cut maneuver also reported that adults with CAI exhibit greater kinematic complexity at the ankle than do controls (Kipp & Palmieri-Smith, 2013). The U group's reliance on the ankle as a primary means of balance control, particularly as tasks become more difficult, may be contributing to their increased sprain reinjury risk.

2.5.2 Dynamic Athletic Maneuvers

The magnitudes of joint angle, moment, and power absorption values reported here are comparable to those reported in other studies (Winter, 1982; Ferber et al., 2003; McLean et al., 2005; Schache et al., 2011). A faster running speed, as reported elsewhere (Novacheck, 1998; Schache et al., 2011), resulted in larger joint angles and greater joint moments and power absorption, but a few group-level differences were also evident. For the Run trials, the U group exhibited greater knee flexion at ground contact and greater peak knee flexion during the landing phase. Since knee flexion during gait and running is a means of absorption during impact (Novacheck, 1998; Buschbacher et al., 2009), landing with greater knee flexion presumably helps the U group to avoid greater ankle joint impulse.

Although for most people the predominant ankle joint moment during running was a plantarflexion moment, several participants exhibited a dorsiflexion moment during early landing phase that transitioned to a PF moment by the end of stance. The U group, however, exhibited a PF moment throughout landing and stance phases. Inspection of the C group results suggests that when running at the slower speed these participants were landing in dorsiflexion and exerting a DF moment, presumably to resist "foot slap" as the foot rotates forward to full contact with the floor. At the faster speed, however, both groups exhibited PF moments throughout the landing and stance phase, and group differences in sagittal plane ankle moment were negligible. Overall, these results suggest that the U group adapted to an increase in running speed by increasing knee flexion, perhaps to limit ankle impact. Note, though, that this is in contrast to the drop landing results, which suggested that the C group utilized their knee for landing control whereas the U group exhibited more changes at the ankle. Measures of ML

GRF for the Run trials were also greater for the U group. Increases in GRFs are expected with increases in running speed (Hamill et al., 1983; Novacheck, 1998), but a greater increase for the U group suggests that more of the ground contact impulse is directed laterally as speed increases. Previous research has suggested that increases in ML GRF during running – particularly in the injured runner population – may be due to foot placement (McClay and Cavanagh, 1994). Runners who exhibit a “cross-over” style running stride (where a right foot fall lands under the left side of the body and vice versa) have a higher ML GRF at a given speed (McClay and Cavanagh, 1994). Although not explicitly measured here, as running speed increased, the U group may have transitioned to a different foot placement pattern compared to the C group, a difference that may expose them to a greater risk of inversion injury.

The U group also utilized more knee flexion to adapt to the faster running speed during JS trials, and had an increase in transverse plane knee power absorption at the faster speed. The C group, in contrast, reduced peak knee flexion at the faster speed and instead reduced their ankle IR angle. A study of drop landing mechanics suggested that landing in a more erect posture (e.g. greater knee extension or hip extension) directs more energy absorption to joints closest to the ground (Decker et al., 2003). Here, the peak ankle power absorption in each of the anatomical planes increased considerably for the C group at the faster speed (Appendix Table A9), although these differences were not statistically significant ($p > 0.11$). It is likely that the reduction in ankle IR with speed helped to mitigate injury by returning the ankle to a more neutral position, allowing absorption to occur primarily in the sagittal plane, and enabling a larger range-of-motion at the ankle. The U group, however, retained a larger IR angle even at the faster speed. Greater knee flexion may have been intended to reduce the impact absorbed at the ankle, but with the ankle still at a larger IR angle more transverse knee power absorption was needed to complete the JS at the faster speed. With increasing difficulty (i.e., faster running speed), this knee strategy alone may not be adequate to resist injury without more neutral alignment of the ankle.

Cut maneuvers also revealed differences in how the two test groups executed a rapid change of direction. While C group controlled landing by exhibiting more knee ER as running speed increased, the U group instead increased knee AB angle. The relevance of these kinematic differences is apparent when comparing differences in joint moments needed to stabilize the leg. The C group generated greater knee IR moment as speed increased. The U group, however, generated an increase in both ankle EV and knee AB moments. The increase in knee

AB and, consequently, knee and ankle frontal plane moments, increases the effort required by those with CAI to avoid ankle inversion injury.

The Shuttle run maneuvers were expected to reveal the biggest differences between groups because of the obvious challenge it imposes on ankle stability. Instead, only running speed elicited significant effects on landing mechanics. As running speed increased, ankle PF angle at GC decreased along with sagittal plane power absorption. Ankle joint angles and moments during landing remained unchanged between speed conditions, although there was an increase in frontal plane power absorption. This increase in frontal plane power absorption likely reflects an increased angular velocity of the shank relative to the foot due to the faster running speed. At the knee, peak ER angle and knee AB moment during landing increased as speed increased. There was also a significant increase in knee frontal plane power absorption. Thus, rather than relying on ankle EV moment or ER of the ankle, both groups demonstrated more reliance on knee mechanics to execute the Shuttle run at the faster speed.

For all maneuvers, excluding the Shuttle run, a greater ML GRF was found in response to an increase in running speed, but no group differences were found. Another study also reported no significant between-group differences in ML GRF for cutting maneuvers (Dayakidis & Boudolous, 2006). As speed increases, the shear forces creating an ankle inversion moment about the ankle increase. Interestingly, the Cut trials demonstrated the greatest ML GRF shear forces rather than the Shuttle trials as expected. This was likely due to larger knee and ankle external rotation angles during the Shuttle maneuvers. Greater external rotation would cause shear ground reaction forces to be distributed between the mediolateral and anteroposterior directions. Presumably, less external rotation during the Cut maneuvers translated to greater magnitudes of ML GRF.

2.5.3 Bivariate Correlations

A second aim of this study was to determine the extent to which quasi-static measures of ankle function relate to dynamic measures. JPS errors for each reference angle were used separately in bivariate correlations, because our analysis revealed a significant effect of reference angle on JPS measures. A few of the correlations between absolute error of ankle GC angles and absolute and true JPS errors were significant (or approached significance), although no specific pattern was evident. Generally, speed, reference angle, and maneuver type affected the magnitudes and directions of the correlations. JPS errors associated with larger PF/DF reference angles had more substantial correlations with measures of PF/DF GC errors for dynamic maneuvers. JPS errors may be a useful technique for predicting motor control during dynamic tasks, though the current results suggest this value to be limited.

More substantial correlations were found between GC angle measures during drop landings and GC angles during dynamic maneuvers. The specific magnitude and direction of the correlations depended on the maneuver with which drop landing measures were compared and the running speed at which the maneuver was executed, though some notable patterns were evident. PF/DF and IR/ER GC angles from forward and side drop landings were typically more highly correlated with corresponding GC angles during dynamic maneuvers, and significant correlations between INV/EV GC angles from drop landings and dynamic maneuvers were less common. Still, faster-speed Cut trials showed moderate to large correlations between GC angles in each of the anatomical planes and corresponding GC angles during both drop landing types. All three GC angles from faster-speed Shuttle trials also had significant, moderate correlations with the same measures during side drop landings. Although the specific correlation magnitudes varied, drop landings had similar GC kinematics to those exhibited during dynamic maneuvers, particularly in the sagittal and transverse anatomical planes. Single-leg drop landings are often used to study non-contact injury potential between two groups. This correlation analysis suggests that GC kinematics measured during drop-landing tests may also offer some insight into GC kinematics affecting injury potential during athletic activity.

Magnitudes of correlations reported here differ somewhat from those in Whatman et al. (2011), who compared peak joint angles during functional screening tests (single knee bend, lunge, step down, etc.) with peak kinematics during jogging. For instance, they reported correlations of 0.60 – 0.78 between peak eversion and inversion angles during a controlled step-down protocol and a jogging stride. Differences between their results and ours may be attributed to differences in the tasks being compared and the specific measures being correlated. Regarding the latter, Whatman et al. (2011) compared peak joint angles whereas the present study focused on joint angles at GC. Both studies, however, provide support to the idea that some quasi-static or clinical techniques (e.g., drop landings) are able to reflect kinematics during more dynamic activities.

2.5.4 Limitations

A limitation of this study is that it only addressed landing biomechanics on solid, flat surfaces. In team-based running and jumping sports, however, sprains are also frequently caused by landing on an opposing player's foot or on uneven ground (Garrick, 1977). The methods here do not address biomechanics of running on uneven ground, and pertain only to sprains obtained through running and athletic maneuvers. Also, this study only explored two quasi-static test methods. These methods were chosen because the dependent measures obtained from each were ankle kinematics and kinetics that seemed to have high potential for correlating well with dynamic measures. Future efforts are needed to validate other quasi-static methods as well.

Another potential limitation of this study is the use of surface markers for estimating joint kinematics. Markers fixed to the skin are susceptible to small movements relative to their intended landmarks, particularly during dynamic tasks, and these movements may influence joint angle estimates. Joint angles can also be affected by the selection of the segment and joint coordinate systems, and the Euler angle rotation sequence used when constructing the necessary rotational matrices. As such, the joint kinematics reported here likely involve some errors. Yet, they are likely a close approximation to true anatomical angles since our methods follow closely those described in other papers (McLean et al., 2005; Dumas et al., 2004), and the magnitudes of the current joint kinematics and kinetics are similar to those reported elsewhere (Winter, 1983; Ferber et al., 2003; McLean et al., 2005; Schache et al., 2011).

Other limitations relates to the lab environment. Dynamic maneuvers performed here were practiced at each speed to help ensure that captured trials involve “natural” performance to the extent possible. We acknowledge, however, that the test protocols were completed in a lab environment with a limited space, which may have influenced how the maneuvers were executed. Lab space also restricted the running speeds that could be tested. Testing a number of different running speeds would have provided more generality to our findings regarding group x speed interaction effects. Participant screening was another potential limitation. Screening for participants with unstable ankles relies heavily on anecdotal evidence and personal perception of stability as reported by ankle stability surveys. Individual differences in stability perception and inaccurate reporting of sprain history could produce a test group consisting of a mix of more severe and less severe ankle stability. Finally, the study may have been underpowered to detect relatively small effect sizes.

2.6 Summary

There were two primary objectives of this study: 1) to determine whether adults with and without ankle instability exhibit differences in joint kinematics/kinetics during the execution of dynamic athletic maneuvers and whether such differences are influenced by speed; and 2) to determine the level of correspondence between quasi-static and dynamic measures of ankle control. Although there were only a few group-level differences in the execution of the maneuvers, the groups appeared to have different strategies in adapting to an increase in speed. The C group more often adjusted ankle or knee internal/external rotation and used ankle and knee torsional efforts to control impact and landing. In contrast, the U group responded to increased speed by increasing knee flexion and control movements more with frontal plane joint angles and moments. The U group's greater reliance on frontal plane control may contribute to their increased risk of sprain reinjury. Remaining uncertain, however, is whether distinct responses in the U group represent differences in pre-existing motor patterns for adults with CAI that may have contributed to repetitive sprain injuries or changes in motor patterns that resulted of repetitive sprains. Regardless, results of this study suggest that assessing motor differences between groups across speed conditions can yield important information regarding injury propensity.

JPS errors have been used in prior work to identify motor deficits between groups, but the relationship between these error measures and kinematic variability during dynamic athletic maneuvers was unclear. Our results suggest that ankle JPS error measures do not have meaningful correlations with ankle GC error measures during dynamic athletic activities. GC angles during drop landings, however, had moderate to large correlations with measures of GC obtained in each of the four dynamic maneuvers. These latter results indicate that quasi-static measures can, to some degree, reflect ankle kinematics under dynamic conditions. The magnitudes of these correlations, however, differed depending on running speed and the specific maneuver being assessed. Typically, drop landings have been used to assess ground contact kinematics relevant to jumping injuries, and the current results provide support for extrapolating kinematic results from drop landings, which are a simpler and more repeatable task, to kinematics during athletic maneuvers performed while running.

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Chapter 3

Differences in Fatigue Development and Ankle Stabilization during Drop Landings Between Controls and Adults with Ankle Instability

3.1 Abstract

This study explored the role of fatigue in recurrent ankle sprains for adults with chronic ankle instability (CAI). Comparisons of ankle strength measures and ratings of perceived exertion obtained during a fatigue protocol revealed that two test groups (stable vs. unstable ankles) developed localized muscle fatigue at similar rates when exercising at equivalent relative effort levels (percentage of strength) and that the two groups reported similar levels of discomfort as fatigue progressed. This indicates that recurrent ankles sprains for those with CAI are likely not due to differences in fatigue resistance or differences in subjective perceptions of fatigue. The two test groups also completed forward and side drop landings before and after a fatigue protocol. Kinematic, kinetic, and EMG measures during drop landings revealed significant group differences both pre- and post-fatigue. For both drop-landing types, the unstable ankle group exhibited more ankle external rotation, which may be protective against inversion sprains. Pre-fatigue comparisons for forward drop landings indicated that the unstable and control groups exerted more control at the ankle and knee, respectively. Side drop landings, however, showed differences in the planar contributions to joint stabilization. Stabilization for the control group was largely achieved through frontal plane exertions. The unstable group exerted frontal plane control at the knee but relied on more transverse plane control at the ankle. Fatigue-induced group differences suggested that ankle sprains in the unstable group may be attributed to relatively smaller biomechanical adjustments following fatigue. Group differences in EMG amplitudes and co-contraction ratios were also observed. Pre-fatigue, adults with stable ankles exhibited similar contraction levels of agonist and antagonist muscle pairs supporting the ankle while those with CAI had greater relative contraction levels of tibialis anterior (TA) and peroneus longus (PL) muscles. Following fatigue, the U group decreased co-contraction of TA and PL muscles indicating greater use of the PL to control ground impact. Overall, this work suggests that ankle sprain injuries in those with CAI may be due to an insufficient magnitude of joint stabilization moments after fatigue, more complex ankle and knee joint control strategies, and a greater reliance on the primary evertor (PL) muscle for ankle stabilization.

3.2 Introduction

Ankle sprains remain a common injury, with 600,000 to 2 million incidents occurring each year in the US (Soboroff et al., 1984; Waterman et al., 2010) and with roughly half resulting from participating in running and jumping sports (Garrick, 1977; Yeung et al., 1994; Hootman et al., 2007; Waterman et al., 2010). These injuries are painful and often (~30-40%) lead to a long-term condition known as chronic ankle instability (CAI) that is characterized by frequent recurrent sprain injuries (Freeman, 1965; Garrick, 1977; Mack, 1982; Schaap et al., 1989; Peters et al., 1991).

Most (~85%) ankle sprains are inversion injuries, in which the sole of the foot rotates medially under the body (Garrick, 1997; Mack 1982). Ankle sprain occurrence is influenced by multiple factors, including ankle joint position and impact force at ground contact, as well as the magnitude of reflexive and voluntary muscular force development to control ankle motion (Konradsen, 2000). Pre-activation of lower leg muscles immediately preceding ground contact may also be important, in that it can enable more rapid reactionary ankle torque production (Konradsen, 2005). Muscular fatigue, in contrast, is a factor that can inhibit joint positioning and adequate muscular response.

In adults with healthy, stable ankles, fatigue adversely affects both passive and active joint position sense (JPS: Forestier et al., 2002; Mohammadi & Roozdar, 2010), possibly due to a fatigue-induced increase in the threshold for muscle spindle firing (Rozzi et al., 2000). Fatigue, by definition, also causes a reduction in maximal voluntary muscle force generation (Hall, 1999), though evidence is mixed as to whether fatigue reduces the amplitude of muscle reflex responses (Wilson & Madigan, 2007; Jackson et al., 2009). However, both compromised force generation capacity and poorer JPS indicate that fatigue has the potential to exacerbate sprain reinjury risk. This assertion is further supported by reports that the majority of sports injuries occur during the latter half of a period of play or during the second half of a competition (Pinto et al., 1999; Gabbett, 2000). Ankle sprains in particular were found to occur with great frequency during the last third of each half of professional soccer matches (Woods et al., 2003).

Given the biomechanical and neuromuscular effects of fatigue, it is likely that the risk of ankle sprain is elevated for all athletes experiencing muscular fatigue. Considering that adults with unstable ankles typically have deficits in joint position sense compared to controls (Jerosch & Bischof, 1996; Hartsell, 2000), resisting ankle sprain may be even more challenging for them.

Participants with CAI reportedly demonstrate poorer postural control following an aerobic fatigue protocol (Hosseinimehr et al., 2010). Poorer postural control following fatigue has also been observed between the involved (history of sprain) and uninvolved (no history of sprain) ankles of adults with unilateral instability (Gribble et al., 2004).

The goal of the present research was to further explore the potential role of fatigue in recurrent ankle sprains among adults with CAI. Single-leg, drop-landing tests were used to measure muscle activation and joint dynamics relevant to ankle stabilization during a rapid, weight-bearing activity. Drop landings were completed before and after a protocol that induced localized muscle fatigue at the ankle. It was hypothesized that fatigue would have differential effects on two groups of adults, those with CAI and healthy controls, in terms of: 1) joint angles in preparation for landing; 2) maximal joint moments and powers stabilizing the ankle and knee following impact; and 3) pre-activation and co-contraction levels of lower leg muscles. Specifically, pre-activation was expected to decrease with fatigue (Konradsen, 2000) requiring greater co-contraction to stabilize the ankle. Due to reports of fatigue affecting JPS among healthy ankles (Forestier et al., 2002; Mohammadi & Roozdar, 2010) and the known deficit in JPS with CAI (Jerosch & Bischof, 1996; Hartsell, 2000), joint angles were expected to be affected by fatigue in both groups, but more hazardous joint angles (e.g. greater ankle joint inversion, plantarflexion, knee extension) were expected for the unstable ankle group. Finally, given that fatigue reduces muscle force capacity and that reaction time may be delayed in adults with CAI (Mitchell et al., 2008), it was expected that joint stabilization moments would increase following fatigue, especially among adults in the unstable group. We also used a controlled fatigue protocol to determine if CAI status influences the relationship between objective and subjective measures of fatigue.

3.3 Methods

3.3.1 Participants and Overview of Experimental Design

Participants included 14 adults (3 males, 11 females) with unstable ankles and 14 controls who were matched individually for age, stature, body mass, and gender (Table 3.1). Ankle instability was determined using the Cumberland Ankle Instability Tool (CAIT) questionnaire and a set of questions addressing medical history. Those in the unstable ankle group (U) had a CAIT score ≤ 25, a history of at least two inversion sprains to the same ankle, and provided anecdotal reports of their ankle “rolling” or “giving way” during activity. Adults in the control group (C) had a history of 0-1 sprains and a CAIT score of ≥ 28. Exclusion criteria included chronic joint pain, prior joint surgeries or neuropathies of the legs, muscle weakness, and vestibular or balance disorders. Indication of current joint pain, participation in physical rehabilitation programs, or regular use of ankle tape or braces were also grounds for exclusion. Written and verbal consent were obtained from all participants, and all experimental protocols were approved by the Virginia Tech Institutional Review Board.

Table 3.1. Mean (SD) participant characteristics for control and unstable ankle groups (n = 14 for each). *P* values are from unpaired t-tests, and the * symbol indicates a significant ($p<0.05$) difference between groups.

Measure	Control (C)	Unstable (U)	*p*
Age (yr)	24.5 (2.3)	24.1 (3.3)	0.79
Stature (cm)	172.7 (11.7)	171.0 (11.6)	0.71
Body Mass (kg)	66.1 (12.5)	67.4 (14.6)	0.80
CAIT Score	29.6 (0.8)	19.9 (3.0)	<0.0001*

For those in the U group, the leg with the lowest CAIT score was tested. The test leg for the C group was that which matched the limb dominance (or non-dominance) of the test leg of their U group counterpart, and the dominant limb was identified as the leg preferred for kicking a ball (Fernandes et al., 2000; Johnson & Johnson, 1993). All participants wore the same model of gender-specific athletic shoe fitted to the nearest whole size. Participants completed fatiguing ankle exercises, during which fatigue development and perception were assessed (Fatigue Protocol 1). Participants also completed a series of single-leg drop landings, both before and after a distinct set of fatiguing exercise (Fatigue Protocol 2). In these, fatigue development, muscle co-activation, and drop landing mechanics were assessed as outcome measures. All procedures were completed in a single experimental session (Figure 3.1).

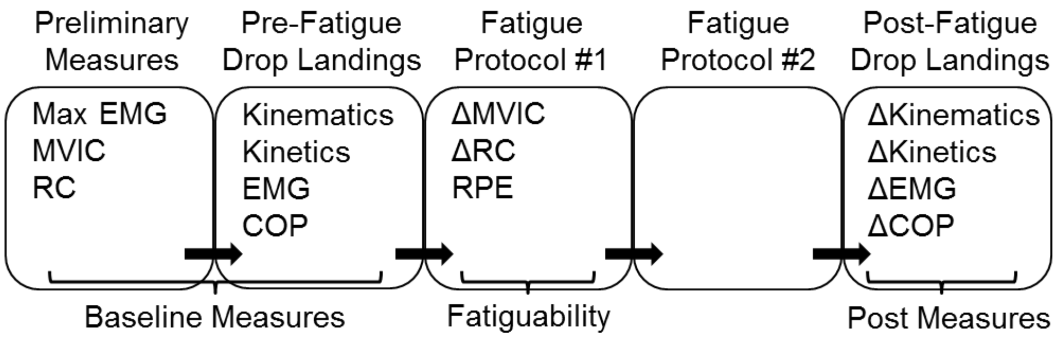

Figure 3.1. Overview of experimental procedures and dependent variables. The symbol "Δ" indicates that measures of interest are changes induced by fatigue. See text for measure definitions.

3.3.2 Preliminary Measurements

Muscle activity of the test leg was monitored throughout all testing protocols. Bipolar surface EMG electrodes (Model # A10012-5S, Vermed, Inc., Bellows Falls, VT) were applied to the skin over three muscles of the lower leg: tibialis anterior (TA), peroneus longus (PL), and the lateral head of the gastrocnemius (GS). Raw EMG was amplified with a 2K gain in hardware (Measurement Systems, Inc., Ann Arbor, MI) and sampled at 2500 Hz. Resting EMG signals were recorded in a relaxed, supine posture, and maximal EMG signals were subsequently recorded for each muscle using recommended manual muscle testing procedures (Kendall et al., 1993; Starkey & Ryan, 2003). Maximum voluntary isometric contractions (MVICs) were also completed in both plantarflexion (PF) and dorsiflexion (DF). Participants were seated in a commercial dynamometer (Biodex System 3, Biodex Medical Systems, Inc., Shirley, NY) in the PF/DF configuration with 50 degrees of knee flexion and the ankle in a neutral position. Straps were used to secure the foot to a foot plate and to hold the leg firmly against the thigh support pad, and a lap belt was tightened across the hips. A minimum of three 3-second MVICs were recorded for each motion, with at least 1 min of rest between each, and torque values were recorded at 2048 Hz. If MVICs were inconsistent or increasing, they were repeated until peak values plateaued. Verbal encouragement was given for all attempts. Peak torque across MVICs was recorded and used to customize each participant's fatigue protocol (see below). Several isometric reference contractions (RCs) were also completed, following MVICs (after ~2 min of rest) and intermittently throughout the remainder of the experiment. RCs were done statically, for 3 sec, at 50% of individual MVIC in both PF and DF. EMG signals obtained during RCs were analyzed to assess fatigue development.

3.3.3 Drop Landings

Participants completed a series of single-leg drop landings prior to and following localized ankle fatigue (Fatigue Protocol #2 below). Reflective markers were positioned bilaterally over select anatomical landmarks (Figure 3.2) of the lower extremities and pelvis, including the first and fifth metatarsal head, lateral and medial malleoli, calcaneus, lateral and medial femoral condyles, greater trochanter, and anterior superior iliac spine. Rigid marker clusters were also attached the thigh, shank, and foot segments of the test leg, and a few additional markers were placed on the trunk to aid in marker labeling. Drop landings were completed by having the participant drop themselves from a platform of 30.5 cm in height onto a force plate embedded in the floor (Model #K20102, Bertec Corp., Columbus, OH). Participants were instructed to place their hands on their waist and land only on their test leg with their contralateral knee bent at 90 degrees. They were asked to maintain single-leg balance for 10 seconds after impact. Practice was given prior to data collection, and drop landings were performed both facing forward (forward drops) and rotated 90-degrees to the side (side drop) where the participant propelled off the platform with a lateral movement onto the test leg. Five forward and five side drops were completed both before and after fatigue. Before fatigue, forward drops were completed first to ensure participants could execute the proper form. Side drops were performed after additional practice. Following fatigue, forward and side drops were alternated to distribute fatigue effects across both types of trials. During each drop landing trial, kinetic and EMG data were recorded at a rate of 2500 Hz in synchronization with recordings of marker positions made using a 6-camera motion capture system (Vicon MX 1.7.1, Vicon, Oxford, UK), at 250 Hz.

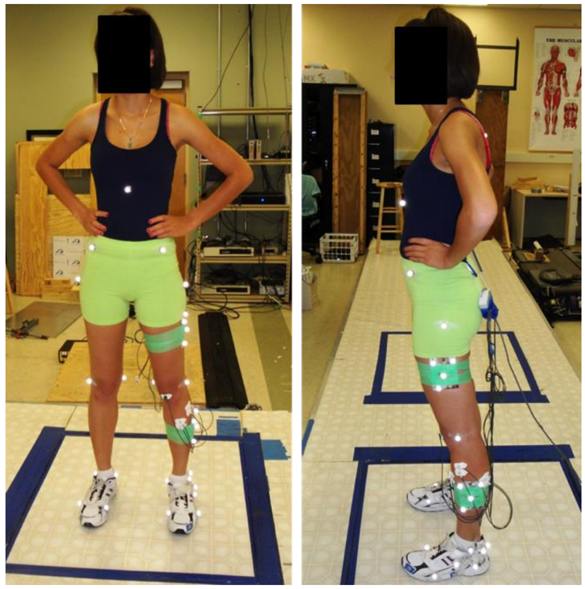

Figure 3.2. Participant fitted with reflective marker set and equipped with EMG electrodes.

3.3.4 Fatigue Protocol #1: Fatigue Development and Fatigue Perceptions

Four 4-minute "rounds" were performed consecutively for a total duration of 16 minutes. Each "round" involved 3 minutes of repetitive, isotonic PF and DF exertions, and in the remaining minute during which MVICs and RCs were completed. Participants sat in the dynamometer in the PF/DF configuration, and ankle range of motion (ROM) was set to 30 degrees of PF and 15 degrees of DF as measured from the neutral position. A counterweight was added to the ankle attachment to ensure a similar level of effort was required to move the ankle in PF and DF motions. Isotonic exertions were performed at a rate of 12 cycles/min through the 45-degree ROM, and isotonic loads were set to 70% and 30% of individual PF and DF MVICs, respectively. Pilot work suggested that these levels were sufficient to produce substantial fatigue (≥20% decrease in MVIC) while still permitting movement through the entire ROM.

A custom LabVIEW™ program (v10.0, National Instruments Corp., Austin, TX) was developed to provide visual feedback and guide participants through the protocol (Figure 3.3). During the isotonic exertions, the program displayed the time-dependent target ankle position, and an indicator dot tracked ankle position in real time. Participants were encouraged to follow the line as closely as possible. Each "spike" prompted the participant to complete one full cycle of plantarflexion and dorsiflexion in three seconds, and two seconds of rest were given between cycles. Such cycles were continued for blocks of three minutes, and the initial exertion direction (lead exertion) was randomized for each participant. Immediately after each 3-minute block of isotonic exertions, participants used the Borg CR-10 scale (Borg, 1998) to rate their perceived level of fatigue in the front (dorsiflexors) and back (plantarflexors) of the lower leg.

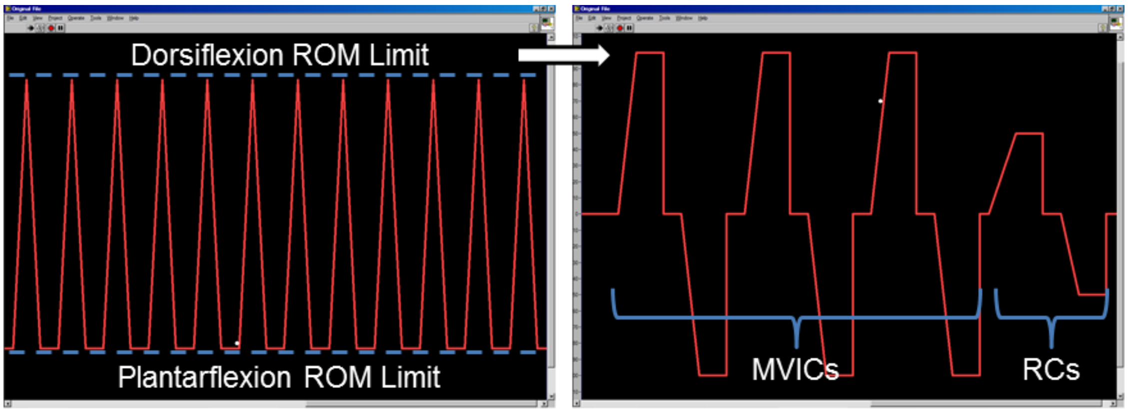

Figure 3.3. Representation of visual feedback provided during Fatigue Protocol 1 (right test leg, dorsiflexion lead exertion). The frame on the left shows the visual display during dynamic isotonic exertions while the frame on the right shows the visual display used for intermittent MVIC and RCs. Positive exertions are dorsiflexion and negative exertions are plantarflexion.

Every fourth minute of the protocol (observation time), the participant's ankle was returned to neutral position and held in place by the dynamometer. In this position, three isotonic MVICs were performed for both PF and DF directions followed by an RC in both directions (Figure 3.3). Brief rest periods were provided between each MVIC and RC, and the direction of the first MVIC and RC corresponded to the noted lead exertion. During these measures, the visual display presented target torque values rather than ankle position (Figure 3.3). MVIC and RC cues displayed target torques equal to 100% and 50%, respectively, of the participant's pre-fatigue peak torque in the relevant direction. Participants were instructed to attempt to ramp up their ankle torque to the MVIC target and to maintain that effort for about 3 seconds. For RCs, they were instructed to follow the target as closely as possible. MVIC and RC data were thus obtained at 4, 8, 12, and 16 minutes during the protocol.

