

**INVESTIGATION OF FOOT LANDING TECHNIQUES  
AND MUSCLE ACTIVATION DURING SINGLE-LEG  
DROP LANDINGS: IMPLICATIONS FOR NON-CONTACT  
ANTERIOR CRUCIATE LIGAMENT INJURIES**

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A thesis submitted to the Nanyang Technological University  
in partial fulfilment of the requirement for the degree of  
Doctor of Philosophy

2017

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## **Acknowledgement**

I would like to express my deep-felt gratitude to my supervisors, Assoc Prof Leong Kah Fai (*School of Mechanical and Aerospace Engineering, Nanyang Technological University*) and Assoc Prof Kong Pui Wah, Veni (*Physical Education and Sports Science Academic Group, National Institute of Education, Nanyang Technological University*) for their valuable guidance, time, understanding and patience over the PhD years.

I would like to thank Mr Long Tien Siew, Laboratory Executive from the School of Mechanical and Aerospace Engineering, Ms Judy Ong, Technical Officer from the National Institute of Education and Mr Terry Chin, Research Assistant from the Institute for Sports Research for their support and assistance in technical support, facility and equipment loaning. I would also like to thank Mr Huang Pengyuan, Mr Gu Kaiyi, Mr Bai Jinhao and Mr Philip Phua for their assistance in data collection. I am also grateful to Mr Frankie Tee for enhancing the image quality of some of the figures in this report. Finally, I am thankful to the Institute for Sports Research for their support in my work and for colleagues who have been so helpful and willing to take part in my pilot studies.

Special thanks to the people who have been a great inspiration in my life: loving and supportive parents, creative and caring sisters, inspiring engineer husband cum mentor and wonderful teachers whom I have the privilege to meet over the years. And to my beloved sons and nephews, may they always find joy in learning.

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## List of Symbols

$\alpha$	Level of significance
$\alpha_c$	Cardan angle representing flexion-extension
$\alpha_E$	ACL elevation angle from the tibial surface
$\alpha_x$	Angular acceleration in the local coordinate system (x axis)
$\alpha_y$	Angular acceleration in the local coordinate system (y axis)
$\alpha_z$	Angular acceleration in the local coordinate system (z axis)
$\beta$	Probability of committing a Type II error
$\beta_c$	Cardan angle representing abduction-adduction
$\beta_D$	ACL deviation angle from the anterior-posterior line of action
$\gamma_c$	Cardan angle representing axial rotation
$\eta^2$	Effect size
$\theta_{PTS}$	Patella tendon-tibia shaft angle
$\mu$	Mean
$\pi$	Pi
$\sigma$	Standard deviation
$\omega_x$	Angular velocity in the local coordinate system (x axis)
$\omega_y$	Angular velocity in the local coordinate system (y axis)
$\omega_z$	Angular velocity in the local coordinate system (z axis)
$\angle AF$	Ankle plantar flexion angles
$a_x$	Segment centre of mass acceleration in the global coordinate system (X axis)

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<b>ay</b>	Segment centre of mass acceleration in the global coordinate system (Y axis)
<b>az</b>	Segment centre of mass acceleration in the global coordinate system (Z axis)
<b>d</b>	Effect size
<b>E</b>	Sum of squares error
<b>F<sub>AS</sub></b>	Anterior shear force
<b>F<sub>ACL</sub></b>	Anterior cruciate ligament force along its longitudinal axis
<b>F<sub>H</sub></b>	Horizontal force component of the quadriceps
<b>∑F<sub>x</sub></b>	Summation of forces in the X axis
<b>∑F<sub>y</sub></b>	Summation of forces in the Y axis
<b>∑F<sub>z</sub></b>	Summation of forces in the Z axis
<b>f(y)</b>	Probability distribution of y
<b>G:H</b>	Gastrocnemius to quadriceps ratio
<b>I</b>	Identity matrix
<b>I<sub>BMI</sub></b>	Internal factor body mass index
<b>I<sub>GJL</sub></b>	Internal factor generalized joint laxity
<b>I<sub>d</sub></b>	Distance from centre of mass to distal joint
<b>I<sub>p</sub></b>	Distance from centre of mass to proximal joint
<b>I<sub>x</sub></b>	Moment of inertia in the local coordinate system (x axis)
<b>I<sub>y</sub></b>	Moment of inertia in the local coordinate system (y axis)
<b>I<sub>z</sub></b>	Moment of inertia in the local coordinate system (z axis)
<b>∠KF</b>	Knee flexion angles
<b>KF<sub>IC</sub></b>	Knee flexion angles at initial contact

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<b>K<sub>FROM</sub></b>	Knee flexion range of motion
<b>K<sub>FVGRF</sub></b>	Knee flexion angles at maximum vertical ground reaction force
<b>M<sub>xd</sub></b>	Distal moments in the local coordinate system (x axis)
<b>M<sub>yd</sub></b>	Distal moments in the local coordinate system (y axis)
<b>M<sub>zd</sub></b>	Distal moments in the local coordinate system (z axis)
<b>m</b>	Mass of a segment
$\vec{P}_1$	Position of the tracking marker in the global coordinate system
$\vec{P}'_1$	Position of the tracking marker in the local coordinate system
<b>p</b>	Probability of the test
<b>Q</b>	Quadriceps muscle force
<b>Q:H</b>	Quadriceps to hamstring ratio
<b>N</b>	Sample points in time
<b>n</b>	Sample size
$\vec{O}_{seg}$	Linear transformation between local and global coordinate system, defining the origin of the local coordinate system, relative to the global coordinate system.
<b>R<sup>2</sup></b>	Coefficient of multiple determination
<b>r<sup>2</sup></b>	Coefficient of determination
<b>R(f<sub>cf</sub>)</b>	Residual at a cut-off frequency
<b>R<sub>Ankle</sub></b>	Ankle joint angle, represented by an orientation matrix
<b>R<sub>Foot</sub></b>	Transformation of the foot local coordinate system to the global coordinate system
<b>R<sub>Knee</sub></b>	Knee joint angle, represented by an orientation matrix
<b>R'<sub>Pelvis</sub></b>	Transpose of the transformation of the pelvis's local coordinate

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	system to the global coordinate system
$\mathbf{R}'_{\text{Seg}}$	Rotation matrix of the segment from local to global coordinate system
$\mathbf{R}_{\text{Shank}}$	Transformation of the shank's local coordinate system to the global coordinate system
$\mathbf{R}'_{\text{Shank}}$	Transpose of the transformation of the shank's local coordinate system to the global coordinate system
$\mathbf{R}'_{\text{Thigh}}$	Transpose of the transformation of the thigh's local coordinate system to the global coordinate system
$\mathbf{R}_{\text{XP}}$	Proximal reaction force in the global coordinate system (X axis)
$\mathbf{R}_{\text{XD}}$	Distal reaction forces in the global coordinate system (X axis)
$\mathbf{R}_{\text{YP}}$	Proximal reaction force in the global coordinate system (Y axis)
$\mathbf{R}_{\text{YD}}$	Distal reaction forces in the global coordinate system (Y axis)
$\mathbf{R}_{\text{ZP}}$	Proximal reaction force in the global coordinate system (Z axis)
$\mathbf{R}_{\text{ZD}}$	Distal reaction forces in the global coordinate system (Z axis)
$\mathbf{R}_{\text{xp}}$	Proximal reaction force in the local coordinate system (x axis)
$\mathbf{R}_{\text{xd}}$	Distal reaction forces in the local coordinate system (x axis)
$\mathbf{R}_{\text{yp}}$	Proximal reaction force in the local coordinate system (y axis)
$\mathbf{R}_{\text{yd}}$	Distal reaction forces in the local coordinate system (y axis)
$\mathbf{R}_{\text{zp}}$	Proximal reaction force in the local coordinate system (z axis)
$\mathbf{R}_{\text{zd}}$	Distal reaction forces in the local coordinate system (z axis)
$s$	Number of markers on the segment
$\mathbf{X}_i$	Raw data at the $i$ th sample
$\hat{\mathbf{X}}_i$	Filtered data at the $i$ th sample using a 4th order zero-lag filter
$x$	X-axis (local coordinate system)

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<b>y<sub>r</sub></b>	Normal random variable
<b>y</b>	Y-axis (local coordinate system)
<b>z</b>	Z-axis (local coordinate system)

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## List of Abbreviation

<b>ACL</b>	Anterior cruciate ligament
<b>ANOVA</b>	Analysis of Variance
<b>ASIS</b>	Anterior superior iliac spine
<b>ASIS_Distance</b>	Distance between left and right anterior superior iliac spine markers
<b>BMI</b>	Body mass index
<b>CAL1</b>	Distal calcaneus
<b>CAL2</b>	Proximal calcaneus
<b>COM</b>	Centre of mass
<b>DC</b>	Direct current
<b>DVRT</b>	Differential variable reluctance transducer
<b>EMG</b>	Electromyography
<b>EMS</b>	Electrical muscle stimulation device
<b>FKBs</b>	Functional knee braces
<b>GCS</b>	Global coordinate system
<b>GJL</b>	Generalized joint laxity
<b>ICC</b>	Intraclass correlation coefficient
<b>KVA</b>	Knee valgus angles
<b>KVM</b>	Knee valgus moments
<b>LCS</b>	Local coordinate system
<b>LHJC</b>	Left hip joint centre marker
<b>MRI</b>	Magnetic resonance imaging

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<b>MCL</b>	Medial collateral ligament
<b>MVC</b>	Maximal voluntary contraction
<b>NBA</b>	National basketball association
<b>NWI</b>	Notch width index
<b>PKBs</b>	Prophylactic knee braces
<b>RHJC</b>	Right hip joint centre marker
<b>SD</b>	Standard deviation
<b>VGRF</b>	Vertical ground reaction force

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## Abstract

Anterior cruciate ligament (ACL) injuries are common during sport, frequently occurring during non-contact situations, caused by the athletes' own manoeuvres. The foot and ankle form the initial parts of the lower extremity kinetic chain during landing. Thus, foot landing positions can affect force transmission from the ankle to the knee. Yet, many studies on ACL injuries have not focused on the ankle and this is considered a gap to the better understanding of non-contact ACL injuries. The roles of the gastrocnemius and hamstring in non-contact ACL injury risk reduction, especially during the pre-landing phase of single-leg landings are also still unclear. Therefore, investigation of foot-landing techniques and muscle activation during single-leg landings is an important area to provide new insights into non-contact ACL injury risk reduction.

The main objective of this thesis was therefore to investigate how non-contact ACL injury risks could be reduced by improved foot landing techniques. The second objective was to study the role of muscle activation, particularly the gastrocnemius, in non-contact ACL injury risk reduction. Four laboratory-controlled experiments were conducted in this thesis. The first three studies were conducted in a motion analysis laboratory, where kinematic and kinetic measurements were taken. The first study provided findings on effects on foot rotation on knee valgus. The second study examined the influence of fore-foot, natural and flat-foot landing techniques on knee flexion angles and knee flexion range of motion. The third study studied the influence of ankle plantar flexion angles and internal risk factors on knee flexion angles at initial contact with the ground. This study

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also measured muscle activation levels and examined the role of the gastrocnemius in non-contact ACL injury risks, as opposed to the hamstrings and quadriceps. Effects of varying levels of gastrocnemius activation on anterior tibial translation and the role of gastrocnemius activation on non-contact ACL injury risk reduction were further investigated in the fourth and last study. A novel methodology using the ultrasound imaging, to measure anterior tibial translation under muscle activation, was incorporated into this last study.

Foot landing techniques were found to have an influence on non-contact ACL injury risks. Athletes should not land with extreme toe-out foot rotation positions to minimize maximum knee valgus moments. Fore-foot landings were better landing techniques than flat-foot landings in terms of force absorption. However, athletes should avoid landing with excessive ankle plantar flexion angles during initial contact, leading to low knee flexion angles, thereby increasing non-contact ACL injury risks. An optimal range of ankle plantar flexion angles was proposed in this study. Ankle plantar flexion angles at initial contact should not exceed 26° and 28° for males and females, respectively. Gastrocnemius was also found to be significantly more activated than hamstrings during the pre-landing phase before initial contact, possibly to compensate for reduced hamstring activation. This study provided insights into the controversial role of the gastrocnemius in ACL injury risks. Results suggest that with sufficient gastrocnemius activation, anterior tibial translation could be reduced, thereby reducing ACL injury risks.

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This research provided new knowledge into how foot landing techniques could be implemented to reduce non-contact ACL injury risks. It also provided a better understanding of the role of muscle activation, particularly the gastrocnemius, in non-contact ACL injury risk reduction. All these could be incorporated into future ACL injury prevention interventions and measures.

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## Chapter 1 - Introduction

### 1.1 Background

Anterior cruciate ligament (ACL) is the tissue connecting the thigh bone to the shank bone (Hartigan *et al.*, 2011) and is the most commonly injured ligament in the general population (Bollen, 2000). ACL plays an important role in providing knee stability, especially in the anterior-posterior direction (Hartigan *et al.*, 2011). An ACL injury frequently occurs in non-contact conditions (Gianotti *et al.*, 2009, Waldén *et al.*, 2011, Dragoo *et al.*, 2012), not caused by contact with another player or thing but by the athlete's own manoeuvres (Marsall *et al.*, 2007). The impact of an ACL injury on athletes is severe (Hewett *et al.*, 2009), resulting in athletes' absence from play for a long period of time (Roi *et al.*, 2006, Waldén *et al.*, 2011) and having greater risks of contracting osteoarthritis in later part of their life (Waldén *et al.*, 2011, Dragoo *et al.*, 2012). Furthermore, anterior/posterior cruciate ligament injury accounts for the greatest mean cost of all knee injuries (Marsall *et al.*, 2007). High ACL surgery costs in the United States rise beyond \$2 billion annually (Ali *et al.*, 2010). As a result, there has been significant amount of work on ACL-related research over the past decade (Rishiraj *et al.*, 2009).

To prevent ACL injuries among healthy athletes, there have been research in the effectiveness of preventive training programmes and knee braces. Preventive training programmes are found to have the potential to reduce ACL injury risks (Griffin *et al.*, 2006). However, the actual techniques that make the training successful are still unclear

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(Griffin *et al.*, 2006). Besides, fatigue that affects neuromuscular control will also be experienced by athletes during competitions (Mohr *et al.*, 2003, Alentorn-Geli *et al.*, 2009a), thus still exposing them to injury risks. On the other hand, the effectiveness of knee braces to prevent knee ligament injuries is still not conclusive and healthy athletes are reluctant to wear them, fearing a reduction of performance (Rishiraj *et al.*, 2009). Alternative solutions to reduce ACL injury risks are therefore necessary, especially when ACL injuries have continued to rise by 50% over the past decade (Donnelly *et al.*, 2012).

Since the foot and ankle form the initial parts of the lower extremity kinetic chain during landing, foot landing positions may affect force transmission from the ankle to the knee (Ali *et al.*, 2010). Many studies on ACL injuries have not focused on the ankle and this is considered a missing link to the better understanding of non-contact ACL injuries (Ali *et al.*, 2010). Yet, the potential effects of foot landing techniques on reducing non-contact ACL injury risk reduction can be seen from the effects of ankle restraints on knee positions (DiStefano *et al.*, 2008, Joseph *et al.*, 2008, Jenkins *et al.*, 2009, Boros *et al.*, 2010), despite not being designed for ACL injury prevention. There have been a few studies that specifically vary foot landing positions during dynamic tasks to study its effects on the lower extremity (e.g. Kovács *et al.*, 1999, Cortes *et al.* 2007, Valenzuela *et al.*, 2016). However, little is known about how these effects could translate to reducing non-contact ACL injury risks. Often, these studies also focused on double-leg landings although higher ACL injuries occur during single-leg landings (Donnelly *et al.*, 2012). Influence of foot-landing techniques on ACL injury risks is therefore an important area of study that has yet to be fully explored.

Injury mechanisms with low knee flexion angles, valgus and internal rotation moments are associated with higher non-contact ACL injury risks (Olsen *et al.*, 2004, Cochrane *et al.*, 2007, Krosshaug *et al.*, 2007, Koga *et al.*, 2010). Low knee flexion angles are often considered to be between 0° to 30° (Myer *et al.*, 2004), especially when such angles are less than 15°, during which hamstring activation becomes less efficient in reducing ACL straining (Li *et al.*, 1999, Yu *et al.*, 2005). These low knee flexion angles occur during single-leg landings at initial contact (Weinhandl *et al.*, 2010), affecting the effectiveness of the hamstring in countering the quadriceps' anterior pull of the tibia (Li *et al.*, 1999, Yu *et al.*, 2005). There is therefore a need to investigate the role of hamstrings in ACL injury risk reduction during the pre-landing phase of single-leg landings. Another knee flexor, the gastrocnemius, has also been a focus of research to examine its role in ACL injuries (Dürselen *et al.*, 1995, Fleming *et al.*, 2001b, Ali *et al.*, 2014, Morgan *et al.*, 2014). However, there are still opposing arguments as to whether the gastrocnemius is an agonist or antagonist of the ACL. It is therefore essential to examine the role of muscle activation, particularly the gastrocnemius, in ACL injury risk reduction.

## 1.2 Motivation

An ACL injury often results in a long absence from play and has a significant impact on the physical, psychological and financial aspects of athletes, especially professional ones. Therefore, preventing healthy athletes from sustaining ACL injuries is of great importance. Hewett *et al.* (2007) also called for a focus on preventive measures to help reduce ACL injuries, especially when current preventive methodologies have not been shown to be effective. There is even a greater urgency to find alternative preventive

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solutions as ACL injuries have continued to increase over the last decade (Donnelly *et al.*, 2012). An ACL injury also frequently occurs in non-contact conditions (Gianotti *et al.*, 2009, Waldén *et al.*, 2011, Dragoo *et al.*, 2012). This research therefore focused on discovering new preventive solutions to reduce non-contact ACL injury risks of healthy athletes. Ankle restraints that were not designed for ACL injury prevention were already showing potential effects of reducing non-contact ACL injury risks (DiStefano *et al.*, 2008, Joseph *et al.*, 2008, Jenkins *et al.*, 2009, Boros *et al.*, 2010). Foot landing positions were also observed to have an influence on knee positions associated with non-contact ACL injury risks (Cortes *et al.* 2007). Further investigation of the optimal foot landing positions could provide new knowledge for non-contact ACL injury prevention. Studies also commonly focused on double-leg landings but more ACL landing injuries occur during single-leg landings (Donnelly *et al.*, 2012). Besides, roles of the gastrocnemius and hamstring in non-contact ACL injury risk reduction, especially during the pre-landing phase of single-leg landings, have also not been well understood. There is therefore a need to look into areas of foot-landing techniques and muscle activation during single-leg landings, to provide better insights into non-contact ACL injury risk reduction.

### **1.3 Research Objectives**

The main objectives of this thesis were:

1. To investigate the influence of foot landing techniques on non-contact ACL injury risks.

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2. To examine the role of muscle activation, particularly the gastrocnemius, in non-contact ACL injury risk reduction.

The hypotheses of this thesis were:

1. Non-contact ACL injury risks could be reduced with decrease of ankle plantar flexion angles and toe-in foot rotation positions.
2. Gastrocnemius would have a significantly higher muscle activation level than the hamstrings, to counter the quadriceps activation during the pre-landing phase of single-leg landings.

## **1.4 Scope**

To achieve the two research objectives, four laboratory-controlled studies were carried out:

### **a. Biomechanics Studies**

The first three studies were conducted in the motion analysis laboratory:

- Effects of Foot Rotation Positions on Knee Valgus
- Influence of Fore-foot, Natural and Flat-Foot Landing Techniques on Non-Contact ACL Injury Risks
- Influence of Ankle Plantar Flexion Angles and Internal Risk Factors on Knee Flexion Angles and Role of the Gastrocnemius during Single-Leg Drop-Landings

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Knee biomechanics were studied with different foot landing positions to examine how these positions might influence non-contact ACL injury risks. These studies focused on single-leg drop landings since more non-contact ACL landing injuries occur during single-leg landings (Donnelly *et al.*, 2012). This thesis focused on investigating non-contact ACL injuries and so only healthy participants were recruited. Gender differences were not the primary focus of this thesis but females were also recruited in the third study for a comparison of findings between genders. Measurement of muscle activation was also carried out in the third study to better understand the role of muscle activation, particularly the gastrocnemius, in non-contact ACL injury risk reduction.

#### **b. Ultrasound Study**

The novel use of ultrasound imaging to measure anterior tibial translation during muscle activation, was incorporated in the fourth and last study, 'Effect of Different Gastrocnemius Muscle Activation Levels on Anterior Tibial Translation'. The influence of different gastrocnemius activation levels on anterior tibial translation was investigated. This provided insights into the arguments regarding the role of the gastrocnemius in non-contact ACL injury risk reduction.

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## 1.5 Organization of the Thesis

This thesis first presents a summary of key existing literature research relating to ACL injuries. This leads to the impetus for this research and highlights research gaps of current work. This is followed by the details of the four studies of this thesis, each organized in their own chapter. Each chapter summarizes the key relevant literature, leading to the aims of each study, then elaborates on the study's methodology, and presents the results, discussion and the conclusion of the key findings. The thesis then ends with two final chapters - an overall discussion and a conclusion. The discussion chapter explains the four studies coming together to address the research gaps identified and their relevance to non-contact ACL injury prevention. The conclusion chapter provides the overall summary of these new findings. It also proposes future work that could be built up on the findings of this thesis for further non-contact ACL injury prevention research.

The organization of this thesis is as follows:

1. Chapter 2 covers a summary of a detailed literature review and analysis of current work, to lay justification for this research. The chapter focused on why research in non-contact ACL injury prevention is important; causes leading to higher non-contact ACL injury risks; gaps in current methodologies for ACL injury prevention; and why foot landing and muscle activation should be studied.
2. Chapter 3 presents the first study, focusing on the effects of foot rotation positions on knee valgus. Effects of foot rotation positions (toe-in, toe-forward, toe-out) on knee valgus, both during initial contact and during landing were studied.

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Recommendations for foot rotation positions leading to lower non-contact ACL injury risks were provided based on the study results.

3. Chapter 4 presents the second study, examining the influence of natural, fore-foot and flat-foot landings on knee flexion angles (at initial contact and maximum vertical ground reaction force) and knee flexion range of motion. This study provided evidence that although fore-foot landings were better for force absorption, the range of fore-foot landings was important to reduce non-contact ACL injury risks.
4. Chapter 5 presents the third study, investigating the influence of ankle plantar flexion angles and internal risk factors on knee flexion angles, and the role of the gastrocnemius in non-contact ACL injury risks. A multiple regression was run to predict knee flexion angles from ankle plantar flexion angles, generalized joint laxity (GJL) and body mass index (BMI). A relationship, to obtain the optimal range of ankle plantar flexion angles associated with reduced non-contact ACL injury risks, was proposed. Binomial logistic regression was also performed to determine how well the model correctly predicts the high and low knee flexion angle cases. Muscle activation levels prior initial contact, among gastrocnemius, hamstring and quadriceps were also studied. This study result suggests that gastrocnemius activation had a role in reducing non-contact ACL injury risks.

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5. Chapter 6 presents the fourth and last study, investigating the influence of different gastrocnemius activation levels on anterior tibial translation. Published literature is still unclear on the definite role of the gastrocnemius as agonists or antagonists of the ACL and this was investigated in this study with the use of ultrasound imaging techniques.
  6. Chapter 7 provides the overall summary of the thesis findings and highlights the research contributions.
  7. Chapter 8 provides the overall conclusion and proposes the future work of this research.

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## Chapter 2 – Literature Review

ACL is the tissue connecting the femur (thigh bone) to the tibia (shank bone) (see Figure 1) (Hartigan *et al.*, 2011). ACL plays an important role in providing knee stability by preventing the tibia from moving anteriorly to (or in front of) the femur (Arnheim *et al.*, 1993). ACL also prevents the tibia from going into extreme internal rotation and provides the secondary role of limiting knee valgus/varus loading (Arnheim *et al.*, 1993). Knee injuries are estimated to contribute up to 60% of all sport injuries, with ACL injuries contributing to around half of these injuries (Rishiraj *et al.*, 2009).



**Figure 1. Position of the anterior cruciate ligament in the knee model.**

### 2.1 Incidence and Impact of the ACL Injury

Around 200,000 ACL injuries are estimated per year in the United States based on national health care surveys (Marsall *et al.*, 2007). Although the ACL injury incidence is not considered high in the general population (Marsall *et al.*, 2007), the impact on

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athletes and society is enormous and is considered to be one of the most severe injuries (Hewett *et al.*, 2009) for the following reasons:

- ACL injuries have a huge impact on the sport careers of athletes, due to the long absence from play (Roi *et al.*, 2006, Waldén *et al.*, 2011). For professional soccer athletes, the mean time to return to competition was almost 8 months after operation (Roi *et al.*, 2006, Waldén *et al.*, 2011). It also took  $5.2 \pm 3.6$  years (range, 1–18 years) to reach pre-injury performance (Roi *et al.*, 2006). In basketball, 98% of National Basketball Association (NBA) players returned to play  $11.6 \pm 4.1$  months after injury (Harris *et al.*, 2013). The long absence from play could also lead to huge financial losses for these athletes. A year's loss of NBA income could amount to an average of US\$6 million (Harris *et al.*, 2013). On average, National Football League ACL-injured players earned \$2,070,521 less than income-matched controls over the 4 years after injury (Secrist *et al.*, 2016). For athletes originally earning less than \$2 million annually, there was also a higher likelihood for them to leave the league than uninjured controls (Secrist *et al.*, 2016), resulting in a potential loss of income.
- The long-term health of athletes is impacted after an ACL injury. It is associated with an increased risk of a fresh knee injury later in life (Waldén *et al.*, 2006, Waldén *et al.*, 2011). It could also have a long-term medical impact such as increasing the risk of osteoarthritis (Waldén *et al.*, 2011, Dragoo *et al.*, 2012).
- Besides, anterior/posterior cruciate ligament injury accounts for the greatest mean cost of all knee injuries (Marsall *et al.*, 2007). Each year, high ACL surgery costs

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in the United States rise beyond \$2 billion (Ali *et al.*, 2010), resulting in a high financial burden on the health care system (Marsall *et al.*, 2007).

As societies promote more sport involvement to maintain public well-being, the number of incidences of sport injuries could correspondingly increase (Engebretsen *et al.*, 2007). Therefore, preventing ACL injuries is impactful to reducing social costs on public health, as well as, to prevent healthy athletes from suffering the ill-effects of the ACL injury.

There is a higher incidence of ACL injury in females than in males (Roos *et al.*, 1995, Bjordal *et al.*, 1997, Agel *et al.*, 2005, Waldén *et al.*, 2011). In basketball, females have injury rates that are 3.5 times higher than males in basketball and 2.7 times higher for soccer (Smith *et al.*, 2012b). However, in terms of absolute numbers, there are still more males suffering from ACL injuries in the general population (Marsall *et al.*, 2007). In a study of cruciate ligament injuries (ACL and posterior cruciate ligament (PCL) combined) among Swiss youth from a national sport organization, males accounted for around 76% of the ACL or PCL injuries (de Loës *et al.*, 2000). New ACL injury preventive measures would benefit a larger number of people if the focus of the study was on males. Therefore, male participants were the focus of the main part of this PhD research. However, for a more comprehensive study, females were also recruited in Chapter 5 to examine if there were any differences as to how foot landing and muscle activation influenced ACL injury risks between genders.

## 2.2 Playing Conditions Associated with High ACL Injury Risks

This research focused on non-contact ACL injuries as ACL injury most frequently occur under non-contact conditions (Chappell *et al.*, 2002, Yu *et al.*, 2007, Hewett *et al.*, 2009), accounting for 70% to 80% of all ACL injuries (Ali *et al.*, 2010). A non-contact ACL injury is defined as one that is not caused by contact with another player or object but by the athlete's own manoeuvres (Marsall *et al.*, 2007). Non-contact ACL injuries commonly occur during jump-landing, cutting or sidestepping (Huston *et al.*, 2007, Donnelly *et al.*, 2012).

Table 1 shows ACL injury incidence rates found in a number of sports and incidence rates ranged from 0.06 to 0.82.

**Table 1. Variation of ACL Injury Incidence by Sport**

<b>Sport</b>	<b>ACL Injury Incidence Rate</b> (ACL Injuries per 1000 athlete-exposures)	<b>Reference</b>
<b>Australian Football</b>	0.82 0.62 (non-contact)	Orchard <i>et al.</i> (2001)
<b>American Football</b>	0.25	Orchard <i>et al.</i> (2001)
<b>Soccer</b>	0.1 (Male) 0.3 - 0.8 (Female)	Marsall <i>et al.</i> (2007)
<b>Basketball</b>	0.06 - 0.2 (Male) 0.1 – 0.5 (Female)	Marsall <i>et al.</i> (2007)

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On comparison, Australian football has the highest recorded incidence rates amongst the sports shown in Table 1. The manoeuvres that result in the highest number of ACL injuries differ with sports. The findings are summarized in Table 2. In Australian football, injuries were mostly found during side-cutting (37%) (Cochrane *et al.*, 2007). Despite being ranked second, landing occurred 32% of the time, which came close to side-cutting (Cochrane *et al.*, 2007). Injuries during landing were also commonly found in basketball and badminton (Krosshaug *et al.*, 2007, Kimura *et al.* 2010). In fact, it was reported that 70% of ACL injuries happen during jump-landing during sports (Ireland, 2007).

**Table 2. Manoeuvres with the Highest ACL Injuries in Sport**

<b>Sport</b>	<b>Manoeuvres with the Highest ACL Injuries</b>	<b>Reference</b>
<b>Australian Football (non-contact)</b>	37% – Side stepping 32% - landing 16% - land and step 10% - stopping/slowing 5% - crossover cut	Cochrane <i>et al.</i> (2007)
<b>Soccer (non-contact)</b>	Change in direction or cutting with deceleration, jump landing, pivoting Deceleration task (Female)	Alentorn-Geli <i>et al.</i> (2009a) Tarmah <i>et al.</i> (2010)
<b>Basketball (contact and non-contact combined)</b>	Mostly (35%) - 1-leg landing (Male) Mostly (41%) - 2-leg landing (Female)	Krosshaug <i>et al.</i> (2007)
<b>Badminton (non-contact)</b>	Mostly (48%) - 1-leg landing after overhead stroke 40% - plant-and-cut	Kimura <i>et al.</i> (2010)

Landing injuries also tend to occur mostly during single-leg landings (Donnelly *et al.*, 2012). Single-leg and double-leg landings result in different knee kinematics and muscle activation levels (Pappas *et al.*, 2007). Single leg-landings were found to result in an increase in knee valgus, significant decrease in knee flexion at initial contact and peak knee flexion, which are all associated with higher ACL injury risks (Pappas *et al.*, 2007).

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Two studies have also suggested that peak ACL strain during landing occurs close to initial contact with the ground (Lamontagne *et al.*, 2008, Taylor *et al.*, 2011), rather than at maximum vertical ground reaction force for cutting manoeuvres (Lamontagne *et al.*, 2008). The Differential Variable Reluctance Transducer (DVRT) was used in the study by Lamontagne *et al.* (2008) and ACL strain could be measured directly during single-leg landing. The second study, estimating ACL loading through a combination of motion capture, magnetic resonance imaging (MRI) and fluoroscopic imaging, showed that ACL loading was highest just before initial impact during double-leg landing (Taylor *et al.*, 2011). This slight discrepancy between the two study results could be because ACL loading was estimated and not measured directly in the second study. Besides, double-leg landing was used in the study by Taylor *et al.* (2011), rather than single-leg landing in the study by Lamontagne *et al.* (2008). The computational modelling study by Laughlin *et al.* (2011) also found peak ACL force occurring at 7-10 ms following initial contact during single-leg landing, instead of at maximum vertical ground reaction force (around 50 ms after initial contact). All these studies suggest that peak ACL strain during landing is close to initial contact and not at the peak vertical ground reaction force as perceived (Ali *et al.*, 2013). Laughlin and colleagues have also proposed to place more attention on lower extremity positions at initial contact to study ACL injury risks (Laughlin *et al.* 2011). However, it is worth noting that many studies do not even consider the initial contact in their studies relating to ACL injuries (Chappell *et al.*, 2002, Garrison *et al.*, 2005, Lin *et al.*, 2012, Ali *et al.*, 2013, Hong *et al.*, 2014).

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Landing with more extended knees during single-leg landings (Pappas *et al.*, 2007) and with the ACL experiencing the highest strain close to initial contact (Lamontagne *et al.*, 2008, Taylor *et al.*, 2011, Laughlin *et al.* 2011), would therefore increase the chances of an ACL injury. This could be a contributory reason as to why non-contact ACL injuries occur more frequently in single-leg, rather than double-leg landings (Donnelly *et al.*, 2012). Thus far, no published literature has been found to look into contributory reasons leading to higher non-contact ACL injuries during single-leg landings, especially at the point of initial contact. Even studies that compared between double-leg and single-leg landings only studied the differences in biomechanical measures, but did not explain why these differences took place (Pappas *et al.*, 2007, Wang, 2011). Research in this could discover new knowledge to reduce non-contact ACL injury risks in single-leg landings. Therefore, single-leg landings underlined the focus of this study, especially at initial contact with the ground. This research mainly used basketball players as participants since most ACL injuries occur during landing in basketball.

### **2.3 Primary Causes of ACL Injuries**

The primary causes of ACL injuries must be identified before any ACL injury prevention can be made effective. In the approach to the better understanding of injury causes, a model has been proposed and it suggests that a sport injury is a result of the interplay of risk factors (internal and external) and the injury mechanism prior injury (see Figure 2) (Bahr *et al.*, 2005). Thus, the various factors should be taken into account for the design of more effective injury prevention programmes (Bahr *et al.*, 2005). This multi-factorial

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approach was also adopted in this PhD research in the understanding of non-contact ACL injuries.

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**Figure 2. Model to show factors leading to a sport injury (BMD: Bone Mass Density; ROM: Range of Motion) (Bahr *et al.*, 2005).**

Based on current literature, internal risk factors leading to increased non-contact ACL injury risks are presented in Section 2.3.1. The injury mechanisms leading to non-contact ACL injuries are presented in Section 2.3.2. External risk factor with the use of protective equipment, leading to injury prevention, will be presented in Section 2.4, together with the review of current preventive training programmes.

### **2.3.1 Internal Risk Factors of ACL Injuries**

The internal risk factors of ACL injuries include anatomical, hormonal and neuromuscular factors (Griffin *et al.*, 2006). Subject differences due to different internal risk factors may contribute to a greater variability of the experimental results. These factors should be considered during studies relating to ACL injuries but not many researchers have done so (Ali *et al.*, 2010). This is probably done to reduce complexity in their studies.

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### 2.3.1.1 Anatomical Risk Factors

#### 2.3.1.1.1 Notch Width, Notch Width Index, Generalized Joint Laxity and Body Mass Index

Current studies still yield contradictory results with regards to anatomical factors' role in increasing the ACL injury risk (Griffin *et al.*, 2006). One widely cited prospective cohort study by Uhorchak *et al.* (2003) followed 1198 new United States Military Academy cadets over 4 years and 859 cadets finished the study. Significant risk factors for both males and females include the notch width, notch width index (NWI) and the generalized joint laxity (GJL).

The notch width is defined as the width of the groove BB' between the condylars (rounded protrusions) at the end of the femur (thigh bone) as shown in Figure 3 (Uhorchak *et al.*, 2003). NWI is defined by notch width (BB') divided by condyle width (AA') as labelled in Figure 3.

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**Figure 3. Position of notch width (labelled as BB') (Adapted from Uhorchak *et al.*, 2003).**

The study by Uhorchak *et al.* (2003) has limitations. This study had a large overall sample size but the actual number of non-contact ACL injured subjects was too few for meaningful statistical comparison. However, as compared to other prospective cohort

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studies in the review by Smith *et al.* (2012a), the study by Uhorchak *et al.* was already one of the bigger prospective cohort studies carried out to date.

Another limitation of Uhorchak *et al.*'s study is that the model was not able to reliably predict any of the non-contact ACL injuries in men (Uhorchak *et al.*, 2003). The most predictive model that consisted only of the notch width and GJL could only account for 15% of the variability of non-contact ACL injuries for males (Uhorchak *et al.*, 2003). For females, the model that included narrow notch width, higher than average body mass index (BMI), and GJL accounted for 62.5% of the variability in noncontact ACL tears (Uhorchak *et al.*, 2003). This model also accurately predicted all the uninjured participants and predicted 75% of the noncontact ACL injuries (Uhorchak *et al.*, 2003). The paper therefore suggested that the risk factors studied here were more applicable to females and remained unknown for males (Uhorchak *et al.*, 2003).

Nevertheless, for GJL, the findings of Ramesh *et al.* also showed that ACL injured subjects tend to have more joint laxity, especially those with knee hyperextension (Ramesh *et al.*, 2005). Furthermore, 8 out of 10 other papers in the review by Smith *et al.* (2012a) also indicated that notch width or NWI were possible risk factors, providing support that these internal factors could contribute to an increase in ACL injury risks.

Two papers in the review by Smith *et al.* (2012a) did not support notch width or NWI as significant internal risks factors of ACL injuries. The first study by Schickendantz *et al.* (1993) did not find any significant differences in NWI or notch width among groups with bilateral ACL injuries, unilateral ACL injuries, or no injuries, using radiograph images.

The other was a case-control study that was carried out for 305 male National Basketball Association players over a long period of 11 years (Lombardo *et al.*, 2005). NWI was shown not to be significantly different between injured and non-injured players (Lombardo *et al.*, 2005). As also discussed, notch width and GJL could only account for 15% of the variability of non-contact ACL injuries for males (Uhorchak *et al.*, 2003). Together, the studies by Lombardo *et al.* (2005) and Uhorchak *et al.* (2003) suggest that while notch width or NWI could be possible ACL risk factors, they could also be less of an impact on males, especially in the sport of basketball. Another study on 50 male and 50 female basketball players also indicated that there were no significant differences in NWI between the two genders (Anderson *et al.*, 2001). The ACL is thought to be impinged against the notch at certain knee positions and at the extremes of joint motion (Smith *et al.*, 2012a) and perhaps these are not positions commonly occurring during basketball. Thus, notch width or NWI was not considered in this PhD thesis, which mainly used basketball players as participants. Furthermore, there are also no consistent methodologies to obtain notch width data (Griffin *et al.*, 2006) and it is not known if higher ACL injury risks are caused by the size and shape of the notch itself, the size of the ACL or an interaction between the two (Smith *et al.*, 2012a).

#### **2.3.1.1.2 Tibial Slope and Foot Pronation**

Based on other research studies, the tibial slope (Smith *et al.*, 2012a) and foot pronation appear to be likely risk factors (Beckett *et al.*, 1992, Woodford-Rogers *et al.*, 1994, Loudon *et al.*, 1996, Allen *et al.*, 2000, Trimble *et al.*, 2002). The tibial slope is found to be a potential new risk factor for ACL injuries among recent papers cited in Smith *et al.*

(2012a)'s review. The tibial slope measures the geometry of the tibial plateau. However, among the papers no commonality can be found about the location of the slope that contributes to the risk. This adds to the complexity of controlling this factor. Sagittal plane loading appears to be a major contributor of the ACL loading (Yu *et al.*, 2007). However, for foot pronation, it is associated with increased internal tibial rotation (Loudon *et al.*, 1996) and does not appear to be correlated to knee and ankle joint kinematics and kinetics in both sagittal and coronal planes (Mitchell *et al.*, 2008). Furthermore, except for the study by Woodford-Rogers *et al.* (1994), the control groups for the other studies were not matched to the injured groups by sport and level of competition (Beckett *et al.*, 1992, Loudon *et al.*, 1996, Allen *et al.*, 2000, Trimble *et al.*, 2002, Smith *et al.*, 2012a). Uninjured subjects might have participated less in sports associated with higher ACL injury risks. It did not consequently mean that they were less vulnerable to ACL injuries. Control groups in the tibial slope review (Smith *et al.*, 2012a) also often had other knee problems and significant differences observed could be due to the existence of these other knee problems. As a result, the studies that determine slope (Smith *et al.*, 2012a) and foot pronation (Beckett *et al.*, 1992, Woodford-Rogers *et al.*, 1994, Loudon *et al.*, 1996, Allen *et al.*, 2000, Trimble *et al.*, 2002) as risk factors are less convincing. These internal factors were therefore not considered in this PhD research.

### **2.3.1.2 Hormonal Risk Factors**

Sex hormones are often studied in an attempt to determine why female athletes have higher ACL injury incidence rates as compared to their male counterparts (Smith *et al.*, 2012b). Although disparity in injury rates at different stages of the female menstrual

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cycle suggests hormonal influence, more research is required to gain better understanding (Griffin *et al.*, 2006, Renstrom *et al.*, 2008, Smith *et al.*, 2012b). Much of the work has also been done with animal subjects and it is not known if results are also applicable to humans (Renstrom *et al.*, 2008). Effects of sex hormones on injury risk are also not clear (Shultz, 2007b). Thus, while sex hormones seem to have an influence on ACL injury risks, not enough knowledge is available to control this effectively. However, some literature seemed to suggest that more ACL injuries occur in the early or late follicular phase rather than in the luteal phase (Griffin LY *et al.*, 2006, Shultz, 2007b). Taking this into consideration and also to avoid discomfort due to premenstrual syndromes (2-3 days before period starts) and menstrual pain (days 1-6), measurements for the females in this study were taken during the late follicular phase (days 7-11, for a 28-day cycle) of their menstrual cycle.

### **2.3.1.3 Neuromuscular Risk Factors**

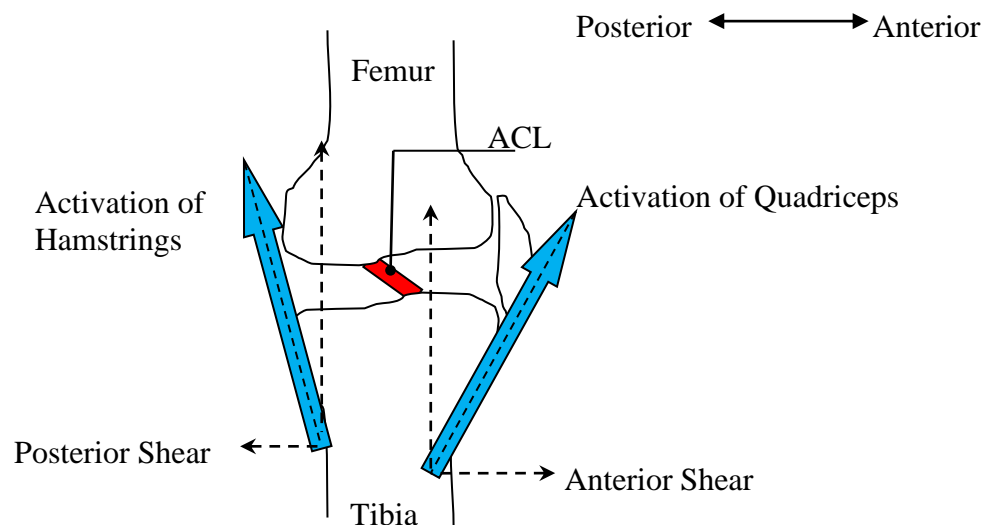
#### **2.3.1.3.1 Effects of Hamstring Strengthening on ACL Injury Risks**

Hamstrings flex the knee and quadriceps extend the knee (Hartigan *et al.*, 2011) and thus together form the opposing muscle groups (see Figure 4). When hamstrings are activated, they produce a posterior tibial shear force (Draganich *et al.*, 1990) that helps reduce ACL loading. On the other hand, quadriceps muscle activation contributes to the anterior shear force, resulting in ACL loading (Yu *et al.*, 2007, see Figure 5). ACL loading induced by the quadriceps become more pronounced at low knee flexion angles (Renström *et al.*, 1986, Draganich *et al.*, 1990, Li *et al.*, 1999, DeMorat *et al.*, 2004). This is when the opposing effect by hamstrings becomes less effective (Renström *et al.*, 1986, Li *et al.*,

1999, Yu *et al.*, 2005) as hamstrings' line of action is only in an ideal position to resist anterior movements when the knee is flexed (Shultz, 2007a). This resulted in increase in knee extension moments, increasing the anterior shear and therefore resulting in large sagittal forces (see Figure 5).

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**Figure 4. Movements of the knee during activation of quadriceps and hamstring muscles (Adapted from Gollogly *et al.*, 2011).**



**Figure 5. Schematic drawing to illustrate quadriceps-hamstring muscles' effect on the anterior cruciate ligament.**

Since ACL loading is mainly caused by the anterior shear force (Yu *et al.*, 2007), weak hamstring activation and strong quadriceps activation will increase anterior pull and therefore increase ACL strain. There is therefore convincing theoretical evidence to suggest that strengthening the hamstrings and encouraging joint positions that improve

hamstrings' efficiency, can be included in ACL injury preventive training programmes (Shultz, 2007a).

However, hamstring activation becomes less efficient in reducing ACL loading when knee flexion is less than  $15^\circ$  (Li *et al.*, 1999, Yu *et al.*, 2005, see Figures 6 and 7), and can occur in single-leg landings (Weinhandl *et al.*, 2010). Hamstring was also observed to have been less activated than the quadriceps in the study by Morgan *et al.* (2014), during the landing phase of single-leg landings. This suggests that when athletes land in low knee flexion angles during single-leg landings, hamstring strengthening used in injury prevention programmes might not fully help to protect the ACL. It is therefore important to further explore how various muscles at the knee act together to reduce ACL injury risks during single-leg landings. This is especially so prior initial contact with the ground, the point at which ACL strain reaches its peak (Lamontagne *et al.*, 2008).

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**Figure 6. In-situ forces in ACL as a function of knee flexion angles. Hamstring co-activation resulted in significant reduction of in-situ forces in the ACL only from  $15^\circ$  to  $60^\circ$ , as compared to the sole activation of the quadriceps (Adapted from Li *et al.*, 1999).**

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**Figure 7. ACL force as a function of hamstring forces and knee flexion angles (males). Hamstring co-activation did not reduce ACL loading when knee flexion angles were less than  $15^\circ$  (Modified from Yu *et al.*, 2005).**

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### 2.3.1.3.2 Effects of Fatigue on ACL Injury Risks

Alentorn-Geli *et al.* (2009a) also suggested that fatigue could have an effect on neuromuscular control, leading to increased ACL injury risks (Wojtys *et al.*, 1996, Chappell *et al.*, 2005, Melnyk *et al.*, 2007). Chappell *et al.* (2005) found that after fatigue had set in for stop-jump tasks, men and women more often landed in positions associated with higher ACL injury risks. The study was carried out on healthy recreational athletes using a method that induced fatigue, similar to those felt during basketball, soccer and volleyball competitions (Chappell *et al.*, 2005). Another result by Melnyk and Gollhofer, using healthy subjects training twice a week, also showed that hamstring fatigue could lead to increased anterior tibial translation (Melnyk *et al.*, 2007), resulting in a rise in ACL strain. Fatigue was found to result in an increase in anterior tibial translation by 32.5% (Wojtys *et al.*, 1996). College basketball players also showed an 18-20% increase in knee laxity after 90 min of training (Steiner *et al.*, 1986). In a separate study on female athletes by Borotikar *et al.* (2008), the combined effects of fatigue and decision-making showed an increase in ACL injury risks

Thus, while there are many neuromuscular training programs, once fatigue sets in, athletes can still lose control of their muscles and be exposed to increased ACL injury risks. Furthermore, professional athletes can also experience fatigue especially towards the end of the game play and temporarily after high intensity manoeuvres during the competition (Mohr *et al.*, 2003). Therefore, this may limit the effectiveness of using preventive training programmes on non-contact ACL injury prevention among athletes.

This will be further discussed in Section 2.4.

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In view of the evidence found in literature, the following internal risk factors were therefore accounted for in this PhD research:

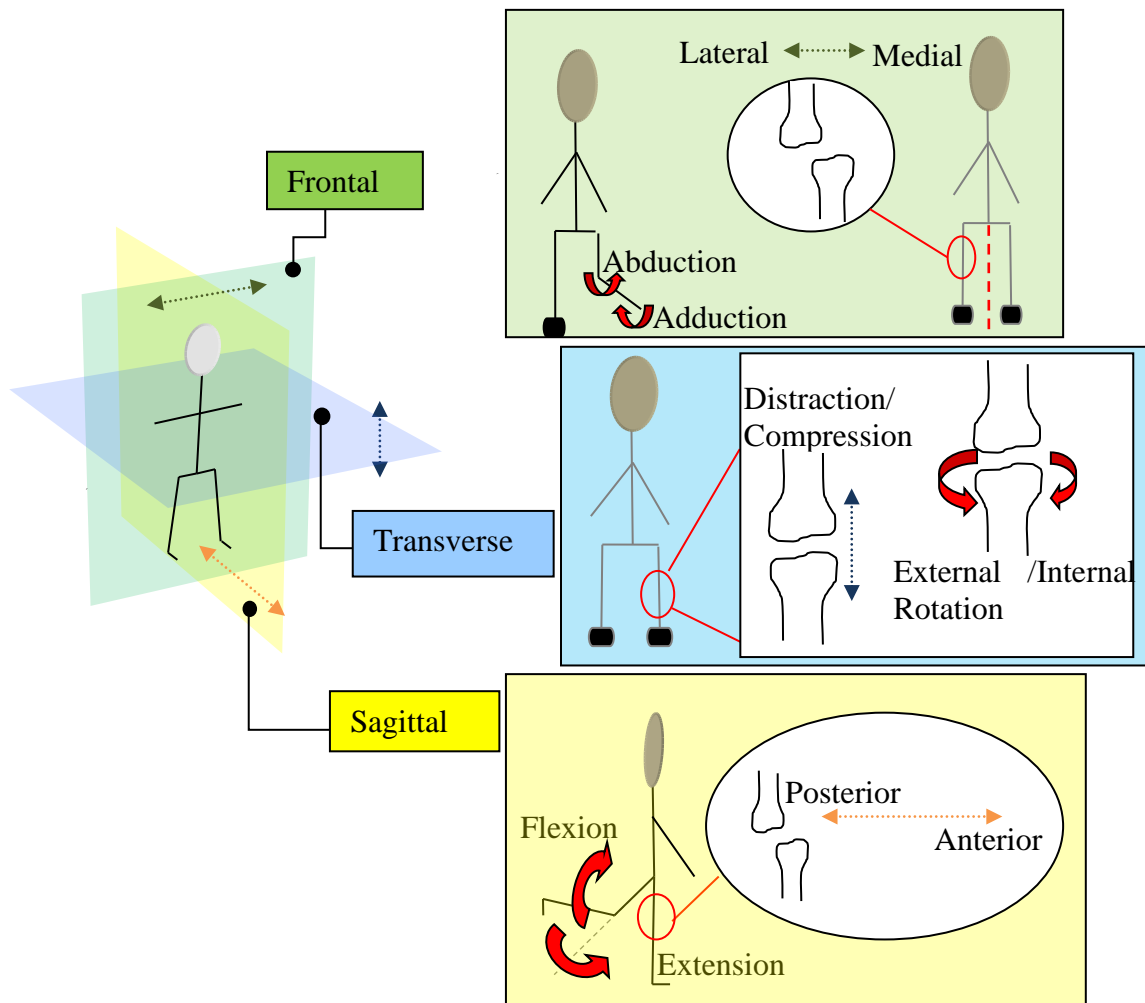
- GJL for both males and females, as well as BMI for females, were found to be internal risk factors of ACL injuries (Uhorchak *et al.*, 2003). GJL and BMI were recorded in this research. GJL was measured using the commonly used Beighton score (Shultz *et al.*, 2007).
- Literature involving hormonal risk factors also suggested higher ACL injuries during the early and late follicular phase rather than in the luteal phase (Griffin *et al.*, 2006, Shultz, 2007b). Taking this into consideration and to avoid discomfort due to premenstrual syndromes (2-3 days before period starts) and menstrual pain (days 1-6), measurements for the females in this study were taken during the late follicular phase (days 7-11, for a 28-day cycle) of their menstrual cycle.
- Hamstring is essential to oppose the anterior tibial force produced by quadriceps but its role is strongly affected by its reduced efficacy at low knee flexion angles during single-leg landings. There is therefore a need to understand how various muscles at the knee act together to reduce ACL injury risks during single-leg landings.
- Fatigue was found to influence neuromuscular control. Participants were therefore allowed proper rest throughout the studies to limit the influence of fatigue on the results.

