DOCTORAL THESIS

The effect of methodological sources ofvariability on the interpretation of change of direction kinematic and kinetic metrics following anterior cruciate ligament reconstruction.

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The Effect of Methodological Sources of Variability on the Interpretation of Change of Direction Kinematic and Kinetic Metrics Following Anterior Cruciate Ligament Reconstruction

by

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0.1 Abstract

Change of direction (CoD) movements are the most common mechanism of anterior cruciate ligament (ACL) rupture during multi-directional field-based sports. ACL-reconstruction (ACLR) is the recommended treatment for athletes intent on returning to sport participation. A high incidence of secondary ACL injury is reported following return to sport. The relationship between technique and knee joint loading during movement tasks associated with injury has been studied extensively in an attempt to identify risk factors for primary and secondary ACL injury. However, there have been limited analyses of CoD in this context. One possible explanation for this is that concurrent methodological and task-specific issues make CoD a challenging movement to study experimentally. This thesis aimed to examine the effect of methodological sources of variability on the interpretation of kinematic and kinetic metrics during a CoD task following ACLR. A cohort of ACLR patients and a non-injured control group completed a 90° CoD task while optical motion capture and ground reaction force data were recorded. Four experimental studies examining issues related to marker placement error, variability in approach velocity and CoD angle during task completion, and calculations of normative kinematic and kinetic inter-limb differences were conducted. Simulated systematic marker placement error within previously reported inter-tester variability ranges caused significant differences in knee abduction moment, hip rotation angle, knee rotation angle, ankle abduction and ankle rotation angle across various periods of stance. Simulated random marker placement error caused large changes in inter-limb difference measures in several variables including hip rotation angle, knee abduction angle and knee abduction moment, severely limiting the ability to monitor these variables and identify ACLR patients with large inter-limb differences relative to a control group. Variability in approach velocity and CoD angle explained 3–60% of the variance in kinematic and kinetic inter-limb differences during CoD stance phase. No method for identifying systematic inter-limb differences in non-injured control groups was successfully identified. Considerable challenges exist in the assessment of CoD as small methodological variation can have a large effect on kinematic and kinetic metrics, altering the subsequent clinical interpretation of data. This thesis can serve as a framework informing best practice in the analysis of CoD tasks following ACLR.

Contents

0.2 Acknowledgments

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0.3 Dissemination of Research

0.3.1 Published Journal Articles

McFadden, Ciarán., Daniels, Katherine, and Strike, Siobhán. 2020. "The Sensitivity of Joint Kinematics and Kinetics to Marker Placement during a Change of Direction Task." Journal of Biomechanics. 101. p.109653.

McFadden, Ciarán., Daniels, Katherine, and Strike, Siobhán. 2021. "The Effect of Simulated Marker Misplacement on the Interpretation of Inter-Limb Differences during a Change of Direction Task." Journal of Biomechanics 116. p. 110184.

McFadden, Ciarán., Daniels, Katherine, and Strike, Siobhán. 2021. "Six Methods for Classifying Lower-Limb Dominance Are Not Associated with Asymmetries during a Change of Direction Task." Scandinavian Journal of Medicine and Science in Sports (September): 32(1). pp. 106-115.

0.3.2 Submitted Journal Articles

McFadden, Ciarán., Siobhán, Strike and Daniels, Katherine,. 2021."Are inter-limb differences in change of direction velocity and angle associated with inter-limb differences in kinematics and kinetics following anterior cruciate ligament reconstruction?". Under review.

0.3.3 Conference Proceedings

Poster Presentations

McFadden, Ciarán., Daniels, Katherine, and Strike, Siobhán. The method used to define limb dominance in uninjured controls affects the resultant limb symmetry index statistics and the identified differences between ACL-reconstructed and uninjured groups. 2018. European College of Sport Science. Dublin, Ireland.

Oral Presentations

McFadden, Ciarán., Daniels, Katherine, and Strike, Siobhán. The sensitivity of joint kinematics and kinetics to marker placement during a change of direction task. 2019. International Society of Biomechanics in Sports. Miami, Ohio, USA.

McFadden, Ciarán., Daniels, Katherine, and Strike, Siobhán., 2020. The effect of marker placement error on the interpretation of inter-limb differences in frontal plane knee loading during a change of direction task. International Society of Biomechanics in Sports. Liverpool, UK [Online due to COVID-19].

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0.4 List of Abbreviations

ACL - Anterior cruciate ligament

ACLR - Anterior cruciate ligament reconstruction

CoD - Change of direction

CGM - Conventional gait model

PiG - Plug-in-Gait

6DOF - Six degrees of freedom

CAST - Calibrated anatomical systems technique

RTS - Return to sport

GRF - Ground reaction force

KAM - Knee abduction moment

CoM - Centre of mass

SPM - Statistical parametric mapping

THI - Lateral thigh marker

KNE - Lateral femoral epicondyle marker

TIB - Lateral shank marker

DLDJ - Double leg drop jump

SLCMJ - Single leg countermovement jump

SLHop - Single leg hop for distance

GCT - Ground contact time **IMU** - Inertial Measurement Unit

Chapter 1

Thesis Introduction and Aims

1.1 Introduction

Anterior cruciate ligament (ACL) rupture is a serious musculoskeletal injury that most commonly occurs during participation in multi-directional field-based sports such as football, basketball, rugby, netball, etc (Geli-Alentorn et al., 2009). Change of direction (CoD) manoeuvres are ubiquitous features of these sports, used by athletes as a means of evading opponents (Condello et al., 2013; Havens and Sigward, 2015*a*). Video analyses identify CoD movements as the most common mechanism of non-contact ACL injuries (Johnston et al., 2018; Olsen et al., 2004). Mechanically, CoD involves the deceleration, redirection, and acceleration of the body's centre of mass (CoM) in a new direction of travel (Dos'Santos et al., 2018; Havens and Sigward, 2015*b*). The ACL's primary function is to resist anterior tibial translation and rotation, thus stabilising the knee joint during dynamic multi-planar movements such as CoD (Barber-Westin and Noyes, 2011). The rapid deceleration involved in CoD can manifest in high stress being applied to the ACL, which, if resulting in excessive strain causes the ligament to rupture (Markolf et al., 1995).

For athletes intent on returning to participation in multi-directional field-based sports following ACL rupture, surgical treatment in the form of an ACL-reconstruction (ACLR) is generally recommended (Kvist, 2004). ACLR typically uses an autograft, normally harvested from either the patellar or hamstring tendons to restore structural stability to the knee joint (Anderson et al., 2006). Surgery induced joint effusion, in combination with graft site disruption, reduced knee joint range of motion and immobilization leads to muscle atrophy and weakness on the operated limb (Thomas et al., 2016). Post-operative rehabilitation aims to restore knee joint function and physical capacity on the operated limb to pre-injury levels, with athletes normally returning to sport (RTS) 9 - 12 months post-surgery (Brophy et al., 2012; Graziano et al., 2017). Despite developments in surgical techniques and rehabilitation practices, outcomes related to RTS, secondary injury and degenerative changes at the knee joint are suboptimal. A systematic review by Ardern et al., (2011) found that while 85% of athletes returned to some form of sports participation following ACLR, only 63% returned to the same level of activity that they participated at pre-injury. Athletes who do RTS are at high risk of sustaining a secondary ACL injury with pooled injury rates of 5.8% for graft failure (ipsilateral injury) and 11.8% for contralateral ACL rupture reported (Wright et al., 2011). Long term outcomes are also poor, with 5 – 10 year follow ups identifying osteoarthritic changes at the knee joint in approximately 50% of patients (Barenius et al., 2014; Claes et al., 2013; Struewer et al., 2012).

Minimising the risk of secondary ACL injury is a central focus of all ACLR rehabilitation programs. Secondary ACL injuries are a devastating occurrence for athletes, necessitating further surgical treatment and rehabilitation following which outcomes related to RTS, knee function and osteoarthritic changes at the knee joint are worse than those after primary injury (Andriolo et al., 2015; Lefevre et al., 2017). A large body of research aimed at identifying risk factors for both primary (Hewett et al., 2005; Krosshaug et al., 2016; Lloyd, 2001; Smith et al., 2012) and secondary injury (Paterno et al., 2010; King et al., 2021a; King et al., 2021b, Zhou et al., 2020, Sward et al., 2012) has developed over the past 30 years. One strand of this research has utilised biomechanical assessments to study the relationship between technique and knee joint loading during movement tasks that challenge the structural integrity of the knee joint (Lloyd, 2001; Hewett et al., 2005; King, Richter, Franklyn-Miller, Daniels, Wadey, Moran and Strike, 2018; Paterno et al., 2010; Sharir et al., 2016; Krosshaug et al., 2016). In these analyses, images captured from multiple perspectives via a synchronised camera system are combined and used to reconstruct body segment positions and orientations in three-dimensional space (Dindaroğlu et al., 2016). Segment positions and orientations are normally defined from the positions of retroreflective markers placed externally on the skin surface of participants, with the specific location and number of markers dependent on the biomechanical model used (Charlton et al., 2004; Rutherford et al., 2014; Robinson et al., 2014). The relative orientations of segments are converted to kinematic variables that describe joint motion. Combining these data with external measures of force, normally quantified using force platforms, allows the forces and moments that produced the observed motion, as well as other kinetic properties such as segmental work and power, to be estimated (Winter, 2009).

Researchers have used biomechanical analyses to identify abnormalities in kinematics and kinetics following ACLR during movement tasks that mimic ACL injury mechanisms (Gokeler et al., 2010; Kuenze et al., 2015; Meyer et al., 2018). The findings from these studies are used to develop hypotheses about the relationship between specific biomechanical variables and clinical outcomes such as secondary injury. If a causal relationship is identified between a modifiable variable and injury risk, the variable can theoretically be monitored and targeted during rehabilitation as a means of reducing the risk of secondary injury upon RTS. Common research designs utilised to identify abnormalities in kinematics and kinetics following ACLR include inter-limb (ACLR limb v non-ACLR limb) (King et al., n.d.; O'Malley et al., 2018; Oberländer et al., 2013; Paterno et al., 2007; Schmitt et al., 2015; Webster et al., 2015; Xergia et al., n.d.) and inter-group (ACLR group v non-injured control group) comparisons (King et al., n.d.; Kuenze et al., 2015; Stearns and Pollard, 2013; Zwolski et al., 2016). Though recent evidence indicates that ACLR can result in bilateral deficits (Dai et al., n.d.), inter-limb comparisons between ACLR and non-ACLR limbs remain commonplace, both in clinical and in research settings (Bishop et al., 2018; King et al., n.d.; Promsri et al., 2020; Schmitt et al., 2015; Sharafoddin-Shirazi et al., 2020; Di Stasi et al., 2013; Webster et al., 2015). As pre-injury levels of function on the ACLR limb are generally unknown, the non-ACLR limb serves as a practical reference for researchers and clinicians to use in order to gauge rehabilitation progress. Inter-group comparisons with non-injured athletes alleviate concerns with respect to bilateral deficits following ACLR, but present unique challenges in forming standardised comparisons between groups when comparing inter-limb difference magnitudes. Quantifying inter-limb differences necessitates one limb to be chosen as a reference for the other to be compared to. In non-injured groups, there is no clear distinction to be made between limbs, with most researchers utilising "limb dominance" defined according to some generic criteria, e.g. preferred kicking limb or the limb that can attain greatest jump height. The method used to distinguish between limbs will thus influence the subsequent inter-group comparisons. These analyses are also logistically more difficult to conduct, requiring the recruitment of non-injured participants.

Despite CoD being identified as the most common mechanism of ACL injury, there has been minimal research quantifying inter-limb differences in CoD technique following ACLR, nor inter-group comparisons of ACLR CoD technique with non-injured groups (King et al., n.d.; Stearns and Pollard, 2013; King, Richter, Franklyn-Miller, Daniels, Wadey, Jackson, Moran and Strike, 2018). Instead, the vast majority of research in this regard has focused on jump/landing tasks, with particular focus given to the bilateral vertical drop jump (Hewett et al., 2005; Paterno et al., 2007; Krosshaug et al., 2016; Kaphingst et al., 2010; Meyer et al., 2018; Paterno et al., 2010; Stephens et al., 2007; King et al., 2021*a*). The reasoning for the focus on the vertical drop jump, over more sport-specific, injury-related tasks such as CoD is likely two-fold. Firstly, the vertical drop jump is an inherently easier task to control and study than CoD. Critical task features that influence kinematics and kinetics - such as landing height - can be easily controlled across participants. Kinematic and kinetic variables analysed during the landing phase are thus not confounded by any features that preceded the onset of the movement. In contrast, critical CoD task features - such as approach velocity and the angle over which participants change direction - are much more challenging to control and can vary significantly between trials, limbs and participants (Daniels et al., 2021; King, Richter, Franklyn-Miller, Daniels, Wadey, Moran and Strike, 2018). Vanrenterghem et al., (2012) demonstrated that approach velocity affected knee flexion angle and knee abduction moment, two variables reported to influence ACL loading, during CoD stance phase (Fagenbaum and Darling, 2003; King et al., n.d.; Myer et al., 2010; Stearns and Pollard, 2013). Inter-limb comparisons of these variables will consequently be influenced by any inter-limb differences in CoD approach velocity that may be present. CoD is thus a more difficult movement task to control experimentally than the vertical drop jump, presenting challenges in conducting and interpreting inter-limb and inter-group comparisons, as well as forming casual relationships between technique features and injury risk.

The second factor explaining the overarching focus on the vertical drop jump is that early research in this field prospectively associated abnormalities in landing mechanics during the vertical drop jump with both primary (Hewett et al., 2005) and secondary (Paterno et al., 2010) ACL injury risk. In a seminal paper, Hewett et al., (2005) found that abnormalities in lower-extremity mechanics during the landing phase of the vertical drop jump predicted primary ACL injury in adolescent females. A cohort of 205 participants completed a vertical drop jump assessment prior to being tracked prospectively over the course of a sporting season, after which 9 primary ACL ruptures were recorded. Athletes who sustained an ACL rupture had larger knee abduction angles at initial contact as well as larger peak knee abduction moments and peak GRFs during the landing phase of the vertical drop jump compared to those athletes who did not sustain an ACL injury. Peak knee abduction moment during landing predicted ACL injury risk with 78% sensitivity and 73% specificity. Subsequent work from this same research group screened a cohort of 56 males and females, ranging in age from 10 - 25, who had undergone primary ACLR in the previous 12 months. Participants completed a vertical drop jump and were prospectively tracked and monitored for incidence of secondary ACL injury. A total of 13 (10 contralateral and 3 ipsilateral) secondary ACL injuries were identified (Paterno et al., 2010). A combination of transverse plane hip kinematics, frontal plane knee kinematics, inter-limb differences in sagittal plane knee moments and deficits in postural stability were found to predict secondary ACL injury with 92% sensitivity and 88% specificity. The authors of these studies strongly recommended the use of the vertical drop jump as a screening tool for both primary and secondary ACL injury risk. These recommendations, in conjunction with the relative ease of studying the vertical drop jump, has led to an overarching focus on landing mechanics within the ACL/biomechanics literature (Clarke et al., 2015; Cowley et al., 2006; Ford et al., 2003; Mclean et al., 2005; Meyer et al., 2018; Pappas and Carpes, 2012; Paterno et al., 2011; Tamura et al., 2017). Particular focus has been given to frontal plane knee motion/loading, with peak knee abduction moment often used as a surrogate measure of "ACL injury risk" across different tasks and populations, despite Hewett's original findings being limited to a very specific, high risk population i.e. adolescent females (Renstrom, 2011). A systematic review by Sharir et al., (2016) noted that the majority of research examining *in vivo* biomechancial risk factors for ACL injuries was associative in nature, whereby variables are associated with previously identified risk factors such as knee abduction moment. The authors of this review highlighted the need for high quality, prospective research to be conducted in order to identify more risk factors for non-contact ACL injury (Sharir et al., 2016).

The need for more research in this area is further demonstrated by persistently high injury/re-injury rates which have remained relatively stable over the previous 20 years (Ardern et al., 2015; Laboute et al., 2010; Pujol et al., 2007; Salmon et al., 2005; King et al., 2021*b*), as well as the findings of recent research questioning the utility of the vertical drop jump as a screening tool for ACL injury risk. In a study explicitly designed to replicate that of Hewett et al., (2005), Krosshaug et al., (2016) screened a cohort of 782 female soccer and handball players using the vertical drop jump and prospectively tracked them for ACL injury occurrence, with a total of 42 non-contact ACL injuries identified at follow up. The authors examined the ability of 5 pre-defined variables, namely knee abduction angle at initial contact, peak knee abduction moment, peak knee flexion angle, peak vertical GRF and medial knee displacement to predict ACL injury risk in this cohort. Unlike Hewett et al., (2005) the authors failed to find any association between these variables and ACL injury risk within their cohort. Similarly, recent work by King et al., (2021a, 2021b) examined the ability of biomechanics testing at 9 months post-ACLR to identify patients who would subsequently sustain a secondary ACL injury following RTS. Inter-limb differences in kinematics and kinetics during the vertical drop jump demonstrated limited ability to distinguish between those who sustained a secondary ACL injury and those who did not (King et al. 2021a, 2021b). While the utility of the vertical drop jump continues to be debated, there is a clear need to use biomechanical analyses to study alternative movement tasks, as this may supplement and develop our current understanding of biomechanical risk factors for both primary and secondary ACL injuries. Despite difficulties from a data collection and interpretation perspective, CoD is a logical choice in this regard given its commonality as an ACL injury mechanism and its relevance to performance in multi-directional field-based sports.

An emerging body of literature points to the presence of inter-limb differences in CoD technique following ACLR that likely persist up to and after RTS. At 9 months post-surgery, ACLR patients systemically reduce their approach velocity and centre of mass (CoM) deflection angle, i.e. the angle over which their centre of mass moves during CoD stance phase, during pre-planned CoD tasks when turning off their ACLR limb (Daniels et al., 2021). Inter-limb differences in approach velocity and CoM deflection angle were present despite no statistical difference in completion times between limbs. This would suggest that in order to achieve the same performance outcome when turning off their operated limb, ACLR patients modify CoD task constraints to make the movement easier to complete. Kinematic and kinetic inter-limb differences are also present at 9 months post-surgery, with King et al., (2018) finding significant inter-limb differences across CoD stance phase in knee flexion angle, frontal plane knee moments, as well as transverse plane knee and ankle moments. When turning off their ACLR limb patients demonstrated smaller knee flexion angles and knee joint moments throughout stance phase. Again, these inter-limb differences were present despite no statistical difference in completion times between limbs. In a follow up study, King et al., (2019) demonstrated that the magnitude of inter-limb differences in kinematics and kinetics were larger in ACLR patients than in uninjured participants. The presence of large inter-limb differences during CoD at 9 months post-surgery suggest that many patients RTS with persistent CoD-specific deficits that may influence secondary injury risk. Sagittal plane deficits are attributed to reduced quadriceps activation/capacity and thought to demonstrate an inability to eccentrically load the ACLR limb during deceleration (King, Richter, Franklyn-Miller, Daniels, Wadey, Moran and Strike, 2018; Schmitt et al., 2012). The quadriceps play an important role in resisting frontal plane knee movement at the knee (Lloyd, 2001), which may be why ACLR patients demonstrate smaller knee abduction moments as they do not have the physical capacity to resist excessive frontal plane loads during CoD.

Although it is reasonable to hypothesise that reduced deceleration capacity and deficits in frontal plane knee control during CoD may increase athletes' susceptibility to injury upon RTS, the only studies thus far examining the ability of biomechanical assessments of CoD to identify ACLR patients at risk of secondary ACL injury found that CoD inter-limb difference measures had limited ability to distinguish between patients who sustained a secondary injury and those who did not (King et al., 2021a, King et al., 2021b). One major factor highlighted by the authors of these studies as a potential explanation for the inability to distinguish between those who re-injured and those who did not was the high variability associated with the kinematic and kinetic metrics used in their statistical model. Variability in kinematic and kinetic inter-limb differences may stem both from methodological sources of error within the data collection process, as well as inter-limb and inter-trial variability in task features such as approach velocity and CoM deflection angle. A combination of these factors may prevent causal relationships between technique features and injury risk being formed. Addressing this variability, as well as concurrent methodological issues in CoD assessments, is necessary in order to determine the utility of CoD assessments as a means of monitoring rehabilitation and informing RTS decision making following ACLR.

With respect to data collection, of chief concern in any analysis utilising marker-based biomechanical models to quantify kinematic and kinetic variables is marker placement error. Marker-based biomechanical models function under various assumptions, one of which is that markers are positioned in a manner that allows the location of otherwise not directly measurable quantities (e.g. joint center positions) to be precisely estimated. In reality, there is always an element of error associated with the position of each marker. Marker placement error is cited as the primary source of variability in biomechanical analyses (Alenezi et al., 2016; Gorton et al., 2009; McGinley et al., 2009). Marker placement error has been widely studied in the context of clinical gait analyses and shown to cause large errors in frontal and transverse plane kinematics (Baker et al., 1999; Fonseca et al., 2020; Groen et al., 2012; Kadaba, Ramakrishnan, Wooten, Gainey, Gorton and Cochran, 1989; Szczerbik and Kalinowska, 2011). The findings of Baker et al., (1999) and Kadaba et al., (1989) demonstrate that the effect of marker placement error may be task specific, with task factors such as sagittal plane range of motion influencing the observed effect on kinematics and kinetics. Despite this, there has been minimal research examining marker placement error on any movement tasks other than walking and thus far none examining its influence on CoD kinematics and kinetics. An exploration of the influence of marker placement error on CoD kinematics and kinetics is currently lacking within the literature. ***This is addressed in Chapters 2 and 3, where the effect of systematic and random marker placement error on CoD kinematics and kinetics are examined.***.

While marker placement error is a major source of variability in any analysis utilising a marker-based biomechanical model, there are also factors unique to CoD that make it challenging to study and may contribute to high levels of variability reported in kinematic and kinetic inter-limb differences (King et al., 2021a, King et al., 2021b). Approach velocity and CoM deflection angle are fundamental CoD task descriptors that reflect the whole body demands of a CoD movement and influence joint kinematics and kinetics during CoD stance phase (Vanrenterghem et al., 2012; Kristianslund et al., 2012). When studying CoD, researchers aim to standardise and control these two factors as much as possible, typically by using specific verbal instructions and/or timing gates for approach velocity and the use of pre-defined angles e.g. 45°, 90° for CoM deflection angle (Bencke et al., 2013; Brown et al., 2014; King et al., n.d.; Pollard et al., 2018). This is in order to reduce the effect that approach velocity and CoM deflection angle have on kinematics and kinetics during CoD stance phase (Dos'Santos et al., 2018; Sigward et al., 2015*a*; Vanrenterghem et al., 2012). It has been shown that following ACLR, systematic inter-limb differences in approach velocity and CoM deflection angle are present during pre-planned CoD tasks (Daniels et al., 2021; King, Richter, Franklyn-Miller, Daniels, Wadey, Moran and Strike, 2018). Inter-limb differences in approach velocity and CoM deflection angle likely contribute to the presence of kinematic and kinetic inter-limb differences during CoD. However, currently it is unclear what proportion of the variance in kinematic and kinetic inter-limb differences during CoD is explained by inter-limb differences in approach velocity and CoM deflection angle. ***This issue is examined in Chapter 4 where the relationship between inter-limb differences in task-level variables and inter-limb differences in kinematic and kinetic inter-limb differences during CoD is investigated***.

Examining the effect of marker placement error and task-level adjustments to approach velocity and CoM deflection angle on kinematic and kinetic variables will provide valuable information with respect to performing both individual and group assessments of CoD kinematic and kinetic inter-limb differences. It is also important to explore issues related to inter-group comparisons between ACLR and non-injured groups, in order to inform best practice with respect to such comparisons. A critical methodological consideration in such analyses is the method used to calculate inter-limb differences in both groups. In ACLR cohorts, the non-operated limb is typically used as a reference for the operated limb to be compared to, with inter-limb differences calculated as the directional difference between limbs

(King, Richter, Franklyn-Miller, Daniels, Wadey, Jackson, Moran and Strike, 2018; Paterno et al., 2007; Sueyoshi et al., 2017). This method allows the direction of the inter-limb difference, i.e. which limb has a greater value of "variable x" to be identified. Though a logical approach in ACLR cohorts, difficulties arise when trying to replicate this in non-injured cohorts as there is no basis on which to make a distinction between limbs. The most common approach has been to distinguish between dominant and non-dominant limbs. However, limb dominance is a poorly defined term, with various different definitions used within the sports medicine literature e.g. preferred kicking limb, limb that attains greatest jump height, limb that attains furthest hop distance, etc. (Carcia et al., 2019; Gabbard and Hart, 1996; van Melick et al., 2017). Recent evidence indicates that the use of different definitions of lower-limb dominance will manifest in different limbs being denoted as dominant and non-dominant (Mulrey et al., 2018). Group inter-limb difference measures and the findings of subsequent inter-group comparisons are therefore contingent on the definition of lower-limb dominance used in a study. One solution to this problem is to compare absolute measures of inter-limb difference magnitudes between groups, as done by King et al., (2019). However, this method does not provide any information about the direction of inter-limb differences. There is a need to determine if there is an appropriate method for calculating directional inter-limb differences during CoD in normative groups, so that the magnitude of inter-limb differences in ACLR cohorts can be properly contextualised and realistic rehabilitative targets can be set. ***This issue is explored in Chapter 5, where the ability of six lower limb dominance definitions to quantify normative directional inter-limb differences is examined***.

1.1.1 Aims and Structure

ACL injuries occur during CoD movements in multi-directional field-based sports. Biomechanical analyses offer the ability to study the relationship between technique and external knee joint loading during movement tasks related with ACL injury, potentially allowing modifiable risk factors for primary and secondary injury to be identified. CoD is a logical task to study in order to improve our understanding of ACL injury mechanisms and risk factors. However, it is also a challenging task to study due to a myriad of methodological challenges related to data collection and analysis. The aim of this thesis is to determine the feasibility of using biomechanical analyses to identify abnormalities in CoD technique following ACLR. To achieve this, four experimental studies, presented here as four separate chapters, were conducted.

1.2 Chapter Layout

Following a short introduction to explain the context of the study within the thesis aims, one experimental study is presented in each of Chapters 2, 3, 4 and 5. For consistency, reference style has been updated to match University of Roehampton requirements, figure and table labels have been updated so that they run consecutively throughout the thesis and paper abstracts have been removed. Other than this, text in each chapter appears as published (Chapters 2, 3 and 5) or as currently formatted for review (Chapter 4). Each chapter thus has its own methodology section where the study participant demographics, data collection and data analysis processes are outlined. Additional information not presented in chapters with respect to participant recruitment and data collection protocols are included in Appendix A.

Chapter 2

Chapter 2 presents one experimental study examining the effect of systematic marker placement error on CoD kinematic and kinetic variable magnitudes. This study has been published as a reviewed article in the Journal of Biomechanics.

McFadden, Ciarán., Katherine. Daniels, and Siobhán Strike. 2020. "The Sensitivity of Joint Kinematics and Kinetics to Marker Placement during a Change of Direction Task." Journal of Biomechanics. 101. p.109653.

Aim – To determine the sensitivity of joint kinematics at the hip, knee and ankle, as well as knee joint moments, to systematic marker displacements across the stance phase of a CoD task.

Chapter 3

Chapter 3 presents one experimental study exploring the effect of random marker placement error on kinematic and kinetic inter-limb differences during CoD, and how this influences the interpretation of differences in inter-limb difference magnitudes between an ACLR and non-injured group. This study has been published as a reviewed article in the Journal of Biomechanics.

McFadden, Ciarán, Katherine Daniels, and Siobhán Strike. 2021b. "The Effect of Simulated Marker Misplacement on the Interpretation of Inter-Limb Differences during a Change of Direction Task." Journal of Biomechanics. 116. p. 110184.

Aim – To determine the effect of random marker displacements on the interpretation of inter-limb differences during a CoD task.

Chapter 4

Chapter 4 presents one experimental study examining the relationship between inter-limb differences in approach velocity and CoM deflection angle during CoD and inter-limb differences in kinematic and kinetic variables. This study is currently under review.

McFadden, Ciarán, Siobhán Strike, Katherine Daniels 2021. "Are change of direction task level inter-limb differences associated with inter-limb differences in kinematic and kinetic variables following anterior cruciate ligament reconstruction?" – currently under review.

Aim - To determine the proportion of variance in inter-limb differences in kinematics and kinetics during a CoD task that can be explained by task-level inter-limb differences in approach velocity and CoM deflection angle.

Chapter 5

Chapter 5 presents one experimental study which attempts to identify a viable method for quantifying directional inter-limb differences in normative groups in order to determine the feasibility of performing inter-group comparisons in inter-limb differences between ACLR and non-injured groups. This study has been published as a reviewed article in the Scandinavian Journal of Medicine and Science in Sport.

McFadden, Ciarán, Katherine Daniels, and Siobhán Strike. 2021a. "Six Methods for Classifying Lower-Limb Dominance Are Not Associated with Asymmetries during a Change of Direction Task." Scandinavian Journal of Medicine and Science in Sports (September): 32(1). pp. 106-115.

Aim – To determine whether five previously used methods of classifying lower-limb dominance and a CoD specific method identified systematic directional kinematic and kinetic inter-limb differences during a CoD task.

1.2.1 Data Collection and Processing Procedures

All ACLR participants used in this thesis were recruited through the Sports Surgery Clinic, a private orthopaedic hospital in Dublin, Ireland. In 2013, the Sports Surgery Clinic created an ACL pathway clinical service for patients who underwent ACLR in the hospital. Following diagnosis of ACL rupture by one of the clinics consultant orthopaedic surgeons, patients were offered the opportunity to enroll in this pathway. As part of this pathway, patients returned to the Sports Surgery Clinic at 3, 6, and 9 months post-surgery to undergo a physical testing battery designed to assess the progress of their rehabilitation. The 6 and 9 month appointments involved the assessment of a series of jump/landing and CoD movement tasks in the Sports Surgery Clinic's biomechanics laboratory, as well as isokinetic dynamometry testing of knee flexor/extensor strength. In order to facilitate this clinical biomechanical testing service, a team of biomechanics laboratory staff were recruited. This team consisted of one laboratory manager and four research assistants. All staff undertook an extensive training period and were assessed by the laboratory manager prior to working with patients to ensure competency and consistency in all data collection processes.

The author of this thesis was a staff member in the Sports Medicine Department of the Sports Surgery clinic and worked in the biomechanics laboratory as a research assistant (2017 - 2020) and as the laboratory manager (2020 - 2022) whilst completing this project. During this period the laboratory saw on average 30 ACLR patients per week for biomechanics testing. The thesis author was involved in data collection for approximately 6 - 10 patients per week between 2017 and 2020. Following this, the author was responsible for training of laboratory research assistants in data collection and processing but did not continue in data collection. For the experimental studies presented in this thesis, a subset of data collected that matched the required patient demographics (male, aged 18-35, multi-directional field-based sports participation, intention of returning to the same level of sports participation following rehabilitation, no history of previous ACL injury and did not require multiple-ligament reconstruction and/or a meniscectomy) were used in this thesis. To aid in data processing, a custom MATLAB application developed by former Sports Surgery Clinic Head of Data and Innovation Dr Chris Richter, was used. This program automated the gap filling process for the biomechanical data used in this thesis as well as allowing for each individual data set to be screened for errors prior to being saved to a data repository for further use. Data processing using this program was completed by laboratory research assistants, with the author of this thesis involved in processing laboratory data from 2017 - 2020. Subsequently, for each experimental study in this thesis, custom MATLAB scripts were written by the thesis author in order to analyse data, perform statistical analyses and for data visualisation purposes.

Chapter 2

The Sensitivity of Joint Kinematics and Kinetics to Marker Placement During a Change of Direction Task

2.1 Chapter Context

Chapter 2 presents an experimental study examining the sensitivity of lower extremity kinematics and knee joint moments to systematic marker placement error during a CoD task. The CGM is a marker-based biomechanical model that has been used in the analysis of various movement tasks, including CoD (Baker et al., 1999; King, Richter, Franklyn-Miller, Daniels, Wadey, Moran and Strike, 2018). The CGM models the lower-body as a series of seven segments (pelvis, x2 femur, x2 shank, x2 foot) linked in a chain by ball joints that allow for three degrees of rotational freedom. Each segment contains an associated orthogonal axes system, the origin and orientation of which are defined from the positions of three non-collinear markers placed externally on specific anatomical landmarks on the skin surface of participants (Fig 2.1). The relative orientations of segments are used to derive joint kinematics, which when combined with external measures of force, allows the forces and joint moments that produced the observed motion to be estimated. Accurate measures of joint kinematics and kinetics are contingent on the correct alignment of segment axes and orientations. Any error in the positions of markers will affect both kinematic and kinetic variables.