3.3.5 Fatigue Protocol #2 and Post-Fatigue Drop Landings

A modified fatigue protocol was used here, designed to induce a comparable level of fatigue in all participants before completing a second set of drop landings. Dynamic isotonic exertions and periodic MVICs were completed, over the same ROM at the same rate in as Protocol #1, though these differed in terms of the isotonic loads and the frequency of MVICs. Here, one "round" consisted of dynamic isotonic exertions for two minutes and MVICs were measured every third minute. Depending on the reduction of MVIC relative to the pre-fatigue MVIC, the isotonic load was adjusted up or down for the next round, similar to the approach in Davidson et al. (2009). This procedure was continued until ankle torques were reduced to under 75% of pre-fatigue MVIC for both PF and DF, and which typically required 4 - 5 rounds. After a delay of ~ 20 sec, while transitioning, participants performed a second set of drop landings using the same procedures and data collection methods described above.

3.3.6 Data Processing and Analyses

Fatigue Protocol #1

Dependent measures included ankle strength and RPE scores collected at each observation time (t = 4, 8, 12, and 16 min). Abbreviations "MVIC" and "RPE" along with subscripts ("PF or "DF") are used to refer to ankle strength and RPE measures in each exertion direction. Additional measures included EMG median frequency (MF) for each muscle during the RCs. Because the PL and GS muscles are primary plantarflexors, the MF was calculated during PF RC efforts. Similarly, because the TA muscle is a primary dorsiflexor, the MF was calculated during DF RC efforts. EMG signals were band-pass filtered (4^{th} order, zero-lag) between 20 and 450 Hz and notch-filtered at 60 Hz. MF was calculated following a Fast Fourier Transform and Welch averaging using 100-ms windows.

Drop Landings

Kinematic and kinetic data for drop landings were low-pass filtered using a 4^{th}-order, zero-lag, Butterworth filter, with respective cutoff frequencies of 20 and 50 Hz. Kinetic data were downsampled to 250 Hz prior to inverse dynamics calculations described below. EMG signals were filtered as described above for Fatigue Protocol #1. Dependent measures derived for drop-landing trials included ankle and knee joint angles, moments, and powers, EMG magnitudes and co-contraction ratios, and center-of-pressure (COP) deviations. Each of these measures was determined with reference to the instant of ground contact (GC), which was determined as the time when the vertical component of the ground reaction force exceeded 10 N, as in earlier work (Hreljac & Stergiou, 2000; Leitch et al., 2011).

Ankle and knee joint kinematics and kinetics were determined using the Euler angle approach (Hamill & Selbie, 2004) and 3D inverse dynamics (Winter 1990; Vaughan et al., 1999; Dumas et al., 2004). Joint angles were determined at GC. In addition, peak joint angles, moments, and powers were obtained separately for both the landing and stance phases following GC. Landing phase was defined as the first 100 ms following the instant of GC (Decker et al., 2003), while stance phase included the remainder of the 10-second trial. Abbreviations are used to identify the anatomical motion or plane to which the measures refer. Ankle motions are defined as plantarflexion (PF) and dorsiflexion (DF), inversion (INV) and eversion (EV); knee motions are defined as flexion (FL) and extension (EX), abduction (AB) and adduction (AD). Both joints also exhibit internal rotation (IR) and external rotation (ER). Joint powers are identified as either absorption (Abs) or generation (Gen) in the sagittal (S), frontal (F), and transverse (T) plane. Mediolateral COP sway range was also calculated, during the stance phase.

Raw EMG magnitudes processed to obtained root mean square (RMS) levels, with a 40 ms moving window (Yeadon et al., 2010), and activation (vs. silence) was defined as an RMS level exceeding two standard deviations of the resting levels (Konradsen et al., 1998; Hopkins et al., 2007). RMS levels were normalized to corresponding maximum values (% Max) obtained during manual muscle testing. Co-contraction ratios (CCR) were determined for each pair of muscles as the ratio of the normalized RMS levels (Padua et al., 2006; da Fonseca, 2006). Muscle abbreviations (TA, PL, GS) are used to define which muscle magnitude served as the numerator and the denominator of the CCR. CCRs = 1 indicate equal activation levels of the two muscles defining the ratio; CCRs greater than or less than 1 indicate greater activation of the muscle in the numerator or denominator, respectively (Padua et al., 2006). Peak values of normalized RMS levels and CCRs were obtained over two distinct time periods: 1) the pre-land phase, defined as 100 ms immediately preceding GC; and 2) the landing phase, defined as 100 ms immediately following GC (Decker et al., 2003; Yi et al., 2003; Weinhandl et al., 2011).

3.3.7 Statistical Analyses

For Fatigue Protocol #1, two-sample *t* tests were used to compare pre-fatigue ankle strength (PF and DF MVICs). Using multivariate analyses of variance (MANOVAs), it was determined that there were no significant order effects ($p > 0.17$) on consecutive MVICs at any observation time in either exertion direction. Post-fatigue measures at each observation time were converted to change scores, as differences from the mean pre-fatigue levels. For each dependent variable (change score), the effects of group and time were assessed using mixed-factor analysis of variance (ANOVA). In cases where sphericity violations were found, a Geisser-Greenhouse correction was used. Rates of change for measures of PF and DF ankle strength and RPE scores were calculated across the four observation times for each participant and trial. Bivariate correlations were then fit, separately for the two groups, between the rates of change in ankle strength and RPE in each direction (PF and DF). Two-sample *t*-tests were used to compare correlation coefficients between groups. Statistical significance was concluded when $p < 0.05$, throughout these and all subsequent analyses.

Analysis for drop landing measures involved several steps. Initially, separate MANOVAs were used to determine if significant order effects were evident for any of the measures across the five pre-fatigue trials, and a similar approach was used for the five post-fatigue trials. These analyses revealed no significant order effects for either set of trials ($p > 0.15$). Two-sample *t* tests were used to determine if there were differences between groups in the pre-fatigue levels of each dependent measure. Subsequently, all post-fatigue measures (at each observation time) were converted to change scores as earlier (i.e., post-fatigue measure – mean of pre-fatigue measures). Main effects of fatigue were assessed using one-sample *t*-tests on the change scores, to determine whether the means were significantly different than zero at each observation time. For each dependent measure, an analysis of covariance (ANCOVA) was then used to determine the effects of group and observation time on change scores, with the pre-fatigue level included as a covariate. All summary statistics are reported as means (SD).

3.4 Results

Complete summary statistics and statistical results for each of the analyses are provided in Appendix B.

3.4.1 Fatigue Protocol #1

Pre-fatigue strengths were similar between groups in both the PF ($t_{(24)} = -0.44$, $p = 0.67$) and DF ($t_{(24)} = -0.52$, $p = 0.60$) directions, with respective values of 43.2 (14.4) and 27.2 (10.6) Nm. At each observation time, both PF and DF ankle strengths were significantly ($p < 0.0001$) less than pre-fatigue levels, and final strength reductions were 21.4% and 37.8%, respectively (Figure 3.4). No significant group differences in strength losses were found at any observation time.

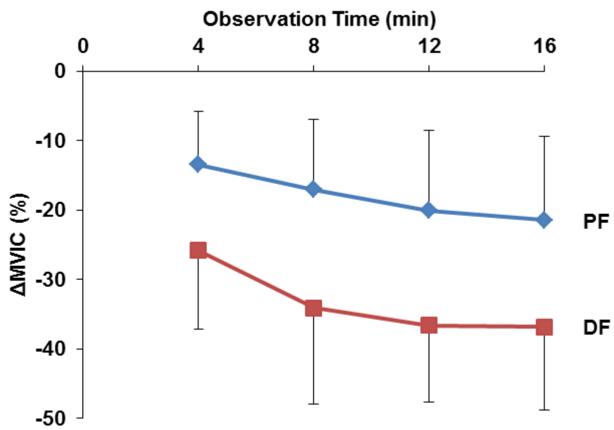

Figure 3.4. Mean strength reductions during Fatigue Protocol #1. Values are presented as percent changes from pre-fatigue levels, and error bars indicate standard deviations.

RPEs were significantly (p < 0.0001) higher post fatigue, and increased over time. Final RPE_{PF} and RPE_{DF} scores were 6.9(1.6) and 5.2(1.9), respectively, and no significant group differences in RPE changes were found at any observation time. Significant changes in MF were only found for the TA muscle. MF_{TA} significantly (p < 0.0001) decreased (by 8.0%) at 4 min, but then increased at each subsequent observation time and at 16 min was no longer different from the pre-fatigue level. There were no significant main effects of group or time on MF for either the PL or GS muscles. No significant group differences were found in the correlations between rates of change in strength and RPE scores (PF: $t_{(1,26)} = 0.0029$, $p = 0.99$; DF: $t_{(1,26)} = -0.41$, $p = 0.68$). Across both groups, respective correlations between PF and DF strength changes and RPE changes were 0.47 (0.31) and 0.57 (0.34).

3.4.2 Drop Landings

Full summary statistics and statistical results for kinematic, kinetic, and EMG measures of drop landings are provided in Appendix B. Significant main effects of fatigue were observed for several dependent measures. Here, and given the goals of the study, the presentation of results focuses on the main and interactive effects of group.

<u>Forward Drop Landing Kinetics and Kinematics</u>

Several pre-fatigue measures at the ankle and knee were significantly different between groups. During the landing phase, the U group had a greater peak ankle ER angle (2.0 deg; $t_{(77)} = -2.65$, $p = 0.0098$) and peak normalized PF moment (1.0 Nm/kg; $t_{(85)} = -4.24$, $p < 0.0001$) than the C group (0.2 deg and 0.3 Nm/kg, respectively). The C group exhibited greater peak normalized ankle ER moment ($t_{(122)}$ = 2.19, p = 0.031) at 0.3 (0.2) Nm/kg than the U group at 0.2 (0.2) Nm/kg, but the U group had a greater peak ankle Abs P_S ($t_{(83)}$ = -4.02, p = 0.0001) than the C group, with respective means of 9.0 (10.4) W/kg and 3.3 (4.3) W/kg. The U group had significantly greater knee AB ($t_{(101)}$ = -2.45, p = 0.016) at 2.0 (5.4) deg compared to the C group's 0.04 (3.2) deg, but the C group exhibited greater peak normalized knee EX moment (6.3 Nm/kg vs. 5.4 Nm/kg; $t_{(109)}$ = -1.99, p = 0.049) and peak knee Gen P_S (40.9 W/kg vs. 33.8 W/kg; $t_{(115)}$ = -2.15, p = 0.034). During stance phase, pre-fatigue group differences were found for peak ankle ER angle ($t_{(81)}$ = -3.05, p = 0.0031) and peak Abs P_S ($t_{(104)}$ = -3.02, p = 0.0032; $t_{(85)}$ = 3.45, p = 0.0009) at both joints (Figure 3.5).

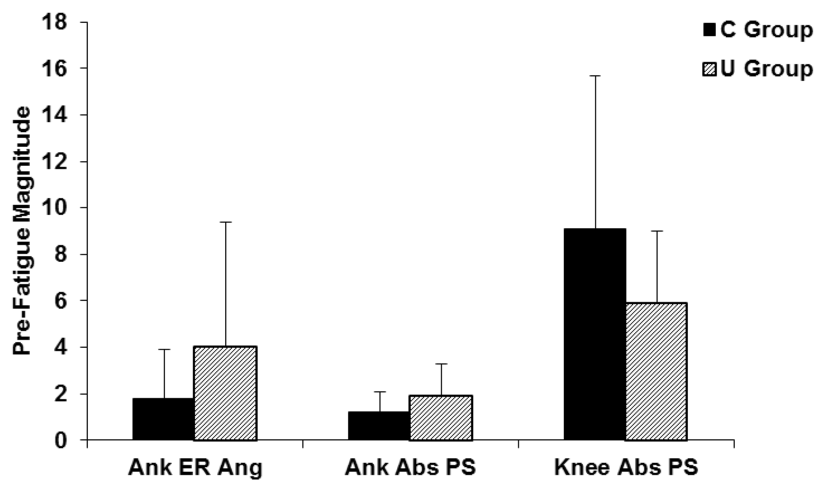

Figure 3.5. Pre-fatigue mean and SD of stance phase measures during forward drop landings. Angle ("Ang") and power ("P") values are in units of degrees and W/kg, respectively. Group differences were significant (p < 0.05) for each measure.

During the landing phase, the two groups had significantly different changes in peak ankle Abs P_T ($F_{(1,23)}$ = 4.79, p = 0.039), peak knee AB angle ($F_{(1,23)}$ = 7.26, p = 0.013), and peak normalized knee EX moment ($F_{(1,23)}$ = 6.60, p = 0.017). Following fatigue, the C group showed increased peak ankle Abs P_T by 0.1 W/kg (20%), whereas it decreased by 0.2 W/kg (33%) in the U group. The C group increased peak knee AB angle with fatigue (0.5 deg; 24%), but the U group showed an increase of 1.9 deg (83%). Both groups increased peak normalized knee EX moment following fatigue, but the magnitude of change was greater for the C group (1.7 Nm/kg; 27%) than for the U group (0.4 Nm/kg; 8%).

During the stance phase, group differences were found for fatigue-induced changes in peak normalized sagittal plane joint moments at both the ankle (DF: $F_{(1,24)} = 7.47$, $p = 0.012$; PF: $F_{(1,24)} = 4.76$, $p = 0.039$) and knee (EX: $F_{(1,24)} = 4.51$, $p = 0.044$; FL: $F_{(1,24)} = 4.65$, $p = 0.041$) as well as peak normalized ankle IR moment ($F_{(1,23)} = 6.29$, $p = 0.019$). With fatigue, both groups increased peak normalized ankle DF and knee EX moments and decreased peak normalized ankle PF and knee FL moments; these changes, however, were more substantial in the C group (Figure 3.6). In the case of peak normalized ankle IR moment, the two groups had fatigue-induced changes in opposite directions.

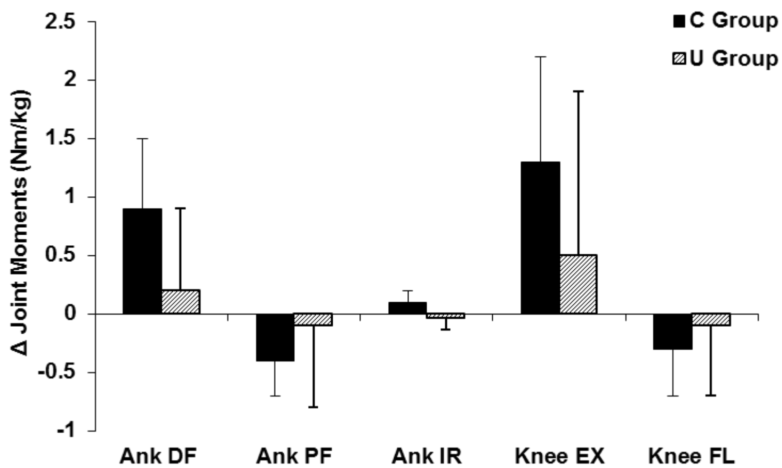

Figure 3.6. Mean fatigue-induced changes in normalized joint moments during the stance phase of forward drop landings. Group differences were significant ($p < 0.05$) for each measure.

Group x covariate interaction effects were found for stance phase peak ankle Abs P_T ($F_{(1,23)} = 6.17$, $p = 0.021$), peak knee Gen P_S ($F_{(1,24)} = 10.47$, $p = 0.0035$), peak knee Abs P_F ($F_{(1,24)} = 17.73$, $p = 0.0003$), and peak knee Abs P_T ($F_{(1,24)} = 32.81$, $p < 0.0001$). For peak ankle Abs P_T and peak knee Gen P_S, a significant covariate effect was found only for the U group ($p = 0.026$ and $p = 0.005$). As pre-fatigue levels of ankle Abs P_T increased, the U group change scores transitioned from small increases to small reductions post-fatigue. As pre-fatigue knee Gen P_S increased for the U group, change scores declined from an increase of roughly 12.5 W/kg (at initial value of 0) to a decrease of about 10 W/kg (at initial value of about 40 W/kg). The covariate effect on peak knee Abs P_F was significant only for the C group ($p = 0.026$), and among that group those with higher pre-fatigue knee Abs P_F had larger reductions in post-fatigue change scores. While larger pre-fatigue knee Abs P_T in the C group was associated with larger increases following fatigue (p =0.0009), larger initial values in the U group were associated with decreased post-fatigue measures.

<u>Side Drop Landing Kinetics and Kinematics</u>

Significant group differences were found for several pre-fatigue measures. At GC, the U group landed with 3.4 (3.4) deg knee AD, which was significantly larger ($t_{(99)}$ = 2.14, p = 0.035) than the 2.3 (2.2) deg exhibited by the C group. During the landing phase, peak ankle ER angle was greater for the U group (1.2 deg; $t_{(124)}$ = -2.26, p = 0.026) than the C group (0.04 deg). The C group had greater peak normalized EV moment (5.1 Nm/kg; $t_{(123)}$ = 3.52, p = 0.0006) and peak ankle Gen P_F (17.1 W/kg; $t_{(124)}$ = -2.56, p = 0.012) compared to the U group values of 3.3 (3.0) Nm/kg and 11.8 (12.1) W/kg. Pre-fatigue peak ankle Abs P_T (0.6 W/kg) during landing was greater for the U group (0.6 W/kg) than the C group (0.3 W/kg). At the knee, pre-fatigue group differences included greater peak knee AD angle ($t_{(105)}$ = 2.04, p = 0.044) and normalized knee EX moment ($t_{(89)}$ = 3.06, p = 0.003) for the U group (13.5 deg and 0.8 Nm/kg, respectively) compared to the C group (10.8 deg and 0.3 Nm/kg). The C group, however, had greater peak normalized knee AB moment ($t_{(107)}$ = 3.01, p = 0.032), at 6.2 (2.5) Nm/kg compared to the U group's value of 4.6 (3.4) Nm/kg. Group differences were also found for landing phase measures of peak knee Gen P_S ($t_{(81)}$ = 3.05, p = 0.0031) and peak knee Abs P_F ($t_{(114)}$ = -2.09, p = 0.039). Both measures were greater for the U group (4.9 W/kg and 0.6 W/kg, respectively) than for the C group (1.6 W/kg and 0.2 W/kg).

Group differences in pre-fatigue stance phase measures at the ankle included peak ankle EV angle ($t_{(125)}$ = -2.58, p = 0.011), peak ankle ER angle ($t_{(124)}$ = -5.06, p < 0.0001), peak normalized PF moment ($t_{(124)}$ = 2.37, p = 0.019), peak normalized ankle EV moment ($t_{(122)}$ = 5.54, p < 0.0001), peak ankle Abs P_S ($t_{(112)}$ = 4.07, p < 0.0001), and peak ankle Gen P_F ($t_{(115)}$ = -3.66, p = 0.0004; Figure 3.7). Stance phase pre-fatigue group differences at the knee were found for peak knee AD angle ($t_{(111)}$ = 2.64, p = 0.0096), peak knee ER angle ($t_{(98)}$ = -2.44, p = 0.016), and peak knee AB moment ($t_{(120)}$ = 4.38, p < 0.0001; Figure 3.8).

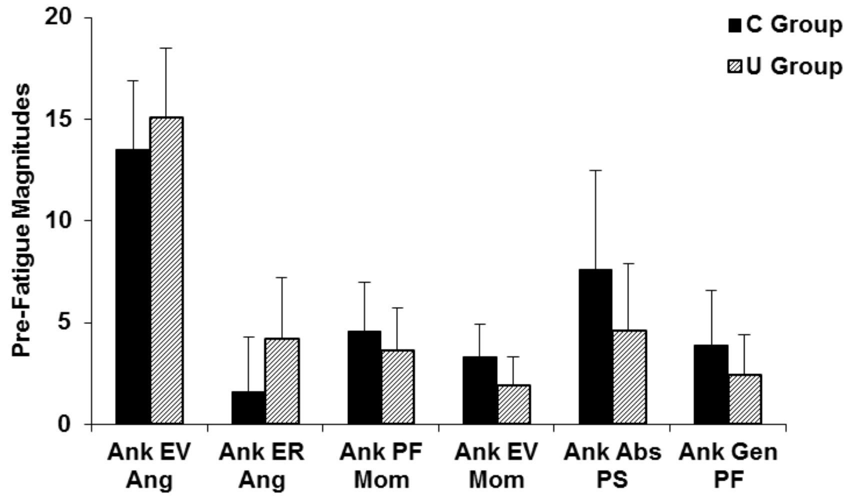

Figure 3.7. Pre-fatigue mean and SD of stance phase ankle measures during side drops. Angle ("Ang"), normalized joint moment ("Mom"), and power ("P") values are in units of degrees, Nm/kg, and W/kg, respectively. Group differences were significant ($p < 0.05$) for each measure.

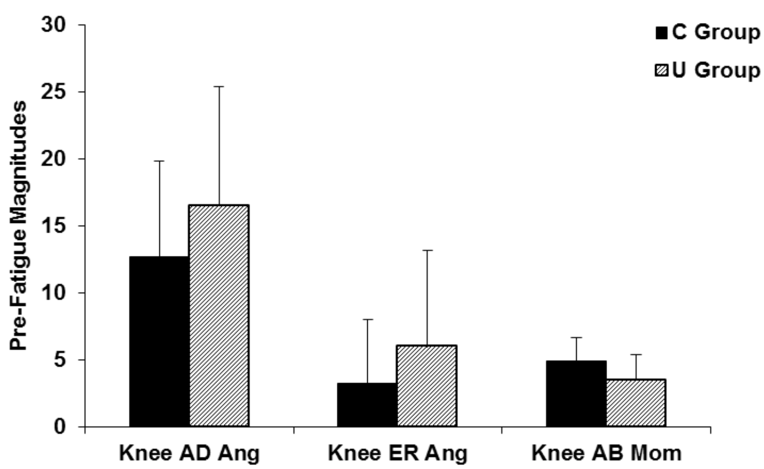

Figure 3.8. Pre-fatigue mean and SD of stance phase knee measures during side drop landings. Angle ("Ang") and normalized joint moment ("Mom") are in units of degrees and Nm/kg, respectively. Group differences were significant ($p < 0.05$) for each measure.

Significant group differences in landing phase change scores were found for peak normalized knee EX moment ($F_{(1,24)} = 11.65$, $p = 0.0023$) and peak knee Gen P_S ($F_{(1,24)} = 16.39$, $p = 0.0005$). Following fatigue, the C group increased (0.4 Nm/kg, 133%) peak normalized knee EX moment, while in the U group it decreased by 0.02 Nm/kg (2.5%). Peak knee Gen P_S increased by 1.9 W/kg (118%) in the C group but decreased by 0.7 W/kg (15%) in the U group. Peak knee Gen P_T during the stance phase increased in both groups (Figure 3.7), but the increase was larger (2.8 W/kg, 44%; $F_{(1,24)} = 16.39$, $p = 0.0005$) in the U group than in the C group (0.1 W/kg, 3%).

Significant group x covariate interactions were found for landing phase measures of peak ankle DF angle ($F_{(1,24)} = 5.29$, $p = 0.03$), peak normalized knee EX and IR moments ($F_{(1,24)} = 20.38$, $p = 0.0001$; $F_{(1,23)} = 4.79$, $p = 0.039$), and peak knee Gen P_S ($F_{(1,24)} = 32.93$, $p < 0.0001$). Covariate effects of peak ankle DF angle and normalized IR moments were only significant in the U group. As initial values increased, post-fatigue change scores were associated with greater reductions in peak DF angle ($p = 0.012$) but greater increases in peak normalized knee IR moment ($p = 0.011$). The groups demonstrated opposite trends regarding covariate effects of peak normalized knee EX moment and peak knee Gen P_S. For both measures, greater initial values were related to larger ($p = 0.0031$; $p = 0.0012$) post-fatigue change scores in the C group and smaller changes ($p = 0.047$; $p = 0.0009$) in the U group. Covariate interactions were also found for stance phase measures of peak ankle Abs P_F ($F_{(1,24)} = 6.86$, $p = 0.015$), and ML COP ($F_{(1,24)} = 4.78$, $p = 0.039$). In the U group, higher initial values of peak ankle Abs P_F were associated with increases in post-fatigue changes ($p = 0.0033$). Also within the U group, higher initial values of ML COP were associated with greater reductions post-fatigue. Interactions at the knee are omitted because a consistent trend was not observed (e.g. most points were clustered with only a 2-3 points defining the line) or they failed to be significant when covariate effects were tested for each group separately.

Drop Landing EMG

The C group had greater pre-fatigue peak TA (27.5% MaxTA) than the U group (20.9% MaxTA; $t_{(125)} = -2.64$, $p = 0.0092$) during the pre-land phase of forward drops. During the landing phase, pre-fatigue peak GS was also significantly greater (37.5% MaxGS; $t_{(128)} = -3.58$, $p = 0.0005$) in the C group than the U group (28.2% MaxGS). Pre-fatigue peak CCR TA/GS ($t_{(122)} = 2.48$, $p = 0.014$) and PL/GS ($t_{(119)} = 3.78$, $p = 0.0002$) during landing, however, were greater in the U group (1.4 and 2.1, respectively) than the C group (1.1 and 1.5, respectively). No significant group differences were found for post-fatigue change scores during forward drop landings. Significant group x covariate interactions indicate that U group change scores for peak pre-land CCR TA/PL ($F_{(1,23)} = 9.43$, $p < 0.0054$) and peak land-phase TA ($F_{(1,24)} = 4.74$, $p < 0.040$) were dependent on pre-fatigue levels. In both cases, a greater pre-fatigue level was associated with a greater reduction of that measure following fatigue.

For side drop landings, the two groups had significant differences in pre-fatigue levels of several measures. Pre-fatigue peak magnitudes of TA ($t_{(113)}$ = -1.98, p = 0.049) and GS ($t_{(125)}$ = -2.02, p = 0.045) muscles during the pre-land phase were significantly greater in the C group (25.6% MaxTA and 57.0% MaxGS) than the U group (20.8% MaxTA and 49.1% MaxGS). Pre-land peak CCR PL/GS, however, was greater in the U group than the C group, at 1.3 (0.7) and 0.9 (0.4), respectively. During landing, the C group had a larger pre-fatigue peak GS ($t_{(117)}$ = -3.80, p = 0.0002) at 39.0 (17.4)% MaxGS than the U group at 28.7 (13.4)% MaxGS. Pre-fatigue peak CCR TA/GS and CCR PL/GS were 0.9 (0.5) and 1.5 (0.6) in the C group but were significantly larger ($t_{(110)}$ = 4.34, p < 0.0001; $t_{(100)}$ = 3.26, p = 0.0015) in the U group, with respectively values of 1.6 (1.1) and 2.0 (1.3). Fatigue-induced changes in peak CCR TA/PL during the landing phase were small (0.03; 4%) in the C group, but the U group had a more substantial decrease (-0.3; 28%; $F_{(1,23)}$ = 6.30, p = 0.020).

3.5 Discussion

3.5.1 Fatigue Protocol #1

The two groups had equivalent pre-fatigue ankle strength in both the PF and DF directions. This was expected, as prior work suggests that ankle instability is not associated with decreased ankle strength (Kaminski et al., 1999; Munn et al., 2003; Kaminski & Hartsell, 2002). Comparable decreases in ankle strength were found during the first fatigue protocol, which indicates that the two groups developed LMF at similar rates when relative effort levels are equal. Further, the two groups exhibited similar correlations between changes in ankle strength and RPE scores. These findings indicate that recurrent ankle sprains in adults with CAI are likely not attributable to either differences in fatigability of muscles controlling ankle stability or a poorer perception of ankle fatigue.

Despite clear losses in ankle strength, EMG MF was ineffective as a measure of fatigue during this particular protocol. Fatigue has been shown to decrease EMG MF during both static and dynamic contractions (e.g., Potvin & Bent, 1997; Masuda et al., 1999; Gutierrez et al., 2007). In these studies, MF was calculated from EMG data collected periodically during continuous performance of either static or dynamic exertions. In our study, however, fatigue was developed using dynamic isotonic exertions, while MVIC efforts were recorded using isometric exertions. These options were selected because pilot work suggested that dynamic exertions helped prevent the foot from losing circulation due to the leg position imposed by the dynamometer, and isometric MVIC exertions provided a consistent, repeatable means of tracking ankle strength loss. Alternating between dynamic isotonic and isometric exertions, however, may have reduced the reliability of fatigue-induced MF changes.

3.5.2 Forward and Side Drop Landings

Kinematic and kinetic measures reported here are similar in magnitude to those reported in other studies using drop-landing tests (Decker et al., 2003; Yeow et al., 2009; Herrington & Munro, 2010; Niu et al., 2011; Yeow et al., 2011). Group differences in drop landing kinematics and kinetics were found for both pre-fatigue measures and post-fatigue change scores. Initial differences in forward drop landings indicated that the U group relied more on the ankle for drop landing execution, as evidenced by greater PF moment and greater sagittal plane ankle power absorption. While reliance on the ankle joint may seem risky for adults with ankle instability, this group also exhibited greater ankle ER, which may actually be a protective mechanism to help resist sprain injury (Garrick, 1977). In contrast, the C group appeared to rely more on the knee for forward drop landing performance, as indicated by greater peak normalized knee EX moment and peak sagittal knee joint powers.

During both of the landing and stance phases, the two groups exhibited similar patterns of post-fatigue changes in forward drop kinematics and kinetics (e.g., increases in knee AB, and peak normalized ankle DF and knee EX moments), but differed in the magnitude of these changes. With regard to study goals, there were no group differences that readily explained one group's predisposition for ankle sprains. Instead, the results observed for stance-phase measures may provide a more general perspective on fatigue-induced motor changes. During stance, the two groups had fatigue-induced increases or decreases in the same joint moment measures, but the magnitude of change was always larger for the C group. It may be that adults with instability exhibit similar biomechanical adjustment in response to fatigue, but perhaps a shortcoming among this group is an insufficient magnitude of that response.

Side drop landings proved to be a more complex task, in that a predominant ankle or knee joint strategy could not be readily identified for either group. Instead, the groups were more clearly distinguished by the planar components of joint dynamics. Specifically, the C group exerted more control in the frontal plane, as evidenced by greater peak normalized ankle EV moment and peak ankle Gen P_F, whereas the U group instead exerted more control in the transverse plane (greater ankle ER angle and greater ankle Abs P_T). At the knee, frontal plane joint moments dominated the landing control strategy for both groups, but frontal plane joint moment (knee AB) was greater for the C group. Considering the ankle and knee together, it appears that the U group avoids frontal plane ankle control at the ankle, presumably to reduce the risk of inversion injury, and instead opts to dissipate frontal plane energy at the knee. During stance phase, the U group exhibited greater frontal and transverse plane joint angles (ankle EV, ankle ER, knee AD, knee ER), but joint stabilization moments were greater for the C group and occurred in the sagittal and frontal planes. Note that the C group utilized a sagittal and frontal plane control strategy for both joints, while the U group used sagittal and frontal plane strategies at the knee and a transverse plane strategy at the ankle. This more complex stabilization method for adults with instability may increase the likelihood of other lower extremity injuries (Yeow et al., 2009). Post-fatigue group differences showed that the C group increased sagittal plane knee power absorption and peak normalized knee EX moment. Such increases were expected, as the knee has been identified as the primary shock absorber during drop landings (Decker et al., 2003; Yeow et al., 2009). The U group, however, exhibited decreases in these measures and an increase in knee Gen P_T. Joints within the lower extremity have specific capacities for energy dissipation in each anatomical plane (Yeow et al., 2009), and thus attempts by those with CAI to protect the ankle may result in biomechanical changes that put other joints at risk for injury.