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### **2.3.2 Injury Mechanisms of the ACL Injury**

Knowledge of the injury mechanisms leading to non-contact ACL injuries is said to play an important part in preventing non-contact ACL injuries (Yu *et al.*, 2007). This section focuses on the injury mechanisms leading to a non-contact ACL injury.

The knee joint moves in all 3 planes (sagittal, frontal and transverse) with 6 degrees of freedom (3 translations and 3 rotations) (Quatman *et al.*, 2010). In the sagittal plane, the knee translates anteriorly and posteriorly, and rotates by flexion and extension (Quatman *et al.*, 2010). In the frontal (or coronal) plane, the knee translates medially and laterally, and rotates by abduction and adduction (Quatman *et al.*, 2010). In the transverse plane, the knee translates by compression and distraction, and rotates by internal and external rotation (Quatman *et al.*, 2010). These six degrees of freedom are illustrated in Figure 8.



**Figure 8. Schematic drawing of the translation and rotation motions of the knee joint.**

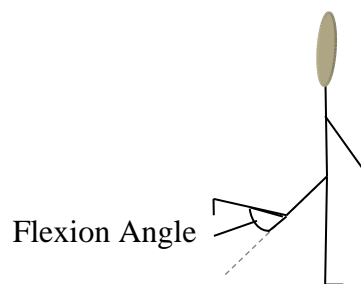
### 2.3.2.1 Loading in the Sagittal Plane

Forces in the three different planes are said to contribute to the ACL loading. In the sagittal plane, the ACL plays the primary role to prevent the tibia from moving anteriorly to the femur (Hartigan *et al.*, 2011). ACL loading is said to be mainly contributed by the anterior shear force (Yu *et al.*, 2007). The anterior shear force can increase when the knee

flexion angle is small; when there is relatively greater quadriceps muscle activation (as compared to the hamstrings); or could be a result of a big posterior ground reaction force (Yu *et al.*, 2007).

Small flexion angles are usually referred to as those that are between  $0^{\circ}$  and  $30^{\circ}$  (Myer *et al.*, 2004). The association of small knee flexion angles to higher ACL injury risks is well supported by video analyses, cadaver studies and biomechanical explanations:

- Small knee flexion angles of less than or equal to  $30^{\circ}$  had been observed in the videos of athletes at initial contact prior onset of ACL injuries (Olsen *et al.*, 2004, Cochrane *et al.*, 2007, Krosshaug *et al.*, 2007, Koga *et al.*, 2010). Small knee flexion angles imply that knees were closer to full extension prior injuries as shown in Figure 9.

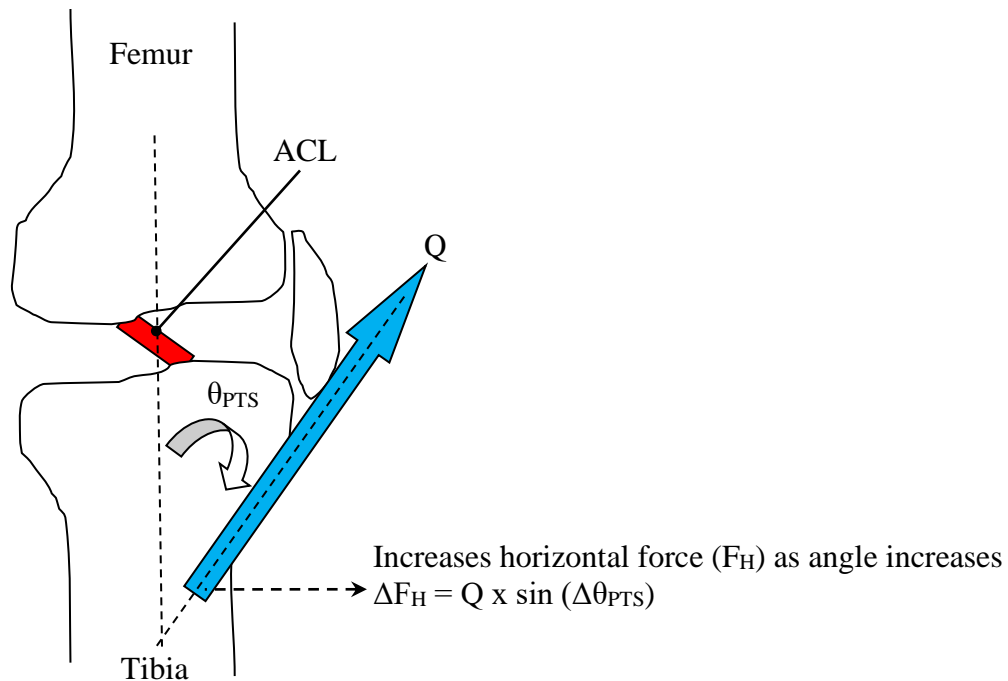


**Figure 9. Definition of the knee flexion angle.**

- Cadaver studies also showed that ACL loading increases with smaller knee flexion angles. Li *et al.* (1999) found that the highest *in-situ* force was at  $15^{\circ}$  knee flexion for human cadaveric knees with no ligament injuries although anterior

- 
- translation was highest at 30°. The study also suggested that hamstring intervention to reduce ACL loading was most efficient between 15° to 60° knee flexion (Li *et al.*, 1999). Hamstring co-activation also did not result in a reduction of ACL loading at knee flexion angles less than 15° in the computer simulation study by Yu *et al.* (2005). In another cadaver study, it was found that co-activation of hamstrings with quadriceps did not help reduce ACL strain from 0° to 30° knee flexion (Renström *et al.*, 1986).
- Biomechanically, there is also theoretical support to explain why smaller knee flexion angles lead to increases in anterior shear forces. Reducing knee flexion angles increases the patella tendon-tibia shaft angle (Yu *et al.*, 2007). Patellar tendon, also known as the patellar ligament, connects the knee cap to the tibia and is extended from the quadriceps (Marieb, 2004). The patella tendon-tibia shaft angle is the angle between the patella tendon and the longitudinal axis of the tibia (Yu *et al.*, 2007). At a fixed quadriceps muscle force (Q), increasing the patella tendon-tibia shaft angle ( $\theta_{PTS}$ ) increases the horizontal force ( $F_H$ ) component that result in an increase of the anterior shear force (Yu *et al.*, 2007, see Figure 10). Equation (1) shows that as  $\Delta\theta_{PTS}$  increases,  $\Delta F_H$  also increases. This contributes to the increase of the anterior shear force.

$$\Delta F_H = Q \times \sin (\Delta \theta_{PTS}) \quad (1)$$



**Figure 10. Schematic drawing showing the effect of an increased patella tendon-tibia shaft angle  $\theta_{PTS}$ .**

- Another biomechanical explanation is that a reduction in knee flexion angles increases ACL force by increasing ACL elevation angle and deviation angle (Li *et al.*, 2005, Yu *et al.*, 2007). The elevation and deviation angles describe the orientation of the ACL as shown in Figure 11.

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**Figure 11. Positions of deviation and elevation angles (Redrawn from Diagram by Li *et al.* (2005)).**

The anterior shear force in the anterior-posterior direction ( $F_{AS}$ ) is related to the ACL loading ( $F_{ACL}$ ) along its longitudinal axis by the following relationship (Li *et al.*, 2005):

$$F_{ACL} = \frac{F_{AS}}{\cos \alpha_E \cos \beta_D} \quad (2)$$

Thus as the elevation ( $\alpha_E$ ) and deviation ( $\beta_D$ ) angles increase, due to reduction of knee flexion angles, the  $(\cos \alpha_E \cos \beta_D)$  term becomes smaller. With the same anterior shear force and a smaller  $(\cos \alpha_E \cos \beta_D)$  term, ACL loading ( $F_{ACL}$ ) increases with reduced knee flexion angles.

As discussed in the previous section (Section 2.3.1), quadriceps muscle activation contributes to the anterior shear force that results in ACL loading (Yu *et al.*, 2007). This effect is also more pronounced at low knee flexion angles (Draganich *et al.*, 1990, Li *et al.*, 1999). This is when the opposing influence by hamstrings to restrain the ACL becomes less effective at these low flexion angles (Renström *et al.*, 1986). All these further highlight the need to understand how non-contact ACL injury risks can still be reduced under weak hamstring activation and strong quadriceps activation during low

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knee flexion angles. It also shows the need to avoid low knee flexion angles in order to reduce non-contact ACL injury risks.

Finally, it was also observed that posterior ground reaction forces create a flexion moment at the knee that can result in greater quadriceps activity to counter the action to keep the leg straight (Yu *et al.*, 2007). This leads to greater anterior tibial shear.

### **2.3.2.2 Multi-Planar Loading**

Loading in the sagittal plane appears to be a major contributor to ACL loading (Yu *et al.*, 2007). However, studies based on video analyses (Olsen *et al.*, 2004, Cochrane *et al.*, 2007, Koga *et al.*, 2010), cadaver studies (Berns *et al.*, 1992, Markolf *et al.*, 1995) and MRI bone bruise scan (Koga *et al.*, 2010) do suggest multi-planar effects could be significant to ACL injuries.

ACL plays a secondary role in restraining abduction/adduction (valgus/varus) moments (Arnheim *et al.*, 1993). In fact, it is also reported that removing the ACL results in loss of valgus and varus steadiness at 20° to 40° knee flexion (Olsen *et al.*, 2004). An often cited study by Markolf *et al.* (1995) showed that the addition of valgus moment to pure anterior loading of the knee, almost doubled ACL force at 30° of knee flexion. Valgus collapse during ACL injury was also recorded based on video analyses (Olsen *et al.*, 2004, Koga *et al.*, 2010) and a 20-year epidemiology study of 1700 athletes (Kobayashi *et al.*, 2010). A cohort study also found greater ACL injury risks in female athletes who jump-landed with greater valgus moments (Hewett *et al.*, 2005). Therefore, increase in

valgus moments would increase ACL injury risks, especially when applied together with anterior shear force at the knee. The study by Markolf *et al.* (1995) also showed an increase in ACL loading with the addition of varus moments to anterior force loading. However, the increase was less pronounced especially at knee flexion angles greater than 20° (see Figure 12). All these once again suggest that the low knee flexion angles could increase the risk of the ACL injury resulting in greater ACL forces experienced.

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**Figure 12. Resultant anterior cruciate ligament force under combined effects of (A) anterior force and internal tibial torque; (B) anterior force and external tibial torque; (C) anterior force and varus moment; (D) anterior force and valgus moment using cadavers (Adapted from Markolf *et al.*, 1995).**

From Figure 12, it can be observed that the addition of internal rotation of the tibia to the anterior force loading increased the ACL force at close to knee flexion angles of less than 30° (Markolf *et al.* 1995). Internal rotation was also observed in the video analyses (Cochrane *et al.*, 2007, Koga *et al.*, 2010). External tibial torque resulted in a decrease in ACL force when external tibial torque was added to anterior force loading (Markolf *et al.* 1995, see Figure 12). However, there were also some other studies that had reported increases in ACL strain through both internal and external rotation (Berns *et al.*, 1992, Fleming *et al.*, 2001a). External rotation was also observed in the video analysis (Olsen *et al.*, 2004). Nevertheless, a MRI study of non-contact ACL injury knees supported the occurrence of an internal rotation during injury (Speer *et al.*, 1992) and it was suggested that external rotation occur only after injury (Koga *et al.*, 2010). Combining the evidence

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from the above-mentioned studies, tibial rotation, especially internal rotation with respect to the femur, increases ACL injury risks.

In conclusion, playing conditions with low knee flexion angles, valgus and internal rotation moments are associated with higher ACL injury risks and have been observed in video analyses prior to ACL injuries (Olsen *et al.*, 2004, Cochrane *et al.*, 2007, Krosshaug *et al.*, 2007, Koga *et al.*, 2010). Low knee flexion angles are often considered to be between 0° to 30° (Myer *et al.*, 2004). Cadaver studies (Berns *et al.*, 1992, Markolf *et al.*, 1995) and MRI scans (Koga *et al.*, 2010) also support these findings. It is also likely that ACL injury may be a result of multi-planar loading (Berns *et al.*, 1992, Markolf *et al.*, 1995) although the forces in the sagittal plane are probably the major contributors (Yu *et al.*, 2007). Yet, many ACL training prevention programmes do not incorporate movements from different planes (Quatman *et al.*, 2010).

This research therefore incorporated multi-planar considerations in the study. Based on Markolf *et al.* (1995)'s study, an addition of valgus moments to anterior shear forces increased ACL loading more than that contributed by the addition of internal rotation moments. Thus, the major contributors to the non-contact ACL injuries, i.e., knee flexion angles and knee valgus, would be further investigated in this study.

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## 2.4 Current Methodologies for Preventing ACL Injury

### 2.4.1 Effectiveness of Preventive Training Programmes

There has been increasing attention focusing on ACL injury prevention in the last 10 years (Rishiraj *et al.*, 2009). Training routines (Renstrom *et al.*, 2008) and knee braces (Lin *et al.*, 2008) have been used to help reduce knee injuries.

Increasing number of training prevention programmes have been developed, focusing on ACL injury prevention (Renstrom *et al.*, 2008). As discussed in Section 2.3.1, internal risk factors, such as anatomical and hormonal factors, influence ACL injury risks. However, these factors are difficult to be changed, unlike neuromuscular and biomechanical factors (Sugimoto *et al.*, 2015). Therefore, improving muscular control and landing techniques are incorporated into preventive training programmes, which include plyometrics, balance, technique training, proprioception, neuromuscular training, agility training, stretching, training for hamstring/quadriceps strength ratio, trunk/core control training, and education to increase awareness of injury mechanisms (Alentorn-Griffin *et al.*, 2006, Geli *et al.*, 2009b, Taylor *et al.*, 2015).

Such training programmes have resulted in a reduction of ACL injuries in the intervention group of several of the studies (Griffin *et al.*, 2006) or in programmes targeting female athletes (Sugimoto *et al.*, 2015, Taylor *et al.*, 2015). However, much is still not known about how these techniques work in reducing ACL injury risks. Studies often focused on training programmes consisting of several training protocols and it is

hard to assess the effectiveness of each training protocol on reducing ACL injuries (Alentorn-Geli *et al.*, 2009b). Therefore, the exact exercises to be included in prevention programmes, the order of these exercises and the duration of the prevention programmes are still uncertain (Griffin *et al.*, 2006, Myklebust *et al.*, 2013, Taylor *et al.*, 2015). The mechanism, by which a training protocol is effective in reducing ACL injury risks, is also not fully understood (Griffin *et al.*, 2006, Donnelly *et al.*, 2012). As discussed in Section 2.3.1, hamstring strengthening used in injury prevention programmes might also not fully help to protect the ACL during single-leg landings. This is because hamstring activation becomes less efficient in reducing ACL loading when knee flexion is less than 15° (Li *et al.*, 1999, Yu *et al.*, 2005), typically occurring during single-leg landings (Weinhandl *et al.*, 2010). Fundamentally, creating an effective prevention training programme also requires substantial knowledge of why ACL injuries occur in the first place. This knowledge, by itself, is also still lacking (Engebretsen *et al.*, 2007, Ali *et al.*, 2010). Besides, few of the studies are randomized controlled trials and their sample sizes are not large enough to yield statistically conclusive results (Griffin *et al.*, 2006). It would also be difficult to ask athletes to land subjectively using ‘soft’ landings, associated with higher knee flexion angles (Devita *et al.*, 1992, Garrett *et al.*, 2007, Laughlin *et al.*, 2011), especially during competitions under stress.

Furthermore, athlete training also often focuses on performance rather than injury prevention (Garrett *et al.*, 2007). This is especially so for the athletes who are in their teenage years, when body size and muscles are drastically changing (Garrett *et al.*, 2007). This is also the possible age period with increased ACL injury risks (Garrett *et al.*, 2007).

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In fact, the study by Sanders *et al.* (2016) also showed that most ACL injuries occur between the age group of 14 to 18 years old for females and between 19 to 25 years old for males. Effectiveness of training programmes would therefore be affected if focus by coaches were mainly on excelling in performance during the teenage years when higher ACL injury risks occur. Furthermore, it is found that 108 individuals are required to prevent a non-contact ACL injury for females by the neuromuscular training (Sugimoto *et al.*, 2012). This could make it harder for coaches to be convinced to work together to reduce ACL injuries (Sugimoto *et al.*, 2012).

Finally, as previously discussed, athletes also experience fatigue (Mohr *et al.*, 2003) which could alter neuromuscular control (Alentorn-Geli *et al.*, 2009a), thus still further putting athletes at risk of an ACL injury. In fact, ACL research was said to have been unsuccessful thus far in helping to reduce ACL injuries that have continued to increase by 50% over the past decade (Donnelly *et al.*, 2012).

Therefore, while training programmes have the potential to reduce risks of ACL injuries, the uncertainties that still exist today in fully understanding the mechanisms, are obstacles to effective implementation of such programmes. Furthermore, when fatigue sets in, the reduction in muscle control would further expose athletes to the risks of ACL injuries. Braces would then be useful to prevent athletes from going into knee positions associated with high ACL injury risks, as identified in Section 2.3.

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## 2.4.2 Effectiveness of Knee Braces

Today, knee braces have been used to provide support to injured or unsteady knees (Rishiraj *et al.*, 2009). They have also been used to prevent or to decrease the seriousness of knee injuries (Rishiraj *et al.*, 2009). Prophylactic knee braces (PKBs) are non-customized braces, aimed to avoid or lessen the impact of knee injuries (Rishiraj *et al.*, 2009). Functional knee braces (FKBs) are personalized braces, used to provide support to unsteady knees (Rishiraj *et al.*, 2009). Both PKBs and FKBs have been studied for their effectiveness in preventing ACL injuries (Rishiraj *et al.*, 2009).

Epidemiologic studies and laboratory experiments had been conducted to study the effectiveness of knee braces. Epidemiologic studies were reported mainly from the mid-1980s to 1990s (Rishiraj *et al.*, 2009). Sitler *et al.* (1990) found that the incidence of knee injuries (medial collateral ligament (MCL) and ACL injuries) decreased with the use of PKBs but provided too small an ACL injured sample size for it to be statistically analysed meaningfully. Severity of ACL injuries was also reported not to be significantly reduced (Sitler *et al.*, 1990). Another report did not find any improvements with brace intervention due to muscle cramping in the triceps surae muscle group (Rovere *et al.*, 1987). These are relatively old studies reported before 2000 that used older brace designs. However, they are still often cited today in assessing brace effectiveness. Brace designs have changed over the years and these studies might no longer be representative of brace effectiveness of other newer brace designs used today.

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There is also insufficient evidence to show that braces can handle all mechanisms of ACL injuries. Besides, studies are often limited to the use of Donjoy braces (Djo LLC, CA, USA) probably due to the lack of other commercial braces that primarily target ACL injury prevention in recent years. The effectiveness of braces in reducing anterior loading is not conclusive. Two studies that used the same Donjoy Legend brace (Djo LLC, CA, USA) have reported differing results (Fleming *et al.*, 2000, Ramsey *et al.*, 2001). Fleming *et al.* (2000) observed a significant reduction of ACL strain under anterior shear loading up to 130N while Ramsey *et al.* (2001) found no consistent decrease in anterior tibial translation using the brace. The differences could be due to the use of different subject pool, dissimilar test methods and different measurement methods. Fleming *et al.* (2000) used healthy participants, carried out the study using a knee loading fixture and measured ACL strain directly using a differential variable reluctance transducer. On the other hand, Ramsey *et al.* (2001) used ACL-deficient participants, carried out the study using a one-legged jump protocol and used Steinmann traction pins attached to the femur and tibia, marker clusters and motion capture as its measurement method.

Based on the Donjoy Legend brace (Djo LLC, CA, USA) and using the knee loading fixture, the brace also did not reduce strain on varus-valgus moments (10 Nm) in both weight bearing and non-weight bearing knees (Fleming *et al.*, 2000).

On the other hand, braces have been reported to be effective in reducing injury risk by increasing knee flexion (Lin *et al.*, 2008, Stanley *et al.*, 2011) and decreasing internal tibial rotation (Fleming *et al.*, 2000, Giotis *et al.*, 2011). However, the braces (Djo LLC,

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CA, USA) that increase knee flexion make use of a ‘knee extension constraint’ feature (Lin *et al.*, 2008, Stanley *et al.*, 2011) that constraints knee extension angles. Thus, this could mean the brace design with the ‘knee extension constraint’ feature is effective in increasing knee flexion. Nevertheless, this result should not be generalized across all brace models or designs that may not have such a feature. In fact, one other study that used a different brace (K300 MX, POD Orthotic Pty Ltd, VIC, Australia) did not detect a significant difference in knee flexion angles at initial contact (Ewing *et al.*, 2016) for landing. Initial contact is when ACL strain is highest during landing (Lamontagne *et al.*, 2008).

As for decreasing internal tibial rotation, using the knee loading fixture, internal rotation of the Donjoy Legend Brace (Djo LLC, CA, USA) is found to be reduced in non-weight bearing conditions only (Fleming *et al.*, 2000). The other study that showed a decrease in internal tibial rotation involves functional activities such as landing and pivoting and uses PKB and patellofemoral braces (Giotis *et al.*, 2011). The brands of these braces were however not named (Giotis *et al.*, 2011). Since brace effectiveness seems to be influenced by brace designs, thus the results reported by Giotis *et al.* (2011) may not be generalized to all other brace models.

In terms of reducing peak vertical ground reaction forces, brace use was effective with the Extreme® ligament knee braces (Össur, BC, Canada) (Rishiraj *et al.*, 2012) but not when K300 MX knee braces were used (POD Orthotic Pty Ltd, VIC, Australia) (Ewing *et al.*, 2016). As before, effectiveness of braces is influenced by specific brace designs.

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Therefore, brace models should always be reported in the assessment of the effectiveness of braces.

In fact, the effectiveness of using braces to prevent knee injuries remains largely inconclusive (Griffin *et al.*, 2006, Pietrosimone *et al.*, 2008, Rishiraj *et al.*, 2009). Critics have also suggested that studies should examine the interactions of internal and external factors (Griffin *et al.*, 2006); should have improved randomized studies (Pietrosimone *et al.*, 2008); should study the mechanics of how PKBs prevent ACL injuries (Alentorn-Geli *et al.*, 2009a); and should ensure that improvements are not due to psychological effects (Alentorn-Geli *et al.*, 2009a). Athletes have also remained unwilling to use a knee brace for fear of a drop in their level of performance (Rishiraj *et al.*, 2009). In fact, the added weight of braces was shown to result in early onset of fatigue and thus affected performance (Highgenboten *et al.*, 1991, Styf *et al.*, 1992, Lundin *et al.*, 1998,) although with practice, speed and agility could still resume after 12-14 hours of brace use (Rishiraj *et al.*, 2011). Migration of braces could also affect performance unless migration is less than 2cm (Singer *et al.*, 2008). However, knee braces were found to migrate easily and in one study, 14 commonly used FKBs were tested and they all of them were reported to have migrated (Brownstein 1998).

The uncertainties of the effectiveness of knee braces in the prevention of ACL injuries are also due to the way effectiveness of braces is being assessed. Currently, there is no commonly accepted method to test the effectiveness of braces. Some studies had used healthy subjects while others studied injured subjects. Brace design varied with time and

not all brands adopted similar designs. Thus, the effectiveness of such braces is hard to be compared. Test procedures or manoeuvres are also different across the various studies. Yet, the results are often put together for comparison and thus it is difficult to make any conclusive statement regarding the effectiveness of knee braces for ACL injury prevention. To further complicate matters, internal risk factors that could result in conflicting results are often not included in studies (Ali *et al.*, 2010). For proper comparison to be made, standardized experimental methodologies should be developed. Studies involving dynamic activities would be more appropriate to simulate conditions of playing situations during ACL injuries. Brace brands or designs have to be provided as effectiveness of each design feature is different. Generalizing brace effectiveness across different brands becomes confusing.

Athletes have been unwilling to use a knee brace for fear of a drop in their level of performance (Rishiraj *et al.*, 2009). It would even be more difficult to persuade these athletes to wear knee braces to prevent ACL injuries since there is still insubstantial evidence that knee braces help prevent ACL injuries. Unlike knee braces, ankle braces are used commonly to prevent ankle sprains (Shaw *et al.*, 2008). Therefore, brace solutions based on knowledge from a new perspective from the foot may bring new insights to ACL injury prevention.

## **2.5 Potential Effects of Foot Landing**

Lateral ankle sprain is the most common injury in sport and occurs mostly during landing in jumping sports like basketball and volleyball (Shaw *et al.*, 2008). Protective equipment

such as lace-up ankle braces is used commonly to prevent injury (Shaw *et al.*, 2008). Studies have also been carried out to determine the effects of ankle braces or foot orthotics on the knee joint, influencing lower extremity biomechanics (DiStefano *et al.*, 2008, Joseph *et al.*, 2008, Jenkins *et al.*, 2009, Boros *et al.*, 2010). Although not specifically designed for ACL injury prevention applications, restraints at the ankle through ankle braces and foot orthotics were found to increase knee flexion (DiStefano *et al.*, 2008), reduce knee valgus (Joseph *et al.*, 2008, Boros *et al.*, 2010) and reduce tibial internal rotation (Jenkins *et al.*, 2009). This suggests that ankle positions can have an effect on knee joint positions and has the potential to be investigated as a new area for non-contact ACL injury risk prevention. Most studies on ACL injuries focused on the knee (e.g. Hewett *et al.*, 2005, Russell *et al.*, 2006), hip (e.g. Pollard *et al.*, 2004, Yu *et al.*, 2006), or trunk (e.g. Sheehan *et al.*, 2012, Sasaki *et al.*, 2015). Since the foot and ankle form the initial parts of the lower extremity kinetic chain during landing, the foot landing positions may have an effect on the transmission of forces from the ankle to the knee (Ali *et al.*, 2010). Thus, studying ACL injuries from the foot landing viewpoint has the potential to provide new insights into reducing non-contact ACL injuries.

It has been reported that in the sagittal plane, landing with the greatest ankle plantar-flexion (fore-foot landing) resulted in the highest force absorption and decrease of peak vertical ground reaction force (Alentorn-Geli *et al.*, 2009a). Greater plantar-flexion moment was also correlated to less initial peak knee extension moment after initial contact, thus lowering ACL injury risks (Shimokochi *et al.*, 2009). Dancers and ice skaters were also found to have less non-contact ACL injuries and one of the possible

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reasons being that they often land on their toes (Griffin *et al.*, 2006). However, Cortes *et al.* (2007) studied the effects of sagittal foot-landing positions (fore-foot, rear-foot and natural) on lower extremity kinematics and found that knee flexion angle was significantly lower at initial contact for fore-foot landing, as compared to natural and rear-foot landing. Lower knee flexion angles have also been associated with higher non-contact ACL injury risks (Yu *et al.*, 2007). This appears to contradict the beneficial effect of fore-foot landing on reducing non-contact ACL injury risks especially resulting in lower knee flexion angles during initial contact, when ACL strain was found to be the highest (Lamontagne *et al.*, 2008). Since natural foot-landings are fore-foot landings (Cortes *et al.* 2007), it would be crucial to further and better understand how fore-foot landings would impact non-contact ACL injury risks.

While the study by Cortes *et al.* (2007) provided an important insight that changes in the sagittal foot landing angles could affect knee flexion angles, their study only focused on double-leg landing (Cortes *et al.* 2007). In fact, many other studies found in literature which specifically monitored foot landing in the sagittal plane to study its effects on the lower extremity during dynamic tasks, were also focused on double-leg landing (Kovács *et al.*, 1999, Valenzuela *et al.*, 2016). Since single-leg and double-leg landings result in different knee kinematics and muscle activation levels (Pappas *et al.*, 2007), double-leg landing results are unlikely applicable to single-leg landings, in which more non-contact ACL injuries are found (Donnelly *et al.*, 2012). Therefore, there seems to be an absence of clear understanding of how fore-foot landings affect ACL injury risks during single-

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leg landings. There is also a need to understand how knee flexion angles could be increased during initial contact, through the study of foot landings in the sagittal plane.

Foot restraints using ankle brace and medial post had also been found to reduce knee valgus angles (Joseph *et al.*, 2008, Boros *et al.*, 2010), suggesting that foot landing could also have an effect on knee valgus. As already discussed in Section 2.3.2, knee valgus is also associated with higher non-contact ACL injuries as athletes often land in knee valgus positions before sustaining ACL injuries (Krosshaug *et al.*, 2007). Few people have reported studies on the effect of foot rotation on knee valgus (Dempsey *et al.*, 2012, Ishida *et al.* 2014, Tran *et al.*, 2016). Ishida *et al.* (2014) studied how foot rotation positions affect shank rotation with respect to the thigh, during knee valgus. Their methodology required participants to shift from a standing position to a knee valgus position and this did not appear to fully represent the dynamic landing conditions (Ishida *et al.* 2014). Tran *et al.* (2016) found hip and knee biomechanics differences for toe-in and toe-out positions, as compared to the toe-forward position during double-leg landing. As before, double-leg landing results may not be applicable for single-leg landings as they result in different knee kinematics and muscle activation levels (Pappas *et al.*, 2007). Dempsey *et al.* (2012) found that maximum knee valgus moment was correlated with toe-out foot position during single-leg landings. However, their study did not examine the different foot rotation positions such as toe-forward and toe-in positions, nor study the effects of foot rotation at the initial contact of landing, whereby ACL strain was the highest (Lamontagne *et al.*, 2008). This presents a need to study the effects of foot rotation on knee valgus to a greater extent. This would in turn provide better

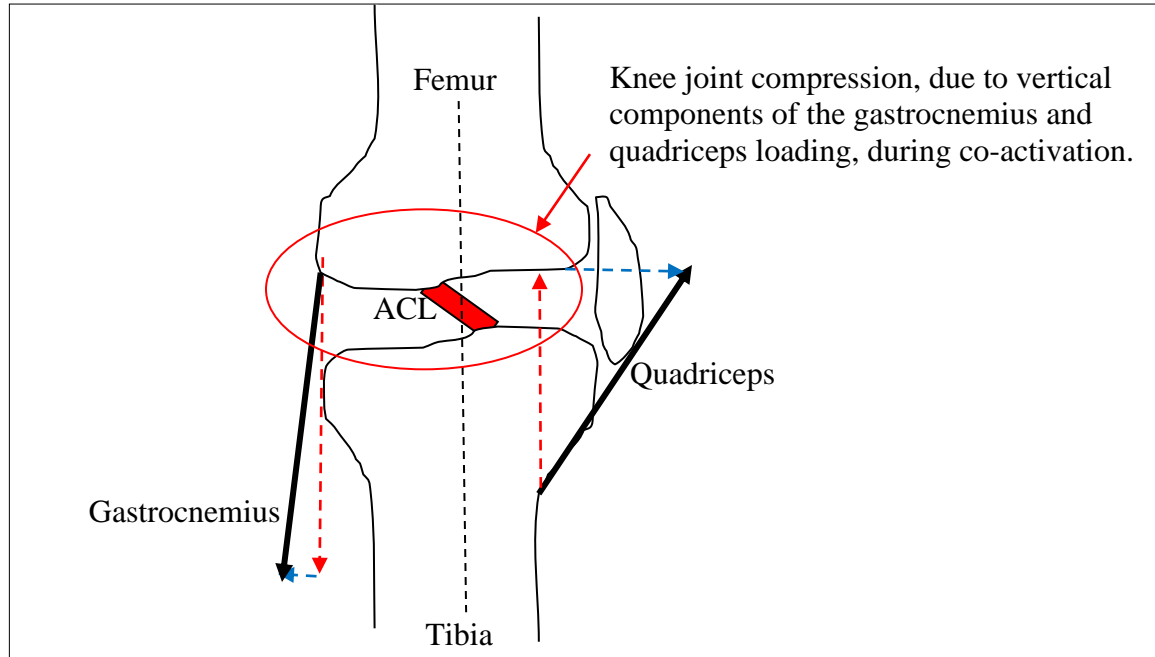
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understanding for non-contact ACL injury prevention, especially during initial contact with the ground. Dempsey *et al.* (2012) did not however find valgus moments to be significantly correlated to ankle inversion/eversion in landing. Thus, ankle inversion/eversion was not the focus of this thesis.

Furthermore, the hamstring is essential to opposing the anterior tibial force produced by quadriceps but its role is strongly affected and its efficacy is reduced at low knee flexion angles, especially for single-leg landings. Another knee flexor, the gastrocnemius, has therefore being a focus of research for a few studies to examine its role in ACL injury prevention (Dürselen *et al.*, 1995, Fleming *et al.*, 2001b, Elias *et al.*, 2003, Pflum *et al.*, 2004, Ali *et al.*, 2014, Morgan *et al.*, 2014, Adouni *et al.*, 2016). The gastrocnemius crosses both the knee joint and the ankle joint (Hartigan *et al.*, 2011). Gastrocnemius, together with the soleus, forms the triceps surae to be the main plantar flexors of the ankle (Martin, 2011). The gastrocnemius connects to the back of the femur and ends at the Achilles tendon (Hartigan *et al.*, 2011). It is less of a knee flexor when the ankle is plantar flexed and creates the most knee flexion moment when the knee is extended (Hartigan *et al.*, 2011). Gastrocnemius could be the reason for Cortes *et al.* (2007) to observe a change in knee flexion angles when plantar flexion angles change.

However, the role of the gastrocnemius in ACL loading has been rather unclear. Morgan *et al.* (2014)'s simulation study using 6 males showed that with the co-contraction of the quadriceps, increased gastrocnemius forces were associated with increased joint compression and reduced ACL forces during single-leg landings. Increased joint

compression limits anterior tibial translation by increasing friction at the joints, thereby providing knee joint stability (Hartigan *et al.*, 2011, Morgan *et al.*, 2014). The greater gastrocnemius forces also compensated for decreased hamstring forces in opposing the quadriceps, highlighting gastrocnemius' part in ACL injury risk reduction and in compensating for the hamstring's reduced efficacy at low knee flexion angles (Morgan *et al.*, 2014). Figure 13 shows the schematic representation of knee joint compression due to the co-activation of the quadriceps and gastrocnemius. A small simulation study using just 3 males and 3 females supports the role of the gastrocnemius in protecting the ACL (Ali *et al.*, 2014) and a cadaver study of 9 knees did not show an increase in ACL strain during gastrocnemius activation (Dürselen *et al.*, 1995).



**Figure 13. Schematic representation of knee joint compression due to the co-activation of the quadriceps and gastrocnemius.**

Yet, Pflum *et al.* (2004) observed a small anterior tibial shear force and Adouni *et al.* (2016) noted an increase in ACL loading under gastrocnemius activation. However, only one sample size was used in each of the two studies (Pflum *et al.*, 2004, Adouni *et al.*, 2016). Six cadavers used in the study by Elias *et al.* (2003) also detected an increase in anterior tibial translation under gastrocnemius activation but the increase was not significant when the knee was flexed at low knee flexion angles of 20°. An *in vivo* study by Fleming *et al.* (2001b) also measured an increase in ACL strain with gastrocnemius activation. The authors suggested that the increased ACL strain was due to effects of gastrocnemius bulk on the tibia, pushing the tibia anteriorly upon activation (Fleming *et al.*, 2001b). Effects of such muscle bulk were not studied in the previous simulation studies (Ali *et al.*, 2014, Morgan *et al.*, 2014) and cadaver study (Dürselen *et al.*, 1995). However, Fleming *et al.* (2001b) studied the gastrocnemius activation at only one level. It is possible that sufficient gastrocnemius activation would result in enough joint compression to reduce the effects of the expanding gastrocnemius bulk, therefore reducing anterior tibial translation and ACL loading. Effects of varying levels of gastrocnemius activation on anterior tibial translation have yet to be studied.

Additionally, the gastrocnemius is attached to the back of the femur and the femur is pulled posteriorly during activation, resulting in an anterior tibial translation (Padua *et al.*, 2005). Figure 14 shows the schematic representation of the biomechanical musculoskeletal knee model. However, the gastrocnemius is around 3° from the tibia's longitudinal axis and thus applies a higher compressive load and lower posterior pull on the femur (Messier *et al.*, 2011). Posterior force of the gastrocnemius on the femur was

also found to be minimal in another study by Mokhtarzadeh *et al.* (2013). Thus, it is likely that this posterior pull on the femur only contribute minimally to ACL loading. Therefore, on comparison, the earlier mentioned effects of joint compression (Morgan *et al.*, 2014) and muscle bulk expansion (Fleming *et al.*, 2001b) during gastrocnemius activation would have greater influence on anterior tibial translation.

<Image removed due to copyright restrictions>

**Figure 14. Schematic drawing of the biomechanical musculoskeletal knee model (Modified from Messier *et al.*, 2011).**

Today, there are still many questions on the effects of foot landing and the role of the plantar flexor, gastrocnemius, on non-contact ACL injury risks. Furthermore, the potential effects of foot landing on reducing non-contact ACL injury risk reduction can also be observed from the effects of ankle restraints on knee positions, even though such restraints are not designed for ACL injury prevention. Foot landing in the sagittal plane seemed to have an influence on knee flexion angles (Cortes *et al.* 2007) and external foot rotation was correlated to increased knee valgus moments (Dempsey *et al.*, 2012). These foot landing positions can thus be directed to avoid knee joint positions associated with high ACL injury risks. It is also unclear if the gastrocnemius, another knee flexor, does play a role in supporting the hamstrings at low knee flexion angles. This is especially so when increased gastrocnemius forces were found to compensate for decreased hamstring forces during the landing phase of single-leg landings (Morgan *et al.*, 2014). Current

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studies on ACL injuries have not focused on the ankle and this can be a barrier to the better understanding of non-contact ACL injuries (Ali *et al.*, 2010). There is also an absence of studies investigating foot landing effects on non-contact ACL injury risks during single-leg landings. This was therefore the primary motivation of this PhD research.

## **2.6 Summary**

In summary, preventing ACL injuries is an important aspect to protect the careers and health of athletes and to reduce huge financial costs for societies. This is especially so when there are many uncertainties to the effectiveness of current preventive methodologies. It is also essential to find alternative preventive solutions as the number of ACL injuries continues to increase despite extensive efforts to prevent them. Foot landing techniques have been identified as having the potential to reduce non-contact ACL injury risks. To realize their potential, it is necessary to fully understand the influence of foot landing techniques on the biomechanics of the knee. Furthermore, the roles of the gastrocnemius and hamstring in non-contact ACL injury risk reduction, especially during the pre-landing phase of single-leg landings should also be explored. Many current studies have not necessarily considered internal risk factors that could predispose athletes to higher non-contact ACL injury risks together with biomechanical measures. Therefore, this also presents a window of opportunity to include internal risk factors in this new study.

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It is therefore opportune to propose a new study to investigate foot-landing techniques and muscle activation during single-leg landing to explore their potential for non-contact ACL injury risk reduction.

This thesis can be broken down into four studies. As there were limited data on the investigation of foot landing techniques in single-leg landings, the first two parts as detailed in Chapters 3 and 4, were proposed and conducted. The first study provided findings on the effects of foot rotation on knee valgus. The second study examined the influence of fore-foot, natural and flat-foot landing techniques on knee flexion angles and knee flexion range of motion. As foot rotation had minimal effect on knee valgus angles during initial contact, foot rotation effects on knee valgus angles were not further explored. The third part was studying the influence of ankle plantar flexion angles and internal risk factors on knee flexion angles; and the role of the gastrocnemius in non-contact ACL injury risks, as opposed to the hamstrings and quadriceps. These were reported in Chapter 5. Effects of varying levels of gastrocnemius activation on anterior tibial translation and the role of gastrocnemius activation on non-contact ACL injury risk reduction were studied as the final and fourth part of the study. This was described in Chapter 6.

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## Chapter 3 - Effects of Foot Rotation Positions on Knee Valgus

### 3.1 Introduction

Very few studies have been reported that had focused on the effects of foot rotation on ACL injuries (Dempsey *et al.*, 2012, Ishida *et al.* 2014, Tran *et al.*, 2016). Ishida *et al.* (2014) studied how foot rotation positions affect shank rotation with respect to the thigh, during knee valgus. Ishida *et al.* (2014) observed that the toe-out foot rotation was associated with the tibia being more externally rotated with respect to the femur. However, their methodology required participants to shift from a standing position to a knee valgus position and this might not fully represent dynamic landing conditions (Ishida *et al.* 2014). Tran *et al.* (2016) found hip and knee biomechanics differences for toe-in and toe-out positions, as compared to the toe-forward position during double-leg landing. Since single-leg and double-leg landings result in different knee kinematics and muscle activation levels (Pappas *et al.*, 2007), double-leg landing results may not be applicable for single-leg landings, in which more non-contact ACL injuries are found (Donnelly *et al.*, 2012). Furthermore, studies by Ishida *et al.* (2014) and Tran *et al.* (2016) had specific markings on the ground for participants to position their feet. While this provided better control of the foot landing positions, this might not reflect how participants naturally landed.

Dempsey *et al.* (2012) found that maximum knee valgus moment was correlated with toe-out foot position during single-leg landings. Their study used a ball-catching landing protocol that was closer to actual play situations. However, their study could not examine

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different foot rotation positions such as toe-forward and toe-in positions, and did not study the effects of foot rotation at the initial contact of landing, whereby ACL strain was the highest (Lamontagne *et al.*, 2008). This presents a need to study the effects of foot rotation on knee valgus in greater extent for single-leg landings. This would in turn provide better understanding for non-contact ACL injury prevention, especially during initial contact.

The purpose of this first part of the thesis was therefore to examine the effects of foot rotation positions (toe-in, toe-forward, toe-out) on knee valgus, both during initial contact and during landing. This work has been published (Teng *et al.*, 2017) and the paper can be found in Appendix X. As of Dempsey *et al.* (2012)'s correlation study result, it was hypothesized that the toe-out position would increase non-contact ACL injury risks, by increasing knee valgus angles (KVA) at initial contact and maximum knee valgus moments (KVM) during landing.

## **3.2 Methods**

### **3.2.1 Participants**

A power analysis was performed using G\*Power (v3.1.9.2, Heinrich-Heine University, Düsseldorf, Germany) based on the results (large effect size in the correlation between external foot rotation and maximum KVM) reported by Dempsey *et al.* (2012). Using a large effect size of 0.40 for a one-way repeated measures Analysis of Variance (Cohen, 1988) a sample size of 12 was required (Power  $1-\beta = .80$ ,  $\alpha = .05$ ). Fifteen participants

were initially recruited but data of 4 participants had to be discarded: 1 dropped out halfway during the study; the other 3 had incomplete data as markers were not adhering well due to their intense perspiration and impact from jumping. Thus, data of the remaining 11 male recreational basketball players (age: mean 23.6 (SD 1.6) years old; height: mean 1.73 (SD 0.07) m; body mass: mean 66.6 (SD 8.6) kg) were used in this study. Participants had no previous ACL injuries or lower extremity injuries in the previous 6 months prior to the study, and played basketball, minimally once a week for at least 2 years before the study. The study was approved by the Institutional Review Board of the Nanyang Technological University. All participants signed written consent before participating in the study.

### **3.2.2 Equipment**

The study by Hong *et al.* (2014) showed that maximum knee valgus moment, maximum anterior tibial translation and maximum knee flexion angle were significantly higher in the shod condition. Joint energetics and energy dissipation were also different between shod and barefoot conditions (Yeow *et al.*, 2011, Shultz *et al.*, 2012). Thus, shoes were used in the study. Participants wore the same type of basketball shoes (Nike Zoom Attero, Nike, Inc., Beaverton, OR, USA) to remove possible confounding effects of variation in footwear. Common shoe sizes of US 9, 10, 11 and 13 were used in the study and participants were pre-screened to ensure that they fell within the size range. Shoes were then snugly tied up to allow minimal relative movements between shoes and feet. When required, additional shoe padding was also provided. This was carried out to minimize relative motion between the shoes and feet.

Retroreflective markers were attached to the dominant leg, as declared by the participants as the leg that kicked a ball the hardest for the ball to travel the longest distance (Hewett *et al.*, 2005; see Figure 15). The markers were attached to the sacrum, anterior superior iliac spine, greater trochanter of femur, mid-thigh, medial and lateral knee epicondyles, tibial tuberosity, head of fibula, anterior aspect of shin, medial and lateral malleoli, distal calcaneus (CAL1), proximal calcaneus (CAL2), head and base of first metatarsal, head and base of fifth metatarsal and toe (between second and third metatarsal heads). The last seven markers were placed on the shoe and steps were taken to locate these positions. Previous studies had also placed markers directly on the shoes (Kovács *et al.*, 1999, Ford *et al.*, 2006, Orishimo *et al.*, 2009, Weinhandl *et al.*, 2010, Laughlin *et al.*, 2011, Dempsey *et al.*, 2012, Ali *et al.*, 2013, Tran *et al.*, 2016). The participants were first told to remove their shoes. CAL1, CAL2, head and base of first metatarsal, head and base of fifth metatarsal and toe positions were palpated without shoes being worn. The metatarsal and toe's palpated positions were held on while the participants were briefed about the placement of markers at these positions on the shoes. Immediately after, the shoes were worn, and testers located the metatarsal and toe positions with the feedback from participants, through the shoe tongue region.

For the CAL1 and CAL2 positions, the markers' proximal positions were estimated when the markers were placed on the shoe. However, the medial-lateral position of the CAL2 marker was aligned to the medial-lateral position of the Achilles tendon. The medial-lateral position of the Achilles tendon could be observed from above the most proximal aspect of the shoe (see Appendix A). Marker CAL1 was placed vertically below CAL2,

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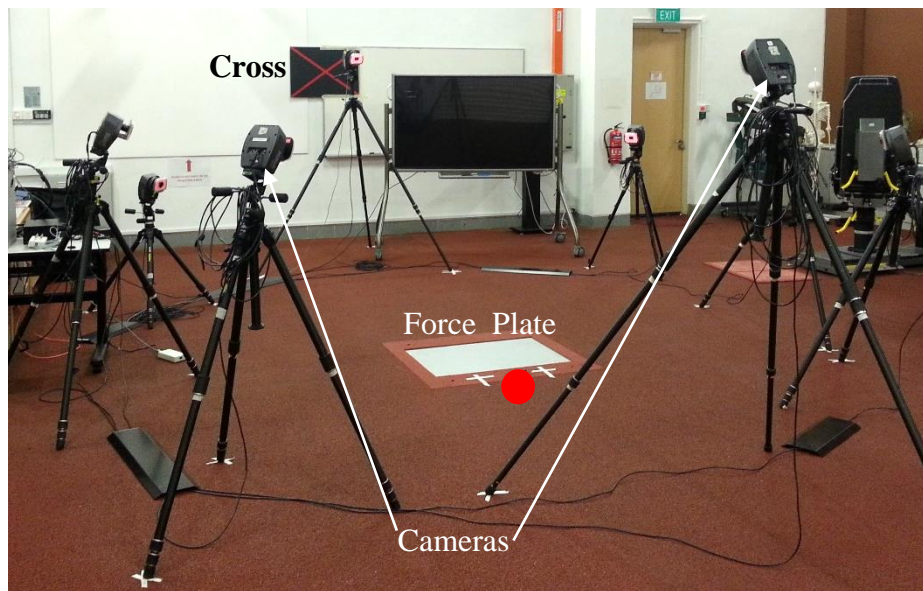
above the heel of the shoe (see Appendix A). Markers were also attached to the anterior superior iliac spine for the estimation of the hip joint location and the greater trochanter of the non-dominant leg. Markers were attached to these key body landmarks using tape (Leukoplast®, BSN medical GmbH, Hamburg, Germany).



Marker	Landmark
1	Sacrum
2	Right anterior superior iliac spine
3	Right greater trochanter of femur
4	Mid-thigh
5	Lateral knee epicondyle
6	Medial knee epicondyle
7	Head of fibula
8	Tibial tuberosity
9	Anterior aspect of shin
10	Lateral malleolus
11	Medial malleolus
12	Base of fifth metatarsal
13	Head of fifth metatarsal
14	Toe (between second and third metatarsal heads)
15	Base of first metatarsal
16	Head of first metatarsal
17	Proximal calcaneus (CAL2)
18	Distal calcaneus (CAL1)
19	Left anterior superior iliac spine
20	Left greater trochanter of femur

**Figure 15. Positions of marker placement.**

Three-dimensional motion data were captured at 200 Hz using 8 digital cameras and the Cortex software (version 1.1.4.368, Motion Analysis Corp, Santa Rosa, CA, USA). Ground reaction forces (GRF) were recorded using a force plate (Kistler type 9287CA, Winterthur, Switzerland) at a sampling rate of 1000 Hz in synchronization with the digital cameras. The entire set-up is shown in Figure 16 and details of how the equipment was set-up and calibrated are included in Appendix B. Kinematic and kinetic data were further processed using Visual3D (v5.00.33, C-Motion, Germantown, PA, USA).



**Figure 16. Laboratory set-up of the cameras, force plate and the board with the cross on the wall. The 30-cm platform was placed at the position, marked by the red circle.**

### 3.2.3 Test Protocol

Before the start of the study, participants each carried out a set of warm-up exercises for 5 min, targeting the lower extremity muscles, since drop-landing trials would be conducted. The exercises were stretching of the hamstrings, quadriceps and calves, tip-

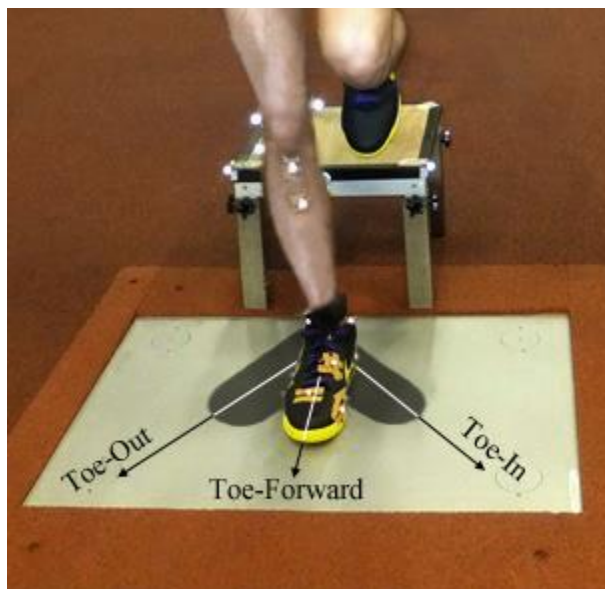
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toe and ankle rotation, squats and low lunges, jumps, jogging on the spot and jogging with high knees. After that, participants practised the drop-landing manoeuvres. Participants were requested to stand with their dominant leg on a 30 cm-height platform, and drop land onto a force plate using the same leg. The platform height was tested during pilot studies to allow the protocol to be carried out without over-exertion or compromise of safety and had been used in previous studies (Cortes *et al.*, 2007, Mokhtarzadeh *et al.*, 2013, Orishimo *et al.*, 2014, Tran *et al.*, 2016, Valenzuela *et al.*, 2016). Furthermore, the platform height was also close to the maximum jump height for males (mean 35.5 (SD 4.5) cm) and females (mean 25.6 (SD 3.5) cm) (Abián *et al.*, 2008).

