The Effect of Ankle Bracing on Lower Extremity Coordination, Coordination Variability, and Neuromuscular Activity in Individuals with and without Chronic Ankle Instability

by

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Abstract

Ankle sprain injuries and ankle bracing act as constraints that can influence natural coordination pattern dynamics. Recently, researchers studying the progression of chronic ankle instability (CAI) have used measures of movement variability to better understand the impact of injury and bracing constraints on movement dynamics; however research in this area is limited. The purpose of this project was to explore the influence of ankle joint bracing on lower extremity coordination, coordination variability and muscular activity in individuals with and without CAI. Results indicate that, across phases of walking and single-leg hopping cycles, lower extremity coordination, coordination variability, and neuromuscular activity did not differ between individuals with and without CAI. Moreover, ankle bracing induced alterations in lower extremity coordination, coordination variability, and neuromuscular activity during walking and hopping tasks. Results from this study suggest that ankle bracing should be implemented with caution, particularly in performance and rehabilitation settings.

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CHAPTER I

INTRODUCTION

It is well known that humans are biologically dynamic and adaptive beings. Evidence of this characteristic can be found in human movement across a lifespan (Clark, 2005; Thelen, 2005). In order for movement adaptations to be beneficial for functional motor behavior, it is necessary that adequate control and coordination of movement be maintained to accomplish specific movement goals, despite perturbation. The emergence of coordinated movement patterns is related to the notion that movement systems are dynamic; the apparent movement pattern depends on the previous state(s) of the movement system, and will influence the emergence of future states (Kelso, 1995; Thelen, 2005). Beginning in the early 1980's, research in motor behavior has supported a dynamic systems theory of control and organization of movement (Kelso, 1995; Kugler, Kelso, & Turvey, 1980). A major tenant of dynamic systems theory is that variability surrounding human movement is an important functional characteristic rather than a detrimental source of error affecting movement outcome (Newell & Corcos, 1993; Slifkin & Newell, 1999). Furthermore, variability within the motor system has been shown to characterize healthy physiological processes (Goldberger et al., 1988; Lipsitz & Goldberger, 1992), facilitate motor learning through spatial-temporal exploration of anatomical degrees of freedom (Wu et al., 2014), and be a useful indicator of movement adaptations resulting from musculoskeletal pain or injury (Hamill, Palmer, & Van Emmerik, 2012; Hamill et al., 1999; Heiderscheit, 2000; Heiderscheit, Hamill, & Van Emmerik, 2002; Herb et al., 2014; Kiefer et al., 2013; Seay, Van Emmerik, & Hamill, 2011). More recent models of motor control (i.e. uncontrolled manifold hypothesis and optimal control theory) have also regarded movement variability as a valuable, rather than detrimental, component of the motor system, allowing for the exploitation of redundant degrees of freedom while controlling task-relevant parameters (Latash, Scholz, & Schöner, 2002; Todorov & Jordan, 2002). Altogether, contemporary interpretations of variability have allowed for a more thorough analysis of both healthy and dysfunctional movement through the use of novel techniques that help describe changes in movement behavior.

From a behavioral standpoint, variability is a critical characteristic of stability and adaptability of movement, indicating transitions between coordinated movement patterns (Clark & Phillips, 1993; Haken, Kelso, & Bunz, 1985; Kelso, 1984), or characterizing rigid behavior associated with movement disorders (Newell, Van Emmerik, et al., 1993; Van Emmerik et al., 1999; Wagennar & Van Emmerik, 1996). Newell (1986), expanding on the dynamic systems construct of Kugler et al., (1980; 1982) advocated that the organization and fluctuation of coordinated movement is dependent on the interaction of constraints (i.e. factors that influence movement outcomes). Organismic, environmental, and task constraints, function to limit the number of possible behavioral actions rather than dictate the selection of action. From this perspective, the understanding of constraints is pivotal for describing movement related adaptations that occur as a result of injury.

The neuromusculoskeletal system acts as a critical organismic constraint on coordinated movement consisting of active (e.g. muscles), passive (e.g. tendons, ligaments, fascia and bones) and neurological (e.g. afferent/efferent neurons and proprioceptive structures) components. Biomechanical models of human locomotion (e.g. inverted pendulum and spring-mass models) interpret active and passive components as sources and dampers of energy surrounding a limb segment. These components, in conjunction with gravitational forces and environmental energy sources, affect parameters such as segment displacement and velocity, movement frequency as well as movement efficiency (Blickhan, 1989; Geyer, Seyfarth, & Blickhan, 2006). Furthermore, neurological components facilitate the temporal regulation of active components and generate sensorimotor feedback on the state of passive and active components, allowing for time-dependent tuning of dynamic (i.e. musculoskeletal and environmental) resources and yielding the emergence of patterned movement (Holt, Wagenaar, & Saltzman, 2010; Taga, 1995). Impairments to these systems represent additional organismic constraints that may interfere with previously established movement repertoires and may lead to negative long-term compensatory adaptations (Ageberg, 2002; Anandacoomarasamy & Barnsley, 2005; Baumeister, 2012; Comerford & Mottram, 2001).

Musculoskeletal injuries present a challenge to movement systems in that damage to tissues can significantly affect functioning of active and passive components of the neuromuscular system. Damaged musculoskeletal structures have been shown to alter the functional muscle length and compliance (Howell, Chleboun, & Conatser, 1993; Proske & Morgan, 2001), cause muscle weakness (Yeung et al., 1994; Zhang & Jordan, 2010) and effect arthrogenic muscle responses (i.e. inhibition or facilitation of voluntary muscle action)(McVey et al., 2005). In addition, neuromusculoskeletal damage can produce sensorimotor impairments which may lead to diminished functional movement capacity (Ageberg, 2002). This has been demonstrated in individuals with both knee and ankle ligamentous injuries, who exhibit altered coordination of limb segments and surrounding muscles (Beard et al., 1996; Delahunt, Monaghan, & Caulfield, 2006a; Doherty et al., 2014; Martin et al., 2013; St-Onge et al., 2004).

Over the past decade, research in neuroscience has looked at the effects of musculoskeletal injury on neural function (Baumeister, 2012; Pelletier, Higgins, & Bourbonnais, 2015). It has been hypothesized that acute neuroplastic changes occurring after musculoskeletal injury act as a response aimed to protect the affected area and re-organize functionality surrounding the affected segments. However, the chronic persistence of neuroplastic adaptations is thought to represent the inability of the neuromotor system to regain normal functioning and processing (Kapreli et al., 2009; Pelletier, Higgins, & Bourbonnais, 2015). Furthermore, chronic maladaptation can be reinforced with inadequate rehabilitative efforts and remain embedded even in the presence of peripheral healing of the injured area (Ageberg & Fridén, 2008). In light of such evidence, categorization of the development of chronic movement dysfunction has begun to shift from a structural-pathology paradigm to include neurophysiological pathology.

From a structural perspective, movement variability has been applied to models predicting overuse injury. Anecdotal evidence suggests that variation in the nature in which loading occurs may result in the distribution of stress across different tissues and increase the time between repetitive loading events. Specifically, variability is thought to yield alterations in the frequency, direction, and magnitude of stress on a given tissue, allowing for tissue remodeling to occur at a greater rate and potentially mitigating the risk of injury due to repeated stress on a single area (James, Dufek, & Bates, 1996; James, Dufek, & Bates, 2000). Globally, variability is often exhibited within common training practices. In an effort to prevent overtraining and reduce injury, principles of periodization and cross training indirectly promote variation of mechanical and physiological stresses over the duration of a training program.

Recently, there has been a push in sports medicine and rehabilitation fields to adopt a dynamic systems/constraints-led approach to better understand how movement variability contributes to continuums of movement function, and recognize the development of movement dysfunction associated with musculoskeletal injury (Davids et al., 2003; Holt, Wagenaar, & Saltzman, 2010; Wikstrom, Hubbard-Turner, & McKeon, 2013). Paramount to this approach is the understanding that physiological, neurological and biomechanical constraints contribute to the organization of movement, and if altered, may disrupt natural movement dynamics. Congruently, human movement and injury rehabilitation paradigms have begun to emphasize the significance of optimal levels of functional variability believed to characterize healthy dynamics of movement systems (Harbourne & Stergiou, 2009; McKeon, 2009; McKeon & Hertel, 2006; Perry, 1998; Stergiou & Decker, 2011; Stergiou, Harbourne, & Cavanaugh, 2006; Wagennar & Van Emmerik, 1996).

Using these theoretical constructs, researchers have assessed movement dynamics in individuals with chronic and acute musculoskeletal injury as well as concussion injury, with results showing reductions in inter-segmental coordination variability, spatial-temporal gait pattern variability, and postural variability, respectively, compared to healthy controls (Cavanaugh et al., 2005; Hamill, Palmer, & Van Emmerik, 2012; Hamill et al., 1999; Moraiti et al., 2007). Thus, it seems evident that measures quantifying movement dynamics can provide insight into impairments brought on by neurological and neuromusculoskeletal damage. Furthermore, such measures may be helpful in determining levels of movement dysfunction that arise from chronic maladaptation to musculoskeletal injury. Recently some effort has been made to apply these concepts to chronic movement dysfunction brought on by lateral ankle injuries (Herb et al., 2014; Hoch & McKeon, 2010).

Ankle injuries are one of the most frequently occurring musculoskeletal injuries among physically active individuals and are often a gateway to the development of chronic movement impairment (Doherty et al., 2013; Fong et al., 2007; Yeung et al., 1994). Interestingly, patients who do not exhibit long-term maladaptations to ankle injuries (i.e. copers) seem to benefit from undetermined mechanisms (e.g. patterns of structural impairment, neuromotor adaptations, or psychological responses) that allow them to manage impairments to produce functional outcomes (Wikstrom et al., 2009). However, research on the long-term effects of ankle injury has shown that a significant number of individuals (i.e. non-copers) exhibit residual symptoms of pain, instability, and functional limitations well after the initial injury (Anandacoomarasamy & Barnsley, 2005; Verhagen, De Keizer, & Van Dijk, 1995; Wikstrom et al., 2009). Chronic ankle instability (CAI) arises from the recurrence of ankle injury and is characterized by either residual mechanical or functional instability of the ankle joint as a result of the initial injury (Hertel, 2002). Mechanical instability refers to pathological laxity following injury, in which excessive ankle joint range of motion has been demonstrated under passive and active loading (Kovaleski et al., 2014a; Monaghan, Delahunt, & Caulfield, 2006). Functional instability represents a wide range of impairment including altered neuromuscular control, diminished postural stability, and reduced capacity to perform functional tasks (Brown & Mynark, 2007; Delahunt, Monaghan, & Caulfield, 2006a; Wikstrom et al., 2012).

Movement function of those with CAI is often diminished and coupled with self-reported movement dysfunction related to activities of daily living and sport (Hale & Hertel, 2005; Wikstrom et al., 2009). Much of the literature surrounding CAI has attributed movement dysfunction to impaired sensorimotor functioning, which reduces the ability of the neuromuscular system to provide proprioceptive feedback and adequately contribute to producing favorable movement outcomes (Bastien et al., 2014; Delahunt, Monaghan, & Caulfield, 2006a). However, little research has been done to investigate the variability characteristics of movement patterns in individuals with CAI. Assessing how CAI affects dynamic characteristics of movement could give an indicator of the severity in movement dysfunction as well as potentially lead to novel means of quantifying progression of movement dysfunction brought on by ankle injury. In addition, identifying the effect of current interventions on movement dynamics could provide new information regarding the efficacy of such interventions.

Joint bracing is commonly implemented within a variety of populations, both healthy and clinical, and is considered a task constraint in that the brace is an implement imposed on the organism. Studies investigating the influence of bracing on biomechanical, sensorimotor, and performance measures have demonstrated differential effects with regards to joint mechanical stability, movement kinematic and kinetic responses, joint proprioception, and functional performance (Brisson & Lamontagne, 2014; Cordova, Ingersoll, & Palmieri, 2002; DeVita & Hortobagyi, 2001; Hume & Gerrard, 1998; McNair, Stanley, & Strauss, 1996; Rishiraj et al., 2009). Due to the high prevalence of ankle injury in normal and athletic populations, the effects of ankle bracing on lower extremity biomechanics and neuromechanics has been a widely studied subject carrying injury prevention and performance implications (Gravlee & Van Durme, 2007; Papadopoulos et al., 2005; Thacker et al., 1999).

From a prophylactic standpoint, the effectiveness of ankle bracing is fairly well established, with several review articles affirming the efficacy of bracing strategies in reducing ankle sprain incidence following implementation (Callaghan, 1997; Hume & Gerrard, 1998; Thacker et al., 1999). Studies examining ankle bracing effects on lower extremity kinematics and neuromuscular activation have demonstrated restricted ankle inversion/eversion range of motion as well as altered timing of discrete kinematic and neuromuscular measures. Furthermore, ankle bracing has been shown to effect sensorimotor functioning by way of altering ankle joint proprioception (Hartsell, 2000; Heit, Lephart, & Rozzi, 1996), however conflicting results warrant further investigation on this topic (Raymond et al., 2012). Thus, the mechanism for the apparent success of ankle bracing is the source of some debate, with researchers attributing bracing efficacy to both enhanced mechanical stability and improvement in neuromuscular functioning by means of enhanced proprioception.

As mentioned, ankle braces are commonly used as a prophylactic implement in healthy and injured populations, and act as a task constraint that may influence the organization of movement. Further research is needed to elaborate on the breadth of these influences as they relate to movement dynamics. Using measures of movement variability as a tool could provide a greater understanding of how bracing strategies impact individuals with chronic movement dysfunction due to musculoskeletal injury.

Summary

Organismic constraints are an important aspect of movement outcomes as well as the dynamic capabilities of human movement, and include components of the neuromusculoskeletal system (Newell, 1986; Wikstrom, Hubbard-Turner, & McKeon, 2013). When alterations to otherwise normal neuromusculoskeletal constraints arise, natural dynamics of movement are negatively affected (Harbourne & Stergiou, 2009; McKeon & Hertel, 2006; Stergiou & Decker, 2011). Measurement of movement variability has proved to be a fruitful assessment of the adaptability of movement patterns, particularly as it pertains to adaptations surrounding musculoskeletal injury (Hamill et al., 1999; Herb et al., 2014; Moraiti et al., 2007). Chronic ankle instability is a common form of movement dysfunction impacting normal and athletic populations alike and is often combated through joint bracing interventions. However, further research is needed to elucidate the association between movement dysfunction and movement variability in order to provide a more comprehensive approach toward recognizing the onset and progression of movement dysfunction related to ankle injuries, and assessing the systematic effects and efficacy of joint bracing interventions commonly used to combat impairments brought on by ankle injury.

Purpose

The purpose of this project was to determine the influence of ankle joint bracing on lower extremity coordination, coordination variability and muscular activity in healthy individuals and individuals with chronic ankle instability (CAI). Specifically, this project aimed to answer the following questions: 1) Does ankle joint bracing influence lower extremity segmental coordination during walking and single-leg hopping in healthy individuals, ankle sprain copers, and in individuals with CAI? 2) Does ankle joint bracing influence lower extremity segmental coordination variability during walking and single-leg hopping in healthy individuals, ankle sprain copers, and in individuals with CAI? 3) Does ankle joint bracing influence neuromuscular activity during walking and single-leg hopping in healthy individuals, ankle sprain copers, and in individuals with CAI?

Specific Aims

1. To assess if individuals with CAI (Non-Copers) exhibit altered lower extremity coordination, coordination variability, and neuromuscular activity compared to healthy individuals and ankle sprain copers during walking and single leg hopping. Specifically, this study is powered to detect differences in coordination between healthy individuals and individuals with chronic ankle instability. Additional between group analyses were deemed exploratory.
2. To assess if ankle joint bracing influences coordination, coordination variability, and neuromuscular activity in healthy individuals, ankle sprain Copers, and Non-Copers during walking and single leg hopping. Specifically, this study is powered to detect

differences in lower leg electromyography measures in individuals with chronic ankle instability with and without an ankle brace. Additional within group analyses were deemed exploratory.

Hypotheses

Null hypotheses for this project are as stated:

H_{01}: Healthy individuals will exhibit unaltered lower extremity coordination, coordination variability, and neuromuscular activity during walking and single-leg hopping when wearing an ankle brace compared to without a brace.

H_{02}: Ankle sprain copers will exhibit unaltered lower extremity coordination, coordination variability, and neuromuscular activity during walking and single-leg hopping when wearing an ankle brace compared to without a brace.

H_{03}: Ankle sprain Non-Copers will exhibit unaltered lower extremity coordination, coordination variability, and neuromuscular activity when wearing an ankle brace compared to without a brace.

H_{04}: Ankle sprain Non-Copers will exhibit unaltered lower extremity coordination, coordination variability, and neuromuscular activity compared to healthy individuals and ankle sprain copers.

Limitations

Limitations of this project are as stated:

1. Inclusion criteria for both lateral ankle sprain copers and non-copers were based on anecdotally reported information acquired from the participants.
2. Familiarization to data collection tasks and procedures were presumed prior to the start of data collection, but after the completion of a familiarization protocol.

Delimitations

1. ASO ankle braces were implemented in this project based on the prevalence of use amongst athletic and clinical populations in addition to the prevalence of use in the ankle bracing literature.

Terms

Constraints – Factors that influence the organization of movement.

Copers – Individuals who do not experience functional limitations following ankle sprain injury.

Non-Copers – Individuals with chronic ankle instability who experience residual functional limitations following ankle sprain injury.

Coordination Dynamics – Changes or fluctuations in a pattern of motion.

Phase Angle – An angle that describes the progress of segment motion through a cycle of movement.

CHAPTER II

REVIEW OF LITERATURE

Over the past several decades, the significance of variability within all levels of the human neuromotor system has become a widely investigated topic. Evidence has been provided to suggest that movement variability is an innate characteristic of a healthy neuromotor system, which characterizes the stability or instability of movement patterns. Movement pattern variability has also been shown to be a mechanism by which the organism functionally adjusts between different movement patterns. These concepts have recently been adopted in the fields of sports medicine and rehabilitation as a theoretical framework for investigating the progression of injury-related dysfunction and effectiveness of intervention protocols.

The goal of this chapter is to present an organized review of the literature surrounding the following topics related to this project: 1) Traditional and contemporary viewpoints regarding movement variability: concepts and assessment of coordination dynamics. 2) Models of movement dysfunction in sports medicine and rehabilitation and the application toward chronic ankle instability research. 3) Bracing as a common implement for injury prevention.

Movement Variability

Movement variability has become increasingly prevalent in the discussion and understanding of how humans develop, learn and control movement. Traditionally, variability has widely been viewed as noise or error within a given motor system (Fitts, 1954). However, the introduction of non-traditional techniques to assess movement variability has afforded a more functional interpretation of movement variability as an important characteristic of movement pattern dynamics (Newell & Corcos, 1993; Stergiou, 2004). The goal of this section is to present an overview of movement variability in light of traditional and contemporary viewpoints; demonstrating that seminal work in human movement output signal analysis and ascending theories of motor control have markedly shifted the perception of variability in the human movement sciences, particularly fields that seek to enhance the understanding of injury related movement pathologies and pertinent forms of intervention.

Traditional Viewpoints on Variability

Since the advent of the information-processing model of human functioning, variability in human movement has largely been associated with noise embedded in the motor signal, producing errors in the execution and outcome of a motor task. This viewpoint was in part influenced by the information theory of communication put forth by Shannon and Weaver (1949) in an attempt to understand the effect of noise as a constituent of a signal being transmitted. The information-processing model or “black box” model has been used to describe human’s response to stimuli resulting in a motor response, with variability being a source of error in the output signal. A general description of this model involves an input of sensory information from the environment to the human, followed by processing of the sensory information and formulation of an output of motor activity (Schmidt & Lee, 1988).

An early study by Fitts (1954) used the information processing theory as a model to address the information capacity of the sensorimotor system. Fitts proposed that increased variability, or noise, embedded in movement attenuates the information capacity of the motor system and can be modulated by adjusting movement and task parameters. This theoretical approach has remained popular and served as a seminal underpinning for future theoretical constructs.

Schmidt et al., (1975) proposed the theory of General Motor Programs or Schema Theory, which expands on the information processing model to explain the recollection of motor skills and recognition of sensory stimuli within the central nervous system that contribute to formulation of output signals to the motor system. When performing a movement, schema theory suggests that the initial conditions, task requirements, sensory feedback and movement outcome are stored for future use when a similar subsequent movement is prescribed. Schmidt et al., (1979) also proposed the impulse-variability theory, which suggests that variability in impulse is directly proportional to the target width for a movement. In other words, increased impulse variability causes increased variability in movement distance. The authors stated that the sources of force-time variability arise from alterations in goal-oriented movement (i.e. changes in motor program selection) between responses or from noise in the motor system beyond selection of the motor program, causing error in the movement outcome over successive trials of a given task. Altogether, these described theories attend to the interpretation of movement invariance as biological order, whereas variability presents as random noise that should be minimized or eliminated (Newell & Slifkin, 1998).

Contemporary Viewpoints on Variability

Under the construct of the aforementioned theories, variability within a motor signal is commonly characterized by the magnitude of its presence (e.g. standard deviation) within a time series. Additionally, the structure of variability within a motor signal has largely been assumed to be white Gaussian noise, characterized by normal amplitude distribution and equal representation of signal frequency within the measured range (Newell, Deutsch, et al., 1993). However, research has shown that often the structure of motor output signal over time can exhibit characteristics of complexity (i.e. fluctuations in a physiological signal that can be predicted), disparate from the structure of white Gaussian noise (Newell & Corcos, 1993; Slifkin & Newell, 1999; Vaillancourt & Newell, 2002). Such has been demonstrated in human movement data including gait kinematics (Hausdorff et al., 1996), postural control (Duarte & Zatsiorsky, 2000), isometric force output (Slifkin & Newell, 2000), and neurological as well as neuromuscular signals (Gupta, Suryanarayanan, & Reddy, 1997). Thus, inherently labeling signals as "noise" may be a misrepresentation of signal frequency characteristics that make up commonly measured variables to describe human movement.

Bernstein (1967) argued that a fundamental problem of the motor system is having to adequately control the vast number of degrees of freedom of human movement. A solution to this problem involves the coordinated organization of movement patterns in the form of movement synergies or classes of movement patterns (Bernstein, 1967; Savelsbergh, Van Der Kamp, & Rosengren, 2006; Sporns & Edelman, 1993). Bernstein (1967) described coordination as the mastering of redundant degrees of freedom into a controllable system, from which a movement goal can be accomplished in a variety of ways within a given movement pattern (Sporns & Edelman, 1993). This concept introduced the notion that variability in the parts of a whole is an innate feature of organized movement patterns. Since Bernstein's work, the role of variability in the formation and control of coordinated synergistic movement has been addressed in contemporary models of motor control.

In contrast to the standpoint that variability is an artifact to be ignored, some contemporary models of motor control suggest that movement variability may provide a better understanding of how biomechanical degrees of freedom are coordinated to produce a desired movement (Latash, Scholz, & Schöner, 2002; Todorov & Jordan, 2002). For example, Latash et al., (2002) described an uncontrolled manifold (UCM) hypothesis which describes the nature of variability with respect to various performance variables for a given task. Additionally, Kugler, Kelso and Turvey (1980) constructed a model using concepts from mechanics and thermodynamics to explain the organization of coordinative movement. From this model, a dynamic systems approach was adopted to describe spontaneous, self-organizing coordinative patterns (Haken, Kelso, & Bunz, 1985; Kelso, 1995) in response to changes in the energy surrounding the sensorimotor system. Within the theoretical framework of dynamic systems, variability in the motor system, across multiple levels, is considered a characteristic trait allowing adaptations of coordinative movement patterns in response to constraints acting on the system (Newell, 1986; Newell & Corcos, 1993).

It has been proposed that the impetus for dynamics surrounding coordinative patterns lies in constraints that act within an organism (internal), environmentally, or through a given set of task parameters (Kugler, Kelso, & Turvey, 1980; Newell, 1986). Categories of constraints, as described by Newell (1986), determine the optimal coordinative arrangement of a given movement: Environmental constraints are those that act external to the organism or reflect the ambient conditions for a task. Task constraints involve the goal of the task, specific action parameters or limitations, and implements or devices that influence the response. Organismic constraints act internally and include components at all levels of analysis (e.g. synaptic formation, joint articulation and anthropometric variables).

Understanding the interplay of psychological and physical constraints is an important concept related to coordination dynamics such that action selection is based on environmental affordances, behavioral repertoires, and intrinsic characteristics (e.g. motivation and experience), which are related reciprocally with physical constraints in the environment, task and organism (Van Ingen Schenau et al., 1995) (Figure 1).

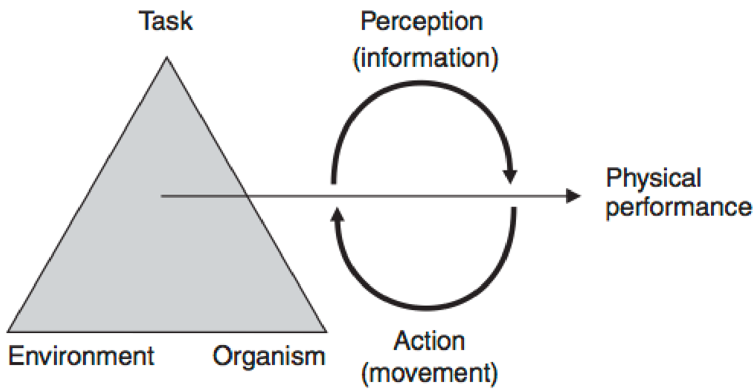

Figure 1: Newell's model of interacting constraints (Davids et al., 2003).

Thus the co-dependency of the movement pattern on psychological and physical constraints makes the understanding of organized movement rather complex and warranting further research, particularly as it relates to changes in movement strategies following injury (Holt, Wagenaar, & Saltzman, 2010).

Coordination Dynamics

Haken's theory of *synergetics* (1983), or the self-organizing structure of open physical, chemical and biological systems (systems capable of exchanging energy, matter or information with the environment) with large numbers of degrees of freedom, provided a foundation for the study of changes in coordinated movement (Haken, Kelso, & Bunz, 1985; Kelso, 1995; Kelso & Schöner, 1988; Schöner & Kelso, 1988). Within this framework it is necessary to identify order parameters, or variables that characterize movement patterns, which are influenced by control parameters, or parameters that induce change in the system. The identification of order parameters provides the means for the quantitative analysis of a small set of variables, taken from a larger set of interrelated components that encompass motor behavioral outcomes, which can describe coordinative patterns as well as the stability of the patterns (Scholz, 1990). Thus, the study of coordination in human movement should consider relevant components that contribute to the movement outcome.

Winter (2009) described synergistic control of movement as the result of convergence from inputs to alpha motor neurons to combined moments generated to carry out a movement goal. Although measurements of these movement-related variables (e.g. kinematics or neuromuscular activity) can be obtained, alone these variables do not represent the coordinated coupling of the segments in question (Scholz, 1990). Rather, a collection of kinematic variables can be incorporated in such a manner to assess the spatio-temporal coupling of coordinated segments.

Relative phase analysis reflects the coordinated relationship between body segments through the reduction of kinematic measurements into a single variable, considered to be an ideal assessment of the dynamic characteristics of a coordinative pattern (Burgess-Limerick, Abernethy, & Neal, 1993). Implementation of relative phase involves constructing a phase portrait (e.g. plotting the state of the order parameter versus the rate of change) for a given set of order parameters in state space (Lamb & Stockl, 2014; Stergiou, 2004). State space refers to a vector space (e.g. the state of a limb segment system may be described by its position vector) where the dynamic system can be defined at any point (Stergiou et al., 2004). From the phase portrait, a phase angle can be determined by calculating the angle formed between a right horizontal line from the origin of the portrait and a vector projected from the origin to a data point of the portrait (Figure 2) (Hamill, Haddad, & McDermott, 2000). The relative phase angle is then determined by calculating the difference between phase angles of two coupled systems (e.g. limb segments) at a given point in time (Stergiou, 2004).

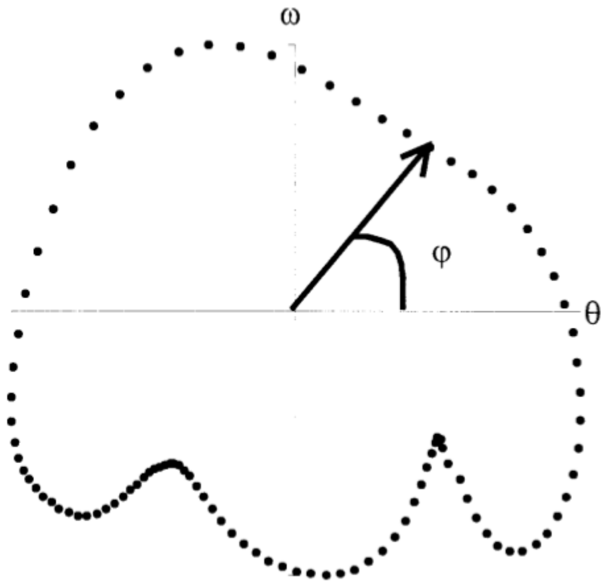

Figure 2: Determining a phase angle from a phase portrait. The phase angle (ϕ) can be determined at any instant in time by calculating the inverse tangent of the ratio of ω to θ (Hamill, Haddad, & McDermott, 2000).

In state space the analysis of pattern dynamics, with respect to an attractor state (i.e. preferred region in state space), can give indication of pattern stability (Van Emmerik & Van Wegen, 2000). When the pattern involves oscillation of two segments in a periodic manner, the attractor can be described as a limit-cycle attractor, in which stable order parameters tend to converge onto a fixed orbit rather than a single point (point attractor) in state space (Van Emmerik & Van Wegen, 2000) (Figure 3).

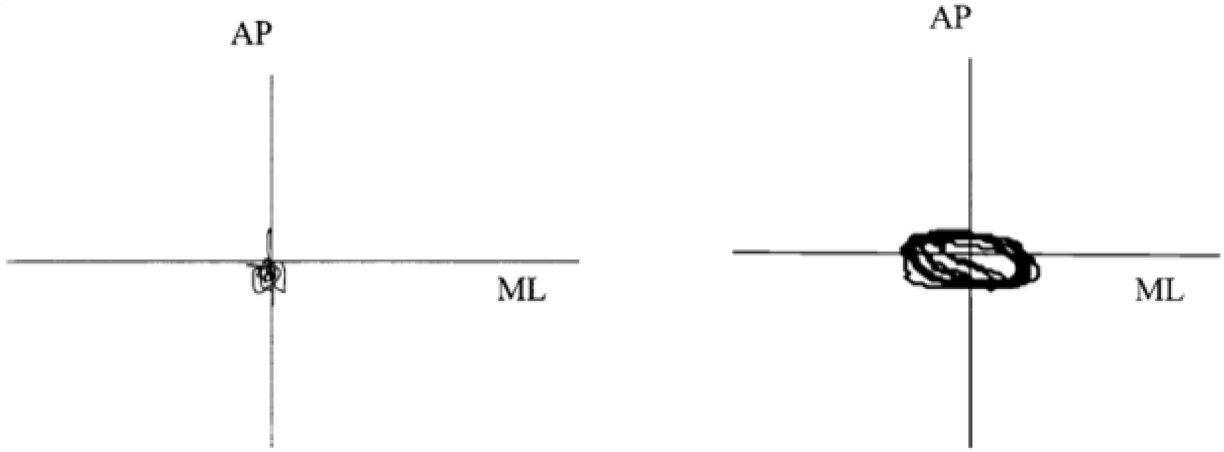

Figure 3: Attractor states demonstrated with center of pressure data. Fixed point attractors (left) converge onto a fixed point in state space. Limit-cycle attractors (right) converge onto an orbit in state space. Adapted from Van Emmerik & Van Wegen (2000).

On the contrary, Shöner and Kelso (1988) proposed that reduced stability of a behavioral pattern is accompanied by elongated relaxation time (restored pattern stability) and increased fluctuation of the order parameter from the attractor sate. Thus, fluctuations or variability of the relative phase has been shown to be a measure of coordination pattern stability as well as an indicator of transition between coordinated patterns.

Analytical tools used for the analysis of coordination dynamics extends the capabilities of researchers beyond discrete measurements of individual movement parameters, and have been fertile in fields seeking to explore mechanisms and characteristics of movement pathologies. Specifically, the study of coordination dynamics has provided insight in areas relating to neurological disorders (Scholz, 1990; Van Emmerik et al., 1999; Wagennar & Van Emmerik, 1996), musculoskeletal conditions (Hamill et al., 1999; Heiderscheit, 2000; Heiderscheit, Hamill, & Van Emmerik, 1999; Seay, Van Emmerik, & Hamill, 2011; Wang et al., 2009) and musculoskeletal injuries (Hamill, Palmer, & Van Emmerik, 2012; Kiefer et al., 2013), where neurological or musculoskeletal impairments can have deleterious effects on fundamental forms of locomotion such as walking and running.

Moving forward, contemporary theories of motor control can provide a framework for studies that aim to understand the role of variability contributing to dysfunctional movement resulting from neuromusculoskeletal impairments. In using a dynamic systems/constraints led approach to study the effects of lateral ankle sprains, one must consider the influence of intra-individual, environmental, and task factors, to provide a more holistic understanding of the variables that give rise to chronic ankle instability, and interventions, such as joint bracing, implemented to mitigate detriments attributed to movement dysfunction.