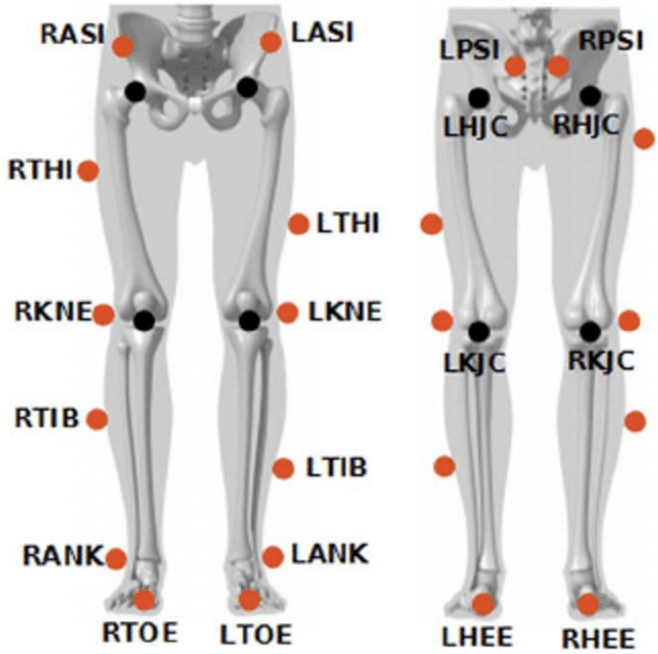


Figure 2.1: Marker positions (red circles) for lower extremity in CGM (Baudet et al., 2014)

The influence of marker placement error on joint kinematics and kinetics has been widely studied in walking. Baker et al., (1999) demonstrated that errors in frontal plane knee kinematics from marker placement were exacerbated by knee flexion during walking, suggesting that task specific features such as sagittal plane range of motion may influence the effect of marker placement error on kinematics. Despite this there is minimal research in activities other than walking and none thus far exploring its influence of CoD kinematics and kinetics. For these metrics to be used in any clinical assessment of ACLR patients, it is necessary to establish how they are influenced by marker placement error and how this affects subsequent clinical interpretation of data. Marker placement error can be categorised into two broad categories, namely systematic error, where there are systematic differences in the positions of markers, when for instance, placed by two different experimenters, and random error, where there are random variations in marker positions, for example when applied on multiple occasions. Systematic error may arise if two experimenters receive different instructions/procedures to follow when identifying anatomical landmarks, or if they interpret instructions/procedures in a different manner to each-other. Such a scenario may preclude the pooling of data collected by different experimenters. This chapter examines the sensitivity of CoD kinematics and kinetics to systematic marker placement error in order to determine which variables are most affected by marker placement, and what magnitude of systematic error is required to cause significant differences in kinematics and kinetics.

2.2 Introduction

The conventional gait model (CGM) refers to several closely related biomechanical models, the data from which are used to analyse human motion, inform clinical decision making and evaluate rehabilitation interventions (Baker et al., 2017). Such models provide an objective record of kinematic and kinetic metrics during movement. Originally developed for and implemented in clinical gait analyses, the CGM's application has been extended to a variety of movements, including a range of change of direction (CoD) tasks (Franklyn-Miller et al., 2017; King, Richter, Franklyn-Miller, Daniels, Wadey, Jackson, Moran and Strike, 2018; Marshall et al., 2014;

O'Malley et al., 2018; Sigward and Powers, 2007).

CoD is the most common mechanism of non-contact anterior cruciate ligament (ACL) rupture, a serious musculoskeletal injury normally requiring surgical intervention (Kvist, 2004). The CGM has been utilised in the analysis of CoD to inform best practice in the prevention and rehabilitation of ACL injury (King, Richter, Franklyn-Miller, Daniels, Wadey, Jackson, Moran and Strike, 2018; Mclean et al., 2005; Sigward and Powers, 2007). Kinematic variables of the hip, knee and ankle have been associated with increased frontal plane knee loading during CoD, considered a key risk factor for injury (Hewett et al., 2005; Sigward and Powers, 2007).

Accurate measures of these variables rely on the correct definition of body segment axes origins and orientations (Kadaba et al. 1989). In the Plug-in-Gait (PiG) model (Vicon, Oxford Metrics, London, UK), a widely used implementation of the CGM, retroreflective markers placed externally on a series of anatomical landmarks define segment origins and orientations. Variation in marker placement is cited as the primary factor in the low reliability indices reported for many kinematic and kinetic variables (Alenezi et al., 2016; Gorton et al., 2009; McGinley et al., 2009).

Inter-tester variability in anatomical landmark location, and subsequently marker placement, makes inferring ACL injury mechanisms based on data collected in different laboratories and by different practitioners challenging. The range of inter-tester variability in anatomical landmark location for marker positions has been reported as 12 – 25 mm (Della Croce et al., 1999). Given their roles in defining the origins and orientations of the femur and shank segments, the lateral thigh (THI), lateral femoral epicondyle (KNEE) and lateral tibia (TIB) markers have the largest effect on model outputs (Kadaba, Ramakrishnan and Wooten, 1989). The deterministic nature of the model indicates that variation in the anterior/posterior positions of these markers will alter joint kinematics and kinetics of the hip, knee and ankle (Kadaba, Ramakrishnan, Wooten, Gainey, Gorton and Cochran, 1989).

Experimental studies confirm the sensitivity of joint kinematics, particularly frontal and transverse plane kinematics, to marker placement error during walking (Baker et al., 1999; Ferrari et al., 2008; Kadaba, Ramakrishnan, Wooten, Gainey, Gorton and Cochran, 1989; Szczerbik and Kalinowska, 2011). Simulated displacements in THI marker position cause large errors in transverse plane hip and frontal plane knee kinematics, both of which have been associated with increased frontal plane knee loading during CoD (Baker et al., 1999; Mclean et al., 2005; Sigward and Powers, 2007). Errors in frontal plane knee kinematics vary non-uniformly throughout the gait cycle, demonstrating analysis of the entire gait cycle may be required to fully understand the effect of marker placement on joint kinematics.

Calculated joint moments of force are also affected by marker placement. Changing the positions of the THI, KNEE and TIB markers alters the locations of the calculated knee (KJC) and ankle joint centres (AJC), affecting the length of the moment arm used to calculate the joint moment.

Simulated displacements in joint centre positions demonstrate this, with 10 mm anterior displacements causing significant differences in net knee moments during walking (Holden and Stanhope, 1998; Stagni et al., 2000).

The specific sensitivity of kinematic and kinetic variables to systematic differences in marker placement remains unclear. The effect of marker placement will vary depending on the variable being reported, the marker in question, the magnitude of displacement and the phase of the movement being analysed. To reliably make inferences related to ACL injury from data collected in different laboratories and by different practitioners, we must establish the sensitivity of lower extremity kinematics and knee moments to systematic differences in marker placement. The aim of this investigation was to determine the sensitivity of joint kinematics of the hip, knee and ankle, as well as knee moments, to systematic displacements in the positions of the THI, KNEE and TIB markers across the stance phase of a CoD task.

2.2.1 Methods

Participants

An a priori power analysis (G*Power, version 3.2.9.2, Universität Düsseldorf, Germany), based on the reported findings of Alenezi et al., (2016) was conducted. Alnenezi et al., (2016) examined the test-retest reliability of a selection of kinematic and kinetic variables of the hip, knee and ankle during a CoD task. From these data, effect sizes for the effect of inter-tester session variation in marker placement were estimated for each variable. The current study was then powered of the variable with the smallest estimated effect size, knee abduction moment. The power analysis indicated that a sample size of 42 participants was required to achieve 80% statistical power with an alpha level of 0.05.

Inclusion criteria for participation were: male, aged 18 – 35, undergone primary ACLR 34 – 43 weeks (mean ± SD: 35.7 ± 2.2 weeks) prior to testing, participation in multi-directional field-based sport prior to ACL injury and intention to return to the same level of participation following rehabilitation. The study received ethical approval from the University of Roehampton, London (LSC 15/122) and the Sports Surgery Clinical Hospital Ethics committee (25AFM010). Participants gave informed, written consent prior to participation in the study.

Data Collection

Testing took place in a biomechanics laboratory, using a ten-camera motion analysis system (200 Hz; Bonita-B10, Vicon, UK), synchronized (Vicon Nexus 2.7) with two force platforms (1000 Hz BP400600, AMTI, USA) recording the positions of 28 reflective markers (14 mm diameter). Markers were secured to the participant's shoe or skin using tape at bony landmarks on the lower limbs, pelvis and trunk according to the PiG marker set

(Marshall et al., 2014).

Prior to data collection, participants undertook a standardised warm-up comprising of a 2-minute jog, 5 bodyweight squats, 2 submaximal and 3 maximal countermovement jumps. A static trial was captured as a reference for the dynamic trials. Each participant completed a pre-planned 90° CoD task. The CoD task followed a wider testing battery that formed part of a larger, ongoing study, in which participants also completed a range of double and single leg jump exercises. The CoD task involved the participants running maximally towards the force platforms then planting their outside foot on the force platform to cut left or right, i.e. planting their left foot to cut to the right. Three valid, maximal effort trials were collected on both the non-operated and operated limb. A full description of the testing protocol is given in King et al., (2018).

Data Processing

Trials in which the participant planted their operated limb on the force platform to complete the CoD task were used for further analysis. Marker trajectory and force data were low-pass filtered using a fourth-order Butterworth filter (cut-off frequency 15 Hz) (Kristianslund et al., 2012). Systematic displacements were then applied in software to the positions of the THI, KNEE and TIB markers. One marker position displacement was applied at a time along the corresponding segment x-axis using:

$$Xk^{'} = T \cdot Xk$$

where $Xk^{'}$ are the new, displaced marker coordinates within the segment coordinate system, T is the translational matrix and Xk are the original marker coordinates within the segment coordinate system (Fig 2.2). Displacements were applied to marker positions in 5 mm increments, to 20 mm anterior and 20 mm posterior from their original positions, resulting in 8 displacement conditions for each marker. Data processing created three separate data sets: A, B and C. Each data set contained displacements of a single marker and were identical except for the position of the corresponding marker.

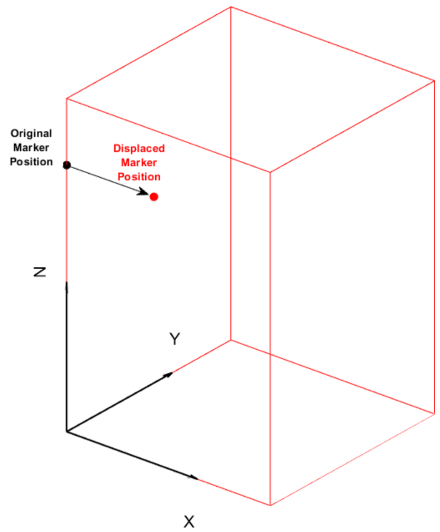


Figure 2.2: Visualisation of marker displacements. Marker coordinates were transformed into local segment coordinates (black circle) and displaced anteriorly/posteriorly along the segment x-axis (red circle).

Stance phase was identified for each trial from when vertical ground reaction force passed above and below 20 N. Tri-planar joint angles at the hip, knee and ankle, as well as tri-planar knee moments were extracted during stance phase for each trial. Kinematic and kinetic signals were time normalised to 101 data points and the mean of each participant's three trials was used for further analysis.

Sensitivity Analysis

One-dimensional statistical parametric mapping (SPM) was used to analyse the effect of marker placement across the entire stance phase of the CoD task (Pataky, 2010; Pataky et al., 2014). Our analysis aimed to simulate a scenario in which we were testing for between group differences in groups which were identical except for the position of the corresponding marker. This would allow us to identify the minimum systematic differences in marker placement required to result in incorrect statistical inferences when making between group comparisons in each variable. For clarity, we will use the example of one data set, data set A, as the process was repeated identically for data sets B and C. Following data processing, nine signals for each variable for each participant were contained in data set A. These corresponded to the original unaltered trial, as well as each of the THI marker displacement conditions (Fig 2.3).

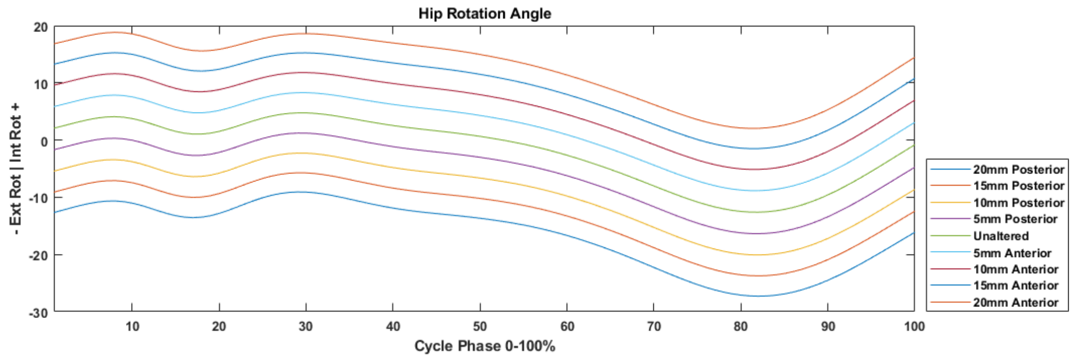


Figure 2.3: Example of kinematic signals produced for one participant following data processing. Image depicts hip rotation angle under each THI marker displacement condition.

Each variable in data set A was submitted to a 1D independent samples SPM t-test between the unaltered condition and each of the displacement conditions. This process produced 8 SPMt curves for each variable, one for each THI marker displacement condition (Fig 2.4). The significance of each SPMt curve was determined topologically using random field theory ($a < 0.05$) (Pataky et al., 2015). Phases of the SPMt curve above the critical-t threshold were identified as significantly affected by the corresponding marker displacement. To aid in interpretation of results, SPMt curves were plotted using image inference surface plots (Fig. 2.5). A variable's "sensitivity" to marker placement was determined by the minimum marker displacement required to cause significant differences, with more sensitive variables significantly affected by smaller marker displacements across larger periods of stance phase.

As we experimentally created the difference between conditions by displacing each marker in a fixed direction from its original position, the changes to outcome variables will be unidirectional and predictable in nature. For

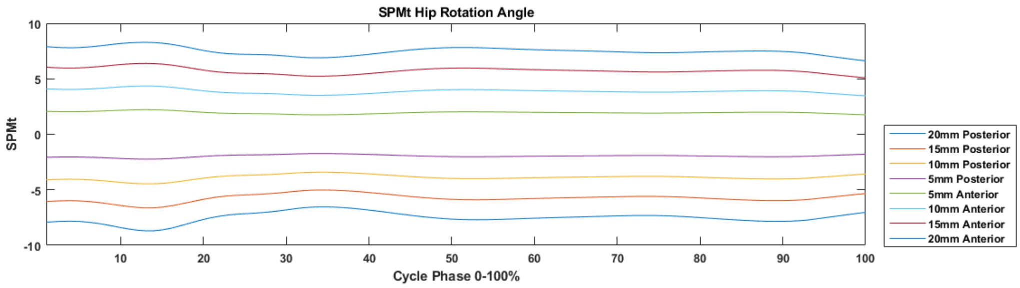


Figure 2.4: Example of SPMt curves for one variable following 1D independent samples SPM t-test between the unaltered condition and each of the marker displacement conditions. Image depicts SPMt for hip rotation angle for each THI marker displacement condition.

example, an anterior displacement of the THI marker will always result in a more internally-rotated calculated position of the thigh segment. The test statistic produced following comparisons between the unaltered condition and each displacement condition is therefore a function of sample size and effect size, meaning that the likelihood of finding a statistically significant differences between conditions is increased at larger sample sizes. In acknowledgment of this, we included sample size as an extra degree of freedom in our analysis. We chose sample sizes of n = 10, n = 25 and n = 50, as these represent the low, mid and upper ranges of sample sizes typically used in biomechanical studies (Besier et al., 2003; Ithurburn et al., 2017; Sankey et al., n.d.; Wen et al., 2018).The sensitivity analysis procedure outlined above was repeated for each variable in data sets A, B and C, at each sample size, resulting in a total of nine sensitivity analyses.

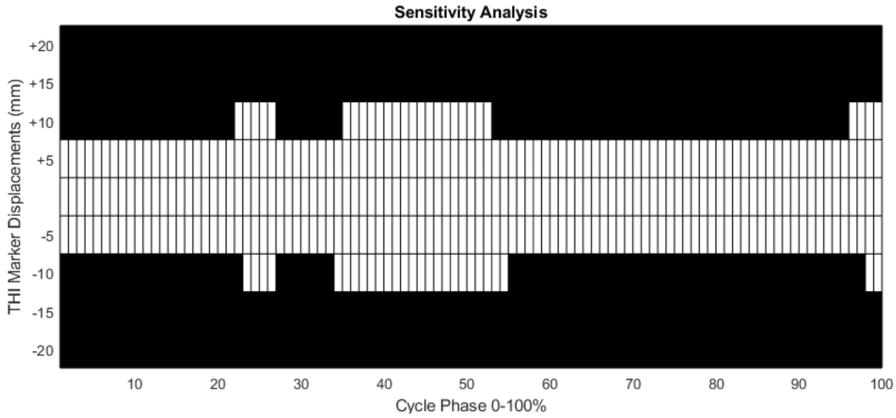


Figure 2.5: Example of sensitivity analysis for hip rotation angle under THI marker displacements. Image depicts each hip rotation angle SPMt as a function of both time and THI marker displacement. Image inference depicting phases of the waveform (x-axis) significantly affected (black) by the corresponding THI marker displacements (y-axis). In this example, hip rotation angle was not significantly affected by THI marker displacements of 5 mm. However, various periods of stance were significantly affected by 10 mm anterior (1-21%, 27-34% and 55-95%) and 10 mm posterior (1-22%, 27-33% and 55-98%) displacements, while displacements of 15 mm and above significantly affected the entire stance phase.

2.3 Results

The results of the sensitivity analyses for the THI, KNEE and TIB markers are presented in Figures 2.6, 2.7 and 2.8 respectively. See supplementary material – Appendix C, for individual sensitivity analyses for each variable. As sample size increased, the magnitude of the marker displacement required to cause significant differences in each variable decreased, and/or the cumulative percentage of stance phase significantly affected by marker displacements increased.

2.3.1 Thigh Marker

No variables were significantly affected by 5mm THI marker displacements. Four variables were significantly affected by displacements of 10 mm and for longer periods of early, mid and late stance (Fig 2.6B, 2.6C). These variables were hip rotation angle, knee abduction angle, ankle abduction angle and ankle rotation angle. Of these, hip rotation and knee abduction angles were most sensitive to THI marker placement, with 10 mm displacements causing significant differences across the entire stance phase at $n = 50$ (Fig 2.6C). At $n = 10$, only hip rotation and knee abduction angles were significantly affected by THI marker displacements of any magnitude. The sensitivity of these variables increased as sample size increased, while at $n = 25$ and $n = 50$, ankle abduction and rotation angles were also significantly affected (Fig 2.6B, 2.7C).

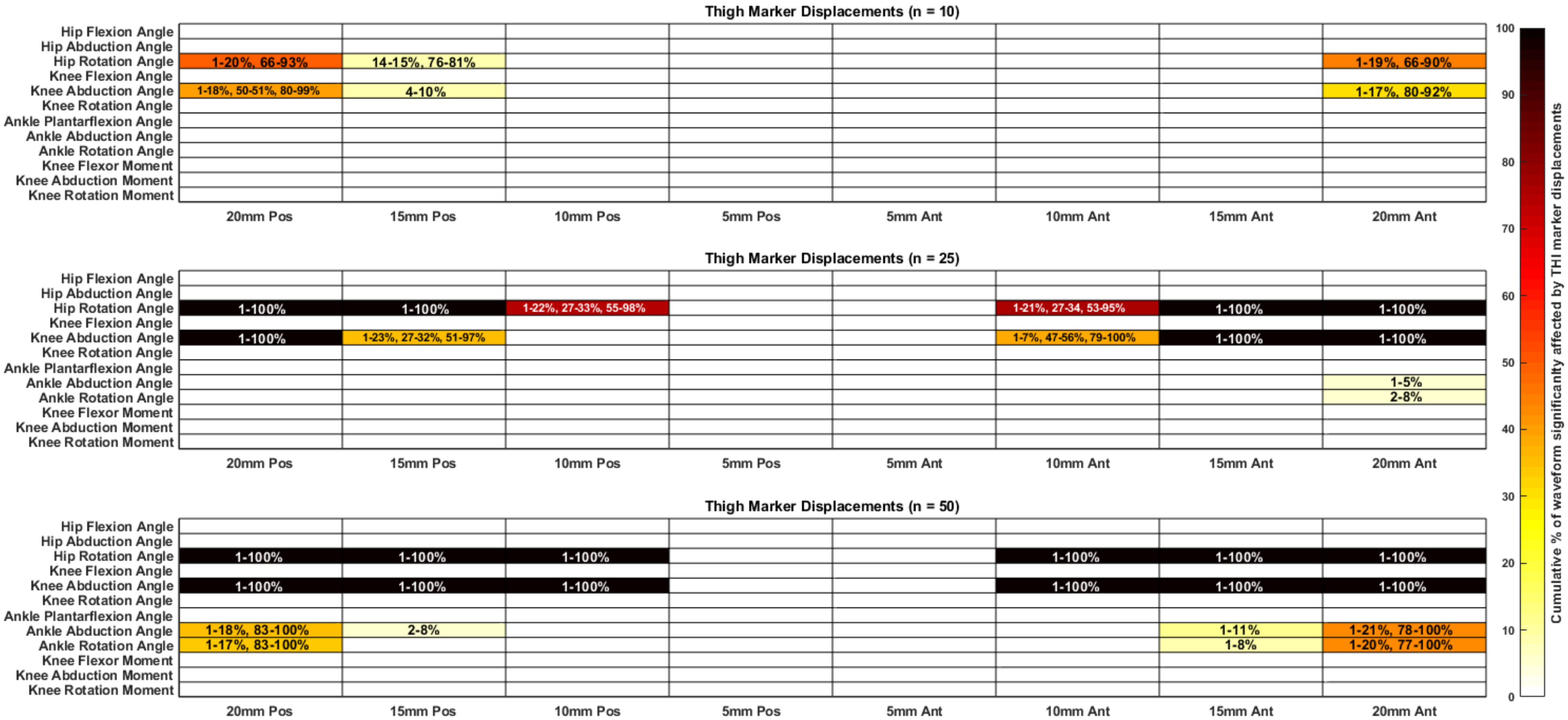


Figure 2.6: THI marker sensitivity analysis for n = 10 (A), n = 25 (B) and n = 50 (C). In-bar text corresponds to the specific phases of the waveform significantly affected by the corresponding marker displacement. Colour corresponds to the cumulative percentage of the signal significantly affected by the corresponding marker displacement.

2.3.2 Knee Marker

No variables were significantly affected by 5 mm KNEE marker displacements (Fig 2.7). Eight variables were significantly affected by KNEE marker displacements of 10 mm and above (Fig 2.7C). These were hip rotation angle, knee flexion angle, knee rotation angle, ankle plantar-flexion angle, ankle abduction angle, knee flexor moment and knee abduction moment (Fig 2.7B, 2.7C). Of these, ankle abduction and rotation angles were most sensitive to KNEE marker displacements, with 10 mm displacements causing significant differences across the first and last 20% of stance (Fig 2.7C). At n = 10, no variables were significantly affected by KNEE marker displacements of any magnitudes. At n = 25, ankle plantar-flexion, ankle abduction, ankle rotation, knee flexor moment and knee abduction moment were significantly affected (Fig 2.7B), while at n = 50, hip rotation, knee flexion, knee abduction and knee rotation angles were also significantly affected (Fig 2.7C).

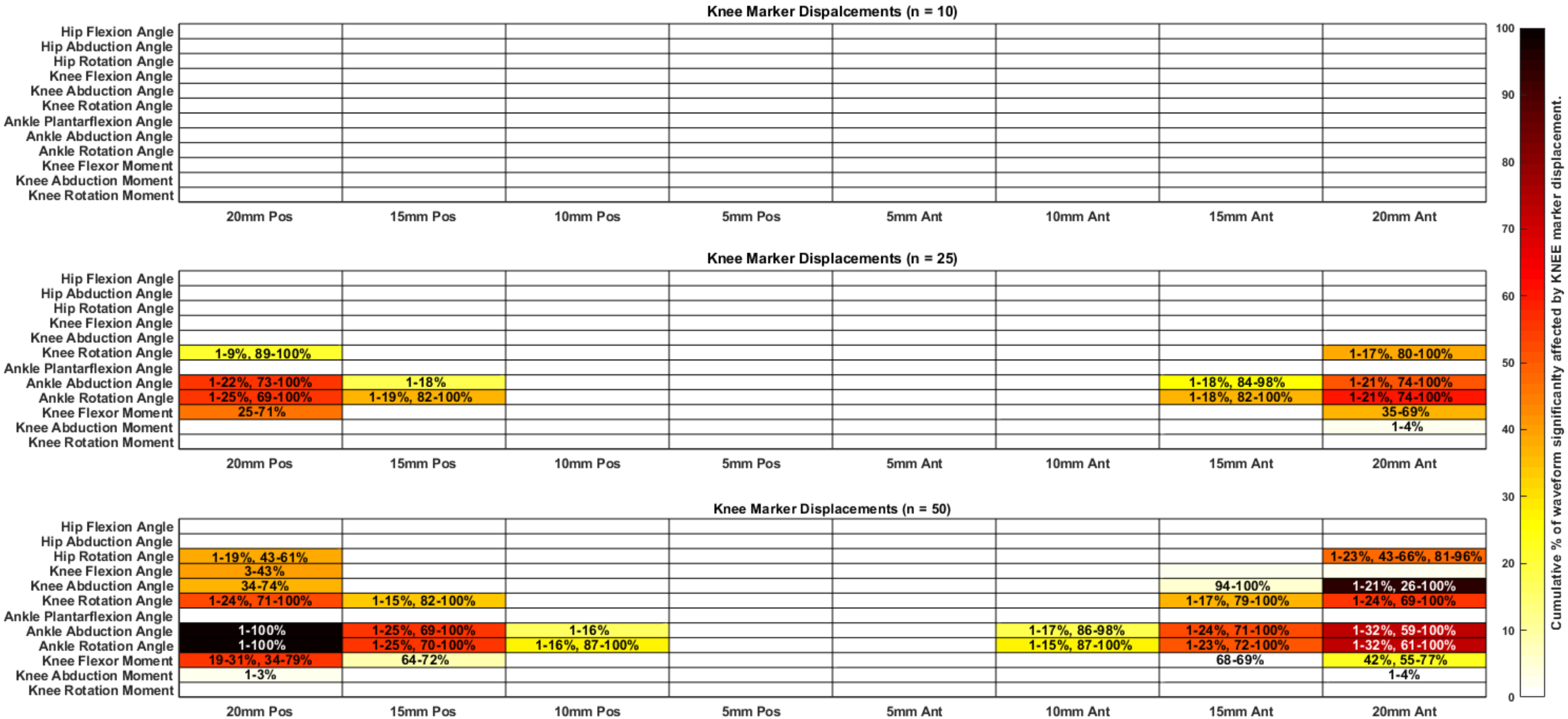


Figure 2.7: KNE marker sensitivity analysis for n = 10 (A), n = 25 (B) and n = 50 (C). In-bar text corresponds to the specific phases of the signal significantly affected by the corresponding marker displacement. Colour corresponds to the cumulative percentage of the waveform significantly affected by the corresponding marker displacement.

2.3.3 Tibia Marker

5 mm TIB marker displacements significantly affected three kinematic variables (Fig 2.8C). These were, knee rotation angle, ankle abduction angle and ankle rotation angle. Displacements of 10 mm and above also significantly affected ankle plantar-flexion angle, knee flexor moment and knee abduction moment (Fig 2.8B, 2.8C). Knee rotation angle was the most sensitive variable to TIB marker displacements, and the only variable to be significantly affected across the entire stance phase by any 5 mm marker displacements (Fig 2.8C). At $n = 10$, knee rotation angle, ankle abduction angle, ankle rotation angle and knee abduction moment were significantly affected by TIB marker displacements (Fig 2.8C). The sensitivity of these variables increased as sample size increased, while ankle plantar-flexion angle and knee abduction moment were also significantly affected at $n = 25$ and $n = 50$ (Fig 2.8B, 2.8C).

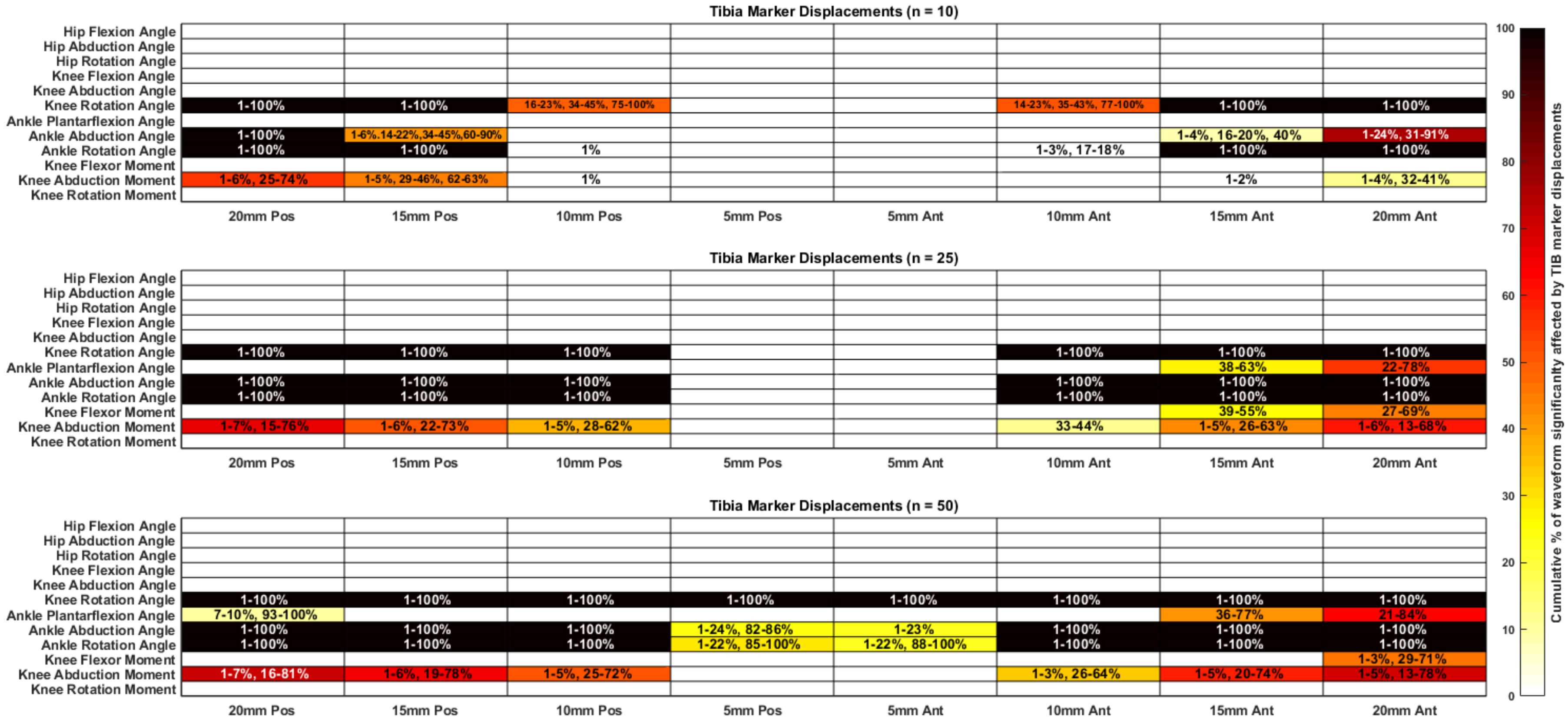


Figure 2.8: TIB marker sensitivity analysis for n = 10 (A), n = 25 (B) and n = 50 (C). In-bar text corresponds to the specific phases of the signal significantly affected by the corresponding marker displacement. Colour corresponds to the cumulative percentage of the signal significantly affected by the corresponding marker displacement..

2.4 Discussion

Inter-tester variability in the anterior/posterior positions of the anatomical landmarks used to define the positions of the THI, KNEE and TIB markers is reported as ranging between 9.3 – 12.5 mm (Della Croce et al., 1999). Several variables previously associated with ACL injury risk and rehabilitation status were significantly affected by marker displacements within, or bordering on, reported inter-tester variability ranges. These were hip rotation angle, knee abduction angle, ankle rotation angle and knee abduction moment (Dempsey et al., 2007; Mclean et al., 2005; Sigward and Powers, 2007).

