SUGIMOTO, YUKI A., Ph.D. The Study of Individual Perception and Neural Control Underlying Movement in Coordinating Postural Control When There Is an Increase in Complexity of Environmental and Task Constraints in CAI Individuals Compared to Healthy Controls. (2021)
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Individuals with Chronic Ankle Instability (CAI) commonly exhibit postural control (stability, adaptation) deficits and altered gait (walking, running) mechanics (Hertel, 2008; Hertel and Corbett, 2019). These impairments in motor behaviors have been hypothesized to be a result of inadequate, yet inherent interactions between individual perception (i.e., sensory systems) and movement (action) integrated at the central nervous system (CNS), resulting in less flexible and adaptable sensorimotor systems. Flexibility and adaptability of sensorimotor systems reflecting on underlying biological noise (movement variability) are critical to coordinate the sensory reweighting system. The sensory reweighting system assigns a relative weight to each sensory system based on the complexity of organismic, environmental, and task constraints to convey redundant and convergent sensory feedback at the CNS. An adequate sensory reweighting system results in sufficient multisensory integration by filtering all potential distractors, the irrelevant sensory information, to the context (e.g., task goals). Successful multisensory integration allows the CNS to integrate the context-relevant sensory information necessary to manage postural control that is the foundation of motor control to achieve suitable performance and adapt to a sudden environmental change. However, the gap exists in the literature to understand the integration phenomenon on how individual elements (i.e., sensory reweighting system, movement variability) contribute to the interaction between perception and movement, especially when environmental and

task constraints increase in the same cohort of participants with and without CAI. Therefore, the primary purpose of this study was to understand the modulation of 1) the sensory reweighting system and postural control, 2) postural adaptation to a sudden change in the environment in the direction of lateral ankle sprain mechanisms, and 3) movement variability, an underlying biological noise pertaining to postural control, when the complexity of environmental and task constraints are manipulated in CAI individuals compared to healthy controls.

A total of 44 physically active individuals, consisting of 22 individuals with CAI (13 females, 9 males; age: 26.09 $\pm$ 5.76 years; height: 172.25 $\pm$ 9.76 cm; weight: 76.18 $\pm$ 14.91 kg) and 22 individuals without CAI (13 females, 9 males; age: 25.41 $\pm$ 5.92 years; height: 169.70 $\pm$ 9.32 cm; weight: 71.98 $\pm$ 14.79 kg) volunteered to participate in this mixed-model repeated-measures study. The NeuroCom Sensory Organization Test (SOT) and Adaptation Test (SMART EquiTest, NeuroCom International Inc., Clackamas, OR) were utilized to examine postural control (equilibrium scores), postural adaptation (sway energy scores), the sensory reweighting system (sensory reweighting ratios), and movement variability (sample entropy) while controlling posture in double- and single-limb (injured, uninjured) stances in individuals with and without CAI. Interestingly, CAI individuals controlled posture very similar to healthy controls. The unique finding of this study was that group differences in the sensory reweighting system depended on both task constraints and sensory systems; CAI individuals upweighted on vestibular feedback when the SOT manipulated somatosensory and visual feedback while controlling posture in the injured-limb. Both groups weighted on somatosensory and visual feedback similarly with continuous emphasis on vision during individual tasks (stance limbs: double, injured, uninjured). Therefore, we contend CAI individuals upweighted on vestibular feedback, which is an independent sole veridical reference to self-motion, when sensory conflicts and task constraints became greater standing in the injured-limb. These findings also imply an effective multisensory integration among CAI.

CAI individuals exhibited respective superior postural adaptation to a sudden environmental change in a support surface with plantarflexion rotation and in the uninjured-limb than healthy controls. Superior postural adaptation is indicative of pre-programmed feedforward motor control. In addition, lower movement variability in postural control was noted in the uninjured- and injured-limbs in CAI. Group differences in movement variability depended on task constraints: those individuals with CAI lowered variability in the uninjured-limb when no sensory feedback was manipulated, and in both the uninjured- and injured-limbs when they were forced to reweight on vestibular feedback with manipulation of somatosensory and visual feedback. Lowered movement variability exhibited with an increase in task constraints in the injured- and uninjured-limbs may be indicative of a mechanism that CAI implemented to provide a boundary to freeze the degree-of-freedom (redundancy in sensory feedback) to achieve effective multisensory integration. Collectively, our findings of superior postural adaptation and lower movement variability in postural control for CAI may imply an existent change in central organization and implementation of supraspinal mechanisms of postural control. Furthermore, postural control, postural adaptation, and movement variability in individuals with and without CAI depended on the environmental or task constraints. Environment- and task-dependent postural control, postural adaptation, and movement variability contribute to motor behaviors throughout the lifespan. Therefore, taking a multisensory-feedback approach by recognizing when to increase environmental and task constraints may optimize rehabilitation intervention to prevent subsequent ankle sprains in individuals with CAI.

THE STUDY OF INDIVIDUAL PERCEPTION AND NEURAL CONTROL
UNDERLYING MOVEMENT IN COORDINATING POSTURAL
CONTROL WHEN THERE IS AN INCREASE IN COMPLEXITY
OF ENVIRONMENTAL AND TASK CONSTRAINTS IN CAI
INDIVIDUALS COMPARED TO HEALTHY CONTROLS

by

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For my parents, *Futoshi* and *Mutsuko*,
who have always propelled me to become an extraordinary human being.

To my grandparents and ancestors,
who worked hard carrying a baton for me to be here and to embrace all of what I am.

APPROVAL PAGE

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"*Gambaru*! - Never ever give up, even, and especially when,
there is no chance of winning."
- *Ann Curry*

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CHAPTER I

INTRODUCTION

Introduction and Statement of Problem

The significance of ankle sprains is overlooked even though it is one of the most common musculoskeletal athletic injuries (McKay, 2001; Fong et al., 2007; Nelson et al., 2007; Waterman et al., 2010; Gribble et al., 2016; Roos et al., 2017). Approximately 23,000 ankle sprains occur daily, and 628,000 ankle sprains are reported annually to the emergency department (Kannus & Renstrom, 1991; Waterman et al., 2010; Gribble et al., 2016). In addition, the annual healthcare costs estimate $6.2 billion for ankle sprain treatments in the United States alone (Knowles et al., 2007; Waterman et al., 2010; Gribble et al., 2016). However, the actual number of ankle sprains and health care costs may be underestimated as greater than 50% of individuals who sustained ankle sprains neglect seeking medical treatment (McKay, 2001; Gribble et al., 2014a). Inadequate treatment and rehabilitation of an initial ankle sprain may contribute to repeated or major injuries (Ekstrand & Gillquist, 1983; McKay, 2001; Murphy et al., 2003). Notably, the current evidence identifies the history of a previous ankle sprain as the strongest predictor for recurrent ankle sprains (Bahr & Bahr, 1997; McKay, 2001; Murphy et al., 2003).

Up to 74% of individuals who sustained an initial ankle sprain experience recurrent ankle sprains and develop lifetime functional disabilities of daily living, such as *chronic ankle instability (CAI)* (Hertel, 2002; Konradsen, 2002; Anandacoomarasamy et al., 2005; Hiller et al., 2011). CAI is not an innocuous injury that can be overlooked because it is the second leading cause of trauma-initiated joint disease, post-traumatic osteoarthritis (PTOA) (Thomas et al., 2017). The early onset of PTOA at the ankle is evident in individuals who have suffered ankle sprains (Lofvenberg et al., 1995; Valderrabano et al., 2006; Golditz, 2014). Approximately 5.6 million clinical cases of the lower extremity PTOA are accounted for in adults aged 25 and older, and the incidence is estimated to double with an increasingly aging population (Thomas et al., 2017). Correspondingly, the lower extremity PTOA is associated with the annual healthcare costs of $11.79 billion in the United States, with direct costs of over $3 billion (Thomas et al., 2017). Therefore, establishing better CAI treatment in conjunction with PTOA prevention is critical to lower the healthcare burden.

CAI individuals who typically suffer pathomechanical impairments (i.e., limited osteokinematic ankle dorsiflexion) from an initial ankle sprain display altered motor behaviors that may expose them to a higher risk of recurrent ankle sprains and subsequent development of PTOA (Hertel, 2008; Hoch et al., 2011, 2012; Hertel & Corbett, 2019). These altered motor behaviors are commonly postural control deficits and maladapted gait. Dysfunctions in postural control with CAI have been presented for several decades in both static (i.e., unipedal stance) and dynamic (i.e., star excursion balance) balance tests compared to healthy individuals (Hertel, 2008; Hertel & Corbett, 2019). Constrained postural control and balance can affect gait biomechanics. Maladapted gait associated with CAI is characterized by less ankle dorsiflexion and excessive inversion (Konradsen, 2002; Nawata et al., 2005; Drewes et al., 2009; Morrison et al., 2010; Schmidt et al., 2011; Chinn et al., 2013; Hoch et al., 2016). Reduced ankle dorsiflexion inhibits the ankle from reaching its stable, close-packed position during the stance phase of gait, failing to provide dynamic joint stability. Likewise, the excessive ankle inversion results in the lateral border of the foot much closer to the ground, setting up for potential collisions during the swing phase of gait (Konradsen, 2002). Additionally, higher pressure under the lateral border of the foot at ground contact increases the moment arm for the ankle into excessive inversion (Hertel, 2002). The subsequent propulsive force on the increased moment arm during the loading phase of gait results in a greater inversion moment, driving the ankle further inward to a vulnerable position and predisposing it to recurrent lateral ankle sprains. In order to prevent the recurrence of ankle sprains, evidence-based clinical evaluation is necessary for organizing effective CAI rehabilitation programs. Therefore, investigating influential factors to motor behaviors and their relationship to those pathomechanical impairments that could be diagnosed with clinical tests may provide new insights into CAI treatments.

There is an inherent dependent relationship between individual perception and movement in coordinating motor behaviors to perform a task goal (Newell et al., 1991; Greeno, 1994; Turvey, 2007; Turvey & Fonseca, 2014). Perception encompasses the body (e.g., sensory systems) and mind (e.g., cognition, emotion), gaining information about the properties of the environment where an individual exists (Greeno, 1994; Hurley, 2001; Turvey, 2007). Individual perception comprises multiple elements that are independent yet functionally redundant at different levels from microscopic (e.g., molecular, cellular, neuronal) to macroscopic (e.g., joint, vision, central organization), known as perceptual redundancy (Newell et al., 1991; Stein, 1998; Davids et al., 2003;

Davids & Glazier, 2010; Glazier, 2017). Perceptual redundancy provides overlapping information to the central nervous system (CNS), then autonomous correction if errors are introduced by one element (Peterka & Loughlin, 2004; Stergiou & Decker, 2011). The CNS integrates the best-suited information (relevant information) necessary for achieving a task goal by filtering all potential distractors (irrelevant information) (Newell, 1991; Greeno, 1994; Horak, 2006; McKeon & Donovan, 2019). The relevance of information is influenced by previous experience in the same or similar task goal (Schmidt, 1975; Newell, 1991; Pacheco et al., 2019). An example of the integration phenomenon of perceptual redundancy is the sensory reweighting system (Nashner & Berthoz, 1978). The sensory reweighting system is the process of the CNS assigning a weight to each sensory feedback (i.e., somatosensory, vision, vestibular) based on its relevance (Horak, 2006). Specifically, the current evidence suggests relevant sensory feedback is given more emphasis (i.e., upweighted) and irrelevant sensory feedback is given less emphasis (i.e., downweighted) (Hwang et al., 2014). Summation in the combination of weighted sensory feedback may determine the accuracy of movement patterns coordinating motor behaviors for a specific task (Peterka, 2002; Stanford et al., 2005). Moreover, flexibility and adaptability of the CNS become crucial to quickly distribute the weight to each sensory feedback, organizing the best combination in achieving a given task (Peterka & Loughlin, 2004; Stergiou & Decker, 2011).

Traditionally, altered motor behaviors in individuals with CAI have been attributed to irrelevant somatosensory feedback, which should have been downweighted, from the foot and ankle complex damaged by an initial ankle sprain (Peterka, 2002;

Hertel, 2008). Despite the conclusion, the recent evidence suggests the existence of the sensory reweighting system in those with CAI. For instance, CAI individuals heavily upweight on the visual sensory feedback, while maintaining posture in a single-limb stance compared to healthy individuals (Song et al., 2016). Yet, altered motor behaviors persist with CAI, even with the pronounced compensatory reliance on the vision acquiring relevant sensory information. The potential assumption for the cause of altered motor behaviors pertains to the flexibility and adaptability of the CNS in distributing the weight to unisensory (i.e., visual sensory feedback), not to each sensory feedback. The unisensory approach to obtaining relevant information may be ambiguous when the environment becomes specific to the sensory system as a primary source (Peterka, 2002). The primary source of relevant information in regards to the spatial orientation of the body while walking in the dark on a stable surface is somatosensory, but it shifts to the vestibular when the surface becomes unstable (Peterka, 2002). Furthermore, the average delay in response to stimuli for the visual sensory system is the longest, estimated as 150 to 200 msec compare to the fastest monosynaptic response (40 to 50 msec), and vestibular which lies somewhat between (Nakamura et al., 1994; Nijhawan, 2008). Under those circumstances, heavy reliance on the visual sensory system may overload the system depending on the consequence of the environment. This is true particularly for these athletes (e.g., soccer) who must partition their vision on the status of opponents and projectiles (e.g., ball) to react to stimuli and organize movements while performing a complex task. However, how the sensory reweighting system contributes to motor behaviors in individuals with CAI is unknown. In essence, the more relevant information an individual can perceive, the greater the precision of motor behaviors will be. Thus, not the dysfunction in one element of perception (e.g., somatosensory deficits) but impaired interplay within multiple elements forming the entire individual perception may tie to altered motor behaviors with CAI.

Perception is context-specific, meaning the definition of irrelevant and relevant information is governed based on three types of constraints and their complexity: organismic that is individual characteristics (e.g., health status), task (e.g., static, dynamic), and environment (e.g., stable, unstable) (Davids et al., 2003). Under normal circumstances, perception constantly develops over time with experience and practice (Newell, 1991; Pacheco et al., 2019). With greater repetition in experience and practice, the task-oriented movement patterns (motor schema) are formed at the CNS (Schmidt, 1975). The movement patterns reduce cognitive demand for the CNS, integrating relevant information from the redundant information collected on the environment at different levels (perceptual redundancy) (Glazier, 2017). Even when new information is perceived in the environment, movement patterns predict relevant information to coordinate motor behaviors. To put it differently, failure in anticipatory motor behaviors to meet the demand of the environment may result in some injuries. Therefore, the determination and correction of perceptual errors to fine-tune movement patterns are important for the CNS in maintaining precise motor behaviors.

The development of new movement requires optimal variation in the neural control in sensorimotor pathways, often referred to as movement variability (Davids et al., 2003). In other words, analyzing movement variability can quantify the state of underlying strategies of movement patterns. Movement patterns are time-dependent that emerge over time, while current movement affects and/or is affected by previous and future movement. Thus, nonlinear variability measures (e.g., sample entropy) that provide a time-evolving aspect of movement patterns are best suited to evaluate movement variability (Harbourne & Stergiou, 2009). Optimal movement variability reveals the flexibility and adaptability of neural control in sensorimotor pathways (Stergiou & Decker, 2011). In detail, healthy individuals have flexible and adaptable neural control in coordinating task goal-oriented motor behaviors, regardless of the type of constraints and their complexity. Conversely, CAI individuals, who have organismic constraints of functional (e.g., decreased movement variability) and structural (e.g., pathomechaincial impairments) insufficiencies from an initial ankle sprain, display less flexible and adaptable neural control as constraints (i.e., task, environment) change, and the level of complexity increases (e.g., task: bipedal to unipedal; environment: stable to unstable). Moreover, less flexible and adaptable neural control in sensorimotor pathways disrupts perception to react to unanticipated changes in environmental and task constraints, resulting in more rigid movement patterns (Stergiou & Decker, 2011). Indeed, CAI individuals exhibit longer postural recovery time from a sudden unexpected ankle inversion perturbation, increasing the risk of recurrent ankle sprains (Hiller et al., 2007). However, the current evidence is limited to understanding how neural control underlying movement modulates the changes in constraints and the level of complexity, contributing to the coordination of motor behaviors in individuals with CAI. Thereupon, investigating neural control in sensorimotor pathways with nonlinear variability measures may provide a better understanding of the interaction between individual perception and movement, generating altered motor behaviors among those individuals with CAI.

Over several decades, evidence-based practice has been emphasized to improve the quality of healthcare. In particular, bridging the gap between laboratory-based findings and clinical practice is necessary, facilitating more effective CAI treatments to prevent a significant rate of recurrent ankle sprains and subsequent development of PTOA. The common pathomechanical impairments (organismic constraints) clinicians are accustomed to examining during the standard clinical evaluation are arthrokinematics abnormalities. Arthrokinematics abnormalities are readily diagnosed with clinical range-of-motion tests such as a reliable weight-bearing lunge test (WBLT) without an expensive piece of laboratory equipment (Bennell et al., 1998). Currently, only a handful of CAI studies have successfully established a positive relationship between the common ankle dorsiflexion range-of-motion deficits displayed in individuals with CAI and their dynamic postural control (Basnett et al., 2013; Gabriner et al., 2015). Chronicity in CAI may cause kinetic alterations at proximal joints (e.g., knee, hip). In support, the most current evidence suggests abnormal hip-ankle coordination is evident in individuals with CAI during the stance phase of gait compared to healthy individuals (Yen et al., 2017). CAI individuals elicit greater hip adduction (lateral shift of pelvis) relative to ankle eversion during loading response (Yen et al., 2017). Laterally shifted pelvis displaces the center-of-mass toward the outside border of the foot, predisposing the ankle for excessive inversion and may delay postural recovery and adaptation to a sudden change in the environment during walking/running or sports, thereby increasing the potential risk of recurrent ankle sprains. In this case, having a well-aligned pelvis in a neutral position among CAI individuals with excessive ankle inversion characteristics may be optimal for controlling center-of-mass oscillations during gait. Despite an increase in anterior pelvic tilt has been suggested to be a significant predictor of the Anterior Cruciate Ligament (knee) injury regardless of sex, no CAI study has examined anterior pelvic tilt with clinical pelvic tilt test with CAI individuals (Hertel et al., 2004). How ankle dorsiflexion range-of-motion deficits and potential proximal hip alterations assessed via clinical WBLT and pelvic tilt test relate to previously discussed individual perception (i.e., sensory reweighting system) and movement (i.e., movement variability), generating altered motor behaviors in those with CAI are unknown. Validating clinical WBLT and pelvic tilt test to the postural recovery, the sensory reweighting system, and movement variability aside from this study may pave an initial step for applying laboratory-based findings into everyday clinical practice.

Statement of Purpose and Hypotheses

Individuals with CAI commonly exhibit postural control (stability, adaptation) deficits and maladapted gait, especially with the increased complexity of environmental and task constraints. These altered motor behaviors may be a result of inadequate, yet inherent interactions between perception and movement, exposing CAI individuals to a higher risk of recurrent ankle sprains and subsequent development of PTOA. Therefore, the primary objective of the current study was to understand the modulation of individual perception and movement in coordinating fundamental motor behaviors of controlling posture when the complexity of the environmental and/or task constraints are manipulated in CAI individuals compared to healthy controls. There are individual elements that contribute to interactions between individual perception and movement in sensorimotor pathways. Consequently, we considered the integration phenomenon of those elements based on the current evidence, examining the sensory reweighting system of perception that assigns a weight to each sensory feedback (somatosensory, visual, vestibular) and variation in the neural control underlying movement. Task complexity was altered by progressing from a double-limb stance to a single-limb stance (injured, uninjured). Whereas environmental complexity was manipulated to affect the sensory feedback with a combination of sway-referenced support and/or surroundings with and without eyes-closed during postural control in double- and single-limb stances. Postural control was examined on the NeuroCom sensory organization test (SOT) (SMART EquiTest, NeuroCom International Inc., Clackamas, OR). The overall theoretical model illustrated in Figure 1.1 presents interactions between individual perception and movement in coordinating motor behaviors and their relationship to pathomechanics diagnosed with clinical tests (i.e., WBLT, pelvic tilt). The current study specifically investigates the interaction between individual perception and movement in postural control in individuals with and without CAI.

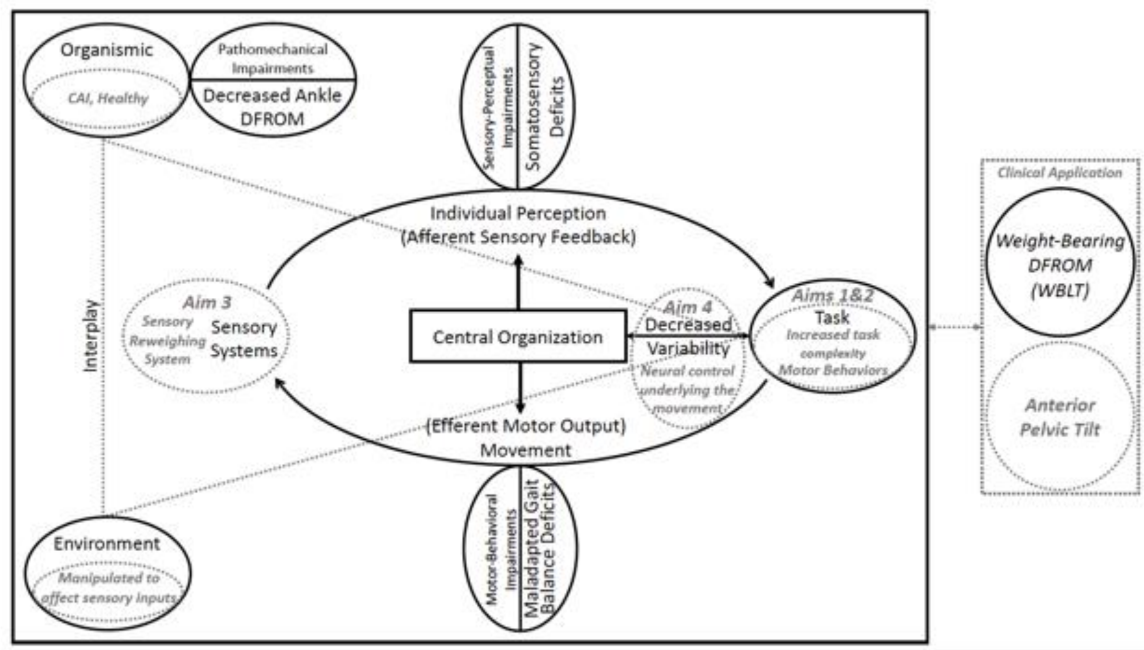

Figure 0.1. Theoretical Model on Individual Aims (1-4).
The black fonts and solid lines present current evidence in the literature. The italicized gray fonts and dotted lines show the gap in the literature.

Aim 1: Assessment of Motor Behaviors (i.e., Postural Stability)

The purpose of the specific Aim 1.1. was to determine group differences in postural control when the complexity of environmental (sensory systems) and task (limbs: double, injured, uninjured) constraints is manipulated while performing the SOT.

Environmental Constraints

<u>Independent Variables</u>:

1.1.Double-limb stance: Group (CAI, healthy controls) × Environment (6 SOT conditions)

1.2.Single-limb stance: Group (CAI, healthy controls) × Environment (6 SOT conditions)

Dependent Variables:

1.1: Equilibrium (EQ) scores (unitless) of SOT conditions

1.2: Equilibrium (EQ) scores (unitless) of SOT conditions

Task Constraints

Independent Variables:

1.3: Group (CAI, healthy controls) × Task (Limbs: double, injured, uninjured)

Dependent Variables:

1.3: Equilibrium (EQ) scores (unitless) of SOT conditions

***Hypothesis 1.1*:** There will be no group differences in EQ scores (unitless) while performing SOT conditions in a double-limb stance.

***Hypothesis 1.2*:** CAI individuals will exhibit lower EQ scores (unitless), especially when somatosensory and visual sensory systems are manipulated while performing SOT conditions in a single-limb (injured, uninjured) stance compared to healthy controls.

***Hypothesis 1.3*:** CAI individuals will exhibit lower EQ scores (unitless) while performing the SOT in the injured-limb compared to their double- and uninjured-limbs, and to double- injured-, and uninjured-limbs of healthy controls.

Aim 2: Assessment of Motor Behaviors (i.e., Postural Adaptation)

The purpose of the specific Aim 2.1. was to determine group differences in the postural adaptation from a sudden unexpected platform tilt (inversion [IN], plantarflexion [PF]) while performing the adaptation test (ADT) in double- and single-limb (injured, uninjured) stances.

Environmental Constraints

Independent Variables:

2.1.Double-limb stance: Group (CAI, healthy controls) × Environment (2 ADT conditions: IN and PF)

2.2.Single-limb stance: Group (CAI, healthy controls) × Environment (2 ADT conditions: IN and PF)

Dependent Variables:

2.1.Double-limb stance: Sway energy (SE) scores (unitless) of ADT conditions

2.2.Single-limb stance: Sway energy (SE) scores (unitless) of ADT conditions

Task Constraints

Independent Variables:

2.3.ADT Inversion/Plantarflexion: Group (CAI, healthy controls) × Task (Limbs: double, injured, uninjured)

Dependent Variables:

2.3: Sway energy (SE) scores (unitless) of ADT conditions

***Hypothesis 2.1*:** There will be no group differences in SE scores (unitless) while performing ADT conditions in a double-limb stance.

***Hypothesis 2.2*:** CAI individuals will exhibit lower SE scores (unitless) while performing ADT conditions in a single-limb (injured, uninjured) stance compared to healthy controls.

***Hypothesis 2.3*:** CAI individuals will exhibit lower SE scores (unitless) while performing the ADT in the injured-limb compared to their double- and uninjured-limbs, and to double- injured-, and uninjured-limbs of healthy controls.

Aim 3: Assessment of Individual Perception (i.e., Sensory Reweighting System)

The purpose of the specific Aim 3.1. was to determine group differences in sensory reweighting on each sensory system (somatosensory, vision, vestibular) in postural controls when the complexity of task (limbs: double, injured, uninjured) constraints is manipulated while performing the SOT.

Task Constraints

Independent Variables:

3.1: Group (CAI, healthy controls) × Sensory Systems (somatosensory, vision, vestibular) × Task (Limbs: double, injured, uninjured)

Dependent Variables:

3.1: Sensory reweighting ratios (unitless) of each sensory system (somatosensory, vision, vestibular) on the SOT

***Hypothesis 3.1*:** CAI individuals will exhibit greater sensory reweighting ratios (unitless) on vision compared to sensory reweighting ratios (unitless) on somatosensory and vestibular while performing the SOT in the injured-limb compared to their double- and uninjured-limbs, and to double- injured-, and uninjured-limbs of healthy controls.

Aim 4: Assessment of Neural Control Underlying Movement (i.e., Movement Variability)

The purpose of the specific Aim 4.1 was to determine group differences in movement variability of COP excursion in postural control when the complexity of environmental (sensory systems) and task (limbs: double, injured, uninjured) constraints is manipulated while performing the SOT.

Environmental Constraints

Independent Variables:

4.1.Double-limb stance: Group (CAI, healthy controls) × Environment (6 SOT conditions)

4.2.Single-limb stance: Group (CAI, healthy controls) × Environment (6 SOT conditions)

Dependent Variables:

4.1: SampEN (unitless) of the COP excursion on SOT conditions

4.2: SampEN (unitless) of the COP excursion on SOT conditions

Task Constraints

Independent Variables:

4.3: Group (CAI, healthy controls) × Task (Limbs: double, injured, uninjured)

Dependent Variables:

4.3: SampEN (unitless) of the COP excursion on SOT conditions

***Hypothesis 4.1*:** There will be no group differences in SampEN (unitless) of the COP excursion while performing SOT conditions in a double-limb stance.

***Hypothesis 4.2*:** CAI individuals will exhibit lower SampEN (unitless) of the COP excursion especially when somatosensory and visual sensory systems are manipulated while performing SOT conditions in a single-limb (injured, uninjured) stance compared to healthy controls.

***Hypothesis 4.3*:** CAI individuals will exhibit lower SampEN (unitless) of the COP excursion while performing the SOT in the injured-limb compared to their double- and uninjured-limbs, and to double- injured-, and uninjured-limbs of healthy controls.

Assumptions

1. Participants with a self-reported history of CAI had the conditions of interest.
2. Participants provided honest and accurate answers on self-administered medical history questionnaires.
3. Participants who were considered physically active regularly exercised at least 150-minutes (2.5 hours) a week of moderate-intensity or 75-minutes (1.15 hours) a week of vigorous-intensity aerobic physical activity.
4. Participants did not obtain new injuries, take a part in rehabilitation, or change their physical activity levels prior to enrolling in the study.
5. Participants understood questions on the Identification of Functional Ankle Instability (IdFAI), Cumberland Ankle Instability Tool (CAIT), Foot and Ankle Ability Measure-Activities of Daily Living/Sports subscales (FAAM-ADL/-Sports) and provided answers that reflect the honest capacity to the best of their abilities.
6. Participants demonstrated consistent best effort in performing tasks during data collection.

7. Participants in different age groups (10s, 20s, and 30s) had the same skill set in completing tasks during data collection.
8. Lab equipment used in data collection was reliable and produced valid measurements during a series of balance tests.
9. The time of day scheduled for individual participant data collection would not affect movement variability.

Limitations

1. This study did not account for race and ethnicity or hormonal risk factors that may have changed underlying neural mechanisms.
2. Biomechanical measures collected in a standard laboratory setting for this study may demonstrate different results from what individuals display in the real-world.
3. Others generalizing the overall results should consider that this study included individuals with a history of bilateral ankle sprains and physically active individuals in the specific age group.

Delimitations

1. 40 physically active participants (females and males) ranging from 16 to 39-year-old were recruited from local and University communities.
2. Participants were free of medically diagnosed concussion within the past six months prior to study enrollment.
3. Participants were free from a history of neurological, vestibular, and/or visual disorders and/or disease (e.g., vertigo, epilepsy, stroke, peripheral neuropathies) that may have influenced postural control.

4. Participants did not have a history of connective tissue disease and/or disorders (e.g., rheumatoid arthritis, Marfan syndrome, Ehlers-Danlos syndrome).
5. Participants had no history of major surgeries in the brain and/or on the lower extremity (e.g., foot, ankle, knee, hip, lower back).
6. Participants were free of ongoing inflammatory symptoms (pain, swelling, etc.) on the lower extremity at least six weeks prior to study enrollment.
7. Participants had no acute injuries to the lower extremity in the last six months prior to study enrollment.
8. Participants did not have chronic musculoskeletal conditions (e.g., OA, ACL deficiency).
9. Healthy participants had no history of ankle sprains prior to study enrollment.
10. Participants with a self-reported history of CAI had sustained a significant initial ankle sprain at least 12-months prior to study enrollment and had not experienced recurrent ankle sprains within the past three-months prior to study enrollment.
11. Participants with a self-reported history of CAI had experienced at least two episodes of previously injured ankle joint "giving way" and/or "feelings of instability" in the last six months prior to study enrollment and/or had a history of recurrent ankle sprains.
12. Participants would be confirmed as healthy controls with following cut-off scores on self-reported ankle instability and function survey instruments (Gribble et al., 2013; Gribble et al., 2014a, 2014b).
    a. Quantified by IdFAI ≤ 11

b. Quantified by CAIT ≥ 28

c. Quantified by FAAM: ADL subscale ≥ 99% and Sports subscale ≥ 97%

13. Participants would be confirmed as CAI individuals with the following cut-off scores on self-reported ankle instability and function survey instruments (Gribble et al., 2013; Gribble et al., 2014a, 2014b).

    a. Quantified by IdFAI > 11

    b. Quantified by CAIT < 24

    c. Quantified by FAAM: ADL subscale < 90%, Sports subscale < 80%

14. Participants performed the SOT barefoot.

15. The current study did not take account of the learning effects of repetitive administration of the SOT and ADT.

16. Chronicity (time since the onset of an initial ankle sprain) was not considered in participants with a self-reported history of CAI.

Operational Definitions

Chronic Ankle Instability (CAI) Individuals: Individuals who 1. had sustained at least two lateral ankle sprains, 2. had a history of a minimum of one significant ankle sprain at least 12-months prior to study enrollment, 3. had experienced two episodes of previously injured ankle joint "giving way" and/or "feelings of instability" within the past six months and/or had a history of recurrent ankle sprains, 4. had not experienced recurrent ankle sprains in the last three months, and 5. met validation as CAI on self-reported ankle instability and function questionnaires (e.g., IdFAI) recommended by International Ankle Consortium (Gribble et al., 2013; Gribble et al., 2014a, 2014b).

Healthy Controls: Individuals who 1. had no history of sustaining ankle sprains prior to study enrollment, and 2. met validation as healthy controls on self-reported ankle instability and function questionnaires recommended by International Ankle Consortium (Gribble et al., 2013; Gribble et al., 2014a, 2014b).

Dominant Limb: The question "if you would kick a ball on a target, which leg would you use to kick the ball?" was asked to define individuals' dominant limbs (van Melick et al., 2017).

Injured Limb for CAI Individuals: The limb with a history of multiple ankle sprains. The limb with the worst IdFAI score was considered the injured-limb for those CAI individuals with bilateral ankle sprains.

Uninjured Limb for CAI Individuals: The contralateral limb to the limb with a history of CAI. The limb with better IdFAI scores was considered the uninjured-limb for those CAI individuals with bilateral ankle sprains.

Injured Limb for Healthy Controls: The matched injured-limb by demographics (i.e., age, sex, height, weight), physical activity level, and limb dominance to the CAI group.

Uninjured Limb for Healthy Controls: The counter limb to the matched injured-limb by demographics (i.e., age, sex, height, weight), physical activity level, and limb dominance to the CAI group.

Sensory Systems: Somatosensory, vision, and vestibular apparatus were considered three primary sensory systems to maintain postural control and to coordinate movement.

Sensory Reweighting System: The process that the CNS assigns relative weight on each sensory feedback based on organismic (e.g., health status), environmental, and task constraints to coordinate motor behaviors during the SOT.

Base-of-Support (BOS): Anterior-posterior (AP) and lateral borders of the feet for double-limb stance and AP and medial-lateral (ML) borders of the foot for single-limb stance.

Center-of-Mass (COM): The COM was the point located at the trunk in individuals, moving the way a single particle of the same mass would move if equivalent external acceleration were applied to it.

Center-of-Gravity (COG): The COG was the COM when the only acceleration applied to individuals is gravity (9.81 m/s^2).

Center-of-Force (COF): The COF was equivalent to the COG.

Center-of-Pressure (COP): The COP was equivalent to the COF on the NeuroCom system.

Postural Stability: The individuals' ability to control COM within their BOS.

Sample Entropy (SampEN): One of the non-linear algorithms that measure repeatability and predictability within a time series (Yentes et al., 2018).

Independent Variables

Groups: Two groups (CAI, healthy controls)

Limbs: Double, single (injured, uninjured)

SOT Conditions: Six conditions (1-6)

ADT Conditions: Two conditions (IN, PF)

Sensory Systems: Somatosensory, vision, vestibular

Dependent Variables

SOT in double- and single-limb stances

Equilibrium (EQ) Scores: Participants' sway of the COG while performing the SOT in double- and single-limb stances.

Sensory Reweighting Ratios: The relative weight assigned to each sensory feedback (somatosensory, vision, vestibular) while performing the SOT in double- and single-limb stances.

Movement Variability: SampEN (unitless) of the COP excursion while performing the SOT in double- and single-limb stances.

ADT in double- and single-limb stances

Sway Energy Scores: The magnitude of ground reaction forces to AP and ML sways of the COG while performing the ADT in double- and single-limb stances.

CHAPTER II

LITERATURE REVIEW

Introduction

The goal of this literature review is to understand the foundation of the dynamic systems theory and provide a background of ankle sprain and chronic ankle instability in the following categories: 1) anatomy of the ankle, 2) etiology of ankle sprains, 3) the significance of individuals with chronic ankle instability and copers, 4) self-reported ankle instability and function, 5) proprioception, 6) muscle strength, 7) muscle latency, 8) stretch reflex, 9) corticospinal excitability, 10) postural control and balance, and 11) gait

The Foundation of the Dynamic Systems Theory

Movement science is reflected in the work of Bernstein and Gibson referred to as the dynamical systems theory. The dynamical systems theory hypothesizes neurobiological systems will self-organize to find the most stable solution based on one of the organismic (individual characteristics), environmental, and task constraints (Davids & Glazier, 2010; Glazier, 2017). Constraints provide boundaries limiting the number of configurations available at different levels of the body (e.g., physiological, anatomical, mechanical), known as the degree-of-freedom (e.g., joint biomechanical degrees, DNA codes), for the central nervous system (CNS) to coordinate motor outputs (Davids et al., 2003; Glazier, 2017). In theory, any change in one of the three constraints can affect motor behaviors (Davids et al., 2003; Glazier, 2017).

Bernstein's Degree of Freedom Problem

Bernstein presented the degree-of-freedom (DOF) problem, known as Bernstein's problem, that there are near-infinite ways of achieving the same performance goal (Newell, 1991; Davids et al., 2003; Davids & Glazier, 2010; Glazier, 2017). The body comprises high-dimensional DOF that contains independent yet often functionally redundant elements at different levels from microscopic (e.g., molecules, cellular, neuronal, motor units) to macroscopic (e.g., joints, muscles), exceeding what the CNS minimally requires performing a presented task goal (Glazier, 2017). Since the CNS cannot continuously micromanage integrating thousands of configurations available for every DOF at all levels, each DOF self-organizes to bind with other DOF forming task goal-specific structural units called coordinative structures (Davids et al., 2003; Davids & Glazier, 2010; Glazier, 2017).

Coordinative structures involve two functional characteristics of dimensional compression and reciprocal compensation, which interact with the organismic, task, and environmental constraints. Dimensional compression refers to the degeneracy of neurobiological systems that convert high-dimension (near-infinite configurations) to low dimension (fewer configurations), yielding stable motor behavioral patterns (Davids et al., 2003; Glazier, 2017). Whereas reciprocal compensation refers to the redundancy of neurobiological systems, assembling solutions to Bernstein's problem (Davids et al., 2003; Davids & Glazier, 2010; Glazier, 2017). Consequently, when one element of neurobiological systems introduces an error, the other elements autonomously make compensatory adjustments with minimal CNS interventions. As an example of the level of muscle and joint complexes, coordinative structures either freeze or unfreeze DOF by rigidly fixating or releasing joints (e.g., ankle, hip) depending on the task and environmental constraints (Glazier, 2017). Upon the formation of coordinative structures, neurobiological systems operate almost autonomously in the exploitation of self-organization, supporting the CNS not to micromanage constituents (Davids & Glazier, 2010; Glazier, 2017). Overall, coordinative structures are structurally stable, yet flexible and adaptable, to the organismic, task, and environmental constraints, organizing best-suited motor behaviors under any circumstances.

Perception and Action

The primary source of tuning coordinative structures are sensory mechanoreceptors that encode relevant proprioceptive and exteroceptive information of the environment (Newell, 1985; Turvey, 1990). Proprioception is the spatiotemporal information of the body segment relative to the entire body, whereas exteroceptive is the information of the object and/or surface in contact with the entire body (Turvey, 2007). The interaction between sensory mechanoreceptors embedded in organisms and properties of the environment, known as *affordance,* is defined as the *perception* from Gibson's affordance theory in ecological psychology that is an essential proposal of motor control (Greeno, 1994).

Although individual (organismic) perception involves different stages of cognitive processing, Gibson's affordance theory concentrates on perception and action (e.g., movement, motor behaviors) coupling to understand whether individuals *learn to move or move to learn* (Newell, 1991; Greeno, 1994). Furthermore, the theory postulates that sensory mechanoreceptors in the ligament, joint capsule, muscle, and cutaneous called *haptic perception* mediate an individual to explore, integrate, and funnel the most relevant affordance in order to achieve task goals (Newell, 1991; Greeno, 1994). Therefore, desired motor behaviors cannot be accomplished with an inadequate haptic perception, ultimately affecting inherent perception and action interactions (Greeno, 1994; Turvey, 2007; Turvey & Fonseca, 2014).

In addition to the role of haptic perception, the visual sensory system obtains exteroceptive information of the environment (Turvey, 1990, 2007; Greeno, 1994). Visual feedback aids haptic perception by fine-tuning individual perception to improve task precision (Turvey, 20017). In an example of gripping a cold-wet water bottle, haptic perception acquires wetness of the water bottle, when visual feedback confirms wetness as relevant information to fine-tune motor acuity in gripping the water bottle. An additional example is walking on the ground after the rain. Haptic perception obtains the presence of a puddle on the ground, while visual feedback validates the affordance to direct an individual's path and foot positioning to avoid the puddle (Turvey, 1990; Turvey & Fonseca, 2016).

Affordance is available at all times but never identical under the ever-changing environment, and similarly, individual characteristics (organismic constraints) are not consistent (Greeno, 1994). The relativity of affordance and perceptual ability is co-defining that affordance depends on either environmental or organismic constraints (Greeno, 1994). Thus, to acquire and maintain movement, an individual may require a series of supplementary practices perceiving affordance. The greater precision of perceiving relevant affordance, the more accurate motor outputs will become, especially under an inherent increase in the organismic, task, and environmental constraints (Newell, 1991; Greeno, 1994).

Sensory Reweighting System

The inherent perception and action interactions in coordinating stable movement patterns involve adequate sensory-motor integration of the CNS. Individuals simultaneously obtain redundant sensory information from multiple elements of the body utilizing three sensory systems (somatosensory, vision, vestibular). Therefore, healthy individuals have the ability to effectively integrate redundant multisensory information, known as multisensory integration. Multisensory integration is processed by identifying the relevancy (relevant, irrelevant) of sensory information to place its emphasis (weight) even with a rapid change in the surrounding environment (Nashner, 1982; Stein & Rowland, 2011). The relevancy of sensory information is context-dependent (e.g., task goals) and proportional to the ever-changing environment (Nashner, 1982; Peterka, 2002; Horak, 2006). For example, healthy individuals with intact sensory systems weight 70% on somatosensory, 20% on vestibular, and 10% on visual sensory information to maintain postural control (Peterka, 2002). Whereas, when somatosensory is disrupted while standing on an unstable surface with eyes-open, those healthy individuals reweight on visual and vestibular feedback to compensate for the disruption of somatosensory feedback to maintain an upright stance (Nashner, 1982). Those flexible multisensory integrations of upweighting the most optimal combination of sensory feedback have been described as the sensory reweighting system (Peterka, 2002).

Inadequate multisensory integration increases with aging and has been reported to contribute to falls in older individuals (Woollacott et al., 1986; Manchester et al., 1989; Teasdale et al., 1991; Whipple et al., 1993; Woollacott, 1993; Judge et al., 1995; Hay et al., 1996; Simoneau et al., 1999). Additionally, researchers found that older fallers utilize unisensory integration rather than multisensory integration compared to healthy older adults (Camicioli et al., 1997). Thus, inadequate multisensory integration may cause unisensory integration. Unisensory integration is less flexible and adaptable, resulting in individuals to emphasize weight on a single sensory system (Peterka & Loughlin, 2004). Therefore, unisensory integration is not only recognized in older individuals with inadequate multisensory integration but also in individuals with compromised sensory systems. Indeed, excessive sensory reweighting on unisensory has been displayed in individuals with somatosensory and vestibular deficits (Cooke et al., 1978; Hafstrom et al., 2004; Lopez et al., 2006; Slaboda et al., 2009; Bonan et al., 2013; Lin et al., 2019). Specifically, individuals with a history of stroke, Parkinson's disease, cervical spondylotic myelopathy, and vestibular disorders elicit greater dependence on visual feedback regardless of their relevance during postural control compared to healthy controls (Bronstein et al., 1990; Bonan et al., 2004; Hafstrom et al., 2004; Lopez et al., 2006; Slaboda et al., 2009; Manor et al., 2010; Lin et al., 2019). In essence, excessive sensory reweighting on a specific sensory system like vision makes an individual more prominent to subsequent injuries.

Tensegrity System

Turvey and Fonseca (2014) proposed that the organism is constructed with a multi-fractal nature of the tensegrity structure (Figure 2.1). Tensegrity structure consists of both compressive and tensile components interconnected by connective tissues (i.e., fasciae), distributing applied force in stress-strain patterns to maintain equilibrium (Turvey & Fonseca, 2014). Thus, the fasciae net within the tensegrity structure may serve the organism to regulate the same objectives (i.e., movement patterns) across many spatial scales (Turvey & Fonseca, 2014). For example, stress-strain patterns recur from the microscopic scale of actin-myosin chains to the macroscopic scale of the musculoskeletal system, where muscles serve as a tensional element, balancing with skeletal bones that serve as a compression element.

According to redundancy in DOF, a similar tension array that comprises the tensegrity system exists at various levels, maintaining an inherent state of pre-stress. The pre-stress transmits any stimulus force imposed within and/or outside organisms through the entire tensegrity structure over dimensions of space and time (Cabe, 2019). In other words, the tensegrity structure intervenes with haptic perception, specifically with muscle spindles to maintain their sensitivity even when organisms are at rest (e.g., breathing, heartbeats, distention of internal organs). Although tensegrity structure is in constant stress, organisms vary in response, contributing to movement variability.

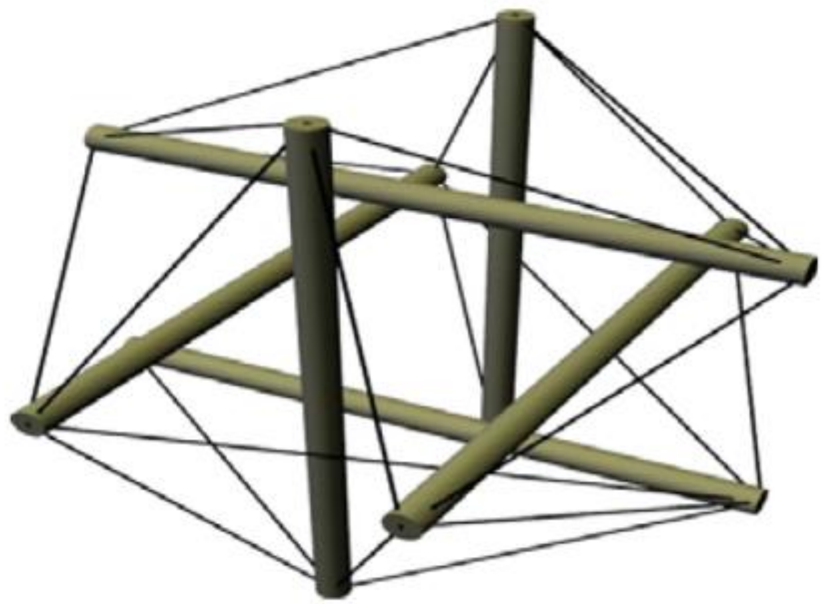

Figure 1.1. Tensegrity Structure (Turvey, 2007).

Movement Variability and the Dynamic Systems Theory

Movement variability is associated with the problem of motor redundancy (Bernstein's problem) that there are too many solutions available to accomplish a task due to redundant DOF. Thus, an increase in movement variability may disrupt neurobiological systems predicting task and environmental constraints. Additionally, pathology like chronic ankle instability (CAI) and anterior cruciate ligament (ACL) reconstruction has been demonstrated to decrease movement variability, altering flexibility and adaptability of neurobiological systems (Stergiou & Decker, 2011).

According to the traditional linear aspect of movement variability, variation in the movement has been intended as the noise (Newell & Corcos, 1993). Either decreased or increased movement variability in linear perspective refers to the magnitude of errors in neurobiological systems, coordinating movement under various tasks and environmental constraints (Stergiou & Decker, 2011). Conversely, the nonlinear perspective of movement variability embraces variation in the movement as a reflection of the learning curve to evolution (Skinner, 1981). In the process of motor learning and skill acquisition, movement variability is an important element for neurobiological systems to provide a wide range of strategies for an individual to explore coordinating task-specific motor behaviors.

The exploration in task-specific movement strategies will be reinforced with experience (i.e., practice and repetition) and stored in memory as a motor schema that autonomously generates motor behaviors. Movement variability typically increases at the beginning of motor learning, plateaus when new skills and behavior emerge, and increases once again when individuals become an expert. For example, skillful elite athletes generate stable movement patterns that are flexible and adaptable to dynamically shifting tasks and environmental constraints. In contrast, less-skilled or injured athletes exhibit unstable (random) or highly stable (rigid) movement patterns that are less flexible and adaptable to constraints. These examples of the movement patterns of different skilled athletes are conceptualized by the dynamic systems theory (DST). The DST suggests that motor learning and skill acquisition depend on interactions between organismic (individual characteristics), task, and environmental constraints (Figure 2.2).

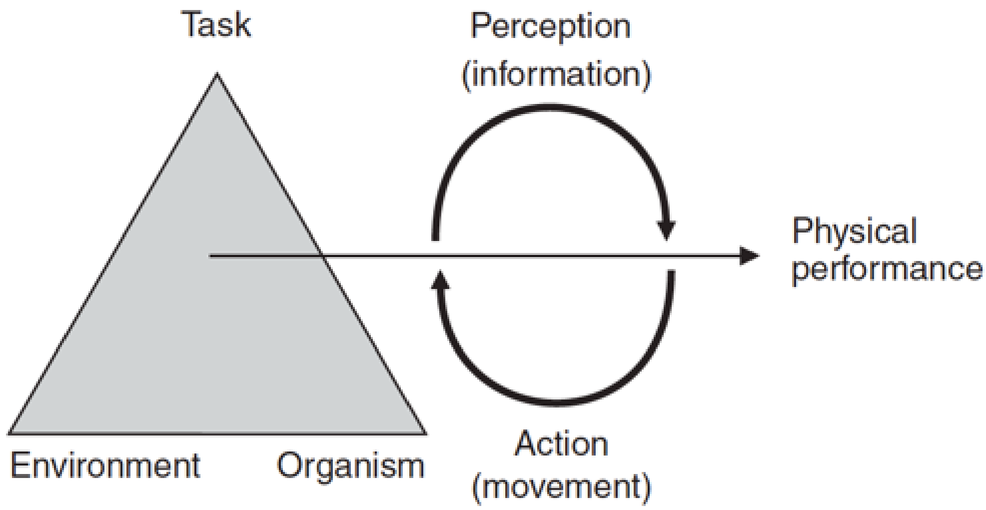

Figure 1.2. The Dynamic Systems Theory (Davids et al., 2003).

The DST is a significant theory that supports movement variability relating to a behavioral transition of dynamic human movements. However, it does not account for the stable behaviors of skillful elite athletes who could perform a task in various ways. Therefore, Harbourne and Stergiou (2009) proposed a new theoretical model that explains movement variability, associating it with mature motor skills, and health status in a concept of movement predictability. The inverted-U shaped theoretical model (Figure 2.3) suggests that optimal movement variability reflects the flexibility and adaptability of neural control in sensorimotor systems of healthy individuals(Stergiou & Decker, 2011).

Optimal movement variability at the uppermost point of the inverted-U shape indicates movement variability is in a healthy state. The healthy state renders the adaptability and flexibility of sensorimotor systems critical in achieving a task goal in an ever-changing environment. Conversely, suboptimal movement variability at below or above the optimal state is associated with a lack of health. The decrease in optimal variability renders more predictable rigid motor behaviors (robotic), whereas an increase in optimal variability renders unpredictable noisy motor behaviors (e.g., frail elders). Consequently, either too much or too little movement variability results in an inflexible adaptation of sensorimotor systems to the task and environmental constraints.

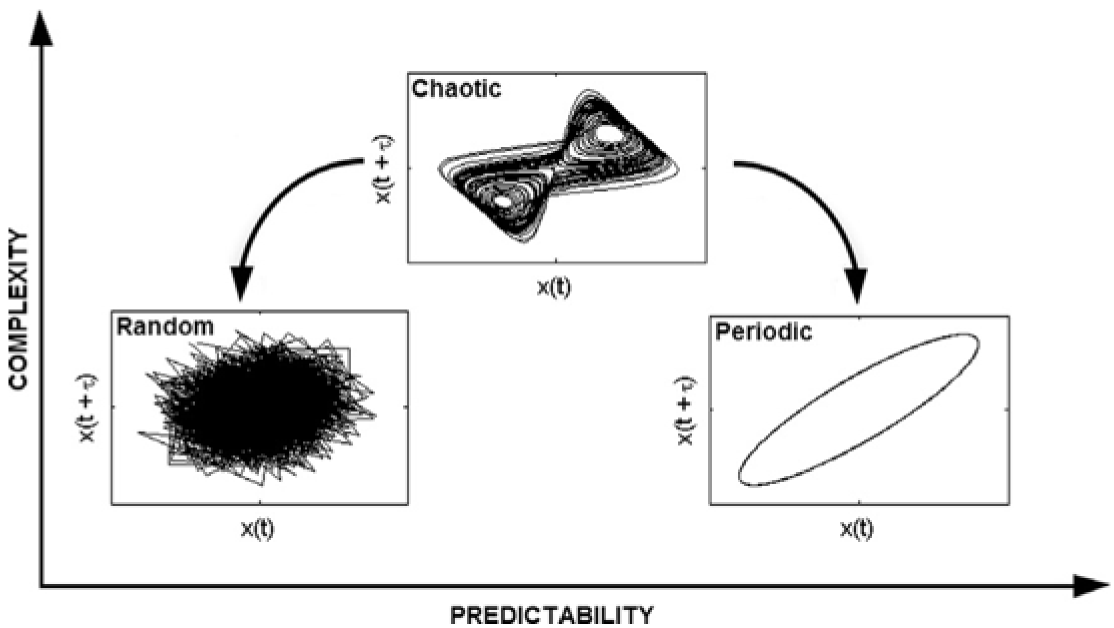

Figure 1.3. Optimal Movement Variability (Stergiou & Decker, 2011).

Movement Variability Measures

The interpretation of movement variability depends on how movement variability is measured. There is an analysis of linear and nonlinear measures to examine different aspects of movement variability. In a traditional linear approach, the quality of the movement is analyzed. The linear analysis includes standard deviation (SD) and coefficient of variations (CV), quantifying the variation around the mean, and interpreted as the standard of performance. The variables away from the mean capture the magnitude of movement variability and are considered performance errors with a motor learning paradigm (Stergiou & Decker, 2011). The mean removes the temporal aspect of variation in the movement. Therefore, the linear analysis does not evaluate the time-evolving nature of the movement. Additionally, the linear analysis assumes the variation between repetitions of the current and past and/or future movements are random and independent (Stergiou & Decker, 2011).

In contrast, nonlinear analysis of movement variability includes entropy (e.g., approximate entropy, sample entropy, Lyapunov exponent), quantifying the evolution of movement emerging over time. This is because movement variability has a deterministic origin in nature, meaning variation between repetitions of the current and past and/or future movements are not random nor independent (Miller et al., 2006; Harbourne & Stergiou, 2009). Movements are generalized from the motor schema (memory) fluctuating between repetitions of movement in time. Consequently, evaluating the temporal aspect of movement variability has been suggested to provide an insight into the explanatory nature of the neurobiological behavior of sensorimotor systems underlying movement (Lipsitz, 2002; Stergiou & Decker, 2011).

The most common entropy algorithms utilized to analyze human movement (e.g., gait) are approximate entropy (ApEN) and sample entropy (SampEN). ApEN was developed to quantify the regularity (likelihood of repeating patterns) within a time series, however, the limitation of ApEN is suggested to include a bias towards regularity

(Pincus, 1991; Richman & Moorman, 2000; Yentes et al., 2018). ApEN also lacks relative consistency as value fluctuates based on the combination of parameters, specifically the selection of the parameter values for the radius (*r*) and the length of the entire data set (N). Thereupon, SampEN was developed to counteract those limitations of ApEN (Richman & Moorman, 2000). Although SampEN demonstrates better relative consistency compared to ApEN in the analysis of gait data, SampEN is also sensitive to the parameter value of the *r*; as the *r* increased, the value of SampEN decreased (Yentes et al., 2018). Overall, both entropy algorithms are sensitive to the selection of parameter values that may affect results. In order to quantify and analyze the complexity (flexibility and adaptability) of movement with ApEN and SampEN, it is recommended to utilize a time series of 200 data points and relative healthy entropy values (e.g., pathological group relative to healthy control) (Yentes et al., 2013).

The Dynamic Systems Theory and Chronic Ankle Instability

In the application of the DST, the health states of individuals with CAI imply an increased organismic constraint (McKeon, 2009; Wikstrom et al., 2013). Specifically, common somatosensory impairments at the ankle joint complex in CAI individuals disrupt the flexible adaptability of healthy sensorimotor systems. In this circumstance, there is a change in the sensorimotor systems to self-organize the most stable motor solution for a task goal while interacting with the environment. According to Bernstein's problem, somatosensory is one element of DOF. There are other elements that are independent yet functionally redundant at various levels of DOF. These neurobiological systems redundancy may make compensatory adjustments for CAI individuals to find a stable solution to achieve a task goal in the ever-changing environment. Indeed, the most recent systematic review with meta-analysis concludes CAI individuals heavily rely on visual feedback to compensate for somatosensory deficits while maintaining a balance in a single-limb stance compared to healthy individuals (Song et al., 2016).

Anatomy of the Ankle

The ankle joint complex comprises three bony articulations that are the talocrural (tibiotalar) joint, subtalar joint, and distal tibiofibular syndesmosis. These joints primarily coordinate cardinal-plane motions (i.e., sagittal-, frontal-, and transverse-planes) of the rearfoot (calcaneus and talus). The cardinal-plane motions are plantarflexion-dorsiflexion (sagittal-plane), inversion-eversion (frontal-plane), and internal-external rotation (transverse-plane). Besides, pronation and supination are coordinated in a unit through each of the cardinal-plane motions around the oblique axis. In the open kinetic chain (OKC), supination comprises calcaneal inversion with plantarflexion and adduction, while pronation comprises calcaneal eversion with dorsiflexion and abduction (Picciano et al., 1993). In the closed kinetic chain (CKC), supination comprises calcaneal inversion and talar dorsiflexion and adduction, while pronation comprises calcaneal eversion and talar plantarflexion and abduction (Picciano et al., 1993). The primary contribution to the static and dynamic ankle joint stability during movements are regulated by ligamentous restraint, musculotendinous units, and bony congruity when the ankle joint is loaded.

The talocrural joint is structured by the fibula (lateral), tibia (medial), and talus (inferior) bones. The fibular and tibia form an inverted U-shaped ankle mortise and sit on the talus providing a stable bony congruity. The talocrural joint primarily allows sagittal-plane motion (dorsiflexion, plantarflexion) and a small amount of rotation around the oblique axis (supination, pronation). On the other hand, the subtalar (talocalcaneal) joint exists between the talus and calcaneus. Two joint cavities that make up anterior and posterior subtalar joints affording supination and pronation of the foot. The anterior subtalar joint consists of the head and anterior-superior facet of the talus, sustentaculum tali of the calcaneus, and the proximal surface of the navicular. The posterior subtalar joint consists of the inferior-posterior facet of the talus and the superior-posterior facet of the calcaneus.

The medial ankle is supported mostly by medial deltoid ligaments that insert proximally to the medial malleolus and distally to the talus (anterior and posterior tibiotalar ligaments), calcaneus (tibiocalcaneal ligament), and navicular (tibionavicualr ligament) bones (Figure 2.4). For instance, medial deltoid ligaments primarily stabilize the ankle against plantarflexion of the talocrural joint. In contrast, the lateral ankle is supported by six ligaments that bind three bones (calcaneus, fibula, talus): anterior talofibular ligament (ATFL), calcaneofibular ligament (CFL), posterior talofibular ligament (PFTL), interosseous talocalcaneal ligament, cervical ligament, and lateral talocalcaneal ligament (LTCL) (Figure 2.5). Among these ligaments, cervical and interosseous talocalcaneal ligaments are deep to peripheral ligaments (CFL, LTCL), stabilizing the subtalar joint.

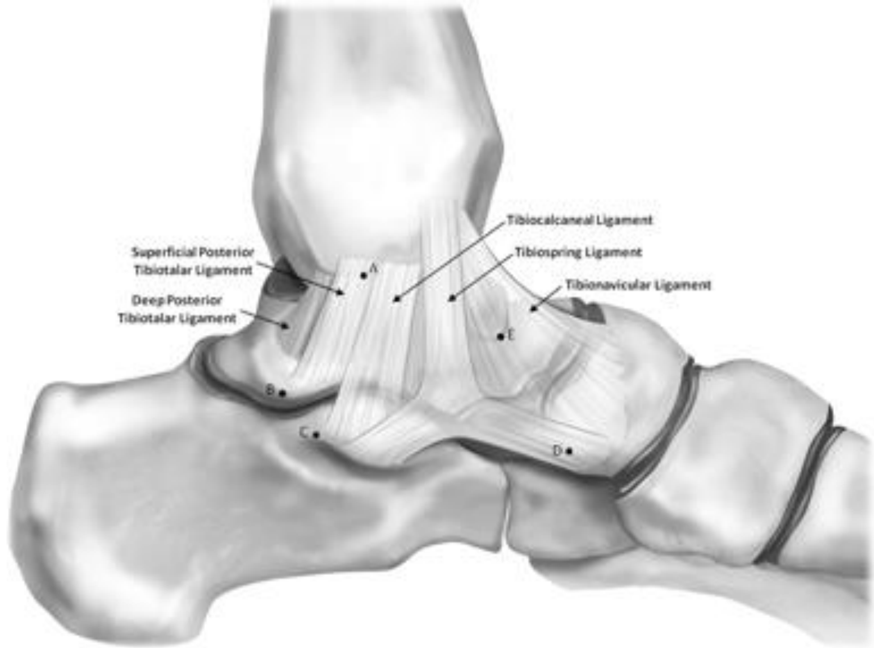

Figure 1.4. The Ligamentous Anatomy of the Medial Ankle (Campbell et al., 2014).

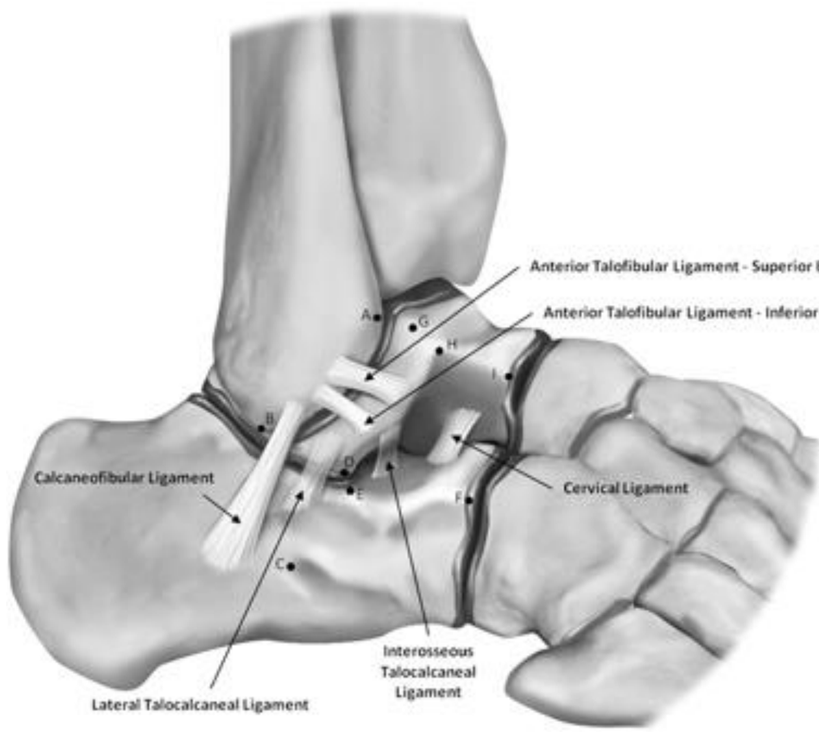

Figure 1.5. The Ligamentous Anatomy of the Lateral Ankle (Clanton et al., 2014).

The ATFL extends anteromedially from the lateral malleolus and inserts to the neck of the talus, preventing anterior displacement of the talus from the ankle mortise (fibula, tibia). Thus, the tensile stress applied to the ATFL increases as the foot moves from dorsiflexion to plantarflexion. Additionally, the ATFL is the weakest of all lateral ankle ligaments, and the most frequently sprained. The PTFL extends posteromedially from the malleolar fossa and inserts the lateral tubercles of the talus. In contrast to the ATFL, the PTFL resists both inversion and internal rotation of the talocrural joint when the ankle is loaded. The CFL extends posteroinferiorly from the tip of the lateral malleolus and inserts the lateral surface of the calcaneus. The CFL resists excessive supination of both talocrural and subtalar joints and excessive inversion and internal rotation of the rearfoot. Since the CFL is a prime stabilizer and most taut when the ankle joint moves into dorsiflexion, it is the second most commonly sprained ligament following the ATFL (Hertel, 2002; Medina McKeon & Hoch, 2019). The LTCL, which runs in parallel to the CFL and posterior to the subtalar joint binding the talus to the calcaneus, is also commonly injured in conjunction with the ATFL.

Musculotendinous units generate stiffness with contraction, providing dynamic stability to the ankle joint complex. The peroneus brevis and longus muscles (peroneals) concentrically evert the ankle and protect excessive supination of the rearfoot. In addition to peroneals, the muscles of the anterior compartment of the lower leg (i.e., anterior tibialis, extensor digitorum longus, extensor digitorum brevis, and peroneus tertius) eccentrically control excessive supination of the rearfoot. These muscles are contracted eccentrically to control the plantarflexion component of supination and may play a critical role in preventing ankle sprains.

Etiology of Ankle Sprains

Epidemiology

The ankle is one of the most common musculoskeletal injury sites, and the lateral ankle sprain (LAS) is the most common injury among physically active individuals of all ages, especially with young children (aged 0 to ≤12 years) and adolescents (aged ≥13 to ≤17 years) (McKay et al., 2001; Doherty et al., 2014; Gribble et al., 2016). The LAS occur due to direct-contact (41.1%), indirect-contact (22.2%), or non-contact (27.4%) such as landing on the court surface (direct), landing on another player's foot (indirect), or cutting (noncontact) (Roos et al., 2017; Medina McKeon & Hoch, 2019). Almost half (45%) of the LAS results from direct or indirect contact with the ankle positioned in excessive inversion and plantarflexion (Freeman, 1965; Freeman et al., 1965; Tropp et al., 1985; McKay et al., 2001). Additional sequences are sharp twisting and turning (30%), collision (10%), falling, others (5%), sudden stopping (2.5%), and tripping (2.5%) (McKay et al., 2001).

Pathomechanics

The LAS is sustained when excessive supination (inversion, plantarflexion, internal rotation) of the not fully loaded rearfoot is coupled with external rotation of the tibia (Wright et al., 2000; Hertel, 2002). The greater manipulation of ankle dorsiflexion and plantarflexion angles occurs following the initial ground contact, further increasing excessive supination of the foot (Wright et al., 2000). This mechanical displacement of excessive supination following the ground contact may increase the moment arm to the ground reaction force, driving the subtalar joint further into excessive supination (Wright et al., 2000). Once the foot contacts the ground, it is harder to initiate corrective movements to prevent injury. Thus, the foot position leading up to the initial ground contact contributes to excessive supination of the foot, advancing the sustainability rate of LAS (Wright et al., 2000).

Peroneal muscles maintain the ankle in a neutral position. However, peroneal muscles may not be quick enough to protect the ankle from excessive supination. For example, peroneal muscles would take at least 126-millisecond (msec) to react to a sudden unexpected inversion perturbation while vertical ground reaction force peaks at 40-msec at jump landing (Ashton-Miller et al., 1996; Konradsen et al., 1997). Upon sustaining an ankle sprain, the most commonly sprained ankle ligament is the ATFL, which is the weakest lateral ankle ligament (Hertel, 2002). The ATFL is damaged in isolation for 65% of the LAS, and the CFL is injured 20% of the time in conjunction with the ATFL (Safran et al., 1999). When the CFL is involved, the LTCL and other subtalar ligaments (e.g., cervical ligament) are most likely to be involved as high as 80% among the LAS (Safran et al., 1999). Damage to the PTFL is seldom with the LAS. In contrast to the ATFL, the PTFL is typically damaged as a result of a severe LAS, often accompanied by fracture and/or dislocation (Safran et al., 1999).

Arthrokinematics Alterations

An ankle sprain has been suggested to alter arthrokienmatics contributing to mechanical instability of the ankle joint complex. Mulligan (1993) first proposed altered arthrokinmeatics result from the positional fault of the distal tibiofibular joint within the ankle mortise following an initial ankle sprain. In subsequent studies, several researchers have found displacement of distal fibular (positional fault) occurs in the anterior direction in individuals with acute LAS, whereas the displacement occurs in posterior and inferior directions in individuals with CAI. Additionally, talar positional fault restriction in the form of anterior talar displacement has been suggested to contribute to limited ankle dorsiflexion range-of-motion commonly exhibited in CAI individuals. Normal ankle glides the talus posteriorly to dorsiflex the ankle. Thus, the anterior-to-posterior talocrural joint mobilization technique has been successful in restoring limited dorsiflexion range-of-motion.

Types of Activities and Populations

An ankle sprain is the major musculoskeletal injury in 33 of 43 documented sports reported from 32 countries (Fong et al., 2007). It is also the most common injury in collegiate sports in the United States. The data captured by National Collegiate Athletic Association (NCAA) Injury Surveillance System (ISS) on 15-collegiate sports over 27 years in the United States indicated that 0.83 ankle sprains occur every 1000 athletic exposures (Hootman et al., 2007; Gribble, 2016). This estimates more than approximately 27,000 ankle sprains, representing 15% of overall sports reported injuries (Hootman et al., 2007; Gribble, 2016).

The sports with the highest incidence of ankle sprains are sports involving jumping, landing, and cutting maneuvers (Yeung et al., 1994). More specifically, individuals who play basketball, soccer, and volleyball report a high incidence of sustaining an ankle sprain (Ekstrand & Gillquist, 1983; Ekstrand & Tropp, 1990; Bahr et al., 1997; McKay et al., 2001; Wang et al., 2006; Fong et al., 2007; Hootman et al., 2007; Roos et al., 2017). Based on the most recent NCAA ISS on 25-collegiate sports over five years in the United States, both women's (9.50 per 10,000 athlete-exposures) and men's basketball (11.96 per 10,000 athlete-exposures) and women's soccer (8.36 per 10,000 athlete-exposure) demonstrated the highest LAS incidence rate compared to other sports (e.g., men's soccer, men's football, women's volleyball) (Roos et al., 2017).

A systematic review with a meta-analysis indicates females and children younger than thirteen years of age have a higher risk of sustaining ankle sprains in indoor and court sports (e.g., basketball, aeroball, volleyball, tennis) compared to males (Doherty et al., 2014). High incidence of ankle sprains in females for court sports is consistent with the report from NCAA ISS on 25-collegiate sports (Roos et al., 2017). For example, female gymnastics (71.4%) and tennis (59.1%) athletes had the highest percentage of the LAS due to surface contact (Roos et al., 2017). In unisex sports among 25 NCAA sports, while females continue to present a higher rate of LAS due to surface contact, males had a higher rate of the LAS due to player contact (Roos et al., 2017). Ankle sprains are also reported to be the most common musculoskeletal injuries in the military population. A few studies have reported that 34.95 and 45.14 ankle sprains occur among active-duty military population per 1000 persons in seven to nine years period, which equates to 0.35 ankle sprains per 1000 exposures (Cameron et al., 2010; Bulathsinhala et al., 2015).

Risk Factors

Risk factors are classified into extrinsic or intrinsic risk factors. Extrinsic risk factors are related to environmental variables, whereas intrinsic factors are related to individual characteristics (Beynnon et al., 2002; Waterman et al., 2010). For example, extrinsic risk factors are shoe type, the level of competition, and fitness levels, while intrinsic risk factors are sex, age, and body size. Some factors such as shoe types and fitness levels are modifiable, however, other factors like sex, age, and body size are non-modifiable. Therefore, targeting the modifiable risk factors for the correction is significant in the prevention of recurrent ankle sprains.

Shoe Type

Types of shoes have been suggested to contribute to the susceptibility of an ankle sprain, but the findings are not consistent. Elite female and male basketball athletes (10,393 athletes) who wore shoes with air cells in the heels are 4.3 times more likely to sustain ankle sprains compared to those athletes who wore shoes without air cells (McKay et al., 2001). Conversely, no relationship between three types of shoes (low top, high top, and high top with inflatable chamber) and ankle sprain incidence among 622 basketball athletes are found (Barrett et al., 1993). This conflict in findings may be due to differences in the design of air cells and chambers that air cells in the heels may decrease rearfoot stability (McKay et al., 2001). A prospective study conducted in 390 male military recruits found no association between the incidence of ankle sprains and types of shoes (Milgrom et al., 1991). The number of LAS incidence was not different between the military population who wore combat boots compared to those who were wearing three-quarter height basketball shoes during basic training (Milgrom et al., 1991). Contribution of shoe types may vary based on subtalar joint alignments whether an individual has a neutral alignment of the calcaneus relative to the tibia: rearfoot (calcaneus) varus or rearfoot valgus. Individuals with a history of ankle sprains typically have an increased varus alignment of the calcaneus to the tibia (rearfoot varus). Hence, certain shoes may expose the subtalar joint to further excessive inversion, setting up for subsequent LAS (Van Bergeyk et al., 2002).

Levels of the Competition and Fitness Levels

The level of competition has been implicated as a risk factor for ankle sprains. More than half of all lower extremity injuries, with ankle and knees being the most common injury sites, occur in games and scrimmages than in practice drills (Prager et al., 1989; Seil et al., 1998; Murphy et al., 2003). Waterman et al. (2010, 2011) consistently reported that intercollegiate athletes at the United States Military Academy (USMA) competing at a high-level had a higher incidence of ankle sprains, which includes syndesmotic and medial ankle sprains, compared to intramural athletes. Additionally, the incidence of ankle sprains in elite soccer matches occurs between 10 and 35 per 1000 playing-hours of competition (Dvorak & Junge, 2000). Athletes who compete at a high-level (i.e., elite, intercollegiate) seem to engage in more risk-taking plays during competitions. In that case, a high proportion of ankle sprains may occur because of foul play. In fact, 63% of foul play recorded during men's world soccer tournaments organized by the Federation Internationale de Football Association (FIFA) has resulted in foot and ankle injuries (Giza et al., 2003).

Several studies have investigated the relationship between the skill level of athletes and lower extremity injuries (Murphy et al., 2003). Less skilled soccer athletes like young athletes (14 to 16-year-olds) reported a higher incidence rate of sustaining ankle sprains despite their shorter playing time compared to highly skilled athletes (Chomiak et al., 2000; Peterson et al., 2000; Willems et al., 2005). In contrast, a few studies have shown that highly skilled netball and basketball athletes are also more likely to sustain ankle sprains than less skilled athletes due to the aggressive intensity of the competition (Hopper et al., 1995; Hosea et al., 2000). This evidence concludes that the incidence rate of ankle sprains not only relates to the level of competition but in combination with athletes' skill levels and the intensity of playing sports.

Sex

A prospective study conducted of 118 Division I collegiate athletes participating in soccer, lacrosse, or field hockey exhibited no differences in the prevalence of ankle sprains between females and males (Beynnon et al., 2001). The finding is supported by a similar prospective study conducted of 145 collegiate athletes who participated in the same sports (soccer, lacrosse, field hockey) (Baumhauer et al., 1995). In contrast, a systematic review with a meta-analysis, including 181 studies concluded that females have a higher risk of obtaining ankle sprains than males (females:13.6 vs. males: 6.94 per 1,000 exposures) (Doherty et al., 2014). Likewise, female military recruits presented a higher incidence rate of ankle sprains compared to male military recruits (females: 96.4 vs. males: 52.7 per 1000 female/male person-years) (Waterman et al., 2010). Researchers who have prospectively investigated the incidence of ankle sprains among intercollegiate basketball players for 2-years displayed that female basketball players have a 25% greater risk of sustaining Grade I ankle sprains than male players (Hosea et al., 2000). However, there were no sex differences for Grade II and III or syndesmotic ankle sprains (Hosea et al., 2000). The current evidence does not fully support why females have a higher risk of sustaining an ankle sprain than males. However, sex-related influences on ankle sprains should be considered when analyzing and/or comparing studies.

Age

There are only a few studies that investigated the incidence of ankle sprains for different age groups: children, adolescents, and adults. A prospective study conducted by Peterson et al. (2000) among football athletes in different age groups presents young athletes, age 14 to15-year-olds had more ankle sprains compared to 16 to18-year-olds. Similarly, another prospective study conducted by McKay et al. (2001) examining risk factors of ankle sprains among elite and recreational basketball athletes reported athletes who sustained ankle sprains are younger (25.02 ± 6.6 years) than athletes who did not (28.0 ± 7.7 years). Those findings are supported by a systematic review with a meta-analysis that concluded children (aged 0 to ≤ 12 years: 2.85 per 1000 exposures) are more likely to sustain an ankle sprain than adolescents (aged ≥ 13 to ≤ 17 years: 1.94 per 1000 exposures) and adults (aged ≥ 18: 0.72 per 1000 exposures) (Doherty et al., 2014). Since children and adolescents are typically in the process of development, they may be at a higher risk of sustaining ankle sprains (Conn et al., 2006).

Body Size

Several studies have reported an association between body size (height, weight) and ankle sprains. For instance, overweight individuals are 3.9 times more likely to sustain ankle sprains compared to individuals with normal weight (Tyler et al., 2006; Fong et al., 2009). Additionally, Watson (1999) reported male soccer athletes who incurred ankle sprains were taller than those athletes who did not. Likewise, male military recruits who sustained ankle sprains during basic training were taller, heavier, and had a greater BMI than uninjured recruits (Milgrom et al., 1991; Waterman et al., 2010). Conversely, Baumhauer et al. (1994) did not find height and weight to be risk factors among collegiate athletes who played soccer, field hockey, or lacrosse. The findings are consistent among elite and recreational basketball athletes (Sitler et al., 1994; McKay et al., 2001). Those findings are confirmed by Beynnon et al. (2002) and Willem et al. (2005), who both revealed no relationship between anthropometrical characteristics (height, weight, BMI) and the occurrence of LAS. Although each of the anthropometrical characteristics may not be an independent risk factor for ankle sprains, an increase in either height or weight is suggested to increase proportionally to the magnitude of the mass moment of inertia of a body (mass x height^2) (Milgrom et al., 1991). Thus, the larger the magnitude of mass moment of inertia for a body may exceed mean supination torque failure of 41- 45 Nm at the subtalar joint sprains (Markolf et al., 1989; Fong et al., 2009; Waterman et al., 2010). It may expose an already hyper-supinated ankle, which individuals with a history of LAS commonly exhibit at the ground contact, to a risk of repetitive ankle sprains.

The Significance of CAI and Copers

Chronic Ankle Instability

Ankle sprains are one of the most common musculoskeletal injuries among physically active individuals and athletes in various age groups at any level. For instance, approximately 23,000 ankle sprains are reported daily, and more than 628,000 ankle sprains are treated annually in emergency departments in the United States alone (Kannus & Renstrom, 1991; Waterman et al., 2010). It costs at a minimum of $10,000 to treat ankle injuries, and the annual health care costs of $6.2 billion in the US have been estimated to treat ankle sprains (Knowles et al., 2007; Gribble et al., 2016). In spite of the frequent incidence of ankle sprains, it is neglected as a minor injury and overlooked. Indeed, more than 50% of individuals who sustain an ankle sprain do not seek proper medical care from allied health professionals (McKay et al., 2001). Therefore, the exact health care costs for the prevalence of ankle sprains may be much higher than what has been estimated. In addition, the non-medically treated ankle may experience further limitation with activities of daily living reported by individuals who sustained an ankle sprain.

Initial ankle sprains result in mechanical and/or functional (perceived) impairments at the ankle. Mechanical instability is described as altered ankle mobility beyond physiological limits (Tropp et al., 1985; Hiller et al., 2011). The cause of mechanical instability includes ligamentous laxity and arthrokinematics (e.g., limited dorsiflexion) restrictions due to the structural damage of an initial ankle sprain (Hertel, 2002; Gribble et al., 2016). In contrast, perceived instability refers to a subjective experience of ankle instability and a "giving way" sensation with no mechanical dysfunction. The common cause of perceived instability is impaired neuromuscular control and proprioceptive deficits (Hertel, 2002; Waterman et al., 2010; Hiller et al., 2011). The mechanical or perceived instability is experienced by at least 30 to 74% of individuals who sustained an initial ankle sprain, contributing to the development of chronic ankle instability (CAI) (Hertel, 2002; Anandacoomarasamy et al., 2005). Those individuals with CAI typically exhibit sensorimotor deficits resulting from long-term consequences of altered proprioceptive perception (e.g., somatosensory deficits) and motor behaviors (e.g., balance deficits, altered gait kinematics) (Hertel, 2008; Hertel & Corbett, 2019).