Movement Dysfunction

One fundamental aspect of the proposed project is to incorporate concepts from dynamic systems theory to describe the characteristics and progression of pathological movement behavior that can arise from lateral ankle sprain injuries. A starting point for this discussion is to assess how current means for classifying movement dysfunction relate to the proposed theoretical approach, such that a more complete understanding of dysfunctional progression can be explained from both clinical and research standpoints. Thus, the first part of this section will discuss how current models of disablement are used in sports medicine and rehabilitation fields to provide a comprehensive understanding of the factors that contribute to disability.

The Disablement Process (Verbrugge & Jette, 1994) describes the effect of acute or chronic conditions on the functioning of body systems and the individual's ability to act in an expected fashion within personal or societal contexts. Disablement models can be implemented to describe movement dysfunction by using a systematic approach to chronicle impairments, functional limitations, and disability (Jette, 1994). From a clinical perspective, disablement models provide a foundational framework to guide research and promote the implementation of evidence-based treatment regiments (Jette, 1994; Snyder et al., 2008). One category of disablement models are biopsychosocial models, which view disability as a consequence of biological, personal and social factors (Jette, 2006).

The Nagi Model of Disablement (1965) is one such model that was derived from sociological theory and was particularly innovative in that it recognized the influence of environmental and societal factors on disability, rather than defining disability based solely on physical limitations. A second model of disablement, the International Classification of Functioning, Disability and Health (ICF), was put forth by the World Health Organization (2001) as a revision of their previous model, the International Classification of Impairments, Disabilities, and Handicaps (ICIDH) (1980). Both models consist of similarly constructed domains that describe facets of the disablement process (Figure 4). The ICF is similar to the Nagi model in that dysfunction is viewed as the result of dynamic interaction between an individual's state of health and various contextual factors (Jette, 2006).

Nagi	**ICF**
Active Pathology—interruption or interference with normal processes, and effort of the organism to regain normal state	*Health Conditions*—diseases, disorders, and injuries
Impairment—anatomical, physiological, mental or emotional abnormalities	*Body Function*—physiological functions of body systems *Body Structures*—anatomical parts of the body *Impairments*—problems in body functions or structure
Functional Limitation—limitation in performance at the level of the whole organism or person	*Activity*—the execution of a task or action by an individual *Activity Limitation*—difficulties an individual may have in executing activities
Disability—limitation in performance of socially defined roles and tasks within a sociocultural and physical environment	*Participation*—involvement in a life situation *Participation Restriction*—problems an individual may experience in involvement in life situations

Figure 4: Nagi and ICF models of disablement with domain definitions. Adapted from Jette, (2006).

Using the ICF framework, many assessment tools have been constructed to gather both clinician and patient reported outcomes on factors surrounding body, activity and participation domains (McLeod et al., 2008; Snyder et al., 2008). In particular, the use of self-assessment tools has become prevalent in clinical and research realms due to their capacity to quantitatively assess functional limitations brought on by musculoskeletal injury. Specifically, several self-assessment surveys have been validated to detect dysfunction associated with foot and ankle injuries (e.g. lateral ankle sprains) (Martin & Irrgang, 2007), which often lead to chronic functional limitations such as chronic ankle instability.

Although disablement models allow for a systematic assessment of factors contributing to movement dysfunction, they do not explicitly define a set of measurement tools necessary to quantify specific movement characteristics that may be associated with dysfunction. Thus, researchers have expressed a need for novel assessment protocols that are sensitive to the dimensional complexity of disability factors in order to improve evidence-based treatments (Jette, 2006; McLeod et al., 2008; Steiner et al., 2002). Subsequently, dynamic systems approaches, along with unique measurement techniques, have been proposed as a model to guide research pertaining to movement dysfunction amongst a variety of clinical populations (Hoch & McKeon, 2010; Holt, Wagenaar, & Saltzman, 2010; McKeon & Hertel, 2006; Perry, 1998; Scholz, 1990).

The advantage of adopting a dynamic systems framework to the study of movement dysfunction is evident when comparing the similarities between domains of disablement models and a constraint led assessment of movement outcomes. Specifically, a dynamic systems viewpoint surrounding movement dysfunction considers how neuromusculoskeletal, psychological, task, and environmental factors interact to produce an observed movement outcome. This systems approach is congruent with treatment approaches and disablement models that attribute movement dysfunction to the interaction of biological, psychological, and social factors (Hoch & McKeon, 2010; Holt, Wagenaar, & Saltzman, 2010; Wikstrom, Hubbard-Turner, & McKeon, 2013).

From a research standpoint, Holt et al., (2010) discussed the limitations of disablement models (Nagi and ICF) to explore ways in which domains contributing to disability might influence one another. In this paper, the authors propose using a dynamic systems/constraints led approach to understand the relationships between levels of impairment and function and how the approach can be used to justify the adoption of novel methods of evaluation. The authors recognized the relationship between the dynamic systems/constraints approach and the Nagi/ICF models, and observed that as analysis shifts from the pathology to actions within the environment, a systems approach requires one to acknowledge the interplay of an increasing number of elements and constraints that guide the emergence of a movement pattern (Figure 5).

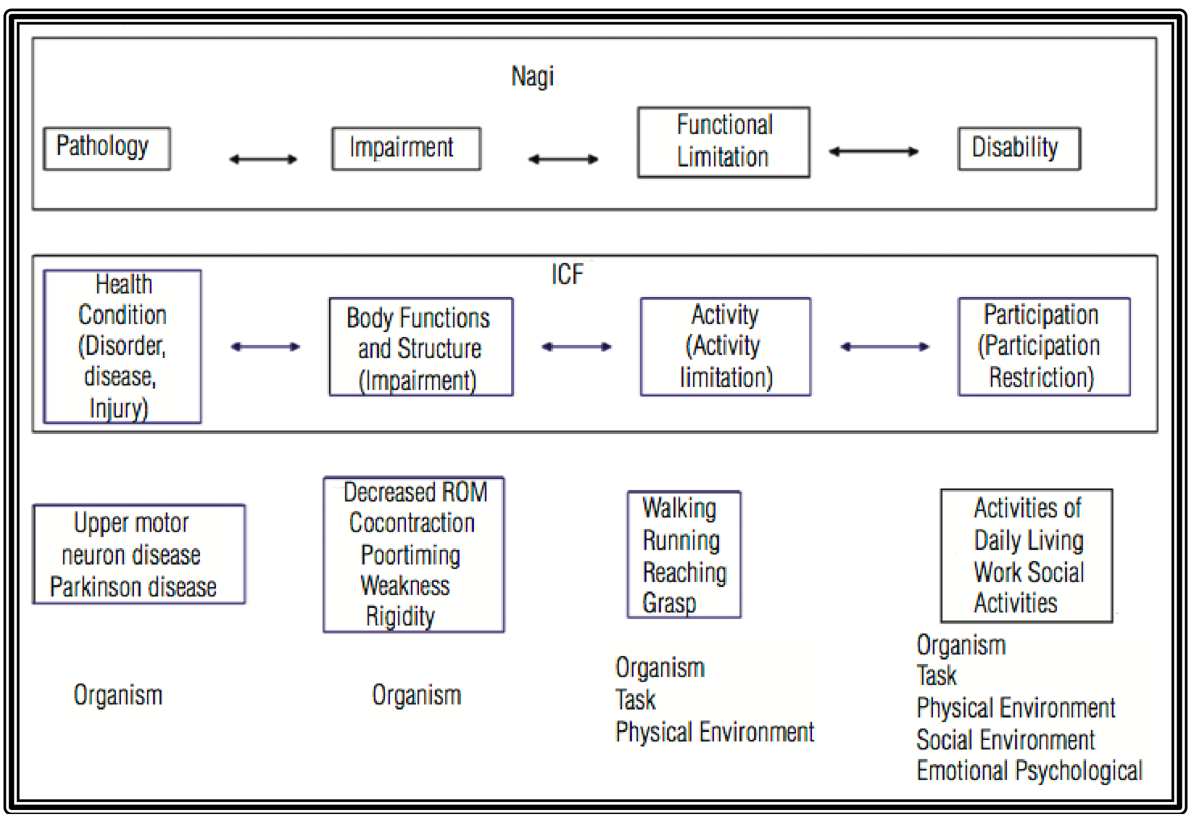

Figure 5: Congruency amongst domains of the Nagi and ICF disablement models and a dynamic systems perspective on constraints. Adapted from Holt et al., (2010)

Furthermore, Holt et al., (2010) proposed the implementation of biomechanical models and assessment of coordination dynamics as modes for understanding the interplay between physical constraints associated with the organism, task, and environment. A central concept surrounding this model is that dynamic constraints based in mechanics are critical to the understanding of movement pattern formation. Two types of dynamic constraints discussed are dynamic resources and environmental constraints. Dynamic resources include properties of the neuromusculoskeletal system that can be used to generate and conserve energy to perform a task; active resources allow for temporally aligned muscle contractions to generate force, while passive resources are features of the musculoskeletal system that allow segments to store and dissipate energy. Environmental constraints are extrinsic sources or dampers of energy that act with dynamic resources to guide pattern formation for a given task.

Alteration of dynamic resources, as the result of impairment or changes in environmental contexts, will lead to changes in the observed movement pattern. Thus, analytical tools related to movement dynamics (e.g. movement variability analysis) may be useful for understanding the association between injury-related impairments and dysfunction (Holt, Wagenaar, & Saltzman, 2010; McKeon & Hertel, 2006; Perry, 1998; Scholz, 1990). The remainder of this section will focus on introducing dynamic systems concepts to the study of movement dysfunction brought on by ankle injury.

Chronic Ankle Instability

Ankle injuries are one of the most commonly occurring types of musculoskeletal injury, with lateral ankle sprains (i.e. injury to the lateral ligaments of the ankle complex) being the most prevalent form of ankle injury (Doherty et al., 2013). Within the general population, emergency room visits due to ankle sprains occur at an estimated rate of 2.15 injuries per 1000 person-years in the United States (Waterman et al., 2010) and account for approximately 3-5% of visits in the United Kingdom (~5,600 per day) (Cooke et al., 2003). The likelihood of an ankle sprain injury is greater in females over the age of 30, children and adolescents, and particularly athletic populations (Doherty et al., 2013; Fong et al., 2007; Yeung et al., 1994). However, the prevalence of ankle sprains may be underrepresented by some epidemiological studies, with some research indicating approximately half of ankle sprain injuries in young athletic populations are not assessed by medical professionals (Smith & Reischl, 1986).

Despite the notion that ankle sprains are isolated incidences with minimal residual effects, individuals commonly experience symptoms well past the acute phases of injury healing (Anandacoomarasamy & Barnsley, 2005). Additionally, a main risk factor for lateral ankle sprains is the previous incidence of lateral ankle ligamentous injury (Barker, Beynnon, & Renström, 1997; Martin et al., 2013; Thacker et al., 1999). The recurrence of lateral ankle sprains has been shown to be as high as 34% (Martin et al., 2013), suggesting that mechanical and functional impairments may exist as a result of the initial injury and contribute to the heightened risk of re-injury and acquired dysfunction.

Chronic ankle instability (CAI) is a condition involving a broad scope of symptoms including bouts of perceived or measured ankle instability, recurrence of ankle sprains, pain, or activity limitations in individuals who have suffered a previous ankle sprain injury (Anandacoomarasamy & Barnsley, 2005; Hertel, 2002; Peters, Trevino, & Renström, 1991; Wikstrom et al., 2012). Individuals with CAI are commonly characterized as ankle sprain non-copers based on the progression of symptoms following initial injury, which differs from individuals (i.e. ankle sprain copers) who return to performing normal activities without recurrent injury or diminished function (Wikstrom et al., 2009). Distinguishing between ankle sprain copers and individuals with CAI (non-copers) is typically based on assessment of a variety of self-assessment tools and outcomes scales used by researchers and clinicians (Wikstrom et al., 2012), however these tools have been considered subjective in nature (Demeritt et al., 2002). Thus, more recent research has begun to investigate biomechanical and performance variables to help distinguish between ankle sprain copers and non-copers (Doherty, Bleakley, et al., 2015a; Doherty, Bleakley, et al., 2015b) in order to better understand the mechanisms that underlie the development of CAI.

Hertel (2002) highlighted mechanical and functional instability as two of the main factors associated with CAI, indicating that both factors likely act mutually to form a continuum of pathology leading to chronicity (Figure 6). Mechanical instability has been defined as ankle joint motion that exceeds normal physiological ranges (Tropp, Odenrick, & Gilquist, 1985) although both hypermobility and hypomobility of the ankle complex may contribute to pathomechanics following injury (Hubbard & Hertel, 2006). Kovaleski et al., (2014a) used an ankle arthrometer to assess mechanical properties of cadaveric lateral ankle complex ligaments [anterior talofibular (ATFL) and calcaneofibular (CFL) ligaments] before and after a simulated ankle sprain injury. Both isolated sectioning of the ATFL and combined sectioning of the ATFL and CFL resulted in greater anterior displacement of the ankle complex under passive anterior loading compared to the intact ankle. In addition, combined sectioning of the ATFL and CFL resulted in greater inversion range of motion under passive inversion loading compared to intact and ATFL-sectioned ankles.

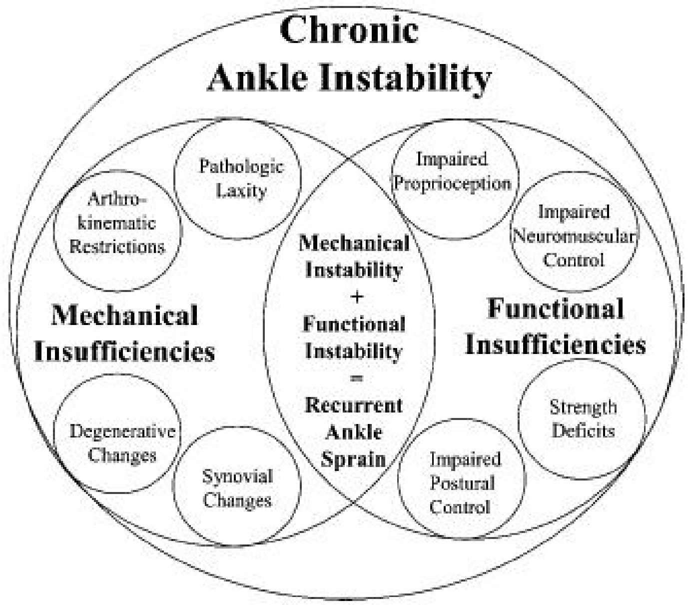

Figure 6: Mechanical and functional factors that contribute to chronic ankle instability (Hertel, 2002).

Kovaleski et al., (2014b) used similar methods to assess joint stability characteristics of the ankle complex in female athletes with a history of lateral ankle sprain injury. Results showed that passive inversion loading produced greater inversion range of motion in ankles that had previously suffered lateral ankle sprain injury compared to healthy ankles. Together, these results demonstrate that damage to ligaments of the lateral ankle complex results in pathomechanical functioning of the joint under loading profiles typically associated with lateral ankle sprains, and may contribute to recurrence of injury due to the diminished capacity of lateral ankle ligaments to prevent excessive displacement of the joint during loading.

Freeman (1965) introduced the concept of functional ankle instability as the tendency for the foot to “give way” as a result of pathological processes that arise proceeding ankle sprain injury. Mechanoreceptors located within lateral ligaments and joint capsules of the ankle complex function to detect strain (e.g. during excessive ankle supination) and send afferent signals to the central nervous system, resulting in an efferent response to muscles that act antagonistically (e.g. peroneal muscles) to mitigate further ligamentous strain. Additionally, muscle (i.e. muscle spindles) and tendon (i.e. golgi tendon organs) receptors act in conjunction with mechanoreceptors to detect changes in joint position. Disruption of ligamentous and muscle sensory receptors may contribute to diminished joint stability and position sense acuity (Hertel, 2000). A cascade of injury-related deficiencies in ankle proprioception (Lentell et al., 1995; Willems et al., 2002), cutaneous receptivity and nerve-conduction velocity (Stoff & Greene, 1982), neuromuscular (Konradsen & Ravn, 1990) and postural control (Doherty, Bleakley, et al., 2015a), or strength (Willems et al., 2002), may diminish the capacity for dynamic support at the ankle and contribute to functional instability (Hertel, 2000; Hertel, 2002).

Mechanical and functional impairments associated with lateral ankle sprains have a confounding effect. Additional complexity is garnered when observed subsets of the population who experience mechanical or functional impairments resulting from lateral ankle sprain do not exhibit characteristics of dysfunction (e.g. perceived instability, injury recurrence, activity limitations) (Wikstrom et al., 2012). Consequently, understanding the factors that drive the progression of CAI is a formidable, yet pivotal process in order to develop effective interventions to combat dysfunctional behavioral patterns. Self-reported outcome instruments (e.g. Foot and Ankle Disability Index, Foot and Ankle Ability Measure) have become increasingly popular among researchers and clinicians seeking to quantify dysfunction associated with CAI (Martin & Irrgang, 2007). Patient self-report instruments provide a comprehensive approach to quantifying dysfunction that is congruent with disablement models (e.g. Nagi and ICF), and help identify sources of functional loss and disability from a patient’s perspective. Yet from a research standpoint, there remains a gap in the understanding of how mechanical and functional impairments relate to self-reported dysfunction associated with CAI (Hoch & McKeon, 2010; Wikstrom, Hubbard-Turner, & McKeon, 2013).

Within the past decade, researchers have begun applying concepts from dynamic systems in an effort to elucidate the cascade from local instability (mechanical and functional impairments) of the ankle following sprain injury, to developing CAI (Hoch & McKeon, 2010; Wikstrom, Hubbard-Turner, & McKeon, 2013). Two concepts from dynamic systems are central to these efforts and include the understanding of constraints that influence movement related behavior following ankle injury and cultivating functional variability to restore function in individuals with CAI. Functional variability refers to the ability of the movement system to (re)organize movement strategies based on the interaction of constraints (Davids et al., 2003). As mentioned previously, injury-related impairments in the form of mechanical or sensorimotor deficiencies can have a direct influence on the available strategies to accomplish a given task. Specifically, diminished mechanical characteristics related to joint function (e.g. laxity) or functional characteristics related to sensorimotor (e.g. proprioception) and neuromuscular function present as novel constraints that limit the ability of the motor system to maintain functional levels of movement variability (Hertel, 2008; Hoch & McKeon, 2010; Wikstrom, Hubbard-Turner, & McKeon, 2013). Using a dynamic systems approach, researchers have begun to assess the effects of injury-related sensorimotor and mechanical constraints on fundamental aspects of locomotion coordination dynamics in individuals with CAI.

Two studies have been conducted to assess coordination between the shank and rearfoot and the associated coordination dynamics in CAI patients. Drewes et al., (2009) assessed the kinematics and coordination dynamics of the shank and rearfoot throughout treadmill walking and running gait cycles in individuals with and without CAI. Continuous relative phase (CRP) provided a measure of segment coordination throughout the gait cycle and was calculated by taking the difference of shank and rearfoot phase angles at each point of the gait cycle. Mean absolute relative phase (MARP) was calculated by averaging the absolute values of the ensemble CRP curve points, which condenses the relative phase values throughout the gait cycle into one value, giving a generalized assessment of the nature of segment coordination during the task. Lower MARP values indicate that the coupling was in phase (i.e. rearfoot motion occurred in concert with shank motion) while higher MARP values indicate that the coupling was out of phase (e.g. rearfoot motion occurred independent of shank motion). Deviation phase (DP) is used to estimate the variability in CRP values during the gait cycle and provide an indication of the pattern stability. DP was calculated by averaging the standard deviations of CRP values across trials, with smaller DP values indicating a highly stable pattern (less variability) and larger DP values indicating a less stable pattern (more variability). Kinematic results from this study showed that the CAI group exhibited more rearfoot inversion and shank external rotation during both walking and jogging. Additionally, the CAI group exhibited a more out of phase shank/rearfoot coupling during terminal swing of walking as well as a more out of phase coupling during terminal stance and terminal swing of jogging. No differences were noted between groups when assessing segment coordination (MARP) or coordination variability (DP) over the entire gait cycle. Kinematic results agree with previous research that suggests CAI groups tend to produce greater inversion angles during terminal swing phase and initial foot contact (Chinn, Dicharry, & Hertel, 2013; Delahunt, Monaghan, & Caulfield, 2006a; Monaghan, Delahunt, & Caulfield, 2006). Furthermore, the authors concluded that the altered segment coupling is an indication of a less coordinated state as the foot nears initial contact, which may be a predisposition to dysfunction in individuals with CAI (Drewes et al., 2009).

With a similar conceptual approach, Herb et al., (2014) used a vector coding method to assess the shank-rearfoot coordination during treadmill walking and running in individuals with CAI. Vector coding quantifies the spatial coupling relationship between two joint position time series from a plot of relative motion (Miller et al., 2010; Tepavac & Carmen, 2001). Specifically, angle-angle plots are generated from segment kinematic data (e.g. shank internal/external rotation angular position vs. rearfoot inversion/eversion angular position) throughout the gait cycle. From the angle-angle plot, a coupling angle is formed by the intersection of a horizontal line to the right of data point *i* and the resultant vector between points *i* and *i*+1 (Figure 7), allowing for interpretations of segment coupling to be made throughout the gait cycle; Coupling angles greater than 45º indicated relatively larger rearfoot excursions compared with shank motion, coupling angles less than 45º indicated relatively larger shank rotation excursions compared with rearfoot motion, and a coupling angle of 45º indicated relatively equal segmental motion.

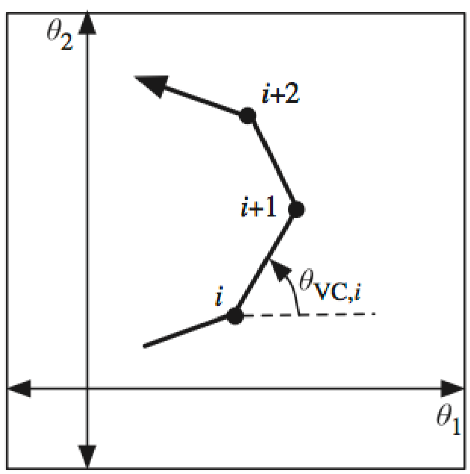

Figure 7: Determining coupling angles (VC) from angle-angle plot. Adapted from Miller et al., (2010).

The magnitude (length) of the resultant vector represented the magnitude of coupled motion. The intersegmental variability coefficient (VCV) was calculated, as the stride-to-stride consistency of magnitude and coupling, to provide insight into sensorimotor behavior in the control of gait (Herb et al., 2014). The authors hypothesized that individuals with CAI would demonstrate greater magnitude, coupling angles, but less variability in coupling during terminal swing and initial contact phases of the gait cycle compared with controls.

During the walking condition, the CAI group exhibited a significantly lower average coupling angle (greater relative rearfoot motion than shank motion) during early swing phase and terminal swing phase compared with the control group. In addition, the CAI group exhibited greater magnitude during swing initiation but lower magnitude during terminal swing phase during walking. Lesser magnitude was said to be indicative of less relative excursion between segments. Coupling variability was less in the CAI group than controls during terminal stance and early swing phases of walking gait, which the authors surmised might be related to less adaptability to constraints by the sensorimotor system. During running, the CAI group had lower coupling angles during early stance, mid-swing and terminal swing phases, similar to the walking condition, indicating greater relative rearfoot motion. No differences in magnitude or variability measures were noted for the jogging condition (Herb et al., 2014).

Both studies demonstrated alterations in relative motion of the shank and rearfoot in CAI patients during walking (Table 1). Furthermore, the fact that group differences in coordination variability did not exist when analyzing the gait cycle as a whole (Drewes et al., 2009), but were evident in specific phases of the gait cycle (Herb et al., 2014) may be an indication that impairments (or organismic constraints) contributing to dysfunction in CAI patients induce heterogeneous movement dynamics throughout the walking gait cycle in relationship to changes in environmental constraints (i.e. the presence or absence of ground reaction forces throughout the gait cycle). It should be noted that due to methodological differences in quantifying relative motion, comparisons between these studies are made with caution (Miller et al., 2010). Based on previous research, it is likely that functional and mechanical impairments that contribute to CAI can negatively affect central aspects of motor control, causing systemic alterations in movement pattern organization. A limitation of the previous research on coordination dynamics in CAI patients is that only shank/rearfoot coordination dynamics were assessed, whereas analysis of segments adjacent to the focal sight would provide more information on systemic maladaptation contributing to global dysfunction (Doherty, Bleakley, et al., 2015b)

Article	Aim	Participants	Task	Dependent Measures	Results	Conclusion
Drewes et al., 2009	To determine if CAI participants demonstrate altered ankle kinematics and shank – rear-foot coupling compared with controls during walking	CAI – 7 Healthy – 7	Treadmill walking (4.83km/h)	<u>Kinematics</u> 3D foot and shank kinematics <u>Segment Coordination</u> Continuous Relative phase (CRP) Mean absolute relative phase (MARP) <u>Coordination Variability</u> Deviation phase (DP)	<u>Kinematics</u>: CAI group exhibited more inversion throughout the gait cycle and more external shank rotation throughout a majority of the gait cycle. <u>Coordination</u>: CAI demonstrated a more out of phase foot – shank coupling during terminal swing phase. No significant difference in MARP. <u>Variability</u>: No significant difference DP measures.	Altered ankle kinematics and joint coupling during terminal swing phase of gait may predispose CAI individuals to ankle-inversion injuries.
Herb et al., 2014	To identify differences in joint coupling and variability between shank internal/external rotation and rearfoot inversion/eversion throughout the gait cycle of CAI subjects and healthy controls.	CAI – 15 Healthy – 13	Treadmill walking (4.83km/h)	<u>Segment Coordination</u> Coupling angle (ø) Magnitude of coupled motion (magnitude) <u>Coordination Variability</u> Coupling variability (VCV)	<u>Coordination:</u> CAI group exhibited lower ø (greater relative rear foot motion compared to the shank) during early and late swing phase. CAI group had lower magnitude (less relative excursion between the foot and shank) during terminal swing phase. <u>Variability</u>: CAI group had less foot – shank variability during late stance phase, toe-of and early swing phase.	CAI may lead to reorganization of the sensorimotor system to adapt to the internal constraints related to the pathology and result in changes in the joint couplings

Table 1: Summary of cited studies pertaining to coordination and coordination variability characteristics in individuals with chronic ankle instability (CAI).

Nonetheless, analysis of coordination dynamics have been shown to be successful in quantifying aspects of locomotion that are likely affected by impairments contributing to CAI, which may be difficult to assess using traditional biomechanical measures or self-report instruments. These types of analyses may provide a new perspective on how a system responds to degradations in sensorimotor functioning as a result of damage to neuromusculoskeletal structures associated with ankle sprain injury. Future research should focus on using similar methodological assessments of coordination dynamics in order to establish sound comparisons to the current literature. In addition, assessing coordination dynamics along a population continuum (i.e. comparing healthy individuals with ankle sprain copers and non copers) may provide a unique insight into the mechanism of pathological progression toward movement dysfunction and disability associated with lateral ankle sprains. Last, interventions used to improve function in individuals with CAI often introduce additional constraints on movement outcome. Assessing how such interventions influence the movement dynamics may provide a better understanding of the systemic effects and efficacy of such protocols.

Joint Bracing

Thus far, a focus of this review of literature has been on the impact of organismic constraints that arise from damage to neuromusculoskeletal structures and influence the organization of movement outcomes. Organismic constraints in the form of mechanical impairments or sensorimotor deficits may drastically influence the centrally mediated organization of movement resulting in functional limitations that lead the development of chronic pathology [e.g. chronic ankle instability (CAI)] (Bastien et al., 2014; Hass et al., 2010). However, task constraints such as task goals, movement specifications, or implements limiting kinematic or dynamic characteristics, should also be considered when assessing movement dynamics in healthy and clinical populations (Newell, 1986). The remainder of this review of literature will focus on the impact of joint bracing, a task constraint, on movement parameters. In particular, discussion of this topic will center on bracing effects on individuals who are predisposed to the development of movement dysfunction as a result of lateral ankle sprain injury.

Joint bracing is a common form of musculoskeletal injury prophylaxis amongst healthy, athletic, and clinical populations. Bracing is often implemented on joints of the lower extremity due to the susceptibility of those structures to injury (e.g. ligamentous sprains or tears) and the development of osteoarthritic conditions, which may be exacerbated by previous injury (Gravlee & Van Durme, 2007). Due to the prevalence of lateral ankle sprain injuries, ankle constraints (e.g. bracing and taping) have become a common prophylactic used to avoid an ankle sprain injury. Ankle bracing, in particular, has been shown to be an effective method of preventing ankle injury, particularly in athletes who have previously suffered an ankle injury (Callaghan, 1997; Gravlee & Van Durme, 2007; Hume & Gerrard, 1998; Papadopoulos et al., 2005; Thacker et al., 1999).

The most commonly reported factor attributed to the efficacy of ankle bracing is improved mechanical stability of the ankle joint. Specifically, ankle bracing has been shown to restrict both sagittal and frontal plane motion of the ankle joint under passive and dynamic loading (Cordova, Ingersoll, & Palmieri, 2002; DiStefano et al., 2008; Shapiro et al., 1994; Simpson, 2013; Zhang et al., 2012), which is ideal for combating the most common mechanism of ankle sprain injury (i.e. excessive ankle inversion, internal rotation and plantarflexion). However, the effect of ankle bracing on additional measures of ankle joint biomechanics (e.g. the rate of change in joint position, or magnitude and rate of joint loading) has been less conclusive and warrants further research (Cordova, Ingersoll, & Palmieri, 2002; Reisberg & Verstraete, 1993).

Additionally, it is not well known how ankle bracing impacts the biomechanics of adjacent segments under dynamic loading. Due to the constrictive nature of ankle bracing, some researchers argue that long term use of ankle bracing may increase the risk of suffering musculoskeletal injury in adjacent structures as a result of linked segments being exposed to greater rates and magnitudes of loading (DiStefano et al., 2008; Mason-Mackay, Whatman, & Reid, 2015; Papadopoulos et al., 2005; Simpson, 2013; Venesky, 2006). A recent review by Mason-Mackay, Whatman, & Reid (2015) examined evidence involving the effects of ankle bracing on lower-extremity landing biomechanics. It was concluded that evidence exists to support altered lower extremity biomechanics during landings as a result of ankle bracing, but that the strength of such evidence depends on the variable of interest. Specifically, the authors reported there is moderate evidence that restricted ankle motion, as a result of bracing, limits knee flexion excursion. Further, there was poor evidence to suggest that sagittal plane ankle restriction affects hip kinematics. Finally, it was reported that evidence is conflicted with respect to changes in the magnitude and rates of ground reaction forces and joint loading rates as a result of ankle bracing. It was postulated that discrepancies in methodological approaches, task parameters, braces tested, within and between subjects variability in landing strategies, and the sole evaluation of biomechanical variables rather than movement patterns may contribute to inconsistencies in findings on biomechanical changes associated with ankle joint bracing (Mason-Mackay, Whatman, & Reid, 2015).

Another purported mechanism by which ankle braces reduce the occurrence of ankle injury is through alterations in sensorimotor and neuromuscular functioning. Research in this area is particularly relevant to CAI populations who often exhibit functional impairments in addition to mechanical impairments. Thus, the effects of ankle bracing on sensorimotor and neuromuscular measures has been an area of interest for researchers seeking to find effective interventions for mitigating the effects of functional impairments that may lead to chronic movement dysfunction in individuals who suffer ankle sprain injuries.

Bracing effects on proprioception have been tested in both healthy individuals and individuals with ankle instability. Heit, Lephart, & Rozzi (1996) investigated the effects of ankle bracing and taping on ankle joint position sense in healthy individuals. Compared to no bracing, sagittal plane (i.e. dorsiflexion/plantarflexion) joint position sense was increased in both the taped and braced conditions, while frontal plane (i.e. inversion/eversion) joint position sense was increased in the taped condition. The authors concluded that the mechanism by which bracing prevents ankle injury is the stimulating of cutaneous receptors or mechanoreceptors in the tissues surrounding the ankle joint, thereby facilitating the muscular stabilization in response to excessive joint motion.

Hartsell (2000) tested the effects of bracing on ankle frontal plane joint position sense awareness in individuals with CAI and healthy controls. The results showed that the CAI group had significantly greater joint position sense error compared to the healthy group. However, joint position sense was significantly improved in the CAI group with the application of an ankle brace. It was concluded that deficits in joint position sense associated with CAI may be mitigated through the use of external braces and that bracing should be recommended for patients with chronically unstable ankles.

Although earlier research, including reviews on the efficacy of prophylactic ankle bracing lend support to the notion that bracing improves proprioceptive capability (Cordova, Ingersoll, &

Palmieri, 2002; Ozer, 2009; Wilkerson, 2002), results pooled from more recent research on this topic indicate that there is little to no benefit of ankle bracing in improving joint position sense acuity in individuals with functional impairments brought on by ankle sprain injury (Raymond et al., 2012). Given that the level of proprioceptive impairments brought on by ankle sprain injury may be different between individuals, it may be difficult to draw significant conclusions regarding the effects of bracing on proprioception in individuals with CAI (Raymond et al., 2012). Furthermore, considering that most measures proprioceptive acuity are usually obtained in an unloaded state in which the joint is displaced either actively or passively, results may have little carry over to tasks including locomotion and drop landings, where many additional factors that can influence proprioception are introduced (e.g. ground reaction forces, synergistic muscles forces, segment inertia, and environmental factors) (Maffiuletti et al., 2005). Altogether, ankle bracing may have some positive effects on proprioception; however it is unlikely that proprioceptive alterations from bracing are the sole contributor to the injury prevention mechanism in individuals with CAI.

