DOES CHRONIC ANKLE INSTABILITY AFFECT LOWER EXTREMITY BIOMECHANICS AND MUSCLE ACTIVATION OF FEMALES DURING LANDINGS?

by

YUMENG LI

(Under the Direction of Kathy J. Simpson)

ABSTRACT

Although much is known about the mechanisms of anterior cruciate ligament (ACL) and other knee-related injuries, it is still unclear how chronic ankle instability (CAI) relates to these injury mechanisms. The purpose of the study, therefore, was to determine if individuals with CAI exhibit atypical knee biomechanics and muscle activation during landing onto a tilted surface. A seven-camera motion analysis system, two force plates and a surface electromyography (EMG) system were used to collect lower extremity biomechanics and EMG of 21 CAI and 21 pair-matched control (CON) participants who performed 10 landings onto a sideward-tilted and flat platform on the CAI/matched and non-test limbs, respectively. Kinematics (joint angles and displacement), kinetics (joint moments and eccentric work) and muscle activation (EMG linear envelope) were generated and compared between the CAI and CON groups using paired t-tests. CAI displayed an increased ankle inversion angle at initial contact; lower ankle inversion moment and eccentric work and increased EMG co-contraction during landing that could be related to their increased peak knee joint extension moment, internal rotation moment, and quadriceps-to-hamstring activation ratio during landing. This shows that CAI group successfully adapt their ankle landing mechanics to prevent the ankle from ‘giving way’ by using greater ankle muscle co-contraction EMG to stabilize the foot. However, the increased co-contraction of ankle muscles could reduce the ankle energy dissipation in the frontal plane and further leaded to altered knee biomechanics and muscle activation. The atypical knee joint biomechanics and muscle activation (i.e., greater knee extension moment, internal rotation moment and quadriceps to hamstring co-contraction) of CAI group have been shown to be related to increased ACL loading. Future studies may need to measure/estimate the ACL loading to confirm that CAI relates to the mechanisms of ACL injury.

INDEX WORDS: electromyography, lower extremity, inverted landing surface, kinetics, kinematics, chronic ankle injury, drop landings

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M.Sc., University of Manitoba, 2012

A Dissertation Submitted to the Graduate Faculty of the University of Georgia in Partial Fulfillment of the Requirements for the Degree

DOCTOR OF PHILOSOPHY

ATHENS, GEORGIA

2016

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August 2016

ACKNOWLEDGEMENTS

First, I would like to thank my major advisor, Dr. Kathy Simpson for her great support and helpful guidance. You have been always a nice, professional and knowledgeable advisor. I would like to thank all my committee members for their critical suggestions and mentorship. I also would like to thank all my graduate colleagues and undergraduate research assistants for helping me recruit participants and collecting data.

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CHAPTER 1

INTRODUCTION

1.1 Introduction

The ankle joint is the most commonly injured joint of the body in sports and physical activities, with ankle sprains accounting for most of the ankle injuries (e.g., Berbert Rosa et al., 2014; Fong, Hong, Chan, Yung, & Chan, 2007; Hootman, Dick, & Agel, 2007). The ankle joint accounts for approximately 10% to 30% of all sports injuries (Hootman, Dick, & Agel, 2007) and up to 80% of basketball and volleyball injuries (Garrick & Requa, 1988). Moreover, ankle sprains accounted for more than 80% of such reported ankle injuries (Fong, Hong, Chan, Yung, & Chan, 2007).

Knee injury also is very common. Researchers recorded injury data of 17,397 patients with 19,530 sports injuries over a 10-year period and observed that 40% of all sport-related injuries were related to the knee joint (Majewski, Susanne, & Klaus, 2006). Amongst all knee injuries, common injuries included “internal knee trauma” (45%), “distortion” (34%), cartilage lesion (11%), contusion (5%) and dislocation (3%).

One knee joint injury of particular concern in sport is damage to the anterior cruciate ligament (ACL) (Majewski et al., 2006). Among the general population, the annual ACL incidence rate is low, ranging from 0.03% (Granan, Bahr, Steindal, Furnes, & Engebretsen, 2008) to 0.04% (Gianotti, Marshall, Hume, & Bunt, 2009). However, among knee injuries reported that occurred during sport activities, ACL injury accounted for 20% of knee injuries and 2.6% (Hootman et al., 2007) to 8% (Majewski et al., 2006) of all sports injuries reported.

Although much is known about mechanisms of knee injury, one factor that is only now beginning to receive attention is the influence of prior ankle injury on ACL injury mechanisms. At present, there is some indirect evidence that there is an association between prior ankle injury and knee injury, although a causal relationship has not yet been demonstrated. Soderman et al. (2001) studied 146 female soccer players and observed that three of the five ACL-injured players also had ankle sprains (Söderman, Alfredson, Pietilä, & Werner, 2001). Moreover, a significant association between lateral ankle sprain history and ACL injury history was observed by Kramer, Denegar, Buckley, and Hertel, (2007). They determined, that among 33 young adult female athletes who had a history of ACL injury, 52% also had a prior ankle sprain, whereas only 32% of non-injured healthy controls had a history of prior ankle history.

If there is an association between prior ankle injury history and risk of knee injury, then it will be important to understand the mechanism of the association. Kramer et al. (2007) reported some common factors that may have contribute to injuring both joints, including increased generalized joint laxity and decreased iliotibial band flexibility. However, this has not been proven.

One reason that the potential link between previous ankle sprain and knee injury is not well established nor the mechanical etiology understood is that the definition of 'previous ankle sprain' is too broad and unspecific. 'Previous ankle sprain' could include a wide range of conditions that may not influence ankle injury (e.g., one prior ankle sprain without further sequelae) to others that could have a significant impact on abnormal lower extremity mechanics.

I believe that having chronic ankle instability (CAI) is one condition whose consequences could affect the potential for acute or chronic knee injury. Chronic ankle instability (CAI) commonly develops following ankle sprains (Hertel, 2002) and has been described "… as an encompassing term used to classify a subject with both mechanical and functional instability of the ankle joint." (Gribble et al., 2013, P.586). Several deficits of individuals with CAI could influence the ankle biomechanics and further affect knee (Gribble & Robinson, 2009, 2010) and hip biomechanics (Brown, Padua, Marshall, & Guskiewicz, 2011; Gribble, Hertel, Denegar, & Buckley, 2004). These deficits could include reduced ankle muscle co-contraction that helps stabilize the ankle joint (Lin, Chen, & Lin, 2011), limited ankle joint range of motion (Denegar, Hertel, & Fonseca, 2002; Hoch, Staton, McKeon, Mattacola, & McKeon, 2012), reduced ankle muscle strength (Willems, Witvrouw, Verstuyft, Vaes, & De Clercq, 2002) and changes to the mechanical properties of ankle joint tissues (e.g., ligament laxity).

Neuromuscular alterations in the lower extremities has been associated with CAI. Reduced open-loop neuromuscular control of peroneus longus was observed in landing (Caulfield, Crammond, O'Sullivan, Reynolds, & Ward, 2004; Delahunt, Monaghan, & Caulfield, 2006) and resulted in a more inverted ankle at initial contact (Delahunt, Monaghan, & Caulfield, 2007). Reduced soleus activity (Caulfield et al., 2004) and increased rectus femoris activity (Delahunt, Monaghan, & Caulfield, 2007) were also observed in various dynamic movements.

Biomechanical alterations were also observed for individuals with CAI during landings. For kinematics, compared to non-injured participants, the CAI ankle was more inverted in the pre-landing phase and less dorsiflexed in landing phase (Delahunt et al., 2006; Lin et al., 2011). CAI also exhibited decreased knee flexion angle at initial contact (Gribble & Robinson, 2009, 2010). However, other studies observed increases (Caulfield & Garrett, 2002) or no differences (Delahunt et al., 2007; Delahunt et al., 2006) in knee flexion angles.

For kinetics, CAI displayed greater peak posterior and vertical GRF (Delahunt et al., 2006) and reduced hip flexion moment and less percentage of knee energy absorption (Delahunt et al., 2006; Terada, Pfile, Pietrosimone, & Gribble, 2013). Therefore, whether or how these biomechanical alterations could influence knee injury predisposition is still unclear.

One previous study has been done to attempt to understand how CAI could relate to mechanism of knee injury. Terada and colleagues (2014) investigated the influence of CAI on knee "anterior shear force" (represented by anterior component of joint reaction force) and lower extremity kinematics during a vertical stop-jump landing (Terada, Pietrosimone, & Gribble, 2014). The 10 male and 9 female CAI participants demonstrated a tendency of reduced ankle dorsiflexion but only a significantly decreased peak knee flexion angle during the landing phase compared to the 19 healthy individuals. CAI could relate to the knee loadings through two steps. First, the tendency of reduced ankle dorsiflexion may be linked to the decreased knee flexion in CAI participants (Terada et al., 2014). Second, decreased knee flexion angle could result in a greater tibial anterior shear force (bone-on-bone force applied on proximal tibia by distal femur).

Terada et al. (2014) did not observe significant difference in knee anterior shear force by examining the resultant joint forces. However, the resultant joint forces provide no information of bone-on-bone force (or joint contact force) or ACL loading due to muscle forces (Winter, 2009). Moreover, the variations between male and female individuals with CAI in landing movement may explain some of the lack of significant differences for other kinematic and kinetic variables in their study. It has been reported that compared to males, females exhibited greater knee abduction and internal rotation angles (McLean et al., 2007), decreased knee flexion angles (Lephart, Ferris, Riemann, Myers, & Fu, 2002) and greater knee anterior shear force and knee extensor moment (Chappell, Yu, Kirkendall, & Garrett, 2002) in landing. These gender differences for knee biomechanics, however, may relate to mechanisms of female ACL injury (Lephart et al., 2002; Shultz, 2015).

To date, therefore, it is still unknown how CAI relates to mechanisms of ACL or other knee injuries. Though a significant association has been observed between ACL injury and ankle sprain history for females (Kramer et al., 2007), biomechanical evidence is needed to understand whether the association is meaningful and to understand the potential underlying mechanisms that may lead to abnormal lower extremity loading and movements that could be involved in the etiology of chronic joint conditions.

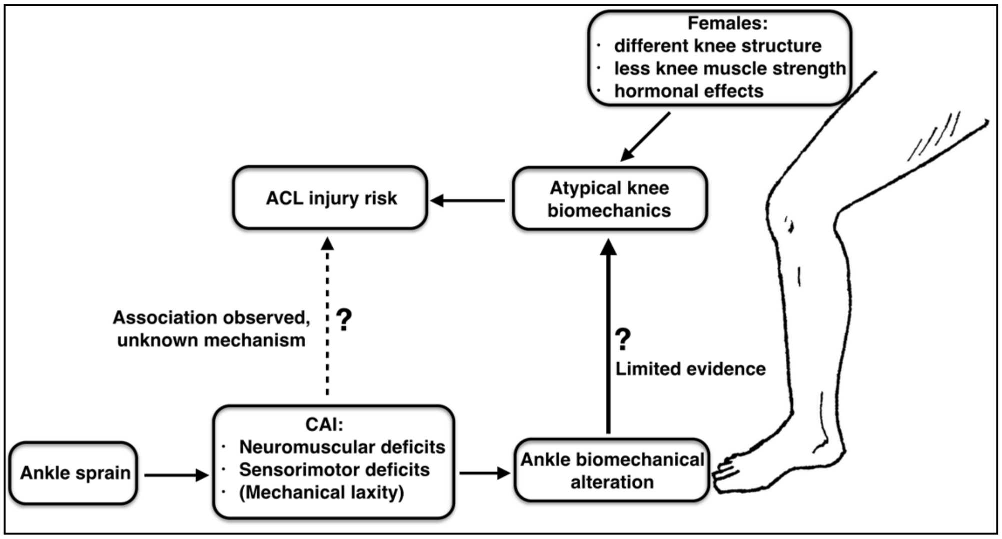

Figure 1.1. Flowchart that shows general relationship between CAI and ACL injury for females.

1.2 Research Questions, Overall Purpose and Aims of the Study

In order to understand how CAI could relate to knee loadings and injuries (as shown by question marks in Figure 1.1), I will compare, between females with and without CAI, the mechanics and muscle activations exhibited during a landing onto a sideward-tilted surface. (Note: the rationales for the choice of genders, task and using a tilted surface are described below). Therefore, I split the overall study into two main research questions, each one to be answered within a corresponding sub-study:

1) Substudy #1: **Does chronic ankle instability influence landing knee biomechanics of females?**

2) Substudy #2: **Does chronic ankle instability influence lower extremity muscle activation of females during landing?**

1.2.1 Purpose of the Overall Study

The purpose of the study is to determine if atypical knee biomechanics and muscular activation that potentially contribute to ACL injury are exhibited by female CAI participants during landing onto a sideward tilted surface.

1.2.2 Aims of the Sub-studies

There is one specific aim for each sub-study:

a) Sub-study #1: to compare knee kinematics and kinetics displayed during landing onto a tilted surface between females with CAI and those with healthy ankles.

b) Sub-study #2: to compare muscle activation of lower extremity muscles during landing onto a sideward tilted surface between females with CAI and those with healthy ankles.

1.3 General Predictions and Rationales

During landing movements, two phases commonly studied are the pre-landing and landing phases (e.g., Brown et al., 2012; Dai, Sorensen, Derrick, & Gillette, 2012; De Ridder, Willems, Vanrenterghem, Robinson, & Roosen, 2014; Delahunt et al., 2006). The primary mechanical goal of pre-landing is to position the body segments, pre-activate muscles to produce ankle joint moments in preparation for a stable landing.

Individuals with CAI in this study may display an increased inverted ankle angle with decreased peroneus longus activations in pre-landing phase, as that has been demonstrated previously (Delahunt et al., 2006). Moreover, CAI may exhibit increased peroneus longus and gastrocnemius medialis activations on the contralateral leg before initial contact (the foot landing onto a flat surface) (Levin et al., 2015). However, it is difficult, for now, to predict whether there is any difference in knee joint angles in pre-landing between CAI and healthy control (CON) participants.

For the landing phase, the overall mechanical goal is to bring the body's vertical velocity to zero and land in a stable position without injury or excessive tissue loading that could lead to chronic injury. Therefore, the roles of the muscles are to control the lower extremity's body movements and to absorb the body's vertical kinetic energy through eccentric muscle work of the extensor muscles about each of the lower extremity joints; and to do this without placing excessive loading on the articulating surfaces and soft tissues surrounding the joint regions. To do this, each joint produces angular work via a self-optimized combination of net extensor muscle moment × joint flexion displacement. I hypothesize that, for the sagittal plane, compared to CON, CAI group will land in a more upright position with a less flexed lower extremity and less flexion displacement, possibly due to limited dorsiflexion of the ankle joint and greater knee extensor muscle activations. With a less flexed knee, the patellar tendon tibia shaft angle will increase, causing the angle of pull of the quadriceps ligament on the tibia to increase; this would result in a greater anterior (shear) force component of the resultant quadriceps muscle force to act on the proximal tibia. Decreased knee flexion displacement could also result in decreased negative work to absorb the kinetic energy by the knee joint muscles, and, therefore, may lead to a greater axial compressive loading on the knee.

In the frontal plane, CAI group was expected to display greater peak ankle inversion angle during landing, with decreased ankle eversion moments and reduced peroneus longus activation as previously observed for flat surface landings (Lin et al., 2011). This could possibly be due to the greater ankle inversion angle positioning during the pre-landing phase, as discussed above. CAI participants also were predicted to exhibit a greater knee adduction angle with greater knee adduction moments to control the lower leg posture in the frontal plane and compensate for the increased ankle inversion angle. The knee adduction was defined as the movement in which the distal lower leg rotating medially in the frontal plane (knee varus). In turn, these atypical knee biomechanics then may potentially relate to mechanisms of knee injuries (Figure 1.2).

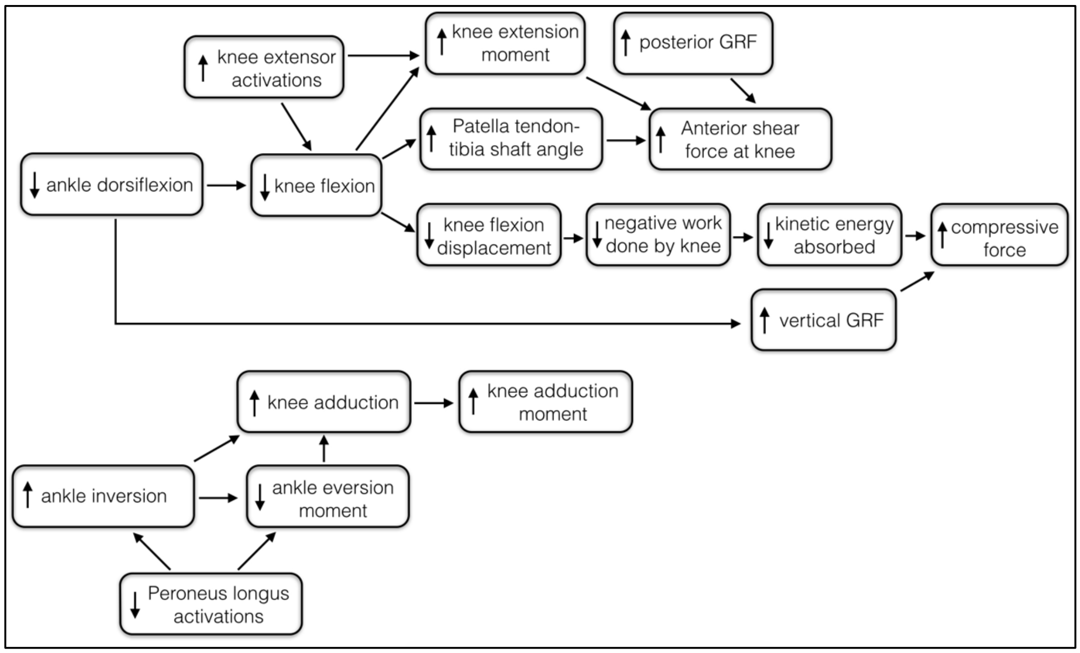

Figure 1.2. Flowchart that shows prediction of how biomechanical alterations of CAI ankle could result in atypical knee biomechanics.

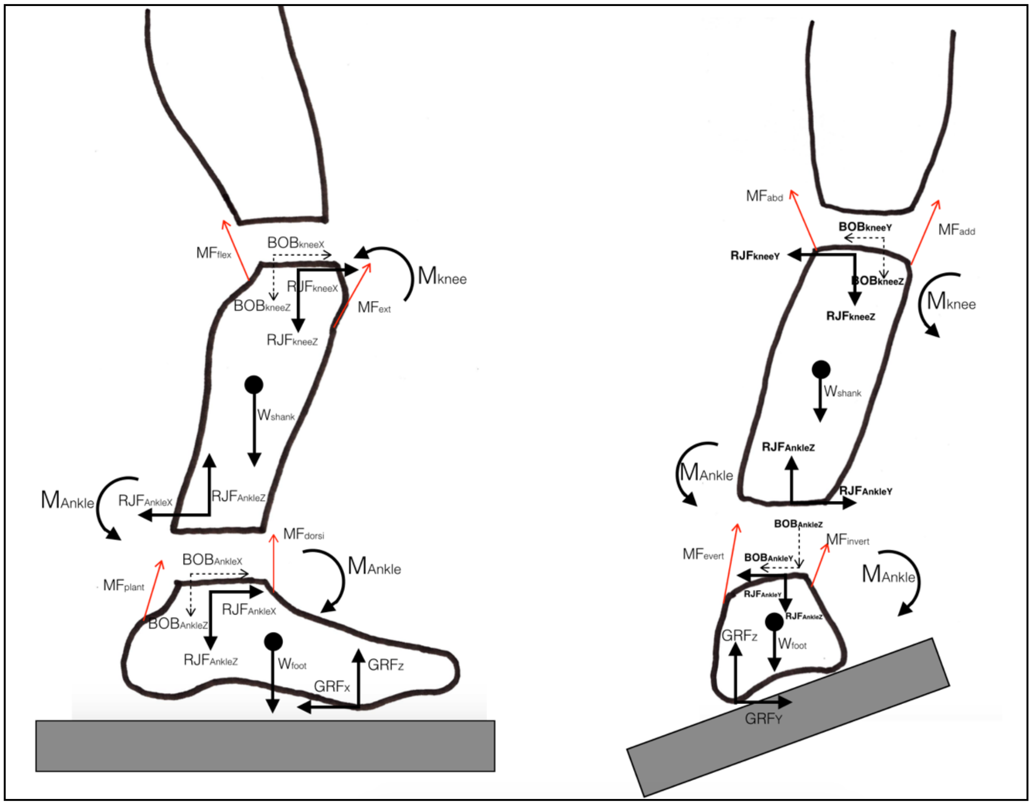

Figure 1.3. Expected free body diagram of a right leg in the sagittal and frontal plane (anterior view) during tilted surface landing. Symbols, lines and abbreviations: black dot = location of center of mass of a given segment; solid black arrows = resultant joint force (RJF_{jointC} where *joint* is the joint at which the force acts on the segment, and *C* is the room coordinate system direction that the force acts), ground reaction (GRF_C), and/or gravitational forces ($W_{segment}$); red arrows = muscle group forces ($MF_{muscle\ group}$); dashed arrows = bone-on-bone (BOB) forces; curved arrows = joint moments (M_{jt}, where jt = the joint about which the joint moment acts on the segment). Note: muscle group forces and bone-on-bone forces for a given joint are shown only on the distal segment for clarity on the diagram; however, each of those forces act on the proximal segment in equal magnitude and opposite direction to that shown on the distal segment.