Pre-fatigue covariates used here represent behaviors when drop landings are executed under normal, rested conditions, and in several cases these were associated with the magnitudes of fatigue-induced effects on behaviors. Several group x covariate interactions were found, and may be of interest in future application. More specifically, for some measures the association between pre-fatigue behaviors and fatigue-induced behaviors were substantially different between groups. These measures might thus be of use in predicting which individuals might be most adversely affected by fatigue, and thereby estimate ankle sprain risk. Future work is needed, though, to substantiate these outcomes (e.g., using larger sample sizes and prospective studies).

3.5.3 EMG

Pre-landing EMG measures describe the neuromuscular preparation for ground impact, while landing phase measures describe the initial attempt at ankle stabilization. EMG amplitudes and CCRs during self-initiated drops are thought to be voluntary and anticipatory rather than reflexive (Santello, 2005; Fu & Hui-Chan, 2007). That is, EMG amplitudes and CCRs are affected by a participant’s perception of the drop height and their expected time to ground contact (Santello & McDonagh, 1998). Here, the C group had greater peak TA during pre-landing, but the U group had greater CCR TA/GS and CCR PL/GS during the landing phase for both drop types. Greater peak TA during pre-landing is considered a protective mechanism (Niu et a., 2011), and the C group was likely better prepared for rapid ankle torque production at the instant of ground contact.

To interpret CCR results, recall that a CCR = 1 indicates equal normalized contraction amplitudes between the two muscles defined by the ratio. As CCR TA/GS and CCR PL/GS during landing phase were closer to 1 in the C group, those with stable ankles demonstrated more equal levels of co-contraction between antagonist (TA/GS) and agonist (PL/GS) muscle pairs. Those with unstable ankles, however, relied more on contraction of the TA and PL muscles to control landing compared to the GS muscle. Covariate effects also suggested that those with instability who exhibit greater pre-fatigue levels of CCR TA/PL or peak TA during forward drop landings will have the greatest decrease in these measures following fatigue. Similar results were obtained for the side drop landings, where CCR TA/PL decreased by about 28% in the U group while remaining more consistent (increase of 3%) in the C group. The potential importance of this effect is evident when considering that pre-fatigue levels of CCR TA/PL in the U group (Appendix B, Table B14) were about 0.5-0.9 depending on drop type and phase. A decrease in this measure indicates a reduction of the relative contraction level of the TA muscle, and therefore a greater dependence on the PL muscle to generate the necessary reactionary ankle torque during impact. As PL reaction times are delayed in adults with instability (Lofvenberg et al., 1995; Mitchell et al. 2008; Kavanaugh et al., 2012), this shift to a greater dependence on the PL muscle following fatigue may contribute to ankle sprain reinjury risk among individuals with unstable ankles.

3.5.4 Limitations

A primary limitation of this study is that the fatigue protocols produced LMF rather than whole-body fatigue. Whole-body fatigue is arguably more relevant to sports injuries, but LMF offered control over the level of fatigue experienced between individuals and ensured that any observed effects were due to fatigue of muscles directly controlling stabilization of the ankle joint. Another limitation, as discussed in Chapter 2, was the use of a marker-based motion capture system for 3D kinematics. Movement of skin relative to anatomical landmarks may cause errors in joint angle calculations (Karlsson & Tranberg, 1999; Benoit et al., 2006). Efforts were made to standardize subject calibration, and segment and joint coordinate systems and rotational matrices were defined in the same manner for all participants. Although joint kinematics reported here may not be exact anatomical angles, they are likely a good approximation and are similar to results reported elsewhere (Winter, 1983; Ferber et al., 2003; McLean et al., 2005; Schache et al., 2011).

Other potential limitations include mental fatigue during the fatigue protocols, the use of surface EMG electrodes, and the sample size. Measures of MVIC recorded at each observation time are subject to some error, as participant motivation and mental fatigue can compromise physical performance (Marcora et al., 2009). To minimize the effects of mental fatigue during the fatigue protocols, verbal motivation was provided here, and multiple MVICs were recorded. In Fatigue Protocol #2, participants were intentionally unaware of the threshold at which the protocol would cease. This was intended to reduce the risk that they would underperform to end the protocol early. Still, individual differences in personal motivation and self-perception of physical ability may have affected MVIC measures. Surface EMG electrodes also introduced a potential limitation. Individual differences in thickness of overlying tissues and variations in musculature could have an effect on the signal quality. To minimize these effects, EMG placement was standardized, and measures were normalized to individual maximums before comparing between groups. Lastly, several dependent measures had effects that approached significance (See Appendix B). A larger number of participants may have helped to confirm statistically significant differences between groups.

3.6 Summary

Strength loss measured during Fatigue Protocol #1 showed that adults with and without ankle instability experience LMF at similar rates when exercising at the same relative effort level (%MVIC). RPE scores and RPE/strength loss correlations also showed that the two groups were similar in their subjective perception of fatigue and in the relationships between these perceptions and decreases in strength. Pre-fatigue comparisons of drop landing kinematics and kinetics revealed different joint stabilization strategies for the two groups. Generally, adults with ankle instability relied more on transverse plane joint control, whereas those with stable ankles exerted more control in the sagittal and frontal planes. During forward drop landings, the two groups also differed in the relative use of the ankle and knee for stabilization. Post-fatigue differences indicate that sprain injuries may be attributed to differences in the magnitude of biomechanical changes following fatigue. EMG measures also revealed significant group differences in muscle activation during drop landing maneuvers. Those with stable ankles exhibited more equivalent levels of co-contraction between muscle antagonists and agonists supporting the ankle. Those with CAI, however, had relatively more activation of the TA and PL muscles as opposed to the GS muscle. When fatigued, those with instability may also transition to a greater reliance on PL muscle contraction for controlling initial ground contact, which may predispose them to ankle inversion injuries. Generally, results indicate that LMF can lead to different motor control adaptations or strategies between groups with and without ankle instability. However, specific fatigue-induced changes and their relevance to sprain injury may also depend on the activity being performed (e.g., forward vs side drop landings).

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Chapter 4

Effect of Localized Ankle Muscle Fatigue on the Stiffness
and Proprioceptive Contribution of Braces

4.1 Abstract

Ankle bracing is a common treatment to reduce ankle sprain recurrence among adults with chronic ankle instability (CAI). Bracing is thought to be effective by providing external joint stiffness and/or improving proprioceptive acuity, thereby helping the user to resist excessive ankle range of motion and have better perception of joint orientation. It is unknown, however, if these posited benefits of ankle bracing are affected when an athlete is fatigued. As fatigue develops, neuromuscular changes occur that oppose the positive effects of bracing. Namely, joint stiffness may be reduced and joint position sense (JPS) declines. The goal of this study was to assess whether braces maintain their effectiveness when an athlete becomes fatigued. Two test braces were selected to enable comparisons between different brace types. JPS and ankle stiffness were measured in a group of adults with CAI, both before and after localized muscle fatigue, in three bracing conditions: no brace, neoprene wrap brace (NW), and a semi-rigid brace (AC). For all brace conditions, absolute JPS errors were significantly increased following fatigue. True JPS error measures, however, suggested that fatigue-induced changes in JPS may differ between portions of the ankle range of motion. Specifically, true JPS error showed the most improvement in the range of small plantarflexion angles while the greatest decrement in JPS error was observed in the range of small dorsiflexion angles. Stiffness results were often gender-specific. Among males, the AC brace resulted in a slight increase in post-fatigue stiffness, and which differed from decreases observed in the NB and AW conditions. Among females, no post-fatigue differences in stiffness were observed. Overall, these results suggest that braces had little influence on joint proprioception in the presence of ankle fatigue, that the AC brace afforded more protection for males, and that either brace type may be appropriate for females. This information may help users make a more informed decision when selecting an ankle brace, though further research is needed to compare across more diverse brace types.

4.2 Introduction

Ankle sprains are a recurrent problem among adults with chronic ankle instability (CAI). Recurrent sprains are often attributed to mechanical or proprioceptive deficits resulting from ligament or nerve damage caused by an initial sprain injury (Hertel et al., 2002; Tropp, 2002;

Bonnel et al., 2010). These deficits likely impede the detection of hazardous ankle joint positioning (e.g., excessive inversion or plantarflexion) and the development of an adequate muscular response to resist ankle inversion. Often, these repeated sprains are of a greater severity than the initial injury (Hawkins et al., 2001) and cause substantial pain and physical activity restriction with each episode.

Among adults with stable ankles, localized muscle fatigue (LMF) can reduce the accuracy of joint position sense (Forestier et al., 2002; Mohammadi & Roozdar, 2010) and decrease voluntary ankle torque capacity (Gutierrez et al., 2007). In adults with CAI, whose joint position sense is already compromised (Jerosch & Bischof, 1996; Hartsell, 2000), fatigue may increase the risk of an ankle sprain injury. In addition to joint position sense and voluntary torque capacity, joint stiffness is also relevant to sprain injury risk, in that it helps to restrict excessive joint motion and to limit tissue strain in response to a perturbation (Duan et al., 1997; Wagner et al., 1999). Results are mixed, however, regarding the effect of fatigue on ankle stiffness. One study suggested that LMF leads to increased antagonistic co-contraction to retain ankle stiffness (Gregory et al., 1998). Others have reported that fatigue causes a decrease in ankle stiffness (Kuitunen et al., 2002; Duquette & Andrews, 2010), possibly due impaired cross-bridge formation from exercise-induced muscle damage (Kuitunen et al., 2002). More research is thus needed to clarify the specific role of fatigue in ankle sprain injuries.

Treatment of unstable ankles typically involves the use of ankle braces. In general, ankle braces can provide external rigid support and/or supplemental cutaneous sensory input to increase mechanical stiffness and reduce ankle range of motion (Eils et al., 2002; Papadopoulous et al., 2005; Cordova et al., 2007; Zinder et al., 2009) and/or improve proprioceptive acuity (Feuerbach et al., 1994; Heit et al., 1996; Hartsell, 2000). The specific benefits of any particular device, though, may depend on its design and construction materials. Only a few studies, however, have quantified the relative stiffness or proprioception benefits afforded by different brace types. Zinder et al. (2009) reported that bracing does not lead to neuromuscular changes in stiffness but provides passive, mechanical stiffness. Others have simply reported that various brace types are effective at reducing ankle range of motion in response to inversion perturbation (Eils et al., 2002). Further, no studies have quantified both stiffness and proprioceptive characteristics, with a goal of identifying the relative benefits afforded by different brace types, though such information might be useful in the selection of ankle braces. Additionally, and because many users of ankle braces are athletes, it is important to consider if and how these positive effects of ankle bracing (improved stiffness and proprioception) are affected by the presence of fatigue which, as discussed, can reduce joint position sense, voluntary muscle strength capacity, and cause changes in physiological ankle stiffness.

The goal of this study was thus to determine whether the noted potential benefits of bracing (increased stiffness and enhanced proprioception) are reduced when there is LMF at the ankle. Ankle proprioception and stiffness were evaluated both before and after a LMF protocol in three different ankle brace conditions (no brace = control, a wrap-style brace, and a rigid shell brace). It was hypothesized that: 1) prior to fatigue, both brace types would improve proprioception and stiffness compared to the control condition; and 2) the relative improvements in proprioception and stiffness provided by a given brace would decrease following the development of ankle LMF. This work was intended to improve our understanding of the mechanisms involved in bracing support during athletic performance, and to help inform brace selection in the future.

4.3 Methods

4.3.1 Participants and Overview of Experimental Design

Twelve young adults (5 males, 7 females) with self-reported ankle instability completed the study. Mean (SD) age, stature, and body mass were 24.1 (3.5) years, 172.8 (11.6) cm, and 71.0 (13.6) kg, respectively. Inclusion required participants to have a history of at least two inversion sprains to the same ankle and answer in the affirmative when asked if they “ever experienced their ankle rolling or giving way during activity”. Participants were also screened using the Cumberland Ankle Instability Tool (CAIT), and were eligible if they had a CAIT score ≤ 27 (Delahunt et al., 2010). The recruited group had scores with mean (SD, range) of 18.7 (3.9, 12 – 25). Other than a history of ankle sprains, participants were excluded if they reported any chronic joint problems (e.g., pain, muscle weakness), prior joint surgeries, limb neuropathies, balance disorders, current participation in physical rehabilitation, or regular use of ankle tape or braces. If both ankles met the eligibility criteria, the ankle with the lowest CAIT score was used as the test ankle. Protocols were approved by the VT Institutional Review Board, and participants provided both written and verbal consent prior to beginning the experiment.

Participants completed three test sessions, on different days, during which one of three brace conditions was used: no brace (NB), a neoprene ankle wrap brace (AW; Model #4547, Mueller Sports Medicine, Inc., Prairie du Sac, WI), and an Aircast® brace (AC; Air Stirrup Model #02XX, DJO Global, Inc., Vista, CA). Specific braces were purposefully selected to include designs composed of neoprene and rigid plastic materials. During each of the sessions, joint position sense and ankle stiffness were measured, both before and after a fatigue protocol. Two sets of the six possible orders of brace conditions were randomly assigned to the 12 participants. An overview of the procedures in each session is provided in Figure 4.1, and which are described in more detail subsequently. The order in which the joint position sense and stiffness tests were performed was counterbalanced across participants.

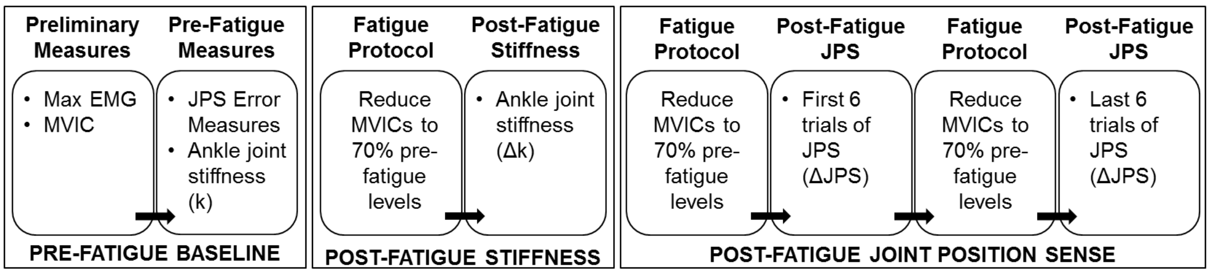


Figure 4.1. Overview of the elements in each experimental session.

4.3.2 Preliminary Measurements and Instrumentation

At the start of each session, maximal voluntary ankle torque was measured for each participant and used subsequently to customize the fatigue protocol. Participants were seated at a commercial dynamometer (Biodex System 3, Biodex Medical Systems, Inc., Shirley, NY) in the ankle plantar/dorsiflexion (PF/DF) configuration. Positioning was standardized for all participants as follows. With the ankle in neutral position, the ankle joint center was aligned with the dynamometer's axis of rotation, the chair was reclined 20 degrees from vertical, and knee flexion was set to 50 degrees. A lap belt was secured across the hips, and additional straps were used to secure the thigh to a support pad and to hold the foot firmly to a foot plate. A stool was provided on which the non-test leg was rested. With the ankle held in the neutral position, multiple three-second maximal voluntary isometric contractions (MVICs) were performed in both the plantarflexion (PF) and dorsiflexion (DF) directions. Rest periods of ~30-60 seconds was provided between consecutive attempts in the same direction, and efforts were repeated until MVIC measures plateaued. Throughout all dynamometer protocols, a counterweight was applied to offset the weight of the foot and foot plate and to ensure that PF and DF motions were performed with an equivalent level of perceived effort (as determined in pilot work). Pairs of surface electromyography (EMG) electrodes (Model # A10012-5S, Vermed, Inc., Bellows Falls, VT) were applied to the skin surface over the tibialis anterior (TA), peroneus longus (PL), and lateral gastrocnemius (GS) muscles following procedures described elsewhere (Soderberg et al., 1991; Fu & Hui-Chan, 2007; Gutierrez et al., 2007). Baseline EMG measures were collected for 5-10 seconds with the participant in a standing posture. EMG measures were used to monitor muscle activation levels immediately preceding ankle stiffness tests as described below.

4.3.3 Joint Position Sense

Ankle joint position sense (JPS) was tested using the dynamometer in the PF/DF configuration (Figure 4.2), and was measured using a passive/active protocol. Range-of-motion (ROM) end stops were set to either ±20 degrees or ±25 degrees from the neutral position depending on the participant’s dorsiflexion limit. For each trial, the participant’s ankle was passively rotated to a reference angle at 2 degrees/sec. The dynamometer held the reference angle for ~10 seconds, and then the participant was given active control. Against a small isotonic load (0.68 Nm), the participant was asked to actively move their ankle to the opposing ROM end stop and then reposition their ankle back to the reference angle. That is, if the reference angle was perceived to be PF, they were asked to move their ankle to the DF end stop, and vice versa. When the participant felt the appropriate reposition angle had been reached, they pressed a stop button to hold the dynamometer and end the trial.

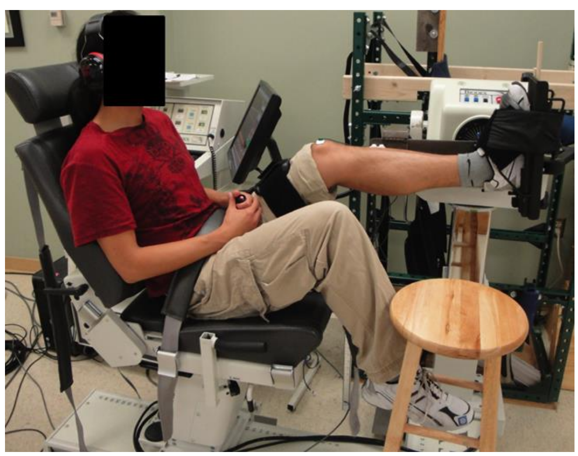


Figure 4.2. A participant shown in the dynamometer prior to beginning a joint position sense test.

Participants practiced the JPS test until they demonstrated an understanding of the protocol and expressed confidence in their control of the dynamometer. All trials were completed with a blindfold and headphones to eliminate supplemental visual or audio cues, respectively. Three small magnitude (5-8 degrees) and three large magnitude (13-16 degrees) reference angles

(Forestier et al., 2002) were tested in both the PF and DF directions, for a total of 12 trials. The sequence of the reference angles was randomized for each participant, and angle data were recorded at 1024 Hz. The specific magnitudes of small and large reference angles were randomized between sessions. In the following, small PF and DF reference angles are termed “SP” and “SD”, while large PF and DF references angles are termed “LP” and “LD”.

4.3.4 Joint Stiffness

Ankle joint stiffness was measured using a transient oscillation (perturbation) test, similar to that described by Zinder et al. (2007). Participants stood with their feet hip-width apart and with their weight evenly distributed between a force plate (AMTI OR-6-7-1000, Advanced Medical Technology, Inc., Watertown, MA) and a rocking cradle platform (Figure 4.3). The cradle had a central axis that was mounted at either end to an aluminum frame, and with bearings to permit free rotation. In-line with this axis was a linear potentiometer (5KΩ, Taiwan Alpha Electronic Co., LTD, Taoyuan, Taiwan) that was used to measure angular orientation of the cradle. The antero-posterior axis of the ankle was aligned with the cradle’s axis, and foot placement was adjusted such that the flat surface of the cradle rested horizontally.

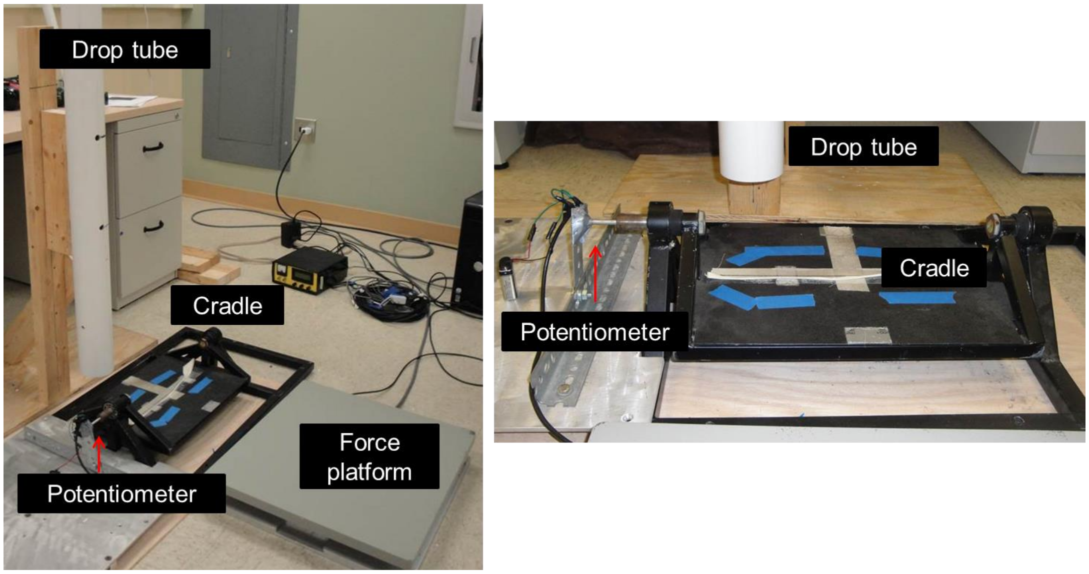


Figure 4.3. Experimental set up for testing ankle joint stiffness.

Perturbations were induced by dropping a 5.08-cm diameter solid metal ball (mass = 0.437 kg, E52100 alloy steel, Grade 50, McMaster-Carr, Robbinsville, NJ) from a height of 1.2 m onto the rear lateral edge of the cradle. The ball was released through a PVC-pipe drop tube to avoid contact with the participant. At maximum displacement, the cradle allowed only 22 degrees of ankle rotation in either direction to prevent ankle inversion injury. Ten perturbations were performed at random intervals over a period of ~3 minutes. Participants were asked to stand in a relaxed state, with gaze directed forward, and to make no attempt to resist the oscillations created by the perturbations. Headphones were worn to minimize auditory cues and anticipation of the ball drop. Throughout the protocol, force plate, EMG, and potentiometer signals were sampled at 2048 Hz. EMG signals were hardware amplified and bandpass filtered between 10 and 500 Hz (Measurement Systems, Inc., Ann Arbor, MI). Real-time visual displays of weight distribution and muscle activity (RMS amplitude) were monitored to ensure that: 1) participants maintained an even weight distribution on both feet (±2.5% body weight); and 2) the ankle muscles were not substantially pre-activated, defined as ≥2 SD over baseline levels (Konradsen et al., 1998; Hopkins et al., 2007) at the time of the perturbation.

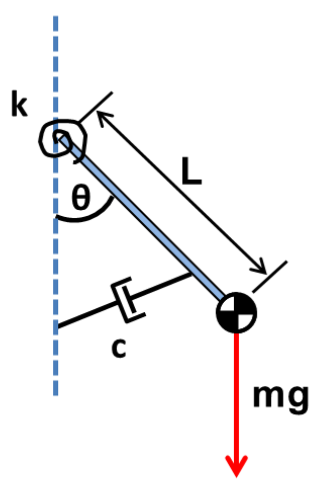
4.3.5 Fatigue Protocol

LMF at the ankle was induced by completion of isotonic PF and DF exertions, using the dynamometer through a range of motion (ROM) spanning 15 degrees of DF and 30 degrees of PF. Visual feedback was provided to ensure all participants adhered to the same work/rest pattern. Exertions were performed at a rate of 12 cycles/min with each cycle involving full movement from one ROM limit to the other and returning. Isotonic loads were initially set to 70% and 30% of PF and DF MVIC, respectively, and were selected after pilot work determined they were difficult but sustainable for several minutes. After two minutes of isotonic exertions, the ankle was returned to the neutral position and MVICs were completed. MVICs were performed alternately in both PF and DF to allow for a brief rest period between exertions in the same direction. For each MVIC, participants were asked to ramp up to their maximal effort over a period of ~ 1 second and sustain this contraction level for ~ 3 seconds. Four MVIC attempts were completed for each direction over a period of 1 minute. Based on the MVIC magnitudes, isotonic loads were adjusted in a manner similar to Davidson et al. (2009) and isotonic exertions were resumed. This process was repeated until both PF and DF MVICs were reduced to ~70% of their pre-fatigue levels. This fatigue protocol was repeated during a session (see Figure 4.1) to ensure a consistent level of fatigue across all post-fatigue measures.

4.3.6 Data Processing and Dependent Measures

Angular data from JPS trials were low-pass filtered using a dual-pass, 4^{th}-order Butterworth filter with a cutoff frequency of 3 Hz. Absolute and true errors (Jerosch & Bischof, 1996; Willems et al., 2002) were derived as dependent measures, respectively defined as the absolute value and the actual (signed) value of the difference between reposition and reference angles. Mean absolute error describes the typical magnitude of the positioning error, whereas mean true error describes whether there was a tendency to overshoot (positive true error) or undershoot (negative true error) the reference angle during the repositioning attempt (Jerosch & Bischof, 1996; Willems et al., 2002).

Ankle joint stiffness was determined using a 2nd-order dynamics modeling approach (Granata et al., 2002; Docherty et al., 2004; Zinder et al., 2007). Cradle angles (from the potentiometer) were first smoothed using a 4th-order, zero-lag Butterworth filter with a low-pass cutoff of 25 Hz. Then, the foot + cradle system was modeled as a simple pendulum (Figure 4.4) consisting of a point mass suspended at a length *L* of 7.4 cm (height of the cradle), and with spring and dashpot elements to represent ankle stiffness (*k*) and a damping coefficient (*c*). Mass of the pendulum, *m*, was the sum of the mass of the cradle (1.66 kg) and the mass of the foot as estimated from anthropometric tables (Dumas, 2007). The inertia (*I*) of the system was calculated using the formula for a point mass (mL^2). A second-order equation of motion was developed describing the pendulum's angular acceleration ($\ddot{\theta}$) as a function of angular position (θ), angular velocity ($\dot{\theta}$), and model parameters.



(Equation 4.1)

$$\ddot{\boldsymbol{\theta}} = \frac{-mgLsin\boldsymbol{\theta} - k\boldsymbol{\theta} - c\dot{\boldsymbol{\theta}}}{I}$$

Figure 4.4. Simple pendulum model for estimating ankle joint stiffness.

For a given perturbation trial, the values of *k* and *c* for each trial were estimated using a least squares optimization algorithm in Matlab® (R2011a, MathWorks, Inc., Natick, MA). Model simulation began at the instant of peak angular displacement following the perturbation, and initial model conditions included angular position and angular velocity at this time *t*. A 4th-order Runge-Kutta method was used to solve the equation of motion with a time step equal to the inverse of the sampling frequency (2048 Hz). Model simulation time was the duration of two full cycles of oscillation. For larger participants, however, two full cycles were not always available. In those cases, the longest duration (1 – 1.5 cycles) available was used. Previous work has also used single cycles for stiffness calculations (Granata et al., 2004). The cost function for the optimization algorithm was the difference between actual (measured) and predicted (modeled) ankle joint angles. From the least-squares fit, *k* values were obtained as measures of ankle joint stiffness in units of Nm/rad. Model fits (R^2) reported in other studies ranged from 0.92 – 0.98 (Winter et al., 2001; Granata et al., 2004; Zinder et al., 2007). Here, a model fit of $R^2 \geq$ 0.95 was required for inclusion in the final analysis, resulting in about 92% of all trials being used and with at least 6 trials for each participant and brace condition. Of the 60 eliminated trials, 39 (65%) were from females while 21 (35%) were from males. Note, however, that the study included 7 females and only 5 males. Across participants, mean (SD) RMS errors between actual and predicted ankle joint angle were 0.0061(0.0029) rad, or (0.35(0.17) deg, and overall measured stiffness was 16.6 (7.0) Nm/rad. Of the maintained trials, mean model fit was R^2 = 0.98(0.005), and representative results are shown in Figure 4.5. Because the simulation was about 350 ms in duration and therefore included passive, reflexive, and intrinsic stiffness components (Sinkjaer et al., 1988), stiffness measures reported here are referred to as *effective stiffness.*

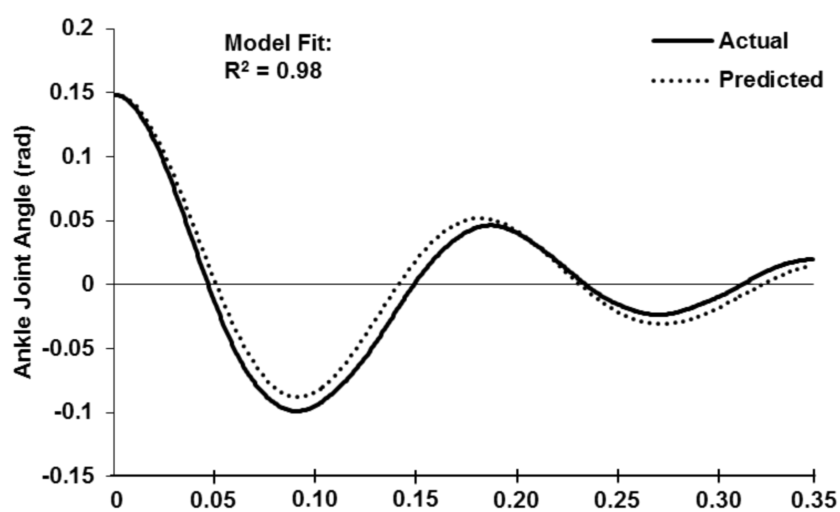


Figure 4.5. Actual oscillation data and corresponding model prediction for a single ankle stiffness perturbation trial.

4.3.7 Statistical Analyses

Statistical analyses were completed using JMP 10.0 software (SAS Institute Inc., Cary, NC). For all statistical tests, significance was determined when $p < 0.05$. Preliminary repeated measures analyses of variance (ANOVAs) were conducted to determine if significant order effects were present for either the pre- or post-fatigue trials for JPS error or effective stiffness measures. No significant order effects were found for JPS error measures ($p > 0.20$), and thus results from all JPS trials were included in subsequent analyses. Similarly, no significant order effects were found for effective stiffness measures ($p > 0.21$), and all qualifying trials (model fit of $R^2 \geq 0.95$) were included in the analyses.