Neuromuscular function has shown to be influenced by ankle bracing, providing additional insight regarding the prophylactic capabilities of ankle bracing in clinical populations. Of particular interest to researchers has been the activation time and reflex amplitude of the peroneal muscle group. Specifically studies measuring peroneus longus latency during sudden inversion have shown ankle constraint (i.e. taping or bracing) is successful in reducing latency in subjects with CAI (Cordova, Ingersoll, & Palmieri, 2002; Wilkerson, 2002). Other research have found no difference in peroneus longus latency with and without ankle constraint during sudden inversion, cutting movements, and walking, however this may be a promising finding indicating bracing does not increase latency in peroneus longus firing which would diminish the dynamic stabilization of the ankle joint (Cordova, Ingersoll, & Palmieri, 2002; Papadopoulos et al., 2005). Furthermore, reflex amplitude of peroneal muscles has shown to be increased with bracing (Nishikawa & Grabiner, 1999), while other studies have found contradicting results including neither inhibition or facilitation of peroneal reflexive activity (Cordova, Ingersoll, & Palmieri, 2002; Sefton et al., 2007).

Other areas of research investigating the effects of ankle bracing have focused on measures of neuromuscular activity during walking and functional exercises (e.g. single limb balance and leg reach, forward lunge, and hopping). Recently, Feger et al., (2014) demonstrated that in individuals with CAI, lower extremity muscles had a collectively lower surface electromyography amplitude while the ankle was braced during single leg exercises compared to no bracing. It was concluded that ankle bracing may elicit an unfavorable response in lower extremity musculature during common rehabilitation exercises for CAI patients, and its use during tasks aimed to maximize neuromuscular responses should be reconsidered. Similarly, Barlow et al., (2015) assessed the effect of ankle bracing on surface electromyography measures during walking in individuals with CAI. Ankle bracing resulted in smaller peroneus longus amplitude prior to foot contact, and later activation of the anterior tibialis, peroneus longus, rectus femoris and gluteus medius relative to foot contact compared to no brace. Further, the peroneus longus and rectus femoris were activated for a shorter percentage of the stride cycle during the braced condition. Contrary to Feger et al., (2014) the authors concluded that the diminished response of muscle activation patterns may improve the efficiency of dynamic ankle stabilizers and the central nervous system, and may contribute to the mechanism in which ankle bracing decreases the recurrence of ankle sprain injury in individuals with CAI (Barlow et al., 2015).

Kautzky et al., (2015) were the first to assess the effects of ankle bracing on lower extremity muscle recruitment variability in healthy and CAI patients during treadmill walking, under the premise that measures of physiological variability provide an indication of the ability of a body to adapt to changing environments and reduce the overuse of specific anatomical structures. CAI patients had reduced amplitude variability in the gluteus medius prior to initial contact, and more variable activation timing in the biceps femoris. In both groups collectively, bracing caused decreases in activation amplitude variability of the shank muscles (i.e. peroneus longus, anterior tibialis, and gastrocnemius) prior to foot contact, with concurrent increases in activation amplitude variability of the thigh muscles (i.e. rectus femoris and gluteus medius). The authors stated that decreases in shank activation variability could be related to more consistent foot alignment prior to foot contact when wearing an ankle brace. It was also suggested that increases in thigh muscle activation variability with ankle bracing could explain alterations in proximal joint kinematics and kinetics that have been observed in previous studies, and could serve as an adaptive mechanism to account for reductions in distal segment muscle activation variability (Kautzky, 2015).

Despite apparent alterations in neuromuscular activity associated with ankle bracing, some research has shown that kinematic and kinetic gait parameters exhibit minimal differences when comparing braced and not braced conditions in healthy and CAI patients (Spaulding, Livingston, & Hartsell, 2003), suggesting that neuromuscular adaptations may not be strong enough to invoke altered movement patterns. However, it may also be true that traditional measures used to assess gait and movement patterns are not particularly effective in detecting movement related adaptations that occur as a result of compounding, and possibly novel, constraints acting on the motor system.

Recently, Herb, Chinn, and Hertel (2016) investigated that effects of ankle taping on the coordination and coordination dynamics of the foot – shank coupling in healthy and CAI individuals using a vector coding analysis similar to Herb et al. (2014). It was hypothesized that the application of tape would decrease the magnitude of coupled motion between the shank and rear foot, decrease the ratio of coupled motion, and decrease the coordination variability during walking and jogging. During walking coordination variability was significantly lower throughout the entire gait cycle in the tape condition for both groups. Additionally, the magnitude of coupled motion was lower for both groups near foot contact in the tape condition. Similar results were reported for the jogging task, however reductions in the magnitude of coupled motion were observed later in the jogging gait cycle (near mid-stance to toe-off and terminal swing phase) for both groups. The investigators postulated that decreases in coupling motion and variability in both groups in response to taping could represent a protective mechanism in which more consistent gait patterns are adopted. It was concluded that more consistent ankle joint coupling could contribute to the effectiveness of prophylactic ankle support to prevent ankle sprains (Herb, Chinn, & Hertel, 2016).

Collectively, the analysis of ankle bracing effects on biomechanical and neuromuscular measures provides some evidence suggesting prophylactic ankle support can have a direct influence on movement parameters (Table 2). Moreover, measures of coordination dynamics have been effective in identifying alterations to foot – shank movement patterns in response to ankle taping (Herb, Chinn, & Hertel, 2016). Considering this, further research should be conducted to explore the effects of prophylactic ankle support using analytical tools that assess the behavior of movement patterns at a more systemic level.

Article	Aim	Participants	Task	Dependent Measures	Results	Conclusion
Feger et al., 2014	To determine the effect of bracing on sEMG amplitudes in participants with CAI during functional exercises.	CAI – 15	Forward lunge Single limb stance (eyes closed) Star excursion balance test Lateral hops	<u>Electromyography</u> sEMG amplitude of lower extremity musculature	<u>Forward lunge</u>: Bracing reduced LG activity prior to foot contact and PL activity after foot contact. <u>Star excursion</u>: Anterior reach – Decreased PL, LG, RF, and GM during braced trials. Posterolateral reach – Bracing significantly reduced GM activity. No differences between braced and no brace conditions during the single limb eyes closed balance, star excursion balance posteromedial reach, or lateral hop exercises.	Clinicians should be aware of the decreased muscle activity that occurs during common rehabilitation exercises when patients with CAI complete those activities while wearing ankle braces.
Barlow et al., 2015	To determine the effect of lace-up ankle braces on sEMG measures during walking in adults with (CAI).	CAI – 15	Treadmill walking (4.83 km/h)	<u>Electromyography</u> sEMG amplitude, time of activation, percent of activation of lower extremity musculature	Lower pre foot contact amplitude of the PL with bracing Later activation of the TA, PL, RF, and GM relative to foot contact with bracing PL and RF were activated for a shorter percentage of the stride cycle with bracing	Braces cause a change in neuromuscular activity during walking. Clinicians should be aware of these changes when prescribing braces, as it may relate to the mechanism in which braces decrease sprains.

Kautzky et al., 2015	To determine whether muscle recruitment variability during walking differs between groups with and without CAI in unbraced and braced conditions.	CAI -15 Healthy -15	Treadmill walking (4.83 km/h)	<u>Electromyography</u> SD of timing of muscle activation relative to foot contact SD of percent activation time across the stride cycle COV of activation amplitude before and after foot contact	CAI group had more variable GM amplitudes prior to foot contact and more variable timing of activation in the BF. Braced conditions resulted in greater variability in time of activation in the PL and less in the RF relative to initial contact. Bracing decreased activation amplitude variability prior to foot contact in the TA, PL, and BF and increased in the RF.	Clinicians should be aware of proximal alterations in muscle recruitment variability in patients with CAI. Neuromuscular changes with bracing enhance the understanding of how ankle braces may prevent injury and provide new insight into proximal neuromuscular alterations with CAI.
Herb et al., 2016	To analyze the ankle-joint coupling during walking and jogging gait with the application of ankle tape and without ankle tape in adults with and without CAI.	CAI -15 Healthy – 11	Treadmill walking (4.83 km/h)	<u>Segment Coordination</u> Coupling angle (ø) Magnitude of coupled motion (magnitude) <u>Coordination Variability</u> Coupling variability (VCV)	Magnitude of coupled motion and VCV were lower in the taped condition compared to no tape in both groups. Lower magnitude (less relative excursion between the foot and shank) was identified near foot contact, while less VCV was identified throughout the gait cycle. No difference in ø between conditions.	A decrease in the magnitude of coupled motion and VCV may represent a protective mechanism of ankle taping in CAI and healthy patients during gait.

Table 2: Summary of cited studies pertaining to ankle bracing effects on coordination, coordination variability, and neuromuscular activation in CAI populations. sEMG = surface electromyography, SD = standard deviation, PL = peroneus longus, BF = biceps femoris, TA = tibialis anterior, RF = rectus femoris, GM = gluteus medius, LG = lateral gastrocnemius.

Summary

Chronic ankle instability (CAI) is a common form of movement dysfunction that arises after ankle injury as a result of mechanical and functional impairments that act as constraints on movement outcomes. Damage to ligamentous or muscle tissue can result in mechanical laxity of the involved joint, which may lead to local movement impairments as well as global biomechanical alterations. Furthermore, damage to sensorimotor elements such as mechanoreceptors and proprioceptors, can result in altered neuromuscular function and central motor control. Combined, these impairments can lead to a cascade of maladaptive compensatory movement strategies that lead to chronic dysfunction (e.g. CAI), or in some cases, may be compensated for to attenuate the progression of movement dysfunction (e.g. ankle sprain copers). Traditional biomechanical analyses has been fruitful for quantifying fundamental aspects of movement that may be altered in CAI populations, but are equivocal when assessing the global deficits associated with movement dysfunction in CAI populations such that progression of pathological movement from acute to chronic states has yet to be explained. In addition, common approaches used to combat residual symptoms following ankle injury (e.g. recurring ankle sprains or feelings of the ankle giving way,) such as ankle bracing, have been shown to be effective, while the mechanism of these effects is not well understood. This has led some researchers to adopt contemporary models of motor control and disablement in an attempt to garner a more systematic understanding of how local and global injury impairments cause long term adaptations in movement and behavioral outcomes, as well as the systemic effect of prophylactic interventions used to combat maladaptive behavior.

Traditionally, variability within the motor system has been viewed as a non-integral factor that should be reduced in order to harbor more favorable movement outcomes. However, from a dynamic systems standpoint, variability is viewed as an innate characteristic of human movement, allowing for transition between coordinated movement patterns that arise from the interplay of organismic, task, and environmental constraints imposed on the movement system. Recently, sports medicine and rehabilitation professionals have adopted dynamic systems approaches in an attempt to explain maladaptive responses to ankle injury that may lead to chronic movement dysfunction. In this light, mechanical and functional impairments are viewed as neuromusculoskeletal constraints that limit an individual's movement repertoire. Further, task constraints imposed by patients, researchers, and clinicians (e.g. rehabilitation exercises or ankle bracing) add to the number of constraints that are imposed on the individual and further complicate the understanding of movement outcomes exhibited during common tasks such as walking or unilateral dynamic stabilization tasks (e.g. single leg hopping).

Measures of movement variability, such as variability in lower leg and foot coordination patterns during walking and running, have been employed in an attempt to understand how levels of constraints impact the organization of movement patterns in CAI populations, but these studies are limited. Moreover, ankle bracing is a common form of injury prophylaxis used by healthy individuals or those who have previously suffered ankle injury, but its effect on how coordinated movement is organized has not extensively been studied. More research is needed to investigate how impairments from ankle injury impact an individual's ability to adapt to changing environmental and task parameters from a systemic perspective. Yet, also of interest is the study of how ankle bracing influences the dynamics of movement patterns, given its purported benefits and noted effects on mechanical and sensorimotor functioning. In doing so, a better understanding of the progression from acute ankle injury to chronic ankle instability as well as the effects and efficacy of ankle joint bracing interventions may be gained.

CHAPTER III

METHODS

The following assessments were made to address the specific aims for this project: 1) Coordination and coordination variability were assessed using relative phase and relative phase variability to determine bracing effects on lower extremity walking and single-leg hopping coordination dynamics in healthy individuals, ankle sprain copers, and ankle sprain non-copers. 2) Neuromuscular activation patterns were assessed during walking and single-leg hopping to provide a further look at the motor system response to joint bracing in healthy individuals, ankle sprain copers, and ankle sprain non-copers.

Participants

Participants recruited for this study included males and females between the ages of 19-30 years. Individuals were recreationally active, partaking in a minimum of 90 minutes of physical activity per week, and satisfied the inclusion criteria for one of three groups: Healthy group, lateral ankle sprain copers (Coper) and chronic ankle instability group (Non-Coper). Individuals without any current lower extremity musculoskeletal injury and without history of neurological impairments, lower extremity surgery or medical condition that presented a contraindication to participation in this study were recruited for the healthy group.

Copers

The ankle sprain Coper group consisted of participants with no residual symptoms following an ankle sprain injury. Specifically, individuals who had suffered at least one unilateral lateral ankle sprain that necessitated immobilization or assisted weight bearing, who currently had no pain, weakness, or instability in the involved ankle, and who had resumed all pre-injury activities without limitation for at least 12 months before testing were recruited as Copers for this study. Individuals had no history of previous neurological impairments, movement disorders, lower extremity surgery, or medical conditions that contraindicated participation in this study.

Non-Copers

Inclusion for the CAI group (Non-Copers) was determined based on the following criteria: 1) History of at least one lateral ankle sprain, occurring greater than 12 months prior to the beginning of this study, requiring a period of assisted weight bearing or immobilization. 2) Chronic weakness, pain, instability or recurrent episodes of giving way in the involved ankle (without injury), attributed to the original injury. 3) Two or more episodes of the involved ankle giving way between three to 12 months of this study. 4) No observed ankle injury or participation in rehabilitation associated with the involved limb within the past three months of this study.

In order to obtain quantitative information related to level of ankle function and stability in each of the three groups, three self-assessment tools were implemented for this project. The Foot and Ankle Ability Measure (FAAM) and Foot and Ankle Ability Measure Sport (FAAM-S) are region-specific assessments designed to quantify dysfunction related to leg, foot and ankle musculoskeletal conditions. The FAAM quantifies function related to activities of daily living while the FAAM-S assess function related to sport or athletic tasks. Martin et al., (2005) investigated the effectiveness of the FAAM and provided evidence of validity for test content, internal structure, score stability and responsiveness. The Ankle Instability Instrument (Docherty et al., 2006) was developed to provide an objective measure of functional ankle instability and obtain information on the presence of instability symptoms. This self-report tool has been shown to be a reliable method of determining if functional ankle instability is present (Docherty et al., 2006).

In addition to the inclusion criteria listed above, individuals in the CAI group met the following scoring criteria modified from Chinn, Dicharry, & Hertel (2013): 1) Three recorded “yes” answers on the Ankle Instability Instrument (AII). 2) Record a score of ≤ 95% on the FAAM. 3) Record a score of ≤ 85% on FAAM-S.

Setting

Data collection took place in the Sport Biomechanics Laboratory (Room #20) at Auburn University.

Materials

An ASO ankle brace (ASO®, Medical Specialties, Inc., Charlotte, NC, USA) was implemented during the bracing condition. . Forty-one 14mm retroreflective markers (MKR-6.4, B&L Engineering, Tustin, California, USA) were attached to anatomical landmarks of the pelvis, thigh, shank and foot using double-sided tape (Duck Tape®, ShurTech Brands, Avon, OH, USA) for individual markers, or foam pre-wrap for rigid marker clusters (Table 3).

Instrumentation

Kinematics

Measurement of segment motion during walking and hopping trials was obtained using a 10-camera Vicon MX optical motion capture system (Vicon®, Los Angeles, CA, USA) with a sampling frequency of 200 Hz. A calibration trial was collected in order to identify marker arrangements and construct a link-segment model from a custom model template; calibration markers were applied solely for calibration trials. Visual 3D software (C-Motion Research

Biomechanics, Germantown, Maryland, USA) was used to calculate absolute segment angular position and angular velocity, and relative phase measures were calculated using custom MATLAB code (MATLAB® R2013b, Mathworks, Inc., Natick, MA).

Marker Name	**Position**	**Segment**
L/RPSIS	Left/Right Posterior Superior Iliac Spine	Calibration
L/RASIS	Left/Right Anterior Superior Iliac Spine	Pelvis
L/RIC	Left/Right Iliac Crest	Pelvis
SAC	Sacrum	Pelvis
L/RTROCH	Left/Right Greater Trochanter	Calibration
L/RTHI1,2,3	Left/Right Lateral Thigh Marker Cluster	Thigh
L/RMKNEE	Left/Right Medial Tibiofemoral Joint	Calibration
L/RLKNEE	Left/Right Lateral Tibiofemoral Joint	Calibration
L/RSHANK1,2,3	Left/Right Lateral Shank Marker Cluster	Shank
L/RMANK	Left/Right Medial Malleolus	Calibration
L/RLANK	Left/Right Lateral Malleolus	Calibration
L/RPHEEL	Left/Right Posterior Calcaneus	Foot
L/RMHEEL	Left/Right Medial Calcaneus	Foot
L/RLHEEL	Left/Right Lateral Calcaneus	Foot
L/RMTP	Left/Right 1st Metatarsophalangeal Joint	Foot
L/RMTP	Left/Right 5th Metatarsophalangeal Joint	Foot
L/RMFT	Left/Right Dorsal Midfoot	Foot

Table 3**:** Locations for retroreflective marker placement. Marker clusters consist of three markers fixed to a rigid plastic plate. Calibration markers were applied solely for calibration trials.

Surface Electromyography

Surface electromyography (sEMG) was used to assess neuromuscular activation during walking and single-leg hopping. Preparation for sEMG electrode placement consisted of procedures recommended by Basmajian and De Luca (1985). Prior to electrode placement, target sites were shaved, abraded, and cleaned using alcohol swabs. Two Ag-AgCl surface electrodes (Red Dot, 3M, St. Paul, MN, USA) were placed, with an interelectrode distance of 2.5cm, over the center of the muscle belly of the following muscle groups of the leg of interest (i.e. affected leg in Non-Coper and Coper group, matched for dominance in the healthy group): peroneus longus, tibialis anterior, lateral gastrocnemius and gluteus medias. During data collection, sEMG signals were sent from sensors attached directly to each electrode pair to a Noraxon Telemyo 2400T-V2 wireless transmitter (Noraxon® U.S.A. Inc., Scottsdale, AZ, USA) and relayed to a Noraxon Telemyo 2400R-World Wide Telemetry receiver (Noraxon® U.S.A. Inc., Scottsdale, AZ, USA), and time synchronized with kinematic data in Vicon Nexus Software (Vicon® Nexus 1.8.5 Software, Vicon®, Los Angeles, CA, USA). Processing of sEMG signals was done using Visual 3D software (C-Motion Research Biomechanics, Germantown, Maryland, USA), and custom MATLAB code (MATLAB® R2013b, Mathworks, Inc., Natick, MA, USA) was used for further analysis of sEMG signals.

Self-Assessment Surveys

The Foot and Ankle Ability Measure (FAAM) and Foot and Ankle Ability Measure Sport (FAAM-S) are comprised of a 21-item scale related to activities of daily living and 8-item scale pertaining to sport respectively; with each subscale receiving scores separately (Appendix B). Each item is scored on a 5-point Likert scale (0 = *unable to do*, 4 = *no difficulty at all*), with score totals ranging from 0-84 for the FAAM and 0-32 for the FAAM-S. Subscale scores were converted to percentage scores, with higher percentage scores indicating higher level of function. In addition, global rating of function scales (represented as a percentage) and categorical rating of function scales are presented at the end of each subscale. The global function ratings ranged from 0% (inability to perform the task listed) to 100% (ability to perform the task meets or exceeds pre-injury level) and the categorical ratings characterize the ankle in a ranged from "normal" to "severely abnormal" (Carcia, 2008).

The Ankle Instability Instrument consisted of 12 items (polar questions), with each item contributing to one of three factors that represent the overall presence of functional ankle instability (Appendix A) (Docherty et al., 2006). The first factor described the severity of initial ankle sprain injury and included four items (2, 2a, 3, and 3a). Factor two represented the history of ankle instability and included five items (1, 4, 4a, 6, and 7). Factor three represented instability while partaking in activities of daily living and included three items (5, 8, and 9).

Design and Procedures

Experimental Design

Experimental design for this project was quasi-experimental, consisting of three groups (Healthy, Coper, Non-Coper) matched for gender and limb dominance. To account for a potential effect of task (i.e. walking and single-leg hopping) and condition [with ankle brace (Brace) and with no ankle brace (No Brace)] order on movement outcome, a fully counterbalanced design consisting of four subgroups was implemented; an equal number of participants from each group were randomly assigned to each of the four subgroups (Table 4).

Subgroup	T1	T2
1	Walking – B, NB	Single-leg hop – B, NB
2	Walking – NB, B	Single-leg hop – NB, B
3	Single-leg hop – B, NB	Walking – B, NB
4	Single-leg hop – NB, B	Walking – NB, B

Table 4: Fully counterbalanced order of task and condition. T1 – *task to be completed first*; T2 – *task to be completed second*; B – *ankle brace condition*; NB – *no brace condition.*

Two a priori power analyses were conducted (G*Power Version 3.1.9.2) to estimate sample size prior to participant recruitment. The first power analysis was conducted to determine the required sample size needed to detect differences in coordination between Healthy and Non-Coper groups. Input parameters included a Cohen's d effect size (1.0), alpha level (0.05), and power (0.80). The effect size input parameter was estimated from group means and standard error from Drewes et al., (2009) who reported a significant difference in relative phase coordination within the walking gait cycle between individuals with and without chronic ankle instability (CAI). Results from the first power analysis indicated a sample size of 17 participants per group was needed to achieve an actual power of 0.80. The second power analysis was conducted to determine the required sample size needed to detect differences in lower leg surface electromyography (sEMG) measures in the Non-Coper group during the brace and no brace conditions. Input parameters included a Cohen's d effect size (0.72), alpha level (0.05), and power (0.80). The effect size input parameter was obtained from Barlow et al., (2015) who reported a significant difference in peroneus longus sEMG activity between braced and no brace conditions in CAI individuals. Results from the second power analysis indicated a sample size of 18 participants per group was needed to achieve an actual power of 0.82. In order to achieve an equal number of participants within each subgroup without oversampling, a total of 42 participants (16 participants per group) were recruited for this study.

Procedures

Forty-two participants reported to the Sport Biomechanics Laboratory for data collection procedures. Prior to data collection, participants voluntarily signed an Auburn University Institutional Review Board approved informed consent form, and completed a health history questionnaire, The Foot and Ankle Ability Measure, The Foot and Ankle Ability Measure – Sport, and the Ankle Instability Instrument. The health history questionnaire served as a screening tool to ensure that individuals met the inclusion criteria for this study and did not possess any malady that would prevent them from safely completing the study. Three self-assessment surveys were used to ensure that participants meet the designated inclusion criteria for the Healthy, Coper or Non-Coper group. Measurements were taken unilaterally from the target limb. For the Non-Coper and Coper groups, the affected or previously injured limb was considered the target limb. Additionally, the target limb was matched based on limb dominance (i.e. dominant or non-dominant). To determine limb dominance, participants were asked which leg was preferably used to kick a ball with; the preferred striking limb was designated as dominant and the contralateral limb was designated as non-dominant (Sadeghi et al., 2000).

To reduce the incidence of marker occlusion and marker movement artifact during data collection, participants were provided compression shorts and tops to wear throughout the experiment. Anthropometric measures of height and weight were taken prior to electrode and marker placement and were included in the group descriptive analysis. Skin of the electrode placement area was shaved and abraded in order to reduce the likelihood of electrical impedance and improve electrode adhesiveness. Electrodes were placed on the surface of the skin at the midpoint of the muscle belly, parallel to the muscle fibers. Wireless sensors were connected to each electrode pair with pinch sensor leads and attached to the segment adjacent to the respective electrode pair.

Following electrode and sensor placement, maximal volitional isometric muscle contractions (MVIC), consisting of resisted manual muscle tests lasting five seconds, were conducted in order to obtain and record signal for normalization of sEMG signals. Manual muscle testing for each muscle group was conducted using the following procedures from Kendall et al., (2005): Gluteus medius – resisted hip abduction while lying on the contralateral side with knees extended; Peroneus longus – resisted ankle eversion while seated with knee extended; Gastrocnemius – resisted ankle plantarflexion while seated with knee extended; Tibialis anterior – resisted ankle dorsiflexion while seated with knee extended. Following MVC testing, retro-reflective markers were placed on the designated anatomical landmarks of the pelvis and lower extremity (Table 1).

Calibration of the ten-camera Vicon system was conducted using a company manufactured rigid L-frame wand fixed with five retro-reflective markers. A static calibration trial was collected and consisted of the participant standing at the center of the capture volume in anatomical neutral. Calibration markers were removed following calibration trials, and the remaining markers served as tracking markers for the dynamic motion capture trials.

For each condition, participants performed both over ground walking and single-leg hopping tasks. Fifteen trials of over ground walking at a preferred normal walking speed were conducted for the No Brace and Brace conditions. A total of 30 over ground walking trials per participant were included for analysis. All walking trials were performed on a 7.6m flat walkway embedded with two vertically aligned force platforms. Walking trials were deemed successful if the foot of the target limb made full contact with the first force platform, and the foot of the contralateral limb made full contact with the second force platform during an uninterrupted cycle of walking gait. For each trial, data from a single gait cycle was obtained for analysis.

The single-leg hop task consisted of participants hopping on the involved or target limb at a self-selected frequency. The use of self-selected frequencies for single leg hopping was chosen to account for differences in participant's mass and preferred hopping frequency, which may alter movement strategies if movement frequencies outside of the preferred are enforced (Granata, Padua, & Wilson, 2002; Hobara et al., 2010). The single-leg hopping task was performed on an embedded force platform and participants were instructed to perform the task with their hands resting on their hips or shoulders. During the single-leg hopping task, a researcher observed foot placement during each block of trials. Footfalls landing outside a designated 30cm x 30cm perimeter were discarded and participants were asked to repeat the block of trials. One trial of single-leg hopping was defined as foot contact to subsequent foot contact. For both AB and NB conditions, participants completed three blocks of hopping, each consisting of nine trials. A minimum of 30 seconds rest was given between blocks. To mitigate effects of movement initiation and cessation, the first and last two trials within each block were discarded. The remaining five trials per block made up a total of 15 single-leg hops per condition performed by each participant. Thus, a total of 30 single-leg hop trials per participant were included for analysis.

Data Processing

A linked-segment model of the pelvis and lower extremity segments was constructed for each participant from tracking and calibration marker locations obtained during a three second static calibration trial. Three-dimensional position data of each tracking marker was collected during subsequent trials of walking and single-leg hopping. Marker trajectories were lowpass filtered (6Hz) and local segment axes were established from global laboratory coordinates using an X-Y-Z rotation sequence. Segment angles and velocities were calculated and used for kinematic assessment of lower extremity coordination during over ground walking and single-leg hopping. Segment angles were determined with respect to global lab coordinates. Kinematic data for over ground walking and single-leg hopping was time-normalized to 100 data points, allowing for between cycle analyses.

Gait cycles for over ground walking were defined as initial foot contact of the target limb to the following ipsilateral foot contact. Foot contact and toe-off events were determined using the automatic gait events function in Visual3D. This function uses vertical and anteroposterior trajectories of the proximal and distal foot at first foot contact and toe-off from the force platform to detect similar subject specific kinematic patterns during a trial (Stanhope et al., 1990). Kinematic and sEMG variables were assessed during intervals of stance and swing phase of the target limb; stance phase was defined as foot contact to toe off. The interval of interest within stance phase was foot contact to mid-stance (1-50% of stance phase). Intervals of interest within swing phase included toe-off to mid-swing (1-50% of swing phase) and terminal swing phase (85-100% of swing phase).

Cycles of single-leg hopping were defined as foot contact to foot contact. For single-leg hopping, foot contact was determined by locating the time point before vertical ground reaction force (GRF) exceeds a threshold of 10N. Intervals within the hopping cycle that were included for analysis consisted of landing (i.e. foot contact to the first maximum in vertical GRF), propulsion (i.e. first maximum in vertical GRF to toe-off), and flight phase (i.e. toe-off to foot contact). Each trial was inspected to in order to detect any erroneous data or missing events within the walking or hopping cycles.

Coordination of lower extremity segment motion during walking and hopping was assessed using relative phase analysis (Table 5). Phase portrait construction consisted of normalizing both segment displacements and velocities to maximum and minimum values and plotting the normalized values, with the segment displacements represented on the abscissa and segment velocities represented on the ordinate. Normalization of the angular displacement to the maxima and minima, and angular velocity to maximum absolute value of the times series allowed the limit-cycle trajectory to be centered on a zero origin and accounted for amplitude and frequency differences between order parameters being assessed, as well as instances where the data were not sinusoidal (Hamill, Haddad, & McDermott, 2000; Kurz & Stergiou, 2002; Lamb & Stockl, 2014; Peters et al., 2003).

#	Coupling	Coupling Abbreviation
1	Foot dorsiflexion/plantarflexion – Shank flexion/extension	Foot DF/PF – Shank F/E
2	Foot dorsiflexion/plantarflexion – Thigh flexion/extension	Foot DF/PF – Thigh F/E
3	Foot inversion/eversion – Shank internal/external rotation	Foot IN/EV – Shank ROT
4	Foot inversion/eversion – Thigh abduction/adduction	Foot IN/EV – Thigh AB/AD
5	Shank flexion/extension – Thigh flexion/extension	Shank F/E – Thigh F/E

Table 5: Segments and segment actions forming the couplings assessed in this study.

From the phase portrait, a phase angle was determined by projecting a ray from the origin of the portrait to each successive data point making up the curve and calculating the angle of the vector from the right horizontal (Figure 8). Specifically, the phase angle calculation is described by

$$\phi(t) = tan^{-1}\left(\frac{\omega(t)}{\theta(t)}\right)$$

where ω is the angular velocity and θ is the segment angle at time point t within the movement cycle (Stergiou, 2004). Relative phase was calculated at each time point within the movement cycle by subtracting the phase angle obtained for the proximal segment from the phase angle for the distal segment, as described by

$$\varphi_{Rel}(t) = \phi_D(t) - \phi_P(t)$$

where ϕ_D is the phase angle of the distal segment and ϕ_P is the phase angle of the proximal segment at time point t (Stergiou, 2004).

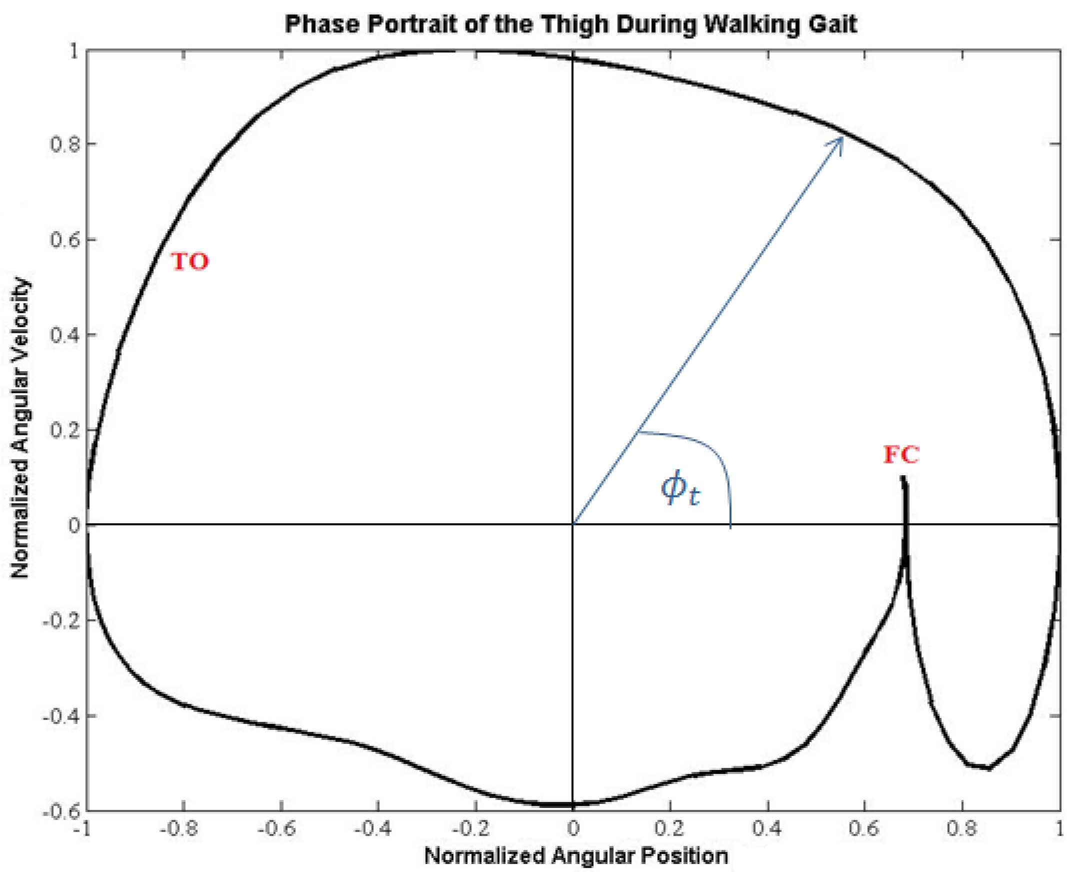

Figure 8: Sample phase portrait of the hip during walking. Normalized sagittal plane thigh angular position (abscissa) and angular velocity (ordinate) are represented on the axes. Progression of the thigh through state space begins at foot contact (FC) and continues clockwise past toe-off (TO) before returning to FC. The phase angle at a given time point [$\phi(t)$] is shown in blue.