CAI individuals commonly experience repetitive ankle sprains. The cause of repetitive ankle sprains has been attributed to articular deafferentation (Freeman, 1965; Freeman et al., 1965). The articular deafferentation theory was first proposed by Freeman (1965) hypothesizing that an ankle sprain damages mechanoreceptors in the joint capsule and/or within ligament disrupting sensory afferent signals to the CNS (Freeman, 1965; Riemann & Lephart, 2002; Hertel, 2008). Limited or inaccurate afferent sensory signals may disrupt individuals' proprioceptive perception in obtaining properties of the environment and the foot movement and position sense relative to the body in space (Hertel, 2008). Additionally, altered proprioceptive perception affects the ability of the CNS to organize the most suited movement toward a task goal while providing ankle stability (Tropp et al., 1985; Konradsen et al., 1993; Hertel, 2002; Hertel & Corbett, 2019). As a result, CAI individuals exhibit altered motor behaviors such as balance and postural control deficits and altered gait kinematics, increasing the susceptibility of recurrent ankle sprains. The chronicity of CAI may also alter the central organization of the movement. Altered central organization decreases the preactivation of peroneal muscles to provide dynamic ankle stability, increasing susceptibility of repetitive ankle sprains (Hertel, 2008; Hass et al., 2010).

A strong linkage between an ankle sprain and osteoarthritis (OA) development has been reported. A study found that the onset of ankle OA is young and progresses faster to the end-stage of OA compared with other lower extremity OA (knee, hip) (Valderrabano et al., 2006). The substantial cause of ankle OA is a perceived instability and recurrent ankle sprains (Saltzman et al., 2005; Valderrabano et al., 2009). Thus, CAI is a leading cause of trauma-initiated joint disease, known as post-traumatic osteoarthritis (PTOA) (Valderrabano et al., 2006; Thomas et al., 2017). The PTOA is a common form of osteoarthritis and causes joint disability, resulting in long-term physical limitations (Valderrabano et al., 2006; Wikstrom & Anderson, 2013; Thomas et al., 2017). Indeed, the lower extremity PTOA is associated with the annual healthcare costs of $11.79 billion with direct costs of over $3 billion in the US (Valderrabano et al., 2009; Thomas et al., 2017). In the United States alone, approximately 5.6 million clinical cases of the lower extremity PTOA are accounted for by adults older than 25 years of age, and the incidence of the PTOA is estimated to double with an increasingly aging population (Thomas et al., 2017). Collectively, conducting research on CAI is critical to prevent recurrent ankle sprains and subsequent development of the PTOA in preserving the health-related quality of life.

Copers

At least 60% of individuals who sustain an ankle sprain become copers (Doherty et al., 2016). Copers are identified as individuals who have suffered an initial ankle sprain yet fully recovered and have returned to at least moderate-level weight-bearing physical activities for at least 12-months without experiencing perceived instability, “giving way” sensations, and recurrent ankle sprains (Wikstrom & Brown, 2014). Previous research report copers have favorable postural control (Wikstrom et al., 2010; Plante & Wikstrom, 2013; McCann et al., 2017), gait kinematics at the ankle and hip (Koldenhoven et al., 2019), isometric hip strength (McCann et al., 2017, 2018), and dorsiflexion range-of-motion (Plante & Wikstrom, 2013) compared to individuals with CAI. In contrast, CAI individuals and Copers displayed similarity in AP ankle laxity (Bowker et al., 2016), change in the ATFL length at the end range ankle inversion (Croy et al., 2012), and frontal-plane forefoot kinematics at initial contact during treadmill walking (Wright et al., 2013). Those findings may suggest copers either did not comprise neuromuscular deficits following an initial ankle sprain or have developed corticomotor plasticity to function as if uninjured. However, the exact reason why some individuals develop CAI, and others become copers remains unknown. Therefore, investigating copers as a comparison group to CAI instead of healthy individuals may provide better insight to elucidate the factors contributing to CAI for directing better rehabilitation programs. Furthermore, rehabilitating CAI individuals to operate more like copers may lead to better prevention of recurrent ankle sprains.

Self-Reported Ankle Instability and Function

The most commonly reported deficits with CAI have been recurrent ankle sprains and episodes/feelings of ankle joint "giving way" (Delahunt et al., 2010). Individuals with CAI also report a decrease in health-related quality of life (HRQOL) (Arnold et al., 2011). One way to quantify functional deficits and HRQOL is by implementing valid and reliable survey instruments (e.g., CAIT, IdFAI, AII, FAAM) of ankle characteristics (i.e., instability) and patient-related outcomes (Hiller et al., 2006; Donahue et al., 2011; Simon et al., 2012; Gurav et al., 2014). Additionally, identification of self-reported instability and functional deficits is significant for clinicians to diagnose CAI and evaluate changes in ankle function from pre-intervention to post-intervention (Hale & Hertel, 2005).

Standardizing selection criteria of CAI individuals may produce consistency and validity in research outcomes and interpretation across studies. Therefore, the International Ankle Consortium (IAC) has advocated, using the work of Delahunt et al. (2010) as a framework, for incorporating specified criteria and survey instruments in defining CAI (Gribble et al., 2013; Gribble et al., 2014a, 2014b). The IAC endorsed self-reported ankle instability questionnaires are the Cumberland Ankle Instability Tool (CAIT), Ankle Instability Instrument (AII), and Identification of Functional Ankle Instability (IdFAI). Additionally, the IAC recommended self-reported ankle function questionnaires are Foot and Ankle Outcome Score (FAOS) and Foot and Ankle Ability Measure (FAAM) (Gribble et al., 2013; Gribble et al., 2014a, 2014b).

Chronic Ankle Instability

Self-Reported Ankle Instability Survey Instruments

All self-reported ankle instability questionnaires (i.e., CAIT, AII, IdFAI) address ankle dysfunction status that is heavily influenced by perceived instability (Docherty et al., 2006; Hiller et al., 2011; Simon et al., 2012). The CAIT places a greater emphasis on self-reported functional ankle instability and management, whereas the AII focuses on previous ankle sprains and the level of ankle function (Docherty et al., 2006; Hiller et al., 2011). The CAIT is a 9-item 30-point scale questionnaire scored on a 5-point Likert scale, with a lower total score indicating more severe functional ankle instability (Hiller et al., 2011). Hiller et al. (2011) originally established a cut-off score of ≤ 27 to discriminate between individuals with and without ankle sprain. Consequently, researchers did not utilize true discrimination of CAI such as the sensation of giving way, which was later suggested by the IAC . With this in mind, a subsequent study conducted by Wright et al. (2014) recalibrated a CAIT cut-off score for CAI classification and reported that a score of ≤ 23 in the dataset that includes copers and a score of at least ≤ 25 in the dataset that excludes copers. Those findings are somewhat consistent with the IAC recommended CAIT cut-off scores of < 24 as an indication of CAI (Gribble et al., 2013; Gribble, et al., 2014a, 2014b).

In contrast to the CAIT, the AII is a 16-item questionnaire consisting of nine yes/no dichotomous questions, six multiple-choice questions, and open-ended questions related to the severity of the initial ankle sprain, history of ankle instability, and instability during activities of daily life (Docherty et al., 2006; Donahue et al., 2011).

Several researchers classified individuals who answered “yes” to five or more yes/no questions as having functional ankle instability (McVey et al., 2005; Sedory et al., 2007). Similarly, the IAC established the AII cut-off score for CAI as an answer of “yes” to at least five yes/no questions (Gribble et al., 2013; Gribble et al., 2014a, 2014b). The limitation of the CAIT and AII is that neither instrument has clearly provided a definition of giving way. For this reason, the CAIT and AII instruments were not able to predict ankle instability status when utilized alone (Donahue et al., 2011). Conversely, researchers have presented that the combined use of the CAIT and AII instruments has provided the highest sensitivity and specificity (0.82, 0.82) in predicting ankle instability status (Donahue et al., 2011). Accordingly, the IdFAI, which comprises elements from the CAIT and AII, may be the best-suited instrument to assess CAI accurately (Simon et al., 2012).

Simon et al. (2012) introduced the IdFAI combining CAIT and AII. The IdFAI is a 10-item 37-point scale questionnaire scored on a 5-point Likert scale, with a higher total score of 11 or higher indicating more severe functional ankle instability (Donahue et al., 2011). That is to say, the IAC cut-off score of the IdFAI for CAI is a total score of $>$ 11 (Gribble et al., 2013; Gribble et al., 2014a, 2014b). Similar to AII, the IdFAI is made up of three factors that include a history of ankle instability (factor 1), initial ankle sprain (factor 2), and instability during activities of daily living (factor 3) (Simon et al., 2012). Additionally, the IdFAI includes a specific definition of “giving way” described as “a temporary uncontrollable sensation of instability of rolling over one’s ankle,” ensuring the IdFAI is completed under the same definition (Freeman, 1965; Freeman et al., 1965;

Donahue et al., 2012; Simon et al., 2012). Researchers have consistently revealed excellent overall test-retest reliability of the IdFAI (ICC = 0.92), including across different age groups (20-60 years: ICC = 0.92-0.98) (Donahue et al., 2012; Gurav et al., 2014). Furthermore, several researchers have confirmed singular use of the IdFAI has higher accuracy in predicting ankle instability status than the combined use of the AII and CAIT (Simon et al., 2014). In summary, the current evidence indicates utilization of the IdFAI as a primary and the CAIT as a secondary survey instrument to evaluate self-reported instability diagnosing CAI accurately.

Self-Reported Ankle Function Survey Instruments

The FAOS and FAAM are valid and reliable survey instruments commonly utilized in conjunction with self-reported instability survey instruments (CAIT, AII, IdFAI) (Roos et al., 2001; Martin et al., 2005; Eechaute et al., 2007). The FAOS evaluates individual symptoms and functional limitations in the previous week (Roos et al., 2001). The FAOS is a 42-item questionnaire consists of five subscales, including 9-items related to pain, 7-items of other symptoms, 17-items related to activities of daily living, 7-items related to sports and reaction function, and 4-items specific to foot and ankle-related quality of life (Roos et al., 2001). Each subscale is scored on a 5-point Likert scale and converted to a percentile with 100% indicating full foot and ankle function (Roos et al., 2001; Donahue et al., 2011). The IAC recommends that a FAOS cut-off score of $<$ 75% in three or more categories is an indication of foot and ankle dysfunction to identify CAI (Gribble et al., 2013; Gribble et al., 2014a, 2014b).

Contrary to FAOS, the FAAM determines self-reported foot and ankle dysfunction status over time. The FAAM consists of two subscales, which emphasize the level of foot and ankle dysfunction during activities of daily living (FAAM-ADL) and sports (FAAM-Sports) (Martin et al., 2005). The FAAM-ADL is a 21-item 84-point scale questionnaire focusing on walking and stepping on a different surface, while FAAM-Sports is an 8-item 32-point scale questionnaire focusing on more extensive activities (i.e., jumping, running) essential to sports (Martin et al., 2005; Cosby & Hertel, 2011; Donahue et al., 2011). Each question is scored on a 5-pint Likert scale, and the raw scores converted to a percentile with 100% implying no foot and ankle dysfunction (Martin et al., 2005; Cosby & Hertel, 2011). The IAC recommended cut-off score for the FAAM-ADL subscale is 90%, and for the FAAM-Sports subscale is 80% (Gribble et al., 2013; Gribble et al., 2014a, 2014b).

Each FAAM subscale has been validated as reliable (ICC = 0.89, 0.87) in detecting self-reported foot and ankle dysfunction in individuals with CAI (Martin et al., 2005; Eechaute et al., 2007). More specifically, Martin et al. (2005) concluded the FAAM (ADL, Sports) is a responsive survey instrument to evaluate a change in self-reported function for individuals participating in physical therapy for foot or ankle injuries. Indeed, several studies have reported significant improvement in FAAM scores with interventions such as ankle joint mobilization (Hoch et al., 2012; Gilbreath et al., 2014; Wikstrom & McKeon, 2017; Feldbrugge et al., 2019;), balance training (Wright et al., 2017), and corrected exercise released by the National Academy of Sports Medicine (Bagherian et al., 2019) in individuals with CAI. Among those findings, studies included

CAI individuals with higher baseline FAAM scores on either subscale than the IAC cut-off recommendations (FAAM-ADL: 96.67%, FAAM-Sports: 89.36%) presented an improvement solely on the FAAM subscale comparable to the IAC cut-off scores (Gilbreath et al., 2014; Feldbrugge et al., 2019). Based on the most recent study, different subgroups of CAI proposed by Hiller et al. (2011), which are recurrent ankle sprains (RAS) and perceived instability (PI), exhibited higher FAAM scores than the IAC recommended FAAM cut-off scores for CAI, ranging from 90 to 97% and 79 to 94%, respectively (Terada et al., 2017). The inclusion of the FAOS and FAAM survey instruments is only considered essential for CAI classification if the self-reported foot and ankle function is important to the research questions (Gribble et al., 2013; Gribble et al., 2014a, 2014b). However, implementation of the self-reported function survey instrument, specifically the FAAM, to differentiate subgroups of CAI may help to understand the difference in characteristics and dysfunctions coordinating intervention programs in individuals with CAI. In conclusion, self-reported ankle function survey instruments in combination with self-reported ankle instability survey instruments may eliminate confounding influence on research outcomes and strengthen the description and understanding of CAI.

Copers

There are no specific selection criteria and cut-off scores for the survey instruments (CAIT, AII, IdFAI, FAOS, FAAM) in defining copers. Wright et al. (2014) recommended a CAIT cut-off score of $>$ 23. Conversely, Wikstrom and Brown (2014) have proposed minimum reporting criteria for self-reported ankle instability and function to be no lower than 28 for CAIT and 99% and 97% for FAAM-ADL/-Sports subscales, respectively. While the CAIT cut-off score of $\geq$ 24 has been widely implemented in previous studies (Doherty, 2015; Doherty et al., 2016; Cao et al., 2019; Cho et al., 2019), a few researchers took a more conservative approach applying the CAIT cut-off score of $\geq$ 28 (Wanner et al., 2019; Rosen et al., 2020). Additionally, Terada et al. (2017) utilized the AII cut-off score of $<$ 5 and the IdFAI cut-off score of $<$ 11 to discriminate copers from subgroups of CAI (RAS, PI). Similarly, Koldenhoven et al. (2019) implemented a more stringent IdFAI cut-off score of $<$ 10. Recognizing the IdFAI is the best-suited survey instrument to diagnose CAI, current evidence may indicate for researchers to utilize the IdFAI cut-off score of $<$ 11 as a primary and the CAIT cut-off score of $\geq$ 24 as a secondary survey instrument for identifying copers.

Proprioception

The proprioceptive deficits associated with CAI have been hypothesized as a cause of recurrent ankle sprains. Freeman et al. (1965) described damaged ankle ligamentous mechanoreceptors creating a void in proprioceptive feedback from the ankle for the CNS to coordinate best-suited movement for a task. Generating adequate motor control (e.g., dynamic joint stability) requires the integration of relevant feedforward and feedback. The feedforward predetermines anticipatory movement, while feedback provides environmental properties to fine-tune motor control. Therefore, proprioception is a feedback phenomenon, and it has been examined by assessing joint position sense, kinesthesia, and force sense.

Joint Position Sense

Glencross and Thornton (1981) first evaluated JPS with a handheld goniometer and revealed that the injured ankle exhibited greater errors than the uninjured ankle. Although the methodology of the study was not rigorous, it suggested an initial ankle sprain can cause substantial impairment in JPS at the ankle. In fact, a systematic review with a meta-analysis has concluded passive and active JPS deficits are evident in individuals with CAI (Konradsen et al., 1993).

Passive Joint Position Sense

Passive JPS recognition errors are typically examined by first introducing index foot positions, then having an individual to detect those positions while the foot is passively moved at a constant velocity. Several researchers have assessed passive JPS recognition errors in inversion, plantarflexion, or a combination of inversion and plantarflexion/dorsiflexion (Munn et al., 2010). For example, Fu and Hui-Chan (2005) examined individuals with bilateral CAI (i.e., multiple ankle sprains) at an index position of 5-degree plantarflexion, starting at a 0-degree neutral position (90-degrees ankle dorsiflexion) with a constant velocity of 1-degree/sec compared to healthy controls. Researchers revealed that CAI individuals demonstrate about 38 to 40% greater JPS recognition errors in bilateral limbs (right [R]: $1.4 \pm 0.7°$, left [L]: $1.1 \pm 0.5°$) compared to healthy controls (R: $1.0 \pm 0.4°$, L: $0.8 \pm 0.2°$). In contrast, Santos and Liu (2008) examined individuals with and without unilateral CAI at an index angle of 30-degree inversion tested from a 0-degree neutral position at a velocity of 5-degree/sec and found no side-to-side differences within-limb or between-group (CAI, healthy controls). Those findings are supported by Brown et al. (2004) who did not find group differences between CAI and healthy controls by utilizing index inversion angles that are 10% and 90% of an individual's total inversion range-of-motion assessed at a velocity of 2-degree/sec. Additionally, no between-group differences in passive JPS recognition errors were identified among additional directions examined in eversion, dorsiflexion, and plantarflexion at their index angles of 10% and 90% of the total range-of-motion (Brown et al., 2004).

Yokoyama et al. (2008) evaluated passive JPS recognition errors in plantarflexion (-10-, 0-, 10-, 20-, 30-degrees) and combination of plantarflexion and inversion (0-, 20-degrees) at a velocity of 4-degree/sec in individuals with and without CAI. Although no group differences were identified in 0- and 20-degrees of inversion, significantly greater passive JPS recognition errors were displayed in 20-degree inversion when coupled with a 30-degree plantarflexion in CAI individuals (Yokoyama et al., 2008). The passive JPS recognition discrepancy in plantarflexion is consistent with the findings reported by Fu and Hui-Chan (2005). The researchers have implied that individuals with CAI may underestimate index angles in plantarflexion rather than inversion compared to healthy controls (Yokoyama et al., 2008). Furthermore, the most recent study of passive JPS recognition errors examined separately in 8-degree pronation and 24-degree supination suggested longitudinal tension of the ligament compresses the fibrous septum (connection between dermis and fascia) space, stimulating mechanoreceptors (Hagen et al., 2018). Therefore, the excitation of mechanoreceptors in the ankle ligaments such as the ATFL is direction-dependent, meaning sensory discharge of the mechanoreceptors in the ATFL is higher during supination in triaxial-plane (frontal-, sagittal-, transverse-planes) compared to inversion in frontal-plane. This may explain why passive JPS recognition errors were more substantial when index angles were combined with plantarflexion and inversion.

Active Joint Position Sense

The active JPS replication enables the examination of an individual's combined ability of afferent angle recognition and efferent motor control (Konradsen & Magnusson, 2000). Most of the studies have evaluated active JPS replication errors in individuals with unilateral CAI by implementing similar parameters of passively positioned index inversion angles tested at a constant velocity for those individuals to actively replicate. Jerosch et al. (1995) assessed three different index inversion angles (5-, 15-, 20-degrees) starting at a 0-degree neutral position (90-degrees ankle dorsiflexion) at a velocity of 4-degree/sec in individuals with unilateral CAI. The study found significantly worse active JPS replication errors in the injured-limb ($2.44 \pm 0.8°$) compared to the uninjured-limb ($2.30 \pm 1.04°$) (Jerosch et al., 1995). This result is supported by Nakasa et al. (2008) who applied similar index inversion angles of 5-, 10-, 15-, 20-, and 30-degrees, starting at 20-degree ankle plantarflexion in individuals with unilateral CAI and healthy controls. The injured-limb with CAI ($3.4 \pm 1.0°$) displayed more active JPS replication errors than the uninjured-limb ($2.3 \pm 0.9°$), and between-group differences were also discovered upon matched healthy controls (Nakasa et al., 2008). Consistently, bilateral differences were reported between the injured-limb ($2.5 \pm 0.4°$) and the uninjured-limb ($2.0 \pm 0.3°$) of CAI and with the matched injured-limb of healthy controls ($1.7 \pm 0.2°$), when examining index inversion angles of 10-, 15-, and 20-

degrees starting at 0-degree neutral in absolute error measures (Konradsen & Magnusson, 2000). However, these findings are not supported in the real error measurement. The real error measurement allows determining the direction of whether active JPS replication errors are underestimated or overestimated. Although the absolute error measurement does not gauge direction, it has been suggested to be the most accurate and precise measurement tool to express the ankle JPS compared with the real error measurement (Fu & Hui-Chan, 2005). Overall, the disparity in active JPS replication errors has been firmly observed in ankle inversion of the injured-limb in CAI individuals.

Passive and Active Joint Position Sense in the Same Cohort

A few studies evaluated the comparison between passive and active JPS recognition/replication errors in the same cohort of participants. Boyle and Negus (1998) examined index angles that are 30, 60, and 90% of an individual's total inversion range-of-motion, starting at 42-degree ankle plantarflexion with a velocity of 5-degree/sec. For passive JPS recognition errors, CAI individuals displayed greater errors in all three index inversion angles compared to healthy controls (Boyle & Negus, 1998). Whereas for active JPS replication errors, CAI individuals only demonstrated significantly greater errors in the index angle of 30% total inversion range-of-motion compared to healthy controls (Boyle & Negus, 1998). Furthermore, CAI individuals showed significantly greater JPS replication errors in the active JPS judgment than the passive JPS judgment in the index angle of 30%, while no such differences were present for 60 and 90% of testing positions (Boyle & Negus, 1998).

Although both passive and active JPS recognition/replication errors have been identified in the same cohort of CAI individuals compared to healthy controls, the active JPS replication method may be superior to the passive JPS recognition method. Gross et al. (1987) examined passive and active JPS recognition/relocation errors at index angles of 10-degree eversion, and 10- and 20-degrees inversion, starting on 0-degree neutral with a velocity of 5-degree/sec in individuals with and without CAI. Although researchers did not find any group differences in either passive or active JPS recognition/relocation errors, CAI exhibited more active JPS replication errors than passive JPS relocation errors (Gross et al., 1987). Willems et al. (2002) tested passive and active JPS recognition/relocation errors at the index angle of 15-degree ankle inversion and maximal active inversion minus 5-degree, starting at 15-degree ankle plantarflexion with a velocity of 5-degree/sec with absolute and real error measurements. In contrast to Gross et al. (1987), a significant difference was revealed in this study that CAI individuals demonstrated greater active JPS replication errors ($-2.96 \pm 2.96°$) in maximal active inversion minus 5-degrees compared to healthy controls ($-0.68 \pm 3.21°$) with real error measurement, indicating CAI underestimated the index angle (Willems et al., 2002).

A meta-analysis has suggested that the active JPS replication method might take into account mechanoreceptors within sensorimotor pathways that the passive JPS recognition method does not (Medina McKeon & McKeon, 2012). Gross et al. (1987) also hypothesized that “active motion mechanoreceptors” (e.g., muscle spindle, GTO) are most concerned detecting joint movement (e.g., velocity, tension), whereas mechanoreceptors in the ligament and joint capsule are most concerned detecting the joint position. Indeed, active JPS replication errors that occurred at 30% ankle inversion may be an indication of a delay in early movement detection by "active motion mechanoreceptors" (Boyle & Negus, 1998). For instance, Konradsen et al. (1993) reported a prolonged reaction time in peroneal muscles when the ankle was suddenly inverted through a 30-degree of motion with CAI. Overall, the current evidence demonstrates that the active JPS replication method is most appropriate to use. However, more research is needed to debate regarding the specific role of each mechanoreceptor in passive or active JPS judgments.

Joint Position Sense under Anesthesia

Some researchers aimed to anesthetize the lateral ankle joint complex, directly impairing the function of mechanoreceptors in the ligament and joint capsule to examine Freeman's articular deafferentation theory (Freeman, 1965). Feuerbach et al. (1994) anesthetized both the anterior talofibular ligament and the calcaneofibular ligament and evaluated active JPS replication errors for index foot positions (degree not specified) in plantarflexion, dorsiflexion, inversion, and eversion with a constant velocity of 30-degree/sec in healthy individuals. No active JPS replication errors were revealed between individuals with and without anesthetized conditions (Feuerbach et al., 1994). The results indicated the least contribution of the ligamentous mechanoreceptors to the active JPS at the ankle and adequate involvement of peripheral afferent feedback from mechanoreceptors in the joint capsule, muscle, musculotendinous units, and cutaneous. Consistently, the presence of the alternative mechanisms to compensate for the loss of afferent stimuli from the region of lateral ankle ligaments is supported by the passive JPS judgment. Hertel et al. (1996) who examined passive JPS recognition errors at index angles of 10-degree eversion, and 20- and 30-degrees of inversion with a velocity of 3-degree/sec displayed a lack of deficits in the passive JPS judgment in individuals with and without anesthetized anterior talofibular ligament and lateral joint capsule.

Konradsen et al. (1993) assessed both passive and active JPS recognition/replication errors by implicating a total anesthetic block to the entire foot and ankle region in the same cohort of healthy individuals and identified significantly greater errors only in the passive JPS judgment. This finding is consistent with Feuerbach et al. (1994) and Hertel et al. (1996). In contrast to those earlier studies (Feuerbach et al., 1994; Hertel et al., 1996), the total anesthetic block over the entire foot and ankle complex Konradsen et al. (1993) utilized in the current study makes it difficult to identify which peripheral afferent feedback from mechanoreceptors in the ligament, joint capsule, muscle, and musculotendinous units attributing to the loss of passive JPS judgment. However, a lack of change in the active JPS judgment may suggest an alteration in the central organization. Specifically, the higher center might have quickly learned to compensate impaired peripheral afferent feedback from the anesthetized foot and ankle complex with mechanoreceptors in the muscle, musculotendinous units, and cutaneous around and above the anesthetized region for the active JPS judgment. Additionally, the method utilized by Konradsen et al. (1993) may also contribute to a lack of change in the active JPS judgment. For example, Konradsen et al. (1993) utilized different velocities (Passive JPS: 2-degree/sec vs. Active JPS: 15-degree/sec) to evaluate passive and active JPS replication errors in index angles of 5-, 10-, 15-, 20-, and 25-degrees inversion, and that the faster velocity might have advantaged muscle spindles to detect JPS. Indeed, Refshauge et al. (2009) found that the threshold of detecting JPS decreases as the testing velocity increases.

Furthermore, cutaneous mechanoreceptors that represent the ability to sense pressure, displacement/stretch, and acceleration of the skin have not received much attention in the JPS literature of CAI. Mildren et al. (2017) anesthetized the posterior skin of the ankle and evaluated passive JPS recognition errors in index angles of 6-, 12-, 18-degrees dorsiflexion and 6-, 12-, 18-degrees of plantarflexion with a velocity of 2-degree/sec in healthy individuals. The study also minimized the effect of muscle spindles by contemplating the slower testing velocity (Mildren et al., 2017). The most interesting finding of this study was a significant contribution of feedback from the posterior skin of the ankle in the passive JPS judgment, specifically during ankle dorsiflexion when the posterior ankle skin is most outstretched (Mildren et al., 2017). Overall, the posterior ankle skin significantly contributes to the passive JPS judgment alongside the intact peripheral afferent feedback from mechanoreceptors in a ligament (i.e., calcaneofibular ligament) and joint capsule.

Kinesthesia

CAI individuals have been thought to present kinesthetic (joint movement detection) impairments in inversion and plantarflexion compared to healthy individuals who have a threshold level of joint movement in less than 2-degree (Konradsen, 2002). Garn and Newton (1988) first examined the kinesthetic ability by passively moving the foot to 5-degree ankle plantarflexion with a velocity of 0.3-degree/sec and found individuals with a history of unilateral ankle sprains exhibit impaired ability to detect passive plantarflexion in the injured-limb compared to the uninjured-limb. The finding is supported by Forkin et al. (1996) who examined the kinesthetic ability to detect passive plantarflexion movement, utilizing the correct hit count rate in collegiate-level gymnasts with a history of unilateral ankle sprains. In detail, the correct hit count rate of detecting passive ankle plantarflexion movement was significantly lower in the injured-limb (142/165) compared to the uninjured-limb (158/165) (Forkin et al., 1996).

A few researchers examined the side-to-side differences of the kinesthetic ability in individuals with unilateral CAI. Lentell et al. (1995) evaluated the kinesthetic ability to detect the passive inversion movement with a velocity of 0.3-degree/sec in individuals with unilateral CAI. The results of the study demonstrated significant deficits in detecting the passive inversion movement in the injured-limb ($4.3 \pm 3.1°$) compared to the uninjured-limb ($3.2 \pm 1.8°$) (Lentell et al., 1995). Conversely, Hubbard and Kaminski (2002) did not find the side-to-side differences in the kinesthetic ability to detect passive inversion and eversion movement with a velocity of 0.5-degree/sec. Kinesthesia is typically measured with a slow velocity targeting the Type II mechanoreceptors embedded in the joint capsule. Mechanoreceptors in the joint capsule have been described as a low threshold, rapidly adapting sensory fibers activated in response to mechanical stress applied to the joint (Lentell et al., 1995). Thus, a lack of consistency in detecting kinesthetic deficits across those studies (Lentell et al., 1995; Hubbard & Kaminski, 2002) might be a result of differences in the velocity utilized (0.2-degree/sec vs. 0.5-degree/sec). Although Hubbard and Kaminski (2002) implemented the velocity of

0.5-degree/sec to minimize the contribution of mechanoreceptors in the muscle and musculotendinous units in movement detection, it might have been too fast, eliminating its effect.

A few researchers have investigated the kinesthetic ability in individuals with unilateral CAI compared to healthy controls, imposing passive inversion-eversion and plantarflexion-dorsiflexion movement at different velocities (0.1-, 0.5-, 2.5-degrees/sec). The velocity of 0.1- and 0.5-degrees/sec was intended to encompass the velocity of normal body sway, while the velocity of 2.5-degree/sec was dedicated to simulating the velocity of sports/activities in which ankle sprains are frequently sustained. De Noronha et al. (2007) implemented velocities of 0.1-, 0.5-, and 2.5-degrees/sec while passively moving the ankle into inversion and eversion and reported no differences in the kinesthetic ability within-limb or between-group. This finding is consistent with Refshauge et al. (2000) who measured the kinesthetic ability at the 70% detection level for plantarflexion and dorsiflexion with a velocity of 0.1-, 0.5-, and 2.5-degrees/sec. A lack of kinesthetic deficits detecting passive inversion-eversion and plantarflexion-dorsiflexion movement in all three velocities may indicate a less contribution of mechanoreceptors embedded in the joint capsule for an individual's kinesthetic ability. Overall, the literature on kinesthesia in CAI does not support Freeman's deafferentation theory that damage to the ligament and joint capsule from an initial ankle sprain alters an individual's kinesthetic ability in detecting movement.

Force Sense

Two components of the force sense are categorized as a central factor (feedforward mechanisms) and peripheral factor (feedback mechanisms), working in unison to detect and generate desired forces (Simon et al., 2014). Force sense judgment is an individual's ability to detect changes in external force applied on the extremity (e.g., joint) before the central organization generates a motor response such as counteracting internal force (e.g., motor units) to stabilize the joint (Hertel, 2008). Such an ability to detect a disparity between a planned and actual muscle force required to generate (force control errors) allows motor control to initiate correction at the time of an event. However, impaired force sense judgment hinders internal force to generate effectively to counteract the external force, leading to joint instability and ankle sprains (Yen et al., 2019). Thus, examining the existence of force sense deficits following an ankle sprain is significant as joint stability partially depends on muscle contraction and appropriate force generation of the involved muscles. Force sense is typically measured by having an individual to detect a target force, which is some portion (N%) of maximum voluntary isometric contraction (MVIC). Examining force sense with N% of MVIC minimizes the involvement of JPS and kinesthetic judgments (Yen et al., 2019).

Force sense has not been extensively investigated, and there are only a limited number of studies examining force sense in CAI. Docherty and Arnold (2008) reported individuals with unilateral CAI demonstrate significantly greater force sense deficits in all three target force loads (10-, 20-, and 30-% of eversion MVIC) applied in the injured-limb compared to the uninjured-limb. This finding is consistent with the study comparing

CAI individuals to healthy controls (Simon et al., 2014). Simon et al. (2014) measured force sense at 30% of eversion MVIC and displayed significantly greater force sense errors in CAI individuals compared to healthy controls. Additionally, there was a general tendency for force sense deficits to increase in variation as target force loads (task complexity) increase (Simon et al., 2014). For example, Docherty and Arnold (2008) revealed that the magnitude (M) and variation (V) of force sense deficits at 10-, 20-, 30-% of eversion MVIC become greater as target force loads increase (M/V at 10-, 20-, 30-%: 2.4N/2.0N vs. 3.4N/2.9N vs. 4.1N/3.5N). Thus, the greater variation in force sense deficits may be associated with the nature of sporadic symptoms of giving-way sensation experienced in CAI individuals. Indeed, Arnold and Docherty (2006) found that self-reported ankle giving-way frequency and force sense errors at 10- and 30-% eversion MVIC are positively associated.

The collateral evidence of the current literature on force sense deficits in CAI individuals may suggest an initial ankle sprain not only damages mechanoreceptors in the ligament and joint capsule but also mechanoreceptors in the muscle and musculotendinous units. In fact, researchers have found a significant correlation between force sense deficits and muscle stiffness suggesting force sense deficits attribute to damaged Golgi tendon organs than secondary endings (Type II) of muscle spindles (Docherty et al., 2004). The relative contribution of mechanoreceptors in the muscle and musculotendinous units also has been found in the active JPS and force sense judgments (Kim et al., 2014). However, no relationship between the active JPS and force sense has been established in CAI. Examining both JPS and force sense in the same cohort of CAI individuals may lead to a better understanding of how feedback and feedforward mechanisms undertake proprioceptive (JPS, force sense) deficits in individuals with CAI.

Muscle Strength

Boisen et al. (1955) first identified peroneal muscle weakness, which diminishes dynamic ankle joint stability, as a significant contributing factor for the recurrence of ankle sprains. Tropp et al. (1986) then displayed significant weakness in concentric ankle evertors in the injured-limb compared to the uninjured-limb in individuals with unilateral CAI. Similar side-to-side differences in concentric ankle invertor weakness was reported in CAI individuals by Wilkerson et al. (1997). Although those earlier studies focused on concentric muscle strength, a dynamic task requires coordination of both concentric and eccentric muscular control. Indeed, it has been suggested that eccentric muscle actions might play a significant role in providing an antagonistic force resisting concentric muscle contraction, preventing overloading of a joint, and regulating foot positioning (Kaminski & Hartsell, 2002; Yildiz et al., 2003).

Joint stabilization can be achieved by coordinating concentric, eccentric, isometric, or kinetic motion of all muscles surrounding the ankle joint complex. Among all, many researchers extensively examined strength quantifying isometric and isokinetic eccentric and concentric strength force productions in both frontal-plane and sagittal-plane at a velocity ranging from 0- to 240-degrees/sec in individuals with CAI (Arnold et al., 2009). Isokinetic testing is widely employed in laboratory settings to measure peak torque (the highest torque produced throughout the imposed range-of-motion). This may be because the isokinetic strength testing offers resistance at a constant velocity and provides very accurate strength easement throughout the entire joint range-of-motion (Kaminski & Hartsell, 2002). Additionally, isokinetic strength testing allows clinicians to evaluate the co-activation of agonist and antagonist muscles at a specific joint, using an isokinetic dynamometer in a clinical setting (Hislop & Perrine, 1967).

A meta-analysis of the ankle evertor strength has concluded that isokinetic concentric evertor weakness is present in individuals with unilateral CAI (Arnold et al., 2009). However, researchers also mentioned that the size of strength deficits between injured and uninjured limbs is small and may not be clinically relevant (Arnold et al., 2009). Indeed, the latter systematic review with a meta-analysis of strength in any plane around the ankle joint axis of motion concluded muscle strength deficits do not distinguish features of CAI (Hiller et al., 2011).

Isometric Strength Deficits

Lentell et al. (1990) demonstrated there are no side-to-side differences in isometric inversion and eversion strengths in individuals with unilateral CAI. Consistently, Kaminski et al. (1999), who evaluated isometric eversion strength while positioning the foot in 0-degree subtalar joint neutral in CAI individuals, found no differences compared to matched healthy controls. In contrast, Termansen et al. (1979), who investigated isometric plantarflexion strength in individuals with unilateral CAI, revealed significantly weaker plantarflexion strength in the injured-limb compared to the uninjured-limb. Those findings of Termansen et al. (1979) do not support the current evidence in the CAI literature that individuals with greater plantarflexion strength have a higher incidence of an inversion ankle sprain (Baumhauer et al., 1995; Kaminski et al., 1999).

Isokinetic Concentric Strength Deficits

Lentell et al. (1990) examined both inversion (IN) and eversion (EV) strengths at 30-degree/sec in individuals with unilateral CAI and found no side-to-side differences in the injured-limb (IN: 20.8 $\pm$ 1.3 Nm, EV: 18.0 $\pm$ 5.5 Nm) compared to the uninjured-limb (IN: 19.9 $\pm$ 1.2 Nm, EV:17.6 $\pm$ 5.4 Nm). Those findings are somewhat supported by Ryan et al. (1994) who also assessed the ankle inversion and eversion strengths at the same velocity of 30-degree/sec in individuals with unilateral CAI. Although the study identified no eversion strength deficits in the injured-limb (18.8 $\pm$ 6.6 Nm) compared to the uninjured-limb (19.2 $\pm$ 5.8 Nm), inversion strength in the injured-limb (22.7 $\pm$ 8.4 Nm) was significantly weaker compared to the uninjured-limb (26.6 $\pm$ 8.5 Nm) (Ryan et al., 1994). Researchers have speculated the unexpected inversion strength deficits are the result of an altered neural drive inhibiting investors from pulling the ankle into inversion, which is one of the mechanisms of an ankle sprain (Ryan et al., 1994).

Gribble and Robinson (2009) evaluated strength in sagittal-plane (plantarflexion, dorsiflexion) along with frontal-plane (inversion, eversion) at a velocity of 60-degree/sec in individuals with unilateral CAI and healthy controls. CAI displayed significantly reduced concentric peak torque production of plantarflexion in the injured-limb (0.63 $\pm$ 0.06 $Nm^{-1}\cdot kg^{-1}$) compared to the uninjured-limb (0.77 $\pm$ 0.07 $Nm^{-1}\cdot kg^{-1}$) of CAI and matched injured-limb of healthy controls (0.70 $\pm$ 0.07 $Nm^{-1}\cdot kg^{-1}$) (Gribble & Robinson, 2009). These findings are consistent with Hubbard et al. (2007) who also showed significantly less plantarflexion strength in relationship to dorsiflexion strength in the injured-limb (76.2 $\pm$ 30.7 Nm/kg) compared to the uninjured-limb (84.2 $\pm$ 28.7 Nm/kg) in peak torque/body weight measure in individuals with unilateral CAI.

Ankle sprains during sporting events occur at a much faster velocity up to 632-degree/sec, therefore several researchers have examined inversion and eversion strengths at different velocities, varying from 60- to 210-degrees/sec (Chu et al., 2010). Lentell et al. (1995) measured inversion and eversion strengths with a velocity of 30-, 90-, 150-, and 210-degrees/sec, while Kaminski et al. (1999) examined eversion strength with a velocity of 30-, 60-, 90-, 120-, 150-, and 180-degrees/sec in individuals with unilateral CAI and healthy controls. Collectively, neither studies have obtained side-to-side or between-group differences in the ankle inversion and/or eversion strength (Lentell et al., 1995; Kaminski et al., 1999).

In 2002, Kaminski et al. (1999) proposed a linear trend for the eversion concentric force-velocity relationship indicating when the velocity of muscle contraction increases, the force production decreases. In other words, the magnitude of the force is velocity dependent and decreases proportionally to the velocity in concentric contraction. During a dynamic task, the agonist muscles produce concentric work to propel, while the antagonist muscles generate eccentric work to control concentric force to prevent overloading of a joint. Thus, examining concentric ankle inversion and eversion strengths might provide limited information on muscle strength in CAI individuals.

Isokinetic Eccentric Strength Deficits

Bernier et al. (1997) evaluated the ankle inversion and eversion strength at 90-degree/sec in individuals with unilateral CAI and demonstrated no side-to-side strength differences. This absence of strength deficits is supported by Kaminski et al. (1999), who also found no side-to-side or between-group differences in eversion strength, tested at a velocity of 30-, 60-, 90-, 120-, 150-, and 180-degree/sec in individuals with CAI and healthy controls. Conversely, Munn et al. (2003) identified inversion strength deficits in the injured-limb compared to the uninjured-limb in individuals with unilateral CAI while assessing inversion and eversion strengths at a velocity of 60- and 120-degrees/sec. Additionally, Fox et al. (2008), who assessed strength in both frontal-plane (inversion, eversion) and sagittal-plane (plantarflexion, dorsiflexion) at a velocity of 90-degree/sec in individuals with and without CAI, displayed significantly reduced eccentric peak torque production of plantarflexion in the injured-limb of CAI (1.50 ± 0.47 Nm/kg) compared to matched injured-limb of healthy controls (1.96 ± 0.66 Nm/kg). Those contradicting findings are the result of a difference in methods and the definition of recruited CAI individuals. Thus, more research is needed to conclude the eccentric strength characteristics of CAI compared to healthy controls.

Isokinetic Eccentric and Concentric Ratios

The eccentric/concentric ratios (E/C ratios) compare agonist eccentric/antagonist eccentric and agonist concentric/antagonist concentric. The ratios generally provide an objective evaluation of muscle coordination (balance, imbalance) and permit comparisons between and within individuals for risk of injury. More specifically, CAI is hypothesized to occur when peroneal muscles are put to work eccentrically in response to high-velocity moments which would reflect in the E/C ratios (Kaminski & Hartsell, 2002). Muscles act in a stretch-shortening cycle, meaning the eccentric-stretching phase facilitates mechanoreceptors in musculotendinous units (GTO, muscle spindles) followed by a concentric contraction. Thus, Kaminski and Hartsell (2002) suggest evaluation of E/C ratios may provide an insight into an individual's neuromuscular performance.

Hartsell and Spaulding (1999) examined maximal E/C ratios for inversion and eversion strengths at a velocity of 60-, 120-, 180-, and 240-degree/sec while the ankle is positioned in 10- to 15-degrees of plantarflexion in CAI individuals compared to healthy controls. Although the study did not find group differences in maximal E/C ratios, CAI individuals exhibited weaker concentric and eccentric ankle inversion and eversion strength than healthy controls (Hartsell & Spaulding,1999) . Typically, muscle strength deficits result in lower E/C ratios, however, the adequate E/C ratios with the ankle strength deficits are also confirmed by Willems et al. (2002). Willems et al. (2002) evaluated E/C ratios for inversion and eversion strengths at a velocity of 30- and 120-degrees/sec with an ankle positioned in 15-degree of plantarflexion in individuals with and without CAI and observed no significant differences between the group (CAI, healthy controls). Movement patterns involve interaction between eccentric and concentric muscle activities, and the magnitude of the moments in those muscle contractions is velocity-dependent. Consequently, the efficiency of the E/C ratios in assessing movement patterns has been challenged. With this in mind, a few researchers have investigated the ankle evertor/invertor strength ratios (E/I ratios) (Yildiz et al., 2003).

Pontaga (2004) demonstrated significantly reduced peak torque E/I ratios mostly at 60-, 90-, and 120-degrees/sec with the ankle positioned in 50- and 60-degrees of inversion in the injured-limb compared to the uninjured-limb in individuals with unilateral CAI. The study also identified that the ankle evertor strength deficits in the injured-limb are most prominent when the ankle is tested in the lowest inversion angular position (Pontaga, 2004). For example, the lowest E/I ratios were displayed when the injured-limb (0.58 ± 0.25) was positioned in 30-degree inversion with a velocity of 60-degree/sec compared to the uninjured-limb (0.75 ± 0.31) (Pontaga, 2004). Overall, eversion strength deficits are expressed at the beginning (30-degree), or the end (50-, 60-degrees) of ankle inversion. Thus, eversion strength deficits at the most extended position (supination) may expose the ankle for repetitive ankle sprains. Considering E/I ratios would change at a different angular position of the foot, analyzing the ratios throughout the entire range-of-motion specifically at 10-degree ankle inversion, where the lateral border of the foot is estimated to collide with the ground, may extend the knowledge on the mechanisms of foot clearance during gait in CAI individuals (Konradsen & Voigt, 2002).

Yildiz et al. (2003) examined the angle-specific eversion eccentric/inversion concentric ratios (Eecc/Icon ratios) at 0-, 5-, 10-, 15-, and 20-degrees of inversion with a velocity of 0-, 5-, 10-, 15-, and 20-degrees/sec in individuals with and without CAI. The Eecc/Icon ratios in the representative of the ankle inversion are defined as the maximal eccentric evertor moment divided by the maximal concentric inverter moment. Researchers found no group differences in concentric inverter peak torque and peak Eecc/Icon strength ratios (CAI: 1.7, H: 1.9) (Yildiz et al., 2003). In spite, the study revealed significantly lower eccentric evertor peak torque (CAI: 28.9 $\pm$ 5.3 Nm, H: 37.3 $\pm$ 5.8 Nm) and Eecc/Icon strength ratios at 15-degree (CAI: 2.2 $\pm$ 0.6, H: 3.9 $\pm$ 1.7) and 20-degree (CAI: 2.6 $\pm$ 0.7, H: 4.9 $\pm$ 2.5) of inversion in CAI individuals compared to healthy controls (Yildiz et al., 2003). The eccentric evertor strength deficits displayed in this study support the findings of Pontaga (2014). Similarly, David et al. (2013) assessed the eversion concentric/inversion eccentric ratios (Econ/Iecc ratios) alongside the Eecc/Icon ratios. CAI individuals exhibited 26% lower Eecc/Icon ratios compared to healthy controls in the study (David et al, 2013), supporting the findings of Yildiz et al. (2003). Besides, David et al. (2013) presented 20% higher Econ/Iecc ratios in CAI individuals and suggested the result provides further depth to the findings of eccentric inversion strength deficits reported by Munn et al. (2003) earlier. Those findings of eccentric and concentric ankle eversion and inversion torque ratios may explain dynamic strength malfunction associated with CAI. Eccentric contraction is significant in controlling the foot position during dynamic tasks. Thus, a change in ratios of dynamic strength might lead to a subsequent fault in the foot positioning. Furthermore, the recurrence of ankle sprains with CAI may result from diminished eccentric evertor strength (Yildiz et al., 2003).

Muscle Strength at Proximal Joints

Altered neuromuscular control at proximal joints (hip, knee) has been reported in individuals with a history of ankle sprains. For instance, Bullock-Saxton et al. (1994) found bilateral deficits in hip muscle activation in individuals with severe ankle sprains. In contrast, the gluteus medius was activated when the hypermobile ankle was forced into unexpected ankle inversion perturbation (Bullock-Saxton, 1994; Beckman & Buchanan, 1995). Therefore, an initial ankle sprain may cause a change in the central organization altering muscle recruitments at other joints proximal to the ankle.

Only a few researchers have investigated the isometric and isokinetic strength of proximal joints (hip, knee) with CAI. Gribble and Robinson (2009) examined concentric peak torque production of hip and knee flexion and extension with a velocity of 60-degree/sec in individuals with unilateral CAI compared to healthy controls. CAI individuals exhibited knee flexion (FL) and extension (EX) strength deficits in the injured-limb (FL: 1.11 ± 0.07 $Nm^{-1} \cdot kg^{-1}$, EX: 1.80 ± 0.12 $Nm^{-1} \cdot kg^{-1}$) compared to the uninjured-limb (FL: 1.15 ± 0.06 $Nm^{-1} \cdot kg^{-1}$, EX: 1.91 ± 0.10 $Nm^{-1} \cdot kg^{-1}$) of CAI and matched injured-limb of healthy controls (FL: 1.36 ± 0.07 $Nm^{-1} \cdot kg^{-1}$, EX: 1.89 ± 0.12 $Nm^{-1} \cdot kg^{-1}$), while no strength deficits were displayed at the hip (Gribble & Robinson, 2009). In contrast, McCann et al. (2017), who assessed isometric strength of hip extension, abduction, and external rotation utilizing a hand-held dynamometer in individuals with and without CAI, demonstrated strength deficits in hip abduction (AB) (1.4 ± 0.5 Nm/kg) and external rotation (ER) (0.5 ± 0.1 Nm/kg) in CAI individuals compared to healthy controls (AB: 1.8 ± 0.9 Nm/kg, ER: 0.7 ± 0.3 Nm/kg).

Aside from the strength deficits, altered kinematics at the knee and hip during the jump landing and the star excursion balance test (SEBT) have also been identified with CAI (Caulfield & Garrett, 2002; Caulfield & Garrett, 2004; Gribble et al., 2007; Gribble & Robinson, 2009; Pope et al., 2011). Additionally, the study conducted by McCann et al. (2017) suggested hip abduction and external rotation strengths explain 25% of the score variance at the ankle in SEBT-Posterior reach measures. Indeed, the implementation of the hip strategy has been commonly noted to compensate for neuromuscular deficits at the ankle during postural control in CAI individuals (Doherty et al., 2015; Rios et al., 2015). Therefore, although the exact mechanisms of strength deficits at proximal joints with CAI are still unknown, the current evidence in the literature may support the hypothesis that CAI individuals manifest a change in the central organization from impaired peripheral afferent feedback at the ankle joint complex, resulting in proximal joint alteration.

Muscle Latency

The initial ankle sprain is speculated to cause neuromuscular control deficits in CAI individuals, provoking postural control discrepancy (Solomonow, 2006). Therefore, muscle latency is often examined in association with postural control. For instance, a few researchers have displayed the association between peroneal muscle (i.e., peroneus longus, peroneus brevis) reaction time and dynamic balance deficits (i.e., single-limb stance, SEBT-Posteromedial/-Lateral) in CAI individuals (Mitchell et al., 2008b; Sierra-Guzmán et al., 2018). Muscles around the ankle joint complex provide a dynamic restraint against excessive loading of the ankle in inversion and plantarflexion, which are the most common mechanisms of a lateral ankle sprain. Likewise, the correct timing of muscular activation is a significant dynamic defense mechanism to protect the ankle from hypersupination. Specifically, activations of peroneal muscles prevent hyperinversion while tibialis anterior limits excessive ankle plantarflexion. Consequently, delay in the onset of muscle activation is hypothesized to contribute to subsequent ankle sprains (Konradsen et al., 1997; Palmieri-Smith et al., 2009).

One of the common measurements examining the dynamic defense mechanisms is the reaction time of peroneal and tibialis anterior muscles to a sudden perturbation (e.g., trapdoor maneuver). The reaction time (RT) assessed with the trapdoor maneuver is defined as the time from the beginning of trapdoor movement or electrical stimulation to the first muscular motor response, typically determined with the surface electromyography (sEMG). The latest systematic review with a meta-analysis concluded CAI individuals exhibit delayed peroneals (peroneal muscles) RT to an inversion perturbation in the injured-limb compared to the uninjured-limb of CAI and matched healthy controls yet different review concluded peroneals RT is not affected with CAI (Munn et al., 2010; Hoch & Mckeon, 2014). Inconsistency in those findings may be due to the heterogeneity characteristics of CAI and/or study design, such as the diversity in the degree (5- to 50-degrees) and the velocity (50- to 700-degrees/sec) utilized to impose ankle inversion or plantarflexion perturbations (Menacho et al., 2010; Munn et al., 2010; Hoch & Mckeon, 2014).

Reaction Time: Peroneus Longus and Brevis

Konradsen and Ravn (1990) first identified significantly delayed RT in peroneus longus (PL) and brevis (PB) muscles in reaction to 30-degree inversion perturbation during standing in individuals with CAI (PL: 82 msec, PB: 84 msec) compared to healthy controls (PL: 65 msec, PB: 69 msec). Conversely, Shima et al. (2005) demonstrated significantly delayed peroneals RT in reaction to 25-degrees inversion perturbation in healthy controls compared to CAI. The researchers did not explain why CAI exhibited shorter peroneals RT than healthy controls (Shima et al., 2005). The great speculations of the result are the preexistence of neurophysiological deficits in healthy controls and/or better neuromuscular control in CAI individuals arising from the completion of rehabilitation. In contrast to group comparisons, several researchers have failed to support side-to-side differences with peroneals RT in reaction to 10- to 35-degrees of inversion perturbation in individuals with and without CAI (Isakov et al., 1986; Lofvenberg et al., 1995; Ebig et al., 1997; Osborne et al., 2001; Schmidt et al., 2005; Donahue et al., 2014). Moreover, researchers have investigated peroneals RT utilizing various angles of perturbation in the same cohort. Fernandes et al. (2000) examined PL RT in reaction to 10-, 15-, and 20-degrees of inversion perturbation in CAI individuals compared to healthy controls. Although no group differences were displayed in PL RT, the study revealed the effect of tilt angles. Specifically, PL RT increased as tilt angles increased (5-degree: 93.75 ± 8.71 msec, 10-degree: 94.65 ± 9.27 msec, 15-degree: 95.08 ± 11.21 msec) (Fernandes et al., 2000). Those findings are supported by Lynch et al.

(1996) who found an increase in acceleration velocity of inversion trapdoor perturbation results in quicker peroneals RT in healthy individuals.

Vaes et al. (2001, 2002) and Mitchell et al. (2008a) both imposed a trapdoor tilt angle in a combination of inversion and plantarflexion. Mitchell et al. (2008a) showed significantly prolonged PL and PB RT in response to 30-degree inversion and 20-degree plantarflexion perturbation in the injured-limb (PL: 62.82 $\pm$ 11.27 msec, PB: 65.46 $\pm$ 11.15 msec) compared to the uninjured-limb (PL: 53.96 $\pm$ 08.91 msec, PB: 51.46 $\pm$ 014.01 msec) of CAI and matched dominant-limb (PL: 54.77 $\pm$ 6.46 msec, PB: 56.86 $\pm$ 8.05 msec) of healthy controls. Vaes et al. (2001, 2002) implemented a trapdoor tilt angle of 50-degree supination to assess PL RT in individuals with unilateral CAI. The initial study conducted by Vaes et al. in 2001 identified significantly longer PL RT in the injured-limb (58.9 $\pm$ 12.0 msec) compared to the uninjured-limb (47.7 $\pm$ 12.1 msec) and shorter total supination time in the injured-limb (109.3 $\pm$ 15.7 msec) compared to the uninjured-limb (124.1 $\pm$ 10.3 msec). The significantly shorter total supination time indicates the injured-limb moved faster during 50-degree supination perturbation and suggests an inadequate ability of the injured-limb to decelerate the supination force to control the speed (Vaes et al., 2001). The follow-up study conducted by Vaes et al. in 2002 discovered that the onset of first deceleration responding to the sudden supination perturbation occurred significantly shorter for the injured-limb (25.5 $\pm$ 5.3 msec) compared to the uninjured-limb (28.4 $\pm$ 5.7 msec). Researchers explained the shorter deceleration time during the supination perturbation in the injured-limb compared to the uninjured-limb of CAI is an indication of the less control of the supination force with the unstable ankle (injured-limb) (Vaes et al., 2002).

There are two deceleration points the body displays during the total supination time, which is the time from the start to the end of the trapdoor tilt movement to avoid any tissue damage (Vaes et al., 2002). The first deceleration point is defined to offer passive control of slowing down of supination. Specifically, soft tissues (e.g., ligament) resist supination to control the speed at the first deceleration point (Vaes et al., 2002). Whereas, the second deceleration point is defined to offer active control (e.g., muscles) of slowing down of supination (Vaes et al., 2002). For instance, peroneals would generate a protective motor response to control supination speed (Raugust, 2006).The functional motor response of PL is calculated as the sum of PL RT and electromechanically delay (EMD) that is the time between the start of EMG activity (latency) and the beginning of the ankle movement (Vaes et al., 2002). Additionally, Flevas et al. (2017) discovered a change in EMD that fluctuates depending on the position of the ankle. Therefore, the shorter time between EMD and the second deceleration point, specifically when the ankle is already inverted with CAI, leads to an assumption that PL would not recruit sufficient motor units. Although no group differences have been found on EMD in studies conducted by Vaes et al. (2001, 2002), several researchers have reported impaired EMD with CAI (Isabelle et al., 2003; Hopkins et al., 2009; Kavanagh et al., 2012).

Ankle sprains rarely happen while standing on a platform in laboratory settings. Thus, a few researchers have investigated PL RT during dynamic activities such as gait and the step-down landing. Knight and Weimar (2011) assessed PL and PB RT and the time to maximum inversion (TMI) during the step-down landing in reaction to 25-degree inversion perturbation upon landing in individuals with and without CAI and found no significant group differences in PL and PB RT and TMI. In contrast, Hopkin et al. (2009) examined PL and PB RT during walking on a runway with a built-in bilateral trapdoor mechanism designed to freely fall to 30-degree inversion and EMD in static double-limb stance in individuals with unilateral CAI and healthy controls. The study found significantly delayed PL RT and EMD in the injured-limb (PL: 106.6 $\pm$ 27.6 msec, EMD: 33.7 $\pm$ 14.3 msec) compared to the uninjured-limb (PL: 74.3 $\pm$ 23.3 msec; EMD: 16.6 $\pm$ 4.0 msec) of CAI and matched controls (PL: 84.6 $\pm$ 18.6 msec, EMD: 19.5 $\pm$ 10.1 msec) (Hopkin et al., 2009). Those findings are consistent with the latter study conducted by Donahue et al. (2014) who also displayed significantly suppressed PL RT during walking on a walkway with a built-in bilateral trapdoor mechanism designed to freely fall to 30-degree inversion in CAI individuals (46.54 $\pm$ 1.52 msec) compared to healthy controls (41.78 $\pm$ 0.82 msec).

Researchers have suggested that the RT is mediated through monosynaptic stretch reflex initiated by mechanoreceptors in the ligament and joint capsule, and possibly muscle spindles of peroneals (Hopkins et al., 2009). Therefore, the discrepancy in any peripheral mechanoreceptors altering reflexive activity may delay muscle activation. Indeed, an anesthetic study conducted by Khin-Myo-Hla (1999) declared that anesthetizing sinus tarsi syndrome, which yields its inhibitory afferent inputs to the gamma-spindle system, resulted in significant improvement of peroneal RT post-anesthetic (71.0 msec) compared to pre-anesthetic (82.0 msec) injection in CAI individuals. Additionally, EMD is suggested to be an indirect indication of muscle stiffness and tone (Hopkins et al., 2009). Specifically, the longer EMD possibly reflects the desensitization of muscle spindles (Isabelle et al., 2003; Hopkin et al., 2009). A delay in PL RT might be a sequence of impairment in short and medium reflexive responses. This may be because partial peripheral deafferentation reduces the activity of the gamma-spindle system and implements slower-adapting Type II fibers instead of the faster-adopting Type Ia fibers. Overall, establishing a way to facilitate the gamma-spindle system may be a way to prevent repetitive ankle sprains with CAI.

Reaction Time: Tibialis Anterior

The primary dynamic defense mechanisms to a sudden inversion perturbation have been described to be peroneals. However, the tibialis anterior muscle (TA) also provides a dynamic defense by resisting forced plantarflexion. There are a handful of researchers investigating TA RT in CAI individuals. Ebig et al (1997) and Osborne et al. (2001) both examined side-to-side TA RT in response to 20-degree inversion trapdoor perturbation during standing in individuals with unilateral CAI and found no bilateral limb differences. Conversely, Mitchell et al. (2008a), who assessed TA RT in reaction to a combination of 30-degree inversion and 20-degree plantarflexion perturbation in individuals with unilateral CAI and healthy controls, revealed significantly slower TA RT in the injured-limb (66.04 $\pm$ 12.88 msec) compared to the uninjured-limb (56.84 $\pm$ 8.84 msec) of CAI and matched dominant-limb of healthy controls (55.75 $\pm$ 9.71 msec). These findings are supported by Mendez-Rebolledo et al. (2015) who also identified significantly delayed TA RT in reaction to 30-degree inversion perturbation during standing in CAI individuals (91 ± 48 msec) compared to healthy controls (47 ± 26 msec). Additionally, Mitchell et al. (2008a) and Mendez-Rebolledo et al. (2015) presented longer peroneals RT in the same cohort of CAI participants. The prolonged reaction time in both peroneals and TA that fulfill significant dynamic defense mechanisms at the ankle joint complex might contribute to the recurrence of ankle sprains.

Reaction Time at Proximal Joints

Beckman and Buchanan (1995) initially revealed significantly faster gluteus medius RT bilaterally in response to the traditional 30-degree ankle inversion trapdoor during standing in individuals with multiple ankle sprains (R: 127.35 ± 6.02 msec, L: 120.71 ± 6.16 msec) compared to healthy controls (R: 150.49 ± 6.49 msec, L: 136.24 ± 5.88 msec). In contrast, Donanue et al. (2014) investigated gluteus medius RT during walking on a walkway with a built-in bilateral trapdoor mechanism designed to freely fall to 30-degree inversion in individuals with and without CAI and found no between-group differences in gluteus medius RT (CAI: 58.19 ± 2.26 msec, H: 59.88 ± 2.30 msec). Those differences in findings may suggest neuromuscular responses at proximal joints differ based on the dynamic perturbation task. Moreover, the lack of faster gluteus medius activation to a sudden inversion perturbation in CAI individuals may fail to provide dynamic stability at the hip for excessive frontal-plane ankle movements associated with CAI to prevent recurrent ankle sprains during gait.

Stretch Reflex

CAI individuals have altered neuromuscular control, especially during dynamic tasks (e.g., postural control, gait). In general, neuromuscular control is governed with spinal or supraspinal mechanisms to modulate adequate reflexive muscle response for maintaining stability at the ankle joint complex. Additionally, reflexive muscle response is initiated by afferent feedback from peripheral mechanoreceptors. Thus, peripheral deafferentation displayed with CAI may predominantly decrease spinal reflex excitability of dynamic ankle stabilizers (e.g., peroneals, soleus, TA). The ongoing reflexive response that alters alpha motoneuron excitability following an initial ankle sprain is referred to as an arthrogenic muscle response (Palmieri et al., 2004; McVey et al., 2005; Sedory et al., 2007; Hertel, 2008). Arthrogenic muscle response (AMR) is characterized by either inhibition (a neurologic decline in muscle activation) or facilitation (a neurologic incline in muscle activation) of neural drive to muscles (McVey et al., 2005). Indeed, muscle weakness reported with CAI has been speculated to be a consequence of arthrogenic muscle inhibition (AMI) increasing susceptibility to subsequent ankle sprains (Palmieri et al., 2004; McVey et al., 2005). AMR may be modulated with presynaptic and postsynaptic mechanisms mediated by inhibitory and excitatory interneurons (Palmieri et al., 2004; Sedory et al., 2007). Although the mechanisms of excitatory interneurons are still unknown, inhibitory interneurons are suggested to play a critical role in the regulation of neuromuscular control (Palmieri et al., 2004). The primary inhibitory mechanisms of AMI are hypothesized to be presynaptic and postsynaptic inhibitions. Hence, paired reflex depression (presynaptic inhibition) and recurrent inhibition (postsynaptic inhibition) measurements are proposed to discriminate between CAI individuals and healthy controls (Hopkins & Ingersoll, 2000; Sefton et al., 2009).

Spinal Reflex Excitability

The Hoffman reflex (H-reflex) is utilized to measure spinal reflex excitability. The electrical stimulation of a peripheral nerve (e.g., sciatic, posterior tibial) is utilized to induce H-reflex, which is an electrical analog of the monosynaptic stretch reflex (Palmieri et al., 2004). The amplitude of H-reflex is indicative of the alpha motor neuron activity of the corresponding motor neuron pool. A decrease in the maximal H-reflex amplitude represents AMI, while an increase in the maximal H-reflex amplitude denotes more eminent muscle activation (McVey et al., 2005). The electrical stimulation of a peripheral nerve also sends action potentials in an orthodromic direction to the neuromuscular junction and causes muscle contraction (M-wave). Therefore, maximal H-reflex is usually normalized to maximal M-wave (H_{max}:M_{max} ratios). The H_{max}:M_{max} ratios quantify the number of alpha motor neurons an individual is capable of activating (H_{max}) within the entire alpha motor neuron pool of the corresponding muscle available to be activated (M_{max}) (Palmieri-Smith et al., 2004). Researchers have suggested lower H_{max}:M_{max} ratios are an indication of AMI (McVey et al., 2005; Hopkins et al., 2009).

McLeod et al. (2015) assessed H_{max}:M_{max} ratios of PL during lying supine in individuals with and without CAI and reported no group differences. Consistently, no group differences have been reported in H_{max}:M_{max} ratios of TA, PL, or soleus during reclining (Doeringer et al., 2009, 2010). In contrast, McVey et al. (2005) identified significantly decreased H-reflex amplitudes (modulation) of TA, PL, and soleus utilizing

H_{max}:M_{max} ratios during prone lying in the injured-limb compared to the uninjured-limb in individuals with unilateral CAI. Therefore, researchers have concluded the presence of AMI in TA, PL, and soleus in CAI (McVey et al., 2005). Subsequent studies also support the notion of AMI in soleus and PL during seated or prone positions (Palmieri-Smith et al., 2009; Otzel et al., 2019). Additionally, Kim et al. (2012) revealed significantly reduced H-reflex modulation of soleus and PL during the transition from simple to more challenging body positions (prone-to-bipedal, bipedal-to-unipedal, prone-to-unipedal) in the injured-limb (Prone: 14.0 6 ± 9.11%, Bipedal: -19.30 ± 15.90%, Unipedal: -2.27 ± 16.16%) compared to the uninjured-limb (Prone: 26.44 ± 11.84%, Bipedal: -9.64 ± 12.97%, Unipedal: 19.23 ± 17.34%) of CAI and matched injured-limb (Prone: 26.38 ± 14.16%, Bipedal: -10.41 ± 11.14%, Unipedal: -18.51 ± 18.22%) and the uninjured-limb of healthy controls (Probe: 27.59 ± 16.55%, Bipedal: -10.03 ± 10.87%, Unipedal: 20.19 ± 19.90%).

In the subsequent study conducted by Kim et al. (2016), they reported significantly decreased down-modulation of H-reflex in soleus and PL from a prone-to-unipedal stance in CAI (soleus: -5.53 ± 38.9%, PL: 12.10 ± 27.50%) compared to healthy controls (soleus: 37.37 ± 19.50%, PL: 33.82 ± 15.60%). The researchers also noted a strong positive relationship between H-reflex modulation of PL and postural control in a unipedal stance with a measurement of TTB minima in CAI compared to healthy controls (Kim et al., 2016). Specifically, a significant reduction in H-reflex modulation of PL in unipedal stance from prone lying was associated with less time needed for the center-of-pressure (COP) to reach the AP boundaries of the base of support in CAI. Additionally, 33% of the variance in postural control deficits among CAI individuals were explained by the alteration in the H-reflex modulation of PL. Conversely, only 19% variance in postural control deficits with CAI was explained by the alteration in H-reflex modulation of the soleus due to its higher variability presented in the H-reflex excitability. Those findings suggest postural control deficits are not only influenced by impaired proprioception associated with CAI but by an inadequate reflexive muscular control. Therefore, it is critical to examine alpha motoneuron excitability (H_{max}:M_{max} ratios) at spinal and supraspinal levels.

Several researchers implied H-reflex amplitude highly depends on task constraints, and greater down-modulation of H-reflex has been identified with an increase in task complexity (e.g., lying to standing, standing to walking, walking to running) (Capaday & Stein, 1986, 1987; Katz & Pierrot-Deseilligny, 1999; Taube et al., 2008; Pinar et al., 2010; Thompson et al., 2016). When transitioning into more complex tasks, motor control at the supraspinal level provides greater precision of movement by minimizing and correcting reflexive oscillations at the ankle. Additionally, a reduction in H-reflex amplitude has been proposed to be an indication of a change in reflexive controls from spinal to supraspinal mechanisms (Taube et al., 2008; Kim et al., 2012). Therefore, decreased down-modulation of H-reflex in soleus and PL during the postural transitions in CAI may be a consequence of the alteration in supraspinal motor control mechanisms (Kim et al., 2012, 2016). Furthermore, the reduced baseline alpha motor neuron pool excitability (MNPE) displayed during prone lying in CAI individuals may have contributed to their inability to suppress H-reflex modulation, especially transitioning from simple to more complex tasks.

Palmieri-Smith et al. (2009) investigated whether AMI suppresses dynamic defense to a sudden inversion perturbation at the ankle during walking. The study examined H-reflex modulation of PL utilizing H_{max}:M_{max} ratios during prone lying and EMG activity in PL during walking on a runway with a built-in bilateral trapdoor mechanism designed to freely fall to 30-degree inversion and relationship between these variables in individuals with unilateral CAI and healthy controls (Palmieri-Smith et al., 2009). Although no relationship was found between H_{max}:M_{max} ratios and dynamic EMG amplitude in PL, AMI in PL existed in individuals with unilateral CAI. Specifically, those individuals with unilateral CAI demonstrated significantly lower H_{max}:M_{max} ratios and PL activation in the injured-limb (H_{max}:M_{max}: 0.323 $\pm$ 0.161, EMG: 1.7 $\pm$ 1.3) compared to the uninjured-limb (H_{max}:M_{max}: 0.399 $\pm$ 0.185, EMG: 3.3 $\pm$ 3.1), while no differences between limbs were noted in either variable for healthy controls (Palmieri-Smith et al., 2009). The electrically stimulated H-reflex is thought to activate small motor neurons, innervating slow-twitch fibers that generate less force compared to large motor neurons, innervating fast-twitch fibers that produce greater force (Knikou, 2008). Consequently, the absence of the relationship between H_{max}:M_{max} ratios and dynamic EMG amplitude in PL was considered due to a difference in the type of motor neurons recruited, such that larger motor neurons are activated to counter a sudden perturbation during walking (Palmieri-Smith et al., 2009). For instance, greater H_{max}:M_{max} ratios of PL existed in healthy individuals when the electrical stimulation was probed in combination with a 30-degree inversion trapdoor test in a bipedal stance compared to standing on a flat surface (Sefton et al., 2007). The electrical stimulus was probed at the initiation of inversion trapdoor perturbation to eliminate the effect of Ia-afferent feedback from a change in muscle length and centrally modulated descending premotor response that supports a significant contribution of large motor neurons, innervating fast-twitch fibers as dynamic defense mechanisms.

A few researchers have investigated the relationship between H-reflex modulation of the dynamic ankle stabilizing muscle (i.e., soleus), utilizing H_{max}:M_{max} ratios and self-reported ankle function measures (FAAM-ADL and Sports subscale) while transitioning from simple to more complex postural control in individuals with CAI (Harkey et al., 2016; Kim et al., 2016). There was a moderate positive relationship between H-reflex modulation of soleus and self-reported ankle function measures that CAI individuals who report lower scores on FAAM-ADL and Sports subscale had less H-reflex modulation of soleus during the postural transition from prone to the unipedal stance (Kim et al., 2016). Additionally, a moderate to the stronger quadratic association between H-reflex modulation of soleus and self-reported ankle function measures were found that CAI individuals with the lowest and the highest H-reflex modulation of soleus expressed the lowest scores on FAAM-ADL and Sports subscale (Harkey et al., 2016). Thus, researchers concluded that either suppression or facilitation of reflexive excitability in soleus results in an increase of self-reported ankle function (Harkey et al., 2016). A study found no group differences in ankle joint laxity in both sagittal (AP) and frontal (ML) planes in CAI individuals (AP: 20.84 $\pm$ 5.66 mm, ML: 60.20 $\pm$ 13.59 mm) compared to healthy controls (AP: 21.22 $\pm$ 4.99 mm, ML: 59.78 $\pm$ 18.90 mm), despite a significant reduction in H-reflex modulation of soleus in these individuals with CAI (0.41 $\pm$ 0.18) relative to healthy controls (0.50 $\pm$ 0.17) (Bowker et al., 2016). Those findings may suggest that spinal reflex excitability relating to self-reported ankle instability is a result of supraspinal mechanisms and peripheral afferent sensory feedback from mechanoreceptors in the cutaneous, musculotendinous units, and muscle (Bowker et al., 2016).

Currently, there are a few studies that investigated changes in motoneuron excitability following focal ankle joint cooling (FAJC) in individuals with and without CAI (Doeringer et al., 2009; Kim et al., 2015). The effect of FAJC was very similar in CAI individuals and healthy controls during prone lying and reclined seating that FAJC immediately increased the H-reflex amplitude of soleus and PL (Doeringer et al., 2009; Kim et al., 2015). However, FAJC did not alter the H-reflex amplitude of soleus and PL during unipedal or bipedal stance (Kim et al., 2015). Those results suggest that peripheral afferent sensory feedback, specifically cutaneous receptors (i.e., thermoreceptors) around the ankle joint complex contributes to spinal reflex activity in non-weight bearing conditions in individuals with CAI and healthy controls (Kim et al., 2015). Unlike maintaining a prone position, postural control in weight-bearing bipedal and unipedal stance may require down-modulation of H-reflex amplitude in the ankle stabilizing muscles to prevent greater reflexive muscle response. Therefore, it is speculated that the supraspinal influence of down-modulating (supraspinal inhibition) the H-reflex amplitude of soleus during more challenging tasks (bipedal and unipedal stance) was greater than the disinhibition of Ia-afferent feedback promoted by FAJC at the spinal level (Kim et al., 2015). Furthermore, the most recent study investigated the effect of whole-body vibration (WBV), which is hypothesized to facilitate acute changes in alpha motoneuron excitability by increasing sensitivity of gamma motor neurons, on H_{max}:M_{max} ratios of soleus in individuals with and without CAI (Otzel et al., 2019). Although CAI individuals demonstrated 25% lower baseline H_{max}:M_{max} ratios of soleus compared to healthy controls, a single WBV session did not change the alpha motor neuron recruitment in either CAI or healthy controls (Otzel et al., 2019). Those findings may suggest the influence of presynaptic or postsynaptic recurrent inhibition to supraspinal inhibition of H-reflex amplitude.

The plasticity of reflexive muscle response is primarily influenced by centrally mediated descending commands and presynaptic and postsynaptic inhibition and facilitation (Sefton et al., 2007, 2007). In theory, presynaptic and postsynaptic inhibition suppress Ia-afferent feedback to regulate undesired reflexive muscle response. A few researchers have investigated inhibitory response measuring the second peak-to-peak H-reflex amplitude (H_2) relative to the initial H-reflex (H_1) (H_2:H_1 ratios) in response to paired stimuli probed utilizing recurrent inhibition (RI) and paired-reflex depression (PRD) protocols in CAI (Sefton et al., 2007, 2007, 2008; Thompson et al., 2019). The RI protocol applies a maximum stimulus (S_{max}) immediately following the initial stimulus to probe alpha MNPE to measure activation of Renshaw cells (RCs) or Ib inhibition interneurons (Ib inhibition), which limits the firing of alpha motor neurons in response to S_{max} (Trimble et al., 2000; Sefton, et al., 2007, 2007; Knikou, 2008). Whereas, the PRD protocol applies paired electrical stimuli in the short intervals of 50- to 80-msec to negate the influence of recurrent RCs or the Ib inhibition on the latter stimulus and to activate pre-synaptic inhibitions, resulting in a depression of the second H-reflex (Trimble et al., 2000; Palmieri et al., 2005; Knikou, 2008; Fioravante & Regehr, 2011). Therefore, an increase in H_2:H_1 ratios in response to RI (postsynaptic inhibition) and PRD (presynaptic inhibition) indicate increased activation of presynaptic and postsynaptic inhibition (Sefton et al., 2007).

Sefton et al. (2008) investigated the physiologic mechanisms (i.e., presynaptic and postsynaptic inhibitions) of the H-reflex modulation of the soleus during a change in stance positions in individuals with and without CAI. No group differences were found in the H-reflex modulation of the soleus, however altered presynaptic and postsynaptic inhibitions were present in individuals with CAI. Specifically, CAI individuals displayed an inability to modulate soleus PRD while transitioning from the bipedal stance (84.10 $\pm$ 11.80 %) to the unipedal stance (83.18 $\pm$ 11.10 %), whereas healthy controls demonstrated a 15% increase in soleus PRD from the bipedal stance (85.20 $\pm$ 11.80 %) to the unipedal stance (70.79 $\pm$ 15.50 %). Similarly, an increase in soleus PRD was displayed while transitioning from the bipedal stance on a stable surface (57.37 $\pm$ 7.43 %) to the unipedal stance on an unstable foam surface (62.48 $\pm$ 8.01 %) in healthy individuals (Sefton et al., 2007b). The study also revealed significantly decreased H-reflex modulation of soleus during the unstable surface condition compared to the stable surface condition (Sefton et al., 2007). Therefore, researchers concluded presynaptic inhibition provides dynamic stability at the ankle by inhibiting greater reflexive muscle response to environmental changes (stable to unstable) while maintaining an upright stance (Sefton et al., 2007b). Given the conclusion, the inability to modulate PRD while transitioning from the bipedal to the unipedal stance to maintain an upright stance may contribute to postural control deficits in CAI individuals. Furthermore, greater RI under both bipedal (DL) and unipedal (SL) stance conditions were exhibited in CAI (DL: 90.75 $\pm$ 5.30 %, SL: 89.80 $\pm$ 3.90 %) compared to healthy controls (DL: 83.40 $\pm$ 12.00 %; SL: 83.91 $\pm$ 8.80 %) (Sefton et al., 2008). Since the physiological mechanisms of postsynaptic inhibition are suggested to provide a more generalized long-lasting modulation, RI was not sensitive to adapt to change in stance conditions in either group (Sefton et al., 2008).

In contrast to Sefton et al. (2007, 2008), Thompson et al. (2019) displayed a significant decrease in PRD (disinhibition of presynaptic inhibition) of soleus that was reduced by 330% in the bipedal stance and 160% in the unipedal stance in CAI individuals (DL: 97.9 $\pm$ 33.7 %, SL: 83.0 $\pm$ 17.2 %) compared to healthy controls (DL: 29.5 $\pm$ 10.5 %, SL: 50.82 $\pm$ 11.0 %). Those conflicts in results may be related to the methodological differences that Thompson et al. (2019) evaluated PRD in the intervals of 100-msec instead of 80-msec utilized in the earlier studies (Sefton et al., 2007, 2008). Implementing 100-msec has been suggested to elicit the response sensitive to both post-activation depression and heteronymous facilitation, resulting in disinhibition of presynaptic inhibition (Thompson et al., 2019). Thompson et al. (2019) also utilized stimulus intensity at 50% of M_{max} ($H_{50\%}$), while 10-30% of M_{max} is typically applied (Kim et al., 2019). Similar to the findings of Sefton et al. (2008), there were no significant group differences in maximal H-reflex modulation of the soleus during maintaining posture in a bipedal and unipedal stance. Instead, the study found CAI individuals significantly increased H-reflex modulation of soleus when spinal reflex excitability was probed at $H_{50\%}$ during unipedal stance in CAI (16.9 $\pm$ 3.5 %) compared to healthy controls (10.36 $\pm$ 3.7 %) (Thompson et al., 2019). Consequently, researchers have concluded that the stimulus intensity of 50% of M_{max} might be a more sensitive measure to detect changes in spinal reflexive response compared to H_{max}:M_{max} (maximal H-reflex to M-wave) response in CAI. Collectively, AMI and dysfunctions (e.g., postural control, muscle weakness, peroneals latency, increased EMD) have been reported with CAI may be a consequence of greater RI of reflexive muscle response compared to healthy individuals. Furthermore, the ability to modulate presynaptic inhibition regulating spinal muscle response, especially when task constraints become more complex, might be critical to providing sufficient dynamic stability at the ankle.

Cutaneous Reflex

Cutaneous reflex is another important aspect of the sensorimotor system. For instance, Futatsubayashi et al. (2013) indicated cutaneous reflex plays an integral role in regulating external perturbation during locomotion, especially to respond to obstacles. Cutaneous reflex can be facilitated by stimulating the skin surface or probing the sural nerve (cutaneous nerve), which innervates the cutaneous mechanoreceptors in the lateral margin of the foot (Duysens & Levin, 2010; Futatsubashi et al., 2013). Futatsubashi et al. (2013) is the only group that has investigated cutaneous reflex by measuring the magnitude of medium latency reflex (MLR), which is implied to fluctuate with the phase of gait in individuals with and without CAI. The MLR (80- to 120-msec post-stimulation) in PL, TA, medial gastrocnemius, and vastus lateralis were measured during maximum isometric contraction test of ankle eversion, plantarflexion, dorsiflexion, and knee flexion in a seated position (Futatsubashi et al., 2013). Researchers found significant suppression of MLR in PL and vastus lateralis around 25 to 30% of the maximum level of EMG activity in the injured-limb of CAI compared to matched injured-limb of healthy controls (Futatsubashi et al., 2013). Because there were no side-to-side differences in MLR in the injured-limb and the uninjured-limb of CAI, the researchers have speculated that the damage to the sural nerve is not the sole cause of the result (Futatsubashi et al., 2013). Therefore, the reduction of MLR in PL and vastus lateralis might be a result of an alteration in spinal and/or supraspinal neural mechanisms (e.g., interneurons mediating MLR).