Participants first stood with the toes facing forward on the platform, toward the front of the lab with a board marked with a red cross. Participants were then verbally instructed to land naturally, similar to the way they usually landed from a jump (“forward”); landing with toes moving to the mid-line of the body (“toe-in”); and toes facing away from the mid-line of the body (“toe-out”). (see Figure 17). No markings were placed on the force plate unlike previous studies (Ishida *et al.*, 2014, Tran *et al.*, 2016) to allow for natural landing as much as possible. Participants were also requested to face the board marked with the red cross, placed around the eye-level at the front of the lab, during landing (see Figure 16). They were also instructed to keep their bodies upright and hands crossed over their chests to reduce any variability from swinging arms (Decker *et al.*, 2003, Fujii *et al.*, 2012). Participants were also to maintain the vertical height across the different jump trials by not bending down or jumping up when they carried out the drop-landing trials

(Decker *et al.*, 2003, Shimokochi *et al.*, 2013). This was monitored after every trial using the z-axis position of the sacrum marker, a proxy for the centre of gravity. Upon landing, participants had to hold still using the dominant leg on the force plate for 2 s to obtain stable readings, especially for the measurement of the maximum KVM (Ford *et al.*, 2006, Shimokochi *et al.*, 2013).

A standing trial was initially captured. Following that, motion capture data of 3 successful trials of each condition were captured and the average of the 3 trials was recorded. The sequence of conditions was randomized for each participant to minimize the influence due to fatigue, learning effects or boredom.



**Figure 17. Three foot rotation positions: toe-forward, toe-in (towards the midline of the body) and toe-out (away from the mid-line of the body).**

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### 3.2.4 Kinematic and Kinetic Analyses

Visual3D (v5.00.33, C-Motion, Inc., Germantown, MD, USA) was used to build the single leg model consisting of the pelvis, thigh, shank and foot segment (see Appendix C). The motion and VGRF data were filtered through a low-pass, 4th order, zero phase shift Butterworth digital filter (Dempsey *et al.*, 2012, Valenzuela *et al.*, 2016) at a cut off frequency of 9 Hz and 50 Hz, respectively (see Appendix D). Initial contact was defined as the first instance that vertical GRF exceeded 10 N (Self *et al.*, 2001, Ford *et al.*, 2006, Fujii *et al.*, 2012). For each landing trial, the two dependent variables were: (1) KVA at initial contact and (2) maximum KVM during the whole phase of landing. The whole landing phase was defined as twice the time taken from initial contact with the ground to the time of maximum vertical GRF following the protocol by Dempsey *et al.* (2012). Zero percent was the point of initial contact and 100% was the end of the landing phase. Data was normalized to 100% using the spline interpolation (see Appendix E). Maximum KVM was normalized to the product of each participant's height (m) and mass (kg) (Dempsey *et al.*, 2012). Normalization is a common practice to remove between-subject differences so that comparison of data among subjects and across different conditions can be made (Selbie *et al.*, 2014).

To verify that participants landed in three distinct foot rotation positions as instructed, foot rotation angles at initial contact were measured with respect to the shank and the pelvis.

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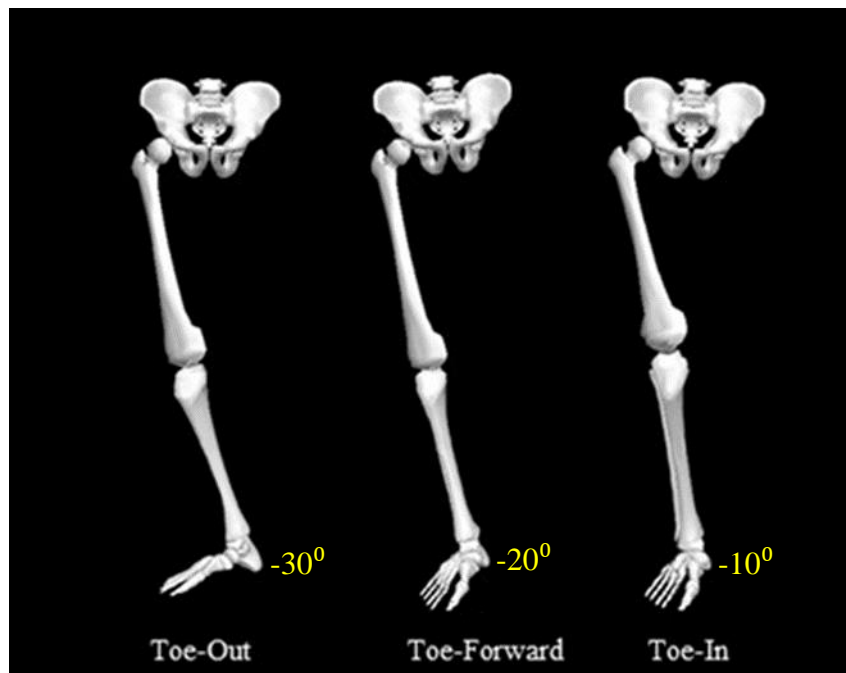
Foot rotation angle with respect to the shank was defined as the orientation of the foot segment relative to the shank segment in the transverse plane. Foot rotation angle with respect to the pelvis was defined as the orientation of the foot segment relative to the pelvis segment in the transverse plane. Knee valgus angle was defined as the external rotation of the shank segment, with respect to the thigh segment in the frontal plane. Knee valgus moment was defined as the external moment of the shank segment, with respect to the thigh segment in the frontal plane. Positive denotes toe-in or knee varus directions. Details of how these measurements were derived are included in Appendix C.

### **3.2.5 Statistical Analysis**

Statistical analysis was carried out using the IBM SPSS 22.0 (IBM Corp., NY, USA). To compare among the foot rotation positions, a one-way repeated measures Analysis of Variance was conducted at  $\alpha = 0.05$ . Tests of assumptions were carried out based on standard statistical protocol (Lund *et al.*, 2015). Normality assumption was tested with the Shapiro-Wilk test ( $p > 0.05$ ). If normality could not be assumed, "reflect and logarithmic" transformation was applied. Assumption of sphericity was tested using the Mauchly's test of sphericity. When the assumption of sphericity was not met, a Greenhouse-Geisser correction was applied. Post-hoc test with Bonferroni adjustment was made with a family-wise  $\alpha = 0.05$ . Effect size was calculated as  $\eta^2$  and values of 0.01, 0.09 and 0.25 were defined as small, medium and large effects, respectively (Hanna *et al.*, 2012). Data are expressed as mean (standard deviation). Definition of normality, sphericity and effect size are found in Appendix F.

### 3.3 Results

Foot rotation angles, when measured with respect to the shank, were not significantly different among the three foot rotation positions ( $p = 0.810$ , see Table 3). When measured with respect to the pelvis orientation, significant differences were found (see Table 3) and participants landed in toe-out positions for all three positions of foot rotation (see Figure 18). However, results showed that toe-in position was significantly less negative than toe-forward position ( $p = 0.004$ ). Similarly, toe-forward position was significantly less negative than the toe-out position ( $p < 0.001$ ). Participants therefore landed in the three distinct foot rotation positions as instructed during the study.



**Figure 18. Actual foot rotation positions that participants landed in: toe-forward, toe-in and toe-out (Image created using Visual3D). Negative angles refer to toe-out foot rotation positions.**

**Table 3. Foot and Knee Biomechanics of Toe-in, Toe-forward and Toe-Out Landings (mean (standard deviation)) (n = 11)**

Variables <sup>1</sup>	Toe-in	Toe-forward	Toe-out	p-value	$\eta^2$
<b>Foot rotation angles<sup>1</sup></b>					
<b>With respect to shank (°)</b>	6(3)	6(3)	6(2)	0.810	0.02
<b>With respect to pelvis (°)</b>	-10(7) <sup>b,c</sup>	-20(5) <sup>a,c</sup>	-30(6) <sup>a,b</sup>	< 0.001	0.74
<b>KVA at initial contact (°)</b>	-2(3) <sup>b,c</sup>	-1(3) <sup>a</sup>	-1(3) <sup>a</sup>	< 0.001	0.66
<b>Maximum KVM (N·m/kg/m)</b>	-0.60(0.29) <sup>c</sup>	-0.67(0.26) <sup>c</sup>	-0.82(0.28) <sup>a,b</sup>	< 0.001	0.55

<sup>1</sup> Negative denotes toe-out or knee valgus directions.

<sup>a</sup> Significantly different from toe-in position.

<sup>b</sup> Significantly different from toe-forward position.

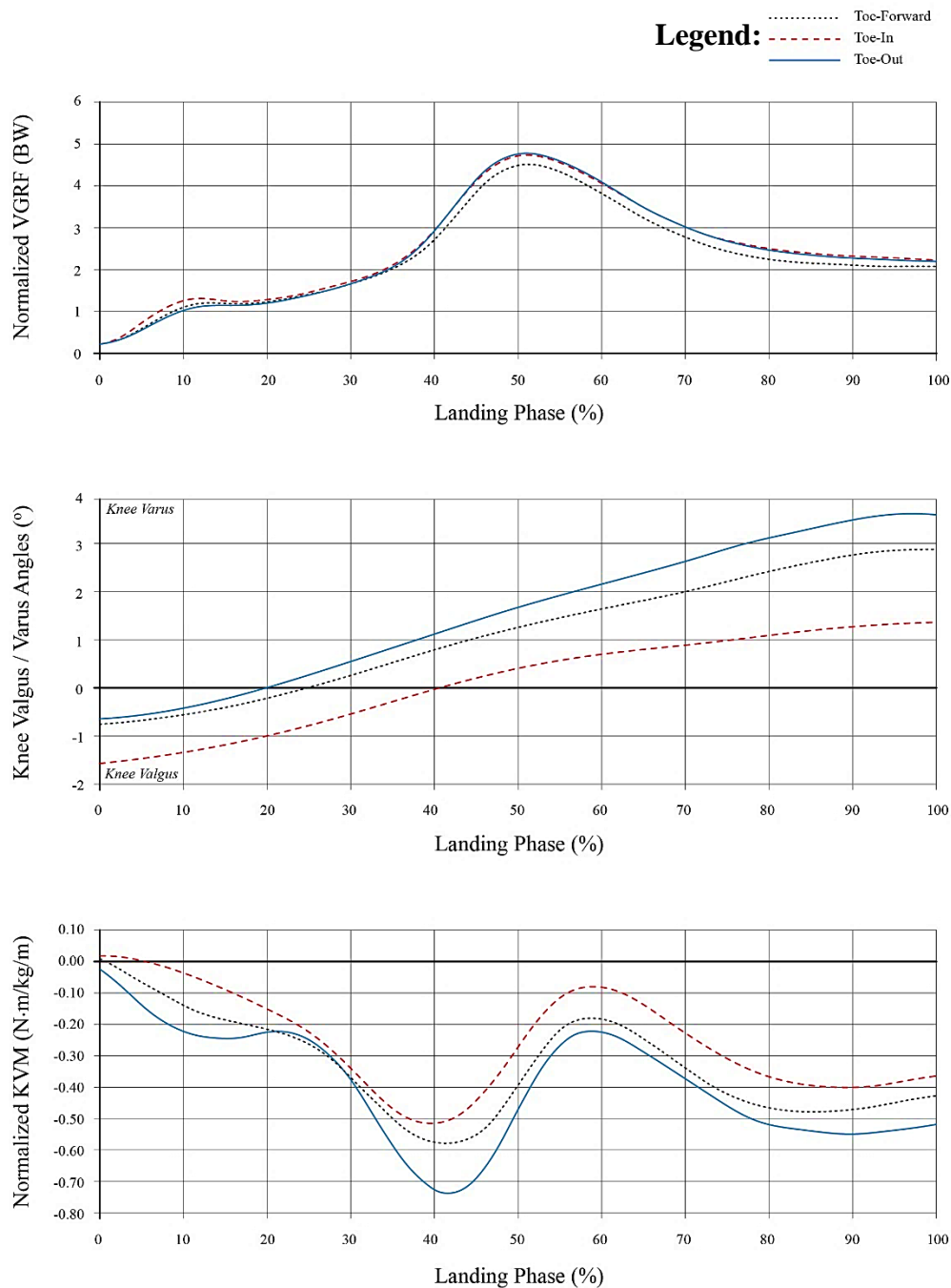
<sup>c</sup> Significantly different from toe-out position.

Figure 19 shows the mean ensemble curves of the vertical ground reaction force (VGRF) (normalized to body weight), KVA and KVM (normalized to the product of height and body mass) over the landing phase, under the three foot rotation positions. Participants landed with knee valgus positions at initial contact under all foot rotation conditions. However, there was a steady decrease of knee valgus angles over the landing phase. Maximum KVM occurred around 40% of the landing phase, just before maximum VGRF took place.

There was a significant effect of foot position on KVA at initial contact ( $p < .001$ , see Table 3). However, the actual difference in angles was small (toe-in vs. toe-forward: 1°, 95% CI [0 to 1],  $p = 0.001$ ; toe-in vs. toe-out: 1°, 95% CI [0 to 2],  $p = 0.003$ ). Maximum

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KVM was significantly different among the three foot rotation positions ( $p < 0.001$ , see Table 3). Post-hoc tests showed greater KVM in the toe-out position than toe-forward ( $p = 0.012$ ) or toe-in conditions ( $p = 0.002$ ). No post-hoc difference in KVM was found between the toe-in and toe-forward positions ( $p = 0.442$ ).



**Figure 19. Mean ensemble curves of vertical ground reaction force (VGRF), normalized to body weight, knee valgus/varus angles and knee valgus moment (KVM), normalized to height and body mass (n = 11). The whole landing phase was defined as twice the time taken from initial contact with the ground (0%) to the time of maximum vertical ground reaction force (100%).**

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## 3.4 Discussion

The aim of this study was to examine the effects of foot rotation positions (toe-in, toe-forward, toe-out) on knee valgus during single-leg drop-landing. This is the first study to experimentally vary various foot rotation positions to better understand how ACL injury risks could be affected by foot rotation during single-leg landings. As hypothesized, toe-out foot rotation position resulted in a significant increase in maximum KVM during landing. Foot rotation also had a significant effect on KVA at initial contact but the difference of the KVA was too small to be practically significant or having “meaningful, practical impact” (Peeters, 2016, p. 84).

### 3.4.1 Effects of Foot Rotation on KVM

This study showed that toe-out foot rotation position resulted in significantly higher maximum KVM than in toe-in and toe-forward positions. This collaborated well with the results of Dempsey *et al.* (2012), which reported a significant correlation between increased KVM and external foot rotation. Since higher KVM is associated with increased ACL injury risks (Hewett *et al.*, 2005, Kimura *et al.*, 2012, Lin *et al.*, 2012), athletes should therefore be instructed to avoid landing in extreme toe-out positions during single-leg landing to minimize ACL injury risks. Since differences in KVM between the toe-in and toe-forward positions were minimal, landing within this range of foot rotation would likely result in similar maximum KVM. These findings suggest that there is no negative effect on knee valgus loading when landing in a toe-in, as compared

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to toe-forward positions. Thus, it may not be essential to stress the ‘toe-forward’ position as usually advised during training (Noyes *et al.*, 2005).

During single-leg landings, the knees flexion angles are typically less than 20° at initial contact (Weinhandl *et al.* 2010). Within this low knee flexion angle range, the “screw-home” mechanism can occur whereby the tibia externally rotates on the femur (Kim *et al.*, 2015, p. 303). Furthermore, the study by Ishida *et al.* (2014) observed that the toe-out foot rotation was associated with the tibia being more externally rotated with respect to the femur. The medial collateral ligament (MCL) helps to hold the knee joint in place by preventing it from going into knee valgus positions and by restraining external tibial rotation (Clippinger *et al.*, 2016). In the toe-out foot rotation position with more external tibial rotation, the MCL could have been twisted (Escamilla *et al.*, 2000) such that it could buckle more easily to constrain less of the KVM. This might indirectly result in greater loading of the ACL, which has the secondary role of restraining knee valgus/varus loading (Arnheim, 1993). Thus, landing in a toe-out position may increase ACL injury risks during single-leg landing.

The maximum KVM values (-0.82 to -0.60 N·m/kg/m) obtained in this study were different from those reported in previous studies on single-leg landing using male subjects (Dempsey *et al.*, 2012, Hong *et al.*, 2014). Dempsey *et al.* (2012) obtained lower KVM mean values ranging from -0.43 (95% CI [-0.53, -0.33]) to -0.23 (95% CI [-0.30, -0.16]) N·m/kg/m, for different landing tasks. Participants had to take off, catch a ball, and land with their preferred leg (Dempsey *et al.*, 2012). Their landing heights might be

different from the fixed 30 cm in this study, resulting in a lower range of KVM. On the other hand, Hong *et al.* (2014) used a similar protocol of single-leg drop landing from a 30-cm platform and obtained a mean maximum KVM of -0.16 N·m/kg/m, 95% CI [-0.20, -0.13] when participants wore shoes. Their different marker set, without the use of the sacrum and anterior superior iliac spine markers, might have resulted in different maximum KVM values (Hong *et al.*, 2014). However, as this is a within-subject study, the conclusion obtained in this study would still apply despite having dissimilar KVM values as other studies.

### **3.4.2 Effects of Foot Rotation on KVA**

This study also studied the effects of foot rotation on KVA at initial contact whereby ACL strain was the highest for single-leg landings (Lamontagne *et al.*, 2008). There were significant but very small differences ( $-1^{\circ}$  to  $-2^{\circ}$ ) among the 3 positions of foot rotation. Similar values of KVA were reported in the video analysis of basketball games during initial contact ( $-3^{\circ}$ , 95% CI [-2 to -4], Krosshaug *et al.*, 2007) and during single-leg landing ( $0^{\circ}$ , 95% CI [-2 to 2]), Orishimo *et al.* 2009;  $1^{\circ}$ , 95% CI [-1 to 2]), Weinhandl *et al.* 2010). The large effect size ( $\eta^2 = 0.66$ ) suggests that 66% of the variance in KVA is explained by foot rotation angles. However, the actual differences in knee valgus angles are too small to be practically significant. Therefore, foot rotation does not considerably affect the risk of ACL injuries at initial contact of single-leg landings.

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### 3.4.3 Foot Rotation Protocol

In the verification of foot rotation angles, foot rotations with respect to the shank were not significantly different among the three positions. This suggests that the foot and shank have likely rotated together as one unit. Participants had possibly attained the toe-in/out positions via internal/external rotations at the knee and hip joints. In fact, Ishida *et al.* (2014) found that the toe-out foot rotation position resulted in a more externally rotated tibia with respect to the femur and the converse was observed for the toe-in foot rotation position.

By contrast and as shown in Table 3, the foot rotation angles with respect to pelvis became increasingly more negative (away from the midline of the body) from toe-in to toe-forward and then toe-out positions. This implies that despite the request for participants to land with various foot rotation conditions (see Figure 17), participants all landed in negative, toe-out positions (see Figure 18). This observation suggests that toe-out foot rotation could be a natural landing position amongst these participants. This was in agreement with the study by Dempsey *et al.* (2012), which also found most participants landing in toe-out positions. More importantly, the purpose of measuring foot rotation angles was to verify if participants landed in three distinct foot rotation groups as instructed. This was verified as when measured with respect to the pelvis, three distinct foot rotation groups were identified, in the correct order as instructed.

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### **3.4.4 Limitations**

A few limitations to this study have been identified. First, only the dominant leg was used in this study. Future studies could also investigate the non-dominant leg since both dominant and non-dominant legs may be similarly exposed to ACL injury risks (Hewett *et al.*, 2010). Studying the non-dominant leg could determine if adopting the toe-in and toe-forward foot rotation positions was still applicable in reducing ACL injury risks. Secondly, only foot rotation angles with respect to knee valgus were studied. Variation of sagittal foot landing positions should also be studied to provide a more complete understanding of foot-landing effects on non-contact ACL injury risk prevention. This study was conducted and results are presented in the next chapter. Thirdly, this study was conducted in a controlled laboratory environment. Effects of the desirable foot rotation (toe-in and toe-forward) positions on reducing maximum KVM might be less distinct during sport. A prospective intervention study could be carried out on a cohort, together with a control group, to study the training effects of foot-landing techniques on reducing non-contact ACL injuries over time.

### **3.5 Conclusions**

The study investigated the effects of foot rotation positions (toe-in, toe-forward, toe-out) on knee valgus, both during initial contact and during landing. A laboratory-controlled motion analysis study was conducted using 11 participants. This study showed that foot rotation positions can influence knee valgus during single-leg drop landing. Particularly, landing in toe-out foot positions resulted in significantly higher maximum KVM, as

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compared to toe-in and toe-forward positions. Although foot rotation positions have an effect on KVA at initial contact, the differences were too small to be practically significant. From the results of this study, during single-leg landings, athletes are advised not to go into extreme toe-out foot rotation positions. This would minimise undesirable knee valgus loading associated with ACL injury risks.

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## Chapter 4 - Influence of Fore-foot, Natural and Flat-Foot Landing Techniques on Non-Contact ACL Injury Risks

### 4.1 Introduction

In Chapter 3, a study was conducted to provide better understanding of various foot rotation positions on knee valgus both at initial contact and during the landing phase. To have a more complete understanding of foot landing effects on non-contact ACL injury risks, foot landing techniques at the sagittal plane should also be considered. In fact, Yu and his colleagues (2007) suggested that knee valgus-varus moments would only result in non-contact ACL injuries when there were large sagittal plane forces. In turn, large sagittal forces were caused by reasons including the increase in knee extension moments, as a result of reduced knee flexion angles (Yu *et al.*, 2007). Low knee flexion angles are often associated with higher non-contact ACL injury risks (Lin *et al.*, 2008) as players commonly land at low knee flexion angles of less than 30° before injury (Krosshaug *et al.*, 2007). Thus, knee flexion angles have previously been used to study ACL injury risks under different test conditions (Pappas *et al.*, 2007, Lin *et al.*, 2008, Stanley *et al.*, 2011).

Cortes *et al.* (2007) studied the effects of foot-landing positions in the sagittal plane (fore-foot, rear-foot and natural) on lower extremity kinematics. They found that knee flexion angle was significantly lower at initial contact for fore-foot landing, as compared to natural and rear-foot landing (Cortes *et al.*, 2007). It was also found that knee flexion angle was significantly lower at maximum vertical ground reaction force (VGRF) for rear-foot landing, as compared to natural and fore-foot landing (Cortes *et al.*, 2007).

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Their study provided an insight that variation of foot landing positions in the sagittal plane could affect knee flexion angles but their study was only on double-leg landing (Cortes *et al.* 2007).

There were only a few other studies reported which specifically varied foot landing positions in the sagittal plane during dynamic tasks to assess its effects on the lower extremity (Kovács *et al.*, 1999, Valenzuela *et al.*, 2016). However, they were also double-leg landing studies (Kovács *et al.*, 1999, Valenzuela *et al.*, 2016). Kovács *et al.* (1999) studied how lower extremity kinematics, kinetics and muscle activation changed due to fore-foot and rear-foot landings. Foot landing positions were found to change the different joint contributions to the total power during drop jumping (Kovács *et al.*, 1999). The study by Valenzuela *et al.* (2016) examined how knee and hip kinematics changed with landing on various landing surfaces, including one of combined plantarflexion and inversion. Plantarflexion was also only fixed at one position (Valenzuela *et al.*, 2016). The combined surface resulted in reduced knee and hip flexion range of motion and increased knee abduction range of motion (Valenzuela *et al.*, 2016). Single-leg and double-leg landings result in different knee kinematics and muscle activation levels (Pappas *et al.*, 2007). Therefore, double-leg landing results are unlikely to reflect accurate knee kinematics and muscle activation levels of single-leg landings, whereby more non-contact ACL injuries occur (Donnelly *et al.*, 2012). Thus, there exists a gap in current literature of similar studies in single-leg landings.

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Therefore, this study aimed to close this gap by studying the influence of fore-foot, natural and flat-foot landing techniques on non-contact ACL injury risks during single-leg drop landings. Knee flexion angles both at initial contact and maximum VGRF, as well as, knee flexion range of motion were studied under the different landing conditions. Similar to the results in double-leg landing by Cortes *et al.* (2007), it was hypothesized that fore-foot and natural landing would result in significantly lower knee flexion angles at initial contact but higher knee flexion angles at maximum VGRF and higher knee flexion range of motion.

## **4.2 Methods**

### **4.2.1 Participants**

A power analysis was performed using G\*Power (v3.1.9.2, Heinrich-Heine University, Düsseldorf, Germany). There were no prior studies studying the influence of fore-foot, natural and flat-foot landing techniques on non-contact ACL injury risks during single-leg drop landings. Therefore, a large effect size was used based on trial study results and based on a medium-to-large effect size  $d$ , that was calculated using the double-leg landing results by Cortes *et al.* (2007). Using a large effect size of 0.40 for a one-way repeated measures Analysis of Variance (Cohen, 1988) a sample size of 12 was required (Power  $1-\beta = .80$ ,  $\alpha = .05$ ). Fifteen participants were initially recruited but data of a participant had to be discarded as he dropped out of the study. Fourteen male recreational basketball players (age: mean 23.4 (SD 1.7) years, height: mean 1.74 (SD 0.07) m, body mass: mean 67.8 (SD 10.4) kg) had agreed to take part in this study. Participants had no

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previous ACL injuries or lower extremity injuries in the previous 6 months prior to the study, and played basketball, minimally once a week for at least 2 years before the study. The study was approved by the Nanyang Technological University Institutional Review Board. All participants signed written consent form before participating in the study.

### **4.2.2 Equipment**

The same equipment and marker positions as Chapter 3 were set up in this study. The location of marker positions on the shoes were carried out in the same way as described in Chapter 3. Participants wore the same type of basketball shoes (Nike Zoom Attero, Nike, Inc., Beaverton, OR, USA). Three-dimensional motion data were captured at 200 Hz using 8 digital cameras and the Cortex software (version 1.1.4.368, Motion Analysis Corp, Santa Rosa, CA, USA). Vertical ground reaction force (VGRF) was recorded using a force plate (Kistler type 9287CA, Winterthur, Switzerland) at a sampling rate of 1000 Hz in synchronization with the cameras. Details of how the equipment was set-up and calibrated are included in Appendix B. Kinematic and kinetic data were further processed using Visual3D (v5.00.33, C-Motion, Germantown, PA, USA).

### **4.2.3 Test Protocol**

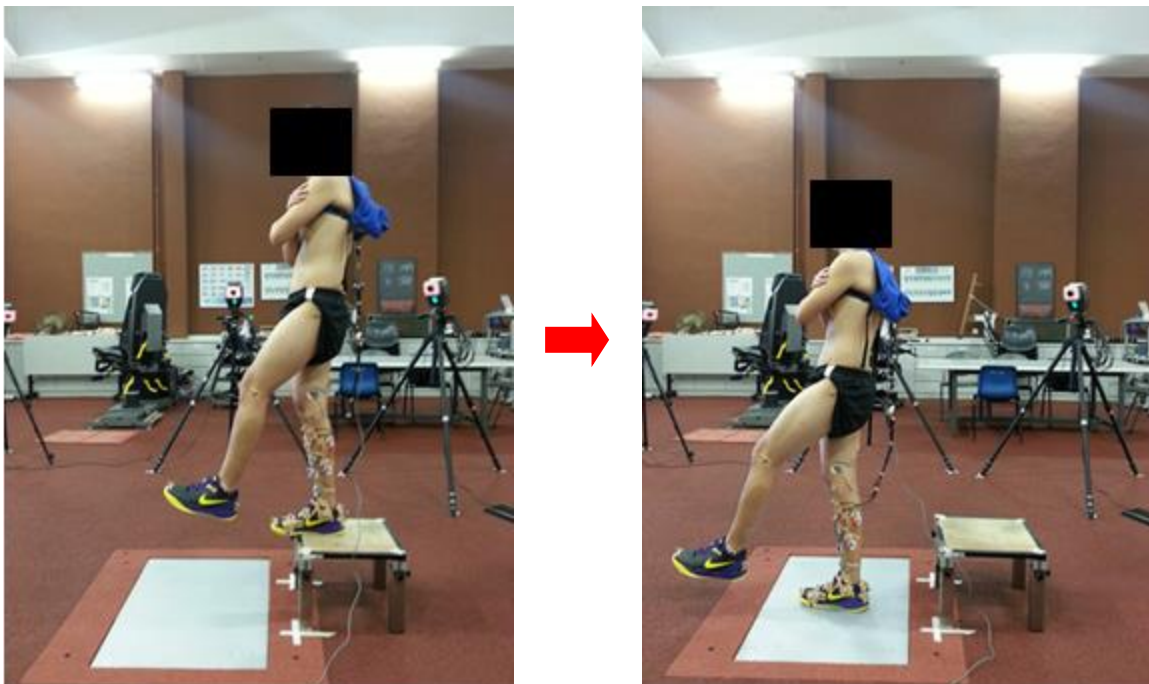
Before the start of the study, participants each carried out a set of warm-up exercises for 5 min, targeting the lower extremity muscles, since drop-landing trials would be conducted. The exercises were stretching of the hamstrings, quadriceps and calves, tip-toe and ankle rotation, squats and low lunges, jumps, jogging on the spot and jogging

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with high knees. After that, participants practised the drop-landing manoeuvres. Participants were requested to stand with their dominant leg on a 30 cm-height platform, and drop land onto a force plate using the same leg (see Figure 20). The platform height was tested during pilot studies to allow the protocol to be carried out without over-exertion or compromise of safety and had been used in previous studies (Cortes *et al.*, 2007, Mokhtarzadeh *et al.*, 2013, Orishimo *et al.*, 2014, Tran *et al.*, 2016, Valenzuela *et al.*, 2016). Furthermore, the platform height was also close to the maximum jump height for males (mean 35.5 (SD 4.5) cm) and females (mean 25.6 (SD 3.5) cm) (Abián *et al.*, 2008).

The protocol was conducted for natural, fore-foot ( $>25^\circ$  ankle flexion angles) and flat-foot ( $<5^\circ$  ankle flexion angles) landings at initial contact, with plantar flexion angles denoting as positive. There was no control over how participants should land during ‘natural’ landing, which was defined as how participants usually landed from a jump. These ankle flexion angles were pre-determined during the pilot study to be representative of each landing position – natural, fore-foot and flat-foot. Measures were taken to ensure that participants landed in the required stipulated positions using the Cortex software (version 1.1.4.368, Motion Analysis Corp, Santa Rosa, CA, USA) and Visual3D (v5.00.33, C-Motion, Germantown, PA, USA) (see Appendix G). Participants were also requested to face a board marked with a red cross, placed around the eye-level at the front of the lab, during landing. They were also instructed to keep their bodies upright and hands crossed over their chests to reduce any variability from swinging arms (Decker *et al.*, 2003, Fujii *et al.*, 2012, see Figure 20).

A standing trial was initially captured. Following that, motion capture data of 3 successful trials of each condition were captured and the average of the 3 trials was recorded. The sequence of conditions was randomized for each participant to minimize the influence due to fatigue, learning effects or boredom.



**Figure 20. Participant drop-landing from the 30-cm platform onto the force plate.**

#### **4.2.4 Kinematics Analyses**

Knee flexion-extension and ankle plantar-dorsi flexion angles were recorded for each participant at initial contact and during maximum VGRF. Knee flexion angle was defined as the orientation of the shank segment, with respect to the thigh segment in the sagittal plane. Ankle flexion angle was defined as the orientation of the foot segment relative to the shank segment in the sagittal plane. Force plate data was collected to detect the initial

contact with the ground, defined as the first point that VGRF exceeded 10 N and maximum VGRF. The motion and VGRF data were filtered through a low-pass, 4th order, zero phase shift Butterworth digital filter (Dempsey *et al.*, 2012, Valenzuela *et al.*, 2016), at a cut off frequency of 9 Hz and 50 Hz, respectively (see Appendix D). Knee flexion and ankle plantar flexion angles were denoted as positive. Cardan rotation sequence xyz was used for the knee joint angle computation and the single-leg model was built in Visual3D (v5.00.33, C-Motion, Germantown, PA, USA). Ankle plantar flexion angle measured during the static standing trial was used as each participant's neutral (zero) alignment. Ankle plantar flexion angle during the drop-landing was measured and referenced to the standing position (Ford *et al.*, 2006). Knee flexion range of motion was computed by subtracting the knee flexion angles at initial contact from the knee flexion angles at maximum VGRF. Maximum VGRF was normalized to body weight (BW). Details of how these measurements were derived are included in Appendix C.

#### **4.2.5 Statistical Analysis**

Statistical analysis, similar to that described in Section 3.2.5, was carried out. To compare among the foot landing positions in the sagittal plane, a one-way repeated measures Analysis of Variance was conducted at  $\alpha = 0.05$  (IBM SPSS 22.0, IBM Corp., NY, USA). Normality assumption was tested with the Shapiro-Wilk test ( $p > 0.05$ ) and normality assumption was not violated in this study. Assumption of sphericity was tested using the Mauchly's test of sphericity. When the assumption of sphericity was not met, a Greenhouse-Geisser correction was applied. Post-hoc test with Bonferroni adjustment was made with a family-wise  $\alpha = 0.05$ . Effect size was calculated as  $\eta^2$  and values of

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0.01, 0.09 and 0.25 were defined as small, medium and large effects, respectively (Hanna *et al.*, 2012). Data are expressed as mean (standard deviation). Definition of normality, sphericity and effect size are found in Appendix F.

### **4.3 Results**

Foot-landing technique in the sagittal plane has a significant effect on knee flexion angles at initial contact ( $p < 0.001$ ; see Table 4) and at maximum VGFR ( $p < 0.001$ ; see Table 4). At initial contact, natural and fore-foot landings resulted in significantly lower knee flexion angles, as compared to flat-foot landings ( $p = 0.002$  and  $p < 0.001$ , respectively). The representative postures of fore-foot and flat-foot landings at initial contact with the ground are shown in Figure 21.

**Table 4. Kinematic Results of Sagittal Foot Landing Techniques (mean (standard deviation)) (n = 14)**

Variables <sup>1</sup>	Natural	Fore-foot	Flat-foot	p-value	$\eta^2$
<b>Initial contact</b>					
<b>Ankle flexion angles (°)</b>	29(8) <sup>b,c</sup>	33(4) <sup>a,c</sup>	-3(4) <sup>a,b</sup>	< 0.001	0.95
<b>Knee flexion angles (°)</b>	13(5) <sup>c</sup>	13(4) <sup>c</sup>	18(4) <sup>a,b</sup>	< 0.001	0.60
<b>Maximum VGRF *</b>					
<b>Maximum VGRF/BW *</b>	4.5(0.4) <sup>c</sup>	4.4(0.4) <sup>c</sup>	5.8(0.5) <sup>a,b</sup>	< 0.001	0.85
<b>Knee flexion angles (°)</b>	31(5) <sup>c</sup>	33(6) <sup>c</sup>	21(5) <sup>a,b</sup>	< 0.001	0.85
<b>Knee Flexion Range of Motion (°) #</b>	18(4) <sup>b,c</sup>	20(4) <sup>a,c</sup>	3(1) <sup>a,b</sup>	< 0.001	0.93

<sup>1</sup> Positive denotes plantar flexion or knee flexion angles.

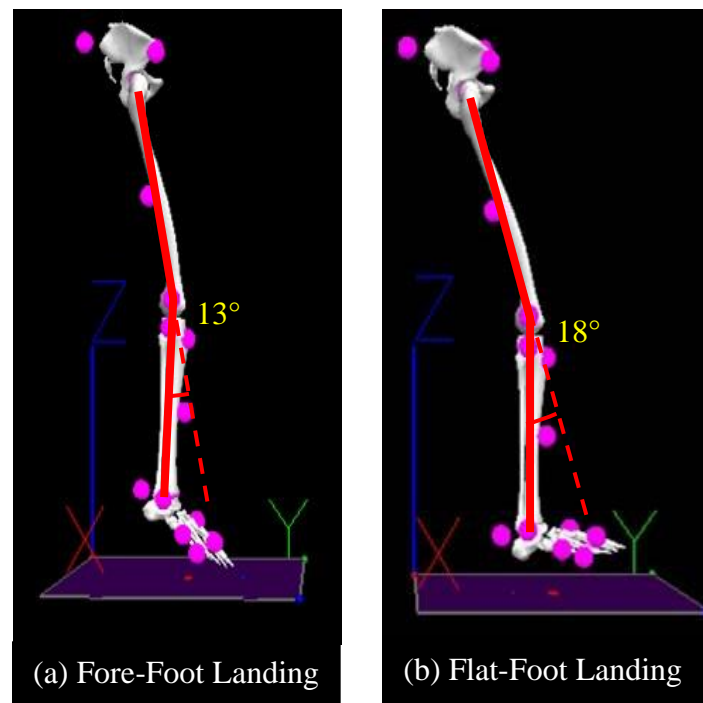
<sup>a</sup> Significantly different from natural landing.

<sup>b</sup> Significantly different from fore-foot landing.

<sup>c</sup> Significantly different from flat-foot landing.

\*VGRF is defined as vertical ground reaction force and BW is defined as body weight.

# Knee flexion range of motion was calculated by subtracting the knee flexion angles at initial contact from the knee flexion angles at maximum VGRF.



**Figure 21. Knee flexion angles at initial contact with the ground for (a) fore-foot landings and (b) flat-foot landings (Image created using Visual3D).**

On the other hand, at maximum VGRF, natural and fore-foot landings resulted in significantly higher knee flexion angles as compared to flat-foot landings at maximum VGRF (both at  $p < 0.001$ ). Natural and fore-foot landings do not result in significantly different knee flexion angles at initial contact ( $p = 1.000$ ) or at maximum VGRF ( $p = 0.102$ ).

Foot-landing technique in the sagittal plane also has a significant effect on knee flexion range of motion ( $p < 0.001$ ; see Table 4). There was significantly higher knee flexion range of motion in natural and fore-foot landings, as compared to flat-foot landings (both

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at  $p < 0.001$ ). There was also significantly higher knee flexion range of motion in fore-foot landing than in natural landing ( $p = 0.018$ ).

Finally, flat-foot landing resulted in significantly higher maximum VGRF, normalized to BW than fore-foot and natural landings ( $p < 0.001$ ). Ankle plantar flexion angles were observed during natural landings, suggesting fore-foot landings (see Table 4).

## **4.4 Discussion**

The aim of this study was to investigate the effects of sagittal foot landing techniques (fore-foot, natural and flat-foot) on knee flexion angles and knee flexion range of motion during single-leg drop-landing. This is the first study to dynamically vary sagittal foot landing positions to study its effects on ACL injury risks during single-leg landings. As hypothesized, fore-foot and natural landing positions resulted in a significant reduction in knee flexion angles at initial contact and a significant increase during maximum VGRF. However, fore-foot landing resulted in significant increase in knee flexion range of motion, as compared to both natural and flat-foot landing.

### **4.4.1 Influence of Natural and Fore-Foot Landings on Non-Contact ACL Injury Risks**

Natural and fore-foot landings resulted in a significant reduction in knee flexion angles at initial contact (see Table 4). The mean knee flexion angles of  $13^\circ$  are also lower than that found in double leg landings in males (Cortes *et al.* 2007, see Table 5). Low knee flexion

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angles were also found during single-leg landings in the literature (Weinhandl *et al.*, 2010, Orishimo *et al.*, 2014, Kim *et al.*, 2016, see Table 5). Low knee flexion angles of less than 30°, especially less than 15°, are often associated with higher ACL injury risks when hamstring activation becomes less effective in reduce ACL loading (Li *et al.*, 1999, Yu *et al.*, 2005). Furthermore, ACL strain was also found to be at the maximum at initial contact for single-leg landings (Lamontagne *et al.*, 2008) when these low knee flexion angles occur. This may be one of the reasons for more non-contact ACL injuries to be found in single-leg, rather than double leg landings (Donnelly *et al.*, 2012). Fore-foot landings have been encouraged during landing to allow for greater force absorption, putting less emphasis on effects on the knee at initial contact (Alentorn-Geli *et al.*, 2009a). However, such a landing technique may ironically be a contributory factor to greater non-contact ACL injury risks during single-leg landings. This is an issue in the current studies on ACL injuries that needs to be further addressed.

**Table 5. Comparisons of Knee Flexion Angles at Initial Contact among Studies**

Study	Knee Flexion Angles at Initial Contact	Comments
<b>This Study</b>	13°, 95% CI [10 to 16]	Natural single-leg landing (Males)
	13°, 95% CI [11 to 15]	Fore-foot single-leg landing (Males)
<b>Cortes <i>et al.</i> (2007)</b>	25°, 95% CI [20 to 29]	Natural double-leg landing (Males)
	18°, 95% CI [13 to 22]	Fore-foot double-leg landing (Males)
<b>Weinhandl <i>et al.</i> (2010)</b>	13°, 95% CI [10 to 16]	Single-leg landing (Males)
<b>Orishimo <i>et al.</i> (2014)</b>	12°, 95% CI [9 to 14]	Single-leg landing (Male athletes)
<b>Kim <i>et al.</i> (2016)</b>	14°, 95% CI [11 to 16]	Single-leg landing (Stable knee side)

When compared to flat-foot landings, natural and fore-foot landing resulted in significantly lower knee flexion angles at initial contact. This was also observed in double leg landing in the study by Cortes *et al.* (2007) who speculated that postural control demands were the cause of it. However, a more in-depth study, possibly looking at the muscle activation patterns should be carried out to have a better understanding of why low knee flexion angles occur and how higher knee flexion angles could be obtained.

#### **4.4.2 Influence of Flat Foot Landing on Non-Contact ACL Injury Risks**

Flat-foot landings resulted in significant lower knee flexion angles at maximum VGRF and significantly less knee flexion range of motion when compared to natural and fore-foot landings. Flat-foot landing also contributed to increasing maximum VGRF. This

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result is expected considering the impulse-momentum relationship. Under the same conditions for all the landings, impulse is the same but for flat-foot landings, the impact time is shorter, resulting in greater impact forces. Consequently, such large forces may be transmitted to the knee which is undesirable. To allow for greater force absorption, flat-foot landing should be avoided.

However, a more important result was that flat-foot landings resulted in higher knee flexion angles at initial contact, as compared to fore-foot landings. In the study by Cortes *et al.* (2007), which observed less ankle plantar flexion angles for natural landings than fore-foot landings, also found significantly higher knee flexion angles. These findings imply that lower fore-foot landing angles may be required for increased knee flexion angles at initial contact to occur. There is therefore a need to determine the optimal range of fore-foot landing angles that are associated with higher knee flexion angles, thereby reducing non-contact ACL injury risks.

#### **4.4.3 Limitations**

The main limitation of this study was that landing was carried out under controlled laboratory settings so that foot-landing in the sagittal plane could be varied and controlled. Participants might not land naturally with the prompting by the tester. Thus, instructions were provided very carefully to participants so that they followed the study protocol but not to the extent that influenced their natural landing strategies. This study also focused on only males. Females could be included in future to determine if it results in similar findings. This study also mainly focused on kinematic variables and not kinetic

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variables. Future studies could also incorporate kinetic variables for a more comprehensive study.

## **4.5 Conclusions**

The influence of foot landing in the sagittal plane on non-contact ACL injury risks during single-leg drop landings was studied. Knee flexion angles both at initial contact and maximum VGRF, as well as, knee flexion range of motion were investigated under the different landing conditions. A laboratory-controlled motion analysis experiment was conducted using 14 participants. Fore-foot landings are better in force absorption, as compared to flat-foot landings. Nevertheless, while fore-foot landing positions are often encouraged for greater foot absorption, participants were found to land at low knee flexion angles of less than 15°. This was associated with increased non-contact ACL injury risks, with ACL strain that was found to be the highest at initial contact for single-leg landings. Thus, it is necessary to study the optimal range of ankle plantar-flexion angles that are associated with higher knee flexion angles during initial contact. This new information could be provided to athletes as a guideline during landings to help minimize non-contact ACL injury risks. Such a study was investigated and the results would be presented in Chapter 5.

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## **Chapter 5 – Influence of Ankle Plantar Flexion Angles and Internal Risk Factors on Knee Flexion Angles and Role of the Gastrocnemius during Single-Leg Drop-Landings**

### **5.1 Introduction**

Fore-foot landings have been encouraged to allow for greater force absorption (Alentorn-Geli *et al.*, 2009a). Nevertheless, it is also reported that fore-foot landings (ankle plantar flexion) have been associated with lower knee flexion angles at initial contact with the ground (Cortes *et al.* 2007). Such low knee flexion angles are often associated with higher non-contact ACL injury risks (Krosshaug *et al.*, 2007). In single-leg landings, low knee flexion angles of less than 15° during initial contact were also observed (Weinhandl *et al.*, 2010). This poses a further increase in non-contact ACL injury risks as hamstring activation becomes less efficient in countering the anterior pull of the quadriceps at such low knee flexion angles (Li *et al.*, 1999, Yu *et al.*, 2005). It is therefore crucial to understand the relationship between ankle plantar flexion angles and knee flexion angles especially during single-leg landings. However, this has so far not been reported in literature.

The study by DiStefano *et al.* (2008) also found that the use of ankle bracing reduced ankle plantar flexion and increased knee flexion angles at initial contact. This further suggests that reduced ankle plantar flexion angles could be associated with increased knee flexion angles. This leads to the belief that there can be an optimal range of ankle plantar flexion angles that are associated with higher knee flexion angles. Determining

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this optimal range would be beneficial for players to land fore-foot safely while reducing non-contact ACL injury risks.

Generalized joint laxity (GJL) is an internal anatomical risk factor that is associated with higher non-contact ACL injury risks (Uhorchak *et al.*, 2003, Ramesh *et al.*, 2005). Uhorchak and colleagues (2003) also found that body mass index (BMI) has a significant effect on ACL injury risks for males and females combined, and for females alone. In the study, those who sustained the ACL injury had significantly higher BMI than those who did not sustain the ACL injury (Uhorchak *et al.*, 2003). However, so far, no known studies have been reported relating these factors with the knee flexion angles during landing. Furthermore, although these factors are associated with higher non-contact ACL injury risks, studies have often focused on studying internal risk factors alone, rather than together with other risk factors, as reviewed by Ali *et al.* (2010). This reduces the comprehensiveness of these studies (Ali *et al.*, 2010) and considering internal factors in biomechanical studies could add further insights into ACL injury risk reduction.

Also, during single-leg landings with low knee flexion angles, hamstring's effectiveness to oppose the anterior tibial force produced by quadriceps is reduced (Li *et al.*, 1999, Yu *et al.*, 2005). Another knee flexor, the gastrocnemius, has therefore being a focus of research for a few studies to examine its role in ACL injury prevention (Dürselen *et al.*, 1995, Fleming *et al.*, 2001b, Elias *et al.*, 2003, Pflum *et al.*, 2004, Ali *et al.*, 2014, Morgan *et al.*, 2014, Adouni *et al.*, 2016). The gastrocnemius crosses both the knee joint and the ankle joint and is both a knee flexor and ankle plantar flexor (Hartigan *et al.*,

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2011). The gastrocnemius was found to increase in activation force, likely to compensate for reduced hamstring forces during the landing phase of single-leg landings (Morgan *et al.*, 2014). However, it is unclear what the role of the gastrocnemius is prior initial contact with the ground, at which ACL strain is the highest (Lamontagne *et al.*, 2008). Besides, the gastrocnemius is less of a knee flexor when the ankle is plantar flexed (Hartigan *et al.*, 2011). It would therefore be important to study the gastrocnemius activation, together with ankle plantar flexion positions as its activation has an impact on the latter.

Therefore, the main purpose of this study was to be able to obtain a relationship between ankle plantar flexion angles and knee flexion angles at initial contact with the ground. In doing so, this study aimed to determine the optimal range of fore-foot landings associated with reduced non-contact ACL injury risks. The internal factors of GJL and BMI were also included to add further insights into their influence on knee flexion angles at initial contact. The first objective was to investigate how ankle plantar flexion angles, GJL and BMI influence knee flexion angles at initial contact during single-leg drop landing. The hypothesis was that reduction of ankle plantar flexion angles would increase knee flexion angles at initial contact, as obtained from results of Chapter 4. A decrease in BMI might increase body flexibility (Cinthuja *et al.*, 2015) and an increase in GJL increases “joint hypermobility” (Shultz *et al.*, 2007, p. 241) and were hypothesized to increase knee flexion angles at initial contact.

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The second objective of this study was to compare muscle activation levels among gastrocnemius, hamstring and quadriceps, prior initial contact. The hypothesis was that gastrocnemius activation would be significantly higher than the hamstring activation but would be significantly lower than the quadriceps activation. A similar result was found in the study by Morgan *et al.*, (2014) during the landing phase of single-leg landings and was therefore also hypothesized in this study for the time prior initial contact.

## **5.2 Methods**

### **5.2.1 Participants**

Twenty-three male recreational basketball players (age: mean 22.8 (SD 2.3) years old, height: mean 1.77 (SD 0.06) m, body mass: mean 70.5 (SD 8.1) kg) and 10 female recreational basketball/netball/handball players (age: mean 20.7 (SD 1.6) years old, height: mean 1.62 (SD 0.07) m, body mass: mean 50.6 (SD 6.6) kg) had agreed to take part in this study. Participants had no previous ACL injuries or lower extremity injuries in the previous 6 months prior to the study, and played their relevant sport regularly, minimally once a week for at least 2 years before the study. To avoid hormonal influence and discomfort due to premenstrual syndromes (2-3 days before period starts) and menstrual pain (days 1-6), measurements for the females in this study were taken during the late follicular phase (days 7-11, for a 28-day cycle) of their menstrual cycle. Therefore, only female participants with regular menstrual cycles were recruited. The study was approved by the Nanyang Technological University Institutional Review Board. All participants signed a written consent before participating in the study.

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The male participants in Chapter 4 and this chapter had the same single-leg drop-landing protocols during the natural landing technique. Therefore, the kinematics (knee flexion and ankle plantar flexion angles), GJL and BMI data of the male participants from both studies were combined (age: mean 23.0 (SD 2.1) years old, height: mean 1.76 (SD 0.06) m, body mass: mean 69.5 (SD 9.0) kg). The combined sample size was  $n = 37$ . As a verification, kinematic variables between both groups of participants were not found to be significantly different (Details are appended in Appendix H).

A separate analysis, consisting of combined male and female participants' data, was also carried out since kinematic variables at initial contact were not found to be significantly different (see Appendix I). This combined sample size was  $n = 47$ . Combining data of male and female participants for regression analyses had been similarly conducted by Uhorchak *et al.* (2003) and Walsh *et al.* (2012).

A power analysis was performed using G\*Power (v3.1.9.2, Heinrich-Heine University, Düsseldorf, Germany). As there were no prior studies in literature, data from Chapter 4 were used to estimate effect size. A large effect size between ankle plantar flexion angles and knee flexion angles was obtained ( $r^2 = 0.48$ ), based on guidelines by Cohen (1988) (Appendix J). Using a large effect size of 0.35 for a linear multiple regression, a sample size of 36 was required (Power  $1-\beta = .80$ ,  $\alpha = .05$ ) for 3 predictors and was achieved in this study.

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## 5.2.2 Equipment

Participants wore the same type of shoes (Mizuno Wave Twister 3, Mizuno Corporation, Osaka, Japan). Common shoe sizes of US 6.5, 8, 9, 10, 12 were used in the study and participants were pre-screened to ensure that they fall in the size range. Shoes were snugly tied up to allow minimal relative movements between shoes and feet. When required, additional shoe padding was also provided. This was carried out to minimize relative motion between the shoes and feet.