Frontal and transverse plane kinematics were most sensitive to marker placement in each marker condition and at every sample size. This is unsurprising given the known limitations of the CGM in assessing frontal and transverse plane kinematics (Baker et al., 1999; Kadaba, Ramakrishnan and Wooten, 1989). Changes in the anterior/posterior positions of the THI, KNEE and TIB markers causes misalignment of the primary and secondary axis of the femur and shank segments. These alterations create a rotational offset, while also resulting in cross-talk between segment axes. This manifests as error in angles calculated in all three planes, and is most pronounced in the frontal and transverse plane kinematics (Baker, Finney, and Orr 1999b). Previous studies using descriptive statistics (Szczerbik and

Kalinowska, 2011), root mean square differences (Groen et al., 2012) and qualitative assessments (Kadaba, Ramakrishnan and Wooten, 1989) to examine the effect of marker placement on joint kinematics during walking report similar findings.

Our findings build on those from previous work and demonstrate the minimum systematic differences in marker placement required to cause statistically significant differences in each variable at three different sample sizes. Utilising a continuous statistical analysis method (SPM) allowed us to identify the specific phases of each kinematic and kinetic signal significantly affected by marker displacements. Statistically significant differences first appeared in many outcome variables across the first and last 20% of stance, indicating these phases are most sensitive to marker placement (Fig 2.6, 2.7, 2.8). As non-contact ACL injuries are believed to occur within the first 20% of stance, discrete kinematic and kinetic measures from this period are regularly reported (Pollard et al., 2007; Sigward and Powers, 2007; Stearns and Pollard, 2013). Increased hip internal rotation, knee abduction and ankle external rotation at initial contact of CoD have been associated with higher peak knee abduction moments (Dempsey et al., 2007; Mclean et al., 2005; Sigward and Powers, 2007). Frontal plane knee loading is considered a key risk factor for ACL injury (Hewett et al., 2005). These findings have thus led to the clinical development of ACL prevention and rehabilitation programs aiming to minimise frontal plane knee loading (Distefano et al., 2011).

Statistical significance is often used to draw clinical inferences in ACL research (Dempsey et al., 2007; Ford et al., 2005; King, Richter, Franklyn-Miller, Daniels, Wadey, Jackson, Moran and Strike, 2018; Sigward and Powers, 2007; Stearns and Pollard, 2013). Previous work has reported statistically significant differences in kinematics and kinetics with respect to gender (Ford et al., 2005), limbs (King, Richter, Franklyn-Miller, Daniels, Wadey, Jackson, Moran and Strike, 2018) and injured/uninjured groups (Stearns and Pollard, 2013), and postulated that these differences may highlight variables of interest in rehabilitation and injury prevention. It should be noted that statistical significance is less relevant than the actual magnitude of differences between groups and how such differences would affect clinical inferences/recommendations. Relative to previously published differences, our findings demonstrate magnitudes approximating or exceeding those reported between groups/conditions (Baker et al., 1999; Ford et al., 2005; King, Richter, Franklyn-Miller, Daniels, Wadey, Jackson, Moran and Strike, 2018; Pollard et al., 2007; Stearns and Pollard, 2013). For example, statistically significant differences in hip rotation angle (5.1°), knee abduction angle (2°) and knee abduction moment (0.21, 0.53 and 1 Nm/kg) during CoD tasks have been reported previously and hypothesised to present clinically relevant differences related to ACL injury (Mclean et al., 2005; Sigward and Powers, 2007; Stearns and Pollard, 2013). Within our data, 10 mm THI marker displacements caused significant differences in hip rotation and knee abduction angle with a mean difference of 3.62° and 2.77°

respectively, while 10 mm TIB marker displacements caused significant differences in knee abduction moment with a mean difference of 0.32 Nm/kg (see supplementary material – Appendix C).

Several limitations can be ascribed to the current study. Firstly, we do not know if the original physical marker positions were optimal. Moving the markers anteriorly/posteriorly may have in fact been moving them closer to the original target positions. However, as the effect of systematic marker displacements on outcome variables is unidirectional, the original marker locations will not affect our general conclusions. Secondly, there is there is likely to be an element of random variation in real-world marker placement, alongside the systematic element investigated here (Osis et al., 2016). Random marker placement error and its effect on kinematics and kinetics requires further research. Also, it is important to note that the specific errors reported in this study are limited to the CoD task analysed, with marker placement likely having a different effect in different tasks (Baker et al., 1999). Lastly, our marker displacements were simplistic in nature and do not directly mimic real world marker placement error. We implemented fixed displacements, meaning markers were moved the same distance relative to the original marker position across all time points of the task. Physically moving markers across a range of $\pm$ 20 mm on the skin would involve a certain amount of medio-lateral in addition to anterior/posterior displacement, as well as different soft tissue artefacts (STA). Different STA's would alter the observed errors in this study, meaning translating our findings directly to real world scenarios is challenging. Separating the effect of marker placement error from that of STA is difficult and the relationship between these two major sources of error is an area that warrants further research. For this study, we chose to focus on simple anterior/posterior displacements, as the model definitions indicate that these are the marker displacements that most substantially effect model outputs (Kadaba, Ramakrishnan and Wooten, 1989). Accounting for the additional effects of medio-lateral displacements and STA went beyond the scope of the current investigation.

Alternative methods for modelling the human body have been developed to mitigate the effect of STA and provide improved anatomical relevance compared to the CGM. These include models that implement the calibration anatomical systems technique (CAST), or models that allow for six degrees of freedom (6DOF) at each joint. Models implementing CAST or 6DOF continue to work on the assumption that marker placement is consistent and repeatable between practitioners (Charlton et al., 2004). Indeed, any model utilising anatomical markers to define joint centres and segment orientations makes this assumption. At present no alternative model or technique has been as widely implemented and validated as the CGM (Baker et al., 2017; Charlton et al., 2004). Research into the sensitivity of alternative modelling techniques to marker placement, and how this compares to the CGM is required prior to any widespread clinical application. While limited in certain aspects, the CGM currently presents a practical, deterministic, extensively validated model that can be easily implemented in routine clinical practice. These factors may explain the continued widespread use of the CGM in contemporary biomechanical research (Cortes et al., 2011; Gore et al., 2016; Lee et al., 2014; Marshall et al., 2015; Mclean et al., 2005; Sigward and Powers, 2007). When utilising the CGM however, it should be done in a manner that openly acknowledges its limitations within the context of the study aims and reported results. If attempting to identify relatively small differences in frontal and transverse plane kinematics for example, it should be made explicitly clear that any identified differences may be attributable to instrumental error such as marker placement.

In conclusion, we have shown that systematic differences in the placement of the THI, KNEE and TIB markers, within or bordering on reported inter-tester variability ranges, can cause statistically significant differences in multiple kinematic and kinetic variables across various periods of CoD stance. Many variables affected have previously been associated with increased frontal plane knee loading during CoD, which is considered a key risk factor for ACL injury. Errors were particularly pronounced across the first 20% of stance, a period from which discrete kinematic and kinetic variables are regularly reported. Our findings demonstrate the minimum systematic differences in marker positions required to cause significant differences in lower extremity kinematics and kinetics. These thresholds can be used by laboratories to establish acceptable levels of inter-tester variability in marker placement. If inter-tester variability is above these thresholds, statistical inferences and corresponding clinical recommendations related to group differences should be made with caution, as marker placement differences may result in invalid conclusions.

Chapter 3

The Effect of Simulated Marker Misplacement on the Interpretation of Inter-Limb Differences During a Change of Direction Task

3.1 Chapter Context

Chapter 3 presents one experimental study examining the effect of random marker placement error on CoD kinematic and kinetic inter-limb differences. This follows on directly from the work presented in chapter 2, which established the sensitivity of CoD kinematics and kinetics to systematic marker placement error across the entirety of CoD stance phase. Chapter 2 identified the minimum systematic differences in marker placement required to cause significant differences in kinematic and kinetic variables at three different sample sizes. While this has important implications for group based analyses using data collected via multiple practitioners, in a real-world setting there will also be an element of random error associated with each marker position.

Random marker placement error has significant implications for repeat assessments of patients. Any between session change in kinematic or kinetic variables may be attributable to random variation in marker positions between assessments, as opposed to genuine changes in these metrics. If biomechanical analyses of CoD kinematics and kinetics are to be used as a means of monitoring individual patient rehabilitation progress following ACLR, the effect of random marker placement error on such assessment must be explored. Inter-limb and inter-group comparisons with non-injured groups are the two most common methods of monitoring rehabilitation progress following ACLR. Thus, this chapter explores this issue by examining how random marker placement error influences the ability to identify and monitor discrete measures of inter-limb differences in kinematics and kinetics during CoD, as well as contextualise the magnitude of these inter-limb differences relative to a normative cohort.

3.2 Introduction

Following anterior cruciate ligament reconstruction (ACLR), inter-limb differences in kinematic and kinetic measures have been observed across various tasks including walking (Wellsandt et al., 2016; White et al., 2013), running (Kline et al., 2016), jumping (Jordan et al., 2015; Orishimo et al., 2010), landing (Gokeler et al., 2010; Orishimo et al., 2010) and change of direction (CoD) (King, Richter, Franklyn-Miller, Daniels, Wadey, Jackson, Moran and Strike, 2018). Individuals exhibiting large inter-limb differences are believed to be at increased risk of negative long-term outcomes, including the development of osteoarthritis and re-injury (Paterno et al., 2010; Wellsandt et al., 2016). Objective assessment of inter-limb differences is proposed as a means of monitoring rehabilitation post-ACLR (Jordan et al., 2015; Myer et al., 2011; Paterno et al., 2007). Implementing such assessments in a clinical environment necessitates that the metrics of interest be reliable and robust to methodological sources of variability.

Data from human motion analysis are prone to error from multiple sources, including instrumental errors, soft tissue artefact and inaccurate placement of anatomical markers (Schwartz and Dixon, 2018). Error in marker placement is recognized as a key source of methodological variability, with low between-session reliability measures attributed to variation in marker positions between sessions (Alenezi et al., 2016; Fonseca et al., 2020; Ford et al., 2007; Gorton et al., 2009; Groen et al., 2012; Kadaba, Ramakrishnan, Wooten, Gainey, Gorton and Cochran, 1989; McGinley et al., 2009; Szczerbik and Kalinowska, 2011). Experimental studies conducted in walking attribute large errors in calculated joint angles and moments to erroneous marker placement (Groen et al., 2012; Szczerbik and Kalinowska, 2011). Extrapolating findings from studies conducted in walking directly to different tasks is challenging. The effect of marker placement error is likely task specific, with features such as sagittal plane range of motion and walking speed shown to influence the observed effect of marker placement (Baker et al., 1999; Cockcroft et al., 2016; Groen et al., 2012; Szczerbik and Kalinowska, 2011). Despite the use of marker-based biomechanical models to study various lower-limb movement tasks, previous research examining marker placement has focused primarily on walking (Baker et al., 1999; Groen et al., 2012; Szczerbik and Kalinowska, 2011).

CoD tasks are commonly examined in studies related to ACL injury and rehabilitation (King et al., n.d.; Mclean et al., 2005; Pollard et al., 2007; Dos'Santos et al., 2018; Sigward and Powers, 2007; Stearns and Pollard, 2013). CoD manoeuvres are ubiquitous in field-based sports, mechanically demanding and reported as the most common mechanism of non-contact ACL injury (Geli-Alentorn et al., 2009; Olsen et al., 2004). Recent research identified multiple inter-limb differences during CoD tasks at 9 months post-ACLR despite no statistical difference in performance times between limbs (King, Richter, Franklyn-Miller, Daniels, Wadey, Jackson, Moran and Strike, 2018). Inter-limb differences were observed in variables associated ACL injury, suggesting their assessment may be relevant to rehabilitation. It is unclear how error in marker placement influences the ability to identify and monitor such inter-limb differences. Were marker placement error to cause substantial changes to inter-limb differences and their interpretation, it could result in an individual returning to play despite the continued presence of deficits that place them at increased risk of injury.

The conventional gait model (CGM) is a widely used marker-based biomechanical model originally developed for use in clinical gait analyses (Kadaba, Ramakrishnan, Wooten, Gainey, Gorton and Cochran, 1989). While alternative modelling techniques are more commonly used for dynamic tasks, the CGM has nevertheless been used in the analysis of a broad range of CoD tasks (Franklyn-Miller et al., 2017; Marshall et al., 2014; Pollard et al., 2007; Stearns and Pollard, 2013). In the CGM, the anterior/posterior positions of markers on the lateral thigh (THI), lateral femoral epicondyle (KNE) and lateral tibia (TIB) directly influence calculated kinematics and kinetics at the hip, knee and ankle (Baker et al., 1999; Groen et al., 2012; Kadaba, Ramakrishnan and Wooten, 1989; Stagni et al., 2000). Previous studies examining these marker positions have tended to implement fixed, systematic marker displacements, whereby a marker's position is moved by a set amount from its original position and the subsequent effect on model outputs is examined (Baker et al., 1999; Cockcroft et al., 2016; Groen et al., 2012; McFadden et al., 2020). For example, Groen et al., (2012) demonstrated that systematic 14 mm anterior displacements to the THI, KNE and TIB markers caused errors greater than 10° in lower extremity joint angles between repeated walk trials.

While systematic differences in marker placement such as those outlined may exist, these study designs fail to account for the inherent randomness to be expected in real world marker placement error (Myers et al., 2015; Osis et al., 2016). Challenges also exist in separating the effect of marker displacements from that of movement variability, as any observed changes in kinematics and/or kinetics will be attributable to both the marker displacement and natural trial-to-trial movement variability. Utilising a simulated approach offers the opportunity to control for movement variability and examine the effect of marker placement in isolation. This approach has been used previously to study marker displacements during walking (Myers et al., 2015) and running (Osis et al., 2016). Simulated marker displacements sampled from the Gaussian distribution have been used to mimic expected real-world variation in marker placement (Myers et al., 2015). Establishing how such marker displacements impact the ability to identify and monitor changes in inter-limb differences will inform the contexts in which their use as objective rehabilitative measures is appropriate, and those in which they are not. Thus, the aim of this investigation was to determine the effect of random THI, KNE and TIB marker displacements on the interpretation of inter-limb differences during a CoD task.

3.3 Methods

3.3.1 Participants

Forty-seven male participants aged 18-35 (mean $\pm$ SD age 24.8 $\pm$ 4.8 years, height 180 $\pm$ 6 cm and mass 84 $\pm$ 6.4 kg) approximately 9 months (8.7 $\pm$ 0.7) post primary ACLR were recruited from the caseload of two orthopaedic surgeons, based in the Sports Surgery Clinic, Dublin, Ireland. Inclusion criteria for ACLR participation in the study were male, aged 18-35, participation in multi-directional field-based sports prior to injury and the intention to return to the same level of participation post rehabilitation. Participants who had multiple ligament reconstructions, meniscal repair or did not intend to return to multidirectional field-based sport were excluded from the study. A matched healthy cohort (NORM) of 50 participants (23.4 $\pm$ 3.7 years, 182.8 $\pm$ 6.38 cm, 81.9 $\pm$ 7.4 kg) with no history of lower limb injury were recruited locally from multi-directional field-based sports teams. Ethical approval was received from the University of Roehampton, London

(LSC 15/122) and the Sports Surgery Clinic Hospital Ethics Committee (25AFM010). Participants gave informed, written consent prior to participation in the study.

Data collection took place in a biomechanics laboratory using a ten-camera motion analysis system recording the positions of 28 reflective markers (14 mm diameter). Markers were secured using tape at bony landmarks on the lower limbs, pelvis and trunk according to a modified Plug-in-Gait marker set (Marshall et al. 2014). Participants completed a pre-planned 90° CoD task, which followed a wider testing battery that formed part of a larger, on-going study. The full testing battery comprised of a standardised warm-up, consisting of a 2-minute jog, 5 bodyweight squats, 2 submaximal and 3 maximal countermovement jumps, followed by a series of double and single leg jump exercises. The CoD task involved the participants running maximally towards the force platforms before planting their outside foot on the force platform to cut left or right, i.e. planting their right foot to cut to the left. The start line was 5 m from the force plates, while the finish line was 2 m from the force plates. Three trials were collected on both the ACLR and contralateral limbs. A full description of the testing protocol is given in King et al. (2018).

3.3.2 Data Processing

A fourth order zero-lag Butterworth filter (cut-off frequency 15 Hz) was used to filter marker trajectory and force data (Kristianslund et al., 2012). Kinematic variables at the hip, knee and ankle have been associated with increased knee loading, quantified in the form of knee joint moments (Dempsey et al., 2007; Mclean et al., 2005). Tri-planar hip, knee and ankle angles, as well as tri-planar knee joint moments, were therefore extracted during stance phase for each trial. Initial contact and toe-off were identified from vertical ground reaction force using a 20 N threshold. Kinematic and kinetic signals were time normalised to 101 data points and the mean of each participant's three trials was used for further analysis. Inter-limb differences were calculated for each variable at 20% of stance. Video analyses of ACL injuries suggest injury occurs within this period (Koga et al., 2010; Krosshaug et al., 2007; Olsen et al., 2004) and thus it is extensively studied in ACL and CoD research (Dempsey et al., 2007; Olsen et al., 2004; Robinson et al., 2014; Stearns and Pollard, 2013). Inter-limb differences were calculated in the ACLR and NORM groups respectively as

$$ACLR - NonACLR$$

$$NonDominant - Dominant$$

The 'dominant limb' was defined in the NORM group as the self-selected preferred kicking leg. The time point of 20% of stance corresponded to 0.065 $\pm$ 0.02 ms and 0.068 $\pm$ 0.01 ms for the ACLR and contralateral limbs respectively, and 0.065 $\pm$ 0.01 ms and 0.064 $\pm$ 0.02 ms for the dominant and non-dominant limbs respectively.

Following initial data processing, we simulated a scenario in which our ACLR cohort underwent repeated testing sessions, with random marker displacements introduced in each session. One hundred copies of each ACLR participant's six original trials (3 x cutting off ACLR limb and 3 x cutting off contralateral limb) were generated, with each set of six trials corresponding to a "simulated" testing session. In each of the one-hundred simulated sessions, random displacements were sampled and applied to each marker (THI, KNE and TIB) on the CoD stance leg in all trials completed on the ACLR and contralateral limbs, mimicking a real-world scenario in which individual marker positions were invariant across trials within a session but the displacement applied to each marker was independent. Unique displacements were generated for each participant and were sampled from the Gaussian distributions created from variance based on previously reported intra-tester variability in anatomical landmark location (Della Croce et al., 1999; Myers et al., 2015). For markers that did not have directly reported intra-tester variability ranges (THI and TIB), the mean variance of anatomical landmarks located on the associated segment was used (Della Croce et al., 1999). Thus, displacements were drawn from distributions with standard deviations of 5.8 mm (THI), 3.9 mm (KNE) and

3.4 mm (TIB). To simulate anterior/posterior positioning error, displacements were applied about the anterior-posterior axis of the corresponding segment coordinate system using:

$$Xk^{'} = R \cdot Xk$$

where $Xk^{'}$ are the new, displaced marker coordinates within the segment coordinate system, R is the translational matrix containing the randomly generated marker displacement and Xk are the original marker coordinates within the segment coordinate system. Within each simulated testing session, mean kinematic and kinetics were extracted as described previously. Using these newly formed mean kinematic and kinetic measures inter-limb differences were recalculated in each of the one-hundred simulated testing sessions. This process produced a total of 4700 inter-limb differences for each variable (47 ACLR participants x 100 inter-limb differences). The displacement process described was completed using a custom written MATLAB script (version2019, The Mathworks Inc, Natick, Massachusetts, USA).

Each ACLR participant's original inter-limb difference was subtracted from their original 100 simulated inter-limb differences for each variable. This produced a distribution of changes in inter-limb differences attributable to marker displacements, from which 95% confidence intervals were estimated. These intervals constituted a range in which the true value of an inter-limb difference was expected to fall on 95% of occasions when accounting for variability introduced from marker placement. Using the identified confidence intervals, the minimal change in inter-limb differences that could be identified with 95% certainty between two assessments was estimated for each variable. This was identified as the point in which % of possible values for an initial observation fell outside a range which contained 95% of possible values for a second observation and corresponded to change of 3.6 SD between two assessments (Fig 3.1).

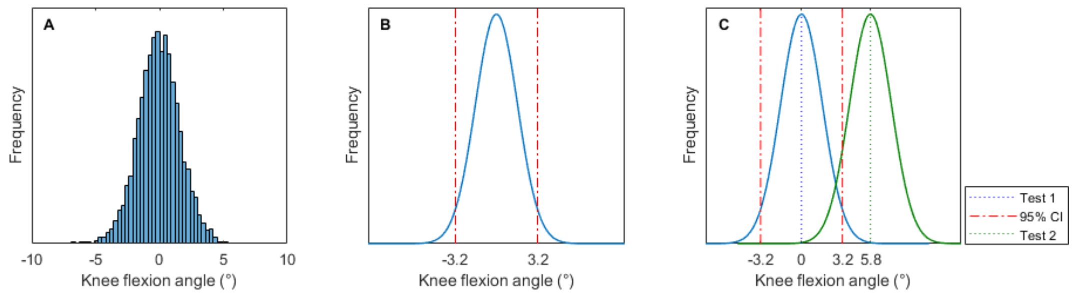


Figure 3.1: Example of confidence interval and minimal identifiable change estimation process. Fig 3.1A depicts the distribution of all observed changes in knee flexion inter-limb differences from marker displacements. This distribution had a standard deviation of 1.6°, from which a 95% confidence interval of ± 3.2° (1.96 x SD) was estimated (3.1B). This constituted a range in which the true value of a knee flexion angle inter-limb difference was expected to fall on 95% of occasions when accounting for variability introduced from marker placement. Using this confidence interval, the minimal change in inter-limb difference that could be identified with 95% certainty between two observations was estimated (Fig 3.1C). For example, if the initial inter-limb difference was 0, possible values of the true inter-limb difference fell within a range of -3.2° < 0 < 3.2°. From this, a minimal change of 5.8° in a subsequent assessment was necessary to be 95% certain that the observed change was not attributable to marker placement. This was the point where 95% of possible values of the second inter-limb difference measure fell outside a range of 95% of possible values of the first inter-limb difference, corresponding to a change of 3.6 SD between two tests.

Following this, descriptive statistics were calculated for NORM inter-limb differences. Each ACLR participant was classified relative to the NORM group as having either a “normal” or “abnormal” inter-limb difference for each variable. ACLR participants with original inter-limb differences between ± 2 SD of the NORM group’s original inter-limb difference were classified as “normal”, while those > ± 2 SD were classified as “abnormal” (Fig 3.1C). ACLR participants were reclassified using each of their 100 simulated inter-limb differences. The percentage of participants whose classification changed from their original in at least one simulation was calculated for each variable (Fig 3.2).

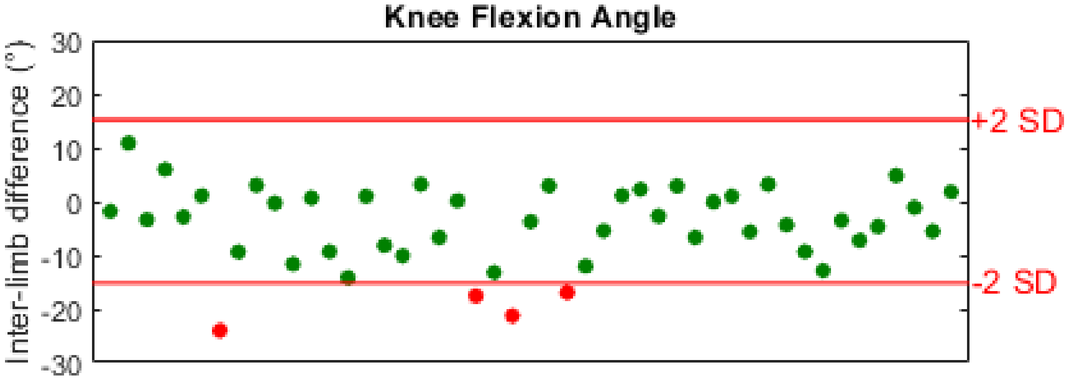


Figure 3.2: Example of inter-limb difference classifications. Image depicts each ACLR participants' original mean inter-limb difference in knee flexion angle at 20% of stance. Participants with a mean inter-limb difference between ± 2SD of NORM group mean inter-limb differences were classified as having "normal" inter-limb differences (green). Participants with mean inter-limb differences above or below 2SD were classified as having an "abnormal" inter-limb difference (red).

3.4 Results

The distribution of changes in inter-limb differences and change in classifications for hip angles, knee angles, ankle angles and knee moments are presented in Figures 3.3, 3.4, 3.5 and 3.6 respectively. The largest minimal identifiable changes and highest percentage of participants to change inter-limb difference classification were in transverse plane hip as well as frontal and transverse plane knee kinematics (Table 3.1).

Table 3.1: Summary results for effect of marker displacements on inter-limb differences. Columns 'SD' and 'Min identifiable change' present the standard deviation and minimal identifiable change in each variable. The percentage of participants who changed their inter-limb difference classification on at least one occasion in the one-hundred simulated testing sessions is presented in column 'Classification change'. For these participants, the median percentage of simulations in which their individual classification changed from its original status is presented in column 'Median classification change' alongside the range of percentages for individual participants.

Variable	SD	Min Identifiable Change	Classification Change	Median Classification Change (Min – Max)
Hip flexion angle	1.3°	4.6°	14.9%	8% (1 – 10%)
Hip abduction angle	0.2°	0.9°	2.1%	15% (-)
Hip rotation angle	6.2°	22.2°	83%	13% (1 – 53%)
Knee flexion angle	1.6°	5.8°	10.6%	9% (5 – 17%)
Knee abduction angle	5.1°	18.3°	87.2%	8% (1-53%)
Knee rotation angle	6.8°	24.6°	91.5%	7% (1-52%)
Ankle plantarflexion angle	1.3°	4.6°	14.9%	2% (1-36%)
Ankle abduction angle	1.8°	6.3°	70.2%	4% (1-50%)
Ankle rotation angle	6.4°	23°	74.5%	9% (1-48%)
Knee flexor moment	0.2 Nm/kg	0.8 Nm/kg	25.5%	18% (1-51%)
Knee abduction moment	0.2 Nm/kg	0.7 Nm/kg	32%	6% (2-44%)
Knee rotation moment	0.01 Nm/kg	0.05 Nm/kg	-	-

Hip Kinematics

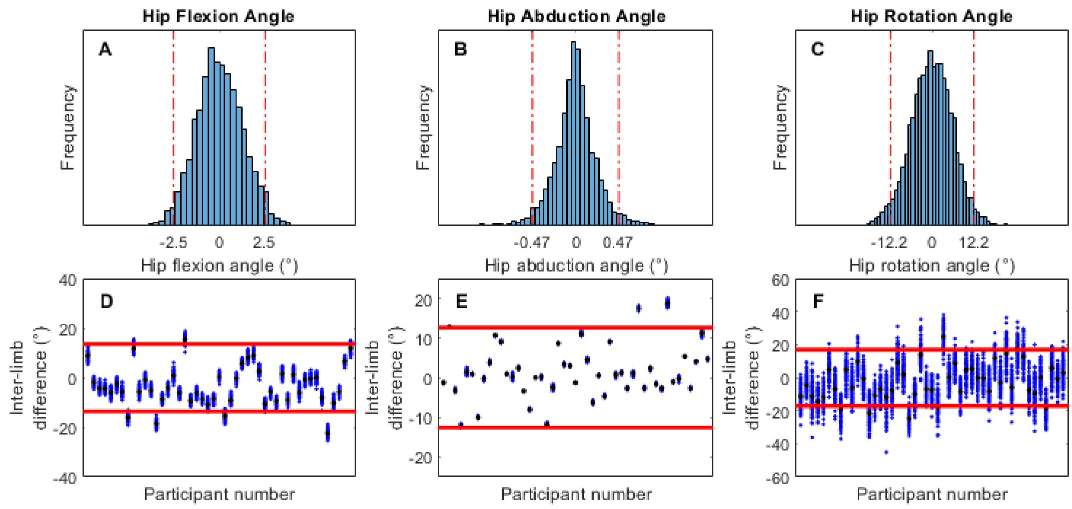


Figure 3.3: Effect of marker displacements on the interpretation of inter-limb differences in hip kinematics at 20% of stance. Top panel depicts the distribution of changes in inter-limb differences and 95% confidence intervals for hip flexion (A), hip abduction (B) and hip rotation (C) inter-limb differences. Bottom panel depicts each ACLR participants inter-limb difference (black = original inter-limb difference, blue = simulated inter-limb differences) relative to the NORM group variability (red lines) for hip flexion (D), hip abduction (E) and hip rotation (F) inter-limb differences.

Marker displacements caused a change in hip flexion inter-limb differences with a standard deviation of 1.3°, from which a 95% confidence interval of -2.5°–2.5° was estimated (Fig 3.3A). Changes in hip abduction angle inter-limb differences had a standard deviation of 0.2° and a confidence interval of -0.47°–0.47° (Fig 3.3B), while for hip rotation angle the standard deviation was 6.2° and confidence interval -12.2°–12.2° (Fig 3.3C). Minimal identifiable changes of 4.6° in hip flexion, 0.9° in hip abduction and 22.2° in hip rotation inter-limb differences were estimated (Table 3.1).

For hip flexion angle inter-limb differences, 40 (85.1%) ACLR participants maintained their original inter-limb difference classification in all one hundred simulations, while 7 (14.9%) participants' classification changed on at least one occasion (Fig 3.3D). The number of simulations in which each participant changed classification ranged from 1 – 10, with a median of 8 (Table 3.1). 46 (97.9%) maintained their original classification for hip abduction angle inter-limb differences in all one hundred simulations, with 1 (2.1%) participant changing classification in 15 simulations (Table 3.1). Lastly, for inter-limb differences, 8 (13%) participants maintained their original classification while 39 (87%) changed classification in at least one simulation (range 1 – 53, median 13) (Table 3.1).

Knee Kinematics

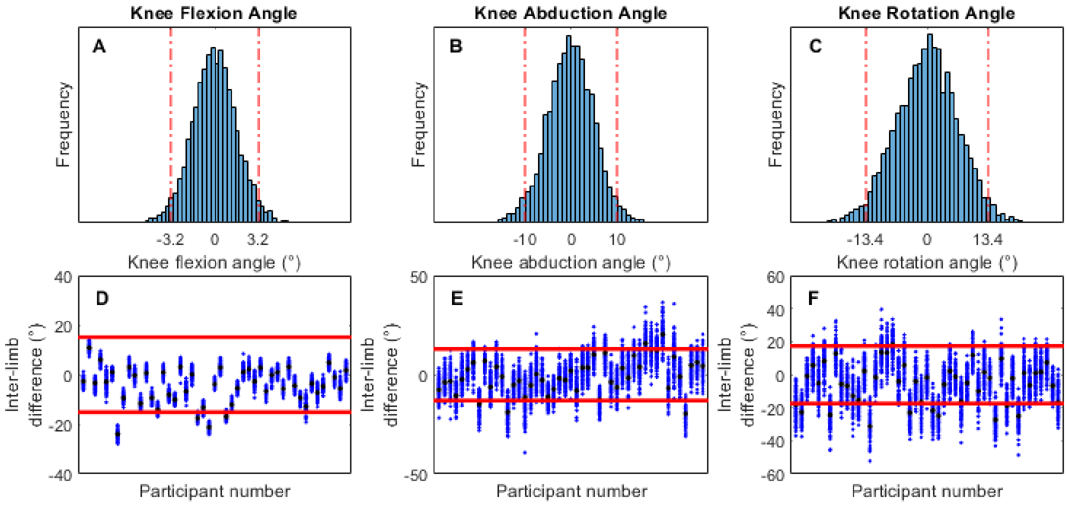


Figure 3.4: Effect of marker displacements on the interpretation of inter-limb differences in knee kinematics at 20% of stance. Top panel depicts the distribution of changes in inter-limb differences and 95% confidence intervals for knee flexion (A), knee abduction (B) and knee rotation (C) inter-limb differences. Bottom panel depicts each ACLR participants inter-limb difference (black = original inter-limb difference, blue = simulated inter-limb difference) relative to the NORM group variability (red lines) for knee flexion (D), knee abduction (E) and knee rotation (F) inter-limb differences.

Marker displacements caused a change in knee flexion inter-limb differences with a standard deviation of 2.3°, from which a 95% confidence interval of -3.2°–3.2° was estimated (Fig 3.4A). The standard deviation for changes in knee abduction inter-limb differences was 5.1° with an estimated confidence interval of -10°–10° (Fig 3.4B), while in knee rotation angle the standard deviation was 6.8° and confidence interval -13.4°–13.4° (Fig 3.4C). Minimal identifiable changes of 5.4° in knee flexion, 18.3° in knee abduction and 24.5° in knee rotation inter-limb differences were estimated (Table 3.1).