Corticospinal (Supraspinal) Excitability

Corticospinal excitability is examined with transcranial magnetic stimulation (TMS) that stimulates the motor cortex to examine descending corticospinal motor pathways in the corresponding muscle (e.g., PL, TA, Soleus). The TMS measurement is typically quantified with a peak-to-peak amplitude of motor evoked potential (MEP) in the targeted muscle, while the direct stimulation is applied to the motor cortex in a seated position. Additionally, TMS intensity parameters are obtained when the targeted muscle is at rest (resting motor threshold [RMT]) or during an active contraction (active motor threshold [AMT]) (Pietrosimone & Gribble, 2012; McLeod et al., 2015). The TMS is also utilized to quantify corticospinal inhibition through the measurement of the cortical silent period (CSP). The CSP is a brief pause in the voluntary muscle activities following the magnetic pulse (Needle et al., 2013). The CSP represents the activity of the inhibitory neurotransmitter, GABAerig, mediating the inhibitory pathway between the motor cortex, basal ganglia, and thalamus (Needle et al., 2013). Thus, the shorter CSP is associated with less inhibition, greater muscle tone, and facilitation in descending drive (Needle et al., 2013).

Pietrosimone and Gribble (2012) investigated the corticospinal excitability of PL in the measurement of RMP in individuals with unilateral CAI and healthy controls. Researchers found significantly higher RMT in the injured-PL (60.8 $\pm$ 8.4 %) and the uninjured-PL (59.1 $\pm$ 8.99 %) of CAI compared to the assigned injured-PL (52.8 $\pm$ 8.56 %) and uninjured-PL (52.0 $\pm$ 7.0 %) of healthy controls, respectively (Pietrosimone & Gribble, 2012). An increase in bilateral RMT implies that greater stimulus was required to excite cortical neurons of PL, meaning there is decreased descending corticospinal excitability. Therefore, the researchers have concluded that CAI individuals have more difficulty generating descending motor commands to PL (Pietrosimone & Gribble, 2012). In contrast, McLeod et al. (2015) examined the corticospinal excitability of PL and observed significantly decreased MEP amplitude while the motor cortex was stimulated at 100% of AMT and 105% of AMT in the injured-PL (100%: 0.014 $\pm$ 0.008, 105%: 0.021 $\pm$ 0.009) and the uninjured-PL (100%: 0.015 $\pm$ 0.007, 105%: 0.023 $\pm$ 0.013) of CAI compared to the assigned injured-PL (100%: 0.023 $\pm$ 0.031, 105%: 0.029 $\pm$ 0.026) and uninjured-PL (100%: 0.021 $\pm$ 0.022, 105%: 0.034 $\pm$ 0.037) of healthy controls. However, no group differences in MEP amplitude were displayed in the injured-PL and the uninjured-PL when the motor cortex was stimulated at higher intensities from 110 to 140% of AMT (McLeod et al., 2015). Reduction in MEP amplitude at lower intensities in CAI individuals compared to healthy controls implies a diminished motor command to PL resulting in less PL motoneuron pool excitability (McLeod et al., 2015). Whereas no group differences in MEP amplitude at higher intensities implies that CAI individuals may require greater corticospinal excitability to achieve the same motor outcome as healthy controls (McLeod et al., 2015).

Terada et al. (2016) supported the findings of McLeod et al. (2015) in the soleus muscle. Researchers have selected soleus instead of PL because the muscle takes a key role in adjusting postural sway over the base of support during standing and gait (Terada et al., 2016). The study assessed CSP in soleus in relation to MEP at 120% of AMT in individuals with and without CAI and demonstrated significantly greater CSP:MEP_{120} ratios in CAI individuals compared to healthy controls. An increase in CSP duration, which signifies the inhibitory GABAerig neurotransmitter, may indicate a potential increase in corticospinal inhibition necessary to regulate unwanted spinal reflex and alpha-gamma coactivation. The potential corticospinal inhibition of soleus in CAI was speculated to be a result of protective mechanisms to maintain the ankle in a more loose, open-packed position to circumvent any associated pain with CAI (Terada et al., 2016). Conversely, the other studies have failed to support the findings of McLeod et al. (2015) and Terada et al. (2016) (Needle et al., 2013; Harkey et al., 2014). Harkey et al. (2014) investigated corticospinal excitability of PL and soleus at 100 and 120% of AMT in individuals with and without CAI and exhibited no group differences in the MEP amplitude of PL and soleus at 100 and 120% of AMT. Similarly, Needle et al. (2013) examined corticospinal excitability of PL, TA, and soleus in relation to ankle joint laxity and CSP (cortical inhibition) in individuals with unilateral CAI and healthy controls. Although no group differences were found on the corticospinal excitability/inhibition and the ankle joint laxity, there was a negative relationship between greater corticospinal excitability of soleus and the higher anterior displacement (ligamentous laxity) of the ankle with CAI (Needle et al., 2013). The relationship might be an indication of a compensatory strategy to heighten descending motor command to increase stiffness in the soleus to resist anterior displacement of the ankle.

TMS mapping was recently employed in a measurement of the corticomotor output area and volume associated with PL to examine the plastic change in the primary motor cortex (Kosik et al., 2017). The corticomotor output area estimates the size of the cortical representation of PL, whereas the volume estimates the total cortical excitability of PL. The smaller area suggests the devotion of fewer cortical neurons to activation of PL, and the lower volume implies the complexity of generating descending motor commands. The TMS mapping displayed that CAI individuals have decreased area and volume of the corticomotor representation of PL compared to healthy controls (Kosik et al., 2017). Those findings indicate individuals with CAI have fewer cortical neurons devoted to activation of PL with increased difficulty in producing descending motor commands (Kosik et al., 2017). Additionally, researchers discovered that greater chronicity from the initial LAS was correlated with greater restriction in the cortical area and volume associated with the PL within the motor cortex (Kosik et al., 2017).

Collectively, the literature may suggest a possible supraspinal alteration with CAI. Furthermore, evaluation of corticospinal excitability during dynamic tasks might reveal a greater insight into changes within the corticospinal pathway.

Postural Control and Balance

Postural control (maintenance of center-of-mass) and balance (maintenance of center-of-gravity) have been suggested as the best predictor for the incidence of an ankle sprain (Wang et al., 2006; Grassi et al., 2017). Strong evidence exists for postural control deficits in individuals with CAI, regardless of the differences in instrumental measurements. Freeman (1965) initially proposed that sustaining ankle sprain damages ligamentous and articular mechanoreceptors and compromises peripheral sensory feedback (deafferentation) resulting in postural control deficits. Indeed, successful maintenance of postural control and balance depends on an individual's perceptual ability to obtain affordance, that is exteroceptive information of the environment (Greeno, 1994). Additionally, sensory mechanoreceptors in the ligament, joint capsule, musculotendinous units, muscle, and cutaneous (haptic perception) mediate perceiving affordance. Therefore, maintaining postural control and balance may depend on somatosensory feedback, especially from the ankle.

Peripheral sensory feedback (somatosensory, vision, vestibular) in healthy individuals initiates involuntary (reflexes) and voluntary muscle contractions to provide dynamic stability at the ankle controlling postural sway. However, somatosensory deficits associated with CAI may alter their muscular recruitment of the ankle by compensating with redundant somatosensory feedback available at proximal joints (knee, hip). For example, alteration in muscle activities at proximal joints (knee, hip) and implementation of hip strategy while maintaining postural control have been reported in CAI (Pintsaar et al., 1996; Caulfield & Garrett, 2002; Gribble et al., 2007; Gribble & Robinson, 2009; Pope et al., 2011) Furthermore, individuals with CAI upregulate reliance on visual feedback to compensate for somatosensory deficits (Song et al., 2016). However, despite those perceptual alterations in obtaining affordance, postural control still deficits persist in individuals with CAI (Arnold et al., 2009; Munn et al., 2010).

Assessment of Static Balance

Traditional Measures

The most common traditional quantification measure of static postural control is the spatial and temporal measurement of the COP in a single-limb stance. The COP is typically assessed utilizing clinical non-instrumented Romberg test protocol on a force plate. In addition, the most common variables of COP categories are the excursion, area, velocity, and total path length in both anteroposterior (AP) and mediolateral (ML) directions. The greater excursion, larger area, faster velocity, and greater path length (radius) of the COP excursion indicate an impaired sensorimotor system associated with poor postural control.

Tropp et al. (1984) examined the COP area during maintaining postural control in a single-limb stance with eyes-open in individuals with unilateral CAI compared to healthy controls. The study presented significantly higher areas in both injured-limb and uninjured-limb of CAI, respectively, compared to healthy controls (Tropp et al., 1984). However, in a subsequent investigation utilizing the same methods, Tropp et al. (1985)

did not find bilateral limb differences (injured, uninjured) within soccer players with unilateral CAI. Similarly, Hubbard et al. (2007) did not identify the between-group and bilateral limb differences examining the COP area during maintaining postural control in a single-limb stance with and without eyes-closed in individuals with unilateral CAI and healthy controls.

Several studies have investigated the variables of COP velocity (Hale et al., 2007; Hubbard et al., 2007) and length (Ross & Guskiewicz, 2004; Lee et al., 2006) when assessing static postural control in CAI. Hale et al. (2007), who examined COP velocity in a single-limb stance with and without eyes-closed in individuals with unilateral CAI and healthy controls, discovered no group differences in bilateral limb, respectively (Hale et al., 2007). Those findings are consistently supported by Hubbard et al. (2007), who did not reveal between group or side-to-side differences in COP velocity in individuals with unilateral CAI and healthy controls. Lee et al. (2006) displayed a significantly greater mean radius of COP excursion during postural control in bilateral limbs (injured, uninjured) with and without eyes-closed in individuals with unilateral CAI, respectively, compared to healthy controls. However, no bilateral limb differences within each group were reported in the study (Lee et al., 2006). Similarly, Ross and Guskiewicz (2004) did not identify group differences in AP and ML mean COP excursion during maintaining postural control in the injured-limb with eyes-open in individuals with unilateral CAI (AP: 0.79 ± 0.25 cm, ML: 0.57 ± 0.11 cm) and matched healthy controls (AP: 0.70 ± 0.18 cm, ML: 0.55 ± 0.09 cm).

COP excursion measures (i.e., velocity, length) have been able to detect postural control deficits associated with acute LAS. Specifically, several researchers have demonstrated a substantial increase in COP excursion measures not only in the injured-limb but in the uninjured-limb following an acute LAS (Friden et al., 1989; Goldie et al., 1994; Leanderson et al., 1996; Holme et al., 1999; Hertel, 2002; Evans et al., 2004). However, those contradictory findings and the lack of group or side-to-side differences found in most of the previous studies in CAI (Tropp et al., 1985; Ross & Guskiewicz, 2004; Hale et al., 2007; Hubbard et al., 2007) may suggest COP measures (excursion, area, velocity, total path length) may not be sensitive enough to detect static postural control deficits associated with CAI. Additionally, a meta-analysis indicates the COP area did not denote postural control deficits in CAI (Arnold et al., 2009). Researchers also concluded that the greatest standard difference of the mean (SDM = 1.818) was revealed for time measures, such as time-to-boundary (TTB) (Arnold et al., 2009). The TTB measures provide information about the COP excursion in relation to the boundaries of the base of support that are not addressed by traditional COP measures (excursion, area, velocity, total path length). Indeed, the TTB measures of postural control were able to detect postural control deficits the traditional COP excursion measures were not able to detect when the plantar sensation was reduced in healthy individuals (McKeon & Hertel, 2008). Therefore, the TTB measures may be more sensitive in detecting postural control deficits associated with CAI.

Nonlinear Measures (Sample Entropy)

Nonlinear measures, especially approximate entropy (ApEN), have been suggested to identify postural control deficits in individuals with pathology (Cavanaugh, 2005). In contrast to linear measures, which analyze the magnitude of variability in COP variables, nonlinear measures focus on the evolutionary properties of COP variables. Specifically, nonlinear measure analysis supports the idea that postural control variables emerge over time through interaction within the physiological element (e.g., underlying neural control) and between task and environmental constraints. Consequently, nonlinear measures are theorized to quantify the flexibility and adaptability of the sensorimotor system underlying postural control (Stergiou & Decker, 2011). For instance, a decrease in nonlinear variability renders a predictable (periodic) sensorimotor system, whereas an increase in nonlinear variability renders an unpredictable (random) sensorimotor system, both resulting in an inflexible adaptation of the sensorimotor system to the task and environmental constraints (Harbourne & Stergiou, 2009).

A few researchers have implemented nonlinear measures (i.e., sample entropy) in COP variables during postural control in individuals with and without CAI. Glass et al. (2014) utilized sample entropy (SampEN) to analyze AP and ML directions of COP velocity in single-limb and double-limb stance. The study found a significant decrease in SampEN of resultant COP velocity in double-limb and single-limb stance, and COP velocity in ML direction in a single-limb stance in CAI individuals compared to healthy controls (Glass et al., 2014). Raffalt et al. (2019) consistently reported a significantly lower SampEN of COP excursion in AP and ML directions during single-limb stance in

CAI compared to healthy controls. Those findings indicate a less flexible and adaptable sensorimotor system in postural control, resulting in CAI individuals to freeze the degree of freedom to restrain excessive movement in the frontal-plane associated with CAI. Further, Raffalt et al. (2019) identified a significant directional effect of COP excursion in SampEN during single-limb stance with and without eyes-closed or wearing an orthosis. More specifically, deprived visual feedback significantly increased SampEN in AP direction and decreased SampEN in ML direction for both CAI and healthy individuals (Raffalt et al., 2019). However, increased somatosensory feedback from wearing an orthosis significantly increased SampEN in the AP direction for CAI and SampEN in the ML direction for healthy controls (Raffalt et al., 2019). Collectively, current evidence suggests different sensory feedback may alter motor strategies in postural control.

Copers may retain adequate sensorimotor system function following an initial ankle sprain without developing dysfunction (Wikstrom & Brown, 2014). The most recent study conducted by Terada et al. (2019), who investigated SampEn of COP excursion during a single-limb stance in individuals with CAI, copers, and healthy controls, did not find between-group differences in SampEN of COP excursion in a single-limb stance. However, a trend was noted that CAI individuals have greater SampEN compared to copers and healthy controls (Terada et al., 2019). Researchers assumed inconsistency in study findings is a result of sampling frequency since lower sampling rate (50 Hz) in COP velocity Glass et al. (2014) utilized has been suggested to produce higher SampEN value (Rhea et al., 2015; Terada et al., 2019). Considering disparities in findings yet exists among Terada et al. (2019) and Raffalt et al. (2019), who both utilized the same sampling frequency of 100 Hz, the inclusion of CAI individuals with a wide range of self-reported functional deficits level may have contributed to the trend of increasing SampEN in CAI compared to copers and healthy controls (Terada et al., 2019). In other words, the difference in self-reported functional deficits level may lead to different sensorimotor responses or compensatory strategies (or both).

Time-to-Boundary Measures

The TTB measures are another common way of evaluating postural control in a single-limb stance and have been suggested to be effective in detecting postural control deficits associated with CAI (Hertel et al., 2006; Hertel & Olmsted-Kramer, 2007). The TTB measures provide spatiotemporal characteristics of postural control in a single-limb stance by evaluating velocity, position, and direction of each COP excursion point in relation to the boundaries (borders) of the base of support (foot). In contrast, the traditional COP measures evaluate the entire time series of the data to represent a mean. However, every COP data point may not be equally important to detect postural control deficits associated with CAI. Thus, the TTB measures selectively focus on the data point that yields minima and to assess instability in the COP time series.

In order to maintain postural control, individuals must control the COP excursion within the boundaries of the foot. Otherwise, individuals would fall or lose balance if the COP excursion crosses or reaches too close to the boundaries. The COP excursion points closer to the boundaries, known as TTB minima, allow less time for the sensorimotor system to make postural corrections while COP excursion continues to move at high-velocity. In theory, evaluation of TTB minima provides insight into strategies CAI individuals utilize to maintain postural control. Additionally, examining the standard deviation of TTB minima demonstrates alteration in the sensorimotor system function (Hertel & Olmsted-Kramer, 2007; McKeon & Hertel, 2008).

There are only a few studies that examined postural control in a single-limb stance with eyes-open utilizing the TTB measures in individuals with and without CAI. Hertel and Olmsted-Kramer (2007) first found the mean of TTB minima in AP and ML directions is significantly lower in the bilateral limb of CAI (AP-INJ: 5.11 $\pm$ 1.89 sec, AP-UNI: 5.72 $\pm$ 1.48 sec, ML-INJ: 1.81 $\pm$ 0.68 sec, ML-UNI: 1.96 $\pm$ 0.47 sec) compared to healthy controls (AP-INJ: 8.99 $\pm$ 2.13 sec, AP-UNI: 8.64 $\pm$ 3.86 sec, ML-INJ: 2.54 $\pm$ 1.17 sec, ML-UNI: 2.57 $\pm$ 0.93 sec), respectively. However, there were no side-to-side differences within the group of CAI and healthy controls (Hertel & Olmsted-Kramer, 2007). Similarly, McKeon and Hertel (2008) reported a significantly lower absolute mean of TTB minima in AP and ML directions during maintaining posture in a single-limb stance with eyes-closed in CAI (AP: 1.36 $\pm$ 0.40 sec, ML: 0.48 $\pm$ 0.10 sec) compared to healthy controls (AP: 1.61 $\pm$ 0.47 sec, ML: 0.53 $\pm$ 0.10 sec). McKeon and Hertel (2008) also assessed postural control with eyes-open and identified no group differences for an absolute mean of TTB minima in AP and ML directions in individuals with and without CAI. When eyes are closed, postural control is influenced more by somatosensory and vestibular feedback (Hugel et al., 1999). Although there are contradictory findings concerning postural control deficits with eyes-open in CAI compared to healthy controls (Hertel & Olmsted-Kramer, 2007; McKeon & Hertel, 2008), findings of McKeon and Hertel (2008) may confirm the significant contribution of vision in the sensorimotor system to maintain postural control.

For the standard deviation of TTB measures, Hertel and Olmsted-Kramer (2007) showed less variability in TTB minima in AP and ML directions during maintaining posture in a single-limb stance with eyes-open in individuals with CAI than healthy controls. Consistently, McKeon and Hertel (2008) found significantly decreased variability in TTB minima in AP direction during postural control with eyes-closed in CAI compared to healthy controls. Pope et al. (2011) also supported the distribution of COP excursion data points in an anterior direction with the removal of vision in CAI. The greater COP excursion in anterior positioning of the foot will put the stance ankle into stable, closed-packed dorsiflexion. Hence, researchers have speculated that increasing task constraints of removing vision during postural control in a single-limb stance led CAI individuals to freeze the degree of freedom at the ankle in anterior position for more stability compared to healthy controls (Pope et al., 2011).

Additionally, a significant decrease of variability in TTB minima in AP and ML directions during maintaining posture in a single-limb stance was denoted in healthy individuals when transitioning from eyes-open to eyes-closed (McKeon & Hertel, 2008). This finding implies the healthy sensorimotor system has flexible adaptability to provide the boundaries to the degree of freedom in involved elements (e.g., joints, motor units) by freezing or unfreezing based on organismic (individual characteristics), task, and environmental constraints. Thus, the greater reduction of variability in the TTB minima measures with CAI compared to healthy controls has been suggested to reflect less flexibility and adaptability of the sensorimotor system in CAI individuals (Hertel & Olmsted-Kramer, 2007). Collectively, the less adaptable and flexible sensorimotor system in CAI may be limiting the ability to find effective postural control strategies, increasing their risk of recurrent ankle sprains.

Sensory Organization Test

In a recent systematic review, researchers concluded CAI individuals rely more on vision while maintaining postural control in a single-limb stance compared to healthy controls (Song et al., 2016). Moreover, researchers have speculated that reliance on vision is to compensate for somatosensory deficits associated with CAI to maintain postural stability in a single-limb stance (Song et al., 2016). The systematic review included 11 studies that determined postural control in single-limb stance with and without eyes-closed utilizing TTB scores, of which only three studies examined both CAI and healthy controls under the same methodological approaches. Maintenance of postural stability involves the interaction of somatosensory, visual, and vestibular feedback of the body orientation to the environmental surface and surroundings. Under normal circumstances with fixed surface, and visual surroundings, somatosensory feedback is responsive to maintain postural stability (Nashner, 1982). Therefore, in order to fully examine an individual's reliance on somatosensory or visual feedback to the maintenance of postural stability, the support surface and visual surroundings must be modified (Nashner, 1982). Specifically, an individual's ability to utilize somatosensory feedback arising from mechanoreceptors in the ankle joint complex, ligament, and plantar cutaneous can be examined by eliminating visual feedback with closed eyes. Whereas, visual feedback can be assessed by altering somatosensory feedback with the freely movable support surface (Nashner, 1982). With this in mind, merely reviewing those studies that only examined postural stability with and without eyes-closed would not imply increased use of visual feedback with CAI. In other words, the systematic review may be more confined to the conclusion that CAI individuals display somatosensory deficits compared to healthy controls.

The Sensory Organization Test (SOT) of the NeuroCom Smart Balance Master is a reliable computer-based balance measure that quantifies which sensory systems (somatosensory, vision, vestibular) an individual emphasizes (weights) while maintaining postural control in a double-limb stance with and without eyes-closed (Ford-Smith et al., 1995). The SOT has been developed based on the posturography to identify postural control deficits resulting from inadequate sensory-motor integration of the CNS (Shumway-Cook & Horak, 1986; Horak, 1997; Riemann et al., 1999). The SOT consists of six postural conditions that systematically manipulate an individual's sensory feedback in a combination of the sway-referenced support surface and visual surroundings with and without eyes-closed (Figure 2.6). Healthy individuals primarily weight on somatosensory feedback from the feet in contact with the support surface to maintain a static posture (Shumway-Cook & Horak, 1990). Thus, the sway-referenced support surface and visual surroundings that provide inaccurate somatosensory or vestibular feedback challenge the CNS to reweight on alternative sensory feedback for postural control. The completion of the SOT determines the equilibrium (EQ) balance score for each condition to allocate a sensory reweighting ratio of each sensory system that individuals utilize for postural control.

Collectively, the SOT would be suited to examine which sensory system is most emphasized (i.e., sensory reweighting system) while maintaining postural stability. Indeed, the SOT has been effective in detecting a change in postural control following a concussion (Mrazik et al., 2000; Riemann & Guskiewicz, 2000; Guskiewicz, 2001; Peterson et al., 2003) and with aging (Cohen et al., 1996). Nevertheless, there are only a few published studies that utilized the SOT, investigating the effect of Kinesiology tape and the elastic bandage on postural control in double-limb stance in individuals with CAI (de-la-Torre-Domingo et al., 2015; Alguacil-Diego et al., 2018; Yin & Wang, 2020). However, only one research group has used a true control condition (no tape) and no researchers reported the full spectrum of outcomes on the SOT, such as EQ balance scores of all six SOT conditions and sensory reweighting ratios (de-la-Torre-Domingo et al., 2015; Alguacil-Diego et al., 2018; Yin & Wang, 2020). In contrast, the study conducted by Sugimoto and Ross (2020) identified individuals with unilateral CAI who utilized sensory feedback similar to healthy controls while maintaining postural control in the injured-limb (i.e., no statistically significant group differences). In exploring the data, effect size values (0.33-0.40) were reported to demonstrate that CAI individuals may utilize less somatosensory and vestibular feedback, but more visual feedback for maintaining postural stability while standing on their injured-limb (Table 2.7). Effect size values for the uninjured-limb were small to moderate (0.22-0.54), suggesting CAI individuals may be associated with utilizing more visual and vestibular feedback for postural stability (Table 2.8) (Sugimoto & Ross, 2020). Despite the existence of a trend, the limitation of the study is the small number of CAI individuals (N = 6) included in the study. Therefore, future research should further investigate the sensory reweighting system with a larger sample size in individuals with CAI and include comparisons to both copers and healthy controls.

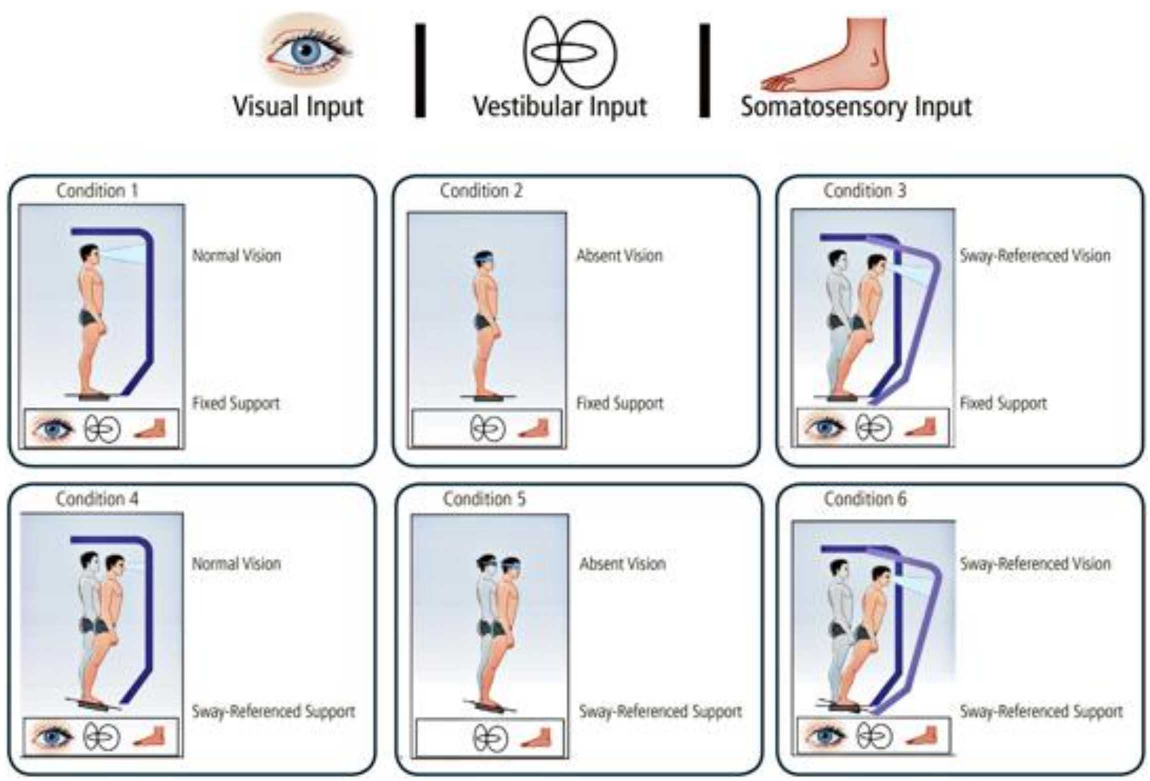

Figure 1.6. Six Conditions of the Sensory Organization Test (NeuroCom® International).

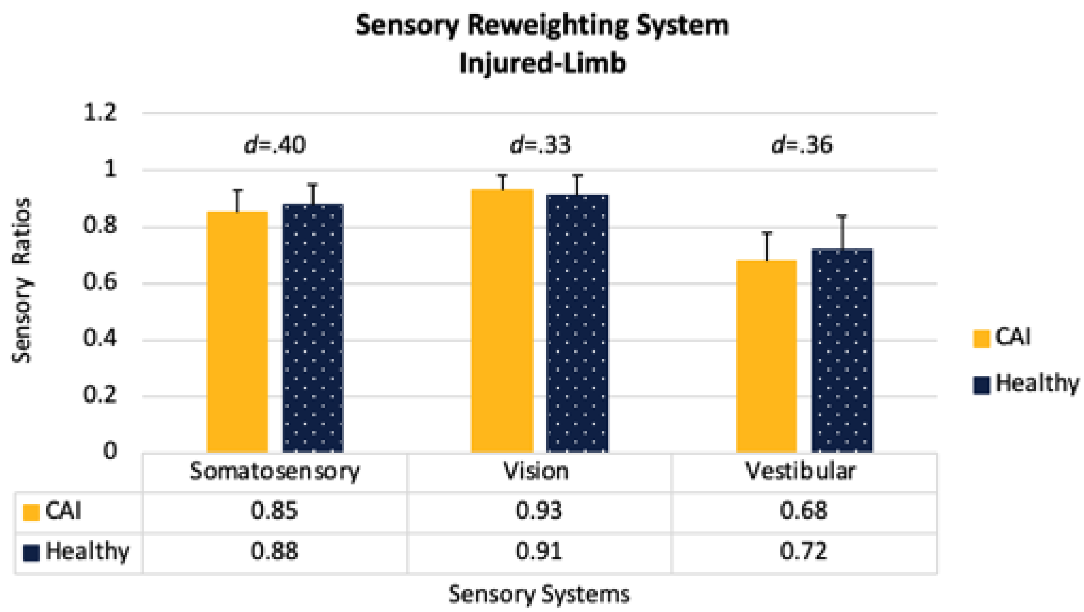

Figure 1.7. Sensory Reweighting Ratios in the Injured-Limb in Individuals with and without CAI (Sugimoto & Ross, 2020).

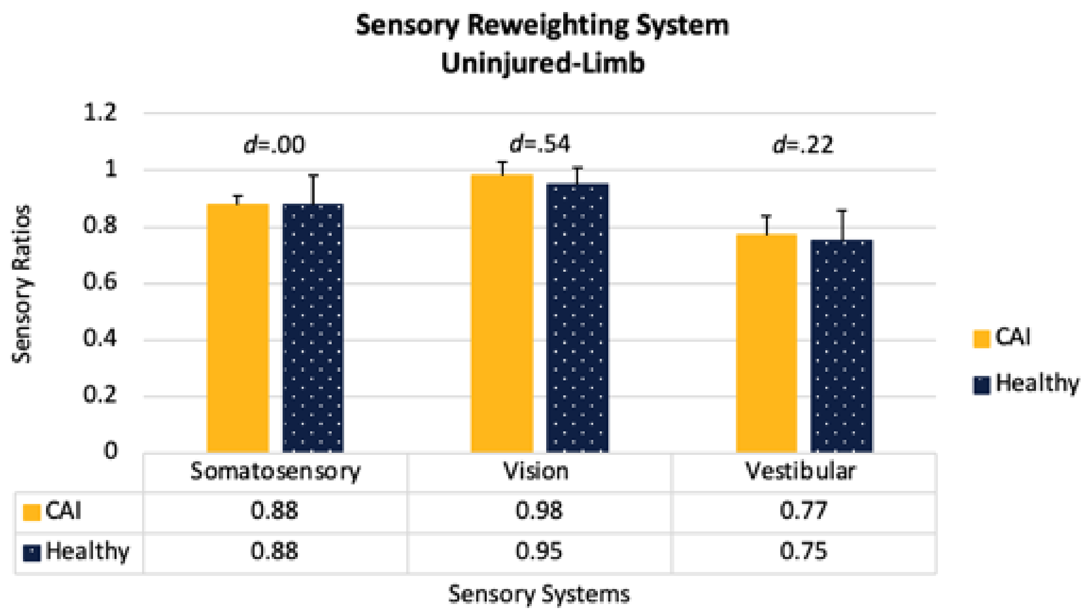

Figure 1.8. Sensory Reweighting Ratios in the Uninjured-Limb in Individuals with and without CAI (Sugimoto & Ross, 2020).

Adaptation Test

The Adaptation Test (ADT) of the NeuroCom Smart Balance Master is a computer-based balance measure that quantifies an individual's ability to react and adapt to a sudden change in the support surface while maintaining postural control in a double-limb stance (Olchowik et al., 2015). The ADT consists of two conditions of support surface perturbations that rotate in toe-up and toe-down directions (Figure 2.9). Each condition comprises five trials of equivalent perturbations of 8-degree amplitude, lasting 400-msec in 3 to 5-seconds randomized intervals. The sway energy score is measured while performing the ADT and extracts the amount of the center-of-gravity (COG) displacement during each trial of conditions. Healthy individuals may display greater postural sway in response to the first trial of each condition, but augment stability in subsequent trials (Nashner, 1976). Thus, a decrease in sway energy scores over trials is an indication of adaptability to unexpected changes in the environment (Nashner, 1976). Whereas the greater sway energy scores indicate insufficient ability to react and adapt to the change in the support surface (environment), resulting in postural instability (Nashner, 1976; Paquette et al., 2016). The ADT toe-up and toe-down conditions both have been effective in discriminating individuals with knee osteoarthritis (ADT toe-down) and with cerebellar ataxia (ADT toe-up) compared to healthy controls (Paquette et al., 2016; Park & Jung, 2018). In addition, a research group has utilized ADT and successfully examined the effect of Kinesiology tape on postural control in double-limb stance in individuals with CAI (Yin & Wang, 2020). Altogether, utilizing the ADT is an effective way to quantify postural recovery from a sudden perturbation without implementing a traditional trap door device.

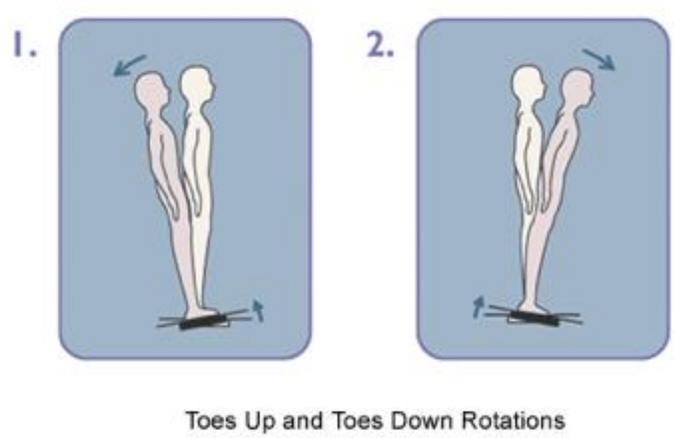

Figure 1.9. Two Conditions of the Adaptation Test (NeuroCom® International).

Assessment of Dynamic Balance

Hop Test

Dynamic functional tests are commonly utilized to evaluate neuromuscular control in the latter phase of rehabilitation in clinical settings (Munn et al., 2002). There are a limited number of studies that utilize different functional tests (e.g., lateral hop test, single hop for distance, 6-m hop for time, 30-m agility hop, triple crossover hop for distance, shuttle run) to evaluate functional performance in CAI individuals. Among those studies, functional performance deficits associated with CAI were not found utilizing most of the functional tests, which failed to stress the lateral ankle joints complex.

Docherty et al., (2005) utilized four unilateral hopping tests (figure-of-eight, side hop, up-down hop, single hop) to examine functional performance in individuals with unilateral CAI and healthy controls. Among those functional studies, only figure-of-eight and side hops produced functional performance deficits in CAI compared to healthy controls. Additionally, those functional tests (figure-of-eight, side hop) were positively related to functional ankle instability (FAI) index researchers utilized to assess self-perceived ankle instability (Docherty et al., 2005). The functional tests (up-down hop, single hop) which did not produce functional performance deficits primarily involved a sagittal-plane of motion (e.g., single hop for distance, 6-m hop for time, 30-m agility hop) (Docherty et al., 2005). Consistently, no group differences were reported utilizing agility hop tests in various face-forward directions in the injured-limb of CAI compared to matched injured-limb of healthy controls (Demeritt et al., 2002). In contrast, Jerosch et al.

(1995), who implemented a slanted hopping course consisting of the uneven surface stimulating frontal and rotational motions to evaluate functional performance, demonstrated a significant reduction in CAI compared to healthy controls. Therefore, considering the typical mechanisms of LAS involving lateral movement, researchers have concluded that the functional tests force an individual to move laterally, placing stress on the lateral ankle joint complex could detect functional performance deficits associated with CAI (Hertel, 2002; Docherty et al., 2005).

Delahunt et al. (2007) and Feger et al. (2014) examined kinematics, kinetics, and/or muscle activities during lateral hop tests in individuals with and without CAI. Delahunt et al. (2007) revealed CAI individuals demonstrate a more inverted ankle position during lateral hopping during the period from 45-msec pre to 95-msec post initial contact (IC) with the ground compared to healthy controls (Delahunt et al., 2007). The current CAI literature suggests the more inverted foot positioning increases the momentum arm for the ankle joint complex, exposing the ankle to excessive supination, especially in weight-bearing (Konradsen & Voigt, 2002). In that case, lateral hopping tasks that stress the lateral ankle joint complex may further increase inversion moment by driving the ankle into greater supination. Additionally, Delahunt et al. (2007) displayed pre-activation of muscles (rectus femoris, TA, soleus) between the period 200-msec before and post IC in CAI compared to healthy controls. Soleus prevents excessive anterior and lateral COP displacement, whereas the tibialis anterior prevents excessive posterior COP displacement (Kim et al., 2003). Therefore, pre-activation of those muscles (rectus femoris, TA, soleus) from before to post IC is a result of a change in pre-programmed feedforward motor control, providing dynamic joint stability for the more inverted ankle during lateral hopping in CAI. In contrast, Feger et al. (Feger et al., 2014), however, denoted a significant decrease in the activity of muscles that act on the ankle (TA, PL, lateral gastrocnemius), knee (rectus femoris, biceps femoris), and hip (gluteus medius) during lateral hopping in CAI individuals compared to healthy controls. Feger et al. (2014) did not investigate kinetics and kinematics to support their results. However, the great speculation of those contradicting findings is due to the absence of ankle instability, not resulting in excessive supination, and/or the complexity of the lateral hopping task is too low to require supraspinal mechanisms in motor control. As previous research presents the existence of AMI in TA, PL, and soleus with CAI, CAI individuals may require greater corticospinal excitability to achieve the same motor outcome as healthy individuals (McVay et al., 2005; Pietrosimone & Gribble, 2012; McLeod et al., 2015).

Time-to-Stabilization and Jump Landings

Some common mechanisms of an ankle sprain are landing and jumping, thus dynamic postural stability tests such as the single-leg jump landing and/or drop jump landing may challenge CAI. Moreover, landing on a force plate in the injured-limb may provide an insight into altered movement strategies associated with CAI during the jump/drop jump landing tasks. The Time-to-stabilization (TTS) which presents moderate to low reliability has been suggested to detect postural control deficits associated with CAI in laboratory settings (Ross et al., 2005). The TTS examines the time an individual takes to stabilize their ground reaction force (GRF) following the jump-landing task

(Ross & Guskiewicz, 2004). In the current CAI literature, significantly longer TTS in both AP (Brown et al., 2004; Ross & Guskiewicz, 2004; Wikstrom et al., 2005; Gribble & Robinson, 2009; Gribble & Robinson, 2010) and ML (Ross & Guskiewicz, 2004) directions have been reported in CAI individuals compared to healthy controls. Those results indicate impairments in dynamic stability and the probable presence of inflexible adaptation to perturbations with CAI. Additionally, longer TTS reflects that an individual took more time organizing postural stability at the jump landing, implying a rigid sensorimotor system that is less flexible and adaptable to the landing impact (perturbation) (Brown & Mynark, 2007).

Delahunt et al., (2006) investigated kinematics, kinetics, and muscle activities during the single-limb drop jump landing task and found CAI individuals have more inverted ankle kinematics accompanied by significantly decreased PL activity during the aerial phase right before landing compared to healthy controls. Those findings may suggest decreased PL activity has failed to maintain the ankle joint in a neutral form (Delahunt et al., 2006). While it is too late for the sensorimotor system to modulate postural correction once the foot contacts the ground, the foot kinematics in the aerial phase may predetermine the loading response and subsequent events following the landing. Although excessive inversion did not persist at the landing in those individuals with CAI, who exhibited more inverted ankle kinematics during the aerial phase, significantly increased vertical GRF with less ankle dorsiflexion was demonstrated following landing compare to healthy controls (Delahunt et al., 2006). A significant increase in vertical GRF with CAI is also supported by Caulfield and Garrett (2004) who assessed changes in GRF during the jump landing task in individuals with and without CAI. Those changes in ankle kinematics before and post landing may fail to absorb increased vertical GRF at the ankle during landing.

Altered kinematics at proximal joints (knee, hip) have also been reported during the jump landing task in CAI individuals compared to healthy control. Caulfield and Garrett (2002) reported increased knee flexion during the period from 20-msec before and 60-msec post landing in the presence of CAI. Increased knee flexion lowers the center-of-mass (COM) closer to the ground and has suggested reducing the vertical GRF impact during landing. Thus, the increase in knee flexion may be attributed to centrally mediated alteration in neuromuscular control to compensate for diminished dynamic stability observed at the ankle. However, in contrast, Gribble and Robinson (2009, 2010) noted a bilateral decrease in knee flexion before (2010) and at (2009) landing in individuals with unilateral CAI. Researchers also revealed TTS deficits in the injured-limb of CAI compared to the matched limb of healthy controls (Gribble & Robinson, 2009, 2010). With no other group differences found in the ankle and hip kinematics, researchers have concluded the decrease in knee flexion noted on both limbs in individuals with unilateral CAI is also a centrally mediated alteration in neuromuscular control contributing to TTS deficits (Gribble & Robinson, 2009, 2010).

Those contradicting findings of Caulfield and Garrett (2002) may be the results of a difference in task complexity that is integrating visual feedback with different flight patterns between the studies (Gribble & Robinson, 2009, 2010). Specifically, Gribble and Robinson (2009, 2010) instructed participants to focus on a vertical object while completing the jump landing task, whereas Caulfield and Garrett (2002) enabled participants to focus on the landing platform throughout the jump landing task. Visual feedback aids other peripheral sensory feedback (somatosensory, vestibular) obtaining exteroceptive information of the environment to fine-tune motor control acuity (Turvey, 1990, 2007; Greeno, 1994). Thus, being unable to focus on landing platforms throughout the jump landing task may lead CAI individuals to freeze DOF of the knee that mediates ankle and hip joints to provide dynamic stability. Although decreased knee flexion may cause a longer time to dissipate the vertical GRF when compared to increased knee flexion. Considering the study results of TTS deficits presented in CAI (Gribble & Robinson, 2009, 2010), the potential protective mechanisms of reducing knee flexion may further increase susceptibility to recurrent ankle sprains with CAI.

Star Excursion Balance Test

The magnitude of errors in SEBT has been utilized to assess functional performance in clinical practice. The SEBT also has been suggested as a highly reliable measurement for examining dynamic postural control associated with CAI in laboratory settings (Plisky et al., 2006; Gribble et al., 2013; Gribble, 2016). The original SEBT consists of eight reach directions: anterior (ANT), anteromedial (AM), anterolateral (AL), medial (MED), lateral (LAT), posterior (POS), posteromedial (PM), and posterolateral (PL) to challenge an individual for postural control (stability), range-of-motion, and proprioception (Hertel et al., 2006). The SEBT is performed to maintain a stable base in a single-limb stance, usually in the injured-limb, with and without eyes-closed, resting both hands on the hip. The functional performance is evaluated while an individual lightly taps at the farthest reach in each reach direction and returns to the starting position with a contralateral stance limb. Functional performance errors are counted when an individual discards the trial, removes hands from the hip, lifts the standing limb, or stands on a contralateral limb while the reach is performed (Ko et al., 2018). Additionally, significantly shorter SEBT in all eight reach distances are associated with postural control deficits in CAI individuals compared to healthy controls (Olmsted et al., 2001; Hertel et al., 2006). Since length and height significantly correlate to the SEBT reach distance, it is typically normalized to an individual's limb length for comparison in individuals with and without CAI (Gribble & Hertel, 2003).

The SEBT protocols have been simplified to ANT, PL, and PM reach directions and are identified to be sensitive in detecting an impaired sensorimotor system related to CAI (Hertel, 2008). Indeed, CAI individuals have been reported to exhibit significantly shorter reach distance in all ANT, PM, and PL reach directions compared to healthy controls (Hoch et al., 2012; Plante & Wikstrom, 2013; McCann et al., 2017). Among those three directions, Hertel et al. (2006) indicated the PM reach direction is the most representative of functional performance in all eight SEBT reach directions. For example, significant functional performance deficits in PM reach direction have been found to discriminate between CAI individuals and healthy controls (Plante & Wikstrom, 2013; Ko et al., 2019). Those findings are also supported by a study presenting significant Area under the curve (AUC) (indicator of classifying having or not having CAI) values in the PM reach direction, and not in the ANT or MED reach direction (Linens et al., 2014).

The primary measure of SEBT performance is the reach of distance. However, ankle, knee, and hip kinematics in sagittal and frontal planes have been suggested to influence the SEBT performance in ANT, PL, and PM reach distances (Gribble et al., 2012). For example, researchers denote ankle dorsiflexion range-of-motion measured in both open-and-closed chains (i.e., lying prone, weight-bearing lunge test) is strongly correlated with the ANT reach distance in healthy individuals that dorsiflexion range-of-motion explains 38% of SEBT performance in the ANT reach (Hoch et al., 2011, 2012; Terada et al., 2014). Moreover, healthy individuals primarily utilized knee and hip flexion to achieve the greater ANT reach distance, explaining 78% of SEBT performance in the ANT reach (Robinson & Gribble, 2008). In contrast, Hoch et al. (2016) found CAI individuals who achieved maximum reach distance in the ANT direction utilized a combination of the frontal-plane ankle, hip, and trunk kinematics. Specifically, the combination of ankle eversion, hip adduction, and lateral trunk flexion in the stance limb explains 81% of the SEBT performance in the ANT reach (Hoch et al., 2016). In a few studies of group comparisons, CAI individuals exhibited less knee flexion and ankle dorsiflexion at the point of maximum reach in the PL direction, and less knee flexion at the point of maximum reach in the PM direction in both stance and contralateral stance limbs compared to healthy controls (Doherty et al., 2016). Similarly, CAI displayed both limited weight-bearing dorsiflexion (WBDF) range-of-motion and deficits in the ANT and PL reach distances compared to healthy controls (Kosik et al., 2019).

Researchers have also investigated the contribution of the strength of the SEBT performance in the ATN, PL, and PM reach distances. For instance, healthy individuals primarily recruited vastus medialis rather than gluteus maximus or medius during performing the SEBT in ANT, MED, and PM reach directions (Norris & Trudelle-Jackson, 2011). In contrast, CAI displayed isometric hip abduction and external rotation strength deficits and SEBT performance deficits in the ANT, PM, PL reach distance (McCann et al., 2017). Specifically, isometric hip abduction and external rotation strength deficits with CAI account for 25% of SEBT performance deficits in PM and PL reach distance (McCann et al., 2017). Additionally, CAI individuals demonstrated deficits in plantarflexion-dorsiflexion strength normalized to their body mass (kg) along with SEBT performance deficits in the PM reach distance and reduced WBDF range-of-motion relative to healthy controls (Plante & Wikstrom, 2013). Those findings may suggest CAI individuals utilize more muscles acting on the ankle and hip while healthy individuals recruit the muscles acting on the knee to coordinate dynamic postural control. In fact, Gabriner et al. (2015) revealed ankle eversion strength and the TTB minima in ML direction account for 28% of the SEBT performance in the PM reach distance and 14% of the SEBT performance in the PL reach distance, while WBDF range-of-motion and plantar cutaneous sensation accounted for 16% of SEBT performance in ANT reach distance. In other words, the SEBT ANT performance is more affected by somatosensory and pathomechanical deficits, whereas the PM and PL reach performance is more influenced by strength. Those findings may explain why improving WBDF range-of-motion with two-weeks of joint mobilization (posterior talar glide) intervention has resulted in the greater SEBT ANT reach distance in CAI individuals (Hoch et al., 2012).

Gait

Gait (walking, running), especially walking, is a necessary part of daily life. An alteration in gait mechanics may affect force absorption and dissipation during the IC and loading phase of gait, resulting in a risk of recurrent ankle sprains and potential long-term consequences. In the last two decades gait kinematics and kinetics, muscle activities, and gait rehabilitation in barefoot or shod conditions have been evaluated in individuals with and without CAI. However, there is a gap in the CAI literature on how CAI individuals utilize sensory systems to achieve dynamic postural control during gait.

Gait Kinematics

Walking

CAI individuals have been reported to exhibit altered sagittal and frontal plane gait kinematics at the ankle during walking. Monaghan et al., (2006) and Delahunt et al. (2006) both demonstrated increased ankle inversion during the terminal swing and early stance phase of gait. Specifically, CAI exhibited a more inverted foot position between 100-msec before and 200-msec post heel strike (HS) and remained inverted throughout the gait compared to healthy controls (Monaghan et al., 2006). Some researchers have assessed walking gait kinematics utilizing a multi-segment foot model and identified the presence of more forefoot (Wright et al., 2013; Northeast et al., 2018) and rearfoot (Drewes et al., 2009; Fraser et al., 2019) inversion in CAI individuals compared to healthy controls. For example, significantly increased forefoot inversion was found at 4 to 16% of the stance phase of gait in the injured-limb compared to the uninjured-limb in individuals with unilateral CAI (Northeast et al., 2018). In between-group comparisons, Wright et al. (2013) denoted significantly greater forefoot inversion at the IC in CAI individuals relative to healthy controls. Conversely, CAI individuals also exhibited increased rearfoot inversion throughout the gait (Drewes et al., 2009) and at 34 to 91% of the stance phase (Fraser et al., 2019) compared to healthy controls.

Although researchers did not investigate mechanisms of those kinematics changes in the ankle joint complex, they may be due to the impaired proprioceptive perception failing to detect the joint position in space and to correct excessive movements (Konradsen et al., 1998; Konradsen, 2002; Konradsen & Voigt, 2002). Additionally, when the ankle continues to move beyond the inversion threshold, the lateral border of the ankle would collide with the ground (Konradsen & Voigt, 2002). For instance, CAI individuals have exhibited a decrease in vertical foot-floor clearance during the terminal swing phase of gait along with excessive inversion of the foot compared to healthy controls (Delahunt et al., 2006). Once the foot is loaded and/or collides with the ground, it is not fast enough for protective reflexes to correct the foot positioning and/or to provide dynamic stability at the ankle joint complex. Indeed peroneals (peroneal muscles), which primarily provide dynamic lateral ankle stability, take approximately 126-msec to react to an unexpected perturbation (Konradsen et al., 1997). Furthermore, the foot positioning during the swing phase of gait seems to predetermine the subsequent loading response at the initial ground contact. Specifically, excessive forefoot and rearfoot inversion at the HS increases the momentum arm for the ankle joint complex, driving the ankle into greater supination for a potential recurrent ankle sprain in CAI individuals (Konradsen, 2002).

A change in sagittal-plane ankle kinematics in CAI individuals was only reported with shod walking on a treadmill (Chinn et al., 2013). Ankle dorsiflexion was approximately 3-degrees less between 42 to 51% (i.e., mid-stance to late terminal stance) of gait in CAI compared to healthy controls (Chinn et al., 2013). These changes in sagittal-plane ankle kinematics may be due to arthrokinematics restrictions at the talocrural joint previously noted with CAI (Hoch et al., 2011, 2012). Sagittal-plane ankle dorsiflexion range-of-motion maintains the ankle joint in a more stable, close-packed position to provide greater protection of the lateral ankle joint complex during a dynamic task such as gait (Drewes et al., 2009; Hoch et al., 2012). Indeed, it has been found individuals with less ankle dorsiflexion range-of-motion have nearly five times greater risk of sustaining ankle sprains than those individuals within healthy ankle dorsiflexion range-of-motion (de Noronha et al., 2006). Furthermore, previous research consistently has reported CAI individuals display laterally deviated plantar pressure during the loading and stance phase of gait compared to healthy controls (Nyska et al., 2003; Nawata et al., 2005; Morrison et al., 2010). Therefore, the presence of decreased ankle dorsiflexion range-of-motion along with laterally deviated plantar pressure may leave the ankle vulnerable to injurious load during the stance phase of gait.

Running

There are only a few studies investigating running gait kinematics with CAI. Drewes et al. (2009, 2009) found more inversion throughout gait and less ankle dorsiflexion range-of-motion during the period of maximal dorsiflexion during jogging in CAI individuals relative to healthy controls. Those findings are supported by Chinn et al.

(2013) who revealed increased inversion from 4 to 96% of the gait and decreased ankle dorsiflexion range-of-motion from 54 to 68% (i.e., mid-swing phase) of the gait compared to healthy controls. In contrast, Lin et al. (2011) reported greater inversion during the terminal-swing phase of the running gait in CAI than healthy controls. In the evaluation of gait kinematics utilizing a multi-segment foot model, greater forefoot inversion was present between the mid-stance and late-stance phase of running in CAI individuals compared to healthy controls (De Ridder et al., 2013). Previous research has found joint position sense deficits are most substantial when the index angles are combined with plantarflexion and inversion (Yokoyama et al., 2008). Therefore, the excessive increase in inversion and decrease in dorsiflexion range-of-motion (more plantar flexion), especially during the swing phase, may result from proprioceptive deficits encouraged with increased task constraints with jogging/running (Yokoyama et al., 2008; Chinn et al., 2013). In addition, it has been suggested CAI individuals decrease down-modulation of H-reflex in PL accounted by supraspinal mechanisms while transitioning from simple to more complex tasks (Taube et al., 2008; Kim et al., 2016). Since down-modulation of the H-reflex is associated with greater postural stability in healthy individuals, altered sagittal and frontal plane kinematics of the ankle joint complex during jogging/running may negatively influence dynamic postural control (Koceja & Mynark, 2000). Collectively, potential change in the central organization may also be a cause of altered gait kinematics with CAI.

Gait Kinetics

Walking

Several researchers have evaluated the COP trajectories during shod walking on a treadmill and barefoot overground walking. CAI individuals exhibited laterally deviated COP trajectories at the initial-HS and the stance phase (early mid-stance) of the gait compared to healthy controls (Nyska et al., 2003; Hopkins et al., 2012). Consistently, CAI individuals displayed 2.9-mm more laterally deviated COP trajectories during the first 10% of the stance phase, and 7.5-mm more laterally deviated COP trajectories from 50 to 60% of the stance phase relative to healthy controls (Koldenhoven et al., 2016). Those findings are consistently supported in barefoot overground walking. Nawata et al. (2005) reported that CAI individuals present more laterally deviated COP trajectories with the foot in increased supination during the mid-stance phase of gait. In contrast, previous research has revealed healthy individuals have laterally deviated COP trajectories at IC yet pronated as a loading response during the mid-stance phase of the gait (Cornwall & McPoil, 1999; Willems et al., 2005; Rice et al., 2013). The foot typically supinates when COP trajectories laterally deviate to the subtalar joint axis and pronate when COP trajectories medially deviate to the subtalar joint axis. Therefore, laterally deviated COP trajectories associated with CAI during the mid-stance phase of gait may decrease dynamic stability in CAI individuals.

Running

Only a few studies have tested the distribution of pressure and force during running with CAI participants. Schmidt et al. (2011) examined gait kinetics utilizing a multi-segment foot model during shod running on a treadmill. Researchers found peak pressure and force distribution in the lateral forefoot and midfoot in CAI individuals compared to healthy controls (Schmidt et al., 2011). A significantly increased peak pressure and force in the lateral column (forefoot, midfoot) of the foot correspond with the laterally deviated COP trajectories exhibited during walking in CAI (Nyska et al., 2003; Hopkin et al., 2012; Koldenhoven et al., 2016). Those findings are consistently supported in barefoot overground running where CAI individuals displayed increased pressure distribution in lateral rearfoot at the HS and large lateral deviation of COP trajectories during the subsequent loading phase of gait (Morrison et al., 2010). CAI individuals also demonstrated a slower loading response in lateral and medial rearfoot and medial midfoot during the early stance phase of gait (Schmidt et al., 2011). Previous prospective research has summarized that healthy individuals who present a greater loading response in medial plantar pressure and lesser loading response in lateral plantar pressure at the IC have a greater risk of sustaining a LAS (Willems et al., 2005). Therefore, slower loading response at the early stance phase of the gait in CAI may be a compensatory mechanism CAI individuals employ to coordinate enough time to provide stability throughout the stance phase, especially at the mid-stance of gait.

Muscle Activities

Walking

<u>Peroneus Longus</u>

Researchers have investigated motor control strategies by analyzing peroneal muscle activities during gait. Peroneal muscles (e.g., PL, PB) prevent excessive inversion (hypersupination) at the ankle joint complex. Konradsen (2002) also indicates PL would activate when greater ankle inversion exists during the mid-swing phase of gait. Additionally, PL is typically activated to support initiating pronation during the mid-stance to the terminal-stance phase of gait and stabilizing the medial column of the foot, specifically the first ray during the propulsive phase (terminal-stance) in healthy individuals (Johnson & Christensen, 1999; Sutherland, 2001; Santilli et al., 2005; Koldenhoven et al., 2016).

Several studies evaluated PL activity during shod walking on a treadmill. Hopkin et al. (2012) revealed significantly increased PL activation at the initial HS and toe-off (TO) with a trend toward decreased activation at the early mid-stance phase of gait in CAI individuals compared to healthy controls. Moreover, significantly greater PL activation was found during 100-msec before the IC and for longer duration throughout the stride (ipsilateral heel-to-heel contact) in CAI compared to healthy controls (Feger et al., 2015; Koldenhoven et al., 2016). Kautzky et al. (2015), however, failed to support those findings with no group differences in PL activity. The heightened PL activation in CAI was also displayed in the previous research, which has demonstrated a more inverted foot position along with laterally deviated COP trajectories (Hopkins et al., 2012; Feger et al., 2015; Koldenhoven et al., 2016). Thus, increased PL activation may have developed an altered feedforward motor control in preparation for the IC and to provide dynamic stability to the lateral ankle joint complex during the stance phase of gait.

Some researchers examined PL activity during barefoot walking on a treadmill. Similar to shod walking on a treadmill, a significant increase in PL activation was observed 40-msec after the HS in CAI individuals compared to healthy controls (Delahunt et al., 2006). Conversely, Louwerens et al. (1995) reported no group differences in PL activation during the stance phase of gait. In a side-to-side comparison, Santilli et al. (2011) found a lack of PL activation in the injured-limb during the stance phase of gait compared to the uninjured-limb in individuals with unilateral CAI. Those findings may suggest CAI individuals have implemented different motor control strategies during barefoot walking compared to shod walking.

Tibialis Anterior

Koholdenhoven et al. (2016) reported significantly increased TA activity 100-msec before the IC during shod walking on a treadmill in CAI compared to healthy controls. Similarly, a few studies have found greater TA activation during the stance phase in CAI relative to healthy controls (Louwerens et al., 1995; Hopkins et al., 2012). However, most researchers did not find group differences in TA activity during shod or barefoot walking on a treadmill (Delahunt et al., 2006; Kautzky et al., 2015, Feger et al., 2016). Decreased ankle dorsiflexion range-of-motion was previously displayed between the mid-stance and the late terminal stance of gait in CAI individuals (Chinn et al., 2013). With this in mind, increased TA activity could be interpreted as a strategy CAI individuals attempt to maintain the ankle joint complex in a more stable, closed-pack position during the stance phase of gait.

Medial and Lateral Gastrocnemius and Soleus

There are mixed findings in gastrocnemius (lateral, medial) and soleus activities during shod and barefoot walking on a treadmill in individuals with and without CAI. No group differences were observed for lateral gastrocnemius (Feger & Hertel, 2016) or soleus (Delahunt et al., 2006) activities throughout gait. However, Koldenhoven et al. (2016) reported significantly increased medial gastrocnemius activity during 100-msec before the IC in CAI individuals compared to healthy controls (Koldenhoven et al., 2016). Additionally, previous research has denoted a decrease in ankle dorsiflexion during the mid-stance and the late terminal stance phase of gait. (Chin et al., 2013). Therefore, preactivation of medial gastrocnemius immediately before the HS may intend for CAI individuals to control increased plantarflexion eccentrically, providing dynamic stability at the ankle.

Rectus Femoris

Healthy individuals activated rectus femoris (RF) as a loading response while pronating the foot at the IC and in the early stance phase of gait to absorb ground impact (Schmidt et al., 2011; Lacquaniti et al., 2012). In contrast, increased RF activity was noted during 108-msec (Feger et al., 2015) or 200-msec (Delahunt et al., 2006) before the HS during shod and barefoot walking on a treadmill in CAI individuals. Only Kautzky et al. (2015) have reported no group differences in RF activity during shod walking. Altered preactivation of RF before the IC may confirm the presence of central adaptation at proximal joints. Although those studies did not observe a change in hip and knee kinematics with CAI, previous research has denoted altered knee kinematics during the jump landing task in CAI individuals compared to healthy controls. Specifically, Caulfield and Garrett (2002) identified increased knee flexion during 20-msec before and 60-msec after the landing in CAI individuals compared to healthy controls. Additionally, individuals with CAI exhibited more laterally deviated COP trajectories while the foot was in increased supination during the mid-stance phase (Nawata et al., 2005). Altered RF activity before the IC may confirm the presence of central adaptation at proximal joints (i.e., knee) with CAI. Specifically, CAI individuals preactivate RF to control the knee movement eccentrically, leading up to the IC to dissipate loading impact more at the knee.

Gluteus Medius and Maximus

There have been inconsistent findings in gluteus medius activities in individuals with and without CAI. Koldenhoven et al. (2016) revealed increased gluteus medius activities during 100-msec before the HS in CAI individuals. In contrast, DeJong et al. (2019) displayed a significant reduction in gluteus medius activities of bilateral limbs (injured, uninjured) during the first 40% (IC to terminal stance) of gait in individuals with unilateral CAI and noted no group differences in gluteus maximus activity throughout gait. Gluteus medius is most active before the HS and during the mid-stance phase of gait in healthy individuals and plays a critical role in correcting the lower extremity alignment during gait (Nguyen et al., 2011). With this in mind, increased gluteus medius activities before the HS could be a mechanism CAI individuals employ to correct excessive inversion, which has been found during the same period of 100-msec before the HS in previous research (Monaghan et al., 2006; Koldenhoven et al., 2016). Furthermore, the reduction in gluteus medius activities between the IC and the terminal stance phase of gait is in line with previous findings of isometric strength deficits in hip AB and external rotation, which has been reported in CAI compared to healthy controls (McCann et al., 2017). CAI individuals also commonly display excessive inversion with laterally deviated COP trajectories during the stance phase of gait (Nawata et al., 2005). Therefore, a lack of dynamic support at the hip to assist excessive frontal-plane movements (e.g., inversion) with laterally deviated COP at the ankle joint complex may contribute to recurrent ankle sprains with CAI.

Running

There is only a single study examining the activity of PL, TA, and lateral gastrocnemius during overground running in CAI individuals compared to healthy controls, and no significant group differences were found (Lin et al., 2011). Therefore, more research is needed to conclude the exact consequences of those results.

Gait Variability and Coordination

Movement variability has been suggested to quantify movement quality and the explanatory nature of neurobiological behaviors (Lipsitz, 2002; Stergiou & Decker, 2011). However, the interpretation of movement variability depends on the utilization of linear and nonlinear tools. Linear analysis (standard deviation, coefficient of variation) quantifies performance errors, whereas nonlinear analysis (entropy) quantifies the flexibility and adaptability of the sensorimotor system underlying movement (Lipsitz, 2002; Stergiou & Decker, 2011). According to the optimal movement variability theory in nonlinear analysis, the healthy sensorimotor system reflects the optimal state, rendering flexibility and adaptability of the sensorimotor system to organismic, task, and environmental constraints (Harbourne & Stergiou, 2009). In contrast, the suboptimal state at below or above the optimal state renders a lack of health (Harbourne & Stergiou, 2009). In other words, movement becomes more predictable and rigid (robotic) with a decrease in optimal movement variability, while movement becomes more noisy and random (frail) with an increase in optimal movement variability (Stergiou & Decker, 2011). For example, significantly decreased stride-to-stride (ipsilateral heel-to-heel) gait variability of frontal-plane ankle kinematics is evident during treadmill walking in CAI individuals compared to healthy control (Terada et al., 2015). This is the only study that examined gait variability in nonlinear analysis (i.e., SampEN), confirming the optimal movement variability theory that CAI individuals have less flexible and adaptable sensorimotor systems relative to healthy controls (Terada et al., 2015).

Several researchers examined movement variability with linear analysis in CAI individuals. Hamacher et al. (2016) examined stride-to-stride ankle kinematics during treadmill running, utilizing standard deviation (SD) as a variability measure in individuals with unilateral CAI compared to healthy controls. CAI exhibited significantly increased variability in frontal-plane ankle kinematics in bilateral limbs (injured, uninjured) during the stance and the late swing phase of gait compared to healthy controls (Hamacher et al., 2016). These findings are supported by Wanner et al. (2019), who also revealed a significant increase in stride-to-stride variability of frontal-plane ankle kinematics during the stance phase of gait, especially during high running velocity compared to moderate running velocity. Additionally, the study denoted a tendency for increased variability in frontal-plane ankle kinematics during the terminal swing phase of gait, partially in agreement with Hamacher et al. (2016). The presence of increased variability during the stance and the late swing phase of gait in CAI individuals with the linear analysis may indicate limited stability of the ankle. Furthermore, CAI individuals displayed laterally deviated COP trajectories (Koldenhoven et al., 2016) and increased stride-to-stride variability in COP distribution (Koldenhoven et al., 2018) during the same period of the first 10% of the stance phase of walking. Those findings may also have contributed to an increase in stride-to-stride variability of frontal-plane ankle kinematics during treadmill running in CAI individuals.

A few studies investigated the movement variability of multi-foot/limb segment coupling during walking or running in CAI individuals. Cornwell et al. (2019) studied tibial (shank) rotation and calcaneal eversion/inversion coupling in individuals with and without moderate or severe CAI during barefoot overground walking. Those individuals with severe CAI displayed significantly decreased variability between shank and rearfoot eversion coupling during the mid-stance phase of gait compared to those individuals with moderate CAI and healthy controls. Consistently, significantly decreased stride-to-stride shank-rearfoot coupling variability was demonstrated during the late swing phase of barefoot walking and running in CAI individuals compared to healthy controls (Drewes et al., 2009). In contrast, Herb et al. (2014), who evaluated shank-rearfoot coupling in individuals with and without CAI during treadmill walking and jogging, showed significantly reduced stride-to-stride shank-rearfoot coupling variability throughout the stance and the swing phase of walking except at the IC in CAI individuals compared to healthy controls. Healthy individuals typically begin to externally rotate the shank as the foot begins to unload during the mid-stance phase of gait. However, those CAI individuals with decreased shank-rearfoot coupling variability moved rearfoot ahead of the shank during the swing and the stance phase of gait (Drewes et al., 2009; Herb et al., 2014; Cornwall et al., 2019). Additionally, the most recent evidence revealed a significant decrease in hip-ankle coupling variability in frontal-plane during the second half of the mid-stance phase (0.45 ± 0.07) of treadmill walking compared to healthy controls (0.54 ± 0.06) (Yen et al., 2016). Therefore, reduction in both shank-rearfoot and hip-ankle coupling variability found in those studies could be a manifestation of an attempt to freeze (restrict) DOF, acting to minimize increased stride-to-stride frontal-plane ankle kinematics associated with CAI during gait (Hamacher et al., 2016; Wanner et al., 2019).

Assessment of Dynamic Postural Control in Healthy Individuals

Healthy individuals can flexibly adapt and recover from an unexpected perturbation, maintaining postural control during locomotion (Rhea & Rietdyk, 2011). This is because healthy individuals adapt to unexpected changes in the environment by implementing compensatory strategies based on constant proprioceptive feedback (Simoneau et al., 1995; Dietz, 2002; Patla et al., 2004). Healthy individuals also apply visual feedback to maintain postural control during gait (Simoneau et al., 1995; Patla, 1997). Visual feedback provides information about the near and far environment to navigate the travel path, coordinating limb trajectories and/or the foot placement specifically over obstacles for postural control during gait (Patla, 1997, 1998; Patla et al., 2002; Patla & Vickers, 2003). Indeed, researchers found removing, blocking, and disturbing visual feedback with goggles and glasses (i.e., liquid crystal goggle, prism glasses) during obstacle crossing overground and on a treadmill significantly increased failure rate and toe clearance variability in healthy individuals (Mohagheghi et al., 2004; Patla & Greig, 2006; Rhea & Rietdyk, 2007; Alexander et al., 2011; Rhea & Rietdyk, 2011). Additionally, inadequate visual feedback of the lower visual field has been found to affect obstacle clearance in healthy individuals (Marigold et al., 2007; Marigold, 2008; Marigold & Patla, 2008). The most recent systematic review concluded CAI individuals heavily rely on visual feedback while maintaining posture in a single-limb stance (Song et al., 2016). However, whether altered gait biomechanics in CAI individuals is the result of postural disturbance along with preexisting somatosensory deficits associated with CAI or failure to apply visual feedback during gait is unknown.

The Locomotor Sensory Organization Test

Chien et al. (2014) have introduced a novel Locomotor Sensory Organization test (LSOT) deriving from the SOT of the NeuroCom Smart Balance Master to measure how each sensory feedback system (somatosensory, vision, vestibular) contributes to postural control during gait. The SOT consists of six conditions to manipulate sensory information by modifying the support surface and visual surroundings with and without eyes-closed while standing in a double-limb stance. Similarly, the LSOT also consists of six conditions to manipulate sensory information by modifying the speed of individually determined preferred walking speed and the virtual reality optic flow projected on a surrounding screen with and without limited vision during gait (Figure 2.10). Specifically, each LSOT condition is matched to the SOT conditions as follows: in condition 1, no sensory information is manipulated; conditions 2 and 5 limit visual information with goggles; condition 3 manipulates visual information with the VR optic flow; conditions 4 and 5 manipulate somatosensory information with treadmill speed; condition 6 manipulates both somatosensory and visual information with the speed of the treadmill and the VR optic flow (Chien et al., 2014). Moreover, the speed of the treadmill and the VR optic flow pseudo-randomly varies between 80 and 120% of PWS during 2-minute walking and is assigned within a 1 to 10 seconds time interval to avoid the learning effect (Chien et al., 2014). The LSOT has been successful in demonstrating certain degrees of similarity in sensory feedback mechanisms involved in postural control, measured with COP net trajectories, during walking relative to sensory feedback mechanisms involved in static postural control assessed with the SOT in healthy individuals (Chien et al., 2014, 2016). The primary differences between the SOT and LSOT findings are a result of variation in the task complexity (static stance vs. walking). For example, somatosensory feedback is preferred for postural control in a static stance, while visual feedback is suited for dynamic postural control during locomotion in healthy individuals (Patla, 1997, 1998; Peterka, 2002). Furthermore, the LSOT is effective in displaying the change in sensory feedback mechanisms involved in postural control while mastoid vibration is induced during walking in healthy individuals (Chien et al., 2016, 2017). Collectively, the LSOT can be utilized to elicit how sensory systems are utilized to achieve dynamic postural control during gait in individuals with and without CAI.

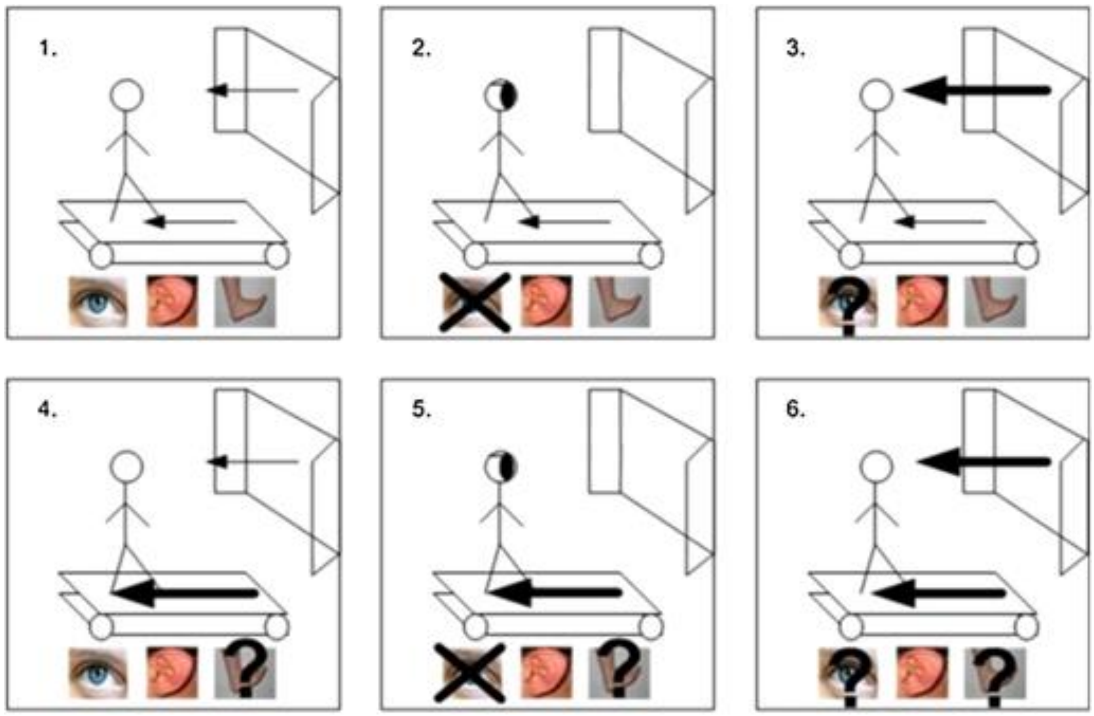

Figure 1.10. Six Conditions of the Locomotor Sensory Organization Test (Chien et al., 2014).

The Margin of Stability

In order for healthy individuals to maintain postural stability, the COM must remain within the boundaries (borders) of the base-of-support (BOS), which is typically the foot in direct contact with the support surface. Unlike maintaining static postural control, the COM and BOS are constantly in motion during locomotion, affecting overall postural control (Pai & Patton, 1997; Hof et al., 2005). The common measure to quantify dynamic stability during locomotion is the margin of stability (MOS). The MOS measures the distance between the extrapolated COM and the boundaries of BOS at any instant in time (Hof et al., 2005, 2007). Hof et al. (2005) introduced the MOS deriving from the inverted pendulum model commonly utilized to describe static postural control with the term "extrapolated COM," or the XcoM. The XcoM refers to the COM position relative to the boundaries of the BOS if the COM would continue on its trajectory at its instantaneous velocity (Hof et al., 2007; Hof, 2008). The MOS can be applied to both AP and ML directions during gait. If the XcoM falls within the boundaries of the BOS at the HS, the MOS is positive and implies an individual is stable. Whereas, if the XcoM falls outside the boundaries of the BOS at the HS, the MOS becomes negative and implies an individual is unstable. In the case of postural instability during gait, healthy individuals rapidly step forward, likely employing shorter step lengths, to change the BOS to maintain postural stability (Hof, 2008). The MOS has been extensively utilized in various gait studies in healthy individuals to quantify dynamic postural stability in AP (Hof et al., 2005; Yang et al., 2009; Lugade et al., 2011; Hak et al., 2012, 2013; McAndrew Young et al., 2012; Caderby et al., 2014; Yang & Pai, 2014; Koyama et al., 2015; Nakano et al., 2015) and ML (Hof et al., 2005, 2010; Sefton et al., 2007; Rosenblatt & Grabiner, 2010; Curtze et al., 2011; Lugade et al., 2011; McAndrew Young et al., 2012; Rosenblatt et al., 2012; Hak et al., 2013; Caderby et al., 2014; Peebles et al., 2016) directions during normal (Hof et al., 2005; Hof et al., 2007; Rosenblatt & Grabiner, 2010; Lugade et al., 2011; Rosenblatt et al., 2012; Hak et al., 2013; Caderby et al., 2014) and perturbed (Hof et al., 2010; Hak et al., 2012; McAndrew Young et al., 2012; Yang & Pai, 2014; Koyama et al., 2015; Nakano et al., 2015) walking. Collectively, examining the MOS may discriminate the degree of flexibility and adaptability of the sensorimotor system during gait, especially with increased task and environmental constraints, in individuals with and without CAI.

Gait Rehabilitation

Maladapted ankle kinematics and alterations in proximal joints have been reported in CAI individuals compared to healthy controls. The alterations in proximal joints may indicate a change in the central organization. Several researchers investigated the effect of neuromuscular training to improve impaired gait parameters reported in gait studies with CAI. McKeon et al. (2009) examined the effect of 4-weeks of novel balance training on ankle kinematics during walking and jogging on a treadmill in CAI individuals. The balance training comprised seven levels of difficulties and challenged stability at landing from a hop in a single-limb stance (McKeon et al., 2009). Although the balance training significantly improved the variability of shank-rearfoot coupling during walking, no training effects on the stability of shank-rearfoot coupling were displayed during jogging (McKeon et al., 2009). These differences may be due to decreased ground contact time during running compared to walking (McKeon et al., 2009). Additionally, completion of balance training did not change shank or rearfoot kinematics during walking or running (McKeon et al., 2009). This lack of balance intervention effects on ankle kinematics is consistent with previous research. In contrast to McKeon et al. (2009), Coughlan and Caulfield (2007) utilized 4-weeks of novel neuromuscular training that contained five levels of difficulties combining the repetition of lunging, hopping, and single-leg exercises, aiming to promote dynamic joint stability. McKeon et al. (2009) have suggested the variability of shank-rearfoot coupling attribute to sensorimotor systems while ankle kinematics associate with mechanical properties. In this case, intervention programs such as joint mobilization techniques, which alter mechanical properties, may provide a better effect in correcting ankle kinematics.

Donovan et al. (2017) evaluated the effect of 4-weeks of balance training utilizing a novel destabilization device on sagittal and frontal plane ankle kinematics, gait kinetics, and muscle activities in PL, PB, TA, and medial gastrocnemius during a treadmill walking in CAI individuals. The destabilization device possessing an articulator was designed to force an individual's foot into plantarflexion, inversion, and internal rotation in a controlled manner during treadmill walking (Donovan & Feger, 2017). Implementation of the destabilization device increased ankle dorsiflexion by approximately 6-degree between the mid-stance to the late stance phase of gait (Donovan et al., 2017). Researchers concluded that increased ankle dorsiflexion was resulting from feedback mechanisms responding to the destabilization device which forces the foot into plantarflexion (Donovan et al., 2017). Indeed, there was a slight reduction in medial gastrocnemius activity during the same period where an increase in dorsiflexion was displayed (Donovan et al., 2017). Significant reduction in PL activities was noted during the early stance and the mid-swing phase of gait, however, no change in frontal-plane ankle kinematics was observed (Donovan et al., 2017). Collectively, a stimulating feedforward response with a destabilization device was not great enough to instill a change in frontal-plane ankle kinematics and kinetics.

Donovan et al. (2016) and Torp et al. (2019) investigated the effect of external sensory feedback intervention strategies to correct increased plantar pressure over the lateral column (forefoot, mid-foot) of the foot and muscle activities in CAI individuals.

Donovan et al. (2016) utilized a custom wearable auditory feedback device that elicits a noise when plantar pressure exceeds a set threshold during walking on a treadmill. Walking not to make a noise on the feedback device significantly reduced peak pressure and pressure distribution time in the lateral column of the foot and increased PL and medial gastrocnemius activities (Donovan et al., 2016). Specifically, the laterally deviated pressure transferred to a more medial hallux region of the foot. Accordingly, researchers speculated PL and medial gastrocnemius have contributed to a medial shift in plantar pressure. In contrast, Torp et al. (2019) tested the effect of a novel wearable laser pointer, which provides visual feedback when the laser pointer deviates from the referenced laser cross-line projected on the wall during treadmill walking in CAI individuals. The implementation of a novel laser device reduced peak pressure and pressure distribution time in the lateral column of the foot to be transferred to a more medial hallux region of the foot. Those changes in lateral to the medial shift of the plantar pressure are in agreement with the effect of the auditory feedback device. Collectively, integrating real-time auditory and visual feedback to gait rehabilitation was successful in correcting the maladapted plantar pressure distribution associated with CAI.

The current evidence suggests the external focus of attention better facilitates the motor learning process relative to an internal focus of attention, however exact mechanisms to why external sensory feedback integrated gait interventions were successful in correcting altered plantar pressure with CAI is unknown (Wulf & Lewthwaite, 2016). According to the DST, motor learning and skill acquisition depends on interactions between organismic (individual characteristics), task, and environmental constraints. Thus, an individual's proprioceptive ability to detect and correct movement errors (i.e., knowledge of results) in the manner of the sensory-motor feedback loop is essential in the motor learning and skill acquisition process (Newell, 1991; Greeno, 1994). Visual feedback plays a significant role in confirming exteroceptive information of the environments relative to task goals in guidance of locomotion (Greeno, 1994; Patla, 1997). Indeed, the most recent conclusion is that CAI individuals heavily rely on visual feedback to compensate for somatosensory deficits presented at the ankle (Song et al., 2016). However, there is no evidence to support that CAI individuals continue to upregulate visual feedback during dynamic tasks such as gait. Therefore, it is worth investigating how sensory feedback contributes to gait biomechanics in CAI individuals relative to healthy controls.