Retroreflective markers were attached to the dominant leg, as declared by the participants as the leg that kicked the ball the hardest for the ball to travel the longest distance (Hewett *et al.*, 2005; see Figure 22). The markers were attached to the sacrum, anterior superior iliac spine, mid-thigh, medial and lateral knee epicondyles, tibial tuberosity, head of fibula, anterior aspect of shin, medial and lateral malleoli, distal calcaneus (CAL1), proximal calcaneus (CAL2), head of first metatarsal, head of fifth metatarsal and toe (between second and third metatarsal heads). The last five markers were placed on the shoe and steps were taken to locate these positions. Previous studies had also placed markers directly on the shoes (Kovács *et al.*, 1999, Ford *et al.*, 2006, Orishimo *et al.*, 2009, Weinhandl *et al.*, 2010, Laughlin *et al.*, 2011, Dempsey *et al.*, 2012, Ali *et al.*, 2013, Tran *et al.*, 2016). The participants were first told to remove their shoes. CAL1, CAL2, head of first metatarsal, head of fifth metatarsal and toe positions were palpated without shoes being worn. The metatarsal and toe's palpated positions were held on while the participants were briefed about the placement of markers at these positions on the

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shoes. Immediately after, the shoes were worn, and testers located the metatarsal and toe positions with the feedback from participants.

For the CAL1 and CAL2 positions, the markers' proximal positions were estimated when the markers were placed on the shoe. However, the medial-lateral position of the CAL2 marker was aligned to the medial-lateral position of the Achilles tendon. The medial-lateral position of the Achilles tendon could be observed from above the most proximal aspect of the shoe (see Appendix A). Marker CAL1 was placed vertically below CAL2, above the heel of the shoe (see Appendix A). Marker was also attached to the anterior superior iliac spine of the non-dominant leg for the estimation of the hip joint location. Markers were attached to these key body landmarks using tape (Leukoplast®, BSN medical GmbH, Hamburg, Germany; 3M™ Durapore™ Surgical Tape, 3M, St. Paul, MN, USA). A surgical netting (Surgifix Size 6, FRA Production S.p.A, Cisterna d'Asti (AT), Italy) was placed over the leg to secure the markers, in addition to the tape that was used to secure them. This also helped to minimize errors due to skin movements.



Marker	Landmark
1	Right anterior superior iliac spine
2	Left anterior superior iliac spine
3	Sacrum
4	Mid-thigh
5	Lateral knee epicondyle
6	Medial knee epicondyle
7	Head of fibula
8	Tibial tuberosity
9	Anterior aspect of shin
10	Lateral malleolus
11	Medial malleolus
12	Head of fifth metatarsal
13	Toe (between second and third metatarsal heads)
14	Head of first metatarsal
15	Proximal calcaneus (CAL2)
16	Distal calcaneus (CAL1)

**Figure 22: Positions of marker placement.**

Three-dimensional motion data were captured at 240 Hz using 8 digital cameras and the Cortex software (version 2.6.2.1169, Motion Analysis Corp, Santa Rosa, CA, USA). Details of how the equipment was set-up and calibrated are included in Appendix B. Ground reaction forces were recorded using a force plate (AMTI Model OR6-7-2000, Advanced Mechanical Technology Inc., Watertown, MA, USA) and electromyography (EMG) sensors (Trigno™ Wireless EMG, Delsys, Inc., Natick, MA, USA) were attached to the muscle bellies of the gastrocnemius medial, quadriceps (vastus lateralis) and hamstring (biceps femoris) (Cram, 2011). EMG sensors were orientated along the pennation angle of the muscles (Ward *et al.*, 2009). For EMG preparation, the skin was shaved and wiped thoroughly with alcohol. Force plate and EMG data were recorded at a sampling rate of 1920 Hz, in synchronization with the cameras. Kinematic and kinetic data were further processed using Visual3D (v5.00.33, C-Motion, Germantown, PA, USA).

### **5.2.3 Test Protocol**

The Beighton score (Shultz *et al.*, 2007) was first taken to monitor GJL. BMI was also calculated using the height and weight measurements provided by participants. Participants then each carried out a set of warm-up exercises for 5 min, targeting the lower extremity muscles, since drop-landing trials would be conducted. The exercises were stretching of the hamstrings, quadriceps and calves, tip-toe and ankle rotation, squats and low lunges, jumps, jogging on the spot and jogging with high knees. Maximal voluntary contraction (MVC) was carried out for the gastrocnemius, quadriceps and hamstring muscles using the methodology as described by Hsu *et al.* (2006) on an

isokinetic dynamometer (Biodex System 4 Pro, Biodex Medical Systems, Inc., Shirley, NY, USA). This isometric MVC methodology concurrently measured the maximum level of muscle activation of different lower limb muscles using two tasks on a dynamometer (Hsu *et al.*, 2006). This methodology was shown to produce the same or higher level of maximum muscle activation, as compared to traditional MVC protocols of individual muscles but was less time-consuming to carry out (Hsu *et al.*, 2006).

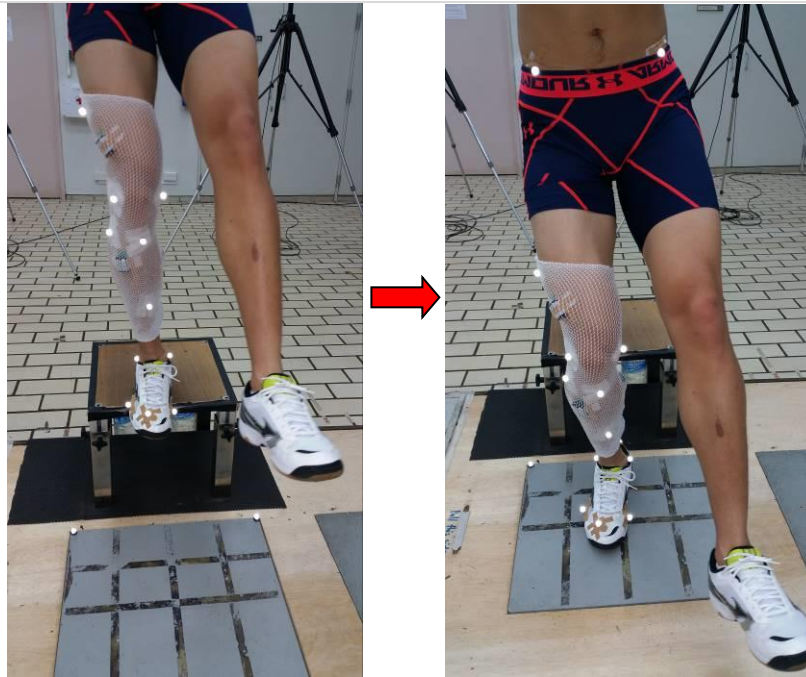
After taking a rest, participants practised the drop-landing manoeuvres before the start of the study. Participants were requested to stand with their dominant leg on a 30 cm-height platform, and drop land naturally onto a force plate using the same leg (see Figure 23). The platform height was tested during pilot studies to allow the protocol to be carried out without over-exertion or compromise of safety and had been used in previous studies (Cortes *et al.*, 2007, Mokhtarzadeh *et al.*, 2013, Orishimo *et al.*, 2014, Tran *et al.*, 2016, Valenzuela *et al.*, 2016). Furthermore, the platform height was also close to the maximum jump height for males (mean 35.5 (SD 4.5) cm) and females (mean 25.6 (SD 3.5) cm) (Abián *et al.*, 2008).

No restrictions were made on whether participants perform fore-foot or flat-foot landings if it was how participants typically land. Participants were requested to face a board marked with a red cross, placed at the front of the lab at around their eye-level, during landing. The participants were also instructed to keep their bodies upright and hands crossed over their chests to reduce any variability from swinging arms (Decker *et al.*, 2003, Fujii *et al.*, 2012). Participants were also requested not to bend down or jump up

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when they carried out the drop-landing trials (Decker *et al.*, 2003, Shimokochi *et al.*, 2013).

A standing trial was initially captured. Following that, data of 2 sets of 3 landing trials were captured. The sequence of conditions was randomized for each participant to minimize the influence due to fatigue, learning effects or boredom. At least a rest time of 5 min was provided between sets of trials. A successful trial was considered when the sacrum marker, a proxy for the centre of gravity, was maintained in position in the z axis just before drop-landing. This ensured that the participant did not bend down or jump up, thus maintaining the vertical height across the different jump trials. A trial was also considered successful only when EMG and marker data were captured and when participants landed within the force plate. Post-processing was used to verify that these conditions were met and data of successful trials were averaged. The Borg scale (0-10) was used to monitor the fatigue level of the participants (Borg, 1982).



**Figure 23. Participant drop-landing from the 30-cm platform onto the force plate.**

#### **5.2.4 Data Reduction**

Knee flexion angles and ankle plantar flexion angles at initial contact, as well as, knee flexion angles at maximum vertical ground reaction force (VGRF) were recorded for each participant. Ankle flexion angle was defined as the orientation of the foot segment relative to the shank segment in the sagittal plane. Knee flexion angle was defined as the orientation of the shank segment, with respect to the thigh segment in the sagittal plane. Details of how these measurements were derived are included in Appendix C. Force plate data were collected to detect the initial contact with the ground and the maximum VGRF. Initial contact was defined as the first point that VGRF exceeded 10 N (Self *et al.*, 2001, Ford *et al.*, 2006, Fujii *et al.*, 2012).

The motion and VGRF data were filtered through a low-pass, 4th order, zero phase shift Butterworth digital filter (Dempsey *et al.*, 2012, Valenzuela *et al.*, 2016), at a cut off frequency of 9 Hz and 50 Hz, respectively (see Appendix D). Knee flexion and ankle plantar flexion angles were denoted as positive. Cardan rotation sequence xyz was used for the knee joint angle calculation and the single-leg model was built in Visual3D (v5.00.33, C-Motion, Germantown, PA, USA). Ankle plantar flexion angle measured during the static standing trial was used as each participant's neutral (zero) alignment. Ankle plantar flexion angle during the drop-landing was measured and referenced to the standing position (Ford *et al.*, 2006). Maximum VGRF was normalized to body weight (BW).

The EMG data of the gastrocnemius, quadriceps and hamstring muscles were band-pass filtered (20 to 450 Hz) during data collection (Trigno™ Wireless EMG, Delsys, Inc., Natick, MA, USA). The raw EMG data during the pre-landing phase, 100 ms before initial contact with the ground (Croce *et al.*, 2004, Russell *et al.*, 2007, Malfait *et al.*, 2016), was taken into consideration in this study to study muscle activation prior initial contact. Initial contact was first identified for each subject. The same sampling rate was set at 1920 Hz for all subjects. Since initial contact can be identified for each subject, EMG data during the pre-landing phase, 100 ms before initial contact with the ground, was calculated for every participant. To remove direct current (DC) offset, the EMG data was high-pass filtered at 10 Hz with a zero-lag Butterworth 4<sup>th</sup> order filter. To obtain the linear envelope, the EMG data was full-wave rectified and low-pass filtered at 10 Hz with zero-lag Butterworth 4<sup>th</sup> order. The average EMG in volts (= integrated

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EMG/integration time) (Aagaard *et al.*, 2002) of each muscle was obtained for 10 ms before initial contact (Zebis *et al.*, 2009, Oliveira *et al.*, 2014). MVC raw data was similarly processed. However, integrated EMG was obtained for a 0.5-s window, centred on the peak value of the EMG burst (Hsu *et al.* 2006) and divided by the integration time period to also obtain the data in volts (Aagaard *et al.*, 2002). Muscle EMG data was then normalized to the MVC data. EMG data reduction was performed using customised programs written in MATLAB (v. R2014a, The Mathworks Inc., Natick, MA, USA).

### **5.2.5 Statistical Analysis**

All statistical analyses were performed using the statistical package SPSS (IBM SPSS 22.0, IBM Corp., NY, USA). Data are expressed as mean (standard deviation).

#### **5.2.5.1 Regression Models**

A multiple regression was run to predict knee flexion angles from ankle plantar flexion angles, GJL and BMI. This provided the relationship needed to obtain the optimal range of ankle plantar flexion angles associated with desirable knee flexion angles of more than 15°. Assumptions required were checked:

- Linearity was checked using partial regression plots, independence of residuals was assessed using the Durbin-Watson statistic and homoscedasticity was verified using visual inspection of a plot of studentized residuals versus unstandardized predictive values.
- There was no multicollinearity found by verifying that tolerance values were greater than 0.1 (Lund *et al.*, 2015). No outliers, high leverage points and highly

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influential points were detected (using studentized deleted residuals to detect greater than  $\pm 3$  standard deviations, checking for leverage values that were greater than 0.2, and Cook's distance that were above 1) (Lund *et al.*, 2015).

- Normality could be assumed, by checking the P-P Plot.

Effect size was calculated as  $R^2$  and values of 0.0196, 0.1300 and 0.2600 were defined as small, medium and large effects, respectively (Cohen, 1988).

Binomial logistic regression was also performed to determine how well the model correctly predicts the high and low knee flexion angle cases. Data were first divided into low and high knee flexion angle groups. The statistical differences in variables (BMI, GJL, kinematic variables at initial contact and maximum VGRF and maximum VGRF) between the groups are appended in Appendix K. Knee flexion angles were considered as 'high' if they were more than  $15^\circ$  and as 'low' if they were less than  $15^\circ$  at initial contact. The  $15^\circ$  threshold was set because hamstring activation becomes less efficient in reducing ACL loading at less than  $15^\circ$  (Li *et al.*, 1999, Yu *et al.*, 2005). Linearity of the continuous variables with respect to the logit of the dependent variable was checked using the Box-Tidwell procedure (Lund *et al.*, 2015). The variables used to determine how well the model predicts high knee flexion angles were:

- **Statistical significance** of model of  $\alpha < 0.05$  shows that the model predicts low and high categories of knee flexion angles well, as compared to having no independent variables.
- **Nagelkerke  $R^2$  value** determines how much the model explained the variance in knee flexion angles.

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- **Sensitivity** shows the percentage of cases that had high knee flexion angles which were correctly predicted by the model.
  - **Specificity** shows the percentage of cases that had low knee flexion angles which were correctly predicted by the model.

#### 5.2.5.2 Analysis for Muscle Activation

One-way Analysis of Variance (ANOVA) with repeated measures was conducted at  $\alpha = 0.05$  to compare muscle activation levels among the gastrocnemius, quadriceps and hamstring muscles. Normality assumption was tested with the Shapiro-Wilk test ( $p > 0.05$ ). As normality could not be assumed and a “logarithmic” transformation was applied. Assumption of sphericity was tested using the Mauchly's test of sphericity and this assumption was not violated. Post-hoc test with Bonferroni adjustment was made with a family-wise  $\alpha = 0.05$ . Effect size was calculated as  $\eta^2$  and values of 0.01, 0.09 and 0.25 were defined as small, medium and large effects, respectively (Hanna *et al.*, 2012). Data are expressed as mean (standard deviation). Definition of normality, sphericity and effect size are found in Appendix F.

Subsequent analyses to determine the differences in muscle activation levels, quadriceps to hamstring ratio (Q:H) and gastrocnemius to quadriceps (G:Q) ratios between the low and knee flexion angles groups were made using the non-parametric test, Mann-Whitney U, at  $\alpha = 0.05$ .

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## 5.3 Results

### 5.3.1 Regression Models

#### 5.3.1.1 Regression Models for Males (n = 37)

The multiple regression model for males was performed to predict knee flexion angles ( $\angle\text{KF}$ ) with ankle plantar flexion angles ( $\angle\text{AF}$ ), as well as, internal factors GJL ( $I_{\text{GJL}}$ ) and BMI ( $I_{\text{BMI}}$ ). BMI was not statistically significant ( $p > 0.05$ ) and was removed from the model. The regression model is described in Equation (3):

$$\angle\text{KF} = -0.45 (\angle\text{AF}) + 0.74 (I_{\text{GJL}}) + 26.33 \quad (3)$$

where  $\angle\text{KF}$  and  $\angle\text{AF}$  are at initial contact with the ground.

The slope coefficients of the ankle plantar flexion angles and GJL were statistically significant ( $p < 0.001$  and  $p = 0.038$ , respectively). The model in equation (3) accounted for 40% of the variation in knee flexion angles with adjusted  $R^2$  of 36%, a large size effect according to Cohen (1988) and significantly predicted knee flexion angles ( $p < 0.001$ ). Fixing GJL as 1 (mean value for male participants, see Appendix I), predictions made from the model show that ankle plantar flexion angle of  $24^\circ$  would result in knee flexion angles of  $16^\circ$ , 95% CI [15 to 18 (see Table 6). This is the maximum ankle plantar flexion angle to achieve at least  $15^\circ$  of knee flexion angle for males, considering the 95% confidence interval.

**Table 6. Predicted Knee Flexion Angles from Ankle Plantar Flexion Angles at Initial Contact (GJL = 1 in this Prediction) (Males) #**

Input	Model Prediction	95% Confidence Interval	
		Lower Bound	Upper Bound
Ankle Plantar Flexion Angles (°)	Knee Flexion Angles (°)		
26	15	14*	17
25	16	14*	17
24	16	15	18

\* Less than 15° of knee flexion angles at initial contact.

# Calculated based on equation (3), with n = 37 and GJL = generalized joint laxity.

Binomial logistic regression for males was then performed using ankle plantar flexion angles ( $p = 0.013$ ) as the only predictor variable as the other variables were not statistically significant ( $p > 0.05$ ). This model was statistically significant ( $p < 0.05$ ) and explained 31% of the variance (see Table 7). Sensitivity was 90% and specificity was 71% (see Table 7).

**Table 7. Binomial Logistic Regression on Effect of Ankle Plantar Flexion Angles on Likelihood of Obtaining High Knee Flexion Angles (Males)**

<b>Model</b>	<b>Significance of Model</b>	<b>Nagelkerke R<sup>2</sup></b>	<b>Sensitivity</b>	<b>Specificity</b>
<b>With Outliers (n = 37)</b>				
<b>Ankle Plantar Flexion (°)</b>	0.002*	31%	90%	71%
<b>Without Outliers (n = 35)</b>				
<b>Ankle Plantar Flexion (°)</b>	<0.001*	69%	90%	81%

\*Model is statistically significant ( $p < 0.05$ ).

Two outliers were found in the analysis. The two outliers were temporarily removed to study the effect of ankle plantar flexion angles on knee flexion angles for the rest of the majority of participants. When these two outliers were excluded, leaving 95% (or  $n = 35$ ) of the rest of the sample size, there was a steep increase in  $R^2$ . The model then explained 69% of the variance (see Table 7) and the predictor variable (ankle plantar flexion angles) was once again statistically significant ( $p = 0.005$ ). Sensitivity was 90% and specificity rose to 81% (see Table 7).

For both the models with or without outliers, the decrease of the ankle plantar flexion angles was associated with the likelihood of having high knee flexion angles ( $\geq 15^\circ$ ).

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### 5.3.1.2 Regression Models for Males and Females Combined (n = 47)

The multiple regression model for male and female participants combined was performed to predict knee flexion angles ( $\angle\text{KF}$ ) with ankle plantar flexion angles ( $\angle\text{AF}$ ), as well as, internal factors GJL ( $I_{\text{GJL}}$ ) and BMI ( $I_{\text{BMI}}$ ). The model is described in Equation (4):

$$\angle\text{KF} = -0.46 (\angle\text{AF}) + 0.68 (I_{\text{GJL}}) - 0.57 (I_{\text{BMI}}) + 39.54 \quad (4)$$

where  $\angle\text{KF}$  and  $\angle\text{AF}$  are at initial contact with the ground.

Ankle plantar flexion angles ( $p < 0.001$ ), GJL ( $p = 0.011$ ) and BMI ( $p = 0.012$ ) were statistically significant. The prediction model accounted for 48% of the variation in knee flexion angles with adjusted  $R^2$  of 45%, a large size effect according to Cohen (1988) and significantly predicted knee flexion angles ( $p < 0.001$ ). Fixing GJL as 1 and BMI as 22 (mean values for male participants, see Appendix I), predictions made from the model show that ankle plantar flexion angle of  $26^\circ$  would result in knee flexion angles of  $16^\circ$ , 95% CI [15 to 17] (see Table 8). This is the maximum ankle plantar flexion angle for males to achieve at least  $15^\circ$  of knee flexion angle, considering the 95% confidence interval.

**Table 8. Predicted Knee Flexion Angles from Ankle Plantar Flexion Angles at Initial Contact (GJL = 1, BMI = 22 in this Prediction) (Males and Females) #**

Input	Model Prediction	95% Confidence Interval	
		Lower Bound	Upper Bound
Ankle Plantar Flexion Angles (°)	Knee Flexion Angles (°)		
27	15	14*	16
26	16	15	17
25	16	15	17

\* Less than 15° of knee flexion angles at initial contact.

# Calculated based on equation (4), with n = 47, GJL = generalized joint laxity, BMI = body mass index.

Fixing GJL as 3 and BMI as 19 (mean values for female participants, see Appendix I) instead, predictions made from the model show that ankle plantar flexion angle of 28° would result in knee flexion angles of 18°, 95% CI [15 to 20] (see Table 9). This is the maximum ankle plantar flexion angle to achieve at least 15° of knee flexion angle for females, considering the 95% confidence interval.

**Table 9. Predicted Knee Flexion Angles from Ankle Plantar Flexion Angles at Initial Contact (GJL = 3, BMI = 19 in this Prediction) (Males and Females) #**

Input	Model Prediction	95% Confidence Interval	
		Lower Bound	Upper Bound
Ankle Plantar Flexion Angles (°)	Knee Flexion Angles (°)		
29	17	14*	20
28	18	15	20
27	18	15	21

\* Less than 15° of knee flexion angles at initial contact.

# Calculated based on equation (4), with n = 47, GJL = generalized joint laxity, BMI = body mass index.

Binomial logistic regression was then performed using ankle plantar flexion angles ( $p = 0.002$ ) as the only predictor variable as the other variables were not statistically significant ( $p > 0.05$ ). The model is statistically significant ( $p < 0.001$ ) and explained 37% of the variance of knee flexion angles (see Table 10). Sensitivity was 85% and specificity was 70% (see Table 10).

**Table 10. Binomial Logistic Regression on Effect of Ankle Plantar Flexion Angles on Likelihood of Obtaining High Knee Flexion Angles (Males and Females Combined)**

Model	Significance of Model	Nagelkerke R <sup>2</sup>	Sensitivity	Specificity
<b>With Outliers (n = 47)</b>				
<b>Ankle Plantar Flexion (°)</b>	<0.001*	37%	85%	70%
<b>Without Outliers (n = 45)</b>				
<b>Ankle Plantar Flexion (°)</b>	<0.001*	67%	89%	74%

\*Model is statistically significant ( $p < 0.05$ ).

The same two outliers as before were found in the analysis. The two outliers were temporarily removed to study the effect of ankle plantar flexion angles on knee flexion angles for the rest of the majority of participants. When these outliers were excluded, leaving 96% (or  $n = 45$ ) of the rest of the sample size, there was an increase in  $R^2$  once again. The model then explained 67% of the variance and the predictor variable (ankle plantar flexion angles) was statistically significant ( $p = 0.001$ ). Sensitivity and specificity rose to 89% and 74%, respectively (see Table 10).

For both models with or without outliers, the decrease of the ankle plantar flexion angles was associated with the likelihood of having high knee flexion angles ( $\geq 15^\circ$ ).

### 5.3.1.3 Generalized Joint Laxity (GJL) Results

There were no significant differences in GJL between low and high knee flexion groups for males, as well as, males and females combined (see Appendix K). There were also no

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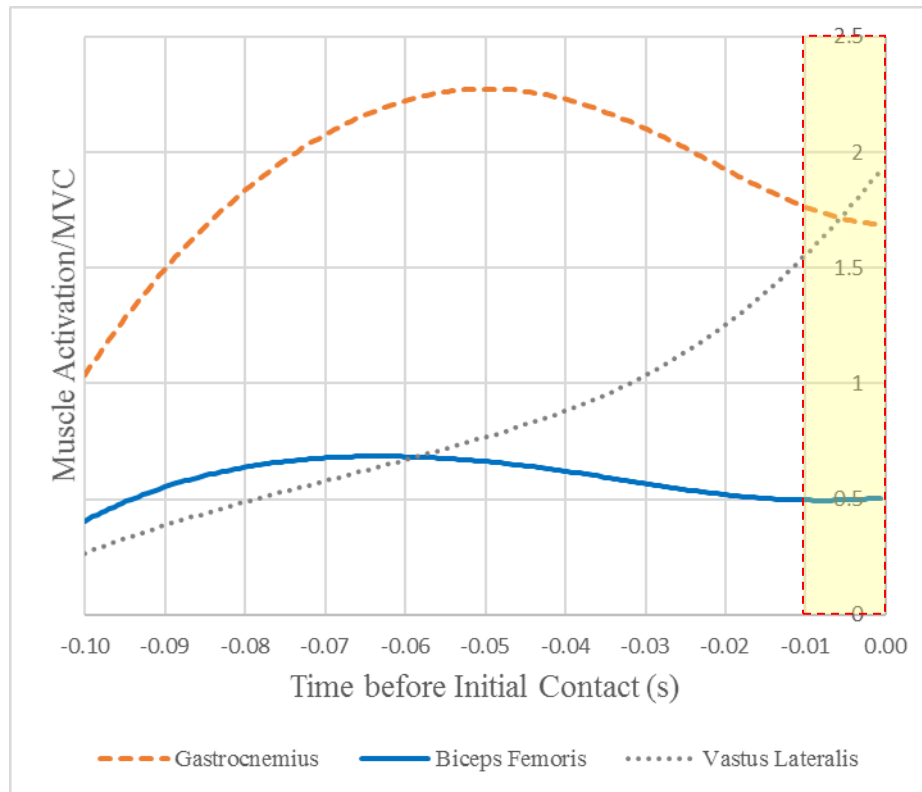
significant differences in GJL between genders (see Appendix I). In terms of GJL values of each individual, no Beighton scores went beyond 6, which defined hypermobility (Shultz *et al.*, 2007).

### 5.3.2 Muscle Activation Prior Initial Contact

MVC measurements were recorded only for participants taking part in this study in Chapter 5. Participants for the study as described in Chapter 4 could therefore not be used in this section. Male and female participants' data were combined in this study since EMG data of gastrocnemius, hamstrings and quadriceps were not found to be significantly different (see Appendix L). Out of the sample size of 33 participants for male and female participants combined, only 18 participants' EMG data could be used in this analysis. Participants were not performing at maximal effect, possibly due to a lack of sleep, muscle soreness or as some were feeling unwell during the study. EMG sensors were also not adhering well for others who were profusely perspiring. All these contributed to exceptionally low MVC readings, resulting in the reduction of usable EMG data (see Appendix M).

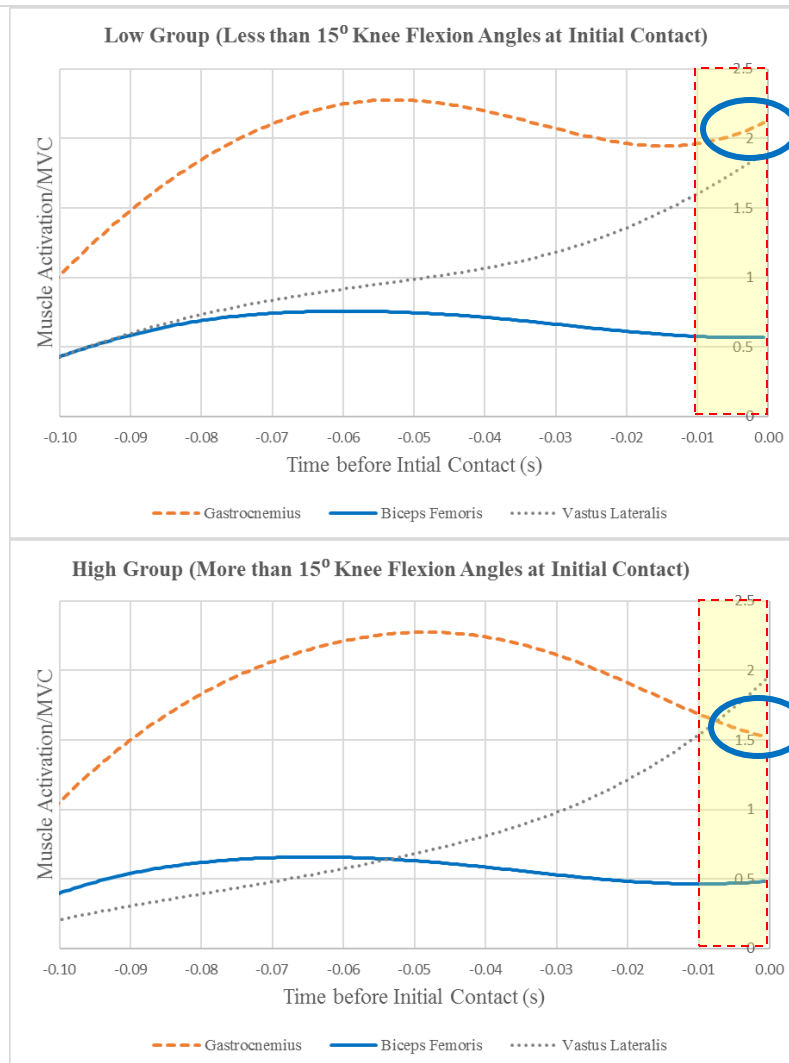
Figure 24 shows the normalized EMG waveforms of the gastrocnemius, hamstring and quadriceps muscles, 100 ms before initial contact with the ground for the combined data of male and female participants. The last 10 ms before initial contact was extracted for data analysis. There were significant differences in average EMG over this last 10 ms before initial contact across the muscles ( $p < 0.001$ ,  $\eta^2 = 0.71$ ). Muscle activation of the

hamstring (mean 0.48 (SD 0.30)) was significantly less than that in the gastrocnemius (mean 1.66 (SD 0.68)) and quadriceps (mean 1.68 (SD 1.04)) (both at  $p < 0.001$ ).



**Figure 24. Normalized electromyography waveforms of the gastrocnemius, hamstring and quadriceps muscles 100 ms before initial contact with the ground (males and females combined) ( $n = 18$ ). Muscle activation was normalized to the maximal voluntary contraction (MVC). The yellow box represents the last 10 ms before initial contact that is extracted for data analysis.**

Figure 25 shows the comparison of normalized EMG waveforms of the gastrocnemius, hamstring and quadriceps muscles 100ms before initial contact, between knee flexion groups with less than ( $n = 5$ ) and more than  $15^\circ$  of knee flexion angles ( $n = 13$ ) at initial contact.



**Figure 25. Comparison of normalized electromyography waveforms of the gastrocnemius, hamstring and quadriceps muscles 100 ms before initial contact with the ground between low ( $n = 5$ ) and high ( $n = 13$ ) knee flexion groups. Muscle activation was normalized to the maximal voluntary contraction (MVC). The yellow box represents the last 10 ms before initial contact that is extracted for data analysis. The blue circle shows the differences in the gastrocnemius activation at the last 10 ms before initial contact.**

The last 10 ms before initial contact was again extracted for data analysis between the low and high knee flexion angle groups. There was no significant difference found in the muscle activation for the 3 muscles, quadriceps to hamstring ratio (Q:H) and

gastrocnemius to quadriceps (G:Q) ratios between the low and high knee flexion angle groups (see Table 11).

**Table 11. Comparison of Muscle Activation Levels (Normalized to Maximal Voluntary Contraction) between the Low and High Knee Flexion Groups (mean (standard deviation))**

<b>Muscle</b>	<b>Low (n = 5)</b>	<b>High (n = 13)</b>	<b>p - Value</b>
<b>Gastrocnemius</b>	1.96(0.76)	1.54(0.64)	0.387
<b>Hamstring</b>	0.55(0.30)	0.45(0.31)	0.503
<b>Quadriceps</b>	1.71(0.98)	1.67(1.09)	0.775
<b>Q:H Ratio<sup>#</sup></b>	3.43(1.32)	4.87(3.25)	0.503
<b>G:Q Ratio<sup>*</sup></b>	1.51(1.22)	1.27(0.78)	1.000

<sup>#</sup>where Q:H refers to the quadriceps to hamstring ratio.

<sup>\*</sup>where G:Q refers to the gastrocnemius to quadriceps ratio.

## 5.4 Discussion

The first aim of this study was to investigate how ankle plantar flexion angles, GJL and BMI influence knee flexion angles at initial contact during single-leg drop landing. The second was to compare muscle activation levels among gastrocnemius, hamstring and quadriceps, prior initial contact. As hypothesized, the reduction of ankle plantar flexion angles and the increase in GJL, increased knee flexion angles at initial contact (see Equations (3) and (4)). However, the reduction of BMI only increased the knee flexion angles at initial contact for the combined data of male and female participants (see Equation (4)). Also as hypothesized, gastrocnemius activation was found to be

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significantly higher than hamstring activation but was not significantly different from quadriceps activation (see Section 5.3.2).

#### **5.4.1 Role of Ankle Plantar Flexion Angles in Increasing Knee Flexion Angles**

Ankle plantar flexion angles was statistically significant both in the multiple regression and in the binomial logistic regression models (see Section 5.3.1). This suggests that ankle plantar flexion angles have an influence on knee flexion angles and as such could be used as a predictor variable to vary knee flexion angles. The regression models showed that an increase of ankle plantar flexion angles decreased the likelihood of increasing knee flexion angles. The gastrocnemius, both a plantar flexor and knee flexor, is less of a knee flexor as the ankle is plantar flexed (Hartigan *et al.*, 2011). Thus, with higher ankle plantar flexion angles, the gastrocnemius activation might not be sufficient to act as a knee flexor. Hamstring, the stronger knee flexor, is also less efficient in countering quadriceps at knee flexion angles of less than 15° (Li *et al.*, 1999, Yu *et al.*, 2005) during the single-leg landings. This might explain why higher ankle plantar flexion angles tend to be associated with lower knee flexion angles.

Including the outliers, the binomial logistic model using only ankle plantar flexion angles already explained 31% of the variance of knee flexion angles for males (see Table 7) and 37% for the combined data for male and female participants (see Table 10). When the two outliers were removed, the models explained almost 70% of the variance of knee

flexion angles (see Table 7 and Table 10). This suggests that for the majority of the participants (at least 95%), controlling ankle flexion angles within the optimal range could contribute substantially to increasing knee flexion angles at initial contact. The models predict high knee flexion angles very well, especially with males' data alone (sensitivity = 90%) (see Table 7 and Table 10). Other studies have investigated how 'soft' landings resulted in higher knee flexion angles (Devita *et al.*, 1992, Laughlin *et al.*, 2011). However, no report has been found to have looked at how ankle plantar flexion angles could be used to increase knee flexion angles, thereby reducing non-contact ACL injury risks. Controlling the ankle plantar flexion angles could therefore be a simple way for players to adopt to increase knee flexion angles, resulting in a reduction of non-contact ACL injury risks.

The multiple regression model further provides information on the optimal range of ankle plantar flexion angles that could be adopted to attain the desirable knee flexion angles of at least 15°. With the ankle plantar flexion angles and internal risk factors put together, the model using data for the combination of male and female participants (adjusted  $R^2 = 45\%$ ) explained about 10% higher variance of knee flexion angles than with males' data alone (adjusted  $R^2 = 36\%$ ). Thus, using the combined data from male and female participants and fixing GJL as 1, ankle plantar flexion angles should not exceed 26° with BMI of 22 for males. Fixing GJL as 3, ankle plantar flexion angles should not exceed 28° with BMI of 19 for females. These are the mean GJL and BMI values of male and female participants respectively (see Appendix I).

Currently in this study for the low knee flexion angle group, males landed with 31°, 95% CI [29 to 33] and the combined data of male and female participants landed with 32°, 95% CI [31 to 33] (see Appendix K). These ankle plantar flexion angles exceeded the optimal range recommended in this study.

Thus, while fore-foot landings have been encouraged to allow for greater force absorption (Alentorn-Geli *et al.*, 2009a), landing beyond the optimal range of ankle plantar flexion angles could in turn contribute to higher non-contact ACL injury risks.

#### **5.4.2 Contributory Reasons Leading to Higher Non-Contact ACL Injuries in Single-leg, rather than Double Leg Landings**

Similar to the results shown in Chapter 4, this study also shows that single-leg landings resulted in lower knee flexion angles than those found in double-leg landings (Cortes *et al.* 2007). The knee flexion angles at initial contact were also found to be significantly lower in single-leg landings, as compared to double-leg landings in the study by Pappas *et al.* (2007). The participants in this study also landed at higher ankle plantar flexion angles than in double-leg landings (Cortes *et al.*, 2007) (see Table 12). As described in Section 5.4.1, the lower knee flexion angles in single-leg landings, rather than double leg landings, could be partially contributed by higher ankle plantar flexion angles. The gastrocnemius, both a plantar flexor and knee flexor, becomes less of a knee flexor as the ankle is plantar flexed (Hartigan *et al.*, 2011).

**Table 12. Comparison of Ankle Plantar Flexion Angles between Double-Leg and Single-Leg Landings**

Landing	Males	Females
Double-Leg (Cortes <i>et al.</i> , 2007)	22°, 95% CI [15 to 28]	25°, 95% CI [19 to 30]
Single-Leg (This study)	28°, 95% CI [26 to 30]	30°, 95% CI [26 to 34]

With lower knee flexion angles during single-leg landing, the hamstring is also less efficient in opposing the quadriceps (Li *et al.*, 1999, Yu *et al.*, 2005). In fact, the results in this study showed significantly less hamstring activation than gastrocnemius and quadriceps activation at 10 ms before initial contact (see Section 5.3.2). This was also collaborated by the observations during the single-leg landing phase by Morgan *et al.* (2014). Figure 24 also shows a continual rise in quadriceps activation during the pre-landing phase of 100 ms before initial contact while the hamstring activation rose slightly before tapering off. Although the study by Lamontagne *et al.* (2008) did not normalize EMG to MVC and EMG levels between muscles cannot be directly compared (Halaki *et al.*, 2012), similar EMG trends were also observed in their study. With this continual rise in quadriceps activation and significantly smaller increase in hamstring activation (see Figure 24), there was likely a net anterior tibial translation, leading to increased ACL strain as pre-landing approaches initial contact (see Figure 5). Current literature also shows that lower knee flexion angles further increase ACL strain by increasing the patella tendon-tibia shaft angle, resulting in increased anterior shear force even for the same quadriceps muscle force (Yu *et al.*, 2007) (see Figure 10). ACL strain is also increased at lower knee flexion angles by increasing ACL elevation angle and deviation

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angle (Li *et al.*, 2005, Yu *et al.*, 2007) (see Figure 11). All these point to further net increase in anterior tibial translation, thereby increasing ACL strain to the maximum at initial contact, before ACL strain reduces as knee flexion angles increase after landing. Therefore, with less knee flexion angles, single-leg landings could result in higher non-contact ACL injury risks, as compared to double-leg landings.

Hamstring-to-quadriceps strengthening are being incorporated in training and prevention programmes (Shultz, 2007a). Since the hamstring is less efficient in opposing the quadriceps at low knee flexion angles (Li *et al.*, 1999, Yu *et al.*, 2005), this implies that hamstring strengthening would be less useful in reducing non-contact ACL injury risks during single-leg landings. This therefore further highlights the need to land in the optimal range of ankle plantar flexion angles that would result in higher knee flexion angles of more than 15°. Hamstring activation would then be effective to create a posterior tibial translation to reduce ACL strain.

In this study, almost half of the participants (see Appendix K) landed at less than 15° of knee flexion angles at initial contact. Therefore, this substantial group of participants are at risk of having higher non-contact ACL injury risks during single-leg landings.

### **5.4.3 Influence of Internal Risk Factors on Knee Flexion Angles**

When internal risk factors were added to the multiple regression models, the models explained a higher variance of knee flexion angles than when ankle plantar flexion angles were used as the only variables (see Appendix N). This was especially so in the case

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when data from males and females were combined. Adjusted  $R^2$  increased 1.5 times from 30% to 45%. This suggests that the GJL and BMI do have influences on the knee flexion angles. Assuming GJL is fixed for each person, when BMI changes, the optimal range of ankle plantar flexion angles would change for the same person. For a person with a higher than average BMI (e.g. obese), ankle plantar flexion angles have to be much lower for the same knee flexion angle. This might thus have an impact on force absorption during landing.

Similarly, based on the proposed model, between two individuals with the same BMI and ankle plantar flexion angles, the one with higher GJL would be able to achieve higher knee flexion angles. This would in turn lower non-contact ACL injury risks. With greater GJL, the knees could flex more easily. However, this seems somewhat contradictory to the fact that GJL is an internal anatomical risk factor that is associated with higher non-contact ACL injury risks (Uhorchak *et al.*, 2003, Ramesh *et al.*, 2005). Athletes with greater GJL may have hyperextension of the knee and less proprioceptive control at close to extended knee position, leading to increased anterior tibial translation (Ramesh *et al.*, 2005). The model in this study was based on participants who did not have hypermobility, since the Beighton scores were not greater than 6 (Shultz *et al.*, 2007). However, athletes with hypermobility may still have to train on their proprioception and should be aware of the risks of having higher joint laxity.

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#### 5.4.4 Role of the Gastrocnemius in Increasing Knee Flexion Angles

The gastrocnemius could act as a secondary knee flexor, but only if ankle plantar flexion angles are in the optimal range, as proposed in Section 5.4.1. This is because the gastrocnemius, both a plantar flexor and knee flexor, is less of a knee flexor as the ankle is plantar flexed (Hartigan *et al.*, 2011). Results showed that its activation was significantly higher than that of the hamstrings. Morgan *et al.* (2014) also suggested that increased gastrocnemius loading could compensate for reduced hamstring loading, resulting in lower ACL forces (see Figure 13). The gastrocnemius could have helped create knee flexion moment during low knee flexion angles since its activation creates the most knee flexion moment when the knee is extended (Hartigan *et al.*, 2011). This could continue to a level until the hamstrings takes over as stronger knee flexors.

Results in this study did not show a significant difference in gastrocnemius activation between the low and high knee flexion angle groups (Section 5.3.2). This could be due to the small sample sizes in the high ( $n = 13$ ) and especially in the low knee flexion angle ( $n = 5$ ) groups. However, from Figure 25, a drop in the gastrocnemius activation is observed, especially in the last 10 ms before initial contact, in the high, as compared to the low knee flexion angle group (blue circle). This could indicate that there might be an optimal range of gastrocnemius activation level associated with the optimal ankle plantar flexion angle range.

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### **5.4.5 Differences in Single-Leg Landing Kinematic Measures Between Genders**

No significant differences were found for the kinematic measures between the male and female participants at initial contact (see Appendix I). This result was somewhat surprising since female participants in this study seemed to have similar landing techniques as male participants although ACL injury rate for females is higher (Agel *et al.*, 2005). However, Cortes *et al.* (2007) also did not see significant gender effects on knee flexion and ankle flexion angles at initial contact. Similarly, Pappas *et al.* (2007) did not find significant gender differences in knee flexion angles at initial contact. This could be because landing with lower knee flexion angles might not be the reason for higher incidences of non-contact ACL injury for females, as compared to males. In fact, knee flexion angles at initial contact between ACL injured and uninjured females were also not found to be significantly different in the prospective study of 205 females by Hewett *et al.* (2005). Since there were no significant differences in kinematic variables during initial contact, the data of males and females were therefore combined in this study to further increase sample size.

The female participants in this study were also age matched to the male participants for comparison. However, the ratio of male to female injury rate also reduced as age increased and as players approached professional level (Renstrom *et al.*, 2008). As this current study focused on preventing non-contact ACL injuries in healthy participants, only participants with no prior ACL injuries were recruited. It could be possible that the

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female participants in this study have landing techniques that were also good enough to prevent them from sustaining ACL injuries during their teenage years when ACL injury rates were higher.

#### **5.4.6 Limitations**

There were a few limitations in this study. Firstly, the study in Chapter 5 was an extension of the previous study in Chapter 4 and thus, kinematics (knee flexion and ankle plantar flexion angles), GJL and BMI data were combined for male participants in the two studies. Due to experimental constraints, although both groups of participants had the same drop-landing protocol, the laboratory, equipment and shoes used were not the same. However, steps were taken to reduce the variability of the two studies. In both studies, cameras used for motion analysis were set up and calibrated to the optimized level, as recommended by the manufacturer (see Appendix B). Verification of force plate accuracy for both studies was also carried out using known weights. Studies have shown the effects of shoes on biomechanical measures, joint energetics and energy dissipation during landing (Yeow *et al.*, 2011, Shultz *et al.*, 2012, Hong *et al.*, 2014). However, to the best of the author's knowledge, there have been no studies that have focused on the effect of shoes prior landing. This could be because shoe effect might be less pronounced without impact forces and friction prior landing. Nevertheless, there might still be a psychological effect of the shoe on how participants prepare for landing. Thus, kinematic variables at initial contact were compared between the two studies and they were not found to be significantly different (see Appendix H). Considering that the differences between these two studies are rather small, the author believes that the benefits of

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increased sample size and hence statistical power resulting from combining data, would outweigh the potential influences due to the confounding factors.

Similarly, male and female participants' data were combined although ACL injury rate for females is higher than males (Agel *et al.*, 2005) and thus landing techniques between genders may be different. Nevertheless, as discussed in Section 5.4.5, kinematic variables were not significantly different between genders (see Appendix I) and knee flexion angles at initial contact might not be a predictor variable of higher ACL injury risks in females (Hewett *et al.*, 2005).

Secondly, the sample size of female participants was smaller than that of male participants. Many female participants did not meet the inclusion and exclusion criteria as many had prior ACL injuries or had irregular menstrual cycles. Most also did not play basketball and female participants playing handball and netball were also invited to the study as landing is a common manoeuvre in these sports. In fact, the second most common manoeuvre to sustain ACL injuries during handball was during single-leg landings (Olsen *et al.*, 2004). For netball, most ACL injuries occur during landing (Stuelcken *et al.*, 2016). Yet, the landing and muscle recruitment strategies might differ among the different sports and this could act as confounding factors in the study. Therefore, a bigger study in future with more female basketball participants would help in a better comparison with the male participants.

Thirdly, the MVC obtained from many participants did not seem to be at the maximal activation level and therefore these data cannot be used for normalization, resulting in many data sets removed from the final analysis. A review by Ball *et al.* (2013) shows that the isometric MVC may not be appropriate for use with high speed dynamic movements. Furthermore, participants are also often not familiar with isometric MVC manoeuvres and this could cause a drop of 20% to 30% in maximal muscle activation unless they are trained (Ball *et al.*, 2013). Despite the removal of several trials, the remaining sample size ( $n = 18$ ) was found to be large enough for the analysis of the EMG across the muscles, extracted from the last 10 ms before initial contact. With the effect size of  $\eta^2 = 0.71$ , observed power was 100%. Furthermore, the trends in the activation levels of the muscles (see Figure 24) were similar to trends observed in the study by Lamontagne *et al.* (2008). However, Lamontagne *et al.* (2008) did not normalize EMG to MVC and EMG levels between muscles cannot be directly compared (Halaki *et al.*, 2012). Results from the study by Morgan *et al.* (2014) also showed significantly less hamstring activation than gastrocnemius and quadriceps activation. Their study only looked at muscle activation levels during the landing phase and not prior initial contact. However, it is crucial to study how muscle activation contributed to the rise of ACL loading prior initial contact with the ground. Thus, despite having less data sets, there was still enough power for EMG analysis to be carried out and trends provided new insights into muscle activation prior initial contact. Nevertheless, when EMG data was split into the low and high knee flexion angle groups, sample sizes were too small to allow meaningful conclusion. Different trends were observed for the gastrocnemius activation in the last 10 ms prior initial contact (see Figure 25) but this should be further validated with bigger sample

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sizes. Future studies could also consider suitable MVC manoeuvres that are more representative of drop-landing for better normalization of muscle activation. This could improve on data quality.

Lastly, landing was carried out under controlled laboratory settings and landing strategies might differ from actual landing conditions. Thus, instructions were provided very carefully to ensure that participants followed the study protocol and yet not vary their natural landing strategies due to prompting by the tester. However, the study could be replicated in manoeuvres closer to landing during sport.

## **5.5 Conclusions**

Results from this study showed that ankle plantar flexion angles at initial contact and internal factors contributed significantly to the prediction of knee flexion angles. Keeping GJL and BMI constant, reduced ankle plantar flexion angles could promote desirable knee flexion angles of more than 15°. To achieve knee flexion angles of at least 15°, ankle plantar flexion angles should not exceed 26° for males and should not exceed 28° for females. Gastrocnemius activation was also found to be significantly higher than hamstrings activation, possibly compensating for reduced hamstring activation during single-leg drop landings. However, the results found in published literature are still unclear on the definite role of the gastrocnemius as agonists or antagonists of the ACL. This was therefore further investigated and results would be presented in the next chapter.

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## Chapter 6 – Effect of Different Gastrocnemius Muscle Activation Levels on Anterior Tibial Translation

### 6.1 Introduction

The ACL plays the primary role in restraining the anterior tibial translation (Hartigan *et al.*, 2011). On the other hand, the hamstring, a knee flexor, decreases ACL strains by creating a reversed, posterior tibial shear force during activation (Draganich *et al.*, 1990). However, hamstring activation becomes less efficient in reducing ACL loading when knee flexion is less than 15° (Li *et al.*, 1999, Yu *et al.*, 2005), often occurring in single-leg landings (Weinhandl *et al.*, 2010). Such low knee flexion angles were also observed in two previous studies that were reported in Chapters 4 and 5 during single-leg landings.

Another knee flexor, the gastrocnemius, has also been a focus of research to examine its role in ACL injuries (Dürselen *et al.*, 1995, Fleming *et al.*, 2001b, Elias *et al.*, 2003, Pflum *et al.*, 2004, Ali *et al.*, 2014, Morgan *et al.*, 2014, Adouni *et al.*, 2016). Morgan *et al.* (2014)'s simulation study using 6 males showed that with the co-contraction of the quadriceps, increased gastrocnemius forces were associated with increased joint compression and reduced ACL forces during single-leg landings. The greater gastrocnemius forces compensated for reduced hamstring forces in opposing the quadriceps, highlighting gastrocnemius' part in reducing ACL injury risk (Morgan *et al.*, 2014). A small simulation study observed a reduction in “peak proximal tibia anterior shear force” with increased gastrocnemius activation, thus supporting the role of the gastrocnemius as an agonist of the ACL (Ali *et al.*, 2014, p. 1602). A cadaver study also

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did not detect a rise in ACL strain during gastrocnemius activation (Dürselen *et al.*, 1995). This further suggests that gastrocnemius activation would not result in increased ACL injury risks.

Yet, small simulation studies by Pflum *et al.* (2004) found a small anterior shear force and Adouni *et al.* (2016) detected an increase in ACL loading under gastrocnemius activation, each using just one sample size. Elias *et al.* (2003) also found significant increase in anterior tibial translation under gastrocnemius activation in an *in vitro* study, at knee flexion angles higher than 20°. A well-cited *in vivo* study by Fleming *et al.* (2001b) also measured an increase in ACL strain with gastrocnemius activation. The authors suggested that the increased ACL strain was due to effects of gastrocnemius bulk on the tibia, pushing the tibia anteriorly upon activation. Effects of such muscle bulk could not have been studied in the previous simulation studies (Ali *et al.*, 2014, Morgan *et al.*, 2014) and cadaver study since muscles were removed (Dürselen *et al.*, 1995). However, Fleming and colleagues (2001b) studied the gastrocnemius activation at only one level. It is possible that sufficient gastrocnemius activation would result in enough joint compression to reduce the effects of the expanding gastrocnemius bulk, therefore reducing anterior tibial translation.