For knee flexion inter-limb differences, 42 (89.4%) participants maintained their original inter-limb difference classification in all simulations, while 5 (10.6%) changed classification in at least one simulation (range 5 – 17, median 9) (Table 3.1). 6 (12.8%) participants maintained their original classification for knee abduction angle inter-limb differences in all simulations, with 41 (87.2%) changing classification on at least one occasion (range 1 – 53, median 8). Lastly, for knee rotation angle inter-limb differences, 4 (8.5%) ACLR participants maintained their original classification and 43 (91.5%) changed classification on at least one occasion (range 1 – 52, median 7).

Ankle Kinematics

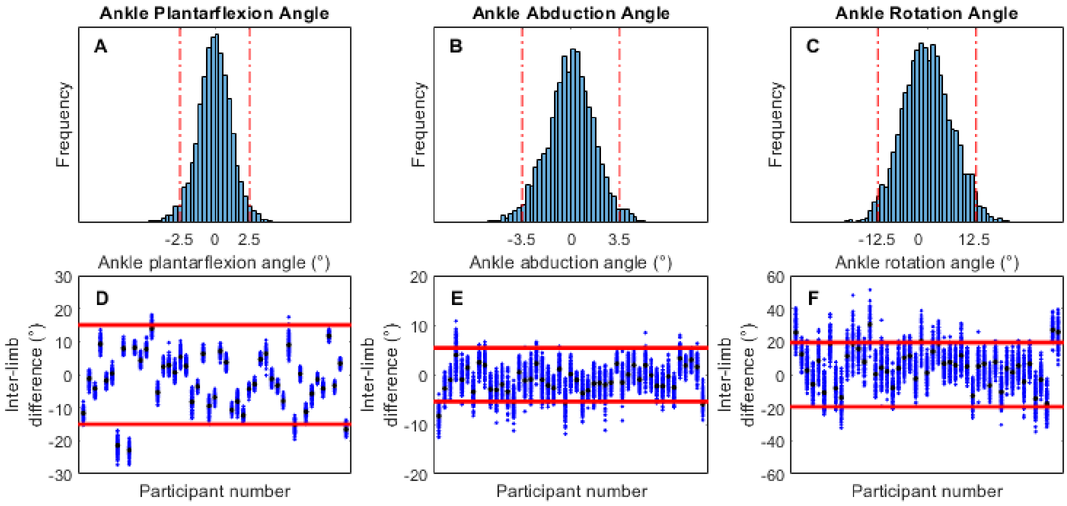


Figure 3.5: Effect of marker displacements on the interpretation of inter-limb differences in ankle kinematics at 20% of stance. Top panel depicts the distribution of changes in inter-limb differences and 95% confidence intervals for ankle plantar-flexion (A), ankle abduction (B) and ankle rotation (C) inter-limb differences. Bottom panel depicts each ACLR participants inter-limb difference (black = original inter-limb difference, blue = simulated inter-limb difference) relative to the NORM group variability (red lines) for ankle plantar-flexion (D), ankle abduction (E) and ankle rotation (F) inter-limb differences.

Marker displacements caused a change in ankle plantar-flexion inter-limb differences with a standard deviation of 1.3°, from which a 95% confidence interval of -2.5°–2.5° was estimated (Fig 3.5A). For ankle abduction angle the standard deviation of changes in inter-limb differences was 1.8° with an estimated 95% confidence interval of -3.5°–3.5° (Fig 3.5B), while in ankle rotation angle the standard deviation was 6.4° and confidence interval -12.5°–12.5° (Fig 3.5C). Minimal identifiable changes of 4.6° in ankle plantarflexion, 6.3° ankle abduction and 23° ankle rotation inter-limb differences were estimated from these distributions (Table 3.1).

Initially, 43 ACLR participants were classified as having normal inter-limb differences in ankle plantarflexion angle and 4 as having abnormal (Fig 3.5D). 40 (85.1%) participants maintained their original inter-limb difference classification throughout all one-hundred simulated testing sessions, with 7 (14.9%) changing classification in at least one simulation (Table 3.1). The number of simulated testing sessions in which participants changed classification ranged from 1 to 36 with a median of 4 simulations. For ankle abduction angle inter-limb differences, 46 ACLR participants were classified as abnormal and 1 as normal (Fig 3.5E). 14 (29.8%) maintained their original classification, while 33 (70.2%) changed classification in at least one simulated testing session (Fig 3.5E). The number of simulations in which each participant changed classification ranged from 1 to 50 with a median of 4 simulations. In ankle rotation, original classifications were 43 normal and 4 abnormal (Fig 3.5F). 12 (25.5%) of participants maintained their original ankle rotation inter-limb difference classification throughout all one hundred simulations, with 35 (74.5%) changing classification on at least one occasion (Fig 3.5F). The number of simulated testing sessions in which participants' classifications changed range from 1 to 48, with a median of 9.

Knee Moments

Marker displacements caused changes in knee flexor moment inter-limb differences with a standard deviation of 0.21 Nm/kg, from which a 95% confidence interval of -0.41–0.41 Nm/kg was estimated (Fig 3.6A). For knee abduction moment inter-limb differences, the standard deviation was 0.2 Nm/kg and estimated confidence interval -0.39–0.39 Nm/kg, while in knee rotation moment, the standard deviation was 0.01 Nm/kg and confidence

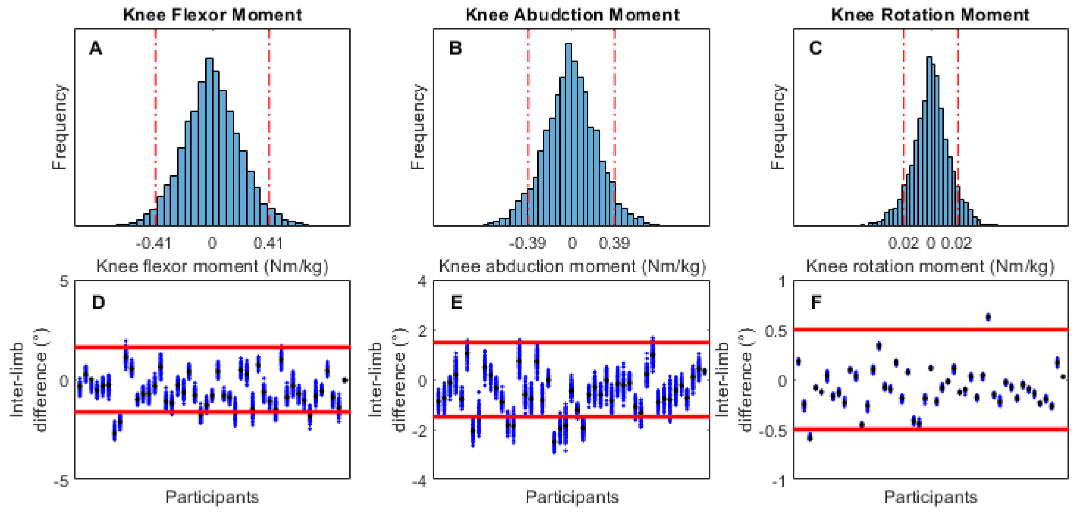


Figure 3.6: Effect of marker displacements on the interpretation of inter-limb differences in knee moments at 20% of stance. Top panel depicts the distribution of changes in inter-limb differences and 95% confidence intervals for knee flexor moment (A), knee abduction moment (B) and knee rotation moment (C) inter-limb differences. Bottom panel depicts each ACLR participants inter-limb difference (black = original inter-limb difference, blue = simulated inter-limb difference) relative to the NORM group variability (red lines) for knee flexor (D), knee abduction moment (E) and knee rotation moment (F) inter-limb differences.

interval of -0.02–0.02 Nm/kg (Table 3.1). From these distributions, the minimal identifiable changes in inter-limb differences were estimated as 0.75 Nm/kg in knee flexor moment, 0.72 Nm/kg in knee abduction moment and 0.05 Nm/kg in knee rotation moment (Table 3.1).

Relative to the NORM cohort, 43 ACLR participants were initially classified as having normal inter-limb differences in knee flexor moment, and 4 as having abnormal (Fig 3.6D). Throughout the one-hundred simulated testing sessions, 35 (74.5%) participants maintained their original classification, while 12 (25.5%) changed classification in at least one simulation. The number of simulations in which participants changed classification ranged from 1 to 51 with a median of 18. Original inter-limb difference classifications for knee abduction moment were 40 normal and 7 abnormal (Fig 3.6E). 15 (32%) participants maintained their original classification while 32 (68%) changed classification throughout the simulated testing sessions. The number of simulated testing sessions in which knee abduction moment inter-limb difference classification changed ranged from 2 to 44 with a median of 6 simulations (Table 3.1). Lastly, in knee rotation moment inter-limb differences, original ACLR classifications were 45 normal and 2 abnormal. All participants maintained their original classification throughout all one-hundred simulated testing sessions.

3.5 Discussion

Our findings highlight challenges in using marker-based biomechanical models such as the CGM to conduct objective assessments of inter-limb differences during CoD. These assessments have been proposed as a means of monitoring rehabilitation progress post-ACLR (Jordan et al., 2015; Myer et al., 2011; Oberländer et al., 2013; Paterno et al., 2007; Di Stasi et al., 2013). Marker displacements caused large changes in inter-limb differences in several variables, which in turn limited the ability to reliably identify participants with large inter-limb differences relative to a NORM cohort (Fig 3.1).

Frontal plane knee and ankle, as well as transverse plane hip, knee and ankle angles were most affected by marker placement error. Previous work examining marker placement reports similar findings (Baker et al., 1999; Groen et al., 2012; McFadden et al., 2020; Myers et al., 2015; Osis et al., 2016; Szczerbik and Kalinowska, 2011), with frontal and transverse plane angles consistently identified as most sensitive to marker placement. Change in the anterior/posterior positions of the THI, KNEE and TIB markers alters the orientation of the femur and shank segments, which manifests as large errors in frontal and transverse plane angles. The ability to identify, monitor and classify inter-limb difference measures in these variables using the CGM appears minimal.

Several of these variables are considered important in the context of CoD and ACL injury, including transverse plane hip and ankle angles as well as frontal plane knee angles and moments. For example, increased knee abduction angle (KAA) during CoD is associated with higher frontal plane knee loading, considered an important risk factor for ACL injury (Mclean et al., 2005; Sigward and Powers, 2007; Stearns and Pollard, 2013). Our data indicate that KAA inter-limb differences are highly sensitive to marker displacements, with an estimated 95% confidence interval of -10°to 10° (Fig 3.3B). A confidence interval of this magnitude presents significant challenges in any assessment of KAA inter-limb differences. Unless the observed difference is outside this range i.e. $>$ 10° or $<$ -10°, the direction, i.e. which limb has the greater value, is unclear. A hypothetical inter-limb difference of 5° may range from -5° to 15°, a range which, as well as encapsulating two alternative interpretations of inter-limb difference direction, also contains the possibility that the true difference is close to zero. This variety of possible values and subsequent interpretations means it is challenging to identify which limb, if any, requires training interventions that may be designed to restore deficits, improve frontal plane alignment and reduce frontal plane loading during CoD (Fox, 2018).

Considerable challenges also exist in the assessment of between session changes in KAA inter-limb differences. A minimal change of 18.3° is necessary to be 95% certain that the observed change is a manifestation of genuine differences in movement as opposed to variability from marker placement (Table 3.1). Depending on the specifics of the population, task and phase being analysed, KAA's of between 1 - 11° have been reported during CoD (Alenezi et al., 2016; Kristianslund et al., 2014; Sigward and Powers, 2007). In a similar 90° CoD task to the one examined in this study, Clark et al., (2019) examined inter-limb differences in ACLR participants who had completed rehabilitation and returned to sport. They reported peak KAA's of 10.2° and 8.9° for the ACLR and contralateral limbs respectively with a mean inter-limb difference of 1.3° (Clark et al., 2019). While inter-limb differences may be more pronounced at earlier stages of rehabilitation, it is unlikely that they will be of the magnitude necessary to be reliably identified and monitored due to the variability introduced by marker placement upon repeated testing.

41 (87.2%) of ACLR participants changed their KAA inter-limb difference classification throughout the simulated testing sessions (Fig 3.3E). Thresholds based on group variability have been used previously as a means of determining ACL injury risk, with individuals with extreme values

thought to be at increased risk of injury (Kristianslund et al., 2014; Robinson et al., 2014; Sigward and Powers, 2007). There appears to be a high probability of mistakenly classifying an individual as having what may be considered normal or abnormal inter-limb differences in KAA due to variability from marker placement (Fig 3.1E). For example, all 5 participants who were initially classified as having “abnormal” inter-limb differences in KAA subsequently changed classification to “normal” in the simulated testing sessions, suggesting that incorrect classifications may arise solely from marker placement error as opposed to changes in movement (Fig 3.3E).

We have chosen to focus on KAA inter-limb differences to convey the implications of this study’s findings. This is because frontal plane knee motion is considered important with respect to ACL injury and CoD (Kristianslund et al., 2014; Mclean et al., 2005; Sigward and Powers, 2007). However, the most sensitive variables to marker placement identified were transverse plane kinematics, with knee rotation angle the variable with the largest confidence intervals, minimal identifiable change, and highest percentage of participants to change classification (Table 3.1). The CGM appears limited in its ability to assess inter-limb differences in these variables as they are sensitive to relatively small variations in marker placement. Alternative techniques for modelling human movement exist and aim to overcome certain limitations ascribed to the CGM. These include models that allow for six degrees of freedom (6DOF) at each joint, those that implement the calibrated anatomical systems technique (CAST) (Cappozzo, Catani, Croce and Leardini, 1995) or utilise optimisation based joint centre positions (Charlton et al., 2004). However, in any model utilising anatomical landmarks locations to define joint centres and/or segment orientations, there is a continued assumption that marker placement is consistent and repeatable between and within practitioners. Indeed, Groen et al., (2012) observed that although implementing the optimized lower limb gait analysis model reduced errors due to marker placement during walking in certain variables, it also exacerbated those in others. Further research investigating the sensitivity of alternative modelling techniques across various movement tasks is warranted to inform their utility in assessing inter-limb differences.

Within the current study design, several limitations necessitate discussion. Marker displacements were sampled from distributions based on previously reported intra-tester variability ranges in anatomical landmark location (Della Croce et al., 2005; Myers et al., 2015). Our results are therefore directly limited to the distributions chosen a priori. It is possible that in some scenarios i.e. differing laboratories and practitioners, that these distributions underestimated the true variation in marker placement, while in others, overestimated it. Additionally, while similar trends and observations are likely in other tasks and at different time points, the specific results observed in this study i.e. confidence intervals and minimal identifiable changes, are limited to the 90° CoD task studied and the distinct time point of 20% of stance. The change in inter-limb differences across the entire stance phase for each variable is included as supplementary material (see Appendix D). Lastly, the implemented marker displacements do not directly mimic real world marker placement error. We implemented anterior/posterior displacements as model definitions indicate that these are the displacements that will have the most substantial effect on model outputs (Kadaba, Ramakrishnan, Wooten, Gainey, Gorton and Cochran, 1989). Real world marker placement will also vary proximally/distally as well as medio-laterally. We also did not account for the variation in soft tissue artefact that would be expected by varying marker positions anteriorly/posteriorly. However, given the relatively small magnitude of displacements it is unlikely these artefacts would substantially affect our findings (Nazareth et al., 2016).

In conclusion, we present an approach to quantify the minimal changes in inter-limb differences that can be reliably identified given realistic marker placement error and demonstrate how the variability resulting from these errors affects inter-limb difference classifications in a post-ACLR population. Our findings highlight challenges in using the CGM in the assessment of inter-limb differences and the critical importance of accurate and repeatable marker placement. Where possible extensive and regular training, as well as standardisation processes should be implemented to try and reduce intra-tester variability in marker placement. However, for several variables this may not be sufficient given the relatively small distributions from which displacements were sampled and the subsequent large effect observed. These include hip rotation angle, knee rotation angle, knee abduction angle, ankle abduction angle and ankle rotation angle (Table 3.1). Definitively alluding to the efficacy of the CGM's continued use in this setting is difficult as it will be dependent on the magnitude of inter-limb difference and subsequent change considered clinically relevant. What constitutes clinically relevant is often difficult to define and may vary for the same variable depending on the task, population and injury being assessed (King, Richter, Franklyn-Miller, Daniels, Wadey, Jackson, Moran and Strike, 2018; Wellsandt et al., 2016). With respect to the specific CoD task examined in this study, the CGM appears limited in the assessment of inter-limb differences for multiple kinematic and kinetic variables.

Chapter 4

The Association Between Task Level and Kinematic/Kinetic Inter-limb Differences During a Change of Direction Task

4.1 Chapter Context

Chapter 4 presents work from an experimental study examining the relationship between task level inter-limb differences and inter-limb differences in kinematic and kinetic variables during CoD. While Chapters 2 and 3 explored a key source of variability fundamental to all biomechanical analyses, this chapter explores sources of variability unique to CoD and examines their influence on joint level kinematic and kinetic inter-limb differences. Approach velocity and CoM deflection angle are fundamental task descriptors that influence the whole body demands of a CoD movement and directly influence kinematics and kinetics during CoD stance phase. When changing direction by turning off their operated limb, ACLR patients have been shown to reduce their approach velocity and CoM deflection angle. It is currently unclear how these task adjustments influence CoD kinematic and kinetic inter-limb differences. This chapter explores the relationship between inter-limb differences in task descriptors (approach velocity and CoM deflection angle) and inter-limb differences in kinematics and kinetics during CoD.

4.2 Introduction

Quantifying inter-limb differences in kinematic and kinetic variables during change of direction (CoD) is proposed as a means of monitoring rehabilitation and informing return to play decision making following anterior cruciate ligament reconstruction (ACLR) (Dingenen and Gokeler, 2017; King et al., n.d.; Meyer et al., 2018). CoD is the most common mechanism of non-contact anterior cruciate ligament (ACL) injury and a major component of post-ACLR rehabilitation is the reintroduction of these movements in the period preceding return to sport (Johnston et al., 2018; Waters, 2012). Approach velocity and change of direction angle are fundamental CoD task descriptors that reflect the whole body demands of every CoD movement, influencing both technique and knee joint loading (Dos'Santos et al., 2018; Havens and Sigward, 2015*a*; Vanrenterghem et al., 2012). In an attempt to control for the effect of approach velocity and angle, studies examining inter-limb differences in kinematics and kinetics during CoD typically instruct participants to change direction at maximal velocity and through the same pre-defined angle when turning off each limb (Bencke et al., 2013; Brown et al., 2014; King et al., n.d.; Pollard et al., 2018).

In ACLR patients, inter-limb differences in kinematics and kinetics during CoD are interpreted as reflecting altered limb-level differences during the completion of equivalent CoD tasks on both limbs. For example, King et al, (2018) identified multiple kinematic and kinetic inter-limb differences associated with reduced knee joint loading on the ACLR limb during 90° CoD tasks. Smaller knee flexion angles and knee joint moments were noted when changing direction from the ACLR limb, despite no statistically significant difference in task completion times between sides. This has been viewed as evidence that, when completing equivalent CoD tasks on both limbs, i.e. at the same velocity and through the same angle, ACLR patients reduce the magnitude of knee joint loading when changing direction from their ACLR limb via modifications to their movement patterns during CoD stance phase, likely as a compensatory mechanism in response to reduced physical capacity and/or psychological deficits that may be present following injury and rehabilitation (King, Richter, Franklyn-Miller, Daniels, Wadey, Moran and Strike, 2018; Kvist et al., 2005; Ardern et al., 2014)

Assumptions of task equivalency during CoD may be unfounded, as recent evidence demonstrates that individuals systematically modify task constraints during CoD after ACLR. Inter-limb differences in both approach velocity and center of mass (CoM) deflection angle during stance have been observed in ACLR patients during pre-planned CoD tasks (Daniels et al., 2021; King, Richter, Franklyn-Miller, Daniels, Wadey, Moran and Strike, 2018). When turning off their ACLR limb, individuals change direction with slower approach velocities and smaller CoM deflection angles compared to when turning off their non-ACLR limb despite being given identical task instructions. Slower approach velocities require smaller posteriorly-directed GRFs and impulses to decelerate, while smaller CoM deflection angles necessitate smaller horizontally-directed GRFs and impulses to redirect the CoM in the intended direction of travel. Changing direction at slower velocities and through smaller angles has been associated with smaller ground reaction forces (GRFs) (Havens and Sigward, 2015*c*), smaller knee flexion angles (Vanrenterghem et al., 2012) and smaller knee joint moments during stance phase in uninjured cohorts (Havens and Sigward, 2015*b*; Vanrenterghem et al., 2012). Task-level modifications to approach velocity and CoM deflection angle may thus be an additional method used by ACLR patients to reduce the magnitude of knee joint loading when changing direction from the ACLR limb.

It is possible that inter-limb differences in approach velocity and CoM deflection angle during CoD contribute, at least in part, to the presence of inter-limb differences in kinematics and kinetics commonly observed following ACLR. Studies examining inter-limb differences in kinematics and kinetics in ACLR patients during CoD report differences consistent with those which would be expected to arise from slower approach velocities and smaller CoM deflection angles when changing direction from the ACLR limb. Smaller GRFs, knee flexion angles and knee joint moments have been identified when turning off the ACLR limb compared to the non-ACLR limb (Daniels et al., 2021; King, Richter, Franklyn-Miller, Daniels, Wadey, Moran and Strike, 2018)). Inter-limb differences in kinematics and kinetics thus likely reflect a combination of task and limb-level modifications following ACLR, although the extent to which inter-limb differences in approach velocity and CoM deflection angle contribute to inter-limb differences in kinematics and kinetics during CoD is currently unknown. Failing to incorporate the effect of task level inter-limb differences on kinematic and kinetic inter-limb differences, may see kinematic and kinetic inter-limb differences incorrectly attributed solely to limb-level compensations, when actually reflecting a combination of task and limb-level differences.

The aim of this study was to determine the proportion of variance in inter-limb differences in kinematics and kinetics during a 90° CoD task that can be explained by task-level inter-limb differences 6-months post ACLR. We hypothesized that inter-limb differences in approach velocity would be associated with inter-limb differences in kinematics and kinetics during CoD, and that adjusting for these differences would reduce the magnitude of joint level differences by a clinically meaningful extent.

4.3 Methods

4.3.1 Participants

A cohort of 192 male participants aged 18-35 years (23.8 ± 3.6) approximately 6 months (6.3 ± 0.4) post primary ACLR were recruited consecutively from the case load of two orthopaedic surgeons based in the Sports Surgery Clinic, Dublin, Ireland. Inclusion criteria were male, aged 18-35, participation in multi-directional field-based sports prior to injury and the intention to return to the same level of participation post rehabilitation. Ethical approval was received from the University of Roehampton, London (LSC 15/122) and the Sports Surgery Clinic Hospital Ethics Committee (25AFM010). Participants gave informed, written consent prior to participation in the study.

4.3.2 Data Collection

Data collection took place in a biomechanics laboratory using a 10-camera motion analysis system (200 Hz; Bonita-B10, Vicon, UK), synchronized with two force platforms (1000 Hz BP400600, AMTI, USA), recording the positions of 28 reflective markers (14 mm diameter). Markers were secured at bony landmarks on the lower limbs, pelvis and trunk according to a modified Plug-in-Gait marker set (Marshall et al., 2014). Each participant completed a pre-planned 90° CoD task in which they ran straight towards the laboratory platform positioned 5 m from the starting point, planted their outside foot on the force platform to cut right or left (i.e. planting their right foot to turn left), turned and ran towards the finish line positioned 2 m from the centre of the force platform at 90° angle to the start line (Fig. 4.1A). Participants were instructed to complete the task as quickly as possible. Trials were considered successful if the participant made a full foot contact with the force platform when turning. Three successful trials were collected when turning off the non-ACLR limb, followed by three successful trials turning off the ACLR limb. A rest period of 30 seconds was given between trials. A fourth order zero-lab Butterworth filter (cut-off frequency 15 Hz) was used to filter marker trajectory and force data (Kristianslund et al., 2012).

4.3.3 Task Level Variables

Initial contact and toe-off were identified in each trial from when vertical GRF went above and below 20 N. Horizontal velocity was defined as CoM resultant velocity at initial contact in the horizontal plane using a moving average filter (5 frame span). CoM deflection angle during stance was calculated as the difference between the orientation of the velocity vector at initial contact and at toe-off in the horizontal plane (Fig 4.1B).

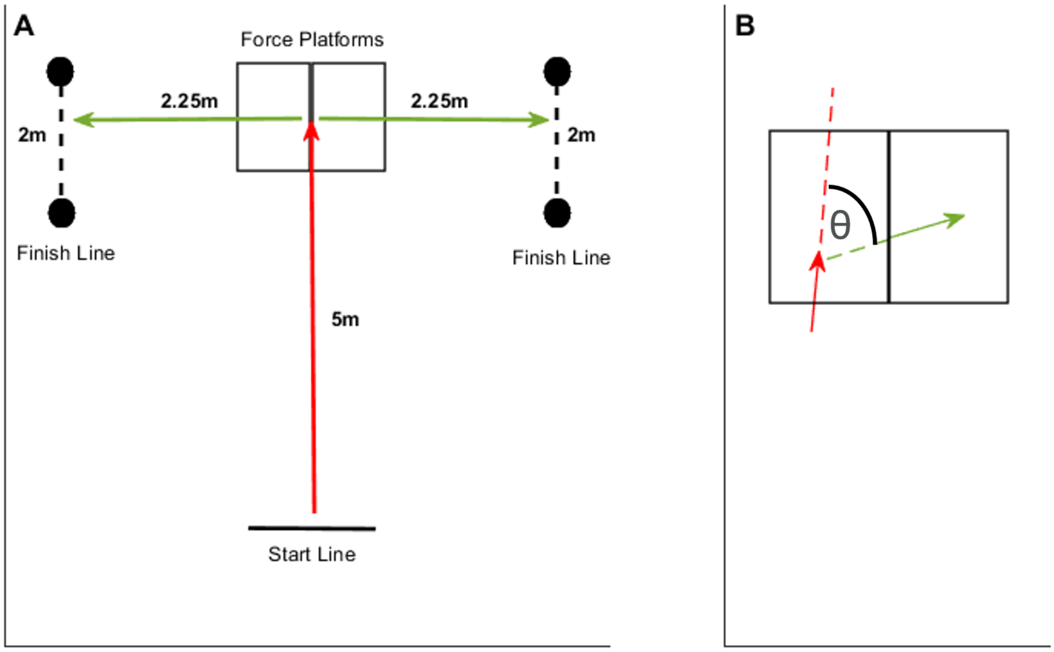


Figure 4.1: Diagram of diagram of the 90° CoD task (A). Participants ran 5m towards the laboratory force platforms before turning to either their right or left and running a further 2.25m from the centre of the force platforms to the finishing line, denoted by timing gates positioned 2m apart. Depiction of CoM deflection angle during stance phase for a change of direction planting on the left limb and turning to the right (B). This was calculated as the difference between the orientation of the velocity vector of the centre of mass at initial contact (red vector) and the orientation of the velocity vector at toe-off (green vector). Figure not to scale.

4.3.4 Kinematic and Kinetic Variables

GRF data were rotated to align with the body's local co-ordinate system before analysis (Havens and Sigward, 2015*c*). Medio-lateral and anterior-posterior impulses were calculated by integrating the rotated medio-lateral and anterior-posterior GRFs respectively. Braking impulse was determined as negative anterior-posterior impulse and propulsion impulse as positive anterior-posterior propulsion impulse.

Joint level kinematic and kinetic variables were extracted during the deceleration phase of the CoD task. Non-contact ACL injuries occur within this phase and it is widely studied in CoD and ACLR literature (Havens and Sigward, 2015*a*; Jones et al., 2016; Kristianslund et al., 2014). The deceleration phase was defined as from initial contact to the point of maximal knee flexion. Peak vertical GRF (vGRF), peak joint angles in each plane at the hip, knee and ankle, as well as peak tri-planar knee joint moments, were extracted from this phase.

4.3.5 Regression Analysis

Mean values for all variables were calculated using values from the three trials collected on each limb. Inter-limb differences were calculated for both task variables and kinematic and kinetic variables as:

$$ACLR\ Limb\ \text{-}\ NonACLR\ Limb$$

Kinematic and kinetic inter-limb differences were submitted to a simple linear regression model against approach velocity and CoM deflection angle inter-limb differences separately. Kinematic and kinetic inter-limb differences with no significant linear regression coefficients for either approach velocity or CoM deflection angle inter-limb differences were excluded from further analysis as this indicated they were not affected by velocity or CoM deflection angle inter-limb differences. For kinematic and kinetic inter-limb differences with significant regression coefficients for either approach velocity or CoM deflection angles, the corresponding linear regression model produced was used for further analysis. Lastly, kinematic and kinetic inter-limb differences with significant linear regression coefficients for both approach velocity and CoM deflection angle inter-limb differences were submitted to a multiple linear regression model with inter-limb differences in approach velocity entered first, followed by inter-limb differences in CoM deflection angle. This mirrored the mechanistic sequence of the CoD task where the approach velocity preceded CoM deflection angle.

Each individual kinematic and kinetic inter-limb difference was then adjusted by removing the variance explained by the predictor variable(s). Inter-limb differences were adjusted using:

$$ILD_{adj} = ILD_{org} - (ILD_{pv1} \cdot \beta_1) - (ILD_{pv2} \cdot \beta_2)$$

where ILD_{adj} is the individual's original inter-limb difference for the kinematic or kinetic variable, ILDpv is the individual's inter-limb difference for the predictor variable (approach velocity and/or CoM deflection angle and β is the beta-coefficient from the model between the predictor variable(s) inter-limb difference(s) and the kinematic or kinetic inter-limb difference (Fig 4.2). Adjusted and unadjusted joint level inter-limb differences were submitted to one-samples t-tests against a value of 0 and Cohens' d effect sizes were calculated for both conditions.

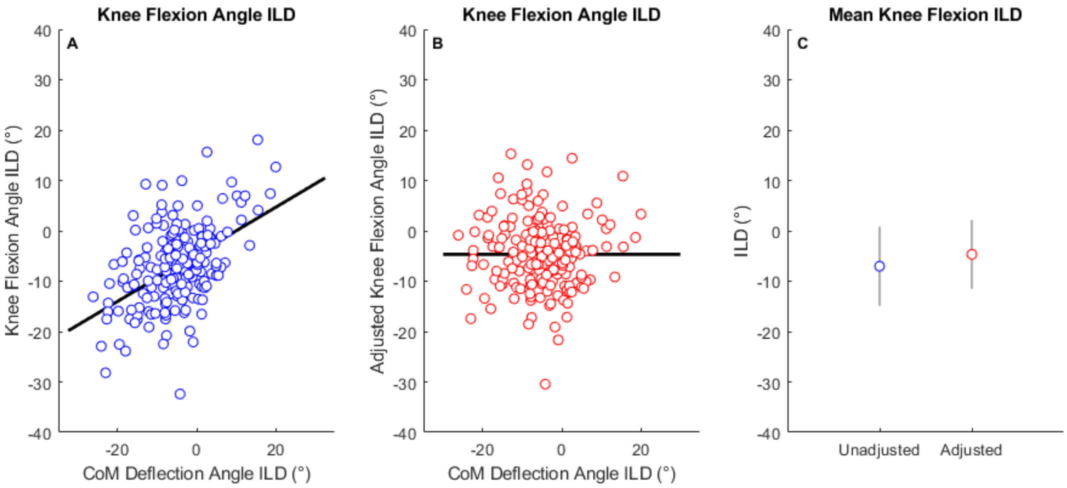


Figure 4.2: Example of adjustment process. Fig 4.2A depicts the linear regression model for CoM deflection angle inter-limb differences and knee flexion angle inter-limb differences. Fig 4.2B depicts knee flexion angle inter-limb differences after the variance attributed to CoM deflection angle inter-limb differences was removed from each data point. Lastly, Fig 4.2C depicts the mean and standard deviation of the original unadjusted knee flexion angle inter-limb differences and the adjusted knee flexion angle inter-limb differences.

4.3.6 Results

Mean approach velocities and CoM deflection angles are presented in Fig 4.3A and 4.3B respectively. Inter-limb differences in 11 variables were found to have significant regression equations with CoD approach velocity and/or CoM deflection angle inter-limb differences. These variables and the corresponding predictor variable(s) and r^2 values are presented in Fig 4.3C.