Conclusion

According to the foundation of the DST, there is inherent perception and action (movement) interaction for an individual to coordinate context-dependent motor behaviors in the ever-changing environment. The presence of CAI, which is an organismic constraint, has been hypothesized to impair individual perception and neural control underlying movements, resulting in altered motor behaviors such as postural control/balance deficits (static task) and maladapted gait (dynamic task) associated with CAI. Adequate perception is significant for an individual to better relate to the ever-changing environment to accomplish a task goal. Specifically, when redundant sensory information is simultaneously perceived, exceeding what the CNS minimally requires to perform a task goal, flexible adaptation is vital to reweight multi-sensory information by identifying its relevance based on the context encountered. Nevertheless, impaired perception associated with CAI rigidly fixates on uni-sensory information (e.g., visual feedback), and how those CAI individuals reweight sensory feedback transitioning from simple to more complex tasks is unknown. Therefore, identifying which external sensory feedback is prominent during balance and gait will allow recommendations for rehabilitation. Specifically, it might be possible to manipulate balance and gait interventions to focus on therapy to restore impaired coordination between individual perception and movement coordination. The goal would be to focus intervention on motor behaviors that have been identified as being most compromised following changing tasks and environments. This study will expand knowledge related to the complex and multifaceted programming of individuals with ankle instability for balance and gait.

CHAPTER III

METHODS

Research Design

The current case-control study was performed in a research laboratory at the University of North Carolina at Greensboro (UNCG). To achieve Aims 1.1.1., 1.1.2., 1.1.3., 2.1.1., 2.1.2, 2.1.3., 4.1.1., 4.1.2., & 4.1.3., a mixed-model design with one between-group factor (CAI, healthy controls) and one within-group factor of the environment and task were utilized. To achieve Aim 3.1.1., a mixed-model design with one between-group factor (CAI, healthy controls) and two within-group factors of sensory systems and tasks were utilized. The overall study design is outlined in Table 3.1.

Table 2.1. Outline of Current Study Design.

PRE-SCREENING								
PRE-SCREENING QUESTIONNAIRES								
ONLINE								
TESTING SESSION								
PART 1: PATIENT ORIENTED OUTCOMES			**PART 2: WARM-UP**	**PART 3: DEMOGRAPHIC MEASURES**	**PART 4: LABORATORY ORIENTED OUTCOMES**			
QUESTIONNAIRES					SOT	Double-limb Injured-limb Uninjured-limb	Counterbalanced	AIM 1: Motor Behaviors AIM 3: Individual Perception AIM 4: Movement Variability
Physical Activity Levels	*Medical History*	*Self-Reported Ankle Instability & Function*			70-minutes			
NASA-PASS	Customized Questionnaires	CAIT IdFAI FAAM: ADL & Sports			ADT	Double-limb Injured-limb Uninjured-limb	Counterbalanced	AIM 2: Postural Adaptation
20-minutes			5-minutes	5-minutes	20-minutes			

NASA-PASS: National Aeronautics and Space Administration Physical Activity Status Scale; CAIT: Cumberland Ankle Instability Tool; IdFAI: Identification of Functional Ankle Instability; FAAM-ADL/Sports: The Foot and Ankle Ability Measure-Activities of Daily Living/Sports Subscales; SOT: Sensory Organization Test; ADT: Adaptation Test.

Participants

40 physically active individuals (females and males) between the ages of 16 to 39 years who presented to be CAI (N = 20) and healthy controls (N = 20) were recruited from a local Piedmont Triad area and university campuses (e.g., UNCG) to participate in this study. All participants reported being physically active, participating in moderate-intensity aerobic activity per week for at least 150-minutes or in vigorous-intensity aerobic activity at least 75-minutes per week. The self-reported level of physical activity over the past month was captured utilizing the National Aeronautics and Space Administration Physical Activity Status Scale (NASA-PASS). Potential participants were excluded if they had a history of 1) medically diagnosed concussion at least six months prior to the study enrollment, 2) neurological, vestibular, and/or visual disorders and/or disease (e.g., vertigo, epilepsy, stroke, peripheral neuropathies), 3) connective tissue disease and/or disorders (e.g., rheumatoid arthritis, Marfan syndrome, Ehlers-Danlos syndrome), 4) major surgeries in the brain and/or on the lower extremity (e.g., ankle, knee, hip, lower back), 5) ongoing inflammatory symptoms (pain, swelling, etc.) on the lower extremity within the past six weeks prior to study enrollment, 6) acute injuries to the lower extremity in the last six months prior to the study enrollment, and 7) chronic musculoskeletal conditions (e.g., OA, ACL deficiency).

Healthy controls were defined as those participants who 1) had no history of sustaining ankle sprains prior to the study enrollment, and 2) scored ≤ 11 on the Identification of Functional Ankle Instability [IdFAI], ≥ 28 on the Cumberland Ankle Instability Tool [CAIT], or $\geq 99\%$ and $\geq 97\%$ on the Foot and Ankle Ability Measure-

Activities of Daily Living [FAAM-ADL]/-Sports subscales (Gribble et al., 2013; Gribble, et al., 2014a, 2014b). CAI individuals were defined as those participants who 1) had sustained at least two lateral ankle sprains, 2) had a history of a minimum of one significant ankle sprain at least 12-months prior to study enrollment, 3) had experienced two episodes of previously injured ankle joint "giving way" and/or "feelings of instability" within the past six months prior to study enrollment and/or had a history of recurrent ankle sprains, 4) had not experienced recurrent ankle sprains in the last 3-months, and 5) scored ≥ 11 on the IdFAI, < 24 on the CAIT, or $< 90\%$ and $< 80\%$ on FAAM-ADL/-Sports subscales (Gribble et al., 2013; Gribble, et al., 2014a, 2014b). Participants with bilateral CAI were allowed to patricipate in the study, and the worst score on the IdFAI was identified as the injured-limb.

Participants in each group (CAI, healthy controls) were matched for sex, age (years, $\pm$ 2), height (cm, $\pm$ 5%), mass (kg, $\pm$ 3%), limb dominance (the leg used to kick a ball), and the NASA-PASS (scale, $\pm$ 1). Thus, the matched CAI group defined the injured-limbs of the healthy controls group. All participants and participants' parents/legal guardians read and signed the consent form. The consent form, pre-screening questionnaires, and a questionnaire package including personal information (e.g., age, date of birth, gender), past medical history, ankle instability and function (i.e., IdFAI, CAIT, FAAM-ADL/-Sports subscales), and physical activity status (i.e., NASA-PASS) were approved by the Institutional Review Boards of UNCG.

Sampling Procedures

All participants visited the Applied Neuromechanics Research group laboratory on the UNCG campus once if they qualified to enroll in the study. In order to be assessed for qualification (Appendix A), participants completed pre-screening questionnaires online prior to attending the laboratory session. Qualified participants completed the session in order of 1) a consent form, 2) a questionnaires package (Appendix B, C, D, E, F, G) on personal information (e.g., age, date of birth, gender), past medical history, ankle instability and function (i.e., IdFAI, CAIT, FAAM-ADL/-Sports), and physical activity status (i.e., NASA-PASS), 3) a 5-minute warm-up on a bike at a self-selected intensity, 4) demographic measures (height, weight), 5) hypermobility tests, 6) lower extremity anatomical alignment measures, and 5) the sensory organization test (SOT), then the adaptation test (ADT) in double- and single-limb stances in counterbalanced order within each group (CAI, healthy controls).

Individual Characteristics

<u>Limb Dominance</u>

Participants' self-reported limb dominance was determined by asking, "if you would kick a ball on a target, which leg would you use to kick the ball?" (van Melick et al., 2017).

Sensory Organization Test (SOT)

The SOT of the NeuroCom system (SMART EquiTest, NeuroCom International Inc., Clackamas, OR) was utilized to examine participants' postural stability within altered sensory environments. The SOT consisted of six-conditions (Table 3.2; Figures

3.1 & 3.2) that were designed to manipulate somatosensory and visual feedback in a combination of the sway-referenced support surface and visual surroundings with and without eyes-closed in a double-limb stance. Each condition comprised three 20-second trials. Conditions 1-3 had a sway-referenced support surface, and conditions 4-6 had sway-referenced visual surroundings. In condition 1, no sensory feedback was manipulated. In conditions 2 and 5, visual feedback was absent with eyes-closed. In condition 3, visual feedback was manipulated with sway-referenced visual surroundings. In conditions 4 and 5, somatosensory feedback was manipulated with the sway-referenced support surface. In condition 6, both somatosensory and visual feedback were manipulated with the sway-referenced support surface and visual surroundings. Participants aligned their bilateral ankles (medial malleoli) perpendicular to the axis of platform rotation while keeping the feet a standardized distance apart based on their height according to manufacturing guidelines for double-limb stance (Figure 3.3). For the single-limb stance, participants positioned each foot (injured, uninjured) in the center of the platform (Figure 3.4). Participants maintained their face forward and stood motionless as possible with arms relaxed by their sides while completing SOT conditions in both double- and single-limb stances, respectively (Figure 3.5). The stance limbs (double, injured, uninjured) were administered in counterbalanced order within each group (CAI, healthy controls) to minimize the learning effect. All participants were permitted to quickly tap down the platform (force plate) with non-stance toes multiple times after 10-seconds to complete the full 20-seconds trials in a single-limb stance (injured, uninjured). However, they were instructed to do their absolute best to maintain postural stability in a single-limb stance to complete each 20-second trial. The trials were stopped and repeated if participants tapped down on non-stance toes before 10-seconds and/or completely stood on a non-stance limb after 10-seconds. Participants were given a 30-second rest between trials and a 1-minute rest between conditions. An additional 1-minute rest was provided after the completion of one stance (e.g., double-limb stance) before transitioning into another stance (e.g., single-limb stance). All SOT data were collected at 100 Hz.

Table 2.2. Description of SOT Conditions.

Conditions	Descriptions	Manipulated Sensory Inputs	Vision
Condition 1	Eyes-open, Fxied sway-referenced support surface	None	Available
Condition 2	Eyes-closed, Fxied sway-referenced support surface	None	Absent
Condition 3	Eyes-open, Fxied sway-referenced support surface, Sway-referenced visual surroundigs	Visual sensory input	Available
Condition 4	Eyes-open, Sway-referenced support surface	Somatosensory input	Available
Condition 5	Eyes-closed, Sway-referenced support surface	Somatosensory input	Absent
Condition 6	Eyes-open, Sway-referenced surface and visual surroundigs	Somatosensory and Visual inputs	Available

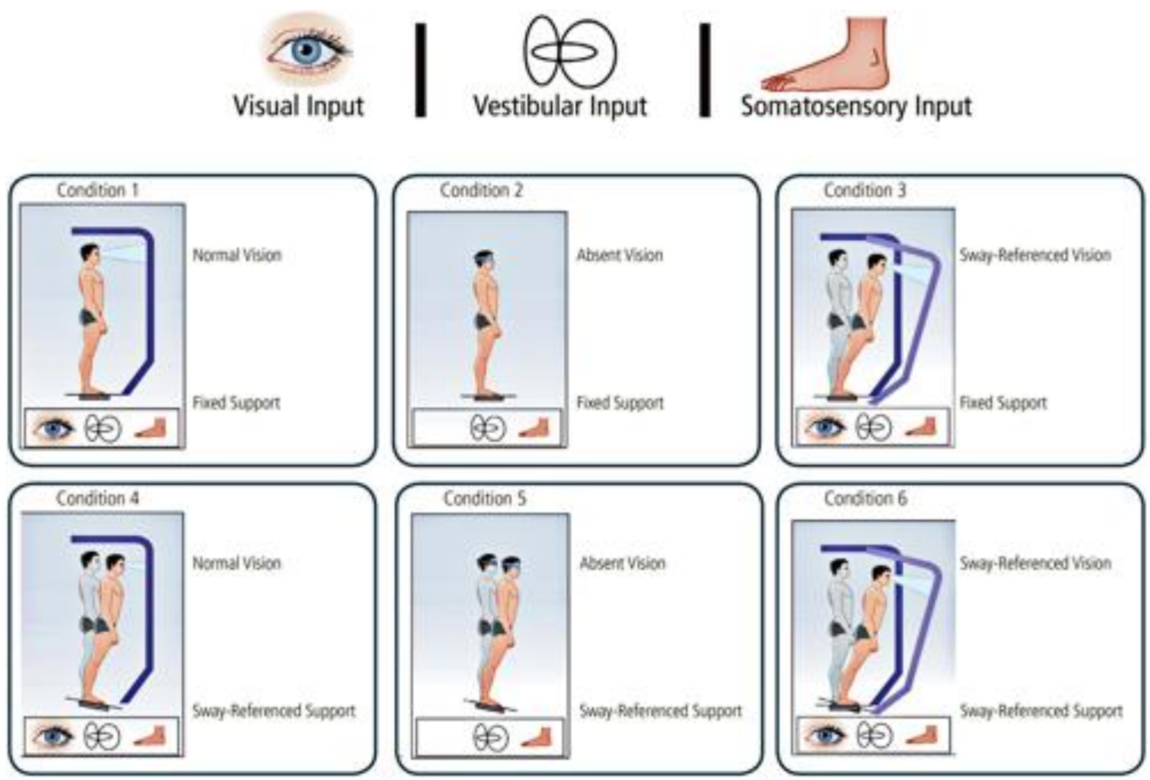

Figure 2.1. Six Conditions of the SOT (NeuroCom® International).

Figure 2.2. The NeuroCom Dynamic Posturography System.

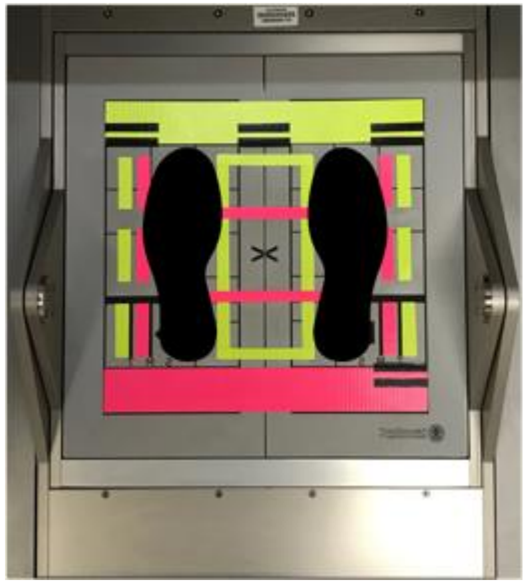

Figure 2.3. Foot Positions for the SOT in a Double-Limb Stance.

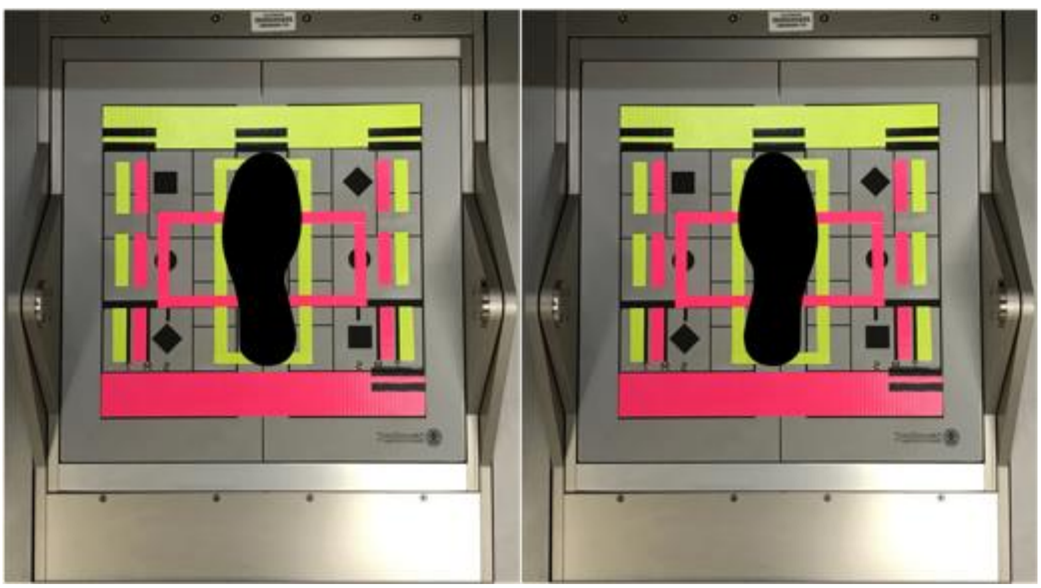

Figure 2.4. Foot Positions for the SOT in a Single-Limb Stance (Injured, Uninjured).

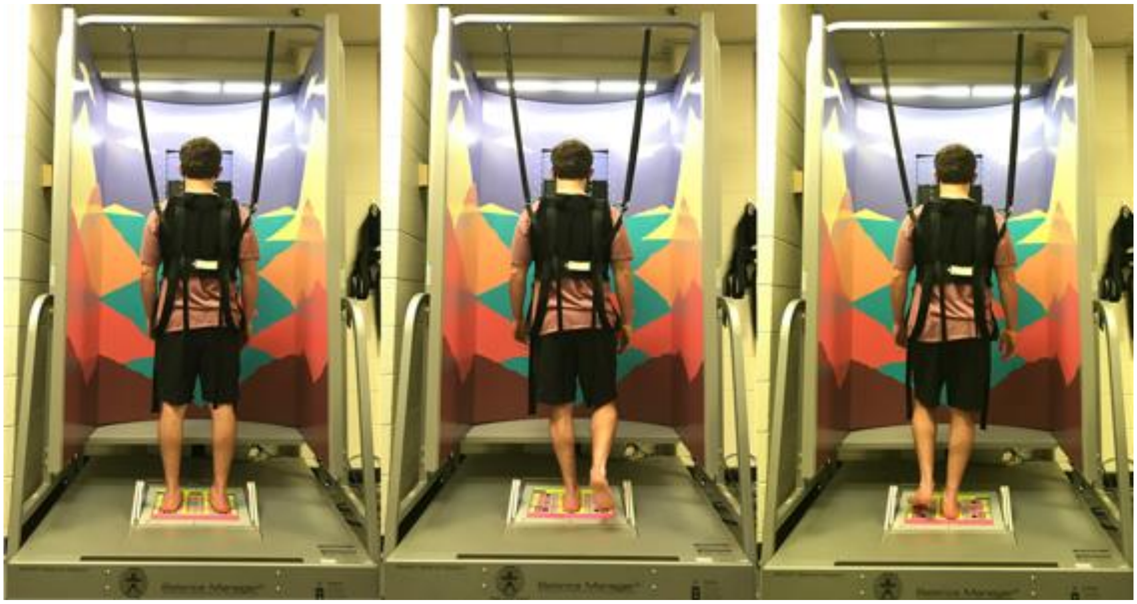

Figure 2.5. Stance Positions for the SOT in Double- and Single-Limb Stances on the NeuroCom.

Adaptation Test (ADT)

The ADT of the NeuroCom system was utilized to examine participants' postural adaptation from unexpected AP and ML rotations (tilts) of the support surface. The ADT consisted of two-conditions (dorsiflexion [toes-up], plantarflexion [toes-down]) which rotate in the AP plane at the amplitude of 8-degrees, each composed of five trials lasting 400 msec (Figure 3.6). In order to examine participants' postural adaptation from unexpected ML rotations (eversion [medial border of foot down], inversion [medial border of the foot up]) of the support surface, participants faced the sidewall of NeuroCom surroundings (Figure 3.7). Participants were instructed to align bilateral ankles (medial malleoli) perpendicular (for AP) or parallel (for ML) to the transverse axis of platform rotation with a standardized distance apart based on their height according to manufacture guidelines for double-limb stance and position the foot in the center of the platform for single-limb stance (Figures 3.8 & 3.9). Participants maintained their face forward and stood as motionless as possible with arms relaxed by sides while completing ADT conditions in double- and single-limb stances, respectively. The stance limbs (double, injured, uninjured) were administered in counterbalanced order within each group (CAI, healthy controls). The trials were stopped and repeated if participants stood on a non-stance limb at the moment of a sudden rotation of the support surface. Participants were given a 1-minute rest between conditions to minimize fatigue. All ADT data were collected at 100 Hz.

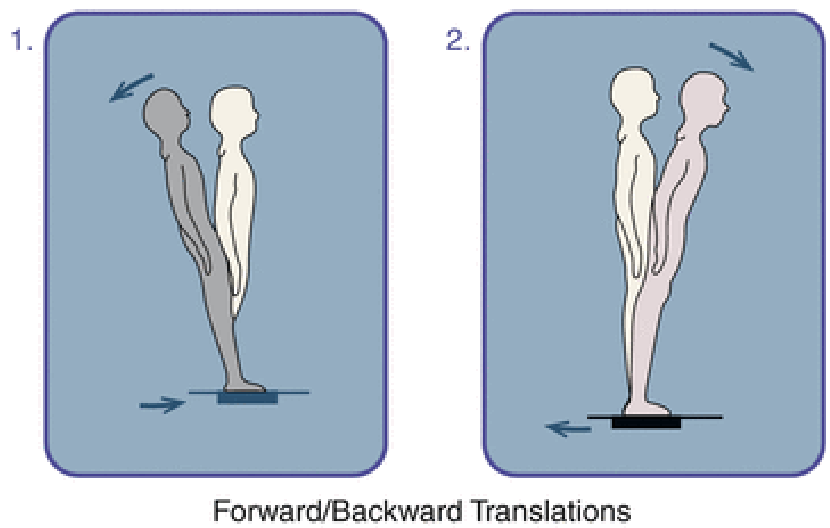

Figure 2.6. Standard Anterior-Posterior Surface Perturbation of the ADT (NeuroCom® International).

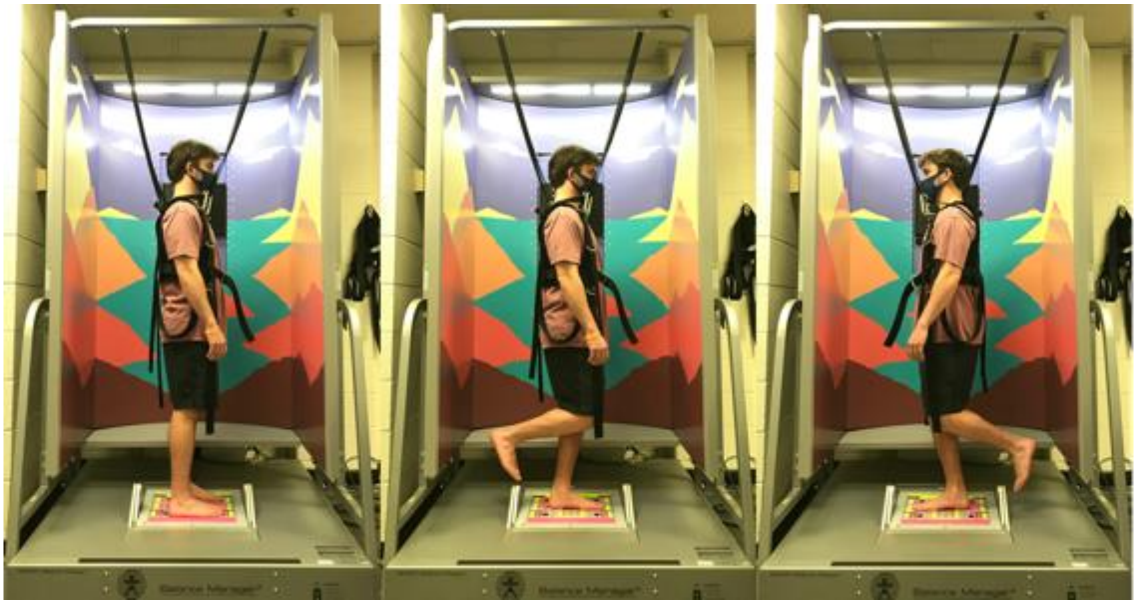

Figure 2.7. Stance Positions for the ADT in ML Direction on the NeuroCom.

Figure 2.8. Foot Positions for the ADT in AP Direction in Double- and Single-limb (Injured, Uninjured) Stances.

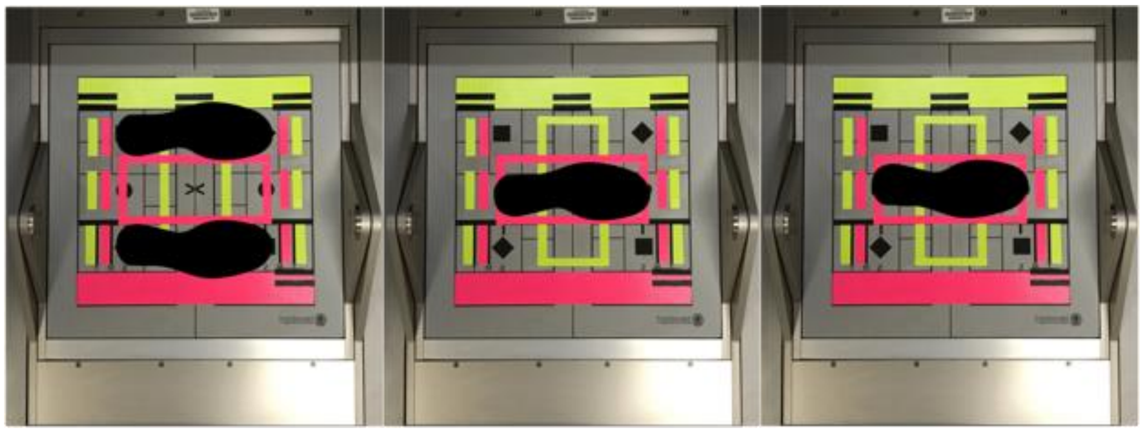

Figure 2.9. Foot Positions for the ADT in ML Direction in Double- and Single-limb Stances.

Data Reduction

National Aeronautics and Space Administration Physical Activity Status Scale (NASA-PASS)

NASA-PASS is an 8-point physical activity scale developed by the National Aeronautics and Space Administration's Johnson Space Center to provide individuals' physical activity for the past month (Jackson et al., 1990). A rating of 0-1 indicates a very low physical activity level; 2-3 indicates a moderate physical activity level; 4-7 indicates a high physical activity level. NASA-PASS has been demonstrated to have a moderate validity (r = 0.58) of aerobic capacity from physical activity (Jackson et al., 1995).

The Cumberland Ankle Instability Tool (CAIT)

The CAIT is a 9-item 30-point scale questionnaire that evaluates the severity of ankle instability independent of reference to the other limb (Hiller et al., 2006). The CAIT has demonstrably excellent reliability (ICC = 0.96) and good validity (α = 0.83) (Hiller et al., 2006).

The Identification of Functional Ankle Instability (IdFAI)

The IdFAI is a 10-item 37-point scale questionnaire that evaluates ankle stability status independent of reference to the other limb (Simon et al., 2012). The IdFAI has demonstratable excellent reliability (ICC = 0.91) and excellent validity (α = 0.98) in individuals who are 20 to 30 years old (Gurav et al., 2014).

The Foot and Ankle Ability Measure (FAAM)

The FAAM is a questionnaire consisting of 21-items, 84-points ADL and 8-items, 32-points Sports subscales used to evaluate the physical function of individuals with foot and ankle related impairments. The Sports subscale examines more difficult tasks that are essential to sports. The FAAM: ADL/Sports have demonstrated good reliability (ADL: ICC = 0.89; Sports: ICC = 0.87) and moderate reliability in individuals with a history of lower extremity injuries (e.g., ankle sprain, plantar fasciitis, Achilles rupture, etc.) (Martin et al., 2005).

Equilibrium Balance (EQ) Scores

The SOT of the NeuroCom system computed EQ scores of each of the three 20-second trials for each condition. Thus, raw data were exported from the NeuroCom, and a custom R program in RStudio (version 4.0.0; RStudio, Inc., Boston, MA) was utilized to calculate EQ_{10} scores based on 10-seconds of each of the three trials for each condition. EQ scores quantify how well the COG sway remains within the expected angular limits of stability, according to equation 3.1. The 12.5° represents the maximum normal AP postural sway, which is an angular limit in the sagittal plane; θ_{max} represents the maximum anterior COG excursion angles (degrees) during a trial; θ_{min} represents the minimum posterior COG excursion angles (degrees) during a trial (Figure 3.10). The θ is computed according to equation 3.2 based on four force transducers measurements (Figure 3.11: left front [LF], right front [RF], left rear [LR], right rear [RR]) that takes into account the 2.3° forward lean of COG from the true vertical when calculating sway from the ankle, by assuming individuals' COG is directly above the center of the foot support. The EQ score of 100 (unitless) represents perfect postural stability, whereas the EQ score of 0 (unitless) represents a loss of postural stability (Figure 3.12). The SOT has demonstrated moderate reliability for the SOT composite scores (ICC = 0.67) in healthy young adults (Wrisley et al., 2007).

$$\frac{12.5° - (\theta_{max} - \theta_{min})}{12.5°} * 100 \qquad (Equation\ 3.1)$$

$$arcsin\left(\frac{P_{COG}}{H_{COG}}\right) - 2.3^{\circ} \qquad (Equation\ 3.2)$$

P_{COG} and H_{COG} are defined as:

$$P_{COG} = COF_y = \frac{(RF + LF) - (LR + RR)}{RF + RR + LF + LR} * 4.2$$

$$H_{COG} = 0.55527 \times Total\ Height\ (inches)$$

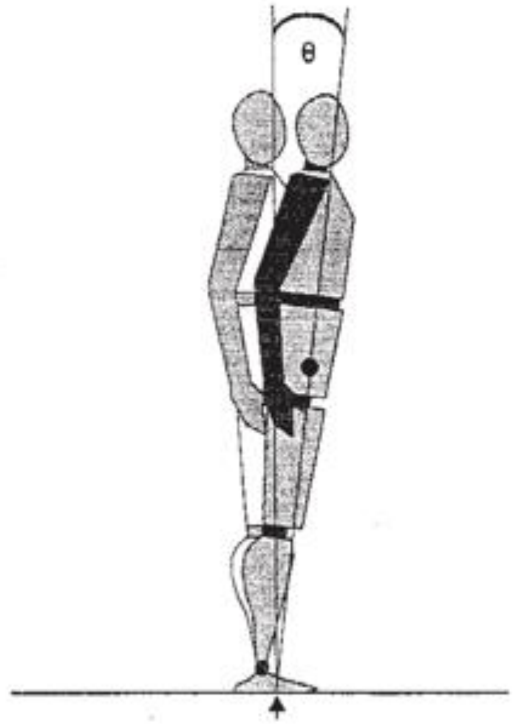

Figure 2.10. The COG sway angle θ (Natus Principles of Operation, 2014).

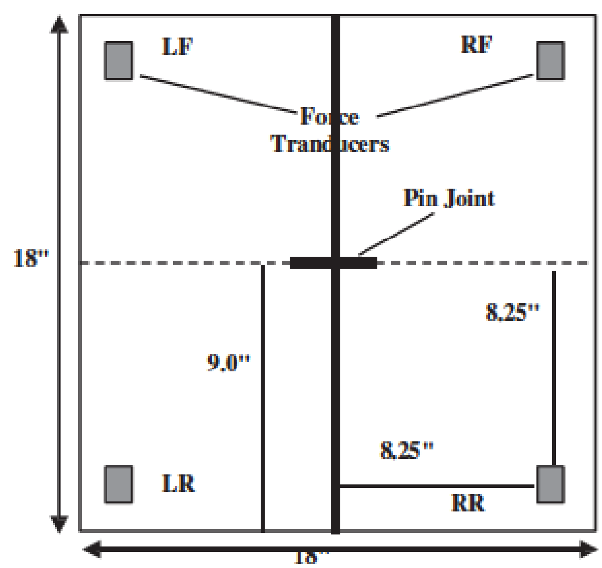

Figure 2.11. The location of four force transducers (loadcell) mounted at each corner of the NeuroCom force plate (Natus Principles of Operation, 2014).

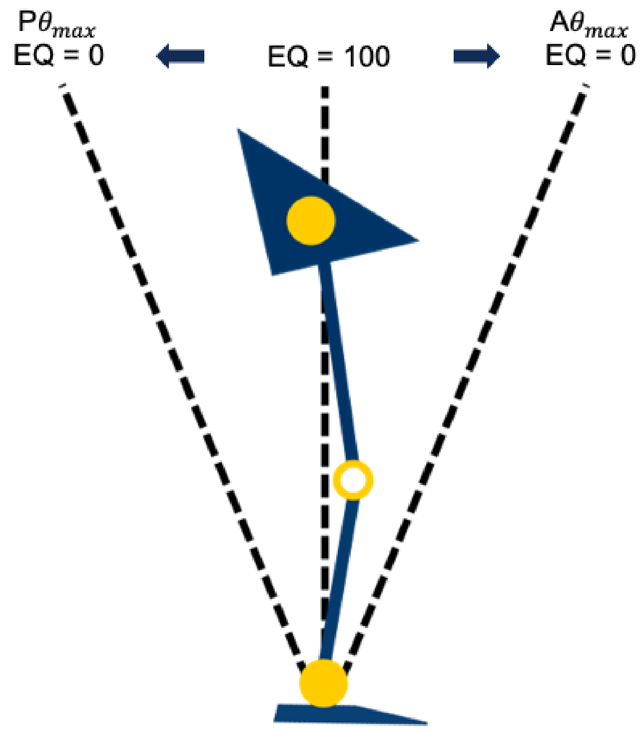

Figure 2.12. The COG sway angle θ and SOT EQ score distributions.
Diagram represents maximum anterior sway ($A\theta_{max}$) and maximum posterior sway ($P\theta_{max}$) relative to the vertical. True vertical represents the EQ score of 100. $A\theta_{max}$ and $P\theta_{max}$ represent the EQ score of 0.

Sensory Reweighting Ratios

Sensory reweighting ratios (unitless) of sensory systems (somatosensory [SOM], vision [VIS], vestibular [VEST]) were calculated on the average EQ_{10} scores from specific pairs of SOT conditions utilizing a custom R program in RStudio. The sensory reweighting ratios quantify the emphasis of sensory systems to maintain postural stability while performing the SOT according to equations 3.3, 3.4, and 3.5. The sensory reweighting ratios of 100 (percentage) represent more emphasis (i.e., upweighted) on the sensory system; the sensory reweighting ratios of 0 (percentage) represent no emphasis (i.e., downweighted) on the sensory system.

$$\frac{Condition\ 2}{Condition\ 1} = SOM \qquad (Equation\ 3.3)$$

$$\frac{Condition\ 4}{Condition\ 1} = VIS \qquad (Equation\ 3.4)$$

$$\frac{Condition\ 5}{Condition\ 1} = VEST \qquad (Equation\ 3.5)$$

Sway Energy (SE) Scores

The ADT of the NeuroCom system computed SE scores (unitless) of every one of the five trials for each condition, and they were exported to spreadsheets (Excel, version 360; Microsoft Corporation, Redmond, WA). Sway energy scores quantity participants' ability to adapt and maintain postural stability to an unexpected surface rotation according to equation 3.6. The magnitude of the y-axis vertical force response overcoming induced instability was differentiated twice (PY', PY'') to calculate the root mean squares (RMS) of given PY' and PY'', then the weighted sum of the RMS velocity PY' and RMS acceleration PY''. The NeuroCom motor control test (MCT), which is very similar to the ADT, has demonstrated excellent reliability for the MCT composite score (ICC = 0.92) in individuals with Parkinson's disease (Harro et al., 2016).

$$C1 * PY'(RMS) + C2 * PY''(RMS) \qquad (Equation\ 3.6)$$

C1 and C2 are weighting constants, giving dimensionless energy value defined as:

$$C1 = \frac{1}{\frac{in}{Sec}}\ C2 = \frac{0.025}{Sec^2}$$

Movement Variability (SampEN) in Postural Control

A custom R program in RStudio was utilized to compute SampEN of COP excursion while completing the SOT in double- and single-limb stances based on the algorithm (equation 3.7) presented by Richman and Moorman (2000). The equation defined as the negative logarithm computes the conditional probability that the number of data points within certain (A) vector lengths (B^m) in time series (N) falling within the relative tolerance limit (set tolerance $[r]$ times the standard deviation [SD] of time series) would be repeated within the vector lengths (A^{m+1}) in time series (N) falling within the relative tolerance limit (r: set tolerance (r) times the standard deviation (SD) of time series) (Raffalt et al., 2019). Parameter values of m, r, and N for the SampEN calculation would be selected based on the data of the current study utilizing a custom R program in RStudio. The SampEN value of 0 corresponds to less flexibility and adaptability and more predictability of neural control in sensorimotor pathways, resulting in more rigid movement patterns; the SampEN value of 2 corresponds to less flexibility and adaptability and unpredictability of neural control in sensorimotor pathways, resulting in more random movement patterns (Stergiou & Decker, 2011; Yentes et al., 2013, 2018).

$$SampEN\ (m, r, N) = -ln \left[\frac{A^{m+1}(r)}{B^{m}(r)}\right] \qquad (Equation\ 3.7)$$

Statistical Approach

All data were first exported to spreadsheets, and statistical analyses were performed in SPSS (version 27; IBM Corp, Armonk, NY, USA) and RStudio software. An a priori alpha level of 0.05 was used to denote statistical significance. Repeated measures analysis of variance (ANOVA) were followed by Tukey post-hoc analyses to ascertain the location of significant findings. Effect sizes were calculated utilizing Cohen's *d* to determine the extent of significant differences. The strength of effect sizes was interpreted as strong (≥ 0.80), moderate ($0.40 < d < 0.80$), and weak (≤ 0.40) (Cohen, 1988). 95% confidence intervals (CIs) for each dependent measure were computed. The following statistical analyses were used to examine each hypothesis:

Aim 1: Assessment of Motor Behaviors (i.e., Postural Stability)

The purpose of the specific Aim 1.1. was to determine group differences in postural control when the complexity of environmental (sensory systems) and task (limbs: double, injured, uninjured) constraints is manipulated while performing the SOT.

Environmental Constraints

Independent Variables:

1.1.Double-limb stance: Group (CAI, healthy controls) × Environment (6 SOT conditions)

1.2.Single-limb stance: Group (CAI, healthy controls) × Environment (6 SOT conditions)

Dependent Variables:

1.1: Equilibrium (EQ) scores (unitless) of SOT conditions

1.2: Equilibrium (EQ) scores (unitless) of SOT conditions

Task Constraints

Independent Variables:

1.3: Group (CAI, healthy controls) × Task (Limbs: double, injured, uninjured)

Dependent Variables:

1.3: Equilibrium (EQ) scores (unitless) of SOT conditions

***Hypothesis 1.1*:** The group differences in EQ scores (unitless) while performing SOT conditions in double-limb stance will be analyzed utilizing 2 × 6 repeated measures ANOVA with one-between (group: CAI, healthy controls) and one-within (6 SOT conditions) factors.

***Hypothesis 1.2*:** The group differences in EQ scores (unitless) while performing SOT conditions in single-limb (injured, uninjured) stance will be analyzed utilizing separate 2 × 6 repeated measures ANOVA with one-between (group:

CAI, healthy controls) and one-within (6 SOT conditions) factors for individual limbs (injured, uninjured).

Hypothesis 1.3**:** The group differences in EQ scores (unitless) transitioning from simple to more complex task constraints while performing the SOT will be analyzed utilizing separate 2 × 3 repeated measures ANOVA with one-between (group: CAI, healthy controls) and one-within (limbs: double, injured, uninjured) factors for individual SOT conditions (1-6).

Aim 2: Assessment of Motor Behaviors (i.e., Postural Adaptation)

The purpose of the specific Aim 2.1. was to determine group differences in the postural adaptation from a sudden unexpected platform tilt (inversion [IN], plantarflexion [PF]) while performing the adaptation test (ADT) in double- and single-limb (injured, uninjured) stances.

Environmental Constraints

<u>Independent Variables</u>:

2.1.Double-limb stance: Group (CAI, healthy controls) × Environment (2 ADT conditions: IN and PF)

2.2.Single-limb stance: Group (CAI, healthy controls) × Environment (2 ADT conditions: IN and PF)

<u>Dependent Variables</u>:

2.1.Double-limb stance: Sway energy (SE) scores (unitless) of ADT conditions

2.2.Single-limb stance: Sway energy (SE) scores (unitless) of ADT conditions

Task Constraints

Independent Variables:

2.1.3.ADT Inversion/Plantarflexion: Group (CAI, healthy controls) × Task (Limbs: double, injured, uninjured)

Dependent Variables:

2.1.3: Sway energy (SE) scores (unitless) of ADT conditions

Hypothesis 2.1.1**:** The group differences in SE scores (unitless) while performing ADT conditions in double-limb stance will be analyzed utilizing 2 × 2 repeated measures ANOVA with one-between (group: CAI, healthy controls) and one-within (2 ADT conditions) factors.

Hypothesis 2.1.2**:** The group differences in SE scores (unitless) while performing ADT conditions in single-limb (injured, uninjured) stance will be analyzed utilizing separate 2 × 2 repeated measures ANOVA with one-between (group: CAI, healthy controls) and one-within (2 ADT conditions) factors for individual limbs (injured, uninjured).

Hypothesis 2.1.3**:** The group differences in SE scores (unitless) transitioning from simple to more complex task constraints while performing the ADT will be analyzed utilizing separate 2 × 3 repeated measures ANOVA with one-between (group: CAI, healthy controls) and one-within (limbs: double, injured, uninjured) factors for individual ADT conditions (inversion, plantarflexion).

Aim 3: Assessment of Individual Perception (i.e., Sensory Reweighting System)

The purpose of the specific Aim 3.1. was to determine group differences in sensory reweighting on each sensory system (somatosensory, vision, vestibular) in postural controls when the complexity of task (limbs: double, injured, uninjured) constraints is manipulated while performing the SOT.

Task Constraints

Independent Variables:

3.1: Group (CAI, healthy controls) × Sensory Systems (somatosensory, vision, vestibular)

× Task (Limbs: double, injured, uninjured)

Dependent Variables:

3.1: Sensory reweighting ratios (unitless) of each sensory system (somatosensory, vision, vestibular) on the SOT

Hypothesis 3.1**:** The group differences in sensory reweighting ratios (unitless) transitioning from simple to more complex task constraints while performing the SOT will be analyzed utilizing 2 × 3 × 3 repeated measures ANOVA with one-between (group: CAI, healthy controls) and two-within (sensory systems: somatosensory, vision, vestibular; limbs: double, injured, uninjured) factors.

Aim 4: Assessment of Neural Control Underlying Movement (i.e., Movement Variability)

The purpose of the specific Aim 4.1 was to determine group differences in movement variability of COP excursion in postural control when the complexity of environmental (sensory systems) and task (limbs: double, injured, uninjured) constraints is manipulated while performing the SOT.

Environmental Constraints

Independent Variables:

4.1.Double-limb stance: Group (CAI, healthy controls) × Environment (6 SOT conditions)

4.1.2.Single-limb stance: Group (CAI, healthy controls) × Environment (6 SOT conditions)

Dependent Variables:

4.1: SampEN (unitless) of the COP excursion on SOT conditions

4.2: SampEN (unitless) of the COP excursion on SOT conditions

Task Constraints

Independent Variables:

4.3: Group (CAI, healthy controls) × Task (Limbs: double, injured, uninjured)

Dependent Variables:

4.3: SampEN (unitless) of the COP excursion on SOT conditions

***Hypothesis 4.1*:** The group differences in SampEN (unitless) of the COP excursion while performing SOT conditions in double-limb stance will be analyzed utilizing 2 × 6 repeated measures ANOVA with one-between (group: CAI, healthy controls) and one-within (6 SOT conditions) factors.

***Hypothesis 4.2*:** The group differences in SampEN (unitless) of the COP excursion while performing SOT conditions in single-limb (injured, uninjured) stance will be analyzed utilizing separate 2 × 6 repeated measures ANOVA with one-between (group: CAI, healthy controls) and one-within (6 SOT conditions) factors for individual limbs (injured, uninjured).

Hypothesis 4.3**:** The group differences in SampEN (unitless) of the COP excursion transitioning from simple to more complex task constraints while performing the SOT will be analyzed utilizing separate 2 × 3 repeated measures ANOVAs with one-between (group: CAI, healthy controls) and one-within (limbs: double, injured, uninjured) factors for individual SOT conditions (1-6).

Power Analysis

An a priori power analysis was performed utilizing the pilot data analyses (3 × 2 repeated measures ANOVA) on stride-to-stride gait variability in bilateral limbs (injured, uninjured) during walking in individuals with CAI, copers, and healthy controls (CAI: IN = 2.10 ± 0.27, UNIN = 2.11± 0.24; Copers: IN = 2.07 ± 0.38, UNIN = 2.05 ± 0.44; Healthy: IN = 2.21 ± 0.49 UNIN =2.03 ± 0.38; effect size: f(U)=0.357) on G*Power (Version 3.1.9.6). The largest effect found in the pilot study across three groups and bilateral limbs (injured, uninjured) was on the group by limb interactions (f(U)=0.357) which was close to a previously published study on stride-to-stride gait variability in frontal plane ankle kinematics in individuals with CAI and healthy controls ($d = -0.59$) (Terada et al., 2015). The estimated total sample size to determine statistical differences was 84, with 80% power and an α level of 0.05. We would recruit 84 participants to ensure adequate statistical power.

The original power analysis was based on a larger data collection methodology that was refined in the dissertation proposal to remove the gait analysis portion of the study due to the COVID-19 pandemic. Therefore, a sample size of 40 (20 per group) was approved for this dissertation. In the latter, an a priori power analysis was performed utilizing G*Power with a large effect size of $f = 0.40$. The estimated total sample size to determine statistical differences was 36 with 80% power and an α level of 0.05.

CHAPTER IV

MANUSCRIPT I

Introduction

Initial ankle sprains result in mechanical and perceived impairments at the ankle. The mechanical or perceived instability is experienced by at least 30% and up to 74% of those individuals who sustained an initial ankle sprain, contributing to the development of chronic ankle instability (CAI) (Hertel, 2002; Anandacoomarasamy et al., 2005). Individuals with CAI suffer subsequent ankle sprains and develop lifetime functional disabilities in daily living. Indeed, CAI is the second leading cause of trauma-initiated joint disease, post-traumatic ankle osteoarthritis (PTAOA) (Lofvenberg et al., 1994; Valderrabano et al., 2005; Goldiz et al., 2014; Thomas et al., 2017). Lower extremity post-traumatic osteoarthritis (PTOA) is associated with $11.79 billion in medical costs in the United States alone, with direct costs of over $3 billion annually (Thomas et al., 2017). However, greater than 50% of individuals who sustain an ankle sprain neglect seeking proper medical treatment. Thus, the health care burden emerging from CAI combined with PTAOA is substantial, especially for those CAI individuals who did not seek medical attention nor complete rehabilitation to restore function.

The cause of subsequent ankle sprains in individuals with CAI has been attributed to articular deafferentation. The articular deafferentation theory hypothesizes that an ankle sprain damages mechanoreceptors in the ankle joint capsule and/or ligament disrupting somatosensory feedback to the central nervous system (CNS) (Freeman, 1965;

Riemann & Lephart, 2002; Hertel, 2008). This disruption can result in a diminished ability to obtain relevant somatosensory feedback. Inability to obtain relevant sensory feedback may contribute to alteration in CAI individuals' overall proprioceptive perception and motor behaviors (e.g., postural control), resulting in less flexible and adaptable sensorimotor systems with chronicity. Therefore, CAI individuals often compensate by employing a rigid movement pattern which is a freezing of the degrees of freedom at the ankle to restrain excessive movement, especially as environmental and task constraints increase (Brown & Mynark, 2007; Pope et al., 2011; Hamacher et al., 2016; Raffalt et al., 2019; Wanner et al., 2019). Furthermore, the strategy to freeze degrees of freedom is not limited to rigid fixation of the ankle joint. It could be achieved by employing unisensory integration by emphasizing a single sensory system, known as unisensory integration, to obtain relevant sensory feedback to accomplish a given task (Bronstein et al., 1990; Bonan et al., 2004; Hafstrom et al., 2004; Lopez et al., 2006; Slaboda et al., 2009; Manor et al., 2010; Lin et al., 2019).

Healthy individuals who present flexible and adaptable sensorimotor systems can freely integrate redundant sensory feedback from three primary sensory systems (somatosensory, vestibular, vision), known as multisensory integration. The multisensory integration processes relevant sensory feedback by placing emphasis (i.e., upweighted) on multiple sensory feedback proportional to the ever-changing environment and task goals. For instance, healthy individuals upweight 70% on somatosensory, 20% on vestibular, and 10% on visual feedback to maintain postural control on a stable surface. When somatosensory feedback becomes disrupted by a sudden change in environmental constraints (from stable to unstable surface), these healthy individuals can also freely reweight emphasis on visual and vestibular feedback (Peterka, 2002). The ability to reweight and seek an optimal combination of sensory feedback based on the complexity of environmental and task constraints has been described as the sensory reweighting system (Peterka, 2002).

In a systematic review with meta-analysis, researchers concluded CAI individuals rely heavily on visual feedback to compensate for somatosensory deficits while maintaining postural control in a single-limb stance compared to healthy controls (Song et al., 2016). Although CAI individuals have a compensatory reliance on visual feedback, postural control deficits persist in individuals with CAI compared to healthy controls (Arnold et al., 2009; Munn et al., 2010). Thus, heavy reliance on visual feedback may be an indication of inadequate multisensory integration resulting in the employment of unisensory integration in CAI. We hypothesize that the cause of altered postural control deficits among CAI pertains to their inability to freely integrate multisensory feedback. How those CAI individuals reweight sensory feedback in certain constraints (environment & task) is unclear. For instance, how CAI individuals reweight sensory feedback transitioning from simple to more complex environmental and task constraints is unknown. Therefore, the primary purpose of this study was to determine group differences in the sensory reweighting system transitioning from a simple double-limb stance to a more complex single-limb stance while maintaining posture with a sway-referenced support surface and/or with visual surroundings. The secondary purpose was to determine group differences in postural control while environmental (sensory systems) and task (stance limbs: double, injured, uninjured) constraints were manipulated. Our primary hypothesis was that participants with CAI would increase reliance on visual feedback while maintaining posture in the injured-limb compared to their uninjured-limb and double-limb stance, and matched stance limbs (double, injured, uninjured) of healthy controls. We also expected those with CAI to present greater postural control impairments while standing in the injured-limb and when somatosensory and visual sensory systems are manipulated.

Methods

Study Design

We implemented a case-control and mixed-model design to examine postural control and the sensory weighting system in individuals with and without CAI.

Participants

A total of 44 physically active individuals, consisting of individuals with CAI (13 females, 9 males; age: 26.09 $\pm$ 5.76 years; height: 172.25 $\pm$ 9.76 cm; weight: 76.18$\pm$14.91 kg; National Aeronautics and Space Administration Physical Activity Status Scale [NASA-PASS]: 6.27 $\pm$ 0.18; IdFAI: 19.09 $\pm$ 5.39) and without CAI (13 females, 9 males; age: 25.41 $\pm$ 5.92 years; height: 169.70 $\pm$ 9.32 cm; weight: 71.98 $\pm$ 14.79 kg; NASA-PASS: 6.27 $\pm$ 1.03; IdFAI: 1.36 $\pm$ 1.81) volunteered to participate in this study. CAI individuals were defined based on the International Ankle Consortium position statement as those participants who 1) had a history of at least one significant ankle sprain at least 12-months prior to study enrollment, 2) had experienced 2 episodes of previously injured ankle joint “giving way” and/or “feelings of instability” within the past 6-months and/or had a history of recurrent ankle sprains, 3) scored $>$ 11 on the Identification of Functional Ankle Instability (IdFAI), and 4) had not experienced recurrent ankle sprains in the last 3-months (Gribble et al., 2013; Gribble et al., 2014a, 2014b). Healthy controls were those participants who had no history of an ankle sprain or "giving way" sensations, and scored $\leq$ 11 on the IdFAI (Simon et al., 2012). All participants were required to be free of 1) inflammatory symptoms and surgeries in the brain and/or on the lower extremity (i.e., ankle, knee, hip, lower back), 2) medically diagnosed concussion within the past 6-months prior to study enrollment, 3) chronic musculoskeletal conditions (e.g., OA, ACL deficiency), 4) connective tissue disease and disorders (e.g., rheumatoid arthritis, Marfan syndrome, Ehlers-Danlos syndrome), and 5) vestibular or visual disorders and peripheral neuropathies that may have influenced postural control. Those individuals with bilateral CAI were allowed to participate, and the limb with worse IdFAI scores was considered the injured limb. Healthy controls were matched to participants with CAI for sex, age (years, $\pm$ 2), height (cm, $\pm$ 5%), mass (kg, $\pm$ 3%), limb dominance (the leg used to kick a ball), and the NASA-PASS (scale, $\pm$ 1), identifying the injured-limb. Participants signed the informed consent approved by the Institutional Review Board at the University of North Carolina Greensboro.

Procedure

Participants who met the study criteria by completing the online pre-screening survey were invited for a single session in our laboratory. All participants completed a standardized health history questionnaire including questions about previous ankle sprains (initial, most recent), injuries on the lower extremity (i.e., knee, hip, lower back), self-reported ankle instability and function (i.e., IdFAI), and physical activity status (i.e., NASA-PASS) upon arrival. Participants then performed a 5-minute warm-up on a bike, at a self-selected intensity, and completed demographic measures (height, weight), joint hypermobility tests, lower extremity (ankle, knee, hip) anatomical alignment measures, and the sensory organization test (SOT). A vest and safety harness were outfitted prior to completing the SOT, barefoot on a computerized NeuroCom dynamic posturography platform (SMART EquiTest, NeuroCom International Inc., Clackamas, OR), in double- and single-limb (injured, uninjured) stances assuring the safety of participants (Figure 4.1).

Sensory Organization Test

The SOT comprises six conditions that are designed to manipulate somatosensory and visual feedback in a combination of the sway-referenced support surface and visual surroundings with and without eyes-closed in a double-limb stance. Conditions 1 to 3 have a sway-referenced support surface and conditions 4 to 6 have sway-referenced visual surroundings (Figure 4.2). All three sensory feedback is enabled in condition 1, visual feedback is absent with eyes-closed in condition 2, visual feedback is manipulated with sway-referenced visual surroundings in condition 3, somatosensory feedback is manipulated with a sway-referenced support surface in condition 4, somatosensory feedback is manipulated with a sway-referenced support surface and visual feedback is absent with eyes-closed in condition 5, and both somatosensory and visual feedback is manipulated with the sway-referenced support surface and visual surroundings in condition 6 (Figure 4.2).

Each participant's bilateral ankles (medial malleoli) were aligned perpendicular to the transverse axis of platform rotation while keeping the feet a standardized distance apart based on their height for double-limb stance (Figure 4.3) and positioning the foot in the center of the platform for single-limb stance (Figure 4.4). We instructed participants to maintain their face forward, keep their arms relaxed by sides, and maintain motionless as possible while completing the SOT in double- and single-limb stances (Figure 4.5). Each SOT condition consisted of three 20-second trials, a total of 18-trials per stance limb (double, single: injured, uninjured). The stance limbs (double, single: injured, uninjured) were assessed in counterbalanced order within each group (CAI, healthy controls) to minimize the learning effect. Participants were given a 30-second rest between trials and a 1-minute rest between conditions. An additional 1-minute rest was provided after the completion of one stance (e.g., double-limb stance) before transitioning into another stance (e.g., single-limb stance). Every participant, not limited to CAI individuals, was allowed to quickly tap down the platform with non-stance toes multiple times after 10-seconds to complete the full 20-seconds trials in single-limb stance (injured, uninjured). However, individuals were directed to do their absolute best to maintain postural stability in a single-limb stance to complete each 20-second trial. The trials were stopped and repeated if participants tapped down on non-stance toes before 10-seconds and/or completely stood on a non-stance limb after 10-seconds.

Balance Measure

The SOT of NeuroCom computes *Equilibrium (EQ) scores* measuring center-of-gravity (COG) at 100 Hz. EQ scores quantify how well the COG sway ($\theta_{max} - \theta_{min}$)

remains within the expected angular limits of stability (12.5°) according to equation 4.1, normalizing by 100. The 12.5° represents the maximum normal anterior-posterior (AP) postural sway, while θ_{max} and θ_{min} represent the maximum anterior and minimum posterior COG excursion angles (degrees), respectively, during a trial.

$$EQ\ Score\text{:}\ \frac{12.5° - (\theta_{max} - \theta_{min})}{12.5°} * 100 \qquad (Equation\ 4.1)$$

If participants exhibit minimum AP postural sway, EQ scores are closer to 100. Yet, if participants exhibit maximum AP postural sway approaching the limits of stability, EQ scores are closer to 0. Therefore, the EQ score of 100 represents perfect postural stability, whereas the EQ score of 0 represents a loss of postural stability in individuals with and without CAI. The raw data from NeuroCom were exported to spreadsheets (Excel, version 360; Microsoft Corporation, Redmond, WA) and imported into a custom R program in RStudio (version 4.0.0; RStudio, Inc., Boston, MA) to compute the EQ score based on the first 10-second trials (EQ_{10}).

Sensory Reweighting Measure

The SOT of NeuroCom also calculates *Sensory Reweighting Ratios* among three primary sensory systems (somatosensory [SOM], vision [VIS], vestibular [VST]) by utilizing the averaged EQ scores from specific pairs of SOT conditions according to equations 4.2, 4.3, and 4.4. The ratios identify participants' ability to emphasize weight on sensory feedback to maintain postural stability while performing the SOT in double- and single-limb stances. The sensory reweighting ratios of 100 represent more emphasis (i.e., upweighted) on the sensory feedback, while the sensory reweighting ratios of 0 represent no emphasis (i.e., downweighted) on the sensory feedback. We have computed sensory reweighting ratios based on EQ_{10} scores using a custom R program in RStudio.

$$SOM: \frac{Condition\ 2}{Condition\ 1} \qquad (Equation\ 4.2)$$

$$VIS: \frac{Condition\ 4}{Condition\ 1} \qquad (Equation\ 4.3)$$

$$VST: \frac{Condition\ 5}{Condition\ 1} \qquad (Equation\ 4.4)$$

Statistical Analysis

An independent *t*-test was conducted to compare group differences in demographic variables (age, height, weight, NASA-PASS). 2 (group) × 3 (sensory systems: somatosensory, vision, vestibular) × 3 (task: double-, injured-, uninjured-limbs) repeated measures analysis of variance (ANOVA) were conducted to examine the sensory reweighting system while maintaining postural stability during the SOT, transitioning from double- to single-limb (injured, uninjured) stances. To examine postural control in transition from simple to more difficult environmental and task constraints while performing the SOT, separate 2 (group) × 6 (environment: SOT conditions 1-6) repeated measures ANOVA were performed examining environmental constraints effect for individual limbs (double, injured, uninjured). Separate 2 (group) × 3 (task: double-, injured-, uninjured-limbs) repeated measures ANOVA were performed to examine task constraints effect for individual SOT conditions (1-6). Tukey post-hoc analyses were performed if significant interactions were found. Cohen's *d* effect sizes (weak $\leq$ 0.40, moderate = 0.40–0.80, strong $\geq$ 0.80) were calculated with a corresponding 95% confidence interval (95% CI) between the group means to determine the range of EQ scores and sensory reweighting ratios (Cohen, 1988). All statistical analyses were performed using SPSS software (version 27; IBM Corp, Armonk, NY, USA) with a priori α level of 0.05.

Results

Independent *t*-tests did not reveal group differences in age, height, weight or physical activity level (*t* range = 0.39-0.94; $P > 0.05$; Table 4.1). CAI individuals had a greater number of ankle sprains, episodes of giving-way within the past 6-months, higher IdFAI scores compared to healthy controls ($P < 0.05$; Table 4.1).

Sensory Reweighting System:

Significant group (CAI, healthy controls) $\times$ sensory systems (somatosensory, vision, vestibular) $\times$ task (stance limbs: double, injured, uninjured) interactions were noted ($F_{4,168} = 3.214$; $P = 0.014$; Table 4.2; Figure 4.6). Tukey post-hoc comparisons revealed CAI individuals (69.91 $\pm$ 2.09) upweighted more on vestibular feedback compared to healthy controls (63.14 $\pm$ 2.09; $P = 0.027$; Table 4.2; Figure 4.6) while controlling posture in the injured-limb. Significant sensory reweighting ratios on somatosensory (97.22 $\pm$ 0.54) compared to vision (81.79 $\pm$ 1.39, $P = 0.000$, Table 4.2, Figure 4.6) and vestibular (81.82 $\pm$ 1.39, $P = 0.000$, Table 4.2, Figure 4.6) were found, respectively, in a double-limb stance for the CAI group. In addition, the CAI group exhibited significant differences in sensory reweighting ratios across all combinations of individual stance limbs for somatosensory (double [DL]: 97.22 $\pm$ 0.54, injured [INJ]: 81.79 $\pm$ 1.39, uninjured [UNI]: 81.82 $\pm$ 1.39; P = 0.000; Table 4.2; Figure 4.6), vision (DL: 93.20 $\pm$ 0.96, INJ: 94.83 $\pm$ 0.94, UNI: 94.41 $\pm$ 0.92, P = 0.000; Table 4.2; Figure 4.6), and vestibular (DL: 71.41 $\pm$ 1.98, INJ: 69.91 $\pm$ 2.09, UNI: 70.13 $\pm$ 1.90; P = 0.000; Table 4.2; Figure 4.6). There was a significant main effect for sensory systems ($F_{2,84}$ = 47.55; P = 0.000; Figure 4.7) and task ($F_{2,84}$ = 466.91; P = 0.000; Figure 4.8) but not for group ($F_{1,42}$ = 1.83; P = 0.18; Figure 4.9). Tukey post-hoc comparisons for sensory systems main effect showed a significantly increased sensory reweighting ratios on somatosensory (87.56 $\pm$ 0.69) compared to vision (80.59 $\pm$ 0.82; P = 0.000; Figure 4.7) and vestibular (80.95 $\pm$ 0.79, P = 0.000; Figure 4.7), respectively. Moreover, Tukey post-hoc comparisons for the task main effect showed significant differences in sensory reweighting ratios across all combination of individual stance limbs: double- (86.22 $\pm$ 0.53), injured- (94.05 $\pm$ 0.49), and uninjured-limbs (68.82 $\pm$ 1.51; All: P = 0.000; Figure 4.8), indicating the greater sensory feedback was utilized to maintain posture in the injured-limb.

Postural Control in Increased Environmental (SOT Conditions) Constraints:

There was a significant group $\times$ environment interaction for the injured-limb ($F_{5,210}$ = 2.62, P = 0.03; Table 4.3; Figure 4.10), but not for double- ($F_{5,210}$ = 0.43, P = 0.83; Table 4.4; Figure 4.11) or the uninjured-limbs ($F_{5,210}$ = 1.96,P = 0.09; Table 4.5; Figure 4.12). Tukey post-hoc comparisons revealed CAI individuals (62.45 $\pm$ 1.96) presented better postural control in the injured-limb for condition 5 compared to healthy controls (56.26 ± 1.96; $P = 0.03$; Table 4.3; Figure 4.10). There was no significant group main effect for all stance limbs ($F_{1,42}$ range = 0.57-1.91, $P > 0.05$; Figure 4.13). However, a significant main effect for environment was found for individual stance limbs; double- ($F_{5,210} = 235.65$, $P = 0.000$; Figure 4.14), injured- ($F_{5,210} = 206.99$, $P = 0.000$; Figure 4.15), and uninjured-limbs ($F_{5,210} = 210.22$, $P = 0.000$; Figure 4.16). Tukey post-hoc comparisons indicated significant differences across all combinations of environment for individual stance limbs (P range = 0.000-0.008) except for comparisons between SOT conditions 2 (92.17 ± 0.41) and 3 (92.17 ± 0.46; $P > 0.05$; Figure 4.14) in double-limb stance and SOT conditions 2 (72.09 ± 0.91) and 6 (69.49 ± 1.28; P > 0.05; Figure 4.15) in the uninjured-limb. Descriptively, individuals with and without CAI displayed the best postural control in the SOT condition 1 (double: 94.74 ± 0.21; injured: 89.15 ± 0.38; uninjured: 88.99 ± 0.43; Figures 4.14-4.16) and the worse postural control in the SOT condition 5 (double: 68.57 ± 1.35; injured: 59.36 ± 1.39; uninjured: 60.18 ± 1.30; Figures 4.14-4.16) among all stance limbs.

Postural Control in Increased Task (Stance Limbs: Double, Injured, and Uninjured) Constraints:

There was no significant group × task interactions across all SOT conditions ($F_{2,84}$ range = 0.00-2.87, $P > 0.05$; Tables 4.6-4.10; Figures 4.17-4.21) except for the SOT condition 5 ($F_{2,84} = 5.40$, $P = 0.006$; Table 4.11; Figure 4.22). Tukey post-hoc comparisons revealed better postural control in the injured-limb in CAI individuals (62.45 ± 1.96) compared to healthy controls (56.26 ± 1.96; $P = 0.031$; Table 4.11; Figure 4.22). For the CAI group, significantly better postural control was presented in double-limb stance (67.64 $\pm$ 1.92) compared to the injured- (62.45 $\pm$ 1.96) and uninjured-limbs (62.41 $\pm$ 1.84, $P = 0.010$; Table 4.11; Figure 4.22), respectively. No significant main effect for the group was found for all SOT conditions ($F_{1,42}$ range = 0.01-1.18, $P > 0.05$; Figure 4.23). Conversely, a significant main effect for task was found for individual SOT conditions (Condition 1: $F_{2,84} = 122.63$, $P = 0.000$; Condition 2: $F_{2,84} = 291.98$, $P = 0.000$; Condition 3: $F_{2,84} = 121.35$, $P = 0.000$; Condition 4: $F_{2,84} = 22.92$, $P = 0.000$; Condition 5: $F_{2,84} = 31.28$, $P = 0.000$; Condition 6: $F_{2,84} = 4.78$, $P = 0.011$; Figure 4.24-4.28). Tukey post-hoc comparisons displayed a significantly better postural control in double-limb stance compared to the injured- and uninjured-limbs for SOT conditions 1 (DL: 94.74 $\pm$ 0.21; INJ: 89.15 $\pm$ 0.38, $P = 0.000$; UNI: 88.99 $\pm$ 0.43, $P = 0.000$; Figure 4.24), 2 (DL: 92.17 $\pm$ 0.41; INJ: 71.62 $\pm$ 0.92, $P = 0.000$; UNI: 72.09 $\pm$ 0.75, $P = 0.000$; Figure 4.25), 3 (DL: 92.17 $\pm$ 0.46; INJ: 80.56 $\pm$ 0.84, $P = 0.000$; UNI: 80.94 $\pm$ 0.75, $P = 0.000$; Figure 4.26), 4 (DL: 88.14 $\pm$ 0.68; INJ: 84.58 $\pm$ 0.66, $P = 0.000$; UNI: 83.84 $\pm$ 0.69, $P = 0.000$; Figure 4.27), and 5 (DL: 68.57 $\pm$ 1.35; INJ: 59.36 $\pm$ 1.39, $P = 0.000$; UNI: 60.18 $\pm$ 1.30, $P = 0.000$; Figure 4.28), respectively. Whereas for the SOT condition 6, postural control in double-limb stance (72.34 $\pm$ 1.41) was significantly better only compared to the injured-limb (68.10 $\pm$ 1.30; $P = 0.004$, Figure 4.29).

Discussion

In the current study, we investigated how the sensory reweighting system transitions from a simple double-limb stance to a more complex single-limb (injured, uninjured) stance while maintaining postural control in individuals with and without CAI. The secondary purpose was to determine group differences in postural control with the increased complexity of environmental and task constraints. This is the first study to our knowledge that examined postural control interactions between group, task, and sensory systems as well as the effect of environmental and task constraints to maintain postural control in individuals with and without CAI. The unique findings were that CAI individuals upweighted on vestibular feedback while controlling posture in the injured-limb compared to healthy controls. Our results also showed somatosensory feedback was significantly emphasized (i.e., upweighted) to maintain postural control in a double-limb stance, yet no differences in the sensory reweighting system were displayed in the injured- or uninjured-limbs for CAI. In addition, CAI upweighted more on somatosensory feedback, then vestibular feedback, when the complexity of task constraints was smaller (i.e., double- and uninjured-limbs). Whereas the visual feedback was upweighted the most when the complexity of task constraints was greater (i.e., injured-limb). The group differences in postural control depended on the environmental and task constraints. Overall, CAI individuals maintained postural control very similar to healthy controls while completing the SOT in double- and single-limb stances.

CAI individuals are well-established to exhibit somatosensory deficits due to potential damage to the foot and ankle complex caused by an initial ankle sprain, resulting in postural control dysfunction (Freeman et al., 1965; Munn et al., 2010). Somatosensory deficits have been also reported in CAI by Song and Wikstrom (2020), who have tested postural control utilizing the SOT, similar to our study. Conversely, our CAI individuals in this study did not display somatosensory deficits and demonstrated similar postural control to healthy controls. Especially when there is greater contact surface area with a wide base of support while maintaining postural control in a double-limb stance, those individuals with CAI significantly upweighted somatosensory feedback more than vestibular and visual feedback. Previous studies that utilized total anesthesia to deprive somatosensory feedback at the foot and ankle complex demonstrated no active ankle joint position sense errors between healthy individuals with and without anesthetized conditions, suggesting an alteration in the central organization (Konradsen et al., 1993; Feuerbach et al., 1994). An altered central organization may influence descending commands to alpha and gamma motor neurons. For instance, the central organization modulates the sensitivity of the gamma-spindle system and discharges fast-adapting Type Ia sensory fibers instead of slow-adapting Type II sensory fibers, innervating alpha motor neurons in the spinal cord. Indeed, Vaes et al. (2001, 2002) did not find group differences in electromechanical delay, which is suggested to be an indirect indication of muscle stiffness and tone in individuals with and without CAI. In addition, altered muscle activities at proximal joints (knee, hip) while maintaining postural control have been reported in CAI (Pintsaar et al., 1996; Caulfield & Garrett, 2002; Gribble et al., 2007; Gribble & Robinson, 2009; Pope et al., 2011). Therefore, it is our hypothesis that the lack of somatosensory deficits observed in our CAI individuals is the result of the contribution of those central mechanisms.

The control of posture requires accurate peripheral sensory feedback from somatosensory, vestibular, and vision for individuals to initiate involuntary (reflexes) and voluntary muscle contractions to provide dynamic stability at the ankle (Horak et al., 1990). Under normal circumstances, healthy individuals obtain redundant and convergent sensory feedback to specify information about the ankle joint position relative to other body segments and the surrounding environment (Horak & Macpherson, 1996). While the exact mechanisms to how the CNS integrates multisensory information into a single coherent perception are still not clear, it has been suggested postural control strategies are context-dependent and determined by assigning relative weight on each sensory system based on organismic (e.g., health status), environmental, and task constraints (Horak & Nasher, 1986; Horak et al., 2006). For instance, the most reliable sensory feedback relative to a task goal is put more emphasis (i.e., upweighted), whereas the least reliable sensory feedback relative to a task goal is put less emphasis (i.e., downweighted) by the CNS. The general consensus is that healthy individuals upweight on Somatosensory feedback from mechanoreceptors in muscles, tendons, ligaments, joint capsules, and skin at the foot and ankle complex and from proximal joints to control postural oscillations on the fixed support surface (Peterka, 2002; Horak, 2006; Horak et al., 2017). Conversely, those individuals reweight on visual and/or vestibular feedback when they maintain posture on an unstable surface, which manipulates somatosensory feedback (Horak, 2006). Overall, the degree by which individuals distribute weight on each sensory feedback is not hard-wired and adaptively changes by constraints.

A systematic review with a meta-analysis concluded that individuals with CAI upregulate reliance on visual feedback to compensate for somatosensory deficits to maintain posture in the injured-limb compared to healthy controls (Song et al., 2016). Likewise, increased emphasis on visual feedback has been reported in individuals with somatosensory and vestibular deficits (Bronstein et al., 1990; Bonan et al., 2004;

Hafstrom et al., 2004; Lopez et al., 2006; Slaboda et al., 2009; Manor et al., 2010; Lin et al., 2019). In contrast, we did not identify group differences in visual reliance, which contradictory findings may be an effect of the absence of somatosensory impairments in CAI. Those participants with CAI displayed a significant upweight on vestibular feedback while maintaining posture in the injure-limb. Our study also revealed how individuals reweight sensory systems depend on the task and individual characteristics (health status). Several researchers who examined individuals with bilateral vestibular loss suggested vestibular information is unnecessary to maintain static postural orientation as long as somatosensory and visual feedback is available (Nashner et al., 1982; Horak et al., 1989, 1990; Shupert et al., 1989; Fitzpatrick et al., 1994). However, vestibular feedback provides independent information about body orientation in space, whereas somatosensory and visual feedback provides relative information (induced by the moving scene/surface) that fluctuates as a visual scene and supports surface change. Therefore, vestibular feedback is the sole source that provides veridical information about self-motion. Indeed, later studies concluded vestibular feedback served as a reference and was upweighted, especially when there was a conflict among feedback from other sensory systems (somatosensory, vision) (Mahboobin et al., 2009; DeAngelis and Angelaki, 2011). The NeuroCom SOT is a reliable computerized dynamic posturography and systematically manipulates individuals' sensory feedback in a combination of the sway-referenced support surface and visual surroundings with and without eliminating vision, creating sensory feedback conflicts (Nashner, 1982). Together, we contend that CAI individuals upweighted on vestibular feedback compared to healthy controls to provide greater precision in controlling postural sway when transitioning from simple to more complex tasks (e.g., double- to uninjured-limb, uninjured- to injured-limb).

We found interactions for the group with environmental and task constraints. Interestingly, CAI individuals revealed better postural control in the injured-limb during the SOT condition 5 compared to healthy controls. Moreover, better postural control was noted in double-limb than in the injured-limb or the uninjured-limb for those with CAI. The SOT condition 5 challenges somatosensory and visual feedback by manipulating a sway-referenced support surface with the elimination of vision to examine how individuals reweight on vestibular feedback. Unisensory integration, such as reliance on vestibular feedback, found in the injured-limb for the CAI group is thought to be a result of inadequate multisensory integration (Woollacott et al., 1986; Teasdale et al., 1991; Whipple et al., 1993; Woollacott, 1993). However, considering a significant emphasis on somatosensory feedback noted in the current study compared to other sensory feedback in CAI for double-limb stance regardless of SOT conditions, better postural control found in double-limb stance may be the result of efficient multisensory integration.

The main effect found for the environmental and task constraints may suggest the implementation of different postural control mechanisms reflected by the complexity of constraints in both groups. For the environment main effect, the same trend in balance scores of individual SOT conditions was exhibited for the injured- and uninjured-limbs. What was common among all stance limbs was that the best balance scores were observed when all sensory feedback is intact in the SOT condition 1, while worse balance scores were observed in ascending order of the SOT conditions 5 and 6. Both conditions

5 and 6 examined reliance on vestibular feedback by manipulating somatosensory feedback, yet vision was eliminated only in condition 5. Despite vestibular feedback providing veridical sensory information, performing the SOT may require more visual feedback compared to other tasks; thus the absence of visual feedback has been a factor contributing to the worst balance found in condition 5 than 6. Notably, emphasis on visual feedback remained constant while performing the SOT in the current and the most recent study, regardless of task constraints transitioning from double- to single-limb stances (Song & Wikstrom, 2020). For the task main effect, both the injured- and uninjured-limbs demonstrated the same trend in balance scores across all SOT conditions. Several researchers implied motor control at the supraspinal level provides greater precision of movement by minimizing and correcting reflexive oscillations at the ankle when transitioning into more complex tasks (Capaday & Stein, 1986, 1987; Katz & Pierrot-Deseilligny, 1999; Taube et al., 2008; Pinar et al., 2010; Thompson et al., 2016). Additionally, a change in reflexive controls from spinal to supraspinal mechanisms has been indicated in CAI individuals (Kim et al., 2012). Although our study did not examine the reflexive excitability of muscles in postural control, similar postural control displayed in the injured- and uninjured-limbs may be a result of greater precision provided at the supraspinal level that both groups implemented. Hence, investigating how visual and vestibular feedback contribute to reflexive muscular control at the supraspinal level may provide further insight into mechanisms of postural control in individuals with and without CAI.

One of the limitations of this study was that we only included healthy young adults who are physically active at the time of study participation. Therefore, the results may not be applicable for individuals who are outside the age group and physical conditions. Another limitation was that we did not consider the chronicity following the initial ankle sprain, including duration, recovery, and recurrence of ankle sprains developing into CAI. The duration of exposure to CAI and completion of rehabilitation programs may impact the sensory reweighting system and postural control with the increased complexity of the environmental and task constraints.

Conclusion

CAI individuals exhibited effective multisensory integration, maintaining posture very similar to healthy controls during the SOT while transitioning from a simple to a more complex environmental and task constraints. Group differences in sensory reweighting depended on the type of sensory systems and task constraints. CAI individuals demonstrated upweighted vestibular feedback while completing the SOT in the injured-limb compared to healthy controls. Moreover, how individuals with and without CAI control posture depended on both environmental and task constraints, displaying similar postural control in the injured- and uninjured-limbs. Environment- and task-dependent postural control contribute to motor behaviors throughout the lifespan. Therefore, taking a multisensory-feedback approach by recognizing when to increase environmental and task constraints may optimize rehabilitation intervention in CAI.

Table 3.1. Participant Demographics (Mean ± SD).

	Group		
	CAI	**Controls**	***P*-values**
N	22 (13 females, 9 males)	22 (13 females, 9 males)	-
Age (years)	26.09 ± 5.76	25.41 ± 5.92	0.84
Height (cm)	172.25 ± 9.76	169.70 ± 9.32	0.61
Weight (kg)	76.18 ± 14.91	71.98 ± 14.79	0.93
NASA-PASS	6.27 ± 0.18	6.27 ± 1.03	0.67
IdFAI	19.09 ± 5.39	1.36 ± 1.81	< 0.001*
# of Ankle of Sprains	6.48 ± 7.08	0.00 ± 0.00	< 0.001*
Episodes of Giving-Way	8.88 ± 21.36	0.00 ± 0.00	< 0.001*

N = Number; NASA-PASS = National Aeronautics and Space Administration Physical Activity Status Scale; IdFAI = the Identification of Functional Ankle Instability; CAI = Chronic Ankle Instability.
* indicates significant group differences (P < 0.05).

Table 3.2. Group × Sensory Systems × Task Interaction for Sensory Reweighting Ratios.

Sensory Reweighting Ratios (Mean ± SD) and Effect Sizes				
Parameter		Group		
Task	**Sensory Systems**	**CAI**	**Control**	**Effect Sizes (95% CI)**
Double	SOM	97.22 ± 2.18 [1]	97.36 ± 2.82	0.06 (-0.54 to 0.30)
	VIS	93.21 ± 5.21 [2]‡	92.87 ± 3.60	-0.08 (-0.67 to 0.30)
	VEST	71.40 ± 9.86 [3]‡	73.30 ± 8.62	0.21 (-0.39 to 0.30)
Injured	SOM	81.78 ± 4.84 [1]	78.88 ± 7.82	-0.45 (-1.04 to 0.31)
	VIS	94.83 ± 4.19 [2]	94.95 ± 4.56	0.03 (-0.56 to 0.30)
	VEST	69.91 ± 7.57 [3]	63.15 ± 11.56 †	-0.69 (-1.30 to 0.31)
Uninjured	SOM	81.82 ± 6.77 [1]	80.25 ± 6.23	-0.24 (-0.83 to 0.30)
	VIS	94.41 ± 3.50 [2]	90.04 ± 5.00	-0.09 (-0.68 to 0.30)
	VEST	70.13 ± 9.56 [3]	65.01 ± 8.23	-0.57 (-1.18 to 0.31)

SD = Standard Deviation; SOM = Somatosensory; VIS = Vision; VEST = Vestibular; CAI = Chronic Ankle Instability; CI = Confidence Interval.
† indicates significant group differences ($P < 0.05$).
[1] indicates significant differences between double- and the injured-limbs, double- and the uninjured-limbs, and the injured- and uninjured-limbs on somatosensory for the CAI group ($P < 0.05$).
[2] indicates significant differences between double- and the injured-limbs, double- and the uninjured-limbs, and the injured- and uninjured-limbs on vision for the CAI group ($P < 0.05$).
[3] indicates significant differences between double- and the injured-limbs, double- and the uninjured-limbs, and the injured- and uninjured-limbs on vestibular for the CAI group ($P < 0.05$).
‡ indicates significant differences between SOM and VIS, and SOM and VEST in a double-limb for the CAI group ($P < 0.05$).

Table 3.3. Group × Environment Interactions for Equilibrium Scores for the Injured-Limb.

Injured-Limb: Equilibrium (EQ) Scores (Mean ± SD) and Effect Sizes			
Parameter	**CAI**	**Control**	**Effect Sizes (95% CI)**
SOT C1	89.26 ± 2.55	89.03 ± 2.46	-0.09 (-0.68 to 0.30)
SOT C2	72.98 ± 4.30	70.26 ± 7.52	-0.44 (-1.04 to 0.31)
SOT C3	80.90 ± 4.24	80.22 ± 6.62	-0.12 (-0.71 to 0.30)
SOT C4	84.63 ± 4.15	84.54 ± 4.56	-0.02 (-0.61 to 0.30)
SOT C5	62.45 ± 7.38	56.26 ± 10.70†	-0.67 (-1.28 to 0.31)
SOT C6	67.86 ± 8.85	68.34 ± 7.73	0.06 (-0.53 to 0.30)

SD = Standard Deviation; SOM = Somatosensory; SOT = Sensory Organization Test; CAI = Chronic Ankle Instability; CI = Confidence Interval.
† indicates significant group differences ($P < 0.05$).