In measuring ACL strain, the differential variable reluctance transducer (Fleming *et al.*, 2001, Lamontagne *et al.*, 2008), the combined use of motion capture with fluoroscopic (Torry *et al.*, 2011) and magnetic resonance imaging techniques (Taylor *et al.*, 2011) have been found in literature. These methods are, however, invasive or very costly.

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Simulation modelling, used by other studies (Pflum *et al.*, 2004, Ali *et al.*, 2014, Morgan *et al.*, 2014, Adouni *et al.*, 2016), is also found to be unsuitable for the studying of muscle bulk expansion effects. On the other hand, ultrasound imaging was found to detect ACL tears with high diagnostic accuracy by measuring anterior tibial translation (Gebhard *et al.*, 1999, Palm *et al.*, 2009). Considering that ultrasound imaging is relatively affordable (Palm *et al.*, 2009) and easy to use without the use of anaesthesia (Suzuki *et al.*, 1991), this technique may be suitable for measuring anterior tibial translation under different muscle activation levels.

This study, therefore, aimed to investigate the effects of different gastrocnemius activation levels on anterior tibial translation, measured using ultrasound imaging. It was hypothesized that anterior tibial translation would increase with low levels of gastrocnemius activation due to the predominant muscle expansion effect, and then decrease at higher activation levels when joint compression effect was greater than muscle expansion effect.

## **6.2 Methods**

### **6.2.1 Participants**

Thirteen male recreational basketball players signed written consent to take part in the study. However, data obtained from 3 of these participants were excluded in the analyses of this study. One dropped out as he could not endure the muscle activation, one had poor ultrasound images that could not be analyzed and the last participant's muscles could not

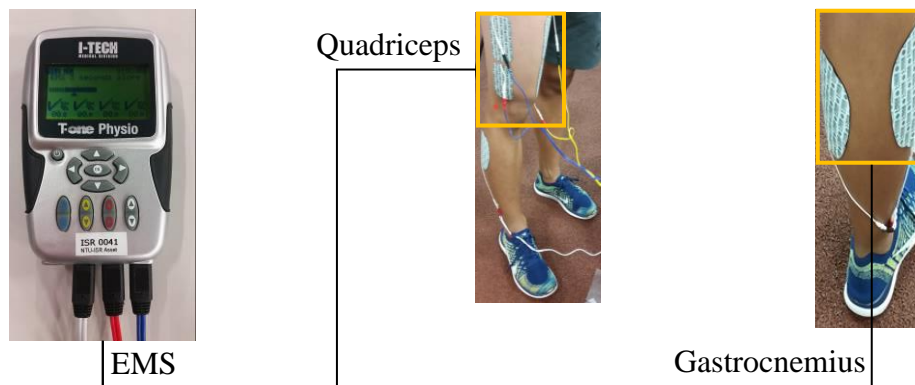
be activated by the muscle activation device probably due to his bigger and heavier build. The remaining 10 participants (age: mean 23.4 (SD 3.8) years old, height: mean 1.78 (SD 0.05) m, body mass: mean 72.4 (SD 6.1) kg) had no inherent health problems, no prior ACL injuries and no lower extremity injuries in the previous 6 months prior to the study. The experiment protocol was approved by the Nanyang Technological University Institutional Review Board.

### **6.2.2 Test Protocol**

The participant was first requested to carry out a 5-min warm-up session, targeting the lower extremity muscles. The exercises were stretching of the hamstrings, quadriceps and calves, tip-toe and ankle rotation, squats and low lunges, jumps, jogging on the spot and jogging with high knees. Electrical muscle stimulation device (EMS) electrodes (T-One Physio, I.A.C.E.R. Srl, Martellago, Italy) were then attached to the quadriceps femoris (vastus lateralis and vastus medialis) and gastrocnemius (gastrocnemius lateral and medial heads) muscles (see Figure 26). Parameters of stimulation were set prior to the study: stimulation frequency of 50 Hz (Doucet *et al.*, 2012) and phase duration of 300  $\mu$ s (Rosso *et al.*, 2014).

The gastrocnemius was activated at four levels: 'low' at the first twitching of the muscle; 'contraction' when toes first began to point down involuntarily; 'mid' at the muscle activation level halfway between 'contraction' and 'high'; and 'high' at each participant's highest tolerable level of muscle activation. Current of the EMS was slowly increased until a level that participants feedback that beyond which, they could no longer endure.

This was defined as the maximum tolerable current setting that participants could endure and would be set as ‘high’ or the maximum gastrocnemius activation level. To produce joint compression, the quadriceps was co-activated at a level that produced 10 Nm of knee extension torque (Fleming *et al.*, 2001b), or at the maximum tolerable level as indicated by the participants. Torque was measured using the isokinetic dynamometer (Humac Norm, CSMI, Massachusetts, USA) as the participant was seated, similar to the method used in Fleming *et al.* (2001b). Knee flexion angle was fixed at 10° and ankle plantar flexion angle was fixed at 30°. This was close to the landing position adopted by participants who landed at less than 15° knee flexion in Chapter 5. The mean current setting in the EMS was 37.6 (SD 10.9) mA for maximum gastrocnemius activation level and was 35.3 (SD 14.8) mA for quadriceps activation level.



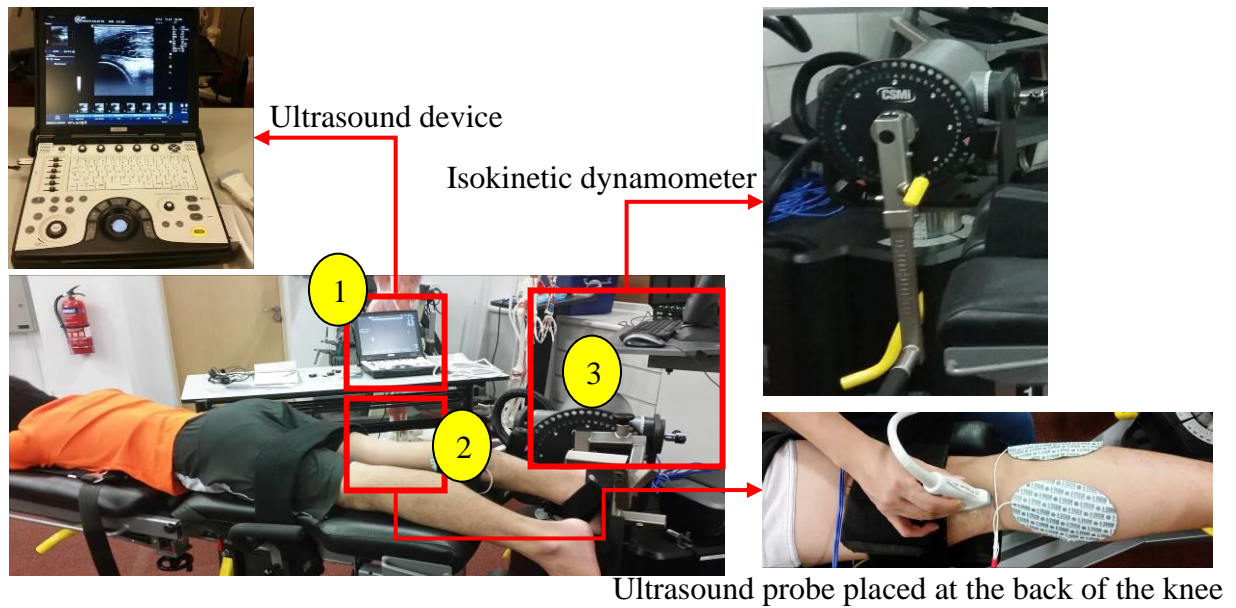
**Figure 26. Positions of electrical muscle stimulation electrode placement.**

The participants then lied in a prone position with the ankle strapped to the isokinetic dynamometer and the thigh strapped to the dynamometer chair during the study (see Figure 27). Knee flexion angle was once again fixed at 10° and ankle plantar flexion

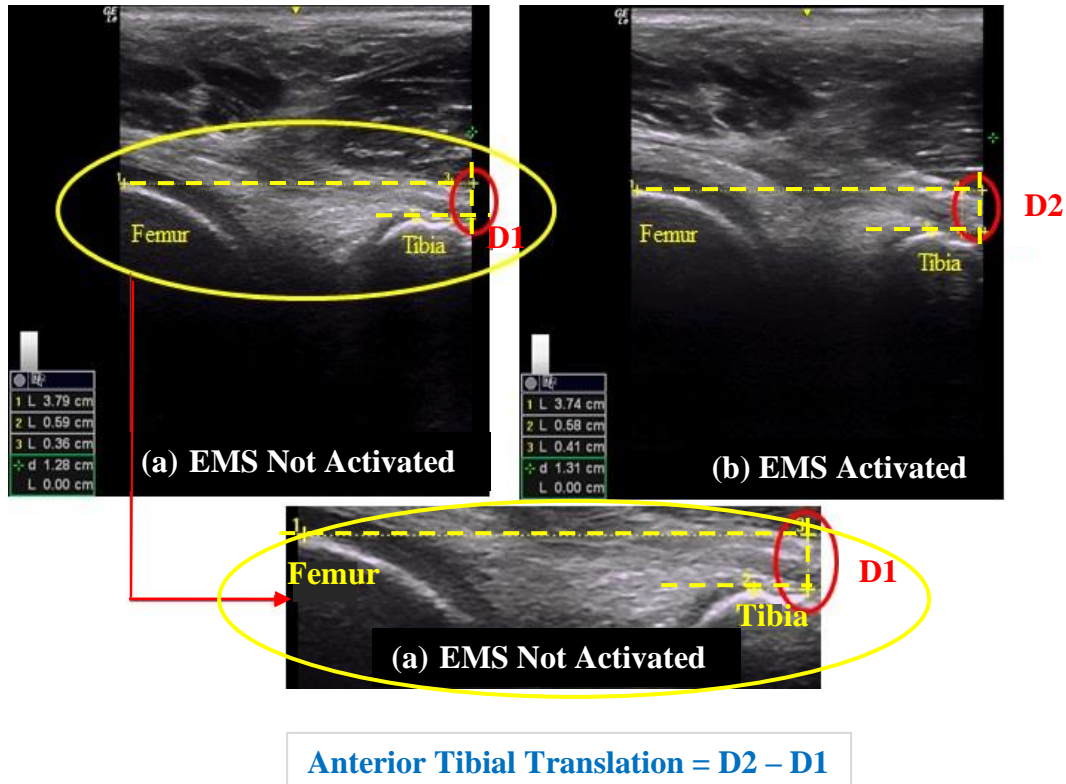
angle was fixed at 30°. Anterior tibial translation was measured using the ultrasound imaging probe set at 12 MHz (LOGIQ e Ultrasound, GE Healthcare, Buckinghamshire, UK), placed at the popliteal fossa at the back of the knee (see Figure 27) (Gebhard et al., 1999, Palm et al., 2009). The distance between the tangent lines to the femoral condyle and the posterior aspect of the tibia was first measured before EMS was activated (labelled as 'D1' in Figure 28). Similarly, the distance between the tangent lines to the femoral condyle and the posterior aspect of the tibia was measured after EMS was activated (labelled as 'D2' in Figure 28). The anterior tibial translation was defined as the difference between D2 and D1 (see Figure 28, Palm *et al.*, 2009).

The current was applied with a 3 s-ramp to the required gastrocnemius stimulation level, maintained for another 4 s to obtain the first ultrasound image. The current was then ramped down in 3 s. Rest duration was 5 s and this stimulation sequence was repeated to the same gastrocnemius activation level and a second ultrasound image was captured. Measurements were then averaged between these two images (data 1). These stimulation steps were carried out from 'low' to 'high' activation sequentially, obtaining an averaged measurement at each level. After which, a rest of at least 5 min was provided. The Borg scale (0-10) was used to monitor the fatigue level of the participants (Borg, 1982). A second session of sequential stimulation following the above-mentioned procedures was then carried out to obtain a second averaged measurement at each stimulation level (data 2). Data 1 and 2 were then averaged across the 2 sessions to obtain the final anterior tibial translation measurement at each gastrocnemius stimulation level. Throughout the simulation, the quadriceps was co-activated at one level, predetermined in the earlier

mentioned procedures. The same ramp, stimulation and rest profiles as the gastrocnemius activation were used.



**Figure 27. Set-up of the study, consisting of the (1) ultrasound device, (2) with the ultrasound probe positioned at the back of the knee and with the participant lying in a prone position on the (3) isokinetic dynamometer chair.**



**Figure 28. Measurement of the anterior tibial translation using ultrasound imaging (EMS refers to the electrical muscle stimulation device).**

### 6.2.3 Statistical Analysis

All statistical analyses were performed using the statistical package SPSS (IBM SPSS 22.0, IBM Corp., NY, USA). Data are expressed as mean (standard deviation).

Intraclass correlation coefficient [ICC (2,1)] was used to assess the intra-rater reliability of the tester in ultrasound image capture over 2 sessions for data 1 and 2. A rest period of at least 5 min was provided between the 2 sessions of sequential muscle activation from ‘low’ to ‘high’. Interpretation of reliability results were based on the following criterion: ICC > 0.75 as good, 0.50 - 0.74 as moderate and < 0.50 as poor (Glaws *et al.*, 2014).

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To compare the effects of the different gastrocnemius activation levels, a one-way repeated measures Analysis of Variance (ANOVA) was conducted at  $\alpha = 0.05$ . Normality assumption was tested with the Shapiro-Wilk test ( $p > 0.05$ ). If normality could not be assumed, "square root" transformation was applied. Assumption of sphericity was tested using the Mauchly's test of sphericity. When the assumption of sphericity was not met, a Greenhouse-Geisser correction was applied. Effect size was calculated as  $\eta^2$  and values of 0.01, 0.09 and 0.25 were defined as small, medium and large effects, respectively (Hanna *et al.*, 2012). Definition of normality, sphericity and effect size are found in Appendix F.

Data were then split into 2 groups, named A and B. A showed a rising trend from 'low' to 'contraction' and then a decreasing trend as activation increased while group B only showed a rising trend across the 4 gastrocnemius activation levels. The Mann-Whitney U test ( $\alpha = 0.05$ ) was conducted to determine if this difference in trends was due to a difference in maximum tolerable gastrocnemius activation levels, quadriceps activation levels or gastrocnemius to quadriceps activation ratios between participants.

### **6.3 Results**

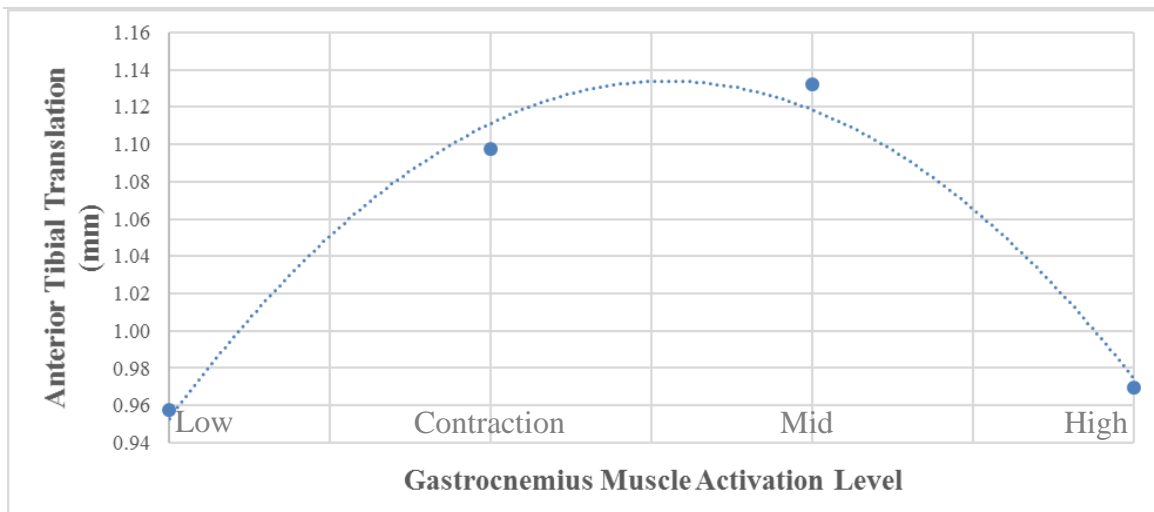
This study investigated the effect of different gastrocnemius activation levels on the anterior tibial translation of male participants. The intra-rater reliability for the tester over 2 sessions was found to be good ( $>0.80$ ) for anterior tibial translation measured before muscle activation, during 'low', 'contraction' and 'mid' gastrocnemius activation levels

but was found to be moderate during ‘high’ gastrocnemius activation level as shown in Table 13.

**Table 13. Comparison of the Intra-Rater Reliability for the Tester over 2 Sessions for the Different Gastrocnemius Muscle Activation Levels**

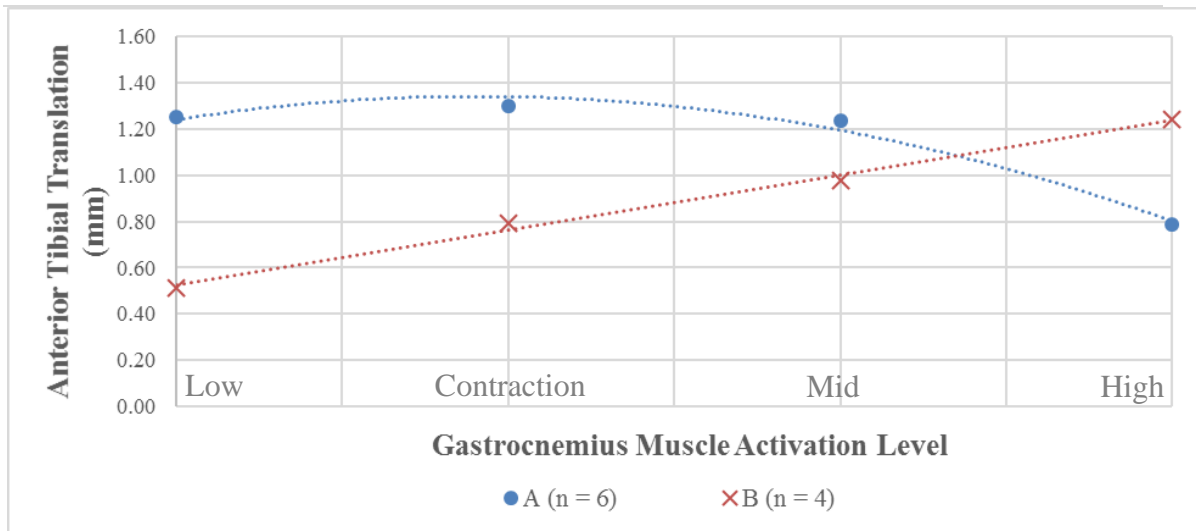
<b>Muscle Activation Level</b>	<b>Intraclass Correlation Coefficient</b>	<b>Interpretation</b>
<b>Before Muscle Activation</b>	0.93	Good
<b>Low</b>	0.89	Good
<b>Contraction</b>	0.87	Good
<b>Mid</b>	0.89	Good
<b>High</b>	0.50	Moderate

Figure 29 shows the non-linear response of the mean anterior tibial translation across all the participants, plotted over each gastrocnemius muscle activation level. Anterior tibial translation increased from ‘low’ to ‘mid’ and then decreased at the level of ‘high’. ANOVA results did not show a significant difference across these 4 levels of muscle activation and effect size was found to be small ( $p = 0.764$ ;  $\eta^2 = 0.04$ ).



**Figure 29. Anterior tibial translation at the four levels of gastrocnemius muscle activation (n = 10).**

Upon further analysis, the participants could be split into two groups since participants exhibited two different trends in terms of anterior tibial translation, over the gastrocnemius muscle activation levels. Group A (n = 6) showed a rising trend from ‘low’ to ‘contraction’ and then a decreasing trend as activation increased. Group B showed a continuous rising trend across the 4 gastrocnemius activation levels (n = 4) (see Figure 30). Participants exhibited a consistent trend within each of the two groups.



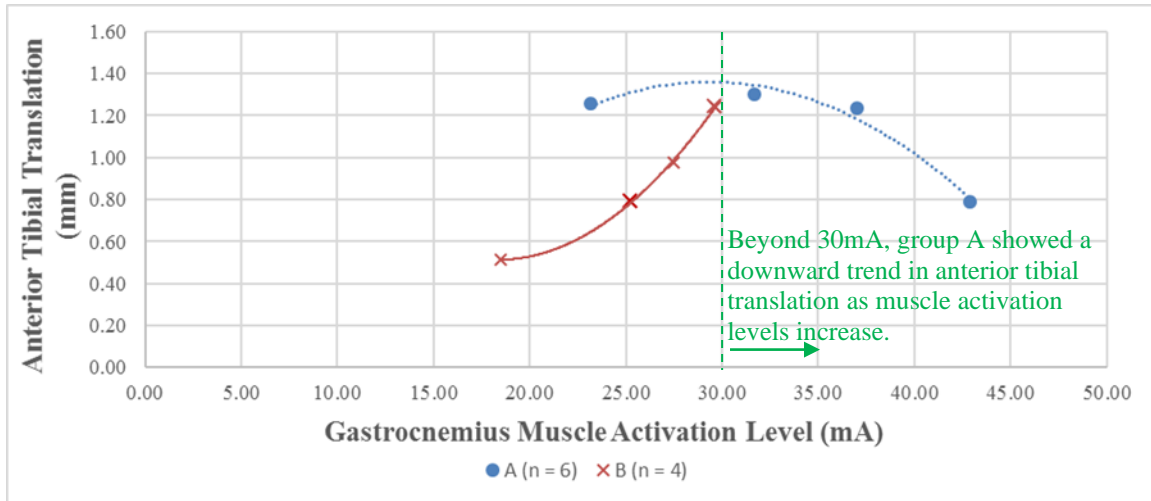
**Figure 30. Difference in anterior tibial translation between groups A and B across the four levels of gastrocnemius muscle activation.**

Focusing on Group A's results, although ANOVA results did not show significant differences across the 4 levels of muscle activation, the effect size was found to be large ( $p = 0.136$ ;  $\eta^2 = 0.30$ ).

Using the Mann-Whitney U test, Groups A and B were found to be significantly different in maximum tolerable gastrocnemius activation level (A: 42.9 (SD 10.2); B: 29.6 (6.3),  $p = 0.038$ ) but not in quadriceps activation level (A: 37.8 (SD 16.3); B: 31.5 (SD 13.4),  $p = 0.762$ ) nor gastrocnemius to quadriceps activation ratio (A: 1.3 (SD 0.4); 1.0 (SD 0.2)),  $p = 0.257$ ). Maximum tolerable gastrocnemius activation level was higher in Group A than Group B, indicating higher activation levels experienced in Group A, as compared to B.

Figure 31 shows the anterior tibial translation of Groups A and B, plotted against the gastrocnemius activation levels (in mA) set in the EMS. Average maximum

gastrocnemius level in Group B (mean: 29.6 (SD 6.3) mA) did not exceed 30 mA but at levels above 30 mA, Group A (mean: 42.9 (SD 10.2) mA) showed a downward trend in anterior tibial translation. The '30 mA' mark is plotted as a green dotted line in Figure 31.



**Figure 31. Difference in anterior tibial translation between groups A and B across the gastrocnemius muscle activation levels set in the electrical muscle activation device (in mA). Beyond 30 mA, group A showed a downward trend in anterior tibial translation.**

## 6.4. Discussion

The aim of this study was to investigate the effects of different gastrocnemius activation levels on anterior tibial translation. As hypothesized, anterior tibial translation increased as the gastrocnemius activation level increased from 'low' to 'mid'; and then decreased as the activation level increased to 'high' level. There are no previous studies found to have investigated the effects of several levels of gastrocnemius activation on anterior tibial translation. This study was also the first to incorporate the use of the ultrasound imaging as a non-invasive device to measure anterior tibial translation.

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### **6.4.1 Effect of Gastrocnemius Muscle Activation Levels on Anterior Tibial Translation**

Gastrocnemius activation resulted in the tibia to move anteriorly to the femur across all muscle activation levels (see Figure 29). This would increase ACL strain, supporting the findings by Fleming *et al.* (2001b) that ACL strain increased with the co-activation of quadriceps and gastrocnemius.

However, anterior tibial translation reduced at higher levels of gastrocnemius muscle activation (see Figure 29), especially for Group A of participants (see Figure 30 and Figure 31). At low gastrocnemius muscle activation levels, the joint compression due to gastrocnemius-quadriceps co-contraction appeared to be not high enough to oppose the anterior tibial translation caused by the gastrocnemius bulk expansion. This may have resulted in an increase in anterior tibial translation. The converse is true at higher gastrocnemius muscle activation levels when greater joint compression effect outweighs the muscle bulk expansion, causing less anterior tibial translation. Unlike the study by Fleming *et al.* (2001b), which only looked at one gastrocnemius activation level, the results in this study suggest that with sufficient gastrocnemius activation, anterior tibial translation could be reduced.

On the other hand, there was a continuous increase in anterior tibial translation across all the gastrocnemius muscle activation levels when considering Group B alone (see Figure 30 and Figure 31).

Comparing Groups A and B, the participants were shown to exhibit two trends of behaviour. Those who could tolerate a higher maximum gastrocnemius activation level were seen to have a rising and then descending trend of anterior tibial translation as current increased (Group A). Those who could tolerate a lower maximum gastrocnemius activation level were seen to be only having a rising trend of anterior tibial translation as current increased (Group B). This could be because the maximum gastrocnemius activation level tolerated was not high enough to counter the effects of muscle bulk expansion for the latter group. Participants exhibited a consistent trend within each of the two groups. These two groups were found to be significantly different in maximum tolerable gastrocnemius activation levels (see ‘Section 6.3 Results’). Figure 31 suggested that anterior tibial translation would start to reduce at levels beyond 30 mA of maximum gastrocnemius activation level. This suggests that a substantial amount of gastrocnemius activation of more than 30 mA may be required to observe the effects of reduced anterior tibial translation, when greater joint compression effect outweighs the muscle bulk expansion.

Focusing on Group A, participants could tolerate a mean current setting of 42.9 mA, closer to levels used by Lenhart *et al.* (2014) ( $\leq 50$  mA) and by Rosso *et al.* (2014) ( $57 \pm 5$  mA), who also used healthy participants. Compared to the study by Rosso *et al.* (2014), which used the same EMS model but seated with the knee flexed at  $90^\circ$ , Group A’s lower maximum tolerable current was expected. Participants were positioned at close to full extension of the knee and ankle, to simulate single-leg landings in this study. Trial studies had shown that intensity levels were less tolerable, at the limits of the range of motion.

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Statistically, Group A's results might not be significantly different across different gastrocnemius muscle activation levels, probably due to the small differences in the actual anterior tibial translation values and small sample size ( $n = 6$ ). However, small changes in anterior tibial translation between conditions were expected, especially for these healthy participants. Anterior tibial translation difference between right and left knees of healthy participants were found to be 0.38 mm and a difference of more than 1 mm was already considered a tear in the ACL (Palm *et al.*, 2009). Furthermore, it is hard for the current to reach the deeper muscle structures when electrodes are positioned on the skin for the EMS (Doucet *et al.*, 2012). Thus, muscle activation may be reduced, as compared to that produced by actual muscles, resulting in less anterior tibial translation values. Despite the small differences in anterior tibial translation, effect size was large for Group A (see 'Section 6.3 Results'). This further suggested that sufficient gastrocnemius activation in Group A could result in a joint compression effect that might be large enough to limit a rise of anterior tibial translation. This effect was practically significant.

In summary, although gastrocnemius activation resulted in anterior tibial translation, thereby increasing ACL strain, substantial amount of gastrocnemius activation of more than 30 mA could limit this increase in anterior tibial translation.

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### **6.4.2 Use of the Ultrasound Imaging in the Measurement of Anterior Tibial Translation**

This study employed a novel method of using non-invasive ultrasound imaging to measure anterior tibial translation under muscle activation.

Ultrasound imaging used in this study demonstrated good intra-rater reliability for anterior tibial translation measured before muscle activation, during ‘low’, ‘contraction’ and ‘mid’ gastrocnemius activation levels. This suggests that ultrasound imaging is a suitable instrument to measure anterior tibial translation. However intra-rater reliability was found to be moderate during ‘high’ gastrocnemius activation level. The moderate ICC result for the condition of ‘high’ could be due to the difficulty of obtaining sharper images when there was a sudden bone shift under the higher muscle activation.

Considering that the tester does not have extensive experience in ultrasound imaging, the good ICC values for most part of the study suggest that ultrasound could be relatively easy to use as a non-invasive measurement tool. For the lower muscle activation settings, tibial translation was more gradual and the ultrasound probe could easily be adjusted to obtain sharp images. Using ultrasound imaging to detect anterior tibial translation under more sudden movements is therefore difficult and requires more experience.

In summary, it was found that ultrasound imaging is suitable for use in future studies to measure anterior tibial translation under muscle activation. It is relatively affordable

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(Palm *et al.*, 2009), easy to use and demonstrates high intra-rater reliability. However, for ‘high’ muscle activation levels, more experience in the usage of ultrasound imaging is required.

### **6.4.3 Limitations**

A major limitation to this study was that muscle activation levels were not randomised. A gradual muscle activation from ‘low’ to ‘high’ was necessary so that the participant would not go into sudden shock. Sudden movements would not be easy for ultrasound imaging to capture and this is difficult to measure anterior tibial translation. However, fatigue effects might be experienced as the study progressed. To minimise the effects of fatigue, the participant was given ample rest as needed. In addition, the Borg scale was used to monitor the fatigue levels throughout the study protocol (Borg, 1982). EMS settings were based on proposed guidelines (Doucet *et al.*, 2012) and constraints in this study. As this was the first study that activated the gastrocnemius under a few muscle activation levels, the general EMS setting guidelines were yet to be established to be optimal and relevant. Although settings, such as EMS activation levels and rest intervals, were also optimized during the pilot study, setting optimization could be further refined and improved in future studies.

A second limitation was that the generalized joint laxity (GJL) and body weight were not considered in the inclusion or exclusion criteria of the study. GJL may be related to knee mobility and may have influenced the tibial movement induced by muscle activation. Body weight can affect the mass and moment inertia of the calf. For a given muscle

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activation level, less tibial movement can be expected for individuals with a heavier and/or longer lower limb. For instance, one participant's data had to be removed as his muscles could not be sufficiently activated by the muscle activation device to induce noticeable anterior tibial translation. The lack of tibial movement may be due to his heavier leg and/or reduced knee mobility. Future studies should monitor generalized joint laxity and the range of body weight of the participants to improve the study outcome.

Another limitation was that study results could not be related back to actual landing situations. EMG was used to measure muscle activation during the pilot study but the signals were distorted when the EMS was switched on. Torque measurement was taken during the study using the isokinetic dynamometer. However, the torque levels produced by the EMS were not sufficiently high to be detected by the dynamometer. This could also be because joint torque could not be accurately measured by the dynamometer due to reasons such as damping of the torque by the paddings in the fixtures. A customized foot fixture to measure torque, similar to the one used by Fleming *et al.* (2001b), could be made for the purpose of the study. Better strap attachments should be included to minimize any leg movements resulting from EMS activation.

Lastly, participants were requested to relax and not resist their muscles during the study. However, there is a possibility that some may experience involuntary resistance of muscles. To minimise this, the protocol was designed to generally increase the muscle activation gradually, especially at the beginning of the study so that participants could adapt to the activation. Alternatively, participants can be anesthetized during the study

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(Fleming *et al.*, 2001b). However, this may not be desired and extra care has to be taken to prevent muscle damage as participants could no longer have any sensations. They would therefore not be able to provide feedback on whether there was pain or whether they would need additional rest.

## **6.5. Conclusion**

This study examined the effect of varying gastrocnemius muscle activation levels on anterior tibial translation. Results showed a non-linear response that anterior tibial translation initially increased at lower levels of gastrocnemius levels, followed by a decrease at higher levels of activation. This suggests that with sufficient joint compression using gastrocnemius activation, anterior tibial translation could be reduced, thereby limiting ACL strain.

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## Chapter 7 – Overall Discussion

This thesis comprised of four studies and mainly focused on studying the influence of foot landing and muscle activation on ACL injury risk reduction during single-leg drop landing. Although there are studies that specifically vary foot landing positions during dynamic tasks to study its effects on the lower extremity (Kovács *et al.*, 1999, Cortes *et al.* 2007, Ishida *et al.* 2014, Valenzuela *et al.*, 2016), little has been reported on how these effects could translate to reducing non-contact ACL injury risks. This thesis also focused on single-leg landings to provide possible reasons into the higher instances of non-contact ACL injuries that could occur during single-leg landings (Donnelly *et al.*, 2012). With the study of the muscle activation of the gastrocnemius, hamstrings and quadriceps, focusing on the time before initial contact with the ground, this study also aimed to better understand why ACL strain could be the highest at initial contact (Lamontagne *et al.*, 2008). Internal ACL injury risk factors (generalized joint laxity (GJL), body mass index (BMI)) were also included in the studies, as these could be contributory factors of higher non-contact ACL injury risks.

### 7.1 Summary of Experimental Results

In Chapter 3, effects of foot rotation positions (toe-in, toe-forward, toe-out) on knee valgus were investigated, both during initial contact and during landing. This study showed that foot rotation positions can influence knee joint loading during single-leg drop landing. Particularly, landing in toe-out foot positions resulted in significantly higher maximum KVM, as compared with toe-in and toe-forward positions. Although

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foot rotation positions also influenced knee valgus angles (KVA) at initial contact, the differences were too small to be clinically meaningful. Since this thesis focused on the pre-landing phase before initial contact, foot rotation effects on KVA at initial contact was therefore not further studied in subsequent chapters. Participants were also found to land in negative, toe-out positions and foot rotation angles became increasingly more negative (away from the midline of the body) from toe-in to toe-forward and then toe-out positions. This observation suggests that toe-out foot rotation is likely to be a natural landing position of these participants.

In Chapter 4, knee and ankle kinematics were compared among landing techniques in the sagittal plane, landing at natural, fore-foot and flat-foot landings. Participants were found to land with fore-foot landings during natural landings. While fore-foot landing positions are often encouraged for greater force absorption, in agreement with the study by Alentorn-Geli *et al.*(2009a), participants were found to land at low knee flexion angles of less than 15° during natural and fore-foot landings. This low knee flexion angles were associated with increased non-contact ACL injury risks (Lin *et al.*, 2008). Less ankle plantar flexion angles seemed to result in higher knee flexion angles during flat-foot landing. However, results in this study showed that natural or fore-foot landings are still better for force absorption, as compared to flat-foot landings. Thus, there seemed to be an optimal range of ankle plantar-flexion angles that could be associated with higher knee flexion angles, leading to lower non-contact ACL injury risks. This was further investigated in Chapter 5.

Chapter 5 investigated how ankle plantar flexion angles, GJL and BMI influence knee flexion angles at initial contact during single-leg drop landing. Muscle activation levels among gastrocnemius, hamstring and quadriceps, prior initial contact were also compared. Results showed that ankle plantar flexion angles at initial contact and internal factors contributed significantly to the prediction of knee flexion angles. Keeping GJL and BMI constant, reduced ankle plantar flexion angles could promote desirable knee flexion angles of more than 15°. To achieve knee flexion angles of at least 15°, ankle plantar flexion angles should not exceed 26° for males and should not exceed 28° for females. These were based on mean GJL and BMI data of male and female participants obtained in this study. Gastrocnemius activation was also found to be significantly higher than hamstrings activation, possibly compensating for reduced hamstring activation during single-leg drop landings. However, there are still uncertain results regarding the role of the gastrocnemius as agonists or antagonists of the ACL as reported in the literature. It is possible that sufficient gastrocnemius activation would result in enough joint compression to reduce the effects of the expanding gastrocnemius bulk, thereby reducing anterior tibial translation. Therefore, the effect of gastrocnemius activation on ACL injury risks was further investigated in Chapter 6.

In Chapter 6, the effects of different gastrocnemius activation levels on anterior tibial translation were examined, to determine if gastrocnemius activation do increase ACL injury risks. Results showed a non-linear response that anterior tibial translation initially increased at lower levels of gastrocnemius levels, followed by a decrease at higher levels of activation. This suggests that with sufficient joint compression using gastrocnemius

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activation of more than 30 mA and with quadriceps' co-activation, anterior tibial translation could be reduced, thereby lowering ACL injury risks.

## **7.2 Discussion and Summary of the Studies**

This PhD research investigated the influence of foot landing techniques on biomechanical measures associated with higher non-contact ACL injury risks. This contributed to new understanding for ACL injury risk reduction during single-leg landings, especially at initial contact with the ground, when peak ACL strain occurs (Lamontagne *et al.*, 2008). A new relationship between ankle plantar flexion angles and knee flexion angles was also examined in this research. As a result, an optimal range of ankle plantar flexion angles could thus be recommended for safer fore-foot landings, associated with less non-contact ACL injury risks. This research also helped to clarify the role of muscle activation, particularly the gastrocnemius, in non-contact ACL injury risk reduction. All these findings could contribute to the better understanding of preventing non-contact ACL injuries for healthy athletes.

### **i. Proposed New Strategy to ACL Injury Risk Reduction using Foot Landing**

Foot landing techniques has an influence on non-contact ACL injury risks. To reduce KVM during the landing phase, toe-in and toe-forward, rather than toe-out positions should be adopted. When focusing on the pre-landing phase before initial contact with the ground, altering ankle plantar flexion angles would be a more promising strategy to increasing knee flexion angles, thereby reducing non-contact ACL injury risks. Ankle plantar flexion angles at initial contact have a significant effect on knee flexion angles based on the binomial logistic regression

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models. A reduction of ankle plantar flexion angles increases the chance of having higher knee flexion angles of more than 15°. Including outliers, the prediction model using just ankle plantar flexion angles could already explain at least 30% of the variance of knee flexion angles. When the 2 outliers were removed, the models explained almost 70% of the variance of knee flexion angles. This suggests that for majority of the participants (at least 95%), controlling ankle flexion angles could contribute substantially to increasing knee flexion angles at initial contact. Controlling ankle plantar flexion angles would be a direct and impactful strategy used during training to teach athletes how increased knee flexion angles could be achieved. It would otherwise be more difficult to ask athletes to land subjectively using ‘soft’ landings, associated with higher knee flexion angles (Devita *et al.*, 1992, Laughlin *et al.*, 2011), especially during competitions under stress.

**ii. New Relationship between Ankle Plantar Flexion Angles and Knee Flexion Angles was Examined**

The multiple regression models relating ankle plantar flexion angles with knee flexion angles were examined in this research and similar studies have not been found in literature. Optimal range of ankle plantar flexion angles leading to knee flexion angles higher than 15° of knee flexion angles at initial contact was proposed. This proposed optimal range could be incorporated into ankle brace or shoe designs used for training or during competition to limit the ankle plantar flexion range of motion that are associated with low knee flexion angles.

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The group of participants who landed with less than 15° of knee flexion angles at initial contact, were also found to land at high ankle plantar flexion angles (31°, 95% CI [29 to 33] using data from the male participants and 32°, 95% CI [31 to 33] using combined data of the male and female participants). The ankle plantar flexion angles are higher than the predicted optimal range of ankle plantar flexion angles of maximum 26° for males and 28° for females, derived in this study.

These multiple regression models account for 45% of the variation of knee flexion angles, a large size effect according to Cohen (1988). Almost half of the participants in this thesis landed with less than 15° of knee flexion angles at initial contact, a level at which hamstrings are less efficient to counter the anterior pull of the quadriceps (Li *et al.*, 1999, Yu *et al.*, 2005). If this sample of 47 participants were representative of the population, almost half of the population could benefit from landing within the optimal range of ankle plantar flexion angles, thereby reducing non-contact ACL injury risks.

### **iii. New Findings to the Fundamental Knowledge of ACL Injury Risks in Single-Leg Drop Landings**

- Players have been encouraged to land with fore-foot landings (Alentorn-Geli *et al.*, 2009a) but there are no recommended optimal range of ankle plantar flexion angles that they should land in. While this thesis supports that fore-foot landings do help in greater force absorption, the results show that excessive ankle plantar flexion angles could in turn reduce knee

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flexion angles to the point that it increases non-contact ACL injury risks during single-leg drop landings.

- Hamstring-to-quadriceps strengthening have also been incorporated in training and prevention programmes (Shultz, 2007a) but gastrocnemius-to-quadriceps strengthening could also be beneficial and essential. This thesis proposes that gastrocnemius also plays an important role during single-leg drop landings when the knees are more extended. Significantly higher gastrocnemius activation than hamstring activation occurred 10 ms before initial contact. The gastrocnemius, both a plantar flexor and knee flexor, is less of a knee flexor when the ankle is plantar flexed (Hartigan *et al.*, 2011). Thus, at the optimal range of ankle plantar flexion angles recommended in this thesis, the gastrocnemius could act as a knee flexor until the hamstrings take a more effective role as knee flexion angles increase.
- This thesis provides new insights into the current arguments regarding the role of the gastrocnemius in ACL injury risks. Although gastrocnemius activation could lead to anterior tibial translation, anterior tibial translation could be reduced should enough gastrocnemius activation and joint compression are initiated.
- This thesis may partly explain why higher non-contact ACL injury risks are found in single-leg, rather than during double-leg landings. As discussed in Chapter 5, lower knee flexion angles occur during single-leg landings, leading to the inefficacy of the hamstring to oppose the

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quadriceps. Excessive ankle plantar flexion could also result in the reduction of the role of the gastrocnemius as the alternative knee flexor. All these could lead to higher non-contact ACL injury risks during single-leg landings. The study in Chapter 5 also showed that there was a continual rise in quadriceps activation during the pre-landing phase of 100 ms before initial contact while the hamstring activation rose slightly and then tapered off. This suggests that hamstring activation may not be sufficient to counter the quadriceps activation during single-leg landings. Lower knee flexion angles also further increase ACL strain by increasing the patella tendon-tibia shaft angle, as well as, ACL elevation angle and deviation angle (Li *et al.*, 2005, Yu *et al.*, 2007). All these could result in a net increase in anterior tibial translation, thereby contributing to the increase of ACL strain to the maximum at initial contact. Therefore, with less knee flexion angles, single-leg landings could possibly result in increased non-contact ACL injury risks, as compared to double-leg landings. Pre-landing phase before initial contact with the ground should not be neglected in future studies looking at ACL injury risks.

**iv. Novel Methodology using the Ultrasound Imaging to Quantify Anterior Tibial Translation**

Current methodologies such as the differential variable reluctance transducer to measure ACL strain is invasive and thus limits its use. The novel method of using ultrasound imaging was explored and incorporated in this thesis to measure anterior tibial translation during muscle activation. An alternative method that

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uses simulation modelling would otherwise not be able to see the effects of muscle bulk expansion. Results show good intra-rater reliability under all the conditions, except during the ‘high’ activation condition. Although the current ultrasound imaging technology may not be used under dynamic conditions such as during landing, this study marks the start of exploring the use of ultrasound imaging in future studies to examine ACL injury risks.

Together, this thesis demonstrated new findings to reduce non-contact ACL injury risks during single-leg drop landing from the foot landing and muscle activation prospective. Data obtained from this study could be further incorporated into training and preventive ankle brace solutions.

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## Chapter 8 – Conclusion and Future Work

### 8.1 Conclusion

This study successfully investigated foot landing techniques and muscle activation during single-leg drop landings, to reduce non-contact ACL injury risks. Foot landing techniques has an influence on non-contact ACL injury risks.

Foot rotation positions can influence knee valgus during single-leg drop landing. Landing in toe-out foot positions resulted in significantly higher maximum knee valgus moments, as compared with toe-in and toe-forward positions. Although foot rotation positions also influenced knee valgus angles at initial contact, the differences were too small to be practically significant. Athletes should therefore avoid extreme toe-out foot rotation positions to minimize maximum knee valgus moments.

Fore-foot landings are still better than flat-foot landings in terms of force absorption. However, there is an optimal range of ankle plantar flexion angles that are associated with increased knee flexion angles, thereby reducing non-contact ACL injury risks. A relationship was examined between knee flexion angles and three independent factors, namely ankle plantar flexion angles, generalized joint laxity and body mass index. Both knee flexion and ankle plantar flexion angles were measured at initial contact with the ground. A reduction of ankle plantar flexion angles increases the chance of having higher knee flexion angles of more than 15°. Keeping generalized joint laxity and body mass index constant, ankle plantar flexion angles at initial contact should not exceed 26° to 28°

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to achieve knee flexion angles of at least 15°. These were based on mean generalized joint laxity and body mass index data of male and female participants obtained in this study.

Gastrocnemius was also found to be significantly more activated, as compared to hamstrings during the pre-landing phase before initial contact, possibly to compensate for reduced hamstring activation. Disagreements regarding the role of the gastrocnemius in ACL injury risks were clarified. Results suggest that with sufficient gastrocnemius activation and therefore joint compression, anterior tibial translation could be reduced, thereby limiting ACL injury risks.

## **8.2 Future Work**

The following should be considered for future work to extend the knowledge of foot landing techniques and muscle activation effects on non-contact ACL injury risk reduction:

1. A similar study could be conducted using manoeuvres closer to landing during different sports. This could be used to further investigate if the relationship between ankle plantar flexion angles and knee flexion angles at initial contact would change with landing during the playing of different sport. The simultaneous manipulation of sagittal and frontal plane foot landing positions could also be conducted to study the effect of its interaction on the knee positions and loading.

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2. Future studies could also examine suitable maximal voluntary isometric contractions manoeuvres that are more representative of drop-landing for better normalization of muscle activation. Following that, a larger scale study on muscle activation could be conducted to further investigate how low and high knee flexion angle groups differ in muscle activation prior initial contact with the ground.
  3. The use of ultrasound imaging could be further explored, measuring ACL strain using the ultrasound elastography method (Cochran *et al.*, 2013). The results in the study by Li *et al.*, (1999) suggested that the greatest anterior tibial translation might not necessarily coincide with the highest ACL loading. Thus, a comparison of results could be made between the anterior tibial translation obtained from this thesis and ACL strain using the ultrasound elastography method to study if similar results were obtained.
  4. A training intervention could be conducted on a group of participants to examine the long-term effect of landing in the recommended range of ankle plantar flexion angles. A prospective cohort could be followed over a few years together with a control group to find out if the training intervention helps in reducing the injury rates of non-contact ACL injury risks.
  5. An ankle brace could be designed based on the recommended optimal range of ankle plantar flexion angles. The ankle brace design should limit the foot from reaching excessive ankle plantar flexion angles associated with small knee flexion angles. To prevent athletes from feeling restricted by wearing ankle braces in the range of motion of ankle plantar flexion angles that is not associated with higher

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non-contact ACL injury risks, the ankle brace should allow little resistance at these angles. The ankle brace could be used during training until athletes are used to landing in the recommended range of ankle plantar flexion angles. It could also be used during competition to prevent athletes from reaching excessive ankle plantar flexion angles that are associated with low knee flexion angles. Lace-up ankle braces have been used to prevent injury (Shaw *et al.*, 2008) and athletes might accept the wearing of ankle braces during competitions more readily than the wearing of knee braces.

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## List of Publications

### Journals:

1. Teng PSP, Leong KF, Kong PW, Halkon BJ, Huang P. The use of rapid prototyping in the design of a customised ankle brace structure for ACL injury risk reduction. *Virtual and Physical Prototyping* 2013; 8(4): 241-247.
2. Bai J, Teng PSP, Gu KY, Leong KF, Kong PW. Effects of single-leg foot-landing techniques on ground reaction force and electromyography. *Chinese Journal of Sports Biomechanics* 2014; 10(S1): 69.
3. Teng PSP, Kong PW, Leong KF. Effects of foot rotation positions on knee valgus during single-leg drop landing: implications for ACL injury risk reduction. *Knee* 2017; 24(3): 547-554.

### Papers Submitted:

1. Teng PSP, Kong PW, Leong KF, Phua PYX. Effect of different gastrocnemius muscle activation levels on anterior tibial translation measured using ultrasound imaging (Submitted to 'Journal of Applied Biomechanics').

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**Conferences:**

1. Teng PSP, Leong KF, Huang P, McLaren J. The effect of a knee-ankle restraint on ACL injury risk reduction during jump-landing. In: Subic A, Fuss FK, Clifton P, Chan KM, editors. The impact of technology on sport V. Proceedings of the 6th Asia-Pacific Congress on Sports Technology; 2013 Sep 18-20; Hong Kong. Melbourne: RMIT University; 2013. p. 300–306.
2. Bai J, Teng PSP, Gu KY, Leong KF, Kong PW. Effects of single-leg foot-landing techniques on ground reaction force and electromyography. 5th Conference of Asian Society of Sports Biomechanics; 2014 Sep 2-4; Taipei, Taiwan.
3. Teng PSP, Kong PW, Leong KF, Gu KY, Bai J. Effects of foot-landing techniques on lower extremity kinematics during single-leg drop landings. 1st International Conference in Sports Science and Technology; 2014 Dec 11-12; Singapore.
4. Teng PSP, Kong PW, Phua PYX, Leong KF. Effect of different gastrocnemius muscle activation levels on anterior tibial translation – a case study. 2nd International Conference in Sports Science and Technology; 2016 Dec 12-13; Singapore.

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## References

Aagaard P, Simonsen EB, Andersen JL, Magnusson P, Dyhre-Poulsen P. Increased rate of force development and neural drive of human skeletal muscle following resistance training. *J Appl Physiol* 2002; 93(4): 1318-1326.

Abián J, Alegre LM, Lara, AJ, Rubio JA, Aguado X. Landing differences between men and women in a maximal vertical jump aptitude test. *J Sports Med Phys Fitness* 2008; 48(3): 305-310.

Adouni M, Shirazi-Adl A, Marouane H. Role of gastrocnemius activation in knee joint biomechanics: gastrocnemius acts as an ACL antagonist. *Comput Methods Biomech Biomed Engin* 2016; 19(4): 376-385.

Agel J, Arendt EA, Bershadsky B. Anterior cruciate ligament injury in national collegiate athletic association basketball and soccer: a 13-year review. *Am J Sports Med* 2005; 33(4): 524-530.

Alentorn-Geli E, Myer GD, Silvers HJ, Samitier G, Romero D, Lázaro-Haro C, *et al.* Prevention of non-contact anterior cruciate ligament injuries in soccer players. Part 1: mechanisms of injury and underlying risk factors. *Knee Surg Sports Traumatol Arthrosc* 2009a; 17(7): 705-729.

Alentorn-Geli E, Myer GD, Silvers HJ, Samitier G, Romero D, Lázaro-Haro C, *et al.* Prevention of non-contact anterior cruciate ligament injuries in soccer players. Part 2: a review of prevention programs aimed to modify risk factors and to reduce injury rates. *Knee Surg Sports Traumatol Arthrosc* 2009b; 17(8): 859-879.

Ali N, Andersen MS, Rasmussen J, Robertson DG, Rouhi G. The application of musculoskeletal modeling to investigate gender bias in non-contact ACL injury rate during single-leg landings. *Comput Methods Biomech Biomed Engin* 2014; 17(14): 1602-1616.

Ali N, Rouhi G. Barriers to predicting the mechanisms and risk factors of non-contact anterior cruciate ligament injury. *Open Biomed Eng J* 2010; 4: 178-189.

Ali N, Rouhi G, Robertson G. Gender, Vertical height and horizontal distance effects on single-leg landing kinematics: implications for risk of non-contact ACL injury. *J Hum Kinet* 2013; 37: 27-38.

Allen MK, Glasoe WM. Metrecom measurement of navicular drop in subjects with anterior cruciate ligament injury. *J Athl Train* 2000; 35(4): 403-406.

Anderson AF, Dome DC, Gautam S, Awh MH, Rennirt GW. Correlation of anthropometric measurements, strength, anterior cruciate ligament size, and intercondylar

notch characteristics to sex differences in anterior cruciate ligament tear rates. *Am J Sports Med* 2001; 29(1): 58-66.