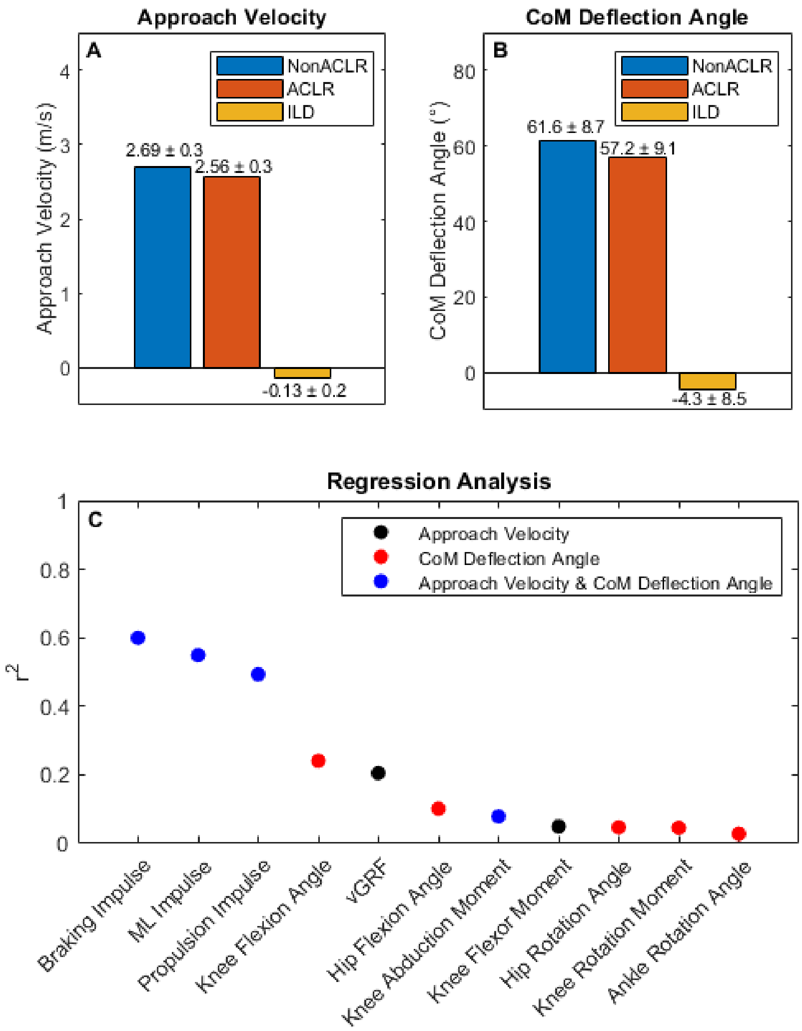


Figure 4.3: Mean approach velocities and CoM deflection angles for the NonACLR and ACLR limbs as well as the corresponding inter-limb difference (ILD) (Fig 4.3A, Fig 4.3B). Fig. 4.3C depicts r^2 values for all variables which had significant regression coefficients for approach velocity and/or CoM deflection angle inter-limb differences.

Regression models and results from one-sample t-tests for adjusted and unadjusted inter-limb differences are presented in Fig. 4.4 (GRF-derived variables), Fig. 4.5 (joint angle variables), and Fig 4.6 (joint moment variables).

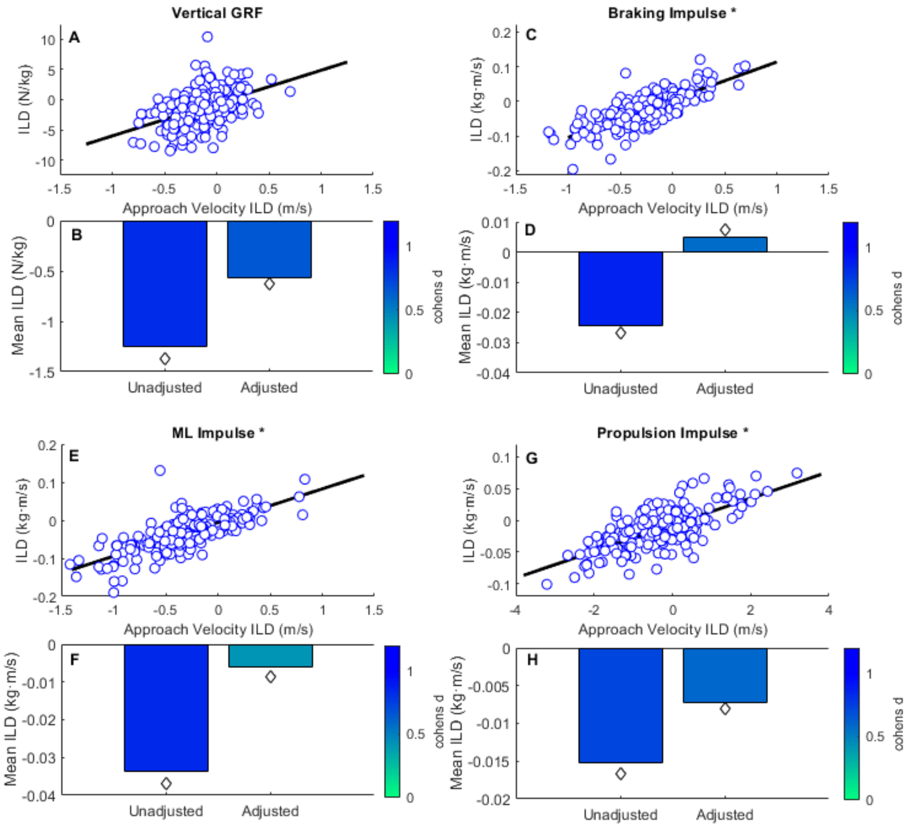


Figure 4.4: Relationship between approach velocity and/or CoM deflection angle inter-limb differences and vertical GRF (Fig 4.4A), braking impulse (Fig 4.4C), medio-lateral impulse (Fig 4.4E) and propulsion impulse (Fig, 4D) inter-limb differences. * in title indicates that a multiple regression model containing both approach velocity and CoM deflection angle was used. Figs 4.4B, 4.4D, 4.4F and 4.4H depict the mean inter-limb difference for vertical GRF, braking impulse, medio-lateral impulse and propulsion impulse respectively, in both unadjusted and adjusted conditions. ◊ indicates statistical significance ($p < 0.05$) and bar face colour corresponds to Cohens' d effect size.

Inter-limb differences in approach velocity explained 21% of the variance in vGRF inter-limb differences (Fig 4.3C), while inter-limb differences in approach velocity and CoM deflection angle explained 60%, 55% and 49% of the variance in braking, medio-lateral and propulsion impulse inter-limb differences respectively (Fig 4.3C). Unadjusted inter-limb differences in each variable were statistically significant ($p < 0.05$) with the direction of inter-limb differences demonstrating lower values when turning off the ACLR limb (Fig 4.4B, Fig 4.4D, Fig 4.4F, Fig 4.4H). Adjusting for inter-limb differences in approach velocity and/or CoM deflection angle reduced the magnitudes of inter-limb differences by 0.68 N/kg (GRF), 0.03 kg·m/s (braking impulse), 0.03 kg·m/s (medio-lateral impulse) and 0.01 kg·m/s (propulsion impulse). Though the interpretation of statistical significance remained consistent when inter-limb differences were adjusted for approach velocity and CoM deflection angle, inter-limb differences in braking impulse were found to be significant in the opposite direction, i.e. a greater value on the ACLR limb.

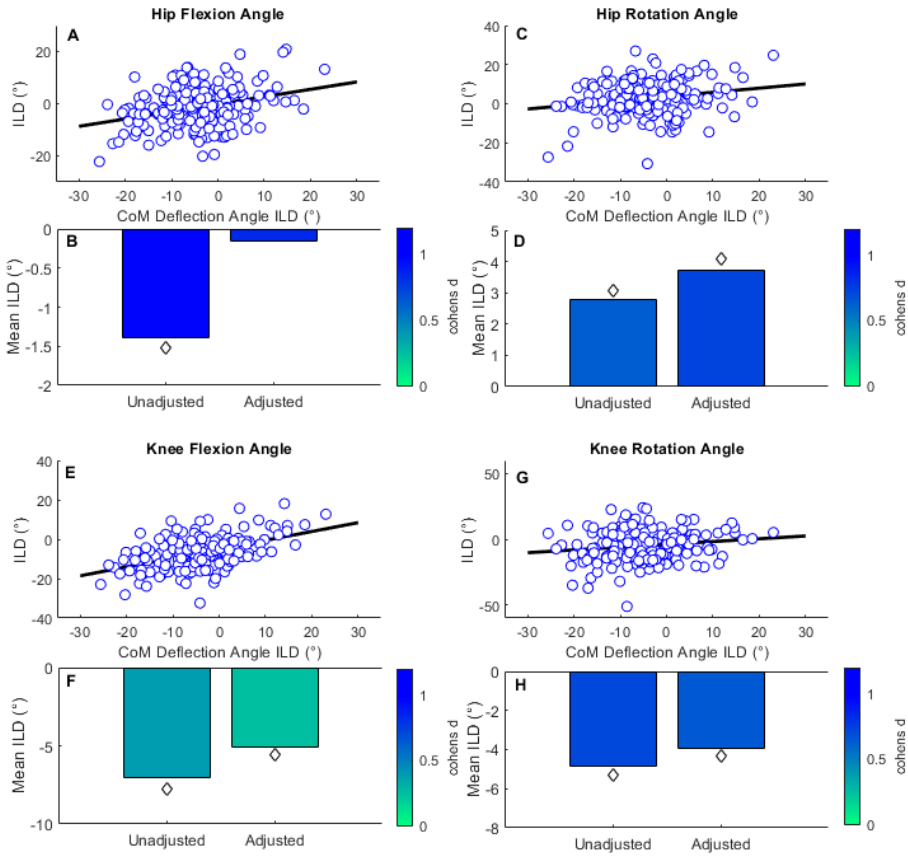


Figure 4.5: Relationship between approach velocity and/or CoM deflection angle inter-limb differences and hip flexion angle (Fig. 4.5A), hip rotation angle (Fig 4.5C), knee flexion angle (Fig 4.5E) and knee rotation angle (Fig 4.5D) inter-limb differences. * in title indicates that a multiple regression model containing both approach velocity and CoM deflection angle was used. Fig 4.5B, 4.5D, 4.5F and 4.5H depict the mean inter-limb difference for hip flexion angle, hip rotation angle, knee flexion angle and knee rotation angle respectively, in both unadjusted and adjusted conditions. ◊ indicates statistical significance ($p < 0.05$) and bar face colour corresponds to Cohens' d effect size.

Inter-limb differences in CoM deflection angle explained 10% of the variance in hip flexion angle inter-limb differences, 5% in hip rotation angle inter-limb differences, 24% in knee flexion angle inter-limb differences and 3% in knee rotation angle inter-limb differences (Fig 4.5C). Unadjusted inter-limb differences in hip flexion, hip rotation, knee flexion and knee rotation angles were all found to be statistically significant ($p < 0.05$), with the direction of the inter-limb differences indicating smaller values when turning off the ACLR limb. Adjusting for inter-limb differences in CoM deflection angle reduced the magnitudes of inter-limb differences by 1.2° (hip flexion), 2° (knee flexion) and 1° (knee rotation) but increased the magnitudes in hip rotation angle by 0.9°. When adjusted for inter-limb differences in CoM deflection angles, the interpretation of statistical significance changed from significant to non-significant for hip flexion angle inter-limb differences (Fig 5B) but remained consistent for the other kinematic variables (Fig 4.5D, Fig 4.5F and Fig 4.5H).

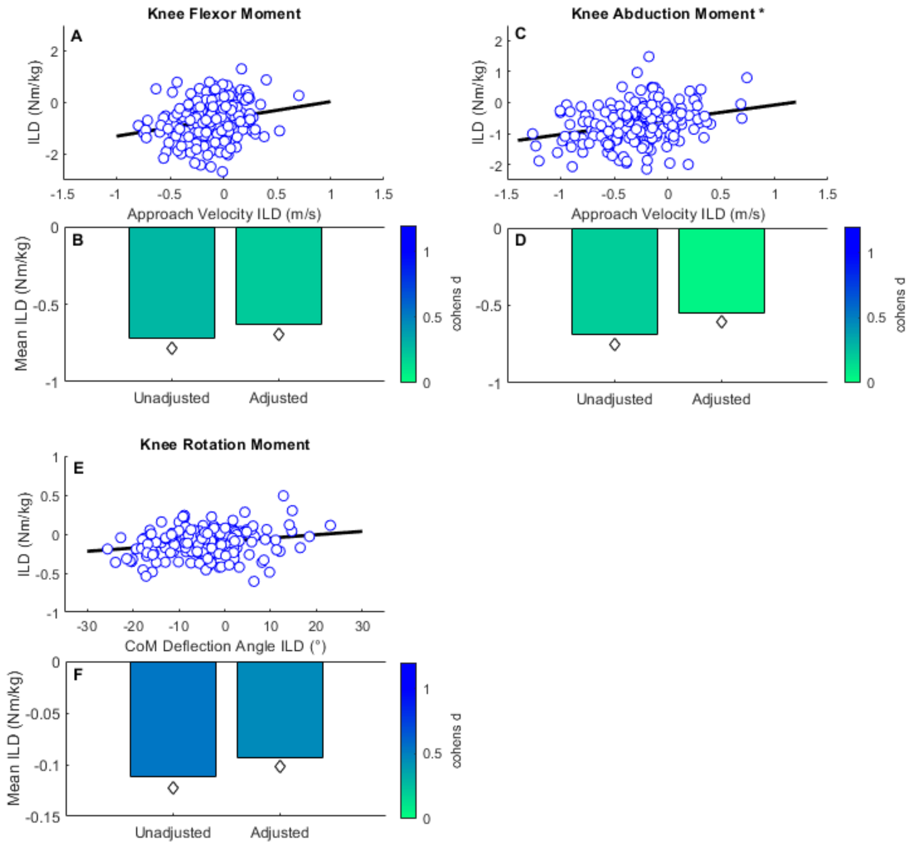


Figure 4.6: Relationship between approach velocity and/or CoM deflection angle inter-limb differences and knee flexor moment (Fig 4.6A), knee abduction moment (Fig 4.6C) and knee rotation moment (Fig 4.6E) inter-limb differences. * in title indicates that a multiple regression model containing both approach velocity and CoM deflection angle was used. Fig 4.6B, 4.6D and 4.6F depict the mean inter-limb difference for knee flexor moment, knee abduction moment and knee rotation moment, in both unadjusted and adjusted conditions. ◊ indicates statistical significance ($p < 0.05$) and bar face colour corresponds to Cohens' d effect size.

Inter-limb differences in approach velocity explained 5% of the variance in knee flexor moment inter-limb differences (Fig 4.6A), while inter-limb differences in approach velocity and CoM deflection angle explained 8% of the variance in knee abduction moment inter-limb differences (Fig 4.6C). Inter-limb differences in CoM deflection angle explained 5% of the variance in knee rotation moment (Fig 3C). Unadjusted inter-limb differences in knee flexor moment, knee abduction moment and knee rotation moment were all found to be statistically significant ($p < 0.05$). Adjusting for inter-limb differences in approach velocity and/or CoM deflection angle reduced magnitudes by 0.08 Nm/kg (knee flexor moment), 0.14 Nm/kg (knee abduction moment) and 0.02 Nm/kg (knee rotation moment). Initial interpretations of statistical significance remained consistent for all three variables when adjusted.

4.3.7 Discussion

Velocity and CoM deflection angle are fundamental task descriptors that characterise the whole-body demands of CoD movements. Our results demonstrate that task level inter-limb differences in approach velocity and CoM deflection angle explained between 3% and 60% of the variance in kinematic and kinetic inter-limb differences during a pre-planned 90° CoD task (Fig 4.3C). Incorporating the effect of inter-limb differences in task-descriptors into analyses involving kinematic and kinetic inter-limb differences during CoD will provide a better understanding of the primary drivers of inter-limb differences in specific kinematic and kinetic variables, i.e. primarily driven by task or by limb-level modifications, or a combination of both.

Inter-limb differences in velocity and CoM deflection angle were consistent with those previously published, both in terms of magnitude and direction (Daniels et al., 2021; King, Richter, Franklyn-Miller, Daniels, Wadey, Moran and Strike, 2018). When turning off their ACLR limb, participants changed direction with slower approach velocities and smaller CoM deflection angles compared to when turning off their non-ACLR limb (Fig 4.3A, Fig 4.3B). Task modifications of this manner appear to be a means of reducing the mechanical demands imposed on the ACLR limb during CoD. For example, the direction of the inter-limb difference for braking impulse switched when adjusted for inter-limb differences in approach velocity and CoM deflection angle, suggesting that the primary mechanism by which ACLR patients reduce deceleration demands during CoD is via task level modifications (Fig 4.4C). While it has been thought that ACLR patients principally control mechanical loading by performing motor tasks with altered movement patterns (Gokeler et al., 2010; King, Richter, Franklyn-Miller, Daniels, Wadey, Moran and Strike, 2018; Paterno et al., 2007), our findings, combined with previous observations of modifications to approach velocity and CoM deflection angles during CoD, demonstrate that ACLR patients also manipulate task constraints as a means of reducing the mechanical demands imposed on their operated limb.

Inter-limb differences in eleven kinematic and kinetic variables were found to have a significant relationship with inter-limb differences approach velocity and/or CoM deflection angle (Fig 4.3C). Visual inspection of r^2 values (Fig 4.3C) indicated a clear differentiation in the magnitude of r^2between braking impulse, medio-lateral impulse, propulsion impulse, vGRF and knee flexion (r^2 > 0.21) and hip flexion, hip rotation, knee rotation, knee flexor moment, knee abduction moment and knee rotation moment inter-limb differences (r^2 < 0.1). Slower approach velocities and smaller CoM deflection angles on the ACLR limb means that deceleration and redirection demands are less than those imposed on the non-ACLR limb. These alterations are associated with smaller GRFs, impulses and knee flexion angles which in turn influences the magnitude of inter-limb differences in these variables.

Adjusting limb-level inter-limb differences for task-level inter-limb differences reduced magnitudes considerably in several variables. For example, mean inter-limb differences of 0.8 N/kg and 5° in GRF and knee flexion angle respectively have been previously identified during CoD tasks in ACLR cohorts (Daniels et al., 2021; King, Richter, Franklyn-Miller, Daniels, Wadey, Moran and Strike, 2018). Such differences are interpreted as clinically meaningful and thought to be indicative of incomplete rehabilitation as they are greater than those observed in non-injured cohorts. Our data indicate that adjusting inter-limb differences in GRF and knee flexion angle for task level inter-limb differences would reduce their magnitudes to 0.12 N/kg and

3° respectively. For context, these magnitudes are comparable to those observed in normative cohorts, where inter-limb differences of 0.1 N/kg and 3° have been reported during CoD tasks (Brown et al., 2014; Greska et al., 2017). This means that without alterations to approach velocity and CoM deflection angle, ACLR participants would be expected to demonstrate inter-limb differences in these variables comparable to those in normative cohorts.

The findings of this study, in combination with those from recent work in this area, highlight the importance of studying CoD as a multi-step movement as opposed to focusing on CoD stance phase in isolation. Daniels et al., (2021) demonstrated that the reduction in CoM deflection angle observed when turning off the ACLR limb was driven primarily via reductions to the CoM heading angle at initial contact of CoD stance phase. This indicates that ACLR patients modify their technique during the steps preceding CoD stance phase as a means of reducing their total CoM deflection angle during stance phase. Similarly, the reduction in CoM velocity at initial contact can only be achieved via a reduced maximal CoM velocity or greater deceleration during the steps preceding stance phase. The current study demonstrates that these pre-stance phase modifications can influence kinematic and kinetic variables during stance phase. Further research examining alterations to the steps preceding CoD stance phase may elucidate the specific mechanisms by which ACLR patients alter their CoM deflection angle.

It is important to note that the CoD task utilised in this study was a pre-planned anticipated CoD task, where participants knew in which direction they were to turn before the onset of the movement. This is not representative of a real world situation where individuals typically change direction in response to external stimuli such as an opposing player or ball position. To better mimic this real world scenario, laboratories often use unanticipated CoD tasks, whereby participants react to an external stimuli during completion of the CoD task (King, Richter, Franklyn-Miller, Daniels, Wadey, Moran and Strike, 2018; Ford et al., 2005; Robinson et al., 2014). It is unclear if the task-level modifications observed in this study are also present during unanticipated CoD tasks. However, given the reduced time to implement adjustments to approach velocity and CoM deflection angle during unanticipated CoD tasks, it is reasonable to hypothesis that if present, the magnitude of task-level adjustments would be smaller than those during anticipated CoD tasks. If this is confirmed experimentally, inter-limb differences during unanticipated CoD would be more reflective of actual limb-level differences during CoD than those present during anticipated CoD tasks.

Joint angle and moment inter-limb differences were less sensitive to task-level inter-limb differences than GRF and impulse variables, with r^2 for these variables (except for hip and knee flexion) all below 0.1. While mechanistically it would be expected that GRF and impulse variables would be more sensitive to task-level inter-limb differences than joint angles and moments, the observed weak relationship with angle and moment inter-limb differences was unexpected. Significant differences have been observed in hip, knee and ankle kinematics, as well as knee joint moments when changing direction at different velocities and through different angles (Havens and Sigward, 2015*b*; Vanrenterghem et al., 2012) suggesting that these variables are influenced by velocity and angle. One explanation for the low r^2 values observed for joint angle and moment inter-limb differences may be the sensitivity of these variables to methodological sources of error such as marker placement. Inter-limb differences in joint angles and moments, particularly those in the frontal and transverse planes, are highly sensitive to marker placement, evidenced in our analyses by the high variability observed in these measures (Fig. 4.5, Fig. 4.6) (McFadden et al., 2020, McFadden et al., 2021). The joint level inter-limb differences which did demonstrate relatively high r^2 values -namely hip and knee flexion angle - were both sagittal plane variables, which are less sensitive to marker placement. Thus, our results may indicate that the variation in these metrics explained by task-level inter-limb differences is masked by the variance explained by marker-placement error and the inability to measure these variables accurately and reliably.

4.3.8 Conclusion

During pre-planned CoD tasks ACLR participants reduce approach velocity and CoM deflection angles as a means of reducing the mechanical demands imposed on the ACLR limb. These task modifications in turn contribute to the presence of inter-limb differences in several kinematic and kinetic variables, primarily those related to GRF and impulse. Where kinematic and kinetic inter-limb differences during CoD have previously been attributed solely to altered limb level differences arising from the ACLR procedure, our study demonstrates that a combination of task and limb level alterations are likely responsible. Task and limb level difference should be considered in tandem when examining and interpreting inter-limb differences in kinematic and kinetic variables in post-ACLR cohorts during CoD.

Chapter 5

An Exploration of Lower-Limb Dominance Definitions and their Relationship with Inter-Limb Differences in Kinematic and Kinetic Variables During a CoD Task

5.1 Chapter Context

Chapter 5 presents an experimental study examining the relationship between six definitions of lower-limb dominance and directional inter-limb differences during a CoD task. Chapters 3 and 4 examined the effect of both methodological and task-related variability on kinematic and kinetic inter-limb differences during CoD. Chapter 3 highlighted challenges in individual monitoring of kinematic and kinetic inter-limb differences, as variability introduced into these assessments from random marker placement error means it is challenging to monitor changes in inter-limb difference magnitudes over time. However, group analyses of kinematic and kinetic inter-limb differences may still be warranted, depending on the magnitude of inter-limb differences considered clinically relevant. For example, Chapter 3 demonstrated that inter-limb differences greater than 3.2° in knee flexion angle during CoD can be identified using the CGM. If inter-limb differences of this magnitude were identified in an ACLR cohort, a logical subsequent analysis in order to contextualise these observations would be to compare this to the magnitude of inter-limb differences present in a non-injured group.

Inter-group comparisons of this nature present additional challenges with respect to forming standardised comparisons between groups. Primary among these is the current ambiguity with respect to the appropriate method for classifying limbs as dominant and non-dominant when calculating directional inter-limb differences in non-injured groups. In Chapter 3, the preferred kicking limb was used to determine limb dominance and calculate normative inter-limb differences in kinematics and kinetics. The preferred kicking limb was chosen as a definition of lower-limb dominance primarily based on the commonality of its use in the sports medicine literature. However, further reading around this issue highlighted inconsistencies within the literature, with various alternative methods also used to define limb dominance e.g. the limb that can attain the highest jump height, the limb that can attain the furthest hop distance, etc.. Recent evidence indicates that the use of different methods for defining limb dominance will result in different limbs being classified as dominant and non-dominant. This means that group measures of inter-limb differences in non-injured groups are contingent on the definition of dominance chosen *a priori*. If the chosen definition classifies limbs in a manner that identifies consistent, systematic inter-limb differences between limbs, the magnitude of group inter-limb differences will approach the magnitude of absolute inter-limb differences. Alternatively, if the definition used bears no relationship to directional inter-limb differences during the task studied, mean group inter-limb differences will approach zero as positive and negative values cancel each other out. In this scenario, inter-group comparisons between injured and non-injured groups may falsely conclude that inter-limb difference magnitudes in injured groups are far in excess of those present in non-injured groups. This chapter explores this problem and examines whether any of six lower limb dominance definition identifies consistent, systematic inter-limb differences in kinematic and kinetic variables during a CoD task.

5.2 Introduction

Inter-limb asymmetry refers to differences in movement and performance between limbs in voluntary motor tasks (Bishop et al., 2018). In the absence of pathology, asymmetry is believed to be driven by differences between dominant and non-dominant limbs, though ambiguity remains with respect to the appropriate method for classifying limbs as dominant and non-dominant. Limb dominance is attributed to functional differences in the two hemispheres of the human brain and is associated with the preferential use of one limb in voluntary motor tasks (Kapreli et al., 2006; Sadeghi et al., 2000). Within many sports the ability to use both limbs effectively in tasks such as kicking, jumping and turning is desirable, meaning that large inter-limb asymmetries may negatively impact athletic performance (Bloomfield et al., 2007; Pollard et al., 2007; De Ruiter et al., 2010). Asymmetries in kinematic and kinetic measures are also associated with increased injury risk (Hewett et al., 2005; Paterno et al., 2010; Zifchock et al., 2006). Quantifying inter-limb asymmetries between dominant and non-dominant limbs is therefore a common research objective aimed at identifying systematic differences between limbs and establishing normative ranges of asymmetry (Kobayashi et al., 2010; Marshall et al., 2015; Pollard et al., 2007; Promsri et al., 2018; van der Harst et al., 2007).

Methods used to classify limbs as dominant and non-dominant include the self-preferred kicking limb (Brown et al., 2014; Marshall et al., 2015), the limb that attains the greatest single-leg countermovement jump height (Kobayashi et al., 2013), the limb that attains the furthest single-leg hop distance (van der Harst et al., 2007), the limb that contacts the ground first when landing from a vertical drop jump (Paterno et al., 2011) and the strongest limb based on isokinetic peak knee extension torque (Coratella et al., 2018). Using different methods will manifest as different limbs being classified as the dominant and non-dominant. Multiple studies have shown that individuals vary their preferred limb across different lower-limb tasks (Huurnink et al., 2014; Mulrey et al., 2018; van Melick et al., 2017), while Mulrey et al. (2018) demonstrated that the limb classified as the dominant differed within the similar hopping tasks of vertical jump height and horizontal hop distance.

The inability to consistently assign a limb classification across different tasks has led to the suggestion that limb dominance should be classified according to the demands of the task being studied (Gabbard and Hart, 1996). Dörge et al. (2002) and Ball et al. (2011) noted significant differences in lower-extremity kinematics between dominant and non-dominant limbs during kicking tasks when classifying limbs according to the self-preferred kicking limb, while Sinclair et al (2014) made similar observations when examining differences in kinematics during jumping between dominant and non-dominant limbs as classified by vertical jump height (Dörge et al., 2002; Sinclair et al., 2014). These methods therefore classify limbs as dominant and non-dominant in a manner that identifies group directional asymmetries during the task being studied, i.e. the dominant limb value is systematically larger or smaller than the non-dominant limb value. The extent to which the method used to classify dominance achieves this can be considered along a continuum ranging from a perfect relationship, where asymmetry direction is consistent across all participants and mean directional asymmetry magnitude is equal to absolute asymmetry magnitude, to no relationship, where asymmetry direction varies randomly across participants and mean directional asymmetries approach zero with positive and negative values cancelling out. Thus, unless the method used to classify dominance in a study relates in some manner to the directional asymmetries during the task studied, movement symmetry may be falsely inferred from low directional asymmetry group means. Large discrepancies between absolute and directional asymmetries would indicate that the chosen dominance definition has not captured the observed asymmetry in the execution of the task.

Large discrepancies between absolute and directional asymmetry magnitudes are apparent in studies examining inter-limb asymmetries during change of direction (CoD) (Bencke et al., 2013; Brown et al., 2009; King et al., n.d.; Marshall et al., 2015; Mok et al., 2018; Pollard et al., 2018). Analyses of inter-limb asymmetries during CoD have gained popularity due to CoDs relevance to sporting performance and its association with anterior cruciate ligament (ACL) injury (Bencke et al., 2013; Brown et al., 2009; Marshall et al., 2015; Pollard et al., 2018). Studying inter-limb asymmetry during CoD is important to quantify performance deficits, identify underlying risk factors for injury and establish normative ranges of asymmetry that can be used to guide rehabilitation programmes. King et al. (2019) compared absolute asymmetries during a CoD task between an injured (post ACL-reconstruction) and healthy control groups and noted relatively large absolute asymmetries within the control group. For example, mean asymmetries of 5.6° in knee flexion angle were observed during CoD stance phase, which is larger than the magnitude of asymmetry observed between operated and non-operated limbs post ACL-reconstruction (King et al., n.d.). In contrast, in non-injured groups, mean directional asymmetries between dominant and non-dominant limbs for knee flexion angle during CoD have been reported as ranging between 0.7° and 2.5° (Brown et al., 2014; Greska et al., 2017; Marshall et al., 2015; Pollard et al., 2018).

The preferred kicking limb is the most common method used for classifying limb dominance when studying CoD asymmetries (Brown et al., 2009; Marshall et al., 2015; Mok et al., 2018; Pollard et al., 2018). The rationale for classifying limbs in this manner when studying CoD is unclear as the demands associated with CoD, particularly in the early deceleration phase where ACL injury occurs, more closely mirror those experienced by the stance limb during a kicking motion than the kicking limb itself (Koga et al., 2010). Alternative methods based on jumping and hopping may be more appropriate when studying CoD due to an overlap in qualities such as strength, power and rapid force generation. There may also be scope for the development and implementation of a task-specific method for classifying lower-limb dominance during CoD. Mechanically, CoD involves the deceleration, reorientation and acceleration of the body's centre of mass (CoM) in the intended direction of travel. From a performance perspective, the ability to complete this process over the shortest time-period is critical. Dominance as classified by these features may provide a useful means of distinguishing between stance limbs during CoD.

Thus, this study had two aims. Firstly, we aimed to determine if five previously-used methods of classifying lower limb dominance and a new task-specific CoD method identified significant inter-limb asymmetries in whole body and joint level mechanics during a 90° CoD task, indicative of systematic directional asymmetries across participants. We hypothesized that dominance as classified by jumping/hopping ability and a task specific CoD definition would identify significant inter-limb asymmetries during CoD due to a relationship between the task studied and the method used. Secondly, we aimed to assess the consistency between the limb dominance classification specified by each definition.

5.3 Methods

5.3.1 Participants

A cohort of 50 male participants (24.8 ± 4.3 years, 182.3 ± 6.38 cm, 83 ± 7.4 kg) with no history of ACL injury or knee injury that required surgery and no lower-limb injuries in the preceding 12 weeks. All participants participated in multi-directional field-based sports (gaelic football, hurling and soccer) at an amateur level. Ethical approval was granted by the University of Roehampton (LSC 15/122) and the Sports Surgery Clinic Hospital Ethics Committee (25AFM010). Participants gave informed, written consent prior to participation in the study. Data collection took place in a biomechanics laboratory using a ten-camera motion analysis system (200 Hz; Bonita-B10, Vicon, UK) recording the positions of 28 reflective markers (14 mm diameter), synchronized (Vicon Nexus 2.3) with two force platforms (1000 Hz BP400600, AMTI, USA). Markers were secured using tape at bony landmarks on the lower limbs, pelvis and trunk according to a modified Plug-in-Gait marker set (Marshall et al., 2014).

5.3.2 Data Collection

Prior to testing participants undertook a standardised warm up consisting of a 2-minute jog, 5 bodyweight squats, 2 submaximal countermovement jumps and 3 maximal countermovement jumps. Following this each participant completed a testing battery consisting of single-leg countermovement jumps (SLCMJ), double-leg drop jumps (DLDJ), single-leg hops (SLHop) and a pre-planned 90° CoD task. Three valid, maximal effort trials were recorded for each task and for single leg exercises (SLCMJ, SLHop and CoD), participants completed three trials on each leg.