Due to the directional nature of the relative phase measure, directional statistics were used to calculate relative phase means and standard deviations for each trial, at each time point over the specified movement intervals (Burgess-Limerick, Abernethy, & Neal, 1991; Lamoth, Beek, & Meijer, 2002). Due to redundancies in the interpretation of relative phase values within a 0-360 degree range, values were transformed to a 0-180 degree range (Burgess-Limerick, Abernethy, & Neal, 1991; Lamb & Stockl, 2014; Lamoth, Beek, & Meijer, 2002). Subsequently, measures of central tendency could be used for mean relative phase (MRP) computations. MRP was calculated to provide a single value describing the coordination between two segments over specified intervals of the walking and hopping cycles; larger MRP values indicate the segmental motion is more out of phase (i.e. independent or asynchronous) while smaller MRP values indicate more in phase motion (i.e. simultaneous or synchronous) (Burgess-Limerick, Abernethy, & Neal, 1993; Lamoth, Beek, & Meijer, 2002).

Relative phase standard deviation (RPD), a measure of the coordinated pattern dynamics, was calculated for each coupling over the specified intervals of the walking and hopping movement cycles. Larger RPD values indicate that the coordinated pattern is less stable or more flexible with respect to patterns with lower RPD values, which are more stable or less flexible (Hamill, Palmer, & Van Emmerik, 2012; Hamill et al., 1999). Assessing the coordination dynamics provide insight on the behavioral response to interacting constraints (e.g. functional impairments and/or ankle bracing) imposed on the system (Van Emmerik, Hamill, & McDermott, 2005).

Surface EMG signals obtained during over ground walking and single-leg hopping trials were used to assess neuromuscular activation of the gluteus medius, peroneus longus, tibialis anterior and gastrocnemius. Signals were lowpass filtered with a cutoff of 500 HZ at the hardware level and sampled at 1000 Hz. In addition, signals were band-pass filtered using a root-mean-square algorithm with a window of 50ms. EMG data collected within the two to four second window of the each manual muscle test were used to determine the average maximal voluntary isometric contraction (MVIC) activation. Once calculated, the average activity for intervals of the movement cycle were normalized relative to the MVIC for the respective muscle and represented as a percentage of MVIC. Percent average activation of each muscle group was calculated for each interval of interest within the gait and hopping cycles.

Statistical Analysis

Statistical analyses were conducted using SPSS software (version 18.0, SPSS Inc, Chicago, IL, USA) and an alpha level of statistical significance was set a priori at $\alpha \leq 0.05$. Shapiro-Wilk tests of normality were conducted to assess the distribution of data points for each dependent variable from each movement interval. To investigate differences in mean relative phase (MRP) and relative phase deviation (RPD), a 3 (group) X 2 (condition) X 5 (coupling) mixed ANOVA, with repeated measures on the within subjects factors, was employed using the MRP and RPD measures for each individual movement phase. To investigate differences in percent normalized sEMG activity (%MVIC), a 3 (group) X 2 (condition) X 4 (muscle) mixed ANOVA, with repeated measures on the within subjects factors, was employed using the percent normalized sEMG measure for each individual movement phase. The between subjects factor was group (Healthy, Coper, and Non-Coper) and the within subjects factors were condition and coupling. Post-hoc analyses were conducted if significant main effects or interactions were observed.

CHAPTER IV

RESULTS

The focus of this chapter is to chronicle the results obtained from methods detailed in Chapter III. Methods for this project were designed to address the following aims: 1) To assess if individuals with chronic ankle instability (Non-Coper) exhibit altered lower extremity coordination, coordination variability, and neuromuscular activity compared to Healthy and Coper groups during walking and single leg hopping. 2) To assess if ankle joint bracing influences coordination, coordination variability, and neuromuscular activity in Healthy, Coper, and Non-Coper groups during walking and single leg hopping. Data processing and statistical analyses were carried out following the completion of data collection procedures for all participants.

The results of this study will be presented in sections. First, demographic data will be presented to provide descriptive information of the sample populations for each group. Next, self-reported outcomes from each of the self-assessment surveys [Foot and Ankle Ability Measure (FAAM), Foot and Ankle Ability Measure – Sport (FAAM-S), and Ankle Instability Instrument (AII)] will be presented. Finally, results for each dependent variable [mean relative phase (MRP), relative phase deviation (RPD), and percent normalized surface electromyography (%MVIC)] will be reported with respect to intervals of single-leg hopping and over ground walking cycles.

Section 1: Demographics

Forty-eight individuals between the ages of 19-30 years participated in this study (Table 6). Each group (Healthy, Coper, and Non-Coper) included 16 individuals with an equal number of females (n = 10) and males (n = 6) in each group. Each participant reported partaking in a minimum of 90 minutes of physical activity per week. In addition, participants were free of neurological impairments, had not undergone major lower extremity surgery, and had no allergy to adhesives.

Group	N	Age (years)	Height (m)	Mass (kg)
Healthy	16	23.1 ± 1.9	1.69 ± 0.08	71.8 ± 12.6
Copers	16	22.4 ± 2.9	1.70 ± 0.08	76.4 ± 19.3
Non-Copers	16	23.3 ± 3.1	1.72 ± 0.09	77.8 ± 17.2

Table 6: Descriptive means and standard deviations (mean ± SD) for each group.

Section 2: Self-Reported Outcomes

Each participant completed three self-assessment surveys. The Foot and Ankle Ability Measure and Foot (FAAM) and Ankle Ability Measure – Sport (FAAM-S) were issued to determine level of function related to activities of daily living and sport respectively. Percentage scores and percent level of perceived function scores were obtained from both the FAAM and FAAM-S. Totaling the Likert scale values obtained from each question, dividing by the highest achievable total and multiplying by 100 determined the percentage score (higher percentage scores indicate higher self-reported function).

In addition, the Ankle Instability Instrument (AII) was issued to determine if ankle instability was present in participants. The AII consists of polar questions (i.e. participant selected either “Yes” or “No” for each question). Inclusion criteria for the Non-Coper group included a score of < %90 and < %85 on the FAAM and FAAM-S respectively, as well as three or more affirmative responses to questions on the AII. Results from the three self-assessment surveys employed for this study are presented in Table 7. As expected, for both the FAAM and

FAAM-S, the group mean percentage score was highest for the healthy participants, followed by the Coper and Non-Coper groups respectively. A similar trend was noted for FAAM and FAAM-perceived percentage score means across groups.

	FAAM		**FAAM-S**		**AII**
	Score (%)	*Perceived (%)*	*Score (%)*	*Perceived (%)*	*Y*
Healthy	99.6 ± 1.2	98.8 ± 2.9	96.5 ± 6.9	96.3 ± 5.9	1 ± 2
Copers	97.6 ± 7.1	99.1 ± 2.0	93.8 ± 11.9	96.0 ± 5.2	3 ± 1
Non-Copers	85.4 ± 5.6	88.9 ± 6.6	68.4 ± 9.6	78.8 ± 11.1	7 ± 1

Table 7: Means and standard deviations (mean ± SD) of outcomes for each self-assessment survey across groups. Score – *Outcome percentage score determined from survey responses.* Perceived – *Self-reported percent level of function*. Y – *Indicates the number of affirmative answers to questions on the Ankle Instability Instrument.*

Section 3: Dependent Measures

Coordination, coordination variability, and neuromuscular activation were assessed using measures of MRP, RPD, and %MVIC respectively. Measures of coordination and coordination variability for five lower extremity segment couplings were included in the analysis. In addition, activation of the peroneus longus, tibialis anterior, lateral gastrocnemius, and gluteus medius was also assessed in the analysis.

Shapiro-Wilk tests of normality were conducted to assess the distribution of MRP, RPD, and %MVIC data obtained from each movement interval. Assumptions of normality were violated for MRP, RPD, and %MVIC across movement intervals; as a result, all data underwent logarithmic transformations and were retested for normality. Follow up tests of normality indicated that the transformed data were normally distributed. Subsequent statistical tests were conducted using the transformed dataset.

In order to test the effects of bracing and group on lower extremity coordination, coordination variability, and neuromuscular activity, a factorial analysis was conducted for each dependent measure. Effects of group, condition, and coupling on MRP and RPD were assessed using a 3(Group) X 2(Condition) X 5(Coupling) repeated measures ANOVA. Effects of group, condition, and muscle on %MVIC activation were assessed using a 3(Group) X 2(Condition) X 4(Muscle) repeated measures ANOVA.

Main effects of coupling or muscle were observed for each dependent measure at each movement phase; observed differences in coordination, coordination variability, and muscle activation between segment couplings/muscles do not directly relate to the specific aims of this study, but served as a validation of expected incongruity between coupling/muscle behaviors. Therefore, the reader will be directed to Appendix C to find the statistical results related to main effects of coupling and muscle. Similarly, condition by coupling and condition by muscle interactions were noted for some dependent measures across several movement phases; observed condition by coupling or condition by muscle interactions do not directly relate to the specific aims of this study, but indicate a response discrepancy of couplings or muscles across conditions for a given dependent measure. Such observations of condition by coupling or condition by muscle interactions could provide insight into the systemic effects of bracing. However, in order to preserve focus on results related to the specific aims, post hoc results related to condition by coupling and condition by muscle interactions are presented in Appendix D.

Related to the specific aims, main effects of condition served as a basis for comparison between dependent measure values observed during trials performed with and without an ankle brace. In addition, group comparisons were made in order to assess any differences in dependent measure values between individuals with and without chronic ankle instability. Post-hoc testing was conducted if significant main effects or interactions were observed. Results of the factorial analyses and post hoc tests relevant to the specific aims will be presented in the following sections.

Section 4: Coordination

Single-Leg Hop Coordination

Results pertaining to the coordination of lower extremity segment couplings during single-leg hopping are presented in this section. MRP was calculated for the landing, propulsion, and flight phases of the single-leg hop cycle (Table 8); results will be presented for each phase of the hop cycle. Results that were not statistically significant are presented in Appendix B.

Landing Coordination

MRP was employed as a dependent measure to assess the coordination of lower extremity segment couplings. Factor analysis of MRP during hop landing (HL) revealed a significant main effect of coupling [$F(1.68, 75.98) = 135.44$, $p < .001$, partial $\eta^2 = .75$]. Pairwise comparisons related to the coupling main effect can be found in Appendix C (Table C-1). Main effects of condition and group were not statistically significant. In addition, no significant interactions were noted.

Propulsion Coordination

Analysis of coordination during the hop propulsion phase indicated a significant main effect of coupling [$F(1.99, 89.92) = 287.21$, $p < .001$, partial $\eta^2 = .86$]. Pairwise comparisons related to the coupling main effect can be found in Appendix C (Table C-2). Main effects of condition and group were not statistically significant. In addition, no significant interactions were noted.

Flight Coordination

Assessment of coordination during the flight phase of single-leg hopping revealed a significant main effect of coupling [$F(2.28, 102.59) = 62.23$, $p < .001$, partial $\eta^2 = .58$], as well as significant condition by coupling interaction [$F(2.69, 121.08) = 6.21$, $p = .001$, partial $\eta^2 = .12$], and condition by coupling by group interaction [$F(8, 45) = 3.45$, $p = .001$, partial $\eta^2 = .13$]. To address the three-way interaction, a follow up two-way repeated measures ANOVA was conducted for each group. For the healthy group, results from the two-way ANOVA revealed a significant simple main effect of coupling [$F(1.77, 26.66) = 18.92$, $p < .001$]. Pairwise comparisons related to the coupling main effect for the healthy group can be found in Appendix C (Table C-3).

For the Coper group, results from the two-way ANOVA revealed a significant simple main effect of coupling [$F(4, 60) = 16.32$, $p < .001$], and a significant condition by coupling simple interaction [$F(2.30, 34.51) = 3.94$, $p = .024$]. To assess the simple effect of bracing between couplings, first, difference scores were calculated for each coupling by subtracting the mean MRP score for the no brace condition from the mean MRP score of the brace condition. Next, post hoc comparisons using the LSD test were made to examine differences between couplings. Overall, differences between three coupling pairs were statistically significant ($p < .05$). Results from this post hoc analysis can be found in Appendix D (Table D-1).

To assess the simple effect of bracing on the coordination of each individual coupling for the Coper group, five one-sample t-tests were conducted. Results from the t-tests revealed that two segment couplings were significantly impacted by bracing during the hop flight phase. Foot DF/PF – shank F/E coordination was significantly more in phase (i.e. foot and shank sagittal plane segment motion was more simultaneous) during hop flight phase of the brace condition compared to no brace [$t(5) = -2.43$, $p = .028$]. Contrarily, foot DF/PF – thigh F/E coordination was significantly more out of phase (i.e. foot and thigh sagittal plane segment motion was more independent) during hop flight phase of the brace condition compared to no brace [t(5) = 2.20, p =.044] (Figure 9).

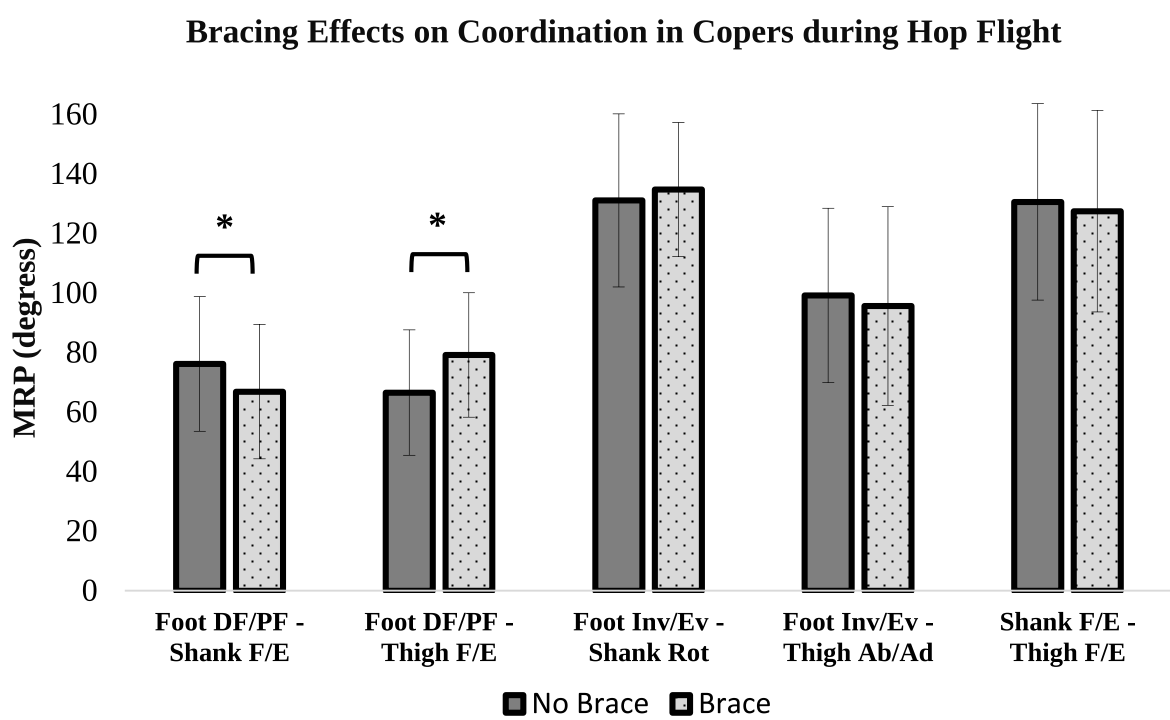

Figure 9: Bracing effects on coordination of lower extremity couplings in Copers during the hop flight phase of the single-leg hopping cycle (***** p <.05).

For the Non-Coper group, results from the two-way ANOVA revealed a significant simple main effect of coupling [F(1.64, 24.66) = 31.25, p < .001] and a significant condition by coupling simple interaction [F(2.55, 38.30) = 10.33, p < .001]. Simple effects of bracing between couplings were assessed by calculating the difference score for each coupling by subtracting the mean MRP score for the no brace condition from the mean MRP score of the brace condition. Post hoc comparisons using the LSD test were made to examine differences between couplings. Overall, differences between six coupling pairs were statistically significant (p < .05). Results from this post hoc analysis can be found in Appendix D (Table D-2).

Simple effects of bracing on the coordination of each individual coupling for the Non-Coper group were assessed using one-sample t-tests. Similar to the Coper group, t-test results from the Non-Coper group revealed that foot DF/PF – shank F/E coordination became more in phase (i.e. simultaneous) [t(5) = -3.57, p =.003], and foot DF/PF – thigh F/E coordination became more out of phase (i.e. independent) [t(5) = 2.92, p =.011] during hop flight when the ankle was braced compared to not braced (Figure 10).

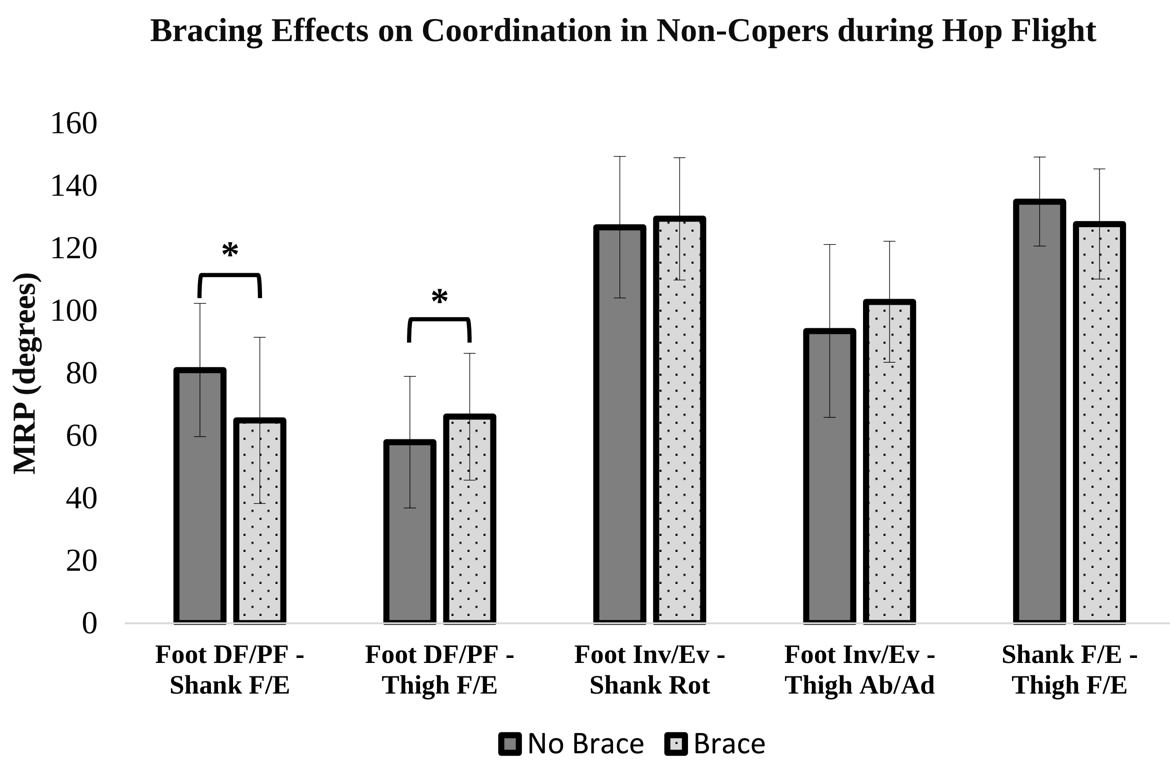

Figure 10: Bracing effects on coordination of lower extremity couplings in Non-Copers during the hop flight phase of the single-leg hopping cycle (***** p <.05)

Single-Leg Hopping MRP Means and Standard Deviations (Mean ± SD)

Coupling	**Healthy** *NB*	**Healthy** *B*	**Coper** *NB*	**Coper** *B*	**Non-Coper** *NB*	**Non-Coper** *B*
Hop Landing						
1	163.86 ± 12.27	166.56 ± 8.87	159.86 ± 11.76	159.12 ± 12.32	161.04 ± 14.52	159.52 ± 15.75
2	44.37 ± 31.61	38.61 ± 34.41	46.69 ± 30.15	46.97 ± 32.92	46.24 ± 21.67	45.77 ± 26.91
3	149.61 ± 21.16	153.00 ± 17.94	153.13 ± 14.14	153.57 ± 14.12	158.10 ± 13.77	151.27 ± 19.37
4	94.36 ± 45.53	88.29 ± 48.16	113.94 ± 39.50	118.14 ± 40.26	108.89 ± 37.46	117.80 ± 35.23
5	147.39 ± 23.37	148.78 ± 28.05	145.28 ± 30.21	144.20 ± 37.72	147.63 ± 8.33	148.47 ± 10.17
Hop Propulsion						
1	158.90 ± 27.45	158.34 ± 29.03	161.87 ± 8.23	162.81 ± 7.58	162.77 ± 8.90	162.51 ± 7.90
2	37.78 ± 37.68	36.26 ± 37.19	29.70 ± 7.60	29.59 ± 7.38	31.66 ± 11.31	29.94 ± 9.42
3	163.40 ± 10.73	163.03 ± 8.76	166.25 ± 7.12	162.42 ± 11.55	165.77 ± 8.01	160.97 ± 12.53
4	105.88 ± 45.38	102.01 ± 49.27	96.07 ± 46.46	100.70 ± 43.01	113.91 ± 37.60	122.71 ± 38.12
5	165.18 ± 5.52	167.29 ± 5.77	167.65 ± 4.83	167.85 ± 6.78	165.09 ± 9.51	166.58 ± 7.87
Hop Flight						
1	76.67 ± 22.41	81.29 ± 31.65	76.20***** ± 22.61	66.87***** ± 22.55	81.02***** ± 21.33	64.88***** ± 26.62
2	65.55 ± 25.43	67.21 ± 31.49	66.53***** ± 21.07	79.21***** ± 20.90	57.94***** ± 21.07	66.08***** ± 20.26
3	135.49 ± 16.98	126.20 ± 25.48	131.12 ± 29.05	134.72 ± 22.53	126.68 ± 22.65	129.40 ± 19.58
4	106.84 ± 21.89	102.32 ± 28.42	99.19 ± 29.27	95.62 ± 33.41	93.50 ± 27.68	102.82 ± 19.39
5	131.28 ± 20.68	131.97 ± 20.39	130.63 ± 33.00	127.50 ± 33.83	134.90 ± 14.26	127.71 ± 17.61

Table 8: Group mean relative phase (MRP) means and standard deviations for each phase of the single-leg hopping cycle across groups. 1 = *Foot dorsiflexion/plantarflexion – Shank flexion/extension*, 2 = *Foot dorsiflexion/plantarflexion – Thigh flexion/extension*, 3 = *Foot inversion/eversion – Shank internal/external rotation*, 4 = *Foot inversion/eversion – Thigh abduction/adduction*, 5 = *Shank flexion/extension – Thigh flexion/extension;* NB = *No brace condition*, B = *Brace condition*
***** = *p* < *.05 for differences between conditions*

Walking Coordination

Results pertaining to the coordination of lower extremity segment couplings during walking are presented in this section. MRP was calculated for foot contact to mid-stance, toe-off to mid-swing and terminal swing phases of the walking cycle (Table 9); results will be presented for each phase of the walking cycle. Results that were not statistically significant are presented in Appendix B.

Stance Coordination

Assessment of coordination from foot contact to mid-stance of over ground walking revealed a significant main effect of coupling [$F(1.16, 52.46) = 656.56$, $p < .001$, partial $\eta^2 = .93$], as well as a significant condition by coupling interaction [$F(1.39, 62.86) = 11.00$, $p = .001$, partial $\eta^2 = .19$]. Simple effects of bracing between couplings were assessed by calculating the difference score for each coupling by subtracting the mean MRP score for the no brace condition from the mean MRP score of the brace condition. Post hoc comparisons using the LSD test were made to examine differences between couplings. Overall, differences between seven coupling pairs were statistically significant ($p < .05$). Results from this post hoc analysis can be found in Appendix D (Table D-4).

Simple effects of bracing on the coordination of each individual coupling, during the walking stance phase interval, were assessed using one-sample t-tests. Results from the t-tests indicated that bracing influenced coordination of two couplings during foot contact to mid-stance of walking. Foot DF/PF – shank F/E coordination became significantly more in phase (i.e. segmental relative motion was more simultaneous) during the braced condition compared to the no brace condition [$t(47) = -7.21$, $p< .001$]. Contrarily, shank F/E – thigh F/E coordination became significantly more out of phase (i.e. segmental relative motion was more independent) during the braced condition compared to the no brace condition [t(47) = 2.48, p = .017] (Figure 11).

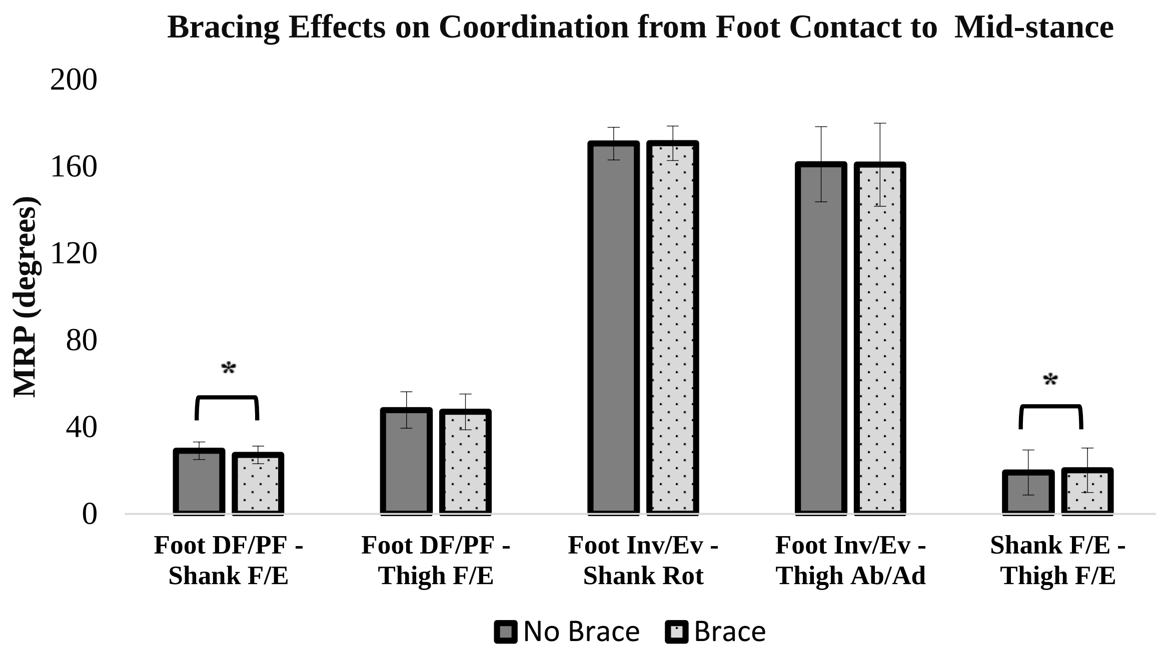

Figure 11: Bracing effects on coordination of lower extremity couplings from foot contact to mid-stance phase of the gait cycle (***** p <.05).

Initial Swing Coordination

Assessment of coordination during the first 50% of swing phase within the walking gait cycle revealed significant main effects of condition [F(1, 45) = 5.67, p = .021, partial η^2 = .11], and coupling [F(2.49, 112.25) = 453.76, p < .001, partial η^2 = .91], as well as significant condition by coupling interaction [F(2.23, 100.64) = 37.03, p < .001, partial η^2 = .45]. Simple effects of bracing between couplings were assessed by calculating the difference score for each coupling by subtracting the mean MRP score for the no brace condition from the mean MRP score of the brace condition. Post hoc comparisons using the LSD test were made to examine differences between couplings. Overall, differences between eight coupling pairs were statistically significant ($p < .05$). Results from this post hoc analysis can be found in Appendix D (Table D-5). Simple effects of bracing on the coordination of each individual coupling, during the toe-off to mid-swing interval, were assessed using one-sample t-tests.

Results from the t-tests indicated that bracing significantly influenced coordination of four couplings during the first 50% of swing phase of the walking cycle. Foot DF/PF – shank F/E coordination became significantly more in phase (i.e. segmental relative motion was more simultaneous) during the braced condition compared to the no brace condition [$t(47) = -12.30$, $p< .001$]. The opposite effect on coordination (i.e. significantly more out of phase or independent segmental relative motion) was observed during the braced condition for the remaining three couplings: Foot DF/PF – thigh F/E [$t(47) = 7.70$, $p< .001$], foot IN/EV – shank ROT [$t(47) = 3.17$, $p = .003$], and foot IN/EV – thigh AB/AD [$t(47) = 5.09$, $p < .001$] (Figure 12).

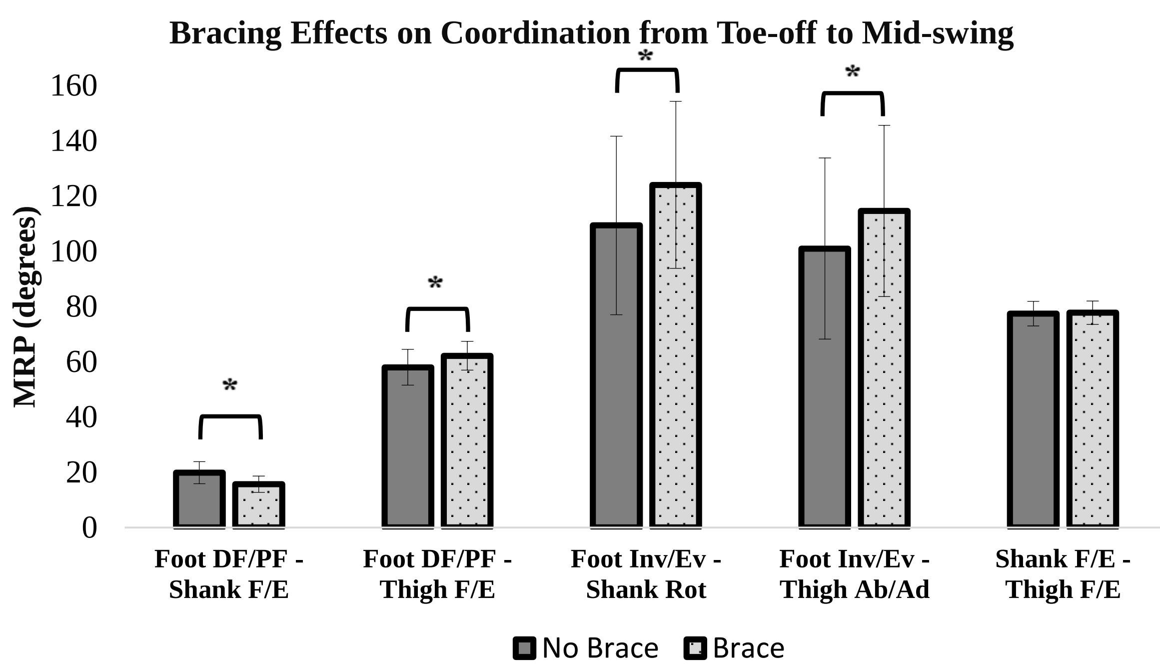

Figure 12: Bracing effects on coordination of lower extremity couplings from toe-off to mid-swing of the gait cycle (***** $p <.05$).

Terminal Swing Coordination

Assessment of coordination during terminal swing phase of over ground walking revealed a significant main effect of coupling [$F(2.11, 95.06) = 754.42$, $p < .001$, partial $\eta^2 = .94$], as well as a significant condition by coupling interaction [$F(2.49, 112.39) = 7.90$, $p < .001$, partial $\eta^2 = .14$]. Simple effects of bracing between couplings were assessed by calculating the difference score for each coupling by subtracting the mean MRP score for the no brace condition from the mean MRP score of the brace condition. Post hoc comparisons using the LSD test were made to examine differences between couplings. Overall, differences between eight coupling pairs were statistically significant ($p < .05$). Results from this post hoc analysis can be found in Appendix D (Table D-6).

Simple effects of bracing on the coordination of each individual coupling, during terminal swing phase were assessed using one-sample t-tests. Results from the t-tests indicated that bracing influenced coordination of two couplings during terminal swing phase of walking. Foot DF/PF – shank F/E coordination became significantly more in phase (i.e. segmental relative motion was more simultaneous) during the braced condition compared to the no brace condition [$t(47) = -3.56$, $p = .001$]. Contrarily, foot IN/EV – thigh AB/AD coordination became significantly more out of phase (i.e. segmental relative motion was more independent) during the braced condition compared to the no brace condition [$t(47) = 2.03$, $p = .047$] (Figure 13).