Arnheim DD, Prentice WE. The knee and related structures. In: *Principles of athletic training*, St. Louis: Mosby-Year Book; 1993, p. 528 - 584.

Bahr R, Krosshaug T. Understanding injury mechanisms: a key component of preventing injuries in sport. *Br J Sports Med* 2005; 39(6): 324-329.

Ball N, Scurr J. Electromyography normalization methods for high-velocity muscle actions: review and recommendations. *J Appl Biomech* 2013; 29(5): 600-608.

Beckett ME, Massie DL, Bowers KD, Stoll DA. Incidence of hyperpronation in the ACL injured knee: a clinical perspective. *J Athl Train* 1992; 27(1): 58-62.

Bell AL, Brand RA. Prediction of hip joint centre location from external landmarks. *Hum Mov Sci* 1989; 8(1): 3-16.

Berns GS, Hull ML, Patterson HA. Strain in the anteromedial bundle of the anterior cruciate ligament under combination loading. *J Orthop Res* 1992; 10(2): 167-176.

Bjordal JM, Arnøy F, Hannestad B, Strand T. Epidemiology of anterior cruciate ligament injuries in soccer. *Am J Sports Med* 1997; 25(3): 341-345.

Bollen S. Epidemiology of knee injuries: diagnosis and triage. *Br J Sports Med* 2000; 34(3): 227-228.

Borg GAV. Psychophysical bases of perceived exertion. *Med Sci Sports Exerc* 1982; 14(5): 377-381.

Boros R, Plumlee E. Effects of ankle immobilization on knee joint biomechanics during an unanticipated cutting maneuver. The 34<sup>th</sup> Annual Meeting of the American Society of Biomechanics; 2010 Aug 18-21; Providence, Rhode Island, USA.

Borotikar BS, Newcomer R, Koppes R, McLean SG. Combined effects of fatigue and decision making on female lower limb landing postures: central and peripheral contributions to ACL injury risk. *Clin Biomech (Bristol, Avon)* 2008; 23(1): 81-92.

Brownstein B. Migration and design characteristics of functional knee braces. *J Sport Rehabil* 1998; 7(1): 33-43.

Chappell JD, Creighton RA, Giuliani C, Yu B, Garrett WE. Kinematics and electromyography of landing preparation in vertical stop-jump: risks for noncontact anterior cruciate ligament injury. *Am J Sports Med* 2007; 35(2): 235-241.

Chappell JD, Herman DC, Knight BS, Kirkendall DT, Garrett WE, Yu B. Effect of fatigue on knee kinetics and kinematics in stop-jump tasks. *Am J Sports Med* 2005; 33(7): 1022-1029.

Chappell JD, Yu B, Kirkendall DT, Garrett WE. A comparison of knee kinetics between male and female recreational athletes in stop-jump tasks. *Am J Sports Med* 2002; 30(2): 261–267.

Chen P-T, Wu C-H, Yu C-W, Wang J-H, Shih T-F, Wang T-G, *et al.* Sonography of the normal anterior cruciate ligament: a preliminary report. *J Med Ultrasound* 2013; 21(1): 16-20.

Cinthuja P, Jayakody JAOA, Perera MPM, Weerarathna WVDN, Nirosha SE, Indeewari DKDC, *et al.* Physical fitness factors of school badminton players in Kandy district. *Euro J Sports Exerc Sci* 2015; 4(2): 14-25

Clippinger KS. The knee and patellofemoral joints. In: *Dance anatomy and kinesiology*, Illinois: Human Kinetics; 2016, p. 163–200.

Cochran AL, Gao Y. An ultrasound elastography method for examining the anterior cruciate ligament. *Natural Science* 2013; 5(8): 23 -31.

Cochrane JL, Lloyd DG, Butfield A, Seward H, McGivern J. Characteristics of anterior cruciate ligament injuries in Australian football. *J Sci Med Sport* 2007; 10(2): 96-104.

Cohen J. *Statistical power analysis for the behavioral sciences*. 2nd ed. Hillsdale: Lawrence Erlbaum Associates; 1988.

Cortes N, Onate J, Abrantes J, Gagen L, Dowling E, Van Lunen B. Effects of gender and foot-landing techniques on lower extremity kinematics during drop-jump landings. *J Appl Biomech* 2007; 23(4): 289-299.

Cram JR. *Cram's introduction to surface electromyography*. 2nd ed. Criswell E, editor. Sudbury: Jones and Bartlett Publishers; 2011.

Croce RV, Russell PJ, Swartz EE, Decoster LC. Knee muscular response strategies differ by developmental level but not gender during jump landing. *Electromyogr Clin Neurophysiol* 2004; 44(6): 339–348.

Decker MJ, Torry MR, Wyland DJ, Sterett WI, Richard Steadman J. Gender differences in lower extremity kinematics, kinetics and energy absorption during landing. *Clin Biomech (Bristol, Avon)* 2003; 18(7): 662-669.

de Loës M, Dahlstedt LJ, Thomée R. A 7-year study on risks and costs of knee injuries in male and female youth participants in 12 sports. *Scand J Med Sci Sports* 2000; 10(2): 90-97.

DeMorat G, Weinhold P, Blackburn T, Chudik S, Garrett W. Aggressive quadriceps loading can induce noncontact anterior cruciate ligament injury. *Am J Sports Med* 2004; 32(2): 477-483.

Dempsey AR, Elliott BC, Munro BJ, Steele JR, Lloyd DG. Whole body kinematics and knee moments that occur during an overhead catch and landing task in sport. *Clin Biomech (Bristol, Avon)* 2012; 27(5): 466-474.

Devita P, Skelly WA. Effect of landing stiffness on joint kinetics and energetics in the lower extremity. *Med Sci Sports Exerc* 1992; 24(1): 108-115.

DiStefano LJ, Padua DA, Brown CN, Guskiewicz KM. Lower extremity kinematics and ground reaction forces after prophylactic lace-up ankle bracing. *J Athl Train* 2008; 43(3): 234-241.

Donnelly CJ, Elliott BC, Ackland TR, Doyle TLA, Beiser TF, Finch CF, *et al.* An anterior cruciate ligament injury prevention framework: incorporating the recent evidence. *Res Sports Med* 2012; 20(3-4): 239-262.

Doucet BM, Lam A, Griffin L. Neuromuscular electrical stimulation for skeletal muscle function. *Yale J Biol Med* 2012; 85(2): 201-215.

Draganich LF, Vahey JW. An in vitro study of anterior cruciate ligament strain induced by quadriceps and hamstrings forces. *J Orthop Res* 1990; 8(1): 57-63.

Dragoo JL, Braun HJ, Durham JL, Chen MR, Harris AHS. Incidence and risk factors for injuries to the anterior cruciate ligament in national collegiate athletic association football: data from the 2004-2005 through 2008-2009 national collegiate athletic association injury surveillance system. *Am J Sports Med* 2012; 40(5): 990-995.

Dürselen L, Claes L, Kiefer H. The influence of muscle forces and external loads on cruciate ligament strain. *Am J Sports Med* 1995; 23(1): 129-136.

Elias JJ, Faust AF, Chu Y-H, 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; 31(2): 241-246.

Engebretsen L. Discussion, summary, and future research goals. In: Hewett TE, Shultz SJ, Griffin LY, editors. *Understanding and preventing noncontact ACL injuries*, Illinois: Human Kinetics; 2007, p. 121-128.

---

Escamilla RF, Lander JE, Garhammer, J. Biomechanics of powerlifting and weightlifting exercises. In: Garrett WE, Kirkendall DT, editors. Exercise and sports science, Philadelphia: Lippincott Williams & Wilkins; 2000, p. 585–615.

Ewing KA, Begg RK, Galea MP, Lee PVS. Effects of prophylactic knee bracing on lower limb kinematics, kinetics, and energetics during double-leg drop landing at 2 heights. *Am J Sports Med* 2016; 44(7): 1753-1761.

Fleming BC, Renstrom PA, Beynnon BD, Engstrom B, Peura G. The influence of functional knee bracing on the anterior cruciate ligament strain biomechanics in weightbearing and nonweightbearing knees. *Am J Sports Med* 2000; 28(6): 815-824.

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* 2001a; 34(2): 163-170.

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* 2001b; 19(6): 1178-1184.

Ford KR, Myer GD, Smith RL, Vianello RM, Seiwert SL, Hewett TE. A comparison of dynamic coronal plane excursion between matched male and female athletes when performing single leg landings. *Clin Biomech (Bristol, Avon)* 2006; 21(1): 33-40.

Fujii M, Sato H, Takahira N. Muscle activity response to external moment during single-leg drop landing in young basketball players: The importance of biceps femoris in reducing internal rotation of knee during landing. *J Sports Sci Med* 2012; 11(2): 255-259.

Garrett WE Jr, Yu B. Congruence between existing prevention programs and research on risk factors and mechanisms of noncontact ACL injury. In: Hewett TE, Shultz SJ, Griffin LY, editors. Understanding and preventing noncontact ACL injuries, Illinois: Human Kinetics; 2007, p. 109-119.

Garrison JC, Hart JM, Palmieri RM, Kerrigan DC, Ingersoll CD. Comparison of knee-joint moments in male and female college soccer players during a single-leg landing. *J Sport Rehabil* 2005; 14(4): 332–337.

Gebhard F, Authenrieth M, Strecker W, Kinzl L, Hehl G. Ultrasound evaluation of gravity induced anterior drawer following anterior cruciate ligament lesion. *Knee Surg, Sports Traumatol Arthrosc* 1999; 7(3): 166-172.

Gianotti SM, Marshall SW, Hume PA, Bunt L. Incidence of anterior cruciate ligament injury and other knee ligament injuries: a national population-based study. *J Sci Med Sport* 2009; 12(6): 622-627.

Giotis D, Tsiaras V, Ristanis S, Zampeli F, Mitsionis G, Stergiou N, *et al.* Knee braces can decrease tibial rotation during pivoting that occurs in high demanding activities. *Knee Surg Sports Traumatol Arthrosc* 2011; 19(8): 1347-1354.

Glaws KR, Juneau CM, Becker LC, Di Stasi SL, Hewett TE. Intra- and inter-rater reliability of the selective functional movement assessment (sfma). *Int J Sports Phys Ther* 2014; 9(2): 195-207.

Gollogly S, Smith JT, West HS Jr. Anatomy of the Knee. ACLSolutions.com. <http://www.aclsolutions.com/anatomy.php>; 2011 [accessed 08.07.16].

Griffin LY, Albohm MJ, Arendt EA, Bahr R, Beynnon BD, DeMaio M, *et al.* Understanding and preventing noncontact anterior cruciate ligament injuries: a review of the Hunt Valley II Meeting, January 2005. *Am J Sports Med* 2006; 34(9): 1512-1532.

Halaki M, Ginn K. Normalization of EMG signals: To normalize or not to normalize and what to normalize to? In: Naik GR, editor. *Computational intelligence in electromyography analysis - a perspective on current applications and future challenges*, Rijeka: InTech; 2012, p. 175-194.

Hamill J, Selbie WS, Kepple TM. Three-Dimensional Kinematics. In: Robertson DE, Caldwell GE, Hamill J, Kamen G, Whittlesey SN. *Research methods in biomechanics*, Illinois: Human Kinetics; 2014, p. 35-59.

Hanna D, Dempster M. *Psychology statistics for dummies*. West Sussex: John Wiley and Sons, Ltd; 2012.

Harris JD, Erickson BJ, Bach BR Jr, Abrams GD, Cvetanovich GL, Forsythe B, *et al.* Return-to-sport and performance after anterior cruciate ligament reconstruction in national basketball association players. *Sports Health* 2013; 5(6): 562-568.

Hartigan E, Lewek M, Snyder-Mackler L. The knee. In: Levangie PK, Norkin CC, editors. *Joint structure and function: a comprehensive analysis*, Philadelphia: F.A. Davis Company; 2011, p. 395-439.

Hewett TE, Beynnon BD, Krosshaug T, Myklebust G. Preventing knee injuries. In: Bahr R, Engebretsen L, editors. *Handbook of sports medicine and science, sports injury prevention*, Chichester: Wiley-Blackwell; 2009, p. 49-71.

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; 33(4): 492-501.

Hewett TE, Johnson DL. ACL prevention programs: fact or fiction? *Orthopedics* 2010; 33(1): 36-39.

Hewett TE, Zazulak BT. The costs associated with ACL injury. In: Hewett TE, Shultz SJ, Griffin LY, editors. *Understanding and preventing noncontact ACL injuries*, Illinois: Human Kinetics; 2007, p. 47-56.

Highgenboten CL, Jackson A, Meske N, Smith J. The effects of knee brace wear on perceptual and metabolic variables during horizontal treadmill running. *Am J Sports Med* 1991; 19(6): 639-643.

Hong YG, Yoon Y-J, Kim P, Shin CS. The kinematic/kinetic differences of the knee and ankle joint during single-leg landing between shod and barefoot condition. *Int J Precis Eng Man* 2014; 15(10): 2193-2197.

Hsu W-L, Krishnamoorthy V, Scholz JP. An alternative test of electromyographic normalization in patients. *Muscle Nerve* 2006; 33(2): 232-241.

Huston LJ. Clinical biomechanical studies on ACL injury risk factors. In: Hewett TE, Shultz SJ, Griffin LY, editors. *Understanding and preventing noncontact ACL injuries*, Illinois: Human Kinetics; 2007, p. 141-153.

Ireland ML. Sport-specific injury mechanisms associated with pivoting, cutting, and landing. In: Hewett TE, Shultz SJ, Griffin LY, editors. *Understanding and preventing noncontact ACL injuries*, Illinois: Human Kinetics; 2007, p. 163-171.

Ishida T, Yamanaka M, Takeda N, Aoki Y. Knee rotation associated with dynamic knee valgus and toe direction. *Knee* 2014; 21(2): 563-566.

Jenkins WL, Williams DS, Durland A, Adams B, O'Brien K. Foot orthotic devices decrease transverse plane motion during landing from a forward vertical jump in healthy females. *J Appl Biomech* 2009; 25(4): 387-395.

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; 36(2): 285-289.

Kim HY, Kim KJ, Yang DS, Jeung SW, Choi HG, Choy WS. Screw-home movement of the tibiofemoral joint during normal gait: three-dimensional analysis. *Clin Orthop Surg* 2015; 7(3): 303-309.

Kim K, Jeon K. Comparisons of knee and ankle joint angles and ground reaction force according to functional differences during single-leg drop landing. *J Phys Ther Sci* 2016; 28(4): 1150-1154.

Kimura Y, Ishibashi Y, Tsuda E, Yamamoto Y, Hayashi Y, Sato S. Increased knee valgus alignment and moment during single-leg landing after overhead stroke as a potential risk factor of anterior cruciate ligament injury in badminton. *Br J Sports Med* 2012; 46(3): 207-213.

Kimura Y, Ishibashi Y, Tsuda E, Yamamoto Y, Tsukada H, Toh S. Mechanisms for anterior cruciate ligament injuries in badminton. *Br J Sports Med* 2010; 44(15): 1124-1127.

Kirtley C. Simple tools for motion and gait analysis. *Clinical gait analysis*. <http://www.clinicalgaitanalysis.com/tools/>; n.d. [accessed 11.05.17].

Kobayashi H, Kanamura T, Koshida S, Miyashita K, Okado T, Shimizu T, *et al.* Mechanisms of the anterior cruciate ligament injury in sports activities: a twenty-year clinical research of 1,700 athletes. *J Sports Sci Med* 2010; 9(4): 669-675.

Koga H, Nakamae A, Shima Y, Iwasa J, Myklebust G, Engebretsen L, *et al.* Mechanisms for noncontact anterior cruciate ligament injuries: knee joint kinematics in 10 injury situations from female team handball and basketball. *Am J Sports Med* 2010; 38(11): 2218-2225.

Kovács I, Tihanyi J, Devita P, Rácz L, Barrier J, Hortobágyi T. Foot placement modifies kinematics and kinetics during drop jumping. *Med Sci Sports Exerc* 1999; 31(5): 708-716.

Krosshaug T, Nakamae A, Boden BP, Engebretsen L, Smith G, Slauterbeck JR, *et al.* Mechanisms of anterior cruciate ligament injury in basketball: video analysis of 39 cases. *Am J Sports Med* 2007; 35(3): 359-367.

Lamontagne M, Benoit DL, Ramsey DK, Caraffa A, Cerulli G. In vivo biomechanical study for injury prevention. In: Hong Y, Bartlett R, editors. *Routledge handbook of biomechanics and human movement science*, Oxon: Routledge; 2008, p. 428-445.

Laughlin WA, Weinhandl JT, Kernozek TW, Cobb SC, Keenan KG, O'Connor KM. The effects of single-leg landing technique on ACL loading. *J Biomech* 2011; 44(10): 1845-1851.

Lenhart RL, Francis CA, Lenz AL, Thelen DG. Empirical evaluation of gastrocnemius and soleus function during walking. *J Biomech* 2014; 47(12): 2969-2974.

Li G, DeFrate LE, Rubash HE, Gill TJ. In vivo kinematics of the ACL during weight-bearing knee flexion. *J Orthop Res* 2005; 23(2): 340-344.

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

Lin C-F, Liu H, Garrett WE, Yu B. Effects of a knee extension constraint brace on selected lower extremity motion patterns during a stop-jump task. *J Appl Biomech* 2008; 24(2): 158-165.

Lin C-F, Liu H, Gros MT, Weinhold P, Garrett WE, Yu B. Biomechanical risk factors of non-contact ACL injuries: A stochastic biomechanical modeling study. *J Sport Health Sci* 2012; 1(1): 36-42.

Lombardo S, Sethi PM, Starkey C. Intercondylar notch stenosis is not a risk factor for anterior cruciate ligament tears in professional male basketball players: an 11-Year prospective study. *Am J Sports Med* 2005; 33(1): 29-34.

Loudon JK, Jenkins W, Loudon KL. The relationship between static posture and ACL injury in female athletes. *J Orthop Sports Phys Ther* 1996; 24(2): 91-97.

Lund A, Lund M. Laerd Statistics. Statistical tutorials and software guides. <https://statistics.laerd.com/>; 2015 [accessed 08.08.16].

Lundin O, Styf JR. Intramuscular pressure in the leg and thigh related to tensile strap force during knee brace wear: an experimental study in man. *Am J Sports Med* 1998; 26(4): 567-570.

Malfait B, Dingenen B, Smeets A, Staes F, Pataky T, Robinson MA, *et al.* Knee and hip joint kinematics predict quadriceps and hamstrings neuromuscular activation patterns in drop jump landings. *PLoS One* 2016; 11(4): e0153737.

Marieb EN. Human anatomy and physiology. 6th ed. San Francisco: Pearson Benjamin Cummings; 2004.

Markolf KL, Burchfield DM, Shapiro MM, Shepard MF, Finerman GAM, Slauterbeck JL. Combined knee loading states that generate high anterior cruciate ligament forces. *J Orthop Res* 1995; 13(6): 930-935.

Marsall SW, Padua D, McGrath M. Incidence of ACL injury. In: Hewett TE, Shultz SJ, Griffin LY, editors. Understanding and preventing noncontact ACL injuries, Illinois: Human Kinetics; 2007, p. 5-29.

Martin RL. The ankle and foot complex. In: Levangie PK, Norkin CC, editors. Joint structure and function: a comprehensive analysis, Philadelphia: F.A. Davis Company; 2011, p. 440-481.

Melnyk M, Gollhofer A. Submaximal fatigue of the hamstrings impairs specific reflex components and knee stability. *Knee Surg Sports Traumatol Arthrosc* 2007; 15(5): 525-532.

Messier SP, Legault C, Loeser RF, Van Arsdale SJ, Davis C, Ettinger WH, *et al.* Does high weight loss in older adults with knee osteoarthritis affect bone-on-bone joint loads and muscle forces during walking? *Osteoarthritis Cartilage* 2011; 19(3): 272-280.

Mitchell LCJ, Ford KR, Minning S, Myer GD, Mangine RE, Hewett TE. Medial foot loading on ankle and knee biomechanics. *N Am J Sports Phys Ther* 2008; 3(3): 133-140.

Mohr M, Krstrup P, Bangsbo J. Match performance of high-standard soccer players with special reference to development of fatigue. *J Sports Sci* 2003; 21(7): 519-528.

Mokhtarzadeh H, Yeow CH, Hong Goh JC, Oetomo D, Malekipour F, Lee PV-S. Contributions of the Soleus and Gastrocnemius muscles to the anterior cruciate ligament loading during single-leg landing. *J Biomech* 2013; 46(11): 1913-1920.

Montgomery DC. Design and analysis of experiments. 7th ed. Asia: John Wiley & Sons (Asia) Pte Ltd; 2009.

Morgan KD, Donnelly CJ, Reinbolt JA. Elevated gastrocnemius forces compensate for decreased hamstrings forces during the weight-acceptance phase of single-leg jump landing: implications for anterior cruciate ligament injury risk. *J Biomech* 2014; 47(13): 3295-3302.

Myer GD, Ford KR, Hewett TE. Rationale and clinical techniques for anterior cruciate ligament injury prevention among female athletes. *J Athl Train* 2004; 39(4): 352-364.

Myklebust G, Skjølberg A, Bahr R. ACL injury incidence in female handball 10 years after the Norwegian ACL prevention study: important lessons learned. *Br J Sports Med* 2013; 47(8): 476-479.

Noyes FR, Barber-Westin SD, Fleckenstein C, Walsh C, West J. The drop-jump screening test: difference in lower limb control by gender and effect of neuromuscular training in female athletes. *Am J Sports Med* 2005; 33(2): 197-207.

Oliveira ASC, Silva PB, Lund ME, Farina D, Kersting UG. Slipping during side-step cutting: Anticipatory effects and familiarization. *Hum Mov Sci* 2014; 34: 128-136.

Olsen O-E, 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; 32(4): 1002-1012.

---

Orchard J, Seward H, McGivern J, Hood S. Intrinsic and extrinsic risk factors for anterior cruciate ligament injury in Australian footballers. *Am J Sports Med* 2001; 29(2): 196-200.

Orishimo KF, Kremenic IJ, Pappas E, Hagins M, Liederbach M. Comparison of landing biomechanics between male and female professional dancers. *Am J Sports Med* 2009; 37(11): 2187-2193.

Orishimo KF, Liederbach M, Kremenic IJ, Hagins M, Pappas E. Comparison of landing biomechanics between male and female dancers and athletes, part 1: influence of sex on risk of anterior cruciate ligament injury. *Am J Sports Med* 2014; 42(5): 1082-1088.

Padua DA, Arnold BL, Carcia CR, Granata KP. Gender differences in leg stiffness and stiffness recruitment strategy during two-legged hopping. *J Mot Behav* 2005; 37(2): 111-125.

Palm H-G, Bergenthal G, Ehry P, Schwarz W, Schmidt R, Friemert B. Functional ultrasonography in the diagnosis of acute anterior cruciate ligament injuries: a field study. *Knee* 2009; 16(6): 441-446.

Pappas E, Hagins M, Sheikhzadeh A, Nordin M, Rose D. Biomechanical differences between unilateral and bilateral landings from a jump: gender differences. *Clin J Sport Med* 2007; 17(4): 263-268.

Peeters MJ. Practical significance: Moving beyond statistical significance. *Curr Pharm Teach Learn* 2016; 8(1): 83-89.

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; 36(11): 1949-1958.

Pietrosimone BG, Grindstaff TL, Linens SW, Uczekaj E, Hertel J. A systematic review of prophylactic braces in the prevention of knee ligament injuries in collegiate football players. *J Athl Train* 2008; 43(4): 409-415.

Pollard CD, Davis IM, Hamill J. Influence of gender on hip and knee mechanics during a randomly cued cutting maneuver. *Clin Biomech (Bristol, Avon)* 2004; 19(10): 1022-1031.

Portney LG, Watkins MP. *Foundations of clinical research: applications to practice*. 3rd ed. Philadelphia: F.A. Davis Company; 2015.

Quatman CE, Quatman-Yates CC, Hewett TE. A 'plane' explanation of anterior cruciate ligament injury mechanisms: a systematic review. *Sports Med* 2010; 40(9): 729-746.

Ramesh R, Von Arx O, Azzopardi T, Schranz PJ. The risk of anterior cruciate ligament rupture with generalised joint laxity. *J Bone Joint Surg Br* 2005; 87(6): 800-803.

Ramsey DK, Lamontagne M, Wretenberg PF, Valentin A, Engström B, Németh G. Assessment of functional knee bracing: an in vivo three-dimensional kinematic analysis of the anterior cruciate deficient knee. *Clin Biomech (Bristol, Avon)* 2001; 16(1): 61-70.

Renström 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; 14(1): 83-87.

Renstrom P, Ljungqvist A, Arendt E, Beynnon B, Fukubayashi T, Garrett W, *et al.* Non-contact ACL injuries in female athletes: An International Olympic Committee current concepts statement. *Br J Sports Med* 2008; 42(6): 394-412.

Rishiraj N, Taunton JE, Lloyd-Smith R, Regan W, Niven B, Woollard R. Effect of functional knee brace use on acceleration, agility, leg power and speed performance in healthy athletes. *Br J Sports Med* 2011; 45(15): 1230-1237.

Rishiraj N, Taunton JE, Lloyd-Smith R, Regan W, Niven B, Woollard R. Functional knee brace use effect on peak vertical ground reaction forces during drop jump landing. *Knee Surg Sports Traumatol Arthrosc* 2012; 20(12): 2405-2412.

Rishiraj N, Taunton JE, Lloyd-Smith R, Woollard R, Regan W, Clement DB. The potential role of prophylactic/functional knee bracing in preventing knee ligament injury. *Sports Med* 2009; 39(11): 937-960.

Roi GS, Nanni G, Tencone F. Time to return to professional soccer matches after ACL reconstruction. *Sport Sci Health* 2006; 1(4): 142-145.

Roos H, Ornell M, Gärdsell P, Lohmander LS, Lindstrand A. Soccer after anterior cruciate ligament injury - an incompatible combination? A national survey of incidence and risk factors and a 7-year follow-up of 310 players. *Acta Orthop Scand* 1995; 66(2): 107-112.

Rosso V, Rainoldi A. Optimization of sEMG electrode positioning in vastus lateralis muscle during neuromuscular electrical stimulation. *Sport Sci Health* 2014; 10(3): 253-260.

Rovere GD, Haupt HA, Yates CS. Prophylactic knee bracing in college football. *Am J Sports Med* 1987; 15(2): 111-116.

Russell KA, Palmieri RM, Zinder SM, Ingersoll CD. Sex differences in valgus knee angle during a single-leg drop jump. *J Athl Train* 2006; 41(2): 166-171.

---

Russell PJ, Croce RV, Swartz EE, Decoster LC. Knee-muscle activation during landings: developmental and gender comparisons. *Med Sci Sports Exerc* 2007; 39(1): 159–170.

Sanders TL, Maradit Kremers H, Bryan AJ, Larson DR, Dahm DL, Levy BA *et al.* Incidence of Anterior Cruciate Ligament Tears and Reconstruction. *Am J Sports Med* 2016; 44(6): 1502-1507.

Sasaki S, Nagano Y, Kaneko S, Imamura S, Koabayshi T, Fukubayashi T. The relationships between the center of mass position and the trunk, hip, and knee kinematics in the sagittal plane: a pilot study on field-based video analysis for female soccer players. *J Hum Kinet* 2015; 45(1): 71-80.

Schickendantz MS, Weiker GG. The predictive value of radiographs in the evaluation of unilateral and bilateral anterior cruciate ligament injuries. *Am J Sports Med* 1993; 21(1): 110-113.

Secrist ES, Bhat SB, Dodson CC. The financial and professional impact of anterior cruciate ligament injuries in national football league athletes. *Orthop J Sports Med* 2016; 4(8): 2325967116663921.

Selbie WS, Hamill J, Kepple TM. Three-Dimensional Kinetics. In: Robertson DE, Caldwell GE, Hamill J, Kamen G, Whittlesey SN. *Research methods in biomechanics*, Illinois: Human Kinetics; 2014, p. 151-176.

Self BP, Paine D. Ankle biomechanics during four landing techniques. *Med Sci Sports Exerc* 2001; 33(8): 1338-1344.

Shaw MY, Gribble PA, Frye JL. Ankle bracing, fatigue, and time to stabilization in collegiate volleyball athletes. *J Athl Train* 2008; 43(2): 164-171.

Sheehan FT, Sipprell WH 3rd, Boden BP. Dynamic sagittal plane trunk control during anterior cruciate ligament injury. *Am J Sports Med* 2012; 40(5): 1068- 1074.

Shimokochi Y, Ambegaonkar JP, Meyer EG, Lee SY, Shultz SJ. Changing sagittal plane body position during single-leg landings influences the risk of non-contact anterior cruciate ligament injury. *Knee Surg Sports Traumatol Arthrosc* 2013; 21(4): 888-897.

Shimokochi Y, Lee SY, Shultz SJ, Schmitz RJ. The relationships among sagittal-plane lower extremity moments: implications for landing strategy in anterior cruciate ligament injury prevention. *J Athl Train* 2009; 44(1): 33-38.

Shultz SJ. Preventive training programs: changing strength ratios versus positions of muscular efficiency. In: Hewett TE, Shultz SJ, Griffin LY, editors. *Understanding and preventing noncontact ACL injuries*, Illinois: Human Kinetics; 2007a, p. 91-101.

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Shultz SJ. Hormonal influences on ligament biology. In: Hewett TE, Shultz SJ, Griffin LY, editors. Understanding and preventing noncontact ACL injuries, Illinois: Human Kinetics; 2007b, p. 219 - 238.

Shultz SJ, Nguyen A-D, Beynon BD. Anatomical factors in ACL injury risk. In: Hewett TE, Shultz SJ, Griffin LY, editors. Understanding and preventing noncontact ACL injuries, Illinois: Human Kinetics; 2007, p. 239-258.

Shultz SJ, Schmitz RJ, Tritsch AJ, Montgomery MM. Methodological considerations of task and shoe wear on joint energetics during landing. *J Electromyogr Kinesiol* 2012; 22(1): 124-130.

Singer JC, Lamontagne M. The effect of functional knee brace design and hinge misalignment on lower limb joint mechanics. *Clin Biomech (Bristol, Avon)* 2008; 23(1): 52-59.

Sitler M, Ryan J, Hopkinson W, Wheeler J, Santomier J, Kolb R, *et al.* The efficacy of a prophylactic knee brace to reduce knee injuries in football: a prospective, randomized study at West Point. *Am J Sports Med* 1990; 18(3): 310-315.

Smith HC, Vacek P, Johnson RJ, Slauterbeck JR, Hashemi J, Shultz S, *et al.* Risk factors for anterior cruciate ligament injury: a review of the literature -- part 1: neuromuscular and anatomic risk. *Sports Health* 2012a; 4(1): 69-78.

Smith HC, Vacek P, Johnson RJ, Slauterbeck JR, Hashemi J, Shultz S, *et al.* Risk factors for anterior cruciate ligament injury: a review of the literature - part 2: hormonal, genetic, cognitive function, previous injury, and extrinsic risk factors. *Sports Health* 2012b; 4(2): 155-161.

Speer KP, Spritzer CE, Bassett FH 3rd, Feagin JA Jr, Garrett WE Jr. Osseous injury associated with acute tears of the anterior cruciate ligament. *Am J Sports Med* 1992; 20(4): 382-389.

Stanley CJ, Creighton RA, Gross MT, Garrett WE, Yu B. Effects of a knee extension constraint brace on lower extremity movements after ACL reconstruction. *Clin Orthop Relat Res* 2011; 469(6): 1774-1780.

Steiner ME, Grana WA, Chillag K, Schelberg-Karnes E. The effect of exercise on anterior-posterior knee laxity. *Am J Sports Med* 1986; 14(1): 24-29.

Stuelcken MC, Mellifont DB, Gorman AD, Sayers MGL. Mechanisms of anterior cruciate ligament injuries in elite women's netball: a systematic video analysis. *J Sports Sci* 2016; 34(16): 1516-1522.

Styf JR, Nakhostine M, Gershuni DH. Functional knee braces increase intramuscular pressures in the anterior compartment of the leg. *Am J Sports Med* 1992; 20(1): 46-49.

Sugimoto D, Myer GD, McKeon JM, Hewett TE. Evaluation of the effectiveness of neuromuscular training to reduce anterior cruciate ligament injury in female athletes: a critical review of relative risk reduction and numbers-needed-to-treat analyses. *Br J Sports Med* 2012; 46(14): 979-988.

Sugimoto D, Myer GD, Barber Foss KD, Hewett TE. Specific exercise effects of preventive neuromuscular training intervention on anterior cruciate ligament injury risk reduction in young females: meta-analysis and subgroup analysis. *Br J Sports Med* 2015; 49(5): 282-289.

Tarmah H, Rahnama N, Khayyambashi K. Video analysis of causes and mechanism of the ACL injuries in the Iranian professional soccer player. *Br J Sports Med* 2010; 44: i3-i4.

Taylor JB, Waxman JP, Richter SJ, Shultz SJ. Evaluation of the effectiveness of anterior cruciate ligament injury prevention programme training components: a systematic review and meta-analysis. *Br J Sports Med* 2015; 49(2): 79-87.

Taylor KA, Terry ME, Utturkar GM, Spritzer CE, Queen RM, Iribarra LA, *et al.* Measurement of in vivo anterior cruciate ligament strain during dynamic jump landing. *J Biomech* 2011; 44(3): 365-371.

Teng PSP, Kong PW, Leong KF. Effects of foot rotation positions on knee valgus during single-leg drop landing: implications for ACL injury risk reduction. *Knee* 2017; 24(3): 547-554.

Torry MR, Myers C, Shelburne KB, Peterson D, Giphart JE, Pennington WW, *et al.* Relationship of knee shear force and extensor moment on knee translations in females performing drop landings: A biplane fluoroscopy study. *Clin Biomech (Bristol, Avon)* 2011; 26(10): 1019-1024.

Tran AA, Gatewood C, Harris AH, Thompson JA, Dragoo JL. The effect of foot landing position on biomechanical risk factors associated with anterior cruciate ligament injury. *J Exp Orthop* 2016; 3(1): 13. Epub 2016 Jun 18.

Trimble MH, Bishop MD, Buckley BD, Fields LC, Rozea GD. The relationship between clinical measurements of lower extremity posture and tibial translation. *Clin Biomech (Bristol, Avon)* 2002; 17(4): 286-290.

Uhorchak JM, Scoville CR, Williams GN, Arciero RA, St. Pierre P, Taylor DC. Risk factors associated with noncontact injury of the anterior cruciate ligament: a prospective four-year evaluation of 859 West Point cadets. *Am J Sports Med* 2003; 31(6): 831-842.

Valenzuela KA, Bhaskaran D, Hummer C, Schefano A, Zhang S. Effects of a combined inversion and plantarflexion surface on knee and hip kinematics during landing. *Sports Biomech* 2016; 15(4): 429-439.

Visual3D Documentation. C-Motion WIKI Documentation. [https://www.c-motion.com/v3dwiki/index.php/Visual3D\\_Documentation](https://www.c-motion.com/v3dwiki/index.php/Visual3D_Documentation); 2017 [accessed 28.05.17].

Waldén M, Hägglund M, Ekstrand J. High risk of new knee injury in elite footballers with previous anterior cruciate ligament injury. *Br J Sports Med* 2006; 40(2): 158-162.

Waldén M, Hägglund M, Magnusson H, Ekstrand J. Anterior cruciate ligament injury in elite football: a prospective three-cohort study. *Knee Surg Sports Traumatol Arthrosc* 2011; 19(1): 11-19.

Walsh M, Boling MC, McGrath M, Blackburn JT, Padua DA. Lower Extremity Muscle Activation and Knee Flexion During a Jump-Landing Task. *J Athl Train* 2012; 47(4): 406-413.

Wang L-I. The lower extremity biomechanics of single- and double-leg stop-jump tasks. *J Sports Sci Med* 2011; 10(1): 151-156.

Ward SR, Eng CM, Smallwood LH, Lieber RL. Are current measurements of lower extremity muscle architecture accurate? *Clin Orthop Relat Res* 2009; 467(4): 1074-1082.

Weinhandl JT, Joshi M, O'Connor KM. Gender comparisons between unilateral and bilateral landings. *J Appl Biomech* 2010; 26(4): 444-453.

Winter DA. *Biomechanics and motor control of human movement*. 4th ed. Hoboken, NJ: John Wiley & Sons, Inc.; 2009.

Wojtys EM, Wylie BB, Huston LJ. The effects of muscle fatigue on neuromuscular function and anterior tibial translation in healthy knees. *Am J Sports Med* 1996; 24(5): 615-621.

Woodford-Rogers B, Cyphert L, Denegar CR. Risk factors for anterior cruciate ligament injury in high school and college athletes. *J Athl Train* 1994; 29(4): 343-346.

Yeow CH, Lee PVS, Goh JCH. Shod landing provides enhanced energy dissipation at the knee joint relative to barefoot landing from different heights. *Knee* 2011; 18(6): 407-411.

Yu B, Lin C-F, Garrett WE. Lower extremity biomechanics during the landing of a stop-jump task. *Clin Biomech (Bristol, Avon)* 2006; 21(3): 297-305.

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Yu B, Garrett WE. Hamstring co-contraction does not necessarily reduce ACL loading. XXth Congress of the International Society of Biomechanics and 29th Annual Meeting of the American Society of Biomechanics; 2005 July 31 – Aug 5; Cleveland, Ohio, USA.

Yu B, Garrett WE. Mechanisms of non-contact ACL injuries. *Br J Sports Med* 2007; 41(Suppl 1): i47-i51.

Zebis MK, Andersen LL, Bencke J, Kjaer M, Aagaard P. Identification of athletes at future risk of anterior cruciate ligament ruptures by neuromuscular screening. *Am J Sports Med* 2009; 37(10): 1967-1973.

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## Appendix A: Location of heel markers on the shoe surface

For the distal calcaneus (CAL1) and proximal calcaneus (CAL2) positions, the markers' proximal positions were estimated when the markers were placed on the shoe. However, the medial-lateral position of the CAL2 marker was aligned to the medial-lateral position of the Achilles tendon. The medial-lateral position of the Achilles tendon could be observed from above the most proximal aspect of the shoe (see Figure A.1). Marker CAL1 was placed vertically below CAL2, above the heel of the shoe (see Figure A.2).



**Figure A.1. Location of the medial-lateral position of the Achilles tendon.**



**Figure A.2. Location of the CAL1 and CAL2 markers.**

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## **Appendix B: Equipment Setting Calibration**

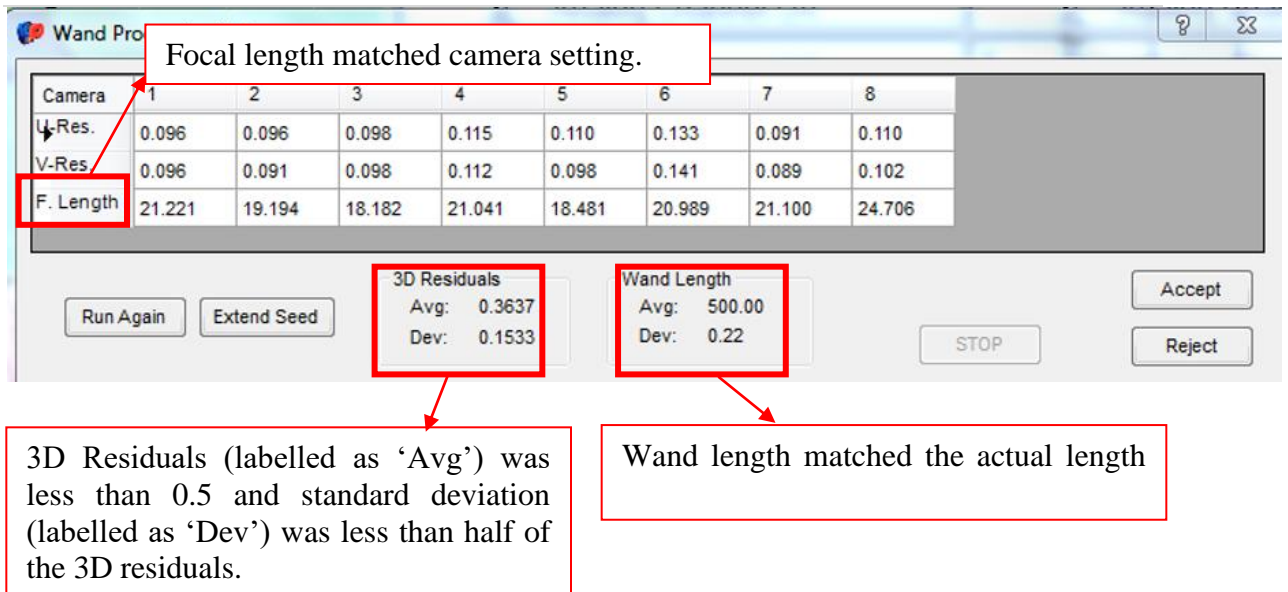
Prior to the pilot, as well as actual study, the lab and cameras were pre-booked for calibration. After calibration, all camera and equipment positions were marked on the ground. The lab was block-booked for a period for the study and no one was to touch or shift the cameras. All participants were scheduled to take part in the study during the period. Each morning and after lunch, the motion analysis cameras were also calibrated using the L-frame and wand. All participants were requested not to hit the camera tripods. Should the camera tripods be hit, the system would be re-calibrated using the L-frame and wand.

The cameras were positioned evenly around the 30-cm platform and force plate region, where the target capture area was. For the study in Chapter 3 and Chapter 4, some of the cameras were positioned to capture an estimated calibration volume from the chest to the feet of a 2-m tall participant standing on the 30-cm platform and on the force plate. Some other cameras were positioned to capture the estimated calibration volume from the chest to the feet of a 2-m tall participant standing on the force plate. In the study in Chapter 5 with cameras of higher resolution, all cameras were positioned to capture an estimated calibration volume both from the chest to the feet of a 2-m tall participant standing on the 30-cm platform and on the force plate. The system was set such that at least 2 cameras could see each marker at every instant. During set-up, markers were placed at extreme positions of the capture area (using a trial participant and additional markers were placed around the 30-cm platform and around the force plate) to ensure that all markers could be

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seen. Camera positions were further optimized. Focal length and focus were adjusted on the cameras; threshold and brightness settings in the motion analysis software, Cortex (version 1.1.4.368 and version 2.6.2.1169, Motion Analysis Corp, Santa Rosa, CA, USA), were fine-tuned to obtain the sharpest possible images of the markers. Any stray light sources that did not come from the markers were removed. Masking was only carried out when cameras across the capture area had to be seen. Care was taken that masked areas would not affect data collection.

Calibration using the L-frame and the wand was carried out during set-up. There were 3 markers on the wand. As recommended by the manufacturer, if any camera captured more than a hundred instances of more than 3 markers, the camera would be re-set up by adjusting the focal length, focus, threshold and brightness. The capture area would also be checked for any stray light and shiny object. The calculated focal length was checked to match the adjusted focal length of each camera. The calculated wand length should match the actual wand length of 500 mm and 3D residuals were maintained below 0.5 (with standard deviation approximately half of the 3D residuals, as recommended by the manufacturer). The calibrated data during set-up is seen in the figure below and all subsequent daily calibration was to maintain similar calibration results.



**Figure B. Print-screen of calibration results obtained during the study.**

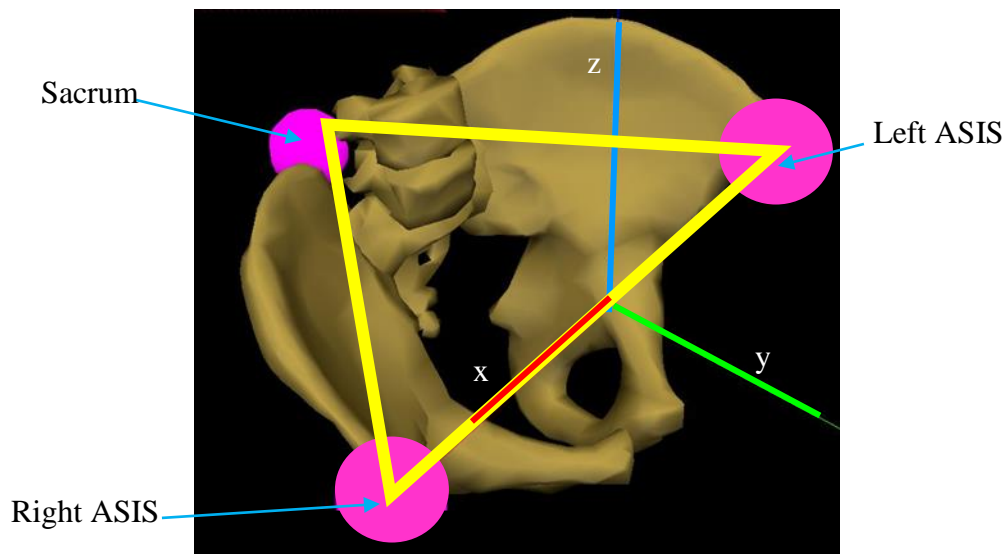
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## **Appendix C: Derivation of Biomechanical Measures**

Visual3D (v5.00.33, C-Motion, Inc., Germantown, MD, USA) was used to build a model consisting of the pelvis, thigh, shank and foot segments. Kinematic and kinetic variables were obtained using Visual3D, using the segment optimization method for three-dimensional kinematics analysis and three-dimensional inverse dynamics analysis for the calculation of kinetic measures (Winter, 2009, Hamill *et al.*, 2014, Selbie *et al.*, 2014, “Visual3D Documentation”, 2017).

### **Step 1: Setting of Local Coordinate System for Each Segment**

First, the local coordinate system (LCS) for each segment in the lower extremity was defined, with respect to the global coordinate system (GCS). The CODA model (Charnwood Dynamics Ltd., Leicestershire, UK) was used to construct the pelvis segment (see Figure C.1) using the sacrum and the anterior superior iliac spine (ASIS) markers. The sacrum marker was input as the right and left posterior-superior iliac spine markers in Visual3D.



**Figure C.1. Pelvis segment local coordinate system.**

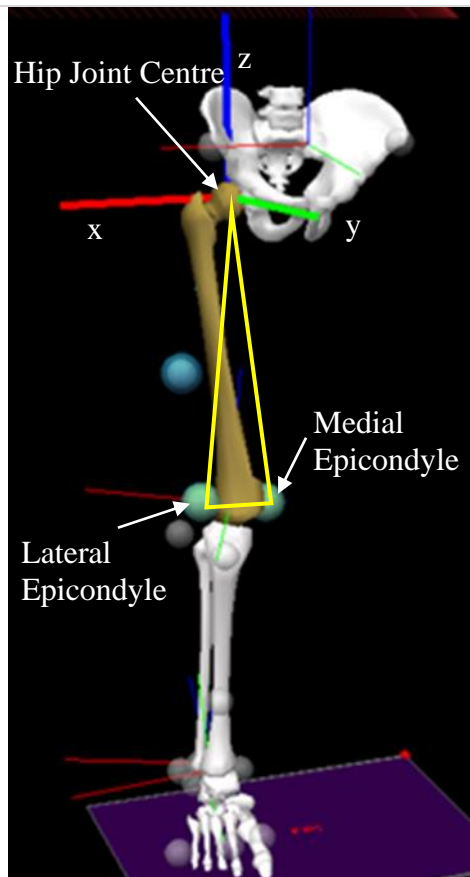
The right and left hip joint centre markers (RHJC and LHJC) were derived using equations by Bell *et al.* (1989) (“Visual3D Documentation”, 2017):

$$\text{RHJC} = (0.36 \cdot \text{ASIS\_Distance}, -0.19 \cdot \text{ASIS\_Distance}, -0.3 \cdot \text{ASIS\_Distance}) \quad (5)$$

$$\text{LHJC} = (-0.36 \cdot \text{ASIS\_Distance}, -0.19 \cdot \text{ASIS\_Distance}, -0.3 \cdot \text{ASIS\_Distance}) \quad (6)$$

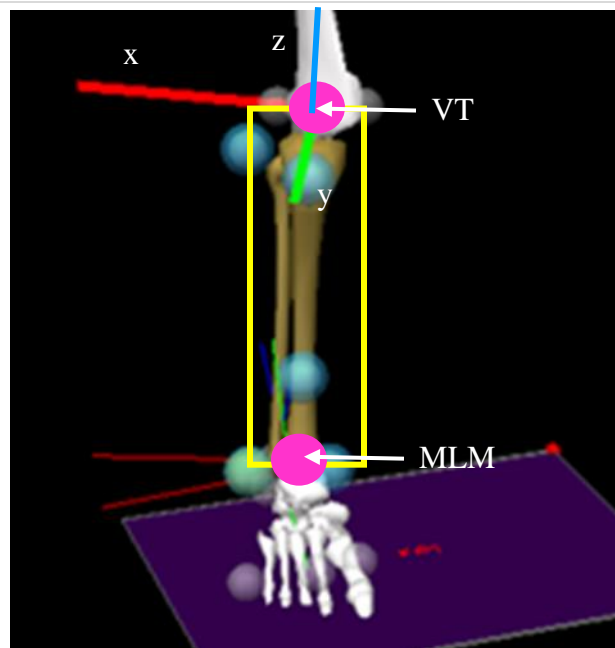
where ASIS\_Distance was the distance between the left and right ASIS markers.

The markers at the hip joint centre, and lateral and medial knee epicondyles, were used to define the frontal plane of the thigh segment (see Figure C.2).



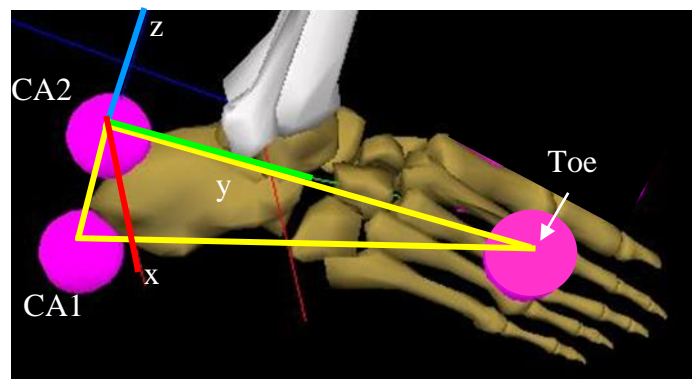
**Figure C.2. Thigh segment local coordinate system.**

A virtual landmark (labelled as 'VT') was created, with an axial location obtained from at a point between the lateral and medial epicondyles. Its anterior-posterior and medial-lateral location was obtained using a point that is created by projecting the tibial tuberosity marker onto the plane created by the head of fibular, and lateral and medial malleoli markers. The frontal plane of the shank segment was defined using this virtual landmark as proximal joint centre and the mid-point of the lateral and medial malleoli markers (labelled as 'MLM') as distal joint centre (see Figure C.3).



**Figure C.3. Shank segment local coordinate system.**

Finally, a single-segment foot model was created using the proximal calcaneus (CAL2) marker as the proximal joint centre and the toe marker as the distal joint centre of the sagittal plane. The distal calcaneus (CAL1) marker was used to define the posterior orientation.



**Figure C.4. Foot segment local coordinate system.**

For all the segments, the frontal plane determined the orientation of the x axis and the sagittal plane determined the orientation of the y axis. The z axis was orthogonal to both x and y axes.