Table 3.4. Group × Environment Interactions for Equilibrium Scores for a Double-Limb.

Double-Limb: Equilibrium (EQ) Scores (Mean ± SD) and Effect Sizes			
Parameter	**CAI**	**Control**	**Effect Sizes (95% CI)**
SOT C1	94.68 ± 1.54	94.81 ± 1.29	0.09 (-0.50 to 0.30)
SOT C2	92.04 ± 2.41	92.30 ± 2.97	0.10 (-0.49 to 0.30)
SOT C3	91.65 ± 3.42	92.69 ± 2.62	0.34 (-0.25 to 0.30)
SOT C4	88.24 ± 5.24	88.04 ± 3.68	-0.04 (-0.63 to 0.30)
SOT C5	67.63 ± 9.64	69.50 ± 8.27	0.21 (-0.39 to 0.30)
SOT C6	71.27 ± 8.22	73.41 ± 10.34	0.23 (-0.36 to 0.30)

SD = Standard Deviation; SOM = Somatosensory; SOT = Sensory Organization Test; CAI = Chronic Ankle Instability; CI = Confidence Interval.

Table 3.5. Group × Environment Interactions for Equilibrium Scores for the Uninjured-Limb.

Uninjured-Limb: Equilibrium (EQ) Scores (Mean ± SD) and Effect Sizes			
Parameter	**CAI**	**Control**	**Effect Sizes (95% CI)**
SOT C1	88.87 ± 3.19	89.10 ± 2.42	0.08 (-0.51 to 0.30)
SOT C2	72.66 ± 5.96	71.52 ± 6.15	-0.19 (-0.78 to 0.30)
SOT C3	81.60 ± 4.48	80.27 ± 5.47	-0.27 (-0.86 to 0.30)
SOT C4	83.91 ± 4.41	83.78 ± 4.74	-0.03 (-0.62 to 0.30)
SOT C5	62.41 ± 9.44	57.94 ± 7.69	-0.52 (-1.12 to 0.31)
SOT C6	71.64 ± 8.57	67.33 ± 8.42	-0.51 (-1.11 to 0.31)

SD = Standard Deviation; SOM = Somatosensory; SOT = Sensory Organization Test; CAI = Chronic Ankle Instability; CI = Confidence Interval.

Table 3.6. Group × Task Interactions for Equilibrium Scores for the SOT condition 1.

SOT Condition 1: Equilibrium (EQ) Scores (Mean ± SD) and Effect Sizes			
Parameter	**CAI**	**Control**	**Effect Sizes (95% CI)**
Double	94.68 ± 1.54	94.81 ± 1.29	0.09 (-0.50 to 0.30)
Injured	89.26 ± 2.55	89.03 ± 2.46	-0.09 (-0.68 to 0.30)
Uninjured	88.87 ± 3.19	89.10 ± 2.42	0.08 (-0.51 to 0.30)

SD = Standard Deviation; SOM = Somatosensory; SOT = Sensory Organization Test; CAI = Chronic Ankle Instability; CI = Confidence Interval.

Table 3.7. Group × Task Interactions for Equilibrium Scores for the SOT condition 2.

SOT Condition 2: Equilibrium (EQ) Scores (Mean ± SD) and Effect Sizes			
Parameter	**CAI**	**Control**	**Effect Sizes (95% CI)**
Double	92.04 ± 2.41	92.30 ± 2.97	0.10 (-0.49 to 0.30)
Injured	72.88 ± 4.30	70.26 ± 7.52	-0.44 (-1.04 to 0.31)
Uninjured	72.66 ± 5.96	71.52 ± 6.15	-0.19 (-0.78 to -0.30)

SD = Standard Deviation; SOM = Somatosensory; SOT = Sensory Organization Test; CAI = Chronic Ankle Instability; CI = Confidence Interval.

Table 3.8. Group × Task Interactions for Equilibrium Scores for the SOT condition 3.

SOT Condition 3: Equilibrium (EQ) Scores (Mean ± SD) and Effect Sizes			
Parameter	**CAI**	**Control**	**Effect Sizes (95% CI)**
Double	91.65 ± 3.42	92.69 ± 2.62	0.34 (-0.25 to 0.30)
Injured	80.90 ± 4.24	80.22 ± 6.62	-0.12 (-0.71 to 0.30)
Uninjured	81.60 ± 4.48	80.27 ± 5.47	-0.27 (-0.86 to 0.30)

SD = Standard Deviation; SOM = Somatosensory; SOT = Sensory Organization Test; CAI = Chronic Ankle Instability; CI = Confidence Interval.

Table 3.9. Group × Task Interactions for Equilibrium Scores for the SOT condition 4.

SOT Condition 4: Equilibrium (EQ) Scores (Mean ± SD) and Effect Sizes			
Parameter	**CAI**	**Control**	**Effect Sizes (95% CI)**
Double	88.24 ± 5.24	88.04 ± 3.68	-0.04 (-0.63 to 0.30)
Injured	84.63 ± 4.15	84.54 ± 4.56	-0.02 (-0.61 to 0.30)
Uninjured	83.91 ± 4.41	83.78 ± 4.74	-0.03 (-0.62 to 0.30)

SD = Standard Deviation; SOM = Somatosensory; SOT = Sensory Organization Test; CAI = Chronic Ankle Instability; CI = Confidence Interval.

Table 3.10. Group × Task Interactions for Equilibrium Scores for the SOT condition 6.

SOT Condition 6: Equilibrium (EQ) Scores (Mean ± SD) and Effect Sizes			
Parameter	**CAI**	**Control**	**Effect Sizes (95% CI)**
Double	71.27 ± 8.22	73.41 ± 10.34	0.23 (-0.36 to 0.30)
Injured	67.86 ± 8.85	68.34 ± 7.73	0.06 (-0.53 to 0.30)
Uninjured	71.64 ± 8.57	67.33 ± 8.42	-0.51 (-1.11 to 0.31)

SD = Standard Deviation; SOM = Somatosensory; SOT = Sensory Organization Test; CAI = Chronic Ankle Instability; CI = Confidence Interval.

Table 3.11. Group × Task Interactions for Equilibrium Scores for the SOT condition 5.

SOT Condition 5: Equilibrium (EQ) Scores (Mean ± SD) and Effect Sizes			
Parameter	**CAI**	**Control**	**Effect Sizes (95% CI)**
Double	67.63 ± 9.64	69.50 ± 8.27	0.21 (-0.39 to 0.30)
Injured	62.45 ± 7.38 ‡	56.26 ± 10.70 †	-0.67 (-1.28 to 0.31)
Uninjured	62.41 ± 9.44 ‡	57.94 ± 7.69	-0.52 (-1.12 to 0.31)

SD = Standard Deviation; SOM = Somatosensory; SOT = Sensory Organization Test; CAI = Chronic Ankle Instability; CI = Confidence Interval.
† indicates significant group differences ($P < 0.05$).
‡ indicates significant differences between double- and the injured-limbs, and double- and the uninjured-limbs for the CAI group ($P < 0.05$).

Figure 3.1. The NeuroCom Dynamic Posturography System.

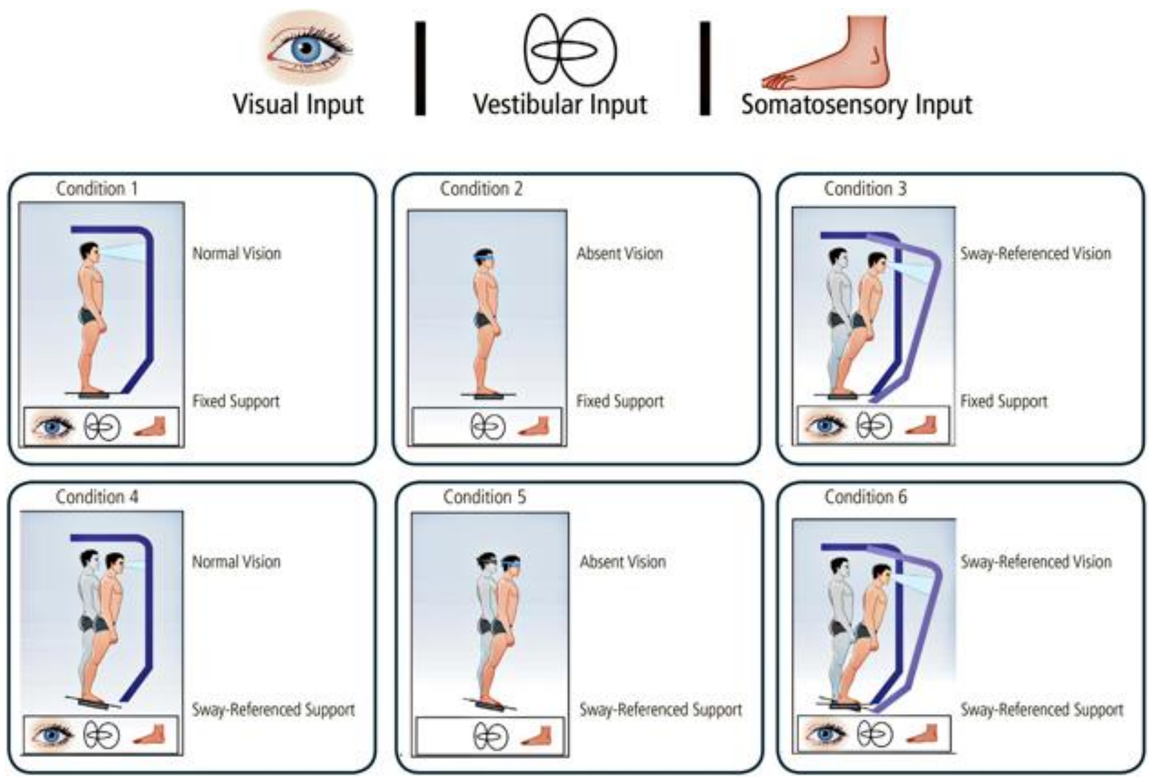

Figure 3.2. Six Conditions of the SOT (NeuroCom® International).

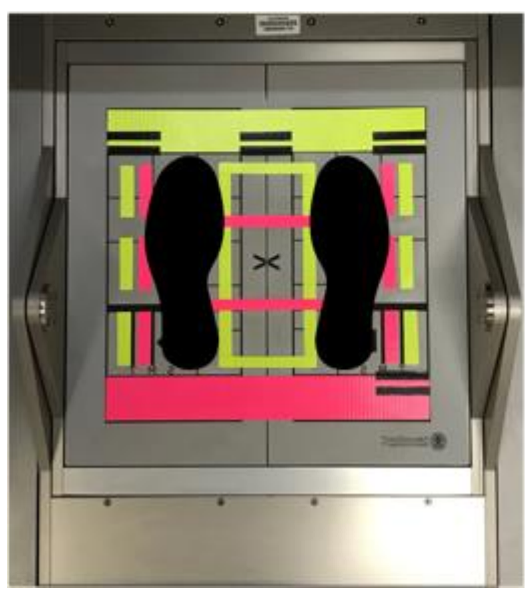

Figure 3.3. Foot Position for the SOT in a Double-Limb Stance.

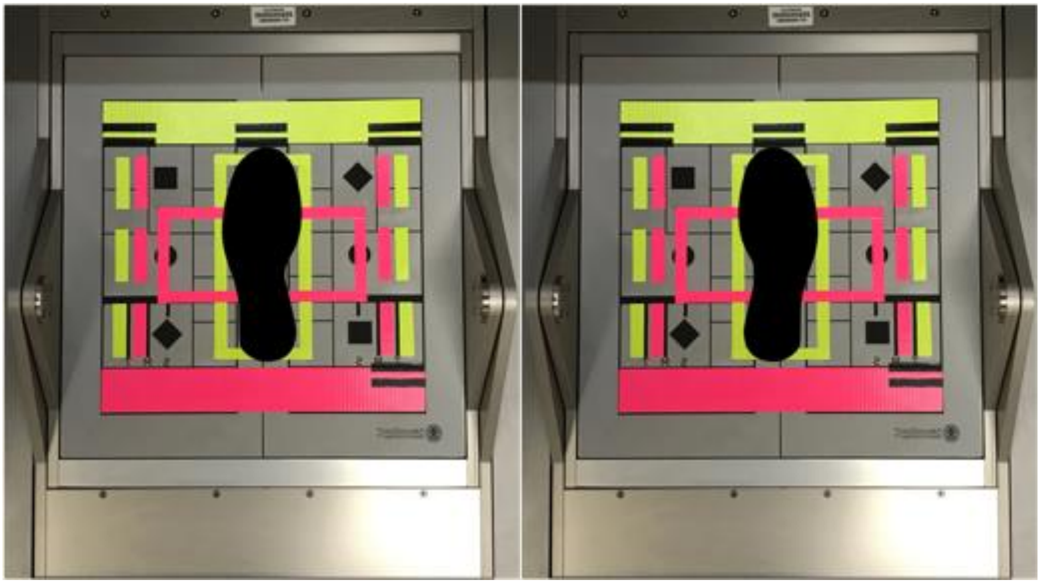

Figure 3.4. Foot Position for the SOT in a Single-Limb Stance (Injured, Uninjured).

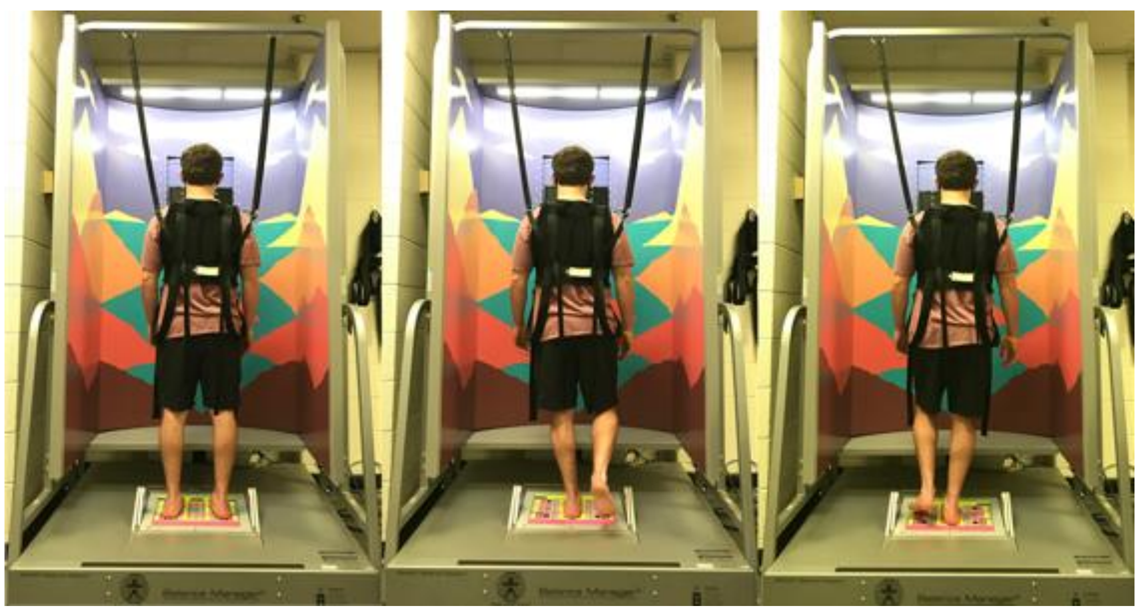

Figure 3.5. Stance Positions for the SOT in Double- and Single-Limb Stances on the NeuroCom.

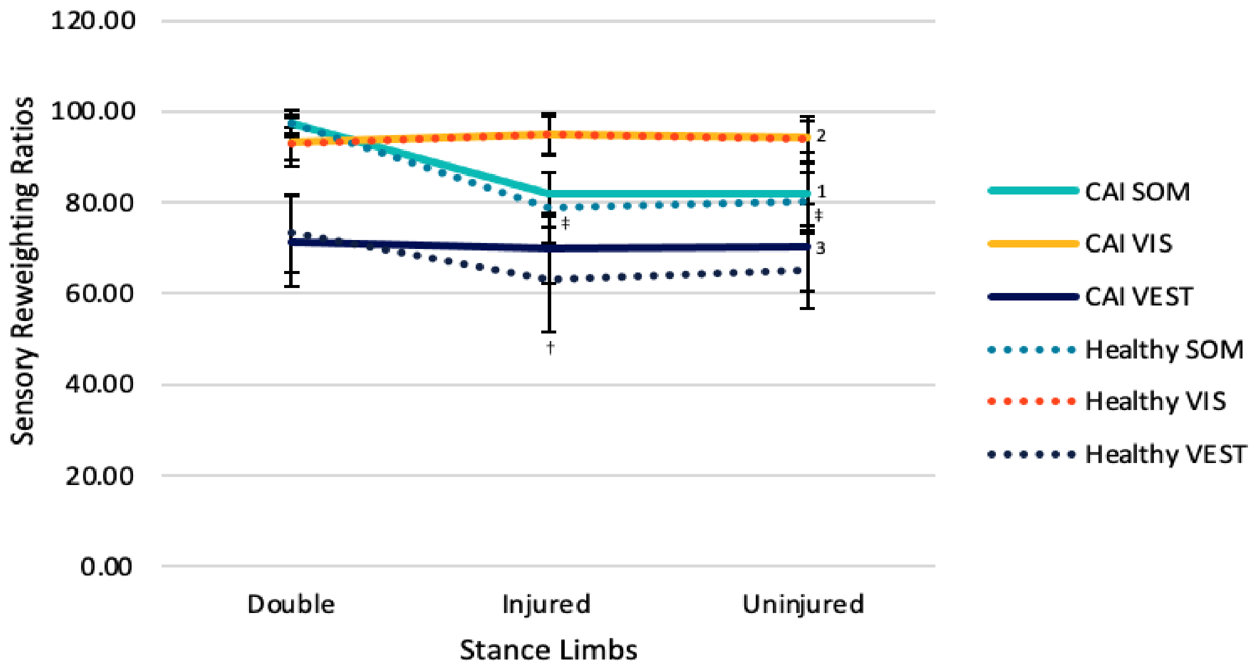

Figure 3.6. Group × Sensory Systems × Task Interaction for Sensory Reweighting Systems.
CAI = Chronic Ankle Instability; SOM = Somatosensory; VIS = Vision; VEST = Vestibular.
† indicates significant group differences ($P < 0.05$).
[1] indicates significant differences between double- and the injured-limbs, double- and the uninjured-limbs, and the injured- and uninjured-limbs on somatosensory for the CAI group ($P < 0.05$).
[2] indicates significant differences between double- and the injured-limbs, double- and the uninjured-limbs, and the injured- and uninjured-limbs on vision for the CAI group ($P < 0.05$).
[3] indicates significant differences between double- and the injured-limbs, double- and the uninjured-limbs, and the injured- and uninjured-limbs on vestibular for the CAI group ($P < 0.05$).
‡ indicates significant differences between SOM and VIS, and SOM and VEST in a double-limb for the CAI group ($P < 0.05$).

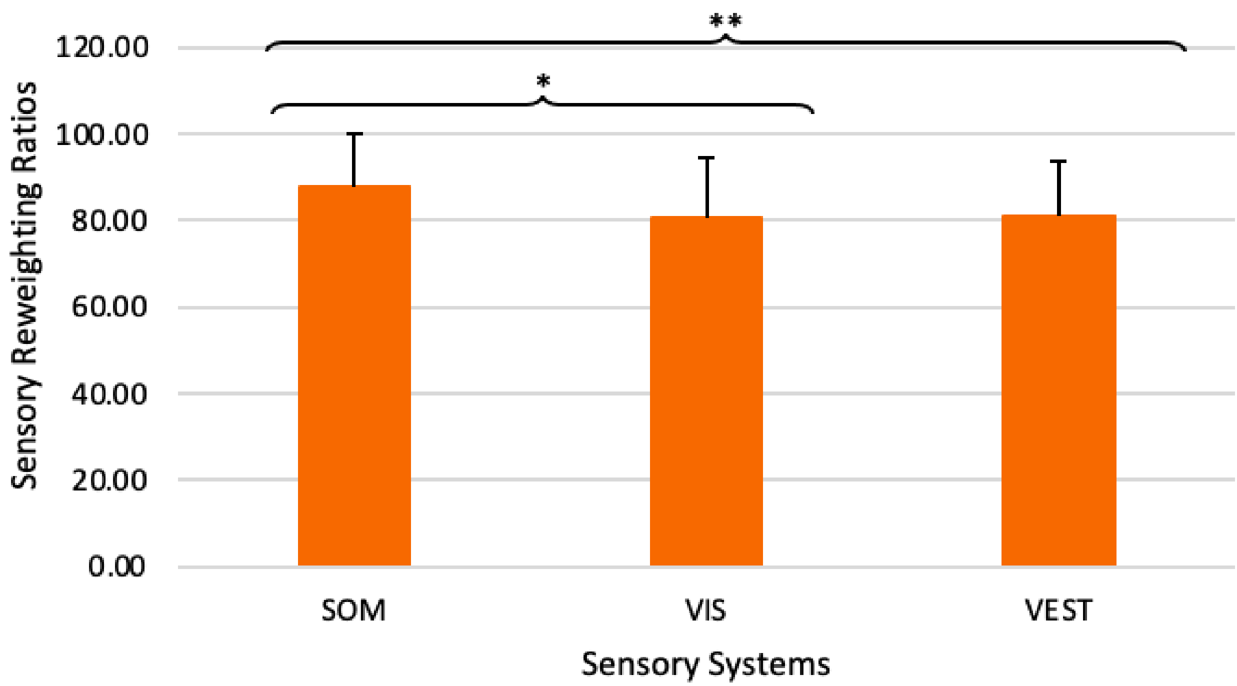

Figure 3.7. Sensory Systems Main Effect for Sensory Reweighting Systems.
SOM = Somatosensory; VIS = Vision; VEST = Vestibular.
* indicates significant differences between SOM and VIS (P < 0.05).
** indicates significant differences between SOM and VEST (P < 0.05).

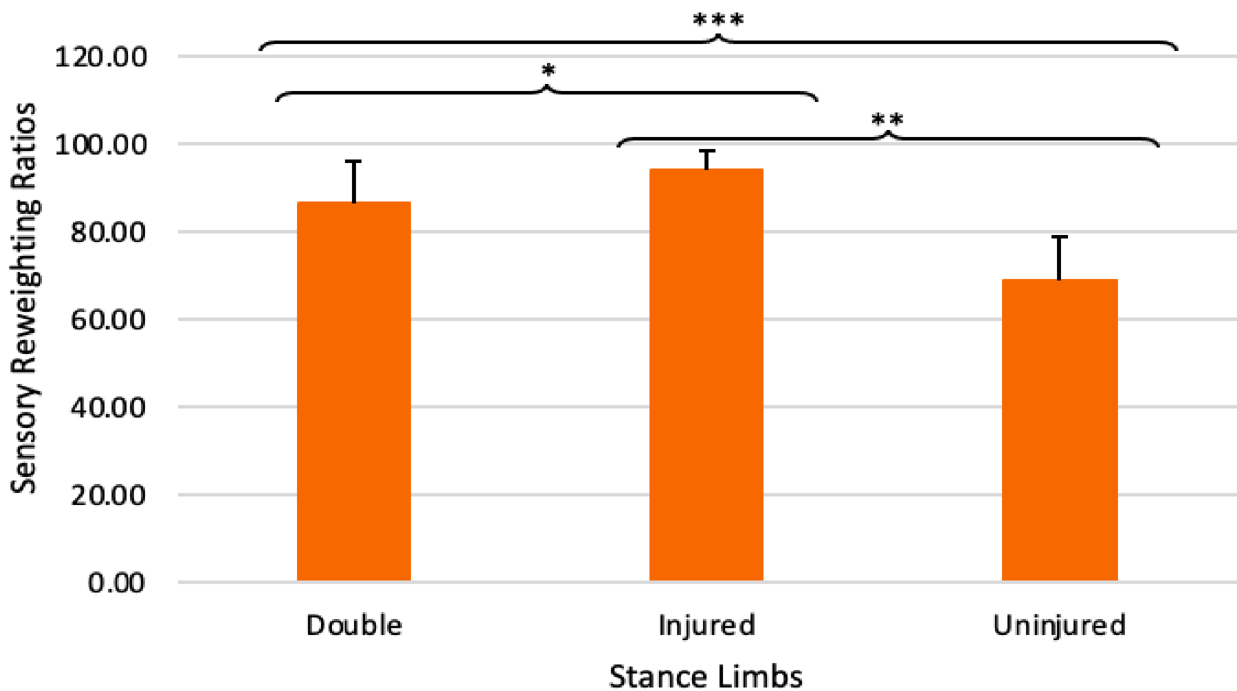

Figure 3.8. Task Main Effect for Sensory Reweighting Systems.
* indicates significant differences between double- and the injured-limbs ($P < 0.05$).
** indicates significant differences between the injured- and uninjured-limbs ($P < 0.05$).
*** indicates significant differences between double- and the uninjured-limbs ($P < 0.05$).

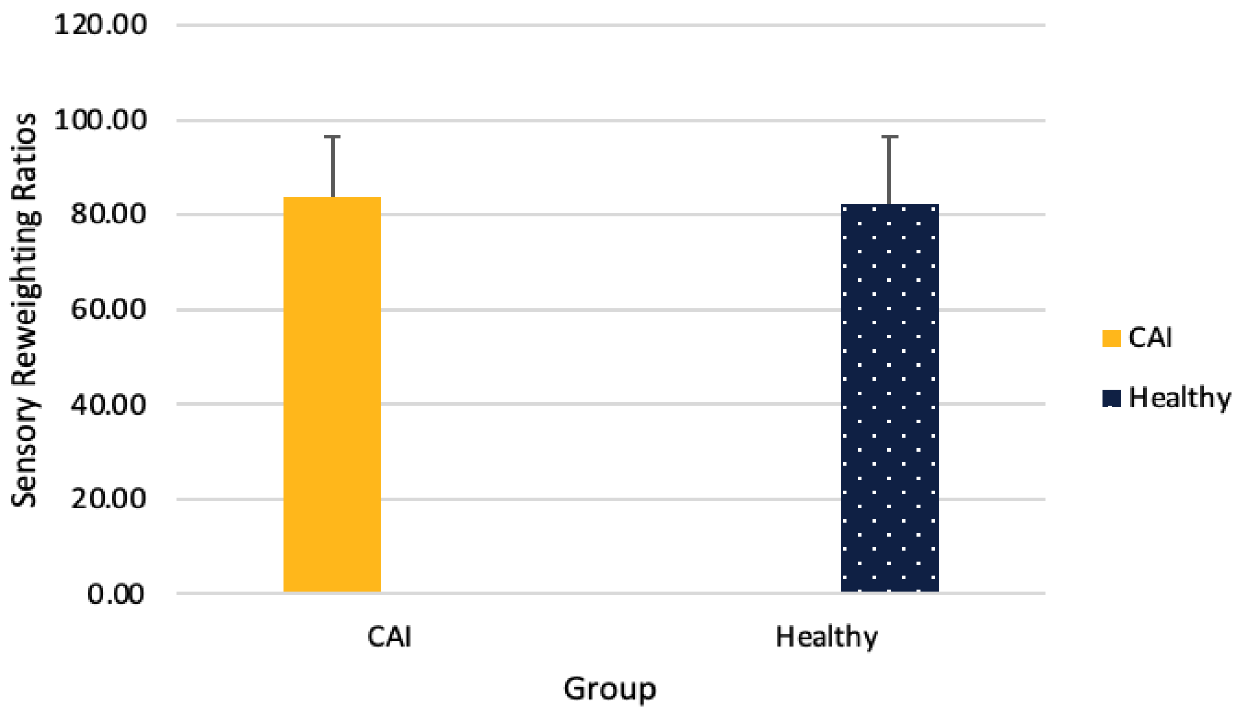

Figure 3.9. Group Main Effect for Sensory Reweighting Systems.

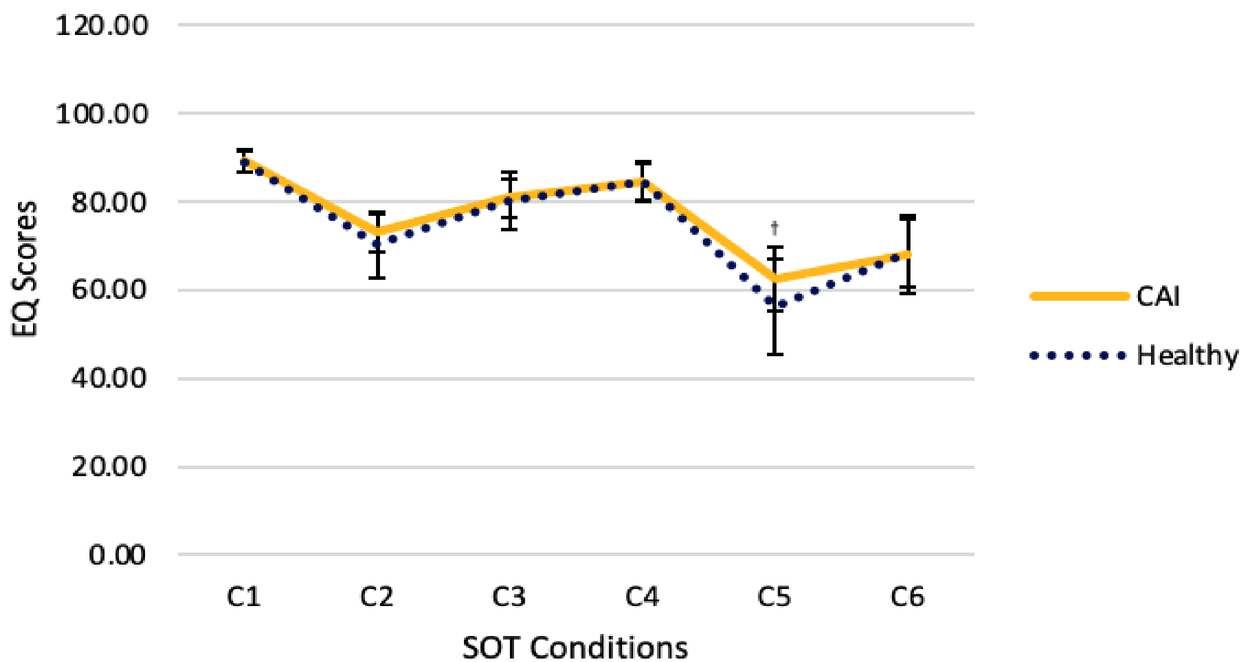

Figure 3.10. Group × Environment Interactions for Equilibrium Scores for the Injured-Limb.
EQ = Equilibrium; C = Condition; SOT = Sensory Organization Test; CAI = Chronic Ankle Instability.
† indicates significant group differences (P < 0.05).

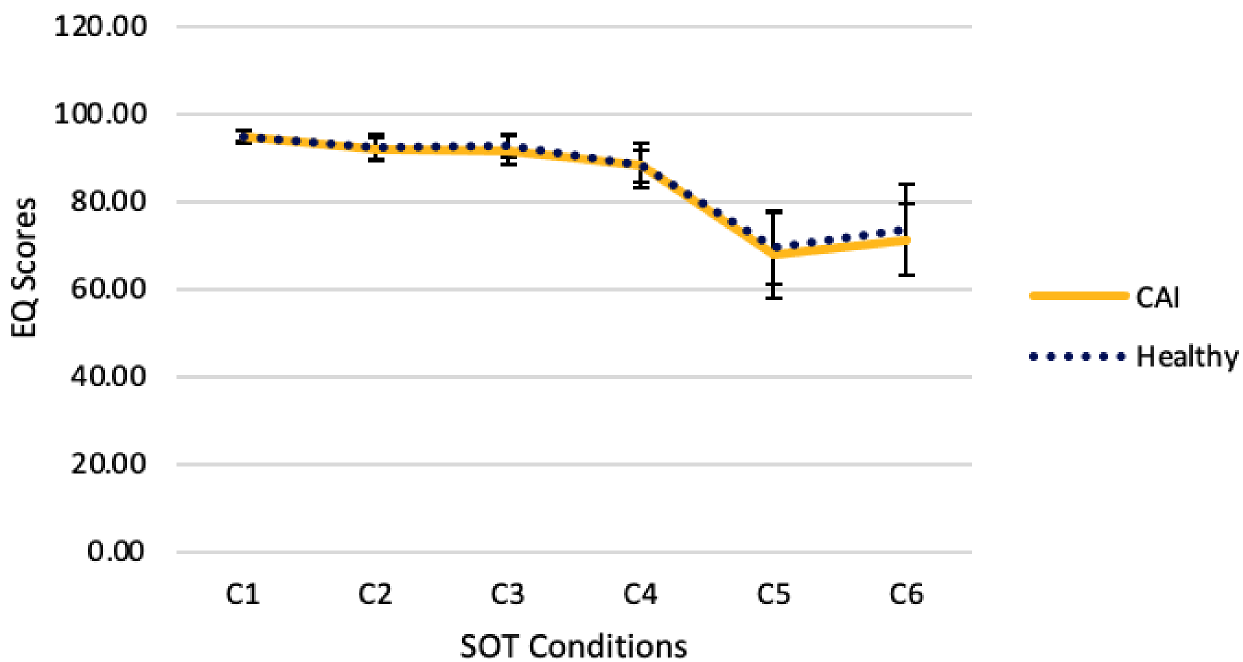

Figure 3.11. Group × Environment Interactions for Equilibrium Scores for a Double-Limb.
EQ = Equilibrium; C = Condition; SOT = Sensory Organization Test; CAI = Chronic Ankle Instability.

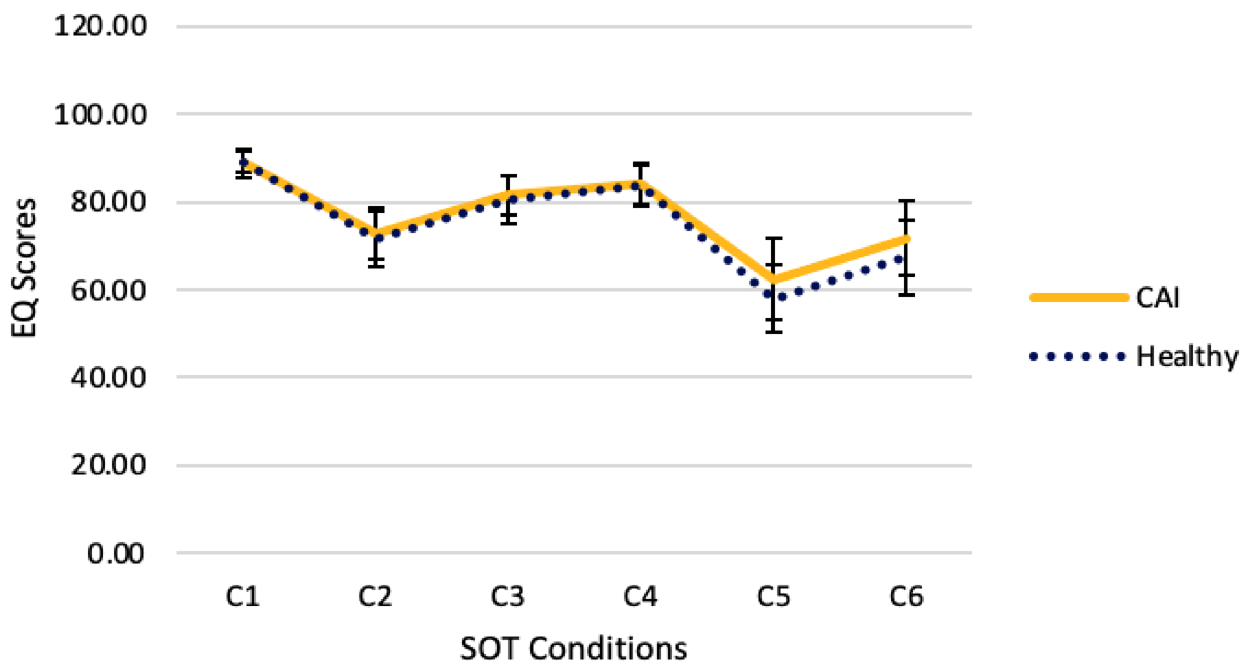

Figure 3.12. Group × Environment Interactions for Equilibrium Scores for the Uninjured-Limb.
EQ = Equilibrium; C = Condition; SOT = Sensory Organization Test; CAI = Chronic Ankle Instability.

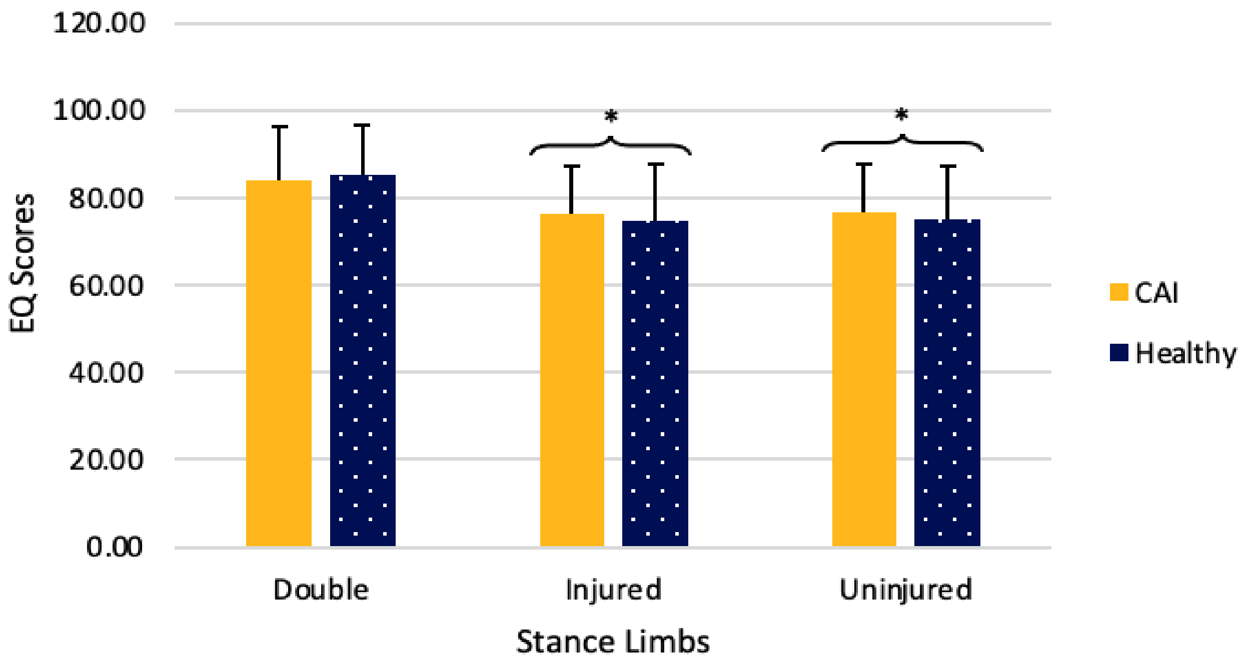

Figure 3.13. Group Main Effect for Equilibrium Scores for Individual Stance Limbs.
EQ = Equilibrium; CAI = Chronic Ankle Instability.
* indicates significant differences between groups in the injured-limb (P < 0.05).
** indicates significant differences between groups in the uninjured-limb (P < 0.05).

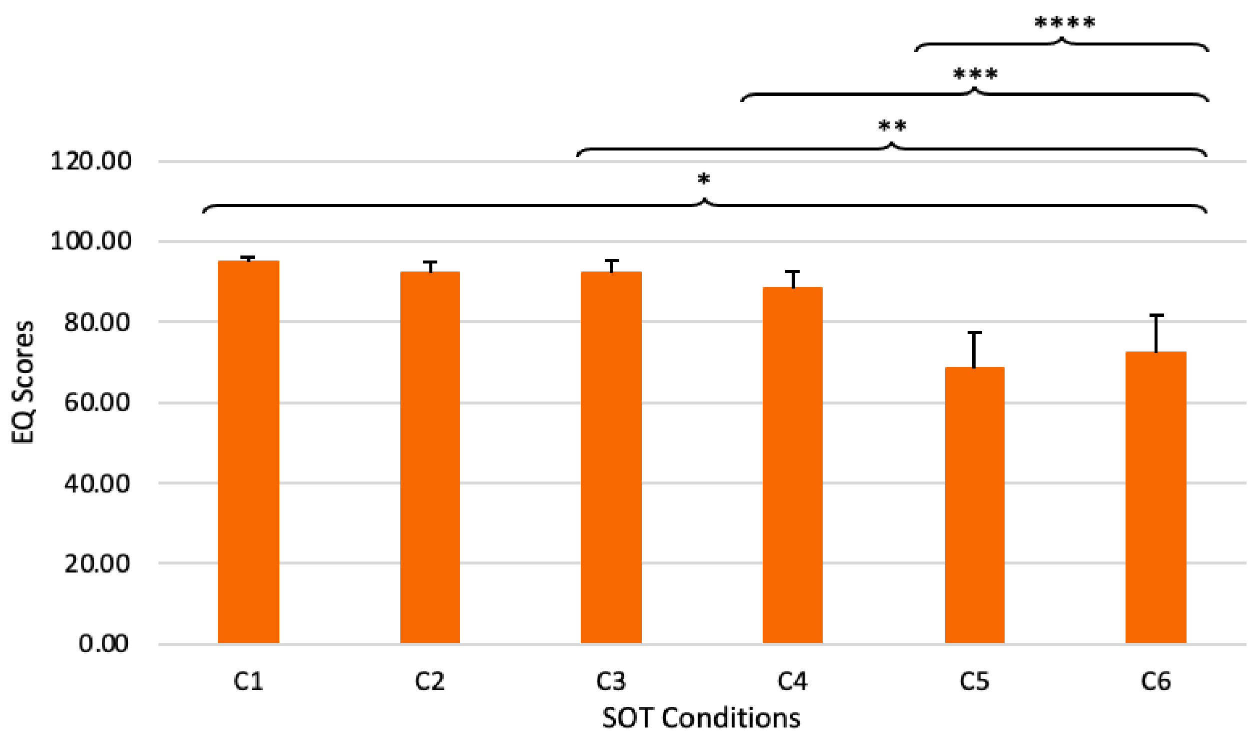

Figure 3.14. Environment Main Effect for Equilibrium Scores for a Double-Limb Stance.

EQ = Equilibrium; CAI = Chronic Ankle Instability.

* indicates significant differences between C1 & C2, C1 & C3, C1 & C4, C1 & C5, C1 & C6 ($P < 0.05$).

** indicates significant differences between C3 & C4, C3 & C5, C3 & C6 ($P < 0.05$).

*** indicates significant differences between C4 & C5, C4 & C6 ($P < 0.05$).

**** indicates significant differences between C5 & C6 ($P < 0.05$).

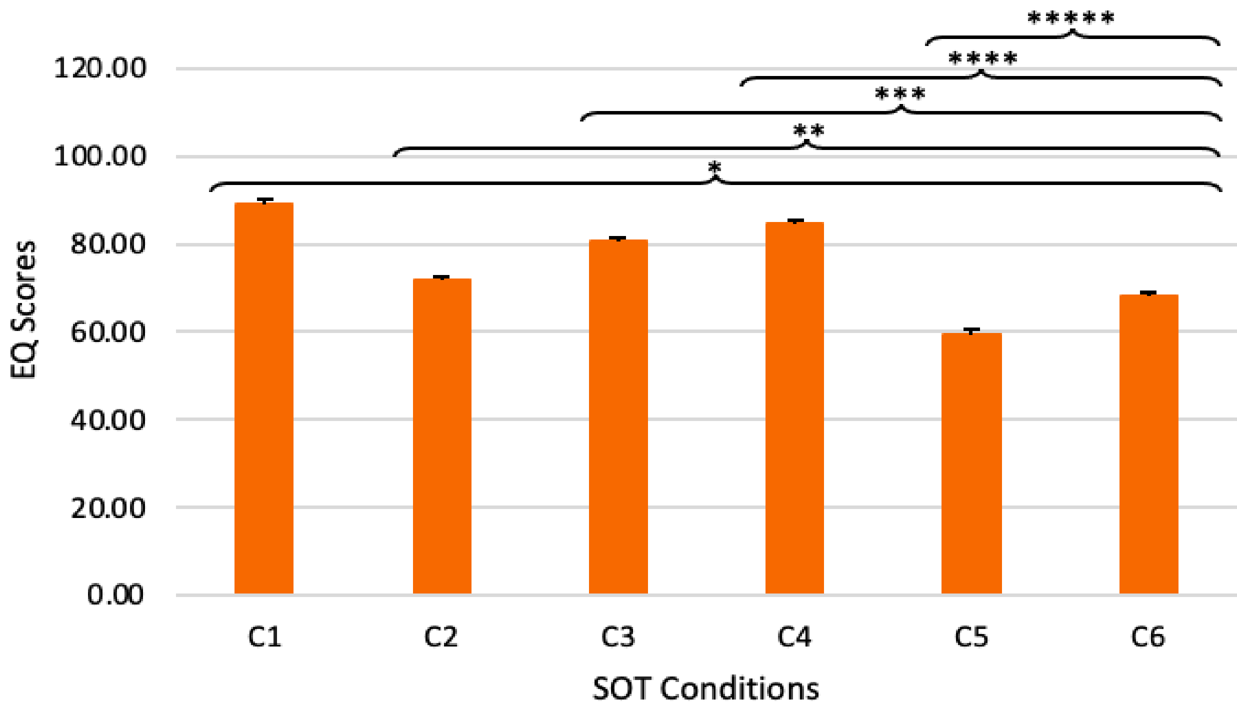

Figure 3.15. Environment Main Effect for Equilibrium Scores for the Injured-Limb.
EQ = Equilibrium; C = Condition; SOT = Sensory Organization Test; CAI = Chronic Ankle Instability.
* indicates significant differences between C1 & C2, C1 & C3, C1 & C4, C1 & C5, C1 & C6 ($P < 0.05$).
** indicates significant differences between C2 & C4, C2 & C5, C2 & C6 ($P < 0.05$).
*** indicates significant differences between C3 & C4, C3 & C5, C3 & C6 ($P < 0.05$).
**** indicates significant differences between C4 & C5, C4 & C6 ($P < 0.05$).
***** indicates significant differences between C5 & C6 ($P < 0.05$).

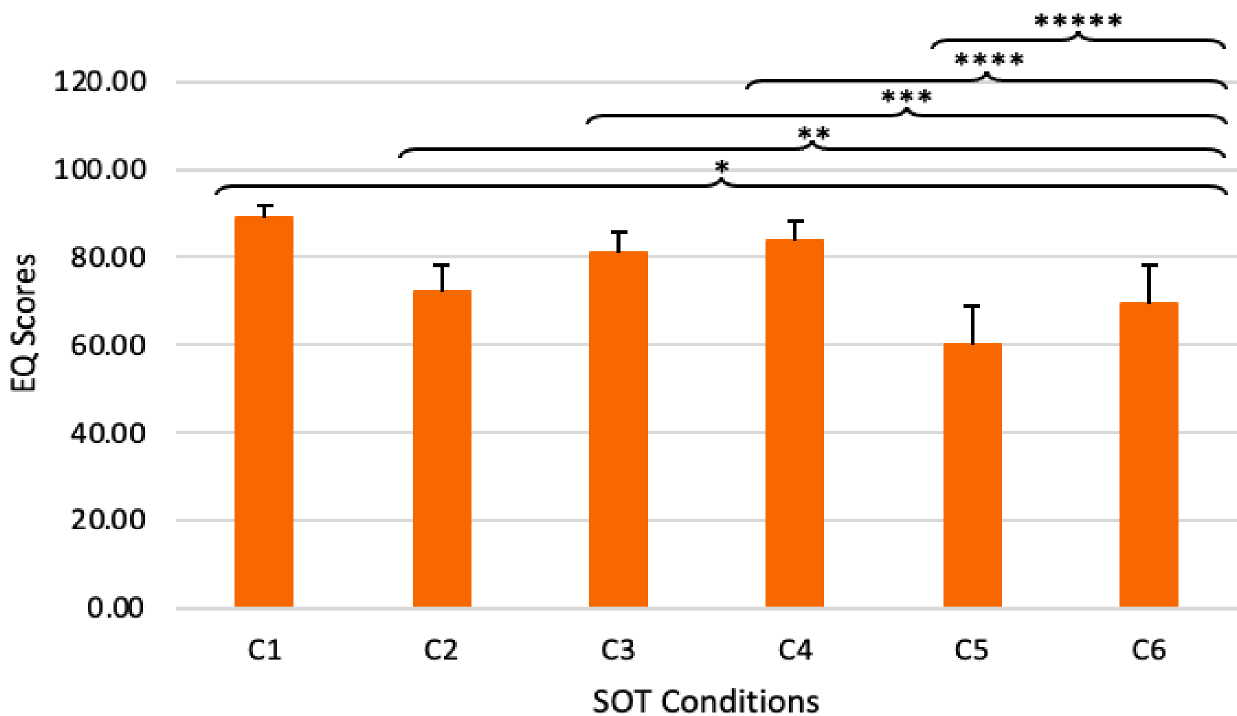

Figure 3.16. Environment Main Effect for Equilibrium Scores for the Uninjured-Limb.
EQ = Equilibrium; C = Condition; SOT = Sensory Organization Test; CAI = Chronic Ankle Instability.
* indicates significant differences between C1 & C2, C1 & C3, C1 & C4, C1 & C5, C1 & C6 ($P < 0.05$).
** indicates significant differences between C2 & C4, C2 & C5, C2 & C6 ($P < 0.05$).
*** indicates significant differences between C3 & C4, C3 & C5, C3 & C6 ($P < 0.05$).
**** indicates significant differences between C4 & C5, C4 & C6 ($P < 0.05$).
***** indicates significant differences between C5 & C6 ($P < 0.05$).

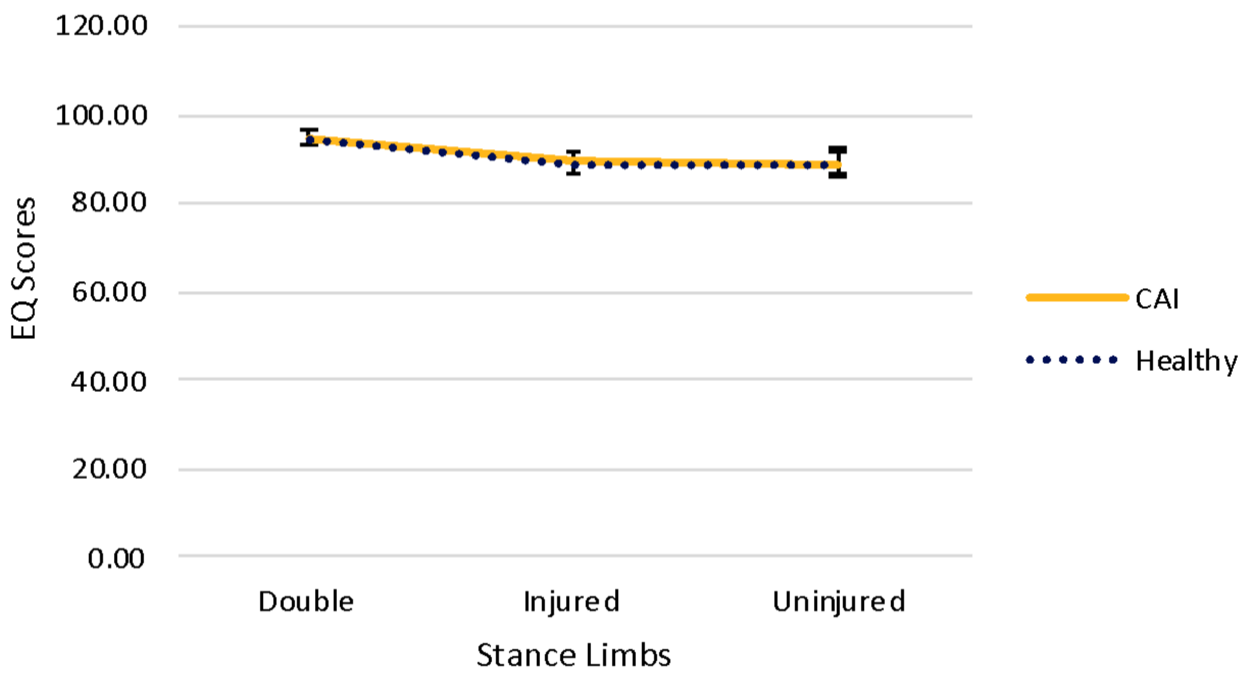

Figure 3.17. Group × Task Interactions for Equilibrium Scores for Individual Stance Limbs for the SOT Condition 1. EQ = Equilibrium; CAI = Chronic Ankle Instability.

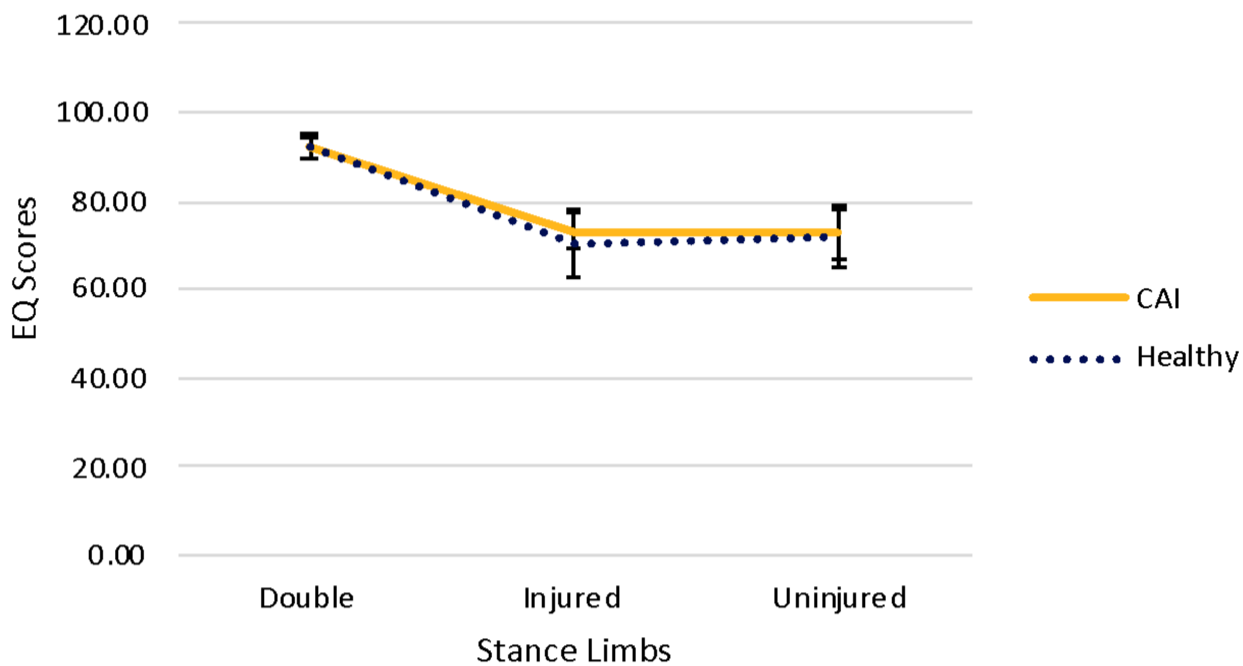

Figure 3.18. Group × Task Interactions for Equilibrium Scores for Individual Stance Limbs for the SOT Condition 2. EQ = Equilibrium; CAI = Chronic Ankle Instability.

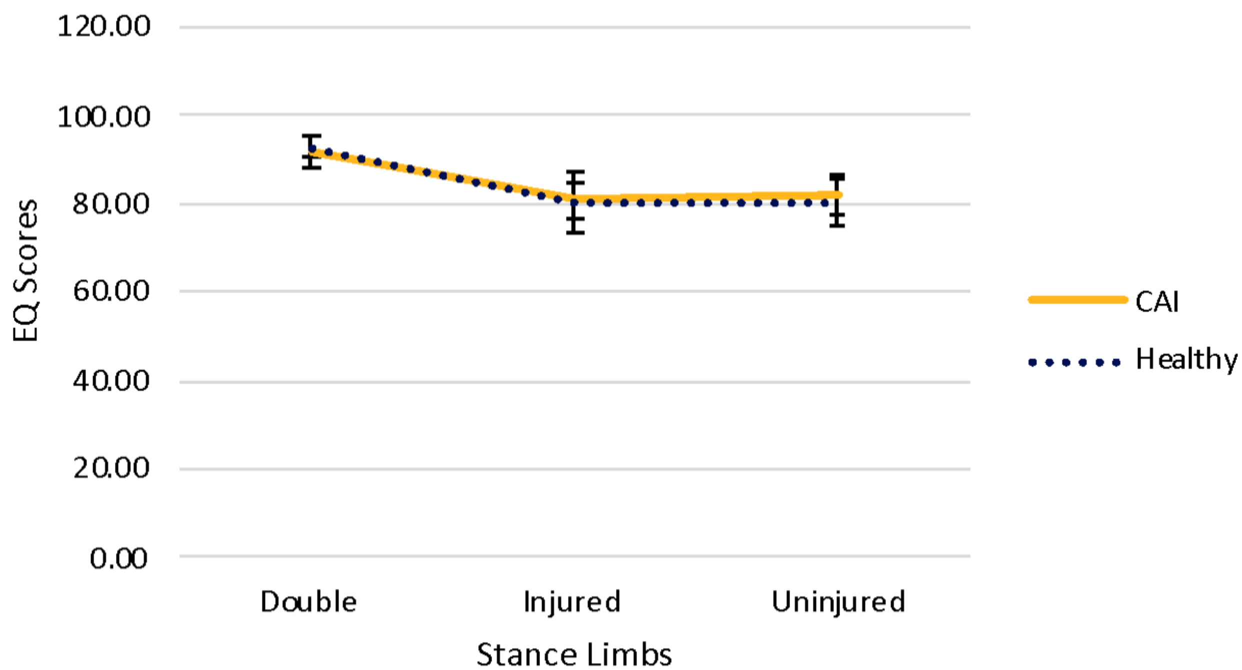

Figure 3.19. Group × Task Interactions for Equilibrium Scores for Individual Stance Limbs for the SOT Condition 3. EQ = Equilibrium; CAI = Chronic Ankle Instability.

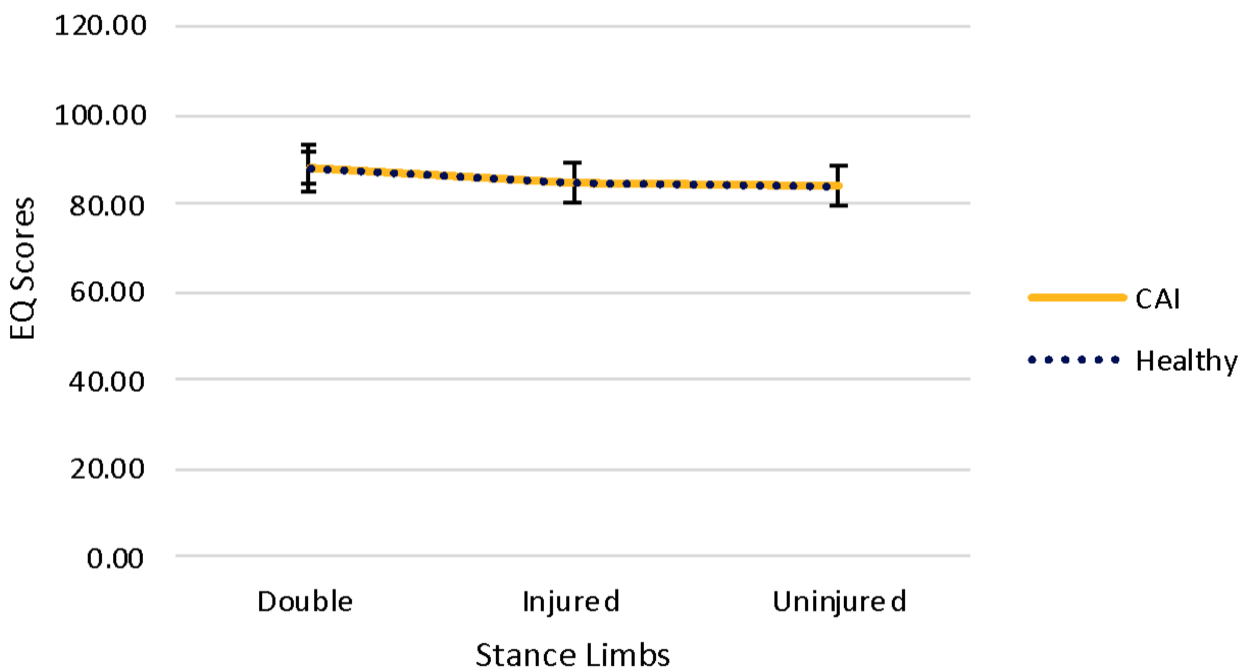

Figure 3.20. Group × Task Interactions for Equilibrium Scores for Individual Stance Limbs for the SOT Condition 4. EQ = Equilibrium; CAI = Chronic Ankle Instability.

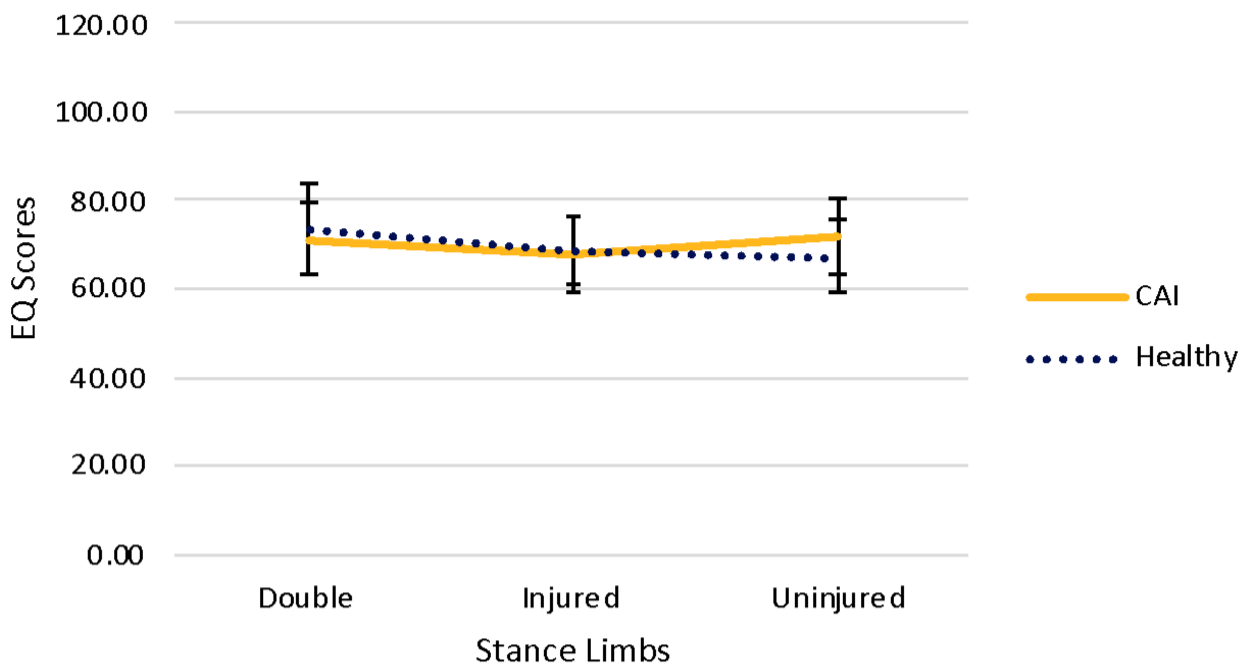

Figure 3.21. Group × Task Interactions for Equilibrium Scores for Individual Stance Limbs for the SOT Condition 6.
EQ = Equilibrium; CAI = Chronic Ankle Instability.

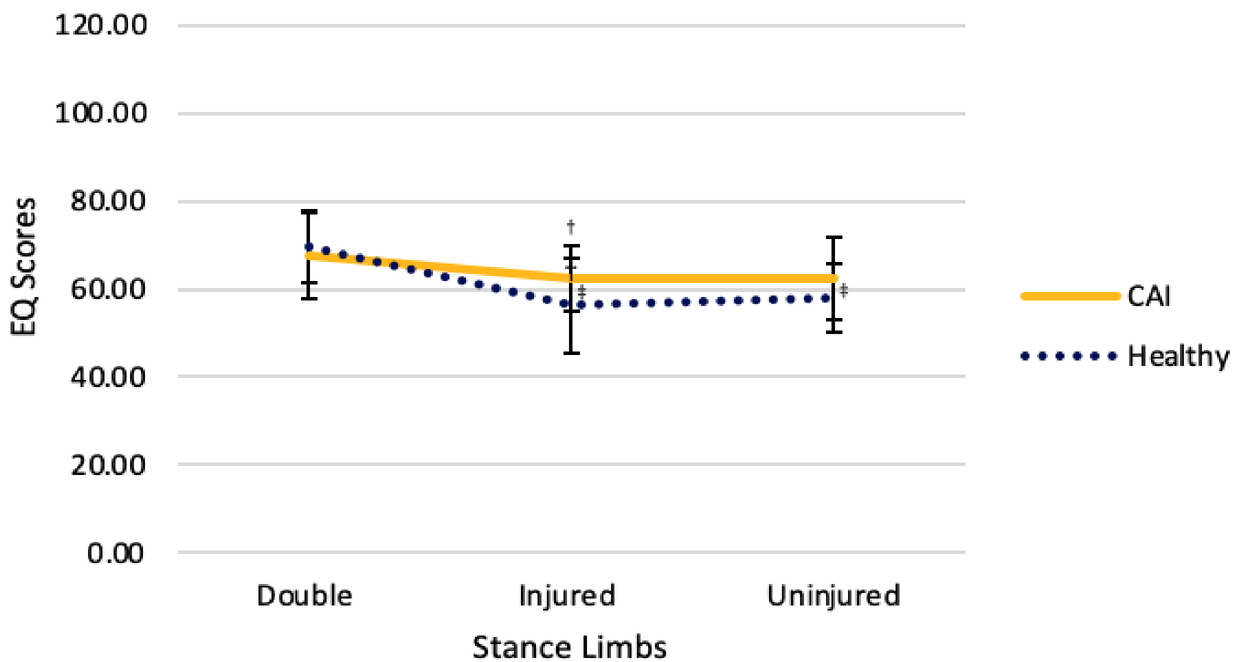

Figure 3.22. Group × Task Interactions for Equilibrium Scores for Individual Stance Limbs for the SOT Condition 5.
EQ = Equilibrium; CAI = Chronic Ankle Instability.
† indicates significant group differences ($P < 0.05$).
‡ indicates significant differences between double- and the injured-limbs, and double- and the uninjured-limbs for the CAI group ($P < 0.05$).

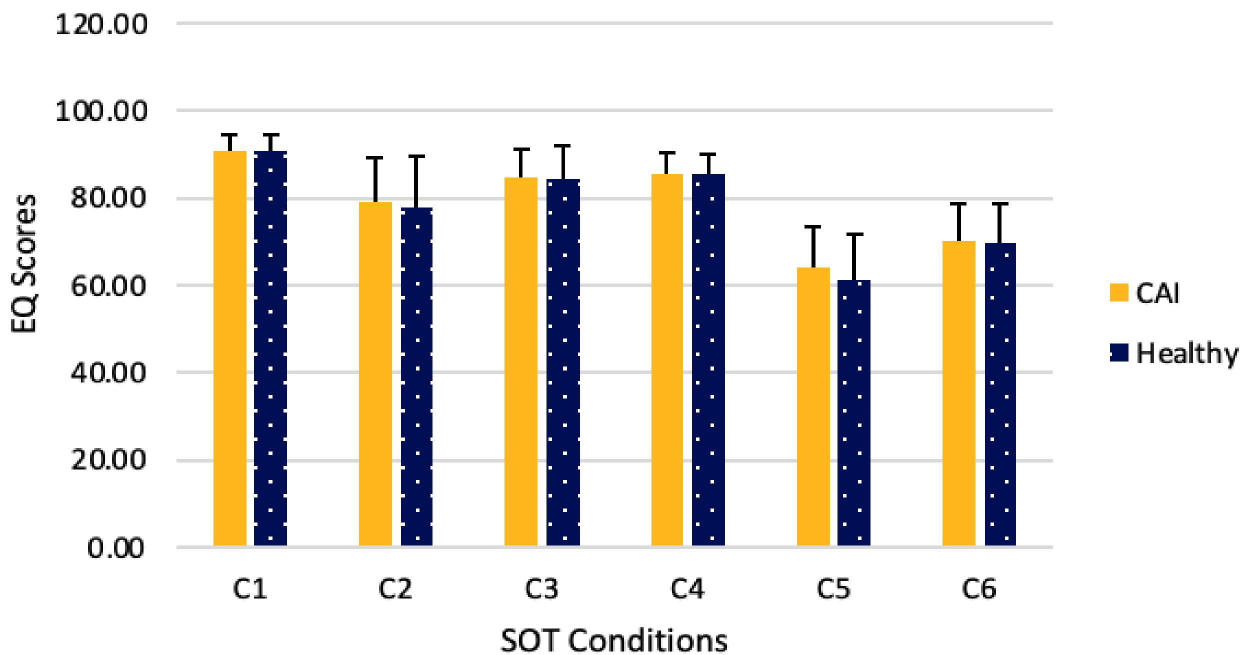

Figure 3.23. Group Main Effect for Equilibrium Scores for Individual SOT Conditions.
EQ = Equilibrium; C = Condition; SOT = Sensory Organization Test; CAI = Chronic Ankle Instability.

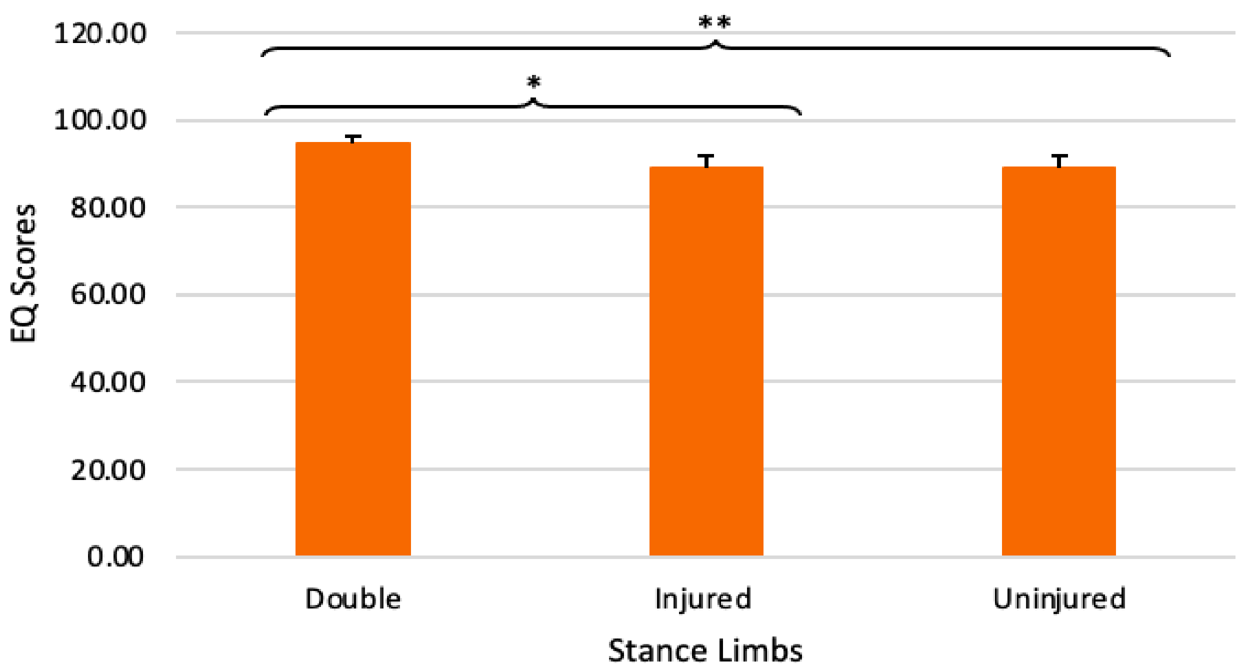

Figure 3.24. Task Main Effect for Equilibrium Scores for the SOT Condition 1.
EQ = Equilibrium; C = Condition; SOT = Sensory Organization Test; CAI = Chronic Ankle Instability.
* indicates significant differences between double- and the injured-limbs ($P < 0.05$).
** indicates significant differences between double- and the uninjured-limbs ($P < 0.05$).

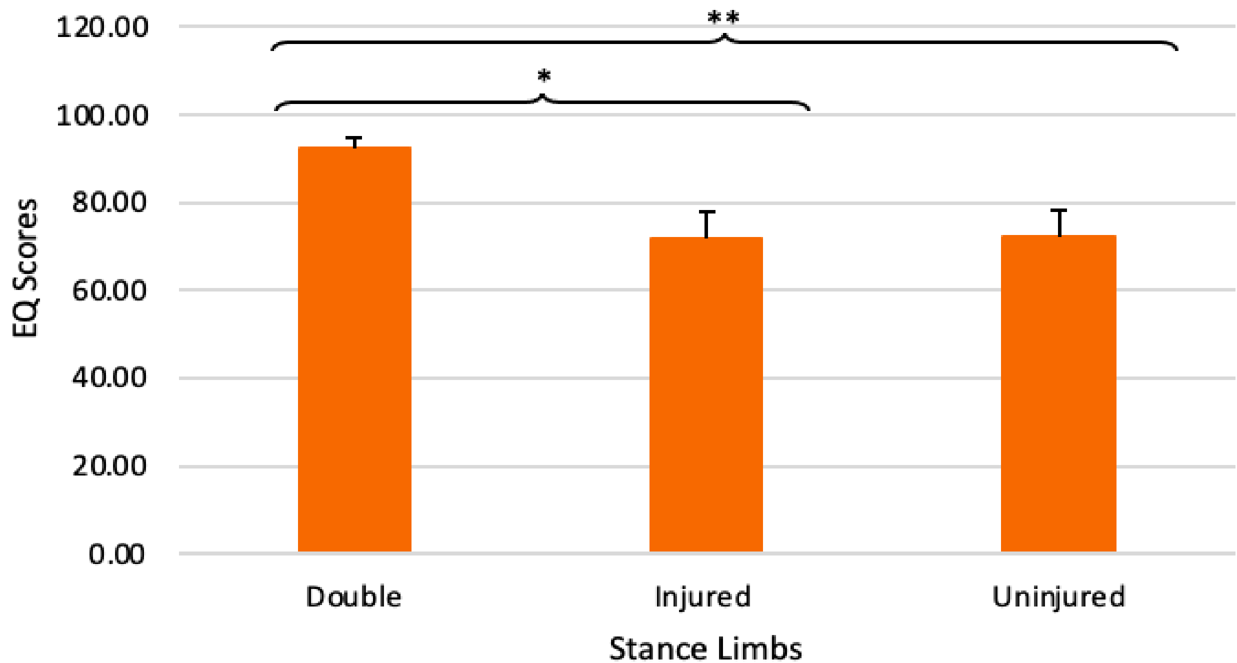

Figure 3.25. Task Main Effect for Equilibrium Scores for the SOT Condition 2.
EQ = Equilibrium; C = Condition; SOT = Sensory Organization Test; CAI = Chronic Ankle Instability.
* indicates significant differences between double- and the injured-limbs (P < 0.05).
** indicates significant differences between double- and the uninjured-limbs (P < 0.05).

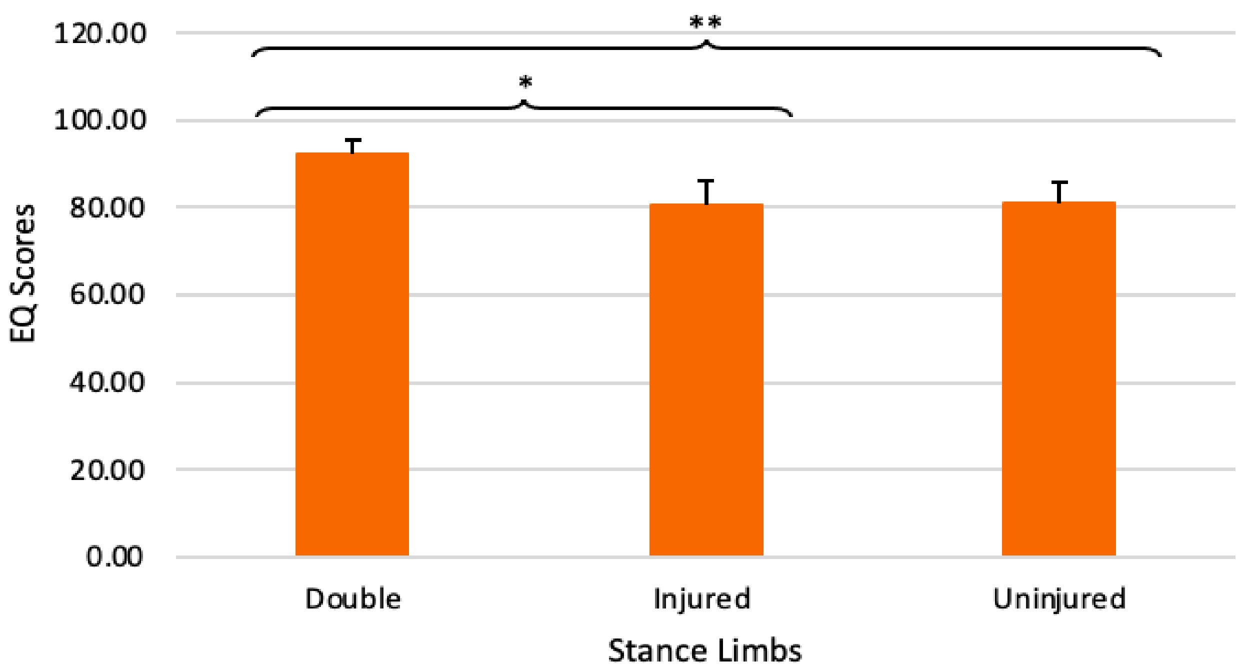

Figure 3.26. Task Main Effect for Equilibrium Scores for the SOT Condition 3.
EQ = Equilibrium; C = Condition; SOT = Sensory Organization Test; CAI = Chronic Ankle Instability.
* indicates significant differences between double- and the injured-limbs ($P < 0.05$).
** indicates significant differences between double- and the uninjured-limbs ($P < 0.05$).

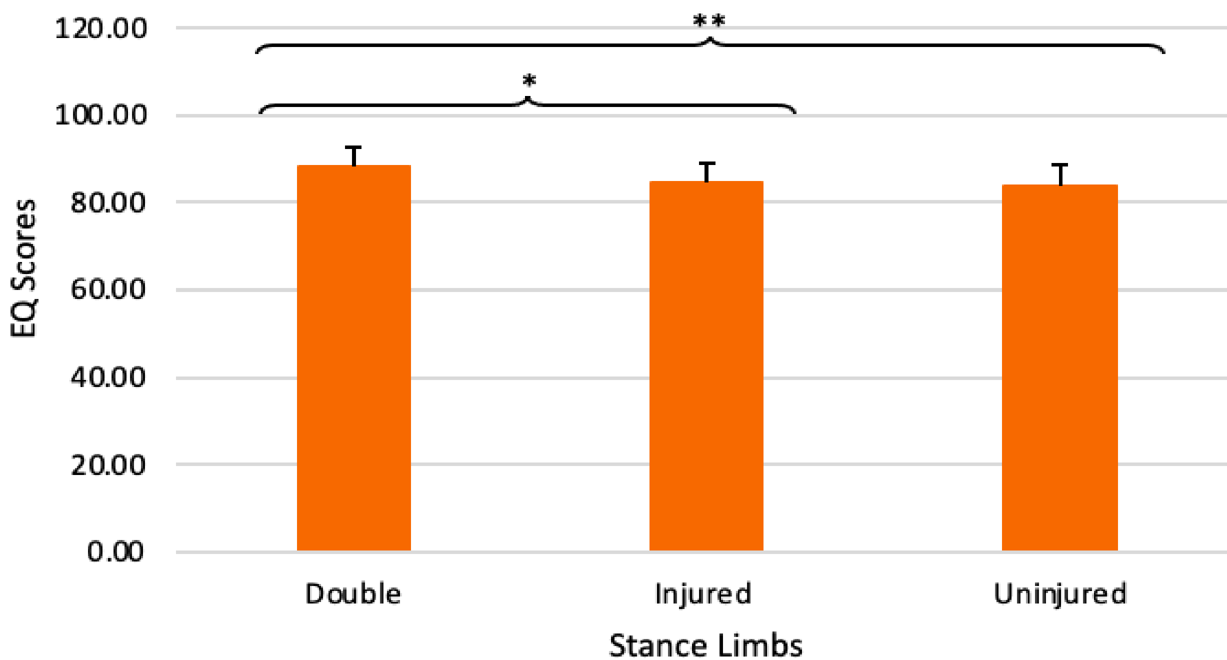

Figure 3.27. Task Main Effect for Equilibrium Scores for the SOT Condition 4.
EQ = Equilibrium; C = Condition; SOT = Sensory Organization Test; CAI = Chronic Ankle Instability.
* indicates significant differences between double- and the injured-limbs ($P < 0.05$).
** indicates significant differences between double- and the uninjured-limbs ($P < 0.05$).