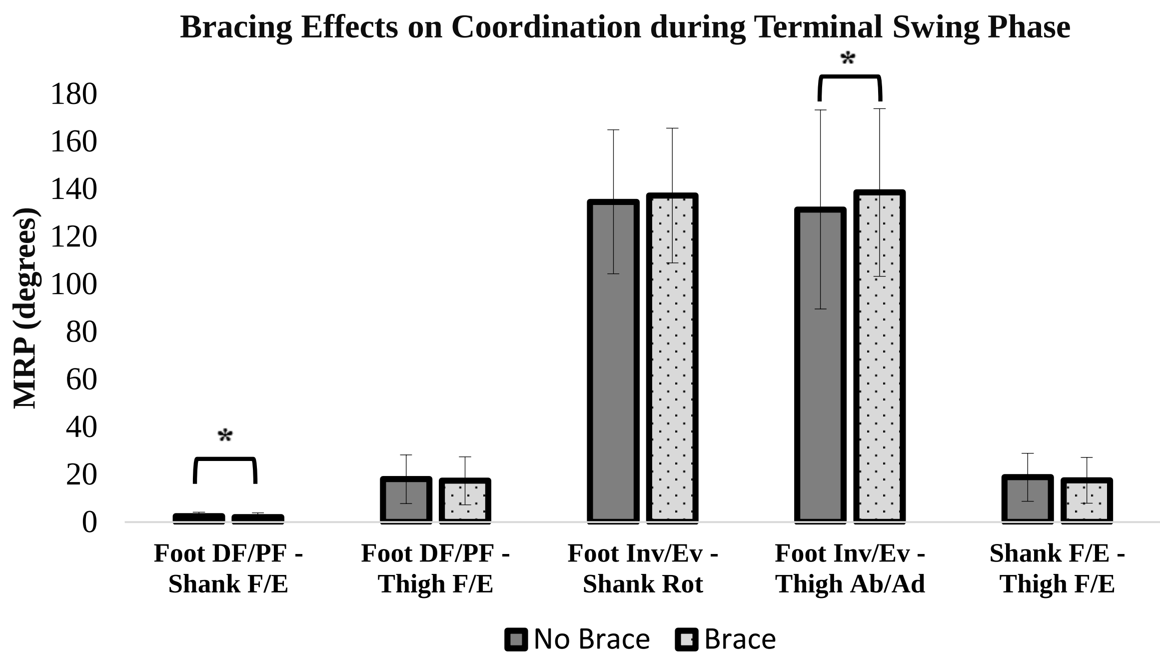

Figure 13: Bracing effects on coordination of lower extremity couplings during the terminal swing phase of the gait cycle (***** $p < .05$).

Walking MRP Means and Standard Deviations (Mean ± SD)

Coupling	Foot Contact to Mid-stance		Toe-off to Mid-swing		Terminal Swing	
	NB	*B*	*NB*	*B*	*NB*	*B*
1	28.99* ± 4.05	27.12* ± 4.09	19.88* ± 4.00	15.68* ± 2.99	2.53* ± 1.56	2.08* ± 1.76
2	47.83 ± 8.35	46.97 ± 8.25	58.02* ± 6.51	62.19* ± 5.22	18.02 ± 10.23	17.35 ± 10.14
3	170.53 ± 7.48	170.70 ± 7.97	109.37* ± 32.30	124.03* ± 30.23	134.59 ± 30.26	137.28 ± 28.35
4	160.99 ± 17.26	160.86 ± 19.14	101.00* ± 32.80	114.64* ± 30.96	131.40* ± 41.85	138.57* ± 35.22
5	19.01* ± 10.34	20.02* ± 10.22	77.43 ± 4.41	77.79 ± 4.25	18.83 ± 10.05	17.53 ± 9.60

Table 9: Mean relative phase (MRP) means and standard deviations collapsed across groups for each phase of the walking gait cycle. 1 = *Foot dorsiflexion/plantarflexion – Shank flexion/extension*, 2 = *Foot dorsiflexion/plantarflexion – Thigh flexion/extension*, 3 = *Foot inversion/eversion – Shank internal/external rotation*, 4 = *Foot inversion/eversion – Thigh abduction/adduction*, 5 = *Shank flexion/extension – Thigh flexion/extension;* NB = *No brace condition*, B = *Brace condition*; ***** = *$p < .05$ for differences between conditions*

Section 5: Coordination Variability

Single-Leg Hop Coordination Variability

Results pertaining to the coordination variability of lower extremity segment couplings during single-leg hopping are presented in this section. RPD was calculated for the landing, propulsion, and flight phases of the single-leg hop cycle (Table 10); results will be presented for each phase of the hop cycle. Results that were not statistically significant are presented in Appendix B.

Landing Coordination Variability

RPD was implemented as a dependent measure to assess the variability of lower extremity segment couplings. For the landing phase of single-leg hopping, initial factor analysis revealed a significant main effect of condition [$F(1, 45) = 5.43$, $p = .024$, partial $\eta^2 = .10$]. Post hoc comparisons using the LSD test indicated that variability across segment couplings was significantly lower in the brace condition compared to the no brace condition. Paired samples t-tests were conducted to the effect of bracing on the coordination variability of each lower extremity coupling during single-leg hop landing. Results of the t-tests indicate that coordination variability of foot IN/EV – shank ROT significantly decreased during the braced condition of the hop landing phase [$t(47) = 2.042$, $p = .047$] (Figure 14). A main effect of coupling was also noted during the hop landing phase [$F(1.83, 82.52) = 80.23$, $p < .001$, partial $\eta^2 = .64$]. Pairwise comparisons related to the coupling main effect can be found in Appendix C (Table C-4). The group main effect on coordination variability during hop landing was not statistically significant. In addition, no significant interactions were noted.

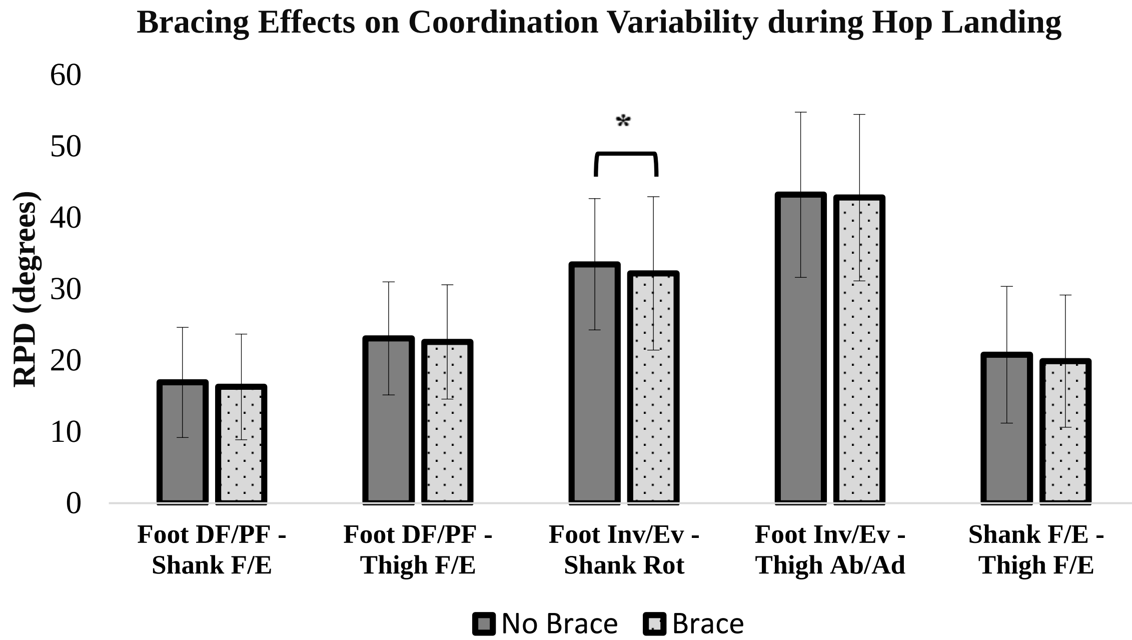

Figure 14: Bracing effects on coordination variability of lower extremity couplings during the hop landing phase of the single-leg hopping cycle (***** $p < .05$).

Propulsion Coordination Variability

Analysis of coordination variability from hop propulsion indicated a significant main effect of coupling [$F(2.82, 127.22) = 114.70$, $p < .001$, partial $\eta^2 = .71$]. Pairwise comparisons related to the coupling main effect can be found in Appendix C (Table C-5). Main effects of condition and group were not statistically significant. In addition, no significant interactions were noted.

Flight Coordination Variability

Assessment of coordination variability during the flight phase of single-leg hopping revealed a significant main effect of coupling [$F(2.92, 131.39) = 44.74$, $p < .001$, partial $\eta^2 = .49$], as well as a significant condition by coupling interaction [$F(3.30, 148.80) = 2.88$, $p = .033$, partial $\eta^2 = .06$]. Post hoc comparisons of RPD mean difference scores between couplings were made using the LSD test. Overall, differences between two coupling pairs were statistically significant (p < .05). Results from this post hoc analysis can be found in Appendix D (Table D-3).

Simple main effects of bracing on the coordination variability of each individual coupling were assessed using one-sample t-tests. Results from the t-tests indicated that coordination variability of one segment coupling was significantly impacted by condition during hop flight. Specifically, during hop flight foot IN/EV – shank ROT coordination variability increased during braced conditions [t(47) = 3.10, p =.003] (Figure 15).

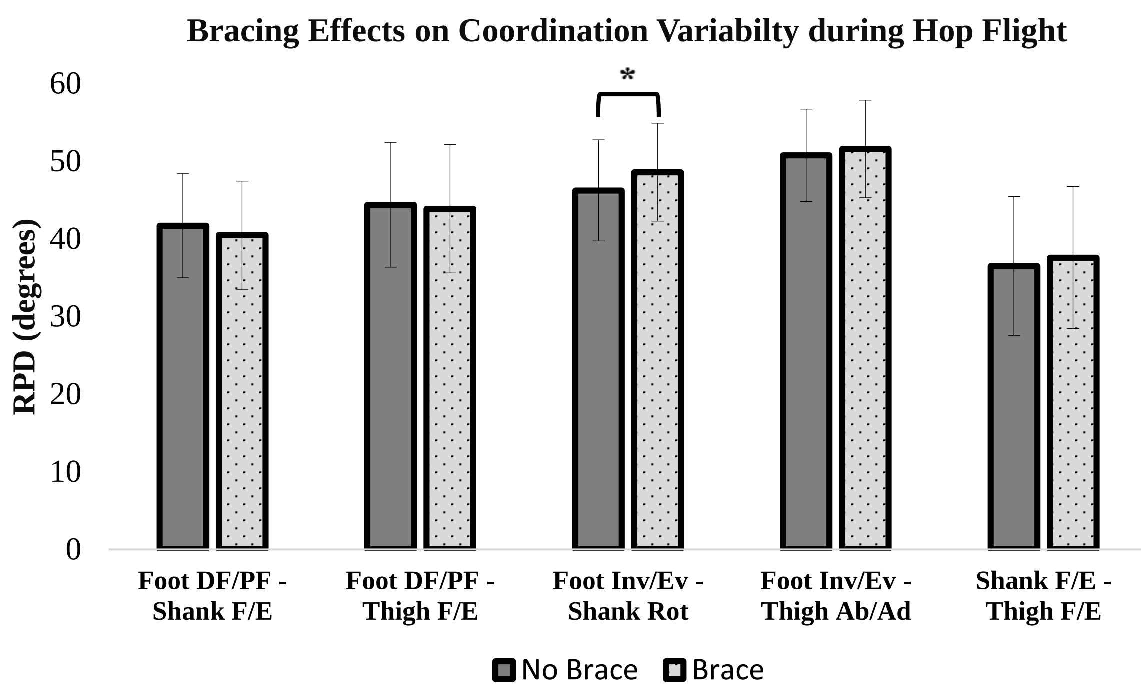

Figure 15: Bracing effects on coordination variability of lower extremity couplings during the hop flight phase of the single-leg hopping cycle (***** p <.05).

Single-Leg Hopping RPD Means and Standard Deviations (Mean ± SD)

Coupling	Hop Landing *NB*	Hop Landing *B*	Hop Propulsion *NB*	Hop Propulsion *B*	Hop Flight *NB*	Hop Flight *B*
1	16.92 ± 7.71	16.29 ± 7.42	19.03 ± 11.51	20.08 ± 12.09	41.66 ± 6.69	40.45 ± 6.97
2	23.09 ± 7.91	22.59 ± 8.00	24.18 ± 11.54	25.02 ± 12.50	44.36 ± 8.01	43.84 ± 8.27
3	33.47***** ± 9.21	32.20***** ± 10.73	21.72 ± 10.00	22.44 ± 10.06	46.21***** ± 6.51	48.57***** ± 6.29
4	43.22 ± 11.58	42.81 ± 11.68	42.44 ± 10.56	42.78 ± 12.05	50.73 ± 5.97	51.57 ± 6.28
5	20.78 ± 9.58	19.90 ± 9.27	12.74 ± 5.44	12.48 ± 5.63	36.47 ± 8.96	37.56 ± 9.16

Table 10: Relative phase deviation (RPD) means and standard deviations collapsed across groups for each phase of the single-leg hopping cycle. 1 = *Foot dorsiflexion/plantarflexion – Shank flexion/extension*, 2 = *Foot dorsiflexion/plantarflexion – Thigh flexion/extension*, 3 = *Foot inversion/eversion – Shank internal/external rotation*, 4 = *Foot inversion/eversion – Thigh abduction/adduction*, 5 = *Shank flexion/extension – Thigh flexion/extension;* NB = *No brace condition*, B = *Brace condition*
***** = *$p < .05$ for differences between conditions*

Walking Coordination Variability

Results pertaining to the coordination variability of lower extremity segment couplings during walking are presented in this section. RPD was calculated for foot contact to mid-stance, toe-off to mid-swing and terminal swing phases of the walking cycle (Table 11); results will be presented for each phase of the walking cycle. Results that were not statistically significant are presented in Appendix B.

Stance Coordination Variability

Analysis of coordination variability from foot contact to mid-stance phase indicated a significant main effect of coupling [$F(2.40, 108.36) = 216.56$, $p < .001$, partial $\eta^2 = .82$]. Pairwise comparisons related to the coupling main effect can be found in Appendix C (Table C-9). Main effects of condition and group were not statistically significant. In addition, no significant interactions were noted.

Initial Swing Coordination Variability

Assessment of coordination variability during the first 50% of swing phase within the walking gait cycle revealed significant main effects of condition [$F(1, 45) = 108.32$, $p < .001$, partial $\eta^2 = .70$], and coupling [$F(2.89, 130.15) = 688.40$, $p < .001$, partial $\eta^2 = .93$], as well as significant condition by coupling interaction [$F(2.90, 130.49) = 83.56$, $p < .001$, partial $\eta^2 = .65$]. Simple effects of bracing between couplings were assessed by calculating the difference score for each coupling by subtracting the mean RPD score for the no brace condition from the mean RPD score of the brace condition. Post hoc comparisons using the LSD test were made to examine differences between couplings. Overall, differences between all ten coupling pairs were statistically significant ($p < .05$). Results from this post hoc analysis can be found in Appendix D (Table D-7).

Simple effects of bracing on the coordination variability of each individual coupling, during the initial swing phase interval, were assessed using one-sample t-tests. Results from the t-tests indicated that bracing significantly influenced coordination variability of four couplings during the first 50% of swing phase of the walking cycle. Bracing significantly diminished coordination variability in three segmental couplings: Foot DF/PF – shank F/E [$t(47) = -16.08$, $p < .001$], foot DF/PF – thigh F/E [$t(47) = -7.23$, $p < .001$], and shank F/E – thigh F/E[$t(47) = -3.69$, $p = .001$]. Contrarily, bracing significantly increased coordination variability in the foot IN/EV – shank ROT coupling [$t(47) = 2.94$, $p = .005$] (Figure 16).

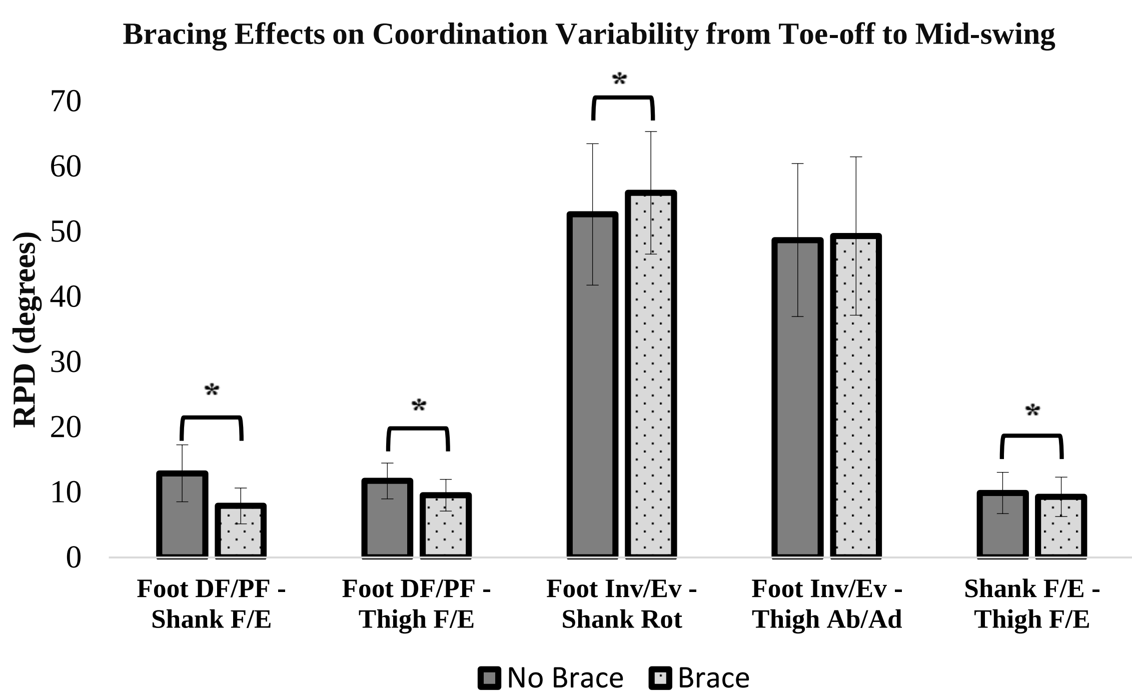

Figure 16: Bracing effects on coordination variability of lower extremity couplings from toe-off to mid-swing of the gait cycle (***** p <.05).

Terminal Swing Coordination Variability

Assessment of coordination variability during terminal swing phase within the walking gait cycle revealed significant main effects of condition [$F(1, 45) = 10.08$, $p = .003$, partial $\eta^2 = .18$], and coupling [$F(2.42, 109.14) = 209.79$, $p < .001$, partial $\eta^2 = .82$], as well as significant condition by coupling interaction [$F(2.57, 115.94) = 9.49$, $p < .001$, partial $\eta^2 = .17$]. Simple effects of bracing between couplings were assessed by calculating the difference score for each coupling by subtracting the mean RPD score for the no brace condition from the mean RPD score of the brace condition. Post hoc comparisons using the LSD test were made to examine differences between couplings. Overall, differences between eight coupling pairs were statistically significant ($p < .05$). Results from this post hoc analysis can be found in Appendix D (Table D-8).

Simple effects of bracing on the coordination variability of each individual coupling, during terminal swing phase, were assessed using one-sample t-tests. Results from the t-tests indicated that bracing significantly diminished coordination variability of the foot DF/PF – shank F/E [$t(47) = -4.99$, $p < .001$], and foot IN/EV – shank ROT [$t(47) = -2.20$, $p = .032$] couplings during terminal swing phase of the walking cycle (Figure 17).

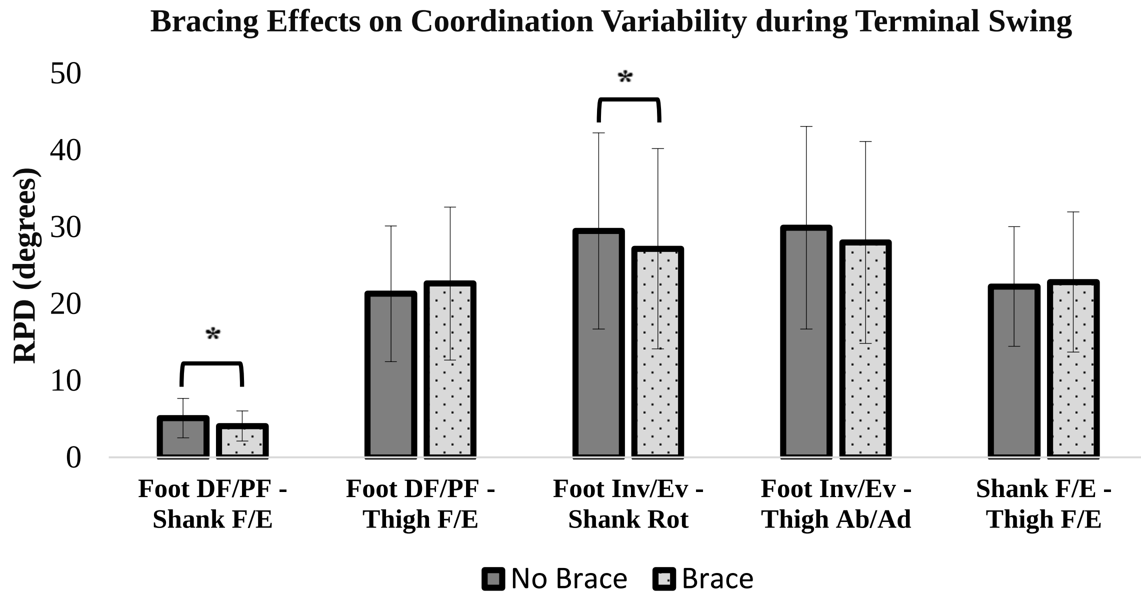

Figure 17: Bracing effects on coordination variability of lower extremity couplings during the terminal swing phase of the gait cycle (***** p <.05).

Walking RPD Means and Standard Deviations (Mean ± SD)

Coupling	**Foot Contact to Mid-stance**		**Toe-off to Mid-swing**		**Terminal Swing**	
	NB	*B*	*NB*	*B*	*NB*	*B*
1	15.64 ± 2.78	15.30 ± 2.61	12.89***** ± 4.36	7.90***** ± 2.75	5.11***** ± 2.57	4.09***** ± 1.97
2	44.03 ± 2.90	44.34 ± 2.63	11.72***** ± 2.76	9.51***** ± 2.43	21.30 ± 8.85	22.65 ± 9.95
3	31.33 ± 7.71	31.06 ± 9.08	52.63***** ± 10.84	55.94***** ± 9.41	29.48***** ± 12.76	27.16***** ± 13.04
4	38.50 ± 12.00	38.82 ± 13.32	48.69 ± 11.74	49.30 ± 12.15	29.90 ± 13.20	28.00 ± 13.14
5	37.64 ± 3.53	38.07 ± 3.06	9.86***** ± 3.18	9.28***** ± 3.04	22.26 ± 7.80	22.84 ± 9.12

Table 11: Relative phase deviation (RPD) means and standard deviations collapsed across groups for each phase of the walking gait cycle. 1 = *Foot dorsiflexion/plantarflexion – Shank flexion/extension*, 2 = *Foot dorsiflexion/plantarflexion – Thigh flexion/extension*, 3 = *Foot inversion/eversion – Shank internal/external rotation*, 4 = *Foot inversion/eversion – Thigh abduction/adduction*, 5 = *Shank flexion/extension – Thigh flexion/extension;* NB = *No brace condition*, B = *Brace condition*
***** = *p < .05 for differences between conditions*

Section 6: Electromyography

Single-Leg Hop Electromyography

Results pertaining to the neuromuscular activation of lower extremity segment couplings during single-leg hopping are presented in this section. %MVIC was calculated for the landing, propulsion, and flight phases of the single-leg hop cycle (Table 12); results will be presented for each phase of the hop cycle. Results that were not statistically significant are presented in Appendix D.

Landing Electromyography

%MVIC was employed as a dependent measure to assess the average activation of lower extremity muscles. During the landing phase of single-leg hopping, a significant main effect of muscle [$F(2.61, 117.62) = 163.59$, $p < .001$, partial $\eta^2 = .78$] was observed. Pairwise comparisons related to the muscle main effect can be found in Appendix C (Table C-6). Main effects of condition and group were not statistically significant. In addition, no significant interactions were noted.

Propulsion Electromyography

Analysis of average neuromuscular activation from hop propulsion indicated a significant main effect of muscle [$F(3, 135) = 73.20$, $p < .001$, partial $\eta^2 = .61$]. Pairwise comparisons related to the muscle main effect can be found in Appendix C (Table C-7). Main effects of condition and group were not statistically significant. In addition, no significant interactions were noted.

Flight Electromyography

Assessment of neuromuscular activation during flight phase of single-leg hopping revealed a significant main effect of muscle [$F(3, 135) = 70.52$, $p < .001$, partial $\eta^2 = .61$]. Pairwise comparisons related to the muscle main effect can be found in Appendix C (Table C-8).

Main effects of condition and group were not statistically significant. In addition, no significant interactions were noted

Single-Leg Hopping % MVIC Means and Standard Deviations (Mean ± SD)

	Hop Landing		**Hop Propulsion**		**Hop Flight**	
Muscle	*NB*	*B*	*NB*	*B*	*NB*	*B*
1	143.63 ± 81.48	139.84 ± 75.55	63.54 ± 48.07	63.67 ± 45.25	82.96 ± 51.15	81.04 ± 47.53
2	37.79 ± 30.79	37.25 ± 27.41	35.86 ± 18.30	35.93 ± 17.53	40.44 ± 22.30	40.08 ± 21.52
3	247.68 ± 144.05	248.81 ± 139.72	87.32 ± 46.73	92.06 ± 48.57	135.23 ± 69.76	137.27 ± 74.48
4	77.58 ± 31.79	80.14 ± 32.01	32.21 ± 16.60	33.32 ± 17.37	56.59 ± 26.59	57.45 ± 28.61

Table 12: Percent normalized electromyography (% MVIC) means and standard deviations collapsed across groups for each phase of the single-leg hopping cycle. 1 = *Peroneus Longus*, 2 = *Tibialis Anterior*, 3 = *Lateral Gastrocnemius*, 4 = *Gluteus Medius*; *NB* = No brace condition, *B* = Brace condition.

Walking Electromyography

Results pertaining to the neuromuscular activation of lower extremity segment couplings during walking are presented in this section. %MVIC was calculated for foot contact to mid-stance, toe-off to mid-swing and terminal swing phases of the walking cycle (Table 13); results will be presented for each phase of the walking cycle. Results that were not statistically significant are presented in Appendix D.

Stance Electromyography

During foot contact to mid-stance, significant main effects of condition [$F(1, 45) = 26.83$, $p < .001$, partial $\eta^2 = .37$] and muscle [$F(2.40, 108.04) = 11.18$, $p < .001$, partial $\eta^2 = .19$] on average neuromuscular activation were noted. No significant main effects of group or significant interactions were observed. Pairwise comparisons related to the muscle main effect can be found in Appendix C (Table C-10). Post hoc comparison using the LSD test was made to examine the main effects of bracing on average neuromuscular activation. Results indicated that bracing significantly decreased neuromuscular activation compared to no brace conditions. Paired-samples t-tests were conducted to determine which muscles were significantly impacted by bracing. Results from the t-tests indicated that activation of the peroneus longus [$t(47) = 2.57$, $p = .013$], tibialis anterior [$t(47) = 5.85$, $p < .001$], and lateral gastrocnemius [$t(47) = 4.03$, $p < .001$] activation significantly decreased during the braced condition compared to the no brace condition (Figure 18).

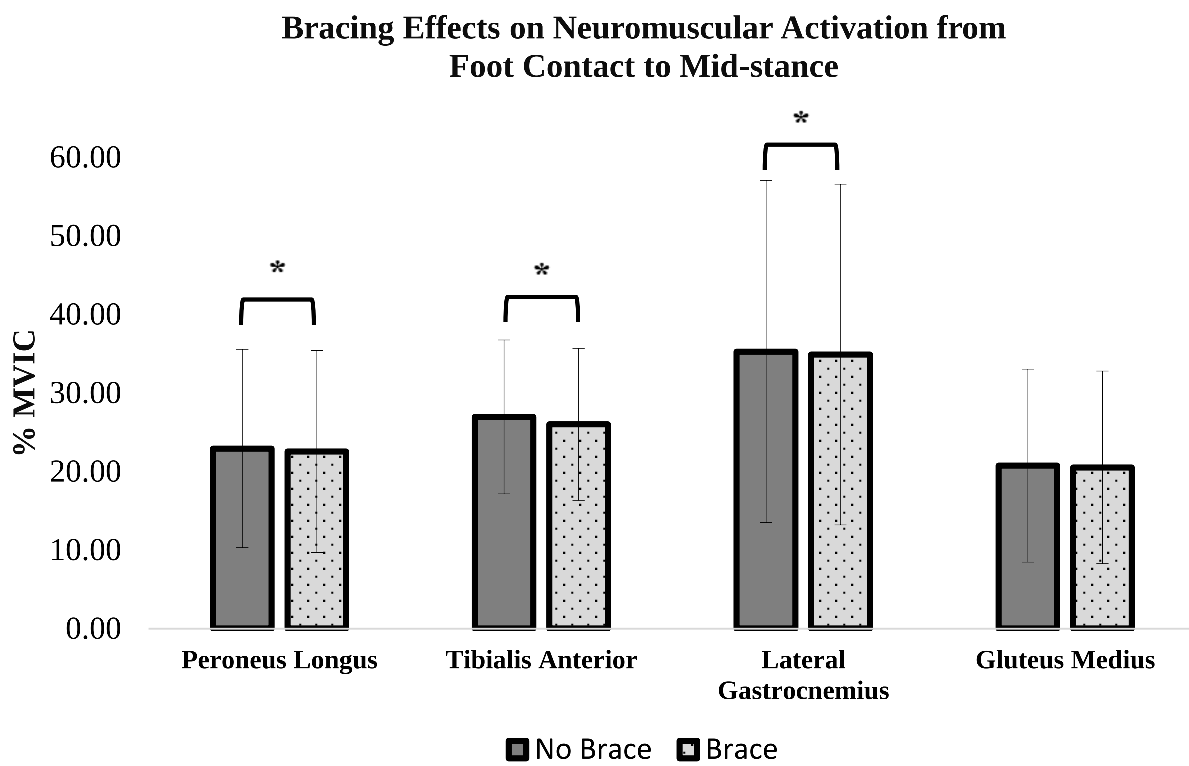

Figure 18: Bracing effects on neuromuscular activation of lower extremity muscles from foot contact to mid-stance phase of the gait cycle (***** $p < .05$).

Initial Swing Electromyography

Assessment of neuromuscular activation during the first 50% of swing phase within the walking gait cycle revealed a significant main effect of muscle [$F(3, 135) = 103.98$, $p < .001$, partial $\eta^2 = .69$]. Pairwise comparisons related to the muscle main effect can be found in Appendix C (Table C-11). Main effects of condition and group were not statistically significant. In addition, no significant interactions were noted.

Terminal Swing Electromyography

Assessment of neuromuscular activation during terminal swing phase of over ground walking revealed a significant main effect of muscle [$F(3, 135) = 112.57$, $p < .001$, partial $\eta^2 = .71$], as well as a significant condition by muscle interaction [$F(2.23, 100.34) = 3.66$, $p = .025$, partial η^2 = .07]. Simple effects of bracing between muscles were assessed by calculating the difference score for each muscle by subtracting the mean %MVIC score for the no brace condition from the mean %MVIC score of the brace condition. Post hoc comparisons using the LSD test were made to examine differences between muscles. Overall, a difference between one coupling pair was statistically significant ($p < .05$). Results from this post hoc analysis can be found in Appendix D (Table D-9). Simple effects of bracing on the neuromuscular activation of each individual muscle, during terminal swing phase, were assessed using one-sample t-tests. Results from the t-tests indicated that bracing significantly diminished the average activation of the peroneus longus during terminal swing phase of walking [$t(47) = -2.28$, $p = .027$] (Figure 19).