### Step 2: Tracking of Local Coordinate System of Each Segment During Motion

After establishing the LCS of each segment, the segment was tracked using the segment optimization method as described by Hamill *et al.* (2014), during motion:

$$\vec{P}_i = R'_{seg} \vec{P}'_i + \vec{O}_{seg} \quad (7)$$

where  $\vec{P}_i$  was the position of the tracking marker in the GCS,

$\vec{P}'_i$  was the position of the tracking marker in the LCS,

$R'_{seg}$  was the rotation matrix of the segment from LCS to GCS,

$\vec{O}_{seg}$  was the linear transformation between LCS and GCS, defining the origin of the LCS, relative to the GCS.

$R'_{seg}$  and  $\vec{O}_{seg}$  were estimated by minimizing the sum of squares error:

$$E = \sum_{i=1}^s \left( \left( \vec{P}_i - R'_{seg} \vec{P}'_i \right) - \vec{O}_{seg} \right)^2 \quad (8)$$

under the orthonormal constraint

$$R'_{seg} R_{seg} = I \quad (9)$$

where s is the number of markers on the segment ( $s > 2$ ).

Tracking markers can be placed at bony areas to reduce soft tissue artifacts using segment optimization (Hamill *et al.*, 2014).

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### Step 3: Calculation of Joint Angles

After that, joint angles could then be obtained. A three-dimensional rotation matrix (to represent the orientation of one LCS with respect to another LCS) was obtained using three successive rotations about each axis (Hamill *et al.*, 2014). Cardan xyz rotation sequence was used as this rotation sequence is commonly used in biomechanics (Hamill *et al.*, 2014, Winter, 2009). This involved a rotation about the lateral axis x, then about the anterior axis y, and finally about the vertical axis z (Hamill *et al.*, 2014, Winter, 2009). A joint angle is the relative orientation of a segment's LCS with another segment's LCS (Hamill *et al.*, 2014). Since each segment has its orientation, with respect to the GCS, computation of the knee joint angle is as follows (Hamill *et al.*, 2014):

$$R_{Knee} = R_{Shank}R'_{Thigh} \quad (10)$$

where  $R_{Knee}$  is the knee joint angle, represented by an orientation matrix,

$R_{Shank}$  is the transformation of the shank LCS to the GCS,

$R'_{Thigh}$  is the transpose of the transformation of the thigh LCS to the GCS.

Three Cardan angles ( $\alpha_c$ ,  $\beta_c$ ,  $\gamma_c$ ) can be obtained from the rotation matrix  $R_{Knee}$ . This could be used to obtain the knee flexion/extension angles (in the x-axis) and knee varus/valgus angles (in the y-axis). From the model, knee extension and knee varus angles were denoted as positive (see Chapter 3). However, for ease of analysis in this study, knee flexion angles were presented as positive in the thesis (see Chapters 4 and 5).

Computation of the ankle joint angle is as follows:

$$R_{Ankle} = R_{Foot}R'_{Shank} \quad (11)$$

where  $R_{Ankle}$  is the ankle joint angle, represented by an orientation matrix,

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$R_{\text{Foot}}$  is the transformation of the foot LCS to the GCS,

$R'_{\text{Shank}}$  is the transpose of the transformation of the shank LCS to the GCS.

The Cardan angles obtained from the rotation matrix  $R_{\text{Ankle}}$  can be used to obtain the foot rotation angles (in the z-axis). Toe-in angles were denoted as positive (see Chapter 3).

Foot rotation angles were also computed, with respect to the pelvis:

$$R_{\text{Ankle}} = R_{\text{Foot}}R'_{\text{Pelvis}} \quad (12)$$

where  $R_{\text{Ankle}}$  is the ankle joint angle, represented by an orientation matrix,

$R_{\text{Foot}}$  is the transformation of the foot LCS to the GCS,

$R'_{\text{Pelvis}}$  is the transpose of the transformation of the pelvis LCS to the GCS.

The Cardan angles obtained from the rotation matrix  $R_{\text{Ankle}}$  can be used to obtain the foot rotation angles (in the z-axis). Toe-in angles were denoted as positive (see Chapter 3).

Ankle plantar flexion angle measured during the static standing trial was used as each participant's neutral (zero) alignment. Ankle plantar flexion angle during the drop-landing was measured and referenced to the standing position (Ford *et al.*, 2006). This is known as the joint angle normalization (Hamill *et al.*, 2014). Since angles are not a vector, angles cannot be added or subtracted (Hamill *et al.*, 2014). To set the ankle plantar flexion angles in the standing trial as zero, the LCS of the foot was aligned precisely with the LCS of the shank, while using tracking markers from the foot ("Visual3D Documentation", 2017). From the model, ankle dorsi flexion angles were denoted as positive. However, for ease of analysis in this study, ankle plantar flexion angles were presented as positive in this thesis (see Chapters 4 and 5).

Knee flexion range of motion ( $KF_{ROM}$ ) was computed by subtracting the knee flexion angles at initial contact ( $KF_{IC}$ ) from the knee flexion angles at maximum VGRF ( $KF_{VGRF}$ ) (see Chapter 4):

$$KF_{ROM} = KF_{VGRF} - KF_{IC} \quad (13)$$

This subtraction method of estimating range of motion was similarly used in previous studies (Walsh *et al.*, 2012, Ewing *et al.*, 2016) and was proposed as one of the methods in the Visual3D documentation (“Visual3D Documentation”, 2017).

#### **Step 4: Calculation of Joint Reaction Forces and Net Moments**

Inverse dynamics was used to compute the joint moments, using the motion analysis system to record the movement of rigid segments and deriving net forces and moments of the joints (Selbie *et al.*, 2014). Angular velocity and angular acceleration were obtained from the first and second time derivative of the Cardan angles respectively (Selbie *et al.*, 2014, Winter, 2009). After which, the following steps were taken to obtain the kinetic measures (Winter, 2009):

First, the Newtonian three-dimensional equations of motion for a segment were used:

$$\sum F_X = ma_X \text{ or } R_{XP} - R_{XD} = ma_X \quad (14)$$

$$\sum F_Y = ma_Y \text{ or } R_{YP} - R_{YD} - mg = ma_Y \quad (15)$$

$$\sum F_Z = ma_Z \text{ or } R_{ZP} - R_{ZD} = ma_Z \quad (16)$$

where  $a_X$ ,  $a_Y$ ,  $a_Z$  were the segment centre of mass (COM) accelerations in the GCS, and  $R_{XP}$ ,  $R_{XD}$ ,  $R_{YP}$ ,  $R_{YD}$ ,  $R_{ZP}$ ,  $R_{ZD}$  were the proximal and distal reaction forces in the GCS.

Secondly, the proximal and distal reaction forces, as well as, the distal moments, were transformed into the anatomical axes, as described by Winter (2009).

Next, Euler's three-dimensional equations of motion of a segment were been used (Winter, 2009):

$$I_x \alpha_x + (I_z - I_y) \omega_y \omega_z = \sum M_x = R_{zd} I_d + R_{zp} I_p + M_{xp} - M_{xd} \quad (17)$$

$$I_y \alpha_y + (I_x - I_z) \omega_x \omega_z = \sum M_y = M_{yp} - M_{yd} \quad (18)$$

$$I_z \alpha_z + (I_y - I_x) \omega_x \omega_y = \sum M_z = -R_{xd} I_d - R_{xp} I_p + M_{zp} - M_{zd} \quad (19)$$

where

$I_x, I_y, I_z$  were moments of inertia in the LCS,

$\omega_x, \omega_y, \omega_z$  were angular velocity in the LCS,

$\alpha_x, \alpha_y, \alpha_z$  were angular acceleration in the LCS,

$M_{xd}, M_{yd}, M_{zd}$  were the transformed distal moments in the LCS,

$R_{xd}, R_{xp}, R_{yd}, R_{yp}, R_{zd}, R_{zp}$  were the proximal and distal reaction forces in the LCS,

$I_p, I_d$  are distances from COM to proximal and distal joints

Knee valgus moment was defined as the external moment of the shank segment, with respect to the thigh segment in the frontal plane (see Chapter 3). Maximum KVM was normalized to the product of each participant's height (m) and mass (kg). Positive denotes knee varus directions.

## Appendix D: Residual Analysis to Obtain Cut-off Frequencies

The motion and GRF data were filtered through a low-pass, 4th order, zero phase shift Butterworth digital filter (Dempsey *et al.*, 2012, Valenzuela *et al.*, 2016). Cut-off frequencies for kinematics (9 Hz) and kinetics (50 Hz) were obtained using the residual analysis based on data obtained during the pilot study (Winter, 2009). Similar cut-off frequencies were also used in jump studies for filtering of kinematics (9 Hz) (Hewett *et al.*, 2005) and ground reaction force data (50 Hz) (Hewett *et al.*, 2005, Laughlin *et al.*, 2011, Kim *et al.*, 2016).

The residual  $R(f_{cf})$  at a cut-off frequency was calculated using the following formula for a signal of  $N$  sample points in time (Winter, 2009) and a template written by Peter Sinclair was used to carry out the residual analysis (Kirtley, n.d.):

$$R(f_{cf}) = \sqrt{\frac{1}{N} \sum_{i=1}^N (X_i - \hat{X}_i)^2} \quad (20)$$

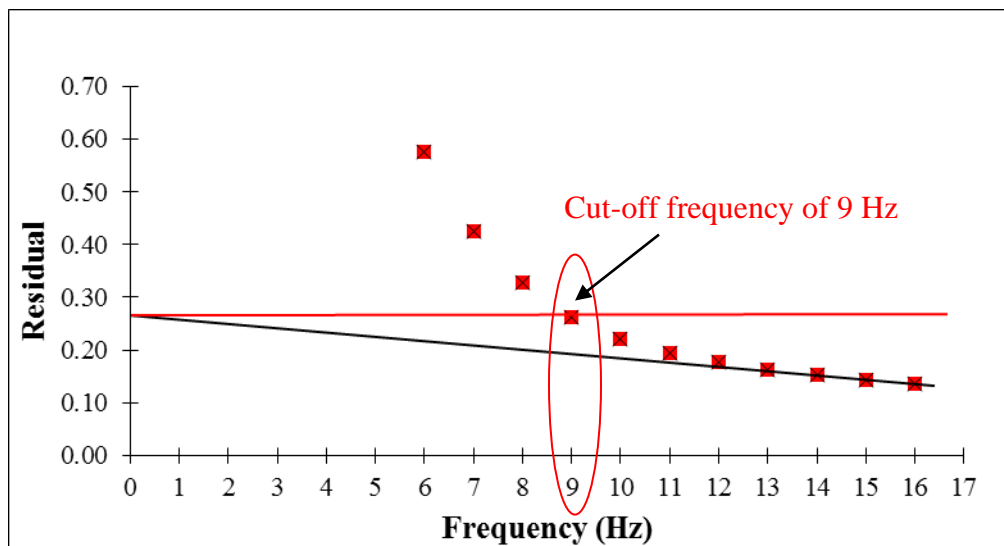
where  $f_{cf}$  is the cut-off frequency of the 4th order dual-pass filter

$X_i$  is the raw data at the  $i$ th sample

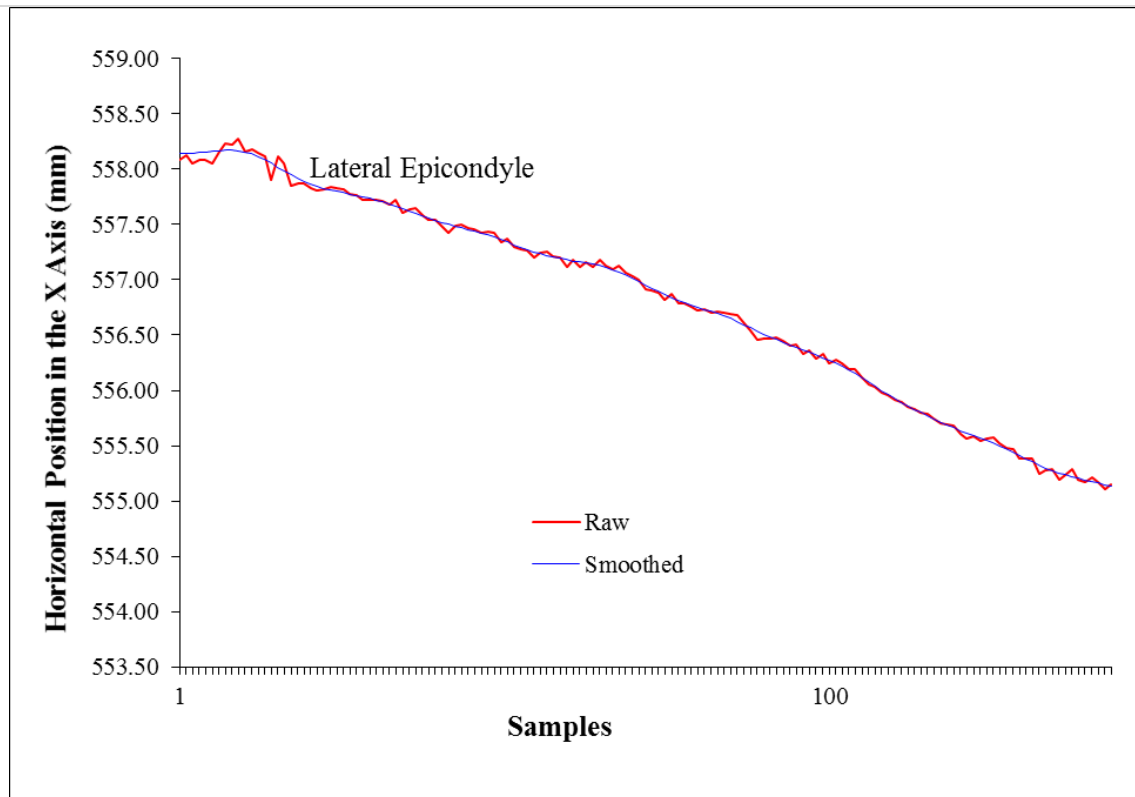
$\hat{X}_i$  is the filtered data at the  $i$ th sample using a 4th order zero-lag filter

The plot of the residual between a filtered and an unfiltered signal with respect to the filter cut-off frequency was plotted for the markers (sacrum, anterior superior iliac spine, lateral knee epicondyles, tibial tuberosity, calcaneus and toe) and ground reaction force.

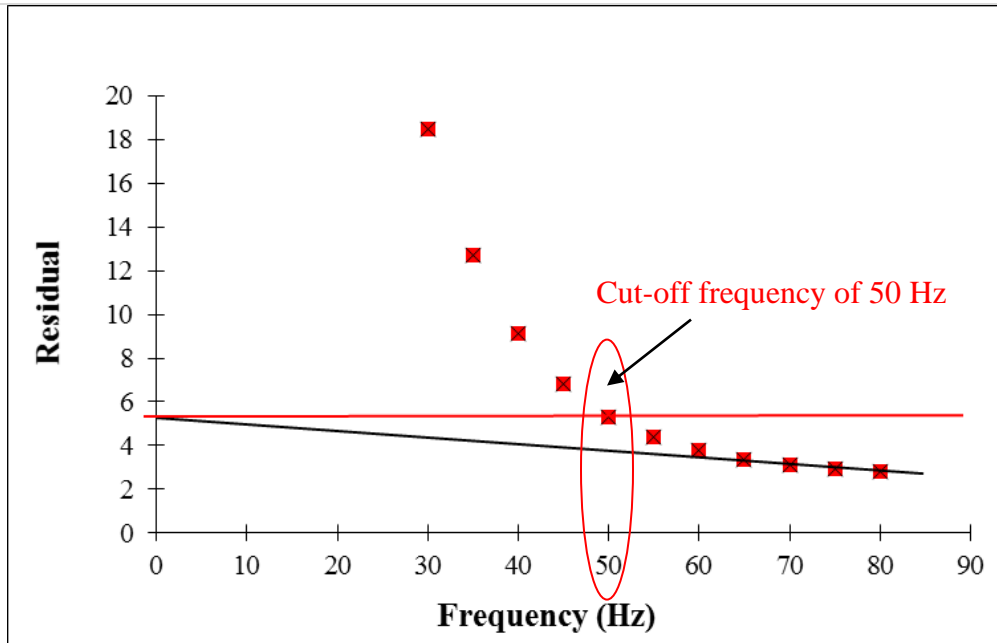
The residual plot of the lateral epicondyle marker and the ground reaction force are illustrated in Figures D.1 and D.3. To obtain an equal amount of signal distortion and noise, the cut-off frequency was determined using the intersection of the horizontal red line and the residual curve (Figures D.1 and D.3) (Winter, 2009). Figures D.2 and D.4 illustrate the effects of the filters on the lateral epicondyle marker and ground reaction force.



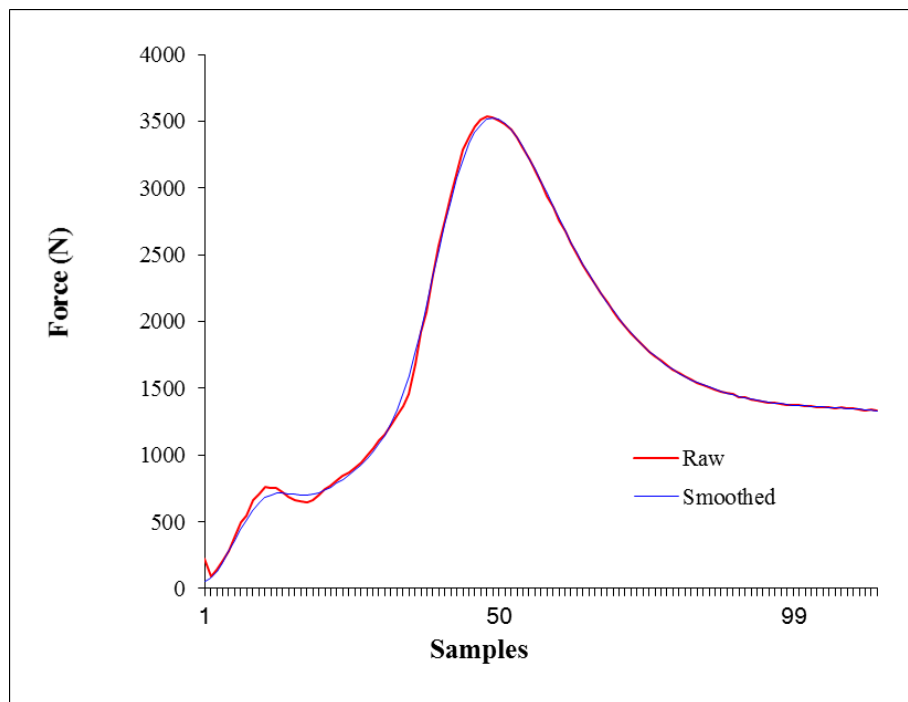
**Figure D.1.** Plot of the residual between a filtered and an unfiltered signal with respect to the filter cut-off frequency (lateral epicondyle marker).



**Figure D.2. Horizontal position of the lateral epicondyle marker during drop-landing. The red line is the unprocessed “raw” data; the blue line is based on the data that has been filtered with a fourth-order zero-lag low-pass Butterworth filter at 9 Hz.**



**Figure D.3.** Plot of the residual between a filtered and an unfiltered signal with respect to the filter cut-off frequency (ground reaction force).



**Figure D.4.** Plot of the ground reaction force during drop-landing. The red line is the unprocessed “raw” data; the blue line is based on the data that has been filtered with a fourth-order zero-lag low-pass Butterworth filter at 50 Hz.

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## Appendix E: Normalization of Landing Phase to 100%

The whole landing phase of each data set was defined as twice the time taken from initial contact with the ground to the time of maximum vertical ground reaction force. The first point for each data set was set as 0% (initial contact) and the last point (twice the time taken from initial contact to the time of maximum vertical ground reaction force) was set as 100% of the landing phase. Data points across participants differ. Spline interpolation was carried out so that all data sets of participants had the same number of points (101 points). The graphs were then plotted based on the average cross all participants at each point of the landing phase. A customized code was written using MATLAB (v. R2014a, The Mathworks Inc., Natick, MA, USA):

```
% MatLab Code
% Clear command window
clear;clc

%% Pext
% Choose file. Must be in .xlsx as stated
filename = uigetfile('*.xlsx', 'Select file to be processed');
% Loading the relevant file.
BF = xlsread(filename, 'PExt');
f = 1;
m = 1;
n = 2;
o = 3;
p = 4;
maxno = 3;
KVAallex = 0;
KVMallex = 0;
VGRFallex = 0;
for i = 1:maxno

% Extract relevant column to process
X = BF(:,m); %Item
```

---

```

KVA = BF(:,n); %Knee Valgus Angle at initial Contact
KVM = BF(:,o); %Maximum Knee Valgus Moment
VGRF = BF(:,p); %Vertical ground reaction force

% Remove 'NaN' no numbers. Else problem with subsequent filtering.
X = X(~any(isnan(X), 2), :);
KVA = KVA(~any(isnan(KVA), 2), :);
KVM = KVM(~any(isnan(KVM), 2), :);
VGRF = VGRF(~any(isnan(VGRF), 2), :);

% To normalize to 0 to 100 as every starting and ending point different.
Xmin = min(X);
Xmax = max(X);
% Interpolation can only be done within the X limits, else becomes
% extrapolation. Converting all to 0 - 100 first.
X = (X-Xmin)*(100/(Xmax-Xmin));

% Interpolation into number of points (set by X2) and by spline.
% Function 'interp1' not 'interp' (Number 'one' not 'L' at the end)
% X and Y(KVA,KVM,VGRF) must have same number of points to plot

X2 = 0:100; %Increment of 1. Standardized for all

KVAout = interp1(X,KVA,X2,'spline');
KVMout = interp1(X,KVM,X2,'spline');
VGRFout = interp1(X,VGRF,X2,'spline');

% Store of data
KVAallex = KVAallex + KVAout;
KVMallex = KVMallex + KVMout;
VGRFallex = VGRFallex + VGRFout;

%f = f+3;
m = m+4;
n = n+4;
o = o+4;
p = p+4;
end % end of for loop

% Mean
KVAmallex = KVAallex/maxno;
KVMmallex = KVMallex/maxno;
VGRFmallex = VGRFallex/maxno;

%% Pint

```

---

```

BF = xlsread(filename, 'PInt');
%f = f+1;
m = 1;
n = 2;
o = 3;
p = 4;
maxno = 3;
KVAallint = 0;
KVMallint = 0;
VGRFallint = 0;
for i = 1:maxno

% Extract relevant column to process
X = BF(:,m); %Item
KVA = BF(:,n); %Knee Valgus Angle at initial Contact
KVM = BF(:,o); %Maximum Knee Valgus Moment
VGRF = BF(:,p); %Vertical ground reaction force

% Remove 'NaN' no numbers. Else problem with subsequent filtering.
X = X(~any(isnan(X), 2), :);
KVA = KVA(~any(isnan(KVA), 2), :);
KVM = KVM(~any(isnan(KVM), 2), :);
VGRF = VGRF(~any(isnan(VGRF), 2), :);

% To normalize to 0 to 100 as every starting and ending point different.
Xmin = min(X);
Xmax = max(X);
% Interpolation can only be done within the X limits, else becomes
% extrapolation. Converting all to 0 - 100 first.
X = (X-Xmin)*(100/(Xmax-Xmin));

% Interpolation into number of points (set by X2) and by spline.
% Function 'interp1' not 'interp1' (Number 'one' not 'L' at the end)
% X and Y(KVA,KVM,VGRF) must have same number of points to plot
X2 = 0:100; % Increment of 1. Standardized for all

KVAout = interp1(X,KVA,X2,'spline');
KVMout = interp1(X,KVM,X2,'spline');
VGRFout = interp1(X,VGRF,X2,'spline');

% Store of data
KVAallint = KVAallint+ KVAout;
KVMallint = KVMallint + KVMout;
VGRFallint = VGRFallint + VGRFout;

```

---

```

%f = f+3;
m = m+4;
n = n+4;
o = o+4;
p = p+4;
end % end of for loop

```

```

% Mean

```

```

KVAmeanint = KVAallint/maxno;
KVMmeanint = KVMallint/maxno;
VGRFmeanint = VGRFallint/maxno;

```

```

%% P

```

```

BF = xlsread(filename, 'P');
%f = f+1;
m = 1;
n = 2;
o = 3;
p = 4;
maxno = 3;
KVAall = 0;
KVMall = 0;
VGRFall = 0;
for i = 1:maxno

```

```

% Extract relevant column to process

```

```

X = BF(:,m); % Item
KVA = BF(:,n); % Knee Valgus Angle at initial Contact
KVM = BF(:,o); % Maximum Knee Valgus Moment
VGRF = BF(:,p); % Vertical ground reaction force

```

```

% Remove 'NaN' no numbers. Else problem with subsequent filtering.

```

```

X = X(~any(isnan(X), 2), :);
KVA = KVA(~any(isnan(KVA), 2), :);
KVM = KVM(~any(isnan(KVM), 2), :);
VGRF = VGRF(~any(isnan(VGRF), 2), :);

```

```

% To normalize to 0 to 100 as every starting and ending point different.

```

```

Xmin = min(X);
Xmax = max(X);
% Interpolation can only be done within the X limits, else becomes
% extrapolation. Converting all to 0 - 100 first.
X = (X-Xmin)*(100/(Xmax-Xmin));
% Interpolation into number of points (set by X2) and by spline.

```

---

```
% Function 'interp1' not 'interpl' (Number 'one' not 'L' at the end)
% X and Y(KVA,KVM,VGRF) must have same number of points to plot
X2 = 0:100; %Increment of 1. Standardized for all
```

```
KVAout = interp1(X,KVA,X2,'spline');
KVMout = interp1(X,KVM,X2,'spline');
VGRFout = interp1(X,VGRF,X2,'spline');
```

```
% Store of data
```

```
KVAall = KVAall + KVAout;
KVMall = KVMall + KVMout;
VGRFall = VGRFall + VGRFout;
```

```
%f = f+3;
m = m+4;
n = n+4;
o = o+4;
p = p+4;
end % end of for loop
```

```
% Mean
```

```
KVAmmean = KVAall/maxno;
KVMmmean = KVMall/maxno;
VGRFmmean = VGRFall/maxno;
```

```
%% Plot overall
```

```
%f = f+1;
figure (f)
plot(X2,KVAmmeanext,'r',X2,KVAmmeanint,'b', X2, KVAmmean,'k');
figure (f+1)
plot(X2,KVMmmeanext,'r',X2,KVMmmeanint,'b', X2, KVMmmean,'k');
figure (f+2)
plot(X2,VGRFmmeanext,'r',X2,VGRFmmeanint,'b', X2, VGRFmmean,'k');
```

```
%% Results Overall
```

```
f2 = 'KVResult';
filename_out = f2;
Tab = 'Ext';
xlswrite(filename_out, 'X2', Tab,'A1:A1');
xlswrite(filename_out, X2, Tab,'B1');
xlswrite(filename_out, 'KVA', Tab,'A2:A2');
xlswrite(filename_out, KVAmmeanext, Tab,'B2');
xlswrite(filename_out, 'KVM', Tab,'A3:A3');
xlswrite(filename_out, KVMmmeanext, Tab,'B3');
xlswrite(filename_out, 'VGRF', Tab,'A4:A4');
```

---

```
xlswrite(filename_out, VGRFmeanext, Tab,'B4');
Tab = 'Int';
xlswrite(filename_out, 'X2', Tab,'A1:A1');
xlswrite(filename_out, X2, Tab,'B1');
xlswrite(filename_out, 'KVA', Tab,'A2:A2');
xlswrite(filename_out, KVAMeanint, Tab,'B2');
xlswrite(filename_out, 'KVM', Tab,'A3:A3');
xlswrite(filename_out, KVMmeanint, Tab,'B3');
xlswrite(filename_out, 'VGRF', Tab,'A4:A4');
xlswrite(filename_out, VGRFmeanint, Tab,'B4');
Tab = 'Forward';
xlswrite(filename_out, 'X2', Tab,'A1:A1');
xlswrite(filename_out, X2, Tab,'B1');
xlswrite(filename_out, 'KVA', Tab,'A2:A2');
xlswrite(filename_out, KVAMean, Tab,'B2');
xlswrite(filename_out, 'KVM', Tab,'A3:A3');
xlswrite(filename_out, KVMmean, Tab,'B3');
xlswrite(filename_out, 'VGRF', Tab,'A4:A4');
xlswrite(filename_out, VGRFmean, Tab,'B4');
```

## Appendix F: Definition of Normality, Sphericity and Effect Size

Assumption of normality tests if residuals are normally distributed (Lund *et al.*, 2015). If

$y_r$  is a normal random variable, the probability distribution of  $y_r$  is

$$f(y_r) = \frac{1}{\sigma\sqrt{2\pi}} e^{-\left(\frac{1}{2}\right)\left[\frac{y_r-\mu}{\sigma}\right]^2} \quad -\infty < y_r < \infty \quad (21)$$

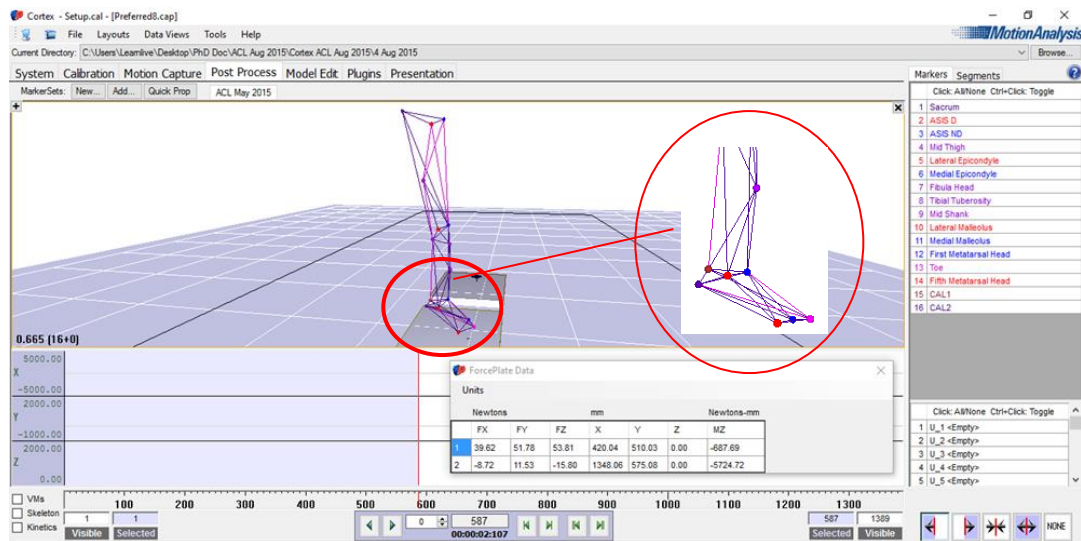
where  $-\infty < \mu < \infty$  is the mean of the distribution and  $\sigma^2 > 0$  is the variance (Montgomery *et al.*, 2009).

Assumption of sphericity requires “all variances of the differences between all combinations of levels of the within-subjects factor” to be the same (Lund *et al.*, 2015, “One-Way Repeated Measures Anova”, p. 7).

Effect size is an estimate of how big the difference is between groups or how big the effect of an intervention is (Portney *et al.*, 2015). It determines if “meaningful change” has taken place (Portney *et al.*, 2015, p. 648).

## Appendix G: Control of Foot-Landing Positions in the Sagittal Plane

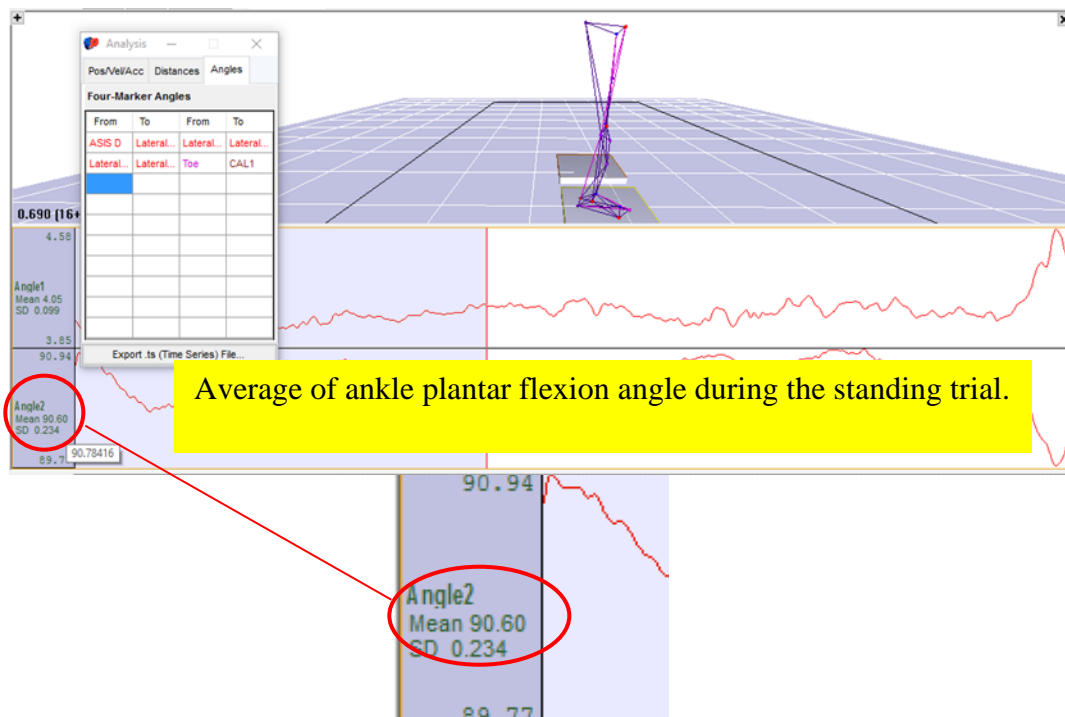
Measures were taken to control the landing technique during the study using three methods. The motion analysis software, Cortex, shows how the person landed based on the marker positions. Firstly, after every trial, the video was analyzed to ensure that fore-foot or flat-foot techniques were used using Cortex (see Figure G.1).



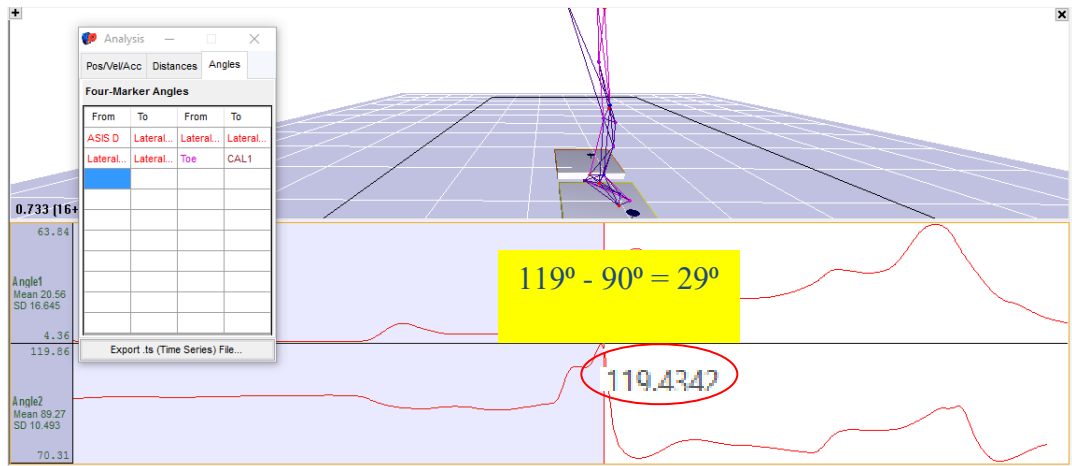
**Figure G.1. Analysis of video to determine fore-foot or flat-foot landing.**

Secondly, to ensure that participants landed in the ankle flexion angle ranges that were requested, 2D data acquisition was made using Cortex after each trial. Markers used to determine the ankle flexion angle through this 2D calculation were the lateral knee epicondyle, lateral malleolus, toe and proximal calcaneus (CAL2). A static standing trial was first collected and the ankle plantar flexion angle was used as each participant's

neutral (zero) alignment. It is at  $90^\circ$  in this example (see Figure G.2). Ankle plantar flexion angle during the drop-landing was measured and was normalized to the standing position. This was estimated by subtracting the ankle flexion angle during the standing trial from the drop-landing trial ( $119^\circ - 90^\circ = 29^\circ$  in this example) (see Figure G.3). This provided an immediate check and feedback to the participants after each drop-landing trial.



**Figure G.2. Ankle plantar flexion angle during the static standing trial.**



**Figure G.3. Ankle plantar flexion angle at initial contact with the ground during the drop-landing trial.**

Thirdly, after data was processed using the Visual3D software, the ankle flexion angles were further checked to ensure that they meet the requirements.

## Appendix H: Comparison of Kinematic Variables between Male Participants from Chapters 4 and 5

One-way Analysis of Variance (ANOVA) was conducted at  $\alpha = 0.05$  to compare kinematic variables between male participants from Chapters 4 and 5. Normality assumption was tested with the Shapiro-Wilk test ( $p > 0.05$ ). Homogeneity of variances was checked using the Levene statistic at  $\alpha = 0.05$ . These assumptions were not violated.

The knee flexion angles ( $p = 0.077$ ), ankle plantar flexion angles ( $p = 0.618$ ) and foot rotation angles ( $p = 0.836$ ) at initial contact, and the knee flexion angles at maximum VGRF ( $p = 0.632$ ) were not found to be significantly different between both groups of male participants. The male participants were therefore combined ( $n = 37$ ). The results are tabulated in Table H.

**Table H. Comparison of Results between Male Participants (n = 37)**

Variables	Chapter 4	Chapter 5	p-Value
<b>At Initial Contact</b>			
<b>Knee Flexion Angles (°) #</b>	13(5)	16(4)	0.077
<b>Plantar Flexion Angles (°) #</b>	29(8)	28(5)	0.618
<b>Foot Rotation Angles (°) #</b>	-20(5)	-19(6)	0.836
<b>At Maximum VGRF<sup>#@</sup></b>			
<b>Knee Flexion Angles (°) #</b>	31(5)	31(5)	0.632

# Positive values denote knee flexion, plantar flexion and internal foot rotation angles.

@ where VGRF refers to vertical ground reaction force.

## Appendix I: Comparison of Results Between Genders

One-way Analysis of Variance (ANOVA) was conducted at  $\alpha = 0.05$  to compare kinematic variables at initial contact between male and female participants. Normality assumption was tested with the Shapiro-Wilk test ( $p > 0.05$ ). Should normality assumption be violated, the non-parametric test, Mann-Whitney U, was conducted instead at  $\alpha = 0.05$ . Homogeneity of variances was checked using the Levene statistic at  $\alpha = 0.05$ . When the variances were unequal, a one-way Welch ANOVA was conducted.

No significant differences were found between genders for the kinematic variables at initial contact (knee flexion angles, ankle plantar flexion angles and foot rotation angles).

The results are tabulated in Table I.

**Table I. Comparison of Results between Genders**

<b>Variables</b>	<b>Males (n = 37)</b>	<b>Females (n = 10)</b>	<b>p-Value</b>
<b>Internal Risk Factors</b>			
<b>BMI<sup>@</sup></b>	22.4(2.0)	19.3(1.7)	<0.001*
<b>GJL<sup>@</sup></b>	1(2)	3(2)	0.117
<b>At Initial Contact</b>			
<b>Knee Flexion Angles (°) <sup>#</sup></b>	15(5)	17(3)	0.186
<b>Plantar Flexion Angles (°) <sup>#</sup></b>	28(6)	30(5)	0.465
<b>Foot Rotation Angles (°) <sup>#</sup></b>	-19(6)	-18(12)	0.674

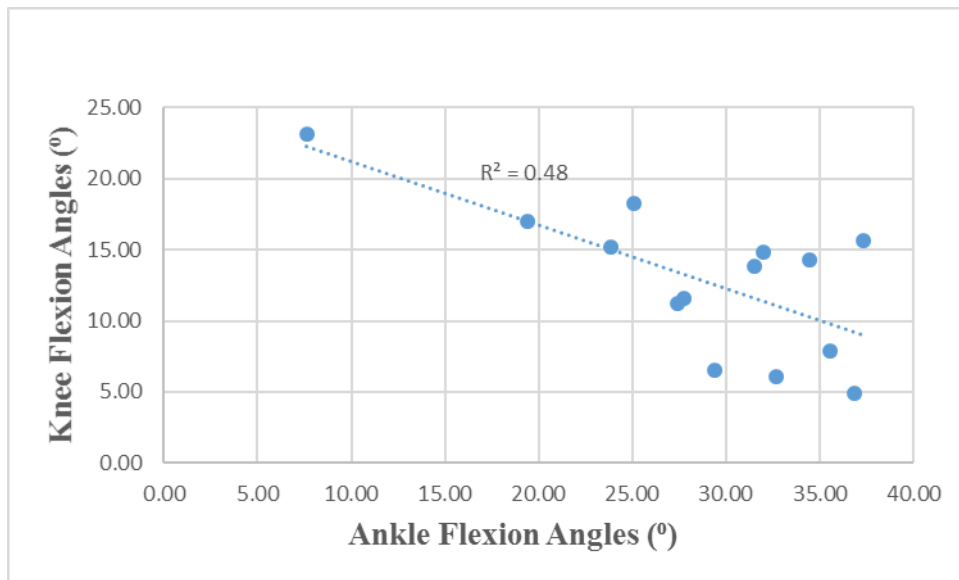
\*Significantly different between low and high knee flexion angles ( $p < 0.05$ ).

<sup>#</sup> Positive values denote knee flexion, plantar flexion and internal foot rotation angles.

<sup>@</sup> where BMI refers to body mass index and GJL refers to generalized joint laxity.

## Appendix J: Linear Regression between Ankle Plantar Flexion Angles and Knee Flexion Angles

A linear regression line between ankle plantar flexion angles and knee flexion angles was plotted. A large effect size between ankle plantar flexion angles and knee flexion angles was obtained ( $r^2 = 0.48$ ), based on guidelines by Cohen (1988).



**Figure J. Plot of a linear regression line between ankle plantar flexion angles (x-axis) and knee flexion angles (y-axis).**

---

## **Appendix K: Comparison of Results Between Low and High Knee Flexion Angle Groups**

One-way Analysis of Variance (ANOVA) was conducted at  $\alpha = 0.05$  to compare variables between low and high knee flexion angle groups. Knee flexion angles were considered to be high if they are more than  $15^\circ$  and low if they are less than  $15^\circ$  at initial contact. The  $15^\circ$  threshold was set because hamstring activation becomes less efficient in reducing ACL loading at less than  $15^\circ$  (Li *et al.*, 1999, Yu *et al.*, 2005). Normality assumption was tested with the Shapiro-Wilk test ( $p > 0.05$ ). Should normality assumption be violated, the non-parametric test, Mann-Whitney U, was conducted instead at  $\alpha = 0.05$ . Homogeneity of variances was checked using the Levene statistic at  $\alpha = 0.05$ . When the variances were unequal, a one-way Welch ANOVA was conducted.

### **K.1 Comparison Between Low and High Knee Flexion Angle Groups for Males**

Participants were divided into low ( $n = 17$ ) and high knee flexion angle ( $n = 20$ ) groups for comparison. Low knee flexion angles are defined as less than  $15^\circ$ . A verification confirmed significant differences in terms of knee flexion angles between the 2 groups (Table K.1). Ankle plantar flexion angles were significantly higher for the group with low knee flexion angles, as compared to the group with high knee flexion angles. Knee flexion angles at maximum VGRF were also significantly lower for the group with low knee flexion angles, as compared to the group with high knee flexion angles. Other results were not found to be significantly different between the 2 groups. The results are tabulated in Table K.1.

**Table K.1. Comparison of Results between Low (n = 17) and High (n = 20) Knee Flexion Angles (Males)**

Variables	Low	High	p-Value
<b>Internal Risk Factors</b>			
<b>BMI<sup>@</sup></b>	22.7(2.6)	22.2(1.3)	0.492
<b>GJL<sup>@</sup></b>	1(2)	2(2)	0.218
<b>At Initial Contact</b>			
<b>Knee Flexion Angles (°) <sup>#</sup></b>	11(3)	18(3)	<0.001*
<b>Plantar Flexion Angles (°) <sup>#</sup></b>	31(5)	25(6)	<0.001*
<b>Foot Rotation Angles (°) <sup>#</sup></b>	-20(5)	-19(6)	0.764
<b>At Maximum VGRF<sup>#@</sup></b>			
<b>Knee Flexion Angles (°) <sup>#</sup></b>	28(5)	34(4)	0.001*
<b>VGRF/BW<sup>@</sup></b>	4.7(0.6)	4.4(0.6)	0.072

\*Significantly different between low and high knee flexion angles ( $p < 0.05$ ).

# Positive values denote knee flexion, plantar flexion and internal foot rotation angles.

@ where BMI refers to body mass index, GJL refers to generalized joint laxity, VGRF refers to vertical ground reaction force and BW refers to body weight.

## **K.2 Comparison of Results Between Low and High Knee Flexion Angle Groups for Males and Females Combined**

Participants were divided into low (n = 20) and high knee flexion angle (n = 27) groups for comparison. Once again, low knee flexion angles are defined as less than 15°. As before, verification confirmed significant differences in terms of knee flexion angles between the 2 groups (Table K.2). Similarly, ankle plantar flexion angles were significantly higher for the group with low knee flexion angles, as compared to the group with high knee flexion angles. Knee flexion angles at maximum VGRF were also

significantly lower for the group with low knee flexion angles, as compared to the group with high knee flexion angles. Other results were not found to be significantly different between the 2 groups. The results are tabulated in Table K.2.

**Table K.2. Comparison of Results between Low (n = 20) and High Knee Flexion Angles (n = 27) (Males and Females)**

Variables	Low	High	p-Value
<b>Internal Risk Factors</b>			
<b>BMI<sup>@</sup></b>	22.1(2.9)	21.5(1.8)	0.438
<b>GJL<sup>@</sup></b>	1(2)	2(2)	0.133
<b>At Initial Contact</b>			
<b>Knee Flexion Angles (°) <sup>#</sup></b>	11(3)	18(2)	<0.001*
<b>Plantar Flexion Angles (°) <sup>#</sup></b>	32(5)	26(5)	<0.001*
<b>Foot Rotation Angles (°) <sup>#</sup></b>	-20(6)	-19(8)	0.726
<b>At Maximum VGRF<sup>#@</sup></b>			
<b>Knee Flexion Angles (°) <sup>#</sup></b>	29(6)	34(4)	0.001*
<b>VGRF/BW<sup>@</sup></b>	4.6(0.6)	4.3(0.5)	0.070

\*Significantly different between low and high knee flexion angles ( $p < 0.05$ ).

<sup>#</sup> Positive values denote knee flexion, plantar flexion and internal foot rotation angles.

<sup>@</sup> where BMI refers to body mass index, GJL refers to generalized joint laxity, VGRF refers to vertical ground reaction force and BW refers to body weight.

---

## Appendix L: Comparison of EMG Between Genders

The non-parametric test, Mann-Whitney U was conducted at  $\alpha = 0.05$  to check if there are any differences in muscle activation levels between genders.

Muscle activation levels were not found to be significantly different for the gastrocnemius ( $p = 0.574$ ), hamstrings ( $p = 0.721$ ) and quadriceps ( $p = 0.192$ ). The results are tabulated in Table L.

**Table L. Comparison of Muscle Activation Levels (Normalized to Maximal Voluntary Contraction) between Genders**

<b>Muscle</b>	<b>Males (n = 37)</b>	<b>Females (n = 10)</b>	<b>p - Value</b>
<b>Gastrocnemius</b>	1.58(0.68)	1.91(0.71)	0.574
<b>Hamstrings</b>	0.47(0.33)	0.53(0.22)	0.721
<b>Quadriceps</b>	1.60(1.15)	2.00(0.46)	0.192

---

## Appendix M: Reasons for Removal of Outlier Points in EMG

### Data

The values of normalized muscle activation to the MVC exceeding 3 were considered as abnormal outlier data since data more than 3 are not found in similar landing studies in literature (Chappell *et al.*, 2007, Pappas *et al.*, 2007, Walsh *et al.*, 2012). Data of normalized muscle activation to the MVC exceeding 3 also coincided with participants of low MVC recordings. Reasons for low MVC recordings were attributed to the following for some of the participants:

- One participant had legs with excessive hair but refused to shave.
- Two participants had flu and were feeling unwell during the study.
- Three were exhausted due to a lack of sleep or had muscle soreness when they arrived at the study.
- EMG sensors were not adhering well to the skin of two participants. This is despite the skin preparation, the extensive taping and use of gauze to hold the sensors in place. It possibly could be that they perspired profusely during the study.
- There was an equipment fault when MVC was measured for one of the participants.

---

## Appendix N: Linear Regression Models between Knee Flexion Angles and Ankle Plantar Flexion Angles

A linear regression was then run to obtain the relationship between knee flexion angles and ankle plantar flexion angles. To check the linearity assumption, a scatterplot of knee flexion angles and ankle plantar flexion angles was plotted. There was also homoscedasticity and normality of the residuals. Effect size was calculated as  $r^2$  and values of 0.01, 0.09 and 0.25 were defined as small, medium and large effects, respectively (Cohen, 1988).

### N.1 Linear Regression Model for Males

The linear regression model was used to predict knee flexion angles ( $\angle$ KF) from ankle plantar flexion angles ( $\angle$ AF):

$$\angle\text{KF} = -0.44 (\angle\text{AF}) + 26.89 \quad (22)$$

Ankle plantar flexion angles significantly predicted knee flexion angles ( $p < 0.001$ ), accounting for 31% of the variation in knee flexion angles with adjusted  $r^2$  of 30%, a large size effect according to Cohen (1988). Predictions made from the model show that ankle plantar flexion angle of  $24^\circ$  would result in knee flexion angles of  $16^\circ$ , 95% CI [15 to 18].

---

**LN.2 Linear Regression Model for Males and Females Combined**

The linear regression model was used to predict knee flexion angles ( $\angle$ KF) from ankle plantar flexion angles ( $\angle$ AF):

$$\angle\text{KF} = -0.43 (\angle\text{AF}) + 27.32 \quad (23)$$

Ankle plantar flexion angles significantly predicted knee flexion angles ( $p < 0.001$ ), accounting for 31% of the variation in knee flexion angles with adjusted  $r^2$  of 30%, a large size effect according to Cohen (1988). Predictions made from the model show that ankle plantar flexion angle of  $25^\circ$  would result in knee flexion angles of  $17^\circ$ , 95% CI [15 to 18].

## Appendix O: Binomial Logistic Regression Models with All Independent Variables Included

Binomial logistic regression for males and combined data of males and females were performed using ankle plantar flexion angles, BMI and GJL. Although only ankle plantar flexion angles were found to be statistically significant ( $p = 0.010$  and  $p = 0.001$ , respectively), the addition of the BMI and GJL improved  $R^2$  to the levels similar to those found in the multiple regression models (Chapter 5). The models were also found to be significant (Table O).

**Table O. Binomial Logistic Regression on Effect of Ankle Plantar Flexion Angles on Likelihood of Obtaining High Knee Flexion Angles**

Model	Significance of Model	Nagelkerke $R^2$	Sensitivity	Specificity
<b>Males (n = 37)</b>				
<b>Ankle Plantar Flexion (°)</b>	0.006*	38%	80%	77%
<b>BMI<sup>@</sup></b>				
<b>GJL<sup>@</sup></b>				
<b>Males and Females (n = 47)</b>				
<b>Ankle Plantar Flexion (°)</b>	<0.001*	47%	85%	75%
<b>BMI<sup>@</sup></b>				
<b>GJL<sup>@</sup></b>				

\*Model is statistically significant ( $p < 0.05$ ).

<sup>@</sup> where BMI refers to body mass index and GJL refers to generalized joint laxity.