1.4 Hypotheses

Based on the general predictions and their supporting rationales, the following is predicted:

Sub-study #1:

Compared to CON participants, the CAI group will exhibit the following knee biomechanics:

a) less ankle, knee, and hip joint flexion displacements during the landing phase.

b) greater peak knee extension moment.

d) less peak ankle eversion and

e) greater knee and hip adduction joint moments.

f) less ankle, knee, and hip eccentric work in the sagittal plane; and less ankle eccentric work in the frontal plane.

Sub-study #2:

Compared to CON participants, the CAI group will also exhibit different electrical muscle activations (measured by averaged linear envelope electromyography):

a) increased knee extensor muscle activations during the landing phase.

b) increased knee co-contraction ratio (knee extensor to knee flexor contractions as measured by rectus femoris/biceps femoris) during the landing phase.

c) decreased peroneus longus activation in pre-landing and landing phases.

d) increased tibialis anterior activation in the landing phase

e) increased co-contraction of tibialis anterior and gastrocnemius lateralis (sagittal plane), and increased co-contraction of tibialis anterior and peroneus longus (frontal plane).

1.5 In-depth Rationales for Choice of Gender, Landing Task, and Tilted Surface

For this study, we will only study female participants, because female athletes have a significantly higher rate of ACL injury relative to male athletes (Agel, Arendt, & Bershadsky, 2005; Shultz, 2015), potentially due to different anatomical structure of the knee (R. Nunley, Wright, Renner, Yu, & Garrett, 2003; Shultz, 2015), lower quadriceps and hamstrings strength (Huston & Wojtys, 1996) and hormonal effects that could reduce load to failure (Slauterbeck, Clevenger, Lundberg, & Burchfield, 1999). In turn, biomechanically, during dynamic movements, females tend to have less knee flexion (Lephart et al., 2002), greater peak knee valgus and internal rotation angle (McLean et al., 2007), and greater anterior shear force acting on the proximal tibia articulating surface and knee extensor moment (Chappell et al., 2002), which may reflect increased tensile stress of the ACL (Lephart et al., 2002; Shultz, 2015). Therefore, we anticipated that, for females with CAI, atypical knee movements and kinetics would occur and could place greater loading on knee joint structures, such as the ACL and/or tibio-femoral contact surfaces.

A landing task was chosen because ACL injury often occurs during landing and landing is a very common element in many sport movements (Agel et al., 2005; Alentorn-Geli et al., 2009; Boden, Dean, Feagin, & Garrett, 2000). It has been reported that over 30% ACL injuries occurred during landing phase (Boden et al., 2000). Moreover, landing is also a commonly used testing protocol to understand causation of ACL injury situations that occur during landings and to explore the biomechanical alterations of CAI individuals (Delahunt, Monaghan, & Caulfield, 2006; Gribble & Robinson, 2010; Lin et al., 2011).

For this study, an inverted surface (a tilted surface that makes the foot roll inward, i.e., invert) rather than a level surface was chosen because landing onto an inverted surface has been suggested as a more demanding and realistic simulation of a lateral ankle sprain (Chen, Wortley, Bhaskaran, Milner, & Zhang, 2012). It is also similar to trail running on an uneven surface, landing on another's foot in jumping related sports or even walking on a tilted sidewalk. Most previous studies on CAI landing focused on level surface landings (Brown et al., 2011; Gribble & Robinson, 2010; Gribble & Robinson, 2009).

To our knowledge, there are only a few published studies comparing inverted landing biomechanics of CAI participants to healthy controls (Gutierrez et al., 2012; Levin et al., 2015). However, these researchers did not examine the knee biomechanics that related to mechanism of injury.

1.6 Significance of Study

The influence of ankle injury on the occurrence of particular knee injuries (e.g., ACL injury) or the development of chronic knee conditions (e.g., knee osteoarthritis), may be more relevant than we previously thought, due, in part, to the high prevalence of ankle injuries that occur during sports and physical activities (Kobayashi & Gamada, 2014). Female athletes with CAI participate in sports activities involving landing (e.g., basketball, volleyball, and soccer). But, are they at a higher risk of knee injury than others without CAI? Do they need additional training to prevent knee injury?

At present, there are no evidence-based answers. The present study will help determine how CAI affects knee biomechanics and muscle activations that, subsequently, will provide a more comprehensive understanding of the mechanisms of loading known to be related to knee injury and the development of chronic knee conditions. In turn, this may help professionals develop improved training programs for knee injury prevention.

This study may also provide a better understanding of how CAI affects knee kinetics due to the potential to obtain more accurate and valid outcomes than that from previous studies. To our knowledge, the proposed study is the first to directly measure the GRF of CAI participants during landings onto an inverted surface.

CHAPTER 2

REVIEW OF LITERATURE

In this chapter, overall, the background of CAI and ACL injury were provided. I began with the background of CAI; then literature that demonstrated how altering the mechanics at the ankle joint influenced the other lower extremity joints. This was followed by a section focused on the anterior cruciate ligament (ACL). The purpose of the CAI background section was to introduce fundamental information about CAI and explore how CAI consequences may influence mechanisms related to knee injury causation. Specifically, for the background of CAI literature, I reviewed a) deficits of CAI and b) biomechanics of CAI in landing.

The biomechanical effects on the lower extremity of altering ankle mechanics" provide rationales underlying my predictions of how CAI ankles affect knee and hip biomechanics. In this chapter, the following topics are reviewed: a) relationship between ankle and knee injuries and b) relationship between ankle and knee biomechanics.

Next, I reviewed ACL injury because ACL injury is common and potentially serious and costly (Brophy, Wright, & Matava, 2009) and there may be a significant association between CAI and ACL injury (Kramer et al., 2007). Thus, presented in the ACL section are first, the anatomy and function of the ACL, to provide the basic background of this ligament and how it is susceptible to injury. Second, the injury rates and consequences of ACL injuries have been reviewed to show the need for greater understanding of the causation of ACL injuries. Third, the loading mechanisms of ACL injuries are presented to examine the biomechanics factors that contribute to ACL loading and that may be relevant to this study.

2.1 Deficits of CAI

CAI commonly develops following the lateral ankle sprain (Hertel, 2002). CAI has been described "as an encompassing term used to classify a subject with both mechanical and functional instability of the ankle joint" (Gribble et al., 2013, P. 586). This section will review the potential deficits of CAI, including reduced proprioception, neuromuscular deficits, postural control deficits that could influence the biomechanics of the lower extremity joints.

2.1.1 Proprioception

I believe the deficits in position and force sensing may increase the ankle injury risk, such as higher possibility putting ankle in an unstable position and resulting in recurrent ankle sprains. Reduced joint position sense has been observed in many previous studies (De Ridder et al., 2014; Lee & Lin, 2008; Nakasa, Fukuhara, Adachi, & Ochi, 2008; Sekir, Yildiz, Hazneci, Ors, & Aydin, 2007; Witchalls, Waddington, Blanch, & Adams, 2012). Willems et al. (2002) tested active and passive joint position sense of ankle inversion and eversion for 10 CAI participants using isokinetic dynamometer. They observed significantly less accurate active position sense for the CAI group compared to healthy control group (Willems et al., 2002). Other studies reported significant less accurate passive position sense in ankle inversion for CAI group (Lee & Lin, 2008; Sekir et al., 2007). Researchers also used goniometer footplate to measure the position sense and also observed a greater error for CAI group (Nakasa et al., 2008; Witchalls et al., 2012).

CAI participants also exhibit deficits in force sensing (Arnold & Docherty, 2006; Docherty, Arnold, & Hurwitz, 2006; Wright & Arnold, 2012). Wright and Arnold (2012) tested the error of being able to reproduce ankle eversion forces using a load cell for 32 CAI participants and 32 healthy controls. The error was the difference between the predetermined target and reproduction forces. As greater errors were observed for the CAI group, Wright and Arnold concluded that individuals with CAI are less accurate in force production of desired magnitudes. Similar findings were also reported for eversion moments production in other studies (Arnold & Docherty, 2006; Docherty et al., 2006).

2.1.2 Neuromuscular Deficits

Ankle neuromuscular deficits of CAI could influence ankle joint mechanics that place the joint at a high risk of recurrent sprain. Studies of neuromuscular deficits have been mainly focused on neuromuscular reaction time and muscle activation level. Most studies used a trapdoor and measured the reaction time of the peroneus longus activation when the ankle was suddenly inverted. Vaes, Duquet and Van Gheluwe (2002) studied the latency of the peroneus longus muscle during a sudden inversion using a trapdoor for 40 CAI participants. The latency was measured as the time lapse between the onset of EMG of peroneous longus muscle and the start of the inversion movement measured by an accelerometer (Vaes, Duquet, & Van Gheluwe, 2002). They did not observe a significant difference in the latency between CAI and control participants. However, they observed shorter first deceleration time during sudden inversion for CAI, which may be due to less control of inversion velocity (Vaes et al., 2002). However, controversy still remains, because other studies observed a significant greater latency of peroneous longus onset for CAI (Karlsson & Andreasson, 1992; Konradsen & Ravn, 1990). This discrepancy among studies may be due to different trap door protocols and criteria of EMG onset used in these studies.

CAI participants also demonstrate atypical muscle activation during various weight-bearing movements (Brown, Ross, Mynark, & Guskiewicz, 2004; Caulfield, Crammond, O'Sullivan, Reynolds, & Ward, 2004; Ty Hopkins, Coglianese, Glasgow, Reese, & Seeley, 2012). Researchers observed decreased peroneus longus activation in the pre-landing phase for CAI compared to CON (Caulfield et al., 2004; Delahunt et al., 2006) and decreased soleus activation in the landing phase (Brown, Ross, Mynark, & Guskiewicz, 2004) of single leg landings. During the lateral hop task, CAI participants demonstrated increased rectus femoris, tibialis anterior and soleus muscle activation in both pre-landing and landing phases (Delahunt, Monaghan, & Caulfield, 2007). CAI individuals also exhibited increased tibialis anterior activation in stance phase and increased peroneus longus in heel contact and toe off sub-phases of walking (Ty Hopkins et al., 2012). Ty Hopkins and colleagues (2012) surmised, although did not prove, that these atypical muscle activations could contribute to recurrent joint instability and may make the ankle vulnerable to excessive external loadings (e.g. GRF).

2.1.3 Postural Control Deficits

Postural control deficits have been observed for CAI in both static and dynamic stability, and may influence the lower extremity and even the trunk biomechanics. Single leg standing is the most commonly used protocol to test static stability in CAI study. Postural sway measured by force plate has been used to quantify the static stability, with greater sway (displacement or sway area) indicating less stability. Most studies of static stability observed increased postural sway for CAI participants (Ben-Ad, 2009;

Leanderson, Wykman, & Eriksson, 1993; Lee & Lin, 2008; McKeon et al., 2008). However, other studies did not observe any significant difference between CAI and CON participants (Hubbard, Kramer, Denegar, & Hertel, 2007; Knapp, Lee, Chinn, Saliba, & Hertel, 2011). The controversy is possibly due to the variability of the inclusion criteria of CAI participants.

The star excursion balance test (SEBT) has been used to assess the dynamic stability. SEBT is a dynamic postural stability test that involves completion of a functional task without compromising one's base of support (Gribble & Hertel, 2003). During the SEBT, a participant will maintain the base of support on the center of a grid with one leg while the other leg is reaching maximally in 8 different directions defined by 8 stripes of tapes at 45-degree angles from the center. The base of support of the stance leg must not change during the test except for rest after each reach direction. The participant should maintain single leg stance with eyes open and hands on the hips. Most studies observed decreases in reaching distance of SEBT for CAI participants. However, decreased distances were detected in different reaching directions, such as all directions (Gribble, Hertel, Denegar, & Buckley, 2004; Olmsted, Carcia, Hertel, & Shultz, 2002), only anterior direction (Hoch, Staton, McKeon, Mattacola, & McKeon, 2012), or posteromedial and anterior direction (Hubbard et al., 2007).

In addition, other tests such as hopping and landing, have been also used to assess dynamic stability. The greater number of times lost balance in hop tests (Eechaute, Vaes, & Duque, 2009) and longer time to stabilization in landing (Brown et al., 2004) indicated CAI exhibited some deficits in dynamic postural control.

2.2 Landing Biomechanics of Individuals with CAI

CAI could influence biomechanics of both ankle and other joints on the same kinetic chain during movement. Landing task is a commonly used testing protocol in examination of performance characteristics of CAI (Delahunt, Monaghan, & Caulfield, 2006; Gribble & Robinson, 2010; Lin et al., 2011). Compared to healthy control participants, though some controversies, CAI participants exhibited alterations in lower extremity kinematics in landing. Delahunt and colleagues (2006) recorded 3D kinematics from 24 CAI and 24 CON participants during a drop landing from a 35-cm high platform. They observed a greater ankle inversion angle and reduced hip external rotation angle for CAI group in pre-landing phase (Delahunt et al., 2006). During the landing phase, CAI group exhibited less ankle dorsiflexion angle compared to CON. However, no significant difference was observed for the knee joint. On the contrary, however, Caulfield and colleagues (2002) observed increased dorsiflexion angle from 10 ms before to 20 ms after initial landing for CAI during drop landing from a 40-cm high platform. Moreover, they observed a greater knee flexion angle for CAI from 20 ms before to 60 ms after initial landing (Caulfield & Garrett, 2002). Increased ankle inversion angle for CAI was also observed by other studies during lateral hopping (Delahunt et al., 2007) and stop jump landing (Lin et al., 2011). For the knee joint, Gribble and colleagues (2009, 2010) observed CAI displayed decreased knee flexion angle during the landing phase of a forward, standing long jump. Brown and colleagues (2011) investigated hip kinematics of mechanical and functional ankle instability individuals (MAI and FAI, respectively) displayed during landings. For hip flexion angles, they observed that the MAI group displayed greater flexion at initial ground contact and at peak hip flexion during the landing phase compared to the coper group (defined as individuals who had a prior lateral ankle sprain but did not subsequently develop CAI) (Brown et al., 2011).

Moreover, CAI group also exhibited different kinetics during landing compared to CON (Delahunt et al., 2006; Doherty et al., 2014; Terada et al., 2013). It was observed that the CAI group exhibited greater peak posterior and vertical GRF (Delahunt, Cusack, Wilson, & Doherty, 2013), reduced hip flexion moment and increased hip stiffness (Doherty et al., 2014) and less percentage of knee energy absorption and greater percentage of ankle energy absorption (Terada et al., 2013). However, it is still unknown whether these altered lower extremity biomechanics could relate to the mechanisms of injury.

2.3 Potential Links between Ankle and Knee Joints Biomechanics

These alterations of lower extremity biomechanics displayed by CAI compared to CON individuals suggest that CAI structural and biomechanical ankle joint abnormalities could also lead to abnormal knee and hip joint biomechanics. This supposition is explored below.

2.3.1 Relationship between Ankle and Knee Injuries

There are a very limited number of studies investigating whether prior ankle injury is related to increased risk of experiencing other lower extremity injuries, that is, knee injuries (Gordon, Distefano, & Denegar, 2014; Kramer et al., 2007). Moreover, the findings are contradictory among studies. Kramer and colleagues (2007) examined the factors associated with ACL injury for young adult female soccer athletes and observed that of the 66 limbs with history of ACL injury, 52% had a previous ankle injury. Though no causal relationship was proved, they suggested that those with a history of ACL injury were more likely to have a previous ankle injury and the factors that may explain this ACL and ankle injury association could be related to joint tissue integrity. By determining the factors that were associated with ankle injury and those factors related to ACL injury in separate analyses of their participant data, then finding which factors were common to both types of injuries, they determined two injury-related factors: increased generalized joint laxity and decreased iliotibial band flexibility.

In contrast to Kramer, et al. (2007) outcomes, Gordon et al. (2014) observed totally different results from their investigation of injury incidences for 246 college and professional women basketball players. They found that the players who had suffered an ankle sprain (*n*=170) were less, not more likely to have suffered an ACL injury (Gordon et al., 2014). Therefore, they suggested that a history of ankle sprain may not be a risk factor of ACL injury (Gordon et al., 2014). The different outcomes between the two studies could possibly attribute to different sports and skill levels.

2.3.2 Relationship between Ankle and Knee Biomechanics

As ankle and knee joints are part of the same, lower-extremity kinetic chain, alterations in ankle biomechanics could influence the knee biomechanics. In fact, this supposition underlies the rationale for using foot orthosis to treat certain forms of knee pain. Maly and colleagues (2002) studied the mechanical effects of foot orthoses on knee adduction moments during walking. However, no significant orthosis effect was observed for knee adduction moments, possibly because participants exhibited a more toe-out foot-landing position while wearing the orthoses (Maly, Culham, & Costigan, 2002). This study may provide some insights that foot progression angle possibly affect the knee loadings, in which the toe-out positioning may reduce medial knee loading (Maly et al., 2002). Therefore, for our study, we may need to control for this variable in order to investigate CAI effects on knee loadings.

Studies focused on the effects of preventative injury braces (worn to stabilize the ankle or knee joint) on lower-extremity landing mechanics also are good examples to demonstrate that the entire lower extremity kinetic chain is affected when the mechanics of one joint are altered.-Several ankle bracing studies have investigated alterations of the lower extremity biomechanics (Simpson et al., 2013; Venesky, Docherty, Dapena, & Schrader, 2006). Simpson and colleagues (2013) observed that when participants were wearing a semi-rigid ankle brace, the restriction of ankle motion led the participants to land in a more flexed knee joint angle. They surmised that this occurred because the participant needed to rotate the tibia about the knee joint to orient the foot into a forefoot landing position. However, the constrained ankle joint motion led to decreased lower extremity joint flexion displacements during the landing phase. Consequently, reduced displacements likely also caused greater peak vertical impact forces, as lower extremity flexion is part of an effective impact force attenuation strategy. Otherwise, the more constrained neutral ankle positioning of the ankle brace tended to cause the lower extremity joints to remain in more neutral positions about all axes during landing. Of interest, also, though, was the finding that individuals vary widely in their landing strategies, as has been commonly reported before (Scholes, McDonald, & Parker, 2012). Individual landing strategies had a greater influence on the outcomes than the presence/absence of the brace. Hence, it will be important in my study to control for the other factors that could influence the landing movement (e.g., trunk posture, landing zone location) and increase the landing trial number for each participant.

Moreover, constraining ankle motion via use of a brace also has been shown to affect knee angular kinetics (Venesky, et al., 2006), although no mention of the influences of the ankle brace on the hip kinematics or kinetics were provided. Moreover, Venesky et al. reported that, for individuals when landing with an ankle brace compared to those landings without bracing, an increased knee external rotation moment and a tendency ($p = 0.08$) toward a reduced peak knee abduction moment were exhibited during single-leg drop landings onto a tilted surface. The behavioral meaning of the outcomes as related to this study is the altered ankle biomechanics or muscle co-contraction of CAI could influence the knee joint loadings.

Knee biomechanical alterations have been observed for the CAI group during landing, though disagreements exist. Caulfield and Garret (2002) detected a greater knee flexion angle for CAI from 20 ms before to 60 ms after initial contact during a single-leg drop landing. Delahunt et al. (2006) used the same task, but no significant difference in knee flexion angle was observed between CAI and control individuals. However, on the other side, Gribble et al. (2009, 2010) observed a reduced knee flexion angle for the CAI group during a single-leg forward jump landing. The controversies among previous studies may be due to different landing protocol. For now, it seems that the biomechanical link between ankle and knee joint is not well understood. Therefore, more evidence may be needed to comprehend the mechanism of how ankle biomechanics could influence knee injuries. One particular knee injury of concern in sports is ACL injury. In the following sections, the background and mechanisms of ACL injury were introduced.

2.4 ACL anatomy and Function

The primary function of the ACL is to resist anterior translation and internal rotation of the tibia relative to the femur (Matsumoto et al., 2001; Sakane et al., 1997). The ACL has been also reported to prevent valgus instability of the knee (Matsumoto et al., 2001). The femoral origin of ACL is at the posterior part of the medial surface of the lateral femoral condyle (Arnoczky, 1983), whereas the tibia insertion of ACL is located in the area between the medial and lateral tibial spine on the tibial plateau (Zantop, Petersen, & Fu, 2005). In general, the ACL has been divided into 2 bundles, anteromedial and posterolateral bundle, based on their tibia insertion locations (Girgis, Marshall, & Monajem, 1975).