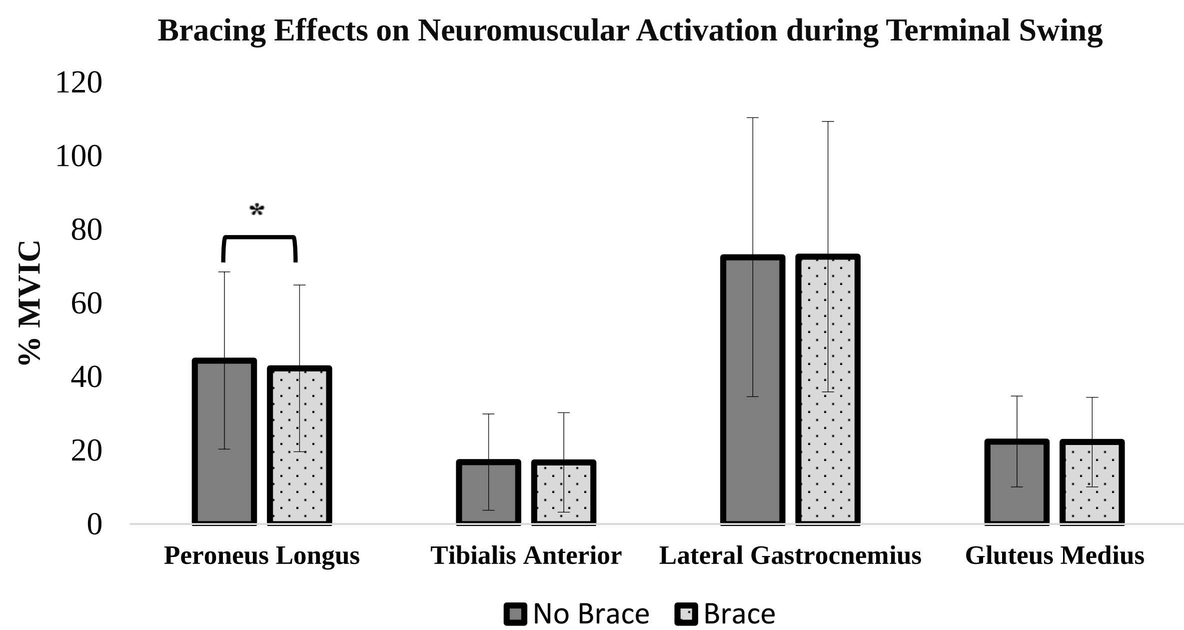

Figure 19: Bracing effects on neuromuscular activation of lower extremity muscles during the terminal swing phase of the gait cycle (***** p <.05).

Walking %MVIC Means and Standard Deviations (Mean ± SD)

Muscle	Foot Contact to Mid-stance		Toe-off to Mid-swing		Terminal Swing	
	NB	*B*	*NB*	*B*	*NB*	*B*
1	22.94***** ± 12.63	22.55***** ± 12.84	42.57 ± 26.60	41.06 ± 24.79	44.41***** ± 24.10	42.32***** ± 22.63
2	26.96***** ± 9.80	26.02***** ± 9.68	16.66 ± 10.23	16.62 ± 11.35	16.85 ± 13.07	16.75 ± 13.52
3	35.28***** ± 21.76	34.90***** ± 21.71	74.46 ± 42.49	75.18 ± 42.15	72.51 ± 37.89	72.65 ± 36.71
4	20.76 ± 12.27	20.53 ± 12.28	23.68 ± 12.51	23.33 ± 12.17	22.43 ± 12.32	22.27 ± 12.13

Table 13: Percent normalized electromyography (% MVIC) means and standard deviations collapsed across groups for each phase of the walking gait cycle. 1 = *Peroneus Longus*, 2 = *Tibialis Anterior*, 3 = *Lateral Gastrocnemius*, 4 = *Gluteus Medius*; NB = *No brace condition*, B = *Brace condition*; ***** = *p < .05 for differences between conditions*

CHAPTER V

DISCUSSION

The purpose of this project was twofold: First, to explore if individuals with chronic ankle instability (CAI) (i.e. Non-Copers) exhibit alterations to lower extremity coordination, coordination dynamics, and neuromuscular characteristics compared to individuals without CAI. Second, to assess if ankle joint bracing influences coordination, coordination dynamics, and neuromuscular activity in individuals with and without CAI. Specifically, this study was powered to detect differences in coordination between healthy individuals and individuals with CAI, as well as differences in lower leg electromyography in individuals with CAI when wearing an ankle brace compared to no brace conditions. However the experimental and statistical designs were constructed to include additional within group and between group analyses, which were deemed exploratory in nature.

The methodological approach for this project was designed with many factors in mind. First, questions on the health history questionnaire as well as self-assessment surveys were chosen in accordance with approaches common in the literature pertaining to CAI (Delahunt et al., 2010). Wikstrom et al. (2012) reported that perception-based outcomes had the greatest ability to distinguish between copers and individuals with CAI. For this study, the Foot and Ankle Ability Measure and sport subscale (FAAM and FAAM-S) were employed to detect self-reported functional deficits related to CAI. Additionally, the Ankle Instability Instrument was employed to detect symptoms of functional ankle instability in Non-Coper participants. Each of these tools has previously been tested for validity and reliability (Carcia, 2008; Docherty et al., 2006; Martin & Irrgang, 2007; Martin et al., 2005), which supports the methodological steps taken to recruit Healthy, Coper, and Non-Coper participants for this study.

Second, experimental tasks used in this study (i.e. single-leg hopping and walking) represent tasks that are ubiquitously implemented in biomechanics research, and are nested in CAI research from which clinical, performance, and behavioral assessments are made. Furthermore, over ground walking was chosen as a task, in particular, to most closely replicate walking movement patterns that could be observed in nature and to avoid the use of a treadmill, which may induce unintended alterations to normal walking patterns (Alton et al., 1998). Furthermore, single-leg hopping is often used to assess ankle function during dynamic tasks (Caffrey et al., 2009; Wikstrom et al., 2009) that involve neuromusculoskeletal demands disparate to walking.

Finally, the dependent measures employed for this study were chosen to address the questions laid out in the specific aims: Mean relative phase angles were calculated as a measure of the coordination between lower extremity segments. Relative phase variability was employed as a measure of the lower extremity coordination dynamics. Electromyography was employed to assess the neuromuscular activation patterns of lower extremity muscles. This chapter will present a detailed discussion of the findings for each dependent measure. The results will be discussed in sections following the form of those in Chapter IV beginning with the dependent measures. Next, a synthesis of the discussion and implications is presented. Last, future research stemming from this project will be proposed.

Section 1: Coordination

Excessive foot inversion and external rotation of the shank places strain on lateral ankle ligaments and is commonly associated with the mechanism for lateral ankle sprain (Hertel, 2002). As a result, foot and shank kinematics during walking and landing tasks have been widely researched in individuals who suffer from recurring ankle sprain injuries (Chinn, Dicharry, & Hertel, 2013; Delahunt, Monaghan, & Caulfield, 2006b; Delahunt, Monaghan, & Caulfield, 2007; Doherty, Bleakley, et al., 2015b; Doherty et al., 2014; Hertel, 2000; Monaghan, Delahunt, & Caulfield, 2006). Less focus has been given to the coordination of the lower extremity in individuals who have suffered one or more ankle sprains (Doherty, Bleakley, Hertel, Caulfield, Ryan, Sweeney, et al., 2015; Doherty et al., 2014; Kipp & Palmieri-Smith, 2012), with even fewer studies focusing on the relative motion of lower extremity segment couplings (Drewes et al., 2009; Herb et al., 2014). Furthermore, studies focusing on the effects of ankle bracing on lower extremity coordination are also limited (Herb, Chinn, & Hertel, 2016; Ozer, 2009). An aim of this study was to elaborate on the lower extremity coordination characteristics exhibited by individuals with and without CAI, as well as understand the effects of ankle bracing on lower extremity coordination across groups.

Coordination of lower extremity segment couplings was assessed over phases of the single-leg hop and over ground walking cycles using mean relative phase (MRP) angles. To begin this discussion, a brief review on the construct and interpretation of relative phase analysis will be presented so that the outcomes are more readily apparent to the reader. Relative phase analysis can be used as a direct measure of the relative motion of two segments throughout multiple cycles of a movement. MRP angles were calculated using phase angles measured from phase portraits created for each segment. Phase portraits were constructed by plotting segment angular displacement (abscissa) and angular velocity (ordinate) values averaged across cycles of single-leg hopping or over ground walking (Figure 20).

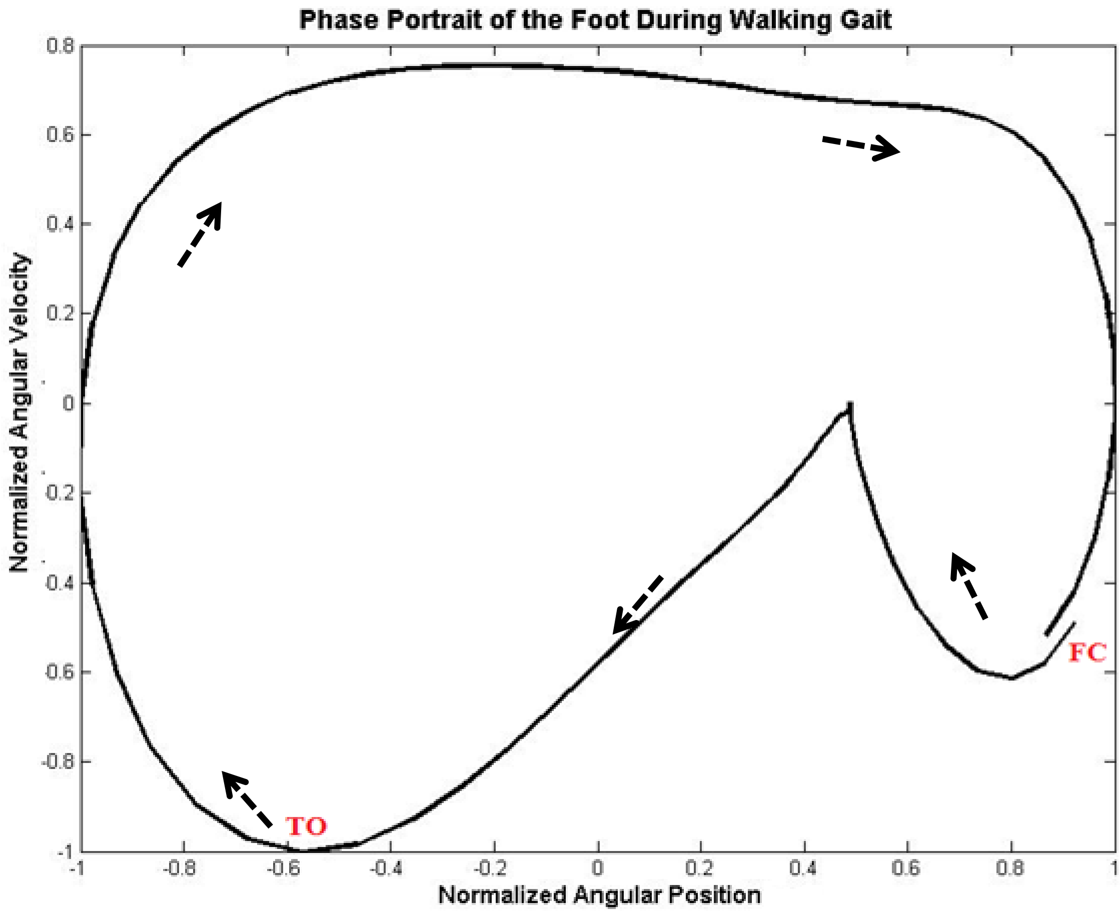

Figure 20: Representative phase portrait of the foot during walking. Normalized sagittal plane angular position and angular velocity data were averaged across 15 cycles of walking. Progression of the foot through state space begins at foot contact (FC) and continues clockwise past toe-off (TO) before returning to FC.

Phase angles represent the progression of segment motion through a given phase of a single cycle of movement. Relative phase angles represent the relative motion of the coupled segments, and fell into a range of 0-180 degrees (0 degrees indicates completely synchronous motion of the coupled segments and 180 degrees indicates complete asynchronous motion of the coupled segments) (Figure 21). In other words, the timing of one segment's motion relative to another segment's motion during a cycle of movement can be quantified using relative phase angles; for this study, the timing disparity between segment motions is proportional to relative phase value. Furthermore, MRP represents the average relative phase angle exhibited for a particular segment coupling (Table 5) over a specified phase of movement.

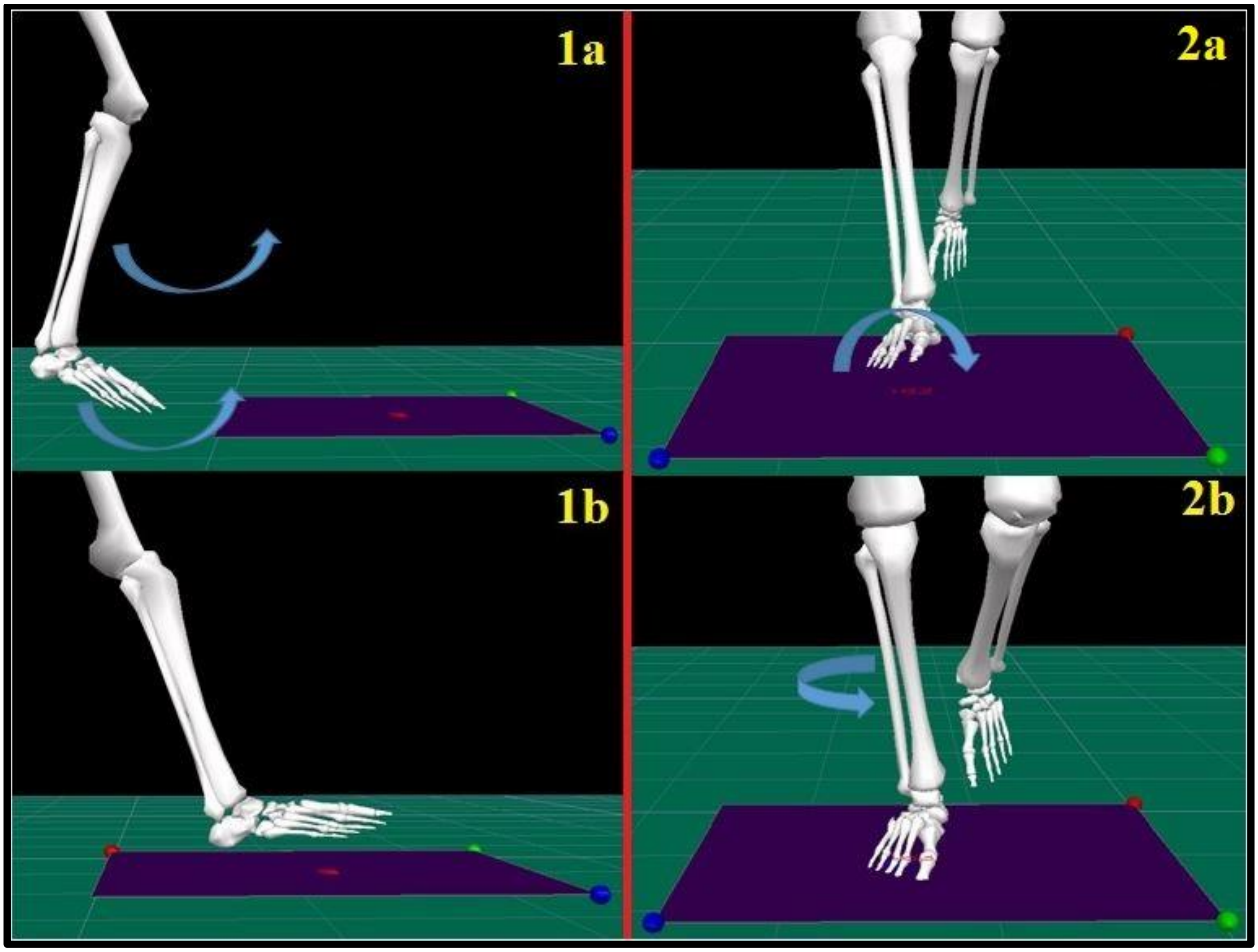

Figure 21: Example of the foot and shank phase relationship during walking. From the start of terminal swing phase (1a), sagittal plane foot – shank motion is in phase (i.e. segment motion is synchronous) until just before foot contact (1b). At foot contact (2a) the foot undergoes eversion followed by shank internal rotation up to mid-swing (2b), representing an out of phase relationship (i.e. segment motion is asynchronous).

Single-Leg Hop Coordination

Hop Flight Coordination

Results from the analysis of coordination during the hop flight phase revealed that bracing significantly affected sagittal plane foot – shank and foot – thigh coordination in Copers and Non-Copers during hop flight, but did not affect the healthy group (Figure 9 & Figure 10).

For this study, the hop flight phase of the single-leg hopping cycle consisted of a brief period between the propulsion and landing phases (i.e. vertical displacement of the body's center of mass was minimal during the hopping task), therefore preparation for foot contact and load response had to occur within a short window of time. Analysis of coordination during hop flight indicated that individuals who had previously sprained their ankles (both Copers and Non-Copers) exhibit more synchronized sagittal plane foot – shank motion but less synchronized foot –thigh motion in preparation for foot contact when braced compared to those who had not previously suffered an ankle sprain injury.

Previous research has shown that sagittal plane ankle range of motion can be significantly limited following ankle sprain injury (Hertel, 2002), and this limitation persists when individuals perform landing tasks (Doherty et al., 2014). Furthermore, ankle bracing has also been shown to limit sagittal plane range of motion of the ankle joint during landing tasks (DiStefano et al., 2008; Mason-Mackay, Whatman, & Reid, 2015; Simpson, 2013; Zhang et al., 2012). Although ankle joint range of motion was not measured for this study, it is possible that both previous ankle sprain injury and ankle bracing have a compounding effect on ankle joint range of motion. This could explain the more synchronous sagittal plane foot – shank motion observed in both Copers and Non-Copers in the braced condition. Furthermore, an increase in knee angle just prior to landing, as was observed by DiStefano et al. (2008) and Simpson (2013), paired with bracing restrictions could cause the greater disparity in sagittal plane foot – thigh motion that was observed in Copers and Non-Copers.

Walking Coordination

Foot Contact to Mid-stance Coordination

Analysis of lower extremity coordination during the foot contact to mid-stance phase showed that bracing, regardless of group, significantly influenced Foot dorsiflexion/plantar flexion (DF/PF) – shank flexion/extension (F/E), and shank F/E – thigh F/E coordination (Figure 11). Studies investigating gait kinematics in individuals with CAI reported decreased plantar flexion throughout the gait cycle (Chinn, Dicharry, & Hertel, 2013), and increased hip flexion after foot contact (Doherty, Bleakley, et al., 2015b). Although no group differences in coordination were observed during this phase, it is possible that kinematic disparities may have existed between groups. Such differences, in concert with bracing effects, could produce marginal alterations to sagittal plane foot – shank and shank – thigh coordination when data were collapsed across groups. It is also possible that any effects occurring strictly during the load response phase were diminished when considering time after load response up to mid-stance, which could explain the minimal differences. However, further analysis of the kinematic data is needed to substantiate these inferences. In any case, the differences noted show that sagittal plane foot and shank motion is slightly more synchronized while braced, and the timing between sagittal plane shank and thigh motion is slightly less synchronized during the first half of stance phase.

Toe-off to Mid-swing Coordination

From toe-off to mid-swing of the walking gait cycle, bracing effects were observed for four lower extremity couplings (Figure 12). Coordination of the foot – shank in the sagittal plane became slightly more in phase. The observed finding for foot – shank coordination in the sagittal plane was similar to that of the first half of stance phase. However, from toe-off to mid-swing phase of walking, sagittal plane foot-shank motion was more tightly coupled compared to stance phase. This is not surprising as the foot is constrained by the ground during after foot fall, and all motion would occur as a result of the motion of the shank about the foot.

At the beginning of the swing phase, foot and shank motion undergoes primarily linear translation, and is largely affected by hip flexion. It is possible that the slight decrease in MRP values observed for the sagittal plane foot – shank coupling during the braced condition was due to a decrease in ankle plantarflexion at toe-off of the walking propulsion phase. Decreased ankle plantarflexion at toe-off would reduce the magnitude of dorsiflexion motion during the initial swing phase as the body prepares for the foot to achieve ground clearance. Likewise, decreased ankle plantarflexion at toe-off during the braced condition could also explain the observed increase in MRP for the sagittal plane foot – thigh coupling. If the foot remains in a less plantarflexed (more dorsiflexion) position when braced, thigh motion would further dominate the sagittal plane foot – thigh coupling and cause the MRP value to increase.

Foot inversion/eversion (IN/EV) – shank rotation (ROT) coordination was also altered during the braced condition during the toe-off to mid-swing phase. Specifically, coupling motion became more out of phase (more asynchronous) during the braced condition. The foot is at peak inversion just after toe-off and everts slightly while approaching mid-swing. Furthermore, the shank is at peak external rotation at toe-off and transitions toward internal rotation throughout swing phase. When braced, frontal plane foot motion has been shown to be restricted (Cordova, Ingersoll, & Palmieri, 2002; DiStefano et al., 2008; Shapiro et al., 1994; Simpson, 2013; Zhang et al., 2012); therefore it is possible that diminished frontal plane foot motion could cause the greater timing disparity in coupling relative motion exhibited during the braced condition.

Finally, it was observed that foot IN/EV – thigh abduction/adduction (AB/AD) coupling became more out of phase during the braced condition of toe-off to mid-swing. The thigh experiences a relatively small degree of frontal plane motion during the first half of swing phase; however, it is possible that greater hip abduction was necessary to achieve ground clearance during swing phase, or was adopted to avoid contact of the brace with the contralateral limb as the braced limb moves through swing phase. This conclusion is supported by Shorter et al. (2008), who reported that subjects exhibit greater step width during ankle brace conditions compared to no brace conditions. Thus, restricted frontal plane foot motion coupled with greater frontal plane thigh excursion could have caused the coupling to become more out of phase during swing phase. Altogether, coordination adaptations during the first half of swing phase could be indicative of either altered propulsive mechanics or preparations made for foot contact.

Terminal Swing Coordination

Analysis of lower extremity coordination during terminal swing phase revealed that coordination of the foot – shank sagittal plane coupling became slightly more in phase (synchronous) during the terminal swing phase of the braced walking trials. In addition, foot IN/EV – thigh AB/AD coordination became slightly more out of phase during the terminal swing phase of the braced walking trials (Figure 12). It is notable to mention that these effects were the same as those observed during toe-off to mid-swing phase. Thus it would appear that the alterations in foot – shank and foot – thigh coordination are sustained from initial swing phase to terminal swing phase. Further, the appearance of these findings immediately after and immediately before ground contact suggests that lower extremity kinetics may be modulated surrounding loading phases of the gait cycle when an ankle brace is applied (Ota et al., 2014).

Terminal swing phase represents the last 15% of swing phase during walking and is a critical period in which the limb is positioned for foot contact and load response. It may be true that a stronger sagittal plane foot and shank coupling contributes to the efficacy of ankle bracing, however alterations to lower extremity coordination during terminal swing phase could have negative ramifications, particularly if unexpected perturbations occur upon foot contact. This is especially true considering the alterations to foot – thigh coordination during swing phase, which signifies that bracing elicited effects up the kinetic chain.

It is uncertain how the observed finding of more out of phase foot IN/EV – thigh AB/AD coordination should be explained. Such a response could indicate that the thigh segment motion is altered in response to a constraint at the ankle joint (e.g. to achieve greater step width as discussed above). However, it is also true that restricted foot motion could drive the coupling to become more out of phase. Nonetheless, it would seem that a weaker coupling of the foot and thigh in the frontal plane could be detrimental if observed at foot contact as ground reaction forces are absorbed up the chain. Although this effect was not noted in the braced condition during the foot contact to mid-stance phase, it is possible that altered foot – thigh frontal plane coordination is sustained from swing phase until just after foot contact. Further analysis conducted during shorter time intervals, specifically during the load response phase within foot contact to mid-stance phase, would elucidate these findings.

Summary of Coordination

A summary of the effects on coordination observed within the single-leg hop and over ground walking cycles is presented in Table 14. Overall, it was shown that ankle bracing significantly impacted lower extremity coordination during hopping and walking. These effects were observed for foot – shank coordination as well as foot – thigh and shank – thigh coordination, signifying that ankle bracing carries a global influence rather than simply local restriction of joint range of motion.

Interestingly, the only group differences observed in this study occurred during the hop flight phase. Altered coordination in the foot – shank and foot – thigh coupling of Coper and Non-Coper participants could signify a maladaptive response to bracing during activities that place high demand on the previously injured limb such as cutting and landing. Specifically, individuals who have previously suffered an ankle sprain may be more susceptible for chronic or acute injury if changes to coordination patterns are not compensated for, or are compensated for in such a manner that puts adjacent structures at risk.

A majority of the observed bracing effects on coordination were during walking. Toe-off to mid-swing phase of gait was shown to be the most affected phase, containing alterations to four couplings. In addition, alterations to sagittal plane foot – shank coordination, and foot IN/EV – thigh AB/AD coordination were sustained through multiple phases of the gait cycle. This suggests that bracing effects on lower extremity coordination are not transient, but persistent through several critical stages of the gait cycle (e.g. pre and post foot contact).

Bracing Effects on Coordination Across Movement Phases

Coupling	**Hop Flight**	**Hop Propulsion**	**Hop Landing**	**FC - Midstance**	**TO - Midswing**	**Terminal Swing**
Foot dorsiflexion/plantarflexion – Shank flexion/extension	C ↓ NC ↓			↓	↓	↓
Foot dorsiflexion/plantarflexion – Thigh flexion/extension	C ↑ NC ↑				↑	
Foot inversion/eversion – Shank internal/external rotation					↑	
Foot inversion/eversion – Thigh abduction/adduction					↑	↑
Shank flexion/extension – Thigh flexion/extension				↑		

Table 14: Summary of bracing effects on coordination across single-leg hopping and walking phases.
↓ - *Indicates bracing significantly decreased MRP values (i.e. segmental motion became more in phase)*
↑ - *Indicates bracing significantly increased MRP values (i.e. segmental motion became more out of phase)*
C – *Coper group*
NC – *Non-coper group*

Section 2: Coordination Variability

Previous research surrounding dynamic systems theory of motor control suggests that variability in the relative motion of segments can be an indication of how stable or adaptable a movement pattern is in response to environmental, organismic, and task constraints (Kelso & Schöner, 1988; Schöner & Kelso, 1988). Researchers studying chronic ankle instability (CAI) have adopted tenets of dynamic systems theory in an attempt to explain functional limitations experienced by those with CAI, and gain a better understanding of the mechanism through which CAI is developed (Hoch & McKeon, 2010; McKeon & Hertel, 2006; Wikstrom, Hubbard-Turner, & McKeon, 2013). One aim of this study was to investigate if individuals who had previously suffered an ankle sprain injury (organismic constraint) exhibited altered lower extremity coordination dynamics (i.e. variability) compared to healthy individuals during single-leg hopping and over ground walking. Specifically, it was of interest to explore the difference between healthy individuals, individuals who have previously suffered an ankle sprain injury but do not experience functional deficits or reoccurring ankle sprains (Copers), and individuals who suffer from reoccurring ankle sprains or have functional limitations as a result of the initial ankle injury (Non-Copers).

Within each phase of the single-leg hopping cycle and walking cycle, no group differences in coordination variability were observed. These results are supported by Drewes et al. (2009), who found no difference in foot – shank coupling variability between individuals with CAI and healthy controls during treadmill walking. Contrarily, Herb et al. (2014) noted differences in foot IN/EV – shank ROT variability during late stance, toe-off, and early swing phase of treadmill walking using a vector coding analysis. One factor that could contribute to these conflicting results is related to the manner in which coordination variability was assessed. Specifically, Drewes et al. (2009) used a single value to measure the average variability over the entire gait cycle. The present study investigated variability over discrete phases of the gait cycle, however some phases were not assessed (i.e. mid-swing to terminal swing and mid-stance to toe-off were not included in the present analysis). Ultimately, it is possible that differences in coordination dynamics between groups exist in narrow windows of time throughout the gait cycle. Expanding the analysis to assess dynamics continuously across the entire movement cycle could reveal such differences, as was reported by Herb et al. (2014). It is also important to note that the measure used to quantify variability differed between the studies mentioned above. Therefore, direct comparisons between this study and those that observed group differences in coordination variability during walking should be made with caution.

A second aim of this study was to assess the effects of ankle bracing (task constraint) on lower extremity coordination dynamics in Healthy, Coper, and Non-Coper groups. The implementation of ankle bracing is widespread in both healthy populations and populations with a history of ankle sprain injury, however, the effects of ankle bracing on lower extremity coordination dynamics have not previously been established. Several significant effects of ankle bracing on lower extremity coordination dynamics across groups were noted for this study; these results will be discussed in the following sub-sections.

Single-Leg Hopping Coordination Variability

Landing Coordination Variability

A main effect of condition on coordination variability was observed during the hop landing phase of the single-leg hop cycle. Follow up analysis revealed that foot IN/EV – shank ROT coupling variability decreased during hop landing as a result of bracing. Single-limb landing tasks involve a critical period of load response as the foot makes contact with the ground. The involved segments must be coordinated in such a manner so that external forces can be safely absorbed and so that task parameters (e.g. center of gravity positioning) are minimally affected. Furthermore, frontal plane motion of the foot and transverse plane motion of the shank are important for dissipating ground reaction forces upon ground contact (Donatelli, 1985). Reduced variability in the foot IN/EV – shank ROT coupling during load response could be attributed to the bracing constraint paired with a closed chain foot configuration. As a result, the foot – shank coupling experiences a diminished capacity to adapt to perturbations (e.g. landing on various surface types) during load response, and could result in a greater demand placed on proximal lower extremity segment couplings to attenuate forces (Mason-Mackay, Whatman, & Reid, 2015; Venesky, 2006).

Flight Coordination variability

For the hop flight phase, foot IN/EV – shank ROT coordination variability significantly increased during braced conditions compared to no brace conditions. This finding is particularly interesting in that bracing had differential effects on foot – shank coordination dynamics from hop flight to hop landing. In addition, it is notable that foot – shank coordination variability increased despite of the restrictive effects of bracing on foot and shank motion. One explanation for these findings is that increased foot – shank variability during the hop flight phase is a compensatory response to the decreased foot – shank adaptability during landing. It is possible that greater flight foot – shank variability allows the motor system to explore segment configurations in order to optimize segment positioning in preparation for landing, where bracing induced a more stable foot – shank configuration.

Walking Coordination Variability

Initial Swing Coordination Variability

Analysis of coordination variability from the toe-off to mid-swing phase of the walking cycle showed that the coordination variability of four couplings was affected by bracing. These findings indicate that bracing had a systemic effect on the dynamics of the lower extremity during toe-off to mid-swing. Specifically, bracing induced reductions in variability across all sagittal plane couplings assessed. Revisiting the coordination results from toe-off to mid-swing, it was observed that foot – shank and foot –thigh sagittal plane coordination was altered during braced conditions. One explanation is that the motor system stabilized these adaptations by reducing pattern variability in order to preserve the task parameters (e.g. foot clearance during swing phase). However, diminished coordination variability could potentially have a deleterious effect particularly if unexpected perturbations are experienced as the lower extremity progresses through swing phase.

Furthermore, increased foot IN/EV – shank ROT variability was noted during toe-off to mid-swing phase. Increased foot - shank variability could signify a more adaptable foot – shank coupling in response to restricted foot and shank motion during braced conditions. When braced, compensatory behavior may be necessary for the foot to achieve ground clearance and for the foot – shank configuration to re-organize prior to terminal swing phase, where variability was shown to be diminished in preparation for foot contact.

Terminal Swing Coordination Variability

Analysis of coordination variability during terminal swing phase of walking indicated that variability of the foot DF/PF – shank F/E and foot IN/EV – shank ROT couplings significantly decreased during braced conditions. Dampening effects of bracing on sagittal plane foot – shank coordination variability during toe-off to mid-swing phase were extended to terminal swing phase of the walking cycle. Conversely, the tendency of foot IN/EV – shank

ROT variability to be affected uniformly across swing phase intervals was not observed. In other words, the increase in foot IN/EV – shank ROT variability observed during toe-off to mid-swing of the braced condition was not extended to terminal swing phase. Rather, during terminal swing phase, a decrease in variability was exhibited with bracing. Herb et al. (2014) postulated that reduced foot – shank variability during terminal swing phase might result in fewer strategies to accomplish the movement task. Thus, it could be said that bracing, while efficacious for reducing the risk of ankle sprain injury, acts as a detriment to lower extremity coordination dynamics during terminal swing phase.

Summary of Coordination Variability

Altogether, bracing affected lower extremity dynamics across multiple phases of both hopping and walking cycles (Table 15). These effects were most apparent during the swing phase of walking, where variability was predominately diminished across foot – shank, foot – thigh, and shank – thigh couplings. Such an effect indicates that ankle bracing adaptations permeate proximally up the extremity from the ankle joint and could influence lower extremity function at ground clearance and limb positioning prior to foot contact during walking. It is plausible that alterations in the coordination dynamics were necessary to preserve function with respect to the movement task. However, these adaptations may diminish the movement system's capacity to respond to additional or unexpected task or environmental constraints.

Bracing Effects on Coordination Variability Across Movement Phases

Coupling	**Hop Flight**	**Hop Propulsion**	**Hop Landing**	**FC - Midstance**	**TO - Midswing**	**Terminal Swing**
Foot dorsiflexion/plantarflexion – Shank flexion/extension					↓	↓
Foot dorsiflexion/plantarflexion – Thigh flexion/extension					↓	
Foot inversion/eversion – Shank internal/external rotation	↑		↓		↑	↓
Foot inversion/eversion – Thigh abduction/adduction						
Shank flexion/extension- Thigh flexion/extension					↓	

Table 15: Summary of bracing effects on coordination variability during single-leg hopping and walking phases.
↓ - *Indicates bracing significantly decreased RPD values (i.e. coupling variability significantly decreased)*
↑ - *Indicates bracing significantly increased RPD values (i.e. coupling variability significantly increased)*
C – *Coper group*
NC – *Non-coper group*

Section 3: Electromyography

In this section, the results from the analysis of neuromuscular activation during hopping and walking trials will be discussed. Average surface electromyography (EMG) measures were obtained for the peroneus longus, tibialis anterior, lateral gastrocnemius, and gluteus medius during hopping and walking cycles. Group and condition effects on neuromuscular activation were of particular interest to this study because of their implications involving the coordination and coordination dynamics of lower extremity segment couplings.