For all jumping exercises, the participants were instructed to complete the task with their hands placed on their hips. The SLCMJ consisted of a maximal vertical jump where the participants were instructed to "stand on one foot, perform a quick dip prior to jumping straight into the air as high as you can". The SLHop was a maximal horizontal jump where the participants were instructed to "stand on one leg and jump horizontally as far as possible while maintaining a balanced landing position". For the DLDJ, participants were positioned upon a 30 cm box and instructed to "drop off the box with both feet simultaneously and upon landing jumped vertically for maximal height and spend as little time as possible on the ground". The box was positioned in a manner that meant that the participant's two feet landed on separate force platforms. Lastly, the CoD task involved the participants running maximally towards the laboratory force platforms before planting their outside foot on the force platform to cut 90° to the left or right, i.e. planting their right foot to cut to the left. The start line was 5 m from the force plates, while the finish line was 2 m from the force plates. Three trials were collected from each leg. A full description of the testing protocol is¬ given in King et al. (2018).

Finally, seated concentric knee extensor and flexor peak torques were assessed at an angular velocity of 60°/s using an isokinetic dynamometer (model Cybex Norm, Computer Sports Medicine Inc, Stoughton, MA) through an angular range of 0-100° knee flexion. Participants completed an initial warm up set consisting of 4 submaximal and 1 maximal repetition followed by two maximal-effort sets each consisting of 5 repetitions. A 60 second rest period was allowed between sets. Participants were instructed to push and pull as hard and fast as possible against the resistance through the full range of motion.

5.3.3 Lower-limb Dominance Classification

Lower-limb dominance was classified for each participant using six methods. These were (1) the self-preferred kicking limb defined by participants response to the question “which limb would you preferentially use to kick a ball with?” (KICK), (2) the limb that attained the greatest vertical jump height calculated using flight time (JUMP), (3) the limb that attained the greatest horizontal hop distance (HOP), (4) the limb that made contact with the force plate first during the initial landing of the DLDJ based on a threshold of 10 N (LAND), (5) the limb that recorded the highest peak knee extension torque during isokinetic dynamometry testing (ISO), and (6) a newly formed task specific method for classifying dominance during CoD

(TURN). For JUMP and HOP, the mean of three trials for vertical jump height and horizontal hop distance were used to classify dominance, while for ISO, peak torque was extracted from both working sets and mean peak torque was calculated. Gravity corrections were applied to all torque values. For LAND, the limb that most frequently made initial contact in the three recorded trials was used.

For TURN, marker and force data from each CoD trial were filtered using a fourth order zero-lag Butterworth filter (cut-off frequency 15 Hz) (Kristianslund et al., 2012). Initial contact and toe-off were identified from when the vertical ground reaction force (GRF) crossed a 20 N threshold. The speed and angle at which an individual changes direction are the two fundamental components of every CoD manoeuvre (Dos'Santos et al., 2018; Havens and Sigward, 2015*c*; Vanrenterghem et al., 2012) thus our TURN method aimed to combine these measures. The change in angle during CoD (CoD Angle) was calculated as the difference between the orientation of the velocity vector of the CoM in the horizontal (x-y) plane at initial contact and toe-off. Ground contact time (GCT) was also extracted from each trial and the rate of change in CoM angle was calculated as:

$$\frac{\Delta CoDAngle}{GCT}$$

The mean of the three trials to each side were calculated and the limb side which attained the largest value was classed as the dominant.

5.3.4 Inter-limb Difference Calculations

Inter-limb differences were calculated for both whole body and joint level mechanical variables. In order to calculate whole body mechanical variables, ground reaction force (GRF) data were rotated to align with the body's local co-ordinate system using a rotation matrix (Havens and Sigward, 2015*a*)(Havens and Sigward 2015). The CoM was used as the origin of the body's local co-ordinate system. Medio-lateral and anterior-posterior impulses were calculated as the integration of the newly rotated medio-lateral and anterior-posterior GRF data. Braking impulse was determined as negative anterior-posterior impulse and propulsion as positive anterior-posterior impulse. Peak vertical ground reaction force during stance phase was also extracted. Lower extremity kinematics at the hip, knee and ankle, as well as knee joint moments were extracted during stance phase for each trial and time normalised to 101 data points.

Inter-limb asymmetries in whole body mechanical variables and lower-extremity kinematics and kinetics were calculated six times, on each occasion using dominance as classified by one of the six methods (KICK, JUMP, HOP, LAND, ISO and TURN). Directional asymmetries were calculated as:

$$NonDominant - Dominant$$

Absolute asymmetries were also calculated as:

$$\sqrt{(Left - Right)^2}$$

For joint level kinematic and kinetics, inter-limb asymmetries were calculated at 20% of stance as this phase is commonly reported in ACL and CoD literature (Dempsey et al., 2007; Stearns and Pollard, 2013) (Fig 5.1). One-sampled t-tests were performed on directional inter-limb asymmetries calculated using each method against a value of 0. The relationship between dominance as classified by each method was assessed using Chi-square tests for independence.

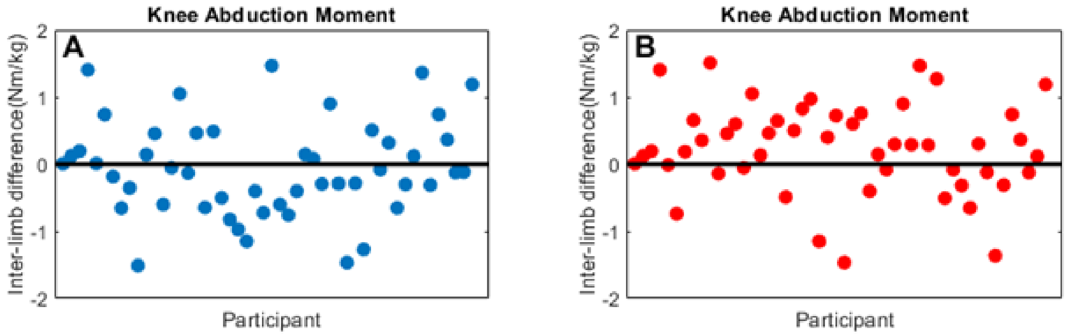


Figure 5.1: Example of inter-limb asymmetries in knee abduction moment at 20% stance in the COD task for each participant using two different definitions of lower limb dominance Fig 5.1A depicts inter-limb differences calculated using the KICK method to assign the dominant limb and 54% of participants had a greater value on their dominant limb with 46% on their non-dominant. Fig 5.1B depicts inter-limb asymmetries calculated using the JUMP method where 32% of participants had a greater value on their dominant limb and 68% on their non-dominant.

5.4 Results

The percentage of participants who were classified as right and left leg dominant under each method is presented in Table 5.1. No statistically significant inter-limb asymmetries were identified in whole body mechanics using the KICK, HOP, ISO and TURN methods. The LAND method identified significant differences in peak vertical GRF ($p = 0.03$, $d = 0.3$) (Fig 5.2), which corresponded to 39.3% of the magnitude of the corresponding mean absolute asymmetries. The JUMP method identified significant inter-limb asymmetries in medio-lateral impulse ($p = 0.03$, $d = 0.31$) (Fig 5.1, 5.2), hip flexion angle ($p = 0.04$, $d = 0.3$) and knee abduction moment ($p = 0.01$, $d = 0.29$) (Fig 5.3). These asymmetries corresponded to 38.7%, 35.5% and 42.6% of the respective mean absolute symmetries. Lastly, using the HOP method, significant inter-limb asymmetries were identified in knee flexor moment ($p = 0.04$, $d = 0.25$), which corresponded to 35.4% the corresponding mean absolute asymmetry.

Table 5.1: Percentage of participants classified as right and left leg dominant using each method.

	KICK	**JUMP**	**HOP**	**LAND**	**ISO**	**TURN**
Right	68%	38%	58%	52%	62%	62%
Left	32%	62%	42%	48%	38%	38%

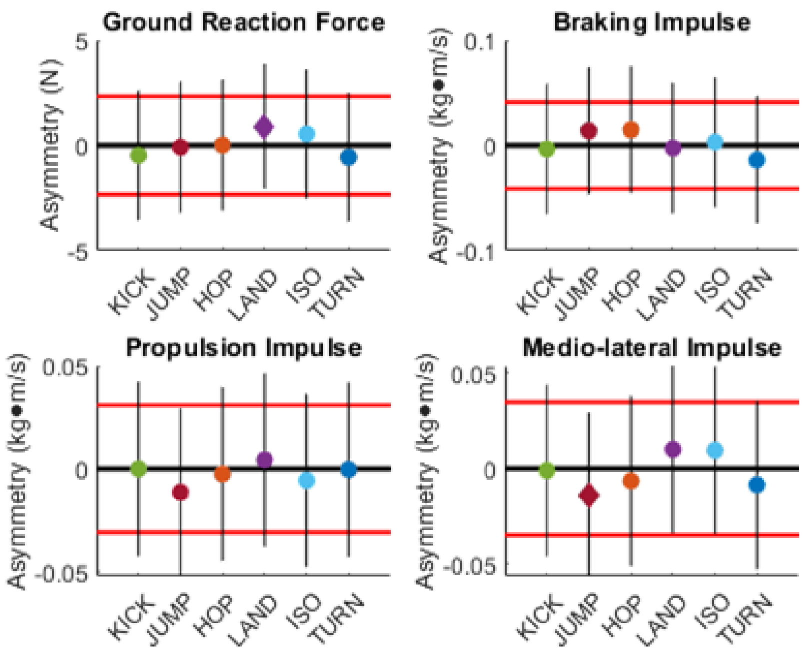


Figure 5.2: Mean difference between dominant and non-dominant limbs for whole body mechanics using each method of classifying limb dominance. Negative values indicate a greater value on the dominant limb, positive valued a greater value on the non-dominant limb and ◊ the identification of a significant inter-limb difference. Horizontal red lines represent absolute asymmetry magnitude for each variable. Negative absolute asymmetries are for illustrative purposes only as all absolute measures were positive by definition.

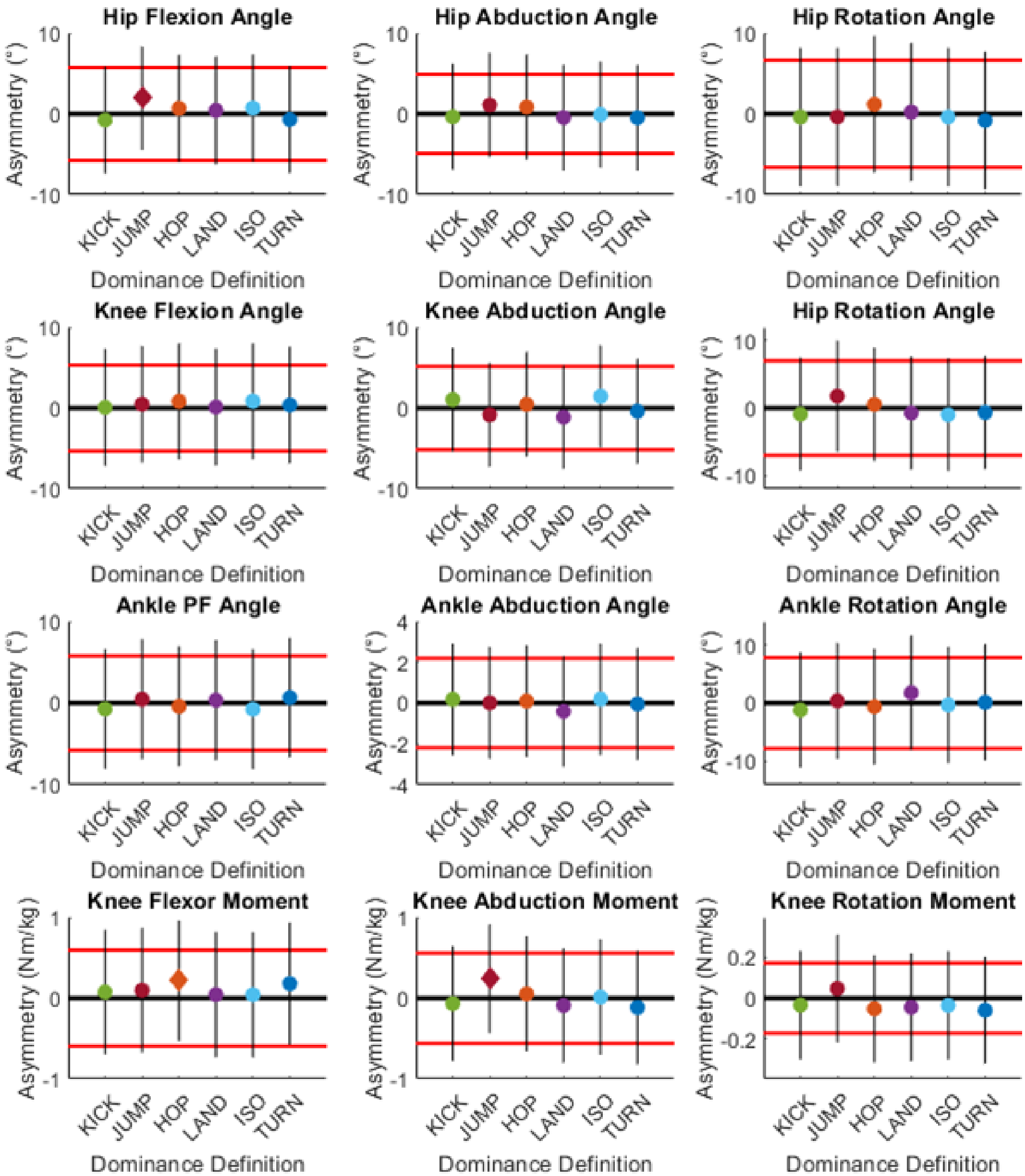


Figure 5.3: Mean difference between dominant and non-dominant limbs for lower extremity kinematics and knee joint moments using each method of classifying limb dominance. Negative values indicate a greater value on the dominant limb, positive valued a greater value on the non-dominant limb and ◊ the identification of a significant inter-limb difference. Horizontal red lines represent absolute asymmetry magnitude for each variable. Negative absolute asymmetries are for illustrative purposes only as all absolute measures were positive by definition.

Chi-square tests for independence did not identify relationships between any of the methods used to classify limb dominance (Table 5.2).

5.5 Discussion

None of the six methods used to classify limb dominance in this study provided a useful means of distinguishing between limbs and quantifying directional asymmetries during CoD. No variable indicated a significant inter-limb asymmetry using the KICK, ISO or TURN methods. While some significant inter-limb asymmetries were identified in variables using the JUMP, HOP and LAND methods, these asymmetries were small relative to the respective absolute asymmetries, consisting of magnitudes between 35.4 – 42.6% of the corresponding absolute values. These findings indicate that "dominance" as classified using each method was not a major factor in the presence of asymmetry within this cohort. The results therefore failed to support our initial hypothesis that methods based on jumping/hopping and the CoD task-specific definition would be related to systematic inter-limb asymmetries.

We did not identify consistent dominant limb classification between any methods (Table 5.2). Despite some methods sharing common physical qualities e.g. jumping and hopping, the limb classified as the dominant varied across participants and methods. This is in agreement with findings by Huurnink et al., (2014) and Mulrey et al., (2018) who noted that limb

Table 5.2: Chi-Square tests for independence for each method for classifying limb dominance.

		JUMP	
		Left	Right
KICK	Left	8	8
	Right	23	11

$X^2 = 0.79$, $p = 0.38$

		HOP	
		Left	Right
KICK	Left	5	11
	Right	16	18

$X^2 = 0.56$, $p = 0.45$

		LAND	
		Left	Right
KICK	Left	4	12
	Right	20	14

$X^2 = 3.72$, $p = 0.05$

		ISO	
		Left	Right
KICK	Left	8	8
	Right	11	23

$X^2 = 0.79$, $p = 0.38$

		ISO	
		Left	Right
KICK	Left	7	9
	Right	12	22

$X^2 = 0.07$, $p = 0.79$

		HOP	
		Left	Right
JUMP	Left	15	16
	Right	6	13

$X^2 = 0.76$, $p = 0.38$

		ISO	
		Left	Right
JUMP	Left	13	18
	Right	6	13

$X^2 = 0.19$, $p = 0.67$

		ISO	
		Left	Right
JUMP	Left	13	18
	Right	6	13

$X^2 = 0.19$, $p = 0.67$

		TURN	
		Left	Right
JUMP	Left	14	17
	Right	5	14

$X^2 = 1.07$, $p = 0.3$

		LAND	
		Left	Right
HOP	Left	12	9
	Right	12	17

$X^2 = 0.66$, $p = 0.42$

		ISO	
		Left	Right
HOP	Left	6	15
	Right	13	16

$X^2 = 0.76$, $p = 0.38$

		TURN	
		Left	Right
HOP	Left	8	13
	Right	11	18

$X^2 = 0.08$, $p = 0.78$

		ISO	
		Left	Right
LAND	Left	7	17
	Right	12	14

$X^2 = 0.89$, $p = 0.34$

		TURN	
		Left	Right
LAND	Left	11	13
	Right	8	18

$X^2 = 0.65$, $p = 0.42$

		TURN	
		Left	Right
ISO	Left	8	11
	Right	11	20

$X^2 = 0.03$, $p = 0.87$

preference and the limb that performed best did not correspond across different tasks. We expand on these findings to demonstrate that a task-specific method of classifying dominance during CoD does so in a manner that is independent of other, more commonly used methods. However, unlike previous research implementing task specific methods, we failed to identify any significant inter-limb asymmetries in kinematics and kinetics. Where previous studies have classified dominance according to the outcome of the task being studied, ball kicking and preferred kicking limb (Dörge et al., 2002), vertical jumping and single-leg countermovement jump height (Kobayashi et al., 2013), hopping and horizontal hop distance (van der Harst et al., 2007), it is not possible to form such a direct classification method for CoD. Classifying limbs solely on task outcomes such as completion time and/or ground contact time, fails to account for any side-to-side differences in the angle over which the CoM passes. Both approach velocity and angle influence CoD biomechanics and it has been shown that at higher approach speeds, individuals deviate more from the intended CoD angle (Dos'Santos et al., 2018; Vanrenterghem et al., 2012). While we attempted to account for both these features in our task specific method, it is possible that they interact differently across participants and that the effect on biomechanics is non-linear, with a larger effect occurring at higher velocities and more acute CoD angles.

Our findings suggest that observations of apparent symmetry between dominant and non-dominant limbs during CoD are likely statistical artefacts as opposed to a true reflection of normative movement. In six previous studies examining inter-limb asymmetry during CoD, three failed to identify any significant inter-limb asymmetries between dominant and non-dominant limbs in kinematics or kinetics (Bencke et al., 2013; Brown et al., 2014; Greska et al., 2017), while three failed to identity significant differences in the vast majority of variables studied (87.9 – 95% of variables) (Marshall et al., 2015; Mok et al., 2018; Pollard et al., 2018). The self-preferred kicking limb was used to classify limb dominance in five of these studies (Bencke et al., 2013; Brown et al., 2014; Marshall et al., 2015; Mok et al., 2018; Pollard et al., 2018). We have shown that the preferred kicking limb is not related to directional asymmetries during CoD and that its use as a means of distinguishing between limbs in this setting is akin to assigning a randomly-selected limb as the dominant. For example, we identified mean absolute asymmetries of 5.3° ± 4.8° in knee flexion angle. This is comparable in magnitude to normative absolute asymmetries reported by King et al. (2019) during a similar CoD task and larger than the magnitude of asymmetry considered clinically relevant between the operated and non-operated limb following ACL-reconstruction, indicating that there are relatively large inter-limb asymmetries present in knee flexion angle in non-injured individuals during CoD. However, using the KICK method to classify limbs, we identified mean inter-limb asymmetries of 0.08° in knee flexion angle, corresponding to just 1.5% of the absolute asymmetry magnitude and suggesting that there was near perfect symmetry between limbs in the cohort. The inconsistency between absolute and directional asymmetries demonstrates that, although individuals completed the CoD with relatively large absolute asymmetries in, for instance, knee flexion angle, the direction of these asymmetries was not captured by the KICK, or indeed any, dominance definition. This was true for the vast majority of variables analysed in this study (Fig 5.2 and 5.3).

Directional symmetries in normative cohorts are regularly compared to those in injured cohorts across various movement tasks (Gardinier et al., 2014; Gokeler et al., 2010; Kuenze et al., 2015; Paterno et al., 2007; Xergia et al., n.d.). These comparisons have been used to contextualise the magnitude of asymmetry in injured cohorts and set rehabilitation targets with respect to their restoration. We have shown that if the method used to define limb dominance does not relate to directional asymmetries in the task studied, conducting such comparisons runs the risk of falsely assuming symmetry within normative cohorts, overinterpreting the magnitude of asymmetry in injured cohorts and setting unattainable targets for injured individuals with respect to restoring asymmetry to normative levels during rehabilitation. King et al. (2019) and O'Malley et al., (2018) raised this issue previously, choosing instead to compare absolute asymmetries due to the inability to make standardised comparisons between groups. The findings of this study further highlight the challenges in making such comparisons and demonstrate the importance giving proper consideration to the method used to classify limb dominance when quantifying directional asymmetries in normative cohorts.

5.5.1 Conclusion

In conclusion, quantification of directional asymmetries in normative cohorts during CoD and, in particular, comparison to injured cohorts should be done with caution until an appropriate method for classifying limb dominance in non-injured individuals is established. These findings relate to CoD and further research is required to determine if it is also true for other movement tasks such as jumping and landing. Until a suitable classification for limb dominance can be determined, we recommend reporting absolute asymmetries as an alternative to directional asymmetries. If directional asymmetries are reported, they should be done in conjunction with the corresponding absolute asymmetries, allowing readers to interpret directional asymmetries with an understanding of the level of asymmetry within the group and assess the probability that they accurately reflect normative movement.

Chapter 6

General Discussion

The aim of this thesis was to determine the feasibility of using biomechanical analyses to identify abnormalities in CoD technique following ACLR. Assessing inter-limb differences in kinematics and kinetics during CoD has been proposed as a means of monitoring rehabilitation progress and informing RTS decision making following ACLR (King, Richter, Franklyn-Miller, Daniels, Wadey, Moran and Strike, 2018). Prior to conducting such assessments, it is necessary to fully explore sources of variability and their influence on clinical outcome measures so that clinicians can have confidence in the results of these assessments and make informed decisions. The thesis aim was explored across four experimental studies, presented in Chapters 2, 3, 4 and 5. The findings of these studies highlight challenges in conducting assessments of CoD kinematics and kinetics, and in particular, in the repeat assessments of inter-limb differences over extended periods. Making definitive statements on the use of biomechanical analyses in the assessment of CoD technique following ACLR is challenging however, as ultimately it will be contingent on the specific analysis being conducted, the kinematic and/or kinetic measures quantified and the magnitude of effect considered clinically relevant. Thus, this thesis is best viewed as a framework by which researchers can assess the practicality of studying CoD under their own experimental conditions, as well as a resource for developing study designs. The manner in which this thesis can be used is dependent on the biomechanical model that is being used by researchers. If using the CGM, the findings of this thesis can act as a directly transferable reference guide with respect to methodological sources of variability and their impact when quantifying inter-limb differences in kinematics and kinetics during CoD. Alternatively, if using a different biomechanical model, this thesis can serve as a protocol to follow in order to examine the utility of any given biomechanical model for studying CoD.

Chapters 2 and 3 explored marker placement error and its influence on lower extremity kinematics and kinetics during CoD. The findings from these chapters present the sensitivity of CoD kinematics and kinetics to marker placement across CoD stance phase, as well as the minimum magnitude of inter-limb differences that can be identified in kinematic and kinetic variables given the expected variability introduced into assessments from marker placement error. These findings can be used to determine the feasibility of conducting an analysis quantifying inter-limb differences in any of the reported kinematic or kinetic variables. For example, knee abduction moment (KAM) is the most widely reported variable in biomechanics/ACL related research (Hewett et al., 2005; Krosshaug et al., 2016; King, Richter, Franklyn-Miller, Daniels, Wadey, Moran and Strike, 2018; Stearns and Pollard, 2013; Sharir et al., 2016; Robinson et al., 2014; Thomas et al., n.d.; McBurnie et al., 2019; Butler et al., 2009; Sigward and Powers, 2007; Sigward et al., 2015*b*). Many researchers/laboratories would likely be interested in quantifying inter-limb differences in KAM during CoD following ACLR and comparing their magnitudes to those present in non-injured groups. The confidence intervals reported in Chapter 3 demonstrate that if using the CGM, the smallest inter-limb differences that can be meaningfully identified in KAM when allowing for variability introduced from marker placement is $\pm$ 0.39 Nm/kg. The use of the CGM remains warranted if identifying inter-limb differences of this magnitude is deemed acceptable for answering a given research question. However, if interested in identifying smaller group effects in KAM inter-limb differences, the CGM should not be used unless intra-tester variability in marker placement is demonstrably lower than the ranges used in Chapter 2.

The process outlined above with respect to the hypothetical analysis of KAM inter-limb differences can be repeated for any of the other kinematic and kinetic variables reported in this thesis, allowing researchers to determine the utility of using the CGM to answer a given research question. Chapter 3 demonstrates that if using the CGM, sagittal plane kinematics are the variables which are least affected by marker placement error and those in which the smallest inter-limb differences can be identified. At late stage rehabilitation and following RTS, when relatively small inter-limb differences may be of interest, the most appropriate means of using the CGM is thus likely in quantifying inter-limb differences in sagittal plane kinematics. Inter-limb differences in knee flexion angles during CoD stance phase are noted following ACLR, with less knee flexion observed when turning off the ACLR limb compared to the non-ACLR limb (King, Richter, Franklyn-Miller, Daniels, Wadey, Moran and Strike, 2018). Knee flexion angle influences ACL loading, with higher ACL loads observed when the knee is in a more extended position (Markolf et al., 1995). Quantifying inter-limb differences in knee flexion angle using the CGM may thus provide information relevant to rehabilitation progress.

Chapters 4 and 5 follow on from the work presented in Chapters 2 and 3, providing practical recommendations to ensure that inter-limb and inter-group comparisons are not confounded by methodological issues if experimenters choose to proceed with these analyses. Chapter 4 demonstrates how variability in the completion of a CoD task influences kinematic and kinetic inter-limb differences during stance phase. Interestingly, there was a clear trend that the variables identified as most sensitive to marker placement in Chapters 2 and 3, were also those found the be least affected by approach velocity and CoM deflection angle inter-limb differences in Chapter 4. It appears that the variability introduced from marker placement error makes it difficult to model the underlying relationships between task level inter-limb differences and inter-limb differences in many kinematic and kinetic variables using the CGM. For example, previous research using alternative modelling techniques have found that higher approach velocities and larger CoM deflection angles are associated with higher KAM during CoD stance phase (Robinson and Vanrenterghem, 2012; Kristianslund et al., 2012). In contrast, in Chapter 4, a relatively weak relationship was observed, with only 8% of the total variance in KAM inter-limb differences explained by inter-limb differences in approach velocity and CoM deflection angle. This highlights further challenges in using the CGM to quantify inter-limb differences in KAM and/or any of the kinematic or kinetic variables identified as highly sensitive to marker placement in Chapters 2 and 3. Due to the inability of the CGM to accurately measure these variables, experimenters will be unable to conclude to what extent their findings are influenced by inter-limb differences in approach velocity and CoM deflection angle that are present following ACLR (Daniels et al., 2021).

For the quantification of inter-limb differences in sagittal plane kinematics and/or GRF-related variables, experimenters should consider controlling for task-level adjustments to approach velocity and CoM deflection angle. There are two mechanisms by which approach velocity and CoM deflection angle can be controlled: experimentally and statistically. Experimentally, researchers can choose to only accept trials that do not deviate substantially from each other in approach velocity and CoM deflection angle. This has been done previously, with ranges of ± 0.2ms and ± 5° in CoD angle used (Dempsey et al., 2007). However, given the relatively small mean inter-limb differences in approach velocity (-0.13 ms) and CoM deflection angle (-4.4°) identified in Chapter 4, collecting a sufficient number of trials that fall within these ranges would likely prove logistically extremely difficult to complete, and also introduce ethical concerns related to participants performing repeated trials of a highly demanding movement. Alternatively, inter-limb differences can be adjusted statistically, as per the methodology presented in Chapter 4. This is an easier process to complete and ensures athlete's natural technique is not altered by experimental constraints. However, as outlined above the effectiveness of this approach will be contingent on the ability of the biomechanical model used to accurately quantify the kinematic and kinetic variables being studied.

While Chapters 2, 3 and 4 explored issues related specifically to the quantification of inter-limb differences in ACLR cohorts, Chapter 5 examined issues related to inter-group comparisons of these metrics between ACLR and non-injured groups. This chapter showed that currently there is no appropriate method available to calculate directional normative inter-limb differences during CoD. The inability of any of the six dominance definitions explored in Chapter 5 to classify limbs in a manner that related to the direction of inter-limb differences in kinematic and kinetic variables, means that group normative inter-limb difference measures will always approximate zero if using any of these definitions of lower-limb dominance. This increases the likelihood of falsely concluding that there are significant inter-group differences if comparing inter-limb differences between ACLR and non-injured groups. While alternative methods for calculating inter-limb differences exist, they suffer from similar limitations with respect to reference limb selection (Bishop et al., 2018). Until an appropriate means of classifying limbs as dominant and non-dominant during CoD is identified, absolute measures of inter-limb differences should be used in inter-group comparisons of injured and non-injured groups. Though these measures do not provide information about the direction of the inter-limb differences in each group, they will provide a more genuine reflection of the actual differences in kinematic and kinetic inter-limb differences that are present between groups.

The methodological issues outlined in this thesis may contribute in part to contradictory findings within the ACL literature as well as the ongoing inability to predict injury risk on an individual basis. Hewett al., (2005) and Krosshaug et al., (2016) are two of the most widely cited studies in research attempting to identify biomechanical risk factors for ACL injury. Hewett's 2005 study associating peak KAM during landing with ACL injury risk in adolescent females has served as the justification and rationale for a vast body of research over the previous 20 years (e.g. Dos'Santos et al., 2017; Sharafoddin-Shirazi et al., 2020; Stearns and Pollard, 2013; Papalia et al., 2015; Bencke et al., 2013; Brown et al., 2014). These studies have assumed that KAM as a risk factor for ACL injury is applicable to alternative populations and movement tasks, and do not consider how methodological differences between experimental setups and movement tasks may affect the calculation of KAM. In failing to replicate the findings of Hewett et al., (2005), Krosshaug et al., (2016) challenged the foundations on which much ACL/biomechanics research has been based. As this thesis focused on CoD, and primarily on inter-limb differences in kinematics and kinetics as opposed to variable magnitudes, the findings cannot be directly applied to the vertical drop jump and peak KAM reported by both Hewett and Krosshaug. However, what this thesis does highlight is that relatively small methodological differences between these two studies could potentially have a large effect on frontal plane kinematics and kinetics. As both studies were large scale (n > 200), intra-tester variability in marker placement likely influenced frontal plane kinematics and kinetics, while it is also possible that systematic differences in marker placement were present between studies. Subtle differences also exist between data collection protocols within both studies. Hewett et al., used a 31cm box,, standardized foot width at 35cm and did not control hand positions during completion of the task. In contrast, Krosshaug et al., used a 30cm box, did not standardise foot position and did not provide information about hand positions throughout the task. Inter-study differences in marker placement and task constraints may be a factor in the conflicting findings of these two studies.

6.1 Future Directions

This thesis can be used to inform best practice with respect to performing inter-limb and inter-group comparisons of CoD kinematics and kinetics following ACLR. However, it is important to note that these recommendations relate to group-based analyses, and that individual assessments and monitoring present further difficulties. An overarching aim of much ACL/biomechanics research has been to identify modifiable biomechanical risk factors for ACL injury through group analyses, and subsequently use these risk factors to screen for ACL injury risk and/or monitor ACLR rehabilitation progress on an individual basis (Hewett et al., 2005; Paterno et al., 2010; Krosshaug et al., 2016; King et al., 2021*b*). Chapter 3 demonstrates that the CGM has limited capacity to be used in such a manner. There may be some scope for monitoring large, gross changes in sagittal plane kinematics, but outside of this the variability introduced into these assessments from marker placement error means that the ability to identify and monitor changes on an individual level is minimal. Thus, while Sharir et al., (2016) were correct in their conclusion of the need for more large scale, high quality, prospective research examining *in vivo* biomechanical risk factors for ACL injury, this thesis demonstrates that prior to this, there is pressing need for research examining methodological considerations in data collection and analysis to ensure that these prospective studies are not confounded by variability from methodological sources. Without this research, the inability to translate the findings of prospective research into techniques for appraising injury risk and monitoring rehabilitation progress on an individual basis will continue.