One-way repeated measures ANOVAs were used to test for an effect of brace condition on pre-fatigue JPS error (absolute/true) and effective stiffness measures. Where a main effect was found, post-hoc paired comparisons were done using Tukey's Honestly Significant Difference (HSD) test. Individual pre-fatigue means for each brace condition were used to convert post-fatigue measures to change scores (post – pre). Change scores for each dependent measure were analyzed using one-sample *t*-tests to assess whether there were main effects of fatigue. Analyses of covariance (ANCOVAs) were then used to assess the main effects of brace type on fatigue-induced change scores using pre-fatigue levels as a covariate. Due to previous reports that specific reference angles can affect JPS errors (Glencross & Thornton, 1981; Forestier et al., 2002; Goble, 2010), reference angle was included as an additional factor in ANCOVA analyses of JPS measures. Similarly, gender has been reported to influence joint stiffness (Granata et al., 2002; Padua et al., 2006; Gabriel et al., 2008) and was therefore included as an additional factor in the ANCOVA analysis for effective joint stiffness.

4.4 Results

Complete summary statistics and statistical results for each of the analyses are provided in Appendix C. Results presented here focus on the main and interactive effects of brace and the main effects of fatigue. Where appropriate, main effects of reference angle and gender are also presented.

4.4.1 JPS

Pre-fatigue JPS errors were not significantly affected by brace type, although the brace effect approached significance for pre-fatigue absolute error measures ($F_{(2,414)} = 2.66$, $p = 0.071$). Mean absolute error with the AC condition was 0.5 deg (13%) and 0.7 deg (18%) less than that with NB and AW, respectively (Figure 4.6). Fatigue led to a significant ($t_{(429)} = 2.45$, $p = 0.015$) increase in absolute errors of 0.3(3.0) deg. Fatigue effects on true errors differed depending on reference angle ($F_{(3,401)} = 6.64$, $p = 0.0002$). The SD reference angle was associated with an increase of 0.5(4.1) deg in overshoot error, which was significantly different ($p < 0.0001$) than the 0.8(4.8) deg reduction in overshoot observed with the SP reference angle. The LD reference angle also resulted in a reduction in overshoot (0.1(3.2) deg), but this was significantly less ($p = 0.036$) than the change observed for the SP reference angle.

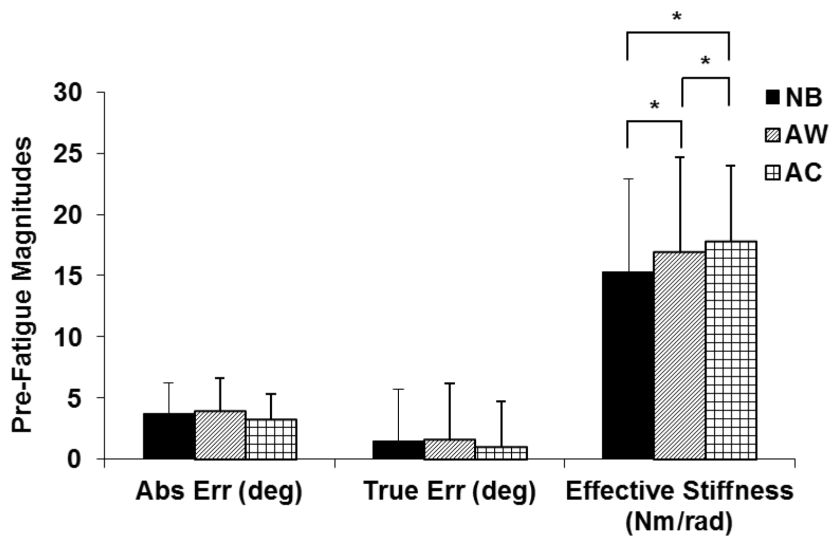


Figure 4.6. Mean and SD of pre-fatigue levels for absolute and true JPS errors and ankle effective stiffness measures. The symbol "*" indicates significant differences ($p < 0.05$).

4.4.2 Stiffness

Pre-fatigue ankle effective stiffness for the AC brace condition (17.8(6.2) Nm/rad) was significantly greater ($F_{(2,313)} = 31.27$; $p < 0.0001$) than the effective stiffness for either the AW (17.0(7.7) Nm/rad) or NB (15.3(7.6) Nm/rad) brace conditions (Figure 4.6). Pre-fatigue effective stiffness with the AW brace was also significantly greater than with the NB brace ($p < 0.0001$). Pre-fatigue levels presented by gender (Table 4.1) reveals that effective stiffness and damping were significantly greater among males ($p < 0.0001$) than females for each brace condition. The main effect of fatigue approached significance ($t_{(329)} = -1.95$, $p = 0.052$), with an overall post-fatigue decrease in effective stiffness of 0.3 (3.0) Nm/rad. Post-fatigue effective stiffness changes were not affected by brace condition alone, but a significant brace x gender effect was found ($F_{(2,318)} = 6.24$, $p = 0.0022$) which revealed that post-fatigue changes in effective stiffness were more substantial for males vs. females (Figure 4.7).

Table 4.1 Mean (SD) of pre-fatigue effective stiffness and damping measures for males and females with each brace condition.

		Pre-Fatigue		
Measures	**Level**	**NB**	**AW**	**AC**
Effective Stiffness, k (Nm/rad)	M	20.7(8.8)	23.3(7.9)	22.9(5.9)
	F	11.4(2.8)	12.2(1.7)	14.0(2.7)
Damping, c (N-s/m)*	M	0.24(0.13)	0.18(0.07)	0.28(0.13)
	F	0.18(0.09)	0.16(0.06)	0.18(0.06)

Damping has units of Newton-seconds/meter.

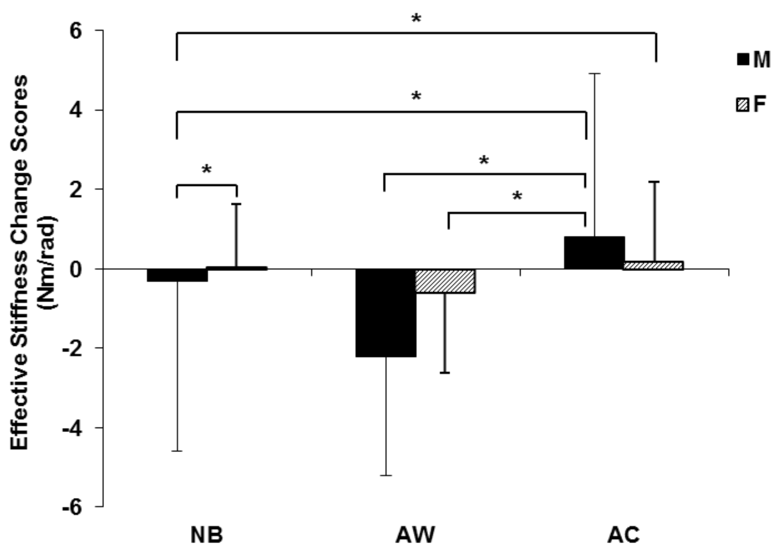


Figure 4.7. Mean and SD of effective stiffness change scores as a function of brace and gender. The symbol "*" indicates significant differences between columns ($p < 0.05$).

Significant brace x covariate effects ($F_{(2,318)} = 7.76$, $p = 0.0005$) showed similar patterns in the NB and AW conditions ($p < 0.0001$), in that greater pre-fatigue effective stiffness levels were associated with greater fatigue-induced decreases in effective stiffness. For the AC condition, however, the relationship was inconsistent, with increases in post-fatigue effective stiffness corresponding to low pre-fatigue effective stiffness for some participants and high pre-fatigue effective stiffness in other cases. The brace x gender x covariate interaction was also significant ($F_{(2,318)} = 14.83$, $p < 0.0001$). For the NB condition, both genders exhibited greater reductions in post-fatigue effective stiffness when pre-fatigue levels were higher ($p < 0.0001$). In the AW condition, pre-fatigue scores had no significant association with fatigue-induced changes in effective stiffness among females ($p = 0.45$), though males showed greater post-fatigue decreases in ankle effective stiffness when pre-fatigue levels were higher ($p < 0.0001$).

In the AC condition, males again showed an inconsistent relationship between pre- and post-fatigue effective stiffness measures. Among females, lower pre-fatigue effective stiffness was associated with small increases in post-fatigue effective stiffness, but higher pre-fatigue effective stiffness values were associated with post-fatigue decreases in effective stiffness.

4.5 Discussion

4.5.1 JPS

Previous work has indicated improvements in joint proprioception with diverse interventions including athletic tape and varying brace designs (Feuerbach et al., 1994; Lohrer et al., 1999; Hartsell, 2000). Because of its neoprene construction and form-fitting contact to the ankle, it was expected that the AW brace would provide the greatest improvement in JPS. However, differences in pre-fatigue JPS errors between brace conditions only approached significance, and actually indicated that the AC brace, rather than the AW brace, best reduced absolute JPS errors. This finding is similar to that of Hartsell (2000), who reported a significant improvement in JPS error with a semi-rigid brace. LMF caused decreases in absolute error here, as has been previously reported (Forestier et al., 2002; Mohammadi & Roozdar, 2010). Neither test brace appeared effective at counteracting the adverse effects of fatigue on JPS.

Reference angles influenced post-fatigue JPS true error. Specifically, larger errors were made following fatigue using small DF reference angles than for small PF reference angles. Forestier et al. (2002) also reported greater error for small DF angles following fatigue. The observed post-fatigue decrease in small PF overshoot errors indicates that participants were able to more closely replicate the reference angle. Thus, this may be seen as a protective change preventing excessive PF, and which could help to reduce the risk of reinjury at ground contact (Konradsen et al., 2000). Although DF angle is not directly related to sprain injury risk, a post-fatigue increase in overshoot error for small DF reference angles may also be viewed as somewhat protective. The ankle is most stable in a “close-packed” position, defined as full dorsiflexion with a flexed knee (Alter, 1996). Perhaps the observed increase in DF overshoot following fatigue reflects neuromuscular changes that cause the ankle to deviate more toward the close-packed position when fatigued. Because these changes were seen across brace types, any changes in JPS performance were likely due to neuromuscular compensations in response to fatigue, rather than supplemental cutaneous input from a given brace condition.

4.5.2 Ankle Stiffness

Here, a kinematic model was used to determine effective stiffness. Some authors have estimated stiffness using a simple ratio of ankle torque to angular displacement (e.g., Winter et al., 2001) while others, using a similar perturbation protocol, related stiffness to the natural frequency of oscillation and externally applied loads (Docherty et al., 2004; Granata et al., 2004; Zinder et al., 2007). Mean stiffness values found here (11 – 24 Nm/rad) are somewhat less than those reported by others (25 – 35 Nm/rad; Docherty et al., 2004; Zinder et al., 2007), though are within the same range (10 – 40 Nm/rad) reported by those studies and elsewhere (Roy et al., 2011). Our analysis, though, was limited to trials in which a high model fit was obtained ($R^2 \geq 0.95$), and the within-session standard error of measurement (SEM) for our current results (0.55 (0.44) Nm/rad) is less than the value of 2.05 Nm/rad reported by Zinder et al. (2007). The current approach may thus be more accurate and repeatable.

Both braces (AW and AC) increased ankle effective stiffness compared to the NB condition. This was somewhat expected, as prior research has suggested that external support ranging from athletic tape to lace-up braces to semi-rigid braces is able to reduce joint range of motion in response to a sudden inversion perturbation (Eils et al., 2002; Cordova et al., 2007). Though the mean difference was small (0.8 Nm/rad), the AC brace did provide greater effective stiffness than the AW brace. This was expected as the AC brace provides a rigid, physical barrier to excessive ankle rotation, whereas the AW brace is considerably more compliant. Kearney & Hunter (1982) found that elastic (stiffness) and viscous (damping) components of an ankle model decrease with increasing displacement amplitude. Thus, while the increases in ankle effective stiffness were different for the AW and AC braces here, testing over a larger range of motion may further distinguish level of effective stiffness provided by these brace types. As reported elsewhere (Granata et al., 2002), males consistently exhibited greater effective stiffness than females for all brace conditions and similar results were obtained for damping.

Fatigue alone did not consistently affect ankle effective stiffness for any brace condition. Instead, interaction effects revealed a complex relationship between brace, fatigue, and gender. Gender differences in lower extremity joint stiffness have been reported elsewhere (Granata et al., 2002; Padua et al., 2006; Gabriel et al., 2008), with males exhibiting greater joint stiffness than females. Similar results were obtained here (Appendix C, Table C4), with males having greater pre-fatigue ankle effective stiffness values than females in each of the brace conditions. Males also had greater post-fatigue changes in effective stiffness in each brace condition, although gender differences were only significant in the NB condition. Both genders showed a post-fatigue decrease in effective stiffness with the AW condition and a slight increase with the AC condition. This latter change in effective stiffness in the AC condition was significant among males, compared to the NB and AW conditions. Two explanations are possible. The first is that physiological ankle effective stiffness decreased following fatigue when using the AW brace but increased when using the AC brace. While previous studies have suggested that males and females may exhibit opposite neuromuscular changes in response to fatigue (Wilson & Madigan, 2007), expecting opposite fatigue-induced changes in neuromuscular measures (stiffness) within in the same gender due to bracing conditions is questionable. Consider that the present post-fatigue changes in effective stiffness in the AW and AC conditions were fairly small, representing changes of 9.4% and 3.5% from pre-fatigue levels, respectively. Perhaps a more plausible explanation of the differences observed among males is that fatigue causes a reduction in physiological stiffness, but the rigidity of the AC brace may help in retaining overall effective stiffness of the ankle/brace complex. Previous research (Zinder et al., 2009) also supports the idea that braces provide a passive, mechanical stiffness rather than provoking neuromuscular changes. As such, the AC brace appears to afford the greatest protection from sprain among males. Among females, however, post-fatigue changes in effective stiffness were not observed. As pre-fatigue levels were higher among females with both brace types vs. the NB condition, the lack of significant fatigue-induced changes may imply that the initial increase in protection is still preserved. Thus, while the AC brace seems more effective among males in terms of added effective stiffness, either the AW or AC brace may provide improved effective stiffness among females.

Significant interaction effects suggest that the pre-fatigue levels of ankle joint effective stiffness were related to the magnitude of post-fatigue changes. Further, the specific relationship between these pre-fatigue levels appeared to differ between genders. The majority of gender/brace combinations demonstrated a decrease in effective stiffness with increasing pre-fatigue effective stiffness. There was, however, a gender difference in the AC condition, with males showing small increases in effective stiffness and females showing greater reductions with higher pre-fatigue effective stiffness levels. These results may reflect gender differences in muscle recruitment following fatigue, though more research is needed to clarify the practical relevance of these effects.

4.5.3 Limitations

As in Chapter 3, a limitation of this study was the use of a LMF protocol rather than a whole body fatigue protocol. Due to the multiple sessions, it was important that the fatigue protocol be controlled and repeatable. The LMF protocol afforded control over the level of fatigue experienced between participants and between sessions for each participant. Whole body fatigue, however, may have had the advantage of greater external validity to sports injury. An additional limitation regarding the fatigue protocol is the potential for personal motivation and mental fatigue to affect measured ankle torque (Marcora et al., 2009). The use of multiple sessions and repeated fatigue protocols within a session made this a relevant concern. Data from the first and second sessions were used as a guide in subsequent sessions. That is, participants were encouraged to complete the same number of rounds in each session, and similar isotonic load adjustments were made (assuming pre-fatigue MVICs were similar) to ensure participants completed a similar amount of work in each session.

Another potential limitation of this study was the construction of the rocking cradle, since the height of the cradle axis was fixed. Small vertical offsets were possible between the cradle axis and the participant's ankle joint center, which could have influenced effective stiffness measures. However, because the study was a within-subjects design, a given participant had the same vertical relationship to the cradle with each brace condition, and thus any such errors would be systematic across sessions (brace conditions). A final limitation was the sample size. The within-subjects design allowed for a relatively smaller sample size, but having more participants could have helped to clarify the significant interaction effects. Also, although it was not a primary interest of the study, significant gender differences were found for ankle effective stiffness measures. A greater number of participants (with even numbers of males and females) could have provided a better understanding of gender differences in ankle effective stiffness.

4.6 Summary

Overall, the present results suggest that the AC brace can provide more consistent ankle support throughout athletic activity compared to NB or AW brace conditions. The AC brace appeared to facilitate improved JPS after initial application, and it helped to maintain ankle effective stiffness even after the development of fatigue. The AW brace may provide an increase in effective stiffness when initially applied, but when an athlete becomes fatigued, this brace may not able to compensate for losses in physiological stiffness, particularly among males. Stiffness results suggest that the AC brace may be a more suitable intervention for males with ankle instability, while either the AW or AC brace may provide improved effective stiffness for females. The results presented here may be applicable to other neoprene and semi-rigid brace designs, and may help those with CAI to make a more informed selection between these two brace types. There are numerous varieties of ankle braces, however, and which include combinations of neoprene, lace-up, and semi-rigid features. More research is clearly needed on diverse brace types, to identify the specific benefits of each brace as they pertain to gender and mode of instability (functional vs. mechanical) to assist consumers in the selection of the brace that will best address their needs.

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Chapter 5

Conclusions and Future Research Directions

5.1 Restatement of Research Goals

Research investigating the causes of ankle sprains in adults with chronic ankle instability (CAI) has been ongoing since the mid-1960s when the condition was first characterized by Freeman (1965). Since that time, several biomechanical deficits related to CAI have been identified, including increased postural sway (Cornwall & Murrell, 1991; Goldie et al., 1994; Docherty et al., 2006; Brown et al., 2007 & 2010) and poorer joint proprioception (Ryan, 1994; Lentell et al., 1995; Jerosch & Bischof, 1996; Refshauge et al., 2000; Hartsell, 2000; Willems et al., 2002), and that are thought to contribute to recurrent sprains among adults with CAI. Despite the frequency with which these ankle sprains occur during sports, very little research has investigated ankle control with CAI under dynamic conditions simulating actual sport activity (Kipp & Palmieri-Smith, 2013). In Chapter 2, participants with and without ankle instability performed a set of quasi-static ankle evaluations and also completed a series of dynamic maneuvers at each of two running speeds. The goal of this study was to determine how running speed affects leg stabilization during simulated sport activities and to assess the level of correspondence between quasi-static measures of ankle motor control and those observed during dynamic athletic tasks. Another gap in the existing literature is research investigating the role of fatigue in ankle sprains. Fatigue is often associated with sports injury (Pinto et al., 1999; Gabbett, 2000; Woods et al., 2003), but it is unknown whether fatigue is more detrimental to stability among adults with CAI compared to stable controls. The second study, described in Chapter 3, addressed whether adults with CAI develop fatigue at a similar rate as those with stable ankles, whether they have a similar perception of fatigue development, and if landing mechanics during a sudden ground impact event differ between groups with and without CAI. Lastly, this dissertation focused on the effectiveness of one treatment for ankle sprains, namely ankle braces. Ankle braces are thought to assist in sprain resistance by increasing stiffness of the ankle (Eils et al., 2002; Papadopoulous et al., 2005; Cordova et al., 2007; Zinder et al., 2009) and improving ankle proprioception (Feuerbach et al., 1994; Heit et al., 1996; Hartsell, 2000). Arguably, the degree to which these benefits are provided may depend on the particular construction of the brace. As such, a third study, presented in Chapter 4, was completed to determine how two specific brace types differ in terms of stiffness and proprioception improvements in adults with CAI and to determine whether these benefits were retained in the presence of localized muscle fatigue (LMF) at the ankle.

5.2 Dynamic vs. Quasi-Static Methods

The control (C) and unstable (U) ankle groups were found to be generally comparable in terms of the results obtained using quasi-static methods, but differences were evident when comparing between groups in their adaptations to changes in running speed. Generally, adults with instability compensated for increases in running speed by increasing knee flexion or frontal plane joint moments. Controls, however, exhibited greater increases in transverse plane joint angles and moments. The Cut step maneuver was particularly effective at identifying group differences: with an increase in speed, the C group increased knee IR moment while the U group increased both ankle EV and knee AB moments. These increases in frontal plane joint moments among individuals with CAI may increase likelihood of injury during this maneuver, particularly if the changes observed here extend to greater increases in speed. Future work, though, should investigate CAI-related differences in kinematics and kinetics over a broader range of running speeds.

Regarding the relationship between quasi-static and dynamic measures of ankle control, JPS measures did not correlate strongly with positioning errors for any of the dynamic maneuvers. Drop landing ground contact angles, however, were relatively highly correlated with corresponding ground contact angles during Cut step maneuvers. These results indicate that kinematic differences observed with drop landing measures may provide a reasonable estimation of kinematics during some types of dynamic athletic activity.

5.3 Fatigue Effects

Adults with and without ankle instability appeared to experience fatigue in a similar manner. Not only did the two groups develop fatigue at a similar rate when working at the same relative effort level, they also reported similar subjective perceptions of fatigue in response to equivalent strength losses. Group differences in joint control strategies and muscle activation (EMG) patterns were found, though, particularly during side drop landings. Fatigue-induced changes in joint stabilization moments during the stance phase of forward drop landings suggested that the two groups had similar adaptations to fatigue, but that individuals with CAI had smaller magnitude of these changes. Fatigue effects during side drop landings, however, revealed group differences in motor control strategies. Specifically, the C group executed side drop landings using sagittal and frontal plane control at both the ankle and knee joints. The U group, in contrast, used sagittal and frontal plane control at the knee but relied on transverse plane control at the ankle. Additionally, EMG measures revealed that the U group has a greater post-fatigue dependence on the peroneus longus (PL) muscle to stabilize the ankle after ground contact. This increased complexity in joint control strategies and a greater reliance on the PL muscle may contribute to the increased frequency of ankle sprains in those with CAI (Lofvenberg et al., 1995; Mitchell et al. 2008; Yeow et al., 2009; Kavanaugh et al., 2012).

Considering the results of Chapters 2 and 3 together, group differences appear to be most evident during more challenging tasks (e.g., Cut step, side drop landings), in particular those that require the participants to either engage in frontal plane control or exhibit an avoidance strategy. As such, a recommendation for future research in ankle instability would be to evaluate group differences using tasks that involve a rapid change of direction or that require substantial mediolateral stabilization effort.

5.4 Bracing and Fatigue Effects

Initially, it was expected that the neoprene ankle wrap brace (AW) would result in greater improvement in JPS and that the semi-rigid Aircast (AC) brace would result in greater improvement in ankle effective stiffness. Instead, results suggested that the AC brace provided more proprioceptive benefit than either the AW or the unbraced (NB) condition. These findings are supported by previous research that has also found an increase in JPS with a semi-rigid brace (Hartsell, 2000), although fatigue was not a consideration in that study. Effective stiffness measures suggested that, prior to fatigue, both braces improved ankle effective stiffness compared to the NB condition. Previously, Zinder et al. (2009) reported an increase in passive, mechanical stiffness with the application of ankle braces, and Eils et al. (2002) found that several brace types were able to restrict ankle motion in response to a sudden inversion perturbation. Following ankle LMF here, stiffness was retained or slightly improved among males with the AC brace condition and was significantly less for males with the NB and AW conditions. Despite demonstrating similar trends, significant post-fatigue differences between brace conditions were not observed among females. However, considering that pre-fatigue effective stiffness with either test brace (AW or AC) was higher than in the NB condition, the current results indicate that the AC brace may provide superior effective stiffness support for males and that either brace condition may provide improved effective stiffness (compared to NB) for females.

5.5 Research Limitations and Future Directions

There are several limitations to this dissertation research that could be addressed in future work. Joint position sense was measured using a dynamometer that imposed on participants a somewhat awkward seated posture, and which was perceived by some to be uncomfortable. The dynamometer also lacked a “free rotation” mode and required substantial practice for some participants to gain confidence in their ability to control foot placement. Additionally, several participants noted that the orientation of the inversion/eversion foot plate restricted them from performing those motions as they would naturally. Although other researchers have used similar dynamometers for joint position sense research (Lee et al., 2003; Chang et al., 2006), a preferred method for future work would be to construct a custom device that allows free rotation and a more comfortable body posture. This would allow participants better control over the device and perhaps more accurately reflect natural joint positioning during real-world activities. Further, a device that would allow weight-bearing during joint position sense testing may be more relevant to sports injury.

Another limitation was the use of an LMF fatigue protocol rather than a whole body fatigue protocol. Kinematic, kinetic, and neuromuscular measures collected during whole-body fatigue could arguably better capture the fatigue-induced compensations made at multiple joints during actual athletic activity. That is, changes observed at any one joint are a result of a coordinated multi-joint effort to cope with general fatigue. Utilizing a whole body fatigue protocol could thus improve the validity of the research with respect to real-world sports injury. Additional potential limitations that should be considered in future work include the use of visual cues during MVIC measurements and the fixed height of the foot rocker cradle. A better method for MVIC measurement may have been to remove the visual display and use only verbal commands to indicate the timing of these efforts. This may have reduced the potential for the participant to stop the protocol prematurely. Also, the effective stiffness measures reported here may have more external validity if the rocker height was adjustable and permitted better alignment of the cradle axis to the participant’s ankle joint.

Results presented in Chapters 2 and 3 suggested that adults with CAI utilize distinct control strategies to execute dynamic athletic maneuvers and fatigued drop landings. Future work may benefit from examining whether specific physical therapy programs – which are often a treatment option for severe sprains – are effective at altering pathological motor control strategies to reduce the risk of reinjury. Additionally, more work is needed on the effects of whole body fatigue on motor control in adults with CAI. If detrimental motor control changes can be identified as a function of fatigue level, it may be possible to identify work-rest periods that help prevent sprain injury. This may be particularly helpful for recreational athletes who plan their own workouts and could help reduce their dependence on ankle taping and bracing for stability. Further, it has been suggested that ankle taping and bracing increase proximal joint kinematics (Santos et al., 2004). Future research may thus explore the long-term effects of bracing on knee and hip kinematics and the prevalence of joint laxity or injury.

As noted in Chapter 1, CAI may be classified as either functional or mechanical in nature, depending on whether the instability arises primarily from nerve damage (functional) or increased tissue laxity and damaged ligaments (mechanical). Currently, however, there is no system in place to help consumers select an ankle brace that would best serve their particular needs. Future work should focus on identifying the specific benefits (e.g., stiffness, improved proprioception) of various brace types as they apply to adults with both functional and mechanical instabilities. Additionally, future work should examine potential gender differences, to determine if particular brace designs are better suited to male or female users.

5.6 Summary

In conclusion, the present research focused on the influence of running speed, fatigue, and bracing on ankle motor control and particular differences among adults with CAI. Results overall indicate that those with CAI exhibit different adaptations to changes in running speed and utilize different joint control strategies in response to fatigue, both of which may contribute to recurrent sprain injuries. When fatigued, a semi-rigid brace may provide greater sprain resistance by retaining effective stiffness at the ankle, but the benefits of bracing may be gender-specific. More research is needed, particularly in the areas of fatigue and ankle brace prescription, to better understand the factors increasing sprain injury risk and to improve preventative treatment.

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Appendix A.

Chapter 2 Statistical Results

Table A1. ANOVA results for the effects of group (G), reference angle (R), and their interaction on JPS error measures in each JPS configuration. Significant effects ($p < 0.05$) are in bold while effects that approached significance ($0.05 < p < 0.1$) are highlighted.

		P/D Configuration			I/E Configuration		
Measures	**Stats**	**G**	**R**	**G x R**	**G**	**R**	**G x R**
Absolute Error (deg)	*F*	1.13	1.70	1.66	0.88	**2.91**	0.91
	p	0.30	0.17	0.18	0.36	**0.03**	0.44
True Error (deg)	*F*	0.15	**8.01**	1.03	0.23	1.85	0.68
	p	0.70	**<0.0001**	0.38	0.63	0.14	0.56

Table A2. Mean (SD) of JPS error measures with respect to group, reference angle, and test configuration.

	P/D Configuration				I/E Configuration			
	Abs Err (deg)		True Err (deg)		Abs Err (deg)		True Err (deg)	
	C	**U**	**C**	**U**	**C**	**U**	**C**	**U**
-15	2.5(1.8)	2.5(1.9)	0.3(31)	0.2(3.2)	2.0(1.4)	2.5(1.4)	1.3(2.1)	0.6(2.8)
-5	3.4(2.2)	2.5(2.5)	-0.3(4.0)	-0.2(3.5)	2.6(2.6)	2.9(2.3)	0.2(3.7)	0.8(3.7)
+5	2.7(2.1)	2.9(2.8)	1.1(3.2)	2.2(3.7)	3.2(2.5)	4.3(3.6)	1.2(4.0)	0.4(5.6)
+15	2.2(1.1)	2.1(1.8)	1.2(2.2)	0.9(2.7)	2.7(1.5)	3.0(2.2)	1.8(2.5)	1.5(3.4)

Table A3. ANOVA results for the effects of group (G) for both drop types on ankle and knee dynamics during drop landings. Significant effects ($p < 0.05$) are in bold while effects that approached significance ($0.05 < p < 0.1$) are highlighted.

Measures	Stats	Forward Drop G - Ankle	Forward Drop G - Knee	Side Drop G - Ankle	Side Drop G - Knee
GC PF/DF or EX/FL angle (deg)	*F*	0.02	0.04	0.17	2.81
	p	0.89	0.84	0.68	0.11
GC INV/EV or AD/AB angle (deg)	*F*	2.12	1.77	0.30	1.75
	p	0.16	0.20	0.59	0.20
GC IR/ER angle (deg)	*F*	0.55	0.38	3.12	0.19
	p	0.47	0.54	0.09	0.67
Peak DF/EX angle (deg)	*F*	0.56	0.08	2.06	2.81
	p	0.46	0.79	0.17	0.11
Peak PF/FL angle (deg)	*F*	0.0001	2.45	0.17	0.06
	p	0.99	0.13	0.68	0.81
Peak INV/AD angle (deg)	*F*	1.72	2.15	0.42	0.94
	p	0.20	0.16	0.52	0.35
Peak EV/AB angle (deg)	*F*	0.45	2.82	1.20	1.74
	p	0.51	0.11	0.29	0.20
Peak IR angle (deg)	*F*	0.10	0.38	2.49	0.34
	p	0.76	0.54	0.13	0.57
Peak ER angle (deg)	*F*	0.24	0.29	2.86	0.13
	p	0.63	0.60	0.10	0.73
Peak DF/EX moment (Nm/kg)	*F*	0.22	0.02	0.004	1.53
	p	0.64	0.89	0.95	0.23
Peak PF/FL moment (Nm/kg)	*F*	0.20	1.00	0.75	0.02
	p	0.66	0.33	0.39	0.90
Peak INV/AD moment (Nm/kg)	*F*	1.28	0.81	0.47	0.96
	p	0.27	0.38	0.50	0.34
Peak EV/AB moment (Nm/kg)	*F*	0.99	0.49	0.41	0.13
	p	0.33	0.49	0.53	0.72
Peak IR moment (Nm/kg)	*F*	0.0007	2.23	0.002	0.46
	p	0.98	0.15	0.97	0.50
Peak ER moment (Nm/kg)	*F*	2.11	0.01	0.78	0.01
	p	0.16	0.94	0.39	0.92
Peak P_S (W/kg)	*F*	2.11	0.67	0.005	0.01
	p	0.16	0.42	0.94	0.91
Peak P_F (W/kg)	*F*	3.22	0.06	1.73	0.09
	p	0.08	0.81	0.20	0.77
Peak P_T (W/kg)	*F*	2.03	0.97	0.01	**7.19**
	p	0.17	0.33	0.94	**0.01**
ML GRF (N/kg)	*F*	0.01	--	0.05	--
	p	0.91	--	0.82	--
COP_{LD} (cm)	*F*	0.02	--	1.06	--
	p	0.89	--	0.31	--

*Abbreviations FL, EX, AB, AD refer to knee flexion, extension, abduction and adduction measures, respectively. The abbreviations DF, PF, INV, and EV refer to ankle dorsiflexion, plantarflexion, inversion, and eversion, respectively.