2.5 Injury Rate of ACL Injuries

ACL injury is a common knee injury in sports activities. The reported incidence rate during games is 8.06 per 10,000 athletes for NCAA football from 2004 to 2008, which is about 10 times higher than that in practice (Dragoo, Braun, Bartlinski, & Harris, 2012). Granan et al. (2008) reported 2714 ACL reconstructions performed from 2004 to 2006 with the annual reconstruction incidence rate of 34 per 100,000 for general populations (Granan et al., 2008). The most frequent activities causing ACL injury in Norway were soccer, team handball and alpine skiing (Granan et al., 2008). Gianotti et al. (2009) reported population-based annual ACL surgery incidence rate was 37 per 100,000 New Zealand residents from 2000 to 2005, in which 65% was sports or recreation related injuries (Gianotti, Marshall, Hume, & Bunt, 2009). Kobayashi et al. (2010) investigated 880 female athletes with ACL injury and observed that about 38% ACL injury occurred in basketball. Other sports in which ACL injury commonly occurred are ski (15.2%), handball (12.0%) and volleyball (9.2%) (Kobayashi, Kanamura, Koshida, Miyashita, & Okado, 2010). In basketball, Krosshaug and colleagues analyzed 39 videos of ACL injury cases and found 23 out of 39 injuries occurred during landing (Krosshaug et al., 2007). In handball, the major injury mechanisms are plant-and-cut movement (60%) and landing (20%) (Olsen, Myklebust, Engebretsen, & Bahr, 2004).

Female athletes have a significantly higher rate of ACL injury compared to comparable male athletes (Agel et al., 2005; Beynnon et al., 2014; Hootman, Dick, & Agel, 2007a). Beynnon et al. 2014 observed that females were more than twice as likely to have a first-time ACL injury compared with males (Beynnon et al., 2014). Specifically, the injury rate of females was about three times as that of males in basketball and soccer (Agel et al., 2005).

2.6 Consequences of ACL injuries

Short and long term consequences were reported after ACL reconstructions. Researchers observed decreased muscle strength in ACL injured participants after ACL reconstructions (Ageberg, Roos, Silbernagel, Thomee, & Roos, 2009; Eitzen, Holm, & Risberg, 2009; Gokeler et al., 2014; Thomas, Villwock, Wojtys, & Palmieri-Smith, 2013). Most studies observed strength deficits of the quadriceps and hamstrings (Ageberg et al., 2009; Gokeler et al., 2014; Lautamies, Harilainen, Kettunen, Sandelin, & Kujala, 2008; Thomas et al., 2013). Other studies also observed a reduction of hip extensor strength (Geoghegan, Geutjens, Downing, Colclough, & King, 2007; Thomas et al., 2013). The deficit in muscle strength after ACL reconstructions was still displayed at the one year follow-up (Bizzini, Gorelick, Munzinger, & Drobny, 2006; Gobbi, Mahajan, Zanazzo, & Tuy, 2003) and could persist two years or longer (Ageberg et al., 2009;

Eitzen et al., 2009; Holsgaard-Larsen, Jensen, Mortensen, & Aagaard, 2014). Eitzen et al. (2009) also reported that over 20% of participants with quadriceps strength deficits had strength deficits that persisted two years after surgery (Eitzen et al., 2009). Therefore, ACL injury and reconstruction could result in a delayed return to sport for athletes.

Participants with ACL reconstruction also exhibited alterations in lower extremities that may lead to a decrease in performance. Hofbauer et al. (2014) investigated ACL-reconstructed knee kinematics during single-legged forward hop landing at 5 months after surgery. They observed a less flexed (20.9° vs 28.4°), more externally rotated (12.2° vs 6.5°) and medially translated (3.8 vs 2.3 mm) knee in landing on the operated side compared to the contralateral intact ACL limb (Hofbauer, Thorhauer, Abebe, Bey, & Tashman, 2014). From 5 to 12 months, these differences significantly decreased. Shabani et al (2014) analyzed kinematic data of 15 unilateral ACL injury patients pre- and post-ACL reconstruction, and 15 healthy control participants during walking. ACL reconstructed knees exhibited a significantly greater knee extension during the entire support phase compared to preoperative knees (ACL injured, but has not yet gotten surgery), but significantly less extended compared to healthy controls during late support and early swing phase (Shabani et al., 2014). Fontenay et al. (2014) compared kinematic difference between the ACL reconstructed limb and contralateral intact-ACL limb during a single-legged jump. A lower jump height with less knee extension and ankle plantar flexion at takeoff was observed for operated limb (De Fontenay, Argaud, Blache, & Monteil, 2014). These less extended joints at takeoff resulted in a 35% less total positive power compared to that for intact-ACL limb side (De Fontenay et al., 2014).

2.7 Loading mechanisms of ACL injuries

In this section, several risk factors of ACL injury that related to movement and loading mechanisms are discussed: anterior shear force at the proximal tibia, knee varus/valgus moments, knee internal rotation moments, axial compressive forces and knee flexion angles.

2.7.1 Anterior Shear Force

ACL loadings have been studied by applying external loads to the knee in vivo (studying a living organism) and in vitro (studying isolated tissue or organ, e.g. cadaver specimens) studies. Previous studies suggested that the anterior shear force at the proximal end of the tibia is the primary contributor to ACL loading (Berns, Hull, & Patterson, 1992; Dai, Herman, Liu, Garrett, & Yu, 2012; Markolf et al., 1995). Berns et al. (1992) measured strain within the anteromedial bundle of the ACL of 13 human knee specimens. They found that the anterior shear force was the primary determinant of strain in the anteromedial bundle of the ACL. Other researchers also observed that ACL tensile force was approximately equal to the applied anterior shear force (100 N) when the knee was at 30° flexion and increased to 1.5 times of the shear force the knee at 0° flexion (Markolf et al., 1995). ACL tensile forces due to the applied anterior shear force increased as knee flexion angle decreased (Sakane et al., 1997). Researchers applied 110 N of anterior shear force on the tibia of 10 cadaver knee joints and observed more than 90 N of ACL tensile force when knee flexion was less than 30° (Sakane et al., 1997). However, the ACL tensile force reduced to 70 N and 59 N when knee flexion was at 60° and 90°, respectively, with the same magnitude of applied anterior shear force (Sakane et al., 1997). These results of the previous studies suggested that ACL may be more vulnerable to excessive anterior shear force at low knee flexion positions.

In a vivo study, Fleming et al. (2001) implanted a strain transducer on the ACL of 11 participants (Fleming et al., 2001). External anterior shear force, ranged from -90 N to 130 N in 10 N increment, was applied on the proximal tibia when the knee was at 20° flexion for both weightbearing and non-weightbearing conditions. During low shear force (< 40 N), the ACL strain values in weightbearing was greater than that in non-weightbearing condition. During high shear force, the strain values of the two conditions became equal. For both conditions, ACL strain increased as the shear force increased (Fleming et al., 2001).

However, many controversies still existed in the literature, such as whether the amount of anterior shear force due to the quadriceps muscle force is large enough to rupture the ACL or whether the posterior GRF during movement will reduce the anterior shear force. Further studies are still needed to provide more evidence to resolve these controversies.

2.7.2 Knee Internal Rotation Moment

Applying a knee internal rotation moment further increases ACL loading when combined with anterior shear force (Berns, Hull, & Patterson, 1992; Markolf et al., 1995). It has been observed in cadaver knees that this loading combination produced dramatic increases of ACL tensile force at full extension and hyperextension (Markolf et al., 1995). This loading combination produced the highest ligament forces recorded in the study and is the most dangerous in terms of potential injury to the ligament. However, the addition of external rotation moment to a knee loaded by anterior shear force actually decreased the ACL tensile force (Markolf et al., 1995) and ACL strain (Kruse, Gray, & Wright, 2012). Markolf et al. (2004) tested effect of 100 N applied quadriceps muscle force on ACL tensile forces. Combined with a 10 Nm knee internal rotation moment, ACL tensile force was approximately twice as that combined with a 10 Nm knee external rotation moment (Markolf, O'Neill, Jackson, & McAllister, 2004). Similarly in a vivo study, an internal rotation moment of 10 Nm strained the ACL when the knee was non-weightbearing while an same amount of external rotation moment did not (Fleming et al., 2001).

2.7.3 Knee Abduction/Adduction Moment

Isolated adduction and abduction moments had very small effect on ACL strain (Fleming et al., 2001), however, combined with anterior shear force or transverse-plane knee load, the adduction and abduction moments increased ACL tensile forces (Gabriel, Wong, Woo, Yagi, & Debski, 2004; Kanamori et al., 2000; Kanamori et al., 2002). The addition of 10 Nm adduction knee moments to the 100 N anterior shear force increased ACL tensile force (to 220 N) when the knee flexion angle was greater than 50° and less than 30°; the addition of 10 Nm abduction moments increased ACL tensile force (to 200 N) when the knee flexion was greater than 5° (Markolf et al., 1995). A combination load of 10 Nm internal rotation and 10 Nm abduction moments could result in a 90 N of ACL tensile forces in vitro studies (Kanamori et al., 2000; Kanamori et al., 2002), whereas a combination load of 10 Nm external rotation and 10 Nm abduction moments only resulted in 30 N of ACL forces (Kanamori et al., 2002).

Computer simulation models have also been used to study ACL loading and showed some advantages in testing the function of individual structures while controlling for all other structures (Pflum, Shelburne, Torry, Decker, & Pandy, 2004; Shelburne, Torry, & Pandy, 2005; Shin, Chaudhari, & Andriacchi, 2011). Shin et al. (2011) tested the influence of combined knee valgus and internal rotation moment on ACL strain during single-leg landing (Shin et al., 2011). The researchers applied in vivo human loading data to a validated computer simulation model. They observed that when the two moments were applied individually, neither of them strained ACL. However, combined moments increased ACL strain significantly. In general, these studies suggested that when excessive valgus and knee internal rotation moments are combined at small knee flexion angles, ACL may be at a higher risk of strain and injury.

2.7.4 Axial Compressive Force

The compressive force (bone on bone force between tibia and femur) with a posterior tibial plateau slope has been considered as an important loading mechanism of ACL (Boden, Torg, Knowles, & Hewett, 2009; Meyer & Haut, 2008; Wall, Rose, Sutter, Belkoff, & Boden, 2012). In a vitro study, Meyer and Haut (2008) applied compression load on 7 pairs of knee joints with repetitive tests at increasing magnitude of load until ACL failure. Before the failure, they observed the femur displaced posteriorly relative to the tibia while the tibia rotate internally (Meyer & Haut, 2008). ACL was ruptured in all knee joints with approximately 5.4 KN of compressive force. Wall et al. (2012) further tested 6 pairs of cadaveric knees (15° flexion) with axial compression and compression combined with a quadriceps force. The reported magnitude of compressive force to rupture the ACL was 10.8 KN and 6.1 KN for the two loading conditions, respectively. They concluded that isolated compressive force could produce ACL injuries. Moreover, the addition of a quadriceps force could significantly reduce the compressive force required for ACL injury, which is more dangerous situation of ACL (Wall et al., 2012).

2.7.5 Knee Flexion Angle

Small knee flexion angles could increase the tibial anterior shear force by increasing the patella tendon tibia shaft angle (Nunley et al., 2003; Yu & Garrett, 2007), defined as the angle between the longitudinal axes of patella tendon and tibia shaft (Fig 2.1). Nunley et al. (2003) measured patella tendon tibia shaft angles of 10 male and 10 female recreational athletes at different knee flexion position using radiographs. A linear relationship between the knee flexion angle and patella tendon tibia shaft (PTT) angle has been reported (Nunley et al., 2003). The reported regression equations to estimate the angle of pull of the quads on the tibia via the patellar tendon for males and females, respectively were:

$$PTT_{male} = 22.03 - 0.3\times\theta$$

$$PTT_{female} = 25.70 - 0.3\times\theta$$

where θ is the knee flexion angle at some instant in time. The averaged PTT angle was about 3.7° greater for females compared to males based on the reported regression equations. The greater patella tendon tibia shaft angle for females resulted in a 13.2% increase in the anterior shear force applied to the tibia. They suggested that patella tendon tibia shaft angle could be a possible risk factor for the ACL injury, especially for female athletes.

Knee flexion also increases the PTT, and thus, also influences the anterior tibial shear force. Figure 2.1 demonstrates that decreasing knee flexion angle could result in a greater patella tendon tibia shaft angle, thus a greater magnitude of the anterior shear force component of a given amount of resultant quadriceps muscle force.

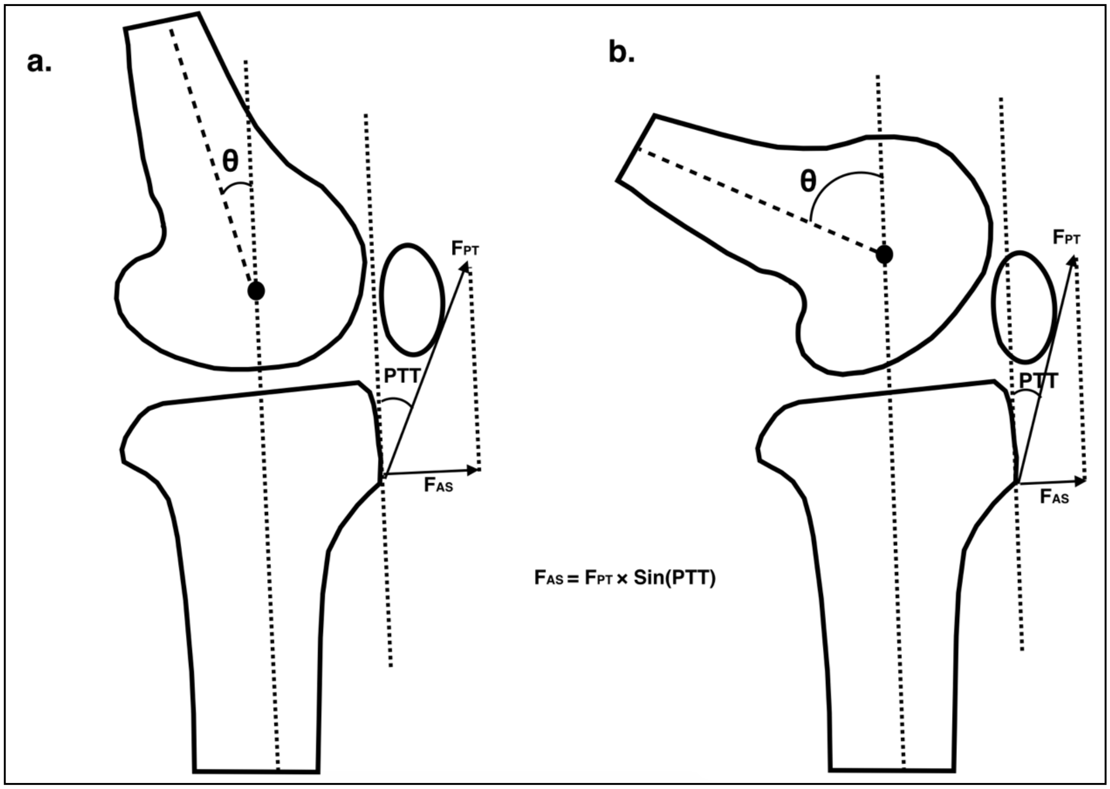

Figure 2.1. Patella tendon tibia shaft angle (PTT) at (a) a small knee flexion angle θ and (b) a great knee flexion angle. F_{PT} = force at patella tendon, F_{AS} = anterior shear force at proximal tibial. With a given F_{PT}, a greater PTT angle could result in a greater F_{AS}.

Small knee flexion angle could also increase ACL tensile force by increasing ACL elevation angle and deviation angle (Guoan Li, DeFrate, Rubash, & Gill, 2005; Yu & Garrett, 2007). The ACL elevation angle is defined as the angle between the longitudinal axis of ACL and the plane of the tibial plateau; and the ACL deviation angle is defined as the angle between the projection of the ACL onto the tibial plateau and the anterior-posterior direction of the tibial (Guoan Li et al., 2005). Li et al (2005) studied elevation and deviation angles of ACL during weight-bearing flexion in 5 human participants using fluoroscopic images and MRI based computer models. Both elevation and deviation angles increased when knee flexion decreased and reached their peak values (65° and 12°, respectively) at full knee extension. Therefore, the resultant force along the ACL would increase at smaller knee flexion angles due to greater elevation and deviation angles of ACL (Fig 2.2).

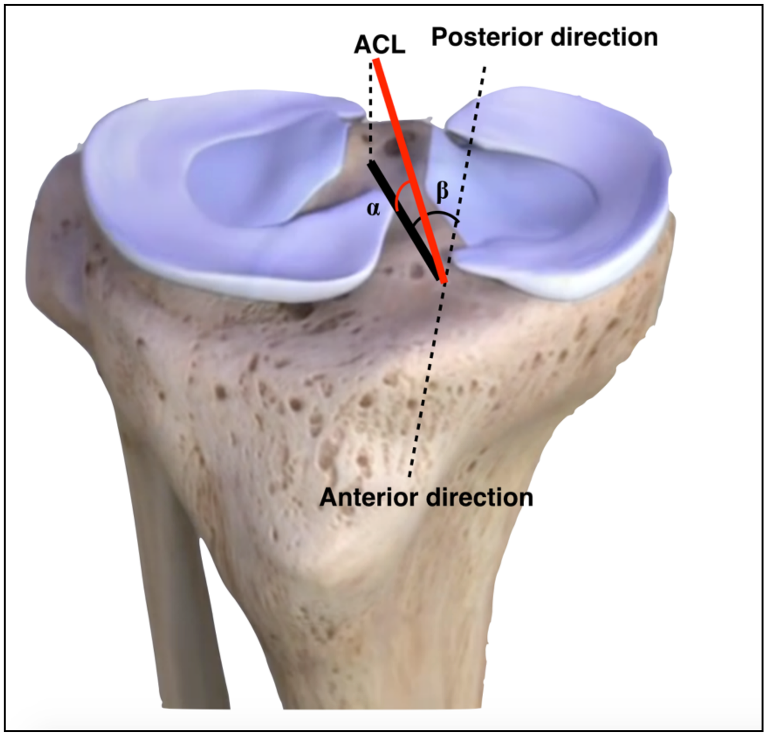

Figure 2.2. Frontal view of ACL elevation angle (α) and deviation angle (β) of a right knee. The red solid line indicates ACL; the black solid line indicates projection of ACL on the plane of tibial plateau.

In conclusion of this chapter, the deficits and biomechanical alterations of CAI and mechanisms of ACL injury have been reviewed. It provided theoretical evidences and strengthened rationale that CAI may influence the knee loading and injuries.

CHAPTER 3

METHODS

3.1 Research design

Cross-sectional, non-randomized, experimental design: the independent variable was the ankle stability group (CAI and CON). There was one overall study in terms of study design and data collection; data reduction and statistical analyses are described separately for the 2 sub-studies.

3.2 Participants

Based on an a priori power analysis (power = .8, Cohen's d = 0.8; GPower®, v. 3.1.9), using data from previous studies (Gutierrez et al., 2012; Liu, 2013), 10 to 19 participants in each of the two ankle stability groups (CAI and CON) were needed. We collected 21 participants for each group to improve the statistical power. This sample size had enough power (> .8) to detect a 4° group difference of joint angles (Gutierrez et al., 2012) and a 0.3 $Nm \bullet kg^{-1}$ difference for knee joint moments (Liu, 2013).

All participants were:

a) female

b) ages 18 to 35 years

c) recreationally active and participating in sports or other physical activities for at least 1.5 hours per week.

d) currently participating at least once a week in an organized sport activity (e.g., practice and competition play in a league or attending a class, rehearsal, or performance in an activity such as ballet, gymnastics, etc.) or have competed at intermediate or advanced level sports team (e.g., high school varsity team, club sports team, etc.) that regularly require jumping and landing tasks such as basketball, volleyball, soccer or have played in an organized sport or taken dance lessons for at least one year;

Participants were separated into CAI and CON groups. The Cumberland Ankle Instability Tool (CAIT) and Identification of Functional Ankle Instability (IdFAI) was used to estimate the severity of functional ankle instability. CON participants also were pair-matched with CAI participants as described below. Additional inclusionary and exclusionary criteria for CAI and CON groups were as follows, using previously published suggestions (Gribble et al., 2013). All ankle sprains and 'giving way' were assessed by participants themselves based on Gribble's suggestion.

CAI inclusionary criteria:

1. A history of at least one significant ankle sprain.

2. A history of the previously injured ankle joint 'giving way', and/or being sprained repeatedly, and/or producing 'feelings of instability'.

3. *Cumberland Ankle Instability Tool* (CAIT): score ≤ 24.