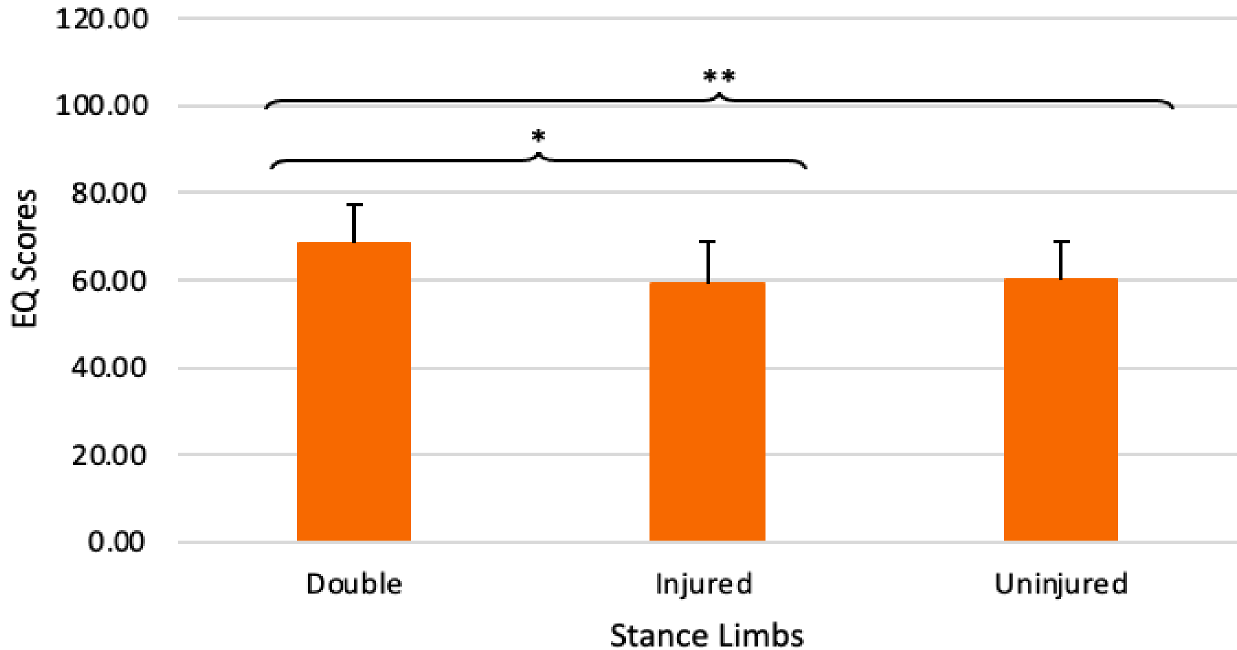

Figure 3.28. Task Main Effect for Equilibrium Scores for the SOT Condition 5.
EQ = Equilibrium; C = Condition; SOT = Sensory Organization Test; CAI = Chronic Ankle Instability.
* indicates significant differences between double- and the injured-limbs (P < 0.05).
** indicates significant differences between double- and the uninjured-limbs (P < 0.05).

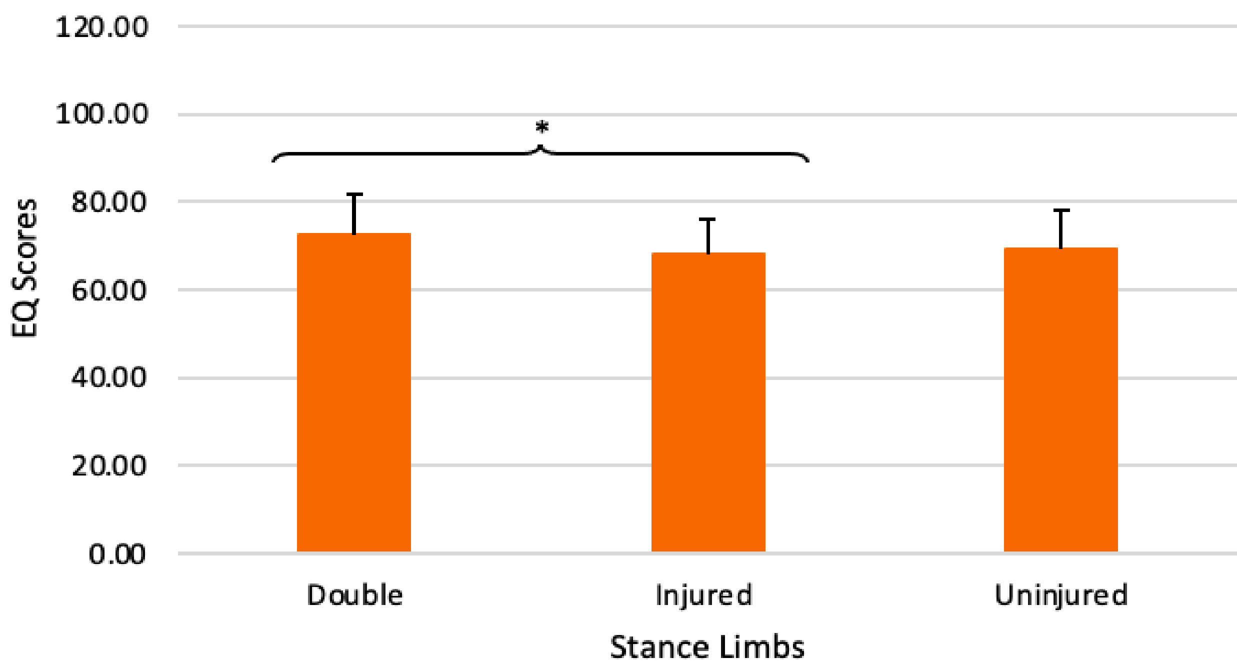

Figure 3.29. Task Main Effect for Equilibrium Scores for the SOT Condition 6.
EQ = Equilibrium; C = Condition; SOT = Sensory Organization Test; CAI = Chronic Ankle Instability.
* indicates significant differences between double- and the injured-limbs ($P < 0.05$).

CHAPTER V

MANUSCRIPT II

Introduction

An ankle sprain is one of the common musculoskeletal injuries in various age groups. Approximately 23,000 ankle sprains occur daily in the United States, and more than 1 million ankle sprains are annually reported in the emergency department (Kannus & Renstrom, 1991; Shah et al., 2016; Wikstrom et al., 2018). Lateral ankle sprain (LAS) accounts for approximately 90% of all health care costs associated with ankle sprains (Shah et al., 2016; Wikstrom et al., 2018). Unfortunately, up to three-quarters of individuals with LAS develop repetitive bouts of ankle instability, known as chronic ankle instability (CAI) (Kannus & Renstrom, 1991; Waterman et al., 2010). CAI is one of the significant contributing factors to the development of post-traumatic ankle osteoarthritis, decreasing quality of life (Valderrabano et al., 2006; Arnold et al., 2012; Thomas et al., 2017). In addition, residual symptoms associated with CAI such as chronic pain, swelling, peroneal muscles weakness, faulty ankle joint positioning sense result in long-term proprioceptive deficits and physical limitations, which are postural instability and maladaptive gait (Hertel, 2008; Hertel & Corbett, 2019). With chronicity in CAI, those sensory-motor limitations may lead to a less flexible and adaptable central nervous system (CNS) to adapt to a sudden change in environment to coordinate motor behaviors (Stergiou & Decker, 2011).

Postural control is an important aspect of daily activity and sports performance, especially in unstable environments. Adequate postural control provides dynamic stability and greater precision by minimizing and correcting reflexive oscillations at the ankle. The dynamic defense mechanisms of the ankle to a sudden change in the environment have been examined with mechanical or sensory perturbation utilizing traditional trapdoor maneuver or the adaptation test (ADT) of the NeuroCom, respectively. Trapdoor maneuver perturbation mechanically displaces the position of the ankle joint, which also displaces total body center-of-mass (COM) or disequilibrium. Displacement of total body COM involves body segments other than at the ankle to counter perturbations, maintaining the COM relative to the base of support. Hence, the trapdoor maneuver perturbation has been commonly examined, measuring the reaction time of muscles (e.g., peroneus longus, brevis, tibialis anterior) around the foot and ankle complex (Munn et al., 2010; Hoch & McKeon, 2014). Whereas the ADT applies sensory perturbation (i.e., somatosensory), imposing perceptual instability rather than disequilibrium by rotating the support surface, which may cause small mechanical displacement of a single body segment (i.e., ankle joint) (Horak et al., 1997; Rasman et al., 2018).

Individuals with CAI are well characterized with somatosensory deficits resulting from structural damage caused by an initial ankle sprain, altering afferent feedback to the CNS (Freeman et al., 1965; Munn et al., 2010). Somatosensory is one of the primary sensory feedback individuals utilize to perceive relevant information of the surrounding environment to organize the most stable motor solution relative to a task goal (Newell, 1991; Greeno, 1994; Hertel, 2008). Therefore, irrelevant somatosensory feedback allows individuals to freeze or unfreeze the degree of freedom that is perceptual redundancy. Perceptual redundancy provides overlapping environmental information from microscopic (e.g., molecules, cellular, neuronal, motor units) to macroscopic (e.g., joints, muscles) physiological elements to the CNS, then autonomous correction to maintain equilibrium status if errors were introduced by one element (Peterka & Loughlin, 2004; Stergiou & Decker, 2011; Glazier, 2017). For instance, healthy individuals can flexibly freeze and unfreeze perceptual redundancy to adapt to the ever-changing environment and task constraints. In contrast, CAI individuals have demonstrated an inability to be as flexible in perceptual redundancy by increasing visual reliance to control posture in a single-limb stance compared to healthy controls (Song et al., 2016).

To the best of our knowledge, there is no study evaluating postural adaptation to sensory perturbation of instability rather than traditional mechanical perturbation with the actual displacement of a body segment position (e.g., foot-ankle complex), especially with the increased complexity of environmental and task constraints. Therefore, the purpose of the current study was to examine postural adaptation to a sudden change in the environment similar to lateral ankle sprain mechanisms (i.e., inversion, plantarflexion) with increased environmental and task constraints in individuals with and without CAI.

Methods

In this case-control and mixed-model design, we examined participants' ability to adapt and sustain proper postural control when there are sudden changes in the environment. The change in environment is stimulated with sudden surface rotation in inversion (IN) and plantarflexion (PF) while participants completed the ADT in double- and single-limb stances.

Participants

We recruited 44 physically active individuals with and without CAI between the ages of 16 and 39. The participants' descriptive information is presented in Table 5.1. Participants were considered physically active if they were engaging in moderate- or vigorous-intensity aerobic activity for at least 150- or 75-minutes, respectively per week. The inclusion criteria for those CAI individuals followed the International Ankle Consortium guidelines (Gribble et al., 2013; Gribble et al., 2014a, 2014b). We ensured potential individuals with CAI 1) had a history of at least one significant ankle sprain at least 12-months prior to study enrollment, 2) had experienced 2 episodes of the previously injured ankle joint "giving way" and/or "feelings of instability" within the past 6-months and/or had a history of recurrent ankle sprains, 3) scored >11 on the Identification of Functional Ankle Instability (IdFAI), and 4) had not experienced recurrent ankle sprains in the last 3-months (Gribble et al., 2013; Gribble et al., 2014a, 2014b). Individuals who did not meet those CAI criteria and scored ≤ 11 on the IdFAI were assigned to the healthy controls group (Simon et al., 2012). Exclusion criteria comprised ongoing inflammatory symptoms; a history of surgeries in the brain and/or on the lower extremity (i.e., foot, ankle, knee, hip, lower back); medically diagnosed concussion within the past six months prior to study enrollment; vestibular and visual disorders and peripheral neuropathies, that may have influenced postural control; connective tissue disease and/or disorders (e.g., rheumatoid arthritis, Marfan syndrome, Ehlers-Danlos syndrome); chronic musculoskeletal conditions (e.g., OA, ACL deficiency). We included those individuals with bilateral CAI and identified the limb with the worst IdFAI score as the injured-limb. In order to assign an injured-limb to healthy individuals, participants in the healthy controls group were matched to participants in the CAI group for sex, age (years, $\pm$ 2), height (cm, $\pm$ 5%), mass (kg, $\pm$ 3%), limb dominance (the leg used to kick a ball), and the National Aeronautics and Space Administration Physical Activity Status Scale [NASA-PASS] (scale, $\pm$ 1), identifying the injured limb. All participants and participants' parents/legal guardians gave their informed consent, which was approved by the Institutional Review Board at the University of North Carolina Greensboro.

Procedure

Pre-screening questionnaires were presented to potential participants using online survey software (Qualtrics, Provo, UT) to ensure individuals met inclusion and exclusion criteria before enrolling in the study. Participants who met study criteria were invited to the lab for a single session and completed a standardized health history questionnaire. The standardized health questionnaire included questions about previous ankle sprains, a history of the lower extremity (i.e., knee, hip, lower back), self-reported ankle instability and function (i.e., IdFAI), and physical activity status (i.e., NASA-PASS). Participants began the protocol with a 5-minute warm-up on a bike at a self-selected intensity, demographic measures (height, weight), joint hypermobility tests, lower extremity (ankle, knee, hip) anatomical alignment measures, the sensory organization test (SOT) as a part of a related study, and the ADT. A vest and safety harness were outfitted on participants prior to completing the ADT barefoot on a computerized NeuroCom dynamic posturography platform (SMART EquiTest, NeuroCom International Inc., Clackamas, OR) in double- and single-limb (injured, uninjured) stances to ensure the safety of participants (Figure 5.1).

Adaptation Test

The ADT of the NeuroCom is designed to examine individuals' postural adaptation to unexpected anterior-posterior (AP) surface rotation (tilt) perturbations in a double-limb stance (Figure 5.2). Unique to our protocol, we have tested the ADT in both AP and medial-lateral (ML) directions in double- and single-limb stances (Figure 5.3 & 5.4). The ADT perturbations have an amplitude of 8-degrees AP surface rotation in either toes-up (dorsiflexion) and toes-down (plantarflexion) positions for five consecutive trials. Each trial lasts 400 msec, and the delay time between trials is between 3-to-5 seconds, independent of participants' sway. For ML surface rotation in eversion (medial border of foot down) or inversion (medial border of the foot up), participants faced the sidewall of the NeuroCom surroundings (Figure 5.4). Participants were instructed to align bilateral ankles (medial malleoli) perpendicular (for AP) or parallel (for ML) to the transverse axis of platform rotation with a standardized distance apart based on their height according to manufacture guidelines for double-limb stance and position the foot in the center of the platform for single-limb stance (Figure 5.5 & 5.6). Participants maintained their face forward, kept their arms relaxed by sides, and stood motionless as possible while completing the ADT in double- and single-limb stances (Figure 5.3 & 5.4). Participants were given a 1-minute rest between conditions to minimize fatigue. The trials were stopped and repeated if participants stood on a non-stance limb at the moment of a sudden rotation of the support surface.

Postural Adaptation Measure

The ADT computes *Sway Energy (SE) scores*. SE scores are quantified by measuring the magnitude of ground reaction forces, collected at 100 Hz, due to individuals' automatic postural response resisting postural disequilibrium caused by unexpected surface rotation perturbation in AP and ML directions. If individuals were able to minimize postural sway when exposed to a sudden surface perturbation, it represents their ability to react efficiently, reflecting smaller SE scores. Therefore, the smaller SE scores are an indication of adaptability to a sudden change in the support surface (environment), resulting in postural stability (Nashner, 1976). Whereas the greater SE scores indicate insufficient ability to react and adapt to the change in the environment, resulting in postural instability (Nashner, 1976; Paquette et al., 2016). As postural response and adaptation are strongly influenced by previous experience over several trials, the SE scores typically become smaller in subsequent trials than in the initial trial (Horak et al., 1989, 1996, 1997). Thus, to control the learning effect, the SE score from the first trial was used for analyses.

Statistical Analysis

SPSS software (version 27; IBM Corp, Armonk, NY, USA) was used to perform all statistical analyses with a priori α level of 0.05. Independent *t*-tests were used for group comparisons on age, height, weight, NASA-PASS scales, and IdFAI scores. We ran three separate 2 (group) $\times$ 2 (environment: ADT IN & PF) repeated measures analysis of variance (ANOVA) to examine the effect of environmental constraints on postural adaptation for individual limbs (double, injured, uninjured). The effect of task constraints on postural adaptation was analyzed with separate 2 (group) × 3 (task: double-, injured-, uninjured-limbs) repeated measures ANOVA for individual ADT conditions (ADT IN & PF). In the event of statistical interactions and main effects, Tukey post-hoc analyses were performed. The Cohen's *d* effect sizes with 95% confidence intervals were calculated between the two means and the pooled standard deviation with values of ≤ 0.40 indicating as small, 0.40–0.80 as moderate, and ≥ 0.80 as large effects (Cohen, 1988).

Results

Descriptive data of all outcome variables (age, height, physical activity level) are presented in Table 5.1 (t range = 0.39-0.94; $P > 0.05$). CAI individuals had a greater number of ankle sprains, episodes of giving-way within the past 6-months, higher IdFAI scores compared to healthy controls ($P < 0.05$; Table 5.1).

Postural Adaptation in Increased Environmental (ADT IN and PF) Constraints:

No significant group × environment interactions were revealed on individual stance limbs (double: $F_{1,42} = 0.21$; injured: $F_{1,42} = 2.07$; uninjured: $F_{1,42} = 0.23$; $P > 0.05$; Tables 5.2-5.4; Figures 5.7-5.9). We observed a significant main effect for group in the uninjured-limb ($F_{1,42} = 6.19$, $P = 0.017$, Figure 5.10) that CAI individuals (39.09 ± 1.47) presented better postural adaptation compared to healthy controls (44.25 ± 1.47), but not for other stance limbs (double: $F_{1,42} = 3.52$; injured: $F_{1,42} = 6.19$; $P > 0.05$; Figure 5.10). There were significant main effects for environment on all stance limbs: double ($F_{1,42}$ =

67.05, $P < 0.001$, Figure 5.11), injured ($F_{1,42} = 55.64$, $P < 0.001$, Figure 5.12), and uninjured ($F_{1,42} = 55.88$, $P < 0.001$, Figure 5.13). Tukey post-hoc comparisons exhibited that postural adaptation was worse to a sudden IN platform perturbation (72.71 $\pm$ 2.61) than to a PF platform perturbation (47.68 $\pm$ 1.63; $P < 0.001$) in a double-limb stance. In contrast, postural adaptation to a sudden PF platform perturbation was worse than postural adaptation to an IN platform perturbation when standing in the injured- (PF: 46.82 $\pm$ 1.31; IN: 37.98 $\pm$ 1.30; P < 0.001) and uninjured-limbs (PF: 47.55 $\pm$ 1.50; IN: 35.80 $\pm$ 1.07; P < 0.001), respectively.

Postural Adaptation in Increased Task (Stance Limbs: Double, Injured, and Uninjured) Constraints:

There were no significant group $\times$ task interactions on individual ADT conditions (IN: $F_{2,84} = 0.65$; PF: $F_{2,84} = 1.06$; P > 0.05; Tables 5.5 & 5.6; Figures 5.14 & 5.15). A statistically significant main effect for the group was found for the ADT PF ($F_{1,42} = 4.93$, P = 0.032, Figure 5.16), indicating CAI individuals (44.68 $\pm$ 1.70) demonstrated better postural adaptation to a sudden PF platform perturbation than healthy controls (50.02 $\pm$ 1.70). Whereas, no significant group differences were found for the ADT IN ($F_{1,42} = 1.69$, P > 0.05, Figure 5.16). A statistically significant main effect for the task was found for the ADT IN ($F_{2,84} = 138.66$, $P < 0.001$, Figure 5.17) but not for the ADT PF ($F_{1,42} = 0.19$, $P > 0.05$, Figure 5.18). Tukey post-hoc comparisons revealed a significant difference in postural adaptation to a sudden IN platform perturbation in double-limb stance (72.71 $\pm$ 2.61) was worse than the injured- (37.98 $\pm$ 1.30, $P = 0.000$) and uninjured-limbs (35.80 $\pm$ 1.07, $P = 0.000$), respectively.

Discussion

This is the first study to our knowledge that has examined postural adaptation in directions of IN and PF to a sudden sensory perturbation without mechanically displacing the ankle joint by utilizing the ADT. Our primary findings were that there was no effect of an increase in complexity of environmental or task constraints on group differences in postural adaptation when a somatosensory perturbation was applied while controlling posture in double- and single-limb (injured, uninjured) stances in individuals with and without CAI. However, there were several group main effects on individual stance limbs and ADT conditions: CAI individuals exhibited better postural adaptation in the uninjured-limb regardless of the direction of somatosensory perturbations (IN, PF), and to the somatosensory PF perturbation regardless of individual stance limbs (double, injured, uninjured) compared to healthy controls. These results imply that CAI individuals had a superior automatic motor response to somatosensory perturbations compared to healthy controls, suggesting the potential implementation of compensatory mechanisms to control posture. Additionally, there were environmental and task main effects regardless of the group. The worst postural adaptation to a sudden somatosensory perturbation in IN was displayed in a double-limb stance, while the worst postural adaptation to a sudden somatosensory perturbation in PF was noted in the injured- and uninjured-limbs. However, the main effect for the task was only present in a single direction: postural adaptation to a sudden somatosensory IN perturbation was worse in double-limb stance than the injured- and uninjured-limbs.

Postural control is critical for achieving suitable performance in an ever-changing environment. In addition, quick postural response and adaptation to environmental change are necessary to prevent a risk of subsequent ankle sprains among CAI individuals. Postural response to unexpected perturbations has been shown to be automatic in humans (Nashner, 1976). The ADT examines individuals' automatic motor systems by quantifying their ability to minimize sway when exposed to an unexpected rotation of the support surface (somatosensory perturbations) in AP and ML directions. Postural adaptation implies a gradual shift from a reactive postural response to perturbations. To prevent recurrent ankle sprains, postural response and adaptation must be context-specific, activating different functionally appropriate muscles to stabilize postural sway. In previous studies, some researchers noted the prolonged reaction time of targeted muscles (e.g., peroneus longus and brevis, tibialis anterior) during a mechanical perturbation with a trapdoor maneuver, especially the perturbation, was imposed in multi-directional supination (IN, PF) (Vaes et al., 2001; Mitchell et al., 2008a, 2008b; Mendez-Rebolledo et al., 2015). Although our study did not examine reaction time of specific muscles, peroneal and tibialis anterior muscles were primarily induced when the support surface rotated in IN and PF directions, respectively (Schieppati & Nardone, 1995; Winter, 1995; Horak et al., 1997; Moore et al., 1998). Thus, similar or better postural adaptation to somatosensory perturbations in IN and PF exhibited by CAI individuals compared to healthy controls in our study may suggest they were able to activate those muscles to an appropriate response without a delay. An adequate postural response is critical to provide dynamic stability at the ankle to prevent subsequent ankle sprains.

Despite the current evidence, why repetitive ankle sprains are still experienced in CAI is still unknown.

Successful automatic postural response and adaptation depend on the integration of somatosensory, visual, and vestibular feedback. Somatosensory feedback, which is often impaired from an initial ankle sprain, has been suggested to be the greatest contribution to postural control during standing (Freeman et al., 1965; Horak et al., 1997; Munn et al., 2010). Similarly, visual feedback plays an important role in guiding and fine-tuning motor outputs (Turvey, 1990, 2007; Greeno, 1994). However, as visual feedback is found to be too slow to provide an influence on postural response and adaptation, the convergence of multisensory integration plays a significant role in postural adaptation (Nagata et al., 2001; Rasman et al., 2018). From the dynamic systems perspective, there is redundancy in sensory feedback and multiple elements across the human body, which comprises a high-dimensional degree of freedom that contains independent, yet often functionally redundant neurobiological elements at different levels from microscopic (e.g., molecules, cellular, neuronal, motor units) to macroscopic (e.g., joints, muscles) (Winter, 1995; Glazier, 2017). This phenomenon, known as Bernstein's problem, suggests there are near-infinite ways of coordinating postural response and adaptation (Newell, 1991; Davids et al., 2003; Davids & Glazier, 2010; Glazier, 2017). If somatosensory is disrupted with surface rotation without elimination of vision, healthy individuals emphasize visual and vestibular feedback to compensate for the disruption of somatosensory feedback to maintain an upright stance (Nashner, 1982). CAI individuals are suggested to increase reliance on visual feedback to compensate for somatosensory deficits caused by an initial ankle sprain (Song et al., 2016). Therefore, better postural adaptation to a sudden somatosensory perturbation displayed in our CAI may be a result of their reliance on visual and vestibular feedback.

Interestingly, CAI individuals revealed better postural adaptation in the uninjured-limb and to the somatosensory perturbation in PF compared with healthy controls. Horak et al. (1997) suggest automatic postural response and adaptation are sharpened by previous experience and preprogrammed muscle activation patterns. Indeed, Delahunt et al. (2007) displayed preactivation of rectus femoris, tibialis anterior, and soleus muscles right before and after initial contact during lateral hopping tests. Preactivation of those muscles is indicative of pre-programmed feedforward motor control, providing dynamic joint stability for the more inverted ankle characterized by CAI. Not only limited to dynamic postural control but preactivation of tibialis anterior muscles has also been reported at right before heel-contact and throughout the stance phase during walking in CAI compared to healthy controls (Louwerens et al., 1995; Hopkins et al., 2012; Koldenhoven et al., 2016). Limited ankle dorsiflexion range-of-motion is commonly exhibited in CAI individuals assessed via weight-bearing lunge tests and walking kinematics (Hoch et al., 2011, 2012; Chinn et al., 2013). Thus, preactivation of tibialis anterior muscles could be interpreted as a strategy CAI individuals attempted to maintain the ankle joint complex in a more stable and close-packed position during dynamic postural control tasks and walking. Tibialis anterior muscles were induced with the somatosensory perturbation in PF in our study. Consequently, preactivation of tibialis anterior muscles may have granted those CAI individuals to achieve better postural adaptation to the somatosensory perturbation in PF compared with healthy controls.

There are a limited number of studies reporting centrally mediated changes in the uninjured-limb among CAI individuals. For example, postural control dysfunction and increased stride-to-stride gait variability (i.e., movement errors) in frontal-plane ankle kinematics have been reported in the uninjured-limb in individuals with unilateral CAI compared to healthy controls (Trop et al., 1984; Hamacher et al., 2016; Wanner et al., 2019). Although our results did not display centrally mediated dysfunctions in the uninjured-limb, current evidence suggests the development of CAI could have affected neuromuscular control of the uninjured-limb. Additionally, Beckman and Buchanan (1995) revealed similar findings to our results at bilateral hips: a significantly faster reaction time of gluteus medius muscles in response to an ankle inversion trapdoor perturbation was observed in both injured- and uninjured-limbs in CAI compared to healthy controls. It has been postulated that the CNS composes postural responses from a combination of synergies (Nashner & McCollum, 1985; Horak et al., 1997). As gluteus medius muscles primarily provide functional stability of lateral trunk sway, preactivation of the gluteus medius is a synergistic muscle response compensating to prevent the unstable ankle from turning into excessive inversion. Consequently, it is our hypothesis that better postural adaptation we found in the uninjured-limb is the result of centrally mediated compensations CAI individuals developed over time, coordinating several synergies at the ankle and hip in preparation for when the injured-limb failed to adapt to a sudden perturbation.

There were environmental and task effects on postural adaptation regardless of the group; how individuals performed postural adaptation depended on individual stance limbs (i.e., double, injured, uninjured) and the direction of somatosensory perturbations (i.e., IN). Specifically, individuals with and without CAI presented worse postural adaptation to the somatosensory IN perturbation in a double-limb stance, whereas those individuals presented worse postural adaptation to the PF somatosensory perturbation in a single-limb (injured, uninjured) stance. The center-of-pressure (COP) excursion point in relation to the lateral border of the base of support is much closer than the anterior border of the base of support while controlling posture. Thus, a large proportion of falls involve failure to compensate for lateral destabilization (Maki & Mcllroy, 1997). Indeed, Maki and colleagues (1994) found quantification of postural response to lateral surface rotation is one of the best predictors of future falls among elderly individuals. Collectively, our assumption was to discover worse postural adaptation to a sudden somatosensory IN perturbation in a single-limb stance, where the base of support is much narrower than double-limb stance. However, better postural adaptation to the IN perturbation we found in a single-limb (injured, uninjured) stance with more complexity may suggest the emergence of compensatory mechanisms, providing dynamic stability.

The onset of postural response shifts as the perturbation direction changes. Response to surface rotation in AP and ML directions involves particular muscle groups. Anterior perturbation activates muscles in order of tibialis anterior, quadriceps, and abdominal, whereas posterior perturbation activates muscles in order of gastrocnemius, hamstrings, and paraspinal (Horak & Nashner, 1986). In contrast, correction for lateral perturbation involves early activation of hip abductors (e.g., gluteus medius) rather than distal-to-proximal muscle activation patterns triggered by a somatosensory perturbation at the feet. The postural response is modulated with perturbation directions and conduction at spinal and supraspinal levels (Moore et al., 1988; Winter, 1995). Automatic response in the limb contralateral to the perturbed limb is mediated at spinal and supraspinal levels and takes longer than an ipsilateral limb. For example, hip abductor muscles were most active in the perturbed limb and active to a lesser extent in the contralateral limb (Moore et al., 1988). Therefore, a worse postural adaptation found to a sudden somatosensory perturbation in double-limb stance may be a consequence of a difference in activation of involved muscles and potential delay in those muscle activations, especially in the contralateral limb to the perturbed limb during double-limb stance.

The current study was not without limitations. Our participants were informed about the perturbation as a part of informed consent. Participants' awareness of perturbations in advance may have influenced their postural response and adaptations. Furthermore, we did not assess muscle activations utilizing EMG sensors while participants were completing the ADT. Thus, it is not clear which muscle activities contributed as mechanisms of postural adaptation in our participants.

Conclusion

Current study findings demonstrated that individuals with and without CAI had similar postural adaptation to a sudden change in the environment in double- and the injured-limbs. However, CAI individuals revealed better postural adaptation in the uninjured-limb and to a sudden PF perturbation compared with healthy controls. Lastly, postural adaptation depended on the environmental and task constraints for both groups.

Table 4.1. Participant Demographics (Mean ± SD).

	Group		
	CAI	Controls	***P*-values**
N	22 (13 females, 9 males)	22 (13 females, 9 males)	-
Age (years)	26.09 ± 5.76	25.41 ± 5.92	0.84
Height (cm)	172.25 ± 9.76	169.70 ± 9.32	0.61
Weight (kg)	76.18 ± 14.91	71.98 ± 14.79	0.93
NASA-PASS	6.27 ± 0.18	6.27 ± 1.03	0.67
IdFAI	19.09 ± 5.39	1.36 ± 1.81	< 0.001*
# of Ankle of Sprains	6.48 ± 7.08	0.00 ± 0.00	< 0.001*
Episodes of Giving-Way	8.88 ± 21.36	0.00 ± 0.00	< 0.001*

N = Numbers; NASA-PASS = National Aeronautics and Space Administration Physical Activity Status Scale; IdFAI = the Identification of Functional Ankle Instability; CAI = Chronic Ankle Instability.
* indicates significant group differences (P < 0.05).

Table 4.2. Group × Environment Interactions for a Double-Limb.

Double-Limb: Sway Energy Scores (Mean ± SD) and Effect Sizes			
Parameter	CAI	Control	Effect Sizes (95% CI)
ADT IN	70.50 ± 14.23	74.91 ± 19.87	0.26 (-0.34 to 0.85)
ADT PF	44.09 ± 8.92	51.27 ± 12.40	0.67 (0.06 to 1.27)

SD = Standard Deviation; ADT = Adaptation Test; IN = Inversion; PF = Plantarflexion; CAI = Chronic Ankle Instability; CI = Confidence Interval.

Table 4.3. Group × Environment Interactions for the Injured-Limb (Mean ± SD).

Injured-Limb: Sway Energy Scores (Mean ± SD) and Effect Sizes			
Parameter	CAI	Control	Effect Sizes (95% CI)
ADT IN	38.23 ± 7.00	37.73 ± 9.93	-0.06 (-0.65 to 0.53)
ADT PF	45.36 ± 8.86	48.27 ± 8.44	0.34 (-0.26 to 0.93)

SD = Standard Deviation; ADT = Adaptation Test; IN = Inversion; PF = Plantarflexion; CAI = Chronic Ankle Instability; CI = Confidence Interval.

Table 4.4. Group × Environment Interactions for the Uninjured-Limb (Mean ± SD).

Uninjured-Limb: Sway Energy Scores (Mean ± SD) and Effect Sizes			
Parameter	CAI	Control	Effect Sizes (95% CI)
ADT IN	33.59 ± 6.86	38.00 ± 7.28	0.62 (0.02 to 1.23)
ADT PF	44.59 ± 6.56	50.50 ± 12.40	0.59 (-0.01 to 1.20)

SD = Standard Deviation; ADT = Adaptation Test; IN = Inversion; PF = Plantarflexion; CAI = Chronic Ankle Instability; CI = Confidence Interval.

Table 4.5. Group × Task Interactions for the ADT IN (Mean ± SD).

ADT IN: Sway Energy Scores (Mean ± SD) and Effect Sizes			
Parameter	**CAI**	**Control**	**Effect Sizes (95% CI)**
Double-Limb	70.50 ± 14.23	74.911 ± 19.87	0.255 (-0.338 to -0.849)
Injured-Limb	38.23 ± 7.00	37.73 ± 9.93	-0.058 (-0.649 to 0.533)
Uninjured-Limb	33.59 ± 6.86	38.00 ± 7.28	0.665 (0.058 to 1.272)

SD = Standard Deviation; ADT = Adaptation Test; IN = Inversion; CAI = Chronic Ankle Instability; CI = Confidence Interval.

Table 4.6. Group × Task Interactions for the ADT PF (Mean ± SD).

ADT PF: Sway Energy Scores (Mean ± SD) and Effect Sizes			
Parameter	**CAI**	**Control**	**Effect Sizes (95% CI)**
Double-Limb	70.50 ± 14.23	74.911 ± 19.87	0.255 (-0.338 to -0.849)
Injured-Limb	38.23 ± 7.00	37.73 ± 9.93	-0.058 (-0.649 to 0.533)
Uninjured-Limb	33.59 ± 6.86	38.00 ± 7.28	0.665 (0.058 to 1.272)

SD = Standard Deviation; ADT = Adaptation Test; PF = Plantarflexion; CAI = Chronic Ankle Instability; CI = Confidence Interval.

Figure 4.1. The NeuroCom Dynamic Posturography System.

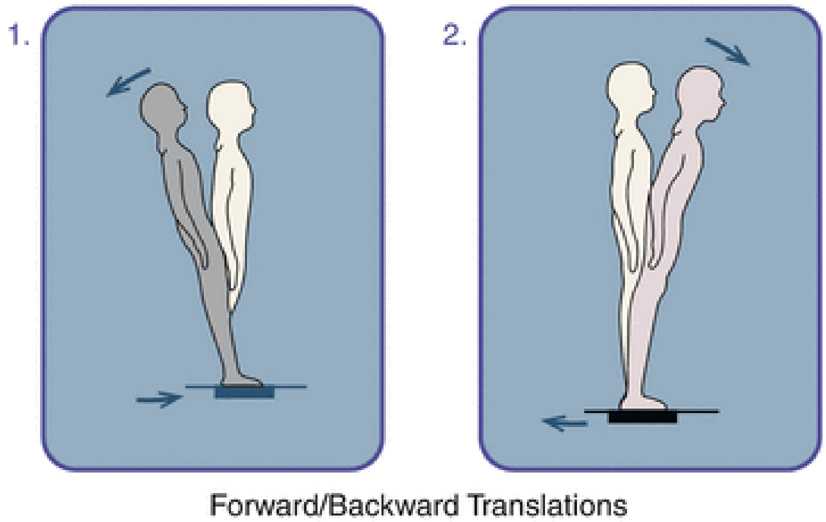

Figure 4.2. Standard Anterior-Posterior Surface Perturbation of the ADT (NeuroCom® International).

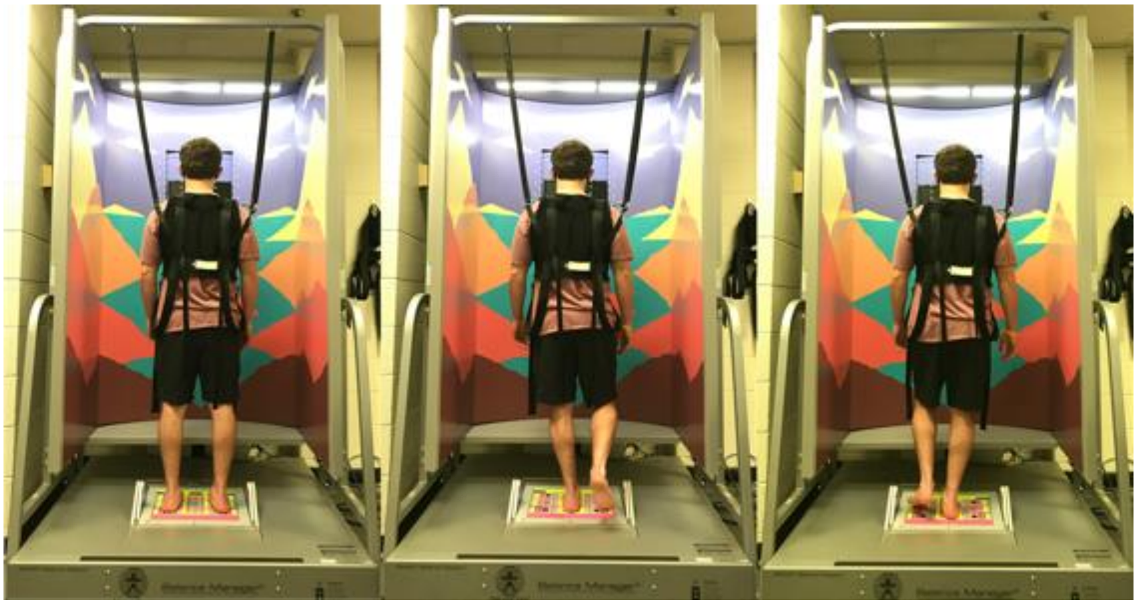

Figure 4.3. Stance Positions for the ADT in AP Direction on the NeuroCom.

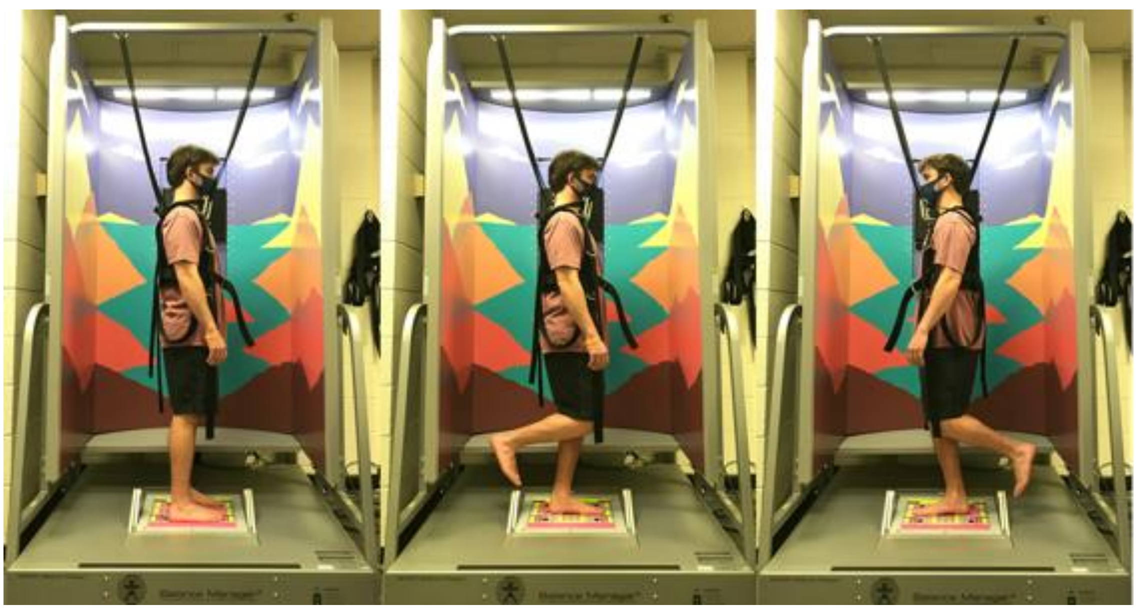

Figure 4.4. Stance Positions for the ADT in ML Direction on the NeuroCom.

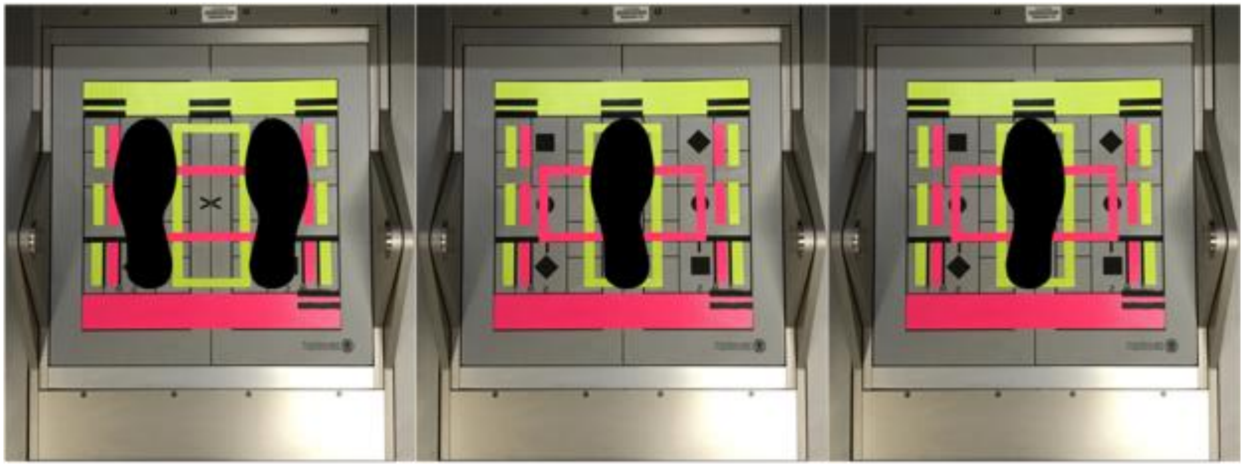

Figure 4.5. Feet Positions for the ADT in AP Direction in Double- and Single-limb (Injured, Uninjured) Stances.

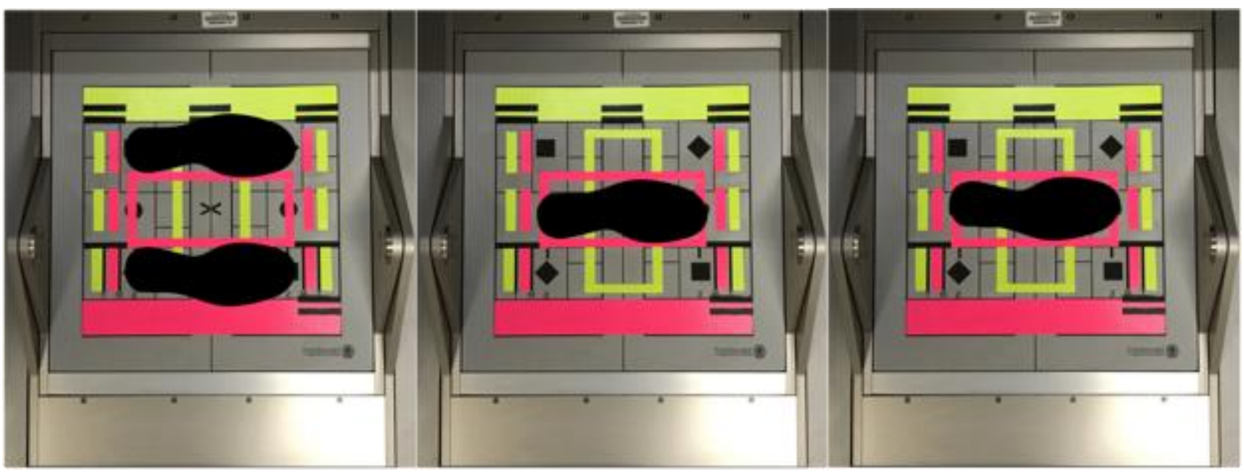

Figure 4.6. Feet Positions for the ADT in ML Direction in Double- and Single-limb Stances.

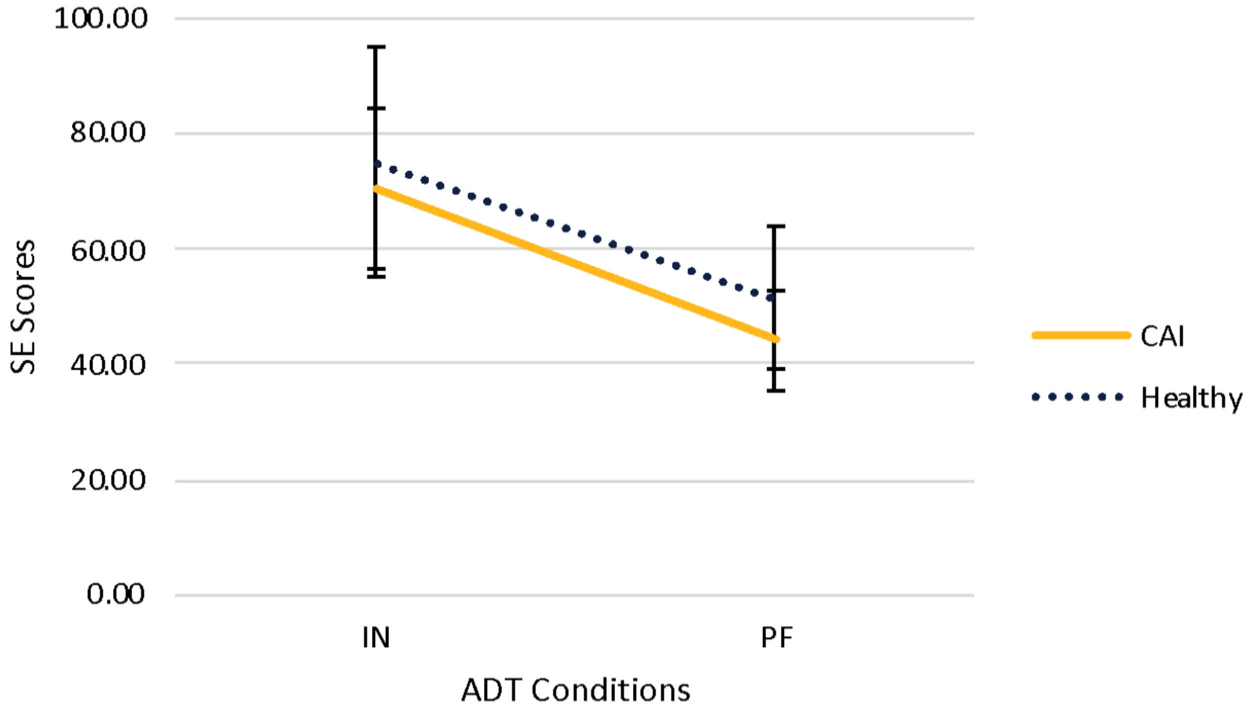

Figure 4.7. Group × Environment Interactions for a Double-Limb.
SE = Sway Energy; IN = Inversion; PF = Plantarflexion; ADT = Adaptation Test; CAI = Chronic Ankle Instability.

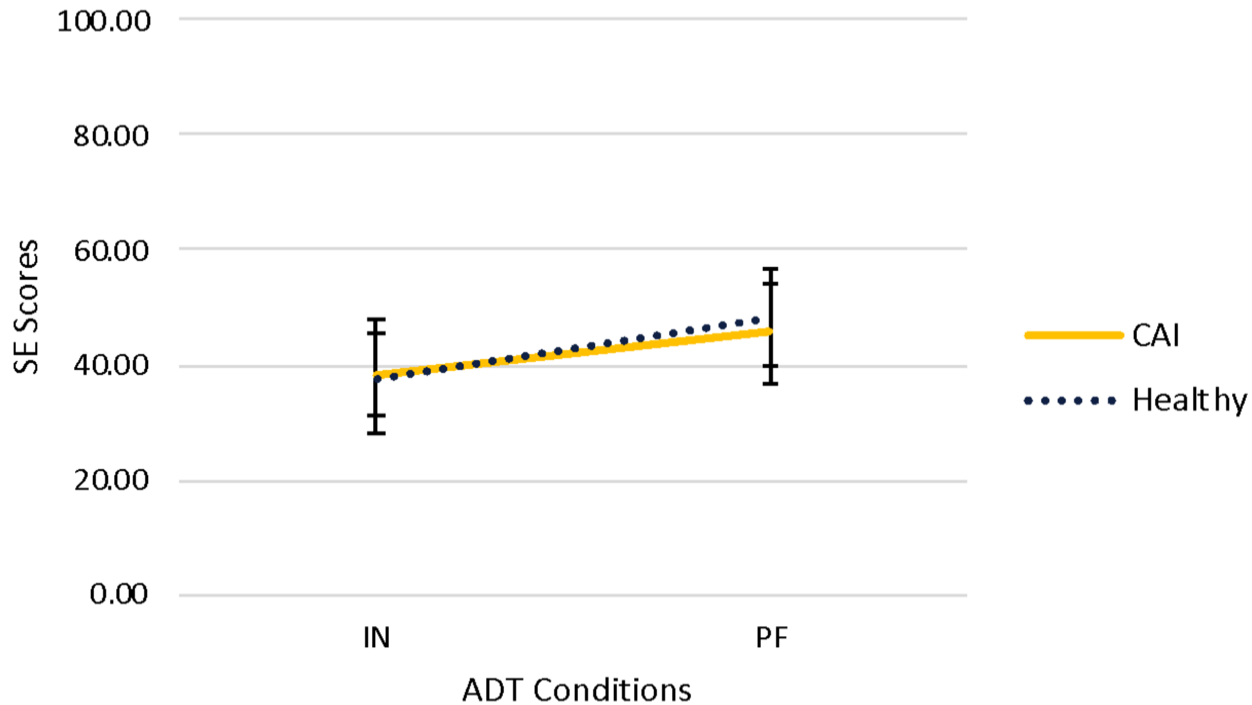

Figure 4.8. Group × Environment Interaction for the Injured-Limb.
SE = Sway Energy; IN = Inversion; PF = Plantarflexion; ADT = Adaptation Test; CAI = Chronic Ankle Instability.

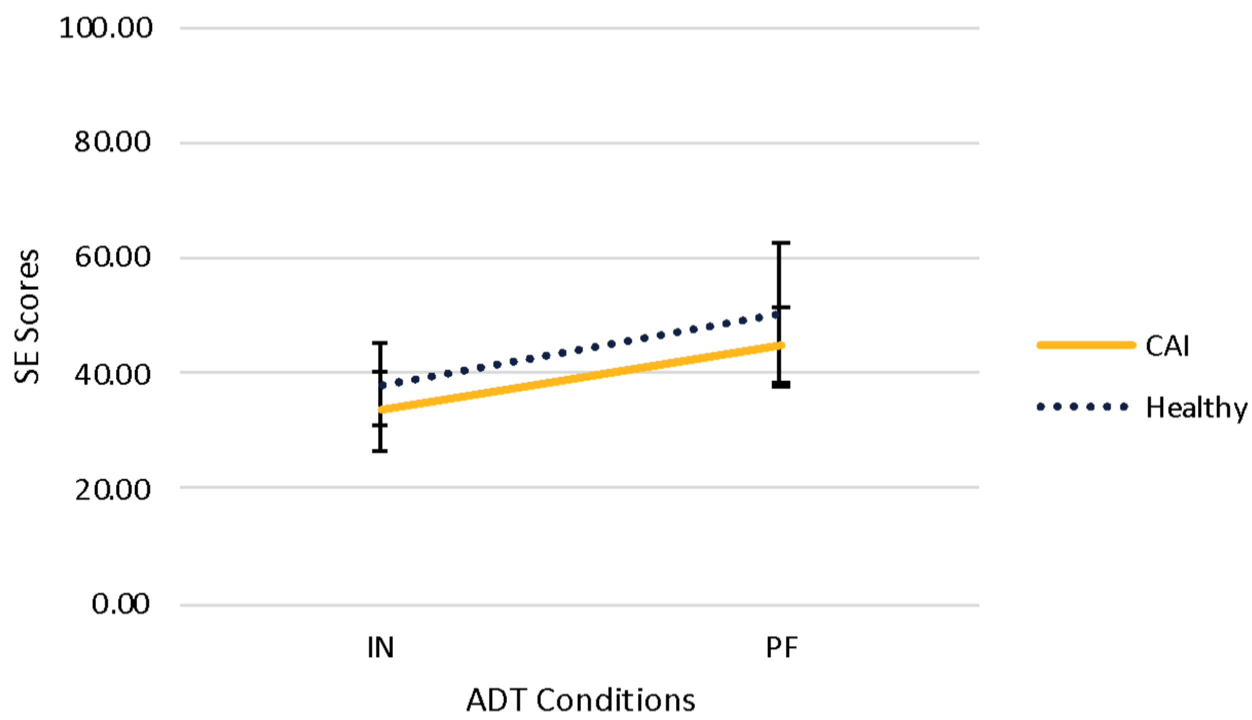

Figure 4.9. Group × Environment Interaction for the Uninjured-Limb.
SE = Sway Energy; IN = Inversion; PF = Plantarflexion; ADT = Adaptation Test; CAI = Chronic Ankle Instability.

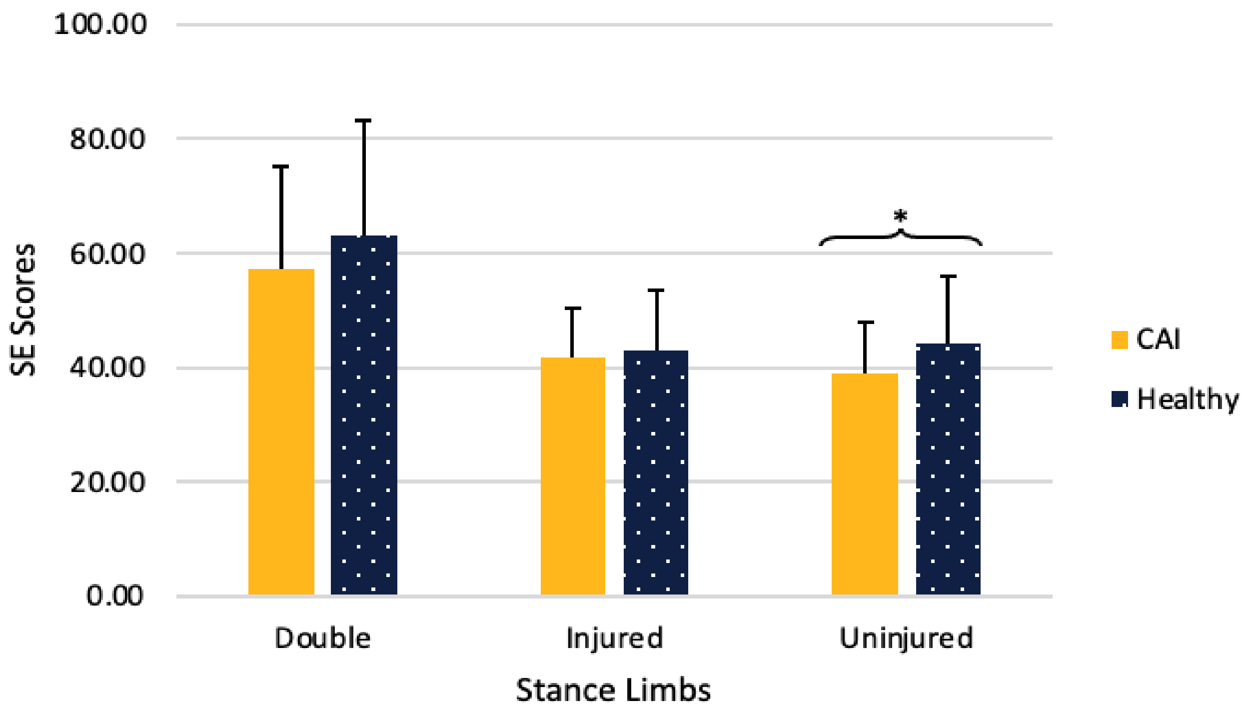

Figure 4.10. Group Main Effect for Individual Stance Limbs.
SE = Sway Energy Scores; CAI = Chronic Ankle Instability.
* indicates significant group differences (P < 0.05).

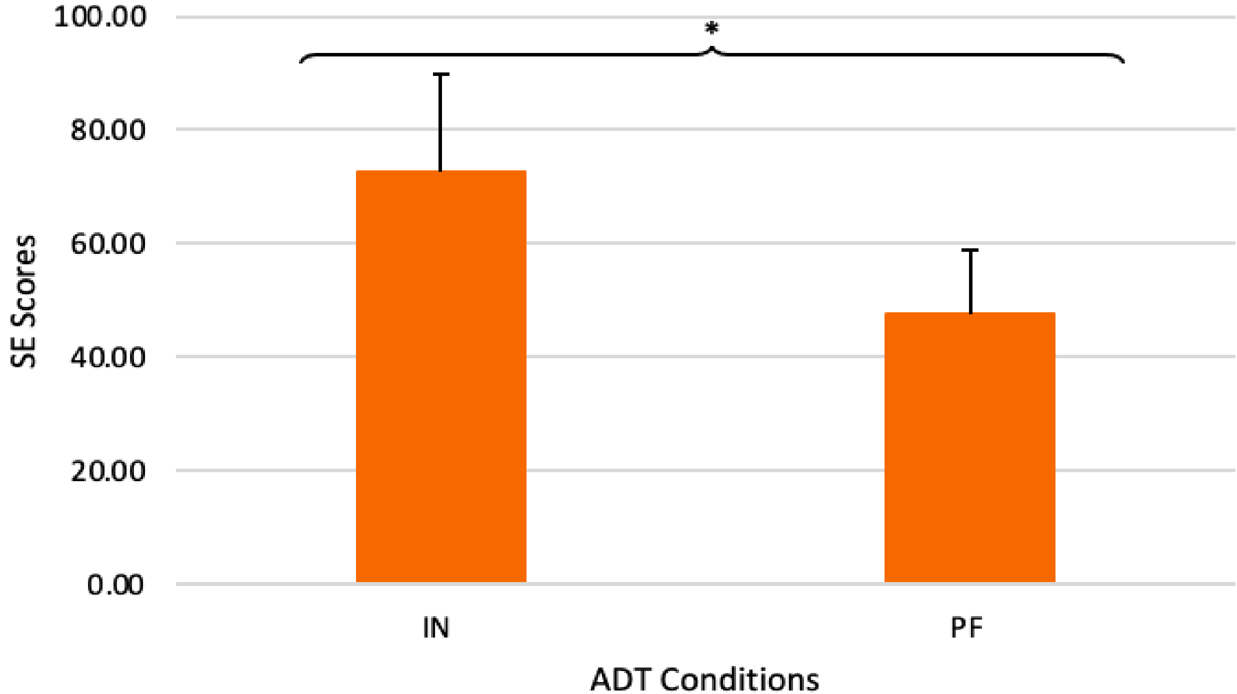

Figure 4.11. Environment Main Effect for a Double-Limb Stance.
SE = Sway Energy Scores; IN = Inversion; PF = Plantarflexion; ADT = Adaptation Test.
* indicates significant differences (P < 0.05).

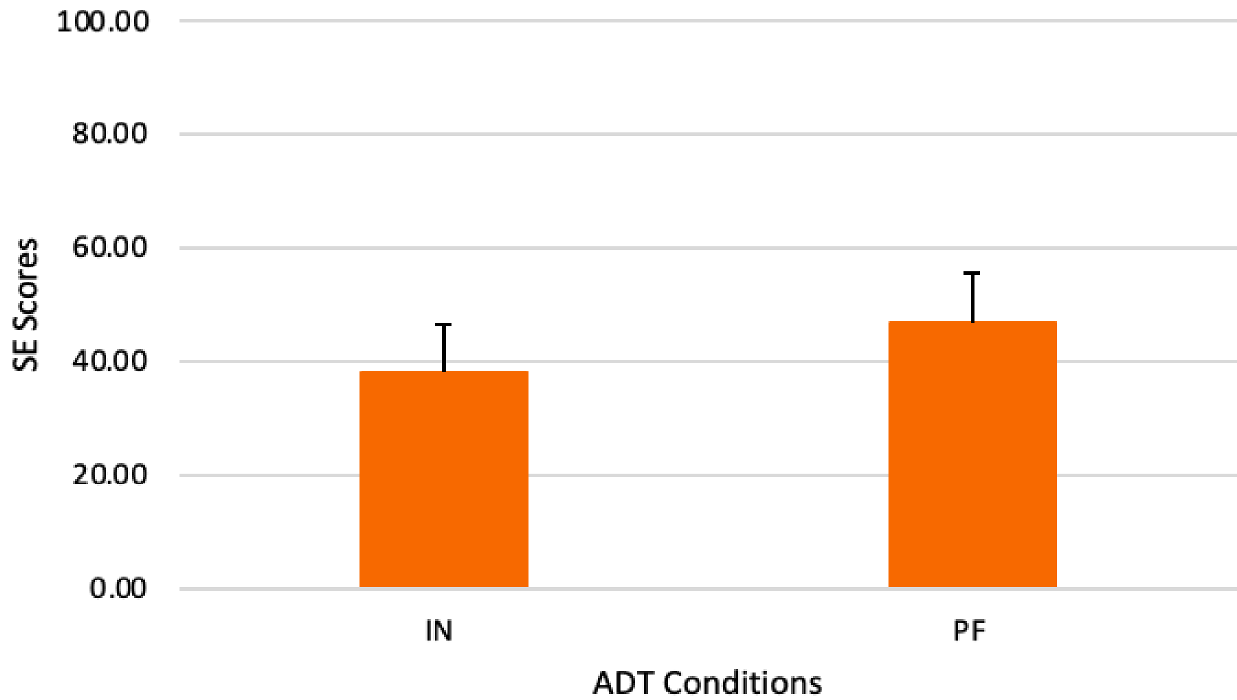

Figure 4.12. Environment Main Effect for the Injured-Limb.
SE = Sway Energy Scores; IN = Inversion; PF = Plantarflexion; ADT = Adaptation Test.
* indicates significant differences ($P < 0.05$).

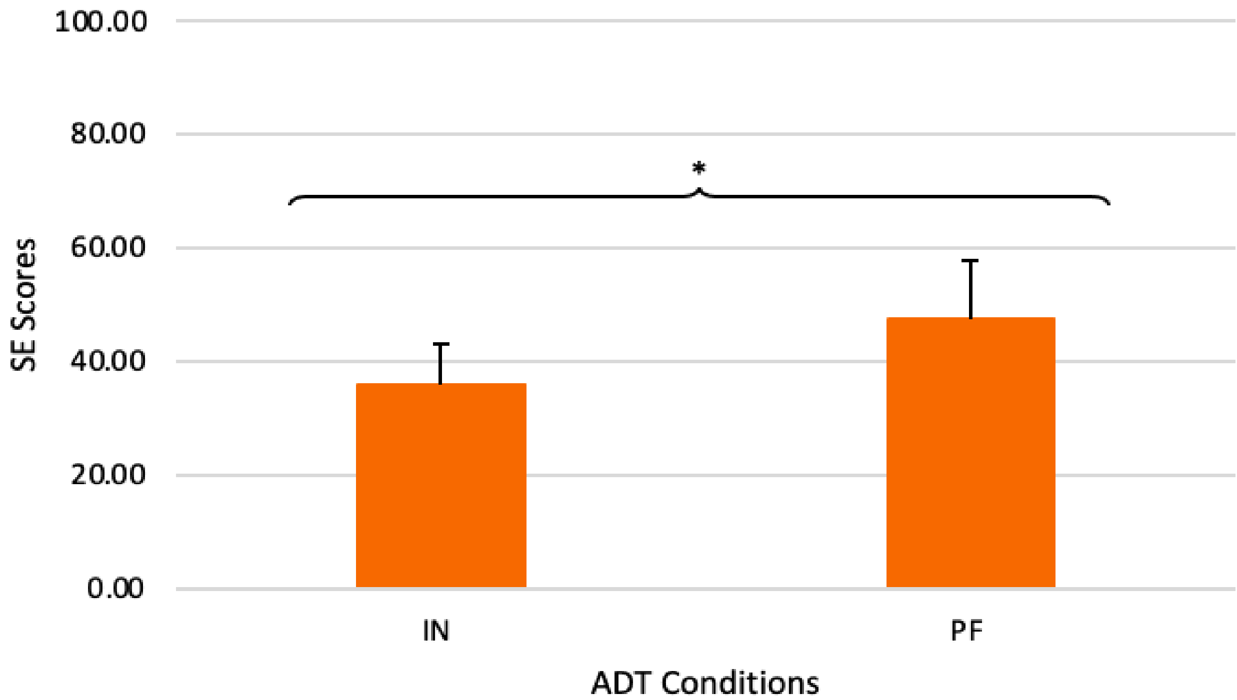

Figure 4.13. Environment Main Effect for the Uninjured-Limb.
SE = Sway Energy Scores; IN = Inversion; PF = Plantarflexion; ADT = Adaptation Test.
* indicates significant differences (P < 0.05).

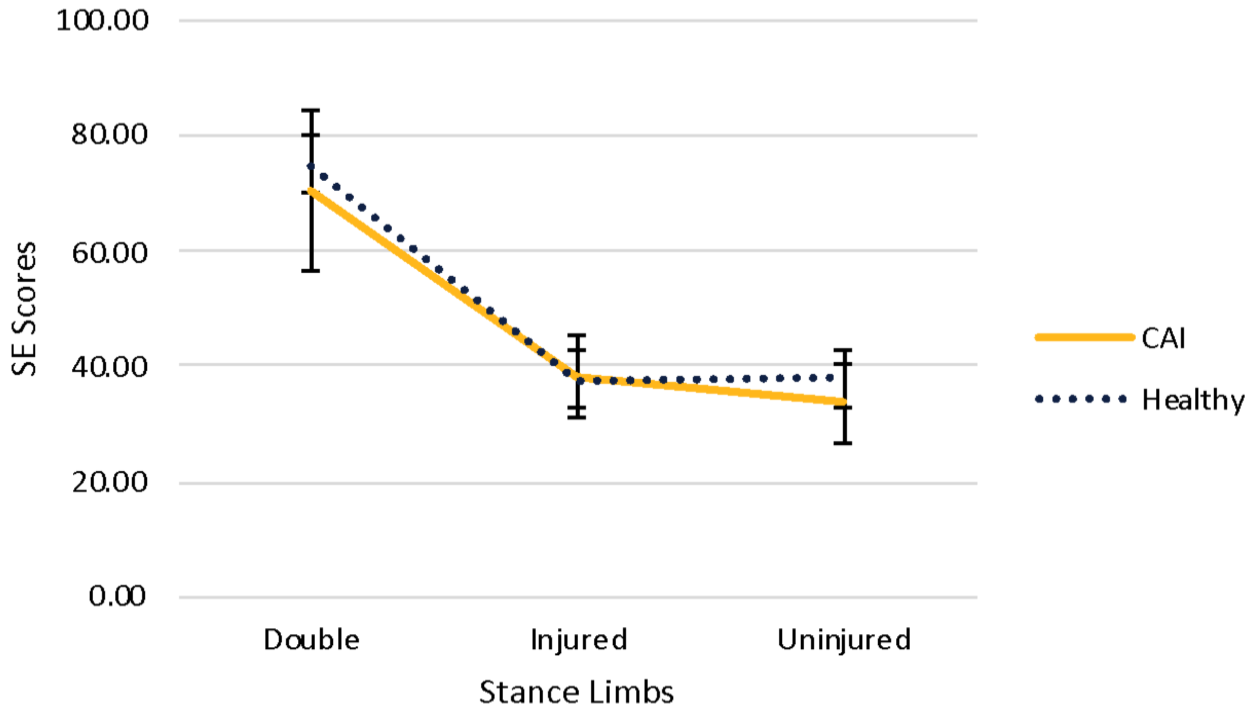

Figure 4.14. Group × Task Interactions for the ADT IN.
ADT = Adaptation Test; IN = Inversion; SE = Sway Energy Scores; CAI = Chronic Ankle Instability.

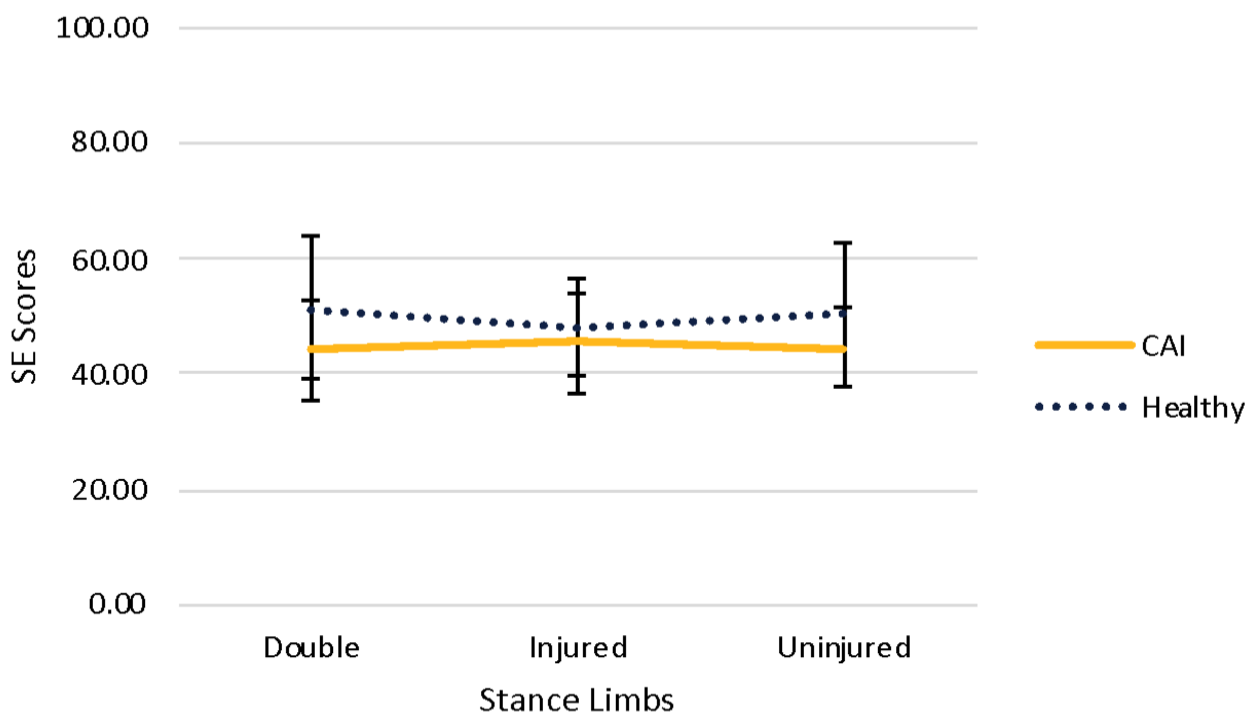

Figure 4.15. Group × Task Interaction for the ADT PF.
ADT = Adaptation Test; PF = Plantarflexion; SE = Sway Energy Scores; CAI = Chronic Ankle Instability.

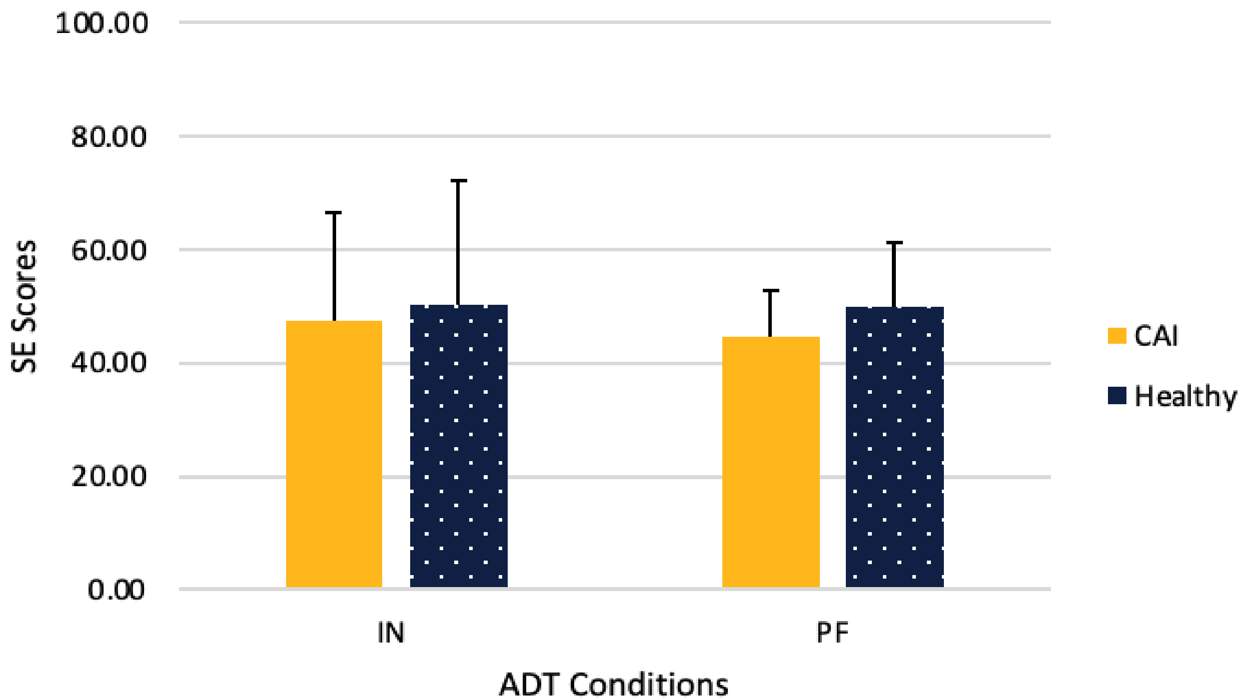

Figure 4.16. Group Main Effect for Individual ADT Conditions.
ADT = Adaptation Test; SE = Sway Energy Scores; IN = Inversion; PF = Plantarflexion; CAI = Chronic Ankle Instability.
* indicates significant group differences ($P < 0.05$).

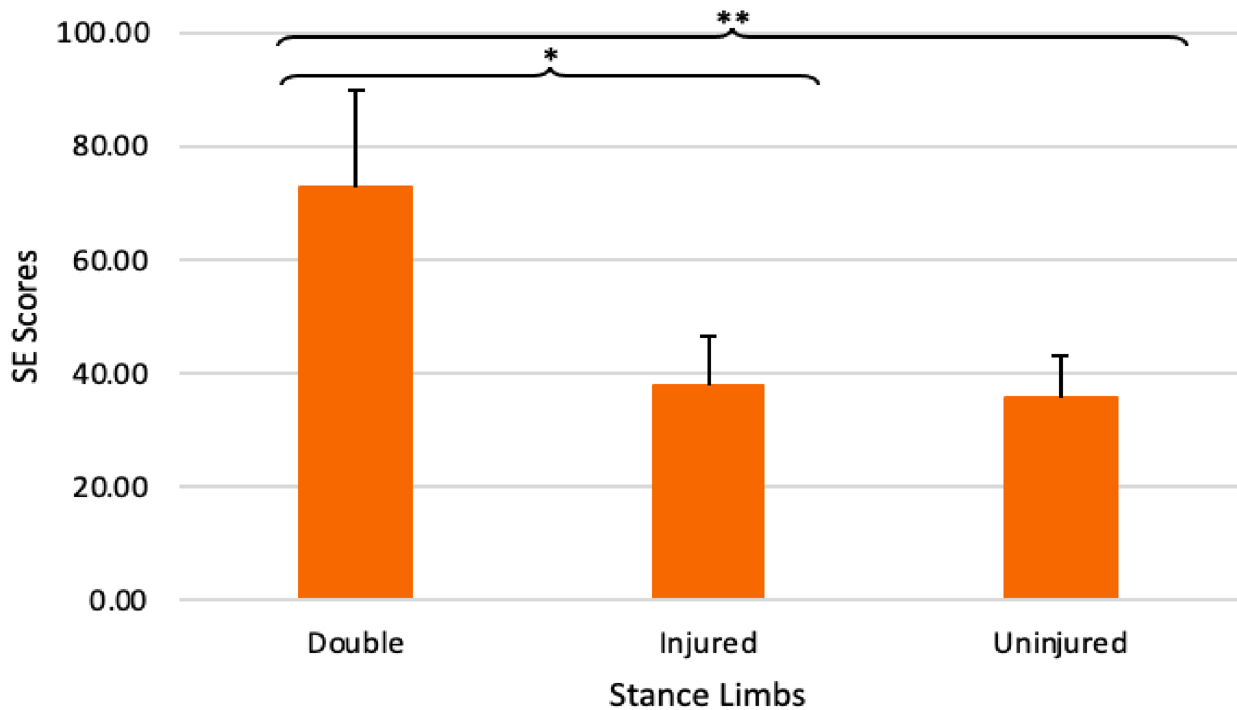

Figure 4.17. Task Main Effect for the ADT IN.
ADT = Adaptation Test; IN = Inversion; SE = Sway Energy.
* indicates significant differences between double- and the injured-limbs (P < 0.05).
** indicates significant differences between double- and the uninjured-limbs (P < 0.05).

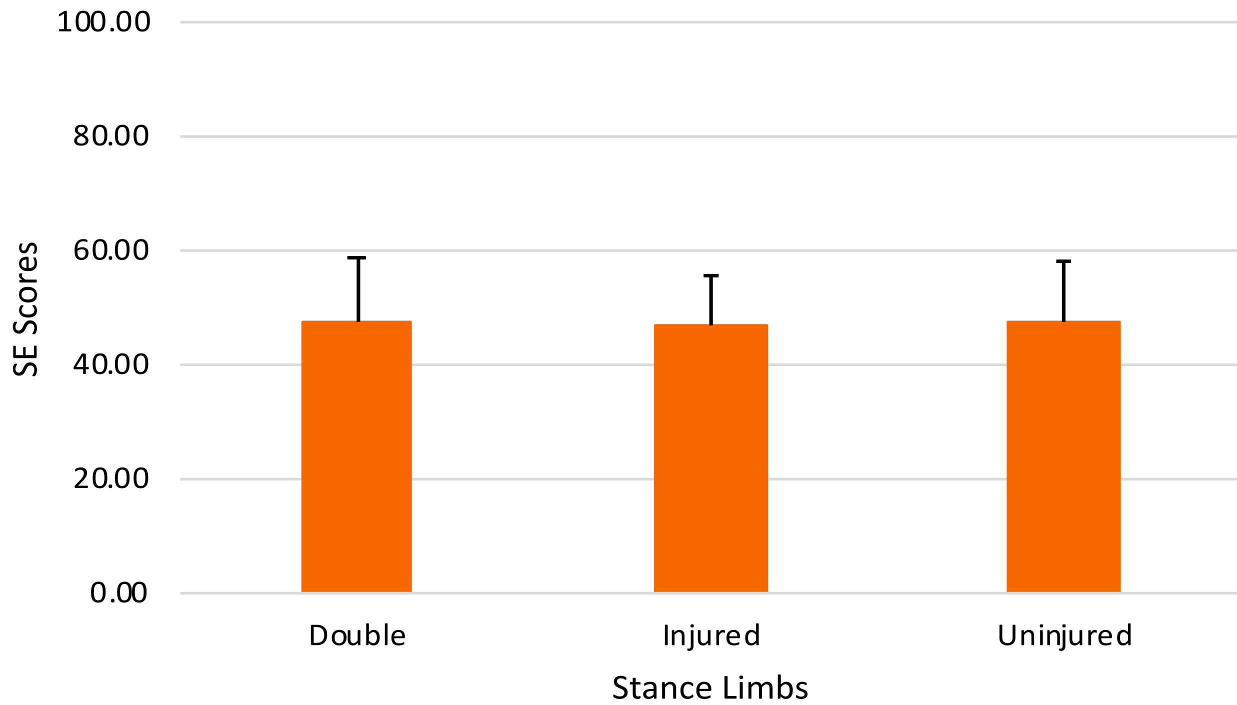

Figure 4.18. Task Main Effect for the ADT PF.
ADT = Adaptation Test; PF = Plantarflexion; SE = Sway Energy.