Table A4. Mean (SD) of kinetic and kinematic measures of ankle and knee joints during drop landings. Results are tabulated as group means for each drop type.

	Forward Drop		Side Drop	
Measures	**C**	**U**	**C**	**U**
Ankle				
GC PF/DF angle	-26.4(9.8)	-25.8(11.3)	-26.0(5.7)	-24.4(7.0)
GC INV/EV angle	0.02(8.9)	3.9(6.3)	9.9(7.8)	11.1(4.4)
GC IR/ER angle	6.2(4.3)	5.3(5.6)	6.5(5.1)	2.1(5.9)
Peak DF angle	7.4(5.8)	9.0(4.6)	14.8(3.1)	13.2(3.0)
Peak PF angle	-26.5(9.5)	-26.5(9.4)	-26.0(5.7)	-24.4(7.0)
Peak INV angle	0.9(7.5)	3.9(6.3)	10.1(7.2)	11.4(4.4)
Peak EV angle	-12.9(5.5)	-11.5(3.8)	-5.4(5.0)	-8.4(4.7)
Peak IR angle	6.5(4.2)	6.5(5.1)	6.9(5.4)	2.9(5.9)
Peak ER angle	-5.5(8.1)	-3.4(8.5)	1.2(3.8)	-2.3(5.5)
Peak DF moment	3.3(1.9)	3.6(3.5)	0.5(1.3)	0.1(0.6)
Peak PF moment	-0.5(0.6)	-0.6(0.7)	-4.3(3.4)	-4.4(2.1)
Peak INV moment	1.4(2.1)	2.7(3.3)	0.04(0.2)	0.01(0.1)
Peak EV moment	-1.0(1.7)	-0.4(0.9)	-6.2(3.7)	-4.8(3.1)
Peak IR moment	0.2(0.3)	0.4(0.4)	0.4(0.5)	0.3(0.4)
Peak ER moment	-0.4(0.3)	-0.4(0.4)	-0.9(0.4)	-0.8(0.3)
Peak P_S	-4.8(5.6)	-7.0(8.2)	-33.2(21.7)	-33.4(13.9)
Peak P_F	-3.0(2.6)	-8.3(11.1)	-0.6(1.2)	-0.2(0.6)
Peak P_T	-0.8(0.7)	-1.2(1.2)	-0.7(0.7)	-0.6(0.6)
ML GRF	2.1(1.2)	2.1(0.9)	3.0(0.7)	3.0(0.8)
COP_{LD} (cm)	2.0(0.7)	2.0(1.0)	1.4(1.4)	1.9(1.4)
Knee				
GC EX/FL angle	-7.4(3.9)	-7.7(3.5)	-8.7(4.1)	-12.3(4.3)
GC AD/AB angle	-0.8(2.6)	0.5(3.2)	0.2(2.3)	1.5(3.0)
GC IR/ER angle	2.6(7.8)	0.9(4.9)	4.5(3.2)	5.4(5.6)
Peak EX angle	-7.3(3.8)	-7.7(3.5)	-8.7(4.1)	-12.3(4.3)
Peak FL angle	-35.6(7.5)	-38.8(5.8)	-37.8(5.8)	-39.0(6.7)
Peak AD angle	0.004(4.2)	2.6(6.3)	5.2(8.2)	8.0(7.5)
Peak AB angle	-7.8(6.3)	-4.1(7.5)	-0.3(2.9)	1.2(3.4)
Peak IR angle	9.0(10.2)	7.4(8.2)	2.2(4.8)	3.2(6.7)
Peak ER angle	-3.5(7.8)	-2.0(5.7)	-4.9(3.4)	-5.6(5.6)
Peak EX moment	5.8(2.3)	5.8(3.6)	2.2(2.4)	1.1(2.2)
Peak FL moment	-0.2(0.3)	-0.3(0.6)	-2.3.(3.0)	-2.5(1.8)
Peak AD moment	1.6(2.4)	2.8(3.4)	0.03(0.2)	-0.02(0.2)
Peak AB moment	-1.0(1.6)	-0.7(1.0)	-6.4(3.6)	-6.0(3.2)
Peak IR moment	0.2(0.3)	0.5(0.5)	0.6(0.5)	0.4(0.5)
Peak ER moment	-0.5(0.3)	-0.5(0.5)	-0.6(0.4)	-0.6(0.3)
Peak P_S	-1.4(2.3)	-1.9(4.1)	-12.6(19.0)	-13.7(12.4)
Peak P_F	-1.3(2.2)	-1.6(2.1)	-1.3(1.9)	-0.9(1.9)
Peak P_T	-0.9(0.9)	-1.4(1.5)	-0.7(0.6)	-0.2(0.3)

Joint angle, moment, and power absorption measures are in units of deg, Nm/kg, and W/kg, respectively. ML GRF are in N/kg, and COP_{LD} in cm. Sign convention is provided to clarify direction. Negative values are FL, AB, PF, EV, ER, and power absorption while positive values are EX, AD, DF, INV, and IR.

Table A5. ANOVA results for the effects of group (G), running speed (S) and their interaction (G x S) on ankle and knee dynamics during **run** trials. Significant effects ($p < 0.05$) are in bold while effects that approached significance ($0.05 < p < 0.1$) are highlighted.

Measures	Stats	Ankle Model Effects G	Ankle Model Effects S	Ankle Model Effects G x S	Knee Model Effects G	Knee Model Effects S	Knee Model Effects G x S
GC PF/DF or EX/FL angle (deg)	*F*	2.50	**5.27**	0.11	0.51	**99.50**	**4.25**
	p	0.13	**0.02**	0.74	0.48	**<.0001**	**0.04**
GC INV/EV or AD/AB angle (deg)	*F*	0.91	2.18	1.27	0.16	**5.15**	0.71
	p	0.35	0.14	0.26	0.69	**0.02**	0.40
GC IR/ER angle (deg)	*F*	0.22	**4.59**	3.37	0.18	0.59	0.30
	p	0.64	**0.03**	0.07	0.68	0.44	0.59
Peak DF/EX angle (deg)	*F*	0.01	35.59	0.97	0.50	99.36	**4.17**
	p	0.93	<.0001	0.33	0.48	<.0001	**0.04**
Peak PF/FL angle (deg)	*F*	2.37	**4.15**	0.01	0.11	**10.32**	**8.91**
	p	0.14	**0.04**	0.93	0.75	**0.002**	**0.003**
Peak INV/AD angle (deg)	*F*	0.75	3.33	0.37	1.30	**7.86**	0.90
	p	0.40	0.07	0.54	0.27	**0.01**	0.35
Peak EV/AB angle (deg)	*F*	0.32	1.16	0.36	0.72	0.01	0.41
	p	0.57	0.28	0.55	0.41	0.91	0.52
Peak IR angle (deg)	*F*	0.66	0.14	0.90	0.84	19.77	1.27
	p	0.43	0.70	0.35	0.37	<.0001	0.26
Peak ER angle (deg)	*F*	0.58	0.90	0.84	0.02	1.47	0.14
	p	0.45	0.35	0.36	0.90	0.23	0.71
Peak DF/EX moment (Nm/kg)	*F*	**4.52**	0.15	**6.44**	2.25	0.01	**8.51**
	p	**0.04**	0.70	**0.01**	0.15	0.91	**0.004**
Peak PF/FL moment (Nm/kg)	*F*	0.95	3.65	**6.27**	2.78	0.02	**4.56**
	p	0.34	0.06	**0.01**	0.11	0.88	**0.03**
Peak INV/AD moment (Nm/kg)	*F*	1.81	1.04	**4.92**	1.16	1.13	**5.63**
	p	0.19	0.31	**0.03**	0.29	0.29	**0.02**
Peak EV/AB moment (Nm/kg)	*F*	0.65	0.44	3.96	0.82	0.03	**4.55**
	p	0.43	0.51	0.05	0.37	0.85	**0.03**
Peak IR moment (Nm/kg)	*F*	0.78	**5.35**	2.57	0.22	**4.98**	1.86
	p	0.39	**0.02**	0.11	0.64	**0.03**	0.18
Peak ER moment (Nm/kg)	*F*	1.31	0.03	1.43	1.29	0.11	1.53
	p	0.26	0.87	0.23	0.27	0.74	0.22
Peak P_S (W/kg)	*F*	0.27	**8.45**	1.77	3.01	0.47	2.87
	p	0.61	**0.004**	0.19	0.10	0.50	0.09
Peak P_F (W/kg)	*F*	0.82	1.75	0.34	0.001	0.01	4.83
	p	0.37	0.19	0.56	0.98	0.94	0.03
Peak P_T (W/kg)	*F*	0.32	0.02	0.66	0.01	0.68	0.79
	p	0.58	0.90	0.42	0.93	0.41	0.38
ML GRF (N/kg)	*F*	0.25	**32.58**	**5.19**	--	--	--
	p	0.62	**<.0001**	**0.02**	--	--	--
COP_{LD} (cm)	*F*	0.69	0.64	0.30	--	--	--
	p	0.41	0.42	0.59	--	--	--

*Abbreviations FL, EX, AB, AD refer to knee flexion, extension, abduction and adduction measures, respectively. The abbreviations DF, PF, INV, and EV refer to ankle dorsiflexion, plantarflexion, inversion, and eversion, respectively.

Table A6. ANOVA results for the effects of group assignment (G), running speed (S) and their interaction (G x S) on ankle and knee dynamics during **jump stop** trials. Significant effects ($p < 0.05$) are in bold while effects that approached significance ($0.05 < p < 0.1$) are highlighted.

		Ankle Model Effects			Knee Model Effects		
Measures	**Stats**	**G**	**S**	**G x S**	**G**	**S**	**G x S**
GC PF/DF or EX/FL angle (deg)	*F*	0.01	2.03	0.47	0.33	0.16	1.87
	p	0.94	0.16	0.49	0.57	0.69	0.17
GC INV/EV or AD/AB angle (deg)	*F*	1.98	0.75	1.28	2.81	0.003	0.44
	p	0.17	0.39	0.26	0.11	0.96	0.51
GC IR/ER angle (deg)	*F*	0.17	1.82	**6.50**	0.02	0.17	0.01
	p	0.68	0.18	**0.01**	0.89	0.68	0.92
Peak DF/EX angle (deg)	*F*	0.83	**17.34**	2.56	0.43	0.02	3.21
	p	0.37	**<.0001**	0.11	0.52	0.90	0.08
Peak PF/FL angle (deg)	*F*	0.10	0.56	0.13	4.09	0.46	**22.81**
	p	0.76	0.46	0.72	0.05	0.50	**<.0001**
Peak INV/AD angle (deg)	*F*	2.00	3.94	3.00	2.74	1.55	0.01
	p	0.17	0.05	0.09	0.11	0.21	0.91
Peak EV/AB angle (deg)	*F*	0.33	0.32	1.15	3.86	0.18	2.11
	p	0.57	0.57	0.29	0.06	0.67	0.15
Peak IR angle (deg)	*F*	0.16	0.37	0.34	0.07	0.44	0.65
	p	0.70	0.54	0.56	0.80	0.51	0.42
Peak ER angle (deg)	*F*	0.55	**4.42**	1.11	0.02	1.21	0.85
	p	0.47	**0.04**	0.29	0.89	0.27	0.36
Peak DF/EX moment (Nm/kg)	*F*	0.03	**17.85**	2.67	0.19	**6.46**	0.36
	p	0.87	**<.0001**	0.10	0.67	**0.01**	0.55
Peak PF/FL moment (Nm/kg)	*F*	0.80	0.08	3.21	0.29	0.32	0.09
	p	0.38	0.78	0.08	0.60	0.57	0.76
Peak INV/AD moment (Nm/kg)	*F*	1.88	2.44	**6.57**	2.11	0.53	**4.61**
	p	0.18	0.12	**0.01**	0.16	0.47	**0.03**
Peak EV/AB moment (Nm/kg)	*F*	1.08	0.23	1.68	1.58	0.0003	0.61
	p	0.31	0.63	0.20	0.22	0.99	0.43
Peak IR moment (Nm/kg)	*F*	0.78	0.60	0.08	0.79	0.01	0.06
	p	0.39	0.44	0.78	0.38	0.93	0.80
Peak ER moment (Nm/kg)	*F*	0.54	2.16	1.48	0.77	0.36	0.19
	p	0.47	0.14	0.23	0.39	0.55	0.66
Peak P_S (W/kg)	*F*	0.19	**20.96**	2.71	0.04	0.66	0.04
	p	0.66	**<.0001**	0.10	0.85	0.42	0.85
Peak P_F (W/kg)	*F*	0.01	1.80	0.36	1.55	0.02	0.20
	p	0.93	0.18	0.55	0.23	0.88	0.66
Peak P_T (W/kg)	*F*	0.03	**7.21**	2.52	0.15	**15.42**	**5.06**
	p	0.86	**0.01**	0.11	0.71	**0.0001**	**0.03**
ML GRF (N/kg)	*F*	0.01	**6.62**	0.41	--	--	--
	p	0.91	**0.01**	0.52	--	--	--
COP_{LD} (cm)	*F*	2.99	2.22	2.68	--	--	--
	p	0.10	0.14	0.10	--	--	--

*Abbreviations FL, EX, AB, AD refer to knee flexion, extension, abduction and adduction measures, respectively. The abbreviations DF, PF, INV, and EV refer to ankle dorsiflexion, plantarflexion, inversion, and eversion, respectively.

Table A7. ANOVA results for the effects of group assignment (G), running speed (S) and their interaction (G x S) on ankle and knee dynamics during **cut step** trials. Significant effects ($p < 0.05$) are in bold while effects that approached significance ($0.05 < p < 0.1$) are highlighted.

		Ankle Model Effects			Knee Model Effects		
Measures	**Stats**	**G**	**S**	**G x S**	**G**	**S**	**G x S**
GC PF/DF or EX/FL angle (deg)	*F*	0.84	0.54	1.54	0.16	1.86	1.54
	p	0.37	0.46	0.22	0.70	0.18	0.22
GC INV/EV or AD/AB angle (deg)	*F*	1.78	**8.33**	0.01	1.18	2.42	3.28
	p	0.19	**0.005**	0.94	0.29	0.12	0.07
GC IR/ER angle (deg)	*F*	1.70	0.21	0.10	0.004	**4.75**	**4.08**
	p	0.20	0.65	0.75	0.95	**0.03**	**0.04**
Peak DF/EX angle (deg)	*F*	0.004	0.04	0.81	0.14	1.85	1.53
	p	0.95	0.83	0.37	0.71	0.18	0.22
Peak PF/FL angle (deg)	*F*	2.23	1.00	0.76	0.06	0.09	0.06
	p	0.15	0.32	0.39	0.80	0.76	0.81
Peak INV/AD angle (deg)	*F*	0.83	**19.68**	3.02	0.68	2.33	2.48
	p	0.37	**<.0001**	0.08	0.42	0.13	0.12
Peak EV/AB angle (deg)	*F*	3.59	3.51	1.27	0.08	**4.42**	**7.60**
	p	0.07	0.06	0.26	0.78	**0.04**	**0.01**
Peak IR angle (deg)	*F*	1.17	0.54	0.67	0.17	0.22	3.27
	p	0.29	0.46	0.41	0.69	0.64	0.07
Peak ER angle (deg)	*F*	1.60	**11.52**	0.68	0.08	**9.73**	**4.01**
	p	0.22	**0.009**	0.41	0.78	**0.002**	**0.04**
Peak DF/EX moment (Nm/kg)	*F*	1.20	0.98	2.96	0.58	0.14	**7.63**
	p	0.28	0.32	0.09	0.45	0.71	**0.01**
Peak PF/FL moment (Nm/kg)	*F*	0.06	1.80	3.43	0.01	4.19	0.14
	p	0.81	0.18	0.07	0.94	0.04	0.71
Peak INV/AD moment (Nm/kg)	*F*	1.66	0.39	**6.86**	1.45	0.47	**9.60**
	p	0.21	0.53	**0.01**	0.24	0.49	**0.002**
Peak EV/AB moment (Nm/kg)	*F*	0.12	**9.82**	**5.37**	0.02	**7.39**	**6.71**
	p	0.73	**0.002**	**0.02**	0.89	**0.01**	**0.01**
Peak IR moment (Nm/kg)	*F*	1.33	0.69	**5.80**	1.12	1.22	**9.74**
	p	0.26	0.41	**0.02**	0.30	0.27	**0.002**
Peak ER moment (Nm/kg)	*F*	0.47	**5.15**	3.75	0.0001	**7.51**	1.55
	p	0.50	**0.03**	0.06	0.99	**0.01**	0.22
Peak P_S (W/kg)	*F*	0.28	**15.48**	2.61	0.11	**4.55**	0.17
	p	0.60	**0.0001**	0.11	0.75	**0.04**	0.68
Peak P_F (W/kg)	*F*	3.32	**11.04**	0.08	1.72	**6.15**	0.41
	p	0.08	**0.001**	0.78	0.20	**0.01**	0.52
Peak P_T (W/kg)	*F*	0.10	**15.16**	2.01	0.34	2.26	1.66
	p	0.76	**0.0002**	0.16	0.57	0.14	0.20
ML GRF (N/kg)	*F*	0.002	**4.67**	0.04	--	--	--
	p	0.96	**0.03**	0.85	--	--	--
COP_{LD} (cm)	*F*	1.63	0.21	1.55	--	--	--
	p	0.21	0.65	0.22	--	--	--

*Abbreviations FL, EX, AB, AD refer to knee flexion, extension, abduction and adduction measures, respectively. The abbreviations DF, PF, INV, and EV refer to ankle dorsiflexion, plantarflexion, inversion, and eversion, respectively.

Table A8. ANOVA results for the effects of group assignment (G), running speed (S) and their interaction (G x S) on ankle and knee dynamics during **shuttle run** trials. Significant effects ($p < 0.05$) are in bold while effects that approached significance ($0.05 < p < 0.1$) are highlighted.

Measures	Stats	Ankle Model Effects G	Ankle Model Effects S	Ankle Model Effects G x S	Knee Model Effects G	Knee Model Effects S	Knee Model Effects G x S
GC PF/DF or EX/FL angle (deg)	*F*	1.26	**16.08**	1.29	0.01	0.09	1.27
	p	0.27	**0.0001**	0.26	0.93	0.76	0.26
GC INV/EV or AD/AB angle (deg)	*F*	0.26	2.01	1.92	3.15	0.12	0.12
	p	0.61	0.16	0.17	0.09	0.73	0.73
GC IR/ER angle (deg)	*F*	0.001	0.56	0.58	0.02	0.12	**4.89**
	p	0.98	0.46	0.45	0.90	0.73	**0.03**
Peak DF/EX angle (deg)	*F*	0.10	1.61	0.03	0.01	0.37	1.19
	p	0.75	0.21	0.87	0.93	0.55	0.28
Peak PF/FL angle (deg)	*F*	0.95	**15.50**	0.51	0.97	0.55	0.01
	p	0.34	**0.0001**	0.48	0.33	0.46	0.91
Peak INV/AD angle (deg)	*F*	0.005	1.52	**4.59**	1.16	1.16	0.03
	p	0.95	0.22	**0.03**	0.29	0.28	0.87
Peak EV/AB angle (deg)	*F*	0.11	1.44	2.60	1.67	0.12	0.53
	p	0.74	0.23	0.11	0.21	0.72	0.47
Peak IR angle (deg)	*F*	0.01	**4.73**	1.08	0.002	0.29	2.46
	p	0.94	**0.03**	0.30	0.97	0.59	0.12
Peak ER angle (deg)	*F*	0.19	0.48	0.06	0.16	**32.27**	0.90
	p	0.66	0.49	0.81	0.69	**<.0001**	0.35
Peak DF/EX moment (Nm/kg)	*F*	0.12	**5.15**	1.81	0.03	0.75	**5.79**
	p	0.73	**0.02**	0.18	0.87	0.39	**0.02**
Peak PF/FL moment (Nm/kg)	*F*	0.20	0.73	0.40	0.80	0.02	1.34
	p	0.66	0.39	0.53	0.38	0.89	0.25
Peak INV/AD moment (Nm/kg)	*F*	2.36	2.14	0.01	3.84	0.15	0.33
	p	0.14	0.15	0.99	0.06	0.70	0.57
Peak EV/AB moment (Nm/kg)	*F*	0.09	**15.77**	3.20	0.25	**5.15**	1.41
	p	0.76	**0.0001**	0.08	0.62	**0.03**	0.24
Peak IR moment (Nm/kg)	*F*	0.02	**9.01**	2.68	0.04	2.98	**5.91**
	p	0.88	**0.003**	0.10	0.84	0.09	**0.02**
Peak ER moment (Nm/kg)	*F*	0.002	1.42	0.80	0.76	0.24	0.34
	p	0.97	0.24	0.37	0.39	0.63	0.56
Peak P_S (W/kg)	*F*	0.001	**5.86**	0.02	0.33	0.01	2.11
	p	0.99	**0.02**	0.90	0.57	0.93	0.15
Peak P_F (W/kg)	*F*	0.07	**12.82**	2.18	0.46	**19.20**	1.49
	p	0.79	**0.001**	0.14	0.50	**<.0001**	0.22
Peak P_T (W/kg)	*F*	0.002	1.72	0.85	0.11	3.11	0.07
	p	0.97	0.19	0.36	0.75	0.08	0.79
ML GRF (N/kg)	*F*	0.18	3.68	3.39	--	--	--
	p	0.67	0.06	0.07	--	--	--
COP_{LD} (cm)	*F*	1.00	0.77	0.23	--	--	--
	p	0.33	0.38	0.64	--	--	--

*Abbreviations FL, EX, AB, AD refer to knee flexion, extension, abduction and adduction measures, respectively. The abbreviations DF, PF, INV, and EV refer to ankle dorsiflexion, plantarflexion, inversion, and eversion, respectively.

Table A9. Mean (SD) for kinetic and kinematic measures for **run** and **jump stop** trials. Results are tabulated as group means at each speed.

	RUN				JUMP STOP			
	Slower Speed (2.5 m/s)		Faster Speed (3.6 m/s)		Slower Speed (2.5 m/s)		Faster Speed (3.6 m/s)	
Measures	**C**	**U**	**C**	**U**	**C**	**U**	**C**	**U**
Ankle								
GC PF/DF angle	8.3(4.6)	3.1(12.6)	-6.5(7.1)	-0.1(14.0)	-6.2(16.0)	-5.8(19.1)	-4.0(10.8)	-4.7(15.2)
GC INV/EV angle	1.3(4.5)	2.2(6.3)	1.3(6.0)	4.0(6.4)	1.3(4.7)	3.5(6.7)	1.1(5.4)	4.6(4.9)
GC IR/ER angle	-0.1(3.8)	1.1(5.2)	1.7(5.0)	1.6(6.0)	3.5(5.7)	1.1(6.0)	1.4(7.4)	2.0(6.9)
Peak DF angle	14.6(3.8)	14.2(3.1)	16.5(5.2)	17.2(3.8)	2.9(6.1)	3.7(4.7)	0.1(5.4)	2.4(4.9)
Peak PF angle	0.6(4.5)	-3.1(9.4)	-0.7(6.0)	-5.2(10.3)	-19.4(10.0)	-20.7(11.1)	-19.0(6.9)	-19.6(8.4)
Peak INV angle	1.5(4.4)	2.5(6.1)	1.9(5.1)	4.0(6.4)	4.4(4.2)	6.2(5.8)	4.6(4.9)	8.0(3.9)
Peak EV angle	-8.9(3.9)	-10.9(6.1)	-8.7(4.5)	-10.3(7.1)	-0.8(3.6)	-0.1(5.5)	-1.1(4.8)	0.6(4.1)
Peak IR angle	4.2(3.0)	5.6(4.1)	4.5(3.9)	5.1(5.1)	5.2(4.8)	4.0(4.2)	4.6(5.6)	4.2(5.7)
Peak ER angle	-0.9(3.3)	-2.1(4.3)	-0.3(3.4)	-2.0(6.2)	0.4(3.5)	-1.7(4.9)	-1.5(6.9)	-2.0(5.2)
Peak DF moment	2.4(1.8)	1.0(1.5)	1.9(2.0)	1.4(1.6)	3.0(1.7)	3.4(1.7)	4.2(2.3)	3.8(1.7)
Peak PF moment	-2.0(2.4)	-3.6(3.1)	-3.7(2.8)	-3.5(3.0)	-0.6(1.3)	-0.2(0.5)	-0.4(1.0)	-0.3(0.9)
Peak INV moment	0.5(1.1)	0.6(1.1)	0.3(0.8)	1.1(1.3)	-0.1(0.1)	-0.1(0.1)	-0.1(0.1)	-0.1(0.1)
Peak EV moment	-2.4(2.3)	-1.9(2.4)	-2.7(2.3)	-1.3(2.0)	-4.2(1.9)	-4.6(2.1)	-4.0(1.0)	-4.9(2.4)
Peak IR moment	0.1(0.2)	0.1(0.2)	0.1(0.2)	0.2(0.3)	-0.003(0.1)	-0.02(0.1)	-0.01(0.1)	-0.03(0.1)
Peak ER moment	-0.7(0.6)	-0.5(0.5)	-0.7(0.6)	-0.4(0.4)	-1.8(0.9)	-1.9(0.8)	-1.8(0.8)	-2.2(1.2)
Peak P_S	-14.3(7.2)	-17.5(13.9)	-21.1(10.0)	-20.9(16.7)	-26.9(24.9)	-25.2(24.4)	-40.1(34.2)	-30.7(22.6)
Peak P_F	-1.6(2.9)	-2.0(2.3)	-2.0(3.6)	-3.0(4.3)	-4.4(5.0)	-3.9(5.6)	-5.1(4.6)	-5.3(5.8)
Peak P_T	-0.6(0.6)	-0.5(0.7)	-0.5(0.7)	-0.5(1.1)	-0.9(1.4)	-1.5(1.9)	-2.4(3.2)	-1.9(2.6)
ML GRF	1.3(0.6)	1.2(0.5)	1.5(0.7)	1.8(0.7)	2.3(1.3)	2.4(1.2)	2.6(1.4)	2.8(1.9)
COP_{LD}	0.5(0.4)	0.6(0.4)	0.6(0.4)	0.8(0.6)	0.9(0.6)	1.3(1.4)	0.9(0.8)	1.7(1.6)
Knee								
GC EX/FL angle	-9.5(6.8)	-10.2(5.1)	-13.5(9.2)	-17.2(6.9)	-16.2(7.4)	-13.3(8.4)	-15.0(7.2)	-14.6(7.6)
GC AB/AD angle	0.6(3.0)	0.9(3.6)	0.9(2.7)	1.9(4.0)	2.0(3.2)	-0.4(4.2)	1.9(4.0)	-0.1(4.3)
GC IR/ER angle	-4.0(4.7)	-2.7(6.3)	-3.5(4.7)	-2.8(6.4)	-4.0(4.5)	-3.8(7.3)	-4.1(5.7)	-3.3(6.4)
Peak EX angle	-9.5(6.8)	-10.2(5.1)	-13.5(9.2)	-17.1(6.8)	-16.2(7.4)	-13.0(7.8)	-15.0(7.2)	-14.5(7.6)
Peak FL angle	-44.5(4.7)	-42.6(4.2)	-44.4(4.5)	-45.6(5.7)	-59.5(6.7)	-52.2(7.5)	-56.3(6.0)	-55.0(5.9)
Peak AD angle	1.6(3.6)	3.1(4.4)	1.9(3.1)	4.3(5.1)	5.2(4.8)	2.2(5.3)	5.9(5.5)	2.9(5.1)
Peak AB angle	-2.4(4.0)	-1.1(4.9)	-2.6(3.7)	-1.0(5.3)	0.7(3.7)	-3.1(5.7)	-0.02(4.8)	-2.7(4.5)
Peak IR angle	0.5(5.7)	2.2(7.4)	2.1(5.2)	4.9(8.1)	-0.7(4.6)	-0.5(7.5)	-0.8(6.2)	-1.1(6.9)
Peak ER angle	-5.6(5.3)	-5.0(6.5)	-4.9(4.9)	-4.6(6.6)	-6.4(5.1)	-7.1(6.8)	-6.6(5.7)	-7.8(6.6)
Peak EX moment	3.5(2.6)	1.7(2.2)	2.5(2.4)	2.4(2.1)	4.1(1.5)	4.3(1.7)	4.8(2.3)	4.8(2.0)
Peak FL moment	-0.3(0.9)	-1.4(1.8)	-0.8(1.4)	-0.9(2.0)	0.1(0.2)	0.2(0.1)	0.1(0.3)	0.2(0.4)
Peak AD moment	0.3(0.7)	0.3(0.6)	0.2(0.7)	0.6(1.0)	-0.1(0.1)	-0.1(0.1)	-0.1(0.1)	-0.1(0.1)
Peak AB moment	-2.9(2.6)	-2.5(2.6)	-3.3(2.5)	-2.0(2.2)	-3.9(2.0)	-4.5(2.1)	-3.6(1.2)	-4.6(2.6)
Peak IR moment	0.1(0.2)	0.1(0.2)	0.1(0.2)	0.2(0.3)	0.01(0.1)	-0.01(0.1)	0.01(0.1)	-0.01(0.1)
Peak ER moment	-0.7(0.6)	-0.6(0.5)	-0.7(0.6)	-0.5(0.5)	-1.5(0.9)	-1.7(0.8)	-1.5(0.8)	-1.8(1.3)
Peak P_S	-1.6(5.5)	-7.2(9.6)	-4.6(7.8)	-6.3(11.5)	0.3(0.8)	0.2(0.6)	0.05(1.9)	-0.0001(2.7)
Peak P_F	-5.0(6.9)	-4.0(4.9)	-4.1(4.8)	-4.8(6.0)	-3.8(3.4)	-6.4(6.3)	-4.1(4.3)	-6.1(7.0)
Peak P_T	-1.0(1.6)	-0.8(0.8)	-1.0(1.3)	-1.0(1.3)	-2.3(2.4)	-1.8(1.8)	2.8(3.3)	-3.8(3.6)

Joint angle, moment, and power absorption measures are in units of deg, Nm/kg, and W/kg, respectively. ML GRF are in N/kg, and COP_{LD} in cm. Sign convention is provided to clarify direction. Negative values are FL, AB, PF, EV, ER, and power absorption while positive values are EX, AD, DF, INV, and IR.