Given the sensitivity of the CGM to marker placement error identified in Chapters 2 and 3, the use of alternative biomechanical models and modelling techniques in the analysis of CoD tasks should be explored. Various marker-based models and modelling techniques are used within the ACL/biomechanics literature, including models which allow for six degrees of freedom (6DOF) at each joint (Cappozzo, Catani, Della Croce and Leardini, 1995), models that implement the calibrated anatomical systems technique

(CAST) and/or use optimization processes in the estimation of joint center positions (Charlton et al., 2004). These models and techniques are thought to overcome some of the limitations ascribed to the CGM. For example, optimisation processes via the use of dynamic functional trials can be used to minimise the effect of marker placement error and provide improved estimates of joint centre positions. Comparison of kinematics and kinetics calculated using the CGM and these alternative models and techniques demonstrates discrepancies in kinematics and kinetics, particularly in frontal and transverse plane kinematics (Ferrari et al., 2008; Collins et al., 2009). Improved inter and intra-tester variability has been reported when using optimised joint positions compared to the CGM method of direct estimation from marker positions, while Groen et al., (2012) found the joint centre optimisation reduced the sensitivity of frontal and transverse plane kinematics to systematic marker placement error (Charlton et al., 2004; Groen et al., 2012). Combined, this evidence suggests that alternative modelling techniques may provide more reliable estimates of CoD kinematics and kinetics during CoD than the CGM. However, as all marker-based models continue to operate on the assumption of repeatable anatomical landmark location, a thorough exploration of the sensitivity of alternative modelling techniques to marker placement error is required prior to making definitive statements about their use for individual and group assessments of CoD kinematics and kinetics.

Exploring the effect of marker placement error on alternative models and techniques will inform as to whether there is currently a method available that allows for more precise monitoring of individual CoD kinematics and kinetics than the CGM. However, this should be viewed as an initial step in the process of using these data in prospective research to identify biomechanical risk factors for ACL injury. There are various additional methodological considerations that warrant investigation in a manner similar to those studied in this thesis. Soft tissue artefact (Pain and Challis, 2006), differences in body segment inertial parameters (Monnet et al., 2010), the use of different filter cut-off frequencies (Kristianslund et al., 2012), analysis techniques (continuous v discrete) (Marshall et al., 2015) and landmark registration (Athif et al., 2020) all have the potential to introduce variability into analyses and/or alter the clinical interpretation of kinematic and kinetic data. Exploring these issues and identifying methods to reduce their influence on kinematics and kinetics (soft tissue artifact, body segment inertial parameters) or developing a consistency in approach (filter cut-off frequencies, analysis techniques, landmark registration) will allow researchers to have greater confidence in the data they collect and the subsequent clinical interpretation. As the ability to identify and monitor individual kinematics and kinetics during CoD improves, the likelihood of identifying associations, if they exist, between biomechanical variables and outcomes such as secondary ACL injury will increase. It may also eventually allow biomechanical risk factors identified via prospective research to be used on an individual basis to appraise injury risk and monitor ACLR rehabilitation progress.

While extensive research needs to be conducted to progress to a point where biomechanical analyses can be used to monitor individual rehab progress and inform return to play decision making following ACLR, there remain ways in which this technology can be used to inform best practice in ACLR rehabilitation. Group-based analyses can still be used to identify statistical associations between kinematic and kinetic variables and clinical outcomes. For example, this thesis demonstrates that currently KAM cannot be measured reliably on an individual level during CoD and thus cannot be used to make any inferences about an individual patient's risk of injury. However, neuromuscular training programs based on the findings of Hewett et al., (2005), aimed primarily at improving frontal plane knee mechanics during landing, have been shown to reduce the incidence of both primary and secondary ACL injury in high risk populations (Gilchrist et al., 2008). Thus, while currently not feasible to use biomechanical analyses to screen for injury risk and/or monitor ACLR rehabilitation progress, it can be used to identify associations between biomechanical variables and outcomes such as secondary injury risk. These associations can be used by researchers and clinicians to develop rehabilitation interventions aimed at improving clinical outcomes, the effectiveness of which can be assessed via randomised control trials.

When using biomechanical analyses in this manner, greater consideration should be given to effectively communicating relevant methodological decisions within any given analysis. Detailed information with respect to marker placement protocols (number of individuals applying markers, laboratory inter and intra-tester variability ranges, etc.), modelling techniques (specific model used, modelling assumptions, joint constraints, etc), as well as a full description of the task studied, the task features which may influence dependent variables and how they were or were not controlled should be supplied in all methodology sections. Removing ambiguity with respect to these features will greatly increase the interpretability of study findings, as well as the ability to compare results between studies.

There are alternative methods for modelling human movement outside of the marker-based biomechanical models used (CGM) and discussed in this thesis thus far (CAST, 6DOF). In recent years both inertial measurement units (IMUs) and markerless motion capture systems have gained popularity (Ghiotti et al., 2018; Wade et al., 2022). IMUs utilise a combination of triaxial accelerometers, gyroscopes and magnetic sensors to estimate segment post (Al-amri et al., 2018), while markerless systems typically combine deep learning and neural networks to estimate segment pose in three-dimensions from two-dimensional video images (Nakano et al., 2020). Both these techniques over-come the limitations of marker-based biomechanical models with respect to accurate and repeatable marker placement, while also allowing for biomechanical assessments to be performed in real-world field-based settings, as opposed to being confined to laboratory environments as with marker-based analyses. However, while IMUs and markerless systems offer exciting alternatives to traditional biomechanical assessments, their use is not without methodological challenges of their own. IMUs can suffer from integration drift, whereby small errors in acceleration and angular velocity will progressively be integrated into larger errors in velocity, angle and position (Al-amri et al., 2018). Markeless based systems utilise various pose-estimation algorithms to identify anatomical landmarks from video footage. These algorithms are trained on large scale data sets, where key anatomical landmarks are manually labelled (Wade et al., 2022). This is an extremely time consuming process and means that the effectiveness of any given pose-estimation algorithm is contingent on the quality of the training data set and the manner in which anatomical landmarks were identified. This issue has been highlighted in recent work where the accuracy of markerless systems in estimating segment pose has been questions (Nakano et al., 2020). Thus, while alternatives to traditional marker-based analyses to exist, they present their own unique methodological challenges that need to be explored prior to their use in clinical settings.

6.2 Conclusion

The aim of this thesis was to examine the feasibility of using biomechanical analyses to identify abnormalities in CoD technique following ACLR. An improved understanding of alterations to CoD technique following ACLR may supplement and develop the wider understanding of biomechanical risk factors for both primary and secondary ACL injuries. The findings from this thesis highlight considerable challenges in conducting such assessments and demonstrate how small differences in methodological processes can have a large effect on kinematic and kinetic variables during CoD. If data collected from different laboratories, under different experimental studies are to be used to further the general understanding of ACL injuries, and ultimately inform clinical practice, a greater emphasis should be placed on understanding how methodological variation influences clinical outcome measures. Together, the four experimental studies presented in this thesis offer a framework for best practice when examining and quantifying inter-limb differences in kinematic and kinetic variables during CoD. Following this framework will allow researchers to have greater confidence in their findings and their subsequent clinical interpretation of data. Individual athlete monitoring of kinematic and kinetic inter-limb differences remains a significant challenge however, with the CGMs sensitivity to marker placement precluding its use in this context for the majority of kinematic and kinetic variables. Alternative biomechanical models may be better placed in this regard, but prior to their use in a clinical context, methodological issues and their influence on clinical outcome measures should be fully explored in a manner similar to this thesis.

Chapter 7

Appendix

7.1 Appendix A - Data Collection Procotols

7.1.1 Participant Recruitment

This thesis used a combination of participants who had undergone primary ACLR (Chapters 2, 3 and 4) as well as healthy non-injured participants (Chapter 3 and 5). ACLR participants were recruited through the Sports Surgery Clinic, Dublin, Ireland, where they had initially attended following suspected ACL rupture. Following diagnosis of ACL rupture by an orthopaedic consultant participants were enrolled in the Sports Surgery Clinic's ACL pathway and ACLR surgery was scheduled with one of two orthopaedic consultants. ACLR was performed using either a bone patellar tendon graft or hamstring graft (semitendinosus/gracilis) harvested from the ipsilateral limb during surgery. As part of the Sports Surgery Clinic's ACL pathway, ACLR participants returned to the clinic at 3, 6 and 9 months post-ACLR to under go a physical testing battery in order to assess the progress of their rehabilitation. The reviews at 6 and 9 months post-ACLR involved the analysis of a series of jump and CoD tasks using the Sport Surgery Clinics' biomechanics laboratory, as well as isokinetic dynamometry assessments of knee flexor/extensor strength. Data collected in these assessments was then filtered down by specific patient criteria i.e. male, aged 18 - 35, participation in multi-directional field based sport at the time of injury and the intention to return to the same level of sports participation following rehabilitation. These data were then used for analyses in Chapters 2 and 3 (9 month assessment data) and Chapter 4 (6 month assessment data). The healthy non-injured cohort used in Chapter 5 were recruited from a collection of multi-directional field based sports teams situated locally to the Sports Surgery Clinic. This cohort were all male, aged 18 - 35 and playing multi-directional field based sports at a competitive level. They underwent a once of physical assessment, identical to that which the ACLR participants completed at 6 and 9 months post-surgery.

The documents presented below are extracted sections from the Sport Surgery Clinic's Biomechanics Laboratory Manual.

7.1.2 Marker Placement

SSC Marker Placement Protocol

The SSC Biomechanics lab uses a modified 'Vicon PlugIn Gait' model which itself is a modified form of the classical Helen Hayes conventional gait model. It is a hierarchical biomechanical model in which markers are placed on joints to define the two adjacent segments. Since body segments are not tracked independently, this may allow errors to propagate 'downstream', cascading from the pelvis through the thigh, shank and foot segments. A consistent and accurate method for marker placement and data acquisition is essential to ensure that data can be compared between patients and with normal values. The accurate placement of kinematic markers is essential if accurate results are to be obtained.

Be ready in the lab before the time a patient is booked in to ensure markers are prepared, testing sheets are ready, etc. When a patient arrives early, marker them early if it is feasible to do so.

General notes

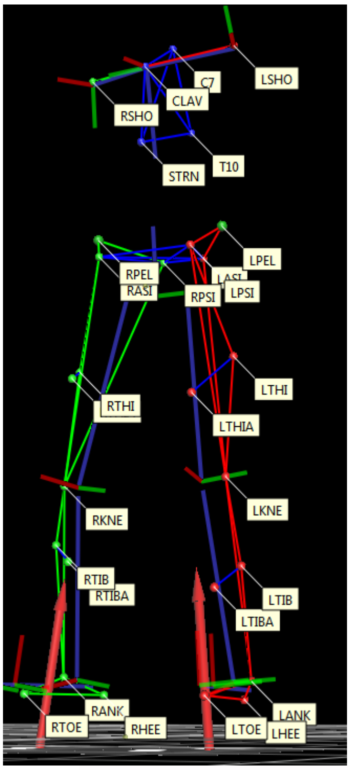
- Ensure you have the correct patient by (discreetly) asking them to confirm their date of birth when you first call them through into the lab.
- Give the patient a brief overview of the full process if this is their first time doing 3D testing at the SSC.
- Get the patient to change into shorts and remove jewellery before you begin.
- Socks should be below ankles so as not to obstruct the ankle markers. Patients may need to remove long or thick socks that cannot be folded down.
- Use zinc oxide tape or masking tape to cover reflective material on runners/clothing if necessary.
- The lab has spare runners, shorts and sleeveless tops available for patients who arrive without appropriate clothing. Ensure these are returned and washed afterwards (you may need to remind the patient to bring the clothing back after they have seen the physio). If markers need to be placed over loose clothing, e.g. on females who have not brought in a sports bra to wear, make a note of this on the testing sheet as the patient will be excluded from the research cohort.
- To help the motion capture system distinguish between the markers on the left side and the right side of the body, place markers on the tibias slightly asymmetrically. As shown in the picture below, markers on the left side are placed more distally.
- Take all subject measurements before placing markers on otherwise the markers may get in the way of this process.
- When finished, the marker positions should move as little as possible and should not be obscured by clothing.
- Make note of the shoe heel to toe drop ('sole delta' on the data collection sheet; see image below). If the patient is wearing raised insoles this should also be recorded.

- Be conscious of patient dignity. Explain what you are doing to the patient before you touch them and get the patient to adjust their own clothing if necessary (e.g. moving their waistband down) rather than doing it for them.

- Be aware of how the previous test is progressing in the lab. If the previous patient is running late you should give your patient questionnaires to complete before they remove their clothing or marker more slowly than usual so the patient is not left waiting around after you finish preparing them for testing.

- If the patient is aged under 18 ensure there are two people present at all times during markering.

- Remember your patient is in a potentially vulnerable position and behave with the utmost professionalism. Some patients may be more comfortable being markered by a biomechanist of the same gender.

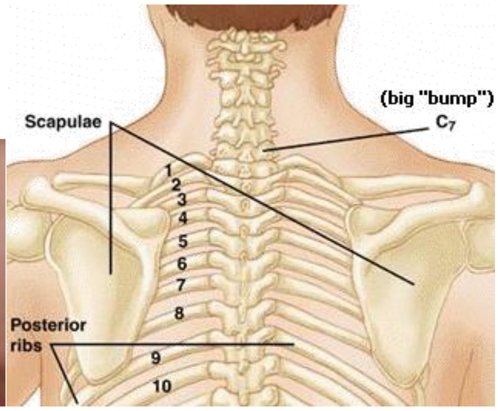
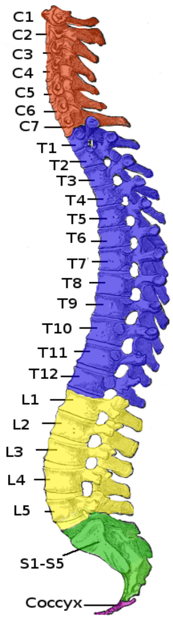


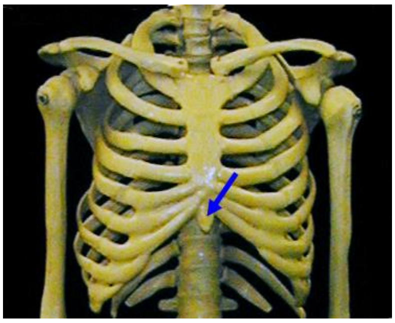


Heel to toe drop = heel height (mm) – toe height (mm)

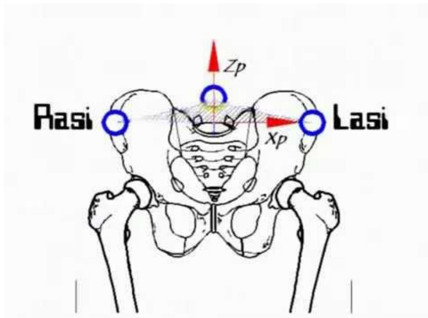
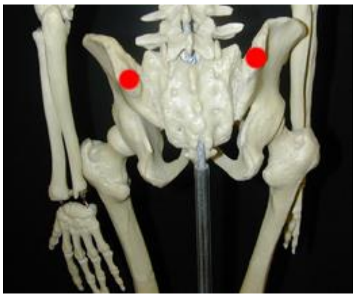
Marker locations

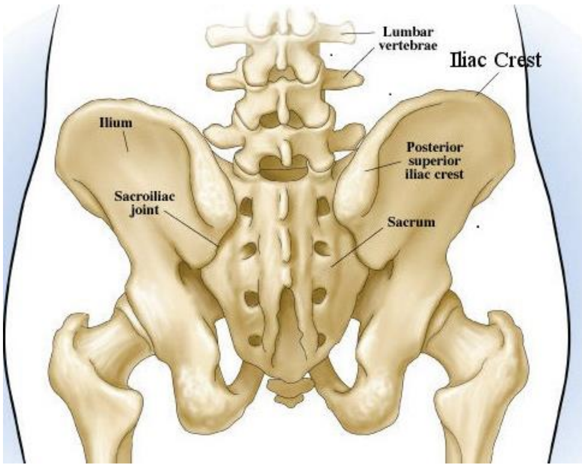
Upper Body

C7	**7th Cervical Vertebra.** Spinous process of 7th cervical vertebrae. The C7 bone is located along the spinal column right where the back of the neck ends. Have the subject bend his or her head down and back up. (small bone jutting out will indicate this).	 
C10	**10th Thoracic Vertebra.** Spinous process of 10th thoracic vertebrae. The T10 is also located along the spinal column. Count 10 vertebrae down from C7. T10 is usually found in line with the inferior angle of the spine of the scapula.	
CLAV	**The Clavicle** is placed in between the two collar bones and below the base of the neck above the jugular notch where the clavicle meets the sternum. The marker should be placed on the bone and not in the jugular notch itself.	

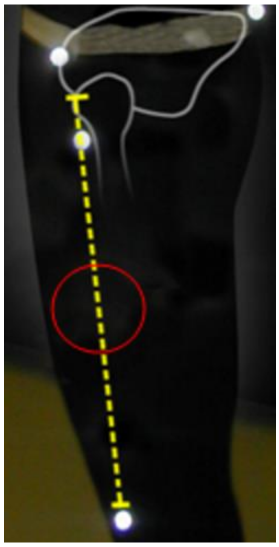
STRN	**Xiphoid process of the Sternum**. This marker must be placed on the bone just above the Xiphoid process. Put the marker on the base of the middle of the ribcage. For women, the marker is placed below the breast**.**	
LSHO/ RSHO	**Shoulder markers** Placed on the acromio-clavicular (AC) joint.	**See figure above**

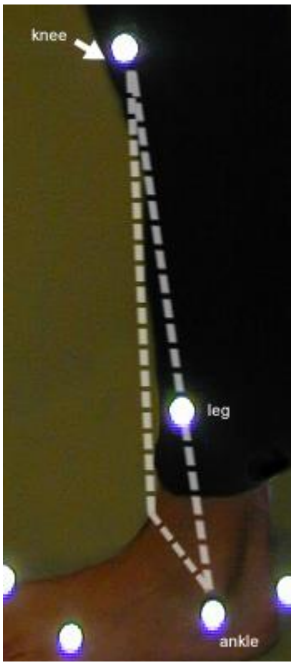
Pelvis

LASI/ RASI	**Anterior Superior Iliac Spine**. Placed on bony prominence of ASIS. Find the area of the pelvis bone that juts out the most and place the marker at the point where the hip flexor tendons meet the bone.	
LPSI/ RPSI	**Posterior Superior Iliac Spine**. Put the marker on the bony prominence that can be felt below the dimples at the point where the spine joins the pelvis. If a participant is difficult to palpate simply ask them to bend forward (but place the markers on the body in an upright position).	

LPEL/ RPEL	**Iliac Crest**. Place markers bilaterally level and in line with the greater trochanter.	

Lower body

LTHI/ RTHI	**Mid-thigh Marker**. Identify and mark the most prominent aspect of the greater trochanter. Mark this and place the marker midway between greater trochanter and knee mark. Greater trochanter can be identified by getting participant to internally/ externally rotate thigh. (to wiggle the foot)	
LKNE/RKNE	**Knee joint marker**. Located on the lateral epicondyle of the knee. Marker is placed 1.5cm above the midpoint of the joint line.	

LTIB/ RTIB	**Tibia marker** should be placed on the lateral tibia, in alignment with the knee marker and the midpoint of lateral malleolus of ankle. Left tibia marker is placed lower (more distally) than right.	
LANK/ RANK	**Ankle marker**. Placed on the middle of the lateral malleolus. Make sure marker is protruding laterally rather than anteriorly/ posteriorly.	
LTOE/RTOE	**Toe marker**. Located on the base of the 2nd metatarsal head. Placed on shoe, feel for the base of the 2nd metatarsal head. Ask the participant to flex their toes to the ceiling to determine this point.	
LHEE/RHEE	**Heel marker**. Placed on shoe, on the calcaneus at the same height as toe marker.	

The marker *placement* process

What you need

- 28 x 14 mm retro-reflective markers on double-sided tape
- Anthropometric callipers
- Wooden rule
- Small ruler/set square
- Pre-tape spray
- Paper towel
- Zinc oxide tape and masking tape
- Patient data collection sheets and consent form

The markering area

Check that the area is tidy and you have everything you need before calling the patient into the lab. Ensure the space is left ready for the next patient (markers and tape prepared, your patient's possessions cleared away) when you are done.

Before you start

See also section 4.

- Ask participant to sign consent form
- Take height (mm) and weight (kg)
- Make note of the shoe heel to toe drop (sole data)
- Put strips of zinc oxide tape on the shoes where you will later place the markers
- Ask the participant all questions on the collection sheet and record responses

Palpation and marker placement process

1. Get the participant to sit down with feet on the step up box whilst marking the lateral and medial malleoli.

Take ankle width with the anthropometric calipers. You are measuring the distance across the skeleton from medial to lateral malleolus. *Record ankle width for each side on the data collection sheet.*

2. Ask patient to step on the second step up box and lean against the wall, shift all the weight on the inside leg and a slight bend in the outside knee to mark it.

What you are looking for:

- Anteriorly: the gap between tibia and femur in the joint line on top of the tibial plateau
- Posteriorly: the last bony prominence of the knee joint (not the tendon) at the same superior-inferior position as the anterior mark.

Mark those two points and draw a line between them with the knee fully-extended.

Take the middle of the line, measure 1.5 cm proximal from this point with the patient's knee full-extended and mark the location. This should be the axis of rotation of the knee and where you stick your marker.

How you can check this: Get participant to flex and extend knee (bring the heel up to the bum).Your identified landmark should not move during the movement. As an alternative, ask the participant to go down into a squat and up again. There will inevitably be some movement of the skin over the joint but the marked position should not be visibly offset from the centre of rotation of the joint.

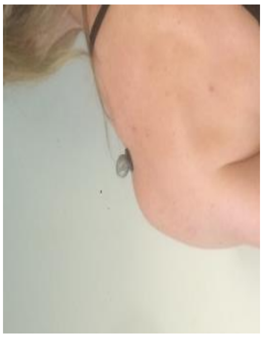
3. Measure knee width, with the participant sitting down and the knee flexed at approximately 90°. The caliper ends should be placed laterally on your knee marker and medially so as to define the flexion-extension axis of rotation of the knee. Ensure you apply sufficient pressure to measure the width of the underlying structure rather than soft tissue around the knee.
Record knee width for each side on the data collection sheet.

4. **Marking the greater trochanter (GT):** participant can step down from the box so that you can tape up the shorts (see picture above). Ask the patient to roll the hem of their shorts up to the waistband and pull the waistband out so you can wrap the leg of the shorts in masking tape.

Get participant to stand with their feet shoulder-width apart and their toes pointing straight forward, facing 12 o'clock. It is important that the feet are perfectly parallel on the ground, otherwise your mark will be off. Now you can start palpating the thighs for the GT. Move the palm of your hand over the thigh from anterior to posterior, superior too inferior and vice versa to get a feeling for where the most prominent part of the GT is. If the GT is challenging to locate, ask the participant to internally and externally rotate the foot with the heel on the ground. You should feel the prominence of the GT pop out under your hand. The GT slopes posterior-laterally so ensure your mark is on the lateral prominence.

5. The next step is to find **LASI/RASI.** Ask the patient to remove their top (confirm they have their sports bra on first if female). In order to palpate the **anterior superior iliac spine (ASIS),** first put your thumb on the hip flexor tendons on the anterior aspect of the pelvis and palpate superiorly until you reach the bony prominence of the ASIS, where the hip flexor tendons insert onto the pelvis. If you are struggling to locate the correct point, ask the participant to take the weight off the ipsilateral leg to reduce tension in the tendons. Do not mark your point until you have located it with the patient standing evenly.

6. Measure the leg length from the ASIS to the ipsilateral medial malleolus. The tape measure should be taught and should pass as directly as possible between the two points. Repeat the measurement at least twice on each side to ensure you are consistent. *Record leg length for each side on the data collection sheet.*



7. The next step is to mark lines on the participant's legs between the GT and the knee and between the knee and lateral malleolus. Get down to the level of the line and ensure you are looking at the leg from a directly-lateral position to avoid errors caused by parallax.

Mark the midway point between GT and knee on each line. Mark a point on the line between the knee and ankle, selecting a more distal point on the left leg than the right. Mark the anterior thigh and tibia marker positions (exact location not important but must be marked to facilitate the accurate replacement of any markers that fall off during testing).

8. Locate RPSI/LPSI at the back of the pelvis. These are bilateral prominences that can be palpated around the inferior-lateral corner of the dimples in the lower back. Ask the participant to bend forward to make the prominences more pronounced if they are difficult to find but always mark the point in a standing position.

9. To mark the shoulders, get participant to stand upright with arms relaxed by their sides. Feel for the plateau behind the bony prominence of the AC joint at the shoulder.

10. The C7 vertebra is located along the spinal column at the base of the neck. The spinous process can be located by getting the subject to flex and extend the cervical spine ("chin down to chest and then back up again"). C7 will remain prominent and will not change orientation when the neck flexes (unlike all other cervical vertebrae). C6, immediately superior to the target vertebra, will disappear when the neck extends and reappear when it is flexed.

11. The T10 vertebra is also located along the spinal column. To approximate the position of T10, get the participant to bring their arm behind the back to make the lower aspect of the scapula protrude. T10 will be at approximately the same level as the inferior aspect of the scapula. You can also count ten vertebrae down from C7 (ask participant to round their back with outstretched arms braced against the wall to open up the spine).

11. To mark the sternum, palpate for the base of the middle of the ribcage. For females the marker will be placed below the breasts, often on the band of the sports bra.

12. To mark the clavicle, feel for where the two collar bones meet below the base of the neck. The marker should be placed on bone at the base of the jugular notch.

13. Use pre tape spray on all lower limb and pelvis markers (also the chest or back if a participant is particularly hairy). Check with the participant that they have no known allergy to pre-tape spray or any other aerosol before applying. Use a sheet of paper towel to remove excess spray and prevent it running down onto clothing.

14. Get the participant to replace shoes.

15. Place markers on lower and upper body. Ensure the marker is exactly on the point you have marked.

The markers on the back will be furthest from the cameras when the participant is standing on the force plate due to the camera configuration. The cleanest and shiniest markers should therefore be placed on L/RPSI, C7 and T10.

Place shoe markers and secure with zinc oxide tape.

Patient is now ready to begin the 3D motion capture testing.

7.1.3 3D Laboratory Data Collection

Data capture protocols

3D motion analysis: Vicon

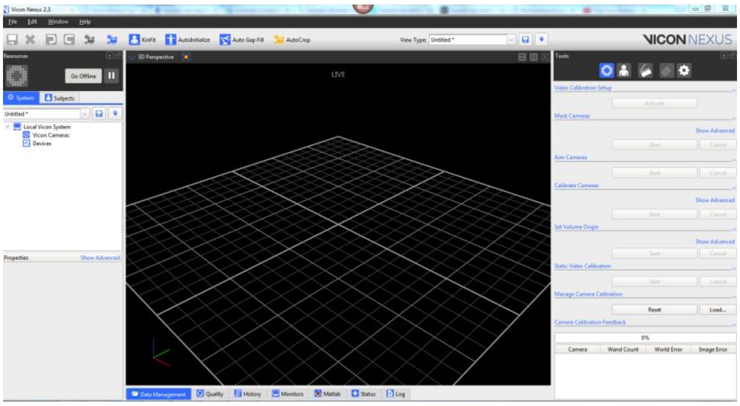
Before testing begins

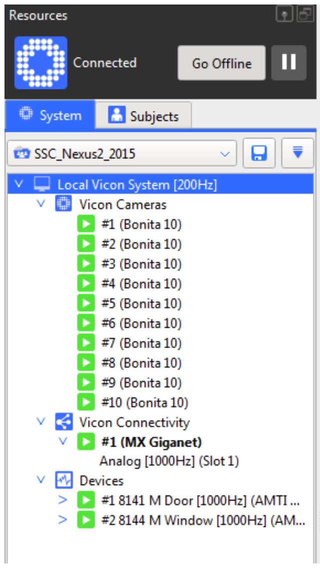
- Look at iMed the previous day for the day's schedule.
- Prepare data collection sheets and consent forms. Groin patients use de-identification codes and therefore name and date of birth should ONLY be placed on consent forms and not on data collection sheets.
- The lab should be clean and tidy. Ensure rubbish has been disposed of and the previous patient's possessions have been removed from the lab.
- 3D testing includes the use of a 30 cm box, a 20 cm step and a 15 cm hurdle. Make sure these are available before testing commences. Some testing protocols may require additional equipment which should also be checked before the patient is brought through into the testing area.
- Ensure SmartSpeed timing gates, SmartSpeed Hub and calibration wand are charged prior to testing.

Calibration

This is the standard calibration procedure for the SSC Biomechanics Lab. It should be adapted appropriately when needed. The cameras should, when possible, be switched on 10 minutes before the calibration is performed to give them time to warm up.

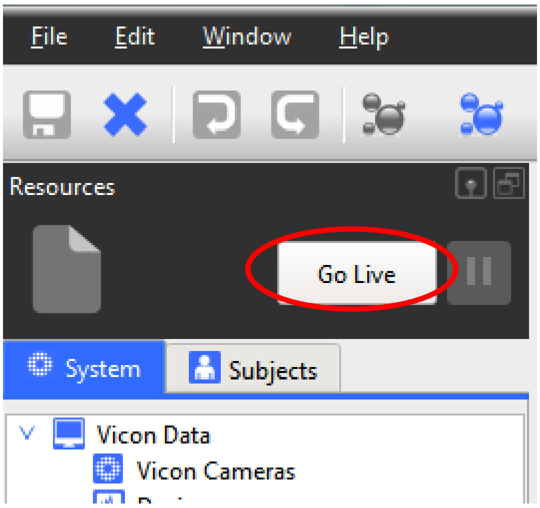
- Ensure curtains are closed and no surplus reflective items are in the lab.
- Ensure PoE switch is switched on.
- Open Vicon Nexus.
- This screen will appear:

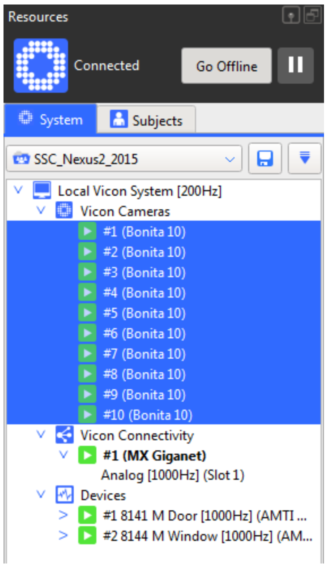




- Right-click ‘Devices’ and select Add Digital Device → AMTI (first option)
- Ensure that all the cameras and force plates are marked with a green icon in the Resources pane. If a camera/force plate is NOT highlighted green, ensure the device is connected, right-click on the icon and select reboot.
- Force synchronisation of the force plates and motion capture system by right-clicking Local Vicon System and selecting Synchronise.
- Switch on the calibration wand.

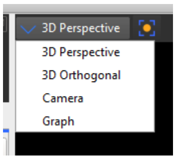
- Select Live button:





- Select all the cameras in the Resources pane.

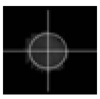
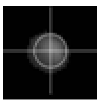
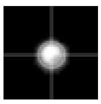
- In the top left hand corner of the "Live" screen you can change your viewing option to Camera.



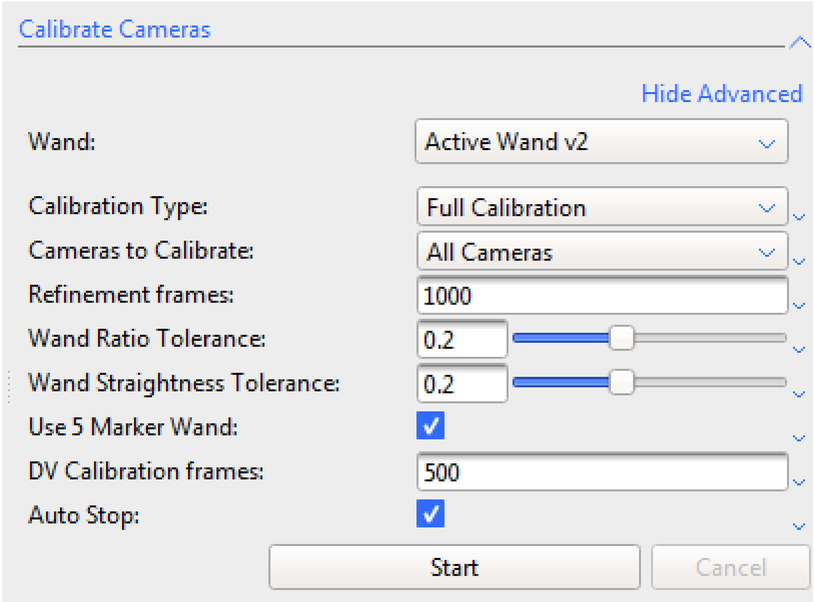
- Check that each camera is viewing only the 5 markers on the wand (5 trajectories).