The first aim of this study, involving neuromuscular activation, was to determine if differences in the activation of lower extremity muscles existed between Healthy, Coper, and Non-Coper groups during phases of hopping and walking cycles. Findings for this study revealed no differences in average EMG activation between groups during intervals of hopping and walking. These findings were supported by results from previous studies showing no group differences in surface EMG amplitude of the peroneus longus (Feger et al., 2015), tibialis anterior (Caulfield et al., 2004; Delahunt, Monaghan, & Caulfield, 2006b; Feger et al., 2015), lateral gastrocnemius, and gluteus medius (Feger et al., 2015) during walking or landing. However, some studies have demonstrated alterations in the temporal response of the peroneus longus in CAI populations (Caulfield et al., 2004; Delahunt, Monaghan, & Caulfield, 2006b; Delahunt, Monaghan, & Caulfield, 2007; Feger et al., 2014; Feger et al., 2015; Santilli et al., 2005). It appears that analysis of the temporal characteristics of lower extremity muscle activation may provide a better indication of the representative characteristics of CAI populations rather than the analysis of average activation over longer time periods.

Despite the lack of group differences, bracing effects on neuromuscular activation were observed. Specifically, differences were noting across groups over phases of the walking gait cycle. These finding will be discussed in the following sub-section.

Walking Electromyography

Stance Electromyography

Analysis of neuromuscular activation during stance phase of walking indicated that average activation of peroneus longus, tibialis anterior, and lateral gastrocnemius muscles was diminished during braced walking trials. From foot contact to mid-stance, diminished activation of muscles that cross the ankle joint could indicate that the restrictive properties of the brace triggered the central nervous system to curtail the stabilizing effort of the peroneus longus, tibialis anterior, and lateral gastrocnemius during this period of load response and single-limb support. Although group differences in muscle activation were not observed for this study, previous studies have shown that activation of the peroneus longus is diminished in CAI populations during walking or functional tasks (Caulfield et al., 2004; Feger et al., 2014; Konradsen & Ravn, 1990; Santilli et al., 2005); results from this study suggest that bracing would perpetuate reductions in peroneus longus activations in CAI populations and could lead to a situation where maladaptive neuromuscular patterns are sustained after the brace is removed.

Terminal Swing Electromyography

From the analysis of electromyography during terminal swing phase of walking it was shown that in the brace condition, average peroneus longus activation significantly decreased during terminal swing phase. These findings are supported by recent studies investigating the effect of ankle bracing on peroneus longus activation in CAI and healthy populations (Barlow et al., 2015; Feger et al., 2014). Considering ankle bracing has been shown to elicit greater foot eversion prior to contact than in no braced conditions, less demand for an eversion moment to position the foot during terminal swing could explain the observed decrease in peroneus longus activation.

Summary of Electromyography

Overall, bracing had a diminishing effect on lower extremity muscles during walking (Table 16). This effect was most prevalent during foot contact to mid-stance phase, where decreases in peroneus longus, tibialis anterior, and lateral gastrocnemius activity were observed. However, the reader should note that differences in EMG activation between conditions were small; therefore clinical significance should not be assumed.

Bracing Effects on Neuromuscular Activation across Movement Phases

Muscle	**Hop Flight**	**Hop Propulsion**	**Hop Landing**	**FS - Midstance**	**TO - Midswing**	**Terminal Swing**
Peroneus Longus				↓		↓
Tibialis Anterior				↓		
Lateral Gastrocnemius				↓		
Gluteus Medius						

Table 16: Summary of bracing effects on average EMG activation during single-leg hopping and walking phases.
↓ - *Indicates bracing significantly decreased %EMG values (i.e. average muscle activation significantly decreased)*
↑ - *Indicates bracing significantly increased %EMG values (i.e. average muscle activation significantly increased)*

Section 5: Synthesis

Group Effects

The development and persistence of aberrant functional limitations characterized by individuals with CAI has been the focus of researchers for some time. Using concepts from dynamic system theory, it has been proposed that the study of movement pattern dynamics (i.e. variability) could help elucidate the behavioral abnormalities observed in CAI populations (Hoch & McKeon, 2010; McKeon & Hertel, 2006; Wikstrom, Hubbard-Turner, & McKeon, 2013). This study showed that the coordinated motion of lower extremity couplings was similar between individuals with and without CAI, and the dynamics of said patterns were not discriminate between groups. Based on these results, it cannot be said that functional limitations surrounding CAI are associated with alterations in lower extremity coordination dynamics in response to added organismic constraints (i.e. ankle sprain injury).

This was not the conclusion drawn by Herb et al. (2014), who postulated that altered foot – shank coordination and diminished foot – shank coupling variability during intervals of the gait cycle may be related to further instances of instability in individuals with CAI. Yet, Drewes et al. (2009), found no difference in foot – shank coordination variability, but did report altered foot – shank coordination during terminal swing phase of walking. Notably, both Herb et al. (2014) and Drewes et al. (2009) focused solely on the coordination and dynamics of foot IN/EV – shank ROT coupling, whereas this study expanded on the coordinative characteristics of several lower extremity couplings, encompassing the behavior of segments proximal to the ankle joint. Ultimately, additional research should be undertaken to elucidate the dynamic characteristics of the lower extremity as a functional unit, and resolve discrepancies in the literature related to the foot – shank coordination dynamics.

Similarly, the absence of group differences with respect to neuromuscular activity is somewhat supported by the literature, in that lower extremity EMG amplitude has been shown to be similar between individuals with and without CAI during phases of the walking cycle (Feger et al., 2015). Although several papers have concluded that altered timing of lower extremity muscular activation, in particular the peroneus longus, is characteristic of individuals with CAI (Caulfield et al., 2004; Delahunt, Monaghan, & Caulfield, 2006b; Feger et al., 2015; Konradsen & Ravn, 1990). Based on the results of this study and previous studies, the focus of research pertaining to neuromuscular activation in CAI populations should be directed to the temporal aspects of muscle activation during critical periods of movement tasks.

Bracing Effects

The implementation of ankle bracing is ubiquitous in both healthy and clinical populations. Although the efficacy of ankle bracing has been well established, the systemic effects of ankle bracing are not well understood. Prior to this study, the added effect of bracing as a task constraint on functional movement pattern characteristics has not been thoroughly explored, particularly in populations most apt to using an ankle brace such as those with CAI. This study provided evidence that bracing has a systemic effect on the coordination, coordination dynamics, and neuromuscular activation in lower extremity.

The effects of prophylactic ankle support on coordination and coordination dynamics in individuals with and without CAI during single-leg hopping have not been investigated. It was shown in this study that implementing an ankle brace induced changes to both the coordination and coordination dynamics of lower extremity segment couplings during the hopping task. Interestingly, during hop flight, participants who had previously sprained their ankle (Copers and Non-Copers) exhibited altered foot – shank and foot – thigh coordination during the braced condition whereas lower extremity coordination of healthy participants was unaffected. This finding represented the only group difference in behavior between groups for this study, and may be related to the heightened task demand (i.e. single limb stabilization and propulsion over short time intervals), coupled with any existing complications (e.g. mechanical or functional deficits) related to previous ankle sprain injury. The absence of neuromuscular adaptations during hop flight suggests that mechanical impairments, paired with the bracing constraint, may be the root cause of coordination adaptations in Coper and Non-Coper groups during hop flight.

Despite the aforementioned result, effects of bracing were uniform across groups during the walking task. These results are loosely supported by Herb, Chinn, and Hertel (2016) who reported that ankle taping affected foot – shank coordination and variability similarly between healthy and CAI individuals throughout the gait cycle. It was concluded that decreased foot – shank relative motion and variability in response to taping might indicate a protective mechanism to create a more consistent gait pattern between strides (Herb, Chinn, & Hertel, 2016). Although this conclusion contains merit, the findings of the present study show that ankle bracing impacts lower extremity coordination beyond foot IN/EV – shank ROT.

To expand on these findings, the synthesis of discussion points related to the bracing effects during walking will be distilled into parts. First, the anticipation and response to loading surrounding foot contact will be discussed. Eils and Rosenbaum (2003) demonstrated that ankle bracing modulates the anticipated response to induced inversion at foot contact during walking gait by restricting inversion and reducing inversion velocity prior to foot contact. Applying these results to the current study would explain the altered coordination and diminished peroneus longus muscle activation exhibited during terminal swing phase of the braced condition. Furthermore, Gehring et al. (2014) showed that in ankle brace conditions, the neuromuscular response after foot contact is diminished. This study yielded analogous responses to bracing in the shank musculature during foot contact to mid-stance in tandem with altered coordination of the foot – shank and shank – thigh couplings. It is apparent that ankle bracing impacts the shank musculature over intervals surrounding foot contact; however it appears that the motor system adheres to these responses prior to perturbation, as fluctuations in the coordination patterns were less apparent. It is also important to note that the restrictions imposed by the ankle brace, paired with less activation of shank musculature, would likely reduce the system's capacity to utilize an ankle strategy for propulsion of the lower extremity into swing phase. Support for this supposition is garnered from studies reporting that plantarflexion and hip extension are negatively impacted during ankle brace conditions (Doherty, Bleakley, et al., 2015b; Shorter et al., 2008; Spaulding, Livingston, & Hartsell, 2003).

Indeed, an adapted propulsion strategy could explain the extensive changes in coordination observed from toe-off to mid-swing during braced conditions, despite the absence of neuromuscular adaptations during this phase. In addition decrease in coordination variability from toe-off to mid-swing further strengthens the argument that the system favors the coordination strategies adopted when the ankle is braced. Altogether, it seems that the motor system relies heavily on bracing adaptations that are induced in anticipation of, and in response to perturbations experienced during load response. However, these compensatory responses were shown to impact the coordination and coordination dynamics throughout swing phase.

In part, the information garnered from the present study related to foot – shank coordination dynamics is contested. Greater fluctuations in the foot IN/EV – shank internal external rotation coupling from toe-off to mid-swing contradicts the Herb, Chinn, and Hertel (2016) findings of decreased fluctuations. Therefore, definitive conclusions regarding foot (frontal plane) and shank (transverse plane) coordination dynamics should be made with caution. Nonetheless, the fact that bracing caused a majority of lower extremity couplings to become more rigid suggests that bracing may be detrimental if applied in rehabilitation settings or situations where an individual is actively recovering from ankle sprain injury. The basis for this conclusion comes from proponents of dynamic systems/constraint-led approaches to rehabilitation, who stress that constraints (i.e. factors that influence the formation of movement patterns) imposed during rehabilitation (e.g. exercise parameters, use of implements, or environmental settings) should be carefully considered, and intended to promote functional variability within the movement system (Hoch & McKeon, 2010; Holt, Wagenaar, & Saltzman, 2010; McKeon, 2009; McKeon & Hertel, 2006; Wikstrom, Hubbard-Turner, & McKeon, 2013).

In closing, findings from this study support the notion that individuals with CAI do not differ from those without CAI with respect to lower extremity coordination, coordination dynamics, and neuromuscular activity during walking and single-leg hopping tasks. Further research is needed to resolve disparities in the literature pertaining to coordination and coordination dynamics in individuals with and without CAI. Moreover, implementing an ankle brace resulted in alterations to lower extremity coordination, coordination dynamics, and neuromuscular activity across groups during walking and single-leg hopping tasks.

From a performance standpoint, strength coaches and athletes (with and without previous ankle sprain injury) should be aware of the apparent alterations to lower extremity musculature experienced when wearing an ankle brace, particularly when training or during competition, as enhanced neuromuscular effort is often required for such activities. Furthermore, ankle bracing may limit the ability of the motor system adapt to added environmental and task constraints, which could be further impacted by acute or chronic limitations brought on by ankle sprain injury. Thus, the use of bracing concurrently through the rehabilitation process or during periods of recovery should be cautioned.

Section 6: Future Research

Several directions can be taken in light of the findings of this study. It is necessary to conduct further research investigating the coordination strategies and coordination dynamics between healthy individuals, ankle sprain copers and those with CAI. Specifically, it is important that future studies be done using similar methodological and analytical approaches, as it is currently difficult to make direct comparisons between studies due to differences in sample populations, movement tasks, and analytical approaches. In addition, it is of interest to explore the relationship between measures of coordination and/or coordination dynamics, and the self-report outcomes commonly used to assess functional limitations in CAI populations. The use of regression analysis could help uncover variables that associate with the observed decline in perceived function (i.e. self-reported outcomes), which characterizes the cascade from initial ankle sprain injury to chronic dysfunction.

It is also warranted to continue to investigate bracing effects on the motor system's dynamics. Specifically, comparing different levels of task constraint (e.g. ankle wrap vs rigid boot) would give an indication of how the system responds to more or less restriction at a given joint. In addition, studies should be conducted to explore how bracing impacts the ability of individuals to adapt to common perturbations (e.g. obstacle clearance or uneven surfaces). Such studies should include analysis across groups, similar to the present study, as it is important to identify how both healthy and clinical populations respond to bracing interventions commonly implemented.

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Appendix A

Forms and Self-Assessment Surveys

INFORMED CONSENT

for a Research Study entitled

Ankle Bracing Effects on Coordination and Coordination Variability in Individuals With and Without Chronic Ankle Instability

You are invited to participate in a research study to understand the effect of ankle bracing on coordination variability characteristics in individuals with and without chronic ankle instability. The study is being conducted by Adam Jagodinsky under the direction of Wendi Weimar in the Auburn University School of Kinesiology. You were selected as a possible participant because you are between the ages of 19-30 years and perform a minimum of 90 minutes of physical activity per week. Exclusion criteria for this study includes: allergy to adhesives, injury to your ankle or participation in ankle rehabilitation within the past 3 months, lower extremity surgery, current or previous neurological impairment or movement disorder (other than ankle dysfunction), or reason to believe that participation in this study may put your health or well-being at risk.

What will be involved if you participate? If you agree to participate, you will be asked if you have ever sustained a lateral ankle sprain. Based on your reply, you will be assigned to one of three groups. The healthy group will consist of individuals who have never sustained a lateral ankle sprain. The ankle sprain coper group will consist of individuals who have sustained a lateral ankle sprain, but do not have any function limitations or instability. The ankle sprain non-coper group will consist of individuals who have sustained a lateral ankle sprain and have functional limitations and/or instability. As a participant, you will be asked to report to the Sport Biomechanics Laboratory for approximately 90 minutes and perform trials of over ground walking and single leg hopping with and without an ankle brace (flexible nylon ankle brace consisting of lace up ties and straps that fold around the foot and ankle).

Are there any risks or discomforts? Potential risk to the participants may include musculoskeletal injury during the performance of the movement tasks as well as muscle fatigue from the test protocol. In addition, muscle soreness may occur immediately and/or several days after the completion of the test protocol. To minimize these risks it is a requirement that you are physically active; therefore it is unlikely that your participation in walking and hopping tasks will increase the likelihood of suffering a musculoskeletal injury. Furthermore, you must not have sustained an ankle injury within the past 3 months, which will reduce the likelihood of any pain or discomfort due to a recent ankle injury. Finally, an athletic trainer will be present or available during each data collection, and an individual with CPR/AED certifications will be present at all times. You are responsible for any costs associated with medical treatment needed as a result of participating in this study. Finally breach of confidentiality poses a risk for participants considering the investigators will be present for the duration of data collection procedures.

If you change your mind about participating, you can withdraw at any time during the study. Your participation is completely voluntary. If you choose to withdraw, your data can be withdrawn as long as it is identifiable. Your decision about whether or not to participate or to stop participating will not jeopardize your future relations with Auburn University or the School of Kinesiology.

Your privacy will be protected. Any information obtained in connection with this study will remain confidential. Any data collected will be stored anonymously onto a computer under the participant number. There will be no document corresponding the participant's name with the data collected. Information obtained through your participation will be included in a dissertation, may be presented at regional and/or national conferences and submitted for publication in scholarly journals pertaining to biomechanics, movement science or human performance.

If you have questions about this study, *please ask them now or* contact Adam Jagodinsky at aej0015@auburn.edu or Dr. Wendi Weimar at weimawh@auburn.edu.

If you have questions about your rights as a research participant, you may contact the Auburn University Office of Research Compliance or the Institutional Review Board by phone (334)-844-5966 or e-mail at IRBadmin@auburn.edu or IRBChair@auburn.edu.

HAVING READ THE INFORMATION PROVIDED, YOU MUST DECIDE WHETHER OR NOT YOU WISH TO PARTICIPATE IN THIS RESEARCH STUDY. YOUR SIGNATURE INDICATES YOUR WILLINGNESS TO PARTICIPATE.

____________________________ Participant's signature Date	____________________________ Investigator obtaining consent Date
____________________________ Printed Name Date	____________________________ Printed Name Date
	____________________________ Co-Investigator Date
	____________________________ Printed Name Date

Participant Health Screening Questionnaire

Please read each question carefully and answer honestly. If you do not understand the question, please ask the investigator for clarification. Check the appropriate answer. **Answering "Yes" to questions 5, 6, 7, 8, and/or 9 will exclude you from participating in this study. Answering "No" to questions 1 and/or 10 will also exclude you from participating in this study.**

<u>**Yes**</u> <u>**No**</u>

___ ___ 1. Are you between the ages of 19-30 and years?

___ ___ 2. Have you suffered at least one lateral ankle sprain injury that required a period of ankle immobilization or assisted weight bearing, more than 12 months ago?

___ ___ 3. Do you currently experience any pain, weakness, instability or feelings of "giving way" in your previously injured ankle?

___ ___ 4. Have you had two or more episodes of your previously injured ankle giving way between 3 to 12 months of this date?

___ ___ 5. Have you suffered an ankle injury or been involved in rehabilitation associated with an ankle injury within the past 3 months?

___ ___ 6. Do you currently or have you had any neurological impairments or movement disorders (other than ankle dysfunction)?

___ ___ 7. Have you ever had a lower extremity surgery?

___ ___ 8. Do you have any reason to believe that your participation in this investigation may put your health or well being at risk?

___ ___ 9. Are you allergic to adhesives?

___ ___ 10. Do you partake in a minimum of 90 minutes of physical activity per week?

Signature of participant ________________________________ Date__________________

Foot and Ankle Ability Measure (FAAM)
Activities of Daily Living Subscale

Please Answer **every question** with **one response** that most closely describes your condition within the past week.
If the activity in question is limited by something other than your foot or ankle mark "Not Applicable" (N/A).

	No Difficulty	Slight Difficulty	Moderate Difficulty	Extreme Difficulty	Unable to do	N/A
Standing	☐	☐	☐	☐	☐	☐
Walking on even Ground	☐	☐	☐	☐	☐	☐
Walking on even ground without shoes	☐	☐	☐	☐	☐	☐
Walking up hills	☐	☐	☐	☐	☐	☐
Walking down hills	☐	☐	☐	☐	☐	☐
Going up stairs	☐	☐	☐	☐	☐	☐
Going down stairs	☐	☐	☐	☐	☐	☐
Walking on uneven ground	☐	☐	☐	☐	☐	☐
Stepping up and down curbs	☐	☐	☐	☐	☐	☐
Squatting	☐	☐	☐	☐	☐	☐
Coming up on your toes	☐	☐	☐	☐	☐	☐
Walking initially	☐	☐	☐	☐	☐	☐
Walking 5 minutes or less	☐	☐	☐	☐	☐	☐
Walking approximately 10 minutes	☐	☐	☐	☐	☐	☐
Walking 15 minutes or greater	☐	☐	☐	☐	☐	☐

Foot and Ankle Ability Measure (FAAM)
Activities of Daily Living Subscale
Page 2

Because of your foot and ankle how much difficulty do you have with:

	No Difficulty at all	Slight Difficulty	Moderate Difficulty	Extreme Difficulty	Unable to do	N/A
Home responsibilities	☐	☐	☐	☐	☐	☐
Activities of daily living	☐	☐	☐	☐	☐	☐
Personal care	☐	☐	☐	☐	☐	☐
Light to moderate work (standing, walking)	☐	☐	☐	☐	☐	☐
Heavy work (push/pulling, climbing, carrying)	☐	☐	☐	☐	☐	☐
Recreational activities	☐	☐	☐	☐	☐	☐

How would you rate your current level of function during you usual activities of daily living from 0 to 100 with 100 being your level of function prior to your foot or ankle problem and 0 being the inability to perform any of your usual daily activities.

__ __ __ . 0 %

Martin, R; Irrgang, J; Burdett, R; Conti, S; VanSwearingen, J: Evidence of Validity for the Foot and Ankle Ability Measure. Foot and Ankle International. Vol.26, No.11: 968-983, 2005.

Foot and Ankle Ability Measure (FAAM)
Sports Subscale

Because of your foot and ankle how much difficulty do you have with:

	No Difficulty at all	Slight Difficulty	Moderate Difficulty	Extreme Difficulty	Unable to do	N/A
Running	☐	☐	☐	☐	☐	☐
Jumping	☐	☐	☐	☐	☐	☐
Landing	☐	☐	☐	☐	☐	☐
Starting and stopping quickly	☐	☐	☐	☐	☐	☐
Cutting/lateral Movements	☐	☐	☐	☐	☐	☐
Ability to perform Activity with your Normal technique	☐	☐	☐	☐	☐	☐
Ability to participate In your desired sport As long as you like	☐	☐	☐	☐	☐	☐

How would you rate your current level of function during your sports related activities from 0 to 100 with 100 being your level of function prior to your foot or ankle problem and 0 being the inability to perform any of your usual daily activities?

__ __ __ . 0%

Overall, how would you rate your current level of function?

☐ Normal ☐ Nearly Normal ☐ Abnormal ☐ Severely Abnormal

Martin, R; Irrgang, J; Burdett, R; Conti, S; VanSwearingen, J: Evidence of Validity for the Foot and Ankle Ability Measure. Foot and Ankle International. Vol.26, No.11: 968-983, 2005.

Ankle Instability Instrument

Instructions

This form will be used to categorize your ankle instability. A separate form should be used for the right and left ankles. Please fill out the form completely. If you have any questions, please ask the administrator of the survey. Thank you for your participation.

1. Have you ever sprained an ankle? ☐ Yes ☐ No
2. Have your ever seen a doctor for an ankle sprain? ☐ Yes ☐ No

   If yes,

   2a. How did the doctor categorize your most serious ankle sprain?

   ☐ Mild (grade I) ☐ Moderate (grade 2) ☐ Severe (grade 3)
3. Did you ever use a device (such as crutches) because you could not bear weight due to an ankle sprain? ☐ Yes ☐ No

   If yes,

   3a. In the most serious case, how long did you need to use the device?

   ☐ 1–3 days ☐ 4–7 days ☐ 1–2 weeks ☐ 2–3 weeks ☐ >3 weeks
4. Have you ever experienced a sensation of your ankle "giving way"? ☐ Yes ☐ No

   If yes,

   4a. When was the last time your ankle "gave way"?

   ☐ <1 month ☐ 1–6 months ago ☐ 6–12 months ago ☐ 1–2 years ago ☐ >2 years
5. Does your ankle ever feel unstable while walking on a flat surface? ☐ Yes ☐ No
6. Does your ankle ever feel unstable while walking on uneven ground? ☐ Yes ☐ No
7. Does your ankle ever feel unstable during recreational or sport activity? ☐ Yes ☐ No ☐ N/A
8. Does your ankle ever feel unstable while going *up* stairs? ☐ Yes ☐ No
9. Does your ankle ever feel unstable while going *down* stairs? ☐ Yes ☐ No

Docherty et al., (2006)

Appendix B

Single-Leg Hop ANOVA Results

Mean Relative Phase (MRP)

Three-way ANOVA Summary: MRP Hop Landing

Source	**MS**	**F**	**df (Effect, Residual)**	**p**	**η_p^2**
Within-Subjects					
Condition	.011	2.49	1, 45	.121	.053
Coupling	16.01	135.44	1.68, 75.98	.000	.751
Condition * Coupling	.010	1.37	2.52, 113.76	.257	.030
Condition * Group	.005	1.19	2, 45	.313	.050
Coupling * Group	.049	.97	8, 180	.456	.042
Condition * Coupling * Group	.006	1.38	8, 180	.204	.058
Between-Subjects					
Group	.099	1.97	2,45	.150	.081

Three-way ANOVA Summary: MRP Hop Propulsion

Source	**MS**	**F**	**df (Effect, Residual)**	**p**	**η_p^2**
Within-Subjects					
Condition	.000	.11	1, 45	.732	.003
Coupling	20.44	287.21	1.99, 89.92	.000	.865
Condition * Coupling	.004	1.06	1.96, 88.45	.350	.023
Condition * Group	.002	.87	2, 45	.425	.037
Coupling * Group	.019	.54	8, 180	.823	.024
Condition * Coupling * Group	.002	1.22	8, 180	.288	.052
Between-Subjects					
Group	.023	.60	2, 45	.550	.026

Three-way ANOVA Summary: MRP Hop Flight

Source	**MS**	**F**	**df (Effect, Residual)**	**p**	**η_p^2**
Within-Subjects					
Condition	.002	.35	1, 45	.557	.008
Coupling	3.55	62.23	2.28, 102.59	.000	.580
Condition * Coupling	.048	6.21	2.69, 121.08	.001	.121
Condition * Group	.001	.24	2, 45	.785	.011
Coupling * Group	.020	.61	8, 180	.764	.027
Condition * Coupling * Group	.018	3.45	8, 180	.001	.133
Between-Subjects					
Group	.011	.35	2, 45	.702	.016

Relative Phase Deviation (RPD)

Three-way ANOVA Summary: RPD Hop Landing

Source	**MS**	**F**	**df (Effect, Residual)**	**p**	**η_p^2**
Within-Subjects					
Condition	.025	5.43	1, 45	.024	.108
Coupling	6.45	80.23	1.83, 82.52	.000	.641
Condition * Coupling	.002	.55	3.04, 137.02	.650	.012
Condition * Group	.012	2.68	2, 45	.079	.107
Coupling * Group	.020	.53	8, 180	.827	.023
Condition * Coupling * Group	.002	.69	8, 180	.693	.030
Between-Subjects					
Group	.063	.63	2, 45	.535	.027

Three-way ANOVA Summary: RPD Hop Propulsion

Source	**MS**	**F**	**df (Effect, Residual)**	**p**	**η_p^2**
Within-Subjects					
Condition	.011	1.74	1, 45	.193	.037
Coupling	5.31	114.70	2.82, 127.22	.000	.718
Condition * Coupling	.006	1.39	2.61, 117.63	.249	.030
Condition * Group	.010	1.58	2, 45	.216	.066
Coupling * Group	.011	.32	8, 180	.955	.014
Condition * Coupling * Group	.004	1.41	8, 180	.191	.059
Between-Subjects					
Group	.002	.010	2, 45	.990	.000

Three-way ANOVA Summary: RPD Hop Flight

Source	**MS**	**F**	**df (Effect, Residual)**	**p**	**η_p^2**
Within-Subjects					
Condition	.003	.94	1, 45	.336	.021
Coupling	.44	44.74	2.92, 131.39	.000	.499
Condition * Coupling	.006	2.88	3.30, 148.80	.033	.060
Condition * Group	.003	1.04	2, 45	.359	.045
Coupling * Group	.003	.37	8, 180	.934	.016
Condition * Coupling * Group	.002	1.28	8, 180	.255	.054
Between-Subjects					
Group	.005	.19	2, 45	.820	.009

Percent Normalized Electromyography (% EMG)

Three-way ANOVA Summary: EMG Hop Landing

Source	**MS**	**F**	**df (Effect, Residual)**	**p**	**η_p^2**
Within-Subjects					
Condition	.001	.26	1, 45	.611	.006
Muscle	13.60	163.59	2.61, 117.62	.000	.784
Condition * Muscle	.004	2.60	2.16, 97.50	.075	.055
Condition * Group	.006	1.54	2, 45	.224	.064
Muscle * Group	.083	1.14	6, 135	.339	.048
Condition * Muscle * Group	.001	.72	6, 135	.630	.031
Between-Subjects					
Group	.062	.38	2, 45	.682	.017

Three-way ANOVA Summary: EMG Hop Propulsion

Source	**MS**	**F**	**df (Effect, Residual)**	**p**	**η_p^2**
Within-Subjects					
Condition	.015	2.41	1, 45	.127	.051
Muscle	3.82	73.20	3, 135	.000	.619
Condition * Muscle	.003	2.42	3, 135	.069	.051
Condition * Group	.008	1.31	2, 45	.278	.055
Muscle * Group	.033	.63	6, 135	.698	.028
Condition * Muscle * Group	.001	1.05	6, 135	.391	.045
Between-Subjects					
Group	.14	.58	2, 45	.563	.025

Three-way ANOVA Summary: EMG Hop Flight

Source	**MS**	**F**	**df (Effect, Residual)**	**p**	**η_p^2**
Within-Subjects					
Condition	.000	.059	1, 45	.809	.001
Muscle	4.71	70.52	3, 135	.000	.610
Condition * Muscle	.000	.21	2.03, 91.58	.810	.005
Condition * Group	.003	.92	2, 45	.403	.040
Muscle * Group	.088	1.31	6, 135	.253	.055
Condition * Muscle * Group	.000	.34	6, 135	.914	.015
Between-Subjects					
Group	.34	1.38	2, 45	.262	.058

Walking ANOVA Results

Mean Relative Phase (MRP)

Three-way ANOVA Summary: MRP FS-Midstance

Source	**MS**	**F**	**df (Effect, Residual)**	**p**	**η_p^2**
Within-Subjects					
Condition	5.81E-5	.039	1, .002	.845	.001
Coupling	66.72	656.56	1.16, 52.46	.000	.936
Condition * Coupling	.037	11.00	1.39, 62.862	.000	.196
Condition * Group	.000	.22	2, 45	.800	.010
Coupling * Group	.003	.10	8, 180	.999	.005
Condition * Coupling * Group	.001	.53	8, 180	.832	.023
Between-Subjects					
Group	.005	.16	2, 45	.852	.007

Three-way ANOVA Summary: MRP TO - Midswing

Source	**MS**	**F**	**df (Effect, Residual)**	**p**	**η_p^2**
Within-Subjects					
Condition	.016	5.67	1, 45	.021	.112
Coupling	16.36	453.76	2.49, 112.25	.000	.910
Condition * Coupling	.19	37.03	2.23, 100.64	.000	.451
Condition * Group	.008	2.64	2, 45	.082	.105
Coupling * Group	.024	1.06	8, 180	.394	.045
Condition * Coupling * Group	.002	.75	8, 180	.646	.032
Between-Subjects					
Group	.022	.85	2, 45	.433	.037

Three-way ANOVA Summary: MRP Terminal Swing

Source	**MS**	**F**	**df (Effect, Residual)**	**p**	**η_p^2**
Within-Subjects					
Condition	.063	3.36	1, 45	.073	.070
Coupling	107.23	754.42	2.11, 95.06	.000	.944
Condition * Coupling	.13	7.90	2.49, 112.39	.000	.149
Condition * Group	.004	.22	2, 45	.798	.010
Coupling * Group	.029	.38	8, 180	.928	.017
Condition * Coupling * Group	.012	1.08	8, 180	.374	.046
Between-Subjects					
Group	.10	.93	2, 45	.400	.040

Relative Phase Deviation (RPD)

Three-way ANOVA Summary: RPD FS-Midstance

Source	MS	F	df (Effect, Residual)	p	η_p^2
Within-Subjects					
Condition	.001	.47	1, 45	.493	.011
Coupling	5.23	216.56	2.40, 108.36	.000	.828
Condition * Coupling	.002	.83	2.55, 115.15	.461	.018
Condition * Group	.004	2.86	2, 45	.067	.113
Coupling * Group	.019	1.32	8, 180	.234	.056
Condition * Coupling * Group	.002	1.19	8, 180	.303	.051
Between-Subjects					
Group	.031	.95	2, 45	.391	.041

Three-way ANOVA Summary: RPD TO - Midswing

Source	MS	F	df (Effect, Residual)	p	η_p^2
Within-Subjects					
Condition	.42	108.32	1, 45	.000	.706
Coupling	20.64	688.40	2.89, 130.15	.000	.939
Condition * Coupling	.31	83.56	2.90, 130.49	.000	.650
Condition * Group	.004	1.00	2, 45	.374	.043
Coupling * Group	.012	.54	8, 180	.823	.024
Condition * Coupling * Group	.004	1.52	8, 180	.152	.063
Between-Subjects					
Group	.009	.17	2, 45	.845	.007

Three-way ANOVA Summary: RPD Terminal Swing

Source	MS	F	df (Effect, Residual)	p	η_p^2
Within-Subjects					
Condition	.10	10.08	1, 45	.003	.183
Coupling	18.05	209.79	2.42, 109.14	.000	.823
Condition * Coupling	.085	9.49	2.57, 115.94	.000	.174
Condition * Group	.002	.23	2, 45	.796	.010
Coupling * Group	.063	1.20	8, 180	.301	.051
Condition * Coupling * Group	.003	.49	8, 180	.857	.022
Between-Subjects					
Group	.14	.99	2, 45	.380	.042

Percent Normalized Electromyography (% EMG)

Three-way ANOVA Summary: EMG FS-Midstance

Source	**MS**	**F**	**df (Effect, Residual)**	**p**	**η_p^2**
Within-Subjects					
Condition	.007	26.83	1, 45	.000	.374
Muscle	1.08	11.18	2.40, 108.04	.000	.199
Condition * Muscle	.001	2.80	2.02, 91.10	.065	.059
Condition * Group	.000	1.22	2, 45	.303	.052
Muscle * Group	.048	.61	6, 135	.714	.027
Condition * Muscle * Group	.000	.74	6, 135	.612	.032
Between-Subjects					
Group	.061	.36	2, 45	.698	.016

Three-way ANOVA Summary: EMG TO - Midswing

Source	**MS**	**F**	**df (Effect, Residual)**	**p**	**η_p^2**
Within-Subjects					
Condition	.002	1.16	1, 45	.286	.025
Muscle	7.58	103.98	3, 135	.000	.698
Condition * Muscle	.001	.76	2.08, 93.61	.471	.017
Condition * Group	.002	1.07	2, 45	.350	.046
Muscle * Group	.086	1.17	6, 135	.324	.050
Condition * Muscle * Group	.000	.31	6, 135	.926	.014
Between-Subjects					
Group	.064	.34	2, 45	.711	.015

Three-way ANOVA Summary: EMG Terminal Swing

Source	**MS**	**F**	**df (Effect, Residual)**	**p**	**η_p^2**
Within-Subjects					
Condition	.003	3.31	1, 45	.075	.069
Muscle	8.06	112.57	3, 135	.000	.714
Condition * Muscle	.002	3.66	2.23, 100.34	.025	.075
Condition * Group	.000	.30	2, 45	.742	.013
Muscle * Group	.067	.92	6, 135	.477	.040
Condition * Muscle * Group	.000	.56	6, 135	.757	.025
Between-Subjects					
Group	.053	.31	2, 45	.729	.014

Appendix C

Pairwise Comparisons for Coupling and Muscle Main Effects

Table C -1: Pairwise Comparisons MRP Hop Landing.