Table A10. Mean (SD) for kinetic and kinematic measures for **cut** and **shuttle run** trials. Results are tabulated as group means at each speed.

	CUT STEP				SHUTTLE RUN			
	Slower Speed (2.5 m/s)		Faster Speed (3.6 m/s)		Slower Speed (2.5 m/s)		Faster Speed (3.6 m/s)	
Measures	**C**	**U**	**C**	**U**	**C**	**U**	**C**	**U**
Ankle								
GC PF/DF angle	-3.1(15.0)	-9.7(15.4)	-2.8(11.4)	-6.6(15.3)	-11.9(6.7)	-10.1(8.1)	-9.3(6.3)	-5.3(10.7)
GC INV/EV angle	3.2(8.4)	6.2(5.5)	6.1(5.8)	8.2(6.0)	12.3(8.8)	15.5(11.0)	12.3(10.7)	12.5(11.7)
GC IR/ER angle	3.0(7.0)	0.4(6.0)	2.7(5.3)	0.5(7.8)	-11.1(6.3)	-11.5(8.7)	-11.2(6.1)	-9.8(10.0)
Peak DF angle	9.5(6.2)	8.8(4.8)	9.6(4.9)	9,6(6.4)	19.7(8.3)	19.2(4.8)	20.7(7.1)	19.3(6.7)
Peak PF angle	-10.5(10.9)	-16.7(10.2)	-11.0(7.9)	-16.3(9.5)	-12.0(6.8)	-10.2(7.7)	-9.3(6.3)	-6.1(9.4)
Peak INV angle	13.7(5.2)	14.1(6.4)	15.4(5.0)	18.5(5.4)	28.0(5.8)	29.8(7.4)	30.8(6.9)	28.8(6.6)
Peak EV angle	1.9(7.8)	5.7(5.6)	3.7(7.0)	7.7(5.6)	11.5(7.9)	12.2(8.5)	12.0(10.5)	9.5(10.1)
Peak IR angle	4.2(6.9)	1.4(5.8)	3.0(4.7)	2.1(5.7)	-0.01(3.6)	-0.5(5.7)	0.5(4.3)	0.5(6.4)
Peak ER angle	-4.1(5.8)	-5.5(4.9)	-6.4(3.8)	-8.5(5.5)	-11.6(5.8)	-12.5(7.4)	-12.1(5.7)	-12.5(7.8)
Peak DF moment	2.1(1.8)	3.0(1.9)	2.4(1.8)	2.9(2.2)	0.5(0.8)	0.5(1.1)	0.5(0.9)	0.9(1.3)
Peak PF moment	-1.6(2.0)	-0.9(1.9)	-1.7(1.9)	-1.5(2.4)	-1.3(1.5)	-1.8(1.6)	-1.5(1.4)	-1.5(1.7)
Peak INV moment	0.7(1.3)	1.7(0.8)	1.0(1.4)	1.2(1.6)	0.1(0.5)	-0.1(0.04)	0.1(0.5)	-0.1(0.05)
Peak EV moment	-1.2(1.4)	-0.4(0.8)	-1.5(1.6)	-1.3(2.1)	-3.2(2.2)	-2.9(1.2)	-3.7(2.1)	-3.8(1.3)
Peak IR moment	0.7(0.8)	1.3(1.1)	0.9(1.0)	1.2(1.1)	0.5(0.7)	0.5(0.9)	0.6(0.9)	0.8(0.9)
Peak ER moment	-0.7(0.8)	-0.4(0.7)	-0.8(0.9)	-0.8(1.1)	-0.9(0.9)	-1.0(0.8)	-1.1(1.0)	-1.0(1.0)
Peak P_S	-14.2(9.4)	-13.3(12.0)	-17.4(10.1)	-21.3(13.6)	-9.8(14.0)	-10.2(10.9)	-7.7(9.1)	-7.5(9.4)
Peak P_F	-2.2(2.5)	-0.8(1.4)	-4.0(5.9)	-2.4(3.2)	-12.1(9.2)	-9.3(10.2)	-15.9(12.9)	17.3(14.8)
Peak P_T	-1.4(1.6)	-1.9(2.1)	-2.7(2.8)	-2.6(2.8)	-3.0(4.1)	-3.0(2.4)	-3.5(3.7)	-3.6(3.0)
ML GRF	6.7(2.0)	6.7(1.7)	7.1(2.1)	7.0(1.7)	2.5(1.5)	2.1(0.9)	2.4(0.9)	2.5(1.1)
COP_{LD}	1.0(0.7)	0.8(0.6)	1.0(0.7)	0.7(0.5)	0.1(1.5)	0.5(2.6)	0.3(1.9)	1.2(4.3)
Knee								
GC EX/FL angle	-12.4(5.7)	-14.0(7.3)	-12.7(6.3)	-12.1(7.4)	-11.2(7.3)	-12.7(6.3)	-12.7(10.1)	-11.4(10.4)
GC AB/AD angle	-2.4(3.7)	-3.2(5.4)	-1.0(6.0)	-3.2(5.7)	-0.5(5.5)	2.7(4.7)	-0.7(5.6)	3.0(4.2)
GC IR/ER angle	-6.7(6.7)	-8.4(6.6)	-8.8(9.2)	-8.7(6.3)	-15.3(4.7)	-13.5(7.2)	-12.9(9.5)	-15.1(8.9)
Peak EX angle	-12.4(5.1)	-14.0(7.2)	-12.7(6.3)	-12.0(7.4)	-11.1(7.3)	-12.4(5.9)	-12.2(9.6)	-11.0(10.2)
Peak FL angle	-47.1(6.1)	-46.7(6.2)	-47.2(7.4)	-45.8(6.2)	-40.0(7.6)	-37.5(6.9)	-39.9(9.6)	-35.6(9.5)
Peak AD angle	-2.2(3.7)	-2.7(5.7)	-0.9(5.9)	-2.5(6.6)	9.3(12.7)	14.1(11.1)	11.0(14.6)	16.8(13.5)
Peak AB angle	-11.2(7.4)	10.3(9.1)	-10.9(6.8)	-12.8(9.2)	-2.4(8.6)	0.5(6.8)	-3.2(8.5)	1.4(5.2)
Peak IR angle	-5.2(6.3)	-7.3(6.8)	-5.6(6.8)	-7.0(6.0)	-14.3(5.3)	-12.5(7.5)	-12.5(9.7)	-13.6(9.0)
Peak ER angle	-12.5(7.3)	-14.8(7.7)	-15.7(9.5)	-15.9(8.2)	-25.4(7.5)	-24.2(8.2)	-29.1(10.0)	-27.5(7.4)
Peak EX moment	3.3(2.4)	4.5(2.5)	4.1(2.3)	4.1(2.7)	1.9(1.3)	1.7(1.7)	1.6(1.4)	2.1(1.8)
Peak FL moment	-0.1(0.6)	-0.1(0.7)	-0.3(1.1)	-0.4(1.3)	-0.2(0.6)	-0.7(1.2)	-0.4(1.1)	-0.6(1.3)
Peak AD moment	1.0(1.5)	2.2(2.1)	1.7(1.6)	1.8(1.9)	0.4(0.7)	0.03(0.2)	0.4(0.7)	0.04(0.2)
Peak AB moment	-1.0(1.2)	-0.3(0.8)	-0.9(1.5)	-1.2(2.2)	-2.4(1.9)	-2.6(1.6)	-2.7(2.1)	-3.3(1.6)
Peak IR moment	1.1(1.2)	1.9(1.3)	1.7(1.2)	1.7(1.4)	1.4(1.0)	1.1(1.2)	1.3(1.1)	1.5(1.2)
Peak ER moment	-0.1(0.4)	0.1(4.2)	-0.2(0.6)	-0.3(0.8)	-0.1(0.3)	-0.3(0.6)	-0.3(0.6)	-0.4(0.8)
Peak P_S	-0.7(4.2)	-1.1(4.2)	-1.7(4.3)	-2.5(6.3)	-1.8(3.3)	-4.3(8.0)	-3.7(7.8)	-3.9(7.3)
Peak P_F	-1.8(2.8)	-0.8(2.2)	-3.5(6.8)	-2.0(4.4)	-2.6(2.6)	-1.6(2.0)	-3.7(3.7)	-3.6(4.0)
Peak P_T	-3.5(4.2)	-4.7(6.1)	-6.2(4.9)	-5.5(5.7)	-1.4(1.4)	-1.7(1.5)	-2.2(3.1)	-2.3(2.3)

Joint angle, moment, and power absorption measures are in units of deg, Nm/kg, and W/kg, respectively. ML GRF are in N/kg, and COP_{LD} in cm. Sign convention is provided to clarify direction. Negative values are FL, AB, PF, EV, ER, and power absorption while positive values are EX, AD, DF, INV, and IR.

Table A11. Bivariate correlations (ρ) between absolute joint positioning errors for JPS reference angles and absolute error for GC angles during dynamic maneuvers. Significant effects ($p < 0.05$) are in bold while effects that approached significance ($0.05 < p < 0.1$) are highlighted.

				JPS Abs Error				JPS True Error			
Speed	**Maneuver**	**JPS Configuration**	**Measures**	**-15**	**-5**	**+5**	**+15**	**-15**	**-5**	**+5**	**+15**
Slower	Run	P/D	GC Abs Error X	0.34	0.05	-0.21	-0.21	-0.29	-0.23	-0.05	-0.28
	Run	I/E	GC Abs Error Y	0.27	0.06	0.24	0.23	0.23	-0.13	-0.27	0.15
	JS	P/D	GC Abs Error X	0.05	0.05	**0.44**	0.33	0.09	-0.22	0.30	0.34
	JS	I/E	GC Abs Error Y	-0.28	0.14	0.13	-0.24	-0.24	0.03	0.11	-0.20
	Cut	P/D	GC Abs Error X	-0.18	-0.08	-0.08	0.35	0.23	0.11	-0.07	-0.06
	Cut	I/E	GC Abs Error Y	0.07	0.06	-0.01	0.11	0.21	-0.06	-0.25	0.01
	Shuttle	P/D	GC Abs Error X	-0.05	-0.24	-0.06	-0.18	-0.12	0.03	0.21	-0.03
	Shuttle	I/E	GC Abs Error Y	-0.19	-0.16	0.13	-0.18	-0.33	-0.14	0.30	-0.09
Faster	Run	P/D	GC Abs Error X	0.26	-0.07	-0.08	-0.15	0.07	0.03	-0.34	-0.28
	Run	I/E	GC Abs Error Y	0.21	0.22	-0.10	-0.10	0.20	-0.06	-0.17	-0.18
	JS	P/D	GC Abs Error X	-0.23	-0.14	0.07	-0.03	0.18	0.07	-0.08	0.08
	JS	I/E	GC Abs Error Y	-0.35	-0.18	-0.05	-0.14	-0.16	0.14	0.34	-0.12
	Cut	P/D	GC Abs Error X	0.06	-0.15	-0.24	-0.23	-0.25	-0.14	-0.04	-0.32
	Cut	I/E	GC Abs Error Y	0.32	0.05	0.03	0.08	**-0.42**	-0.35	-0.004	**-0.49**
	Shuttle	P/D	GC Abs Error X	0.32	-0.26	-0.15	0.13	-0.02	0.32	-0.16	-0.13
	Shuttle	I/E	GC Abs Error Y	-0.12	0.06	0.06	-0.02	0.09	-0.25	-0.09	-0.10

Measures refer to mean joint positioning error at the instant of GC with “X” and “Y” referring errors about the PF/DF and INV/EV axes of the ankle, respectively.

Table A12. Correlation coefficients (ρ) for **ankle** ground contact angles of forward drop and side drop landings and Run and JS dynamic athletic maneuvers performed at slower (2.5 m/s) and faster (3.6 m/s) running speeds. Significant effects ($p < 0.05$) are in bold while effects that approached significance ($0.05 < p < 0.1$) are highlighted.

			Drop Landing Measure		
Drop Type	**Maneuver/Speed**	**Dynamic Measure**	GC PF/DF angle	GC INV/EV angle	GC IR/ER angle
Forward	Run/Slower	GC PF/DF angle	0.38	0.03	-0.03
		GC INV/EV angle	0.01	**0.44**	-0.16
		GC IR/ER angle	-0.35	0.07	**0.41**
	JS/Slower	GC PF/DF angle	**0.56**	0.12	-0.09
		GC INV/EV angle	-0.14	0.38	0.01
		GC IR/ER angle	-0.39	0.05	**0.40**
	Run/Faster	GC PF/DF angle	**0.47**	0.12	-0.16
		GC INV/EV angle	-0.05	0.23	0.10
		GC IR/ER angle	-0.41	-0.05	**0.53**
	JS/Faster	GC PF/DF angle	0.34	0.07	-0.06
		GC INV/EV angle	-0.14	0.35	-0.09
		GC IR/ER angle	-0.35	0.13	0.33
Side	Run/Slower	GC PF/DF angle	**0.50**	0.02	0.16
		GC INV/EV angle	-0.02	0.08	-0.22
		GC IR/ER angle	-0.47	-0.03	**0.40**
	JS/Slower	GC PF/DF angle	**0.59**	-0.25	0.01
		GC INV/EV angle	-0.02	0.27	-0.04
		GC IR/ER angle	-0.34	0.30	**0.42**
	Run/Faster	GC PF/DF angle	**0.66**	-0.02	0.10
		GC INV/EV angle	-0.11	0.05	-0.06
		GC IR/ER angle	-0.52	0.08	**0.67**
	JS/Faster	GC PF/DF angle	**0.54**	-0.23	-0.02
		GC INV/EV angle	0.03	0.24	-0.20
		GC IR/ER angle	-0.19	0.32	0.34

Table A13. Correlation coefficients (ρ) for **ankle** ground contact angles of forward drop and side drop landings and Cut and Shuttle dynamic athletic maneuvers performed at slower (2.5 m/s) and faster (3.6 m/s) running speeds. Significant effects ($p < 0.05$) are in bold while effects that approached significance ($0.05 < p < 0.1$) are highlighted.

			Drop Landing Measure		
Drop Type	**Maneuver/Speed**	**Dynamic Measure**	GC PF/DF angle	GC INV/EV angle	GC IR/ER angle
Forward	Cut/Slower	GC PF/DF angle	**0.68**	-0.25	0.03
		GC INV/EV angle	0.24	0.22	0.03
		GC IR/ER angle	0.03	-0.04	**0.42**
	Shuttle/Slower	GC PF/DF angle	**0.46**	0.08	0.04
		GC INV/EV angle	-0.14	0.27	0.03
		GC IR/ER angle	0.21	-0.12	0.38
	Cut/Faster	GC PF/DF angle	**0.50**	0.03	-0.13
		GC INV/EV angle	-0.04	**0.58**	-0.10
		GC IR/ER angle	-0.16	0.13	**0.43**
	Shuttle/Faster	GC PF/DF angle	0.12	0.30	-0.35
		GC INV/EV angle	-0.50	**0.46**	0.01
		GC IR/ER angle	0.03	0.20	0.06
Side	Cut/Slower	GC PF/DF angle	**0.56**	-0.34	-0.17
		GC INV/EV angle	0.45	0.18	-0.14
		GC IR/ER angle	0.05	0.16	**0.57**
	Shuttle/Slower	GC PF/DF angle	**0.52**	0.09	0.00
		GC INV/EV angle	-0.33	0.34	-0.13
		GC IR/ER angle	0.22	0.02	**0.44**
	Cut/Faster	GC PF/DF angle	**0.58**	-0.22	-0.06
		GC INV/EV angle	0.15	**0.41**	-0.02
		GC IR/ER angle	-0.13	0.31	**0.48**
	Shuttle/Faster	GC PF/DF angle	**0.42**	0.28	0.01
		GC INV/EV angle	-0.44	**0.43**	-0.02
		GC IR/ER angle	0.16	0.36	**0.47**

Appendix B.

Chapter 3 Statistical Results

Throughout this appendix, abbreviations FL, EX, AB, AD refer to knee flexion, extension, abduction and adduction measures, respectively. The abbreviations DF, PF, INV, and EV refer to ankle dorsiflexion, plantarflexion, inversion, and eversion, respectively. For both joints, IR is internal rotation and ER is external rotation. Column labels indicate whether the statistics refer to pre-fatigue group effects ("Pre"), fatigue effects ("Main"), or ANOVA or ANCOVA results for change scores.

Table B1. Statistical results for the effects of group (G), observation time (T), and their interaction (G x T) on ankle strength, RPE scores and MF measures during Fatigue Protocol #1. Significant effects ($p < 0.05$) are in bold while effects that approached significance ($0.05 < p < 0.1$) are highlighted.

			Pre	ANOVA		
PF/DF	**Measures**	**Stats**	**G**	**G**	**T**	**G x T**
PF	Ankle Strength ($\Delta MVIC_{PF}$)	*t,F*	-0.44	0.02	**8.93**	2.26
		p	0.67	0.90	**0.0002**	0.11
	RPE Score (ΔRPE_{PF})	*t,F*	--	1.68	**32.22**	1.28
		p	--	0.21	**<0.0001**	0.28
	PL MF ($\%\Delta MF_{PL}$)	*t,F*	--	1.75	0.32	0.76
		p	--	0.20	0.81	0.53
	GS MF ($\%\Delta MF_{GS}$)	*t,F*	--	0.02	0.54	1.90
		p	--	0.89	0.66	0.16
	Strength/RPE Correlation (R^2)	*t*	--	0.003	--	--
		p	--	0.99	--	--
DF	Ankle Strength ($\Delta MVIC_{DF}$)	*t,F*	-0.52	1.88	**31.54**	0.84
		p	0.60	0.17	**<0.0001**	0.46
	RPE Score (ΔRPE_{DF})	*t,F*	--	2.39	**29.06**	0.21
		p	--	0.13	**<0.0001**	0.77
	TA MF ($\%\Delta MF_{TA}$)	*t,F*	--	3.08	**9.46**	0.91
		p	--	0.091	**0.003**	0.45
	Strength/RPE Correlation (R^2)	*t*	--	-0.41	--	--
		p	--	0.68	--	--

Table B2. Mean (SD) of change scores for measures of ankle strength, RPE, and MF during Fatigue Protocol #1. Results are tabulated as individual group means at each observation time t.

	t = 4 min		t = 8 min		t = 12 min		t = 16 min	
Measures*	**C**	**U**	**C**	**U**	**C**	**U**	**C**	**U**
$MVIC_{PF}$	-5.1(3.6)	-6.3(4.8)	-7.8(7.6)	-7.5(6.0)	-9.3(7.9)	-8.9(6.3)	-8.2(7.2)	-10.7(6.9)
RPE_{PF}	3.1(1.0)	2.6(1.5)	4.4(1.0)	3.3(1.8)	5.0(1.5)	4.0(2.2)	5.1(1.9)	4.6(2.1)
MF_{PL}	0.8(7.6)	7.1(14.6)	3.7(12.9)	5.2(11.6)	0.6(5.0)	4.9(15.1)	0.3(6.8)	6.5(10.9)
MF_{GS}	1.8(11.7)	4.7(8.9)	4.1(12.6)	1.5(11.6)	1.0(7.4)	1.4(15.1)	4.1(6.8)	1.3(10.1)
PF R^2	--	--	--	--	--	--	0.47(0.29)	0.47(0.34)
$MVIC_{DF}$	-8.2(5.2)	-7.0(4.9)	-10.5(6.8)	-9.3(6.9)	-11.9(6.4)	-9.5(5.7)	-11.3(6.4)	-9.7(6.4)
RPE_{DF}	4.4(2.0)	3.6(1.9)	5.9(1.7)	5.1(1.5)	6.6(1.3)	5.6(1.4)	6.8(1.9)	6.3(1.5)
MF_{TA}	-8.0(8.1)	-14.0(10.4)	-6.2(9.3)	-9.3(10.2)	-1.8(10.0)	-10.0(10.1)	-1.8(11.5)	-6.3(6.3)
DF R^2	--	--	--	--	--	--	0.60(0.38)	0.55(0.32)

*MVIC and MF measures are in units of Nm and Hz while RPE and R^2 measures are unitless.

Table B3. Statistical results for the effects of group (G), fatigue (F), covariate (C), and the group x covariate interaction (G x C) on ankle and knee GC angles during forward and side drop landings. Significant effects (p < 0.05) are in bold while effects that approached significance (0.05 < p < 0.1) are highlighted.

			ANKLE					KNEE				
			Pre	**Main**	**ANCOVA**			**Pre**	**Main**	**ANCOVA**		
Drop	**Measures**	**Stats**	**G**	**F**	**G**	**C**	**G x C**	**G**	**F**	**G**	**C**	**G x C**
FWD	GC DF/PF or EX/FL ang	*t,F*	-1.68	**-11.36**	0.01	**8.65**	0.40	-1.07	**-2.15**	0.02	1.53	0.22
		p	0.096	**0.0001**	0.91	**0.01**	0.53	0.29	**0.041**	0.90	0.23	0.64
	GC INV/EV or AD/AB ang	*t,F*	1.54	**-2.83**	0.13	3.58	0.77	0.17	-1.89	0.70	1.02	0.13
		p	0.13	**0.009**	0.72	0.07	0.39	0.87	0.070	0.41	0.32	0.72
	GC IR/ER ang	*t,F*	-0.66	1.06	0.22	2.30	0.30	-1.11	**2.56**	2.01	1.79	0.87
		p	0.51	0.30	0.65	0.14	0.59	0.27	**0.017**	0.17	0.19	0.36
SIDE	GC DF/PF or EX/FL ang	*t,F*	0.88	-3.92	0.01	0.58	0.61	-0.31	-0.005	0.24	0.35	1.23
		p	0.38	0.0005	0.92	0.46	0.44	0.76	0.99	0.63	0.56	0.28
	GC INV/EV or AD/AB ang	*t,F*	0.85	1.57	2.81	0.41	0.25	**2.14**	**2.41**	0.29	0.11	0.43
		p	0.40	0.13	0.11	0.53	0.62	**0.035**	**0.023**	0.59	0.75	0.52
	GC IR/ER ang	*t,F*	-1.69	1.63	0.04	**12.17**	0.76	-1.86	1.85	1.48	3.41	0.33
		p	0.094	0.11	0.85	**0.0019**	0.39	0.065	0.075	0.24	0.077	0.57

Table B4. Mean (SD) for pre-fatigue levels of GC angles during drop landings.

		Ankle		Knee	
Drop Type	**Measures**	**C**	**U**	**C**	**U**
Forward	GC DF/PF or EX/FL ang	-27.6(9.3)	-29.9(5.9)	-9.5(5.0)	-10.4(4.5)
	GC INV/EV or AD/AB ang	7.5(6.8)	9.1(4.8)	0.8(1.8)	0.8(2.4)
	GC IR/ER ang	4.5(2.1)	4.0(5.0)	1.1(4.8)	-0.2(8.1)
Side	GC DF/PF or EX/FL ang	-25.7(8.5)	-24.5(6.2)	-10.9(5.6)	-11.2(5.3)
	GC INV/EV or AD/AB ang	15.3(6.1)	16.3(7.5)	2.3(2.2)	3.4(3.4)
	GC IR/ER ang	5.5(3.2)	4.5(4.0)	0.4(5.3)	-1.4(5.4)

Table B5. Mean (SD) of change scores for GC angles during drop landings.

		Ankle		Knee	
Drop Type	**Measures**	**C**	**U**	**C**	**U**
Forward	ΔGC DF/PF or EX/FL ang	-5.5(2.4)	-5.3(2.6)	-1.1(3.0)	-1.4(3.1)
	Δ GC INV/EV or AD/AB ang	-1.5(3.7)	-2.2(3.3)	-0.3(1.7)	-0.9(1.7)
	Δ GC IR/ER ang	0.9(2.6)	0.2(2.8)	0.6(2.9)	2.9(3.7)
Side	Δ GC DF/PF or EX/FL ang	-3.1(4.7)	-3.3(4.1)	-0.2(1.9)	0.2(2.2)
	Δ GC INV/EV or AD/AB ang	-0.1(4.5)	3.1(5.5)	0.7(2.2)	1.2(1.9)
	Δ GC IR/ER ang	0.8(2.7)	1.2(3.7)	1.6(2.6)	0.6(3.7)

Table B6. Statistical results for the effects of group (G), fatigue (F), covariate (C), and the group x covariate interaction (G x C) on ankle and knee kinematic and kinetic measures during **landing phase** of **forward** drop landings. Significant effects ($p < 0.05$) are in bold while effects that approached significance ($0.05 < p < 0.1$) are highlighted.

		ANKLE					KNEE				
		Pre	Main	ANCOVA			Pre	Main	ANCOVA		
Measures	**Stats**	**G**	**F**	**G**	**C**	**G x C**	**G**	**F**	**G**	**C**	**G x C**
Peak DF/EX ang	*t,F*	-0.03	**-4.35**	0.11	**5.17**	0.001	-1.07	**-2.15**	0.02	1.53	0.22
	p	0.97	**0.0002**	0.74	**0.03**	0.98	0.29	**0.040**	0.90	0.23	0.64
Peak PF/FL ang	*t,F*	-1.65	**-11.30**	0.01	**8.65**	0.40	0.50	-0.27	3.74	**4.98**	0.0049
	p	0.10	**<0.0001**	0.91	**0.01**	0.53	0.62	0.79	0.065	**0.035**	0.95
Peak INV/AD ang	*t,F*	1.54	**-2.81**	0.14	3.65	0.76	0.002	**-2.30**	2.73	0.47	3.48
	p	0.13	**0.009**	0.71	0.07	0.39	0.99	**0.029**	0.11	0.50	0.074
Peak EV/AB ang	*t,F*	-1.22	0.15	0.83	0.08	1.62	**-2.45**	-1.60	**7.26**	0.51	0.96
	p	0.23	0.88	0.37	0.79	0.22	**0.016**	0.12	**0.013**	0.48	0.34
Peak IR ang	*t,F*	-0.34	1.47	0.16	2.19	0.01	1.46	**2.81**	0.06	2.72	0.38
	p	0.74	0.15	0.70	0.15	0.93	0.15	**0.0095**	0.80	0.11	0.54
Peak ER ang	*t,F*	**-2.65**	0.10	0.18	0.02	0.88	-0.23	1.38	0.51	**5.41**	0.10
	p	**0.0098**	0.92	0.68	0.89	0.36	0.82	0.18	0.48	**0.029**	0.75
Peak DF/EX mom	*t,F*	-1.85	**4.03**	2.06	0.43	0.14	**-1.99**	**3.42**	**6.60**	0.89	1.28
	p	0.087	**0.004**	0.16	0.52	0.71	**0.049**	**0.0021**	**0.017**	0.36	0.27
Peak PF/FL mom	*t,F*	**-4.24**	**2.29**	1.29	**5.10**	0.02	**-4.15**	1.20	2.13	**5.06**	0.03
	p	**<0.0001**	**0.03**	0.27	**0.03**	0.88	**<0.0001**	0.24	0.16	**0.034**	0.87
Peak INV/AD mom	*t,F*	-0.41	-0.08	0.07	**6.36**	0.69	-0.45	-0.50	0.18	**13.61**	0.89
	p	0.69	0.93	0.80	**0.02**	0.41	0.66	0.62	0.68	**0.0013**	0.36
Peak EV/AB mom	*t,F*	1.73	-0.38	0.97	1.58	0.06	1.69	0.31	0.47	1.77	0.47
	p	0.087	0.71	0.33	0.22	0.80	0.093	0.76	0.50	0.20	0.50
Peak IR mom	*t,F*	-1.09	-0.38	1.59	**7.01**	0.07	-1.23	0.47	1.43	3.69	0.03
	p	0.28	0.71	0.22	**0.01**	0.79	0.22	0.65	0.24	0.067	0.87
Peak ER mom	*t,F*	**2.19**	-1.74	0.33	0.49	2.11	1.44	-0.71	0.09	1.02	1.18
	p	**0.031**	0.09	0.57	0.49	0.16	0.15	0.49	0.76	0.32	0.29
Peak Abs P_S	*t,F*	**-4.02**	**2.24**	1.55	**5.34**	0.0003	**-4.01**	1.26	2.11	**7.85**	0.07
	p	**0.0001**	**0.03**	0.23	**0.03**	0.99	**0.0001**	0.22	0.16	**0.010**	0.79
Peak Gen P_S	*t,F*	-1.13	**3.57**	0.60	0.34	0.04	**-2.15**	**2.57**	0.65	2.50	0.03
	p	0.26	**0.001**	0.44	0.57	0.84	**0.034**	**0.016**	0.43	0.13	0.88
Peak Abs P_F	*t,F*	0.59	0.98	1.93	**8.44**	0.0004	0.58	-1.55	1.71	1.32	0.41
	p	0.56	0.34	0.18	**0.01**	0.98	0.57	0.13	0.20	0.26	0.53
Peak Gen P_F	*t,F*	-1.87	-0.31	2.34	2.21	0.72	-0.97	-0.87	0.0001	3.20	3.11
	p	0.065	0.76	0.14	0.15	0.40	0.33	0.39	0.99	0.086	0.091
Peak Abs P_T	*t,F*	-0.02	0.50	**4.79**	**4.73**	1.49	2.15	-1.14	0.0009	0.0049	0.0035
	p	0.98	0.62	**0.04**	**0.04**	0.23	0.034	0.26	0.98	0.94	0.95
Peak Gen P_T	*t,F*	-0.21	1.31	0.13	0.74	0.39	1.56	-0.42	0.23	31.83	0.27
	p	0.83	0.20	0.72	0.40	0.54	0.12	0.68	0.64	<0.0001	0.61
ML COP	*t,F*	-1.48	-1.10	0.16	2.29	0.39	--	--	--	--	--
	p	0.14	0.28	0.69	0.14	0.54	--	--	--	--	--

Table B7. Statistical results for the effects of group (G), fatigue (F), covariate (C), and the group x covariate interaction (G x C) on ankle and knee kinematic and kinetic measures during **stance phase** of **forward** drop landings. Significant effects (p < 0.05) are in bold while effects that approached significance (0.05 < p < 0.1) are highlighted.