4. Identification of Functional Ankle Instability (IdFAI): score ≥ 11.

CON inclusionary criteria:

1. Pair-matched to a CAI participant on age (± 3 yr), height (± 2.5 cm), body mass (± 4.5 kg) and physical activity level (± 2 hr/wk in moderate and vigorous activities using our 'Physical Activity Questionnaire'), and has experience in at least one of the same landing-related sports as self-reported by the corresponding CAI participant.

2. Cumberland Ankle Instability Tool: score ≥ 28.

Exclusionary criteria for both groups:

1. A history of previous surgeries to the musculoskeletal structures in either leg.

2. A history of a fracture in either leg.

3. Acute injury to the joints of either leg or the spine or trunk in the previous six months.

    a. For the CON group only, cannot have any history of ankle sprain to either limb within 6 months, nor any history of having had more than one ankle sprain to either limb.

    b. For the CAI group, cannot have any significant ankle sprain within three months.

4. Reports experiencing any unusual symptoms, such as dizziness, nausea, problems with balance, discomfort or pain.

3.3 Instrumentation

A 7-camera Vicon MX™ motion-capture system (120 Hz; Vicon Motion System Ltd., UK) was used to record the spatial locations of retro-reflective markers on the participant's body. The marker set was based on a modified Helen Hayes model of the Vicon Plug-In-Gait model (Vicon®, 2002) as well as some additional markers we use to ensure capturing a sufficient number of marker locations for the segments. Twenty-nine 14 mm-diameter markers were be placed on the skin, clothing and shoes of both lower extremities and the trunk. Markers were attached on the spinal process of the 7th cervical vertebrae and 10th thoracic vertebrae, jugular notch of the manubrium, xiphoid process of the sternum, and center of the right scapula. On both sides of the body, markers were placed on the anterior superior iliac spine and posterior superior iliac spine, peak of iliac crest, lateral surface of the thigh, lateral epicondyle of the femur, lateral surface of the shank, center of lateral malleolus of the ankle, second metatarsal head of the foot and calcaneus.

Ground reaction force (GRF) signals were obtained using two force plates (Bertec 4060-NC; sample rate = 1200 Hz). Each force plate was securely mounted on the top of a customized mounting structure (see Figure 3.1). The test foot landed upon the force platform that was tilted downwards 25° in the lateral direction; the other force platform surface was flat for the other foot to land upon. The centers of the two force plates were at the same height. In the frontal plane, the landing target zone for each force plate was 13 cm wide, bisecting the anterior-posterior axis of the force plate; the medial border of the target zone was 7 cm away from the medial edge of the force platform. This landing target ensured that the participant landed safely upon the force platforms while also allowing the feet to land at natural medio-lateral locations (as determined from our pilot testing). The person stood on a 'start' box whose height was 30 cm higher than the center of the force platforms and was located directly posterior to the nearest edges of the force platforms. A 30 cm drop height is a commonly used height for drop-landing studies (Earl, Monteiro, & Snyder, 2007; Gutierrez et al., 2012; Hewett, Myer, & Ford, 2004) and is a common landing height in sports activities such as soccer and basketball.

A wireless EMG system (sampling rate = 2040 Hz, CMRR > 80 dB; Delsys Trigno™ System, USA) was used to measure the lower leg and thigh muscle activations. Surface EMG electrodes (37mm x 27 mm rectangular shape with four silver bar contacts) were attached on the muscle belly of anterior tibialis (ANT-TIB), gastrocnemius lateralis (GAS-LAT), peroneus longus (PER-LON), rectus femoris (REC-FEM), vastus lateralis

(VAS-LAT), biceps femoris (BIC-FEM) and tensor fascia latae (TNS-FAS) (Cram, Kasman, & Holtz, 1998) on the test limb.

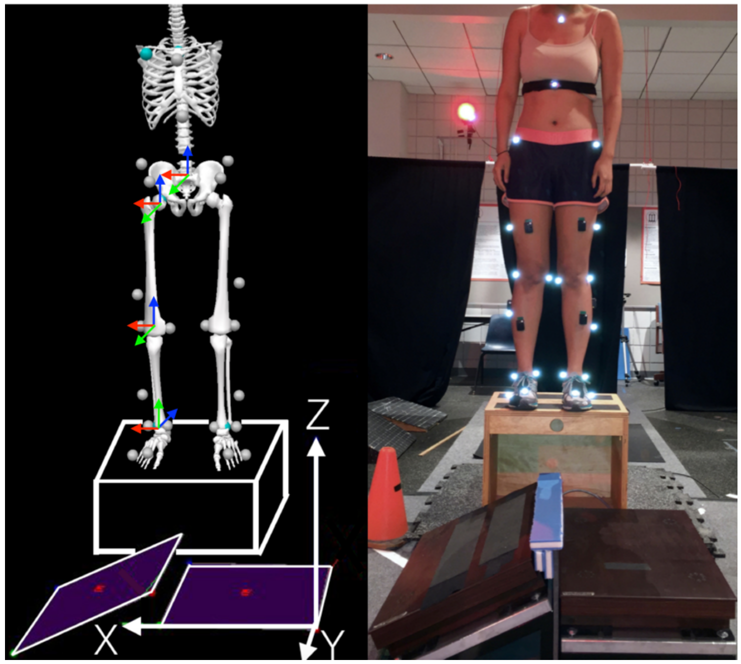

Figure 3.1. Experimental setup and marker placement. The orientation of segmental coordinate system of the pelvis, thigh, shank and foot was shown on the left (red = x-axis, green = y-axis, blue = z-axis).

3.4 Data collection

3.4.1 Eligibility screening

The initial screening of the potential participant's eligibility was conducted via phone including explanation of study, initial screening on participant's health and medical conditions, physical activity level. The data collection date was set up if participant appears to be eligible.

3.4.2 Preparation

Upon arrival to the Biomechanics Laboratory, informed consent was obtained, then the participant completed the 'Pre-Participation and Health Status Questionnaire' and 'Physical Activity Questionnaire' as part of the final eligibility screening and to obtain more detailed information about physical activity history (Appendix A & B, respectively). The CAIT and IdFAI questionnaires were administered to assess the participant's ankle stability of both ankles. If the participant's severity of ankle instability also met the ankle instability eligibility criteria, the participant continued, undergoing the pre-test tasks next. Anthropometrics, including height, body mass, leg length, joint width, and other quantities were obtained (Hanavan, 1964) to later estimate the participant's joint centers and inertial parameters needed to calculate kinematic and kinetic quantities. Reflective markers and surface EMG electrodes were placed on participant's body on the locations described earlier. Before the electrodes were attached, each electrode site location was prepared by shaving any hair, then wiping the skin area with isopropyl alcohol to remove oils and surface residuals. The location of electrodes were based on published guidelines (Cram et al., 1998).

3.4.3 Maximum voluntary isometric contraction (MVIC)

In order to normalize the EMG data to compare across participants, 'maximum' voluntary isometric contraction (MVIC) tests were conducted. For all MVIC tests, the participant was placed in the test position and a hand-held dynamometer (FCE Series Medical Dynamometer, AMETEK, Inc., PA, USA) was used to create isometric resistance and obtain the resistance force. A five-second EMG signal was captured for each muscle MVIC test (Dai, Sorensen, et al., 2012); and a 30-second break was taken between the tests. For TIB-ANT, the participant was in a supine position with the ankle joint relaxed, and knee and hip joints in the neutral (straight) position (Hsu, Krishnamoorthy, & Scholz, 2006); and the resistance was applied on the instep of the foot, 10 cm distal to the ankle joint center. For GAS-LAT, participant was in a prone position with neutral joint positions (Hsu et al., 2006); and the resistance was applied on the ball of the foot, 10 cm distal to the ankle joint center. For PER-LON, participant was sitting on a table with the lower leg hanging from the table, and knee and hip joints at 90° flexion (Escamilla et al., 2006); and the resistance was applied on the lateral side of the heel, 3 cm below the ankle joint center. For REC-FEM and VAS-LAT, participant was in the same position as for PER-LON test (Escamilla et al., 2006, 2010); the resistance was applied on the front of the tibia, 25 cm distal from the knee joint center. For BIC-FEM, participant was in a prone position with the knee joint flexed at 90° (Burnett et al., 2012); the resistance was applied on the posterior side of the lower leg, 25 cm distal from the knee joint center.

Landing Task Protocol

The participant did a five-minute warm-up of walking and running at self-selected speeds on a treadmill to be physically prepared for the landing tasks. A subject's static calibration was captured before the landing test in order to define anatomical and segmental local coordinate system. Participant stood and held for three seconds in a natural position with arms pointing out.

The participant then performed three practice landings on the inverted testing surface to become familiar with the task. The test limb was the limb with the lower CAIT scores for a given CAI participant. In case of equal CAIT scores on both limbs, the dominant limb was the test limb. The dominant limb was defined as the limb a participant prefers to use kicking a soccer ball (Palmieri-Smith, Hopkins, & Brown, 2009; Yeow, Lee, & Goh, 2010). For the matching CON participant, the test limb was the limb that was pair-matched with the limb dominancy of the CAI participant's test limb.

For a given practice or test trial, the participant stood on the start box with the toes aligning with the front edge of the box and arms at sides (Figure 3.1). Next, the take-off phase consisted of the performer stepping forward with the test leg, then bringing the non-test foot forward approximately to the same anterior location as the test foot, without flexing the knee to prevent a 'step down'. The person then landed with one foot onto each force platform within the target zone, with the test limb landing upon the inverted surface. Upon completion of the landing, the person was instructed to slowly stand up and remain in an upright standing position for approximately two seconds. The arms remained in their natural functional position during the entire movement to prevent arm motion to confound the lower extremity outcomes. A spotter was ready to assist the participant as needed.

Fifteen acceptable drop-landing trials were performed. The participant was not allowed to jump up or out, hop or step down from the box (Kulas, Schmitz, Schultz, Watson, & Perrin, 2006), lean the body to one side and the tested foot landed firstly with toes pointing straight ahead. The acceptable trials were initially determined based on visual observations by the investigators, and self-reporting by the participant who was asked to inform the researchers when the movement was not performed as required or realized that the landing was unstable or awkward. Approximately 20-second rests were given between trials to reduce fatigue effect.

3.4.6 Ankle Range of Motion Test

In order to test the ankle function of participants and partially explain the ankle and knee biomechanics, active ankle range of motion in dorsiflexion/plantarflexion and inversion/eversion direction was tested. Each direction was measured three times and averaged values were used. All range of motion tests were measured by the same investigator to ensure consistency of measurement. For ankle dorsiflexion/plantarflexion range of motion, the participant was sitting on a clinical bed with the lower leg hanging from the table, and knee and hip joints at 90° flexion. The ankle joint remained relax and dorsiflexed and plantarflexed as much as possible during the test. For ankle inversion/eversion range of motion, the participant was instructed to be in a prone position and extend 10 cm over the end of the clinical bed (Menadue, Raymond, Kilbreath, Refshauge, & Adams, 2006). The ankle joint remained in neutral and inverted and everted as much as possible during the test. A goniometer was used to measure the inversion range of motion by aligning the center with the midpoint between the malleoli on the posterior part of the ankle; two arms aligning with the midline of the lower leg and midline of the calcaneus, respectively (Menadue et al., 2006).

3.5 Data reduction

Ten trials for each participant were selected for analysis. The trials to be processed should meet the following criteria: a) test foot landed before non-test foot, with the time difference between the initial contact of the two limbs less than 50 ms; and b) the maximum vertical GRF should be greater for the test limb versus the non-test limb. The phases of interest for analysis were pre-landing and landing. The pre-landing phase was defined as the interval of time from 50 ms prior to initial contact (IC) until contact. The 50 ms before IC was commonly used to represent the pre-landing phase (Gribble et al., 2013; Kipp & Palmieri-Smith, 2013) and was appropriate for the present study based on the pilot study. The landing phase began from the IC and ended at the first instant when the center of mass (COM) reached its lowest height (Kulas et al., 2006; Zhang, Bates, & Dufek, 2000). Initial contact was defined as the instant in time when the vertical GRF magnitude first reaches a value of 10 N or greater (Terada et al., 2013). The COM was grossly estimated, as the arms were not included in the calculations, using Dempster's anthropometric data (Dempster, 1955). This estimate should be sufficient for the sole purpose of defining the end of the landing phase.

For the GRF data, a 4th-order Butterworth low-pass filter (cutoff frequency = 15 Hz) was used to remove noise from the raw GRF data. The filtered GRF data were used for inverse dynamic calculations.

<u>Sub-study #1:</u> For kinematic data, a 4th-order Butterworth low-pass filter (15 Hz) was used to remove noise from the raw marker coordinates. We were using Butterworth filtering because it is commonly used in landing studies (Brown et al., 2011; De Ridder et al., 2014; Earl et al., 2007) and has sharp 'roll off' so that we can keep more of the signal content without allowing much high-frequency noise passing through in the frequency band near the cutoff frequency. The cutoff frequency was determined using the residual analysis (Winter, 2009). Then the segmental coordinate system for each segment was constructed based on the filtered marker coordinates and calculated joint center coordinates (Figure 3.1). The joint centers were calculated using the 'chord' function (Vicon®, 2002). Joint angles were calculated using Cardan-Euler angles in a rotation order of X (flexion-extension), Y (adduction-abduction), Z (internal-external rotation) for knee joint; and X (dorsiflexion-plantarflexion), Y (adduction-abduction), Z (inversion-

eversion) for ankle joint. Joint angles in all three directions for each trial were determined, including the knee and ankle angles in pre-landing and at initial landing, and maximum knee and ankle angles, and maximum angular displacement during the landing phase.

Net joint moments were calculated using an inverse dynamics method (Ramakrishnan, Kadaba, & Wootten, 1987). The segment masses were based on Dempster's data (Dempster, 1955); centers of mass and radii of gyration for each segment of the lower extremity of the landing leg was calculated using Hanavan's data (Hanavan, 1964). Maximum knee extensor and adductor moments; and ankle plantarflexor and evertor moments during the landing phase was determined for each trial. Negative work of the knee joint was calculated to assess the knee function in energy absorption in landing. In order to calculate negative work, the joint power curve was calculated and negative part of the curve was integrated over time.

<u>Sub-study #2:</u> For each muscle of the test limb, the DC offset of the raw EMG signals for the MVIC and each trial was removed by detrending (i.e., by removing the mean value of the EMG signal of the entire trial from the raw signal). Then the signal was filtered using a Butterworth band-pass filter (20 - 450 Hz). Then the filtered data were rectified and the linear envelope generated (low-pass filter of cutoff frequency = 10 Hz) (Chappell, Creighton, Giuliani, Yu, & Garrett, 2007; Dai, Sorensen, et al., 2012). The average of the maximum 1-second MVIC linear envelope was used as the magnitude of MVIC. The linear envelope data was averaged for the 50 ms before initial landing (Nagano, Ida, Akai, & Fukubayashi, 2007) and for the first 100 ms of the landing phase (Dai,

Heinbaugh, Ning, & Zhu, 2014)) and normalized to the corresponding muscle's magnitude of MVIC to represent the magnitude of muscle activation during these phases. The first 100 ms after IC was chosen because the ACL injuries usually occur within that period based on injury case analyses (Koga et al., 2010; Krosshaug et al., 2007).

Co-contraction ratio (CCR) of REC-FEM to BIC-FEM (Q/H = quadriceps/hamstring) activation was calculated (Equation 1) because it could influence the ACL loading (Li et al., 1999). Co-contraction index (CCI) in the sagittal plane between the TIB-ANT and GAS-LAT and frontal plane between TIB-ANT and PER-LON was calculated as shown in Equation 2 (Lin et al., 2011; Suda, Amorim, & de Camargo Neves Sacco, 2009). CCR was calculated only for the knee joint muscles, whereas CCI was calculated only for the ankle joint muscles.

$$CCR = \frac{EMG_{REC-FEM}}{EMG_{BIC-FEM}} \qquad \text{(Equation 1)}$$

$$CCI = \frac{EMG_{Antagonist}}{(EMG_{Antagonist} + EMG_{Agonist})/2} \qquad \text{(Equation 2)}$$

in which $EMG_{Antagonist}$ and $EMG_{Agonist}$ represents the average linear envelope values for a given phase for the antagonist and agonist muscles, respectively. The CCI of the ankle joint muscles was assessed because it relates to joint stability (Lin et al., 2011). Time to peak EMG linear envelope was calculated in order to better understand the strategy pattern of muscle activation (Suda, Cantuaria & Sacco, 2008).

3.6 Statistical analysis

Sub-study 1: Participants' leg length, ankle plantar/dorsiflexion range of motion and ankle instability scores of the test limb were compared between groups using paired *t*-tests ($p < 0.05$). Joint angles at IC and peak joint angles, joint angular displacements in landing, peak net joint moments and negative work was averaged respectively across the ten trials for each participant for statistical tests. All group comparisons of kinematic and kinetic variables also using paired *t*-tests ($p < 0.05$). Possible tendency of significance was defined ($0.05 \leq p < 0.1$). Outliers were detected using Chauvenet's criterion. Statistical analyses were processed in SPSS 22 (IBM Corporation, USA).

Sub-study 2: Participants' joint strength and ankle instability scores were compared between groups using paired *t*-tests ($p < 0.05$). For each muscle and relevant phase(s), the ten trial average of a given EMG variable for each participant was used for statistical tests. To detect potential outliers, Chauvenet's criterion was applied. Group comparisons of EMG variables were run using paired *t*-tests ($p < 0.05$). Possible tendency toward significance was defined as $0.05 \leq p < 0.10$. Effect size (Cohen's *d*) was reported. Statistical analyses were processed in SPSS™ v.22 (IBM Corporation, USA).

CHAPTER 4

Manuscript #1: Does chronic ankle instability influence landing knee biomechanics of females?

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[1] Li, Y., Ko, J., Brown, C. N., Schmidt, J. D. Kim, S.-H. and Simpson, K. J. To be submitted to *Journal of Applied Biomechanics*

ABSTRACT

Although much is known about the mechanisms of anterior cruciate ligament (ACL) and other knee-related injuries, it is still unclear how chronic ankle instability (CAI) relates to these injury mechanisms. The purpose of the study, therefore, was to determine if individuals with CAI exhibit atypical knee biomechanics during landing onto a tilted surface. A seven-camera motion analysis system, two force plates were used to collect lower extremity biomechanics of 21 CAI and 21 pair-matched control (CON) participants who performed 10 landings onto a sideward-tilted and flat platform on the CAI/matched and non-test limbs, respectively. Lower extremity joint angles, joint angular displacements, joint moments and eccentric work was calculated. Paired t-tests were used to compare between-group differences ($p < 0.05$). We observed that CAI displayed a significantly increased ankle inversion angle at initial contact; lower ankle inversion moment and eccentric work in the frontal plane during landing that could be related to their increased peak knee joint extension moment and internal rotation moment. However, these atypical knee joint biomechanics of CAI individuals have been shown to be related to increased ACL loading. We conclude that these alterations of ankle biomechanics could lead to alterations in knee biomechanics that are related to increased ACL strain. Future studies may need to measure or estimate the ACL loading to confirm whether CAI relates to the mechanism of ACL injury.

KEY WORDS: lower extremity, inverted landing surface, kinetics, kinematics, drop landings

4.1 Introduction

Knee joint injury is very common in sports and physical activities; about 40% of all sport injuries are related to the knee joint (Majewski et al., 2006). One knee injury of particular concern in sport is damage to the anterior cruciate ligament (ACL). ACL injury accounts for 20% of knee injuries and 2.6% to 8% of all sports injuries (Hootman et al., 2007b; Majewski et al., 2006). In addition, female athletes have a significant higher rate of ACL injury compared to male athletes (Agel et al., 2005; Beynnon et al., 2014; Hootman et al., 2007b). Purported reasons for the higher female rate include different anatomical structure of the knee (Nunley et al., 2003; Shultz, 2015), less quadriceps and hamstrings strength (Huston & Wojtys, 1996) and hormonal effects that could reduce load to failure (Slauterbeck et al., 1999).

Although much is known about mechanisms of knee and ACL injury, one factor that is only now beginning to receive attention is the influence of previous ankle injury on ACL injury mechanisms. Two groups of researchers suggested that having had a prior ankle injury may be related to ACL injury causation. Soderman et al. (2001) studied 146 female soccer players and observed that three of the five ACL-injured players also had ankle sprains (Söderman et al., 2001). Moreover, a significant association between lateral ankle sprain history and ACL injury history was observed by Kramer, Denegar, Buckley, and Hertel, (2007). They determined, that for young adult female athletes, among the 66 limbs that had a history of ACL injury, 52% also had a prior ankle sprain, whereas only 32% of non-injured knees had a history of prior ankle history.

However, whether there is a causal relationship between prior ankle injury and subsequent knee injury was not proven in those studies. Moreover, if the mechanisms by which a prior ankle injury could be related to knee injury are not well understood. One possible reason is that the definition of ‘previous ankle sprain’ is too broad and unspecific. ‘Previous ankle sprain’ could include a wide range of conditions that may not influence ankle injury (e.g., one prior ankle sprain without further sequelae) that could have a significant impact on abnormal lower extremity mechanics.

We suggest that chronic ankle instability (CAI) is one specific condition that likely could affect the potential for acute or chronic knee injuries (Terada et al., 2013, 2014). CAI commonly develops following ankle sprains (Hertel, 2002) and has been described “as an encompassing term used to classify a subject with both mechanical and functional instability of the ankle joint.” (Gribble et al., 2013, p586). Individuals with CAI exhibit reduced ankle muscle co-contraction and ankle muscle strength (Willems et al., 2002) that decreases ankle joint stability (Lin et al., 2011), limited ankle joint range of motion (Denegar et al., 2002; Hoch et al., 2012), and changes to the mechanical properties of ankle joint tissues (e.g., ligament laxity). These properties, therefore, could influence the ankle biomechanics and thereby affect knee biomechanics (Gribble and Robinson, 2009a, 2009b) and hip biomechanics (Brown, Padua, Marshall, & Guskiewicz, 2011; Gribble, Hertel, Denegar, & Buckley, 2004).

However, to our knowledge, only one study exists attempting to investigate how CAI could relate to mechanism of knee injury. Terada, Pietrosimone, and Gribble (2014) investigated the influence of CAI on knee anterior shear force (estimated using the anterior component of joint reaction force, which cannot fully represent the joint loading) and lower extremity kinematics during a vertical stop jump landing. Compared to the corresponding control groups, the ten male and nine female CAI participants displayed a decreased peak knee flexion angle during the landing phase, but no other differences. They believed that the nonsignificant tendency of reduced ankle dorsiflexion could have been linked to the decreased knee flexion angle in CAI participants (Terada et al., 2014). Consequently, the reduced knee flexion could result in a greater tibial anterior shear force by increasing the patella tendon tibia shaft angle (see Figure 1 in Nunley, Wright, Renner, Yu, & Garrett, 2003; Yu & Garrett, 2007).

Terada et al. (2014), however, did not observe significant difference in knee kinetics. Landing biomechanics vary between males and females (Lephart et al., 2002; Nagano et al., 2007), thus, differences between CAI and control individuals may not have been detectable. The unique landing biomechanics of females (e.g., reduced knee flexion angle, increased knee abduction and internal rotation angle, etc.) may result in unique influences of CAI on knee mechanics.

To date, it is still unknown how CAI relates to mechanisms of ACL or other knee injuries. Understanding the influences of CAI on knee biomechanics could provide important information about knee injury prevention and help developing safe movement strategies particularly for individuals with CAI. Therefore, the purpose of the study was to determine if atypical knee biomechanics that potentially contribute to ACL injury are exhibited during the landing. Based on the deficits of CAI dorsiflexion range of motion (described earlier), we hypothesize that CAI compared to CON could exhibit reduced ankle dorsiflexion displacement. In order to maintain upright posture, CAI could also exhibit reduced knee and hip flexion displacement, which leads to greater knee and hip extension moment to absorb the energy. We further hypothesize that CAI could display increased ankle inversion before and at landing due to deficits of proprioception and reduced peroneal activities observed in previous studies (Caulfield, Crammond, O'Sullivan, Reynolds, & Ward, 2004; Sekir, Yildiz, Hazneci, Ors, & Aydin, 2007). Reduced peak ankle eversion moment and eccentric work was expected. CAI was also expected to exhibit a greater knee adduction angle with greater adduction moment to control the lower leg orientation due to increased ankle inversion angle.

4.2 Methods

4.2.1 Participants

Based on a priori power analysis (power = .8, Cohen's *d* = 0.8; GPower®, v.3.1.9), using data from previous studies (Gutierrez et al., 2012; Liu, 2013), ten to nineteen participants in each of the two ankle stability groups (CAI and CON) were needed. Twenty-one female participants with CAI were recruited to improve the statistical power. The inclusionary and exclusionary criteria for CAI participants came from previously published suggestions (Gribble et al., 2013). Twenty-one healthy control participants were recruited to pair-match with the CAI participants on gender, height, body mass and physical activity level. All participants are physical active and have experience in jumping related sports (e.g., basketball, volleyball, soccer, etc.). The participant characteristics are presented in Table 4.1.

Table 4.1. Demographical Data (Mean ± SD) of the Participants

Variable	CON	CAI
Sample size	21	21
Body mass (kg)	64.4 ± 11.9	64.4 ± 12.4
Height (cm)	165 ± 6	164 ± 6
Age (yr)	21 ± 2	21 ± 2

Note: CON = healthy control participants; CAI = chronic ankle instability participants.

4.2.2 Instrumentation and Experimental Setup

A seven-camera Vicon system (MX40, Vicon Motion Systems Ltd., Oxford, UK) was used to capture spatial locations of the reflective markers placed on trunk and lower body at 120 Hz. Ground reaction forces (GRF) were collected using two Bertec® force plates (4060-NC, Bertec Corporation, Ohio, USA) at 2040 Hz. The two force plates were located side-by-side with one tilted downward 25° (inverted) in the lateral direction and the other one flat such that the centers of the two force plates were at the same height (Figure 4.1). One inverted surface was chosen because landing onto an inverted surface has been suggested as a more demanding and realistic simulation of a situation in which one foot lands onto an uneven surface (Chen et al., 2012). Marker coordinates and ground reaction forces data were captured and synchronized using Vicon Nexus® 2.2 software (Vicon Motion Systems Ltd., Oxford, UK).

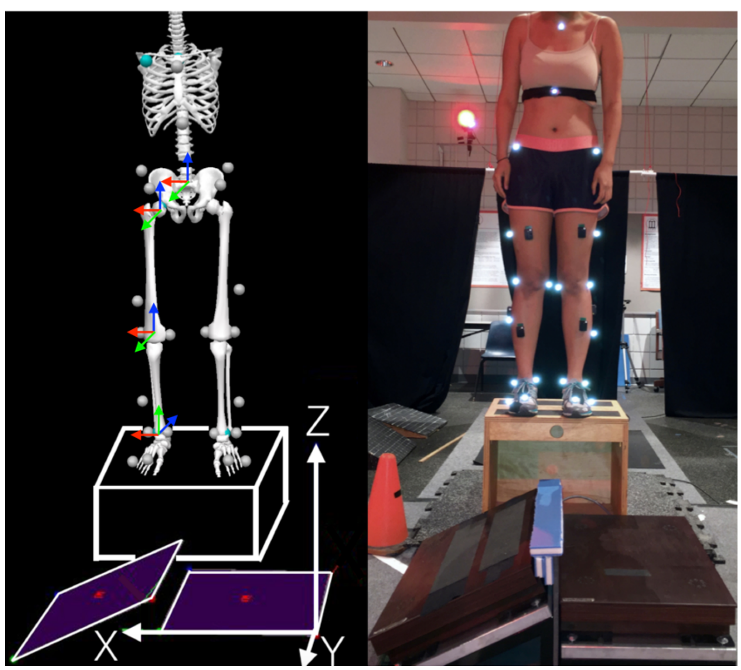

Figure 4.1. Experimental set up and marker placement. The orientation of segmental coordinate system of the pelvis, thigh, shank and foot was shown on the left (red = x-axis, green = y-axis, blue = z-axis).

4.2.3 Procedures

The study was approved by our institutional review board, and all participants provided informed consent before data collection. Participants completed the health status and physical activity questionnaires, IdFAI and CAIT. The limb with less ankle stability was chosen as the test limb for CAI. For CON, the test limb was the limb that is pair-matched with the limb dominancy of the CAI individual's test limb. The dominant limb was defined as the limb a participant prefers to use kicking a soccer ball (Palmieri-Smith, McLean, Ashton-Miller, & Wojtys, 2009; Yeow et al., 2010). Anthropometrics (height, mass, leg length) were obtained to estimate the inertial parameters (Hanavan, 1964) needed to calculate kinetic quantities. Twenty-nine retroreflective markers (14mm diameter, Figure 4.1) were placed on the trunk (spinal process of C7 and T10, xiphoid process of the sternum, jugular notch of the manubrium, and center of the right scapula) and both sides of the lower extremity (ASIS, PSIS, peak of the iliac crest, lateral surface of the thigh, lateral epicondyle of the femur, lateral surface of the shank, center of lateral malleolus of the ankle, second metatarsal head of the foot and posterior calcaneus).

A five-minute jogging warm-up was performed, then the drop landing testing. For a given landing trial, the participant stood on a box 30 cm above the landing zones, then stepped forward with the test limb then the other limb and landed with test foot on the tilted force plate and the other foot on the flat force plate. To ensure consistent performance, the participant was not allowed to jump, hop or step down from the box

(Kulas et al., 2006). Upon completion of the landing, the participant was instructed to slowly stand up and remain in an upright standing position for approximately two seconds. After the landing test, active range of motion of dorsflexion/plantarflexion and inversion/eversion was assessed for both ankles using a manual goniometer following the previous protocol (Menadue et al., 2006) by the same investigator. For each participant, ten acceptable trials were collected for analysis.

4.2.4 Data Analysis

The phases of interest for analysis included the pre-landing and landing phase. The pre-landing phase was from 50 ms prior to contact (-50 ms) to the instant of initial contact (‘IC’, vertical GRF > 10 N). The landing phase was from the IC to the first instant when the center of mass reaches its lowest height (Kulas et al., 2006). Three-dimensional coordinates of the markers were reconstructed (Vicon Nexus 2.2) and filtered with a 4th order low-pass Butterworth filter at 15 Hz. Lower extremity segmental coordinate systems and joint centers were defined and joint angles were calculated (Visual 3D, C-Motion, Inc.) using a Cardan (X-Y-Z) sequence (Figure 4.1). The hip joint center was defined based on Bell et al. (1990) study. The knee joint center was defined as the midpoint between the lateral and medial femoral epicondyles. The ankle joint center was defined as the midpoint between the lateral and medial apex of malleoli. The foot landing angle at IC and full contact (instant of the lowest heel marker position) was analyzed to understand the orientation of the foot relative to the landing surface during the landing phase. The foot landing angle was calculated as the projected angle onto the horizontal plane and formed between the longitudinal axis of the foot and anterior-posterior axis of the global coordinate system.

GRF data were filtered the same as the markers to minimize the impact artifact at the knee moment (Bisseling & Hof, 2006; Kristianslund, Krosshaug, & Van den Bogert, 2012). GRF data were down-sampled at 120 Hz to synchronize with coordinates data. Net joint moments at the lower extremity were calculated using an inverse dynamic method (Ramakrishnan et al., 1987) and normalized to participants' body mass. The segment masses were based on Dempster's data (Dempster, 1955); center of mass and radii of gyration were using Hanavan's data (Hanavan, 1964). Negative work of the joints was calculated based on integrating the negative part of the power curve and also normalized to body mass.

4.2.5 Statistical Analysis

Participants' leg length, ankle plantar/dorsiflexion range of motion and ankle instability scores of the test limb were compared between groups using paired *t*-tests ($p < 0.05$). Joint angles at IC and peak joint angles, joint angular displacements in landing, peak net joint moments and negative work was averaged respectively across the ten trials for each participant for statistical tests. All group comparisons of kinematic and kinetic variables also using paired *t*-tests ($p < 0.05$). Possible tendency of significance was defined ($0.05 \leq p < 0.1$). Four outliers were detected among all the tested biomechanical variables by Chauvenet's criterion. Statistical analyses were processed in SPSS 22 (IBM Corporation, USA).

4.3 Results

Participants' leg length, ankle ranges of motion and ankle instability (CAIT and IdFAI scores) are presented in Table 4.2. No significant differences were observed for leg length and inversion and eversion range of motion between groups. CAI group displayed less ankle instability (less CAIT and greater IdFAI scores) and ~4° dorsiflexion and ~8° less plantarflexion range of motion.

Table 4.2. Leg Length, Ankle Instability and Range of Motion (ROM)

Variable	CON	CAI	*t*-value	*p*-value	Effect size
Leg length (cm)	86.5 ± 4.7	85.3 ± 4.2	1.31	0.205	0.27
CAIT score	29.5 ± 0.9	19.3 ± 6.0	7.37	**0.000***	2.38
IdFAI score	1.3 ± 2.1	22.2 ± 9.2	9.53	**0.000***	3.13
ROM dorsiflexion (°)	20 ± 7	16 ± 7	2.36	**0.028***	0.57
ROM plantarflexion (°)	36 ± 9	28 ± 8	2.60	**0.017***	0.94
ROM inversion (°)	34 ± 6	33 ± 5	0.25	0.806	0.18
ROM eversion (°)	20 ± 4	19 ± 5	0.38	0.705	0.22

Ensemble curves for the knee, ankle and hip joints are shown in Figures 4.2 – 4.4, respectively. For the knee joint (Table 4.3), the CAI group exhibited significantly greater values for flexion displacement (difference: ~9°), peak extension moment (~0.3 Nm/kg), peak internal rotation moment (~0.12 Nm/kg) and (~0.25 J/kg) sagittal plane negative work compared to CON during the landing phase. No other significant differences were found (p = 0.510 – 0.544).

For the ankle joint (Table 4.3), CAI group exhibited an ~5° greater inversion angle at IC, an ~8° less inversion displacement, an ~0.2 Nm/kg lower peak eversion moment and an ~0.15 J/kg less frontal plane negative work during the landing phase. Other ankle variables were not significantly different between groups (p = 0.187 – 0.876). In addition, no differences were found for the foot landing angles at IC and full contact (p = 0.135 and 0.577, respectively).

For the hip joint, the only significant difference detected was for hip flexion displacement (Table 4.3). The CAI exhibited an ~11° greater hip flexion displacement during the landing. Other variables were not different between groups (p = 0.119 – 0.944).

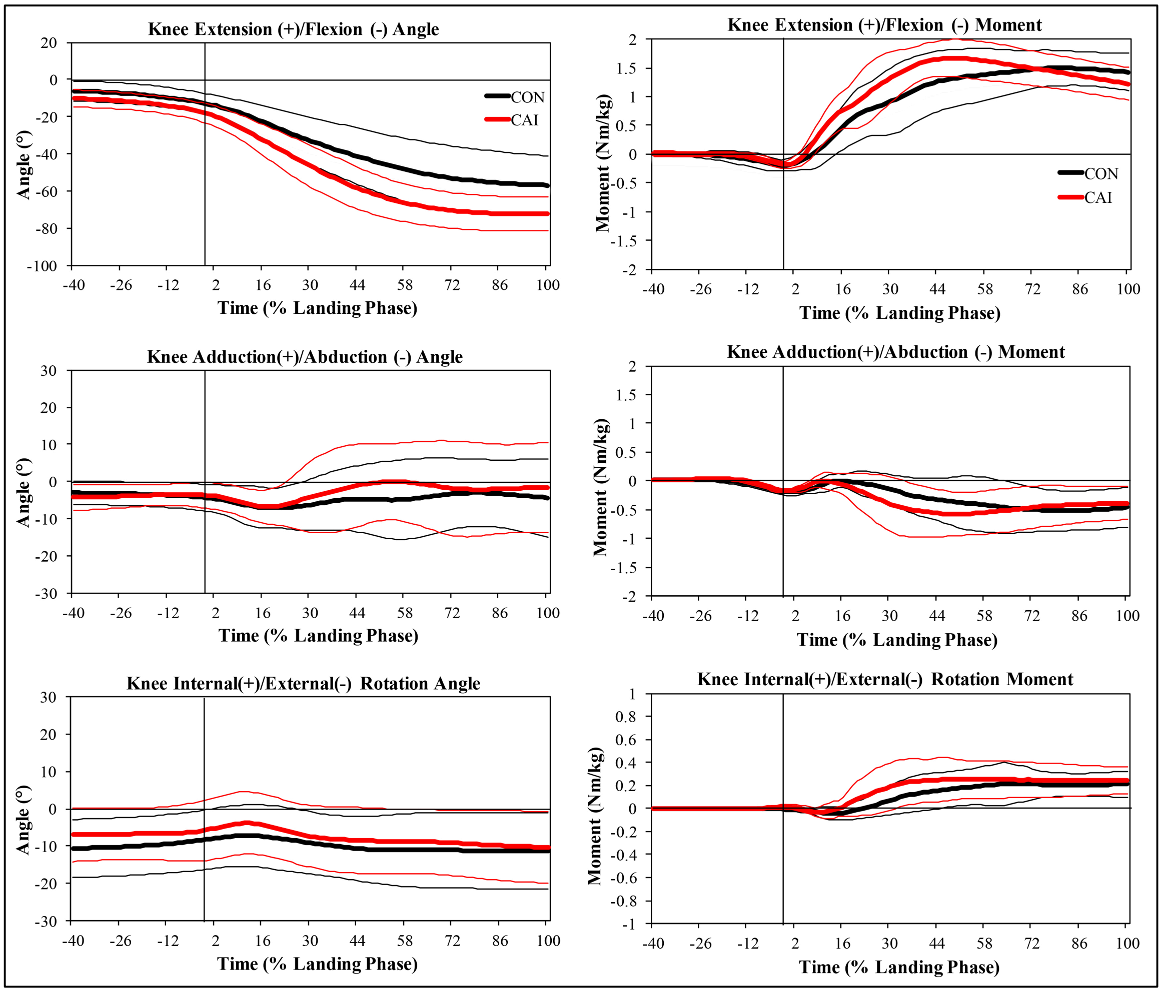

Figure 4.2. Ensemble average curves of the knee joint angles (left column) and knee joint moment (right column). The vertical line indicates the instant of initial contact (0%). A portion of the pre-landing phase to the end of the landing phase (100%) is shown.

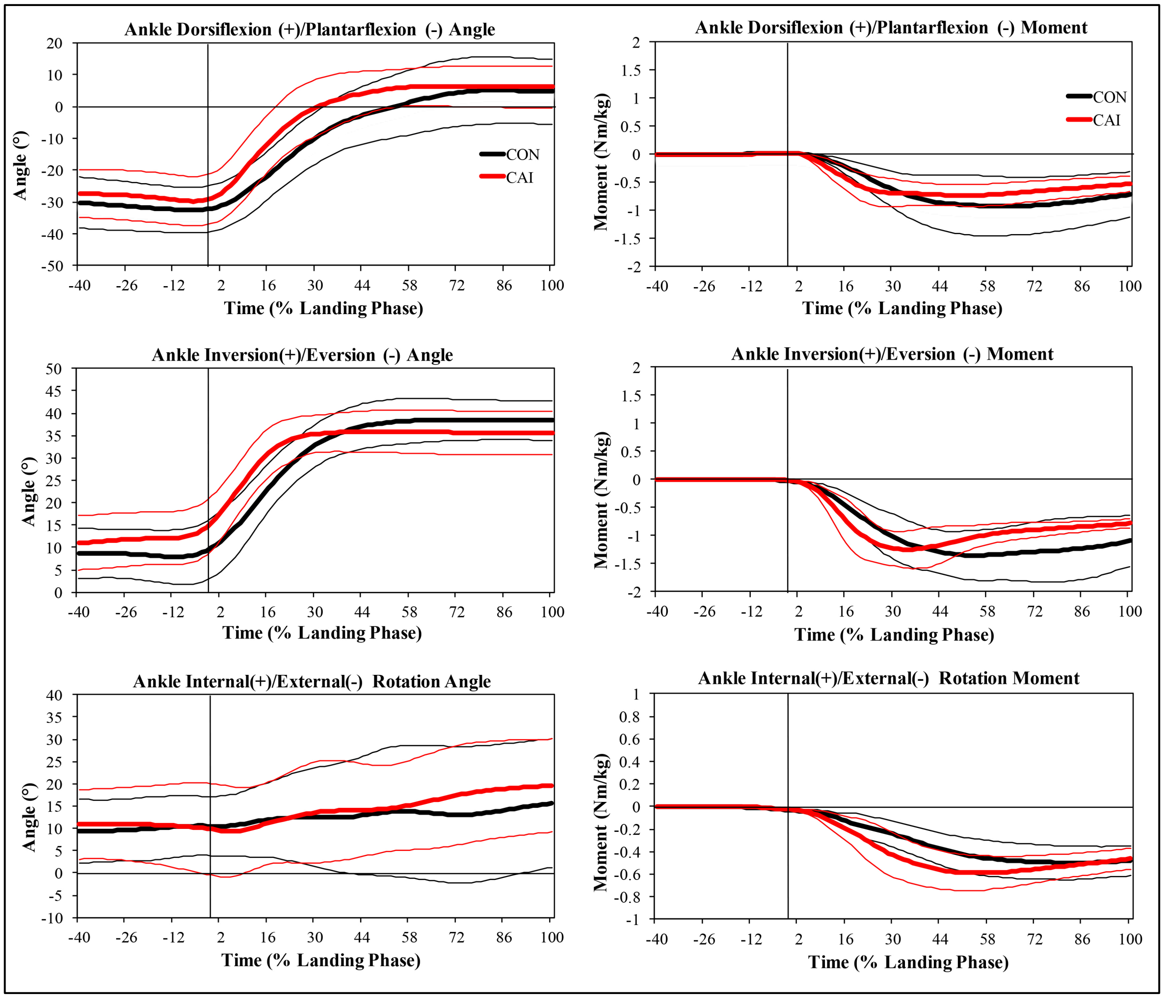

Figure 4.3. Ensemble average curves of the ankle joint angles (left column) and ankle joint moment (right column). The vertical line indicates the instant of initial contact (0%). A portion of the pre-landing phase to the end of the landing phase (100%) is shown.

Table 4.3. Descriptives (mean ± SD) and Between-Group Comparison Outcomes of the Lower Extremity Kinematics and Kinetics.

		Group				
Joint	Variable	CON	CAI	*t*-value	*p*-value	Effect size
Knee	Flexion Displacement (°)	-43.4 ± 14.7	-52.2 ± 7.8	2.10	**0.048***	0.75

	Peak Flexion Angle (°)	-55.8 ± 12.8	-71.5 ± 9.7	3.86	**0.001***	0.75
	Flexion Angle at IC (°)	-16.8 ± 6.3	-21.1 ± 4.6	2.18	**0.042***	0.78
	Peak Abduction Angle (°)	-11.9 ± 9.9	-10.2 ± 5.6	0.62	0.544	0.21
	Peak Extension Moment (Nm/kg)	1.79 ± 0.35	2.06 ± 0.30	2.87	**0.010***	0.85
	Peak Abduction Moment (Nm/kg)	-0.76 ± 0.32	-0.77 ± 0.36	0.67	0.510	0.33
	Peak Internal Rotation Moment (Nm/kg)	0.33 ± 0.17	0.45 ± 0.18	2.77	**0.012***	1.26
	Sagittal Plane Negative work (J/kg)	-0.83 ± 0.42	-1.07 ± 0.26	2.11	**0.048***	0.69
Ankle	Dorsiflexion Displacement (°)	38.7 ± 9.7	38.3 ± 7.8	0.16	0.876	0.05
	Inversion Displacement (°)	30.5 ± 7.1	23.0 ± 6.7	3.86	**0.001***	1.09
	Peak Inversion Angle (°)	40.3 ± 4.7	38.4 ± 3.6	1.46	0.162	0.31
	Inversion Angle at IC (°)	14.7 ± 7.4	19.8 ± 6.7	2.80	**0.011***	0.72
	Peak Plantarflexion Moment (Nm/kg)	-1.08 ± 0.48	-0.93 ± 0.24	1.37	0.187	0.53
	Peak Eversion Moment (Nm/kg)	-1.85 ± 0.41	-1.63 ± 0.26	2.37	**0.028***	0.85
	Sagittal Plane Negative Work (J/kg)	-0.39 ± 0.24	-0.37 ± 0.03	0.40	0.692	0.12
	Frontal Plane Negative Work (J/kg)	-0.34 ± 0.15	-0.20 ± 0.08	3.87	**0.001***	1.16
Foot	Foot landing angle at initial contact	-1.6 ± 6.3	1.0 ± 4.9	1.56	0.135	0.46
	Foot landing angle at full contact	1.8 ± 8.2	2.9 ± 5.9	0.59	0.577	0.15
Hip	Flexion Displacement (°)	18.7 ± 9.8	30.0 ± 11.3	2.89	**0.009***	1.07
	Peak Adduction Angle (°)	-5.3 ± 5.1	-5.7 ± 3.6	0.23	0.823	0.09
	Peak Extension Moment (Nm/kg)	-1.52 ± 0.54	-1.51 ± 0.71	0.07	0.944	0.02
	Peak Abduction Moment (Nm/kg)	-0.92 ± 0.52	-0.81 ± 0.42	0.66	0.520	0.23
	Sagittal Plane Negative Work (J/kg)	-0.17 ± 0.17	-0.29 ± 0.24	1.64	0.119	0.58

4.4 Discussion

We predicted that CAI group would display some differences in lower extremity kinematics and kinetics compared to CON in landing. Specifically, we predicted that CAI group would display a more erect leg posture during landings (e.g., reduce ankle, knee and hip flexion) compared to CON due to the limited ankle range of motion. However, our hypothesis was not supported, suggesting that CAI group may display some different strategies to compensate for altered ankle biomechanics than we expected. As we expected, greater peak knee extension moment was observed for CAI group. Unexpectedly, a greater peak knee internal rotation moment was also observed for CAI group. These greater knee moments may relate to a greater ACL loading. We further hypothesized that CAI would display reduced peak ankle eversion moment and eccentric work and further influence the knee biomechanics in the frontal plane. Though the hypothesis of ankle variables was supported, no differences were found for frontal plane knee variables between groups. For all the statistical between-group differences, the effect size was moderate to large (Cohen's $d = 0.69 – 1.26$).

4.4.1 The knee joint

For the kinematics of the knee joint, our hypothesis that compared to CON, CAI would display less knee flexion displacement was not supported. Although the knee joint was slightly more flexed (~5° more) at IC, CAI group displayed a 10° more knee flexion displacement, thus ending the landing phase in much greater (~16° more) flexed position. One explanation for this outcome may be that the CAI participants employed greater knee flexion displacement to absorb the kinetic energy due to reduced energy dissipation by the ankle joint (described above).

A second potential explanation is that CAI participants may use a "protective strategy" during the landing (Caulfield et al., 2002). Caulfield et al. observed that the greater knee flexion angle (from 20 ms before IC to 60 ms after IC) associated with greater ankle dorsiflexion angle at IC, which may be a 'protective strategy' of CAI group to stabilize the ankle joint. Though they did not provide detailed explanation about how this association occurred, the CAI group could improve the ankle stability by increasing the dorsiflexion. The dorsiflexion motion could be achieved by rotating the lower leg forward and thus increase the knee flexion angle. However, other studies did not observe this more flexed knee or ankle joint (Delahunt et al., 2006; Gribble & Robinson, 2010). More studies may be needed to confirm whether CAI group use this protective strategy during landings.

Our hypothesis of greater peak knee moments generated by the CAI compared to the control group was mostly supported. CAI group displayed 17% and 36% greater peak knee extension and internal rotation moments, respectively. However, no frontal plane moment differences were detected. In the sagittal plane, the greater knee extension moment and flexion displacement resulted in a greater eccentric work at the knee joint for CAI group. This may be a compensatory strategy for CAI to absorb the body's kinetic energy. During the landing, the eccentric work in the sagittal and frontal planes contribute to reducing the body's kinetic energy in the vertical direction. Because the CAI ankle joint did comparable eccentric work in the sagittal plane but less eccentric work in the frontal plane, the knee joint muscles may need to do more work to compensate. The greater knee eccentric work of CAI would benefit dissipation of the knee compressive impact forces that could lead to cartilage lesions and osteoarthritis (Childs, Sparto, Fitzgerald, Bizzini, & Irrgang, 2004).

However, the greater knee extension moment and internal rotation moment has been related to greater ACL strain (Fleming et al., 2001). The greater knee extension moment of CAI is possibly due to greater quadriceps activation that imply greater quadriceps anterior shear force component acting on the tibia and posterior shear force on the distal femur (DeMorat, Weinhold, Blackburn, Chudik, & Garrett, 2004). Greater knee internal rotation moment of CAI could further increase the ACL loading (Berns et al., 1992; Markolf et al., 1995). Moreover, greater quadriceps activation could also relate to the mechanism of patellar tendonitis (Rosen et al., 2015).

Our results are somewhat different from the previous findings by Terada et al. (2013). They observed that CAI group exhibited a significantly lower proportion of the total lower extremity eccentric work was done by knee extensors and a higher proportion of work done by ankle extensors (Terada et al., 2013). The discrepancy may attribute to several factors: different task and landing protocols, calculations (relative percentage of the total eccentric work by the lower extremity versus absolute values of the eccentric work), and a landing phase of analysis (100 ms after IC vs. landing phase definition in our study). Moreover, combined results of males and females in their study may also contribute the discrepancy between their study and the present study.

4.4.2 The ankle joint

The outcomes of the sagittal plane ankle variables did not support our hypothesis of reduced dorsiflexion angle during landing for CAI, as we did not observe group differences for dorsiflexion displacement or peak dorsiflexion angle for the landing phase. Researchers have been attributed the reduced peak dorsiflexion angle of CAI group that occurs during landing to their limited range of motion (Delahunt et al., 2006). Indeed, we assessed the CAI group's ankle range of motion and they exhibited reduced (~ 4°) dorsiflexion active range of motion compared to CON. However, one reason that significant group difference existed for dorsiflexion angle was that the demands of dorsiflexion (peak dorsiflexion angle = ~ 7°) in our landing protocol may not exceed or be even close to CAI group's active range of motion (~16°).

Individuals with CAI did exhibit a greater inverted foot position at landing, as predicted. Descriptively, they also appeared to invert their CAI foot further prior to IC (Figure 4.3), which was also observed in previous studies (Delahunt et al., 2006; Lin et al., 2011). The increased inversion angle of CAI during landing has been attributed to deficits of ankle proprioception (Lin et al., 2011) and reduced peroneus muscle activities (Caulfield, Crammond, O'Sullivan, Reynolds, & Ward, 2004; Delahunt et al., 2006; Li et al., 2016, Study II). This ankle position at IC has also been suggested to increase the risk of ankle roll-over and sprain injury due to the sub-talar instability (Yamamoto et al., 1998). However, due to the inverted landing surface used in this study, our CAI group may be preparing for landing onto the inverted surface by inverting the ankle joint to achieve a more frontal plane 'flat-foot' landing relative to the surface instead of a 'medial-border foot' landing. This strategy may reduce the inversion torque exhibited by GRF on the foot by laterally shifting the point of application of the GRF and reducing the moment arm (Figure 4.4). In addition, having a greater plantar contact surface during early landing phase could also help improving foot and postural stability.

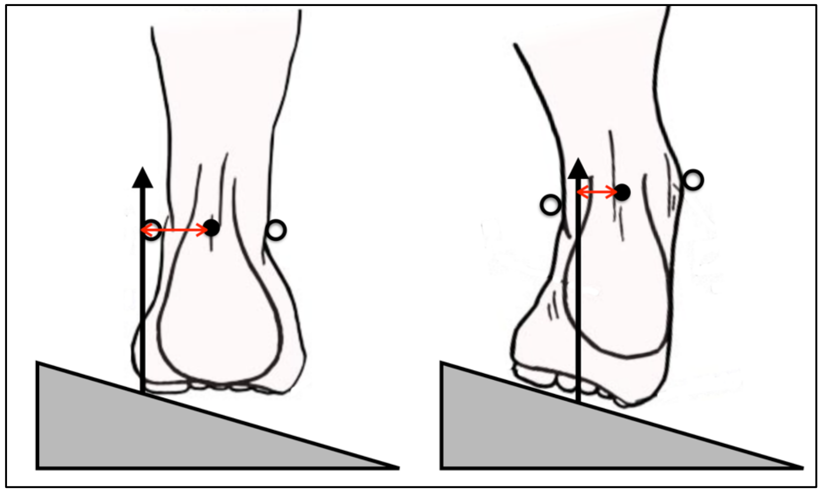

Figure 4.4. Graphical representation of vertical ground reaction force (black arrow) and moment arm (red line) during a 'medial-border foot' landing (left) and a 'flat-foot' landing with ankle inversion (right).

In the landing phase, CAI compared to CON individuals exhibited reduced peak eversion moment, which was expected. However, they displayed less inversion displacement, which indicates that they used some mechanical strategies (possibly greater co-contraction of ankle joint muscles or earlier peak peroneus muscle activation) other than the peak eversion moment to hinder the ankle inversion movement and also prevent excessive ankle inversion. The increased ankle inversion angle at IC and reduced displacement of CAI could lead to reduced ability of energy dissipation by the ankle joint in the frontal plane. Therefore, compensation (increased energy absorption by the knee joint) has been done by CAI group as described above.

4.4.3 The hip joint

For hip joint kinematics, contrary to our hypothesis, CAI compared to CON displayed greater hip flexion displacement. Due to the more flexed knee position and greater knee flexion displacement of CAI during the landing phase, the upper leg could be in a more flexed position (rotated forward) around the hip joint thus resulted in a more hip flexion. The hip joint has been investigated in the previous studies of CAI landing (Brown et al., 2011; Delahunt et al., 2006; Gribble & Robinson, 2010), because it could influence the postural stability and knee biomechanics in landing. However, contradicting results were reported (Brown et al., 2011; Delahunt et al., 2006; Gribble & Robinson, 2010). Therefore, the influence of CAI on hip biomechanics may be complicated, because the influence may be dependent on landing protocols and hip biomechanics displayed large individual variation.

4.4.4 Summary of discussion

These lower extremity kinematic and kinetic results suggest two main interpretations and associated behavioral meanings, overall. Relative to control individuals, CAI participants demonstrated: a) different landing kinematics (e.g., increased ankle inversion at IC) that improve their ankle joint and posture stability when landing on an inverted surface; b) reduced energy dissipation at the ankle joint and thus greater knee extension and internal rotation moments about the knee joint. Moreover, these altered knee kinetics are related to the mechanisms of ACL loading.

4.4.5 Limitations

There are several limitations in the present study. Skin movement artifact may have influenced the lower extremity kinematics and kinetics. However, the use of very lightweight markers reduced skin movement; a least squares error approach to form the tracking coordinate system was used to minimize error (McLean et al., 2007). We only analyzed the data of the test limb. Limb dominance affects landing mechanics of limbs differently (McPherson, Dowling, Tubbs, & Paci, 2016); however, the control's test limb corresponded to the same limb dominance as the pair matched CAI participant. Lastly, differences between CAI and control participants' perception of landing onto tilted surfaces could lead to their landing mechanics differently. For example, as one feature of CAI is the tendency to roll over the ankle, CAI may have perceived the tilted surface as a dangerous situation and thus exhibit some protective strategies to prevent the roll-over.

4.5 Conclusion

In conclusion, individuals with CAI may display some kinematic strategies to improve their stability while landing onto an inverted surface, such as a 'flat-foot' landing pattern. However, the greater ankle inversion at IC, less ankle inversion displacement and less ankle eccentric work of CAI could lead to alterations in knee biomechanics (i.e., greater peak extension and internal rotation moment) that are related to increased ACL strain. In order to minimize the negative effects of altered knee biomechanics, we suggest that individuals with CAI may benefit from absorbing the kinetic energy using the ankle joint muscles. Future studies may need to measure or estimate the ACL loading to confirm whether CAI could relate to the mechanism of ACL injury.

CHAPTER 5

MANUSCRIPT II

Manuscript #2: DOES CHRONIC ANKLE INSTABILITY INFLUENCE LOWER EXTREMITY MUSCLE ACTIVATION OF FEMALES DURING LANDING?[1]

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[1]Li, Y., Walker, M., Brown, C. N., Schmidt, J. D. Kim, S.-H. and Simpson, K. J. To be submitted to *Journal of Electromyography and Kinesiology*

ABSTRACT

Much remains unclear that how chronic ankle instability (CAI) could affect knee muscle activations and interact with knee biomechanics and ACL loading. Therefore, the purpose of the present study was to assess the influence of CAI on the lower extremity muscle activation during the landing onto tilted surfaces. A surface electromyography (EMG) system and two force plates were used to collect lower extremity muscle activation of 21 young female individuals with CAI and 21 pair-matched control participants (CON) during a double-leg landing with test limb landed on the tilted surface. EMG linear envelope, time to peak EMG linear envelope, co-contraction index of ankle muscles in the frontal and sagittal plane, and co-contraction ratio of quadriceps to hamstring (Q/H CCR) was generated. Between-group differences were assessed by paired t-tests ($p < 0.05$). In the pre-landing phase, compared to CON, CAI displayed a reduced ankle evertor activation that could place CAI at a high risk of giving way or sprain injury. In the landing phase, the increased TIB-ANT activation of CAI leaded to increased co-contraction of ankle muscles in the sagittal and frontal plane, which increase the ankle stability during landing but may hinder the ability of ankle energy absorption especially in the frontal plane. A greater Q/H CCR of CAI group could relate to a higher ACL loading. In conclusion, CAI group displayed differences of lower extremity muscle activations and some atypical knee muscle activation may relate to the mechanism of ACL injury.

INDEX WORDS: electromyography, lower extremity, inverted landing surface, drop landings, ACL loading

5.1 Introduction

Chronic ankle instability (CAI) usually develops after an initial acute ankle sprain (Hertel, 2002; van Rijn et al., 2008). The common symptoms of CAI include pain, feeling of instability, episode of giving way and ankle weakness (Hubbard et al., 2007; Mitchell, Dyson, Hale, & Abraham, 2008). These symptoms could be related to changes in tissues (e.g., elongation of the anterior talo-fibular ligament and damage to the cartilage, (Hintermann, Boss, & Schäfer, 2002)), deficits in proprioception (Lee & Lin, 2008; Witchalls et al., 2012), and/or reduced ankle muscle strength (Willems et al., 2002). CAI may be a more serious condition than initially thought because it could be related to functional impairment (Simon, Donahue, & Docherty, 2012) and decreased physical activity levels (Hubbard-Turner & Turner, 2015).

Moreover, CAI may be related to knee injury (e.g., ACL injury: Kramer, Denegar, Buckley, & Hertel, 2007; Söderman, Alfredson, Pietilä, & Werner, 2001) due to the alterations of lower extremity biomechanics during high-impact movements (Gribble & Robinson, 2010; Gribble & Robinson, 2009; Terada, Pietrosimone, & Gribble, 2014). In our previous, related study of knee biomechanics of CAI (Study I), compared to healthy controls (CON), reduced ankle energy dissipation could lead to greater eccentric knee extensor moment and work and increased internal rotation moments that are related to the mechanisms of ACL strain (DeMorat et al., 2004; Fleming et al., 2001).

One possible reason for altered biomechanics is differences in neuromuscular control. Neuromuscular differences of ankle muscles for CAI group compared to CON have been observed, though conflicting findings exist. First, CAI group have demonstrated atypical muscle activation magnitudes during various weight-bearing movements (Brown, Ross, Mynark, & Guskiewicz, 2004; Delahunt, Monaghan, & Caulfield, 2007; Lin, Chen, & Lin, 2011; Suda, Amorim, & de Camargo Neves Sacco, 2009; Ty Hopkins, Coglianese, Glasgow, Reese, & Seeley, 2012). Increased tibialis anterior activation was found for CAI during the stance phase of walking (Louwerens, van Linge, de Klerk, Mulder, & Snijders, 1995; Ty Hopkins et al., 2012). Among CAI studies utilizing landing activities, the researchers observed reduced pre-landing peroneal activation (Caulfield, Crammond, O'Sullivan, Reynolds, & for CAI group compared with CON. However, for the landing phase, conflicting results among studies were reported for soleus (Brown et al., 2004; Delahunt et al., 2007), peroneal (Lin et al., 2011; Suda et al., 2009) and tibialis anterior activation magnitudes (Delahunt et al., 2007; Suda et al., 2009). Second, differences of neuromuscular reaction time between CAI and CON have been investigated. A trapdoor device was commonly used to measure the reaction time of the peroneal activation when the ankle was suddenly inverted. A greater latency of peroneal onset was found for CAI (Karlsson & Andreasson, 1992; Konradsen & Ravn, 1990); however, others did not observe timing differences between CAI and CON individuals (Vaes et al., 2002). Third, reduced eversion muscle strength has been reported for CAI (Rottigni & Hopper, 1991; Willems et al., 2002).

The ankle neuromuscular alterations of CAI could influence the ankle biomechanics and further interact with the knee biomechanics and muscle activation patterns of knee joint muscles, because the function and dysfunction at one joint can affect the function of the adjacent proximal joint of the kinetic chain (Kaminski & Hartsell, 2002; Terada et al., 2013, 2014). In general, we predict that a greater ankle muscle co-contraction would be exhibited by CAI compared to CON in order to stabilize the ankle joint during landing onto a sideward-tilted surface. The increased co-contraction could lead to reduced inversion displacement and decreased energy dissipation at the ankle joint that we observed in Study I. Consequently, greater knee extensor activation would be needed to achieve a greater extensor moment to increase the energy dissipation at the knee joint.

However, the above predictions cannot be confirmed based on previous studies, because conflicting results were found for knee kinematics when comparing CAI with CON (Caulfield & Garrett, 2004; Delahunt et al., 2006; Gribble & Robinson, 2009); and few studies observed knee kinetic differences (Terada et al., 2013 & Study I). Moreover, only one study investigated knee muscle activity of CAI group (Delahunt et al., 2007). The researchers reported greater rectus femoris activity in pre-landing and landing phases of the lateral hopping, but did not provide any detailed explanation for this observation.

Therefore, much remains unclear as to how CAI could affect knee muscle activations and their effects on knee biomechanics and ACL loading. Therefore, the purpose of the present study was to assess the influence of CAI on the lower extremity muscle activation patterns during landings onto tilted surface.

Based on our overall prediction above and the biomechanical outcomes observed in Study I, compared to CON, we hypothesize that CAI group will exhibit increased pre-landing tibialis anterior activation to invert the foot. In addition, based on our finding in Study I of the negligible magnitude of pre-landing net ankle inversion/eversion moment, we expected increased or equivalent ankle evertor (peroneus longus) activity of CAI to somewhat counteract the inversion moment created by tibialis anterior. Increased tibialis anterior and peroneus longus activity also were expected because the increased co-contraction could provide a more stable ankle for CAI (Lin et al., 2011). For the landing phase, we hypothesized that CAI group will exhibit increased co-contraction of tibialis anterior and gastrocnemius lateralis (sagittal plane), and increased co-contraction of tibialis anterior and peroneus longus (frontal plane) to stabilize the ankle joint. In addition, based on the increased knee extensor moment observed in Study I, we hypothesized that the co-contraction ratio of rectus femoris to biceps femoris activation would be greater for CAI group.

5.2 Methods

5.2.1 Participants

The same twenty-one females with CAI that participated in Study I (Li et al., 2016) also participated in this portion of the project. The inclusionary and exclusionary criteria for CAI were the published criteria for identifying CAI (Gribble et al., 2013) and assessed from the participants' answers on the laboratory health status and physical activity questionnaire reported and described in greater detail in Study I.

Correspondingly, the same twenty-one healthy control participants that were pair-matched with the CAI participants for gender, height, body mass and physical activity level in the first study also consented to be in the present study. All participants were healthy, without having had a serious lower extremity injury or dysfunction, and had experience in landing-related sports (e.g., basketball, volleyball, soccer, etc.). The participant characteristics are presented in Table 5.1.

Table 5.1. Demographical Data (mean ± SD) of the Participants

Variable	CON	CAI

Sample size	21	21
Body mass (kg)	64.4 ± 11.9	64.4 ± 12.4
Height (cm)	165 ± 6	164 ± 6
Age (yr)	21 ± 2	21 ± 2

Note: CON = healthy control participants; CAI = chronic ankle instability participants.

5.2.2 Instrumentation and Experimental Setup

A wireless surface EMG system (sampling rate = 2040 Hz, CMRR > 80 dB; Delsys Trigno^{TM} System, MA, USA) was used to measure the EMG of lower extremity muscles. Surface EMG electrodes (37 mm W by 27 mm L), with four silver bar-shaped contacts, were attached on the muscle belly of tibialis anterior (TIB-ANT), gastrocnemius lateralis (GAS-LAT), peroneus longus (PER-LON), rectus femoris (REC-FEM), vastus lateralis (VAS-LAT), biceps femoris (BIC-FEM) and tensor fascia latae (TNS-FAS) of the test limb.

As data for Study 1 were collected simultaneously with these data, and in order to define the instant of initial contact, vertical ground reaction forces (GRF) of each foot were collected using one Bertec force plate per foot (4060-NC, Bertec Corp., OH, USA) at 2040 Hz. The two force plates were located as shown in Figure 5.1; one side was tilted downwards 25° (inverted) in the lateral direction and the other one flat such that its height was the same as the plate's antero-posterior axis. EMG and GRF data were captured and synchronized using Vicon Nexus^{TM} 2.2 software (Vicon Motion Systems Ltd., Oxford, UK). The motion capture data of Study I were not utilized.

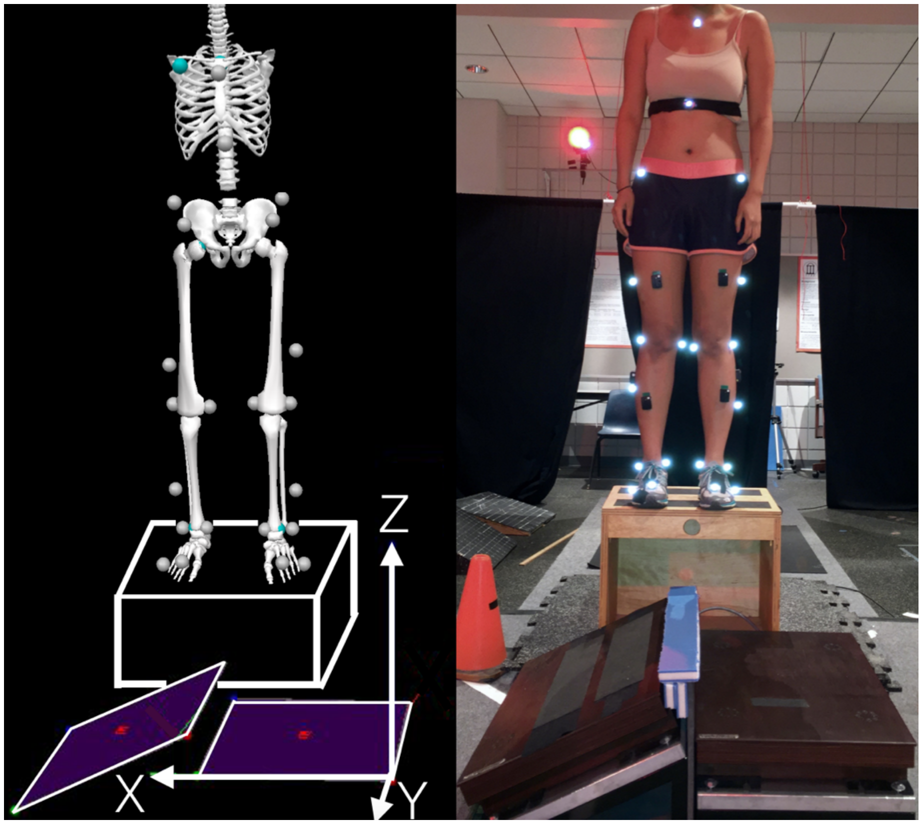

Figure 5.1. Experimental set up and surface EMG electrode placement.

5.2.3 Procedures

The study was approved by our institutional review board, and all participants provided informed consent before data collection. Participants completed the health status and physical activity questionnaire; and the Identification of Functional Ankle Instability (IdFAI) questionnaire and the Cumberland Ankle Instability Tool (CAIT) to determine level of chronic ankle instability of each limb. From the self-reported answers, if the participant was deemed eligible to participate, the limb with less ankle stability (greater IdFAI and lower CAIT score) was chosen as the test limb for CAI. For a pair-matched CON, the test limb was the limb that had the same limb dominance as the corresponding the CAI test limb. The dominant limb was classified as the limb that the participant verbally choose as their ‘kicking’ leg (Palmieri-Smith, McLean, et al., 2009; Yeow et al., 2010).

Before the electrodes were attached, each electrode site location was prepared by shaving any hair, then wiping the skin area with isopropyl alcohol to remove oils and surface residuals. The electrodes were placed on the participant at the locations suggested in Cram et al. (1998) guidelines.

In order to normalize the EMG data to compare across participants, ‘maximum’ voluntary isometric contraction (MVIC) tests were conducted. For all MVIC tests, the participant was placed in the test position and a hand-held dynamometer (FCE Series Medical Dynamometer, AMETEK, Inc., PA, USA) was used to create isometric resistance and obtain the resistance force value. A five-second EMG signal was captured for each muscle MVIC test (Dai, Sorensen, et al., 2012); and a 30-second break was taken between the tests. For TIB-ANT, the participant was in a supine position with the ankle joint relaxed, and knee and hip joints in the neutral (straight) position (Hsu et al., 2006); and the resistance was applied on the instep of the foot, 10 cm distal to the ankle joint center. For GAS-LAT, participant was in a prone position with neutral joint positions (Hsu et al., 2006); and the resistance was applied on the ball of the foot, 10 cm distal to the ankle joint center. For PER-LON, participant was sitting on a table with the lower leg hanging from the table, and knee and hip joints at 90° flexion (Escamilla et al., 2006); and the resistance was applied on the lateral side of the heel, 3 cm below the ankle joint center. For REC-FEM and VAS-LAT, participant was in the same position as for PER-LON test (Escamilla et al., 2006, 2010); the resistance was applied on the front of the tibia, 25 cm distal from the knee joint center. For BIC-FEM, participant was in a prone position with the knee joint flexed at 90° (Burnett et al., 2012); the resistance was applied on the posterior side of the lower leg, 25 cm distal from the knee joint center. The distance of applied resistant force to the joint center was chosen based on our pilot study.

After five minutes of treadmill jogging, the participant practiced performing the drop landing task, then performed ten acceptable drop landing trials. For a given landing trial, the participant stood on a box 30 cm above the foot landing targets, then stepped forward with the test limb, then the other limb and landed with test foot on the tilted force plate and the other foot on the flat force plate. To ensure consistent performance, the participant was not allowed to jump, hop or step down from the box (Kulas et al., 2006). Upon completion of the landing, the person was instructed to slowly stand up. The criteria for acceptable trials are described in Study I.

5.2.4 Data Analysis

The phases of interest for analysis included the pre-landing and landing phase. The pre-landing phase was defined as the interval from 50 ms prior to contact (-50 ms) to the instant of initial contact (IC, vertical GRF > 10 N). The landing phase started from the IC and ended at the first instant when the center of mass (COM) reached its lowest height (Kulas et al., 2006). The COM height was determined from Study I data. For the EMG data, the DC offset was removed by subtracting the mean value of the EMG signal of the entire trial from the raw data. Then the signal was filtered using a Butterworth band-pass filter (20-450 Hz). Filtered EMG data were rectified and the linear envelope was generated using a low-pass filter at 10 Hz (Dai et al., 2014; Dai, Sorensen, et al., 2012). The maximum one-second average of the MVIC linear envelope was used as the MVIC amplitude (Dai, Sorensen, et al., 2012). For the drop landing EMG data, the linear envelope data were averaged for the 50 ms before IC (Nagano et al., 2007) and for the first 100 ms after IC (Dai et al., 2014) and normalized to the corresponding muscle's MVIC amplitude to represent the amplitude of muscle activation during the pre-landing and landing phases, respectively. The first 100 ms after IC was chosen because ACL injuries usually occur within that period, according to injury case analyses (Koga et al., 2010; Krosshaug et al., 2007). In order to estimate the joint strength, the maximum MVIC moment was calculated by multiplying the maximum resistant force (measured by the dynamometer) and the moment arm (distance from the location of applied resistance to the joint center).

Co-contraction ratio (CCR) of REC-FEM to BIC-FEM (Q/H = quadriceps/hamstring) activation was calculated (Equation 1) because it could influence the ACL loading (G Li et al., 1999). Co-contraction index (CCI) in the sagittal plane between the TIB-ANT and GAS-LAT and frontal plane between TIB-ANT and PER-LON was calculated as shown in Equation 2 (Lin et al., 2011; Suda et al., 2009). The CCI of the ankle joint muscles was assessed because it relates to joint stability (Lin et al., 2011). The antagonist was TIB-ANT in both sagittal and frontal plane; the agonist was GAS-LAT and PER-LON in the sagittal and frontal plane, respectively.

$$CCR = \frac{EMG_{REC-FEM}}{EMG_{BIC-FEM}} \qquad \text{(Equation 1)}$$

$$CCI = \frac{EMG_{Antagonist}}{(EMG_{Antagonist}+EMG_{Agonist})/2} \qquad \text{(Equation 2)}$$

in which $EMG_{Antagonist}$ and $EMG_{Agonist}$ represents the average linear envelope values for a given phase for the antagonist and agonist muscles, respectively. Time to peak EMG linear envelope was calculated in order to reveal the neuromuscular control and muscle activation strategy during landings.

5.2.5 Statistical Analysis

Participants' maximum MVIC moment and ankle instability scores were compared between groups using paired *t*-tests ($p < 0.05$). For each muscle and relevant phase(s), the ten trial average of a given EMG variable for each participant was used for statistical tests. To detect potential outliers, Chauvenet's criterion was applied. Five outliers were found among all the tested EMG variables. Group comparisons of EMG variables were run using paired *t*-tests ($p < 0.05$). Possible tendency toward significance was defined as $0.05 \leq p < 0.10$. Effect size (Cohen's d) was reported. Statistical analyses were processed in SPSS™ (v.22; IBM Corporation, USA).

5.3 Results

Participants' joint strength and ankle instability (CAIT and IdFAI scores) are presented in Table 5.2. Compared to CON, CAI group displayed less ankle stability (lower CAIT and greater IdFAI scores) and ~4 Nm and ~1.6 Nm lower max MVIC moments for ankle dorsiflexion and eversion, respectively, and a tendency of reduced plantarflexor strength (~5 Nm less, $p = 0.054$). No significant differences were observed for knee or hip joint max MVIC moments ($p = 0.151 – 0.590$).

Table 5.2. Ankle Instability Scores and Maximum MVIC Moments (Nm).

Variable		CON	CAI	*t*-value	*p*-value	Effect size
Ankle Instability Score	CAIT score	29.5 ± 0.9	19.3 ± 6.0	7.37	**0.000***	2.38
	IdFAI score	1.3 ± 2.1	22.2 ± 9.2	9.53	**0.000***	3.13
Max MVIC	Ankle dorsiflexor	18.0 ± 5.3	14.1 ± 3.5	3.40	**0.003***	0.87

Moment	Ankle plantarflexor	22.2 ± 10.6	17.1 ± 4.7	2.06	**0.054^**	0.62
	Ankle evertor	6.3 ± 2.8	4.7 ± 1.7	3.00	**0.007***	0.69
	Knee extensor	95.7 ± 39.7	76.5 ± 33.4	1.48	0.156	0.52
	Knee flexor	33.0 ± 8.2	31.7 ± 8.3	0.55	0.590	0.16
	Hip abductor	46.7 ± 16.7	40.8 ± 8.0	1.50	0.151	0.45

Note: CON = healthy control participants; CAI = chronic ankle instability participants. * indicates statistical significant group comparison ($p < 0.05$); ^ indicates a tendency of statistical significance ($0.05 \leq p < 0.1$).

In the pre-landing phase, compared with CON, CAI group exhibited ~70% MVIC reduced PER-LON activation, a significantly ~15% MVIC greater GAS-LAT, and a tendency of ~10% MVIC increased VAS-LAT activation ($p = 0.083$). A significantly greater ankle muscle co-contraction in the frontal plane was displayed by CAI (CCI=0.77 ± 0.36) compared to CON (CCI=0.52 ± 0.35). During the pre-landing phase (Figure 5.2 & 5.3 and Table 5.3), No other significant between-group differences were found (p = 0.166 – 0.992).

In the landing phase, CAI compared to CON individuals exhibited a ~60% MVIC greater TIB-ANT activation, a ~80% MVIC greater REC-FEM activation and 53% MVIC greater VAS-LAT, and a tendency of ~17% MVIC reduced BIC-FEM activation ($p = 0.068$) (Figure 5.3). A greater CCI was found in both sagittal (0.94 ± 0.20) and frontal planes (0.75 ± 0.24) for CAI compared to CON individuals (0.78 ± 0.27 and 0.56 ± 0.21, respectively). Moreover, the quadriceps to hamstring CRR was significantly greater (~2.7) for CAI group. Other muscles exhibited no significant differences for linear envelope EMG, CCI, or CCR between groups (p = 0.540 – 0.776).

Time to peak EMG linear envelope is presented in Table 5.4. Almost all the peak muscle activations occurred within 100 ms after IC. The only significant between-group difference was observed for tensor fascia latae. Compared to CON, CAI exhibited peak tensor fascia latae linear envelope EMG ~30 ms later. Other temporal variables were not significantly different (p = 0.106 – 0.617).

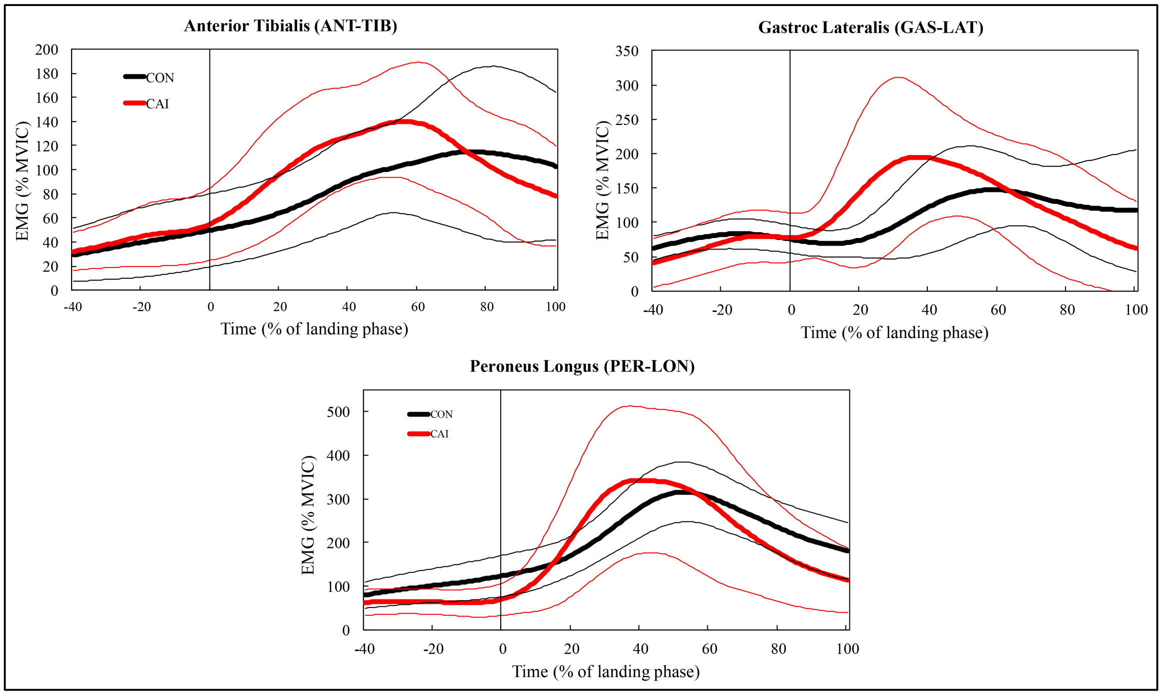

Figure 5.2. Group ensemble average curves of the linear envelope EMG (+/- SD) of ankle muscles. The vertical line indicates the instant of initial contact (0%). A portion of the pre-landing phase to the end of the landing phase (100%) is shown.

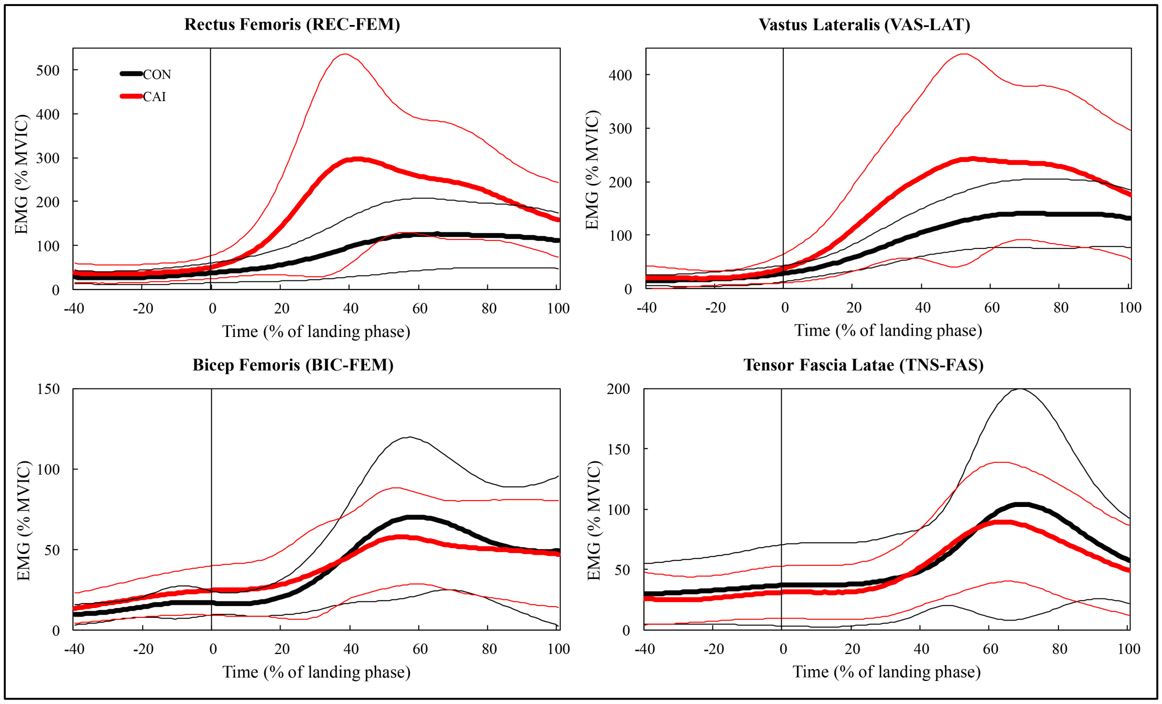

Figure 5.3. Group ensemble average curves of the linear envelope EMG (+/- SD) of knee and hip muscles. Initial contact = 0% and end of landing phase = 100%. A portion of the pre-landing phase also is shown.

Table 5.3. Mean ± SD (% MVIC) of EMG Linear Envelope, CCR and CCI.

	Muscle	CON	CAI	*t*-value	*p*-value	Effect size
Pre-landing phase	TIB-ANT	31.3 ± 19.7	39.5 ± 25.9	0.95	0.355	0.35
	GAS-LAT	80.9 ± 51.0	95.6 ± 48.2	3.04	**0.008***	0.31
	PER-LON	143.8 ± 112.7	72.9 ± 50.3	3.28	**0.005***	0.81
	REC-FEM	30.9 ± 25.1	47.8 ± 39.9	1.39	0.183	0.71
	VAS-LAT	36.2 ± 18.2	45.9 ± 10.0	1.85	**0.083^**	0.66
	BIC-FEM	21.4 ± 18.2	36.5 ± 35.7	1.45	0.166	0.53
	TNS-FAS	75.4 ± 67.5	79.8 ± 88.4	0.32	0.756	0.05
	CCR Q/H	1.58 ± 1.40	1.58 ± 1.10	0.01	0.992	0.00
	CCI sagittal	0.61 ± 0.38	0.58 ± 0.26	0.30	0.768	0.09

	CCI frontal	0.52 ± 0.35	0.77 ± 0.36	2.81	**0.013***	0.70
Landing phase	TIB-ANT	84.2 ± 33.1	123.5 ± 40.5	3.16	**0.006***	1.06
	GAS-LAT	132.5 ± 51.2	138.9 ± 50.0	0.39	0.701	0.13
	PER-LON	226.3 ± 70.0	217.6 ± 77.4	0.29	0.776	0.12
	REC-FEM	96.8 ± 62.1	177.4 ± 70.0	2.76	**0.015***	1.22
	VAS-LAT	102.5 ± 48.6	155.5 ± 46.7	2.80	**0.013***	1.11
	BIC-FEM	54.8 ± 31.4	38.1 ± 17.7	1.96	**0.068^**	0.65
	TNS-FAS	48.1 ± 20.8	44.0 ± 20.6	0.63	0.540	0.20
	CRR Q/H	2.31 ± 1.82	4.97 ± 3.16	2.56	**0.023***	1.03
	CCI sagittal	0.78 ± 0.27	0.94 ± 0.20	2.13	**0.049***	0.67
	CCI frontal	0.56 ± 0.21	0.75 ± 0.24	2.12	**0.049***	0.84

Note: CON = healthy control participants; CAI = chronic ankle instability participants. * indicates statistical significant ($p < 0.05$); ^ indicates a tendency of statistical significant ($0.05 \le p < 0.1$).

Table 5.4. Mean (+/- SD) of Time to Peak EMG Linear Envelope (ms) during the Landing Phase.

Muscle	CON	CAI	*t*-value	*p*-value	Effect size
TIB-ANT	88 ± 25	84 ± 23	0.51	0.617	0.17
GAS-LAT	65 ± 24	76 ± 15	1.64	0.119	0.55
PER-LON	72 ± 15	77 ± 16	0.94	0.360	0.32
REC-FEM	89 ± 17	83 ± 13	1.01	0.327	0.40
VAS-LAT	92 ± 21	100 ± 33	0.82	0.427	0.29
BIC-FEM	81 ± 29	99 ± 26	1.89	0.106	0.65
TNS-FAS	63 ± 36	93 ± 14	3.18	**0.006***	1.10

Note: CON = healthy control participants; CAI = chronic ankle instability participants. * indicates statistical significant ($p < 0.05$).

5.4 Discussion

Our general predictions were that CAI increased ankle muscle co-contraction to stabilize the ankle joint; limited ankle inversion displacement and eversion moment (presented in Study 1) led to greater knee extensor activation to do greater negative work; and thus greater quadriceps to hamstring co-contraction ratio. These predictions were supported by the outcomes of the present study. The primary findings were that the CAI group displayed some differences of lower extremity linear envelope EMG magnitudes in both pre-landing and landing phases. Reduced peroneal activity in pre-landing of CAI may place the ankle at vulnerable position to giving way and sprain injury. The increased co-contraction of ankle muscles could lead to altered knee muscle activations (e.g., increased knee extensors activities and CCR) that may relate to the mechanism of ACL loading.

5.4.1 Pre-landing phase

<u>Activation of ankle joint muscles:</u>

CAI group exhibited reduced PER-LON activations and may result in an increased ankle co-contraction in the frontal plane compared to CON, which contradicts our hypothesis. The primary function of PER-LON is to produce ankle eversion or control/resist foot-ankle inversion, thus, to prevent excessive ankle inversion when landing on the tilted surface. The decreased PER-LON activation may result in an increased pre-landing ankle inversion angle for CAI group and reduce the ankle stability, observed in the Study I and previous studies (Caulfield et al., 2004; Delahunt et al., 2006).

There are two potential negative consequences to reduced peroneal activation that could place the ankle at a high risk of ankle giving way or even sprain injury. First, the preparatory peroneal activity is important to protect the ankle joint when encountering a sudden inversion torque (Isakov, Mizrahi, Solzi, Susak, & Lotem, 1986; Konradsen, Voigt, & Hojsgaard, 1997). Due to the electromechanical delay, peroneal activation at IC and during the landing may not be early enough to generate sufficient eversion moment to control the ankle inversion/eversion position. This is particularly true for landing onto the tilted surface that requires a relatively greater eversion moment compared to a flat surface landing. Therefore, in real landing situations, CAI group may be more vulnerable to inversion moment created by the GRF when landing unexpectedly on a sideward tilted surface, such as someone's foot. Second, the increased ankle inversion angle possibly due to decreased peroneal activity in pre-landing and at IC may laterally shift the subtalar joint axis relative to the line of action of GRF thus an inversion moment generated from the GRF (see Tropp, 2002: Figure 4). Our results were also supported by other studies, in which they observed the reduced pre-landing PER-LON activity for CAI group (Caulfield, Crammond, O'Sullivan, Reynolds, & Ward, 2004; Delahunt et al., 2006; Suda, Amorim, & de Camargo Neves Sacco, 2009).

Different from our hypothesis in which we surmised that no differences were displayed in GAS-LAT activation, CAI group exhibited increased pre-landing GAS-LAT EMG compared to CON. There are two potential explanations for this outcome. First, because of the limited plantarflexion range of motion (~28° observed in Study I), CAI group may need to use a greater plantarflexor muscle activation during pre-landing to position the foot in a plantarflexion angle (~29° observed in Study I) sufficient to allow for enough dorsiflexion displacement to dissipate the body's kinetic energy (Terada et al., 2013) during the landing phase.

Second, the increased GAS-LAT EMG may counteract with the increased VAS-LAT activation in pre-landing and relate to the increased knee flexion angle (observed in

Study I and Caulfield et al. 2002). Our finding may be supported by Delahunt et al (2007), in which they reported similar findings for the increased soleus muscle activation of CAI group in the pre-landing.

Activation of knee joint muscles:

No significant differences of EMG linear envelope for muscles that cross the knee joint were observed between groups, which was consistent with our prediction, except for a tendency of increased CAI VAS-LAT activation. There are two potential explanations for this tendency. First, CAI group tended to increase VAS-LAT activation to increase the frontal plane knee stability/stiffness (Loui & Mote, 1987; Olmstead, Wevers, Bryant, & Gouw, 1986) to prepare for a stable landing (Dai, Sorensen, et al., 2012) onto the tilted surface. Second, there was a tendency that CAI group increased knee extensor activation to counteract the increased GAS-LAT activation to prevent excessive knee flexion angle in the pre-landing phase. This would allow for sufficient knee flexion displacement to absorb energy in the landing phase (Norcross et al., 2013). However, we did not observe a significant difference or a tendency of the other knee extensor muscle (REC-FEM) possibly due to large standard deviation within the group thus a higher type II error to mask the true difference.

Whether there is a significantly increased pre-landing knee extensor activation for CAI still needs to be confirmed by future studies, because contradicting results also have been reported by previous studies. Delahunt and colleagues observed an increased REC-FEM activity of CAI with increased TIB-ANT and soleus muscles during the pre-landing phase of the lateral hopping (Delahunt, Monaghan, & Caulfield, 2007). They suggested that CAI may use the increased muscle activations during hopping to control the center of gravity (COG) upon IC. However, it is unclear how to control the COG by these muscles, because COG trajectory cannot be altered by internal forces (e.g., muscle contraction) in the pre-landing phase. Other studies did not find any significant difference of the knee extensor activity (Delahunt, Monaghan, & Caulfield, 2006; Santos & Liu, 2008).

5.4.2 Landing phase

<u>Ankle muscle activation:</u>

The increased TIB-ANT activation was observed for CAI compared to CON participants. The major mechanical purpose of the increased dorsiflexor activity could be that CAI used the dorsiflexor activation to increase the ankle stability. This could be a strategy that employing active stabilizers (i.e., TIB-ANT muscle) to compensate for the limited ability of passive stabilizers (e.g., ligaments) of CAI group. In addition, due to the limited ankle dorsiflexion range of motion of CAI (observed in Study I and Denegar, Hertel, & Fonseca, 2002; Hoch, Staton, McKeon, Mattacola, & McKeon, 2012) and reduced maximum MVIC moment of ankle dorsiflexor that may associate with joint strength (observed in the present study), increased dorsiflexor activation may be necessary to bring the ankle into the ‘close-packed’ position. The ‘close-packed’ position could achieve higher ankle stability because of the shape of talus dome (Norkus & Floyd, 2001). The ‘close-packed’ position may be especially necessary during landing onto the tilted surface when a large inversion moment is exerted by the vertical GRF. Moreover, the increased TIB-ANT activity could also contribute to laterally shifting the center of pressure (Kim, Uchiyama, Kitaoka, & An, 2003) that would, in turn, reduce the VGRF moment arm. Consequently, the close-packed foot-ankle position would create a lower inversion VGRF moment.

Increased TIB-ANT activation was also observed in even-surface landing from previous studies (Delahunt et al., 2007; Louwerens, van Linge, de Klerk, Mulder, & Snijders, 1995), which could support our explanations. Delahunt el al. suggested that the increased TIB-ANT activation for both pre-IC and post-IC aimed to controlling the center of gravity at IC. However, no detailed description nor explanation was provided. Future studies may need to examine the influence of TIB-ANT activation of CAI on the center of gravity or the center of pressure at landing and investigate how it could influence the postural stability.

As we expected, the ankle muscle co-contraction index was greater in the sagittal and frontal planes for CAI group. The increased co-contraction was due to the increased TIB-ANT (antagonist in the sagittal and frontal plane) activation. We believe that the increased ankle muscle co-contraction in the frontal plane of CAI could reduce the energy dissipation at the ankle joint based on two rationales. First, the increased ankle muscle co-contraction combined with the evidence of increased ankle inversion at IC (observed in Study I) could limit the ankle inversion displacement during the landing phase of CAI group. Second, the increased ankle muscle co-contraction could also decrease the eversion moment due to increased TIB-ANT activation. Consequences of both rationales could impede the eccentric work done by the ankle muscles, thus limiting the energy absorption of these muscles. The reduced energy absorption at the ankle joint could influence the knee biomechanics (observed in Study I) and knee muscle activations (described below).

Knee muscle activation:

As we expected, CAI group exhibited a greater quadriceps to hamstrings CCR due to greater knee extensor activation but reduced knee flexor activation. The greater Q/H CCR has been related to anterior tibial translation and ACL strain when the knee flexes less than 45° (Arms et al., 1984). The higher knee extensor activity could increase the anterior shear force applied at the proximal end of the tibia, which is the primary contributor of ACL loading (Berns et al., 1992; Dai, Herman, et al., 2012; K L Markolf et al., 1995). A previous research group observed that "aggressive" quadriceps contraction could produce significant anterior translation of the tibia and ACL strain (DeMorat et al., 2004). On the other side, the reduced knee flexor activity could reduce the forces that counteract the anterior shear force (Ko, Yang, Ha, Choi, & Kim, 2012; Yoon et al., 2013). Therefore, CAI group may undergo greater ACL loading, especially during the early landing phase (100 ms post IC) with little knee flexion. However, as we did not measure ACL loading, this conjecture needs to be confirmed in the future.

The direct cause of the increased Q/H CCR of CAI is not known to us. One plausible explanation could be that CAI group used more knee extensor activity to generate greater eccentric work in order to compensate the deficits of the ankle joint in dissipating the energy (see Study I). However, no other studies have reported this increased CRR, because most studies investigating muscle activity of CAI focused on the ankle, not knee joint muscles (Brown et al., 2004; Caulfield et al., 2004; Palmieri-Smith, Hopkins, & Brown, 2009). One study found increased rectus femoris activity of CAI during 200 ms post IC, but no CCR ratio was reported (Delahunt et al., 2007). Moreover, the greater VAS-LAT activity could possibly increase the contraction force on the lateral knee and predispose the knee joint to excessive valgus position (knee abduction) that increase the ACL strain (Letafatkar, Rajabi, Tekamejani, & Minoonejad, 2014; Palmieri-Smith, Hopkins, et al., 2009). However, without the data of vastus medialis activation, this cannot be confirmed. Moreover, no increased knee abduction was found for CAI group. Therefore, we suggest that future study may investigate both anterior-posterior and medial-lateral co-contraction ratio to better understand the knee muscle activity of CAI group and how it could relate to the mechanism of ACL injury.

5.4.3 Summary of discussion

Combined with the findings of Study I, we believe that CAI could potentially increase the ACL loading during the landing onto the inverted surface through several possible mechanisms. First, we propose that due to some neuromuscular and proprioception deficits, reduced pre-landing peroneal activity and a more inverted ankle joint of CAI could place the ankle in a less stable position. Second, in order to stabilize the ankle joint in the landing, CAI group increased the ankle muscle co-contraction ratio in the sagittal and frontal planes by increasing the ANT-TIB activation. Third, the increased co-contraction in the frontal plane along with increased ankle inversion angle at IC reduced the ability of the ankle joint to dissipate the energy by decreasing the ankle inversion displacement and eversion moment. Fourth, reduced ankle energy dissipation leads to an increased knee eccentric work; CAI increased knee extensor moment through greater knee extensor activation and reduced knee flexor activation to achieve a greater knee energy absorption. Finally, the increased knee extensor activation and tendency of reduced knee flexor activation could result in a greater anterior shear force at the proximal tibia thus a higher ACL tensile loading.

5.4.4 Limitations

We used a single muscle to represent the ankle dorsiflexor, plantarflexor, evertor and knee flexor muscles, respectively. Other muscles within the same muscle group (e.g., peroneus brevis) may act differently than the muscle we tested (e.g., peroneus longus). A hand-held dynamometer was used to create and measure the participant's force. Hand holding this instrument may not have resulted in the participant generating her true maximal isometric force. As is typical during isometric MVIC tests, the magnitude of EMG measured is not the maximal amount that can be generated during actual movements (Halaki & Ginn, 2012). This was evident for the peroneal EMG linear envelope values. However, the MVIC test protocol was consistent for each participant; thus the results should not be biased. Another limitation is that the participants expected to land on a tilted surface during the study, which may reduce the external validity to generalize these findings for landings that are unexpected. However, these findings likely are generalizable to those scenarios of expected landings onto sideward tilted surfaces (e.g., trail running).

5.5 Conclusion

In conclusion, CAI group displayed some differences of lower extremity muscle activation compared to CON. In the pre-landing phase, reduced ankle evertor activation could place a CAI ankle at a higher risk of giving way. The increased TIB-ANT activation of CAI could increase the ankle stability in the landing but may hinder the ability of ankle energy absorption and further adversely influence the knee biomechanics and muscle activations. A greater quadriceps to hamstring co-contraction ratio due to greater knee extensor activation and a tendency of reduce knee flexor activation of CAI could relate to a higher ACL loading, especially during the early landing phase (100 ms post initial contact). We conclude that the alterations in ankle muscle activation may result in some atypical knee muscle activation that related to the ACL loading. Relevant training programs (e.g., increasing pre-landing peroneal activation, optimizing activation ratio of quadriceps to hamstrings) may help individuals with CAI improve ankle stability and reduce atypical knee loading. However, future study may need to confirm whether these altered lower extremity muscle activation could increase the ACL loading.

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APPENDIX A

Pre-Participation and Health Status Questionnaire

For researcher use only
Participant ID _______
Date ______________
Researcher: _____________________

Pre-Participation and Health Status Questionnaire

When is your birthday? Month: _______ Day ________ Year _______

HEALTH AND MEDICAL CONDITIONS

If you now have or have had one of the conditions listed, put an X in the box provided. If you are unsure, put a question mark next to box.

- ☐ Heart problem
- ☐ Lung problem
- ☐ Trouble breathing or asthma
- ☐ Broken bones
- ☐ Sprains, or hurt an ankle, shoulder, hip, or knee
- ☐ Injury requiring major medical attention
- ☐ Inner ear problem
- ☐ Pain lasting more than 2 weeks
- ☐ Balance problem
- ☐ Blurred or bad eyesight or other eye problem not corrected
- ☐ Surgery
- ☐ Other medical condition(s)

CURRENT HEALTH STATUS

Have you had during the past 2 weeks or have today, any of the following: (check all that apply)

- ☐ Discomfort or pain to any part of body
- ☐ Chest pain or tightness, tingling in arm
- ☐ Trouble breathing
- ☐ Injury
- ☐ Illness
- ☐ Any other health problems
- ☐ Feeling sick to your stomach
- ☐ Trouble with balance
- ☐ Trouble seeing
- ☐ Had any medical or dental procedures
- ☐ Feeling dizzy or light-headed
- ☐ Soreness

1. __Yes __No Are you taking or using any medications? If yes, list each, including those you can buy without a nonprescription:

____

2. If you are female, is there any chance that you are pregnant? __Yes __No ___Don’t know

3. Is there anything else that we should know concerning your health?

___ No ___ Yes ___ Maybe

APPENDIX B
Physical Activity Questionnaire

1. What sports do you usually play?
   - ☐ Basketball
   - ☐ Volleyball
   - ☐ Soccer
   - ☐ Rugby
   - ☐ Tennis
   - ☐ Other ______________________________

2. How often do you participate in the jumping related sports (e.g. basketball, volleyball, soccer, handball)?
   - ☐ Never
   - ☐ Once a year
   - ☐ Once a month
   - ☐ Once a week
   - ☐ Almost every day
   - ☐ Other _______________________________

3. How many hours per week do you participate in light activities?
   (Light activities = your heart beats slightly faster than normal; you can talk and sing. e.g. walking, stretching, light yard work)
   - ☐ 0 – 0.5 hrs
   - ☐ 0.5 – 1.0 hrs
   - ☐ 1.0 – 1.5 hrs
   - ☐ 1.5 – 2.0 hrs
   - ☐ 2.0 – 2.5 hrs
   - ☐ Other _______________________________

4. How many hours per week do you participate in moderate activities?
   (Moderate activities = your heart beats faster than normal; you can talk but not sing. e.g. fast walking, aerobics class, strength training, swimming gently)
   - ☐ 0 – 0.5 hrs
   - ☐ 0.5 – 1.0 hrs
   - ☐ 1.0 – 1.5 hrs
   - ☐ 1.5 – 2.0 hrs
   - ☐ 2.0 -2.5 hrs
   - ☐ Other _______________________________

5. How many hours per week do you participate in vigorous activities?
   (Vigorous activities = your heart rate increases a lot; you cannot talk or your talking is broken up by large breaths. e.g. jogging or running, intensive sports, stair machine)
   - ☐ 0 – 0.5 hrs
   - ☐ 0.5 – 1.0 hrs
   - ☐ 1.0 – 1.5 hrs
   - ☐ 1.5 – 2.0 hrs
   - ☐ 2.0 -2.5 hrs
   - ☐ Other _______________________________

APPENDIX C

CUMBERLAND ANKLE INSTABILITY TOOL

Please mark the ONE statement in EACH question that BEST describes your ankles. Participant ID ______

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

APPENDIX D
IDENTIFICATION OF FUNCTIONAL ANKLE INSTABILITY (IDFAI)

Participant ID _______________

Instructions: Please fill out the form completely and if you have any questions, please ask the administrator.

	Left	Right
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	☐	☐
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)	☐	☐
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	☐	☐
5. When was the last time you had ***"giving way"*** in your ankle?		
Never	☐	☐
Once a year	☐	☐
Once a month	☐	☐
Once a week	☐	☐
Once a day	☐	☐
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	☐	☐
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	☐	☐
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	☐	☐
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	☐	☐
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	☐	☐