CHAPTER VI

MANUSCRIPT III

Introduction

An ankle sprain is one of the most common musculoskeletal injuries among the general public and athletes at all levels participating in physical activities and athletics, affecting up to two million people in the United States annually (Medina McKeon & Hoch, 2019). Among ankle sprains, lateral ankle sprain (LAS) accounts for approximately 92% of physician office visits for ankle sprains, and more than 628,000 LAS are treated in the emergency department (Kannus & Renstrom, 1991; Waterman et al., 2010; Wikstrom et al., 2018). The majority of those individuals with a history of an initial LAS suffer residual symptoms and develop repetitive bouts of subsequent ankle instability known as chronic ankle instability (CAI) (Anandacoomarasamy & Barnsley, 2005; Thomas et al., 2017). Moreover, CAI significantly affects the quality of life, a decrease in physical activity, and the development of long-term deficits in activities of daily living (Arnold, et al., 2011). The consequent impairments are oftentimes reflected as postural control dysfunction and maladapted gait among CAI individuals.

Much research has been done on postural control, examining center-of-pressure (COP) excursion, area, and velocity in CAI. However, there has been inconsistent evidence identifying group differences in postural control deficits in single-limb stance associated with CAI compared to healthy controls (Tropp et al., 1984; Ross & Guskiewicz, 2004; Lee et al., 2006; Hale et al., 2007; Hubbard et al., 2007). The lack of consistency may be because traditional COP measures have been suggested not to specify temporal proximity to stability boundaries that individuals work to control posture in the upright stance (McKeon et al., 2008). In contrast, a nonlinear approach such as time-to-boundary and sample entropy (SampEN) has been suggested to show promise in detecting postural control deficits related to CAI and various given pathological conditions (Glass et al., 2014; Hoffman et al., 2017; Lee, et al., 2017). SampEN is designed to quantify the amount of regularity of fluctuations in time-series data and detects subtle physiological changes (i.e., neural control) with ever-changing environments and task goals. Therefore, nonlinear analysis of SampEN is theorized to quantify the flexibility and adaptability of sensorimotor systems underlying postural control (Stergiou & Decker, 2011). Deploying the nonlinear approach of SampEN as a dependent variable takes into account the incredibly complex nature of the human body with multiple network interactions among internal and external properties, which traditional COP measures previously overlooked.

In order for individuals to achieve optimal performance in an unpredictable ever-changing environment, they must be able to perceive relevant information about the environment and task demands using various combinations of somatosensory, visual, and vestibular information. However, an initial ankle sprain induces somatosensory dysfunction and CAI individuals exhibit a less flexible and adaptable sensory-motor feedback loop circuit (Freeman, 1965; Freeman et al., 1965). Those physiological characteristics of CAI may cause individual perception to either diversify or isolate sensory systems. Indeed, CAI individuals present heavy reliance on visual information to maintain posture in a single-limb stance compared to healthy controls (Song et al., 2016). The degree of diversity or isolation of sensory systems reflects the regularity (readiness) and adaptability of movement. For instance, the greater degree of sensory diversity (multisensory) provides complex information, whereas the greater degree of sensory isolation (unisensory) provides limited information for the central nervous system (CNS) to configure coordinating movement. Therefore, too much and too few interactions among physiological elements (e.g., sensory systems) affect individuals' adaptive capability to task demands and environmental constraints resulting in either more random or restricted (rigid) movement patterns.

Current evidence suggests CAI individuals display rigid postural control during double- and single-limb stances compared to healthy controls (Glass et al., 2014; Raffalt et al., 2019). Rigid movement patterns are also reported in CAI during more dynamic tasks, such as walking (Terada et al., 2015). This movement rigidity found in CAI may interfere with performance, especially with increased task demands and environmental constraints, contributing to a risk of recurrent ankle sprains. However, it is currently unclear how movement (neural mechanisms) underlying postural control emerge with an increase in complexity of environmental and task constraints with manipulation of somatosensory, visual, and vestibular feedback. Therefore, the purpose of this study was to evaluate movement variability of COP excursion to examine underlying biological noise pertaining to postural control during an increased environmental and task complexity, with manipulation of sensory feedback in individuals with and without CAI.

Methods

Participants

A total of 22 participants with CAI (13 females, 9 males) and 22 healthy controls (13 females, 9 males) without a history of CAI were voluntarily enrolled in this study. Participants' demographics are shown in Table 6.1. Participants were recruited from local universities and communities and pre-screened for inclusion and exclusion criteria to participate in the study. Participants who met the following criteria identified by the International Ankle Consortium position statement were assigned to the CAI group: 1) had a history of at least one significant ankle sprain at least 12-months prior to study enrollment, 2) had a history of 2 episodes of previously injured ankle joint “giving way” and/or “feelings of instability” within the past 6-months and/or had a history of recurrent ankle sprains, 3) had scored $>$ 11 on the Identification of Functional Ankle Instability (IdFAI), and 4) had not experienced recurrent ankle sprains in the last 3-months (Gribble et al., 2013; Gribble et al., 2014a, 2014b). Participants who did not meet those CAI criteria and scored $\leq$ 11 on the IdFAI were allotted to the healthy controls group (Simon et al., 2012). Individuals who reported any of the following conditions were excluded from the study: ongoing inflammatory symptoms; a history of surgeries in the brain and/or on the lower extremity (i.e., foot, ankle, knee, hip, lower back); medically diagnosed concussion at least six months prior to study enrollment; neurological, vestibular, and/or visual disorders and disease (e.g., vertigo, epilepsy, stroke, peripheral neuropathies); connective tissue disease and/or disorders (e.g., rheumatoid arthritis, Marfan syndrome, Ehlers-Danlos syndrome); chronic musculoskeletal conditions (e.g., OA, ACL deficiency). If participants presented bilateral CAI, we chose the ankle with the worst score on the IdFAI. Healthy controls were matched by age (years, $\pm$ 2), sex, height (cm, $\pm$ 5%), weight (kg, $\pm$ 3%), physical activity level (National Aeronautics and Space Administration Physical Activity Status Scale [NASA-PASS], $\pm$ 1), and limb dominance (the leg used to kick a ball) to the CAI individuals, identifying the injured-limb. All participants read and signed an informed consent form approved by the Institutional Review Board at the University of North Carolina Greensboro before participating in the study.

Procedure

After providing informed consent, participants completed a standardized health history questionnaire package including questions on the previous injury on the lower extremity, self-reported ankle instability and function (i.e., IdFAI), and physical activity status (i.e., NASA-PASS). Participants warmed up for 5-minutes on a bike at a self-selected intensity, were scaled for height and weight, performed joint hypermobility tests, and completed lower extremity anatomical alignment measures (ankle, knee, hip). We used the NeuroCom sensory organization test (SOT) (SMART EquiTest, NeuroCom International Inc., Clackamas, OR) to examine postural control (Figure 6.1).

Sensory Organization Test

Participants were positioned barefoot on the NeuroCom dynamic posturography platform. We outfitted participants with a vest attached to the safety harness of the NeuroCom to assure safety and prevent injury from falls. Individuals' bilateral ankles (medial malleoli) were positioned perpendicular to the transverse axis of the platform rotation with a standardized distance apart based on their height according to manufacture SOT guidelines for double-limb stance (Figure 6.2) and positioning unilateral foot right center of the platform, respectively for single-limb stance (Figure 6.3). All participants were instructed to maintain their face forward, their arms relaxed by their sides, and motionless as possible while completing the SOT (Figure 6.4). Participants completed six SOT conditions (three 20-second trials each, a total of 18-trials per stance limb) in individual stance limbs (double, injured, uninjured). The six SOT conditions are designed to systematically manipulate somatosensory and vestibular feedback by altering the sway-referenced support surface and visual surroundings in conjunction with the elimination of vision in a double-limb stance (Table 6.2, Figure 6.5). Participants were given a 30-second rest between trials and a 1-minute rest between conditions. At the completion of each stance limb (e.g., double-limb stance), an additional 1-minute rest was provided before the next testing stance (e.g., single-limb stance) was initiated following a counterbalanced order within the group (CAI, healthy controls) to minimize the learning effect. All participants were allowed to quickly tap down the platform multiple times with non-stance toes while performing the SOT in single-limb stance (injured, uninjured) after 10-seconds to complete the full 20-second trials. However, we encouraged participants to do their absolute best to complete each 20-second trial. The trials were stopped and repeated if participants tapped down on non-stance toes before 10-seconds and/or completely fell and stood on a non-stance limb after 10-seconds.

Movement Variability Measure

The SOT of the NeuroCom sampled COP coordinates for anteroposterior (AP) and mediolateral (ML) components at 100 Hz for every three trials of individual SOT conditions. The raw data from the NeuroCom were exported to spreadsheets (Excel, version 360; Microsoft Corporation, Redmond, WA) and imported into a custom R program in RStudio (version 4.0.0; RStudio, Inc., Boston, MA) to compute path length following equation 6.1, where N represents the number of data points and i is each successive data point (Rhea et al., 2014). Path length was calculated by summing the magnitude of the distance change of the COP at every time point from the resultant vector created from the combined COP AP and ML time based on the first 10-second trials (Rhea et al., 2014).

The variety of m (2 to 6) and r (0.01 to 0.25) value combinations were examined based on guidelines defined by Lake et al. (2002) to obtain parameters m (template length) and r (tolerance level). The efficacy of the metric was derived by setting the maximal relative error less than 0.05, corresponding to a 95% confidence interval, which is a 10% sample entropy estimation. We selected the values of $m = 3$ data points and $r = 0.2 \times SD$ of the time series for COP path length. The SampEN values were then computed using a custom R program based on an algorithm proposed by Richman and Moorman (2000). SampEN values typically range from 0 to 2 in human biological systems. Both lower and greater SampEN values correspond to less flexibility and adaptability of neural control in sensorimotor pathways (Stergiou & Decker, 2011). However, lower SampEN values result in more predictable rigid movement patterns, whereas, the greater SmpEN Values result in more unpredictable random movement patterns (Stergiou & Decker, 2011; Yentes et al., 2013, 2018).

$$Path\ Length: \sum_{i=1}^{N=1} \sqrt{(AP_{i+1} - AP_i)^2 + (ML_{i+1} - ML_i)^2} \qquad (Equation\ 6.1)$$

Statistical Analysis

Independent t-tests were used to compare between-group differences in demographic characteristics (age, height, weight), physical activity level (NASA-PASS), and IdFAI scores. Separate 2 (group) × 6 (environment: SOT conditions 1-6) repeated measures analysis of variance (ANOVA) were performed, examining environmental constraints effect on neural mechanisms underlying postural control for individual limbs (double, injured, uninjured). Similarly, separate 2 (group) × 3 (task: double-, injured-, uninjured-limbs) repeated measures ANOVA were performed examining task constraints effect on neural mechanisms underlying postural control for individual SOT conditions (1-6). Tukey post-hoc analyses were applied if significant interactions were found. Cohen's *d* effect sizes with 95% confidence intervals were calculated to estimate the magnitude of significant group differences with values of ≤ 0.40 interpreted as small, 0.40–0.80 as moderate, and ≥ 0.80 as large effects (Cohen, 1988). All statistical analyses were performed using SPSS software (version 27; IBM Corp, Armonk, NY, USA) with a priori α level of 0.05.

Results

No significant group differences were detected on age, height, weight or physical activity level (*t* range = 0.39-0.94; $P > 0.05$; Table 6.2). CAI individuals had a greater number of ankle sprains, episodes of giving-way within the past 6-months, higher IdFAI scores compared to healthy controls ($P < 0.05$; Table 6.2).

Movement Variability in Increased Environment (SOT Conditions) Constraints:

There was no significant group by environment interactions for individual stance limbs (double: $F_{5,210} = 0.62$; injured: $F_{5,210} = 0.39$; uninjured: $F_{5,210} = 1.27$; $P > 0.05$; Tables 6.3-6.5; Figures 6.6-6.8). A significant main effect for group was found in the injured- ($F_{1,42} = 4.98$, $P = 0.031$, Figure 6.9) and uninjured-limbs ($F_{1,42} = 12.23$, $P = 0.001$, Figure 6.9), but not in double-limb stance ($F_{1,42} = 2.75$, $P > 0.05$, Figure 6.9), indicating lower SampEN values for CAI individuals in both the injured- (CAI: 1.48 $\pm$ 0.03; Healthy: 1.56 $\pm$ 0.03; Figure 6.8) and uninjured-limbs (CAI: 1.48 $\pm$ 0.03; Healthy: 1.60 $\pm$ 0.03; Figure 6.8) compared to healthy controls. A significant main effect for environment was found in individual stance limbs: double- ($F_{5,210} = 243.58$, $P = 0.000$, Figure 6.10), injured- ($F_{5,210} =318.14$, $P = 0.000$, Figure 6.11), and uninjured-limbs ($F_{5,210} = 316.26$, $P = 0.000$, Figure 6.12). Tukey post-hoc test revealed a significant difference in SampEN values across all combinations of six SOT conditions in double- (Mean $\pm$ SD range = 1.01 $\pm$ 0.02-1.35 $\pm$ 0.02; $P < 0.001$; Figure 6.10), injured- (Mean $\pm$ SD range = 1.27 $\pm$ 0.0-1.81 $\pm$ 0.02; $P < 0.001$; Figure 6.11), and uninjured-limbs (Mean $\pm$ SD range = 1.29 $\pm$ 0.02-1.83 $\pm$ 0.02; $P < 0.001$; Figure 6.12) except between conditions 2 and 3 in double-limb stance ($P > 0.05$, Figure 6.10).

Movement Variability in Increased Task (Stance Limbs: Double, Injured, and Uninjured) Constraints:

There was a significant group by task interactions for SOT conditions 1 ($F_{2,84}$ = 3.41, *P* = 0.038, Table 6.6, Figure 6.13) and 5 ($F_{2,84}$ = 3.59, *P* = 0.032, Table 6.7, Figure 6.14), but not for other SOT conditions (2, 3, 4, & 6: $F_{2,84}$ range = 0.96-1.78; *P* > 0.05; Tables 6.8-6.11; Figures 6.15-6.18). Tukey post-hoc test revealed a significant decrease in SampEN in the uninjured-limb (CAI: 1.23 $\pm$ 0.03, Healthy: 1.35 $\pm$ 0.03, Table 6.6, Figure 6.13) for the SOT condition 1 and in both the injured- (CAI: 1.76 $\pm$ 0.03, Healthy: 1.86 $\pm$ 0.03, Figure 6.14) and uninjured-limbs (CAI: 1.75 $\pm$ 0.03, Healthy: 1.92 $\pm$ 0.03, Table 6.7, Figure 6.14) for the SOT condition 5 among CAI individuals compared to healthy controls. Significantly lower SampEN values also were found for the CAI group in double-limb stance (condition 1 [C1]: 1.00 $\pm$ 0.03, Table 6.6, Figure 6.13; condition 5 [C5]: 1.32 $\pm$ 0.02, Table 6.7, Figure 6.14) compared to the injured- (C1: 1.24 $\pm$ 0.03, Table 6.6, Figure 6.13; C5: 1.76 $\pm$ 0.03, Table 6.7, Figure 6.14) and uninjured- limbs (C1: 1.23 $\pm$ 0.03, Figure 6.13; C5: 1.75 $\pm$ 0.03, Table 6.7, Figure 6.14). Significant main effects for group ($F_{1,42}$ range = 4.76-11.01; *P* range = 0.002-0.009; Figure 6.19) and task ($F_{2,84}$ range = 125.13-780.34, *P* = 0.000; Figures 6.20-6.25) for all six SOT conditions were found. The group main effect revealed CAI individuals presented significantly lower SampEN values than healthy control while completing individual SOT conditions (Figure 19). Additionally, the Tukey post-hoc test for task main effect revealed significantly lower SampEN values in double-limb stance compared to the injured- (*P* =

0.000) and uninjured-limbs ($P = 0.000$), respectively, across all SOT conditions (Figures 6.20-6.25).

Discussion

The primary finding was that CAI individuals presented lower movement variability compared to healthy controls. The unique finding of the current study was that group differences in movement variability depended on task constraints, but not on environmental constraints. Specifically, CAI individuals revealed lower movement variability in the uninjured-limb for the SOT condition 1 and both the injured- and uninjured-limbs for the SOT condition 5 compared with healthy controls. Significantly lower movement variability was also noted for the CAI group in double-limb stance compared to the injured- and uninjured-limbs, respectively, for SOT conditions 1 and 5. The main effect found for environmental constraints revealed the lowest movement variability when all sensory feedback was intact and displayed the largest movement variability when somatosensory feedback was manipulated with the elimination of visual feedback across individual stance limbs (double, injured, uninjured). Whereas for the group main effect, CAI individuals displayed lower movement variability in the injured- and uninjured-limbs regardless of SOT conditions compared with healthy controls.

The lower movement variability displayed among CAI individuals in our study remains consistent with current literature. There are only a few studies that implement nonlinear measures (i.e., SampEN). Similar to our findings, those studies have consistently reported decreased variability on COP variables (velocity, excursion) while maintaining posture in single- and double-limb stances compared to healthy controls

(Glass et al., 2014; Raffalt et al., 2019). Decreased variability is also reported with more dynamic tasks in which CAI individuals displayed decreased stride-to-stride gait variability in frontal-plane ankle kinematics while walking at a self-selected speed (Terada et al., 2015). Specific to gait kinematics, CAI individuals are known to present lower vertical foot-floor clearance during the terminal swing phase of gait, along with excessive inversion of the foot compared to healthy controls (Delahunt et al., 2006). Hence, researchers implied decreased movement variability found during both static and dynamic tasks is a result of compensatory mechanisms in which CAI individuals constricted the degree-of-freedom at the ankle to coordinate a stable motor solution (Glass et al., 2014; Terada et al., 2015).

Humans are complex with inherent variability in neurobiological systems, allowing individuals to coordinate motor behaviors in an ever-changing environment to accomplish a given task. Therefore, movement variability is like the various choices in the toolbox, yielding flexible strategies to adapt to various constraints. Variability typically increases at the beginning of motor learning, plateaus when new skills and behavior emerge, then increases once again when individuals become experts in the motor skills (Skinner, 1981). Consequently, significant main effects found for the environment and task in our study may be a reflection of the skill acquisition phase. For the environment main effect, variability increased as manipulation of sensory feedback became more complex in the SOT condition 5 than when all sensory feedback was intact in the SOT condition 1. Similarly, for task main effect and its impact on group differences, variability increased as task complexity transitioned from double- to single-limb (injured, uninjured) stances, especially for individuals with CAI. Overall, we contend increased variability in the current study implies an explanatory phase of our participants experiencing the SOT for the first time, searching to coordinate task-and-environment specific postural control.

The status of health such as whether individuals are healthy or injured also reflects on movement variability (Stergiou & Decker, 2011). Current literature suggests skillful and/or healthy individuals have flexible strategies to adapt dynamically shifting environmental and task constraints, generating stable movement patterns (Cavanaugh et al., 2006; Stergiou & Decker, 2011). In contrast, less-skilled and/or injured individuals (e.g., concussion) present a loss of flexible strategies to adapt constraints, exhibiting either unstable (random) or highly stable (rigid) movement patterns (Cavanaugh et al., 2005; Stergiou & Decker, 2011). Harborne and Stergiou (2009) have proposed a theoretical model that explains movement variability, associated with mature motor skills and status of health in a concept of movement predictability. Their inverted-U-shaped theoretical model (Figure 6.26) implies that optimal movement variability reflects flexibility and adaptability of neural control in sensorimotor systems of healthy individuals (Stergiou & Decker, 2011). More specifically, the uppermost point of the inverted-U shape refers to optimal variability considered to be a healthy state. Conversely, the point below the uppermost point of the inverted-U shape designates suboptimal variability associated with a lack of health (Harborne & Stergiou, 2009). The decrease in optimal variability renders more predictable (rigid) motor behaviors like a robot, whereas an increase in optimal variability renders unpredictable (random) motor behaviors like a frail baby. Consequently, either too much or too little movement variability results in less flexible adaptation in sensorimotor systems to environmental and task constraints (Stergiou & Decker, 2011). This evidence may suggest the development of CAI in our participants has disrupted the healthy state of optimal variability, lowering SampEN values among CAI individuals compared to healthy controls. Furthermore, not only limited to CAI, decreased SampEN values specific to postural control and gait mechanics have been reported in individuals with neurological deficits (i.e., proprioceptive impairments, concussion, ACL injury) (Cavanaguh et al., 2005; Moraiti et al., 2007; Manor et al., 2010; Sosnoff et al., 2011).

Our study revealed CAI individuals decreased movement variability only in the uninjured-limb during the SOT condition 1, yet displayed decreased movement variability in both the uninjured- and injured-limbs during the SOT condition 5 compared to healthy controls. This may be an indication of supraspinal alteration in CAI. Pietrosimone and Gribble (2012) who examined corticospinal excitability of peroneus longus (PL) muscles revealed CAI individuals required greater stimuli to excite the PL in both the injured- and uninjured-limbs compared to healthy controls. Although the study was conducted in a seated position, it may support our hypothesis to register that decreased movement variability we observed in the uninjured-limb during postural control reflects centrally mediated compensatory alterations that CAI employed through activities of daily living, such as gait. Furthermore, freezing both uninjured- and injured-limbs may have been the only compensatory postural strategy available to maintain stability, transitioning from simple to more complex task constraints when CAI individuals were required to rely on vestibular feedback during the SOT condition 5 (manipulation of somatosensory feedback with the elimination of vision) in the current study.

Researchers surmise that the long-time reduction in variability may result in abnormal configurations of the sensory cortex (Merzenich et al., 1993; Nudo et al., 1996; Byl et al., 2002). Although whether the abnormal configuration of the sensory cortex pre-existed prior to sustaining an injury is still unknown, several studies reported a change in sensory feedback configuration by increasing dependence on visual feedback among individuals with somatosensory and vestibular deficits (Cooke et al., 1978; Hafstrom et al., 2004; Lopez et al., 2006; Slaboda et al., 2009; Bonan et al., 2013; Lin et al., 2019). Similarly, a meta-analysis suggested CAI individuals upweighted their visual feedback as a compensatory mechanism for somatosensory deficits while maintaining posture in the injured-limb compared to healthy controls (Song et al., 2016). In addition, somatosensory deficits in CAI individuals are firmly confirmed by studies displaying diminished joint position and force senses compared to healthy controls (Jerosch et al., 1995; Konradsen & Magnusson, 2000; Docherty & Arnold, 2008; Nakasa et al., 2008; Munn et al., 2010; Simon et al., 2014). Diminished proprioceptive feedback from the injured-limb and abnormal sensory cortex configurations may be another factor resulting in centrally mediated alterations in the uninjured-limb, an outcome existing in our data. For instance, postural control deficits are not only displayed in the injured-limb but exhibited in the uninjured-limb in individuals with unilateral CAI (Lee et al., 2006).

Altered postural control strategies at proximal knee joints are also reported with more dynamic functional tests such as jump landing tasks with and without altering visual focus in CAI (Caulfield & Garrett, 2002; Gribble & Robinson 2009, 2010). Caulfield and Garrett (2002) reported increased knee flexion right before and after landing when they instructed participants to focus on the landing platform in CAI compared to healthy controls. Conversely, Gribble and Robinson (2009, 2010) noted a bilateral decrease in knee flexion before (2010) and at (2009) landing along with longer time-to-stabilization when instructing individuals with unilateral CAI to focus on a vertical object while completing the jump landing task. Visual feedback aids other peripheral sensory feedback (somatosensory, vestibular) in obtaining exteroceptive information of the environment to fine-tune motor control acuity (Greeno, 1994; Turvey, 1990, 2007). Thus, the inability to focus on landing platforms throughout the jump landing task manipulating visual feedback may have led CAI individuals to freeze the degree-of-freedom at the knee, which mediates ankle and hip joints, to provide dynamic stability. The NeuroCom SOT systematically manipulates individuals' sensory feedback in a combination of the sway-referenced support surface and visual surroundings with and without eliminating vision, creating somatosensory and visual feedback conflicts (Nashner, 1982). Therefore, we speculate CAI individuals in our study restricted the degree-of-freedom to perform the SOT successfully, especially the complexity of task constraints increased by controlling posture in the uninjured- and injured-limbs during SOT conditions 1 and 5.

There were a few limitations to this study. We had five CAI individuals with a self-reported history of bilateral ankle sprains. Therefore, decreased movement variability displayed in the uninjured-limb among CAI may be driven by those individuals with a history of bilateral CAI. Another limitation was that we had individuals who have participated in various sports, not only limited to the collegiate level. Therefore, the results may not be applicable for individuals who have different sports histories within or outside the recruited age group. Lastly, we can only speculate centrally mediated changes potentially resulting in the uninjured-limb were not present before sustaining an initial ankle sprain and/or induced by the development of CAI.

Conclusion

Group differences in movement variability during postural control depended on task constraints. CAI individuals demonstrated decreased strategies in postural control in the uninjured-limb when all sensory feedback was intact, whereas in both the uninjured- and injured-limbs when they were forced to rely on vestibular feedback while manipulating somatosensory feedback with the elimination of vision. Future studies should investigate the contribution of vestibular feedback to postural control and its relation to movement variability. For clinicians, it is important to recognize when to increase the complexity of task constraints to optimize rehabilitation protocols to prevent subsequent ankle sprains in individuals with CAI.

Table 5.1. Participant Demographics (Mean ± SD).

	Group		
	CAI	**Controls**	***P*-values**
N	22 (13 females, 9 males)	22 (13 females, 9 males)	-
Age (years)	26.09 ± 5.76	25.41 ± 5.92	0.84
Height (cm)	172.25 ± 9.76	169.70 ± 9.32	0.61
Weight (kg)	76.18 ± 14.91	71.98 ± 14.79	0.93
NASA-PASS	6.27 ± 0.18	6.27 ± 1.03	0.67
IdFAI	19.09 ± 5.39	1.36 ± 1.81	< 0.001*
# of Ankle of Sprains	6.48 ± 7.08	0.00 ± 0.00	< 0.001*
Episodes of Giving-Way	8.88 ± 21.36	0.00 ± 0.00	< 0.001*

N = Number; NASA-PASS = National Aeronautics and Space Administration Physical Activity Status Scale; IdFAI = the Identification of Functional Ankle Instability; CAI = Chronic Ankle Instability.
* indicates significant differences ($P < 0.05$).

Table 5.2. Descriptions of SOT Conditions.

Conditions	Descriptions	Manipulated Sensory Inputs	Vision
Condition 1	Eyes-open, Fxied sway-referenced support surface	None	Available
Condition 2	Eyes-closed, Fxied sway-referenced support surface	None	Absent
Condition 3	Eyes-open, Fxied sway-referenced support surface, Sway-referenced visual surroundigs	Visual sensory input	Available
Condition 4	Eyes-open, Sway-referenced support surface	Somatosensory input	Available
Condition 5	Eyes-closed, Sway-referenced support surface	Somatosensory input	Absent
Condition 6	Eyes-open, Sway-referenced surface and visual surroundigs	Somatosensory and Visual inputs	Available

Table 5.3. Group × Environment Interactions for a Double-Limb Stance.

Double-Limb: Movement Variability (SampEN) Values (Mean ± SD) and Effect Sizes			
Parameter	**CAI**	**Control**	**Effect Sizes (95% CI)**
SOT C1	0.99 ± 0.13	1.04 ± 0.13	0.37 (-0.22 to 0.97)
SOT C2	1.03 ± 0.13	1.09 ± 0.11	0.45 (-0.14 to 1.05)
SOT C3	1.02 ± 0.16	1.08 ± 0.12	0.34 (-0.26 to 0.93)
SOT C4	1.05 ± 0.14	1.13 ± 0.13	0.59 (-0.01 to 1.20)
SOT C5	1.32 ± 0.10	1.39 ± 0.12	0.59 (-0.02 to 1.19)
SOT C6	1.25 ± 0.12	1.29 ± 0.10	0.39 (-0.21 to 0.99)

SampEN = Sample Entropy; SD = Standard Deviation; CI = Confidence Interval; SOT = Sensory Organization Test; C = Condition; CAI = Chronic Ankle Instability.

Table 5.4. Group × Environment Interactions for the Injured-Limb.

Injured-Limb: Movement Variability (SampEN) Values (Mean ± SD) and Effect Sizes			
Parameter	**CAI**	**Control**	**Effect Sizes (95% CI)**
SOT C1	1.24 ± 0.14	1.30 ± 0.12	0.50 (-0.10 to 1.10)
SOT C2	1.65 ± 0.14	1.73 ± 0.12	0.60 (0.00 to 1.20)
SOT C3	1.43 ± 0.14	1.50 ± 0.14	0.49 (-0.11 to 1.09)
SOT C4	1.30 ± 0.14	1.37 ± 0.12	0.55 (-0.05 to 1.15)
SOT C5	1.76 ± 0.17	1.86 ± 0.13	0.65 (0.04 to 1.26)
SOT C6	1.50 ± 0.15	1.60 ± 0.13	0.68 (0.07 to 1.29)

SampEN = Sample Entropy; SD = Standard Deviation; CI = Confidence Interval; SOT = Sensory Organization Test; C = Condition; CAI = Chronic Ankle Instability.

Table 5.5. Group × Environment Interactions for the Uninjured-Limb.

Uninjured-Limb: Movement Variability (SampEN) Values (Mean ± SD) and Effect Sizes			
Parameter	**CAI**	**Control**	**Effect Sizes (95% CI)**
SOT C1	1.23 ± 0.14	1.35 ± 0.10	0.94 (0.31 to 1.56)
SOT C2	1.64 ± 0.14	1.76 ± 0.12	0.97 (0.34 to 1.59)
SOT C3	1.44 ± 0.14	1.55 ± 0.14	0.84 (0.22 to 1.46)
SOT C4	1.28 ± 0.14	1.41 ± 0.12	1.01 (0.38 to 1.64)
SOT C5	1.75 ± 0.14	1.92 ± 0.15	1.18 (0.54 to 1.82)
SOT C6	1.53 ± 0.16	1.63 ± 0.17	0.59 (-0.01 to 1.19)

SampEN = Sample Entropy; SD = Standard Deviation; CI = Confidence Interval; SOT = Sensory Organization Test; C = Condition; CAI = Chronic Ankle Instability.

Table 5.6. Group × Task Interactions for the SOT Condition 1.

SOT Condition 1: Movement Variability (SampEN) Values (Mean ± SD) and Effect Sizes			
Parameter	**CAI**	**Control**	**Effect Sizes (95% CI)**
Double	1.04 ± 0.13	0.99 ± 0.13	-0.37 (-0.97 to 0.22)
Injured	1.30 ± 0.12 ‡	1.24 ± 0.14	-0.50 (-0.10 to 1.10)
Uninjured	1.35 ± 0.10 ‡	1.23 ± 0.14 †	-0.94 (-1.56 to 0.34)

SampEN = Sample Entropy; SD = Standard Deviation; CI = Confidence Interval; SOT = Sensory Organization Test; C = Condition; CAI = Chronic Ankle Instability.
† indicates significant differences between groups ($P < 0.05$).
‡ indicates significant differences between double- and the injured-limbs and double- and the uninjured limbs for the CAI group ($P < 0.05$).

Table 5.7. Group × Task Interactions for the SOT Condition 5.

SOT Condition 5: Movement Variability (SampEN) Values (Mean ± SD) and Effect Sizes			
Parameter	**CAI**	**Control**	**Effect Sizes (95% CI)**
Double	1.32 ± 0.10	1.39 ± 0.12	0.59 (-0.01 to 1.19)
Injured	1.76 ± 0.17 ‡	1.86 ± 0.13 †	0.65 (0.04 to 1.26)
Uninjured	1.75 ± 0.14 ‡	1.92 ± 0.15 †	1.18 (0.54 to 1.82)

SampEN = Sample Entropy; SD = Standard Deviation; CI = Confidence Interval; SOT = Sensory Organization Test; C = Condition; CAI = Chronic Ankle Instability.
† indicates significant differences between groups ($P < 0.05$).
‡ indicates significant differences between double- and the injured-limbs and double- and the uninjured-limbs for the CAI group ($P < 0.05$).

Table 5.8. Group × Task Interactions for the SOT Condition 2.

SOT Condition 2: Movement Variability (SampEN) Values (Mean ± SD) and Effect Sizes			
Parameter	**CAI**	**Control**	**Effect Sizes (95% CI)**
Double	1.09 ± 0.11	1.03 ± 0.13	-0.45 (-1.05 to 0.15)
Injured	1.73 ± 0.12	1.65 ± 0.14	-0.60 (-1.20 to 0.00)
Uninjured	1.76 ± 0.12	1.64 ± 0.14	-0.97 (-1.59 to -0.34)

SampEN = Sample Entropy; SD = Standard Deviation; CI = Confidence Interval; SOT = Sensory Organization Test; C = Condition; CAI = Chronic Ankle Instability.

Table 5.9. Group × Task Interactions for the SOT Condition 3.

SOT Condition 3: Movement Variability (SampEN) Values (Mean ± SD) and Effect Sizes			
Parameter	**CAI**	**Control**	**Effect Sizes (95% CI)**
Double	1.02 ± 0.12	1.07 ± 0.136	0.34 (-0.26 to 0.93)
Injured	1.43 ± 0.14	1.50 ± 0.14	0.49 (-0.11 to 1.09)
Uninjured	1.44 ± 0.14	1.55 ± 0.14	0.84 (-0.22 to 1.46)

SampEN = Sample Entropy; SD = Standard Deviation; CI = Confidence Interval; SOT = Sensory Organization Test; C = Condition; CAI = Chronic Ankle Instability.

Table 5.10. Group × Task Interactions for the SOT Condition 4.

SOT Condition 4: Movement Variability (SampEN) Values (Mean ± SD) and Effect Sizes			
Parameter	**CAI**	**Control**	**Effect Sizes (95% CI)**
Double	1.05 ± 0.14	1.13 ± 0.13	0.59 (-0.01 to 1.20)
Injured	1.30 ± 0.14	1.37 ± 0.12	0.55 (-0.05 to 1.15)
Uninjured	1.28 ± 0.14	1.41 ± 0.12	1.01 (0.38 to 1.64)

SampEN = Sample Entropy; SD = Standard Deviation; CI = Confidence Interval; SOT = Sensory Organization Test; C = Condition; CAI = Chronic Ankle Instability.

Table 5.11. Group × Task Interactions for the SOT Condition 6.

SOT Condition 6: Movement Variability (SampEN) Values (Mean ± SD) and Effect Sizes			
Parameter	**CAI**	**Control**	**Effect Sizes (95% CI)**
Double	1.25 ± 0.12	1.29 ± 0.10	0.39 (-0.21 to 0.99)
Injured	1.50 ± 0.15	1.60 ± 0.13	0.68 (0.07 to 1.29)
Uninjured	1.53 ± 0.16	1.63 ± 0.17	0.59 (-0.01 to 1.19)

SampEN = Sample Entropy; SD = Standard Deviation; CI = Confidence Interval; SOT = Sensory Organization Test; C = Condition; CAI = Chronic Ankle Instability.

Figure 5.1. The NeuroCom Dynamic Posturography System.

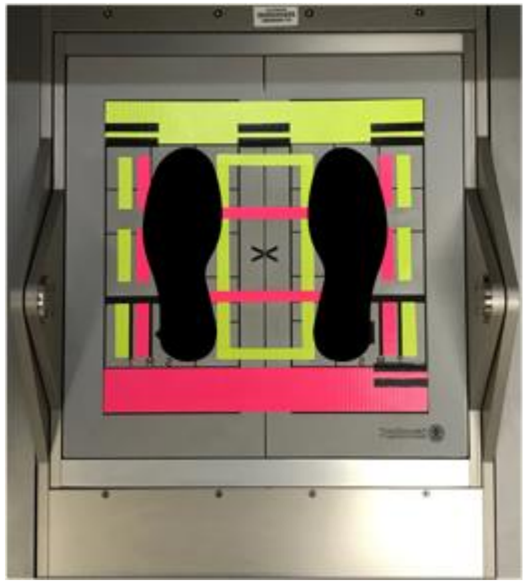

Figure 5.2. Foot Positions for the SOT in a Double-Limb Stance.

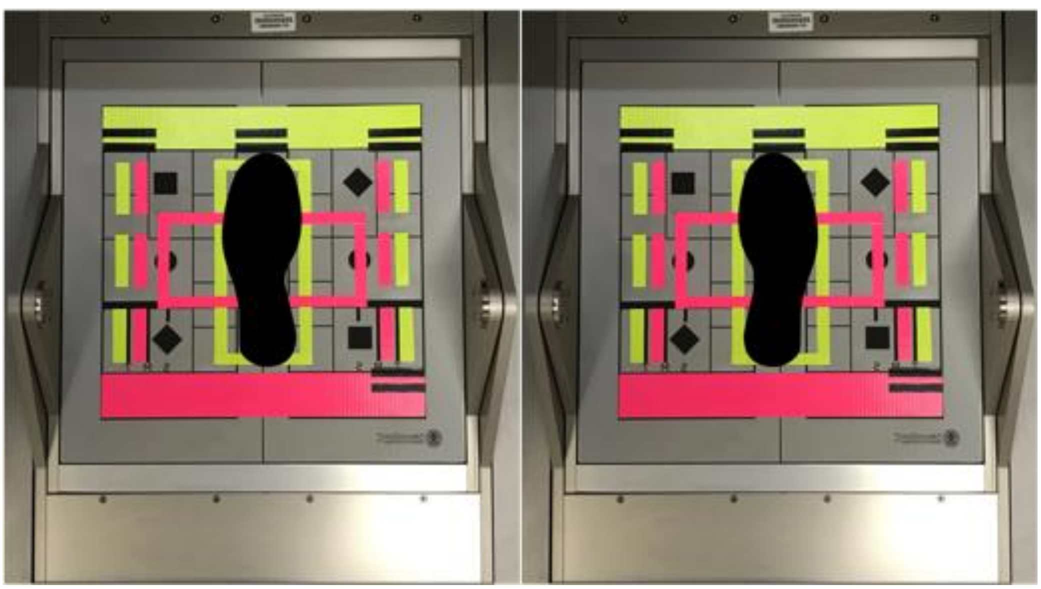

Figure 5.3. Foot Positions for the SOT in Single-Limb Stance (Injured, Uninjured).

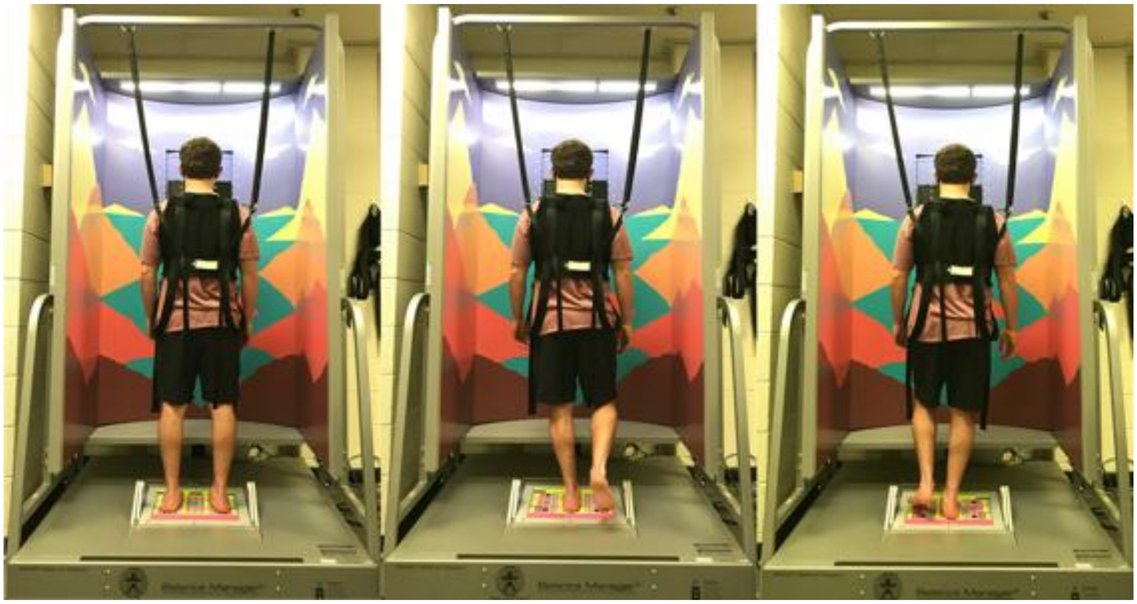

Figure 5.4. Stance Positions for the SOT in Double- and Single-Limb Stances on the NeuroCom.

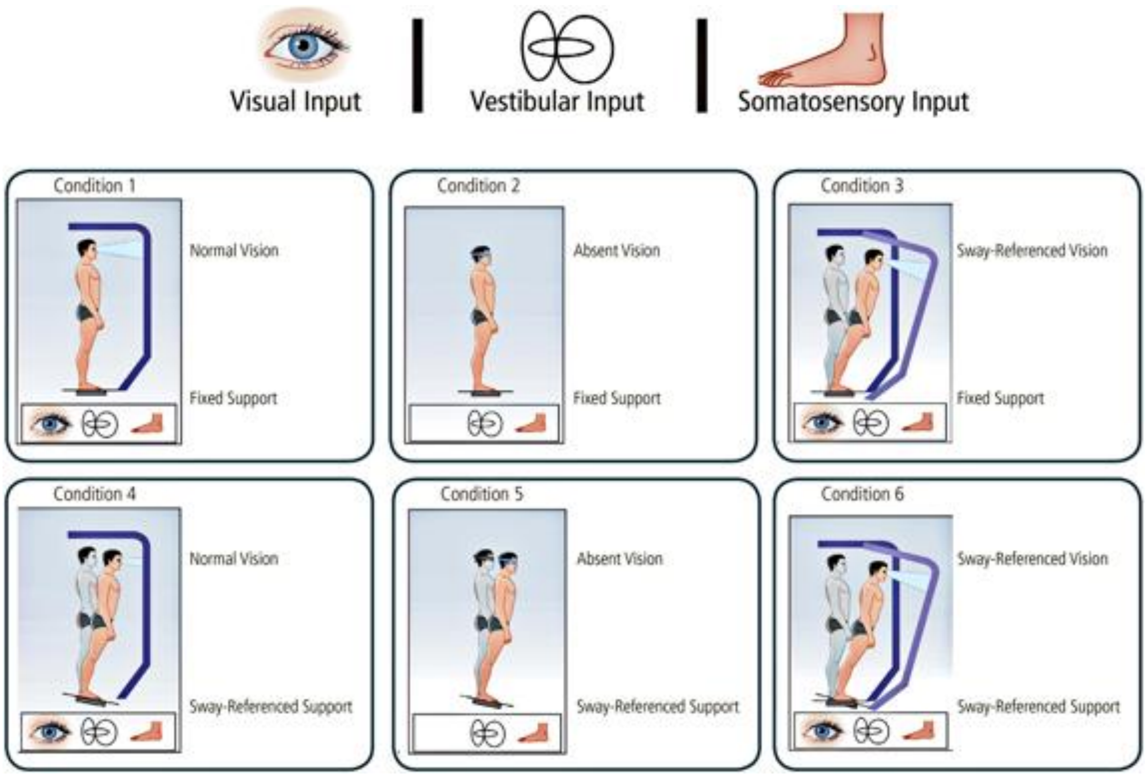

Figure 5.5. Six Conditions of the SOT (NeuroCom® International).

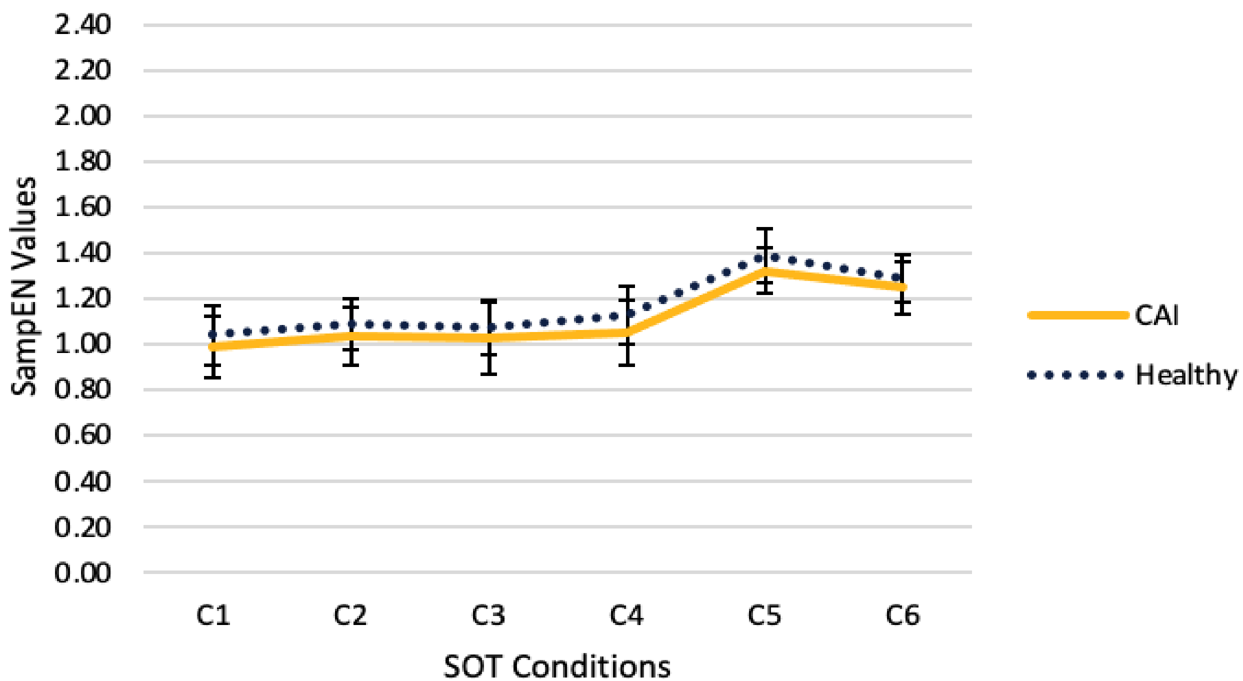

Figure 5.6. Group × Environment Interactions for a Double-Limb.
SampEN = Sample Entropy; C = Condition; SOT = Sensory Organization Test; CAI = Chronic Ankle Instability.

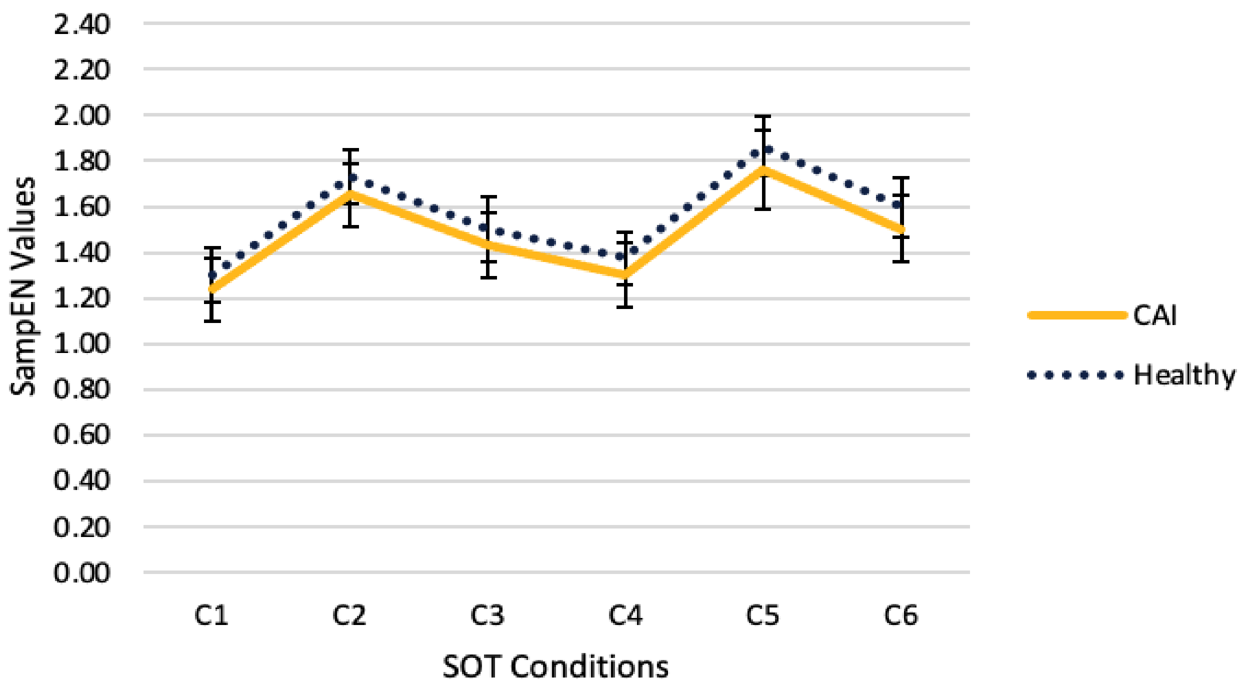

Figure 5.7. Group × Environment Interactions for the Injured-Limb.
SampEN = Sample Entropy; C = Condition; SOT = Sensory Organization Test; CAI = Chronic Ankle Instability.

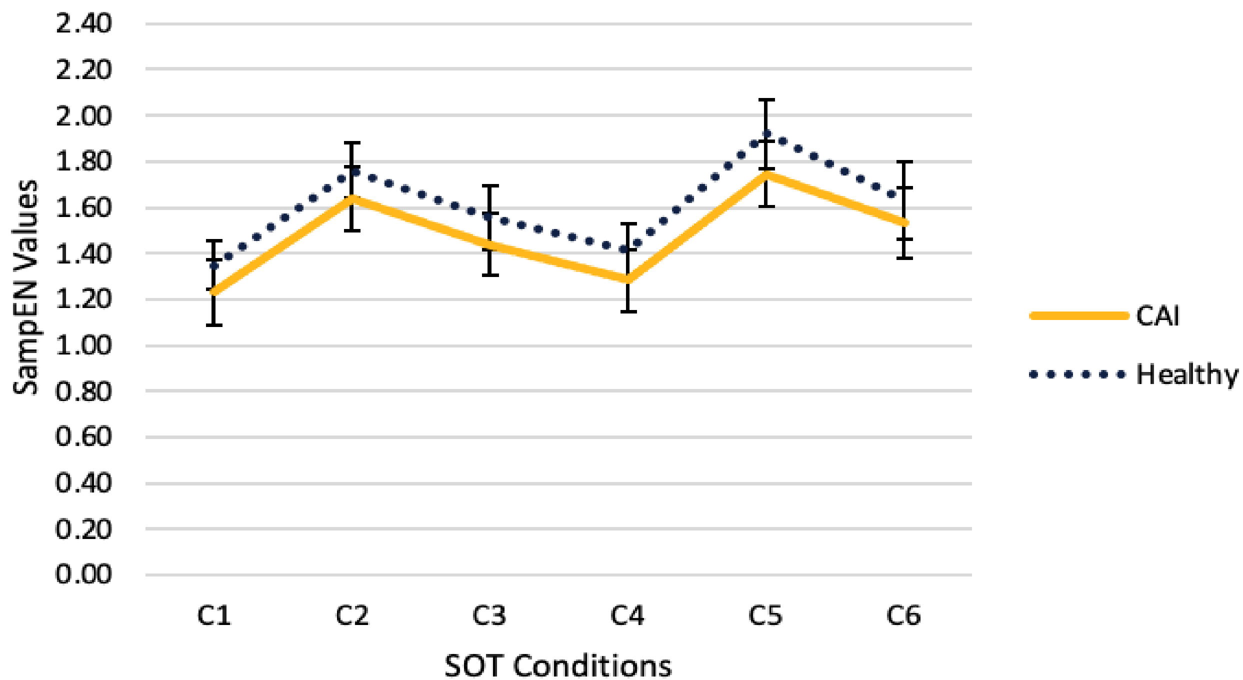

Figure 5.8. Group × Environment Interactions for the Uninjured-Limb.
SampEN = Sample Entropy; C = Condition; SOT = Sensory Organization Test; CAI = Chronic Ankle Instability.

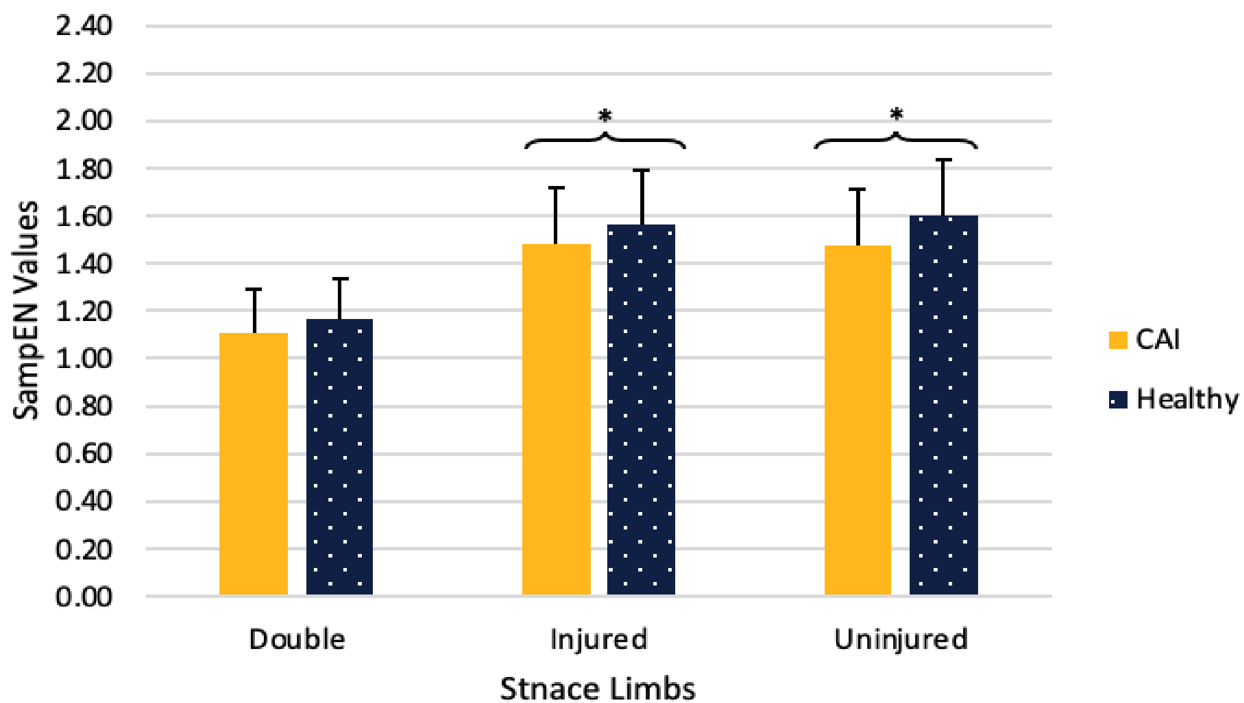

Figure 5.9. Group Main Effect for Individual Stance Limbs.
SampEN = Sample Entropy; CAI = Chronic Ankle Instability.
* indicates significant group differences (*P* < 0.05).

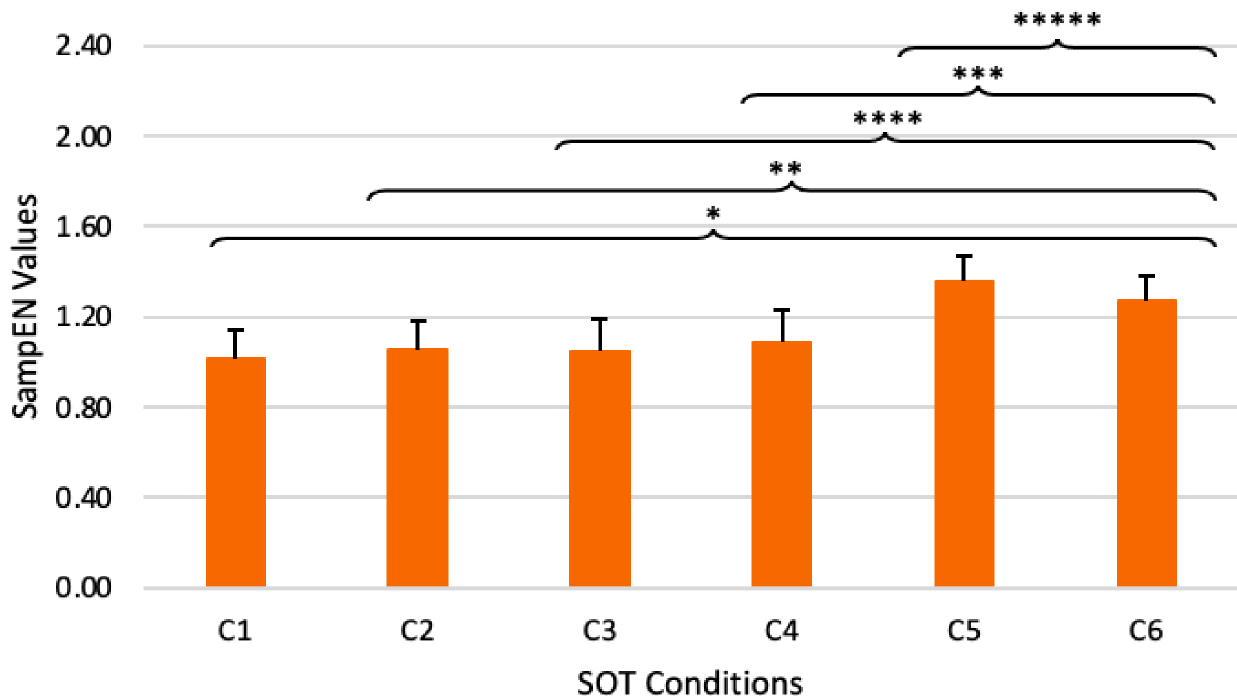

Figure 5.10. Environment Main Effect for a Double-Limb.
SampEN = Sample Entropy; C = Condition; SOT = Sensory Organization Test.
* indicates differences between C1&C2, C1&C3, C1&C4, C1&C5, C1&C6 ($P < 0.05$).
** indicates significant differences between C2&C4, C2&C5, C2&C6 ($P < 0.05$).
*** indicates significant differences between C3&C4, C3&C5, C3&C6 ($P < 0.05$).
**** indicates significant differences between C4&C5, C4&C6 ($P < 0.05$).
***** indicates significant differences between C5&C6 ($P < 0.05$).

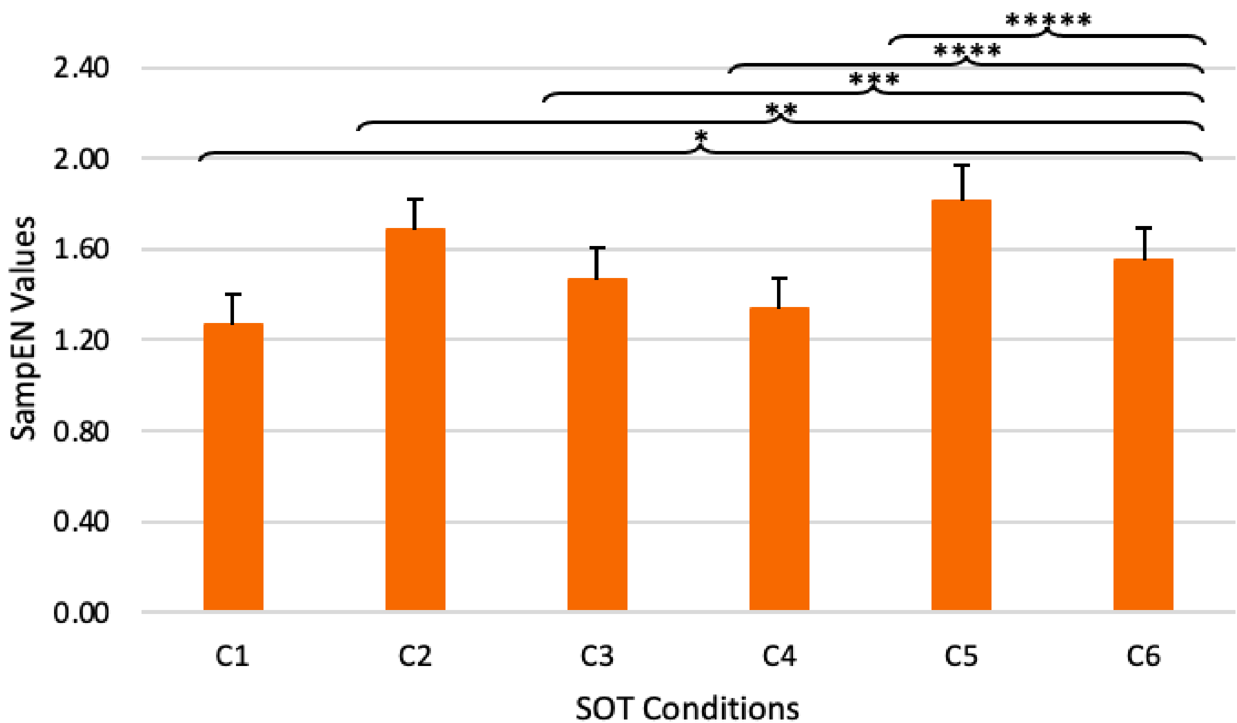

Figure 5.11. Environment Main Effect for the Injured-Limb.
SampEN = Sample Entropy; C = Condition; SOT = Sensory Organization Test.
* indicates significant differences between C1 & C2, C1 & C3, C1 & C4, C1 & C5, C1 & C6 ($P < 0.05$).
** indicates significant differences between C2 & C4, C2 & C5, C2 & C6 ($P < 0.05$).
*** indicates significant differences between C3 & C4, C3 & C5, C3 & C6 ($P < 0.05$).
**** indicates significant differences between C4 & C5, C4 & C6 ($P < 0.05$).
***** indicates significant differences between C5 & C6 ($P < 0.05$).

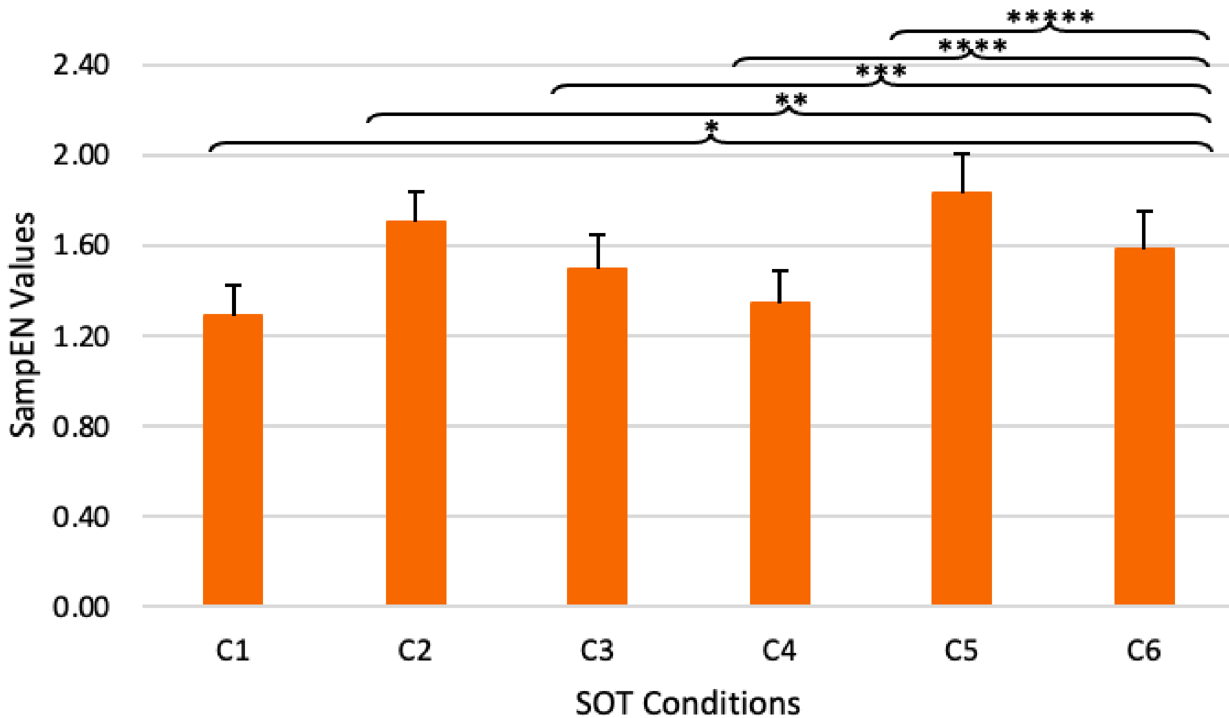

Figure 5.12. Environment Main Effect for the Uninjured-Limb.
SampEN = Sample Entropy; C = Condition; SOT = Sensory Organization Test.
* indicates significant differences between C1 & C2, C1 & C3, C1 & C4, C1 & C5, C1 & C6 ($P < 0.05$).
** indicates significant differences between C2 & C4, C2 & C5, C2 & C6 ($P < 0.05$).
*** indicates significant differences between C3 & C4, C3 & C5, C3 & C6 ($P < 0.05$).
**** indicates significant differences between C4 & C5, C4 & C6 ($P < 0.05$).
***** indicates significant differences between C5 & C6 ($P < 0.05$).

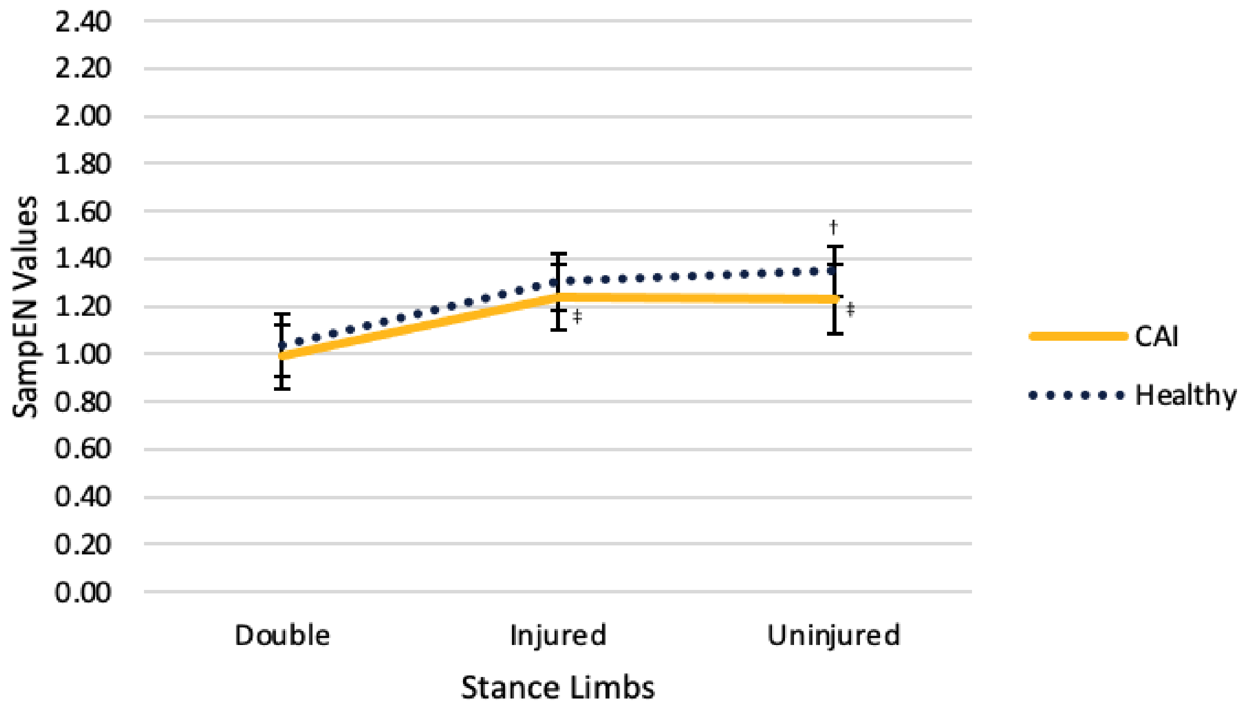

Figure 5.13. Group × Task Interactions for the SOT Condition 1.
SampEN = Sample Entropy; CAI = Chronic Ankle Instability.
† indicates significant group differences ($P < 0.05$).
‡ indicates significant differences between double- and the injured-limbs and double- and the uninjured-limb for the CAI group ($P < 0.05$).

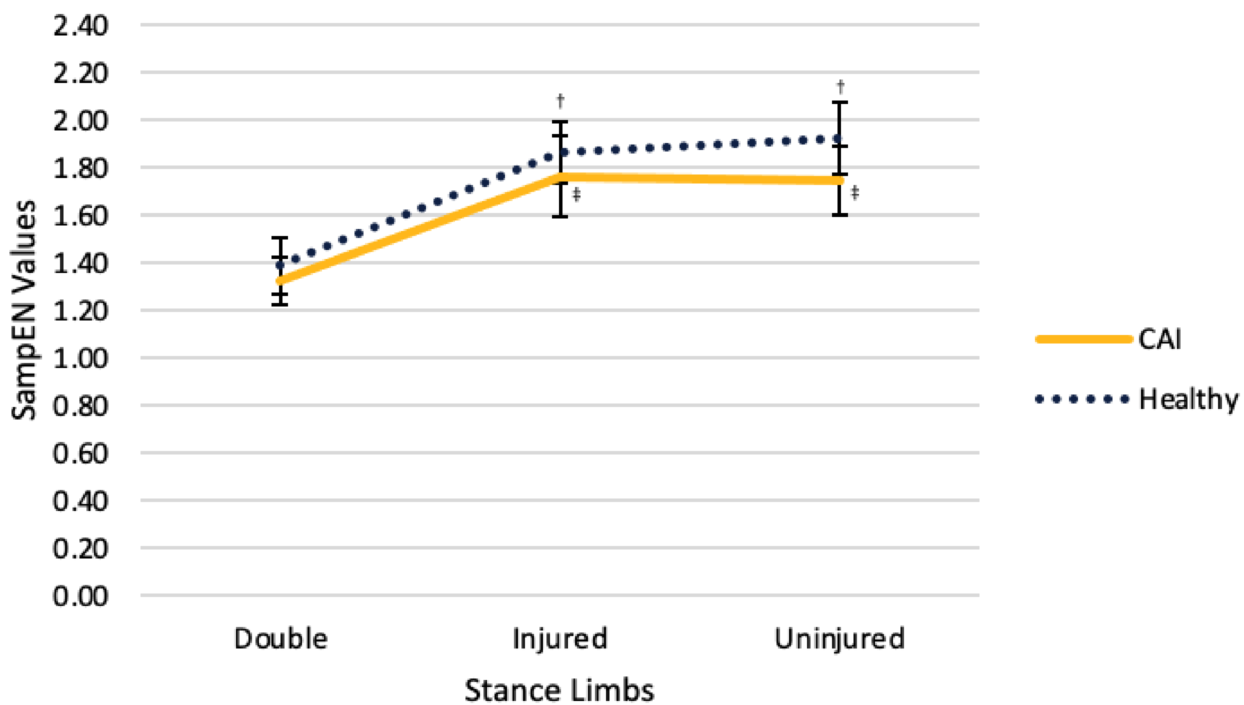

Figure 5.14. Group × Task Interactions for the SOT Condition 5.
SampEN = Sample Entropy; CAI = Chronic Ankle Instability.
† indicates significant group differences ($P < 0.05$).
‡ indicates significant differences between double- and the injured-limbs and double- and the uninjured-limb for the CAI group ($P < 0.05$).

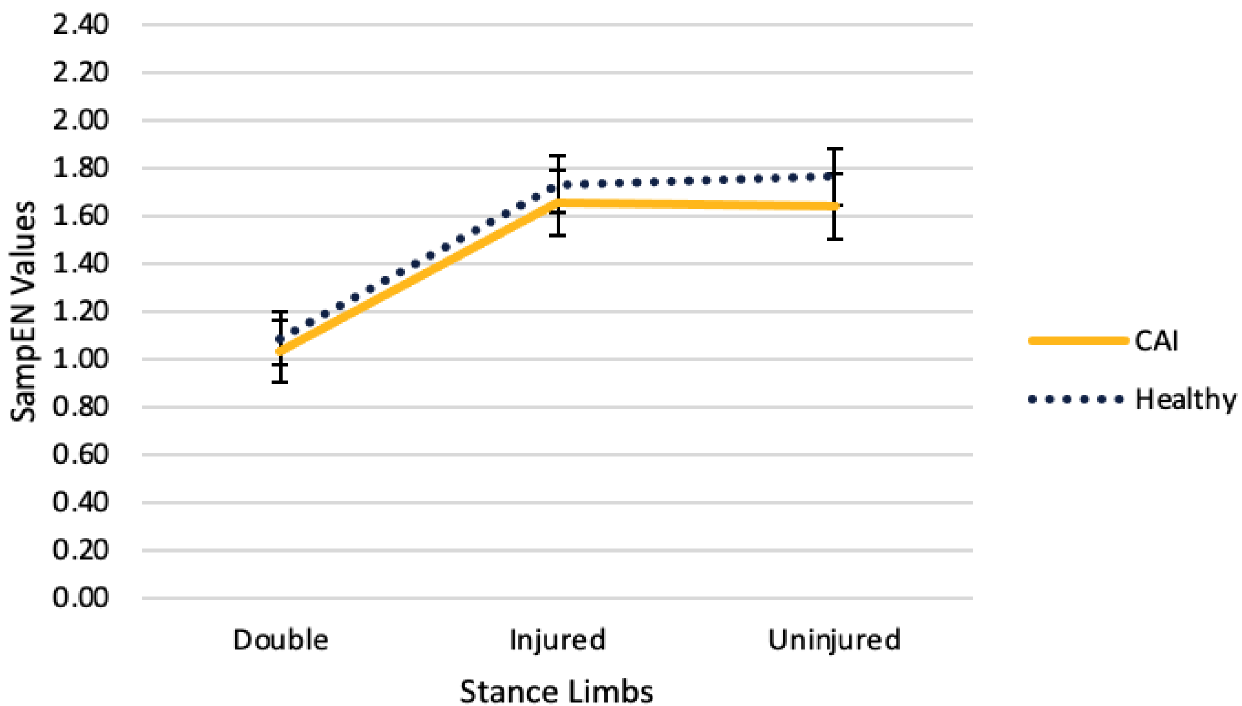

Figure 5.15. Group × Task Interactions for the SOT Condition 2.
SampEN = Sample Entropy; CAI = Chronic Ankle Instability.

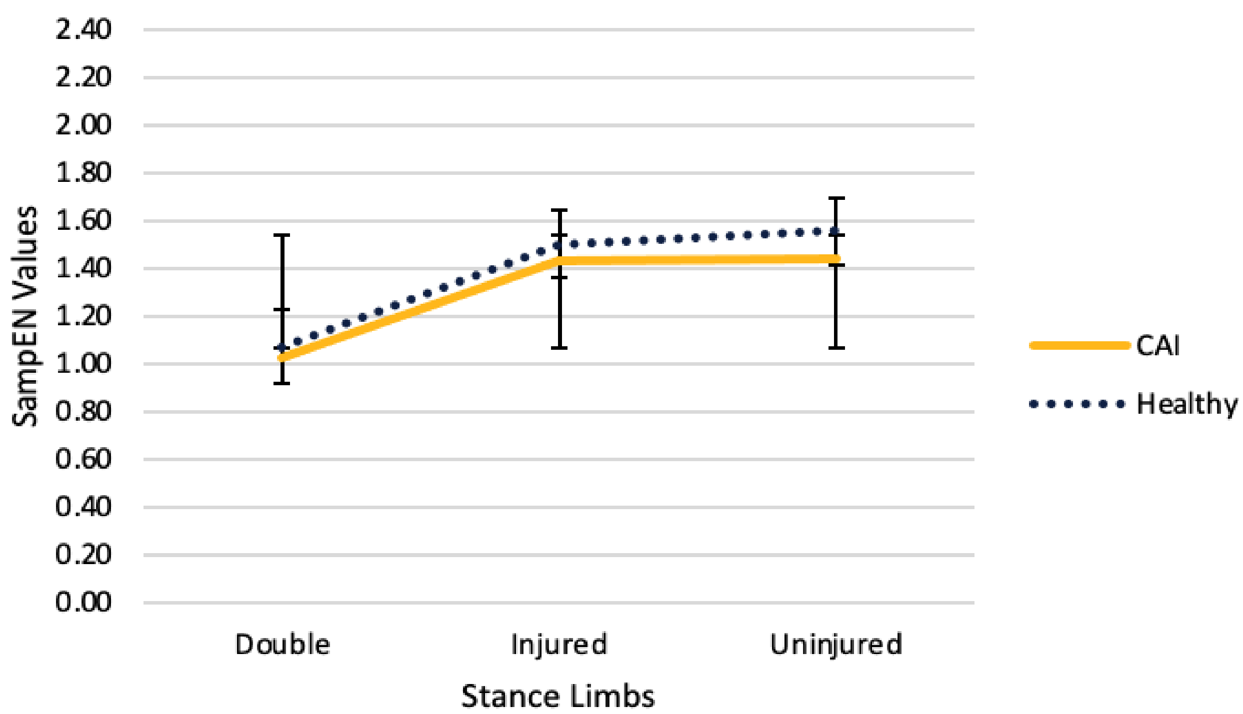

Figure 5.16. Group × Task Interactions for the SOT Condition 3.
SampEN = Sample Entropy; CAI = Chronic Ankle Instability.

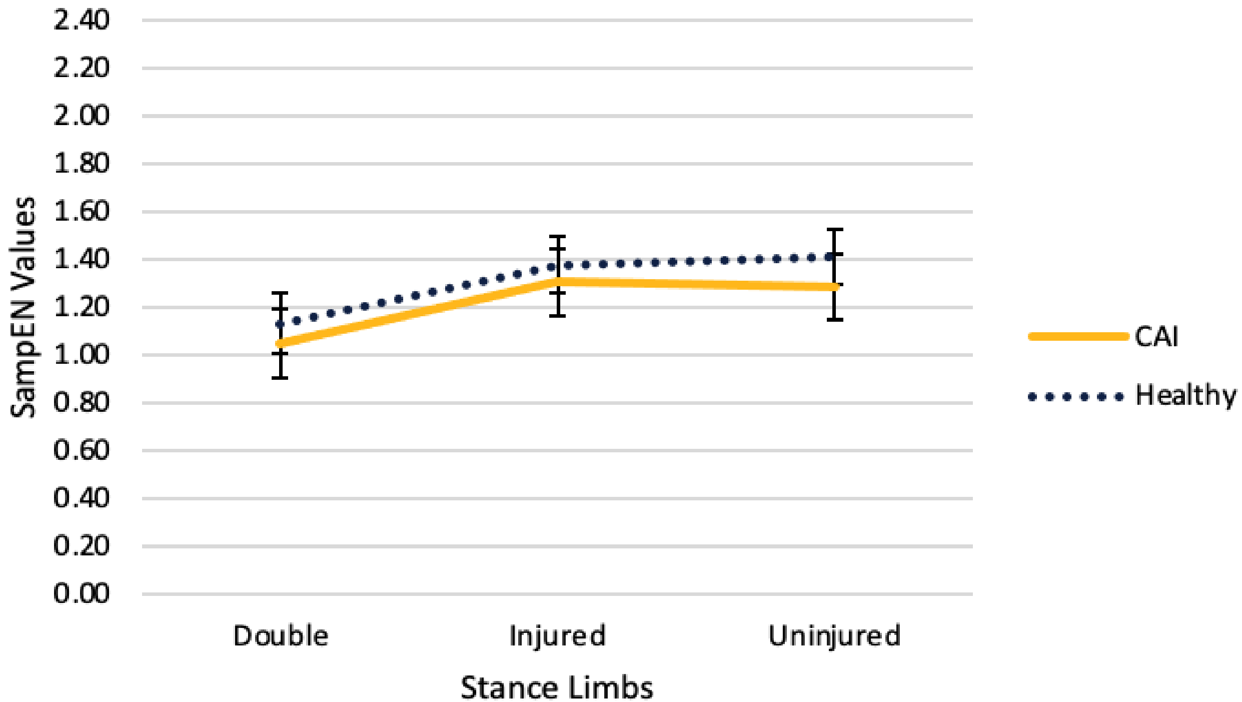

Figure 5.17. Group × Task Interactions for the SOT Condition 4.
SampEN = Sample Entropy; CAI = Chronic Ankle Instability.