		ANKLE					KNEE				
		Pre	Main	ANCOVA			Pre	Main	ANCOVA		
Measures	Stats	G	F	G	C	G x C	G	F	G	C	G x C
Peak DF/EX ang	*t,F*	0.09	**-2.76**	0.78	2.91	1.39	1.67	1.34	0.05	2.58	1.32
	p	0.93	**0.010**	0.38	0.10	0.25	0.097	0.19	0.82	0.12	0.26
Peak PF/FL ang	*t,F*	-0.82	**-4.26**	0.35	1.52	0.75	0.04	**-2.64**	4.01	3.55	0.001
	p	0.41	**0.0002**	0.56	0.23	0.39	0.97	**0.014**	0.057	0.072	0.98
Peak INV/AD ang	*t,F*	1.07	1.05	1.59	0.31	8.12	-0.27	-1.51	0.35	3.96	0.63
	p	0.29	0.30	0.22	0.58	0.0091	0.78	0.14	0.56	0.059	0.43
Peak EV/AB ang	*t,F*	-0.97	-1.67	0.74	0.13	3.47	-1.78	**-3.07**	3.38	0.21	2.48
	p	0.33	0.11	0.40	0.72	0.075	0.077	**0.005**	0.080	0.65	0.13
Peak IR ang	*t,F*	0.34	**2.27**	1.20	1.31	2.34	0.62	**4.53**	1.41	4.03	0.91
	p	0.73	**0.031**	0.28	0.26	0.14	0.54	**0.0001**	0.25	0.056	0.35
Peak ER ang	*t,F*	**-3.05**	**-2.47**	0.19	1.96	2.81	0.20	-1.32	1.62	3.44	1.29
	p	**0.0031**	**0.021**	0.67	0.18	0.11	0.84	0.20	0.22	0.076	0.27
Peak DF/EX mom	*t,F*	-0.64	**3.86**	**7.47**	0.23	0.13	-1.19	**3.73**	**4.51**	3.28	2.02
	p	0.52	**0.0006**	**0.012**	0.63	0.73	0.24	**0.0009**	**0.044**	0.083	0.17
Peak PF/FL mom	*t,F*	-1.67	**2.55**	**4.76**	**6.57**	0.0002	-1.55	2.00	**4.65**	**7.84**	0.01
	p	0.097	**0.017**	**0.039**	**0.017**	0.99	0.12	0.056	**0.041**	**0.010**	0.92
Peak INV/AD mom	*t,F*	0.28	-0.23	0.33	2.99	0.05	0.66	0.51	0.16	1.24	0.72
	p	0.78	0.82	0.57	0.10	0.83	0.51	0.61	0.69	0.28	0.40
Peak EV/AB mom	*t,F*	1.29	-1.21	0.69	2.47	3.65	1.72	-0.03	0.86	3.03	0.46
	p	0.20	0.24	0.42	0.13	0.068	0.088	0.98	0.36	0.095	0.50
Peak IR mom	*t,F*	0.95	1.31	**6.29**	3.20	0.81	0.31	**2.60**	4.60	0.13	0.04
	p	0.34	0.20	**0.019**	0.087	0.38	0.75	**0.015**	0.043	0.72	0.85
Peak ER mom	*t,F*	0.45	-1.74	1.00	3.18	2.78	-0.50	**-2.21**	0.10	**4.56**	2.91
	p	0.66	0.093	0.33	0.087	0.11	0.62	**0.036**	0.75	**0.043**	0.10
Peak Abs P_S	*t,F*	**-3.05**	-1.46	0.01	3.93	0.41	**3.45**	-1.01	0.01	1.44	0.22
	p	**0.0029**	0.16	0.92	0.059	0.53	**0.0009**	0.32	0.92	0.24	0.64
Peak Gen P_S	*t,F*	0.95	**4.12**	3.45	0.28	1.29	-0.17	**4.45**	**6.56**	1.93	**10.47**
	p	0.35	**0.0003**	0.075	0.60	0.27	0.87	**0.0001**	**0.017**	0.18	**0.0035**
Peak Abs P_F	*t,F*	0.49	**-2.78**	1.33	**5.20**	0.91	1.79	-0.25	2.91	**5.09**	**17.73**
	p	0.63	**0.0098**	0.26	**0.032**	0.35	0.075	0.80	0.10	**0.034**	**0.0003**
Peak Gen P_F	*t,F*	-1.16	**3.35**	0.001	**4.34**	7.98	-1.39	1.33	1.71	2.16	0.49
	p	0.25	**0.0026**	0.97	**0.049**	0.0099	0.17	0.19	0.20	0.16	0.49
Peak Abs P_T	*t,F*	**-2.26**	**13.52**	**5.62**	0.08	**6.17**	0.60	**-2.13**	**9.30**	0.93	**32.81**
	p	**0.026**	**<0.0001**	**0.027**	0.77	**0.021**	0.55	**0.043**	**0.0055**	0.34	**<0.0001**
Peak Gen P_T	*t,F*	0.64	**2.22**	2.60	0.42	1.97	-1.35	**3.32**	0.78	2.16	1.26
	p	0.53	**0.035**	0.12	0.52	0.17	0.18	**0.028**	0.38	0.15	0.27
ML COP	*t,F*	-0.12	**2.27**	0.14	**6.46**	0.10	--	--	--	--	--
	p	0.90	**0.032**	0.71	**0.018**	0.76	--	--	--	--	--

Table B8. Statistical results for the effects of group (G), fatigue (F), covariate (C), and the group x covariate interaction (G x C) on ankle and knee kinematic and kinetic measures during **landing phase** of **side** drop landings. Significant effects ($p < 0.05$) are in bold while effects that approached significance ($0.05 < p < 0.1$) are highlighted.

		ANKLE					**KNEE**				
		Pre	**Main**	**ANCOVA**			**Pre**	**Main**	**ANCOVA**		
Measures	**Stats**	**G**	**F**	**G**	**C**	**G x C**	**G**	**F**	**G**	**C**	**G x C**
Peak DF/EX ang	*t,F*	-0.75	**-6.06**	0.52	2.18	**5.29**	-0.31	-0.005	0.24	0.35	1.23
	p	0.46	**<0.0001**	0.48	0.15	**0.03**	0.76	0.99	0.63	0.56	0.28
Peak PF/FL ang	*t,F*	0.88	**-3.85**	0.02	0.60	0.57	0.24	**2.99**	0.0001	0.72	1.93
	p	0.38	**0.0007**	0.90	0.45	0.46	0.81	**0.0061**	0.99	0.41	0.18
Peak INV/AD ang	*t,F*	0.85	1.64	2.61	0.34	0.31	**2.04**	1.90	0.05	0.09	5.15
	p	0.40	0.11	0.12	0.56	0.58	**0.044**	0.069	0.82	0.77	0.03
Peak EV/AB ang	*t,F*	-0.56	-0.29	1.36	**5.50**	2.50	1.61	**2.27**	0.48	0.21	1.14
	p	0.58	0.77	0.26	**0.03**	0.13	0.11	**0.031**	0.49	0.65	0.30
Peak IR ang	*t,F*	-0.85	1.45	0.15	**8.35**	0.61	-0.34	0.65	0.33	1.85	2.11
	p	0.40	0.16	0.70	**0.01**	0.44	0.73	0.52	0.57	0.19	0.16
Peak ER ang	*t,F*	**-2.26**	-0.61	0.10	0.62	4.11	-1.81	1.90	0.94	2.50	1.44
	p	**0.026**	0.55	0.75	0.44	0.054	0.072	0.068	0.34	0.13	0.24
Peak DF/EX mom	*t,F*	**2.21**	**2.16**	1.48	0.01	0.05	**3.06**	1.66	**11.65**	**8.24**	**20.38**
	p	**0.031**	**0.040**	0.24	0.92	0.82	**0.003**	0.11	**0.0023**	**0.0084**	**0.0001**
Peak PF/FL mom	*t,F*	1.24	**2.31**	1.24	**9.63**	1.31	0.67	1.34	0.56	3.95	3.24
	p	0.22	**0.029**	0.28	**0.0049**	0.26	0.51	0.19	0.46	0.060	0.086
Peak INV/AD mom	*t,F*	**3.66**	0.46	0.0005	0.69	1.53	**2.93**	1.77	1.12	0.42	0.002
	p	**0.0005**	0.65	0.98	0.42	0.23	**0.0047**	0.090	0.30	0.53	0.97
Peak EV/AB mom	*t,F*	**3.52**	-0.02	0.06	0.26	0.06	**3.01**	-0.038	0.01	1.22	0.0002
	p	**0.0006**	0.99	0.80	0.61	0.81	**0.0032**	0.97	0.93	0.28	0.95
Peak IR mom	*t,F*	-1.49	**3.29**	0.03	1.13	2.71	-1.84	**7.30**	0.12	1.57	**4.79**
	p	0.14	**0.0028**	0.87	0.30	0.11	0.070	**<0.0001**	0.73	0.22	**0.039**
Peak ER mom	*t,F*	-0.37	**2.82**	1.62	3.84	0.01	1.74	**13.37**	0.27	0.90	0.08
	p	0.71	**0.0089**	0.22	0.062	0.93	0.086	**<0.0001**	0.61	0.35	0.78
Peak Abs P_S	*t,F*	1.54	1.12	0.46	0.22	1.07	-0.10	1.13	0.61	1.85	2.79
	p	0.13	0.27	0.50	0.64	0.31	0.92	0.27	0.44	0.19	0.11
Peak Gen P_S	*t,F*	1.78	2.02	4.12	0.02	0.73	**3.05**	0.92	**16.39**	**7.66**	**32.93**
	p	0.079	0.053	0.054	0.90	0.40	**0.0031**	0.37	**0.0005**	**0.011**	**<0.0001**
Peak Abs P_F	*t,F*	**-3.31**	-0.43	0.07	0.50	0.84	**-2.09**	0.05	0.64	0.04	0.01
	p	**0.0016**	0.67	0.80	0.49	0.37	**0.039**	0.96	0.43	0.84	0.92
Peak Gen P_F	*t,F*	**-2.56**	0.45	0.0001	0.62	2.27	-0.39	1.03	0.03	2.48	0.06
	p	**0.012**	0.66	0.96	0.44	0.14	0.70	0.31	0.86	0.13	0.81
Peak Abs P_T	*t,F*	**-2.02**	-1.32	2.12	0.08	0.67	-1.23	-0.87	0.76	**7.19**	0.65
	p	**0.047**	0.20	0.16	0.78	0.42	0.22	0.39	0.39	**0.013**	0.43
Peak Gen P_T	*t,F*	0.54	1.13	0.06	0.01	0.89	-1.57	0.38	2.25	0.20	1.02
	p	0.59	0.27	0.81	0.92	0.36	0.12	0.71	0.15	0.66	0.32
ML COP (cm)	*t,F*	-0.67	-0.33	0.06	1.33	0.06	--	--	--	--	--
	p	0.50	0.74	0.80	0.26	0.81	--	--	--	--	--

Table B9. Statistical results for the effects of group (G), fatigue (F), covariate (C) and the group x covariate interaction (G x C) on ankle and knee kinematic and kinetic measures during **stance phase** of **side** drop landings. Significant effects ($p < 0.05$) are in bold while effects that approached significance ($0.05 < p < 0.1$) are highlighted.

		ANKLE					**KNEE**				
		Pre	**Main**	**ANCOVA**			**Pre**	**Main**	**ANCOVA**		
Measures	**Stats**	**G**	**F**	**G**	**C**	**G x C**	**G**	**F**	**G**	**C**	**G x C**
Peak DF/EX ang	*t,F*	-1.48	**-5.37**	0.52	2.86	3.32	1.34	0.91	0.64	0.0004	0.09
	p	0.14	**<0.0001**	0.48	0.10	0.081	0.18	0.37	0.43	0.98	0.77
Peak PF/FL ang	*t,F*	-1.57	**-10.12**	2.57	0.43	0.24	0.79	-0.50	0.10	0.90	**4.95**
	p	0.12	**<0.0001**	0.12	0.52	0.63	0.43	0.62	0.76	0.35	**0.036**
Peak INV/AD ang	*t,F*	1.53	1.67	0.32	**6.63**	0.01	**2.64**	1.92	0.15	0.12	3.04
	p	0.13	0.11	0.58	**0.017**	0.92	**0.0096**	0.065	0.70	0.73	0.094
Peak EV/AB ang	*t,F*	**-2.58**	-1.42	1.16	2.00	2.12	-1.65	1.39	0.87	3.89	1.02
	p	**0.011**	0.17	0.29	0.17	0.16	0.10	0.18	0.36	0.061	0.32
Peak IR ang	*t,F*	-0.14	-0.10	0.98	**5.01**	0.13	1.31	**3.86**	0.65	1.82	0.01
	p	0.89	0.92	0.33	**0.035**	0.72	0.19	**0.0006**	0.43	0.19	0.93
Peak ER ang	*t,F*	**-5.06**	-1.40	0.0001	0.02	1.91	**-2.44**	0.24	0.46	**5.76**	1.23
	p	**<0.0001**	0.17	0.99	0.88	0.18	**0.016**	0.81	0.50	**0.025**	0.28
Peak DF/EX mom	*t,F*	**2.42**	1.57	1.09	**4.35**	0.0001	**2.69**	1.24	1.22	2.00	1.60
	p	**0.017**	0.13	0.31	**0.048**	0.99	**0.0083**	0.23	0.28	0.17	0.22
Peak PF/FL mom	*t,F*	**2.37**	**2.14**	2.15	**5.17**	1.70	1.74	0.51	0.74	**6.82**	2.54
	p	**0.019**	**0.041**	0.16	**0.032**	0.20	0.084	0.61	0.40	**0.016**	0.12
Peak INV/AD mom	*t,F*	**4.98**	**2.71**	0.31	2.30	0.18	**4.14**	1.99	1.88	3.14	0.04
	p	**<0.0001**	**0.011**	0.58	0.14	0.67	**<0.0001**	0.058	0.18	0.090	0.85
Peak EV/AB mom	*t,F*	**5.54**	1.82	0.58	0.68	0.0001	**4.38**	1.19	0.11	**5.27**	0.12
	p	**<0.0001**	0.080	0.45	0.42	0.99	**<0.0001**	0.25	0.74	**0.031**	0.73
Peak IR mom	*t,F*	1.93	-0.16	0.0025	1.29	1.61	1.79	-1.22	2.04	**72.94**	**15.69**
	p	0.057	0.87	0.96	0.27	0.22	0.077	0.23	0.17	**<0.0001**	**0.0006**
Peak ER mom	*t,F*	-1.77	0.57	0.0001	3.09	0.08	-1.57	**19.35**	0.13	**13.17**	**4.67**
	p	0.080	0.57	0.99	0.094	0.78	0.12	**<0.0001**	0.72	**0.0013**	**0.041**
Peak Abs P_S	*t,F*	**4.07**	0.43	0.37	0.66	0.22	1.91	0.39	0.16	0.50	**8.70**
	p	**<0.0001**	0.67	0.55	0.42	0.65	0.060	0.70	0.69	0.49	**0.0079**
Peak Gen P_S	*t,F*	-1.94	0.76	0.01	1.70	0.0001	1.32	0.17	0.40	2.74	0.14
	p	0.055	0.45	0.93	0.21	0.99	0.19	0.87	0.53	0.11	0.71
Peak Abs P_F	*t,F*	-1.60	-1.26	0.73	1.44	**6.86**	0.84	-1.28	0.002	0.004	2.33
	p	0.11	0.22	0.40	0.24	**0.015**	0.40	0.21	0.97	0.95	0.14
Peak Gen P_F	*t,F*	**-3.66**	-0.13	0.72	3.08	2.77	-1.08	1.03	0.39	2.83	0.01
	p	**0.0004**	0.89	0.41	0.092	0.11	0.28	0.31	0.54	0.11	0.91
Peak Abs P_T	*t,F*	-1.86	-0.69	0.12	0.51	0.27	-1.55	-0.65	2.29	4.20	3.54
	p	0.067	0.50	0.74	0.48	0.61	0.13	0.52	0.15	0.054	0.074
Peak Gen P_T	*t,F*	1.85	1.43	0.03	0.14	0.21	1.61	1.66	**7.40**	0.71	**16.96**
	p	0.069	0.16	0.86	0.71	0.65	0.11	0.11	**0.014**	0.41	**0.0006**
ML COP (cm)	*t,F*	1.79	0.56	2.12	2.46	**4.78**	--	--	--	--	--
	p	0.076	0.58	0.16	0.13	**0.039**	--	--	--	--	--

Table B10. Mean (SD) for pre-fatigue levels of ankle and knee kinematic and kinetic measures during drop landings. Results are tabulated as individual group means for each drop type.

	Forward Drop Landings				Side Drop Landings			
	Ankle		Knee		Ankle		Knee	
Measures	**C**	**U**	**C**	**U**	**C**	**U**	**C**	**U**
Land Phase								
Peak DF/EX ang	8.3(4.3)	8.3(5.1)	-9.5(5.0)	-10.4(4.5)	12.1(4.7)	11.5(5.5)	-10.9(5.6)	-11.2(5.3)
Peak PF/FL ang	27.6(9.1)	29.9(5.9)	39.4(7.8)	38.7(7.3)	25.7(8.5)	24.5(6.2)	37.5(7.8)	37.1(7.9)
Peak INV/AD ang	7.5(6.8)	9.1(4.8)	4.7(4.6)	4.7(5.9)	15.3(6.1)	16.3(7.5)	10.8(6.0)	13.5(8.3)
Peak EV/AB ang	10.5(2.7)	11.2(3.5)	0.04(3.2)	2.0(5.4)	8.5(4.1)	9.0(5.0)	-2.3(2.3)	-3.2(3.9)
Peak IR ang	5.3(1.8)	5.1(3.3)	5.5(6.1)	7.4(8.1)	5.6(3.3)	5.0(4.5)	8.0(6.3)	7.6(5.7)
Peak ER ang	0.2(1.7)	2.0(5.1)	1.3(6.3)	1.6(6.0)	-0.04(2.9)	1.2(3.2)	0.4(5.3)	2.2(5.8)
Peak DF/EX mom	3.8(2.1)	2.9(3.1)	6.3(2.0)	5.4(3.1)	-0.2(0.1)	-0.1(0.4)	0.3(0.6)	0.8(1.3)
Peak PF/FL mom	0.3(0.5)	1.0(1.1)	0.1(0.2)	0.5(0.7)	5.3(2.8)	4.7(2.9)	3.4(2.5)	3.1(2.7)
Peak INV/AD mom	1.2(2.2)	1.1(1.2)	0.8(1.7)	0.7(1.0)	-0.03(0.1)	0.3(0.8)	-0.03(0.1)	0.2(0.6)
Peak EV/AB mom	1.3(1.9)	0.7(1.5)	2.0(2.3)	1.3(1.7)	5.1(2.6)	3.3(3.0)	6.2(2.5)	4.6(3.4)
Peak IR mom	0.3(0.3)	0.2(0.2)	0.3(0.3)	0.2(0.2)	0.3(0.4)	0.2(0.4)	1.7(6.0)	0.4(0.5)
Peak ER mom	0.3(0.2)	0.2(0.2)	0.4(0.3)	0.3(0.2)	0.8(0.3)	0.8(0.4)	1.9(6.1)	0.6(0.4)
Peak Abs P_S	3.3(4.3)	9.0(10.4)	0.4(1.1)	2.4(3.8)	37.6(19.8)	32.4(18.5)	16.9(12.4)	17.2(19.6)
Peak Gen P_S	16.1(11.2)	13.3(15.5)	40.9(15.6)	33.8(21.0)	-1.0(0.8)	-0.6(1.7)	1.6(3.5)	4.9(7.4)
Peak Abs P_F	4.5(8.2)	3.8(3.6)	1.1(2.3)	0.9(1.4)	-0.1(0.6)	1.8(4.5)	0.2(1.1)	0.6(1.3)
Peak Gen P_F	4.1(7.0)	2.1(4.2)	2.7(4.1)	2.1(2.3)	17.1(10.9)	11.8(12.1)	13.8(10.3)	13.0(12.4)
Peak Abs P_T	0.6(0.7)	0.6(0.6)	0.9(0.8)	0.7(0.4)	0.3(0.4)	0.6(1.0)	0.4(0.4)	0.5(0.7)
Peak Gen P_T	0.4(0.4)	0.4(0.6)	0.4(0.5)	0.6(0.8)	0.7(0.7)	0.8(0.8)	1.6(1.1)	1.3(1.0)
ML COP	1.2(1.7)	1.0(1.4)	--	--	2.2(1.5)	2.1(1.3)	--	--
Stance Phase								
Peak DF/EX ang	11.7(4.6)	11.8(5.4)	-8.6(8.3)	-6.2(8.0)	15.0(5.4)	13.5(6.0)	-7.8(7.6)	-5.9(8.6)
Peak PF/FL ang	-0.04(4.3)	0.7(5.3)	43.9(9.0)	43.8(7.5)	1.5(4.7)	2.9(5.5)	44.8(9.0)	43.6(8.2)
Peak INV/AD ang	-7.0(3.4)	-6.2(4.6)	7.3(5.8)	7.0(6.9)	-4.7(5.7)	-3.2(6.0)	12.7(7.2)	16.5(8.9)
Peak EV/AB ang	13.6(2.4)	14.1(3.2)	1.9(4.7)	3.9(7.2)	13.5(3.4)	15.1(3.4)	-1.5(3.4)	-0.2(4.8)
Peak IR ang	2.9(3.0)	3.1(3.5)	12.4(7.6)	13.3(7.8)	4.1(3.9)	3.9(5.4)	16.4(6.6)	18.0(7.0)
Peak ER ang	1.8(2.1)	4.0(5.4)	5.4(6.9)	5.2(7.4)	1.6(2.7)	4.2(3.0)	3.3(4.7)	6.0(7.2)
Peak DF/EX mom	1.9(0.8)	1.8(1.3)	4.6(1.6)	4.2(1.9)	-0.7(0.9)	-0.3(0.9)	-0.1(1.2)	0.6(1.5)
Peak PF/FL mom	0.2(0.7)	0.5(1.2)	-0.7(0.6)	-0.4(0.9)	4.6(2.4)	3.6(2.1)	2.1(2.2)	1.5(1.4)
Peak INV/AD mom	0.7(1.3)	0.7(0.9)	0.1(1.1)	0.2(0.8)	-0.7(0.6)	0.02(1.0)	-1.1(0.6)	-0.5(1.1)
Peak EV/AB mom	0.7(1.4)	0.4(0.9)	1.7(1.8)	1.3(1.2)	3.3(1.6)	1.9(1.4)	4.9(1.8)	3.5(1.9)
Peak IR mom	0.2(0.2)	0.2(0.2)	0.2(0.2)	0.2(0.2)	0.6(2.0)	1.8(4.6)	2.6(8.0)	6.3(13.9)
Peak ER mom	0.2(0.1)	0.2(0.1)	0.2(0.2)	0.2(0.2)	1.3(2.0)	2.3(4.4)	3.1(8.5)	6.1(12.7)
Peak Abs P_S	1.2(0.9)	1.9(1.4)	9.1(6.6)	5.9(3.1)	7.6(4.9)	4.6(3.3)	7.6(8.9)	5.2(4.6)
Peak Gen P_S	2.1(1.7)	2.5(2.0)	17.4(10.4)	17.0(10.8)	3.7(2.4)	2.9(1.9)	4.6(3.6)	5.5(4.2)
Peak Abs P_F	0.7(0.8)	0.6(0.6)	1.9(1.8)	1.3(1.5)	1.2(1.1)	1.6(1.5)	4.5(2.6)	4.1(2.7)
Peak Gen P_F	0.6(0.8)	0.5(0.5)	2.1(3.4)	1.4(1.4)	3.9(2.7)	2.4(2.0)	8.3(5.5)	7.1(6.6)
Peak Abs P_T	0.05(0.05)	0.08(0.08)	0.7(0.8)	0.6(0.7)	0.5(1.2)	1.9(5.8)	3.3(6.6)	6.2(10.0)
Peak Gen P_T	0.1(0.1)	0.1(0.1)	0.4(0.4)	0.3(0.3)	0.5(1.2)	1.9(6.2)	3.9(6.5)	6.3(9.3)
ML COP	4.1(5.1)	4.1(5.0)	--	--	5.8(1.8)	6.6(3.0)	--	--

Joint angle measures are in units of degrees, moments are in Nm/kg, joint powers are W/kg, and ML COP is in cm. Negative signs indicate that a peak in given dependent measure was not observed and the value reported is a minimum value for the opposing motion.

Table B11. Mean (SD) of change scores for ankle and knee kinematic and kinetic measures during **forward** drop landings. Results are tabulated as individual group means for each drop type.

	Ankle		Knee	
Measures	**C**	**U**	**C**	**U**
Land Phase				
Peak DF/EX angle	-2.5(2.7)	-2.3(3.2)	-1.1(3.0)	-1.4(3.1)
Peak PF/FL angle	5.4 (2.4)	5.3(2.6)	-1.0(3.0)	1.3(3.6)
Peak INV/AD angle	-1.5(3.7)	-2.2(3.3)	-0.5(3.6)	-2.4(3.0)
Peak EV/AB angle	0.5(3.6)	-0.7(3.7)	-0.5(1.9)	1.9(2.5)
Peak IR angle	0.8(2.5)	0.7(3.0)	1.8(2.9)	1.7(3.7)
Peak ER angle	0.7(2.9)	-0.8(4.6)	-0.5(3.0)	-1.6(4.8)
Peak DF/EX moment	1.7(1.4)	0.8(1.8)	1.7(1.2)	0.4(1.7)
Peak PF/FL moment	-0.2(0.2)	-0.3(0.8)	-0.1(0.01)	-0.04(0.4)
Peak INV/AD moment	0.2(0.7)	-0.2(0.8)	0.05(0.6)	-0.2(0.5)
Peak EV/AB moment	-0.2(0.8)	0.3(1.2)	-0.3(1.1)	0.1(1.1)
Peak IR moment	0.03(0.1)	-0.05(0.1)	0.05(0.1)	-0.02(0.1)
Peak ER moment	0.02(0.1)	0.04(0.1)	0.004(0.1)	0.02(0.1)
Peak Abs P_S	-2.0(2.3)	-2.5(7.2)	-0.5(0.7)	-0.3(2.3)
Peak Gen P_S	7.8(8.3)	5.2(11.0)	8.0(12.0)	5.6(16.1)
Peak Abs P_F	0.4(3.0)	-1.4(2.1)	0.8(1.7)	0.1(1.2)
Peak Gen P_F	-1.5(3.4)	1.1(4.0)	-0.3(1.4)	-0.3(1.7)
Peak Abs P_T	0.1(0.4)	-0.2(0.4)	0.1(0.5)	0.1(0.4)
Peak Gen P_T	0.1(0.4)	0.1(0.4)	0.1(0.3)	-0.1(0.4)
ML COP	-0.2(0.9)	-0.1(0.5)	--	--
Stance Phase				
Peak DF/EX angle	-2.1(2.7)	-1.2(3.7)	1.3(4.7)	0.6(2.5)
Peak PF/FL angle	3.2(3.7)	4.1(5.4)	0.6(3.2)	3.2(4.0)
Peak INV/AD angle	0.2(4.9)	1.6(4.4)	-0.7(3.5)	-1.2(3.0)
Peak EV/AB angle	1.8(3.8)	0.6(4.0)	1.1(5.2)	5.3(5.1)
Peak IR angle	2.0(3.8)	1.4(4.1)	3.4(4.0)	5.2(5.9)
Peak ER angle	1.5(3.8)	1.3(1.7)	2.3(5.4)	0.2(4.3)
Peak DF/EX moment	0.9(0.6)	0.2(0.7)	1.3(0.9)	0.5(1.4)
Peak PF/FL moment	-0.4(0.3)	-0.1(0.7)	-0.3(0.4)	-0.1(0.6)
Peak INV/AD moment	0.1(0.5)	-0.1(0.5)	0.1(0.5)	0.03(0.5)
Peak EV/AB moment	0.01(0.5)	0.3(0.7)	-0.2(0.8)	0.2(0.7)
Peak IR moment	0.1(0.1)	-0.04(0.1)	0.1(0.1)	0.004(0.1)
Peak ER moment	0.01(0.1)	0.05(0.1)	0.03(0.1)	0.04(0.1)
Peak Abs P_S	0.5(0.8)	0.1(1.3)	0.6(2.5)	0.6(3.6)
Peak Gen P_S	2.0(2.0)	0.8(1.3)	9.0(5.9)	3.4(7.8)
Peak Abs P_F	0.3(0.4)	0.1(0.4)	-0.3(1.5)	0.4(0.7)
Peak Gen P_F	0.2(0.3)	0.2(0.3)	-0.01(0.8)	0.5(1.2)
Peak Abs P_T	0.1(0.1)	0.01(0.1)	0.6(0.9)	0.03(0.6)
Peak Gen P_T	0.1(0.1)	0.02(0.1)	0.1(0.3)	0.2(0.2)
ML COP	0.5 (1.4)	0.8(1.7)	--	--

Joint angle measures are in units of degrees, moments are in Nm/kg, joint powers are W/kg, and ML COP is in cm. Positive and negative change scores indicate an increase or decrease, respectively, from pre-fatigue levels.

Table B12. Mean (SD) of change scores for ankle and knee kinematic and kinetic measures during **side** drop landings. Results are tabulated as individual group means for each drop type.