---

## **Appendix P: Multiple Regression Model for Males and Females Combined using Participants from Chapter 5 (n = 33)**

The multiple regression model for male and female participants combined was performed to predict knee flexion angles ( $\angle\text{KF}$ ) with ankle plantar flexion angles ( $\angle\text{AF}$ ), as well as, internal factors GJL ( $I_{\text{GJL}}$ ) and BMI ( $I_{\text{BMI}}$ ). The model is described as:

$$\angle\text{KF} = -0.51 (\angle\text{AF}) + 0.47 (I_{\text{GJL}}) - 0.53 (I_{\text{BMI}}) + 40.73 \quad (24)$$

where  $\angle\text{KF}$  and  $\angle\text{AF}$  are at initial contact with the ground.

The prediction model accounted for 36% of the variation in knee flexion angles with adjusted  $R^2$  of 29%, a large size effect according to Cohen (1988) and significantly predicted knee flexion angles ( $p = 0.005$ ). However, only ankle plantar flexion angles ( $p = 0.001$ ) was statistically significant. GJL ( $p = 0.125$ ) and BMI ( $p = 0.081$ ) were not statistically significant, probably because sample size was not sufficient as a sample size of 36 was required, based on the power analysis, performed using G\*Power (v3.1.9.2, Heinrich-Heine University, Düsseldorf, Germany). Fixing GJL as 1 and BMI as 22 (mean values for male participants, see Appendix I), predictions made from the model show that ankle plantar flexion angle of  $26^\circ$  would result in knee flexion angles of  $16^\circ$ , 95% CI [15 to 18] (see Table P.1). This is the maximum ankle plantar flexion angle for males to achieve at least  $15^\circ$  of knee flexion angle, considering the 95% confidence interval.

**Table P.1. Predicted Knee Flexion Angles from Ankle Plantar Flexion Angles at Initial Contact (GJL = 1, BMI = 22 in this Prediction) (Males and Females) #**

Input	Model Prediction	95% Confidence Interval	
		Lower Bound	Upper Bound
Ankle Plantar Flexion Angles (°)	Knee Flexion Angles (°)		
27	16	14*	17
26	16	15	18
25	17	15	18

\* Less than 15° of knee flexion angles at initial contact.

# Sample size n = 33, GJL = generalized joint laxity, BMI = body mass index.

Fixing GJL as 3 and BMI as 19 (mean values for female participants, see Appendix I) instead, predictions made from the model show that ankle plantar flexion angle of 30° would result in knee flexion angles of 17°, 95% CI [15 to 19] (Table P.2). This is the maximum ankle plantar flexion angle to achieve at least 15° of knee flexion angle for females, considering the 95% confidence interval.

**Table P.2. Predicted Knee Flexion Angles from Ankle Plantar Flexion Angles at Initial Contact (GJL = 3, BMI = 19 in this Prediction) (Males and Females) #**

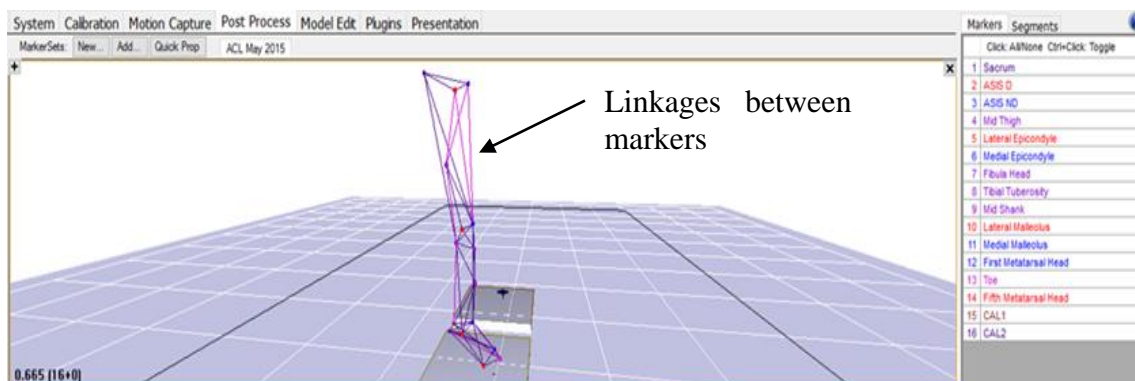
Input	Model Prediction	95% Confidence Interval	
		Lower Bound	Upper Bound
Ankle Plantar Flexion Angles (°)	Knee Flexion Angles (°)		
31	16	14*	18
30	17	15	19
29	17	15	19

\* Less than 15° of knee flexion angles at initial contact.

# Sample size n = 33, GJL = generalized joint laxity, BMI = body mass index.

## Appendix Q: Monitoring of Marker Movement

Marker movement was monitored in the motion analysis software, Cortex software (version 1.1.4.368 and version 2.6.2.1169, Motion Analysis Corp, Santa Rosa, CA, USA). During the set-up of template in the Cortex software, linkages were created between markers (see Figure Q). Should markers be swapped during capture or moved, this could easily be identified as wrongly connected linkages in the motion capture. Furthermore, the Visual3D (v5.00.33, C-Motion, Germantown, PA, USA) software uses the ‘segment optimization’ method’ and additional tracking markers during motion could be placed at areas that minimize soft tissue artefact (Hamill *et al.*, 2014). This software was used in the processing of kinematic and kinetic data. A surgical netting (Surgifix Size 6, FRA Production S.p.A, Cisterna d'Asti (AT), Italy) was placed over the leg to secure the markers in Chapter 5, in addition to the tape that was used to secure them. The netted wrap also further helped reduce any possible skin movements that might result in the soft tissue artifacts.



**Figure G.3. Linkages (coloured lines) between markers to track marker movement.**

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## Appendix R: Removal of $R^2$ in the Graphs of Chapter 6

The main purpose of Figures 29 to 31 was to observe the trend of the anterior tibial translation under different gastrocnemius activation levels:

- Figure 29: To present the overall trend of anterior tibial translation for all the participants, under activation levels ‘low’, ‘contraction’, ‘mid’ and ‘high’ ( $R^2 = 0.98$ ).
- Figure 30: To compare the trend of the anterior tibial translation between Groups A ( $R^2 = 0.98$ ) and B ( $R^2 = 0.99$ ), under activation levels ‘low’, ‘contraction’, ‘mid’ and ‘high’.
- Figure 31: To compare the trend of the anterior tibial translation between Groups A ( $R^2 = 0.97$ ) and B ( $R^2 = 1.00$ ), under the average gastrocnemius activation levels (in mA) set in the EMS, at each level of activation.

To statistically compare the effects of the different gastrocnemius activation levels, a one-way repeated measures Analysis of Variance (ANOVA) was conducted at  $\alpha = 0.05$ . Effect size was also calculated. To minimize effects of fatigue, 4 levels of gastrocnemius activation was used. This results in only 4 data points. Each point was an average of the participants at each level of activation. Therefore, with 4 data points and a curve fitting polynomial of order 2,  $R^2$  might not be providing more useful information.  $R^2$  had also not been used in the analysis. As such,  $R^2$  was not included in the figures to avoid confusion.

## Appendix S: Subject Data

Subject ID	Age (years)	Height (m)	Mass (kg)	BMI	GJL	Max VGRF/Body Weight
3	24	1.75	62.00	20.24	0	4.62
4	24	1.78	73.40	23.17	1	4.31
5	21	1.78	64.70	20.42	0	3.97
7	26	1.81	73.70	22.50	0	4.33
8	22	1.77	69.10	22.06	0	5.09
9	24	1.75	60.80	19.85	0	4.78
11	21	1.84	88.00	25.99	0	4.76
12	23	1.59	60.50	23.93	0	5.03
13	21	1.80	91.00	28.09	0	4.65
15	25	1.67	60.50	21.69	0	3.98
16	25	1.70	64.20	22.21	0	4.33
17	25	1.75	56.70	18.51	0	3.96
19	22	1.66	63.80	23.15	0	4.35
23	24	1.73	60.80	20.31	0	4.97
M1	22	1.71	67.00	22.91	1	4.88
M2	22	1.75	69.00	22.53	0	4.81
M3	23	1.79	72.00	22.47	0	4.73
M4	23	1.72	62.00	20.96	6	4.06
M6	25	1.72	70.00	23.66	3	4.25
M7	24	1.76	73.00	23.57	0	6.16
M8	22	1.68	56.00	19.84	1	4.21
M9	23	1.76	73.00	23.57	3	5.03
M10	24	1.82	78.40	23.67	4	4.92
M11	22	1.84	92.00	27.17	2	4.40
M12	21	1.77	68.00	21.71	4	4.72
M13	23	1.79	63.50	19.82	3	3.30
M14	22	1.78	66.00	20.83	0	4.70
M15	22	1.89	83.00	23.24	6	5.25
M16	22	1.68	63.00	22.32	3	3.37
M17	23	1.75	64.00	20.90	3	4.26
M18	24	1.75	72.00	23.51	0	5.47
M19	32	1.80	80.00	24.69	0	4.09
M20	21	1.78	76.00	23.99	2	5.81
M21	21	1.72	66.00	22.31	5	4.01
M22	21	1.69	60.00	21.01	2	4.13
M23	21	1.83	75.00	22.40	3	4.02
M24	21	1.86	73.00	21.10	0	3.67
F1	22	1.80	67.00	20.68	5	4.42
F2	24	1.59	55.50	21.95	4	4.41
F3	20	1.59	50.00	19.90	3	3.85
F4	22	1.65	52.00	19.10	6	3.64
F5	20	1.62	46.00	17.53	1	4.06
F6	20	1.60	44.00	17.19	0	4.37
F7	20	1.67	50.00	17.93	0	3.56
F8	20	1.61	46.00	17.75	0	4.10
F9	18	1.57	51.00	20.69	4	4.24
F10	21	1.53	48.00	20.50	2	3.87

where BMI = body mass index, GJL = generalized joint laxity measured using the Beighton score, Max VGRF = maximum vertical ground reaction force.

## Appendix T: Data for Chapter 3

Subject	Foot Rotation Angles (°)						Knee Valgus Angles at Initial Contact (°)			Maximum Knee Valgus Moments (N-m/kg/m)		
	With respect to pelvis			With respect to shank			Toe-out	Toe-in	Toe-forward	Toe-out	Toe-in	Toe-forward
	Toe-out	Toe-in	Toe-forward	Toe-out	Toe-in	Toe-forward						
3	-22.74	-6.81	-17.32	5.36	6.90	6.28	4.09	2.72	3.65	-1.43	-1.24	-1.27
4	-33.52	-19.55	-20.17	7.12	4.10	8.57	0.29	-0.15	0.85	-1.06	-0.58	-0.70
5	-23.39	-7.38	-11.68	2.14	3.48	4.94	0.08	-2.36	-0.52	-0.62	-0.40	-0.45
7	-34.48	-5.28	-22.49	7.68	6.66	8.43	-2.09	-2.94	-2.11	-0.68	-0.41	-0.78
9	-28.81	-8.37	-22.28	8.29	8.82	6.98	1.01	-0.16	0.80	-0.75	-0.73	-0.68
11	-27.22	-8.79	-18.54	9.17	9.06	9.86	-3.79	-5.02	-4.19	-0.82	-0.44	-0.67
12	-37.82	-26.96	-29.74	2.29	-0.07	-1.61	-2.67	-3.58	-2.18	-0.73	-0.49	-0.75
15	-23.80	-10.95	-19.27	6.89	5.97	5.98	1.08	1.49	1.71	-1.06	-1.03	-0.81
16	-39.55	-4.35	-21.22	7.50	7.87	6.87	-6.24	-6.79	-6.89	-0.42	-0.40	-0.32
19	-27.14	-9.26	-14.44	5.23	4.49	4.72	2.69	1.60	2.28	-0.85	-0.53	-0.66
23	-26.96	-6.47	-24.01	4.85	5.56	2.59	-1.38	-2.42	-1.67	-0.55	-0.32	-0.31

## Appendix U: Kinematics Data

Subject ID	Knee Valgus		Knee Flexion Angles (KF), Ankle Plantar Flexion (AF) and Foot Rotation Angles (FR)			
	KVA Initial Contact (°)	Max KVM (Nm/kgm)	KF Max VGRF (°)	AF Angle (°)	KF Angle (°)	FR (°)
3	3.65	-1.27	32.08	19.39	16.98	-17.32
4	0.85	-0.70	34.63	25.08	18.29	-20.17
5	-0.52	-0.45	32.79	31.99	14.81	-11.68
7	-2.11	-0.78	34.80	37.30	15.68	-22.49
8	-4.31	-0.55	26.10	27.74	11.63	-22.81
9	0.80	-0.68	28.28	27.43	11.26	-22.28
11	-4.19	-0.67	21.15	29.42	6.51	-18.54
12	-2.18	-0.75	30.09	35.53	7.84	-29.74
13	-6.04	-0.64	20.39	36.84	4.94	-13.19
15	1.71	-0.81	33.38	7.63	23.11	-12.74
16	-6.89	-0.32	34.73	23.85	15.15	-21.22
17	-0.23	-0.53	38.27	34.47	14.28	-25.69
19	2.28	-0.66	35.34	31.53	13.86	-14.44
23	-1.67	-0.31	26.32	32.66	6.12	-24.01
M1	-1.76	-0.64	30.97	20.07	17.88	-5.81
M2	2.31	-1.17	29.25	33.67	8.36	-19.68
M3	4.11	-0.69	31.74	26.81	16.66	-25.65
M4	3.96	-0.83	33.10	38.76	10.34	-17.94
M6	-1.26	-0.54	34.53	28.04	19.59	-17.57
M7	2.26	-0.97	19.94	26.85	9.00	-17.87
M8	3.97	-0.93	31.07	27.73	17.15	-17.52
M9	-2.22	-0.85	21.57	34.37	10.94	-27.66
M10	0.97	-0.62	26.45	24.33	17.41	-23.89
M11	-0.91	-0.48	27.41	14.90	14.20	-20.67
M12	2.86	-0.63	38.49	26.51	25.83	-29.33
M13	1.24	-0.60	41.61	28.16	18.32	-16.09
M14	-1.78	-0.63	35.64	24.53	16.09	-14.04
M15	3.66	-0.89	26.76	25.66	18.43	-16.59
M16	0.58	-0.37	40.54	31.22	17.80	-27.72
M17	2.77	-0.64	29.17	28.86	11.06	-16.94
M18	0.57	-0.67	27.89	28.06	17.47	-27.61
M19	-0.51	-0.56	33.20	26.48	20.22	-11.07
M20	-0.60	-0.97	29.33	30.62	14.63	-14.44
M21	4.37	-0.79	38.89	28.83	15.31	-22.77
M22	5.02	-0.86	29.94	30.87	9.34	-18.90
M23	-2.45	-0.36	35.61	24.76	18.57	-19.70
M24	2.78	-1.03	31.26	24.23	17.53	-15.37
F1	-0.07	-0.55	35.11	28.34	19.50	-8.35
F2	5.16	-0.81	31.98	23.11	21.46	-31.03
F3	2.29	-0.81	37.83	30.15	18.44	-19.81
F4	-9.13	-0.28	36.58	23.18	18.97	1.73
F5	4.56	-0.84	35.09	36.97	14.71	-27.34
F6	0.78	-0.67	37.56	28.67	16.17	-14.94
F7	-4.06	-0.40	40.94	34.48	13.17	-4.16
F8	3.24	-0.65	34.51	25.45	18.68	-16.90
F9	0.98	-0.70	33.21	37.63	9.97	-23.45
F10	5.22	-0.95	35.14	30.94	17.25	-34.11

where KVA = knee valgus angles and Max KVM = maximum knee valgus moments

## Appendix V: Electromyography Data

Subject ID	EMG ( $\mu\text{V}$ )			MVC ( $\mu\text{V}$ )			EMG/MVC		
	Gastrocnemius (G)	Biceps Femoris (BF)	Vastus Lateralis (VL)	G_MVC	BF_MVC	VL_MVC	G_N	BF_N	VL_N
M1	135.38	56.08	232.69	123.34	52.80	403.25	1.10	1.06	0.58
M6	78.69	20.42	149.53	46.78	141.25	191.23	1.68	0.14	0.78
M7	69.92	31.63	81.20	53.49	54.19	40.65	1.31	0.58	2.00
M10	61.24	20.34	230.53	46.22	66.49	63.23	1.32	0.31	3.65
M11	94.13	21.63	186.33	40.50	78.79	143.30	2.32	0.27	1.30
M14	106.35	18.23	221.05	103.98	76.66	194.15	1.02	0.24	1.14
M15	169.75	20.28	143.05	208.44	65.30	187.80	0.81	0.31	0.76
M17	234.92	21.02	124.48	85.06	91.68	159.27	2.76	0.23	0.78
M18	120.32	53.94	186.56	87.00	59.58	49.28	1.38	0.91	3.79
M19	79.70	11.43	63.47	67.92	72.57	42.06	1.17	0.16	1.51
M20	210.17	75.90	146.45	206.83	83.51	124.37	1.02	0.91	1.18
M21	108.26	10.86	73.05	95.00	99.35	140.33	1.14	0.11	0.52
M22	188.53	58.26	505.05	78.43	76.21	154.32	2.40	0.76	3.27
M23	157.31	42.02	173.62	57.66	75.24	181.38	2.73	0.56	0.96
F2	68.23	21.16	85.01	74.98	89.04	43.39	0.91	0.24	1.96
F3	246.70	27.09	79.67	101.05	42.61	45.32	2.44	0.64	1.76
F8	84.58	29.37	173.96	44.57	61.23	107.33	1.90	0.48	1.62
F10	136.79	18.09	123.34	56.98	24.06	46.50	2.40	0.75	2.65

where EMG = electromyography, MVC = maximal voluntary contraction

## Appendix W: Muscle Activation and Anterior Tibial Translation Data

Subject ID	Age (years)	Height (m)	Mass (kg)
1	21	1.69	60.00
2	21	1.72	66.00
3	21	1.83	75.00
4	21	1.78	76.00
5	23	1.79	72.00
6	21	1.86	73.00
7	22	1.76	80.00
8	32	1.80	80.00
9	28	1.78	70.00
10	24	1.75	72.00

Subject	Muscle Activation (mA)						Anterior Tibial Translation (mm)			
	Quadriceps	Low	Contraction	Mid	High		Low	Contraction	Mid	High
1	28.00	18.00	26.00	27.00	27.50		0.51	0.88	0.78	0.90
2	36.00	22.00	34.00	37.00	40.00		3.25	2.90	3.03	1.85
3	24.00	17.00	24.00	28.00	36.00		0.63	0.70	1.03	0.73
4	25.50	20.00	27.00	30.00	33.50		0.24	0.10	0.03	0.15
5	25.00	36.00	39.00	41.00	45.00		0.47	0.50	0.40	0.00
6	57.00	28.00	35.00	50.00	62.00		1.14	2.51	1.88	0.90
7	18.00	14.00	18.00	20.00	23.00		0.25	0.65	1.60	1.70
8	50.00	27.00	32.00	36.00	38.00		0.90	1.05	1.08	1.08
9	59.00	16.00	31.00	36.00	41.00		1.80	1.10	1.05	1.10
10	30.00	15.00	25.00	27.00	30.00		0.39	0.59	0.47	1.29

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## **Appendix X: Journal Submission to ‘The Knee’**

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The Knee



## Effects of foot rotation positions on knee valgus during single-leg drop landing: Implications for ACL injury risk reduction

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### ARTICLE INFO

*Article history:*  
Received 21 September 2016  
Received in revised form 9 January 2017  
Accepted 13 January 2017

*Keywords:*  
Non-contact  
Moments  
Ankle  
Kinematics  
Injury prevention  
Initial contact

### ABSTRACT

**Background:** Non-contact anterior cruciate ligament (ACL) injuries commonly occur when athletes land in high risk positions such as knee valgus. The position of the foot at landing may influence the transmission of forces from the ankle to the knee. Using an experimental approach to manipulate foot rotation positions, this study aimed to provide new insights on how knee valgus during single-leg landing may be influenced by foot positions.

**Methods:** Eleven male recreational basketball players performed single-leg drop landings from a 30-cm high platform in three foot rotation positions (toe-in, toe-forward and toe-out) at initial contact. A motion capture system and a force plate were used to measure lower extremity kinematics and kinetics. Knee valgus angles at initial contact (KVA) and maximum knee valgus moments (KVM), which were known risk factors associated with ACL injury, were measured. A one-way repeated measures Analysis of Variance was conducted ( $\alpha = 0.05$ ) to compare among the three foot positions.

**Results:** Foot rotation positions were found to have a significant effect on KVA ( $p < 0.001$ ,  $\eta^2 = 0.66$ ) but the difference between conditions (about  $1^\circ$ ) was small and not clinically meaningful. There was a significant effect of foot position on KVM ( $p < 0.001$ ,  $\eta^2 = 0.55$ ), with increased moment observed in the toe-out position as compared to toe-forward ( $p = 0.012$ ) or toe-in positions ( $p = 0.002$ ).

**Conclusions:** When landing with one leg, athletes should avoid extreme toe-out foot rotation positions to minimise undesirable knee valgus loading associated with non-contact ACL injury risks.

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### 1. Introduction

Anterior cruciate ligament (ACL) injuries commonly occur during sports such as basketball, soccer, and Australian Football [1,2]. These injuries may result in long absence from play [3] and thus negatively impact the careers of athletes. ACL injuries often happen in manoeuvres such as sidestepping and single-leg landing during non-contact situations [4], in which players are not hit by an object or another player [1]. Instead, from the video analyses of ACL injuries, athletes frequently land in high risk positions such as knee valgus positions [5–7] whereby one or both knees are seen moving towards the midline of the body [8]. In the prospective cohort study of female athletes by Hewett et al. [9], athletes who sustained ACL injuries landed at

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higher knee valgus angles (KVA) at initial contact during a drop vertical jump, as compared to the uninjured athletes. In addition, their study showed that knee valgus moments (KVM) predicted ACL injuries with 73% specificity and 78% sensitivity [9]. Among badminton players, higher KVA and KVM were also associated with higher ACL injury risks [10]. Based on a stochastic biomechanical model, Lin et al. [11] similarly considered high KVM as a risk factor of non-contact ACL injury. Using the magnetic resonance imaging scan, Speer et al. [12] further supported valgus as one of the mechanisms of ACL injuries. Collectively, these studies suggest that higher KVA at initial contact and greater KVM could increase the risks of ACL injury. To reduce these risks, knee valgus positions should therefore be avoided when landing from a jump.

Although ACL injury prevention has been studied extensively, there is still a rising trend of the occurrence of ACL injuries [4]. Most studies on ACL injuries examined the knee [9,13], hip [14,15], or trunk [16,17]. Since the foot and ankle form the initial parts of the lower extremity kinetic chain during landing, the position of the foot may influence the transmission of forces from the ankle to the knee [18]. Thus, studying ACL injuries from the foot perspective may provide new knowledge on preventive strategies for non-contact ACL injuries.

In the literature, very few studies have considered the role of the foot in influencing ACL injuries [19–24]. Ishida et al. [21] studied the effect of foot rotation positions on knee rotation during knee valgus. Their protocol which requested participants to move from a standing position to a knee valgus position, however, might not sufficiently simulate the loading during dynamic landing [21]. Cortes et al. [19] studied the effects of sagittal foot-landing positions (fore-foot, rear-foot and preferred) on lower extremity kinematics during double-leg landing and found that knee flexion angle was significantly lower at initial contact for fore-foot landing. Tran and colleagues [23] found differences in hip and knee biomechanics during double-leg landing in toe-in and toe-out positions in comparison to a neutral reference. Since unilateral and bilateral landings are characterised by different knee kinematics and muscle activation levels [25], the results reported by previous studies [19,22–24] on double-leg landing may not be applicable for single-leg landings whereby more non-contact ACL injuries occur [4]. One study on single-leg landing found that maximum knee valgus moment was correlated with toe-out foot position [20]. This study, however, did not compare different foot rotation positions such as toe-forward and toe-in positions, or examine the ACL risk factors at the initial contact of landing whereby ACL strain was the highest [26]. Thus, it will be useful to further examine the influence of foot rotation positions on ACL risk factors both at initial contact and during landing.

The purpose of this study was therefore to investigate the effects of foot rotation positions (toe-in, toe-forward, toe-out) on knee valgus, during single-leg drop-landing. Based on the correlation results reported by Dempsey et al. [20], it was hypothesized that the toe-out position would be characterised by greater knee valgus angles (KVA) at initial contact and higher maximum knee valgus moments (KVM) during landing. These would translate into increased risks of non-contact ACL injury.

## 2. Methods

### 2.1. Participants

A power analysis was performed using G\*Power (v3.1.9.2, Heinrich-Heine University, Düsseldorf, Germany) based on the results (large effect size in the correlation between external foot rotation and maximum KVM) reported by Dempsey et al. [20]. Using a large effect size of 0.40 for a one-way repeated measures Analysis of Variance [27], a sample size of 12 was required ( $\beta = .80$ ,  $\alpha = .05$ ). Fifteen participants were initially recruited but data of four participants had to be discarded: one dropped out halfway of the study; the other three had incomplete data as markers were not adhering well due to their intense perspiration. Thus, data of the remaining 11 male recreational basketball players (age: mean 23.6 (standard deviation (SD) 1.6) years old; height: mean 1.73 (SD 0.07) m; body mass: mean 66.6 (SD 8.6) kg) were used in this study. Participants had no prior ACL injuries or lower extremity injuries in the last six months, and played basketball at least once a week for a minimum of two years prior to the study. The study was approved by the Institutional Review Board of the university. All participants signed a written consent form before the study began.

### 2.2. Equipment

The same model of basketball shoes (Nike Zoom Attero, Nike, Inc., Beaverton, OR, USA) was provided to all participants to eliminate the influence of footwear. Retroreflective markers were attached to key body landmarks of the dominant leg using tape (Leukoplast®, BSN medical GmbH, Hamburg, Germany). The dominant leg was self-reported by the participants as the leg that kicked the ball the hardest. The landmarks included sacrum, anterior superior iliac spine, greater trochanter of femur, mid-thigh, medial and lateral knee epicondyles, tibial tuberosity, head of fibular, anterior aspect of shin, medial and lateral malleoli, bottom of the calcaneus (CA1), Achilles tendon insertion point to calcaneus (CAL2), head and base of first metatarsal, head and base of fifth metatarsal and toe (between second and third metatarsal heads). Markers were also attached to the anterior superior iliac spine and greater trochanter of the non-dominant leg for the estimation of the hip joint location. Three-dimensional motion data were captured at 200 Hz using eight digital cameras and the Cortex software (version 1.1.4.368, Motion Analysis Corp, Santa Rosa, CA, USA). Ground reaction forces (GRF) were recorded using a force plate (Kistler type 9287CA, Winterthur, Switzerland) at a sampling rate of 1000 Hz in synchronization with the cameras. Kinematic and kinetic data were further processed using Visual3D (v5.00.33, C-Motion, Germantown, PA, USA).

### 2.3. Test protocol

This study investigated the biomechanics of single-leg drop-landing by experimentally manipulating foot rotation positions. Participants were asked to carry out a warm-up exercise for five minutes and were given time to practise the drop-landing manoeuvres before data collection commenced. Participants were requested to stand with their dominant leg on a 30 cm-high platform, and drop land onto a force plate using the same leg. Drop-landings were performed in three foot rotation positions presented in a randomized manner: toe-forward, toe-in and toe-out (Figure 1). Participants were requested to face a board, marked with a red cross and hung on a wall at their eye-level, while landing. Upon landing on the force platform with the dominant leg, participants were required to maintain the position for two seconds. A trial was considered successful when the participants dropped off from the raised platform without jumping up, and landed with the entire foot on the force plate, and maintained balance for at least two seconds. Three successful trials of each condition were collected.

### 2.4. Kinematic and kinetic analyses

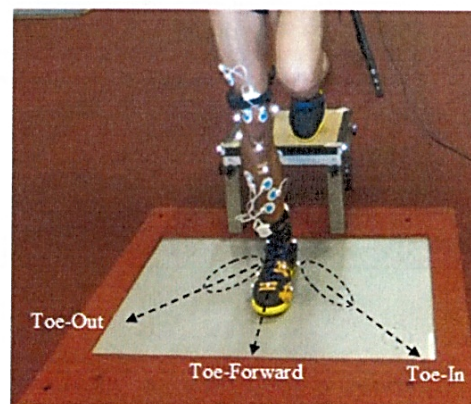
Visual3D (v5.00.33, C-Motion, Inc., Germantown, MD, USA) was used to build a model consisting of the pelvis, thigh, shank and foot segments. The CODA model (Charnwood Dynamics Ltd., Leicestershire, UK) was used to construct the pelvis segment and the hip joint centre was derived using equations by Bell and Brand [28]. The sacrum marker was input as both the right and left posterior-superior iliac spine markers in Visual3D. The markers at the hip joint centre, and lateral and medial knee epicondyles, were used to define the frontal plane of the thigh segment. A virtual landmark was created by projecting the tibial tuberosity marker onto the plane created by the head of fibular, and lateral and medial malleoli markers. Using this virtual landmark, the frontal plane of the shank segment was defined as proximal joint centre and the mid-point of the lateral and medial malleoli markers as distal joint centre. Finally, a single-segment foot model was created using CAL2 as the proximal joint centre and the toe marker as the distal joint centre of the sagittal plane. CA1 was used to define the posterior orientation. For all the segments, the frontal plane determined the orientation of the X axis and the sagittal plane determined the orientation of the Y axis. The Z axis was orthogonal to both X and Y axes. Cardan/Euler rotation sequence was used for the knee joint angle calculation.

The motion and GRF data were filtered through a low-pass Butterworth digital filter at cut-off frequencies of 9 Hz and 50 Hz, respectively. Initial contact was defined as the first instant that vertical GRF exceeded 10 N. For each landing trial, two variables were calculated to indicate ACL injury risk: (1) KVA at initial contact, and (2) maximum KVM during the whole phase of landing. The whole landing phase was defined as twice the time taken from initial contact with the ground to the time of maximum vertical GRF, according to Dempsey et al. [20]. Maximum KVM was normalised to the product of each participant's height (m) and mass (kg) [20].

To verify that participants had landed in three distinct foot rotation positions as instructed, foot rotation angles at initial contact were measured with respect to the shank and also the pelvis.

### 2.5. Statistical analysis

The variables of interest include KVA at initial contact, maximum KVM and foot rotation angles. Values of the three trials per foot rotation condition were averaged and then used in statistical analysis. Normality assumption was first checked with the



**Figure 1.** Three foot rotation positions: toe-out (away from the mid-line of the body), toe-forward, and toe-in (towards the midline of the body).

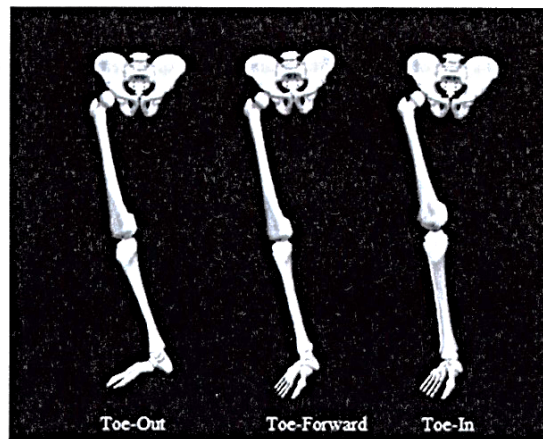


Figure 2. Representations of actual foot rotation positions that participants landed in: toe-out, toe-forward, and toe-in.

Shapiro-Wilk test ( $p > 0.05$ ). If the data were not normally distributed, "reflect and logarithmic" transformation was applied. To compare among the foot rotation positions, a one-way repeated measures Analysis of Variance was conducted at  $\alpha = 0.05$  (IBM SPSS 22.0 – IBM Corp., NY, USA) for each variable of interest. Assumption of sphericity was tested by Mauchly's test of sphericity. When assumption of sphericity was not met, a Greenhouse-Geisser correction was applied. Post-hoc test with Bonferroni adjustment was made with a family-wise  $\alpha = 0.05$ . Effect size was calculated as  $\eta^2$  and values of 0.01, 0.09 and 0.25 were interpreted as small, medium and large effects, respectively [29]. Data are expressed as mean (SD).

### 3. Results

Figure 2 illustrates the typical postures when participants were asked to land in toe-in, toe-forward, and toe-out positions. Foot rotation angles, when measured with respect to the shank, did not differ among the three foot positions ( $p = 0.810$ , Table 1). When measured with respect to the pelvis orientation, significant differences were noted ( $p < .001$ , Table 1). Post-hoc results showed that toe-in position was significantly less negative (i.e. less outward) than toe-forward position ( $p = 0.004$ ). Similarly, toe-forward position was significantly less negative (i.e. less outward) than the toe-out position ( $p < 0.001$ ).

Plotted over the entire landing phase, Figure 3 shows the mean ensemble curves of the vertical GRF, normalised to the body weight, KVA and KVM. There was a significant effect of foot position on KVA at initial contact ( $p < .001$ , Table 1). However, the actual difference in angles was small (toe-in vs. toe-forward:  $1^\circ$ , 95% confident intervals (CI) [0 to 1],  $p = 0.001$ ; toe-in vs. toe-out:  $1^\circ$ , 95% CI [0 to 2],  $p = 0.003$ ). Maximum KVM significantly differed among the three foot positions ( $p < 0.001$ , Table 1). Post-hoc tests showed greater KVM in the toe-out position, as compared to toe-forward ( $p = 0.012$ ) or toe-in conditions ( $p = 0.002$ ). No post-hoc difference in KVM was found between the toe-in and toe-forward positions ( $p = 0.442$ ).

Table 1  
Foot and knee biomechanics of toe-in, toe-forward and toe-out single-leg landings.

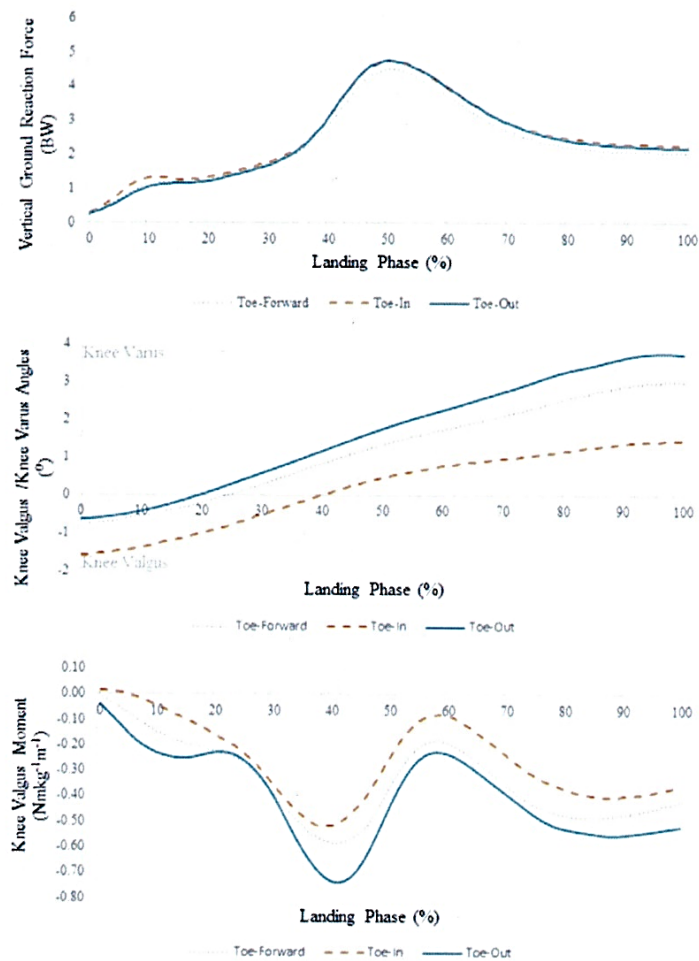
Variables <sup>1</sup>	Toe-in	Toe-forward	Toe-out	p-Value	$\eta^2$
Foot rotation angles <sup>1</sup>					
With respect to shank ( $^\circ$ )	6 (3)	6 (3)	6 (2)	0.810	0.02
With respect to pelvis ( $^\circ$ )	-10 (7) <sup>bc</sup>	-20 (5) <sup>ac</sup>	-30 (6) <sup>ab</sup>	<0.001	0.74
Knee valgus angles at initial contact ( $^\circ$ )	-2 (3) <sup>bc</sup>	-1 (3) <sup>a</sup>	-1 (3) <sup>a</sup>	<0.001	0.66
Maximum knee valgus moments (Nm kg <sup>-1</sup> m <sup>-1</sup> )	-0.60 (0.29) <sup>c</sup>	-0.67 (0.26) <sup>c</sup>	-0.82 (0.28) <sup>ab</sup>	<0.001	0.55

<sup>1</sup> Negative denotes toe-out or knee valgus directions. Effect sizes ( $\eta^2$ ) of 0.01, 0.09 and 0.25 were interpreted as small, medium and large effects, respectively.

<sup>a</sup> Significantly different from toe-in position.

<sup>b</sup> Significantly different from toe-forward position.

<sup>c</sup> Significantly different from toe-out position.



**Figure 3.** Mean vertical ground reaction force (normalized to the body weight (BW)), knee valgus angles and knee valgus moments (normalized to the product of height and body mass) under toe-forward, toe-in and toe-out foot rotation conditions. The whole landing phase was defined as twice the time taken from initial contact with the ground to the time of maximum vertical ground reaction force.

#### 4. Discussion

The main objective of this study was to investigate the effects of foot rotation positions (toe-in, toe-forward, toe-out) on knee valgus during single-leg drop-landing. To our best knowledge, this is the first study to experimentally manipulate various foot rotation positions during single-leg landings. This provides a better understanding as to how knee biomechanics can be influenced by foot orientations. In support of our hypothesis, the toe-out position resulted in a significant increase in maximum KVM during landing. Foot rotation also had a significant effect on KVA at initial contact but the difference was too small to be practically meaningful.

##### 4.1. Effects of foot rotation on KVM

The present study showed that toe-out foot rotation position resulted in significantly higher maximum KVM than toe-in and toe-forward positions. This finding confirmed the results of the study by Dempsey et al. [20] which found a significant correlation

between increased KVM and external foot rotation. Since higher KVM is associated with increased ACL injury risks [9,10,11], athletes should be advised to avoid landing in extreme toe-out positions during single-leg landing to minimise ACL injury risks. Given the lack of differences in KVM between the toe-in and toe-forward positions, landing within this range of foot rotation (about 10°, Table 1) would likely result in similar maximum KVM. These findings suggest that there is no detrimental effect on knee valgus loading when landed in a toe-in compared to toe-forward positions. Thus, it may not be necessary to emphasize the 'toe-forward' position as previously perceived [30].

During single-leg landings, the knees are generally at less than 20° of flexion at initial contact [31]. Within this range of small knee flexion angles, the "screw-home" mechanism can take place whereby the tibia externally rotates on the femur [32]. Furthermore, one previous study demonstrated that the toe-out foot rotation was associated with the tibia being more externally rotated with respect to the femur [21]. The medial collateral ligament (MCL) helps stabilise the knee joint by preventing it from going into knee valgus positions and by restraining external tibial rotation [33]. In the toe-out foot rotation position with more external tibial rotation, the MCL could have been twisted [34] such that it could buckle more easily to constrain less of KVM. This might in turn increase the loading of the ACL, which has the secondary role of restraining knee valgus/varus loading [35]. Thus, landing in a toe-out position may thereby increase ACL injury risks during single-leg landing.

The maximum KVM values ( $-0.82$  to  $-0.60$  Nm kg<sup>-1</sup> m<sup>-1</sup>) obtained in our study were quite different from those reported in previous studies on single-leg landing using male subjects [20,36]. Dempsey et al. [20] obtained lower KVM mean values ranging from  $-0.43$  (95% CI [ $-0.53$ ,  $-0.33$ ]) to  $-0.23$  (95% CI [ $-0.30$ ,  $-0.16$ ]) Nm kg<sup>-1</sup> m<sup>-1</sup>, for different landing tasks. Participants had to take off, catch a ball, and land with their preferred leg [20]. Their landing heights may differ from the fixed 30 cm in our study, resulting in lower range of KVM. On the other hand, Hong et al. [36] used a similar protocol of single-leg drop landing from a 30-cm platform and obtained a mean maximum KVM of  $-0.16$  Nm kg<sup>-1</sup> m<sup>-1</sup>, 95% CI [ $-0.20$ ,  $-0.13$ ] using shod condition. Their different marker sets, especially without the use of the sacrum and anterior superior iliac spine markers, might have resulted in different maximum KVM values [36]. Since this study adopted a within-participant design, comparisons among the three foot rotation positions were made only within the same individual. Thus, the conclusion obtained would be robust regarding the influence of foot rotation positions, despite having different KVM values compared with previous studies.

#### 4.2. Effects of foot rotation on KVA

This study also investigated the effects of foot rotation on KVA at initial contact and found significant but very small differences ( $-1^\circ$  to  $-2^\circ$ ) among the three levels of foot rotation. Similar values of KVA were reported during single-leg landing ( $1^\circ$ , 95% CI [ $-1$  to  $2$ ]) [31], and in the video analysis of basketball games during initial contact ( $-3^\circ$ , 95% CI [ $-2$  to  $-4$ ]) [5]. Despite the statistically significant difference and large effect size ( $\eta^2 = 0.66$ ), the actual differences in angles are too small to be practically meaningful. Therefore, foot rotation does not substantially influence the risk of ACL injuries at initial contact of single-leg landings.

#### 4.3. Foot rotation manipulation protocol

In the verification of foot rotation angles, foot rotation with respect to the shank did not differ among the three conditions. This suggests that the foot and shank tend to rotate together as one unit. Participants likely achieved the toe-in/out positions via internal/external rotations at the knee and hip joints. Ishida et al. [21] also demonstrated that toe-out foot rotation position resulted in a more externally rotated tibia with respect to the femur, whereas toe-in foot rotation position resulted in a more internally rotated tibia.

As shown in Table 1, the foot rotation angles with respect to pelvis progressively became more negative (outwards) from toe-in to toe-forward and then toe-out positions. It is interesting to note that all three foot rotation angles, including the toe-in position, were negative. This implies that despite the verbal instruction to land with various foot rotation conditions (Figure 1), participants all landed in negative, toe-out positions (Figure 2). This observation is somewhat unexpected; suggesting that toe-out foot rotation could be a natural landing position of these participants.

#### 4.4. Limitations

There were a few limitations to our study. Firstly, only the dominant leg was examined in our study and future studies could also look at the non-dominant leg. With a higher reliance on the dominant leg, it may be exposed to higher forces and moments during high loading circumstances [37]. On the other hand, the non-dominant leg may be less trained to dissipate loading and moments [37]. Thus, both dominant and non-dominant legs can be susceptible to ACL injuries. Secondly, only foot rotation angles with respect to knee valgus were studied. A simultaneous manipulation of sagittal and frontal plane foot landing positions (e.g. [24]) could also be carried out. This could provide a more comprehensive understanding of foot-landing effects on non-contact ACL injury prevention. Thirdly, this study was carried out in a controlled laboratory environment using standardised drop-landing tasks. It is possible that the desirable effects of toe-in and toe-forward foot positions on maximum KVM might be less pronounced in actual playing conditions. A prospective intervention study could be conducted on a cohort, together with a control group, to provide insights as to whether the training of foot-landing techniques can help in reducing non-contact ACL injuries over a period of time.

## 5. Conclusions

This study showed that foot rotation positions can influence knee joint loading during single-leg drop landing. Specifically, landing in toe-out positions resulted in significantly higher maximum KVM compared with toe-in and toe-forward positions. Although foot rotation positions also influenced KVA at initial contact, the differences were too small to be clinically meaningful. When landing with one leg, athletes should avoid extreme toe-out foot rotation positions to minimise undesirable knee valgus loading associated with non-contact ACL injury risks.

## Grants and funding

This study was supported by the Institute for Sports Research at the Nanyang Technological University.

## Conflicts of interest

The authors declare that there is no conflict of interest.

## Acknowledgments

The authors would like to thank Gu Kaiyi and Bai Jinhao for their contribution in data collection.

## References

- [1] Marsall SW, Padua D, McGrath M. Incidence of ACL injury. In: Hewett TE, Shultz SJ, Griffin LY, editors. Understanding and preventing noncontact ACL injuries. Illinois: Human Kinetics; 2007. p. 5–29.
- [2] Orchard J, Seward H, McGivern J, Hood S. Intrinsic and extrinsic risk factors for anterior cruciate ligament injury in Australian footballers. *Am J Sports Med* 2001; 29(2):196–200.
- [3] Waldén M, Häggglund M, Magnusson H, Ekstrand J. Anterior cruciate ligament injury in elite football: a prospective three-cohort study. *Knee Surg Sports Traumatol Arthrosc* 2011;19:11–9.
- [4] Donnelly CJ, Elliott BC, Ackland TR, Doyle TLA, Beiser TF, Finch CF, et al. An anterior cruciate ligament injury prevention framework: incorporating the recent evidence. *Res Sports Med* 2012;20(3–4):239–62.
- [5] Krosshaug T, Nakamae A, Boden BP, Engebretsen L, Smith G, Slauterbeck JR, et al. Mechanisms of anterior cruciate ligament injury in basketball: video analysis of 39 cases. *Am J Sports Med* 2007;35:359–67.
- [6] Koga H, Nakamae A, Shima Y, Iwasa J, Myklebust G, Engebretsen L, et al. Mechanisms for noncontact anterior cruciate ligament injuries: knee joint kinematics in 10 injury situations from female team handball and basketball. *Am J Sports Med* 2010;38(11):2218–25.
- [7] Olsen O-E, 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;32(4):1002–12.
- [8] Myer G, Stroube B, DiCesare C, Brent J, Ford K, Heidt R, et al. Augmented feedback supports skill transfer and reduces high-risk injury landing mechanics: a double-blind, randomized controlled laboratory study. *Am J Sports Med* 2013;41(3):669–77.
- [9] Hewett TE, Myer GD, Ford KR, Heidt Jr RS, 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;33(4):492–501.
- [10] Kimura Y, Ishibashi Y, Tsuda E, Yamamoto Y, Hayashi Y, Sato S. Increased knee valgus alignment and moment during single-leg landing after overhead stroke as a potential risk factor of anterior cruciate ligament injury in badminton. *Br J Sports Med* 2012;46(3):207–13.
- [11] Lin C-F, Liu H, Gros MT, Weinhold P, Garrett WE, Yu B. Biomechanical risk factors of non-contact ACL injuries: a stochastic biomechanical modeling study. *J Sport Health Sci* 2012;1(1):36–42.
- [12] Speer KP, Spritzer CE, Bassett III FH, Feagin Jr JA, Garrett Jr WE. Osseous injury associated with acute tears of the anterior cruciate ligament. *Am J Sports Med* 1992; 20(4):382–9.
- [13] Russell KA, Palmieri RM, Zinder SM, Ingersoll CD. Sex differences in valgus knee angle during a single-leg drop jump. *J Athl Train* 2006;41(2):166–71.
- [14] Pollard CD, Davis IM, Hamill J. Influence of gender on hip and knee mechanics during a randomly cued cutting maneuver. *Clin Biochem* 2004;19:1022–31.
- [15] Yu B, Lin C-F, Garrett WE. Lower extremity biomechanics during the landing of a stop-jump task. *Clin Biochem* 2006;21:297–305.
- [16] Sasaki S, Nagano Y, Kaneko S, Imamura S, Koabayashi T, Fukubayashi T. The relationships between the center of mass position and the trunk, hip, and knee kinematics in the sagittal plane: a pilot study on field-based video analysis for female soccer players. *J Hum Kinet* 2015;45(1):71–80.
- [17] Sheehan FT, Sipprell WH, Boden BP. Dynamic sagittal plane trunk control during anterior cruciate ligament injury. *Am J Sports Med* 2012;40(5):1068–74.
- [18] Ali N, Rouhi G. Barriers to predicting the mechanisms and risk factors of non-contact anterior cruciate ligament injury. *Open Biomed Eng J* 2010;4:178–89.
- [19] Cortes N, Onate J, Abrantes J, Gagen L, Dowling E, Van Lunen B. Effects of gender and foot-landing techniques on lower extremity kinematics during drop-jump landings. *J Appl Biomech* 2007;23(4):289–99.
- [20] Dempsey AR, Elliott BC, Munro BJ, Steele JR, Lloyd DG. Whole body kinematics and knee moments that occur during an overhead catch and landing task in sport. *Clin Biochem* 2012;27(5):466–74.
- [21] Ishida T, Yamanaka M, Takeda N, Aoki Y. Knee rotation associated with dynamic knee valgus and toe direction. *Knee* 2014;21(2):563–6.
- [22] 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; 31(5):708–16.
- [23] Tran AA, Gatewood C, Harris AH, Thompson JA, Drago J. The effect of foot landing position on biomechanical risk factors associated with anterior cruciate ligament injury. *J Exp Orthop* Dec 2016;3(1):13 [Epub 2016 Jun 18].
- [24] Valenzuela KA, Bhaskaran D, Hummer C, Schefano A, Zhang S. Effects of a combined inversion and plantarflexion surface on knee and hip kinematics during landing. *Sports Biomech* 2016;31:1–11.
- [25] Pappas E, Hagins M, Sheikhzadeh A, Nordin M, Rose D. Biomechanical differences between unilateral and bilateral landings from a jump: gender differences. *Clin J Sport Med* 2007;17(4):263–8.
- [26] Lamontagne M, Benoit DL, Ramsey DK, Caraffa A, Cerulli G. In vivo biomechanical study for injury prevention. In: Hong Y, Bartlett R, editors. Handbook of biomechanics and human movement science. Oxon, New York: Routledge; 2008. p. 428–45.
- [27] Cohen J. Statistical power analysis for the behavioral sciences. 2nd ed. Hillsdale: Lawrence Erlbaum Associates; 1988.
- [28] Bell AL, Brand RA. Prediction of hip joint centre location from external landmarks. *Hum Mov Sci* 1989;8:3–16.
- [29] Hanna D, Dempster M. Effect sizes and power. Psychology statistics for dummies. West Sussex: John Wiley and Sons, Ltd; 2012. p. 165–83.
- [30] Noyes FR, Barber-Westin SD, Fleckenstein C, Walsh C, West J. The drop-jump screening test: difference in lower limb control by gender and effect of neuromuscular training in female athletes. *Am J Sports Med* 2005;33(2):197–207.

- [31] Weinhandl JT, Joshi M, O'Connor KM. Gender comparisons between unilateral and bilateral landings. *J Appl Biomech* 2010;26(4):444–53.
- [32] Kim HY, Kim KJ, Yang DS, Jeung SW, Choi HG, Choy WS. Screw-home movement of the tibiofemoral joint during normal gait: three-dimensional analysis. *Clin Orthop Surg* 2015;7(3):303–9.
- [33] Clippinger KS. The knee and patellofemoral joints. *Dance anatomy and kinesiology*. Illinois: Human Kinetics; 2016. p. 163–200.
- [34] Escamilla R, Lander J, Garhammer J. Biomechanics of powerlifting and weightlifting exercises. In: Garrett WE, Kirkendall DT, editors. *Exercise and sports science*. Philadelphia: Lippincott Williams & Wilkins; 2000. p. 585–616.
- [35] Arnheim DD, Prentice WE. The knee and related structures. *Principles of athletic training*. St. Louis: Mosby-Year Book; 1993. p. 528–84.
- [36] Hong YG, Yoon Y-J, Kim P, Shin CS. The kinematic/kinetic differences of the knee and ankle joint during single-leg landing between shod and barefoot condition. *Int J Precis Eng Manuf* 2014;15(10):2193–7.
- [37] Hewett TE, Johnson DL. ACL prevention programs: fact or fiction? *Orthopedics* 2010;33(1):36–9.