(I) coupling	(J) coupling	Mean Difference (I-J)	Std. Error	Sig.[b]	95% Confidence Interval for Difference[b]	
					Lower Bound	Upper Bound
1	2	.634*	.041	.000	.552	.716
	3	.025*	.009	.008	.007	.043
	4	.226*	.033	.000	.160	.292
	5	.049*	.013	.000	.023	.075
2	1	-.634*	.041	.000	-.716	-.552
	3	-.609*	.037	.000	-.684	-.534
	4	-.408*	.031	.000	-.471	-.345
	5	-.585*	.048	.000	-.682	-.487
3	1	-.025*	.009	.008	-.043	-.007
	2	.609*	.037	.000	.534	.684
	4	.201*	.032	.000	.137	.265
	5	.024	.016	.145	-.009	.057
4	1	-.226*	.033	.000	-.292	-.160
	2	.408*	.031	.000	.345	.471
	3	-.201*	.032	.000	-.265	-.137
	5	-.177*	.038	.000	-.253	-.100
5	1	-.049*	.013	.000	-.075	-.023
	2	.585*	.048	.000	.487	.682
	3	-.024	.016	.145	-.057	.009
	4	.177*	.038	.000	.100	.253

Table C-1: Pairwise Comparisons MRP Hop Landing. 1 = *Foot dorsiflexion/plantarflexion – Shank flexion/extension*, 2 = *Foot dorsiflexion/plantarflexion – Thigh flexion/extension*, 3 = *Foot inversion/eversion – Shank internal/external rotation*, 4 = *Foot inversion/eversion – Thigh abduction/adduction*, 5 = *Shank flexion/extension – Thigh flexion/extension; * = the mean difference is significant at the .05 level;* b = *Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).*

Table C-2: Pairwise Comparisons MRP Hop Propulsion

(I) coupling	(J) coupling	Mean Difference (I-J)	Std. Error	Sig.[b]	95% Confidence Interval for Difference[b]	
					Lower Bound	Upper Bound
1	2	.743*	.036	.000	.670	.816
	3	-.010	.011	.379	-.032	.012
	4	.221*	.035	.000	.151	.292
	5	-.018	.011	.109	-.040	.004
2	1	-.743*	.036	.000	-.816	-.670
	3	-.752*	.028	.000	-.809	-.696
	4	-.521*	.032	.000	-.586	-.456
	5	-.761*	.029	.000	-.818	-.703
3	1	.010	.011	.379	-.012	.032
	2	.752*	.028	.000	.696	.809
	4	.231*	.031	.000	.168	.294
	5	-.008	.005	.077	-.017	.001
4	1	-.221*	.035	.000	-.292	-.151
	2	.521*	.032	.000	.456	.586
	3	-.231*	.031	.000	-.294	-.168
	5	-.240*	.031	.000	-.302	-.177
5	1	.018	.011	.109	-.004	.040
	2	.761*	.029	.000	.703	.818
	3	.008	.005	.077	-.001	.017
	4	.240*	.031	.000	.177	.302

Table C-2: Pairwise Comparisons MRP Hop Propulsion. 1 = *Foot dorsiflexion/plantarflexion – Shank flexion/extension*, 2 = *Foot dorsiflexion/plantarflexion – Thigh flexion/extension*, 3 = *Foot inversion/eversion – Shank internal/external rotation*, 4 = *Foot inversion/eversion – Thigh abduction/adduction*, 5 = *Shank flexion/extension – Thigh flexion/extension; * = the mean difference is significant at the .05 level;* b = *Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).*

Table C-3: Pairwise Comparisons MRP Hop Flight

(I) coupling	(J) coupling	Mean Difference (I-J)	Std. Error	Sig.[b]	95% Confidence Interval for Difference[b] Lower Bound	Upper Bound
1	2	.015	.054	.781	-.099	.130
	3	.046	.038	.243	-.035	.128
	4	.036	.046	.447	-.063	.135
	5	.007	.033	.829	-.064	.078
2	1	-.015	.054	.781	-.130	.099
	3	.031	.032	.341	-.036	.099
	4	.021	.039	.600	-.063	.105
	5	-.008	.032	.806	-.075	.060
3	1	-.046	.038	.243	-.128	.035
	2	-.031	.032	.341	-.099	.036
	4	-.010	.026	.703	-.065	.045
	5	-.039	.026	.151	-.094	.016
4	1	-.036	.046	.447	-.135	.063
	2	-.021	.039	.600	-.105	.063
	3	.010	.026	.703	-.045	.065
	5	-.029	.041	.490	-.116	.058
5	1	-.007	.033	.829	-.078	.064
	2	.008	.032	.806	-.060	.075
	3	.039	.026	.151	-.016	.094
	4	.029	.041	.490	-.058	.116

Table C-3: Pairwise Comparisons MRP Hop Flight. 1 = *Foot dorsiflexion/plantarflexion – Shank flexion/extension*, 2 = *Foot dorsiflexion/plantarflexion – Thigh flexion/extension*, 3 = *Foot inversion/eversion – Shank internal/external rotation*, 4 = *Foot inversion/eversion – Thigh abduction/adduction*, 5 = *Shank flexion/extension – Thigh flexion/extension; * = the mean difference is significant at the .05 level;* b = *Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).*

Table C-4: Pairwise Comparisons RPD Hop Landing

(I) coupling	(J) coupling	Mean Difference (I-J)	Std. Error	Sig.[b]	95% Confidence Interval for Difference[b] Lower Bound	Upper Bound
1	2	-.152*	.016	.000	-.184	-.119
	3	-.316*	.035	.000	-.386	-.247
	4	-.434*	.035	.000	-.505	-.363
	5	-.087*	.018	.000	-.124	-.050
2	1	.152*	.016	.000	.119	.184
	3	-.165*	.031	.000	-.227	-.103
	4	-.283*	.029	.000	-.340	-.225
	5	.064*	.014	.000	.036	.093
3	1	.316*	.035	.000	.247	.386
	2	.165*	.031	.000	.103	.227
	4	-.118*	.019	.000	-.156	-.079
	5	.229*	.034	.000	.161	.297
4	1	.434*	.035	.000	.363	.505
	2	.283*	.029	.000	.225	.340
	3	.118*	.019	.000	.079	.156
	5	.347*	.034	.000	.278	.415
5	1	.087*	.018	.000	.050	.124
	2	-.064*	.014	.000	-.093	-.036
	3	-.229*	.034	.000	-.297	-.161
	4	-.347*	.034	.000	-.415	-.278

Table C-4: Pairwise Comparisons RPD Hop Landing. 1 = *Foot dorsiflexion/plantarflexion – Shank flexion/extension*, 2 = *Foot dorsiflexion/plantarflexion – Thigh flexion/extension*, 3 = *Foot inversion/eversion – Shank internal/external rotation*, 4 = *Foot inversion/eversion – Thigh abduction/adduction*, 5 = *Shank flexion/extension – Thigh flexion/extension; * = the mean difference is significant at the .05 level;* b = *Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).*

Table C-5: Pairwise Comparisons RPD Hop Propulsion

(I) coupling	(J) coupling	Mean Difference (I-J)	Std. Error	Sig.[b]	95% Confidence Interval for Difference[b]	
					Lower Bound	Upper Bound
1	2	-.115*	.011	.000	-.138	-.092
	3	-.066*	.031	.039	-.128	-.004
	4	-.375*	.032	.000	-.439	-.312
	5	.168*	.025	.000	.118	.219
2	1	.115*	.011	.000	.092	.138
	3	.049	.028	.086	-.007	.105
	4	-.260*	.027	.000	-.314	-.207
	5	.283*	.018	.000	.246	.320
3	1	.066*	.031	.039	.004	.128
	2	-.049	.028	.086	-.105	.007
	4	-.309*	.030	.000	-.369	-.250
	5	.234*	.028	.000	.178	.291
4	1	.375*	.032	.000	.312	.439
	2	.260*	.027	.000	.207	.314
	3	.309*	.030	.000	.250	.369
	5	.544*	.025	.000	.494	.593
5	1	-.168*	.025	.000	-.219	-.118
	2	-.283*	.018	.000	-.320	-.246
	3	-.234*	.028	.000	-.291	-.178
	4	-.544*	.025	.000	-.593	-.494

Table C-5: Pairwise Comparisons RPD Hop Propulsion. 1 = *Foot dorsiflexion/plantarflexion – Shank flexion/extension*, 2 = *Foot dorsiflexion/plantarflexion – Thigh flexion/extension*, 3 = *Foot inversion/eversion – Shank internal/external rotation*, 4 = *Foot inversion/eversion – Thigh abduction/adduction*, 5 = *Shank flexion/extension – Thigh flexion/extension; * = the mean difference is significant at the .05 level;* b = *Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).*

Table C-6: Pairwise Comparisons EMG Hop Landing

(I) muscle	(J) muscle	Mean Difference (I-J)	Std. Error	Sig.[b]	95% Confidence Interval for Difference[b]	
					Lower Bound	Upper Bound
1	2	.583*	.045	.000	.492	.674
	3	-.243*	.041	.000	-.325	-.161
	4	.228*	.031	.000	.165	.291
2	1	-.583*	.045	.000	-.674	-.492
	3	-.826*	.044	.000	-.915	-.737
	4	-.355*	.037	.000	-.430	-.280
3	1	.243*	.041	.000	.161	.325
	2	.826*	.044	.000	.737	.915
	4	.471*	.032	.000	.406	.537
4	1	-.228*	.031	.000	-.291	-.165
	2	.355*	.037	.000	.280	.430
	3	-.471*	.032	.000	-.537	-.406

Table C-6: Pairwise Comparisons EMG Hop Landing. 1 = *Peroneus Longus*, 2 = *Tibialis Anterior*, 3 = *Lateral Gastrocnemius*, 4 = *Gluteus Medius*; * = *the mean difference is significant at the .05 level;* b = *Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).*

Table C-7: Pairwise Comparisons EMG Hop Propulsion

(I) muscle	(J) muscle	Mean Difference (I-J)	Std. Error	Sig.[b]	95% Confidence Interval for Difference[b] Lower Bound	Upper Bound
1	2	.222*	.031	.000	.159	.286
	3	-.160*	.034	.000	-.229	-.091
	4	.268*	.030	.000	.207	.329
2	1	-.222*	.031	.000	-.286	-.159
	3	-.382*	.031	.000	-.445	-.319
	4	.046	.036	.215	-.028	.119
3	1	.160*	.034	.000	.091	.229
	2	.382*	.031	.000	.319	.445
	4	.428*	.034	.000	.360	.496
4	1	-.268*	.030	.000	-.329	-.207
	2	-.046	.036	.215	-.119	.028
	3	-.428*	.034	.000	-.496	-.360

Table C-7: Pairwise Comparisons EMG Hop Propulsion. 1 = *Peroneus Longus*, 2 = *Tibialis Anterior*, 3 = *Lateral Gastrocnemius*, 4 = *Gluteus Medius*; * = *the mean difference is significant at the .05 level;* b = *Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).*

Table C-8: Pairwise Comparisons EMG Hop Flight

(I) muscle	(J) muscle	Mean Difference (I-J)	Std. Error	Sig.[b]	95% Confidence Interval for Difference[b]	
					Lower Bound	Upper Bound
1	2	.294*	.039	.000	.217	.372
	3	-.228*	.038	.000	-.304	-.152
	4	.140*	.031	.000	.077	.203
2	1	-.294*	.039	.000	-.372	-.217
	3	-.522*	.043	.000	-.609	-.435
	4	-.154*	.040	.000	-.235	-.074
3	1	.228*	.038	.000	.152	.304
	2	.522*	.043	.000	.435	.609
	4	.368*	.032	.000	.303	.432
4	1	-.140*	.031	.000	-.203	-.077
	2	.154*	.040	.000	.074	.235
	3	-.368*	.032	.000	-.432	-.303

Table C-8: Pairwise Comparisons EMG Hop Flight. 1 = *Peroneus Longus*, 2 = *Tibialis Anterior*, 3 = *Lateral Gastrocnemius*, 4 = *Gluteus Medius*; * = *the mean difference is significant at the .05 level;* b = *Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).*

Table C-9: Pairwise Comparisons RPD FC to Midstance

(I) coupling	(J) coupling	Mean Difference (I-J)	Std. Error	Sig.[b]	95% Confidence Interval for Difference[b]	
					Lower Bound	Upper Bound
1	2	-.462*	.010	.000	-.483	-.441
	3	-.295*	.015	.000	-.326	-.264
	4	-.379*	.023	.000	-.426	-.333
	5	-.394*	.014	.000	-.422	-.366
2	1	.462*	.010	.000	.441	.483
	3	.167*	.016	.000	.135	.199
	4	.083*	.021	.000	.040	.125
	5	.068*	.005	.000	.057	.079
3	1	.295*	.015	.000	.264	.326
	2	-.167*	.016	.000	-.199	-.135
	4	-.084*	.021	.000	-.126	-.042
	5	-.099*	.018	.000	-.135	-.062
4	1	.379*	.023	.000	.333	.426
	2	-.083*	.021	.000	-.125	-.040
	3	.084*	.021	.000	.042	.126
	5	-.015	.022	.500	-.059	.029
5	1	.394*	.014	.000	.366	.422
	2	-.068*	.005	.000	-.079	-.057
	3	.099*	.018	.000	.062	.135
	4	.015	.022	.500	-.029	.059

Table C-9: Pairwise Comparisons RPD FC to Midstance. 1 = *Foot dorsiflexion/plantarflexion – Shank flexion/extension*, 2 = *Foot dorsiflexion/plantarflexion – Thigh flexion/extension*, 3 = *Foot inversion/eversion – Shank internal/external rotation*, 4 = *Foot inversion/eversion – Thigh abduction/adduction*, 5 = *Shank flexion/extension – Thigh flexion/extension; * = the mean difference is significant at the .05 level;* b = *Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).*

Table C-10: Pairwise Comparisons EMG FC to Midstance

(I) muscle	(J) muscle	Mean Difference (I-J)	Std. Error	Sig.[b]	95% Confidence Interval for Difference[b]	
					Lower Bound	Upper Bound
1	2	-.091*	.024	.000	-.140	-.043
	3	-.154*	.043	.001	-.241	-.066
	4	.060	.040	.145	-.021	.141
2	1	.091*	.024	.000	.043	.140
	3	-.063	.040	.125	-.143	.018
	4	.151*	.038	.000	.074	.229
3	1	.154*	.043	.001	.066	.241
	2	.063	.040	.125	-.018	.143
	4	.214*	.050	.000	.113	.314
4	1	-.060	.040	.145	-.141	.021
	2	-.151*	.038	.000	-.229	-.074
	3	-.214*	.050	.000	-.314	-.113

Table C-10: Pairwise Comparisons EMG FC to Midstance. 1 = *Peroneus Longus*, 2 = *Tibialis Anterior*, 3 = *Lateral Gastrocnemius*, 4 = *Gluteus Medius*; * = *the mean difference is significant at the .05 level;* b = *Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).*

Table C-11: Pairwise Comparisons EMG TO to Midswing

(I) muscle	(J) muscle	Mean Difference (I-J)	Std. Error	Sig.[b]	95% Confidence Interval for Difference[b]	
					Lower Bound	Upper Bound
1	2	.390*	.038	.000	.314	.467
	3	-.249*	.040	.000	-.330	-.169
	4	.244*	.037	.000	.171	.318
2	1	-.390*	.038	.000	-.467	-.314
	3	-.640*	.035	.000	-.710	-.569
	4	-.146*	.044	.002	-.234	-.058
3	1	.249*	.040	.000	.169	.330
	2	.640*	.035	.000	.569	.710
	4	.493*	.040	.000	.413	.574
4	1	-.244*	.037	.000	-.318	-.171
	2	.146*	.044	.002	.058	.234
	3	-.493*	.040	.000	-.574	-.413

Table C-11: Pairwise Comparisons EMG TO to Midswing. 1 = *Peroneus Longus*, 2 = *Tibialis Anterior*, 3 = *Lateral Gastrocnemius*, 4 = *Gluteus Medius*; * = *the mean difference is significant at the .05 level;* b = *Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).*

Appendix D

Post Hoc Results for Condition x Coupling and Condition x Muscle Interactions

Table D-1: Pairwise Comparisons MRP Coper Hop Flight

(I) coupling	(J) coupling	Mean Difference (I-J)	Std. Error	Sig.[c]	95% Confidence Interval for Difference[c]	
					Lower Bound	Upper Bound
1	2	-.142*	.057	.025	-.264	-.020
	3	-.081*	.032	.022	-.148	-.013
	4	-.035	.029	.246	-.097	.027
	5	-.046	.031	.158	-.112	.020
2	1	.142*	.057	.025	.020	.264
	3	.061	.038	.132	-.021	.143
	4	.107*	.045	.030	.012	.202
	5	.096	.046	.054	-.002	.194
3	1	.081*	.032	.022	.013	.148
	2	-.061	.038	.132	-.143	.021
	4	.046	.030	.150	-.019	.110
	5	.035	.025	.181	-.018	.088
4	1	.035	.029	.246	-.027	.097
	2	-.107*	.045	.030	-.202	-.012
	3	-.046	.030	.150	-.110	.019
	5	-.011	.039	.786	-.093	.072
5	1	.046	.031	.158	-.020	.112
	2	-.096	.046	.054	-.194	.002
	3	-.035	.025	.181	-.088	.018
	4	.011	.039	.786	-.072	.093

Table D-1: Pairwise Comparisons MRP Coper Hop Flight. 1 = *Foot dorsiflexion/plantarflexion – Shank flexion/extension*, 2 = *Foot dorsiflexion/plantarflexion – Thigh flexion/extension*, 3 = *Foot inversion/eversion – Shank internal/external rotation*, 4 = *Foot inversion/eversion – Thigh abduction/adduction*, 5 = *Shank flexion/extension – Thigh flexion/extension; * = the mean difference is significant at the .05 level;* c = *Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).*

Table D-2: Pairwise Comparisons MRP Non-Coper Hop Flight.

(I) coupling	(J) coupling	Mean Difference (I-J)	Std. Error	Sig.[c]	95% Confidence Interval for Difference[c]	
					Lower Bound	Upper Bound
1	2	-.182*	.042	.001	-.270	-.093
	3	-.124*	.034	.002	-.197	-.052
	4	-.166*	.038	.001	-.248	-.084
	5	-.088*	.027	.006	-.146	-.030
2	1	.182*	.042	.001	.093	.270
	3	.058	.032	.090	-.010	.125
	4	.016	.041	.701	-.071	.103
	5	.094*	.020	.000	.052	.136
3	1	.124*	.034	.002	.052	.197
	2	-.058	.032	.090	-.125	.010
	4	-.042	.023	.090	-.091	.007
	5	.036	.022	.116	-.010	.083
4	1	.166*	.038	.001	.084	.248
	2	-.016	.041	.701	-.103	.071
	3	.042	.023	.090	-.007	.091
	5	.078*	.032	.028	.010	.147
5	1	.088*	.027	.006	.030	.146
	2	-.094*	.020	.000	-.136	-.052
	3	-.036	.022	.116	-.083	.010
	4	-.078*	.032	.028	-.147	-.010

Table D-2: Pairwise Comparisons MRP Non-Coper Hop Flight. 1 = *Foot dorsiflexion/plantarflexion – Shank flexion/extension*, 2 = *Foot dorsiflexion/plantarflexion – Thigh flexion/extension*, 3 = *Foot inversion/eversion – Shank internal/external rotation*, 4 = *Foot inversion/eversion – Thigh abduction/adduction*, 5 = *Shank flexion/extension – Thigh flexion/extension; * = the mean difference is significant at the .05 level;* c = *Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).*

Table D-3: Pairwise Comparisons RPD Hop Flight.

(I) coupling	(J) coupling	Mean Difference (I-J)	Std. Error	Sig.[b]	95% Confidence Interval for Difference[b]	
					Lower Bound	Upper Bound
1	2	-.008	.010	.437	-.029	.013
	3	-.036*	.012	.003	-.060	-.013
	4	-.021	.012	.088	-.045	.003
	5	-.027	.014	.055	-.055	.001
2	1	.008	.010	.437	-.013	.029
	3	-.028*	.011	.012	-.050	-.007
	4	-.013	.012	.295	-.036	.011
	5	-.019	.014	.165	-.046	.008
3	1	.036*	.012	.003	.013	.060
	2	.028*	.011	.012	.007	.050
	4	.016	.009	.086	-.002	.033
	5	.009	.013	.500	-.017	.035
4	1	.021	.012	.088	-.003	.045
	2	.013	.012	.295	-.011	.036
	3	-.016	.009	.086	-.033	.002
	5	-.007	.015	.658	-.036	.023
5	1	.027	.014	.055	-.001	.055
	2	.019	.014	.165	-.008	.046
	3	-.009	.013	.500	-.035	.017
	4	.007	.015	.658	-.023	.036

Table D-3: Pairwise Comparisons RPD Hop Flight. 1 = *Foot dorsiflexion/plantarflexion – Shank flexion/extension*, 2 = *Foot dorsiflexion/plantarflexion – Thigh flexion/extension*, 3 = *Foot inversion/eversion – Shank internal/external rotation*, 4 = *Foot inversion/eversion – Thigh abduction/adduction*, 5 = *Shank flexion/extension – Thigh flexion/extension; * = the mean difference is significant at the .05 level;* c = *Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).*

Table D-4: Pairwise Comparisons MRP FC to Midstance.

(I) coupling	(J) coupling	Mean Difference (I-J)	Std. Error	Sig.[b]	95% Confidence Interval for Difference[b]	
					Lower Bound	Upper Bound
1	2	-.022*	.006	.000	-.033	-.011
	3	-.030*	.004	.000	-.039	-.021
	4	-.029*	.006	.000	-.040	-.017
	5	-.064*	.016	.000	-.096	-.033
2	1	.022*	.006	.000	.011	.033
	3	-.008	.005	.105	-.018	.002
	4	-.007	.006	.250	-.019	.005
	5	-.042*	.012	.001	-.066	-.019
3	1	.030*	.004	.000	.021	.039
	2	.008	.005	.105	-.002	.018
	4	.001	.004	.728	-.007	.010
	5	-.034*	.014	.020	-.063	-.006
4	1	.029*	.006	.000	.017	.040
	2	.007	.006	.250	-.005	.019
	3	-.001	.004	.728	-.010	.007
	5	-.036*	.014	.017	-.065	-.007
5	1	.064*	.016	.000	.033	.096
	2	.042*	.012	.001	.019	.066
	3	.034*	.014	.020	.006	.063
	4	.036*	.014	.017	.007	.065

Table D-4: Pairwise Comparisons MRP FC to Midstance. 1 = *Foot dorsiflexion/plantarflexion – Shank flexion/extension*, 2 = *Foot dorsiflexion/plantarflexion – Thigh flexion/extension*, 3 = *Foot inversion/eversion – Shank internal/external rotation*, 4 = *Foot inversion/eversion – Thigh abduction/adduction*, 5 = *Shank flexion/extension – Thigh flexion/extension; * = the mean difference is significant at the .05 level;* c = *Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).*

Table D-5: Pairwise Comparisons MRP TO to Midswing.

(I) coupling	(J) coupling	Mean Difference (I-J)	Std. Error	Sig.[b]	95% Confidence Interval for Difference[b]	
					Lower Bound	Upper Bound
1	2	-.133*	.011	.000	-.155	-.110
	3	-.162*	.021	.000	-.205	-.120
	4	-.166*	.014	.000	-.195	-.138
	5	-.103*	.009	.000	-.121	-.086
2	1	.133*	.011	.000	.110	.155
	3	-.030	.020	.142	-.070	.010
	4	-.034*	.014	.022	-.062	-.005
	5	.029*	.003	.000	.024	.035
3	1	.162*	.021	.000	.120	.205
	2	.030	.020	.142	-.010	.070
	4	-.004	.021	.847	-.046	.038
	5	.059*	.020	.004	.019	.099
4	1	.166*	.014	.000	.138	.195
	2	.034*	.014	.022	.005	.062
	3	.004	.021	.847	-.038	.046
	5	.063*	.014	.000	.036	.091
5	1	.103*	.009	.000	.086	.121
	2	-.029*	.003	.000	-.035	-.024
	3	-.059*	.020	.004	-.099	-.019
	4	-.063*	.014	.000	-.091	-.036

Table D-5: Pairwise Comparisons MRP TO to Midswing. 1 = *Foot dorsiflexion/plantarflexion – Shank flexion/extension*, 2 = *Foot dorsiflexion/plantarflexion – Thigh flexion/extension*, 3 = *Foot inversion/eversion – Shank internal/external rotation*, 4 = *Foot inversion/eversion – Thigh abduction/adduction*, 5 = *Shank flexion/extension – Thigh flexion/extension; * = the mean difference is significant at the .05 level;* c = *Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).*

Table D-6: Pairwise Comparisons MRP Terminal Swing.

(I) coupling	(J) coupling	Mean Difference (I-J)	Std. Error	Sig.[b]	95% Confidence Interval for Difference[b] Lower Bound	Upper Bound
1	2	-.101*	.040	.013	-.181	-.022
	3	-.131*	.035	.001	-.201	-.060
	4	-.157*	.036	.000	-.229	-.084
	5	-.087*	.037	.022	-.160	-.013
2	1	.101*	.040	.013	.022	.181
	3	-.029	.027	.293	-.084	.026
	4	-.055*	.027	.046	-.109	-.001
	5	.015	.008	.058	-.001	.030
3	1	.131*	.035	.001	.060	.201
	2	.029	.027	.293	-.026	.084
	4	-.026	.022	.239	-.070	.018
	5	.044	.028	.120	-.012	.100
4	1	.157*	.036	.000	.084	.229
	2	.055*	.027	.046	.001	.109
	3	.026	.022	.239	-.018	.070
	5	.070*	.028	.017	.013	.127
5	1	.087*	.037	.022	.013	.160
	2	-.015	.008	.058	-.030	.001
	3	-.044	.028	.120	-.100	.012
	4	-.070*	.028	.017	-.127	-.013

Table D-6: Pairwise Comparisons MRP Terminal Swing. 1 = *Foot dorsiflexion/plantarflexion – Shank flexion/extension*, 2 = *Foot dorsiflexion/plantarflexion – Thigh flexion/extension*, 3 = *Foot inversion/eversion – Shank internal/external rotation*, 4 = *Foot inversion/eversion – Thigh abduction/adduction*, 5 = *Shank flexion/extension – Thigh flexion/extension; * = the mean difference is significant at the .05 level;* c = *Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).*

Table D-7: Pairwise Comparisons RPD TO to Midswing.

(I) coupling	(J) coupling	Mean Difference (I-J)	Std. Error	Sig.[b]	95% Confidence Interval for Difference[b]	
					Lower Bound	Upper Bound
1	2	-.121*	.013	.000	-.146	-.095
	3	-.242*	.017	.000	-.276	-.209
	4	-.216*	.019	.000	-.255	-.178
	5	-.186*	.013	.000	-.213	-.159
2	1	.121*	.013	.000	.095	.146
	3	-.122*	.016	.000	-.153	-.091
	4	-.096*	.018	.000	-.132	-.060
	5	-.065*	.014	.000	-.093	-.037
3	1	.242*	.017	.000	.209	.276
	2	.122*	.016	.000	.091	.153
	4	.026*	.011	.020	.004	.048
	5	.057*	.014	.000	.028	.085
4	1	.216*	.019	.000	.178	.255
	2	.096*	.018	.000	.060	.132
	3	-.026*	.011	.020	-.048	-.004
	5	.031*	.015	.047	.000	.061
5	1	.186*	.013	.000	.159	.213
	2	.065*	.014	.000	.037	.093
	3	-.057*	.014	.000	-.085	-.028
	4	-.031*	.015	.047	-.061	.000

Table D-7: Pairwise Comparisons RPD TO to Midswing. 1 = *Foot dorsiflexion/plantarflexion – Shank flexion/extension*, 2 = *Foot dorsiflexion/plantarflexion – Thigh flexion/extension*, 3 = *Foot inversion/eversion – Shank internal/external rotation*, 4 = *Foot inversion/eversion – Thigh abduction/adduction*, 5 = *Shank flexion/extension – Thigh flexion/extension; * = the mean difference is significant at the .05 level;* c = *Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).*

Table D-8: Pairwise Comparisons RPD Terminal Swing.

(I) coupling	(J) coupling	Mean Difference (I-J)	Std. Error	Sig.[b]	95% Confidence Interval for Difference[b]	
					Lower Bound	Upper Bound
1	2	-.125*	.026	.000	-.177	-.073
	3	-.063*	.028	.028	-.119	-.007
	4	-.070*	.025	.008	-.121	-.019
	5	-.105*	.022	.000	-.150	-.060
2	1	.125*	.026	.000	.073	.177
	3	.062*	.022	.008	.017	.107
	4	.055*	.022	.017	.010	.099
	5	.020*	.007	.009	.005	.034
3	1	.063*	.028	.028	.007	.119
	2	-.062*	.022	.008	-.107	-.017
	4	-.007	.016	.660	-.039	.025
	5	-.042*	.020	.045	-.084	-.001
4	1	.070*	.025	.008	.019	.121
	2	-.055*	.022	.017	-.099	-.010
	3	.007	.016	.660	-.025	.039
	5	-.035	.021	.095	-.077	.006
5	1	.105*	.022	.000	.060	.150
	2	-.020*	.007	.009	-.034	-.005
	3	.042*	.020	.045	.001	.084
	4	.035	.021	.095	-.006	.077

Table D-8: Pairwise Comparisons RPD Terminal Swing. 1 = *Foot dorsiflexion/plantarflexion – Shank flexion/extension*, 2 = *Foot dorsiflexion/plantarflexion – Thigh flexion/extension*, 3 = *Foot inversion/eversion – Shank internal/external rotation*, 4 = *Foot inversion/eversion – Thigh abduction/adduction*, 5 = *Shank flexion/extension – Thigh flexion/extension; * = the mean difference is significant at the .05 level;* c = *Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).*

Table D-9: Pairwise Comparisons EMG Terminal Swing.

(I) muscle	(J) muscle	Mean Difference (I-J)	Std. Error	Sig.[b]	95% Confidence Interval for Difference[b]	
					Lower Bound	Upper Bound
1	2	-.010	.007	.192	-.025	.005
	3	-.020*	.006	.004	-.033	-.007
	4	-.015	.008	.059	-.030	.001
2	1	.010	.007	.192	-.005	.025
	3	-.010	.005	.073	-.021	.001
	4	-.005	.004	.264	-.014	.004
3	1	.020*	.006	.004	.007	.033
	2	.010	.005	.073	-.001	.021
	4	.005	.005	.326	-.005	.015
4	1	.015	.008	.059	-.001	.030
	2	.005	.004	.264	-.004	.014
	3	-.005	.005	.326	-.015	.005

Table D-9: Pairwise Comparisons EMG Terminal Swing. 1 = *Peroneus Longus*, 2 = *Tibialis Anterior*, 3 = *Lateral Gastrocnemius*, 4 = *Gluteus Medius*; * = *the mean difference is significant at the .05 level;* c = *Adjustment for multiple comparisons: Least Significant Difference (equivalent to no adjustments).*