	Ankle		Knee	
Measures	**C**	**U**	**C**	**U**
Land Phase				
Peak DF/EX angle	-3.9(2.8)	-3.1(3.4)	-0.2(1.9)	0.2(2.2)
Peak PF/FL angle	3.0(4.8)	3.3(4.1)	-1.5(2.7)	-1.5(2.5)
Peak INV/AD angle	0.1(4.5)	3.1(5.4)	1.5(4.1)	1.8(5.0)
Peak EV/AB angle	0.8(2.6)	-0.4(3.6)	-0.5(2.5)	-1.5(2.1)
Peak IR angle	0.8(2.8)	0.8(3.3)	0.8(3.6)	0.1(3.9)
Peak ER angle	0.3(2.8)	0.5(4.0)	-1.5(3.1)	-0.7(3.2)
Peak DF/EX moment	0.1(0.2)	0.03(0.2)	0.4(0.6)	-0.02(0.4)
Peak PF/FL moment	-1.1(1.8)	-0.3(1.3)	-0.6(1.6)	-0.1(1.1)
Peak INV/AD moment	0.1(0.3)	-0.02(0.4)	0.03(0.1)	0.1 (0.2)
Peak EV/AB moment	-0.1(1.4)	0.1(1.1)	-0.1(1.5)	0.1(1.4)
Peak IR moment	0.1(0.2)	0.1(0.2)	0.2(0.2)	0.2(0.2)
Peak ER moment	-0.1(0.2)	-0.05(0.1)	-0.1(0.1)	-0.04(0.1)
Peak Abs P_S	-3.4(8.5)	-0.8(10.3)	-2.8(9.9)	-1.0(7.8)
Peak Gen P_S	0.6(0.8)	0.1(0.9)	1.9(3.6)	-0.7(3.0)
Peak Abs P_F	0.5(1.9)	-0.2(2.2)	-0.1(0.7)	0.1(1.0)
Peak Gen P_F	0.2(6.7)	0.8(5.3)	1.5(7.7)	1.3(6.8)
Peak Abs P_T	0.3(0.4)	-0.02(0.5)	0.1(0.3)	-0.02(0.3)
Peak Gen P_T	0.1(0.4)	0.02(0.3)	-0.2(0.8)	0.3(0.8)
ML COP	-0.03(1.00)	-0.09(0.99)	--	--
Stance Phase				
Peak DF/EX angle	-3.8(3.2)	-2.8(3.4)	1.2(3.3)	0.05(4.2)
Peak PF/FL angle	4.7(1.9)	3.3(2.1)	0.1(2.9)	0.5(3.2)
Peak INV/AD angle	1.7(5.0)	1.4(5.0)	1.6(4.8)	2.2(5.9)
Peak EV/AB angle	1.8(3.0)	0.1(3.8)	0.1 (2.9)	-1.5(2.7)
Peak IR angle	0.3(1.9)	-0.4(1.8)	4.0(4.4)	2.4(4.3)
Peak ER angle	0.8(2.6)	0.7(3.0)	0.9(3.6)	-1.3(4.3)
Peak DF/EX moment	0.3(0.7)	0.001(0.4)	0.3(0.7)	-0.02(0.6)
Peak PF/FL moment	-1.0(1.5)	-0.1(0.9)	-0.4(1.3)	0.1(0.7)
Peak INV/AD moment	0.2(0.3)	0.2(0.5)	0.1(0.2)	0.2(0.4)
Peak EV/AB moment	-0.4(0.7)	-0.1(0.7)	-0.4(0.9)	0.04(0.8)
Peak IR moment	0.05(1.4)	-0.1(1.4)	1.2(8.0)	-1.9(5.4)
Peak ER moment	0.3(2.1)	-0.2(1.4)	-1.4(8.0)	-1.5(4.4)
Peak Abs P_S	-0.6 (2.4)	0.2(1.5)	-0.8(3.8)	0.3(2.3)
Peak Gen P_S	0.1(1.7)	0.3(0.9)	0.5(3.5)	-0.3(2.1)
Peak Abs P_F	0.4(1.0)	0.1(0.9)	0.4(1.6)	0.4(1.5)
Peak Gen P_F	-0.4(1.2)	0.3(1.1)	0.1(3.6)	1.2(3.2)
Peak Abs P_T	0.3(0.8)	-0.1(1.1)	0.2(4.4)	1.1(5.7)
Peak Gen P_T	0.3(0.7)	0.1(0.9)	0.1(3.5)	2.8(5.0)
ML COP	0.61(1.04)	-0.30(1.69)	--	--

Joint angle measures are in units of degrees, moments are in Nm/kg, joint powers are W/kg, and ML COP is in cm. Positive and negative change scores indicate an increase or decrease, respectively, from pre-fatigue levels.

Table B13. Statistical results for the effects of group (G), fatigue (F), covariate (C) and the group x covariate interaction (G x C) on peak EMG magnitudes and CCRs in the 100 ms preceding (Pre-Land) and following GC (Land) for forward and side drop landing trials. Significant effects (*p* < 0.05) are in bold while effects that approached significance (0.05 < *p* < 0.1) are highlighted.

		Forward Drop Landings					Side Drop Landings				
		Pre	**Main**	**ANCOVA**			**Pre**	**Main**	**ANCOVA**		
Measures	**Stats**	**G**	**F**	**G**	**C**	**G x C**	**G**	**F**	**G**	**C**	**G x C**
Pre-Land											
Peak TA (% MaxTA)	*t,F*	**-2.64**	**-3.91**	0.97	**9.38**	3.74	**-1.98**	**-3.32**	0.27	3.26	1.18
	p	**0.0092**	**0.0006**	0.33	**0.0053**	0.065	**0.049**	**0.0028**	0.61	0.085	0.29
Peak PL (% MaxPL)	*t,F*	0.56	-0.55	0.31	0.30	0.49	0.82	-0.71	1.55	0.24	3.13
	p	0.57	0.58	0.58	0.59	0.49	0.41	0.48	0.23	0.63	0.090
Peak GS (% MaxGS)	*t,F*	-1.04	**-3.45**	0.20	**22.99**	0.17	**-2.02**	**-2.48**	0.44	1.23	1.60
	p	0.30	**0.002**	0.66	**<0.0001**	0.68	**0.045**	**0.020**	0.51	0.28	0.22
Peak CCR TA/PL	*t,F*	-1.68	**-3.16**	4.23	**19.13**	**9.43**	-1.84	-1.46	4.10	2.33	2.55
	p	0.096	**0.0039**	0.051	**0.0002**	**0.0054**	0.069	0.16	0.055	0.14	0.12
Peak CCR TA/GS	*t,F*	-0.16	-1.71	0.63	**33.63**	1.16	1.25	0.81	2.32	0.64	0.01
	p	0.88	0.099	0.44	**<0.0001**	0.29	0.21	0.42	0.14	0.43	0.91
Peak CCR PL/GS	*t,F*	1.68	0.92	1.48	**8.29**	0.91	**3.92**	1.79	0.74	0.16	0.84
	p	0.099	0.37	0.24	**0.0085**	0.35	**0.0002**	0.085	0.40	0.69	0.37
Land											
Peak TA (% MaxTA)	*t,F*	-1.09	**-2.97**	3.56	**18.48**	**4.74**	0.35	**-2.06**	3.03	0.38	0.22
	p	0.28	**0.0061**	0.071	**0.0002**	**0.040**	0.73	**0.049**	0.095	0.55	0.64
Peak PL (% MaxPL)	*t,F*	0.29	-1.18	0.74	0.23	0.08	-0.68	-1.21	0.39	**11.18**	0.90
	p	0.78	0.25	0.40	0.64	0.78	0.50	0.24	0.54	**0.0027**	0.35
Peak GS (% MaxGS)	*t,F*	-3.58	1.04	0.22	0.34	**4.76**	**-3.80**	0.35	1.80	**4.48**	0.07
	p	0.0005	0.31	0.64	0.57	**0.040**	**0.0002**	0.73	0.19	**0.045**	0.79
Peak CCR TA/PL	*t,F*	-1.44	-0.16	0.49	1.74	0.40	1.27	-1.70	**6.30**	**26.80**	0.10
	p	0.15	0.88	0.49	0.20	0.54	0.21	0.10	**0.020**	**<0.0001**	0.75
Peak CCR TA/GS	*t,F*	**2.48**	**-2.12**	0.36	**23.42**	1.27	**4.34**	-2.04	0.12	**5.50**	3.98
	p	**0.014**	**0.043**	0.56	**<0.0001**	0.27	**<0.0001**	0.051	0.74	**0.028**	0.058
Peak CCR PL/GS	*t,F*	**3.78**	-1.22	1.30	3.32	0.37	**3.26**	-0.17	0.23	2.74	9.69
	p	**0.0002**	0.23	0.27	0.081	0.55	**0.0015**	0.87	0.64	0.11	0.0051

Table B14. Pre-fatigue mean(SD) of peak EMG magnitudes and CCRs in the 100 ms preceding (Pre-Land) and following GC (Land) for forward and side drop landings.

		Forward Drop Landings		Side Drop Landings	
Time	**Measures**	**C**	**U**	**C**	**U**
Pre-Land	Peak TA (% MaxTA)	27.5(16.2)	20.9(12.4)	25.6(15.5)	20.8(11.2)
	Peak PL (% MaxPL)	47.1(24.2)	49.4(21.1)	49.9(26.4)	53.3(19.4)
	Peak GS (% MaxGS)	59.8(26.1)	55.0(26.6)	57.0(23.6)	49.1(21.0)
	Peak CCR TA/PL	0.7(0.5)	0.6(0.5)	0.6(0.4)	0.5(0.4)
	Peak CCR TA/GS	0.5(0.4)	0.5(0.6)	0.5(0.3)	0.6(0.6)
	Peak CCR PL/GS	0.9(0.6)	1.1(0.9)	0.9(0.4)	1.3(0.7)
Land	Peak TA (% MaxTA)	36.5(21.0)	32.9(16.4)	33.8(22.3)	35.0(15.4)
	Peak PL (% MaxPL)	51.9(30.8)	50.6(20.3)	53.4(28.5)	50.2(24.5)
	Peak GS (% MaxGS)	37.5(15.8)	28.2(14.1)	39.0(17.4)	28.7(13.4)
	Peak CCR TA/PL	0.9(0.6)	0.7(0.5)	0.8(0.7)	0.9(0.7)
	Peak CCR TA/GS	1.1(0.7)	1.4(0.8)	0.9(0.6)	1.6(1.1)
	Peak CCR PL/GS	1.5(0.8)	2.1(1.0)	1.5(0.6)	2.0(1.3)

Table B15. Mean(SD) of change scores for measures of peak EMG and CCRs in the 100 ms preceding (Pre-Land) and following GC (Land) for forward and side drop landings. Results are tabulated as individual group means for each drop type.

		Forward Drop Landings		Side Drop Landings	
Time	**Measures**	**C**	**U**	**C**	**U**
Pre-Land	Peak TA (% MaxTA)	-4.7(7.5)	-5.4(6.4)	-3.6(6.8)	-4.0(5.1)
	Peak PL (% Max PL)	-2.5(11.8)	0.1(11.8)	-3.3(9.1)	0.7(8.6)
	Peak GS (% Max GS)	-8.6(11.8)	-7.3(12.0)	-7.0(12.8)	-3.2(8.0)
	Peak CCR TA/PL	-0.1(0.2)	-0.2(0.3)	0.002(0.2)	-0.1(0.1)
	Peak CCR TA/GS	-0.1(0.2)	-0.1(0.2)	0.1(0.2)	-0.03(0.1)
	Peak CCR PL/GS	-0.01(0.3)	0.1(0.3)	0.04(0.2)	0.1(0.2)
Land	Peak TA (% MaxTA)	-3.1(8.2)	-7.7(10.5)	-0.7(14.2)	-8.7(8.3)
	Peak PL (% MaxPL	-4.9(13.1)	-0.5(10.9)	-5.2(16.1)	-0.7(8.3)
	Peak GS (% MaxGS)	3.2(13.3)	1.9(12.3)	1.6(10.0)	-0.3(7.1)
	Peak CCR TA/PL	0.03(0.4)	-0.1(0.4)	0.03(0.3)	-0.3(0.3)
	Peak CCR TA/GS	-0.1(0.4)	-0.6(1.2)	-0.03(0.4)	-0.3(0.5)
	Peak CCR PL/GS	-0.2(0.5)	-0.1(0.7)	0.1(0.4)	-0.1(0.4)

Appendix C.

Chapter 4 Statistical Results

Table C1. Statistical results for the main effects of brace (B) and fatigue (F) on JPS error and effective stiffness measures. Significant effects (p < 0.05) are in bold while effects that approached significance (0.05 < p < 0.1) are highlighted.

Measures	Stats	Pre-Fatigue B	Main F
Abs Err (deg)	*t,F*	2.66	**2.45**
	p	0.071	**0.015**
True Err (deg)	*t,F*	0.96	-0.53
	p	0.38	0.60
Effective Stiffness, k (Nm/rad)	*t,F*	**31.27**	-1.95
	p	**<0.0001**	0.052

Table C2. Statistical results for the effects of brace (B), an additional factor (R), and the covariate (C) on JPS error and ankle effective stiffness change scores. Significant effects (p < 0.05) are in bold while effects that approached significance (0.05 < p < 0.1) are highlighted.

Measures	Stats	B	R	C	B x R	B x C	R x C	B x R x C
Abs Err (deg)	*F*	2.10	2.38	**18.57**	0.69	2.74	1.08	1.43
	p	0.12	0.069	**<0.0001**	0.66	0.066	0.36	0.20
True Err (deg)	*F*	0.069	**6.64**	**24.20**	0.45	1.04	**4.48**	1.58
	p	0.93	**0.0002**	**<0.0001**	0.84	0.35	**0.0042**	0.15
Effective Stiffness, k (Nm/rad)	*F*	1.15	5.43	**23.04**	**6.24**	**3.65**	3.20	**16.02**
	p	0.32	0.058	**<0.0001**	**0.0022**	**0.027**	0.078	**<0.0001**

The additional factor R represents the effect of reference angle for JPS error measures and the effect of gender on ankle stiffness measures.

Table C3. Mean (SD) of pre-fatigue levels and post-fatigue change scores for measures of JPS error and ankle effective stiffness. Means are presented for each brace type.

	Pre-Fatigue			Post-Fatigue Change Scores		
Measures	**NB**	**AW**	**AC**	**NB**	**AW**	**AC**
Abs Err (deg)	3.7(2.5)	3.9(2.8)	3.2(2.2)	0.4(2.8)	0.4(3.5)	0.2(2.6)
True Err (deg)	1.5(4.2)	1.6(4.6)	1.0(3.8)	-0.2(4.5)	-0.01(4.6)	-0.1(3.5)
Effective Stiffness, k (Nm/rad)	15.3(7.6)	17.0(7.7)	17.8(6.2)	-0.1(3.0)	-1.3(2.6)	0.5(3.0)

Table C4. Mean (SD) of pre-fatigue levels and post-fatigue change scores for measures of JPS error and ankle effective stiffness. Means are presented for each brace type and each level of the reference angle (JPS) or gender (stiffness) factor.

		Pre-Fatigue			Post-Fatigue Change Scores		
Measures	**Level**	**NB**	**AW**	**AC**	**NB**	**AW**	**AC**
Abs Err (deg)	LP	3.2(2.6)	3.2(2.2)	3.3(2.1)	1.2(3.0)	1.6(3.9)	0.5(2.2)
	SP	3.5(2.8)	4.1(3.2)	3.0(2.2)	0.2(2.9)	-0.1(3.6)	0.5(2.8)
	SD	4.5(2.5)	4.2(3.3)	4.0(2.7)	0.2(3.1)	0.8(3.7)	-0.4(2.9)
	LD	3.3(2.1)	4.1(2.4)	2.7(1.7)	0.003(1.8)	-0.6(2.1)	0.2(2.3)
True Err (deg)	LP	0.1(4.1)	-0.3(3.9)	0.04(3.9)	0.3(5.0)	0.5(4.9)	-0.9(3.3)
	SP	0.8(4.5)	0.4(5.2)	0.2(3.7)	-1.0(4.5)	-0.6(5.9)	-0.9(3.9)
	SD	3.2(4.1)	3.3(4.2)	2.5(4.2)	0.2(4.8)	0.5(4.3)	0.8(3.1)
	LD	1.9(3.5)	2.8(3.9)	1.2(3.1)	-0.4(3.4)	-0.5(2.6)	0.6(3.3)
Effective Stiffness, k (Nm/rad)	M	20.7(8.8)	23.3(7.9)	22.9(5.9)	-0.3(4.3)	-2.2(3.0)	0.8(4.1)
	F	11.4(2.8)	12.2(1.7)	14.0(2.7)	0.02(1.6)	-0.6(2.0)	0.2(2.0)
Damping, c (N-s/m)	M	0.24(0.13)	0.18(0.07)	0.28(0.13)	-0.005(0.14)	0.07(0.13)	0.03(0.08)
	F	0.18(0.09)	0.16(0.06)	0.18(0.06)	-0.05(0.06)	-0.03(0.05)	-0.01(0.08)

Appendix D.

Informed Consent Document

VIRGINIA POLYTECHNIC INSTITUTE AND STATE UNIVERSITY

Informed Consent for Participation
in Human Subjects Research

Title of Project: Effects of Running Speed, Fatigue, and Bracing on Motor Control of Chronically Unstable Ankles

Investigators: Dr. Maury Nussbaum, Ms. Courtney Haynes

I. Purpose

The objective of this project is to improve our knowledge of risk factors for sprain injuries among adults with chronic ankle instability. Specifically, we will focus on three aspects of athletic performance – running speed, fatigue, and ankle bracing – that are not often considered in studies of ankle control. You will help us obtain data to address three primary concerns: 1) whether typical stationary measures of ankle control sufficiently describe ankle control in a running athlete, 2) if fatigue occurs more quickly and is potentially a more dangerous condition for unstable ankles, and 3) how ankle stiffness and position sense are affected by fatigue and bracing.

II. Procedures

This project involves 3 unique studies of which you may choose to complete one or more. Each study is described below. For each study, an introduction session (~ 1 hr) will familiarize you with lab equipment and protocols. Data collection will occur during the test sessions, which are expected to last approximately 2 hours each. Studies #1 and #2 will have a single test session whereas Study #3 will require 3 test sessions.

Study #1: You will perform a set of stationary and dynamic (moving) tasks. Stationary tasks will include multiple trials of a joint position sense (JPS) test and a single-leg drop landing from a height of about 30 cm (1 foot). Dynamic tasks include performing trials involving a run, a cut step, a jump stop, and a shuttle run at each of two running speeds. For each test, you will be fitted with reflective markers using tape applied to the skin and clothing, and these will be used to estimate postures and motions. Forces exerted during these tasks will also be recorded.

Study #2: You will be fitted with reflective markers, and electrodes will be attached to the skin over muscles of the lower leg; the latter, called electromyography (EMG) is used to measure the small electrical signals generated by muscles when they contract. In this study, a series of single-leg drop landings will be performed as in Study #1. Following this, a simple task will be peformed to fatigue the ankle muscles over approximately 20 minutes. During this fatiguing task, maximum ankle strength measures will be collected periodically along with ratings of how hard you feel you are working. You will be fatigued to half your initial ankle strength and complete a second set of drop landings. Forces, motion data, and muscle signals will be recorded for each trial.

Study #3: You will complete multiple trials of a JPS test and an ankle stiffness test under 3 bracing conditions (no brace, ankle wrap brace, semi-rigid brace) before and after fatigue development. EMG electrodes will again be placed over muscles of the lower leg. JPS and stiffness will first be measured before the fatigue protocol (pre-fatigue). Then, a fatigue protocol as described in Study #2 will be used to reduce strength, and either the JPS or stiffness procedures will be repeated. You will return briefly to the fatigue protocol to again reduce your strength, and the remaining JPS or stiffness protocol will be completed. One test session is required for each of the 3 bracing conditions.

III. Risks and Benefits

The risks of this study are considered minimal. You may experience minor muscle weakness and muscle pain from fatigue protocols.   Because you are asked to perform some dynamic activity, there is a small risk of ankle sprain or strain due to a misstep. Note that the required activities are similar to those that might be performed during normal running exercise. Additionally, these protocols have been used in previous research without report of injury. Should an injury occur, you will be responsible for the associated costs of treatment. Treatment costs are not the responsibility of the research team or Virginia Tech.

Although there are no direct benefits promised to you, your participation will help improve knowledge of ankle sprain risk factors as well as address how ankle bracing support changes as an athlete fatigues.

IV. Extent of Anonymity and Confidentiality

Your personal information and identity will be kept confidential. A unique study ID code will be assigned to you, and all data, questionnaire responses, and experiment check sheets will be identified using only this study ID code. Your name and any personal information you provide will never be connected with your unique data set. All individual information will be collected in a file and locked when not being used. Only the investigators have access to the data. It is possible that the Institutional Review Board (IRB) may view this study's collected data for auditing purposes. The IRB is responsible for the oversight of the protection of human subjects involved in research.

V. Informed Consent

You will receive two copies of this informed consent document. One will be signed and kept on file with the research team, and the second is for your records.

VI. Compensation

You will be compensated for your participation at a rate of $10 per hour. Compensation will be limited to time spent in the experimental session (e.g., you will not be compensated for your travel to or from the study). Your total payment will vary, depending on the length of time for your testing, and portions of an hour will be compensated by rounding up to the nearest half hour.

VII. Freedom to Withdraw

You are free to withdraw from this study at any time without giving a reason, and there will be no penalty for doing so. If you choose to withdraw, you will be compensated for the testing time you've already completed. Furthermore, you are free not to answer any questions or to choose not to respond to experimental situations without penalty. There may be circumstances under which the investigator may determine that the experiment should not be continued. In this case, you will be compensated for the portion of the project completed.

VIII. Approval of Research

The Department of Industrial and Systems Engineering has approved this research, as well as the Institutional Review Board (IRB) for Research Involving Human Participants at Virginia Tech.

IX. Participant's Responsibilities

I voluntarily agree to participate in this study. I have the following responsibilities:

1. To read and understand the above instructions.
2. To answer questions, surveys, etc. honestly and to the best of my ability.
3. Be aware that I am free to ask questions or end my participation at any point in time.

X. Participant's Permission

I have read and understand the Informed Consent and conditions of this research project. I have had all my questions answered. I hereby acknowledge the above and give my voluntary consent for participation in this project.

If I participate, I reserve the right to withdraw at any time without penalty. I agree to fulfill the responsibilities, noted above, to the best of my ability, or to inform the investigators if I am unable to do so.

____

Participant's Signature Date

____

Experimenter's Signature Date

Should I have any questions about this research or its conduct, and research subjects' rights, and whom to contact in the event of a research-related injury to the subject, I may contact:

Courtney Haynes 540-231-2380/ chaynes7@vt.edu
Investigator Telephone/e-mail

Dr. Maury A. Nussbaum 540-231-6053/ nussbaum@vt.ed
Faculty Advisor Telephone/e-mail
Professor
Department of Industrial and Systems Engineering
521 Whittemore Hall (0118)
Blacksburg, VA 24061

David M. Moore 540-231-4991/moored@vt.edu
Telephone/e-mail
Chair, Virginia Tech Institutional Review Board for the Protection of Human Subjects
Office of Research Compliance
2000 Kraft Drive, Suite 2000 (0497)
Blacksburg, VA 24061

Appendix E.

Medical History Questionnaire

DEMOGRAPHIC SURVEY/MEDICAL HISTORY

Demographics

Participant ID: ________________ **Date:** _________________ **Age:** ______________

Height: __________________ **Weight:** ________________ **Gender:** M / F

Medical History (Please check one.)

For the RIGHT ANKLE:			
1) Have you ever experienced an ankle sprain?	[ ] NO	[ ] YES	
IF YES:			
a) How many ankle sprains have you had?	[ ] 1	[ ] 1-3	[ ] >3
b) Did you seek medical attention?	[ ] NO	[ ] YES	
If yes, how many times was medical attention sought?	[ ] Once	[ ] 2-3 times	[ ] For each sprain
c) Did the injury cause you to restrict normal activity?	[ ] NO	[ ] YES	
d) About how long ago was the last sprain?	[ ] ≤ 1 year	[ ] 1-2 years	[ ] > 2 years
e) Did you complete physical rehabilitation to treat your last ankle sprain?	[ ] NO	[ ] YES	
f) Do you currently use athletic tape or an ankle brace?	[ ] NO	[ ] YES	
g) Are you currently participating in a physical rehabilitation program?	[ ] NO	[ ] YES	
For the LEFT ANKLE:			
2) Have you ever experienced an ankle sprain?	[ ] NO	[ ] YES	
IF YES:			
a) How many ankle sprains have you had?	[ ] 1	[ ] 1-3	[ ] >3
b) Did you seek medical attention?	[ ] NO	[ ] YES	
If yes, how many times was medical attention sought?	[ ] Once	[ ] 2-3 times	[ ] For each sprain
c) Did the injury cause you to restrict normal activity?	[ ] NO	[ ] YES	
d) About how long ago was the last sprain?	[ ] ≤ 1 year	[ ] 1-2 years	[ ] > 2 years
e) Did you complete physical rehabilitation to treat your last ankle sprain?	[ ] NO	[ ] YES	
f) Do you currently use athletic tape or an ankle brace?	[ ] NO	[ ] YES	

g) Are you currently participating in a physical rehabilitation program?	[ ] NO	[ ] YES	
General Information/Medical History			
3) How often do you exercise each week?	[ ] 2-3 times	[ ] 3-5 times	[ ] > 5 times
4) How would you describe exercise routine?	[ ] Mostly cardio	[ ] Mostly weights	[ ] 50/50

5) Are you currently experiencing pain or have chronic pain in:			
a) Either ankle?	[ ] NO	[ ] YES	
b) Either knee?	[ ] NO	[ ] YES	
c) Either hip?	[ ] NO	[ ] YES	
d) Back?	[ ] NO	[ ] YES	
e) Neck?	[ ] NO	[ ] YES	
6) Have you had surgery on any joints of the lower limb (hip, knee, ankle)?	[ ] NO	[ ] YES	
7) Have you ever been diagnosed with neuropathy (loss of sensation in limbs)?	[ ] NO	[ ] YES	
8) Are you currently experiencing muscle weakness in the lower limbs?	[ ] NO	[ ] YES	
9) Have you been diagnosed with a vestibular or balance disorder?	[ ] NO	[ ] YES	
10) Do you have any current injuries?	[ ] NO	[ ] YES	
11) Are you taking any medications that interfere with balance?	[ ] NO	[ ] YES	

For Researcher's Use:

CAIT Score: ________________

Group Assignment: Stable Ankle / Unstable Ankle

Appendix F.

Cumberland Ankle Instability Tool*

Please tick the ONE statement in EACH question that BEST describes your ankles.

	LEFT	RIGHT	Score
1. I have pain in my ankle			
Never	☐	☐	5
During sport	☐	☐	4
Running on uneven surfaces	☐	☐	3
Running on level surfaces	☐	☐	2
Walking on uneven surfaces	☐	☐	1
Walking on level surfaces	☐	☐	0
2. My ankle feels UNSTABLE			
Never	☐	☐	4
Sometimes during sport (not every time)	☐	☐	3
Frequently during sport (every time)	☐	☐	2
Sometimes during daily activity	☐	☐	1
Frequently during daily activity	☐	☐	0
3. When I make SHARP turns, my ankle feels UNSTABLE			
Never	☐	☐	3
Sometimes when running	☐	☐	2
Often when running	☐	☐	1
When walking	☐	☐	0
4. When going down the stairs, my ankle feels UNSTABLE			
Never	☐	☐	3
If I go fast	☐	☐	2
Occasionally	☐	☐	1
Always	☐	☐	0
5. My ankle feels UNSTABLE when standing on ONE leg			
Never	☐	☐	2
On the ball of my foot	☐	☐	1
With my foot flat	☐	☐	0
6. My ankle feels UNSTABLE when			
Never	☐	☐	3
I hop from side to side	☐	☐	2
I hop on the spot	☐	☐	1
When I jump	☐	☐	0
7. My ankle feels UNSTABLE when			
Never	☐	☐	4
I run on uneven surfaces	☐	☐	3
I jog on uneven surfaces	☐	☐	2
I walk on uneven surfaces	☐	☐	1
I walk on a flat surface	☐	☐	0
8. TYPICALLY, when I start to roll over (or "twist") on my ankle, I can stop it			
Immediately	☐	☐	3
Often	☐	☐	2
Sometimes	☐	☐	1
Never	☐	☐	0
I have never rolled over on my ankle	☐	☐	3
9. After a TYPICAL incident of my ankle rolling over, my ankle returns to "normal"			
Almost immediately	☐	☐	3
Less than one day	☐	☐	2
1–2 days	☐	☐	1
More than 2 days	☐	☐	0
I have never rolled over on my ankle	☐	☐	3

NOTE. The scoring scale is on the right. The scoring system is not visible on the subject's version.

*__Source__: Hiller CE, Refshauge KM, Bundy AC, Herbert RD, Kilbreath SL. (2006) The Cumberland Ankle Instability Tool: A report of validity and reliability testing. Arch Phys Med Rehabil 87(9): 1235-1241. Used under fair use, 2013.

Appendix G.

Borg CR-10 Scale*

rating	description
0	NOTHING AT ALL
0.5	VERY, VERY LIGHT
1	VERY LIGHT
2	FAIRLY LIGHT
3	MODERATE
4	SOMEWHAT HARD
5	HARD
6	
7	VERY HARD
8	
9	
10	VERY VERY HARD (MAXIMAL)

***Source**: Borg, G. (1998). *Borg's Perceived Exertion and Pain Scales*. Champaign, IL: Human Kinetics. Used under fair use, 2013.

Appendix H.

Annotated List of Figures

Chapter 2:

Figure 2.1. – This figure illustrates experimental set up for testing I/E and P/D joint position sense (the former movement configuration is illustrated).

Figure 2.2. – This figure provides still shots of a participant performance the forward drop landing maneuver.

Figure 2.3. – This figure presents a bar graph showing between- and within-group differences in peak knee flexion angle during a jump stop performed at two different running speeds.

Figure 2.4. – This figure presents a bar graph of selected peak joint angles for each group during the landing phase of the cut step maneuver performed at two running speeds.

Figure 2.5. – This figure presents a bar graph of selected peak joint moments for each group during the landing phase of the cut step maneuver performed at two running speeds.

Chapter 3:

Figure 3.1. – This figure is a flow chart describing experimental procedures for Chapter 3.

Figure 3.2. – This figure provides an image of a participant fitted with the reflective markers and the EMG electrodes needed for Chapter 3 protocols.

Figure 3.3. – This figure is an image of the visual displays provided to the participants during completion of the fatigue protocols.

Figure 3.4. – This figure presents a graph of strength loss over time during Fatigue Protocol #1.

Figure 3.5. – This figure is a bar graph showing pre-fatigue values of select kinematic and kinetic measures during the stance phase of forward drop landings.

Figure 3.6. – This figure is a bar graph showing the fatigue-induced changes in normalized joint moments during the stance phase of forward drop landings.

Figure 3.7. – This figure is a bar graph showing pre-fatigue values of select ankle kinematic and kinetic measures during stance phase of side drop landings.

Figure 3.8. – This figure is a bar graph showing pre-fatigue values of select knee kinematic and kinetic measures during stance phase of side drop landings.

Chapter 4: