Tibial Torsion Influences Lower Body Joint Kinematics and Kinetics in Women with Anterior Cruciate Ligament Injury During Squatting and Landing

by

Michael George Chizewski

A thesis submitted in partial fulfillment of the requirements for the degree of

Doctor of Philosophy

Faculty of Physical Education and Recreation

University of Alberta

ABSTRACT

External tibial torsion is an individual anatomical variation in the leg segment that alters musculoskeletal alignment which may increase risk of ACL rupture; however, the effect of external tibial torsion on lower body kinematics in healthy and ACL injured populations is underreported. The purpose of this dissertation was threefold: first to examine tibial torsion as a potential mediator of tri-planar leg rotation during a partial squat, second to compare tibial torsion angles between women with and without ACL injury; and third to investigate the influence of tibial torsion on cross-planar segmental contributions to lower body joint motion in women with and without ACL injury during squatting and landing. To accomplish these objectives, three studies were completed. Study 1: fourteen physically active men (29.4 ± 7.9 years old, 1.78 ± 0.06 m tall, 77.9 ± 9.2 kg body mass) and sixteen physically active women (25.1 ± 6.0 years old, 1.68 ± 0.07 m tall, and 63.5 ± 6.7 kg body mass) volunteered to be in the study. Each participant performed three consecutive partial squats to maximum dorsiflexion for motion analysis. External tibial torsion was significantly correlated with transverse plane leg (R^2=0.459), knee (R^2=0.262) and thigh (R^2=0.158) angular excursions. Study 2: fifteen physically active women with ACL injury (24.6 ± 7.3 years old, 1.68 ± 0.08 m tall, and 67.6 ± 11.6 kg body mass), and fifteen physically active women without ACL injury (25.9 ± 5.9 years old, 1.69 ± 0.07 m tall, and 63.7 ± 6.0 kg body mass) volunteered to be in the study. Tibial torsion was measured using motion analysis, in the ACL injured and non-injured participants, for the left and right limbs. ACL injured women had significantly greater external tibial torsion angles (19 ± 4 degrees) compared to the healthy control participants (12 ± 4 degrees). Cohen's effect size value (*d*=1.79) suggested a high practical significance. Study 3: fourteen physically active women with reconstructed unilateral ACL injury (23.2 ± 2.4 years old, 1.71 ± 0.07 m tall, and 67.3 ± 10.8 kg body mass), and thirty-four physically active women, divided in groups based on external tibial torsion, without ACL injury (23.8 ± 4.1 years old, 1.68 ± 0.05 m tall, and 64.9 ± 9.4 kg body mass) participated in a three group case-control study design. The three groups consisted of ACL injured participants, high tibial torsion control participants, and low tibial torsion control participants. Participants performed three body weight partial squats, six two-foot vertical jump landings, and six step-off box landings (three left foot lead and three right foot lead) for motion analysis. Transverse plane leg medial rotation was higher in the low torsion group than the high torsion group (d=1.06) and the ACL injured group (d=0.42). During the jump and box landings, ankle plantar flexor net joint moment was greater in the low torsion group than the high torsion group (d=1.43), and the ACL injured group (d=0.82). During the jump landings, frontal plane peak leg abduction was greater in the high torsion group compared to the low torsion group (d=0.54) and the ACL group (d=0.61). In conclusion, these data suggest there are similarities between kinematics known to be ACL injury risk factors and lower body kinematics that have been linked to individuals who demonstrate greater external tibial torsion. External tibial torsion must be considered when discussing differences in lower body motion and ACL injury for squatting and landing.

PREFACE

This thesis is an original work by Michael Chizewski. The research projects, of which this thesis is a part, received research ethics approval from the University of Alberta Research Ethics Board as follows:

- The effect of rear-foot motion and leg anatomy on direction of lower leg rotation during walking and squatting, Pro00028411, approved February 29, 2012.
- Tibial torsion in ACL injured women, Pro00028419, approved February 29, 2012.
- Comparison of different methods of measuring tibial torsion, Pro00046837, approved March 18, 2014.
- The influence of tibial torsion on lower body mechanics in women with ACL injury, Pro00048284, approved May 21, 2014.

ACKNOWLEDGEMENTS

I would like to thank my supervisor, Dr. Loren Chiu; as well as my supervisory committee members, Dr. Brian Maraj and Dr. Samer Adeeb, for their guidance and patience.

I would like to thank Dr. Eric Parent and Dr. John (Pierre) Baudin for serving as external examiners. I would like to thank Dr. Andrew Harrison for agreeing to serve as external reviewer and Dr. Dan Syrotuik for agreeing to serve as Chair of the Examination Committee.

Thank you to all the study participants, and to my fellow graduate students, who volunteered their time. Your commitment to my project was integral throughout the process.

I would like to thank my wife and children (Ruth, William, Timothy and Daniel) for providing daily reminders of what is, and always will be, truly important in this life.

Table of Contents

CHAPTER 1 1
1.1 Introduction 1
1.2 Segment rotation and joint motion 1
1.3 Segmental contributions to sagittal plane ankle motion 6
1.4 Non-sagittal movement patterns 9
1.5 Tibial torsion and knee joint pathology 10
1.6 Summary 13
1.7 Purposes and Hypotheses 14
1.8 Significance 16
1.9 Delimitations 17
1.10 Limitations 17
1.11 References 18
CHAPTER 2 20
2.1 Introduction 20
2.2 Methods and Measurement 23
2.2.1 *Participants* 23
2.2.2 *Experimental Protocol* 24
2.2.3 *Motion Analysis* 25
2.2.4 *Statistical Analyses* 28
2.3 Results 28
2.4 Discussion 31
2.5 Summary 33
2.6 References 35
CHAPTER 3 37
3.1 Introduction 37
3.2 Methods and Measurement 40
3.2.1 *Participants* 40
3.2.2 *Experimental Protocol* 41
3.2.3 *Statistical Analyses* 44
3.3 Results 45

3.4 Discussion 46
3.5 Summary 49
3.6 References 50
CHAPTER 4 54
4.1 Introduction 54
4.2 Methods and Measurements 57
4.2.1 *Participants* 57
4.2.2 *Experimental Protocol* 59
4.2.3 *Motion Analysis* 62
4.2.4 *Data Reduction* 64
4.2.5 *Statistical Analysis* 65
4.3 Results 65
4.3.1 *Transverse Plane Leg and Foot Mechanics* 72
4.3.2 *Ankle Plantar Flexor, Knee Extensor and Hip Extensor Net Joint Moments* 77
4.3.3 *Frontal Plane Leg Mechanics* 78
4.4 Discussion 78
4.5 Summary 85
4.6 References 87
CHAPTER 5 91
5.1 Introduction 91
5.2 Tibial Torsion Influences Non-sagittal Motion 91
5.3 Tibial torsion and ACL Injury 93
5.4 Tibial Torsion Influences Non-sagittal Motion in Women with ACL Injury 95
5.5 Limitations 96
5.6 Practical Applications and Future Directions 99
5.7 References 101

List of Tables

Table 4.1 Kinematic and kinetic comparisons for weight bearing partial squat........................... 67

Table 4.2 Kinematic and kinetic comparisons for two-foot jump landing 68

Table 4.3 Kinematic and kinetic comparisons for involved limb step off box landing................ 69

Table 4.4 Kinematic and kinetic comparisons for non-involved limb step off box landing 70

Table 4.5 Hip, knee, ankle internal net joint moments (NJM) for landings 71

List of Figures

Figure 1.1 Diagram showing positive translations and rotations about XYZ axes. 2
Figure 1.2 Diagram showing three possible variations of squat kinematics................................. 4
Figure 1.3 Free body diagram depicting the influence of increased forward leg inclination on knee extensor internal net joint moment. 5
Figure 1.4 Image (medial view) of left foot and distal leg bones 6
Figure 1.5 Diagram of measurement of tibial torsion (θ) 12
Figure 2.1 Diagram of superior view of the tibia with lesser and greater tibial torsion 21
Figure 2.2 Diagram of posterior view of the leg with lesser and greater tibial torsion 22
Figure 2.3 Diagram of superior view of the leg with lesser and greater tibial torsion 23
Figure 2.4 Image showing retro-reflective marker placement and partial squat task. 25
Figure 2.5 Proximally and distally biased leg segments used to calculate tibial torsion angle 26
Figure 2.6 Transverse plane leg rotation as a function of external tibial torsion......................... 29
Figure 2.7 Transverse plane knee rotation as a function of external tibial torsion. 29
Figure 2.8 Transverse plane thigh rotation as a function of external tibial torsion 30
Figure 2.9 Frontal plane thigh rotation measured as a function of external tibial torsion 30
Figure 3.1 Diagram showing the TFA2 (thigh foot angle through second metatarsal) and the TFAM (thigh foot angle through a line perpendicular to the trans-malleolar axis)..................... 42
Figure 3.2 Diagram showing the INC (inclinometer) method of determining tibial torsion. 43
Figure 4.1 Image showing retro-reflective marker placement, partial squat task and positive directions for translations and rotations following the right hand rule. 60
Figure 4.2 Image showing box landing... 61
Figure 4.3 Scatter plot demonstrating left leg transverse plane leg rotation as a function of left leg external tibial torsion 72
Figure 4.4 Scatter plot demonstrating right leg transverse plane leg rotation as a function of right leg external tibial torsion. 72
Figure 4.5 Scatter plot demonstrating left foot progression angle as a function of left leg external tibial torsion 73
Figure 4.6 Scatter plot demonstrating right foot progression angle as a function of right leg external tibial torsion 73

Figure 4.7 Scatter plot demonstrating left ankle plantar flexor net joint moment as a function of left leg external tibial torsion. 74
Figure 4.8 Scatter plot demonstrating right ankle plantar flexor net joint moment as a function of right leg external tibial torsion 74
Figure 4.9 Scatter plot demonstrating left leg peak abduction angle as a function of left leg external tibial torsion 75
Figure 4.10 Scatter plot demonstrating right leg peak abduction angle as a function of right external tibial torsion 75

CHAPTER 1

Literature Review

1.1 Introduction

Lower extremity segment movement patterns are complex and interrelated. Lower limb alignment and musculoskeletal injury are understood to be interactive, but these reports are sometimes conflicting. The ultimate goal of this project was to provide observational evidence of the effect of tibial torsion, as an anatomic variation, on lower body segment kinematics. The author was interested in the lower limb variations that may occur during static and dynamic human movement tasks. The research project investigated how this variation may influence kinematic and kinetic parameters in three-dimensional segment motion. Particular consideration was given to movements that may present a potential risk to the integrity of the anterior cruciate ligament (ACL) in women.

1.2 Segment rotation and joint motion

The evaluation of three-dimensional (3D) joint kinematics in biomechanics research entails the description of the position and orientation of one bony segment relative to another segment. Additionally, 3D joint kinetics describe the forces (and torques) causing relative segment translations or rotations. For example, the interpretation and understanding of the knee joint comes from understanding the movement of the thigh and leg segments during the movement of interest. The human body is therefore described as a series of linked (joints) rigid bodies (segments), where each segment may move independently in 3D space. Each segment is considered to have six degrees of freedom: three linear, along the X, Y, and Z axes and three rotational about those same axes (1). The cross product of the +X and +Y axes is the +Z axis. The right hand rule states that if the thumb of the right hand points in the positive direction of an axis, the fingers will curl around the axis in the direction of positive rotation about that axis. All movements are translations along and/or rotations about these coordinate axes. Conventions for this project place the X (medio-lateral) and Y (anterior-posterior) axes on the horizontal plane, and the Z axis (superior-inferior) is vertical. Figure 1.1 depicts the positive conventions assumed for this research in terms of translation and rotation, following the right hand rule.

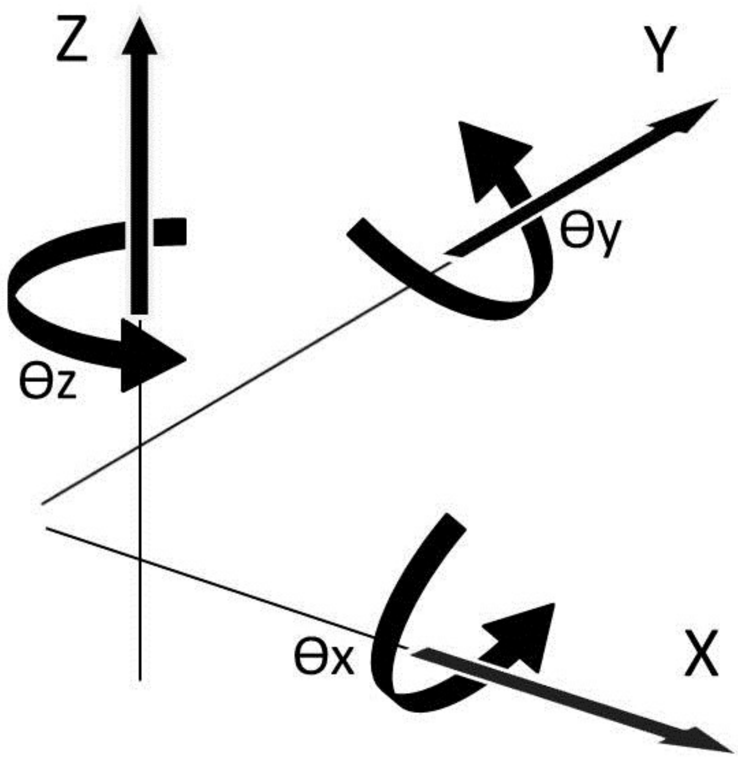

Figure 1.1 Diagram showing positive translations and rotations about XYZ axes.

Bony segments are linked by joints. The definition of a joint, however, is different in anatomy when compared to a mechanical definition. Anatomically speaking, a joint is most often defined as the union or junction where two or more bones articulate. It is often described as the area between bones or rigid parts of the skeleton and is treated as its own entity (2). In mechanics, a joint is defined as the connection between two (or more) rigid bodies that may impose constraints on the involved segments' relative motion. The joint itself is the connection, or link maintaining contact, between the segments. This distinction is important because it is the segments that move about the joint (as in the mechanical definition) that is most important, as that is what ultimately determines the kinematics of the joint. The relative joint angle is not an independent value, but an indirect result of the segments' motion. In practice, joint angular motion is defined as the resultant net angular displacement between two segments (1). Therefore, joint rotations or joint angle measurements, are in fact the rotation of one segment relative to a second segment. These joint angles are often reported in biomechanics research, without consideration of the rotation of all constituent segments. This method of determining joint angles usually designates the second, often distal, segment as a reference segment. This determination assumes the distal segment is fixed or non-moving (3). However, in most human movement, it is plausible that the reference segment moves as well, therefore rotations of both segments may contribute to joint angle (4). To accurately interpret joint rotations, it is important to evaluate the rotations of both segments, which may provide a greater understanding of movement than by investigating joint rotations alone.

This can be observed in a 2D paradigm through a sagittal plane example which demonstrates the need for segment rotations to be included with joint rotations. A common exercise used in resistance-based training is the parallel squat, where the thigh segment flexes about the knee in the sagittal plane until it is parallel with the ground (horizontal). Sometimes used synonymously to describe a parallel squat is a ninety degree (90^{o}) squat. This refers to the

90^o of knee flexion required to accomplish a parallel thigh segment, which can only be possible if the leg segment is perfectly vertical. If the leg segment is not vertical, then greater than 90^o of knee flexion is required for every degree of forward leg segment inclination (in the sagittal plane). Figure 1.2 illustrates three different squats where the knee is flexed 90^o. Only one of these is a parallel squat, which can only be achieved if the leg segment does not rotate forward, otherwise greater than 90^o of knee flexion is required, and no longer a 90^o squat. Without knowing the thigh and leg rotations in the sagittal plane, parallel cannot be assumed to be 90^o of knee flexion, nor can a 90^o squat be assumed to be a parallel squat.

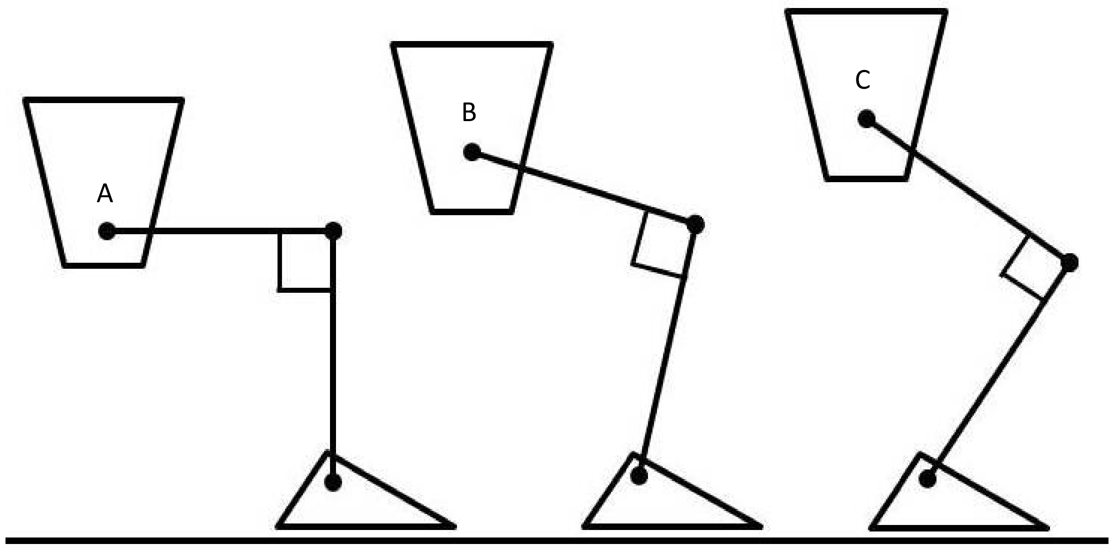

Figure 1.2 Diagram showing three possible variations of squat kinematics. All examples demonstrate 90 degrees of knee flexion (A, B, C), only A demonstrates a parallel squat.

This kinematic interpretation of the joints and relevant segments is especially important when considering the kinetics of the movement task. When calculating the net joint moments acting on a segment, the joint reaction forces are applied to the proximal and distal ends of the segment being analyzed. In most upright or weight bearing tasks, the vertical joint reaction force vector often bears the greatest magnitude. Therefore, as a squat is performed, the leg dorsiflexes forward (about the talocrural joint) and the moment arm for the vertical joint reaction force vector will increase. This increase in moment arm length causes the internal joint reaction torque (about the centre of mass of the leg), to increase; resulting in a greater knee extensor moment. Figure 1.3 illustrates the importance of correctly interpreting the segment motion for kinetics discussion. In these two static situations, the only difference is the amount of forward leg inclination (ten degrees more on the right). This difference results in a greater net joint moment acting on the proximal end of the leg segment. Forward leg inclination and position of the superior segments (thighs, upper body) influence the net joint moment at the knee, not knee joint angle.

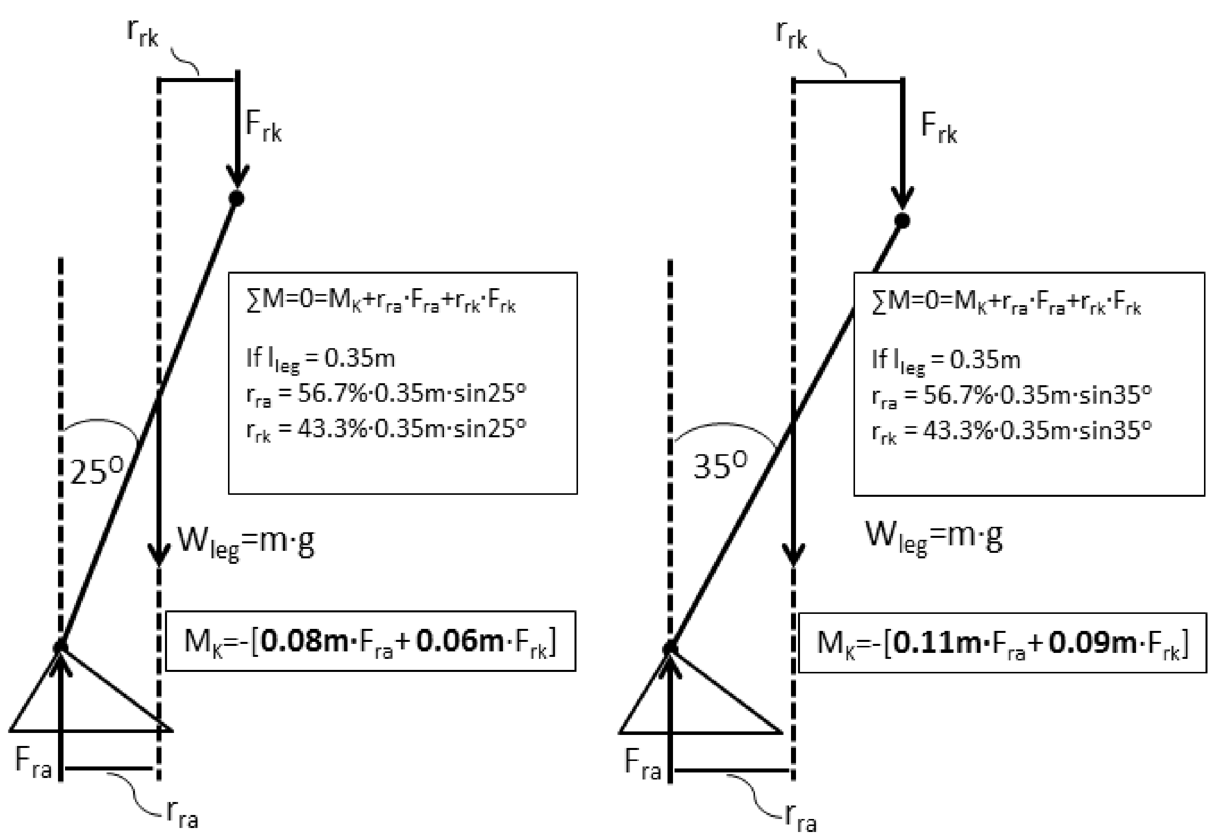

Figure 1.3 Free body diagram depicting the influence of increased forward leg inclination on knee extensor internal net joint moment. Note the greater moment arm lengths for vertical reaction forces on the right when the leg is at 35 degrees of inclination.

1.3 Segmental contributions to sagittal plane ankle motion

Anatomically, the primary rotation at the ankle occurs predominantly at the talocrural joint in the sagittal plane, and is referred to as plantarflexion and dorsiflexion. The secondary rotation occurs predominantly at the subtalar joint in the frontal plane, and is referred to as inversion and eversion. The two posterior bones of the foot are often called the rearfoot, and include the calcaneus and the talus (Figure 1.4). The talocrural joint is a mortise shaped joint where the talus articulates with the tibia and the fibula of the lower leg (referred to as leg in this project). The majority of the forces (90-95%) are acting at the talo-tibial portion of the joint, with the remainder of the forces acting at the talo-fibular portion of the joint (5). Due to the widening of the superior talar surface, this mortise shaped joint provides a bony stability during talocrural dorsiflexion and is limited primarily to sagittal plane motion.

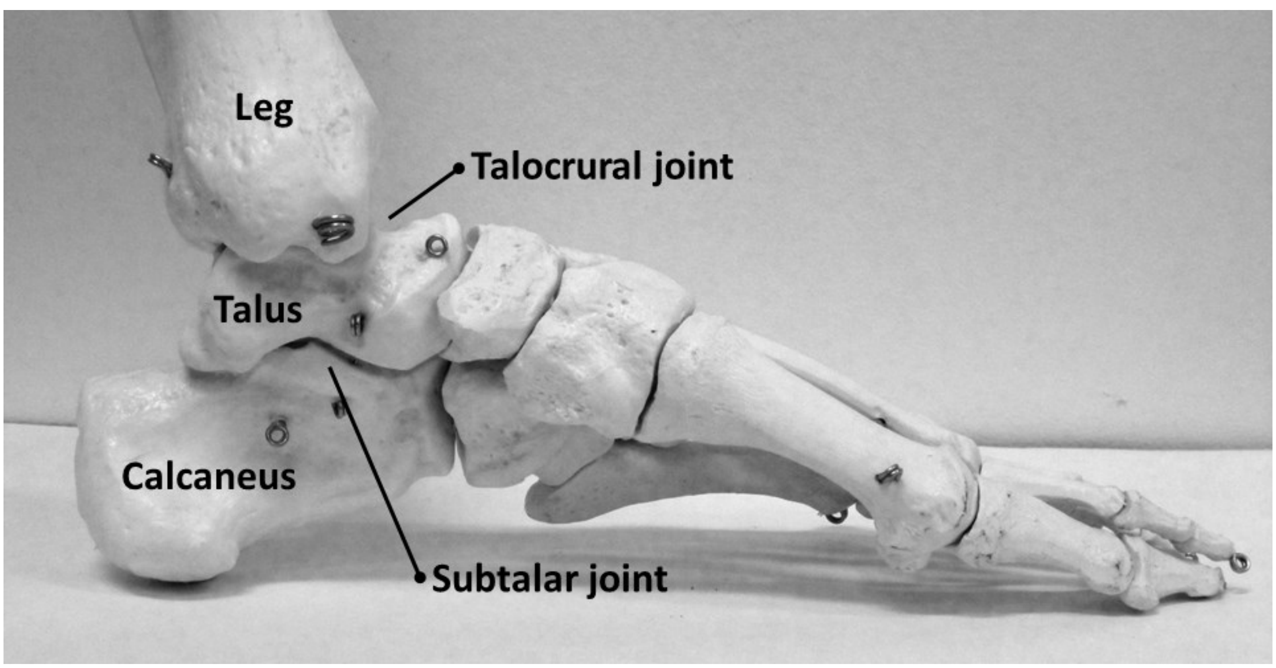

Figure 1.4 Image (medial view) of left foot and distal leg bones; talus and calcaneus inferior to leg segment (tibia and fibula), talocrural joint, and subtalar joint.

In weight bearing tasks, it is assumed that the foot is fixed and non-moving. Therefore, talocrural dorsiflexion is often measured as rotation of the leg relative to the ground. However, numerous investigations have found that variations in loading of bone and muscle in the lower extremity can cause translations and rotations within the foot. Initial research for this project investigated sagittal plane ankle kinematics to quantify the contribution of the leg and calcaneus segments to talocrural dorsiflexion in a partial squatting task (4). This author found the calcaneus segment rotated anteriorly ten degrees (10^{o}) in the sagittal plane (calcaneal plantarflexion) while the leg segment rotated anteriorly 30 degrees (30^{o}) (leg dorsiflexion) during the weight bearing squat. The angle for talocrural dorsiflexion was measured at nineteen degrees (19^{o}) for the task. Neither segment alone can accurately describe the position of the ankle, but when considered together, the actual kinematics occurring can be interpreted correctly. Multiple regression modelling found that in fact, anterior leg rotation was a significant positive contributor, while anterior calcaneus rotation was a significant negative contributor to talocrural dorsiflexion.

This discovery supports the importance of considering segment rotations instead of joint rotations alone when interpreting kinematic data. It is not uncommon to find ankle (talocrural) dorsiflexion angles upwards of 40 degrees (40^{o}) reported in the literature (6). These values occur when the ankle joint is modeled by the leg (tibia and fibula) as the proximal segment and the foot (malleoli to metatarsal heads) as the distal segment. However, the issue here is that this definition of the foot assumes the foot is a rigid structure. Inter-segmental motion between the tarsals and metatarsals may alter the orientation of the talus and the talocrural joint (7). It is well documented in the anatomic literature that the talocrural joint, due to its bony morphology, limits the ankle to approximately 20-25 degrees of dorsiflexion (8). This discrepancy motivated these authors to study the ankle joint using a non-rigid foot model. While the entire foot segment is constrained by the floor in weight bearing motion, movement of the bones within the foot may occur. Therefore, the distal segment of the talocrural joint was modeled using the calcaneus. This method had been previously used by Stacoff et al. (9) while investigating movement coupling at the ankle during running. As the calcaneus rotates anteriorly (calcaneal plantarflexion), its motion re-orients the talar dome, thus allowing the leg segment to dorsiflex further than if no calcaneal motion occurs. This explains how the joint angle, measured as the leg relative to the foot or ground, could regularly exceed the anatomic limitation of the joint.

The calcaneal motion is significant enough, at up to 5-15 degrees in the sagittal plane, to imply that any structure attaching to this bone will be affected by its movement. The largest of these are the calcaneal tendon directing the tensile force of soleus and gastrocnemius to the posterior calcaneus (causing anterior rotation) while the plantar aponeurosis' inferior and anterior attachment on the calcaneus result in negative (or posterior) rotation of the calcaneus when tension in the structure increases (10). Increased tensile stress on the plantar aponeurosis may result in Achilles tendon strain, while repetitive soleus and gastrocnemius contraction may result in plantar fasciae strain (11, 12). Unless segmental contribution is included alongside joint rotations, readers are unable to ascertain the actual kinematics occurring. It is ultimately possible to apply this type of reporting model to all lower body joints and segments in all three cardinal planes of motion. This inclusive and comprehensive type of reporting model will be a constant focus of this research. This leads to another major consideration, which is to evaluate the starting position, or alignment of each segment relative to the next segment prior to analysis. Anatomic alignment may predispose a segment to rotate in a certain direction and/or may affect the magnitude of that motion.

1.4 Non-sagittal movement patterns

The same obstacles that appear in 2D modeling are present when interpreting 3D kinematics. However, the challenges are confounded as there are an increased number of possible rotations. In 2D, only rotations about a single plane, such as sagittal, are considered. In 3D, rotations about the sagittal, frontal and transverse axes must be considered. Human movement is multi-planar, as the actions of muscles may cause rotations in multiple planes. For example, the gluteus maximus extends and externally rotates the thigh segment about the hip as well as assisting in adducting an already abducted thigh (2). Another example is adductor longus, which adducts and flexes the thigh segment about the hip and may assist in returning the thigh to anatomic position after being axially rotated. The multi-planar nature of human movement is captured through some anatomical terminology, such as pronation and supination. Pronation and supination are terms describing foot, ankle and possibly leg motion (8). Pronation results in eversion, dorsiflexion, and abduction of the foot about the ankle joints. Supination is inversion, plantarflexion, and adduction of the foot about the ankle. Segment motion contributes proximally and distally to these measures at the ankle joints.

As human movement is multi-planar, it is possible for the same resultant rotation to be accomplished by different combinations of constituent rotations. This is commonly illustrated using Codman's paradox, which refers to a specific pattern of motion of the humerus or upper arm. In Codman's paradox, axial or transverse plane rotation occurs about the longitudinal axis of the upper arm following two or three sequential arm rotations about the shoulder joint that do not involve rotation about the long axis of the segment (13). To address this paradox, human motion can be described using a method of quantifying angular positions using independent angles known as Cardan/Euler angles (14). These angles are measured by an ordered sequence of XYZ rotations. For example, if an XYZ Cardan sequence is chosen, the modeled segment rotates about the X axis by a given angle, then rotates about a rotated Y axis, and then rotates about a twice rotated Z axis. For any given movement pattern, different Cardan sequences may influence the angular calculations, and therefore interpretation of these rotations (15). Although used extensively in the field of biomechanics and the analysis of human motion, the effect of altering the sequence of rotations has not been fully investigated in this field (16).

For joint rotations, the proximal or distal segment must be chosen to be the moving segment while the other segment is considered to be fixed. This expands the number of Cardan/Euler rotations to twelve for a given joint. As the proximal and distal segments of a joint can rotate independently in all three planes, the selection of which segment is fixed is critical (17). The greatest challenge in describing joint rotations is when both segments rotate, since it is difficult to interpret a single measurement value when reporting the data. One solution is to describe the rotations of both segments forming a joint (4). A further step may be to investigate the possibility that segment movement is nearly always multi-planar, with sagittal movement and alignment influencing non-sagittal movement and alignment of each segment involved in the joint rotation.

1.5 Tibial torsion and knee joint pathology

Tibial torsion is the rotational alignment offset between the proximal and distal ends of the leg, generated by a transverse plane twisting of the collective tibia and fibula from proximal to distal. In late fetal development and in infants, the feet may be turned inward and appear to be "pigeon-toed", representing internal tibial torsion (18). As part of normal growth and development, tibial torsion changes from internal to external between infanthood and adolescence (19). In the adult population, the lateral malleolus is almost invariably posterior and inferior to the medial malleolus when standing with feet facing forward. When referenced to a line made between the proximal tibial condylar prominences, the angle of the line between the malleoli represents external tibial torsion (Figure 1.5). The normal range of angles for tibial torsion in adults is approximately five to thirty degrees, and values exceeding thirty degrees have been suggested to be excessive (20). These absolute values may vary depending on the anatomical landmarks and methods used to determine tibial torsion, although relative variance appears to be consistent across the methods.

Excessive tibial torsion has been linked with abnormal gait mechanics in cerebral palsy (21) and pain in Osgood-Schlatter disease (22). Excessive external tibial torsion may also influence dynamic transverse plane leg rotation in non-clinical populations (23). However, there has been little study of how variation in tibial torsion influences transverse plane leg rotation during weight bearing activities such as squatting or landing from a jump or a pre-determined height. Tibial torsion has been linked to greater external foot progression angles, as well as decreased ability of soleus to extend the knee (20, 24). Our research (Chapter 2) was able to link tibial torsion to frontal plane ankle, transverse plane leg, and transverse plane thigh rotations. This is consistent with the joint motions demonstrated by analyzing the cross-planar segmental contributions. Tibial torsion must be considered as a possible mediator in lower body kinematics and will be investigated further in this project.

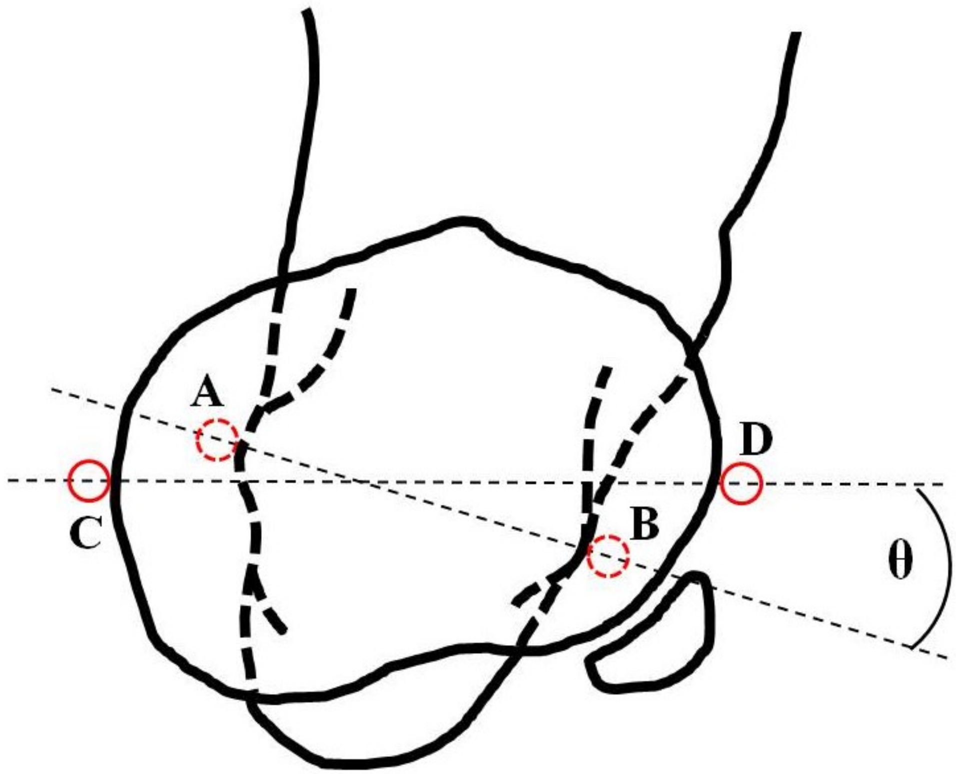

Figure 1.5 Diagram of measurement of tibial torsion (θ) (superior view). Markers/anatomic landmarks: A (medial malleolus), B (lateral malleolus), C (medial tibial condylar prominence), D (lateral tibial condylar prominence).

An interesting crossover is that many of the same factors that have been linked to tibial torsion are also considered when discussing ACL injury. External leg rotation when weight bearing may cause excessive loading of the medial collateral ligament, medial meniscus, and ACL (25, 26). External leg rotation may also change the alignment of the tibial tuberosity (anterior tibia), thereby affecting patellofemoral loading. Furthermore, external leg rotation has been described as a mechanism of non-contact ACL rupture (27). Increased tibial torsion may also decrease or alter the functionality of the soleus muscle (20, 24). Because of its proximal and posterior attachment to the leg, the soleus muscle may act as an ACL protagonist, or play a role in dynamically stabilizing the knee (28) by decreasing anterior tibial rotation (leg dorsiflexion).

If the calcaneus is rotated externally due to excessive external tibial torsion, the distal attachment of the posterior leg muscles would be more medial than without excessive external tibial torsion. These muscles would include: soleus, tibialis posterior, flexor digitorum, and flexor hallucis longus (2). This would not only decrease the posteriorly directed force vector of these muscles, but would likely increase the medially directed force vector, thus contributing to medially directed frontal plane motion of the leg (leg inversion) (24). Leg inversion, or knee valgus, is often reported as a prominent ACL risk factor. ACL research from this project (Chapter 3) supports this speculation, whereby tibial torsion angles were found to be greater in ACL injured participants over control subjects.

1.6 Summary

Segmental contributions to joint motion may be affected by abnormalities, disorders, injury, disease, or anatomical variations. These may result in non-sagittal segmental motion occurring during predominantly sagittal lower body joint rotations. External tibial torsion is a possible explanation for certain uncharacteristic or non-sagittal segmental contributions to joint motion. This research has identified similarities between ACL injury risk factors and certain kinematics that are linked to individuals demonstrating greater external tibial torsion. It is necessary to consider external tibial torsion values when investigating the mechanics of ACL injured individuals versus healthy participants; therefore, further research is warranted.

1.7 Purposes and Hypotheses

Study 1 (Chapter 2): **Tibial torsion influences non-sagittal leg kinematics during partial squatting.**

The primary purpose of this investigation was to determine whether tibial torsion influences non-sagittal plane lower extremity segment and joint rotations, particularly the direction of frontal and transverse plane leg rotation in physically active men and women. This investigation sought to evaluate whether an increase in external tibial torsion would influence the magnitude and direction of leg segment rotation in the frontal and transverse planes.

It was hypothesized that individuals with greater external tibial torsion would demonstrate an increase in frontal plane leg inversion (knee valgus). It was also hypothesized that an increase in external tibial torsion would be associated to an increase in transverse plane external leg rotation (negative rotation determined by the right hand rule).

Study 2 (Chapter 3): **An examination of tibial torsion measurement reliability and a comparison of external tibial torsion angles between women with uninjured and previously injured anterior cruciate ligaments.**

The primary purpose of this study was to examine the intra-rater and inter-rater reliability of four different methods of measuring external tibial torsion. The secondary purpose of this study was to compare external tibial torsion angles in physically active women with and without ACL injury, as well as to compare tibial torsion between the involved and non-involved limbs of the individuals with ACL injury.

It was hypothesized that there would be no difference between the four methods or the two observers. It was also hypothesized that external tibial torsion angle would be greater in individuals with ACL injury than in uninjured control participants. Further to the secondary hypothesis, it was hypothesized that external tibial torsion angle would be greater in the involved limb of individuals with ACL injury when compared to the non-involved limb.

Study 3 (Chapter 4): **Tibial torsion influences non-sagittal lower body joint kinematics and kinetics in women with ACL injury during squatting and landing tasks.**

The purpose of this study was to determine whether lower extremity mechanics differed between women with healthy and damaged ACLs in a pattern consistent with the functional consequences of varying degrees of external tibial torsion.

Three main hypotheses were considered for study participants with greater external tibial torsion. Each of these also proposed a similar corollary hypothesis for ACL injured participants. It was hypothesized that participants with greater external tibial torsion (high torsion control participants) would demonstrate: 1) increased external leg rotation (or decreased internal leg rotation) during squat tasks, and greater foot progression angles at ground contact during landing tasks; 2) decreased internal ankle plantarflexor net joint moments during all tasks; and 3) increased net joint moments acting to invert the leg segment and increased leg segment abduction angles. The corollary hypotheses for participants with ACL injury were identical to the three supposed for participants with greater external tibial torsion when compared to participants with lesser external tibial torsion (low torsion control participants).

1.8 Significance

Evaluating sagittal and non-sagittal segmental contributions to joint motion along with tibial torsion may shed light on unresolved issues. Excess tibial torsion may result in a reduction of the internal plantarflexor net joint moments in the sagittal plane, which may affect the capacity of the associated muscles to perform movement. If there is a cross-planar contribution to joint motion, instead of only in the sagittal plane, then it is possible that frontal or transverse plane segmental and joint motion occurs as a result of altered musculoskeletal alignment. Furthermore, it is possible that excess tibial torsion alters the dynamic interaction between the muscles involved and the skeletal system during tasks such as squatting or landing from a height. The multi-articular nature of the human body means that tibial torsion, a transverse plane misalignment of the lower leg, is a possible contributor to altering movement in other planes as well as more proximal joints, like the knee and hip.

If there is evidence to support the congruence between the kinematics associated with increased tibial torsion and those known to be contributors to ACL injury, then it may provide a previously unidentified link to non-contact ACL rupture. The studies in this thesis will investigate tibial torsion as a potential mediator of transverse plane leg rotation and cross-planar segmental contributions during a partial squat, evaluate inter-rater consistency of tibial torsion measurements and compare tibial torsion angles between women with and without ACL injury, and investigate the influence of tibial torsion and muscular effort on cross-planar segmental contributions to lower body joint motion in women with and without ACL injury during squatting and landing tasks.

1.9 Delimitations

Participants between the ages of 18 and 35 years in good health without any medical conditions that may confound the movement patterns in the lower body were recruited in all three studies. Study 1 involved men and women without history of lower body surgery or current serious musculoskeletal injury to the feet, ankles, knees, or hips. Study 2 involved only women and included athletic individuals with surgically repaired ACLs. Study 3 also involved only women and included minimum six month post-operative athletic individuals with unilateral ACL rupture repaired by hamstring graft only. In all three experiments, practice and collection trials were consistent, as well as investigator dialogue used for prompting activity by the participants. Consistency was maintained, with respect to both the investigator(s) conducting the data collection and with the equipment and protocols used in order to maximize validity and reliability.

1.10 Limitations

Inherent limitations within the studies include the absence of controlling height and weight of active participants, and physical activity controls for the participants prior to meeting with the investigator for data collection. Furthermore, particularly for maximal effort or range of motion tasks, it was difficult to ascertain whether physical or psychological factors may have affected the participants' performances. Another limitation for the ACL injured participants was the lack of restriction on individual pre- and post-operative training or rehabilitation that may have affected the observed results.

1.11 References

1. Robertson DGE. Research methods in biomechanics. Champaign, IL: Human Kinetics; 2004.

2. Moore KL, Dalley AF, Agur AMR. Clinically oriented anatomy. 6th ed. Philadelphia: Wolters Kluwer/Lippincott Williams & Wilkins; 2010.

3. Özkaya N, Nordin M. Fundamentals of biomechanics: equilibrium, motion, and deformation. 2nd ed. New York: Springer; 1999.

4. Chizewski MG, Chiu LZ. Contribution of calcaneal and leg segment rotations to ankle joint dorsiflexion in a weight-bearing task. Gait Posture. 2012 Feb 13.

5. Neumann DA. Kinesiology of the musculoskeletal system: foundations for rehabilitation. 2nd ed. St. Louis, Mo.: Mosby/Elsevier; 2010.

6. Hemmerich A, Brown H, Smith S, Marthandam SS, Wyss UP. Hip, knee, and ankle kinematics of high range of motion activities of daily living. J Orthop Res. 2006 Apr;24(4):770-81.

7. Lundberg A. Kinematics of the ankle and foot. In vivo roentgen stereophotogrammetry. Acta Orthop Scand Suppl. 1989;233:1-24.

8. Lundberg A, Goldie I, Kalin B, Selvik G. Kinematics of the ankle/foot complex: plantarflexion and dorsiflexion. Foot Ankle. 1989 Feb;9(4):194-200.

9. Stacoff A, Nigg BM, Reinschmidt C, van den Bogert AJ, Lundberg A, Stussi E, et al. Movement coupling at the ankle during the stance phase of running. Foot Ankle Int. 2000 Mar;21(3):232-9.

10. Gefen A. Stress analysis of the standing foot following surgical plantar fascia release. J Biomech. 2002 May;35(5):629-37.

11. Cheung JT, Zhang M, An KN. Effect of Achilles tendon loading on plantar fascia tension in the standing foot. Clin Biomech (Bristol, Avon). 2006 Feb;21(2):194-203.

12. Cheung JT, Zhang M, An KN. Effects of plantar fascia stiffness on the biomechanical responses of the ankle-foot complex. Clin Biomech (Bristol, Avon). 2004 Oct;19(8):839-46.

13. Cheng PL. Simulation of Codman's paradox reveals a general law of motion. J Biomech. 2006;39(7):1201-7.

14. Schache AG, Wrigley TV, Blanch PD, Starr R, Rath DA, Bennell KL. The effect of differing Cardan angle sequences on three dimensional lumbo-pelvic angular kinematics during running. Med Eng Phys. 2001 Sep;23(7):493-501.

15. Sinclair J, Taylor PJ, Edmundson CJ, Brooks D, Hobbs SJ. Influence of the helical and six available Cardan sequences on 3D ankle joint kinematic parameters. Sports Biomech. 2012 Sep;11(3):430-7.

16. Crawford NR, Yamaguchi GT, Dickman CA. A new technique for determining 3-D joint angles: the tilt/twist method. Clin Biomech (Bristol, Avon). 1999 Mar;14(3):153-65.

17. Tupling SJ, Pierrynowski MR. Use of cardan angles to locate rigid bodies in three-dimensional space. Med Biol Eng Comput. 1987 Sep;25(5):527-32.

18. Hutter CG,Jr, Scott W. Tibial torsion. J Bone Joint Surg Am. 1949 Jul;31A(3):511-8.

19. Kristiansen LP, Gunderson RB, Steen H, Reikeras O. The normal development of tibial torsion. Skeletal Radiol. 2001 Sep;30(9):519-22.

20. Hicks J, Arnold A, Anderson F, Schwartz M, Delp S. The effect of excessive tibial torsion on the capacity of muscles to extend the hip and knee during single-limb stance. Gait and Posture. 2007;26(4):546-52.

21. Turner MS. The association between tibial torsion and knee joint pathology. Clin Orthop Relat Res. 1994 May;(302)(302):47-51.

22. Gigante A, Bevilacqua C, Bonetti MG, Greco F. Increased external tibial torsion in Osgood-Schlatter disease. Acta Orthop Scand. 2003;74(4):431-6.

23. MacWilliams BA, McMulkin ML, Baird GO, Stevens PM. Distal Tibial Rotation Osteotomies Normalize Frontal Plane Knee Moments. J Bone Joint Surg Am. 2010 December 1;92(17):2835-42.

24. Schwartz M, Lakin G. The effect of tibial torsion on the dynamic function of the soleus during gait. Gait Posture. 2003 4;17(2):113-8.

25. Fleming BC, Renstrom PA, Ohlen G, Johnson RJ, Peura GD, Beynnon BD, et al. The gastrocnemius muscle is an antagonist of the anterior cruciate ligament. J Orthop Res. 2001 Nov;19(6):1178-84.

26. Fleming BC, Renstrom PA, Beynnon BD, Engstrom B, Peura GD, Badger GJ, et al. The effect of weightbearing and external loading on anterior cruciate ligament strain. J Biomech. 2001 Feb;34(2):163-70.

27. Park H-, Wilson NA, Zhang L-. Gender differences in passive knee biomechanical properties in tibial rotation. Journal of Orthopaedic Research. 2008;26(7):937-44.

28. Sherbondy PS, Queale WS, McFarland EG, Mizuno Y, Cosgarea AJ. Soleus and gastrocnemius muscle loading decreases anterior tibial translation in anterior cruciate ligament intact and deficient knees. J Knee Surg. 2003 Jul;16(3):152-8.

CHAPTER 2

Tibial torsion influences non-sagittal leg segment kinematics during partial squatting

2.1 Introduction

Limb segments exhibit structural variations which may alter muscle forces and this presents a challenge in evaluating segment and joint rotations. In the leg segment, one structural variation is external tibial torsion (1-3). External tibial torsion is the rotational alignment offset between the proximal and distal ends of the leg in the transverse plane (4), generated by a predominantly distal twisting of the collective tibia and fibula. The potential for coronal plane irregularities (knee varus and valgus) to influence leg and knee pathology is well recognized, whereas rotational deformities, like tibial torsion, are often disregarded or unreported (5). Extreme tibial torsion, whether internal or external, may be associated with serious hindrances to normal movement patterns (6). Although tibial torsion is most often evaluated in orthopedic settings, it may also affect the movements of otherwise healthy individuals. Variations in tibial torsion are believed to influence the magnitude and direction of the frontal and transverse plane rotations of the leg segment (7).

The amount of tibial torsion may affect the force vector for the posterior leg muscles, namely: soleus, tibialis posterior, flexor digitorum longus, and flexor hallucis longus. Each of these muscles originates with an attachment on the posterior leg. In a limb with no tibial torsion, the force vector from these muscles is predominantly directed inferior and posterior (8). With increasing external tibial torsion, this force vector would likely change (8). This altered direction of the force vector is due to the lateral twisting of the leg and concomitant change in the position of the malleoli and the calcaneus (due to the morphology of the leg and tarsal bones). It is plausible that external tibial torsion could decrease the posteriorly directed force while increasing the medially directed force vector (Figure 2.1). As a consequence, leg rotations may be affected. Specifically, a reduction in the posterior force would decrease the plantar flexor moment applied to the leg segment. In addition to a reduction in the posterior force, external tibial torsion may reduce the muscles' moment arms (8), further reducing the plantar flexor moment.

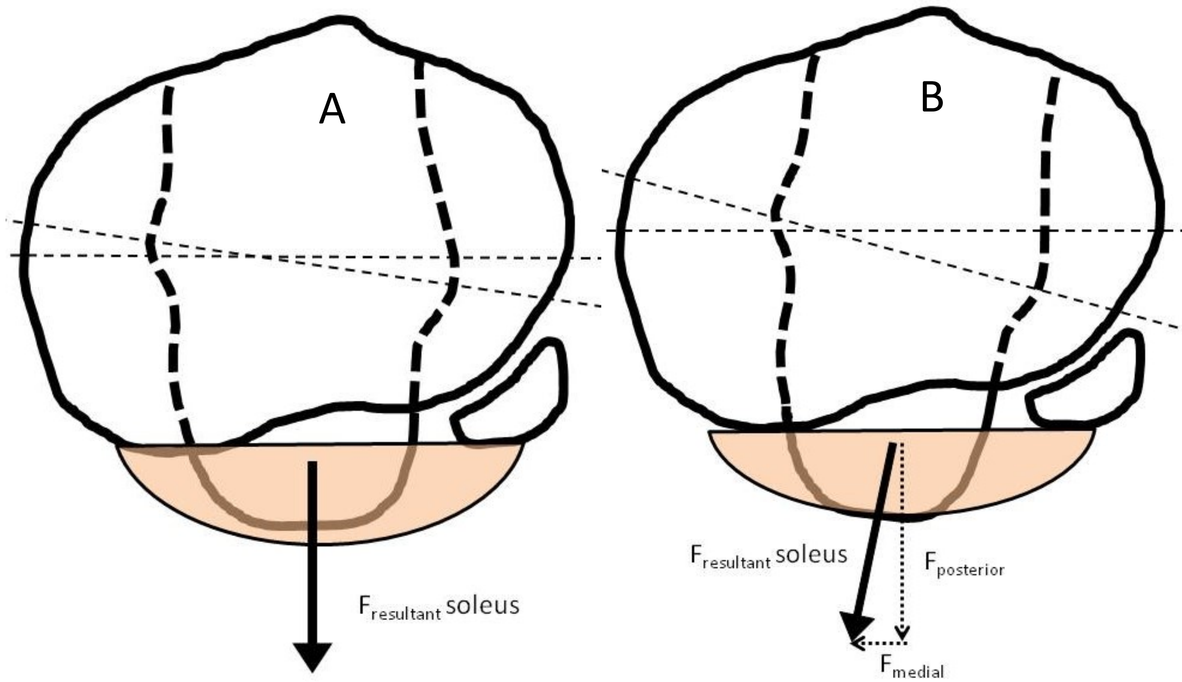

Figure 2.1 From a superior view of the tibia, the diagram shows the mainly posterior force vector of soleus when a lesser degree of tibial torsion is present (A), and the decreased posterior force vector accompanied by a medial force vector when a greater degree of tibial torsion is present (B).

While the posterior directed force is reduced with increased external tibial torsion, the medial directed force increases (8). In the frontal plane, an increase in the medial force acting proximally would cause an ankle inversion moment (Figure 2.2), potentially increasing leg abduction and knee valgus.

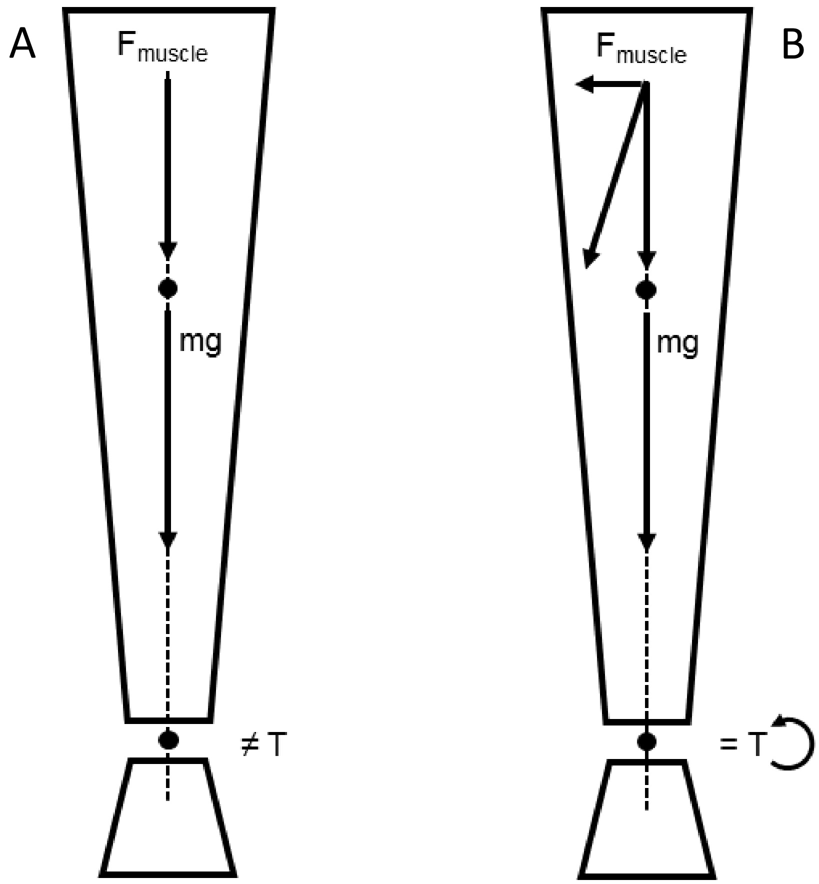

Figure 2.2 From a posterior view of the right leg, the diagram shows the inferior directed force vector of the posterior leg muscles when a lesser degree of tibial torsion is present (A) creating no torque, and the inferior directed force vector accompanied by a medial directed force vector when a greater degree of tibial torsion is present (B) creating a knee abductor moment.

A medial directed force vector could also generate a moment in the transverse plane (27). A superior view of the leg segment (Figure 2.3) illustrates that a medial directed force from these muscles has a lateral rotator moment arm. Taken together, greater external tibial torsion may increase leg abduction and lateral rotation. An association between external tibial torsion and the previously described kinematics has been established in clinical populations (5, 9-11), however, this has not been studied in healthy adults.

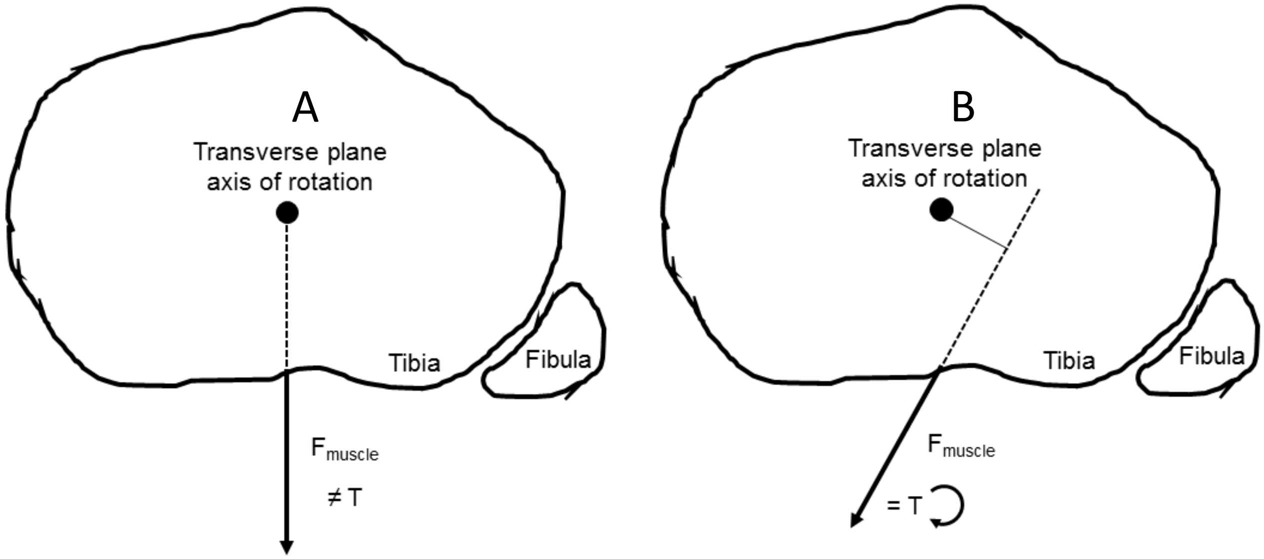

Figure 2.3 From a superior view of the right leg segment, the diagram shows the posterior directed force vector of the posterior leg muscles with a lesser degree of external tibial torsion (A) creating no torque, and the increase in a more medial directed vector when a greater degree of external tibial torsion is present (B) creating a leg lateral rotation moment.

The primary purpose of this research was to examine the association between tibial torsion and non-sagittal plane lower extremity segment and joint rotations during a weight bearing squat. Two primary hypotheses were proposed for this investigation. The first hypothesis was that greater external tibial torsion would be related to increased leg abduction and knee valgus. The second hypothesis was that greater external tibial torsion would be related to increased leg lateral rotation.

2.2 Methods and Measurement

2.2.1 Participants

A non-probability consecutive convenience sample of men (n=14) and women (n=16) participants was recruited from the university population using recruitment flyers on notice boards and electronic mailing lists to Faculty and students in and around the Faculty of Physical Education and Recreation buildings at the University of Alberta. All participants provided written informed consent as approved by a Research Ethics Board at the authors' institution (Ethics ID: Pro00028411). Participants were excluded if they had a self-reported history of lower body segment or joint problems that required medical treatment. Participants were also excluded if they had reconstructive surgical intervention in the lower body. Participants were not excluded if they had pes planus, pes cavus or dynamic flexible flatfoot. Men were 29.4±7.9 years old, 1.78±0.06 m tall, and 77.9±9.2 kg body mass. Women were 25.1±6.0 years old, 1.68±0.07 m tall, and 63.5±6.7 kg body mass.

2.2.2 Experimental Protocol

Participants performed a partial squat task, where they were instructed to reach maximum dorsiflexion without allowing their trunk to flex forward (Figure 2.4; (12)). Participants practiced the task until they were comfortable with the procedures before data collection. The partial squat was selected for this investigation as it allowed foot placement and foot rotational alignment to be controlled. The feet were aligned such that an imaginary line through the center of the heel and the second ray pointed forward. Foot width was standardized by placing the feet such that this imaginary line was equal to the measured inter-distance width of the femoral greater trochanters. For each trial, participants stood motionless for a minimum of two seconds (baseline position) prior to commencing the squatting task, then squatted to their maximum dorsiflexion position and held this position for three to four seconds (squat position). Three consecutive trials were recorded for motion analysis. Retro-reflective markers (9 mm diameter) were placed on the participant's lower body (Figure 2.4) and recorded at 120 Hz using an eight-camera optoelectronic motion capture system (Qualisys ProReflex MCU240; Qualisys, Sweden).

2.2.3 Motion Analysis

A six-degree-of-freedom marker set was used, involving calibration markers to identify the proximal and distal ends of segments during a standing calibration trial and tracking markers for the dynamic squat task (Figure 2.4). Data processing was performed with Visual 3D (version 4.82.0. C-Motion, Inc., Germantown, MD, USA).

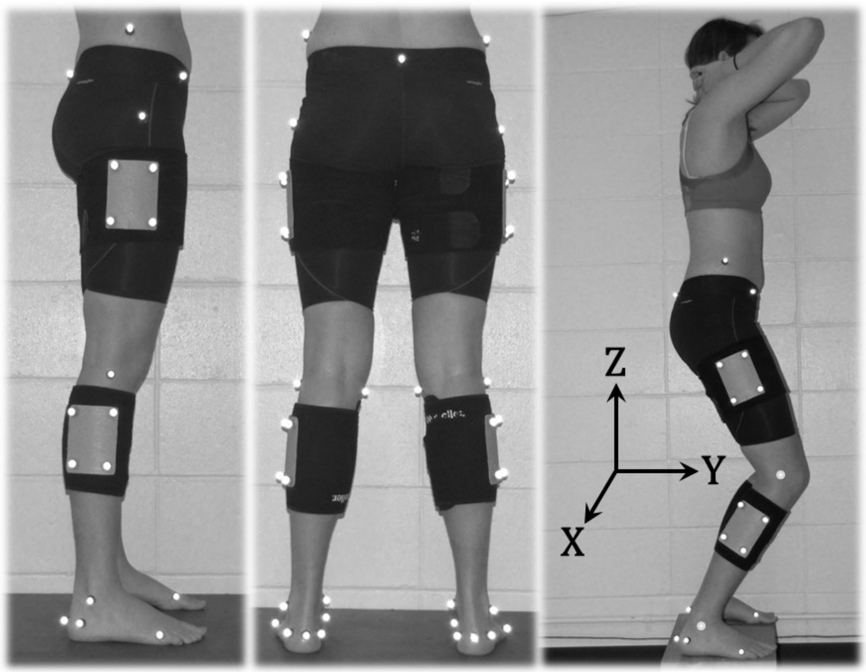

Figure 2.4 Diagram showing retro-reflective marker placement, partial squat task and positive directions for translations, with positive directions for rotations following the right hand rule.

The leg was modeled two ways: 1) proximal-biased and 2) distal-biased. The proximal-biased leg was modeled proximally using the medial and lateral tibial condyle markers, and distally as the midpoint between the medial and lateral malleoli. The use of a virtual marker at the distal leg segment mid-point was required for analysis to remove the influence of tibial torsion on the leg segment coordinate system. The distal-biased leg was modeled proximally using the mid-point between the tibial condyle markers and distally using the medial and lateral malleoli markers. The longitudinal axes (Z) of the segments ran from the proximal to distal ends of the segment. The transverse (X) and sagittal (Y) axes were orthogonal to each other and to the longitudinal axis. The local coordinate system for each segment was located at the proximal end of the segment. The angle about the Z-axis between the proximal- and distal-biased legs was used to calculate the measure of tibial torsion (Figure 2.5). Only the proximal-biased leg was used for kinematic analysis of the squat task. A cluster of four markers on a molded thermoplastic plate, mounted on an elastic fabric wrap, was used to track the leg segment during squatting.

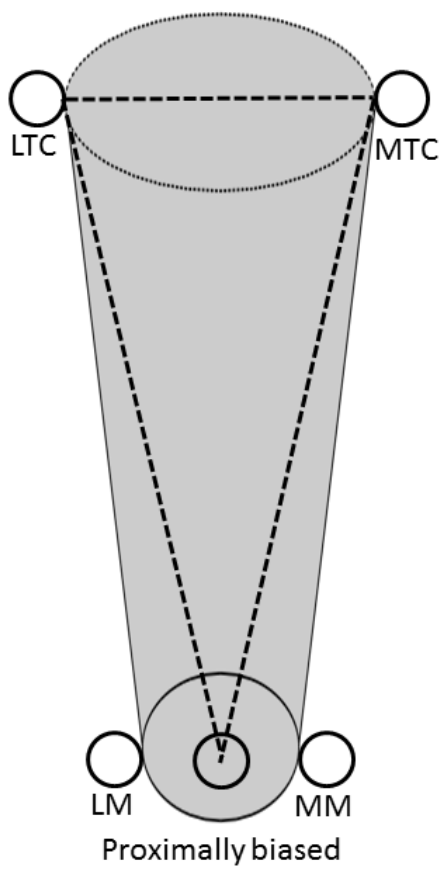

Figure 2.5 Proximally and distally biased leg segments (anterior view of right leg) used to calculate tibial torsion angle; LTC = lateral tibial condyle, MTC = medial tibial condyle, LM = lateral malleolus, MM = medial malleolus.

The thigh segment was modeled proximally using the femoral greater trochanter markers. The proximal parameters of the thigh were considered to be one quarter the distance from the ipsilateral to the contralateral marker. The distal parameters of the thigh were modeled as the mid-point between markers on the medial and lateral tibial condylar prominences. A cluster of four markers fixed to a molded thermoplastic plate was used to track the thigh during the squat task.

Segment rotations were determined using a ZYX Cardan sequence relative to the laboratory reference frame to calculate mechanical rotations consistent with anatomically defined sagittal (X), frontal (Y) and transverse (Z) planes (13). Joint rotations were determined using an XYZ Cardan sequence with the distal segment as the reference segment. All coordinate systems conformed to the right hand rule (Figure 2.4). For left limb data the signs for rotations about the Y and Z axes were reversed to conform to positive and negative continuum of the right limbs. Rotations of segments and joints were determined by the mean value in the squat position compared to the mean value in the standing (baseline) position. Positive segment rotations are plantarflexion/extension (X), eversion/adduction (Y) and medial rotation (Z) [leg]. Positive joint rotations are plantarflexion (X), eversion (Y) and lateral rotation (Z) [ankle]; and flexion (X), abduction (Y) and lateral rotation (Z) [knee].

The angle of the leg and thigh segments, and knee joint were determined prior to initiating the squat and at peak dorsiflexion. Segment and joint angular excursions were calculated as the angle at the bottom of the squat minus the angle prior to squatting. Three trials were recorded and data were averaged between trials, as well as between left and right limbs for analysis.

2.2.4 Statistical Analyses

To determine the association between external tibial torsion with segment and joint rotations, Pearson product moment correlations were used. Beta coefficients (slope of regression lines) were also analyzed as a measure of the influence of external tibial torsion on segment and joint rotations. Statistical calculations were performed in SPSS (version 11.0; SPSS Inc. Chicago, IL), and alpha was set *a priori* (α=0.05).

2.3 Results

External tibial torsion was significantly correlated with transverse plane leg (r=-0.678; r^2=0.459; *b*=-0.477; *p*<0.001; Figure 2.6), knee (r=0.512; r^2=0.262; *b*=0.301; *p*=0.004; Figure 2.7), and thigh (r=-0.398; r^2=0.158; *b*=-0.156; *p*=0.03; Figure 2.8) rotations. External tibial torsion was also significantly correlated with frontal plane thigh (r=0.485; r^2=0.235; *b*=0.193; *p*=0.007; Figure 2.9), but not frontal plane leg or knee rotations.

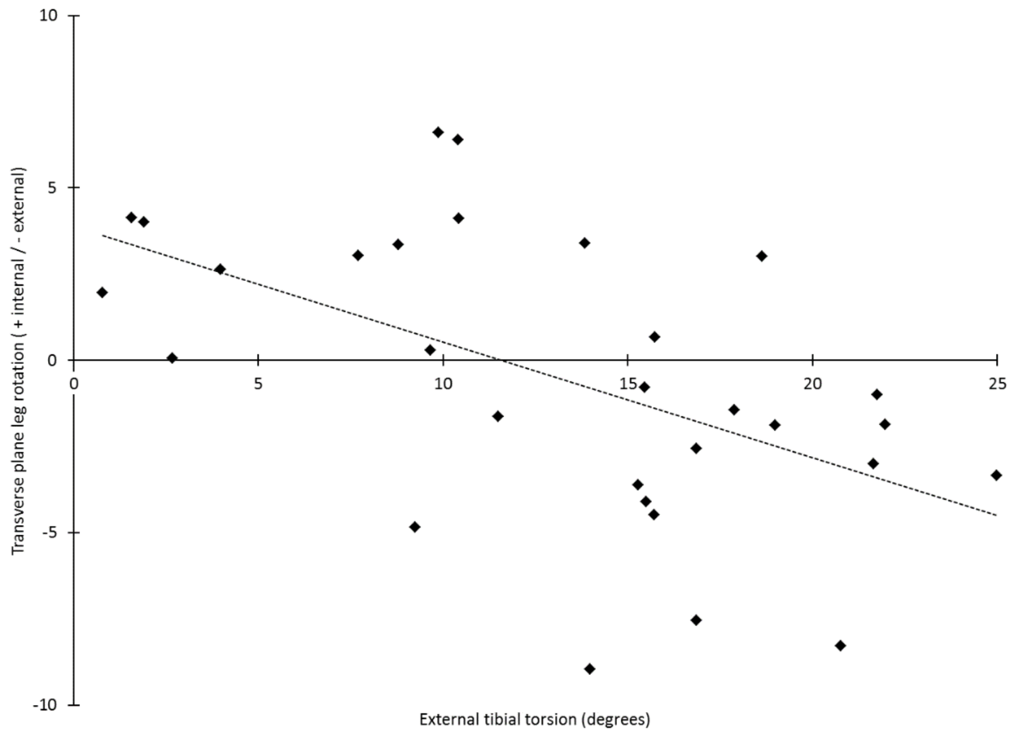

Figure 2.6 Transverse plane leg rotation measured in degrees (+ internal / - external) as a function of external tibial torsion (degrees); r=-0.678; r^2=0.459; *b*=-0.477; *p*<0.001.

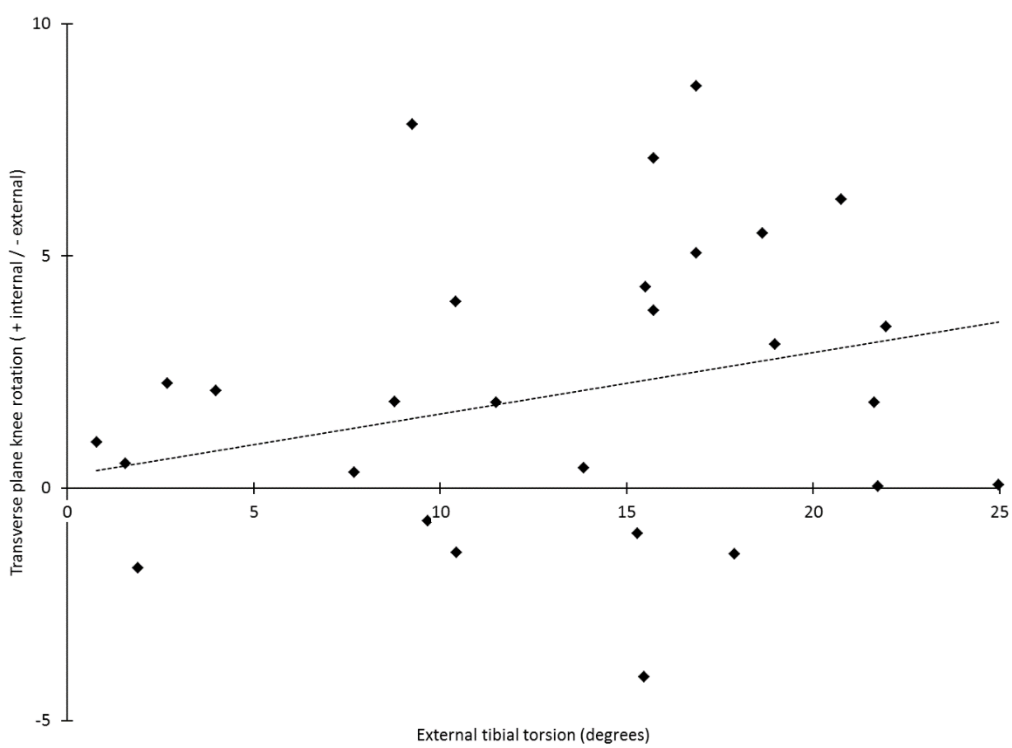

Figure 2.7 Transverse plane knee rotation measured in degrees (+ internal / - external) as a function of external tibial torsion (degrees); r=0.512; r^2=0.262; *b*=0.301; *p*=0.004.

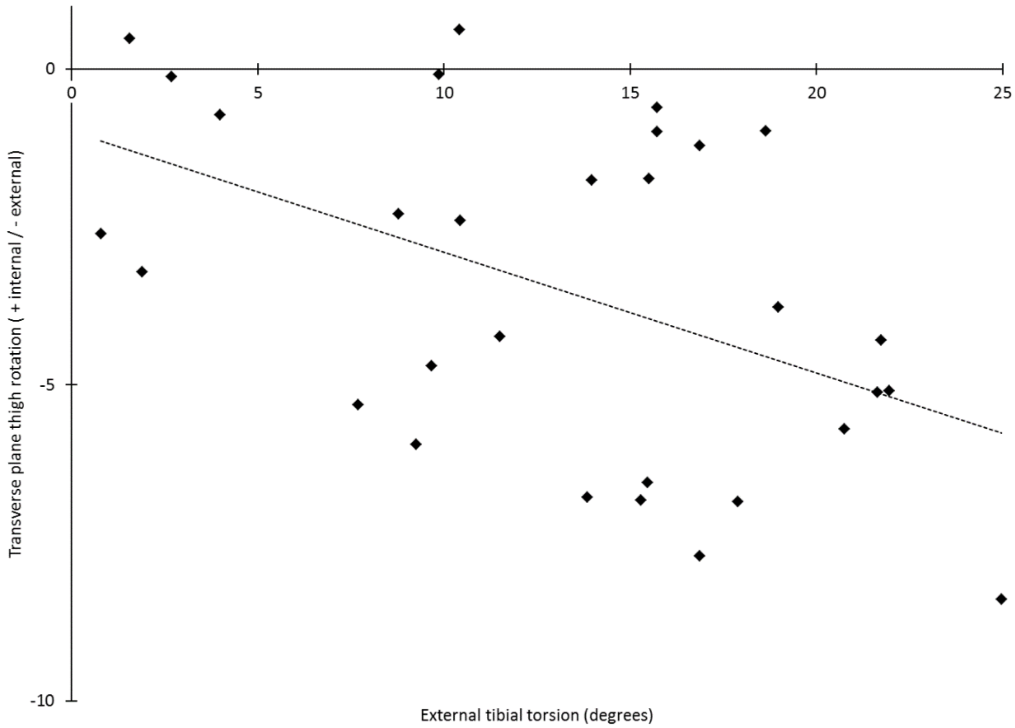

Figure 2.8 Transverse plane thigh rotation measured in degrees (+ internal / - external) as a function of external tibial torsion (degrees); r=-0.398; r^2=0.158; *b*=-0.156; *p*=0.03.

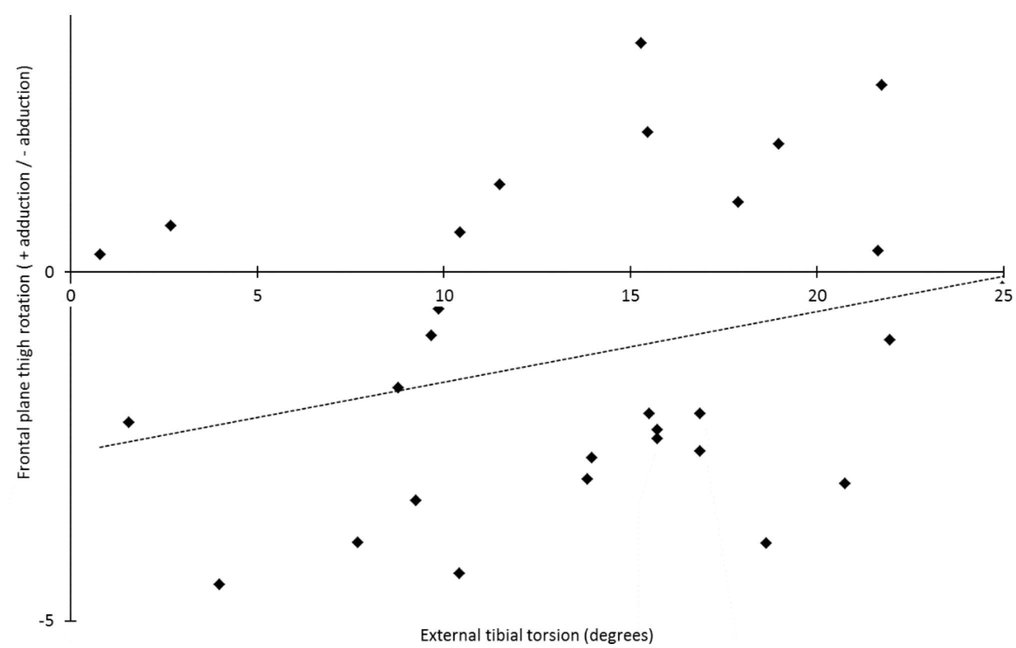

Figure 2.9 Frontal plane thigh rotation measured in degrees (+ adduction / - abduction) as a function of external tibial torsion (degrees); r=0.485; r^2=0.235; *b*=0.193; *p*=0.007.

2.4 Discussion

The most important finding of this investigation is that the magnitude of external tibial torsion was associated with transverse plane leg angular excursion. As external tibial torsion increases, there is a strong negative correlation (r=-0.678) to leg rotation (if medial rotation is positive and lateral rotation is negative). Individuals with greater external tibial torsion had either lesser medial or greater lateral rotation of the leg when squatting, and 46% of the variance in leg rotation could be explained by external tibial torsion angle. This finding supported our second hypothesis that proposed a relation between increasing external tibial torsion and transverse plane leg rotation, which may be explained by a shift from a posterior- to medial-directed force vector for the posterior leg muscles (soleus, tibialis posterior, flexor digitorum, and flexor hallucis longus). Slope analysis revealed that for every degree increase in external tibial torsion, lateral leg rotation increased by 0.48 degrees, which is of strong practical significance. This data suggests the magnitude of tibial torsion influences transverse plane leg segment rotation. This is supported by research by Stefko et al. (14), who found tibial de-rotation surgery (decreasing external tibial torsion angles) in children with cerebral palsy altered the direction of leg rotation during gait. Prior to surgery, a large external tibial torsion angle was exhibited and associated with leg lateral rotation. Following surgery, external tibial torsion was reduced and the leg medially rotated. These results provide evidence that tibial torsion may influence non-sagittal joint rotations in healthy individuals performing low intensity movement tasks. This is the first investigation to associate tibial torsion with differences in lower limb kinematics for healthy, uninjured adult participants. Previous studies have demonstrated the association between tibial torsion and patellofemoral instability (5, 15, 16), overuse injuries (17), and anterior knee pain (18, 19).

Thus, greater external tibial torsion appears to influence transverse plane leg in non-clinical populations. This effect of external tibial torsion to alter lower extremity kinematics may lead to movement dysfunction and injury. Excessive tibial torsion has been associated with abnormal gait mechanics in cerebral palsy (20), pain in Osgood-Schlatter disease (21) and development of tibio-femoral osteoarthritis (22). Findings of this investigation identify important implications for sagittal and non-sagittal plane movement. Numerous injuries are associated with mal-rotations of the frontal and transverse plane leg and knee. These include posterior tibial tendon dysfunction (23); and meniscal tear, anterior cruciate ligament rupture, patellofemoral pain syndrome (24) and Osgood-Schlatter disease (21). Further research is warranted to determine if tibial torsion contributes to these injuries and identifying the specific mechanisms of this contribution. The current results, and previous studies, suggest that tibial torsion be considered as a structural factor that may predispose individuals to lower extremity musculoskeletal injury.

There was little evidence to support the primary hypothesis of the study, which suggested that external tibial torsion would be associated with frontal plane leg and concomitant knee excursions. Although frontal plane leg and knee angular excursions were not related to tibial torsion, frontal plane thigh rotation was found to be related to increased external tibial torsion. Increased magnitudes of external tibial torsion were associated to increased frontal plane thigh adduction. Since proximal and distal segments contribute to joint motion, thigh adduction could potentially influence knee abduction angle, or knee valgus. A relation between external tibial torsion and knee abduction was hypothesized, but as a result of leg abduction rather than thigh adduction. Thus, the relation between external tibial torsion and thigh adduction was not expected and the mechanisms underlying this relation are not known.

Leg abduction, thigh adduction, or simultaneous occurrence of both, are principle variables contributing to dynamic knee valgus in studies of knee injury, including anterior cruciate ligament rupture (25). There are two challenges in understanding dynamic knee valgus. First, the specific segment rotations contributing to dynamic knee valgus are not known. Second, both proximal (i.e. thigh and hip) and distal (i.e. foot and ankle) influences are potential causes of dynamic knee valgus. For example, ankle eversion may be associated with leg abduction and thigh adduction (26). Alternately, hip adduction and thigh internal rotation have been implicated in dynamic knee valgus (27). Although this investigation did not provide evidence to support our first hypothesis, the intensity of the task may be limiting factor. The weight-bearing partial squat may not have been dynamic enough of a task to illicit observable differences in frontal plane motion. Future studies should consider landing from a jump or an elevated platform to further investigate the possible relationship between tibial torsion and non-sagittal plane mechanics.

2.5 Summary

Transverse plane leg and knee rotations during weight-bearing activity were correlated to tibial torsion. These kinematic variations were predicted from free body diagram analyses of the expected changes in soleus and deep posterior leg muscle force vectors as a consequence of increasing external tibial torsion. This finding supports research associating tibial torsion with injury, and highlights that leg segment morphology may influence both ankle and knee joint function. While this anatomical variation has previously been considered in clinical populations, external tibial torsion appears to influence lower extremity kinematics in healthy individuals.

Therefore, further study of external tibial torsion is warranted to understand its role in non-clinical human movement and injury.

2.6 References

1. Schwartz M, Lakin G. The effect of tibial torsion on the dynamic function of the soleus during gait. Gait Posture. 2003 4;17(2):113-8.

2. Song KMMD, Concha MCBS, Haideri NFME. Effects of Lower Limb Torsion on Ankle Kinematic Data During Gait Analysis. Journal of Pediatric Orthopaedics. 2001 November/December;21(6):792-7.

3. Radler C, Kranzl A, Manner HM, Höglinger M, Ganger R, Grill F. Torsional profile versus gait analysis: Consistency between the anatomic torsion and the resulting gait pattern in patients with rotational malalignment of the lower extremity. Gait and Posture. 2010;32(3):405-10.

4. Valmassy R, Stanton B. Tibial torsion. Normal values in children. J Am Podiatr Med Assoc. 1989 Sep;79(9):432-5.

5. Turner MS. The association between tibial torsion and knee joint pathology. Clin Orthop Relat Res. 1994 May;(302)(302):47-51.

6. Hutter CG,Jr, Scott W. Tibial torsion. J Bone Joint Surg Am. 1949 Jul;31A(3):511-8.

7. Yoshioka Y, Siu DW, Scudamore RA, Cooke TD. Tibial anatomy and functional axes. J Orthop Res. 1989;7(1):132-7.

8. Hicks J, Arnold A, Anderson F, Schwartz M, Delp S. The effect of excessive tibial torsion on the capacity of muscles to extend the hip and knee during single-limb stance. Gait and Posture. 2007;26(4):546-52.

9. Yagi TM, D. Tibial Torsion in Patients With Medial-Type Osteoarthrotic Knees. Clinical Orthopaedics & Related Research. 1994 May;302:52-6.

10. Lincoln TL, Suen PW. Common rotational variations in children. J Am Acad Orthop Surg. 2003 Sep-Oct;11(5):312-20.

11. Silvers HJ, Mandelbaum BR. Prevention of anterior cruciate ligament injury in the female athlete. Br J Sports Med. 2007 Aug;41 Suppl 1:i52-9.

12. Chizewski MG, Chiu LZ. Contribution of calcaneal and leg segment rotations to ankle joint dorsiflexion in a weight-bearing task. Gait Posture. 2012 Feb 13.

13. Baker R. Pelvic angles: A mathematically rigorous definition which is consistent with a conventional clinical understanding of the terms. Gait and Posture. 2001;13(1):1-6.

14. Stefko RM, de Swart RJ, Dodgin DA, Wyatt MP, Kaufman KR, Sutherland DH, et al. Kinematic and kinetic analysis of distal derotational osteotomy of the leg in children with cerebral palsy. J Pediatr Orthop. 1998 Jan-Feb;18(1):81-7.

15. Turner MS, Smillie IS. The effect of tibial torsion of the pathology of the knee. J Bone Joint Surg Br. 1981;63-B(3):396-8.

16. Delgado ED, Schoenecker PL, Rich MM, Capelli AM. Treatment of severe torsional malalignment syndrome. J Pediatr Orthop. 1996 Jul-Aug;16(4):484-8.

17. Krivickas LS. Anatomical factors associated with overuse sports injuries. Sports Med. 1997 Aug;24(2):132-46.

18. Eckhoff DG, Brown AW, Kilcoyne RF, Stamm ER. Knee version associated with anterior knee pain. Clin Orthop Relat Res. 1997 Jun;(339)(339):152-5.

19. Eckhoff DG, Johnson KK. Three-dimensional computed tomography reconstruction of tibial torsion. Clin Orthop. 1994(302):42-6.

20. Turner MSMC, F.R.C.S. The Association Between Tibial Torsion and Knee Joint Pathology. Clinical Orthopaedics & Related Research. 1994 May;302:47-51.

21. Gigante A, Bevilacqua C, Bonetti MG, Greco F. Increased external tibial torsion in Osgood-Schlatter disease. Acta Orthop Scand. 2003;74(4):431-6.

22. Krackow KA, Mandeville DS, Rachala SR, Bayers-Thering M, Osternig LR. Torsion deformity and joint loading for medial knee osteoarthritis. Gait Posture. 2011 Apr;33(4):625-9.

23. Rabbito M, Pohl MB, Humble N, Ferber R. Biomechanical and clinical factors related to stage I posterior tibial tendon dysfunction. J Orthop Sports Phys Ther. 2011 Oct;41(10):776-84.

24. Powers CM, Chen PY, Reischl SF, Perry J. Comparison of foot pronation and lower extremity rotation in persons with and without patellofemoral pain. Foot Ankle Int. 2002 Jul;23(7):634-40.

25. Hewett TE, Myer GD, Ford KR, Heidt RS,Jr, Colosimo AJ, McLean SG, et al. Biomechanical measures of neuromuscular control and valgus loading of the knee predict anterior cruciate ligament injury risk in female athletes: a prospective study. Am J Sports Med. 2005 Apr;33(4):492-501.

26. Joseph M, Tiberio D, Baird JL, Trojian TH, Anderson JM, Kraemer WJ, et al. Knee valgus during drop jumps in National Collegiate Athletic Association Division I female athletes: the effect of a medial post. Am J Sports Med. 2008 Feb;36(2):285-9.

27. Shultz SJ, Schmitz RJ. Effects of transverse and frontal plane knee laxity on hip and knee neuromechanics during drop landings. Am J Sports Med. 2009 Sep;37(9):1821-30.

CHAPTER 3

An examination of tibial torsion measurement reliability and a comparison of external tibial torsion angles between women with uninjured and previously injured ACL

3.1 Introduction

Anterior cruciate ligament (ACL) injuries are prevalent in athletic individuals, with greater incidence in young women than men (1, 2). Previous research of the etiology of ACL injury suggests anatomical (i.e. structural) and neuromuscular (i.e. functional) factors may influence the risk for injury. There are two prevalent theories proposed as mechanisms for ACL injury: 1) anterior tibial translation (3) and 2) dynamic knee valgus (4). The tibial translation mechanism suggests an imbalance between ACL protagonists, muscles which cause posterior translation of the leg relative to the thigh; and ACL antagonists, muscles that cause anterior translation of the leg relative to the thigh (5). Although the hamstrings are considered the primary ACL protagonist, the soleus may also fulfill this function as both generate posterior directed forces on the leg segment (6). Dynamic knee valgus is described as multi-planar rotations which cause a medially directed collapse of the knee (4). Both anterior tibial translation and dynamic knee valgus are often reported to strain the ACL. Although anterior tibial translation and dynamic knee valgus are often described as competing mechanisms of ACL injury, a combined mechanism, in which both excessive anterior tibial translation and multi-planar rotations are present is possible and could be expected to place the greatest strain on the ACL.

Structural and functional factors at the hip and ankle may influence risk of ACL injury, under the concept that motion at joints proximal and/or distal to the knee will affect knee mechanics (7). One structural factor that may influence knee function is tibial torsion (8). Tibial torsion is the transverse plane rotational offset between the proximal and distal ends of the leg segment (Figure 1.5). Our recent research (Chapter 2) found an association between the direction of transverse plane leg rotation and increasing magnitude of external tibial torsion. Individuals with greater external tibial torsion had leg segments that rotated laterally – rather than medially – during a weight bearing partial squat. Similarly, Stefko et al. (9) reported that the direction of leg rotation changed from lateral to medial following tibial de-rotation surgery aimed at reducing the magnitude of external tibial torsion. A laterally rotated leg is consistently reported in analyses of ACL injuries (10, 11). Lateral rather than medial leg rotation may be the result of a shift in the soleus' force vector – from posterior to medial directed.

Musculoskeletal modelling (12, 13) has found that increasing external tibial torsion alters the soleus' function during weight bearing activity. While weight bearing, the soleus is an ankle plantar flexor, which indirectly extends the knee (14). However, the soleus' force vector changes direction with increasing external tibial torsion, diminishing its ability to plantar flex the ankle and extend the knee (12). Schwartz and Lakin (13) describe the soleus, in a leg with a high degree of tibial torsion, as losing its ability to generate a rearward thrust of the leg, a role complementary to acting as an ACL protagonist. Although not previously described, the decreased posterior directed force with greater external tibial torsion should be accompanied by an increase in the medial directed force acting on the leg. An increased medial directed force would pull the proximal aspect of the leg medially resulting in leg abduction. When the leg is abducted, external knee abduction moment increases (15), which is observed in individuals with excessive external tibial torsion (16). As a result, greater external tibial torsion appears to increase valgus stress at the knee – the second mechanism of ACL injury.

Taken together, greater external tibial torsion may result in: 1) reduced anterior tibial translation restraint, 2) increased lateral leg rotation and 3) increased external knee abduction moment. Each of these phenomena alone is associated with ACL injury, which suggests all three combined may place high tensile and torsional stresses on the ACL (4, 17, 18). As such, these functional consequences of greater external tibial torsion suggest that this anatomic variation may predispose an individual to ACL injury. The primary objective of this research was to examine tibial torsion in individuals with and without ACL injury, as well as to compare tibial torsion between the involved (injured) and non-involved (healthy) limbs of individuals with ACL injury. It was hypothesized that external tibial torsion angle would be greater in individuals with ACL injury than those without. Further, it was hypothesized that external tibial torsion angle would be greater in the involved limb of individuals with ACL injury compared to the non-involved limb.

The secondary objective of this research was to evaluate the reliability and validity of methods to measure tibial torsion. Although motion analysis was previously used (Chapter 2) to determine tibial torsion, a clinical method is desirable because motion analysis is not always available or required. Clinical tests for tibial torsion can have multiple uses, including: patient management, screening, surveillance, or epidemiological studies. Tibial torsion is commonly evaluated by the orthopedist, pediatrician, or physical therapist (19-21). Measurement techniques vary from the use of a standard goniometer (22, 23), calipers (20, 24), computed tomography (CT) (25, 26), and magnetic resonance imaging (MRI) (27, 28). Since specialized equipment is seldom available or desired in a standard clinical setting, the need to establish the reliability of simple clinical methods and validity against direct measures such as motion analysis was needed prior to proceeding with this investigation. Therefore, this chapter describes: [1] an assessment of the reliability and validity of three methods of measuring tibial torsion, and [2] a comparison of tibial torsion in individuals with and without ACL injury.

3.2 Methods and Measurement

3.2.1 Participants

[1] To assess the reliability and validity of tibial torsion measurements, a non-probability consecutive convenience sample of men (n=10) and women (n=5) was recruited from the university population using recruitment flyers on notice boards and electronic mailing lists to Faculty and students in and around the Faculty of Physical Education and Recreation buildings at the University of Alberta. All participants provided written informed consent as approved by a Research Ethics Board at the author's institution (ID: Pro00046837). Participants were healthy adults aged 18-40 years without injury. Men were 23.8±5.2 years old, 1.80±0.07 m tall, and 86.1±17.6 kg body mass. Women were 21.2±0.75 years old, 1.64±0.05 m tall, and 60.9±4.6 kg body mass.

[2] To compare tibial torsion in individuals with (n=15) and without (n=15) ACL injury, a non-probability consecutive convenience sample of women (n=30) was recruited from the university population using recruitment flyers on notice boards and electronic mailing lists to Faculty and students in and around the Faculty of Physical Education and Recreation buildings at the University of Alberta. Non-injured participants were excluded if they had a self-reported history of serious foot or ankle problems that required medical treatment or surgical intervention. Injured participants fulfilled our inclusion criteria if they had a noncontact or minimal contact mechanism of injury, but were excluded if they had a (full) contact ACL tear (29). All participants provided written informed consent as approved by a Research Ethics Board (ID: Pro00028419).

Participants with ACL injury were 24.6±7.3 years old, 1.68±0.08 m tall, and 67.6±11.6 kg body mass. Control participants without ACL injury were 25.9±5.9 years old, 1.69±0.07 m tall, and 63.7±6.0 kg body mass.

3.2.2 Experimental Protocol

[1] Three methods of measuring tibial torsion were compared to our previously developed motion analysis method. These methods were: thigh foot angle (through the second ray of the foot) using a standard goniometer (TFA2), thigh foot angle (through a line perpendicular to the transmalleolar axis) using a standard goniometer (TFAM), and tibial torsion measured with a custom designed fixed inclinometer clamp system (INC). Two observers collected measurements of all methods from participants during separate visits. To better ensure inter-rater reliability, participants were tested by each of the two observers without knowledge of the other observer's results. Inter-method and inter-rater reliability was determined using intra-class correlation coefficients. Limits of agreement (Bland-Altman) were also calculated for inter-method comparisons.

The thigh-foot angle (Figure 3.1) using the second ray of the foot (TFA2) was determined with a goniometer as the angle between the axis of the foot through the second metatarsal (with the foot held in subtalar neutral) and the long axis of the thigh when the knee is flexed 90 degrees (26-28). A key limitation of this measure is that it represents a composite measure of: tibial torsion, structural alignment of the foot distal to the talocrural joint, and alignment between the leg and thigh segments. Thus, irregularities proximal or distal to the leg segment may affect the accuracy of measuring tibial torsion. The participant was placed prone with the hip in neutral position, and the active knee flexed to 90 degrees. A goniometer was aligned to the bisection of the long axis of the femur and a line that bisected the calcaneus and the ray of the second metatarsal (Figure 3.1).

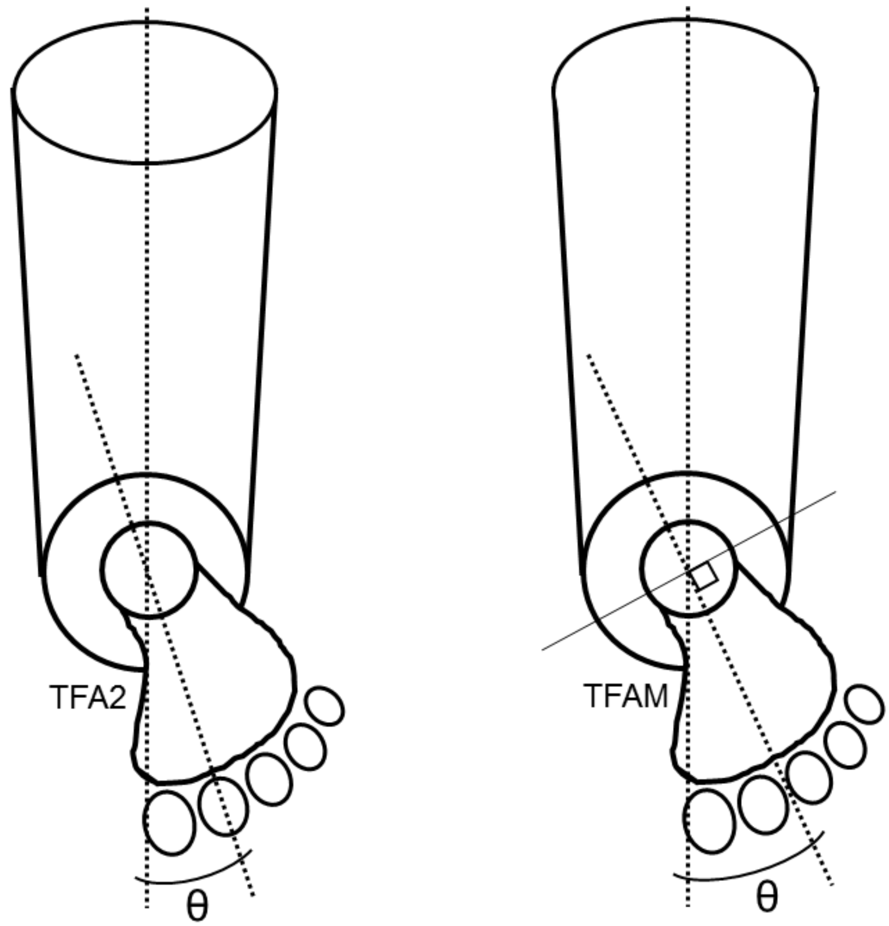

Figure 3.1 Diagram showing the TFA2 (thigh foot angle through second metatarsal) and the TFAM (thigh foot angle through a line perpendicular to the trans-malleolar axis) methods of determining tibial torsion (Θ) in study participants.

The thigh-foot angle using the trans-malleolar axis (TFAM) was determined with a goniometer as the angle between the axis of the foot through a line perpendicular to the trans-malleolar axis (with the foot held in subtalar neutral) and the long axis of the thigh when the knee is flexed 90 degrees (Figure 3.2). This method reduces the possibility of error associated with malalignments in the foot. It also removes any assumption that the second metatarsal head is associated to the alignment of the trans-malleolar axis. However, tibio-femoral alignment may still influence the accuracy of measuring tibial torsion.

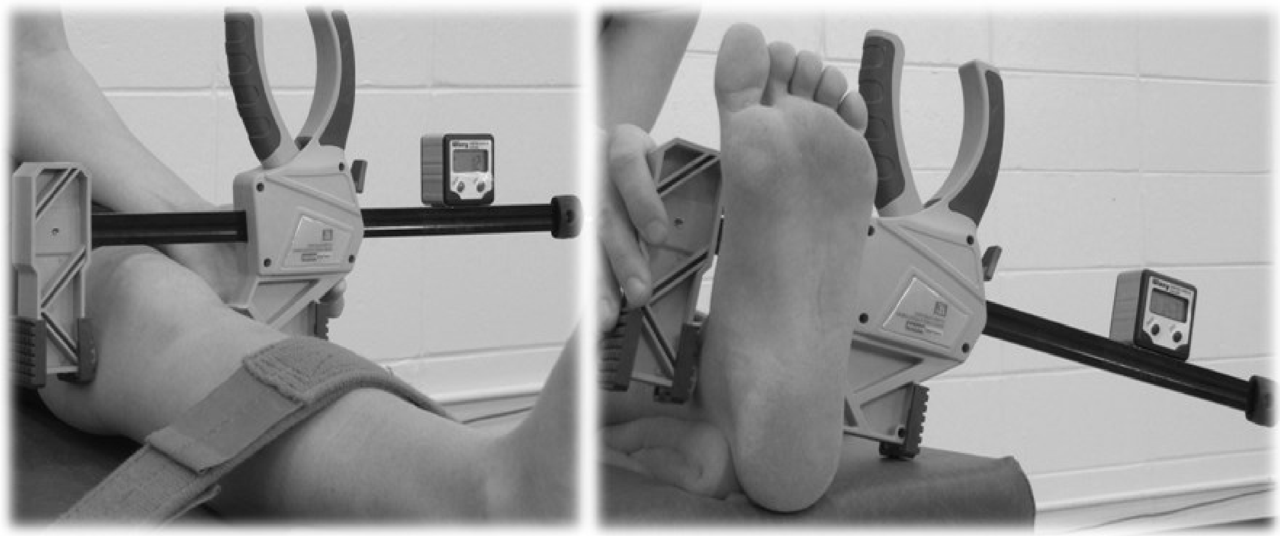

Figure 3.2 Diagram showing the INC (inclinometer) method of determining tibial torsion.

To determine tibial torsion using the inclinometer (INC) method, participants were seated with the hip flexed to 90 degrees and the knee extended, with padding if necessary, to ensure the leg segment was parallel to the horizontal. Markings were made on the lateral and medial tibial condylar prominences, and the lateral and medial malleoli. The leg was then secured to the bench with strapping to minimize any translation or rotation of the leg segment, while measurements were taken with the foot in anatomic neutral (sole of the foot 90 degrees relative to the leg). A digital angle gauge (Wixey; Sanibel, FL) was fixed to a large bar clamp (Haussmann; Boucherville, Canada) (Figure 3.2).   By aligning the clamp centers with the proximal and distal landmarks, the angle of these landmarks relative to horizontal was calculated. The clamp was secured in position by the examiner throughout the procedure, and was not released while the measures were taken.

The difference in inclinometer measures represented the transverse plane rotation between the proximal and distal ends of the leg segment, or external tibial torsion.

The three previous methods were compared to a 3D motion analysis method. Previously reported in Chapter 2 (Figure 2.4), this method involves markers placed on the lateral tibial condylar prominence, the medial tibial condylar prominence, the lateral malleolus, and the medial malleolus. A proximally biased leg segment is created along with a distally biased leg segment (Figure 2.5). A comparison of these two leg segments in the transverse plane about the longitudinal (Z) axis is the external tibial torsion angle.

[2] Tibial torsion was measured in the ACL injured and non-injured female participants for left and right limbs using the inclinometer method previously described. Three measurements were taken on each limb and the average was used for analysis.

3.2.3 Statistical Analyses

[1] Inter-method and inter-rater comparisons between the methods of measurement were made using an intra-class correlation (ICC) analysis for all comparisons. A Bland-Altman analysis was used to assess the level of agreement between the methods to compare the new techniques to the established reference standard of motion analysis. Alpha was set a priori (α=0.05) and all statistical calculations were conducted using SPSS (version 11.0; SPSS Inc. Chicago, IL).

[2] For healthy controls, tibial torsion was averaged between limbs. For ACL injured participants, 14 had an ACL injury on one limb, and one had ACL injuries in both limbs. Within this group, limbs were categorized as involved and non-involved. Data for the participant with ACL injury in both limbs was averaged. Tibial torsion was compared between the ACL injured group's involved limb and the healthy control group's averaged data using unpaired t-tests. For the

14 participants with injury on one limb only, tibial torsion was compared between involved and non-involved limbs using paired t-tests. To determine the magnitude of differences, Cohen's *d* effect size (ES) was calculated. Alpha was set a priori (α=0.05) and all statistical calculations were conducted using SPSS (version 11.0; SPSS Inc. Chicago, IL).

3.3 Results

[1] A high degree of inter-method reliability was found between visits. The inter-method reliability of the left limb average measures ICC was .954 with a 95% confidence interval from .909 to .976 and p<.001. The inter-method reliability of the right limb average measures ICC was .962 with a 95% confidence interval from .930 to .980 and p<.001. A high degree of inter-rater reliability was also found between testers. The inter-rater reliability of the left limb average measures ICC was .963 with a 95% confidence interval from .937 to .977 and p<.001. The inter-rater reliability of the right limb average measures ICC was .968 with a 95% confidence interval from .938 to .982 and p<.001.

The Bland-Altman analysis indicated that the 95% limits of agreement: between the MAN and TFA2 methods ranged from -8.2 to 22.2 degrees (bias=7.6); between the MAN and TFAM methods ranged from -11.8 to 5.8 degrees (bias=4.5); and between MAN and INC methods ranged from -9.3 to 3.4 degrees (bias=3.1).

[2] When comparing the involved limb of the ACL injured to the healthy controls, significant differences were observed between groups for tibial torsion (p<0.0001; ES=1.75 SD). ACL injured persons had a tibial torsion angle of 19±4 degrees versus 12±4 degrees in the healthy controls. Tibial torsion angle in participants with ACL injury on one limb only was 20±4 degrees

in the involved limb and 18±4 in the non-involved limb, a difference which approached significance (p=0.08; ES=0.5 SD). The participant who had bilateral ACL injuries had external tibial torsion angles of 22 (left) and 26 (right) degrees.

3.4 Discussion

[1] This investigation found the inclinometer method of determining tibial torsion to be the most strongly related to a motion analysis method. High inter-method reliability between visits, and high inter-rater reliability between testers was also demonstrated during the investigation. The average discrepancy (bias) between the two methods (MAN / INC) was 3.1 degrees, which is considerably small when defining kinematic measures. The main concern is that the limits of agreement are 12.7 degrees wide, which suggests some ambiguity between the measures, and presents a notable limitation to the inclinometer method as a substitute for motion analysis. While this method does not use equipment (bar clamp and digital angle gauge) commonly found in clinical settings, this equipment is low cost and readily available. There are two other major benefits for using this method that helped to justify its inclusion in the project. Firstly, it is an advantage that the measurement is taken directly on the leg segment. Instead of an indirect estimate of leg anthropometrics based on thigh and foot structure, the inclinometer method is able to identify specific landmarks on the proximal and distal leg segment for analysis. Second, the landmarks for the inclinometer method are the identical landmarks used to define the proximal and distal leg parameters in the motion analysis method. This similarity in structural landmarks further supports the use of the inclinometer method, if motion analysis is considered as the reference ‘gold’ standard.

[2] The primary objective of this investigation was to compare tibial torsion in individuals with and without ACL injury. As hypothesized, external tibial torsion was greater in ACL injured individuals than healthy controls; this difference had a very large effect size (1.75). Effect size expresses the mean difference between two groups in standard deviation units. The meaning of effect size varies by context, but the standard interpretation offered by Cohen (1988) is: 0.2 = small (1/5 of a standard deviation), 0.5 = moderate (1/2 of a standard deviation), and 0.8 = large (4/5 of a standard deviation unit). Based on these criteria, the effect size value of 1.75 suggests a high to very high practical significance. A significant difference was not found for tibial torsion in the affected versus unaffected limb in individuals who had unilateral ACL injury. While the effect size was moderate (0.5), it is unclear whether such a difference, although statistically significant, would be practically meaningful. Nonetheless, these data support our theory that greater external tibial torsion may be a risk factor for ACL injury. Excessive external tibial torsion has also been associated with other knee pathologies, including Osgood-Schlatter's (30), patellofemoral pain syndrome and tibio-femoral osteoarthritis (31). Although there is no study of tibial torsion specific to knee mechanics in ACL injury, the functional mechanical consequences of greater external tibial torsion provide insight into why this anatomical variable may increase risk of ACL injury. Greater magnitudes of external tibial torsion may: 1) influence the direction of leg rotation, 2) alter the soleus force vector and 3) increase valgus stress at the knee.

Our previous research found that the magnitude of tibial torsion correlates with whether the leg segment internally or externally rotates (Chapter 2). Individuals with greater external tibial torsion (>14 degrees) externally rotated their leg during a squat task compared to those with less external tibial torsion (<14 degrees) whose legs internally rotated. A similar correlation between tibial torsion and transverse plane knee rotation during gait was reported by Radler et al. (32).

Although leg external rotation does not appear to strain the ACL to the same extent as leg internal rotation (33, 34), external rotation has been described as a mechanism of non-contact ACL rupture (11). Moreover, video analyses of ACL injuries typically describe the leg as being externally rotated (35). The risk of ACL injury from external leg rotation may be due to impingement of the ACL on the intercondylar prominence. Although external leg rotation does not appear to load the ACL as much as internal rotation (33, 34), ACL impingement becomes a greater risk with a leg that is externally rotated and abducted such as during landing or a plant and cut maneuver (35). Cadaver studies have demonstrated that the lateral notch wall (36) and the notch roof (37) are possible areas of impingement, particularly during external rotation of the leg or internal rotation of the thigh, due to the attachment sites of the proximal ACL bundle.

Excessive external tibial torsion has also been demonstrated using musculoskeletal modeling to reduce the ability of the soleus to extend the knee during gait (12). The soleus muscle stabilizes the knee, when tibial torsion angle is small, by exerting more posterior-directed force on the proximal leg which limits excessive anterior tibial translation. A model by Schwartz (13) suggests that extreme external tibial torsion decreases the ability of the soleus to extend the knee. Although not described, the loss in posterior-directed force from the soleus should be accompanied by an increase in medial-directed force. Therefore, in addition to reducing the ability for soleus to plantar flex the ankle and extend the knee, greater external tibial torsion would be hypothesized to increase the medial-directed force acting on the proximal leg, which would contribute to abduction of the knee.

External tibial torsion has been linked to increased external valgus knee moments (38). The lateral malleolus is positioned more posterior and inferior relative to the medial malleolus in most individuals. This difference is more pronounced when greater external tibial torsion is present and results in greater abduction of the foot in the transverse plane (39). Consequently, an increase in tibial torsion is proposed to increase foot progression angle (12). Video studies of ACL injury have also reported external rotation of the leg and/or foot in cases of rupture (35, 40). The soleus muscle is responsible for contributing to knee stability, and when tibial torsion angle is lower, imparts a more posteriorly directed force on the (proximal) leg.

3.5 Summary

Individuals with greater external tibial torsion demonstrate altered lower extremity mechanics in squatting. Despite the associations between external tibial torsion and knee pathologies, minimal research exists on the functional mechanics of individuals with varying external tibial torsion. In particular, the differences in lower extremity mechanics and the effect on joint stresses should be determined in individuals with high and low external tibial torsion. In relation to ACL injuries, future research should compare lower extremity mechanics between ACL injured versus healthy individuals with high external tibial torsion. This comparison may allow determination of "coping strategies" that protect the ACL in uninjured individuals. This research also demonstrates that reliable measurement of tibial torsion, with high inter-rater reliability, can be made using a simple bar clamp and digital angle gauge. This method may facilitate future study of and clinical screening for tibial torsion.

3.6 References

1. Arendt E, Dick R. Knee injury patterns among men and women in collegiate basketball and soccer. NCAA data and review of literature. Am J Sports Med. 1995 Nov-Dec;23(6):694-701.

2. Wojtys EM, Huston LJ, Schock HJ, Boylan JP, Ashton-Miller JA. Gender differences in muscular protection of the knee in torsion in size-matched athletes. J Bone Joint Surg Am. 2003 May;85-A(5):782-9.

3. Berchuck M, Andriacchi TP, Bach BR, Reider B. Gait adaptations by patients who have a deficient anterior cruciate ligament. J Bone Joint Surg Am. 1990 Jul;72(6):871-7.

4. Hewett TE, Myer GD, Ford KR, Heidt RS,Jr, Colosimo AJ, McLean SG, et al. Biomechanical measures of neuromuscular control and valgus loading of the knee predict anterior cruciate ligament injury risk in female athletes: a prospective study. Am J Sports Med. 2005 Apr;33(4):492-501.

5. Boden BP, Dean GS, Feagin JA,Jr, Garrett WE,Jr. Mechanisms of anterior cruciate ligament injury. Orthopedics. 2000 Jun;23(6):573-8.

6. Renstrom P, Arms SW, Stanwyck TS, Johnson RJ, Pope MH. Strain within the anterior cruciate ligament during hamstring and quadriceps activity. Am J Sports Med. 1986 Jan-Feb;14(1):83-7.

7. Astephen JL, Deluzio KJ, Caldwell GE, Dunbar MJ. Biomechanical changes at the hip, knee, and ankle joints during gait are associated with knee osteoarthritis severity. J Orthop Res. 2008 Mar;26(3):332-41.

8. Krackow KA, Mandeville DS, Rachala SR, Bayers-Thering M, Osternig LR. Torsion deformity and joint loading for medial knee osteoarthritis. Gait Posture. 2011 Apr;33(4):625-9.

9. Stefko RM, de Swart RJ, Dodgin DA, Wyatt MP, Kaufman KR, Sutherland DH, et al. Kinematic and kinetic analysis of distal derotational osteotomy of the leg in children with cerebral palsy. J Pediatr Orthop. 1998 Jan-Feb;18(1):81-7.

10. Park H-, Wilson NA, Zhang L-. Gender differences in passive knee biomechanical properties in tibial rotation. Journal of Orthopaedic Research. 2008;26(7):937-44.

11. Fung DT, Zhang LQ. Modeling of ACL impingement against the intercondylar notch. Clin Biomech (Bristol, Avon). 2003 Dec;18(10):933-41.

12. Hicks J, Arnold A, Anderson F, Schwartz M, Delp S. The effect of excessive tibial torsion on the capacity of muscles to extend the hip and knee during single-limb stance. Gait and Posture. 2007;26(4):546-52.

13. Schwartz M, Lakin G. The effect of tibial torsion on the dynamic function of the soleus during gait. Gait Posture. 2003 4;17(2):113-8.

14. Neumann DA. Kinesiology of the musculoskeletal system: foundations for rehabilitation. 2nd ed. St. Louis, Mo.: Mosby/Elsevier; 2010.

15. Kristianslund E, Faul O, Bahr R, Myklebust G, Krosshaug T. Sidestep cutting technique and knee abduction loading: implications for ACL prevention exercises. Br J Sports Med. 2014 May;48(9):779-83.

16. Aiona M, Calligeros K, Pierce R. Coronal plane knee moments improve after correcting external tibial torsion in patients with cerebral palsy. Clin Orthop Relat Res. 2012 May;470(5):1327-33.

17. Fukuda Y, Woo SL, Loh JC, Tsuda E, Tang P, McMahon PJ, et al. A quantitative analysis of valgus torque on the ACL: a human cadaveric study. J Orthop Res. 2003 Nov;21(6):1107-12.

18. Gabriel MT, Wong EK, Woo SL, Yagi M, Debski RE. Distribution of in situ forces in the anterior cruciate ligament in response to rotatory loads. Journal of Orthopaedic Research. 2004;22(1):85-9.

19. Engel GM, Staheli LT. The natural history of torsion and other factors influencing gait in childhood. A study of the angle of gait, tibial torsion, knee angle, hip rotation, and development of the arch in normal children. Clin Orthop Relat Res. 1974 Mar-Apr;(99)(99):12-7.

20. Staheli LT, Engel GM. Tibial torsion: a method of assessment and a survey of normal children. Clin Orthop Relat Res. 1972 Jul-Aug;86:183-6.

21. Turner MSMC, F.R.C.S. The Association Between Tibial Torsion and Knee Joint Pathology. Clinical Orthopaedics & Related Research. 1994 May;302:47-51.

22. King HA, Staheli LT. Torsional problems in cerebral palsy. Foot Ankle. 1984 Jan-Feb;4(4):180-4.

23. Luchini M, Stevens DB. Validity of torsional profile examination. J Pediatr Orthop. 1983 Feb;3(1):41-4.

24. Herold HZ, Marcovich C. Tibial torsion in untreated congenital clubfoot. Acta Orthop Scand. 1976 Feb;47(1):112-7.

25. Jakob RP, Haertel M, Stussi E. Tibial torsion calculated by computerised tomography and compared to other methods of measurement. J Bone Joint Surg Br. 1980 May;62-B(2):238-42.

26. Laasonen EM, Jokio P, Lindholm TS. Tibial torsion measured by computed tomography. Acta Radiol Diagn (Stockh). 1984;25(4):325-9.

27. Muhamad AR, Freitas JM, Bomar JD, Dwek J, Hosalkar HS. CT and MRI lower extremity torsional profile studies: measurement reproducibility. J Child Orthop. 2012 Oct;6(5):391-6.

28. Diederichs G, Kohlitz T, Kornaropoulos E, Heller MO, Vollnberg B, Scheffler S. Magnetic resonance imaging analysis of rotational alignment in patients with patellar dislocations. Am J Sports Med. 2013 Jan;41(1):51-7.

29. Griffin LY, Agel J, Albohm MJ, Arendt EA, Dick RW, Garrett WE, et al. Noncontact anterior cruciate ligament injuries: risk factors and prevention strategies. J Am Acad Orthop Surg. 2000 May-Jun;8(3):141-50.

30. Gigante A, Bevilacqua C, Bonetti MG, Greco F. Increased external tibial torsion in Osgood-Schlatter disease. Acta Orthop Scand. 2003;74(4):431-6.

31. Lohmander LS, Englund PM, Dahl LL, Roos EM. The long-term consequence of anterior cruciate ligament and meniscus injuries: osteoarthritis. Am J Sports Med. 2007 Oct;35(10):1756-69.

32. Radler C, Kranzl A, Manner HM, Höglinger M, Ganger R, Grill F. Torsional profile versus gait analysis: Consistency between the anatomic torsion and the resulting gait pattern in patients with rotational malalignment of the lower extremity. Gait and Posture. 2010;32(3):405-10.

33. Miyasaka T, Matsumoto H, Suda Y, Otani T, Toyama Y. Coordination of the anterior and posterior cruciate ligaments in constraining the varus-valgus and internal-external rotatory instability of the knee. J Orthop Sci. 2002;7(3):348-53.

34. Markolf KL, Burchfield DM, Shapiro MM, Shepard MF, Finerman GA, Slauterbeck JL. Combined knee loading states that generate high anterior cruciate ligament forces. J Orthop Res. 1995 Nov;13(6):930-5.

35. Olsen OE, Myklebust G, Engebretsen L, Bahr R. Injury mechanisms for anterior cruciate ligament injuries in team handball: a systematic video analysis. Am J Sports Med. 2004 Jun;32(4):1002-12.

36. LaPrade RF, Burnett QM,2nd. Femoral intercondylar notch stenosis and correlation to anterior cruciate ligament injuries. A prospective study. Am J Sports Med. 1994 Mar-Apr;22(2):198,202; discussion 203.

37. Norwood LA, Cross MJ. Anterior cruciate ligament: functional anatomy of its bundles in rotatory instabilities. Am J Sports Med. 1979 Jan-Feb;7(1):23-6.

38. MacWilliams BA, McMulkin ML, Baird GO, Stevens PM. Distal Tibial Rotation Osteotomies Normalize Frontal Plane Knee Moments. J Bone Joint Surg Am. 2010 December 1;92(17):2835-42.

39. Seber S, Hazer B, Kose N, Gokturk E, Gunal I, Turgut A. Rotational profile of the lower extremity and foot progression angle: computerized tomographic examination of 50 male adults. Arch Orthop Trauma Surg. 2000;120(5-6):255-8.

40. Ebstrup JF, Bojsen-Moller F. Anterior cruciate ligament injury in indoor ball games. Scand J Med Sci Sports. 2000 Apr;10(2):114-6.

CHAPTER 4

Tibial torsion influences lower body joint kinematics and kinetics in women with ACL injury during squatting and landing

4.1 Introduction

Anterior cruciate ligament (ACL) injuries are prevalent in sport and activities that require landing, sudden deceleration or change-in-direction manoeuvres (1). Sequelae of ACL injury include knee joint instability and lower extremity muscle weakness, or compensation strategies which may impair physical function in the short and long term (2-5). Young women have a higher incidence of ACL injury, regardless of mechanism of injury, than young men (6, 7). This discrepancy between sexes is particularly pronounced for non-contact ACL rupture (8). When ACL injury occurs without physical contact to another athlete or immovable object (other than the ground or landing surface), they are referred to as non-contact (1, 9). Mechanically, the ACL is injured from excessive tensile stress, which may occur when anterior shear force is exerted on the proximal tibia (10, 11), or through frontal and transverse plane moments of force exerted on the segments comprising the knee joint (7). Although ACL rupture can be described anatomically, this knowledge, at present, cannot be translated to *in vivo* human movement (12). As a consequence, the exact mechanism(s) of ACL injury during functional activities is not exactly known. Further research is warranted to identify anatomical, neuromuscular and motor control factors contributing to ACL injury which may lead to developing strategies that reduce the incidence of these injuries.

Multiple anatomical risk factors are proposed to contribute to ACL risk, including femoral intercondylar notch width, Q-angle, and excessive foot pronation. Anatomical risk factors are also known as structural factors, as they are variations in skeletal structure morphology that may influence segmental motion, and subsequently stress placed on muscle and connective tissue.

Previous research by this author (Chapter 3) has identified another potential anatomical risk factor for ACL injury. Specifically, external tibial torsion is greater in individuals with ACL injury versus healthy individuals. Tibial torsion is the rotational offset of the distal to the proximal articulating axes of the tibia/fibula complex in the transverse plane (13). Musculoskeletal modeling studies have found that increasing external tibial torsion alters force vector directions for muscles acting on the leg segment. In particular, there appears to be a shift in the ability of mono-articular plantar flexors, such as soleus, to plantar flex the ankle and extend the knee (14, 15) through a reduction in the posterior-directed force vector. It is hypothesized that as external tibial torsion increases, the change in these muscles' line of pull not only decreases the posterior-directed force vector, but also increases the medial-directed force vector. An increased medial-directed force vector would contribute to moments causing leg abduction and lateral rotation, while leg abduction has been reported to increase the external knee valgus moment (16, 17). Individuals with excessive external tibial torsion have larger external knee valgus moments (18), moreover, surgical reduction of excessive external tibial torsion has been shown to reduce these moments (19). Greater external tibial torsion has been associated to increased leg segment lateral rotation (Chapter 2). Lateral rotation of the leg has been demonstrated, in vitro, to result in increased impingement of the ACL against the lateral wall of the intercondylar notch (20, 21) with an applied valgus moment. These reports suggest that external tibial torsion may predispose the lower body to movements that increase the risk of ACL rupture.

A decreased posterior force vector for the posterior leg muscles may have other consequences that relate to ACL injury. Specifically, decreasing this force vector reduces the ability to generate an ankle plantar flexor moment. Mechanically, this may be accomplished by shifting the centre of pressure away from the forefoot and towards the heel (22). When athletic activities, such as landing from a jump, are performed with the centre of pressure near the heel, the time to achieve knee flexion is greater and vertical ground reaction forces are higher (23). Further, less work is performed by the ankle plantar flexors and more work by the knee extensors (24). Finally, peak knee extensor moment is achieved earlier in heel landing versus forefoot landing, corresponding with smaller knee flexion angles. At smaller knee flexion angles (less than 35 degrees), the quadriceps muscles have been implicated in ACL injury because of their anteriorly directed force vector acting on the tibia (25), which places greater tensile stress on the ACL.

ACL injured persons may have increased external tibial torsion (26) and external tibial torsion may cause mechanics that increase stress on the ACL. However, there has been no study of lower extremity mechanics in ACL injured persons in comparison to uninjured individuals with more or less external tibial torsion. An understanding of how mechanics differ between these populations may provide insight into how ACL injuries occur during functional tasks. The purpose of this investigation was to compare lower extremity mechanics in ACL injured persons with healthy individuals who have either low or high external tibial torsion. As ACL injured persons may have different mechanics that are the result of: 1) greater external tibial torsion and 2) ACL injury itself, a three group between subjects case-control study design was employed. This design categorized eligible participants as either: ACL injured (ACL), high external tibial torsion (HTT), or low external tibial torsion (LTT). If the mechanics demonstrated by the HTT and LTT groups were not different from each other, but different from the ACL group, this would suggest the mechanics are not due to external tibial torsion. If the mechanics demonstrated by the HTT and

ACL groups were not different from each other, but different from the LTT group, this would suggest the mechanics are due to a higher magnitude of external tibial torsion.

It was hypothesized that individuals with ACL injury (and high tibial torsion) would have mechanics that differed from low external tibial torsion individuals. These hypotheses were predicted from how external tibial torsion affects muscle force vectors (as previously described). The three hypotheses were:

Hypothesis 1 – Individuals with greater external tibial torsion (both ACL injured and healthy) will have greater lateral leg rotation (or lesser medial leg rotation) during squatting and landing, and greater foot progression angles during landing.

Hypothesis 2 – Individuals with greater external tibial torsion (both ACL injured and healthy) will have lesser internal ankle plantar flexor net joint moments for squatting and landing, and a corollary increase in knee extensor and hip extensor net joint moments.

Hypothesis 3 – Individuals with greater external tibial torsion (both ACL injured and healthy) will have greater leg abduction for squatting and landing.

4.2 Methods and Measurements

4.2.1 Participants

A non-probability consecutive convenience and quota sample of women, aged 18-35 years, with (n=14) and without (n=34) ACL injury, was recruited from the university population using recruitment flyers on notice boards and electronic mailing lists to Faculty and students in and around the Faculty of Physical Education and Recreation buildings at the University of Alberta.

All participants provided written informed consent as approved by a Research Ethics Board at the University of Alberta (Ethics ID: Pro 00048284). Potential participants were excluded if they had a self-reported history of hip, ankle or foot problems that required repeated medical treatment or surgery. ACL participants were required to fulfill inclusion criteria which involved the following requirements: being physically active in sport or recreation, having one ACL reconstructed once by hamstring graft, mechanism of injury being non-contact (1) except with the ground, being at least six months post-surgery and having no other major surgeries on the lower body. Healthy controls were screened prior to enrollment and divided into two groups: high external tibial torsion (n=14) or low external tibial torsion (n=20). An inclinometer manual measurement of tibial torsion was used (Figure 3.2) and individuals with more than 17 degrees external tibial torsion were classified as high and those with less than 14 degrees external tibial torsion as low. Participants were excluded if the screening determined their tibial torsion to be between 14-17 degrees. This mid-range was chosen based on previous research from our laboratory as being unrepresentative of high or low tibial torsion. To determine this, pilot participants were grouped based on the direction of transverse plane leg rotation during a squat, as medial or lateral rotators. Medial rotators consistently measured below 14 degrees of torsion, and lateral rotators measured above 17 degrees. Inconsistencies were most frequent between 14-17 degrees, providing the basis for exclusion. The pre-screening was evaluated using the inclinometer measure of tibial torsion as this method showed the strongest correlation to our laboratory method (motion analysis) for determining tibial torsion (Figure 2.5). Participants with ACL injury (n=14) were 23.2±2.4 years old, 1.71±0.07 m tall, and 67.3±10.8 kg body mass. Healthy control participants (n=34) were 23.8±4.1 years old, 1.68±0.05 m tall, and 64.9±9.4 kg body mass.

4.2.2 Experimental Protocol

Participants visited the laboratory twice, approximately one week apart. On the first visit, control participants were screened for their external tibial torsion (for grouping purpose) and anthropometric measurements were completed on all participants. The first visit also provided the opportunity for all participants to practice the tasks which would be completed as part of the motion analysis on the second visit. Participants viewed an instructional video demonstrating the tasks, and were then asked to practice five trials of each task. The tasks included: a weight bearing partial squat, two-foot vertical jump and landing, and box landings (stepping off the box both with the right foot first and the left foot first). Practice trials were not recorded, and minimal feedback was given to assist the participants in performing the tasks safely. Feedback was given for the squatting task, to standardize the foot position, and to limit trunk and pelvis movement (27). For the jump landing and box landing tasks, feedback was only offered if the participants felt unsafe with the task since it was important not to alter the participants' individual movements. The researcher would repeat the same instructions for each trial. For the two-foot jump task: "Place one foot on each force plate, take off with both feet, land on both feet and jump as high as you can with your eyes forward". For the box landing task: "Step off the box with one foot, land with two feet and keep your eyes forward".

During the second session, all tasks were recorded for motion analysis. Participants performed a body weight partial squat (3 repetitions). For the partial squat, participants were instructed to reach a position of maximum dorsiflexion without allowing their trunk to flex forward (Figure 4.1) or their pelvis to rotate. The partial squat was selected for this investigation as it allowed foot placement and foot rotational alignment to be controlled, and has previously been used to determine the association between lower extremity mechanics and tibial torsion (Chapter 2). The feet were aligned such that an imaginary line facing forward was through the centre of the heel and the second ray of the foot. Foot width was standardized by placing the feet so that this imaginary line was equal to the measured inter-distance width of the femoral greater trochanters. For each trial, participants stood motionless for a minimum of two seconds (baseline position). The participants then squatted to their maximum dorsiflexion position and held this position for two to three seconds (squat position).

Figure 4.1 Image showing retro-reflective marker placement, partial squat task and positive directions for translations and rotations following the right hand rule.

Two-foot maximal vertical jump landings were chosen for the participants, as this task mimics a dynamic task present in most activities that involve landings. The task also provides an inherent safety measure, since the participants are landing from a self-regulated height. The participants would be comfortable landing from their own maximal effort jumps, and would hopefully demonstrate their normal mechanics. For the vertical jump and landing (six repetitions), participants performed a maximal effort double leg take-off vertical jump. Participants were instructed to reach upward with both hands, and to jump as high as possible. Participants were only instructed to place one foot on each of the force platforms, but were not given any constraints to foot placement or landing technique throughout the data collection. Rest was provided ad libitum between repetitions.

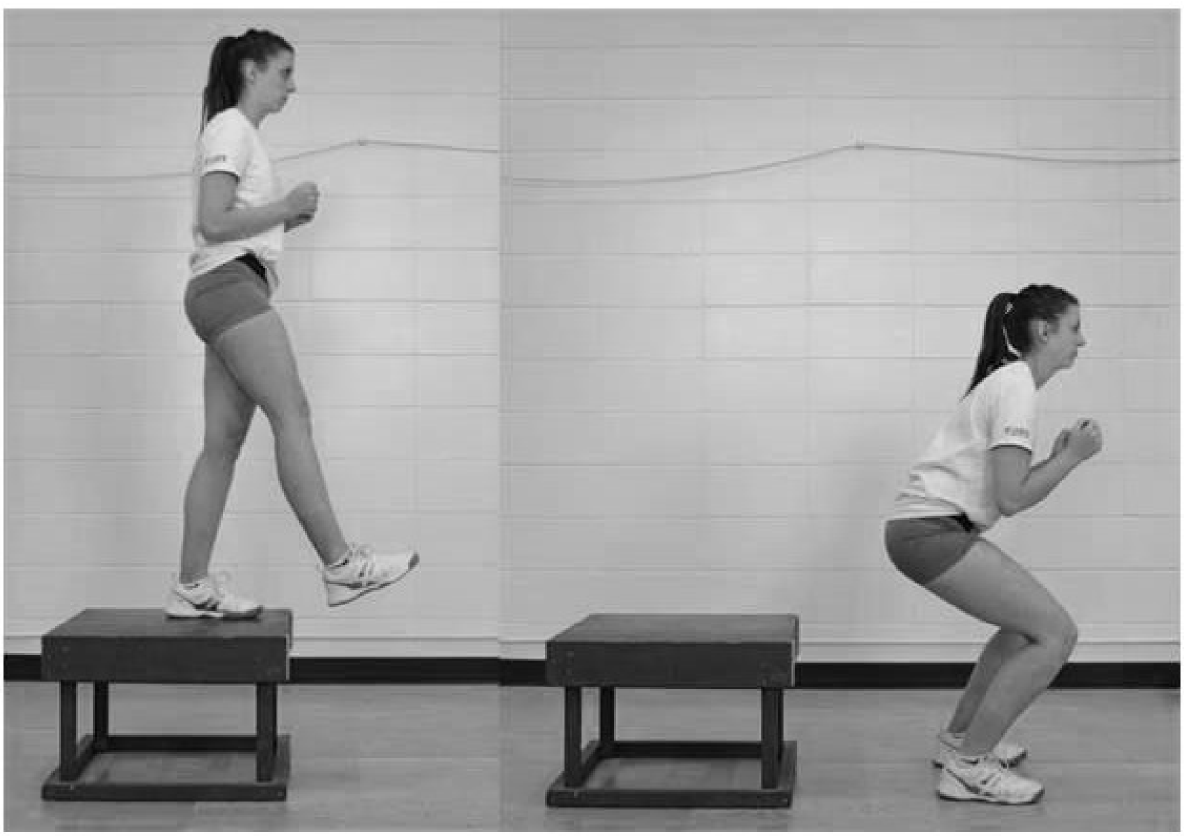

Figure 4.2 Image showing box landing; non-involved (right foot) step off with two-foot landing.

The box landing task incorporated standardized equipment (40 cm platform), in order to observe the mechanics for all individuals landing from the same height. This was important due to the large variability in jump height between participants in the maximal effort jump landing. For the box landing task (six repetitions for each left and right step off), all participants stepped off a 40 cm platform and performed a double leg landing (Figure 4.2). Participants were instructed to step off the platform with one foot, land with two feet and to keep eyes forward or upward to avoid tracking the force platforms during landing. Arm motion and foot placement were not constrained, and participants landed on the force platforms during data collection. Three left foot step off trials were conducted, followed by three right foot step off trials, and then this order was repeated. Rest was provided ad libitum between repetitions.

4.2.3 Motion Analysis

A six degree-of-freedom marker set was used (Figure 4.1). This involved calibration markers to identify the proximal and distal ends of the segments and tracking markers during a standing calibration trial, and tracking markers only for the dynamic tasks. Marker trajectories were recorded using seven optoelectronic cameras (ProReflex MCU240; Qualisys, Gothenburg, Sweden) sampling at 120 Hz. Two AMTI force platforms (OR6-6; AMTI, Watertown, MA) sampling at 1200 Hz were used to collect ground reaction force data. The pelvis was modeled from proximal markers placed on the iliac crests and distal markers on the femoral greater trochanters. The thigh was modeled proximally from the femoral greater trochanter markers as one quarter the distance from the ipsilateral to the contralateral marker. The distal thigh was modeled as the mid-point between markers on the medial and lateral femoral condyles. Pelvis tracking markers included the iliac crests, anterior superior iliac spines, and the intervertebral disc at L5/S1, while clusters of four markers fixed to a molded thermoplastic plate were used to track the thigh during all tasks.

The leg was modeled proximally as the mid-point between the medial and lateral tibial condylar eminence markers. The distal leg was modeled as the mid-point between the medial and lateral malleolus. The foot was modeled from proximal markers placed on the medial and lateral malleoli and distal markers on the 1st and 5th metatarsal heads. Clusters of four (leg) and three (foot) markers were fixed to these segments with a molded thermoplastic plate to track them during all tasks. The longitudinal axes (Z) of all segments connected the proximal and distal ends of the segment. The sagittal (X) and frontal (Y) axes were orthogonal to each other and to the longitudinal axis. The segment coordinate system for each segment was located at the proximal end of the segment. All segments were modeled as conical frusta for analysis. Proximal and distal markers on segments were used to define appropriate segment length and the radii of each end of the frusta. Segment mass was determined as a percentage of total body mass using Dempster's anthropometric data (28).

To determine tibial torsion angle, the tibia's transcondylar axis was modeled using the proximal tibial markers and the transmalleolar axis modeled using the markers on the medial malleolus (tibia) and lateral malleolus (fibula). Two separate segments were created using motion analysis: a proximally biased leg segment and a distally biased leg segment (Figure 2.5). The proximally biased leg segment used the medial and lateral tibial markers as the proximal parameters and the mid-point between the malleoli as the distal parameters. The distally biased leg segment used the mid-point between the tibial condyles as the proximal parameters and the medial and lateral malleoli as the distal parameters. A comparison of these proximally and distally biased leg segments in the transverse plane about the longitudinal (Z) axis was reported as tibial torsion angle.

Reflective marker data were processed in Visual 3D software (Version 4.82; C-Motion, Germantown, MD) to determine segment and joint kinematics. Marker and force platform data were filtered using a low-pass fourth order recursive Butterworth with an 8 Hz cut-off frequency. Segment rotations were determined using a ZYX Cardan sequence relative to the laboratory reference frame to calculate mechanical rotations consistent with anatomically defined sagittal (X), frontal (Y) and transverse (Z) planes. Joint rotations were determined using an XYZ Cardan sequence with the distal segment as the reference segment. All coordinate systems conformed to the right hand rule. For left limb data, the signs for rotations about the Y and Z axes were reversed to conform to the right limbs. Positive segment rotations are dorsiflexion (X), inversion (Y) and internal rotation (Z) [foot]; plantarflexion/extension (X), eversion/adduction (Y) and medial rotation (Z) [leg]; flexion (X), adduction (Y) and medial rotation (Z) [thigh]; and extension (X), ipsilateral obliquity (Y) and contralateral axial rotation (Z) [pelvis]. Positive joint rotations are plantarflexion (X), eversion (Y) and external rotation (Z) [ankle]; flexion (X), abduction (Y) and external rotation (Z) [knee]; and extension (X), abduction (Y) and external rotation (Z) [hip]. Inverse dynamics procedures were used to calculate the internal net joint moment (NJM) at the ankle, knee, and hip with moments expressed in the coordinate system of the distal segments.

4.2.4 Data Reduction

For the partial squat, data was extracted based on the following events: standing (baseline), peak leg dorsiflexion, and peak leg abduction. Transverse plane leg excursion was measured from standing to peak leg dorsiflexion. Internal net joint moment metrics for the partial squat were measured at peak leg dorsiflexion. The two-foot jump and box landing data were recovered based on the following events: ground contact, foot flat, peak knee flexion, peak leg dorsiflexion, and peak leg abduction. Foot progression angles were evaluated at ground contact and foot flat. Transverse plane leg excursions were measured from ground contact to foot flat, and foot flat to peak knee flexion. Internal net joint moment metrics for the two-foot jump and box landings were measured at foot flat and peak knee flexion.

4.2.5 Statistical Analysis

To investigate the effect of tibial torsion on lower limb kinematics and kinetics, a two-way mixed ANOVA was performed on this data using case-control group (ACL injured vs. high torsion control vs. low torsion control) as the between subjects variable and side (involved/left vs. non-involved/right) as the within subjects repeated measures variable for squatting and landing tasks. For this project, involved will refer to the injured limb for ACL participants and left limb for control participants. Non-involved will refer to the non-injured limb for ACL participants and right limb for control participants. Scheffé's method was used for ANOVA post hoc comparisons to help control Type 1 error, and was determined most appropriate due to its conservative method of controlling experiment wide error rate with multiple comparisons. Statistical calculations were performed in SPSS (Version 11.0; SPSS Inc. Chicago, IL), and alpha was set *a priori* (α=0.05).

4.3 Results

The complete results from the ANOVA statistical analysis and post hoc tests are summarized in: Table 4.1 (weight bearing partial squat to maximum leg dorsiflexion), Table 4.2 (two-foot jump landing), Table 4.3 (involved limb step off box landing), Table 4.4 (non-involved limb step off box landing), and Table 4.5 (sagittal plane hip, knee and ankle internal net joint moments in two-foot jump landings and step off box landings). Effect size, which expresses the mean difference between two groups in standard deviation units, is also reported alongside key differences in the results. The meaning of effect size varies by context, but the standard interpretation offered by Cohen (1988) is: d = 0.2 = small (1/5 of a standard deviation), d = 0.5 =

moderate (1/2 of a standard deviation), and d = 0.8 = large (4/5 of a standard deviation unit). Data from key dependent variables was also plotted against external tibial torsion by group and are summarized in Figures 4.3-4.10.

There was a significant difference in external tibial torsion between groups as determined by two-way mixed ANOVA ($F(2,45) = 70.387$, $p<0.001$). Post-hoc analysis revealed that external tibial torsion was significantly higher in the high torsion group (M=20.1 degrees, SD=4.2, median=18.8, range=17.4 to 31.7) than the ACL group (M=16.8 degrees, SD=3.1, median=16.1, range=10.8 to 20.4, $p=0.05$, $d=1.35$) and the low torsion group (M=6.6 degrees, SD=4.1, median=7.2, range=0.5 to 13.2, $p<0.001$, $d=3.77$). The ACL group was also significantly higher than the low torsion group ($p<0.001$, $d=2.76$). There was no difference between limbs for the low or high torsion groups, but the difference in tibial torsion between the involved (M=18.4 degrees, SD=2.9) and non-involved limb (M=15.2 degrees, SD=3.4) in the ACL group was found to be significant ($p<0.001$). Effect size ($d=1.0$) also suggested that the difference between involved and non-involved limbs had a high practical significance.

Table 4.1 Kinematic and kinetic characteristics of participants with unilateral ACL reconstruction, and control participants with high (>17 degrees) and low (<14 degrees) variations of external tibial torsion; weight bearing partial squat to maximum dorsiflexion.

Task - Squat		**ACL (n=14)**		**HTT (n=14)**		**LTT (n=20)**		**Group**		**Side**		**Interaction**		**Post Hoc**
Variables (units)		**INV**	**NON**	**INV**	**NON**	**INV**	**NON**	***F*** **(2,45)**	***p***	***F*** **(1,45)**	***p***	***F*** **(2,45)**	***p***	**Testing***
External tibial torsion	$\bar{x}$	18.4	15.2	19.9	20.4	6.5	6.7	70.387	< 0.001	2.901	0.095	5.027	0.011	HTT > ACL > LTT
(degrees)	SD	2.9	3.4	3.8	4.5	4.8	3.4							(ACL) INV > NON
Foot progression angle	$\bar{x}$	6.8	8.0	8.0	7.8	7.1	7.5	0.464	0.632	1.136	0.292	0.548	0.582	
(degrees)	SD	3.1	3.4	2.4	2.0	2.8	3.4							
Leg excursion (transverse)	$\bar{x}$	0.03	3.2	0.05	0.6	4.2	4.0	6.685	0.003	11.528	0.001	9.092	< 0.001	LTT > HTT
(degrees; + medial, - lateral)	SD	2.2	2.8	4.1	3.1	3.5	3.5							(ACL) NON > INV
Peak leg dorsiflexion	$\bar{x}$	30.2	28.7	33.6	34.3	32.6	32.6	1.298	0.283	0.681	0.414	3.585	0.036	
(degrees)	SD	6.8	5.5	7.8	6.9	8.9	8.4							
Peak leg abduction	$\bar{x}$	5.7	5.1	6.4	4.4	6.0	5.0	0.011	0.989	11.870	0.001	1.403	0.256	(HTT) INV > NON
(degrees)	SD	2.2	2.3	3.2	3.2	3.0	3.0							
Ankle plantarflexor moment	$\bar{x}$	0.22	0.20	0.28	0.28	0.26	0.26	1.761	0.184	0.648	0.425	1.421	0.252	
(N·m/kg body mass)	SD	0.09	0.10	0.09	0.08	0.13	0.12							
Ankle evertor moment	$\bar{x}$	0.03	0.01	0.10	0.05	0.01	0.00	5.957	0.005	5.626	0.022	1.852	0.169	HTT > ACL,LTT
(N·m/kg body mass)	SD	0.06	0.04	0.08	0.11	0.06	0.06							(HTT) INV > NON
Ankle lateral rotator moment	$\bar{x}$	0.07	0.06	0.08	0.08	0.05	0.04	8.209	0.001	0.836	0.365	0.859	0.431	HTT > LTT
(N·m/kg body mass)	SD	0.12	0.03	0.02	0.04	0.03	0.03							

* ACL = ACL injured participants; HTT = high tibial torsion controls; LTT = low tibial torsion controls
INV: Involved limb in ACL participants, left limb in control participants
NON: Non-involved limb in ACL participants, right limb in control participants

Table 4.2 Kinematic and kinetic characteristics of participants with unilateral ACL reconstruction, and control participants with high (>17 degrees) and low (<14 degrees) variations of external tibial torsion; two-foot jump landing.

Task - Jump landing		ACL (n=14)		HTT (n=14)		LTT (n=20)		Group		Side		Interaction		Post Hoc
Variables (units)		**INV**	**NON**	**INV**	**NON**	**INV**	**NON**	***F* (2,45)**	***p***	***F* (1,45)**	***p***	***F* (2,45)**	***p***	**Testing***
Foot progression angle at GC	x̄	15.4	14.7	18.3	22.5	12.5	14.5	7.105	0.002	2.678	0.109	1.405	0.256	HTT > ACL,LTT
(degrees)	SD	5.4	5.3	3.4	8.7	4.5	9.3							
Foot progression angle at FF	x̄	14.3	14.9	16.4	19.5	12.6	14.0	4.514	0.016	2.742	0.105	0.480	0.622	HTT > LTT
(degrees)	SD	4.2	5.2	4.0	7.4	3.7	8.0							
Leg excursion (transverse) GC→FF	x̄	2.5	2.5	1.9	1.9	6.9	6.2	17.225	< 0.001	0.207	0.651	0.197	0.822	LTT > ACL,HTT
(degrees; +medial/-lateral)	SD	4.1	2.7	3.4	2.2	3.3	3.0							
Leg excursion (transverse) FF→PKF	x̄	0.8	1.0	0.7	1.1	0.03	-0.2	2.002	0.147	0.140	0.710	0.300	0.742	
(degrees; +medial/-lateral)	SD	1.5	2.3	1.8	2.0	1.4	2.7							
Peak leg dorsiflexion	x̄	37.6	37.8	35.3	37.0	36.1	36.7	0.547	0.583	2.353	0.132	0.621	0.542	
(degrees)	SD	3.3	4.0	4.8	4.9	4.1	6.1							
Peak leg abduction	x̄	7.8	7.5	11.5	9.9	6.7	7.1	5.172	0.010	2.243	0.141	3.047	0.057	HTT > ACL,LTT
(degrees)	SD	3.1	2.8	3.4	4.0	4.2	3.9							
Leg abduction at GC	x̄	8.0	8.1	10.2	8.6	8.6	7.9	3.536	0.037	4.972	0.031	1.926	0.158	HTT > ACL,LTT
(degrees)	SD	1.9	1.2	2.1	1.9	1.7	2.1							(HTT) INV > NON
Ankle lateral rotation at GC	x̄	9.9	10.3	11.0	15.3	3.0	6.1	14.761	< 0.001	5.179	0.028	0.979	0.383	ACL,HTT > LTT
(degrees)	SD	5.6	6.1	7.8	5.5	4.6	6.7							(HTT) NON > INV
Ankle plantarflexor moment at FF	x̄	1.07	1.06	0.89	1.02	1.23	1.27	7.143	0.002	5.067	0.029	2.347	0.107	LTT > HTT
(N·m/kg body mass)	SD	0.23	0.19	0.20	0.19	0.24	0.34							(HTT) NON > INV
Ankle plantarflexor moment at PKF	x̄	0.67	0.60	0.58	0.66	0.66	0.69	0.459	0.635	0.316	0.577	4.416	0.018	
(N·m/kg body mass)	SD	0.15	0.14	0.22	0.20	0.18	0.25							
Ankle evertor moment at FF	x̄	0.27	0.24	0.31	0.26	0.17	0.19	3.776	0.030	1.007	0.321	1.185	0.315	HTT > LTT
(N·m/kg body mass)	SD	0.17	0.10	0.19	0.09	0.12	0.10							
Ankle evertor moment at PKF	x̄	0.14	0.13	0.19	0.18	0.09	0.10	6.802	0.003	0.010	0.922	0.214	0.808	HTT > ACL,LTT
(N·m/kg body mass)	SD	0.07	0.06	0.10	0.10	0.07	0.07							
Ankle lateral rotator moment at FF	x̄	0.16	0.13	0.18	0.17	0.12	0.12	0.113	0.092	0.214	0.646	0.060	0.942	
(N·m/kg body mass)	SD	0.08	0.06	0.13	0.17	0.08	0.11							
Ankle lateral rotator moment at PKF	x̄	0.09	0.06	0.09	0.09	0.07	0.03	2.911	0.065	4.203	0.046	1.056	0.356	(ACL,LTT) INV > NON
(N·m/kg body mass)	SD	0.06	0.05	0.06	0.08	0.05	0.07							

* ACL = ACL injured participants; HTT = high tibial torsion controls; LTT = low tibial torsion controls
GC = Ground contact; FF = Foot flat; PKF = Peak knee flexion
INV: Involved limb in ACL participants, left limb in control participants
NON: Non-involved limb in ACL participants, right limb in control participants

Table 4.3 Kinematic and kinetic characteristics of participants with unilateral ACL reconstruction, and control participants with high (>17 degrees) and low (<14 degrees) variations of external tibial torsion; involved limb step off box landing.

Task - Involved step off box landing		**ACL (n=14)**		**HTT (n=14)**		**LTT (n=20)**		**Group**		**Side**		**Interaction**		**Post Hoc**
Variables (units)		**INV**	**NON**	**INV**	**NON**	**INV**	**NON**	***F*** **(2,45)**	***p***	***F*** **(1,45)**	***p***	***F*** **(2,45)**	***p***	**Testing***
Foot progression angle at GC	$\bar{x}$	12.1	16.5	14.7	21.3	9.2	18.4	4.601	0.015	48.230	< 0.001	2.132	0.130	(ALL) NON > INV
(degrees)	SD	4.2	5.8	5.4	5.1	4.8	6.3							
Foot progression angle at FF	$\bar{x}$	10.2	10.2	13.9	15.8	8.5	13.8	4.378	0.018	9.061	0.004	4.132	0.023	HTT > ACL
(degrees)	SD	4.1	4.3	6.2	4.4	4.5	6.8							(LTT) NON > INV
Leg excursion (transverse) GC→FF	$\bar{x}$	4.5	6.5	3.0	6.7	5.8	8.7	3.225	0.049	32.592	< 0.001	0.916	0.407	LTT > ACL,HTT
(degrees; +medial/-lateral)	SD	3.9	3.2	3.3	3.0	3.1	3.5							(ALL) NON > INV
Leg excursion (transverse) FF→PKF	$\bar{x}$	-1.1	0.1	-1.7	0.4	-1.8	1.0	0.042	0.959	10.980	0.002	0.652	0.526	(HTT,LTT) NON > INV
(degrees; +medial/-lateral)	SD	3.5	3.4	2.7	3.2	2.9	3.0							
Peak leg dorsiflexion	$\bar{x}$	34.7	35.2	34.6	36.2	33.9	35.4	0.059	0.943	2.797	0.101	0.230	0.795	
(degrees)	SD	5.9	6.2	4.8	5.2	7.9	8.5							
Peak leg abduction	$\bar{x}$	7.9	8.3	10.1	8.9	7.8	7.9	0.808	0.452	0.200	0.657	0.917	0.407	
(degrees)	SD	3.9	4.1	3.9	3.6	4.1	5.0							
Leg abduction at GC	$\bar{x}$	7.8	4.4	9.6	4.0	8.7	5.1	0.763	0.472	88.629	< 0.001	2.301	0.112	(ALL) INV > NON
(degrees)	SD	2.3	2.3	2.5	1.9	2.5	3.1							
Ankle lateral rotation at GC	$\bar{x}$	9.7	10.5	10.5	12.4	4.8	6.2	8.698	0.001	1.534	0.222	0.074	0.929	ACL,HTT > LTT
(degrees)	SD	5.4	4.7	6.8	6.5	4.8	6.5							
Ankle plantarflexor moment at FF	$\bar{x}$	1.03	0.91	0.89	0.75	1.17	0.78	2.097	0.135	21.378	< 0.001	3.937	0.027	(ALL) INV > NON
(N·m/kg body mass)	SD	0.32	0.26	0.33	0.22	0.38	0.15							
Ankle plantarflexor moment at PKF	$\bar{x}$	0.65	0.60	0.62	0.58	0.73	0.66	1.166	0.321	5.366	0.025	0.113	0.894	(LTT) INV > NON
(N·m/kg body mass)	SD	0.23	0.22	0.20	0.21	0.22	0.14							
Ankle evertor moment at FF	$\bar{x}$	0.27	0.20	0.25	0.14	0.17	0.09	4.354	0.019	12.418	0.001	0.450	0.640	ACL,HTT > LTT
(N·m/kg body mass)	SD	0.16	0.10	0.16	0.15	0.11	0.13							(ALL) INV > NON
Ankle evertor moment at PKF	$\bar{x}$	0.16	0.13	0.18	0.15	0.10	0.12	1.807	0.176	0.703	0.406	1.215	0.306	
(N·m/kg body mass)	SD	0.11	0.08	0.12	0.14	0.08	0.08							
Ankle lateral rotator moment at FF	$\bar{x}$	0.16	0.07	0.19	0.03	0.16	0.02	0.923	0.405	56.709	< 0.001	1.532	0.227	(ALL) INV > NON
(N·m/kg body mass)	SD	0.07	0.06	0.11	0.08	0.11	0.05							
Ankle lateral rotator moment at PKF	$\bar{x}$	0.07	0.04	0.09	0.04	0.07	0.03	0.743	0.481	14.321	< 0.001	0.158	0.854	(ALL) INV > NON
(N·m/kg body mass)	SD	0.07	0.06	0.04	0.08	0.04	0.04							

* ACL = ACL injured participants; HTT = high tibial torsion controls; LTT = low tibial torsion controls
GC = Ground contact; FF = Foot flat; PKF = Peak knee flexion
INV: Involved limb in ACL participants, left limb in control participants
NON: Non-involved limb in ACL participants, right limb in control participants

Table 4.4 Kinematic and kinetic characteristics of participants with unilateral ACL reconstruction, and control participants with high (>17 degrees) and low (<14 degrees) variations of external tibial torsion; non-involved limb step off box landing.

Task - Non-involved step off box landing		ACL (n=14)		HTT (n=14)		LTT (n=20)		Group		Side		Interaction		Post Hoc
Variables (units)		**INV**	**NON**	**INV**	**NON**	**INV**	**NON**	***F* (2,45)**	***p***	***F* (1,45)**	***p***	***F* (2,45)**	***p***	**Testing***
Foot progression angle at GC	X̄	16.9	10.7	20.2	16.3	15.7	13.0	4.411	0.018	17.967	< 0.001	1.106	0.340	HTT > ACL,LTT
(degrees)	SD	5.6	4.4	3.8	5.9	7.3	5.2							(ALL) INV > NON
Foot progression angle at FF	X̄	10.4	10.0	13.0	15.2	10.0	12.4	2.596	0.086	3.283	0.077	1.291	0.285	
(degrees)	SD	4.8	4.3	3.9	6.4	5.2	6.6							
Leg excursion (transverse) GC→FF	X̄	7.0	5.3	6.4	2.3	8.7	5.0	3.039	0.058	25.786	< 0.001	1.338	0.273	(ALL) INV > NON
(degrees; +medial/-lateral)	SD	4.3	4.1	3.2	3.2	3.7	3.4							
Leg excursion (transverse) FF→PKF	X̄	-0.9	-1.2	0.0	-1.1	1.7	-1.2	1.443	0.247	6.908	0.012	2.247	0.117	(HTT,LTT) INV > NON
(degrees; +medial/-lateral)	SD	3.7	3.1	2.9	2.3	2.4	2.8							
Peak leg dorsiflexion	X̄	35.3	35.3	33.7	35.4	34.4	36.4	0.093	0.911	7.405	0.009	1.727	0.189	(HTT,LTT) NON > INV
(degrees)	SD	6.2	5.9	5.8	4.9	6.3	6.9							
Peak leg abduction	X̄	8.0	8.4	9.7	5.3	8.9	7.4	0.211	0.811	10.472	0.002	5.299	0.009	(HTT,LTT) INV > NON
(degrees)	SD	3.5	3.7	3.8	2.1	3.9	4.4							
Leg abduction at GC	X̄	4.2	8.0	5.4	6.7	5.7	7.4	0.336	0.716	37.366	< 0.001	3.864	0.028	(ALL) NON > INV
(degrees)	SD	1.9	1.4	2.2	2.3	3.0	2.2							
Ankle lateral rotation at GC	X̄	11.6	10.9	10.5	14.6	4.1	8.1	9.017	0.001	4.616	0.037	1.870	0.166	ACL,HTT > LTT
(degrees)	SD	6.0	5.5	9.4	5.1	4.0	6.2							(HTT,LTT) NON > INV
Ankle plantarflexor moment at FF	X̄	0.92	1.12	0.80	0.93	0.92	1.20	3.844	0.029	34.328	< 0.001	1.324	0.276	ACL,LTT > HTT
(N·m/kg body mass)	SD	0.21	0.23	0.15	0.26	0.13	0.33							(ALL) NON > INV
Ankle plantarflexor moment at PKF	X̄	0.66	0.67	0.55	0.66	0.77	0.77	5.148	0.010	2.663	0.110	2.354	0.107	LTT > HTT
(N·m/kg body mass)	SD	0.13	0.15	0.16	0.28	0.11	0.20							
Ankle evertor moment at FF	X̄	0.19	0.29	0.16	0.23	0.08	0.16	7.333	0.002	21.360	< 0.001	0.183	0.834	ACL > HTT > LTT
(N·m/kg body mass)	SD	0.10	0.13	0.08	0.12	0.10	0.12							(ALL) NON > INV
Ankle evertor moment at PKF	X̄	0.17	0.15	0.20	0.20	0.08	0.10	6.670	0.003	0.003	0.954	0.479	0.623	ACL,HTT > LTT
(N·m/kg body mass)	SD	0.09	0.09	0.10	0.16	0.07	0.09							
Ankle lateral rotator moment at FF	X̄	0.07	0.16	0.10	0.16	0.06	0.09	5.032	0.011	13.231	0.001	1.486	0.241	HTT > LTT
(N·m/kg body mass)	SD	0.05	0.05	0.11	0.08	0.08	0.08							(ALL) NON > INV
Ankle lateral rotator moment at PKF	X̄	0.06	0.05	0.06	0.09	0.06	0.05	0.528	0.593	0.018	0.893	0.961	0.390	(HTT) NON > INV
(N·m/kg body mass)	SD	0.05	0.07	0.09	0.08	0.05	0.07							

* ACL = ACL injured participants; HTT = high tibial torsion controls; LTT = low tibial torsion controls
GC = Ground contact; FF = Foot flat; PKF = Peak knee flexion
INV: Involved limb in ACL participants, left limb in control participants
NON: Non-involved limb in ACL participants, right limb in control participants

Table 4.5 Hip, knee, ankle internal net joint moments (NJM) of participants with ACL reconstruction, and control participants with high (>17 degrees) and low (<14 degrees) variations of external tibial torsion; two-foot jump and box landings.

Task - Landings Variables (units)	ACL (n=14)	HTT (n=14)	LTT (n=20)	Group *F*(2,45)	 *p*	Post Hoc Testing*
Ankle plantar flexor NJM (N·m/kg body mass) at FF						
Jump landing (two-foot landing, involved limb)	1.07 ± 0.21	0.89 ± 0.20	1.23 ± 0.24	9.143	0.001	LTT > HTT
Jump landing (two-foot landing, non-involved limb)	1.06 ± 0.19	1.02 ± 0.19	1.27 ± 0.34	6.954	0.003	LTT > ACL,HTT
Box landing (involved limb step off, involved limb)	1.03 ± 0.32	0.89 ± 0.33	1.17 ± 0.38	6.423	0.008	LTT > HTT
Box landing (non-involved limb step off, non-involved limb)	1.12 ± 0.23	0.93 ± 0.26	1.20 ± 0.33	4.257	0.022	LTT > HTT
Knee extensor NJM (N·m/kg body mass) at FF						
Jump landing (two-foot landing, involved limb)	1.22 ± 0.28	1.16 ± 0.19	0.99 ± 0.31	5.501	0.013	ACL > LTT
Jump landing (two-foot landing, non-involved limb)	1.08 ± 0.24	1.10 ± 0.23	1.06 ± 0.29	0.851	0.477	
Box landing (involved limb step off, involved limb)	1.40 ± 0.41	1.27 ± 0.32	1.19 ± 0.37	3.314	0.034	ACL > LTT
Box landing (non-involved limb step off, non-involved limb)	1.27 ± 0.27	1.22 ± 0.31	1.13 ± 0.33	2.446	0.098	
Hip extensor NJM (N·m/kg body mass) at FF						
Jump landing (two-foot landing, involved limb)	0.89 ± 0.18	0.94 ± 0.27	0.73 ± 0.23	3.637	0.031	HTT > LTT
Jump landing (two-foot landing, non-involved limb)	0.84 ± 0.22	0.90 ± 0.20	0.77 ± 0.29	1.168	0.289	
Box landing (involved limb step off, involved limb)	0.82 ± 0.28	0.84 ± 0.30	0.68 ± 0.28	2.521	0.089	
Box landing (non-involved limb step off, non-involved limb)	0.65 ± 0.25	0.68 ± 0.31	0.53 ± 0.19	2.939	0.076	

* ACL = ACL injured participants; HTT = high tibial torsion controls; LTT = low tibial torsion controls
Internal net joint moments were calculated at foot flat (FF) for two-foot jump and box landings
INV: Involved limb in ACL participants, left limb in control participants
NON: Non-involved limb in ACL participants, right limb in control participants

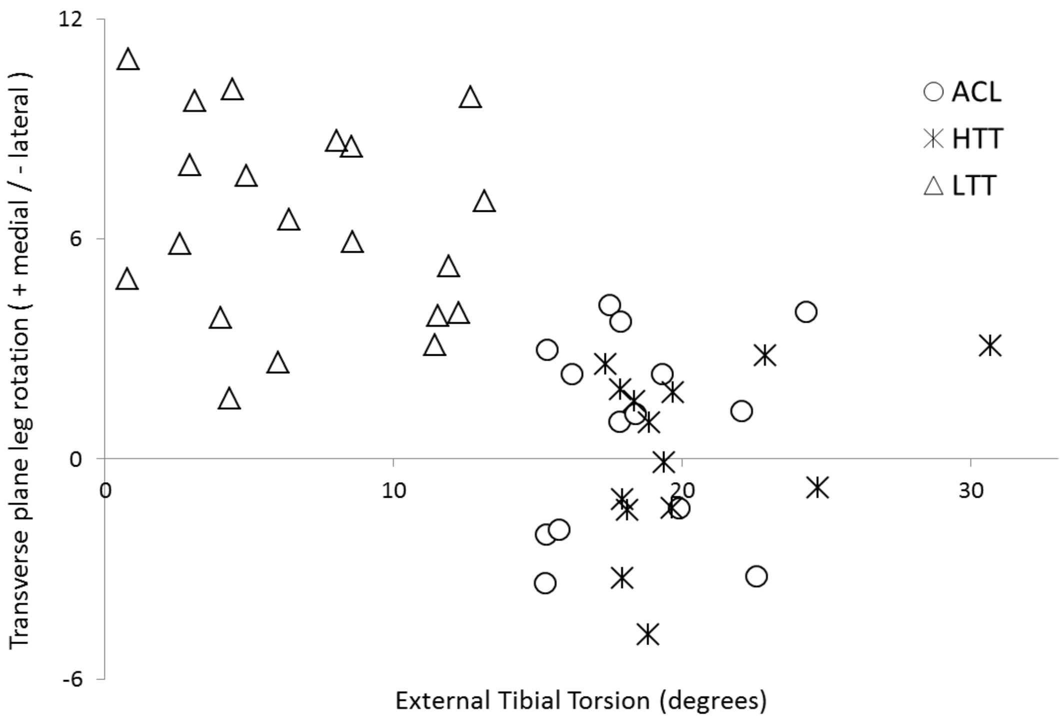

Figure 4.3 Scatter plot demonstrating left leg transverse plane leg rotation, from ground contact to foot flat, during jump landing (degrees) as a function of left leg external tibial torsion (degrees) for ACL injured participants, high torsion control participants, and low torsion control participants.

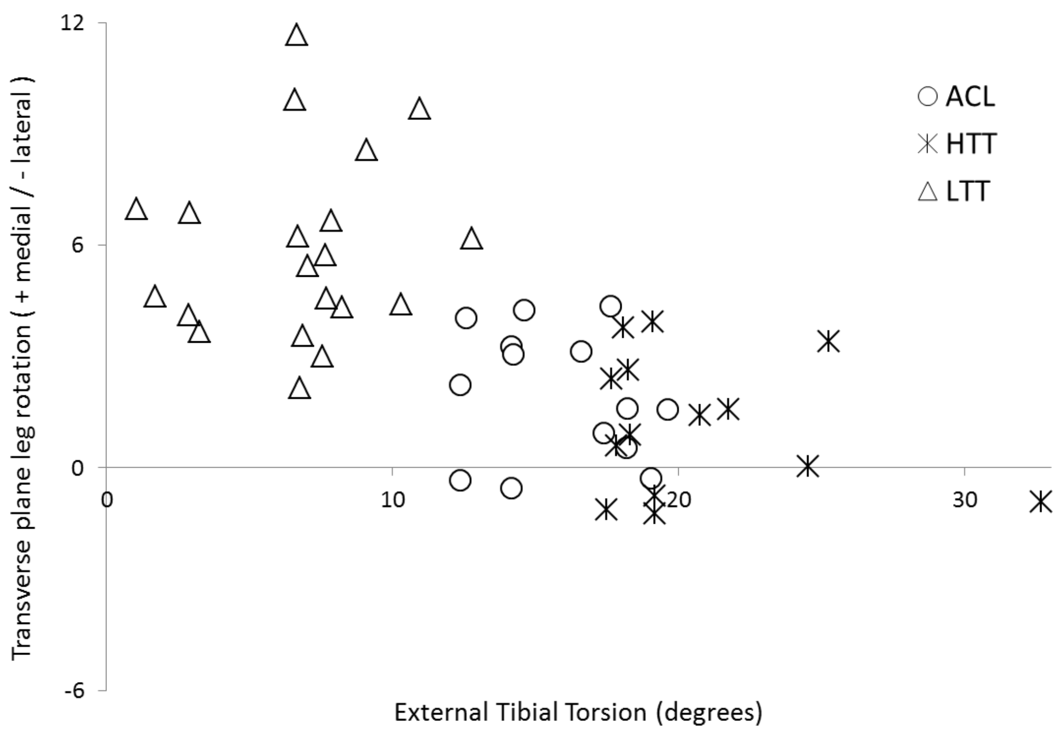

Figure 4.4 Scatter plot demonstrating right leg transverse plane leg rotation, from ground contact to foot flat, during jump landing (degrees) as a function of right leg external tibial torsion (degrees) for ACL injured participants, high torsion control participants, and low torsion control participants.

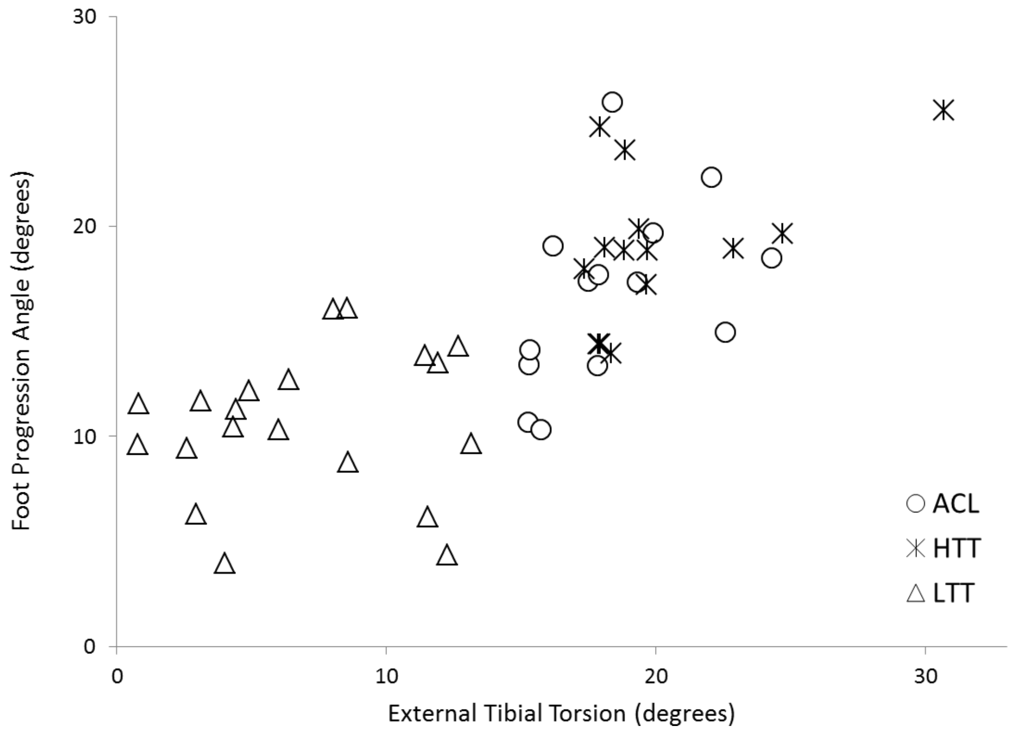

Figure 4.5 Scatter plot demonstrating left foot progression angle during jump landing (degrees) as a function of left leg external tibial torsion (degrees) for ACL injured participants, high torsion control participants, and low torsion control participants.

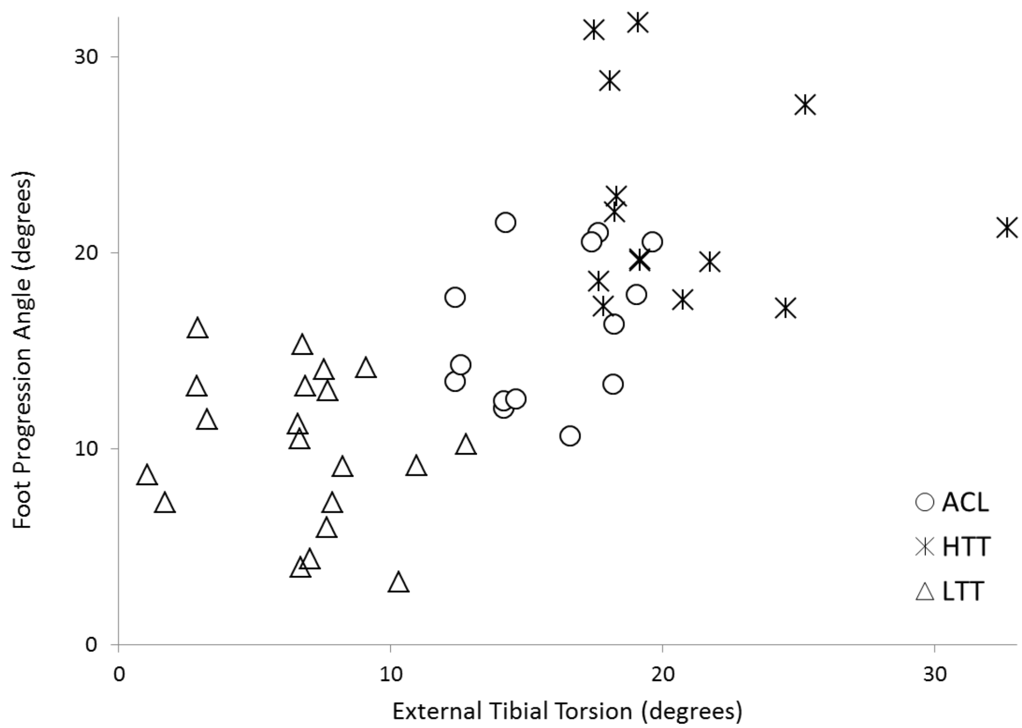

Figure 4.6 Scatter plot demonstrating right foot progression angle during jump landing (degrees) as a function of right leg external tibial torsion (degrees) for ACL injured participants, high torsion control participants, and low torsion control participants.

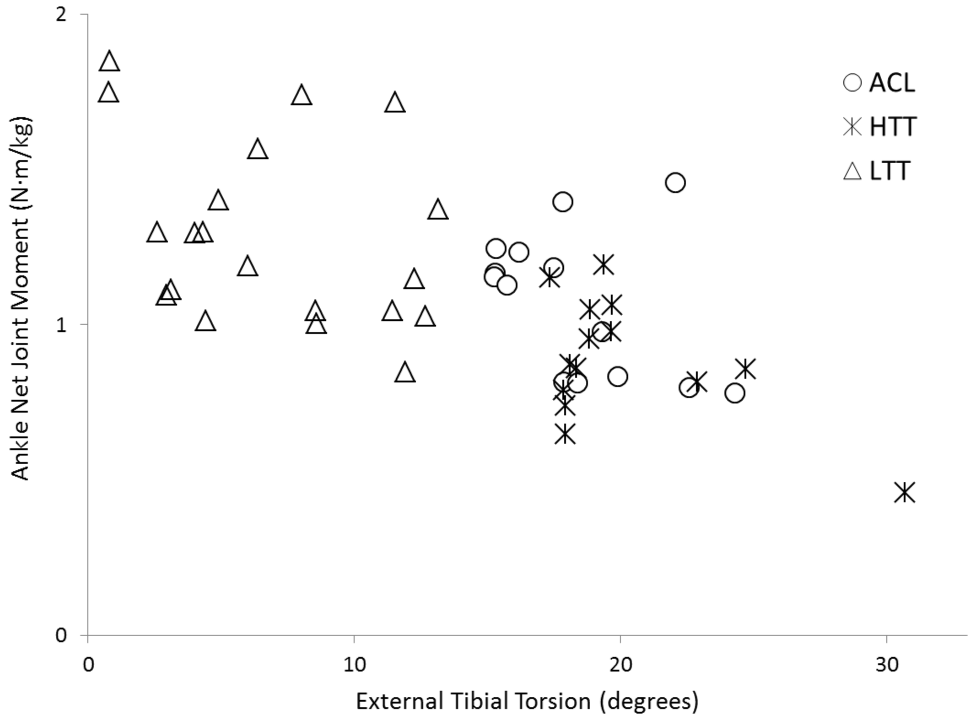

Figure 4.7 Scatter plot demonstrating left ankle plantar flexor net joint moment during jump landing (N·m/kg body weight) as a function of left leg external tibial torsion (degrees) for ACL injured participants, high torsion control participants, and low torsion control participants.

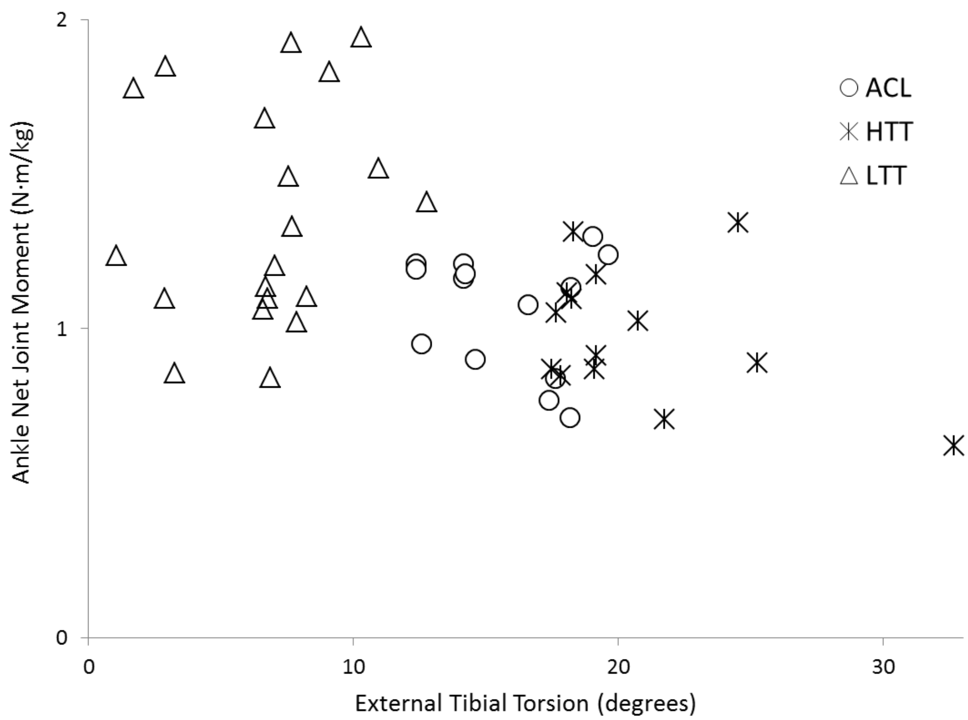

Figure 4.8 Scatter plot demonstrating right ankle plantar flexor net joint moment during jump landing (N·m/kg body weight) as a function of right leg external tibial torsion (degrees) for ACL injured participants, high torsion control participants, and low torsion control participants.

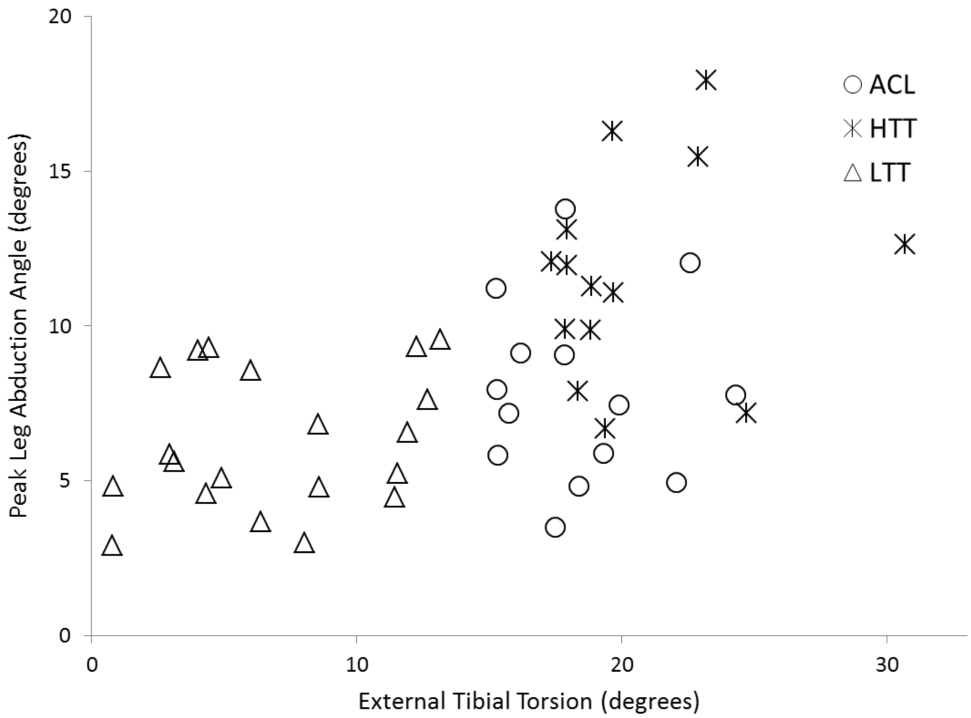

Figure 4.9 Scatter plot demonstrating left leg peak abduction angle during jump landing (degrees) as a function of left leg external tibial torsion (degrees) for ACL injured participants, high torsion control participants, and low torsion

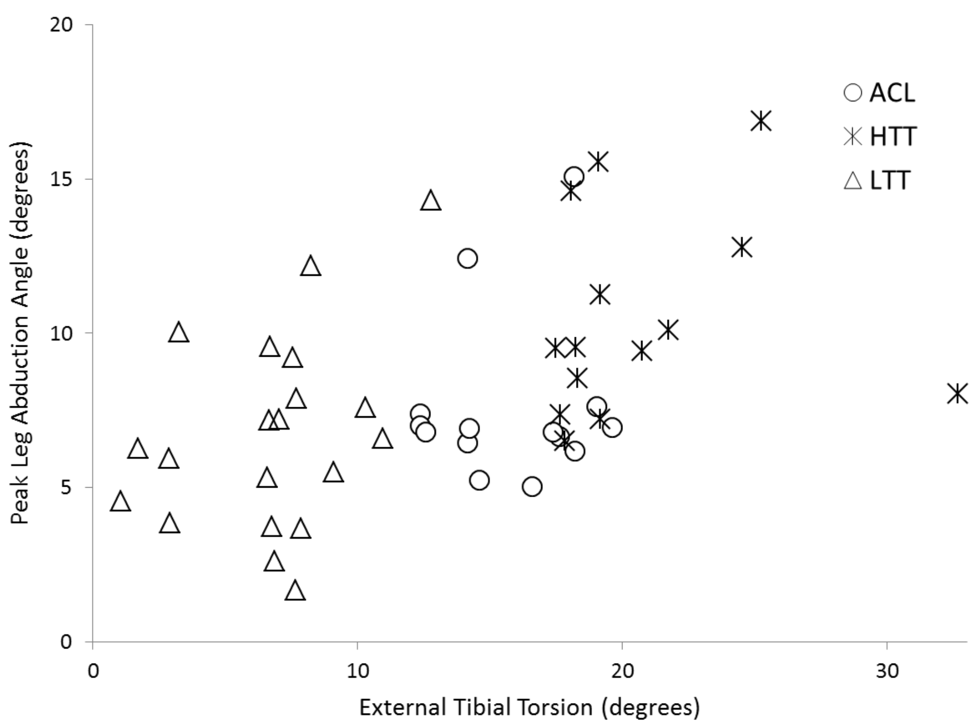

Figure 4.10 Scatter plot demonstrating right leg peak abduction angle during jump landing (degrees) as a function of right external tibial torsion (degrees) for ACL injured participants, high torsion control participants, and low torsion control participants.

4.3.1 Transverse Plane Leg and Foot Mechanics

During the partial squat, there was a group main effect for leg excursion in the transverse plane as determined by two-way mixed ANOVA ($F(2,45) = 6.685$, p=0.003). Post-hoc analysis revealed that transverse plane medial leg rotation was significantly higher in the low torsion group (4.1 ± 3.3 degrees) when compared to the high torsion group (0.3 ± 3.4 degrees, d=1.06), but not significantly higher when compared to the ACL group (1.6 ± 2.4 degrees, d=0.42). Within the ACL group, the non-involved limb had significantly higher leg medial rotation (3.2 ± 2.8 degrees) compared to the involved limb (0.03 ± 2.2 degrees, d=1.01). During the jump landing task, there was a group main effect for leg excursion in the transverse plane (from ground contact to foot flat), ($F(2,45) = 17.225$, $p<0.001$). Post hoc analysis revealed that medial leg rotation was significantly greater in the low torsion group (6.5 ± 3.2 degrees) compared to the high torsion group (1.9 ± 2.8 degrees, d=0.98) and the ACL group (2.5 ± 3.5 degrees, d=0.99). During the involved limb step off box landing, there was a group main effect for leg excursion in the transverse plane (from ground contact to foot flat), ($F(2,45) = 3.225$, p=0.049). Post hoc analysis revealed that medial leg rotation was significantly greater in the low torsion group (7.3 ± 3.3 degrees) compared to the high torsion group (4.9 ± 3.2 degrees, d=0.77) and the ACL group (5.5 ± 2.2 degrees, d=0.63).

During the jump landing, there was a group main effect for foot progression angle (at ground contact) as determined by two-way mixed ANOVA ($F(2,45) = 7.105$, p=0.002). Post-hoc analysis revealed that foot progression angle at ground contact was significantly greater in the high torsion group (20.4 ± 6.1 degrees) when compared to the ACL group (15.1 ± 5.4 degrees, d=0.74), and the low torsion group (13.5 ± 6.9 degrees, d=1.02). During the step-off box landing tasks, the trailing limb consistently showed significantly greater foot progression angles (18.8 ± 5.6 degrees) than the leading step-off limb (12.2 ± 4.7 degrees) in all groups.

4.3.2 Ankle Plantar Flexor, Knee Extensor and Hip Extensor Net Joint Moments

During the two-foot jump landing, there was a group main effect for ankle plantar flexor net joint moment (at foot flat), ($F(2,45) = 7.143$, $p=0.002$). Post hoc analysis revealed that ankle plantar flexor net joint moment was greater in the low torsion group (1.25 ± 0.29 N·m/kg body mass) when compared to the high torsion group (0.96 ± 0.20 N·m/kg body mass, $d=1.43$). During the non-involved limb step-off box landing, there was a group main effect for ankle plantar flexion net joint moment (at foot flat), ($F(2,45) = 3.844$, $p=0.029$). Post hoc analysis revealed that ankle plantar flexor net joint moment was greater in the low torsion group (1.20 ± 0.33 N·m/kg body mass) and the ACL group (1.12 ± 0.23 N·m/kg body mass) when compared to the high torsion group (0.93 ± 0.26 N·m/kg body mass).

When considering the more proximal internal net joint moments during the two-foot jump landing task, there was a group main effect for knee extensor net joint moment (at foot flat), ($F(2,45) = 5.501$, $p=0.013$). Post hoc analysis revealed that knee extensor net joint moment was significantly greater in the ACL group (1.22 ± 0.28 N·m/kg body mass) compared to the low torsion group (0.99 ± 0.31 N·m/kg body mass, $d=1.08$). During the involved limb step-off box landing, there was a group main effect for knee extensor net joint moment (at foot flat), ($F(2,45) = 3.314$, $p=0.034$). Post hoc analysis revealed that knee extensor net joint moment was greater in the ACL group (1.40 ± 0.41 N·m/kg body mass) compared to the low torsion group (1.19 ± 0.37 N·m/kg body mass, $d=0.88$). There was a significant group main effect on hip extensor net joint moment in the involved limb during the jump landing ($F(2,45) = 3.637$, $p=0.031$), where Scheffé post hoc testing revealed that hip extensor net joint moment was greater in the high torsion group (0.94 ± 0.27 N·m/kg body mass) compared to the low torsion group (0.73 ± 0.23 N·m/kg body mass, $d=0.81$). During the jump landing, there was a group main effect for hip extensor net joint moment

($F(2,45) = 3.637$, $p=0.031$). Post hoc analysis revealed that hip extensor net joint moment was greater in the high torsion group (0.94 ± 0.27 N·m/kg body mass) when compared to the low torsion group (0.73 ± 0.23 N·m/kg body mass, $d=0.89$).

4.3.3 Frontal Plane Leg Mechanics

During the two-foot jump landing, there was a group main effect for peak leg abduction ($F(2,45) = 5.172$, $p=0.01$) and leg abduction at ground contact ($F(2,45) = 3.536$, $p=0.037$). Post hoc analysis revealed that leg abduction was greater in the high torsion group (10.7 ± 3.7 degrees, 9.4 ± 2.0 degrees) compared to the low torsion group (6.9 ± 4.2 degrees, $d=0.54$, 8.3 ± 1.9 degrees, $d=0.33$) and the ACL group (7.6 ± 3.0 degrees $d=0.61$, 8.1 ± 1.6 degrees, $d=0.35$). During the step off box landings, the lead leg, whether involved or non-involved, always demonstrated significantly greater leg abduction at ground contact than the trailing leg.

4.4 Discussion

The study design grouped participants as ACL injured (n=14) or control participants without ACL injury (n=34). The control participants were then grouped by degree of external tibial torsion; high (n=14) or low (n=20). This grouping allowed the researcher to investigate the potential relation between magnitude of torsion and mechanics known to influence the risk of non-contact ACL injury. Theoretical rationale was based on previous research by this author (Chapters 2, 3) and limited research that has suggested that ACL injured persons may have increased external tibial torsion (29). The difference in external tibial torsion measured between the groups was as anticipated. The high torsion controls had greater tibial torsion than the ACL group, and the ACL group had greater tibial torsion than the low torsion controls. This was an important finding as it replicates previous work from the investigator (Chapter 2), and provides support for the grouping parameters used to classify participants as high or low in the control groups. It was interesting that the involved limbs in the ACL group displayed greater external tibial torsion than the non-involved limbs. This also provides further evidence to support previous findings (Chapter 3) and presents auxiliary evidence to consider tibial torsion as an important inter- and intra-individual anatomic risk factor for non-contact ACL rupture in women.

The first objective of the first hypothesis was to examine the potential influence of tibial torsion on transverse plane leg rotation during squatting and landing tasks. The second objective of the first hypothesis was to compare foot progression angles during the landing tasks between the groups. As hypothesized, ACL injured and high torsion participants had lesser medial leg rotation (or greater lateral leg rotation) for all tasks; and foot progression angles were also greatest in the landing tasks in the ACL injured and high torsion groups.

It appears that external tibial torsion is greater in ACL injured individuals (29). Moreover, for individuals with unilateral injury, external tibial torsion appears to be greater in the injured limb (Chapter 3). Increased tibial torsion is associated with less medial leg rotation, or even lateral leg rotation during human motion that requires weighted knee flexion and ankle dorsiflexion (squatting and landing). This variation is likely, in part, due to the altered line of pull of the posterior leg muscles (soleus, tibialis posterior, flexor digitorum longus, and flexor hallucis longus) in individuals with increasing magnitudes of external tibial torsion.

An association between external tibial torsion and leg lateral rotation has been previously established in clinical populations (30, 31), and children (32), as well as being implicated in ACL injured individuals (33). Theoretical rationale for the association is based on the premise that there is a progressive increase in the medial directed force vector from the posterior leg muscles as the magnitude of external tibial torsion increases. This increased medial-directed force vector has the ability to generate a lateral rotation moment in the leg segment (Figure 2.3). The low torsion group demonstrated the greatest medial leg excursion (or least lateral leg rotation) during all tasks. The ACL group demonstrated leg rotation that was similar to the high torsion group. However, this leg rotation in the ACL group was not always significantly greater than the low torsion group.

While many non-contact ACL injuries result from direct tensile loading, the ACL may also be injured due to impingement against the intercondylar notch (20, 34). Although external leg rotation does not appear to load the ACL with the same degree of loading as internal leg rotation (10, 35), ACL impingement injury is more likely to occur when the tibia is abducted and externally rotated (36). Individuals with smaller notch dimensions appear to be at greater risk of impingement injury than those with larger notch dimensions (37), and notch width index has been reported to be smaller in women (38). If external tibial torsion is related to lateral leg rotation, which is commonly demonstrated during landing or change of direction manoeuvres, then tibial torsion should be considered as a possible predictor for mechanics that increase the risk of ACL injury in women.

Foot progression angles were not different between groups during the squatting task, and only varied by 2 degrees (6-8 degrees of turnout). This was as expected since foot placement and rotational alignment were controlled for in the squat. When the control was removed in the landing tasks, participants' foot progression angles changed in accordance with their external tibial torsion. Specifically, individuals with greater external tibial torsion demonstrated increased foot progression angles at ground contact and foot flat. The greater foot progression angles may contribute to non-sagittal plane leg rotation when landing with a flexed knee (39). During this type of landing, combined tibial axial rotation and translation affects the force distribution of the ligamentous structures in the knee (40), and may result in subsequent transfer of higher loads to the ACL. Excessive transverse plane rotation, medial or lateral, with a flexed knee can lead to excessive stress on the ACL (41), which is a common mechanism of non-contact ACL injuries (42). This combination of kinematics may be further intensified with increased foot progression angles during landing, which appear to be increased in individuals with greater external tibial torsion. Taken together, individuals with increased foot progression angles, possibly mediated by external tibial torsion, are at a mechanical disadvantage for preventing ACL injury.

The objective of the second hypothesis was to examine the relation between internal plantar flexor net joint moments and tibial torsion during the squatting and landing tasks. As hypothesized, the low torsion group demonstrated the highest internal plantar flexor net joint moments for all landing tasks. The less dynamic nature of the squat task likely contributed to the non-significant differences in plantar flexor net joint moments; as ground reaction forces were likely not great enough to elicit group effects in sagittal moments. However, in the landing tasks, there did appear to be a shift in the ability of the mono-articular musculature to plantar flex the ankle as tibial torsion increased, as previously proposed (14, 15). If the posterior leg muscles have a decreased posterior directed force vector, this would help explain the reduced ability to generate an ankle plantar flexor net joint moment in humans with high external tibial torsion. Since the quadriceps muscles have an anterior directed force on the leg segment during landing (25), the reduced posterior soleus force may result in a higher net anterior leg force which increases tensile stress on the ACL. If there are less posterior directed forces on the leg segment to counteract the effect of the quadriceps, there is mechanical rationale to suggest that the ACL would be at greater risk of tensile rupture individuals with greater external tibial torsion.

Moreover, a decrease in the ability to generate an internal plantar flexor net joint moment may affect the muscular effort in the more proximal knee and hip joints. During landing tasks, the contribution of kinetic and potential energy to the total energy of the system is ultimately determined by the vertical displacement (jump height or box height). As the landing occurs, muscular work must be performed to reduce the vertical velocity of the falling body to zero. The best estimate of muscular work performed is often achieved by summing the work performed on the segments and the work performed by the muscle (43), where the work performed by the muscle can be estimated through inverse dynamic techniques. This means that for any given landing, a certain amount of muscular work is required.

If, by anatomic variation of external tibial torsion, the ankle plantar flexors' muscular effort on the leg segment is decreased, then it is plausible that the knee extensors and hip extensors would see an increase in muscular effort. Zhang (44) and Moolyk (24) have reported this relationship in landing and weightlifting tasks. Hashemi et al. (45) also reported that an increase in hip extensor and knee extensor net joint moments would contribute to the posterior force on the femur, and the anterior force on the tibia, thereby increasing the risk of ACL rupture. Internal ankle plantar flexor, knee extensor and hip extensor net joint moments for this investigation followed similar outcomes to these reports (Table 4.5). Low tibial torsion participants consistently demonstrated the highest internal plantar flexor net joint moments. Moreover, knee extensor and hip extensor net joint moments in the low torsion group were consistently lower, though not always significantly, than the ACL injured and high tibial torsion participants.

When summing the net joint moments within the study groups, similar totals were demonstrated, although greater extensor moments were observed at the knee and hip when ankle plantar flexor contribution decreased. This suggests that the relative contribution of knee and hip extensor musculature increases as ankle plantar flexor contribution decreases. This alteration of lower body muscular effort during landing means that the knee extensors (quadriceps) and hip extensors (gluteus maximus) may be more active in individuals with higher magnitudes of external tibial torsion when compared to those with low tibial torsion. If there is less posterior directed force on the leg segment (due to a reduction of the contribution of the mono-articular ankle plantar flexors), more anterior directed force on the leg segment (due to an increase of the contribution of the knee extensors), and more posterior directed force on the thigh segment (due to an increase of the contribution of the hip extensors), there is even further mechanical rationale to suggest that the ACL would be at greater risk of injury in individuals with greater external tibial torsion.

The objective of the third hypothesis was to examine the relation between frontal plane leg abduction and tibial torsion during the squatting and landing tasks. No differences were found between the groups during the squatting task. The less dynamic nature of the squat task and the constraints on foot placement may have contributed to the non-significant differences in frontal plane leg abduction. The results from the jump landing tasks were generally in support of the a priori hypothesis. The high torsion group demonstrated greater leg abduction angles than the ACL injured and low torsion groups during the jump landing. This discrepancy may be due to increased external tibial torsion in the high torsion group. With a decrease in the ability of the soleus muscle to plantar flex the leg, it is plausible to consider that the remaining mono-articular plantar flexors would have a greater contribution to sagittal force generation. Tibialis posterior, flexor digitorum longus and flexor hallucis longus all pass posterior to the medial malleolus to an eventual insertion on the base of the foot (navicular, base of the distal phalanges II-IV, base of distal phalanx I). In a limb with an increasing magnitude of tibial torsion, the force vector from these muscles would become increasingly medial and less inferior/posterior. A decreased ankle plantar flexor moment may be associated to an increase in leg abduction during landing, although this was not consistently demonstrated in the current investigation. A possible explanation could be unidentified individual differences in the static alignment of the proximal segments which may contribute to faulty dynamic alignment and influence rotational moments about the knee and ankle. Previous research has reported larger external knee valgus moments with increased external tibial torsion (18) and increased leg abduction (16, 17). Functional valgus collapse at the knee, also known as faulty dynamic alignment (46), during landing activities is a mechanism associated with non-contact ACL injury (47, 48). Leg abduction, along with tibial rotation, has also been reported more often in women than in men during landing activities (49, 50). Women have also demonstrated greater non-sagittal knee moments than men while performing activities that increase ACL risk of injury (51). There is enough evidence in the jump landing data to warrant further investigation in the effects of external tibial torsion on frontal plane motion known to increase risk of ACL injury.

The step off box landing results did not fully support the hypothesis of greater leg abduction with an increasing tibial torsion angle. Although leg abduction was consistently greater in the step off limb compared to the trailing limb, there was no significant relation between leg abduction and tibial torsion. This may have been partially due to the foot placement of the lead leg in the box landings, where the foot progression angles remained lower as the participants landed predominantly on the step off limb. The step-off box landings may not mimic functional single leg landing mechanics, as initially forecast by the investigation. It appears that further investigation is warranted to study the effects of tibial torsion on one-legged landings and how they influence frontal plane leg segment motion.

This investigation is limited in terms of generalizability to all non-contact ACL rupture, since many ACL injuries do not occur during the landings demonstrated. Hyperextension mechanisms, one-footed landings from a greater height, and plant and cut type manoeuvres are some of the activities where ACL injury occurs and must be investigated and compared to these results for a better understanding of how tibial torsion may contribute to ACL risk factors. This investigation also acknowledges the limitation that the controls and cases were not matched with respect to anthropometrics and exposure to activities (type, intensity and duration) that may have contributed to their ACL injury. Increased body mass index has been associated with increased risk of ACL rupture, and the greater time spent in activities known to contribute to ACL injury risk, the more likely ACL injury occurrence would be. Study design may have also introduced limitations in the interpretation, most notably dividing the control participants into high torsion and low torsion groups. This division of the control participants may have resulted in greater differences than if a single control group had been used. Lastly, manual measurements used in screening, and landmark identification used in the motion analysis process are subject to observer inconsistencies which may contribute to error in data interpretation.

4.5 Summary

ACL injured and high torsion participants consistently demonstrated greater medial leg rotation (or lesser lateral leg rotation) than low torsion participants during the squats and landings in this investigation. ACL injured and high torsion participants consistently demonstrated lesser internal ankle plantar flexor net joint moments than low torsion participants during the squats and landings in this investigation. High torsion participants demonstrated greater leg abduction during the jump landings than the ACL injured and low torsion participants. The mechanics demonstrated by the HTT and ACL injured groups were often similar and statistically different from the LTT group.

This suggests the variation in mechanics is related to a higher magnitude of external tibial torsion, and not caused by the ACL injury (or surgical intervention). This investigation highlights the importance of considering the inter- and intra-individual anatomic variation of tibial torsion as a tangible risk factor when considering non-contact ACL rupture in active women. This study identifies external tibial torsion as an important influence on the direction of future research in the area of preventing ACL injuries, or potential reduction of repeated injury risk in an ACL repaired limb.

4.6 References

1. Boden BP, Dean GS, Feagin JA,Jr, Garrett WE,Jr. Mechanisms of anterior cruciate ligament injury. Orthopedics. 2000 Jun;23(6):573-8.

2. Lohmander LS, Englund PM, Dahl LL, Roos EM. The long-term consequence of anterior cruciate ligament and meniscus injuries: osteoarthritis. Am J Sports Med. 2007 Oct;35(10):1756-69.

3. Salem GJ, Salinas R, Harding FV. Bilateral kinematic and kinetic analysis of the squat exercise after anterior cruciate ligament reconstruction. Arch Phys Med Rehabil. 2003 Aug;84(8):1211-6.

4. Hsiao SF, Chou PH, Hsu HC, Lue YJ. Changes of muscle mechanics associated with anterior cruciate ligament deficiency and reconstruction. J Strength Cond Res. 2014 Feb;28(2):390-400.

5. Rahnemai-Azar AA, Naendrup JH, Soni A, Olsen A, Zlotnicki J, Musahl V. Knee instability scores for ACL reconstruction. Curr Rev Musculoskelet Med. 2016 Mar 15.

6. Ireland ML, Ott SM. Special concerns of the female athlete. Clin Sports Med. 2004 Apr;23(2):281,98, vii.

7. Hewett TE, Myer GD, Ford KR, Heidt RS,Jr, Colosimo AJ, McLean SG, et al. Biomechanical measures of neuromuscular control and valgus loading of the knee predict anterior cruciate ligament injury risk in female athletes: a prospective study. Am J Sports Med. 2005 Apr;33(4):492-501.

8. Arendt E, Dick R. Knee injury patterns among men and women in collegiate basketball and soccer. NCAA data and review of literature. Am J Sports Med. 1995 Nov-Dec;23(6):694-701.

9. Feagin JA,Jr, Lambert KL. Mechanism of injury and pathology of anterior cruciate ligament injuries. Orthop Clin North Am. 1985 Jan;16(1):41-5.

10. Markolf KL, Burchfield DM, Shapiro MM, Shepard MF, Finerman GA, Slauterbeck JL. Combined knee loading states that generate high anterior cruciate ligament forces. J Orthop Res. 1995 Nov;13(6):930-5.

11. Berns GS, Hull ML, Patterson HA. Strain in the anteromedial bundle of the anterior cruciate ligament under combination loading. J Orthop Res. 1992 Mar;10(2):167-76.

12. Lorenz A, Rottgerkamp H, Bobrowitsch E, Leichtle CI, Leichtle UG. Tibial rotation influences anterior knee stability - a robot-aided in-vitro study. Clin Biomech (Bristol, Avon). 2016 Feb;32:131-7.

13. Hutter CG,Jr, Scott W. Tibial torsion. J Bone Joint Surg Am. 1949 Jul;31A(3):511-8.

14. Hicks J, Arnold A, Anderson F, Schwartz M, Delp S. The effect of excessive tibial torsion on the capacity of muscles to extend the hip and knee during single-limb stance. Gait and Posture. 2007;26(4):546-52.

15. Schwartz M, Lakin G. The effect of tibial torsion on the dynamic function of the soleus during gait. Gait Posture. 2003 4;17(2):113-8.

16. Kristianslund E, Faul O, Bahr R, Myklebust G, Krosshaug T. Sidestep cutting technique and knee abduction loading: implications for ACL prevention exercises. Br J Sports Med. 2014 May;48(9):779-83.

17. Kristianslund E, Krosshaug T. Comparison of drop jumps and sport-specific sidestep cutting: implications for anterior cruciate ligament injury risk screening. Am J Sports Med. 2013 Mar;41(3):684-8.

18. Turner MSMC, F.R.C.S. The Association Between Tibial Torsion and Knee Joint Pathology. Clinical Orthopaedics & Related Research. 1994 May;302:47-51.

19. Dodgin DA, De Swart RJ, Stefko RM, Wenger DR, Ko JY. Distal tibial/fibular derotation osteotomy for correction of tibial torsion: review of technique and results in 63 cases. J Pediatr Orthop. 1998 Jan-Feb;18(1):95-101.

20. Fung DT, Zhang LQ. Modeling of ACL impingement against the intercondylar notch. Clin Biomech (Bristol, Avon). 2003 Dec;18(10):933-41.

21. Fung DT, Hendrix RW, Koh JL, Zhang LQ. ACL impingement prediction based on MRI scans of individual knees. Clin Orthop Relat Res. 2007 Jul;460:210-8.

22. McCaw ST, DeVita P. Errors in alignment of center of pressure and foot coordinates affect predicted lower extremity torques. J Biomech. 1995 Aug;28(8):985-8.

23. Kovacs I, Tihanyi J, Devita P, Racz L, Barrier J, Hortobagyi T. Foot placement modifies kinematics and kinetics during drop jumping. Med Sci Sports Exerc. 1999 May;31(5):708-16.

24. Moolyk AN, Carey JP, Chiu LZ. Characteristics of Lower Extremity Work during the Impact Phase of Jumping and Weightlifting. J Strength Cond Res. 2013 Feb 25.

25. Renstrom P, Arms SW, Stanwyck TS, Johnson RJ, Pope MH. Strain within the anterior cruciate ligament during hamstring and quadriceps activity. Am J Sports Med. 1986 Jan-Feb;14(1):83-7.

26. Griffin LY, Agel J, Albohm MJ, Arendt EA, Dick RW, Garrett WE, et al. Noncontact anterior cruciate ligament injuries: risk factors and prevention strategies. J Am Acad Orthop Surg. 2000 May-Jun;8(3):141-50.

27. Chizewski MG, Chiu LZ. Contribution of calcaneal and leg segment rotations to ankle joint dorsiflexion in a weight-bearing task. Gait Posture. 2012 May;36(1):85-9.

28. Winter DA, Winter DA. Biomechanics and motor control of human movement. 2nd ed. New York: Wiley; 1990.

29. Griffin LY, Agel J, Albohm MJ, Arendt EA, Dick RW, Garrett WE, et al. Noncontact anterior cruciate ligament injuries: risk factors and prevention strategies. J Am Acad Orthop Surg. 2000 May-Jun;8(3):141-50.

30. Turner MS. The association between tibial torsion and knee joint pathology. Clin Orthop Relat Res. 1994 May;(302)(302):47-51.

31. Yagi TM, D. Tibial Torsion in Patients With Medial-Type Osteoarthrotic Knees. Clinical Orthopaedics & Related Research. 1994 May;302:52-6.

32. Lincoln TL, Suen PW. Common rotational variations in children. J Am Acad Orthop Surg. 2003 Sep-Oct;11(5):312-20.

33. Silvers HJ, Mandelbaum BR. Prevention of anterior cruciate ligament injury in the female athlete. Br J Sports Med. 2007 Aug;41 Suppl 1:i52-9.

34. LaPrade RF, Burnett QM,2nd. Femoral intercondylar notch stenosis and correlation to anterior cruciate ligament injuries. A prospective study. Am J Sports Med. 1994 Mar-Apr;22(2):198,202; discussion 203.

35. Miyasaka T, Matsumoto H, Suda Y, Otani T, Toyama Y. Coordination of the anterior and posterior cruciate ligaments in constraining the varus-valgus and internal-external rotatory instability of the knee. J Orthop Sci. 2002;7(3):348-53.

36. Fung DT, Hendrix RW, Koh JL, Zhang LQ. ACL impingement prediction based on MRI scans of individual knees. Clin Orthop Relat Res. 2007 Jul;460:210-8.

37. Ireland ML, Ballantyne BT, Little K, McClay IS. A radiographic analysis of the relationship between the size and shape of the intercondylar notch and anterior cruciate ligament injury. Knee Surg Sports Traumatol Arthrosc. 2001 Jul;9(4):200-5.

38. Souryal TO, Freeman TR. Intercondylar notch size and anterior cruciate ligament injuries in athletes. A prospective study. Am J Sports Med. 1993 Jul-Aug;21(4):535-9.

39. Tillman MD, Bauer JA, Cauraugh JH, Trimble MH. Differences in lower extremity alignment between males and females. Potential predisposing factors for knee injury. J Sports Med Phys Fitness. 2005 Sep;45(3):355-9.

40. Ahmed AM, Burke DL, Duncan NA, Chan KH. Ligament tension pattern in the flexed knee in combined passive anterior translation and axial rotation. J Orthop Res. 1992 Nov;10(6):854-67.

41. Coplan JA. Rotational motion of the knee: a comparison of normal and pronating subjects. J Orthop Sports Phys Ther. 1989;10(9):366-9.

42. Emerson RJ. Basketball knee injuries and the anterior cruciate ligament. Clin Sports Med. 1993 Apr;12(2):317-28.

43. Robertson DGE. Research methods in biomechanics. Champaign, IL: Human Kinetics; 2004.

44. Zhang SN, Bates BT, Dufek JS. Contributions of lower extremity joints to energy dissipation during landings. Med Sci Sports Exerc. 2000 Apr;32(4):812-9.

45. Hashemi J, Chandrashekar N, Mansouri H, Gill B, Slauterbeck JR, Schutt RC,Jr, et al. Shallow medial tibial plateau and steep medial and lateral tibial slopes: new risk factors for anterior cruciate ligament injuries. Am J Sports Med. 2010 Jan;38(1):54-62.

46. Nguyen AD, Shultz SJ, Schmitz RJ. Landing biomechanics in participants with different static lower extremity alignment profiles. J Athl Train. 2015 May;50(5):498-507.

47. Olsen OE, Myklebust G, Engebretsen L, Bahr R. Injury mechanisms for anterior cruciate ligament injuries in team handball: a systematic video analysis. Am J Sports Med. 2004 Jun;32(4):1002-12.

48. Ireland ML. Anterior cruciate ligament injury in female athletes: epidemiology. J Athl Train. 1999 Apr;34(2):150-4.

49. Shultz SJ, Schmitz RJ. Effects of transverse and frontal plane knee laxity on hip and knee neuromechanics during drop landings. Am J Sports Med. 2009 Sep;37(9):1821-30.

50. Hewett TE, Myer GD, Ford KR, Heidt RS,Jr, Colosimo AJ, McLean SG, et al. Biomechanical measures of neuromuscular control and valgus loading of the knee predict anterior cruciate ligament injury risk in female athletes: a prospective study. Am J Sports Med. 2005 Apr;33(4):492-501.

51. Sigward SM, Powers CM. The influence of gender on knee kinematics, kinetics and muscle activation patterns during side-step cutting. Clin Biomech (Bristol, Avon). 2006 Jan;21(1):41-8.

CHAPTER 5

General Discussion

5.1 Introduction

This project sought to investigate how an anatomical variation, increased external tibial torsion, would influence kinematic and kinetic parameters in human three-dimensional motion. Segment rotations were considered in conjunction with joint rotations, since this analysis may provide additional detail into the muscular contributions to movement. The author was particularly interested in non-sagittal plane segment rotations that could be explained by the variation in external tibial torsion between and within individual participants. The investigation ultimately undertook the objective to determine whether the kinematics and kinetics commonly reported as ACL injury risk factors, would be exhibited by participants with greater external tibial torsion.

5.2 Tibial Torsion Influences Non-Sagittal Motion

A challenge in evaluating motion analysis results for segment and joint rotations; and interpreting the causes of these rotations, is that muscle force vectors may be altered by anatomical structural variations. One such variation that has been shown to affect lower body motion is tibial torsion (1-3). Generated by a predominantly distal twisting of the tibia and fibula collective complex, tibial torsion is often described as the rotational alignment offset in the transverse plane of the proximal and distal leg (4). Although often evaluated in the presence of patellofemoral instability (5, 6), overuse injuries (7), and anterior knee pain (8, 9), the effect(s) of tibial torsion in otherwise healthy individuals is generally underreported. It was predicted from prior research (1,

10) combined with free body diagram analyses of the expected changes in soleus (and deep posterior leg) muscle force vectors, that greater external tibial torsion would increase leg lateral rotation (transverse plane), decrease internal ankle plantar flexor net joint moment, and increase leg abduction (frontal plane) during dynamic analysis. These altered mechanics were hypothesized to be mainly a result of the reduction in the posterior directed force of the posterior leg muscles because of the external tibial torsion. An association between external tibial torsion and the previously described kinematics has been established in clinical populations (11-13), but had not been investigated in healthy adults.

An examination of individuals with varying external tibial torsion performing a partial squat task (Chapter 2) revealed a relation between increasing magnitudes of torsion and non-sagittal plane motion in otherwise healthy individuals. Men and women with greater external tibial torsion showed increased lateral rotation of the leg and thigh. However, likely due to the lesser magnitude of ground reaction forces required for this particular task, frontal plane leg excursion was not related to varying degrees of tibial torsion. The link between tibial torsion and lateral leg rotation has been reported (14) in distal tibial de-rotational osteotomy cases, where the leg segment would actually change its direction of rotation (from lateral to medial) when the external tibial torsion was surgically altered (decreased). These initial results provided adequate evidence that tibial torsion influences non-sagittal joint rotations, and supported continuing the investigation in more dynamic tasks. The investigators felt that tibial torsion should be considered as a structural risk factor that may predispose individuals to injury-related lower body mechanics. Whereas internal tibial torsion tends to improve with time, external tibial torsion often worsens as the natural chronological progression appears to be towards increasing external tibial torsion (6). The ability to compensate for increased external tibial torsion likely depends on other anatomic variations such as subtalar inversion/eversion, or femoral anteversion/retroversion. The initial findings supported further research investigating tibial torsion in otherwise healthy individuals. More specifically, it was important to study the influence of tibial torsion on the prospective action of the posterior leg muscles and their corollary influence on the leg segment during dynamic tasks.

5.3 Tibial torsion and ACL Injury

Approximately 70% of ACL injuries are non-contact, and occur during side cutting manoeuvres or when landing from a height, such as a jump landing (15). Prevention of these injuries includes strategies that limit motion between the thigh and leg segments (knee joint) to protect the various tissues and ligaments from excessive loading. Two key factors in the stabilization of the knee are lower body muscle strength and muscle recruitment patterns (16). When considering single-leg landings, the ACL is primarily responsible for limiting anterior tibial translation if relevant muscles are not able to stabilize the knee (17). Muscles that cross the knee joint, such as the hamstrings and quadriceps, are known to play a major role in affecting anterior tibial translation and concomitant ACL strain. Multiple studies have demonstrated that the quadriceps, due their common insertion on the anterior tibia, may act to increase anterior tibial translation and ACL strain (18-21). These same reports consider the hamstrings as an ACL agonist, or muscles that restrict anterior tibial translation due to their posterior tibial insertions, and thereby assist in reducing tensile ACL stress.

The muscles in the posterior compartment of the leg are also active during these same activities known to place the ACL at risk, such as cutting manoeuvres and landing from a jump. Recent investigations have provided evidence which suggests that muscles which stabilize the ankle joint(s) may also contribute to knee stability and thereby influence the promotion or prevention of ACL injuries. Cadaver study by Elias et al. (22) concluded that soleus acts as an agonist, and gastrocnemius as an antagonist to the ACL. Sherbondy et al. (23) found, through arthrometer and cadaver specimens, that both soleus and gastrocnemius decreased anterior tibial translation in ACL injured and healthy knees. Theoretical reports (24) and in vivo experimental studies (25) have reported gastrocnemius as an ACL antagonist, although other in vitro experiments have suggested that gastrocnemius activation strains the posterior cruciate ligament (PCL) rather than the ACL (26). The debate over the effect of gastrocnemius on ACL loading is largely in part to its bi-articular function at the knee and ankle. The position of the thigh, leg and foot segments will all contribute to the active or passive insufficiency of gastrocnemius, thereby making interpretations situation specific and difficult to generalize. Since the soleus muscle does not cross the knee joint, it has, until recently, been largely ignored as an influential muscle on ACL loading. Although there is much debate and inconsistency around gastrocnemius' effect on ACL strain, it is clear that with the foot planted, the force exerted by the soleus muscle would resist forward rotation of the tibia. This resistance to anterior tibial translation (relative to the distal femur) would certainly be a protective mechanism to ACL injuries. The soleus muscle appears to be more important in many simulations, as its activity begins earlier and continues longer than the gastrocnemius (23). These observations suggest that the soleus muscle has a significant stabilizing effect on the ACL by supplying a dynamic posterior force that resists anterior tibial translation. This led us to question whether anatomic variations that may reduce the effect of the soleus muscle would contribute to an individual's risk of ACL injuries more than previously appreciated.

With the understanding that structural and functional parameters at the foot and ankle may influence risk of ACL injuries by altering knee mechanics, the focus shifted to examining external tibial torsion as a possible link between lower body motion and ACL risk factors. The tibial torsion study in women with and without ACL injury (Chapter 3) demonstrated that ACL injured women had higher external tibial torsion (19$\pm$4 degrees) than uninjured women (12$\pm$4 degrees). These findings were important in identifying a significant association between external tibial torsion and providing rationale to examine this phenomenon further. Despite the associations previously reported between external tibial torsion and knee pathologies, minimal research existed on the actual functional mechanics of individuals with varying magnitudes of external tibial torsion. It was important to continue the research by comparing lower extremity mechanics between ACL injured individuals and individuals with healthy ACLs and varying degrees of external tibial torsion.

5.4 Tibial Torsion Influences Non-Sagittal Motion in Women with ACL Injury

The participant selection at this point of the research was deliberately restricted to women. Epidemiological studies consistently reveal that women have a higher incidence of non-contact ACL injuries than men (27). Women also demonstrate greater frontal and transverse plane knee moments than men during dynamic activities that potentially place the ACL at greater risk of injury (28, 29). Although these differences in lower extremity motion are proposed to increase the risk of ACL injury in women (30), much is still unknown about which factors contribute to these high risk lower body mechanics. High torsion and ACL-injured participants demonstrated key similarities when compared to low torsion participants during squatting and landing (Chapter 4). Uninjured participants with low external tibial torsion demonstrate less lateral rotation of the leg, smaller foot progression angles in two-foot landings, greater ankle plantar flexor moments, and less frontal plane leg abduction in two-foot landings. It has been established that external tibial torsion influences lower body mechanics. Future research is required to corroborate these findings in simple dynamics tasks, as well as more complex movements that simulate human movement in a less predictable setting.

5.5 Limitations

Inherent limitation of the present study is that the participants were not classed or grouped by their anthropometric data or activity levels. The controls and cases (Chapter 4) were not matched with respect to their size (height and mass) or their specific exposure to modes and methods of physical activity (frequency, intensity, duration and type), although inclusion criteria did require that participants were engaged in moderate to active physical activity levels through sport or recreation. Since these factors contribute to ACL injury risk, the study is limited in its findings. Future studies should attempt to group participants based on similar anthropometrics and physical activity exposure, or provide stricter guidelines on potential inclusion criteria.

Another limitation to the current dissertation is that the investigations included various types of non-contact ACL rupture mechanisms (Chapters 3 and 4), but only examined the mechanics during squatting, two-foot landings, and controlled step off landing protocols. ACL rupture is not restricted to these tasks, and may include hyperextension scenarios, one-footed landings from various heights, or most notably, change of direction manoeuvres, which were not part of the investigation. The current dissertation also restricted post-operative participant inclusion to hamstring graft reconstruction (Chapter 4). Even within hamstring graft patients, the reconstruction techniques not only differ by their fixation devices, but also have considerable differences with respect to fixation methods and graft configuration (31). Future research must consider the type of reconstruction as a potentially influential factor in post-operative functional performance.

Disruption of the ACL results in not only a mechanical disturbance, but also a loss of joint sensation. This deficit is often reported to be caused by a deafferentation of peripheral mechanoreceptors (sensory receptors) (32, 33). This deafferentation may alter reflex pathways to muscle and higher motor centres (34). Disrupted pathways would likely contribute to altered muscle activity post-ACL injury. It is estimated that one-third of ACL-deficient individuals have innate compensatory mechanisms that would present as remarkably different sequelae of muscle activity compared to their muscle activation before the injury (35). This dissertation did not consider these mechanisms when analyzing the lower body motion of ACL-reconstructed participants. Future research is required to consider these disruptions in the selection of participants and the methods of data collection.

A potential limitation which has not been discussed is with regards to the tasks chosen for the participants (Chapters 2 and 4). Participants' normal foot placement was constrained in the squat task in an effort to standardize the protocol. This constraint may have contributed to error in interpretation of the data, since there is no way of knowing the level of constraint based on each individual. During the two-foot jump landings, the arm motion was partially constrained, but the participants were permitted to use a slight counter movement. This lack of specific protocol may have contributed to error in interpretation of the data, since the counter movement can affect the height jumped and the subsequent forces required in landing.

Furthermore, it is important to note that the current dissertation incorporated manual measurements in the screening process (Chapter 3) and landmark identification in the motion analysis process (Chapters 2,3 and 4). High between day reliability of joint and segment angles has been established in our laboratory (1-3 degrees) during the squatting task used in this project, but these methods are subject to observer bias and inconsistencies that may affect the data.

To further assist in adequately addressing limitations, the following quality appraisal checklist was considered for the final study in the project:

1. Is the objective of the study clearly stated?
2. Are the characteristics of the participants included in the study described?
3. Are the eligibility criteria (inclusion and exclusion criteria) for entry into the study explicit and appropriate?
4. Are the recruitment criteria for the study appropriate?
5. Was the study adequately powered?
6. Did case participants enter the study at a similar point in the disease/recovery?
7. Are the measurements clearly defined in the introduction or methods section?
8. Were relevant outcomes appropriately measured with objective and /or subjective methods?
9. Were the statistical tests used to assess the relevant outcomes appropriate?
10. Are the conclusions of the study supported by the results?

The objective of the study is explicitly stated in the introduction and the rationale of the study is clear. Throughout the project, the characteristics of the participants is clearly described and consistently included in the methodology. Eligibility criteria are clearly outlined in the methodology and are appropriate for the aim of the study. The methods of participant recruitment are appropriate, although the limited placement of recruitment flyers may affect generalizability of the data. The study exceeds minimum power requirements, which improves the ability of the study to detect actual differences between the groups. It is unknown whether case participants (ACL injured) were entering the study at a similar point in their recovery. Although inclusion criteria required that participants were a minimum of six months post-operative, the exact length of time since surgery was not reported. This presents a limitation, since recovery protocol and duration may alter the mechanics of the individual. All measurements are clearly defined in the methodology, and are explicit and repeatable based on the information given. Relevant outcomes were appropriately measured with reliable and valid methods that limited subjectivity throughout data collection and extraction. Statistical testing was appropriate for the outcomes measured. As mentioned, the sub-grouping of the control participants may have limited the study based on the statistical tests used to assess differences. The conclusions of the study are supported by the results presented.

5.6 Practical Applications and Future Directions

External tibial torsion must be considered when discussing differences in lower body kinematics, kinetics and ACL injury. Increasing magnitudes of external tibial torsion may alter the loading on the supportive tissues and ligaments of the lower body, and may alter the functionality of the leg muscles, such as soleus, in particular. This dissertation examined the sagittal and non-sagittal segmental contributions to joint motion in participants with and without ACL injury, and with varying degrees of external tibial torsion. These results suggest there are similarities between ACL injury risk factors and certain kinematics that are linked to individuals who demonstrate greater external tibial torsion. Further research is warranted to examine tibial torsion and the influence of individual anthropometrics, mechanism of ACL injury, graft type, proprioception, and conditioning protocols on the pre- and post-injury muscle activity in a variety of functional dynamic tasks.

5.7 References

1. Schwartz M, Lakin G. The effect of tibial torsion on the dynamic function of the soleus during gait. Gait Posture. 2003 4;17(2):113-8.

2. Song KM, Concha MC, Haideri NF. Effects of lower limb torsion on ankle kinematic data during gait analysis. Journal of Pediatric Orthopaedics. 2001;21(6):792-7.

3. Radler C, Kranzl A, Manner HM, Hoglinger M, Ganger R, Grill F. Torsional profile versus gait analysis: consistency between the anatomic torsion and the resulting gait pattern in patients with rotational malalignment of the lower extremity. Gait Posture. 2010 Jul;32(3):405-10.

4. Valmassy R, Stanton B. Tibial torsion. Normal values in children. J Am Podiatr Med Assoc. 1989 Sep;79(9):432-5.

5. Turner MSMC, F.R.C.S. The Association Between Tibial Torsion and Knee Joint Pathology. Clinical Orthopaedics & Related Research. 1994 May;302:47-51.

6. Delgado ED, Schoenecker PL, Rich MM, Capelli AM. Treatment of severe torsional malalignment syndrome. J Pediatr Orthop. 1996 Jul-Aug;16(4):484-8.

7. Krivickas LS. Anatomical factors associated with overuse sports injuries. Sports Med. 1997 Aug;24(2):132-46.

8. Eckhoff DG, Brown AW, Kilcoyne RF, Stamm ER. Knee version associated with anterior knee pain. Clin Orthop Relat Res. 1997 Jun;(339)(339):152-5.

9. Eckhoff DG, Johnson KK. Three-dimensional computed tomography reconstruction of tibial torsion. Clin Orthop. 1994(302):42-6.

10. Hicks J, Arnold A, Anderson F, Schwartz M, Delp S. The effect of excessive tibial torsion on the capacity of muscles to extend the hip and knee during single-limb stance. Gait and Posture. 2007;26(4):546-52.

11. Turner MS. The association between tibial torsion and knee joint pathology. Clin Orthop Relat Res. 1994 May;(302)(302):47-51.

12. Yagi TM, D. Tibial Torsion in Patients With Medial-Type Osteoarthrotic Knees. Clinical Orthopaedics & Related Research. 1994 May;302:52-6.

13. Lincoln TL, Suen PW. Common rotational variations in children. J Am Acad Orthop Surg. 2003 Sep-Oct;11(5):312-20.

14. Stefko RM, de Swart RJ, Dodgin DA, Wyatt MP, Kaufman KR, Sutherland DH, et al. Kinematic and kinetic analysis of distal derotational osteotomy of the leg in children with cerebral palsy. J Pediatr Orthop. 1998 Jan-Feb;18(1):81-7.

15. Griffin LY, Agel J, Albohm MJ, Arendt EA, Dick RW, Garrett WE, et al. Noncontact anterior cruciate ligament injuries: risk factors and prevention strategies. J Am Acad Orthop Surg. 2000 May-Jun;8(3):141-50.

16. Mokhtarzadeh H, Yeow CH, Hong Goh JC, Oetomo D, Malekipour F, Lee PV. Contributions of the soleus and gastrocnemius muscles to the anterior cruciate ligament loading during single-leg landing. J Biomech. 2013 Jul 26;46(11):1913-20.

17. Zantop T, Herbort M, Raschke MJ, Fu FH, Petersen W. The role of the anteromedial and posterolateral bundles of the anterior cruciate ligament in anterior tibial translation and internal rotation. Am J Sports Med. 2007 Feb;35(2):223-7.

18. Hewett TE, Myer GD, Ford KR. Anterior cruciate ligament injuries in female athletes: Part 1, mechanisms and risk factors. Am J Sports Med. 2006 Feb;34(2):299-311.

19. Pflum MA, Shelburne KB, Torry MR, Decker MJ, Pandy MG. Model prediction of anterior cruciate ligament force during drop-landings. Med Sci Sports Exerc. 2004 Nov;36(11):1949-58.

20. Li G, Rudy TW, Sakane M, Kanamori A, Ma CB, Woo SL. The importance of quadriceps and hamstring muscle loading on knee kinematics and in-situ forces in the ACL. J Biomech. 1999 Apr;32(4):395-400.

21. Renstrom P, Arms SW, Stanwyck TS, Johnson RJ, Pope MH. Strain within the anterior cruciate ligament during hamstring and quadriceps activity. Am J Sports Med. 1986 Jan-Feb;14(1):83-7.

22. Elias JJ, Faust AF, Chu YH, Chao EY, Cosgarea AJ. The soleus muscle acts as an agonist for the anterior cruciate ligament. An in vitro experimental study. Am J Sports Med. 2003 Mar-Apr;31(2):241-6.

23. Sherbondy PS, Queale WS, McFarland EG, Mizuno Y, Cosgarea AJ. Soleus and gastrocnemius muscle loading decreases anterior tibial translation in anterior cruciate ligament intact and deficient knees. J Knee Surg. 2003 Jul;16(3):152-8.

24. O'Connor JJ. Can muscle co-contraction protect knee ligaments after injury or repair? J Bone Joint Surg Br. 1993 Jan;75(1):41-8.

25. Fleming BC, Renstrom PA, Ohlen G, Johnson RJ, Peura GD, Beynnon BD, et al. The gastrocnemius muscle is an antagonist of the anterior cruciate ligament. J Orthop Res. 2001 Nov;19(6):1178-84.

26. Durselen L, Claes L, Kiefer H. The influence of muscle forces and external loads on cruciate ligament strain. Am J Sports Med. 1995 Jan-Feb;23(1):129-36.

27. Beck JL, Wildermuth BP. The female athlete's knee. Clin Sports Med. 1985 Apr;4(2):345-66.

28. Shultz SJ, Schmitz RJ. Effects of transverse and frontal plane knee laxity on hip and knee neuromechanics during drop landings. Am J Sports Med. 2009 Sep;37(9):1821-30.

29. Sigward SM, Powers CM. The influence of gender on knee kinematics, kinetics and muscle activation patterns during side-step cutting. Clin Biomech (Bristol, Avon). 2006 Jan;21(1):41-8.

30. Hewett TE, Myer GD, Ford KR, Heidt RS,Jr, Colosimo AJ, McLean SG, et al. Biomechanical measures of neuromuscular control and valgus loading of the knee predict anterior cruciate ligament injury risk in female athletes: a prospective study. Am J Sports Med. 2005 Apr;33(4):492-501.

31. Scheffler SU, Sudkamp NP, Gockenjan A, Hoffmann RF, Weiler A. Biomechanical comparison of hamstring and patellar tendon graft anterior cruciate ligament reconstruction techniques: The impact of fixation level and fixation method under cyclic loading. Arthroscopy. 2002 Mar;18(3):304-15.

32. Skinner HB, Barrack RL, Cook SD, Haddad RJ,Jr. Joint position sense in total knee arthroplasty. J Orthop Res. 1984;1(3):276-83.

33. Johansson H, Sjolander P, Sojka P. A sensory role for the cruciate ligaments. Clin Orthop Relat Res. 1991 Jul;(268)(268):161-78.

34. Swanik CB, Lephart SM, Swanik KA, Stone DA, Fu FH. Neuromuscular dynamic restraint in women with anterior cruciate ligament injuries. Clin Orthop Relat Res. 2004 Aug;(425)(425):189-99.

35. Noyes FR, Mooar PA, Matthews DS, Butler DL. The symptomatic anterior cruciate-deficient knee. Part I: the long-term functional disability in athletically active individuals. J Bone Joint Surg Am. 1983 Feb;65(2):154-62.