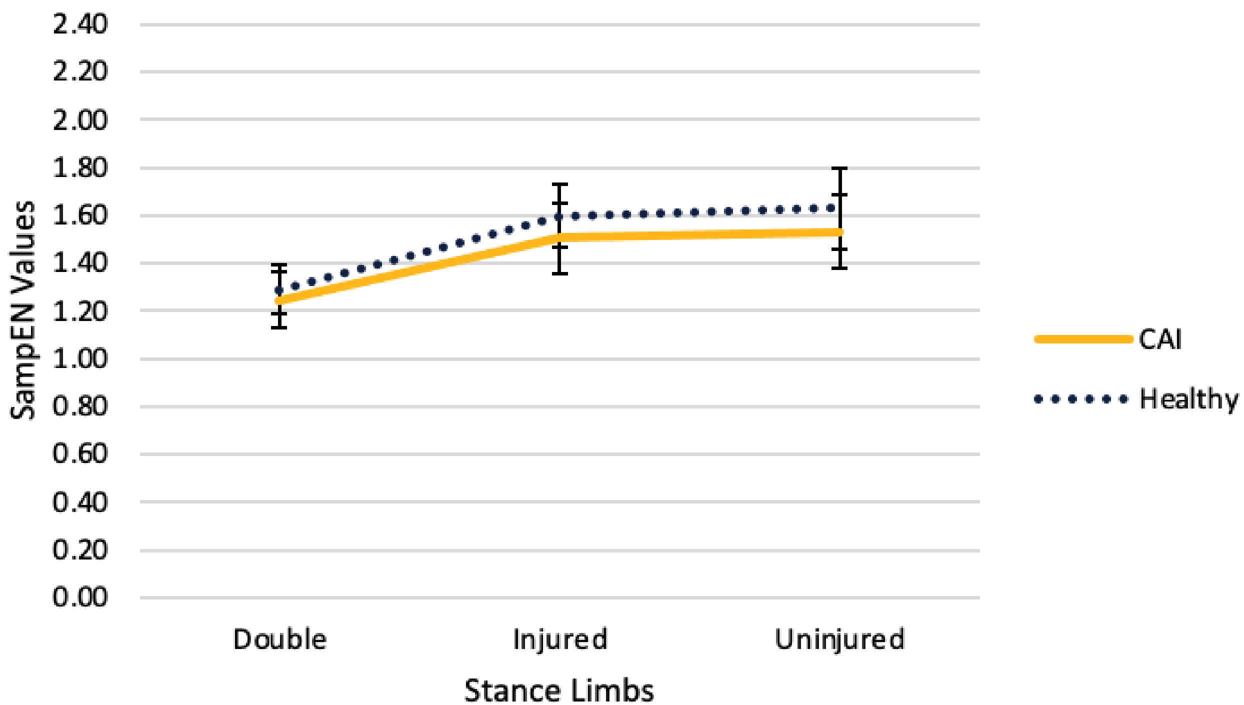

Figure 5.18. Group × Task Interactions for the SOT Condition 6.
SampEN = Sample Entropy; CAI = Chronic Ankle Instability.

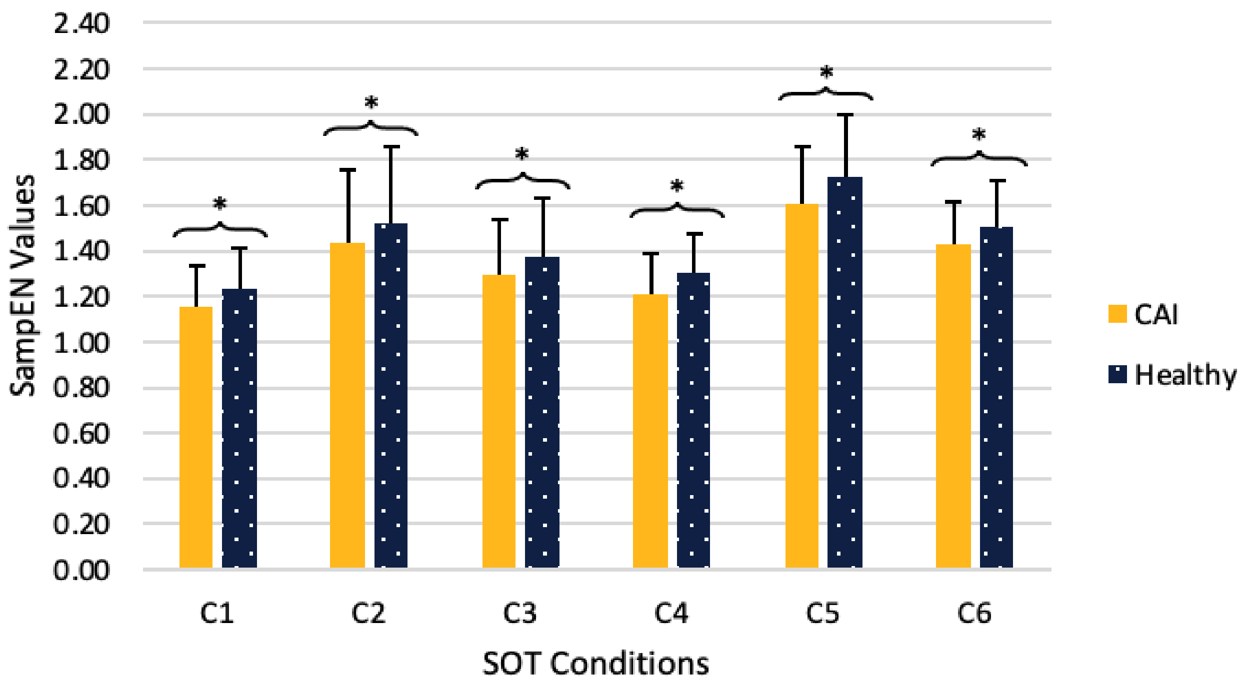

Figure 5.19. Group Main Effect for Individual SOT Conditions.
SampEN = Sample Entropy; C = Condition; SOT = Sensory Organization Test; CAI = Chronic Ankle Instability.
* indicates significant differences between groups ($P < 0.05$).

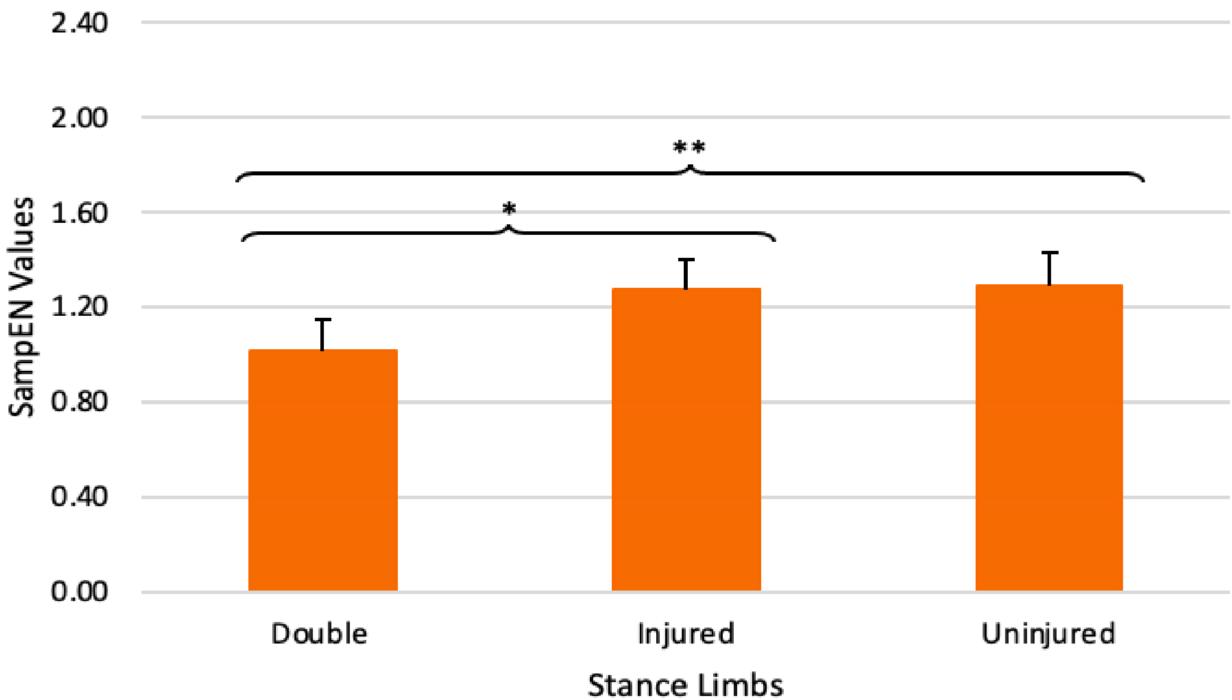

Figure 5.20. Task Main Effect for the SOT Condition 1.
SampEN = Sample Entropy.
* indicates significant differences between double- and the injured-limbs (*P* < 0.05).
** indicates significant differences between double- and the uninjured-limbs (*P* < 0.05).

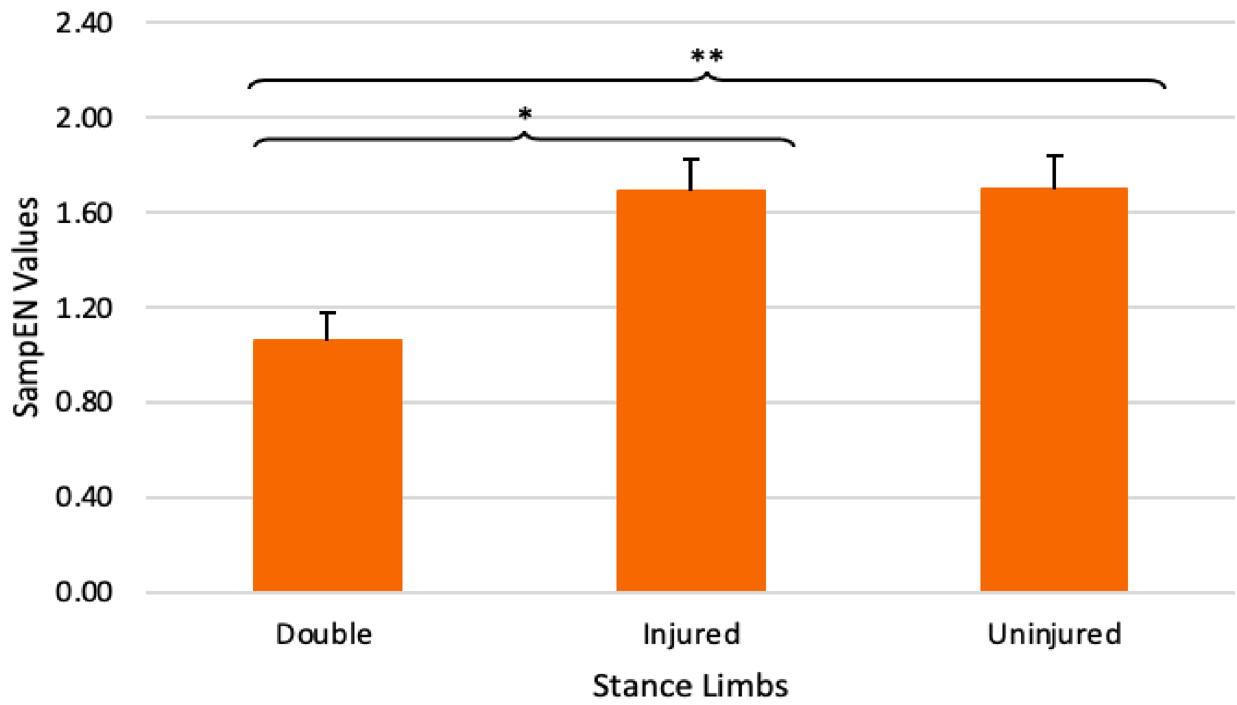

Figure 5.21. Task Main Effect for the SOT Condition 2.
SampEN = Sample Entropy.
* indicates significant differences between double- and the injured-limbs ($P < 0.05$).
** indicates significant differences between double- and the uninjured-limbs ($P < 0.05$).

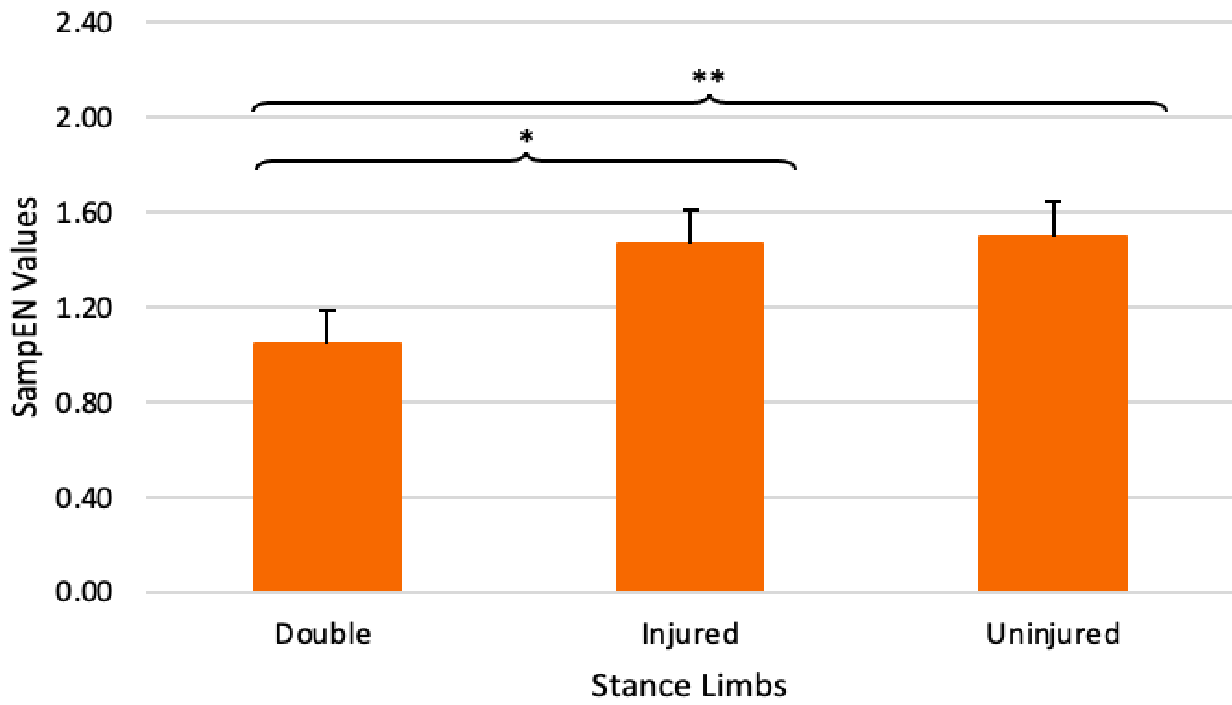

Figure 5.22. Task Main Effect for the SOT Condition 3.
SampEN = Sample Entropy.
* indicates significant differences between double- and the injured-limbs ($P < 0.05$).
** indicates significant differences between double- and the uninjured-limbs ($P < 0.05$).

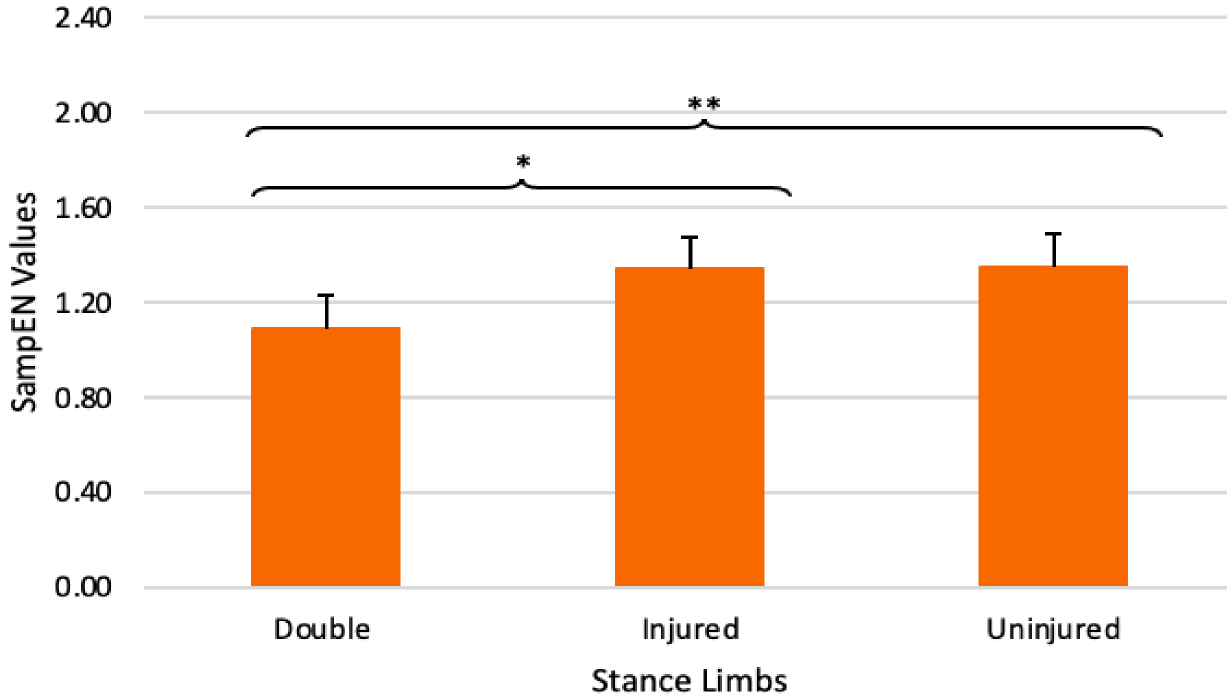

Figure 5.23. Task Main Effect for the SOT Condition 4.
SampEN = Sample Entropy.
* indicates significant differences between double- and the injured-limbs ($P < 0.05$).
** indicates significant differences between double- and the uninjured-limbs ($P < 0.05$).

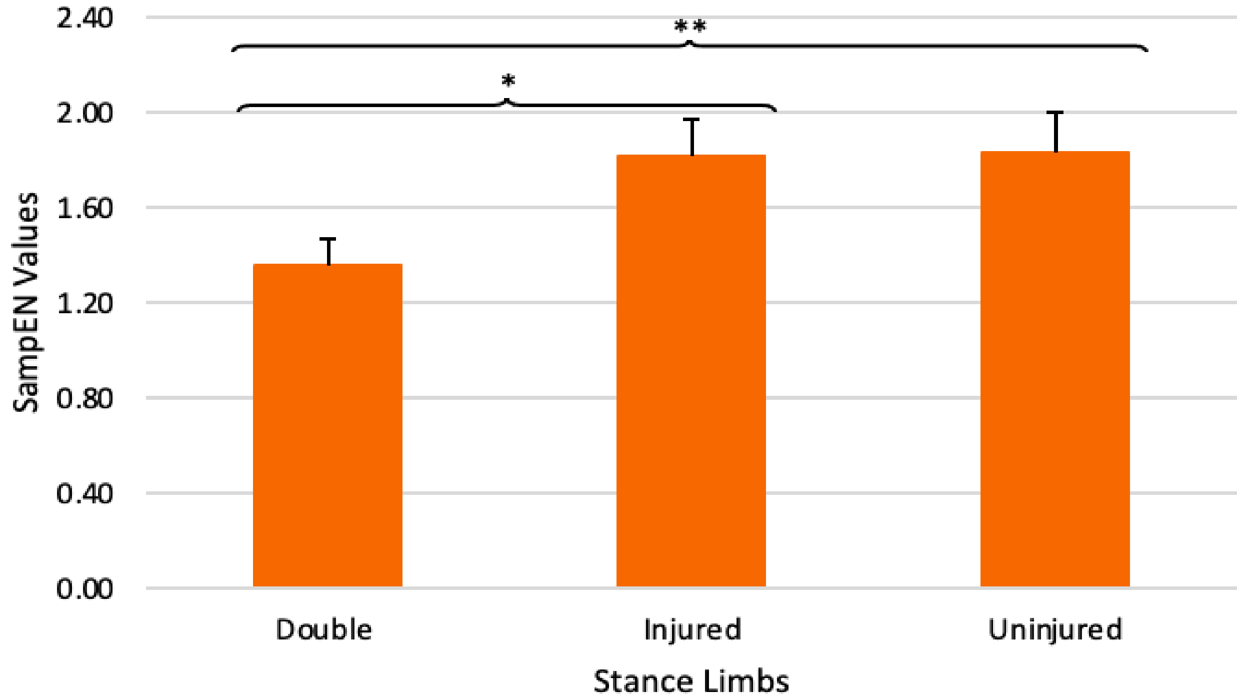

Figure 5.24. Task Main Effect for the SOT Condition 5.
SampEN = Sample Entropy.
* indicates significant differences between double- and the injured-limb (*P* < 0.05).
** indicates significant differences between double- and the uninjured-limb (*P* < 0.05).

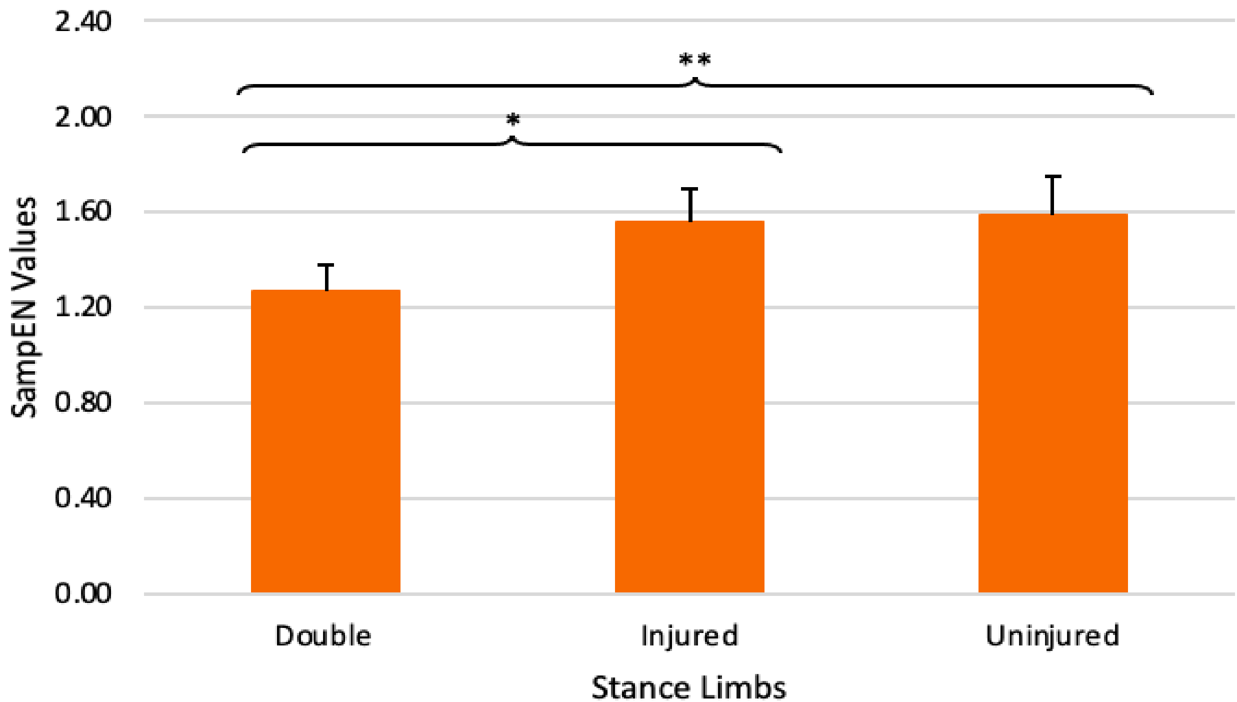

Figure 5.25. Task Main Effect for the SOT Condition 6.
SampEN = Sample Entropy.
* indicates significant differences between double- and the injured-limbs (*P* < 0.05).
** indicates significant differences between double- and the uninjured-limbs (*P* < 0.05).

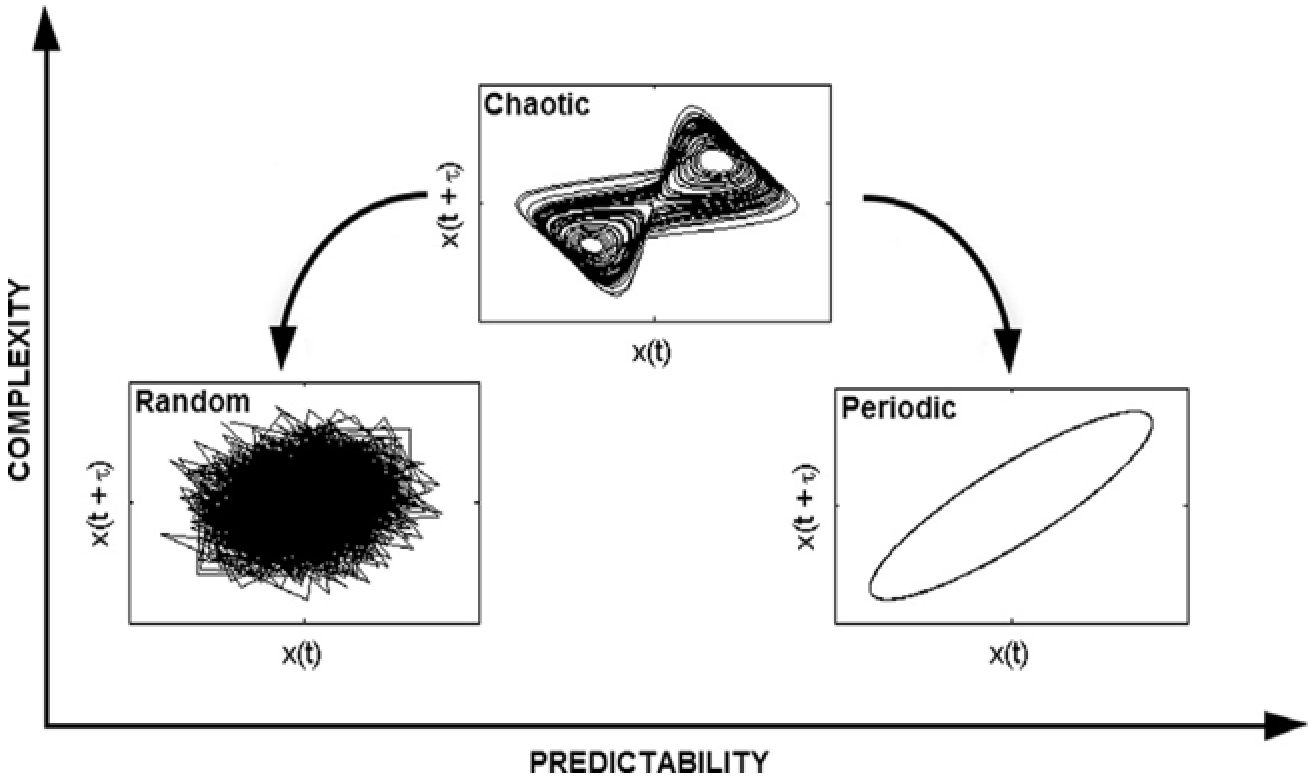

Figure 5.26. Theoretical Model of Optimal Movement Variability (Stergiou & Decker, 2011).

CHAPTER VII

EXECUTIVE SUMMARY

Individuals with Chronic Ankle Instability (CAI) commonly exhibit postural control (stability, adaptation) deficits and altered gait (walking, running) mechanics (Hertel, 2008; Hertel & Corbett, 2019). These impairments in motor behaviors may be a result of inadequate, yet inherent interactions between individual perception (i.e., sensory systems) and movement (action) integrated at the central nervous system (CNS), resulting in less flexible and adaptable sensorimotor systems. Flexibility and adaptability of sensorimotor systems which reflect on underlying biological noise are critical to coordinate the sensory reweighting system and adapt to the complexity of the environmental and task constraints. However, there is no study to our knowledge that has examined the integration phenomenon of individual elements (i.e., sensory reweighting system, movement variability) contributing to interactions between individual perception and movement of sensorimotor pathways in the same cohort of participants with and without CAI. Therefore, the primary objective of the current study was to understand the modulation of 1) the sensory reweighting system and postural control, 2) postural adaptation to a sudden change in the environment in the direction of lateral ankle sprain mechanisms, and 3) movement variability, an underlying biological noise pertaining to postural control, when the complexity of environmental and task constraints are manipulated in CAI individuals compared to healthy controls.

Postural control is critical for achieving suitable performance, especially with an increase in the complexity of task constraints in an ever-changing environment. In addition, adequate automatic postural response and adaptation that is an ability to minimize sway when exposed to an unexpected perturbation (e.g., change in environment) are necessary to prevent a risk of subsequent ankle sprains and development of post-traumatic osteoarthritis at the ankle among CAI. Successful postural control and adaptation depend on context-specific integrations of redundant and convergent somatosensory, visual, and vestibular feedback by assigning relative weight on each sensory system based on organismic (e.g., health status), environmental, and task constraints (Horak & Macpherson, 1996). However, excessive reliance on unisensory, a specific sensory system like vision, has been displayed in individuals with somatosensory and vestibular deficits (Cooke et al., 1978; Hafstrom et al., 2004; Lopez et al., 2006; Slaboda et al., 2009; Bonan et al., 2013; Lin et al., 2019). Likewise, a systematic review with a meta-analysis concluded those individuals with CAI upregulate reliance on visual feedback to compensate for somatosensory deficits, resulting from an initial ankle sprain to control posture in the injured-limb compared to healthy controls (Song et al., 2016). Consequently, unisensory integration is thought to be a result of inadequate multisensory integration (Woollacott et al., 1986; Teasdale et al., 1991; Whipple et al., 1993; Woollacott, 1993).

Our study demonstrated that CAI individuals have effective multisensory integration to control posture very similar to healthy controls while performing the sensory organization test (SOT), transitioning from a simple to a more complex environmental and task constraints. Interestingly, both groups similarly distributed weight on somatosensory and visual feedback while controlling posture in double- and single-limb (injured, uninjured) stances. The main effect found for environmental and task constraints may imply the implementation of different postural control mechanisms and sensory reliance reflected by the complexity of constraints in both groups. For instance, the same trend in postural control was noted in the injured- and uninjured-limbs compared with a double-limb stance for individual SOT conditions. What was common regardless of stance limbs was that both groups demonstrated better postural control when all sensory feedback was intact in the SOT condition 1, and worse posture was observed in the SOT condition 5, followed by the SOT condition 6. Both conditions 5 and 6 examined reliance on vestibular feedback, yet vision was eliminated only in the SOT condition 5. This may suggest performing the SOT requires more visual feedback compared to other tasks. Emphasis on visual feedback remained constant while performing the SOT in the current study, regardless of task constraints transitioning from double- to single-limb stances.

Somatosensory feedback has been suggested to be the greatest contribution to postural control during static stance (Freeman et al., 1965; Horak et al., 1997; Munn et al., 2010). Similarly, visual and vestibular feedback contribute to postural control, particularly when somatosensory feedback is disrupted with the unstable surface (Nashner, 1982; Horak, 2006). However, visual feedback may be too slow to provide an influence on postural control and adaptation (Nashner et al., 1982; Lestienne et al., 1997; Nagata et al., 2001; Rasman et al., 2018).

Additionally, visual feedback provides relative information (induced by the moving scene/surface) that fluctuates as the visual scene and support surface change, while vestibular feedback provides independent information about body orientation in space (Hwang et al., 2014). With the absence of somatosensory deficits commonly demonstrated in those with CAI and persistent visual reliance during the SOT, the only group differences we noted were in vestibular feedback in the injured-limb during the SOT condition 5 (manipulation of somatosensory feedback with the elimination of vision). Group differences in the sensory reweighting system depended on both sensory systems and task constraints. Accordingly, our CAI individuals upweighted on vestibular feedback as a sole veridical reference to self-motion when task constraints are the greatest in the injured-limb while the SOT systematically manipulated other sensory feedback (i.e., somatosensory, vision), creating sensory conflicts (Mahboobin et al., 2009; DeAngelis & Angelaki, 2012).

Current literature suggests supraspinal mechanisms provide greater precision of movement by minimizing and correcting reflexive oscillations at the ankle when transitioning from simple to more complex tasks (Capaday & Stein, 1986, 1987; Katz & Pierrot-Deseilligny, 1999; Taube et al., 2008; Pinar et al., 2010; Thompson et al., 2016). Notably, a change in reflexive controls from spinal to supraspinal mechanisms has been indicated in CAI individuals (Kim et al., 2012). Furthermore, our findings of superior postural adaptation and decreased movement variability in CAI compared to healthy controls may support a change in central organization and implementation of supraspinal mechanisms of postural control. Postural adaptation depended on environmental and task constraints, respectively. CAI individuals exhibited a superior postural adaptation in the uninjured-limb and to somatosensory perturbation toward plantarflexion (PF) compared with healthy controls in the current study. This may suggest those with CAI could activate tibialis anterior muscles induced with the PF perturbation to an appropriate response without a delay (Schieppai & Nardone, 1995; Winter, 1995; Horak et al., 1997; Moore et al., 1998). Indeed, preactivation of tibialis anterior muscles has been reported during the lateral hopping test and walking in CAI compared to healthy controls (Louwerens et al., 1995; Delahunt et al., 2007; Hopkins et al., 2012; Koholdenhoven et al., 2016). Horak et al. (1997) suggest automatic postural response and adaptation are sharpened by previous experience and preprogrammed muscle activation patterns. CAI individuals are well-characterized with limited ankle dorsiflexion range-of-motion assessed via weight-bearing lunge test and walking kinematics (Hoch et al., 2011, 2012; Chinn et al., 2013). Preactivation of tibialis anterior muscles that is indicative of pre-programmed feedforward motor control could be interpreted as a strategy CAI individuals developed to maintain the ankle joint complex in a more stable and close-packed dorsiflexion position during dynamic postural control tasks and walking.

The dynamical systems theory hypothesizes neurobiological systems will self-organize to find the most stable solution based on one of the organismic, environmental, and task constraints (Davids & Glazier, 2010; Glazier, 2017). For example, we found the process by which both groups control posture depended on environmental (SOT conditions) and task (stance limbs) constraints. There are near-infinite ways to self-organize. Thus, constraints provide boundaries limiting the number of configurations available at different levels of the body (e.g., sensory systems), known as the degree of freedom (e.g., sensory reweighting system), for the CNS to coordinate motor outputs (Davids et al., 2003; Glazier, 2017). The degree of diversity or isolation of sensory systems reflects the regularity (readiness) and adaptability of movement. For instance, the greater degree of sensory diversity (multisensory) provides complex information, whereas the greater degree of sensory isolation (unisensory) provides limited information for the CNS to configure coordinating movement. Therefore, too much and too few interactions among physiological elements (e.g., sensory systems) affect individuals' adaptive capability to the environmental and task demands, resulting in either more random or restricted movement patterns, that are respectively increases or decreases in movement variability. CAI individuals in our study demonstrated lower movement variability in the uninjured-limb for the SOT condition 1, and in both the uninjured- and injured-limbs for the SOT condition 5 compared with healthy controls. There was also a trend (ES = 0.50) of lower movement variability in the injured-limb for the SOT condition 1 in CAI. Group differences in movement variability depended on task constraints, thus lower movement variability is a result of compensatory mechanisms that those with CAI implemented to provide dynamic stability, specifically when the task was more challenging in a single-limb stance. Although we did not find significant interactions between group and environmental constraints, there was a strong trend (ES range = 0.49-1.18) suggesting lower movement variability found in CAI across all SOT conditions in the uninjured- and injured-limbs depended on environmental constraints. Consequently, we contend lower movement variability is another mechanism CAI participants implemented to provide a boundary to freeze the degree-of-freedom (redundancy in sensory feedback) to achieve effective multisensory integration.

Our collective findings of superior postural adaptation and lower movement variability displayed in CAI compared to healthy controls may imply an existent change in central organization and implementation of supraspinal mechanisms of postural control. CAI individuals exhibited effective multisensory integration to control posture very similar to healthy controls. Therefore, our study confirms the integration phenomenon of individual elements (i.e., sensory reweighting system, movement variability) contributing to interactions between individual perception and movement, especially when the complexity of environmental and task constraints increase. Postural control, postural adaptation, and movement variability in individuals with and without CAI depended on environmental or task constraints. Environment- and task-dependent postural control and adaptation and movement variability contribute to motor behaviors throughout the lifespan. Therefore, taking a multisensory-feedback approach by recognizing when to increase environmental and task constraints may optimize rehabilitation intervention to prevent subsequent ankle sprains in individuals with CAI.

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APPENDIX A.

PRE-PARTICIPATION CRITERIA QUESTIONNAIRE

1. Are you between 16-34 years old? YES / NO

2. Do you regularly exercise at least 2.5 hours (150-minute) a week of **moderate**-intensity or 1.15 hours (75-minute) a week of **vigorous**-intensity aerobic physical activity? YES / NO

3. How many hours and days do you participate in physical activities per week?

   HOURS: ________________ DAYS: ________________

4. Do you exercise (**moderate**-intensity or **vigorous**-intensity aerobic physical activity) at least 3 times per week? YES / NO

5. Have you been medically diagnosed with a concussion during the last 6-months? YES / NO

   a. If **yes**, when were you diagnosed with a concussion? DATE:

   MONTH: _____________ DAY: _____________ YEAR: _____________

6. Do you have any medical history of neurological, vestibular (inner ear), and visual disorders and/or disease (e.g., vertigo, epilepsy, stroke, peripheral neuropathies) that may influence your balance and gait (walking, running)? YES / NO

7. Do you have any medical history of connective tissue disease and/or disorders (e.g., rheumatoid arthritis, Marfan syndrome, Ehlers-Danlos syndrome)? YES / NO

8. Do you have any history of surgeries in the brain and/or lower extremity (e.g., lower back, hip, thigh, knee, lower leg, ankle, foot)? YES / NO

9. Have you had any acute injuries experiencing either pain, swelling, redness, and/or loss of functions in the lower extremity (e.g., lower back, hip, thigh, knee, lower leg, ankle, foot) during the last 6-months? YES / NO

10. Have you felt “giving way” and/or instability at the knee or hip while exercising (**moderate**-intensity or **vigorous**-intensity aerobic physical activity) during the last 6-months? YES / NO

11. Have you sprained your ankle in the past? YES / NO

    If you answered **yes**:

    a. Is this the **only** ankle sprain you have experienced in a lifetime? YES/NO

    b. Has it been at least 12-month since you sustained an initial ankle sprain that was associated with inflammatory symptoms (e.g., pain, swelling, loss of function, etc.)? YES / NO
    c. Have you experienced at least 2 episodes of your ankle “giving way’’ (excessive ankle movement) and/or “feelings of instability” during the last 6-month? YES / NO

    d. Have you sustained **recurrent ankle sprains** (≥ 2 ankle sprains to the same ankle)? YES / NO

        i. If you answered **YES**, has the most recent recurrent ankle sprain occurred during the last 3-month? YES / NO

        ii. If you answered **NO**, have you been successfully participating in weight-bearing activities for the last 12-month without recurrent injury, episodes of “giving way,” and/or “feelings of instability”? YES / NO

12. Do you have any general health problems or illness? (e.g., diabetes, respiratory disease) YES / NO
    a. If **yes**, please describe and provide detailed information, including diagnosis:

    ____

    ____

13. Do you smoke? YES / NO
    a. If **yes**, how many times a ***day*** and a ***week***?:

    DAY: ________________ WEEK: ________________

APPENDIX B.

HEALTH HISTORY QUESTIONNAIRE FOR HEALTHY CONTROLS

What is your name?: ____________________

What is your date of birth?: ____________________

Gender? Female Male

Age?: ____________________

Which is your dominant foot? (Which foot do you kick a ball with?) RIGHT / LEFT

What is your email address?: ____________________

Can we contact you to participate in future research? YES / NO

1. Have you ever suffered significant "***lower limb***" (e.g., hip, thigh, knee, lower leg, ankle, foot) pain and/or injury which interrupted your participation in physical activities and/or sports? YES / NO

a.) If **yes**, which lower limb structure did you experience pain and/or sustain an injury? Select all that apply: HIP / THIGH / KNEE / LOWER LEG / ANKLE / FOOT

i. Please describe and provide detailed information, including diagnosis:

____

____

ii. How severe was the lower limb pain and/or injury?

___ 0 (no pain): ___ 1 (very light pain): ___ 2 (light pain): ___ 3 (moderate pain):

___ 4 (strong pain): ___ 5 (unbearable pain): ___ Unknown

b.) When did you sustain lower limb pain and/or injury? Specify the DATE:

MONTH: _____________ DAY: _____________ YEAR: _____________

c.) Was the lower limb pain and/or injury diagnosed by a healthcare professional? YES / NO

d.) What was the mechanism of lower limb pain and/or injury? (How did it happen?):

____

____

e.) Which side of the lower limb did you experience the pain and/or sustain the injury? RIGHT / LEFT / RIGHT + LEFT

f.) How many days of non-weight bearing and/or weight-bearing immobilization did you experience with the lower limb pain and/or injury? Specify the number of days:

NON-WEIGHT BEARING: LEFT: _____________ RIGHT: _____________

WEIGHT BEARING: LEFT: _____________ RIGHT: _____________

g.) Did you complete rehabilitation following the lower limb pain and/or injury with a healthcare professional? YES / NO

i. If **yes**, how many weeks of supervised rehabilitation did you complete?

# of WEEKS: ________________

ii. When was the last day that you have completed the rehabilitation? Specify the DATE:

MONTH: ___________ DAY: ___________ YEAR: ___________

h.) Did you undergo any additional (surgical and/or nonsurgical) treatment with the lower limb pain and/or injury? YES / NO

i. If **yes**, please describe and provide detailed information:

____

____

2. Have you ever suffered a significant ***"lower back"*** pain and/or injury which interrupted your participation in physical activities and/or sports? YES / NO

i. If **yes**, please describe and provide detailed information, including diagnosis:
____
____

ii. How severe was the lower back pain and/or injury?

___ 0 (no pain): ___ 1 (very light pain): ___ 2 (light pain): ___ 3 (moderate pain):

___ 4 (strong pain): ___ 5 (unbearable pain): ___ Unknown

a.) When did you sustain lower back pain and/or injury? Specify the DATE:

MONTH: _____________ DAY: _____________ YEAR: _____________

b.) Was the lower back pain and/or injury diagnosed by a healthcare professional? YES / NO

c.) What was the mechanism of the lower back pain and/or injury? (How did it happen?):
____
____

d.) Which side of the lower back did you experience the pain and/or sustain the injury? RIGHT / LEFT / RIGHT + LEFT / CENTER (RIGHT ON SPINE)

e.) How many days of non-weight and/or weight-bearing immobilization did you experience with the lower back pain and/or injury? Specify the number of days:

NON-WEIGHT BEARING: LEFT: _____________ RIGHT: _____________

WEIGHT BEARING: LEFT: _____________ RIGHT: _____________

f.) Did you complete rehabilitation following the lower back pain and/or injury with a healthcare professional? YES / NO

i. If **yes**, how many weeks of supervised rehabilitation did you complete?

# of WEEKS: ________________

ii. When was the last day that you have completed the rehabilitation? Specify the DATE:

MONTH: ___________ DAY: ___________ YEAR: ___________

g.) Did you undergo any additional (surgical and/or nonsurgical) treatment with the lower back pain and/or injury? YES / NO

i. If **yes**, please describe and provide detailed information:
____
____

3. Do you feel unsteady and/or experience a loss of balance when ***walking***? YES / NO

a.) If **yes**, how frequently?

___ Very frequently: ___ frequently: ___ Occasionally: ___ Rarely:

___ Very rarely: ___ Never: ___ Unknown

b.) When was the most recent event? Specify the DATE:

MONTH: _____________ DAY: _____________ YEAR: _____________

4. Do you feel unsteady and/or experience a loss of balance when ***running***? YES / NO

a.) If **yes**, how frequently?

___ Very frequently: ___ frequently: ___ Occasionally: ___ Rarely:

___ Very rarely: ___ Never: ___ Unknown

b.) When was the most recent event? Specify the DATE:

MONTH: _____________ DAY: _____________ YEAR: _____________

5. Do you feel unsteady and/or experience a loss of balance when ***participating in physical activities and/or sports***? YES / NO

a.) If **yes**, how frequently?

___ Very frequently: ___ frequently: ___ Occasionally: ___ Rarely:

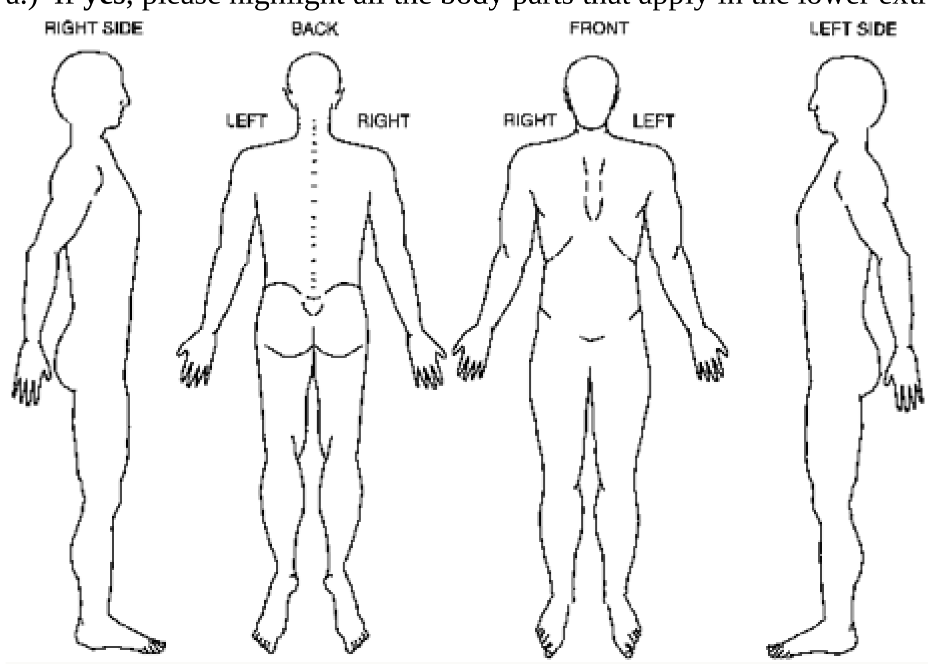
___ Very rarely: ___ Never: ___ Unknown

b.) When was the most recent event? Specify the DATE:

MONTH: _____________ DAY: _____________ YEAR: _____________

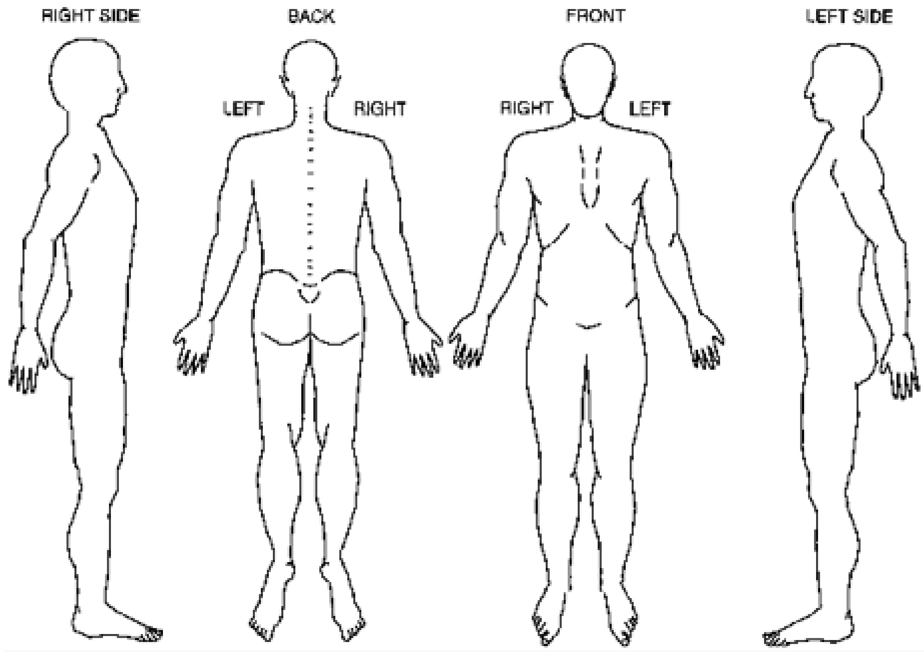
6. Do you feel and/or experience muscle weakness in the lower extremity (i.e., lower back, hip, thigh, knee, lower leg, ankle, foot) when ***walking***? YES / NO

a.) If **yes**, please highlight all the body parts that apply in the lower extremity:



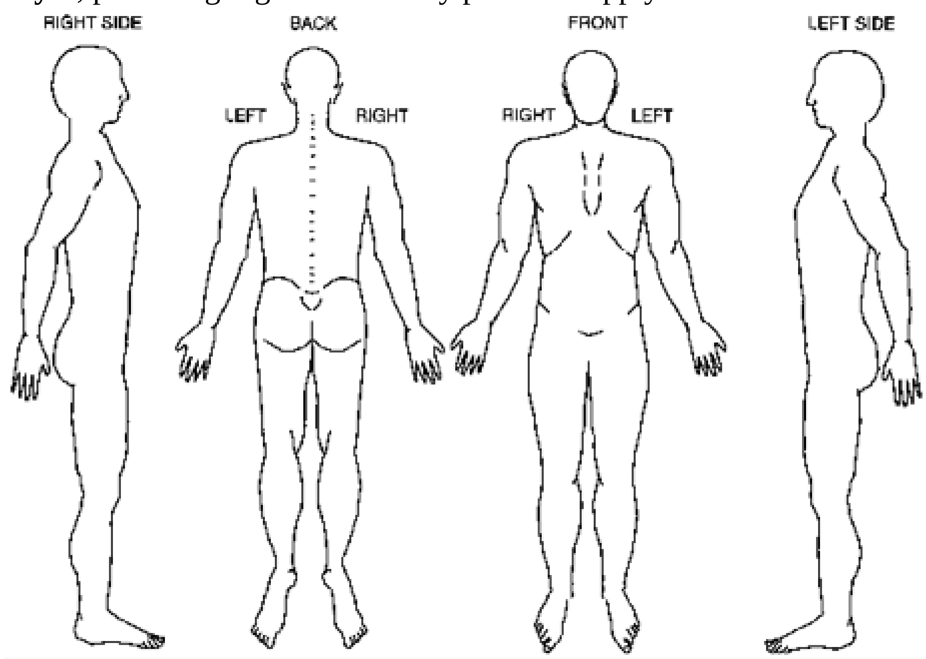
7. Do you feel and/or experience muscle weakness in the lower extremity (i.e., lower back, hip, thigh, knee, lower leg, ankle, foot) when ***running***? YES / NO

a.) If **yes**, please highlight all the body parts that apply in the lower extremity:



8. Do you feel and/or experience muscle weakness in the lower extremity (i.e., lower back, hip, thigh, knee, lower leg, ankle, foot) when ***participating in physical activities and/or sports***? YES / NO

a.) If **yes**, please highlight all the body parts that apply in the lower extremity:



9. Please list other medical conditions/concerns that you feel we should be aware of:

____

____

APPENDIX C.

HEALTH HISTORY QUESTIONNAIRE FOR CAI AND COPERS

What is your name?: ____________________

What is your date of birth?: ____________________

Gender? Female Male

Age?: ____________________

Which is your dominant foot? (Which foot do you kick a ball with?) RIGHT / LEFT

What is your email address?: ____________________

Can we contact you to participate in future research? YES / NO

1. When did you sustain the ***initial ankle sprain***? Specify the DATE:

MONTH: _____________ DAY: _____________ YEAR: _____________

a.) Which ankle did you initially sprain? RIGHT / LEFT / RIGHT + LEFT

b.) What was the mechanism of the initial ankle sprain? (How did it happen?):

____

____

c.) Was the initial ankle sprain diagnosed by a healthcare professional? YES / NO

d.) How severe was the initial ankle sprain?

___ 0 (no pain): ___ 1 (very light pain): ___ 2 (light pain): ___ 3 (moderate pain):

___ 4 (strong pain): ___ 5 (unbearable pain): ___ Unknown e.) How many days of non-weight bearing and/or weight-bearing immobilization did you experience with the initial ankle sprain? Specify the number of days:

NON-WEIGHT BEARING: LEFT: _____________ RIGHT: _____________

WEIGHT BEARING: LEFT: _____________ RIGHT: _____________

f.) Did you complete rehabilitation following the initial ankle sprain with a healthcare professional? YES / NO

i. If **yes**, how many weeks of supervised rehabilitation did you complete?

# of WEEKS: ________________

ii. When was the last day that you have completed the rehabilitation? Specify the DATE:

MONTH: ___________ DAY: ___________ YEAR: ___________

g.) Did you undergo any additional (surgical and/or nonsurgical) treatment with the initial ankle sprain? YES / NO

i. If **yes**, please describe and provide detailed information:
____
____

h.) How many repeated episodes of "giving way" (excessive ankle movement) have you experienced following the initial ankle sprain?

LEFT: ________________ RIGHT: ________________

2. Have you ever suffered a significant "***lower leg or foot"*** pain and/or injury which interrupted your participation in physical activities and/or sports ***following the initial ankle sprain***? YES / NO

a.) If **yes**, which lower limb structure did you experience pain and/or sustain an injury? Select all that apply. LOWER LEG / FOOT

i. Please describe and provide detailed information, including diagnosis:

____

____

ii. How severe was the lower leg or foot pain and/or injury?

___ 0 (no pain): ___ 1 (very light pain): ___ 2 (light pain): ___ 3 (moderate pain):

___ 4 (strong pain): ___ 5 (unbearable pain): ___ Unknown

b.) When did you experience lower leg or foot pain and/or injury? Specify the DATE:

MONTH: ______________ DAY: ______________ YEAR: ______________

c.) Was the lower leg or foot pain and/or injury diagnosed by a healthcare professional? YES / NO

d.) What was the mechanism of the lower leg or foot pain and/or injury? (How did it happen?):

____

____

e.) Which side of the lower leg or foot did you experience the pain and/or sustain the injury? RIGHT / LEFT / RIGHT + LEFT

f.) How many days of non-weight bearing and/or weight-bearing immobilization did you experience with the lower leg or foot pain and/or injury? Specify the number of days:

NON-WEIGHT BEARING: LEFT: ______________ RIGHT: ______________

WEIGHT BEARING: LEFT: ______________ RIGHT: ______________

g.) Did you complete rehabilitation following the lower leg or foot pain and/or injury with a healthcare professional? YES / NO

i. If **yes**, how many weeks of supervised rehabilitation did you complete?

# of WEEKS: ________________

ii. When was the last day that you have completed the rehabilitation? Specify the DATE:

MONTH: ___________ DAY: ___________ YEAR: ___________

h.) Did you undergo any additional (surgical and/or nonsurgical) treatment with the lower leg or foot pain and/or injury? YES / NO

i. If **yes**, please describe and provide detailed information:

____

____

3. Have you ever suffered a significant ***"proximal joint"*** (i.e., knee, hip, or lower back) pain and/or injury which interrupted your participation in physical activities and/or sports ***following the initial ankle sprain***? YES / NO

a.) If **yes**, which proximal joint did you injure and/or experience pain? Select all that apply. KNEE / HIP / LOWER BACK

i. Please describe and provide detailed information, including diagnosis:

____

____

ii. How severe was the proximal joint pain and/or injury? Select one applies:

___ 0 (no pain): ___ 1 (very light pain): ___ 2 (light pain): ___ 3 (moderate pain):

___ 4 (strong pain): ___ 5 (unbearable pain): ___ Unknown

b.) When did you sustain the proximal joint pain and/or injury? Specify the DATE:

MONTH: _____________ DAY: _____________ YEAR: _____________

c.) Was the proximal joint pain and/or injury diagnosed by a healthcare professional? YES / NO

d.) What was the mechanism of the proximal joint pain and/or injury? (How did it happen?):

____

____

e.) Which side of the proximal joint did you experience the pain and/or sustain the injury? RIGHT / LEFT / RIGHT + LEFT / BETWEEN RIGHT & LEFT (CENTER)

f.) How many days of non-weight bearing and/or weight-bearing immobilization did you experience with the proximal joint pain and/or injury? List the number of days.

NON-WEIGHT BEARING: LEFT: _____________ RIGHT: _____________

WEIGHT BEARING: LEFT: _____________ RIGHT: _____________

g.) Did you complete rehabilitation following the proximal joint pain and/or injury with a healthcare professional? YES / NO

i. If **yes**, how many weeks of supervised rehabilitation did you complete?

# of WEEKS: ________________

ii. When was the last day that you have completed the rehabilitation? Specify the DATE:

MONTH: ___________ DAY: ___________ YEAR: ___________

h.) Did you undergo any additional (surgical and/or nonsurgical) treatment with the proximal joint pain and/or injury? YES / NO

i. If **yes**, please describe and provide detailed information:

____

____

4. Have you experienced **recurrent ankle sprains**? YES / NO

a.) If **yes**, have you successfully returned to participating in weight-bearing activities for the last **12-months** without episodes of "giving way," and/or "feeling of instability"? YES / NO

b.) Do you wear any supportive devices (e.g., ankle brace, tape)? YES / NO

i. If **yes**, please describe when you wear them:

____

____

5. When did you sustain the ***most recent recurrent ankle sprain***? DATE:

MONTH: ______________ DAY: ______________ YEAR: ______________

a.) Which ankle did you recently sprain? RIGHT / LEFT / RIGHT + LEFT

b.) What was the mechanism of the most recent ankle sprain? (How did it happen?):

____

____

c.) Was the most recent ankle sprain diagnosed by a healthcare professional? YES / NO

d.) How severe was the most recent ankle sprain?

___ 0 (no pain): ___ 1 (very light pain): ___ 2 (light pain): ___ 3 (moderate pain):

___ 4 (strong pain): ___ 5 (unbearable pain): ___ Unknown

e.) How many days of non-weight bearing and/or weight-bearing immobilization did you experience with the most recent ankle sprain? List the number of days:

NON-WEIGHT BEARING: LEFT: ______________ RIGHT: ______________

WEIGHT BEARING: LEFT: ______________ RIGHT: ______________

f.) Did you complete rehabilitation following the most recent ankle sprain with a healthcare professional? YES / NO

i. If **yes**, how many weeks of supervised rehabilitation did you complete?

# of WEEKS: ________________

ii. When was the last day that you have completed the rehabilitation? Specify the DATE:

MONTH: ___________ DAY: ___________ YEAR: ___________

g.) Did you undergo any additional (surgical and/or nonsurgical) treatment with the most recent ankle sprain? YES / NO

i. If **yes**, please describe and provide detailed information:

____

____

h.) How many repeated episodes of "giving way" have you experienced following the most recent ankle sprain?

LEFT: ________________ RIGHT: ________________

6. Have you ever suffered a significant "***lower leg or foot"*** pain and/or injury which interrupted your participation in physical activity and/or sports ***following the most recent ankle sprain***? YES / NO

a.) If **yes**, which lower leg or foot did you experience the pain and/or sustain an injury? Select all that apply. LOWER LEG / FOOT

i. Please describe and provide detailed information, including diagnosis:

____

____

ii. How severe was the lower limb pain and/or injury?

___ 0 (no pain): ___ 1 (very light pain): ___ 2 (light pain): ___ 3 (moderate pain):

___ 4 (strong pain): ___ 5 (unbearable pain): ___ Unknown

b.) When did you sustain lower leg or foot pain and/or injury? Specify the DATE:

MONTH: _____________ DAY: _____________ YEAR: _____________

c.) Was the lower leg or foot pain and/or injury diagnosed by a healthcare professional? YES / NO

d.) What was the mechanism of the lower leg or foot pain and/or injury? (How did it happen?):

____

____

e.) Which side of the lower leg or foot did you experience the pain and/or sustain the injury? RIGHT / LEFT / RIGHT + LEFT

f.) How many days of non-weight and/or weight-bearing immobilization did you experience with the lower limb pain and/or injury? Specify the number of days:

NON-WEIGHT BEARING: LEFT: _____________ RIGHT: _____________

WEIGHT BEARING: LEFT: _____________ RIGHT: _____________

g.) Did you complete rehabilitation following the lower leg or foot pain and/or injury with a healthcare professional? YES / NO

i. If **yes**, how many weeks of supervised rehabilitation did you complete?

# of WEEKS: ________________

ii. When was the last day that you have completed the rehabilitation? Specify the DATE:

MONTH: ___________ DAY: ___________ YEAR: ___________

h.) Did you undergo any additional (surgical and/or nonsurgical) treatment with the lower leg or foot pain and/or injury? YES / NO

i. If **yes**, please describe and provide detailed information:

____

____

7. Have you ever suffered a significant ***"proximal joint"*** (i.e., knee, hip, or lower back) pain and/or injury which interrupted your participation in physical activity and/or sports ***following the most recent ankle sprain***? YES / NO

a.) If **yes**, which proximal joint did you injure and/or experience pain? Select all that apply. KNEE / HIP / LOWER BACK

i. Please describe and provide detailed information, including diagnosis:

____

____

ii. How severe was the proximal joint pain and/or injury?

___ 0 (no pain): ___ 1 (very light pain): ___ 2 (light pain): ___ 3 (moderate pain):

___ 4 (strong pain): ___ 5 (unbearable pain): ___ Unknown

b.) When did you sustain proximal joint pain and/or injury? Specify the DATE:

MONTH: _____________ DAY: _____________ YEAR: _____________

c.) Was the proximal joint pain and/or injury diagnosed by a healthcare professional? YES / NO d.) What was the mechanism of the proximal joint pain and/or injury? (How did it happen?):

____

____

e.) Which side of the proximal joint did you experience the pain and/or sustain the injury? RIGHT / LEFT / RIGHT + LEFT / BETWEEN RIGHT & LEFT (CENTER)

f.) How many days of non-weight and/or weight-bearing immobilization did you experience with the proximal joint pain and/or injury? List the number of days:

NON-WEIGHT BEARING: LEFT: ______________ RIGHT: ______________

WEIGHT BEARING: LEFT: ______________ RIGHT: ______________

g.) Did you complete rehabilitation following the proximal joint pain and/or injury with a healthcare professional? YES / NO

i. If **yes**, how many weeks of supervised rehabilitation did you complete?

# of WEEKS: _________________

ii. When was the last day that you have completed the rehabilitation? Specify the DATE:

MONTH: ____________ DAY: ____________ YEAR: ____________

h.) Did you undergo any additional (surgical and/or nonsurgical) treatment with the proximal joint pain and/or injury? YES / NO

i. If **yes**, please describe and provide detailed information:

____

____

8. Please list the **total number** of ankle sprains you have sustained on each leg in the past.

LEFT: ________________ RIGHT: ________________

9. Have you ever experienced 2 or more repeated episodes of your ankle "giving way'' during the last **6-month**? YES / NO

a.) If **yes**, when was the last time you have experienced your ankle "giving way"?

MONTH: _____________ DAY: _____________ YEAR: _____________

b.) Please describe and provide detailed information on the experience of your ankle "giving way":

____

____

c.) How many episodes of "giving way" have you experienced during the last **6-month**?

LEFT: ________________ RIGHT: ________________

10. Please list the **total episodes** of ankle "giving way" you have experienced in the past.

LEFT: ________________ RIGHT: ________________

11. Have you modified your activity due to the ankle injury? YES / NO

a.) If **yes**, please describe and provide detailed information:

____

____

12. Do you feel a risk of injury because of your ankle status when participating in physical activities and/or sports? YES / NO

13. Are you concerned about environmental conditions, such as uneven surfaces, because of your ankle status when participating in physical activities and/or sports? YES / NO

14. Do you feel unsteady and/or experience a loss of balance when ***walking***? YES / NO

a.) If **yes**, how frequently?

___ Very frequently: ___ frequently: ___ Occasionally: ___ Rarely:

___ Very rarely: ___ Never: ___ Unknown

b.) When was the most recent event? Specify the DATE:

MONTH: ______________ DAY: ______________ YEAR: ______________

15. Do you feel unsteady and/or experience a loss of balance when ***running***? YES / NO

a.) If **yes**, how frequently?

___ Very frequently: ___ frequently: ___ Occasionally: ___ Rarely:

___ Very rarely: ___ Never: ___ Unknown

b.) When was the most recent event? Specify the DATE:

MONTH: ______________ DAY: ______________ YEAR: ______________

16. Do you feel unsteady and/or experience a loss of balance when ***participating in physical activity and/or sports***? YES / NO

a.) If **yes**, how frequently?

___ Very frequently: ___ frequently: ___ Occasionally: ___ Rarely:

___ Very rarely: ___ Never: ___ Unknown

b.) When was the most recent event? Specify the DATE:

MONTH: ______________ DAY: ______________ YEAR: ______________

17. Besides ankle sprain, have you ever suffered a significant "***lower limb***" (e.g., hip, thigh, knee, lower leg, foot) pain and/or injury which interrupted your participation in physical activities and/or sports? YES / NO

a.) If **yes**, which lower limb structure did you experience pain and/or sustain an injury? Select all that apply. HIP / THIGH / KNEE / LOWER LEG / FOOT

i. Please describe and provide detailed information, including diagnosis:

____

____

ii. How severe was the lower limb pain and/or injury?

___ 0 (no pain): ___ 1 (very light pain): ___ 2 (light pain): ___ 3 (moderate pain):

___ 4 (strong pain): ___ 5 (unbearable pain): ___ Unknown

b.) When did you sustain lower limb pain and/or injury? Specify the DATE:

MONTH: ______________ DAY: ______________ YEAR: ______________

c.) Was the lower limb pain and/or injury diagnosed by a healthcare professional? YES / NO

d.) What was the mechanism of lower limb pain and/or injury? (How did it happen?):

____

____

e.) Which side of the lower limb did you experience the pain and/or sustain the injury? RIGHT / LEFT / RIGHT + LEFT f.) How many days of non-weight bearing and/or weight-bearing immobilization did you experience with the lower limb pain and/or injury? List the number of days:

NON-WEIGHT BEARING: LEFT: _____________ RIGHT: _____________

WEIGHT BEARING: LEFT: _____________ RIGHT: _____________

g.) Did you complete rehabilitation following the lower limb pain and/or injury with a healthcare professional? YES / NO

i. If **yes**, how many weeks of supervised rehabilitation did you complete?

# of WEEKS: ________________

ii. When was the last day that you have completed the rehabilitation? Specify the DATE:

MONTH: ___________ DAY: ___________ YEAR: ___________

h.) Did you undergo any additional (surgical and/or nonsurgical) treatment with the lower limb pain and/or injury? YES / NO

i. If **yes**, please describe and provide detailed information:

____

____

18. Have you ever suffered significant ***"lower back"*** pain and/or injury, which interrupted your participation in physical activities and/or sports? YES / NO

i. If **yes**, please describe and provide detailed information, including diagnosis:

____

____

ii. How severe was the lower back pain and/or injury?

___ 0 (no pain): ___ 1 (very light pain): ___ 2 (light pain): ___ 3 (moderate pain):

___ 4 (strong pain): ___ 5 (unbearable pain): ___ Unknown b.) When did you sustain lower back pain and/or injury? Specify the DATE:

MONTH: ______________ DAY: ______________ YEAR: ______________

c.) Was the lower back pain and/or injury diagnosed by a healthcare professional? YES / NO

d.) What was the mechanism of the lower back pain and/or injury? (How did it happen?):

____

____

e.) Which side of the lower back did you experience the pain and/or sustain the injury? RIGHT / LEFT / RIGHT + LEFT / CENTER (RIGHT ON SPINE)

f.) How many days of non-weight bearing and/or weight-bearing immobilization did you experience with the lower back pain and/or injury? Specify the number of days:

NON-WEIGHT BEARING: LEFT: ______________ RIGHT: ______________

WEIGHT BEARING: LEFT: ______________ RIGHT: ______________

g.) Did you complete rehabilitation following the lower back pain and/or injury with a healthcare professional? YES / NO

i. If **yes**, how many weeks of supervised rehabilitation did you complete?

# of WEEKS: _________________

ii. When was the last day that you have completed the rehabilitation? Specify the DATE:

MONTH: ____________ DAY: ____________ YEAR: ____________

h.) Did you undergo any additional (surgical and/or nonsurgical) treatment with the lower back pain and/or injury? YES / NO

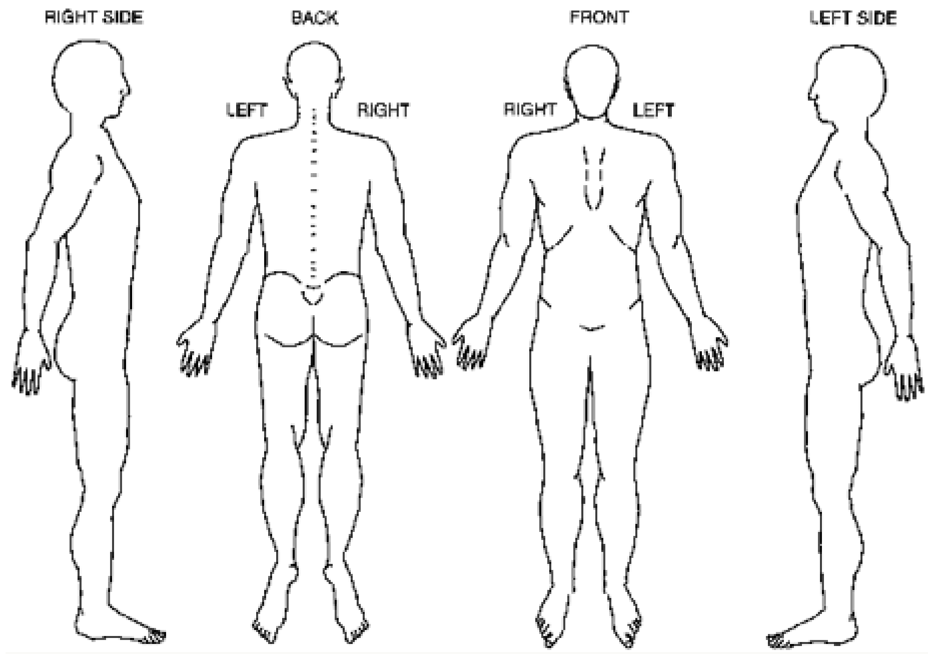
i. If **yes**, please describe and provide detailed information:

____

____

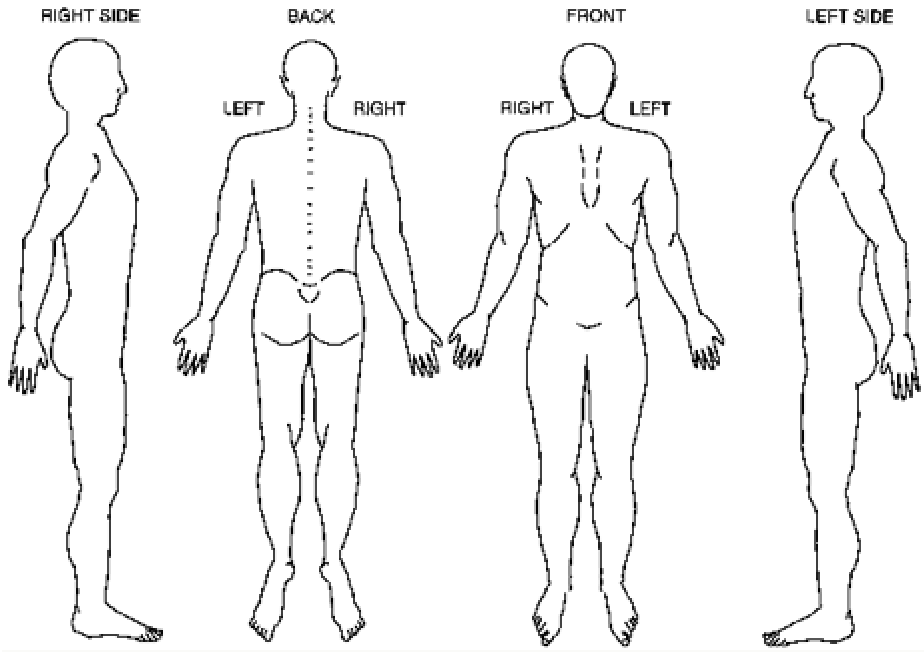
19. Do you feel and/or experience muscle weakness in the lower extremity (lower back, hip, thigh, knee, lower leg, ankle, or foot) when ***walking***? YES / NO

a.) If **yes**, please highlight all the body parts that apply in the lower limb:



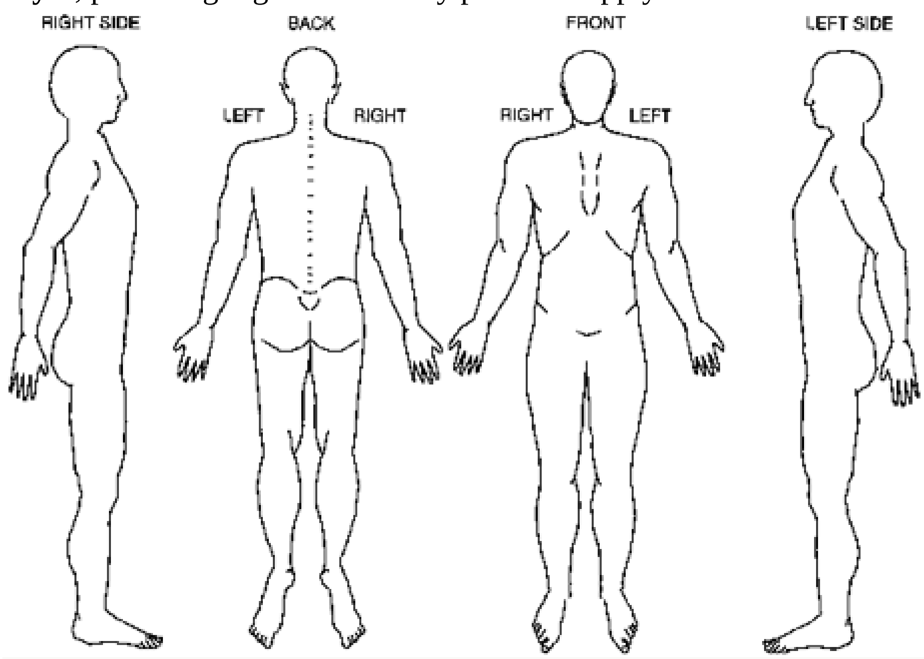
20. Do you feel and/or experience muscle weakness in the lower extremity (lower back, hip, thigh, knee, lower leg, ankle, or foot) when ***running***? YES / NO

a.) If **yes**, please highlight all the body parts that apply in the lower limb:



21. Do you feel and/or experience muscle weakness in the lower extremity (lower back, hip, thigh, knee, lower leg, ankle, or foot) when ***participating in physical activity and/or sports***? YES / NO

a.) If **yes**, please highlight all the body parts that apply in the lower limb:



22. Please list other medical conditions/concerns that you feel we should be aware of:

____

____

APPENDIX D.

CUMBERLAND ANKLE INSTABILITY MEASURE

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

	LEFT	RIGHT	Score
1. I have pain in my ankle			
Never			5
During sport			4
Running on uneven surfaces			3
Running on level surfaces			2
Walking on uneven surfaces			1
Walking on level surfaces			0
2. My ankle feels UNSTABLE			
Never			4
Sometimes during sport (not every time)			3
Frequently during sport (every time)			2
Sometimes during daily activity			1
Frequently during daily activity			0
3. When I make SHARP turns, my ankle feels UNSTABLE			
Never			3
Sometimes when running			2
Often when running			1
When walking			0
4. When going down the stairs, my ankle feels UNSTABLE			
Never			3
If I go fast			2
Occasionally			1
Always			0
5. My ankle feels UNSTABLE when standing on ONE leg			
Never			2
On the ball of my foot			1
With my foot flat			0
6. My ankle feels UNSTABLE when			
Never			3
I hop from side to side			2

I hop on the spot			1
When I jump			0
7. My ankle feels UNSTABLE when			
Never			4
I run on uneven surfaces			3
I jog on uneven surfaces			2
I walk on uneven surfaces			1
I walk on a flat surface			0
8. TYPICALLY, when I start to roll over (or “twist”) on my ankle, I can stop it			
Immediately			3
Often			2
Sometimes			1
Never			0
I have never rolled over on my ankle			3
9. After a TYPICAL incident of my ankle rolling over, my ankle returns to “normal”			
Almost immediately			3
Less than one day			2
1-2 days			1
More than 2 days			0
I have never rolled over on my ankle			3

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

APPENDIX E.

IDENTIFICATION OF FUNCTIONAL ANKLE INSTABILITY

IDENTIFICATION OF FUNCTIONAL ANKLE INSTABILITY (IdFAI)

Instructions: This form will be used to categorize your ankle stability status. A separate form should be used for the right and left ankles. Please fill out the form completely and if you have any questions, please ask the administrator. Thank you for your participation.

Please carefully read the following statement:
"Giving way" is described as a temporary uncontrollable sensation of instability or rolling over of one's ankle.

I am completing this form for my **RIGHT/LEFT** ankle (circle one).

1.) Approximately how many times have you sprained your ankle? ________

2.) When was the last time you sprained your ankle?

☐ Never	☐ > 2 years	☐ 1-2 years	☐ 6-12 months	☐ 1-6 months	☐ < 1 month
0	1	2	3	4	5

3.) If you have seen an athletic trainer, physician, or healthcare provider how did he/she categorize your most serious ankle sprain?

☐ Have **not** seen someone	☐ Mild (Grade I)	☐ Moderate (Grade II)	☐ Severe (Grade III)
0	1	2	3

4.) If you have ever used crutches, or other device, due to an ankle sprain how long did you use it?

☐ Never used a device	☐ 1-3 days	☐ 4-7 days	☐ 1-2 weeks	☐ 2-3 weeks	☐ >3 weeks
0	1	2	3	4	5

5.) When was the last time you had ***"giving way"*** in your ankle?

☐ Never	☐ > 2 years	☐ 1-2 years	☐ 6-12 months	☐ 1-6 months	☐ < 1 month
0	1	2	3	4	5

6.) How often does the ***"giving way"*** sensation occur in your ankle?

☐ Never	☐ Once a year	☐ Once a month	☐ Once a week	☐ Once a day
0	1	2	3	4

7.) Typically when you start to roll over (or 'twist') on your ankle can you stop it?

☐ Never rolled over	☐ Immediately	☐ Sometimes	☐ Unable to stop it
0	1	2	3

8.) Following a typical incident of your ankle rolling over, how soon does it return to 'normal'?

☐ Never rolled over	☐ Immediately	☐ < 1 day	☐ 1-2 days	☐ > 2 days
0	1	2	3	4

9.) During "Activities of daily life" how often does your ankle feel ***UNSTABLE***?

☐ Never	☐ Once a year	☐ Once a month	☐ Once a week	☐ Once a day
0	1	2	3	4

10.) During "Sport/or recreational activities" how often does your ankle feel ***UNSTABLE***?

☐ Never	☐ Once a year	☐ Once a month	☐ Once a week	☐ Once a day
0	1	2	3	4

Version 1.0

APPENDIX F.

FOOT AND ANKLE ABILITY MEASURES

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.

APPENDIX G.

NATIONAL AERONAUTICS AND SPACE ADMINISTRATION
PHYSICAL ACTIVITY STATUS SCALE

1. Use the appropriate number (0 to 7) which best describes your general ACTIVITY LEVEL for the PREVIOUS MONTH. Please choose only one option below.
DO NOT PARTICIPATE REGULARLY IN PROGRAMMED RECREATION SPORT OR HEAVY PHYSICAL ACTIVITY

(0) - Avoid walking or exertion, e. g., always use elevator, drive whenever possible instead of walking.
(1) - Walk for pleasure, routinely use stairs, occasionally exercise sufficiently to cause heavy breathing or perspiration.

PARTICIPATED REGULARLY IN RECREATION OR WORK REQUIRING MODEST PHYSICAL ACTIVITY, SUCH AS GOLF, HORSBACK RIDING, CALISTENICS, GYMNASTICS, TABLE TENNIS, BOWLING, WEIGHT LIFTING, YARD WORK.

(2) - 10 to 60 minutes per week.
(3) - Over one hour per week.

PARTICIPATE REGULARLY IN HEAVY PHYSICAL EXERCISE SUCH AS RUNNING OR JOGGING, SWIMMING, CYCLING, ROWING, SKIPPING ROPE, RUNNING IN PLACE OR ENGAGING IN VIGOROUS AEROBIC ACTIVITY TYPE EXERCISE SUCH AS TENNIS, BASKETBALL OR HANDBALL.

(4) - Run less than one mile per week, walk 1.5 miles per week, or spend less than 30 minutes per week in comparable physical activity.
(5) - Run one to five miles per week or spend 30 to 60 minutes per week in comparable physical activity.

(6) - Run five to ten miles per week, walk 7-14 miles per week, or spend 1 to 3 hours per week in comparable physical activity.
(7) - Run over ten miles per week, walk over 14 miles per week, or spend over 3 hours per week in comparable physical activity.