- Remove anything else the camera may be picking up (reflections from shiny surfaces, etc.)

- Screen each of the cameras individually. Ensure that it sees each of the 5 markers as depicted below. The images below show a marker as viewed by camera that is focused too far away (a), too near (b) and well-focused (c). If this is not the case, the camera may need to be adjusted.

A  B  C 

- Whilst it is the lab default, check that 'Active Wand v2' is selected in the Tools pane (Calibrate Cameras → Show Advanced) and that the following parameters are selected:

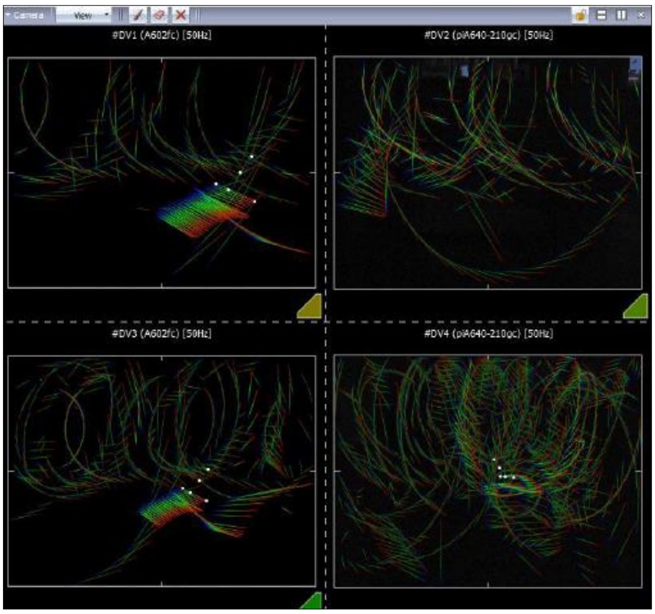


- Highlight all 10 cameras. Pick up the wand and turn the computer screen around to the calibration area.

- Start the dynamic calibration by selecting Start under Calibrate Cameras in the Tools pane.

- Be sure to cover the entire volume where your subject will be during capture and wave the wand all the way down to the floor and up to the height of your subjects.

- Check the coloured triangle in the lower right corner of each camera view. As more frames are captured by each camera the triangles will change from red to green. When calibration for that camera is complete the triangle will disappear.



- Once enough data is captured Nexus will begin processing it automatically. The calibration is done in two passes, so the Camera calibration bar will progress from 0 - 100% twice.

- SSC Lab guidelines are that only an image error of less than **0.35** should be accepted. If this is not achieved the dynamic calibration should be repeated.

- The purpose of this calibration is to define a reference frame for each camera relative to all the other cameras. The next stage is to set the origin and axis orientation for the desired laboratory reference frame. Set the volume origin and axes by carefully placing the wand in between the two force plates ensuring it is level and that it is turned on. In the Tools pane select start and then Set Volume Origin.

- Switch back to 3D perspective view and ensure that the origin is in the correct location with respect to the force plates.

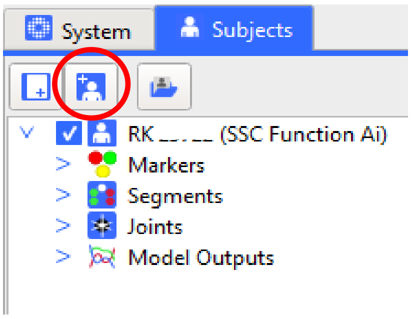
- Remove wand, switch off and replace on the wall, being careful not to knock or otherwise damage it.

- To calibrate the two AMTI force platforms press 'zero' on the AMTI amplifier boxes then right-click on the force plates and select "Zero Level" for each.

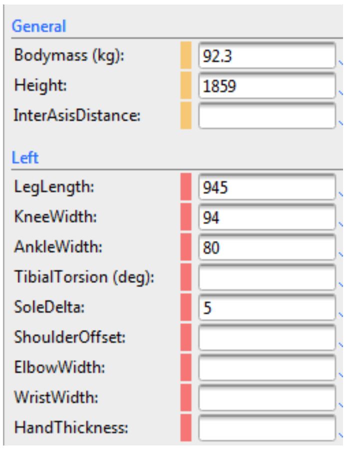
Create subject and enter anthropometric data

- To create a new subject press F2 to display the Data Management window.
- Select the correct classification. This will normally be the current month (e.g. May2016) except for norms or special experimental populations.
- Select new Subject () and name using patient's initials and SSC number (if not groin patient) or de-identification code (if groin patient).
    - When creating an ACL subject it is important to note whether the patient is a 6 or 9 month test and whether they have had a previous ACL reconstruction. This information will have been recorded on the data collection sheet. If you are unsure how to name a patient's data collection folder please ask a colleague – it is better to ensure it is correct at the time than to have to go back and attempt to identify issues later on.
        - A 6 month patient (Test 1) who has never had a previous ACL reconstruction on either side will just be named as 'initials + SSC number', e.g. AB 123456

        - A 9 month patient (Test 2) who has never had a previous ACL reconstruction on either side will be named as ‘initials, SSC number, Retest’, e.g. AB 123456 Retest
        - ‘6 month’ and ‘9 month’ in this context refer to *the review file on iMed*, not the actual dates from surgery. For example, a patient who is booked in for Test 1 (6 month review) but had their surgery 9 months ago will still be named AB 123456, and a patient who is booked in for Test 2 (9 month review) but never attended their Test 1 will still be named AB 123456 Retest.
        - The initials entered should be the patient’s initials as per their iMed file. If the patient’s iMed file is under the name ‘Emily Jane Smith’, the initials you save the data under should be EJS. Do not use hyphens, fadas or any other punctuation in the file name. If the patient has ‘Og’ or ‘Ní’ as part of their name, treat it as you would any other initial (Fred Og Ryan = FOR; Ainé Ní Nullian = ANN).
        - There are two exception cases in which you would enter more than the first initial of each word in the name: Mac and Mc. In order to help distinguish between patients, each of these is entered as the full prefix. David McFadden would therefore be named DMcF, and Sam Og McGovern would be named SOMcG.
        - The words ‘Revision’ and ‘Contra’ are used after the SSC number to indicate that the patient has a previous ACL reconstruction on the same side (Revision) or the other side (Contra). A 6-month ACL patient who has had a previous ACL reconstruction on the same leg will be named ‘AB 123456 Revision’. A 6-month ACL patient who has had a previous ACL reconstruction on the contralateral side will be named ‘AB 123456 Contra’. If the patient has had both (i.e. this is their third reconstruction), they can be named ‘… Revision Contra’. ‘Retest’, if applicable, always goes at the end: ‘AB 123456 Revision Retest’.
    - Groin pain patients will have de-identification codes and will be named as such, e.g. “1234”. A groin patient returning for a retest will be named “1234 Retest”.
- Fill in the required database information at Subject level – age, gender, dominant side, injured side and cohort, NFR/Ext, Taken for cleaning.
- Double click the subject and create a new Session () folder.
- Make sure you are in the new session and then return to the capture screen
- In the Resources pane create a new subject from a labelling skeleton and select **SSC Function Ai** from the drop-down menu**:**



- Click on the subject name (‘RK XXXXX’ in the example above) and enter the Subject anthropometric details. Nexus fills in default values for some of these parameters which you will need to replace with the correct values for your patient. InterAsisDistance will autocomplete later on so delete the default value from this box and leave it blank. Scroll down to do the same for the left leg. Upper body parameters are all left blank as these are not included in our model.



Key information for patient

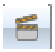
When the patient is first brought out of the markering area into the motion capture space the following key information should be given before testing continues:

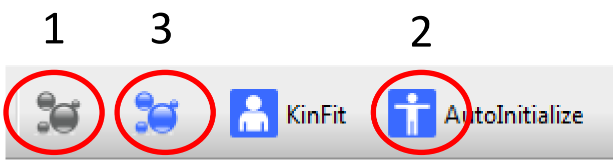
- The patient should be **introduced to the testing space** if they haven’t been in the lab before – “you can see the cameras all around with the red rings of light – that light reflects off the markers and that’s how we track how you are moving through the space” or similar. DO NOT point out the force plates: their presence should be de-emphasised to avoid targeting during the cutting exercises. If the patient asks why they are required to do the exercises in a particular location tell them this is where the cameras are focused.
- A **brief overview should be given of the exercises** to be undertaken – “Three set-up trials to start with then some squats, some jumps and some running and cutting” so the patient is clear about what will be asked of them
- It should be made clear that these are all **performance tests** and the patient will be asked to complete exercises to the best of their ability…
- … but they must be told to notify the tester if they **feel any pain or are uncomfortable with any of the exercises**. It is not uncommon for ACL patients to experience some mild stiffness or soreness in their knee and initial groin patients are likely to have some discomfort in cutting exercises but this should be monitored and any unusual pain, acute pain or increase in pain may require the testing to be discontinued. If in any doubt, refer to a Senior Biomechanist.

Set-up trials

Static Trial

The static trial enables the Plug-in Gait model (SSC Function Ai) to associate captured markers with known positions or labels and to calculate certain key parameters that are used during dynamic trials (marker placement should be screened visually by the tester as well as a double-check of greater trochanter markings).

- On the force plates, have the subject stand in a stationary pose (“looking straight ahead, standing very still with hands on bum”), to enable the Vicon system to determine the location of key markers. The patient should have feet placed shoulder-width apart, toes pointing straight forwards and toes level in the anterior-posterior direction. Check foot position and get the patient to adjust if necessary. Ensure all markers are visible in Nexus Live view and that there no additional ‘ghost markers’ caused by reflections off clothing or shoes (cover with tape if required).
- In the Tools pane select Capture () and capture a static trial by typing in the trial name “StaticTrial” and pressing Start. Capture approximately 600 frames (three seconds) and then press Stop. Repeat if the patient moves visibly during the trial. Tell the subject that he/she can relax their arms but to keep their feet where they are (in case you need to repeat the trial).



- Press F2 to open up the Data Management window and double-click your static trial to open it. Reconstruct the marker positions (shortcut 1 below in the top toolbar) and check that all markers are visible in the reconstruction.

- Run AutoInitialize (shortcut 2). This applies marker labels based on the selected template to the reconstructed marker positions. Check that all these labels are correct and make corrections if necessary by selecting Label () in the Tools pane and manually labelling the reconstructed markers.
- Once complete, select Pipelines () in the Tools pane and run **Plug-in Gait Static**. This will visually create segment volumes on the static trial. SAVE YOUR TRIAL.
- You can now call the patient over and show them the image on the screen. Point out the different body parts and explain what the model allows us to see.

Range of Motion trial (ROM)

The Range of Motion (ROM) trial facilitates movement tracking by associating trajectories with marker and hence joint positions and improving Nexus reconstruction and labelling performance. The ROM trial can also be used to compute functional joint centres rather than those defined by marker locations.

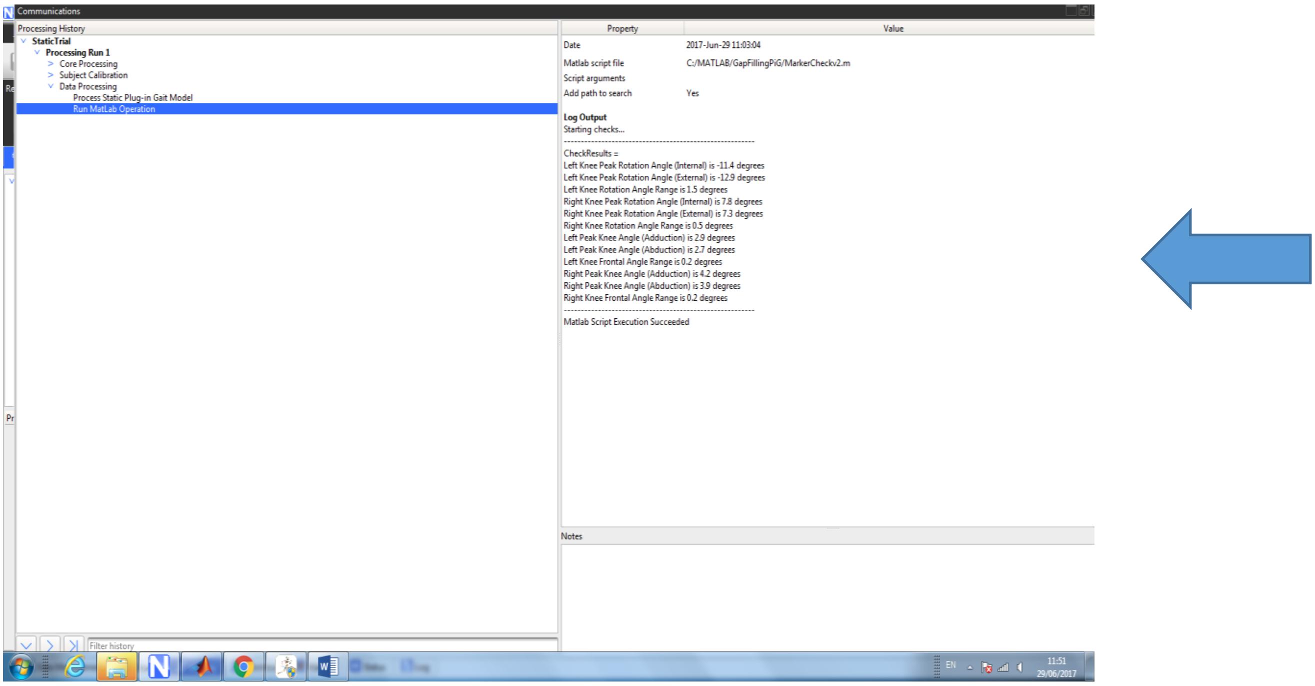
- The patient should begin standing in neutral position (as for the static trial) but with their hands on their hips (ensure the hands are not covering up any markers) and should then perform the following sequence of actions:
  - Circumduction of the hip (both sides one after the other): “circle with your left/right leg”
  - Squat – “squat down and up”
  - Circular movement of the waist area – “circle with the hips”
  - Flex the hip joints with both feet on the ground – “tip your upper body forwards”
  - Extend the hip joints and lumbar/thoracic spine – “tip your upper body backwards”
  - Return to the neutral starting position – “return to an upright position”
- The trial should be named ‘ROM’. It doesn’t matter if the patient makes small adjustments to foot or arm position during this process as long as markers are not obscured. The process should be demonstrated to the patient before recording commences and they should be cued through the individual movement components during recording.
- Press F2 to open the Data Management window as before and double-click the ROM trial to open it. Reconstruct and label the trial (shortcut 3 in toolbar image), check labelling and correct manually if necessary. Small gaps in marker trajectories are acceptable but any labelling errors/failure to identify a marker should be corrected.
- Once complete, select Pipelines and run **Calibrate Labelling Skeleton ROM**. This can take a couple of minutes so the patient can be doing their warm-up in this time. SAVE YOUR TRIAL.

Walk Trial

Collect a trial named ‘Walk1’ :

- Patients starts standing on the force plates facing the back curtain, walks straight to the back curtain, touches it, turns around, walks straight back to the force plates and stops. Try to get them to walk straight, then turn at the wall, then walk straight again - i.e. not gradually turning during the whole far-end region of the trial
- Reconstruct+Label and run Calibrate Labelling Skeleton ROM as normal, save

- Reconstruct+Label Walk1 then model the trial
- With the Walk1 trial open, run CheckMarkers pipeline.
- Open History tab in Communications toolbar and check that the CheckResults values have been written in.



- The positions of the markers are used to apply a model to determine joint angles, etc. The MarkerCheck routine allows the model that has been applied to your markers to be checked. The markers that define the knee joint are most commonly misplaced – by screening knee angles before testing starts we are able to identify and correct these errors. Two variables are extracted: knee internal rotation angle range and knee internal rotation angle inter-limb asymmetry. All values for the first should be less than **20 degrees** and all for the second should be less than **10 degrees**.
- If your values fall outside the desired range, check your marker positions and lines (or get your teammate to do do), adjust as necessary and the Walk trial (save as 'Walk2').
- Collect a second static trial with the new marker positions (no need to repeat ROM). If the MarkerCheck test is not passed on the second attempt proceed with the testing but put a '1' in the MarkerCheckFail column at Patient level in the Nexus database.

Testing protocols

Before commencing all testing protocols the patient should undertake a two-minute warm-up jogging around the lab (unless different warm-up defined for a particular experimental protocol).

ACL

A demonstration and set verbal cues are given before each test.

Two sub-max practice trials are completed for each test followed by three recorded acceptable trials. Always test the un-injured leg first for each single leg test.

Label	Exercise	Verbal cues
Squat	Squat. Five performed one after the other with one recorded out of the middle. **Pass criteria:** Following cues, hands on hips.	Hands on hips, sitting back into it, keeping an upright trunk and squatting down to maximum parallel.
CMJ (1,2,3)	Double-Leg Counter Movement Jump. **Pass criteria:** Landing on the force plates/ one fluid motion/ legs straight in the air.	Same position with the hands, squatting down and jumping up in one fluid motion, trying to hit the top of your head off the ceiling.
SLCMJ (Left, Right, 1,2,3)	Single-Leg Counter Movement Jump. The participants will be asked to put their hands on their hips, and to bend their knee and do a maximal jump in one fluid motion. They will be instructed to try to hit their head off the ceiling. They will stand with one foot on the force plate, contralateral leg flexed behind, jump vertically and land with the same foot on the force plate. No instruction will be given on how to land, other than to stick and hold the landing until told to relax. **Pass criteria:** Keeping leg straight when in the air (e.g. no flicking heel behind, as height is calculated from flight time off the force plate). Sticking and holding the landing within the force plate.	Starting on one leg, free leg behind, hands on hips, I want you to hit the top of your head off the ceiling. Stick and hold the landing until told to relax.
DLDJ (Left, Right, 1,2,3)	Double-Leg Drop Jump. The participant will start with both feet on the box (30cm), feet slightly over the edge. They will be instructed to drop off the box, land with both feet simultaneously and immediately upon landing perform a maximal jump. No instruction will be given on how to land. **Pass criteria:** Following cues and landing within the force plates. Legs straight when in the air.	Your performance is based on your jump height and the minimum time spent on the ground. Start on top of the box, hands on hips, feet slightly over the edge. Drop off the box and, immediately upon landing, spending as little time as possible on the ground, jump as high as you can - try to bang the top of your head off the ceiling.

SLDJ (Left, Right, 1,2,3)	Single-Leg Drop Jump. The participant will start with one foot on the step (20cm), toe slightly over the edge and contralateral leg flexed behind them. They will be instructed to drop off the box and immediately upon landing perform a maximal jump. No instruction will be given on how to land. **Pass criteria:** Following cues and landing within the force plate. Take-off leg straight when in the air.	Same as before but this time single legged. Hands on hips, foot slightly over the edge. Drop off the box and, immediately upon landing, spending as little time as possible on the ground, jump as high as you can - try to bang the top of your head off the ceiling.
HH (Left, Right, 1,2,3)	Medial Hurdle Hop. The patient will be instructed to hop medially over 15cm hurdle hop, laterally back and then stick the landing. This will be performed as quickly as they can. **Pass criteria:** Following cues. The final landing should be stuck or nearly-stuck – a little shuffle is acceptable but do not accept the trial if the patient is not able to control their landing at all.	Hands on hips, jumping over and back as quick as you can. Spending as little time on the ground as possible and stick and hold the landing at the end.
SLHop (Left, Right, 1,2,3)	Single-Leg Hop For Distance. Start on force plate, hands on hips, hop as far as possible and land on same foot. The participant will be instructed to imagine themselves jumping all the way to the wall. Two jumps will be performed to set the distance, and a cone will be placed where the patient landed. The patient will then be asked to start from the cone and jump back towards the force plate, maximal jump and hold the landing. **Pass criteria:** Following cues and being able stick and hold landing position until told to relax.	Set the distance: Hands on hips, imagine yourself hopping all the way to the wall. Stick and hold your landing position until told to relax. Recording: Starting from the cone, jump towards the force plate, again keeping the hands on the hips, maximal jump and holding the landing until told to relax.
Cut (Left, Right, 1,2,3)	The participants are instructed to run as fast they can towards a dummy defender. They will be instructed to run all the way up to the defender and past the cones (to ensure that they land on the force plates) before cutting (90°) either left or right starting with their uninjured leg. The participants will be instructed to complete the task as quickly as possible running through the first gate all the way through the second gate as they are timed. **Pass criteria:** Running as fast as they can all the way past the second gate. If they start slowing down before, this will not count as a good trial. Patient needs to turn rather than perform a side step.	Starting by the red cones, run as fast as you can toward the defender, run all the way up to the defender passing the yellow cones before you cut in the direction of the light. The time starts as you pass the first gate and ends when you pass the second gate, so run as quickly as you can from the first set of gates all the way through the second set of gates.

IndecisionCut (Left, Right, 1,2,3)	The set-up will be identical to the planned cutting task. The participants are instructed to run as fast they can towards a dummy defender. As soon as the participant passes through the first timing gate either the left of the right timing gate will be triggered and will flash. The participant will be instructed to run all the way up to the defender and past the cones (to ensure that they land on the force plates) before cutting (90°) either left or right depending on which light is triggered. The participants will be instructed to complete the task as quickly as possible. **Pass criteria:** Running as quickly as they can all the way through the second set of gates. If they start slowing down before, this will not count as a good trial. Patient needs to turn rather than perform a side step.	Starting by the red cones, run as fast as you can toward the defender, as you pass the first set of gates it will trigger a light to the left or the right, run all the way up to the defender passing the yellow cones before you cut in the direction of the light. The time starts as you pass the first gate and ends when you pass the second gate, so run as quick as you can from the first set of gates all the way through the second set of gates.

7.1.4 Isokinetic Dynomometry

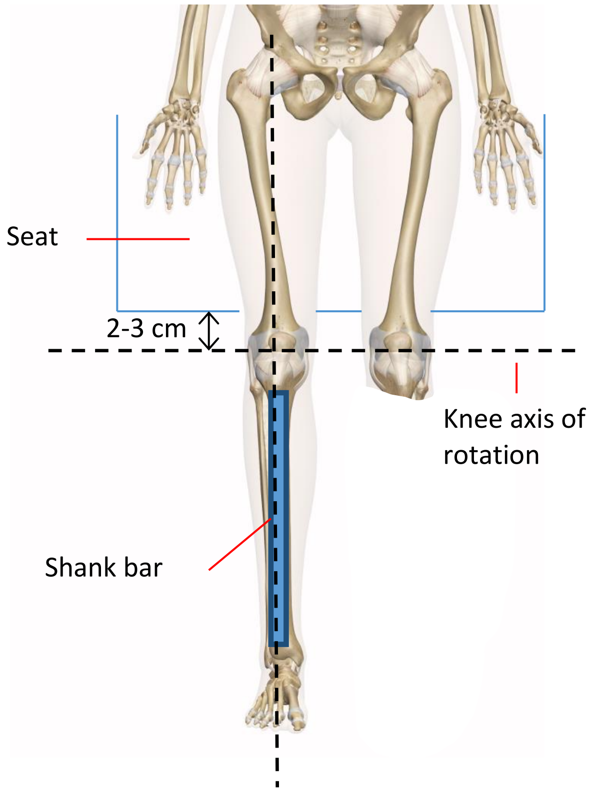
Isokinetic dynamometry

Standard ACL protocol

- Open HUMAC programme and perform Power-On test (follow instructions on screen).
- Uninjured (non-involved) leg to be tested first. Adjust chair and dynamometer to right or left side.
- Add a 'new' patient (or search an existing patient and select 'edit') and add their anthropometric information (height, weight, date of birth, involved leg, preferred leg etc.) and press 'OK'.
- Check that Standard Cybex Norm settings are in place:
    - Hip flexion: 85°
    - Chair rotation: 40°
    - Dyna tilt: 0°
    - Dyna height: 8
    - Dyna rotation: 40°

- Once the patient's information is in the system, press 'OK'.
- A list of different testing protocols will appear. You will select the 1st 'knee' test for flexion/ extension then press 'OK'.
- Select test method: '60 ACL Rehab test'. This testing procedure should include 3 sets of 5 reps with 60s break in between sets. Press 'OK'.
- On the next screen, you will choose which leg you are testing first and then click 'All Sets'. The next screen to pop up will be a reminder of standard Cybex Norm settings.
- Ensure the patient is sitting right back in the chair
- Adjust the chair back so that the patient's back is upright against the back of the chair and the patient's knee is about 3 cm off the edge of the seat to allow for free knee flexion. Both thighs should be parallel with knees pointing straight forwards.
- Put on chest straps and adjust as necessary.
- Adjust the monorail position as necessary so that the patient's leg is aligned with the shank bar on the anti-shear device. Ask the patient to extent the leg and ensure the shank bar is parallel to the tibia and that a virtual line extending beyond it would pass through the centre of the knee and hip joints.

- Adjust the chair position forwards or backwards to visually align the centre of the knee joint (as marked during 3D) with the centre of the dynamometer axis. See image below for bird's-eye view of patient correctly aligned for testing.



- Adjust the input arm length so that the lower pad is above the medial malleolus and the participant can fully dorsiflex and adjust the upper pad position so it is over the tibial tuberosity.
- Attach test leg's thigh and shank straps (straps should be firm but not uncomfortable). Leg not tested should be placed behind the leg block.
- Start procedure by asking the patient to fully extend their leg SLOWLY and set as 0°.
- Set ROM to extension: 0° and flexion: 100° (you will manually let the patients' leg fall and stop once 100° ROM is achieved). Make sure mechanical end stops are set.
- Set up for gravity correction, place the subject's limb at 45° (same position that the weight was taken during the dynamic calibration) and press the lock button. Then press weigh to get a static weight of the limb.
- Explain that the first set is a warm up and familiarisation set. Five repetitions of extension and flexion to be performed, starting at 60% of max and increasing by 10% each repetition so last rep is max effort (increasing strength intensity each time by looking at the feedback on screen). Emphasise that the patient should not pause between efforts but to work continuously from rep to rep.
- During the rest period, explain that the next 2 sets are maximal extension and flexion, pushing and pulling as hard and fast they can against the shin pad, holding on to the handle

bars on the sides and keeping their back against the back rest. Give the patient verbal support during testing.

- Two max sets of concentric flexion and extension with 60 sec rest in between. Turn the screen away from the patient during the testing – you should be focusing on watching them and they should be focusing on putting maximum effort into each rep.
- Repeat procedure on involved leg.
- Print out two copies of the report. This contains the peak torque and work values obtained in each set for flexors and extensors. All values are also reported as a percentage of body mass and ratios/deficits are also given.
- Select the set to be included in the main report according to the decision tree below. Put a star on the left hand side of the page next to the chosen set.
- Give one copy to the patient to pass on to their physio/S&C and keep one for our records.
- Under normal circumstances, do not talk the patient through the results or express an opinion on their performance – tell them that the physio/S&C coach will go through the results with them. This is to avoid either repetition or conflict of information.
- Export the data for all research patients: Database → Export data → Export

7.2 Appendix B - Ethics

The research for this project was submitted for ethics consideration under the reference LSC 15/122 in the Department of Life Sciences and was approved by the University of Roehampton's Ethics Committee on March 25th 2015.

Participant consent forms and information booklets are presented below.

3D Biomechanics Assessment Consent Form

For 3D assessment and collection and subsequent use of data and video

PLEASE READ BEFORE COMPLETING FORM

Please ensure that all the following steps are completed before you are collected for your appointment:

1. All personal information is completed.
2. Each statement is read and the corresponding box is initialled if consent is given. If consent is not given, please leave the corresponding box blank.
3. Signature and date at the bottom of the form is completed.

NOTE: For patients under the age of 18, boxes need to be initialled by both the patient and guardian. A guardian's signature is also required at the bottom of this form.

Full Name:

Date of Birth: **Date of assessment:**

Email:

Telephone:

PLEASE READ THE STATEMENTS BELOW AND INITIAL BOX IF CONSENT IS GIVEN

Assessment: I consent to carry out physical testing in the movement testing laboratory and have read the attached information sheet. **Assessment** ☐

Collection: I give my permission for video and 3D data to be taken of me at the 3D Movement Analysis Lab, Sports Surgery Clinic, as part of my assessment today. **Collection** ☐

Health Records: I consent to the use of my health records related to my injury or condition to be used for research purposes. **Health records** ☐

Further Use: I give my permission for the data collected and video taken of me today at the 3D Biomechanics Assessment Lab, Sports Surgery Clinic, to be used:

1. For teaching purposes. **Teaching** ☐
2. In articles written by staff from the Sports Surgery Clinic for publication in Professional or Scientific journals or books, and conference or Laboratory posters. **Publication** ☐

Patient Signature: **Date:**

Guardian Signature: **Date:**

Countersigned for Laboratory: **Date:**

3D Biomechanical testing

It is important that we recreate sport specific and challenging movements in the assessment of your limitations, injury mechanisms and underlying movement deficiencies which have caused injury. These analyses are used to guide the rehabilitation process to get you back playing as soon as possible.

We will ask you to perform a maximal countermovement jump, a step down hop, a hurdle hop, maximal distance jump and cutting maneuvers, all designed to stress the body in sport specific manner. We may add in additional tests specific to your presentation but these will be explained to you. You will have the test demonstrated to you and be allowed to practice the test prior to testing. If you have any questions or concerns please raise them before completing the test. We repeat the tests 3 times on each leg to provide
comparison and maximize accuracy. We use an online web hosting site which will allow you secure access to the video files of your testing today.

What are the risks?

In any testing programme there is a risk of injury. Care has been taken in the design of the tests to apply no more load than required or encountered in normal training. If you feel pain, or cannot complete the testing, please tell your biomechanist who will cease the testing. There is always the risk in any training exercise of acute injury. While we minimize these risks as much as possible you should understand that injury can occur. Should injury occur, testing will be terminated, and first aid will be administered using the PRICE (protect, rest, ice, compress, elevate) principles. A Charted Physiotherapist at the SSC will then asses the injury and advise in its treatment.

Visual Recordings at the Sports Surgery Clinic Information

The Sports Surgery Clininbmbnc has adopted a policy that gives you the right to control the use of visual recordings taken of you. Please read the information below before completing the consent form and ask questions if you are unsure of anything. ***You may at any time during or after the recording withdraw your consent for any use other than that relating to your medical care.***

Who will see my recordings?

Healthcare staff involved in your care may have access to the recordings. This may include staff at the Sports Surgery Clinic, Dublin and any other centre where you receive treatment or consultation.

Sometimes staff may wish to use the recordings for teaching or in publications. You can choose whether or not this happens. All recordings will be treated as confidential and only made available to people unrelated to your medical care with your consent.

If I consent, how may my recordings be used for teaching?
Recordings are useful for teaching staff, clinicians and students about injury and about how to do movement assessments and understand the results. They may be used in presentations both on-screen and on paper.

Please be aware that if you withdraw your consent for this use later on it may not be possible to withdraw all of the recordings, or copies of, from use. For example, if hand-outs are provided. ghyjtgh

If I consent, how may my recordings be used in Professional or Scientific Publications?
Occasionally staff at the Sports Surgery Clinic may wish to use all or part of a recording in articles, books, talks and poster presentations. These are normally of interest to other healthcare providers and scientists but the information may be accessible to the public. Again, please be aware that if you withdraw your consent for this use later on it may not be possible to withdraw all of the recordings, or copies of, from use.

Will they be able to tell if it is me?
You can choose to have your face hidden in any photographs or videos used for teaching purposes or in publications. Please be aware that it still may be possible to identify you by your clothing.

Can I see my recordings?
Yes. You may ask to see your recordings if the opportunity does not arise during the session.

Can I get a copy of my recordings?
You may request a copy of your videos for you to keep. This will not include any clinical report.

Investigator Contact Details:

Name: Ciarán McFadden Department: Dept of Life Sciences University Address: Erasmus House, Roehampton Lane, London Postcode: SW15 5PU Email: mcfaddec@roehampton.ac.uk Telephone: +35315262030

Please note: if you have a concern about any aspect of your participation or any other queries please raise this with the investigator (or if the researcher is a student you can also contact the Director of Studies.) However, if you would like to contact an independent party please contact the Head of Department.

Director of Studies Contact Details: Name Siobhan Strike University Address: Whitelands College, Roehampton Lane, London Email : s.strike@roehampton.ac.uk Telephone: +44 (0)20 8392 3546

Head of Department Contact Details: Name: Dr. Caroline Ross University Address: Whitelands College, Roehampton Lane, London Email: c.ross@roehampton.ac.uk Telephone: +44 (0)20 8392 3529

7.3 Appendix C - Chapter 2 Supplementary Material

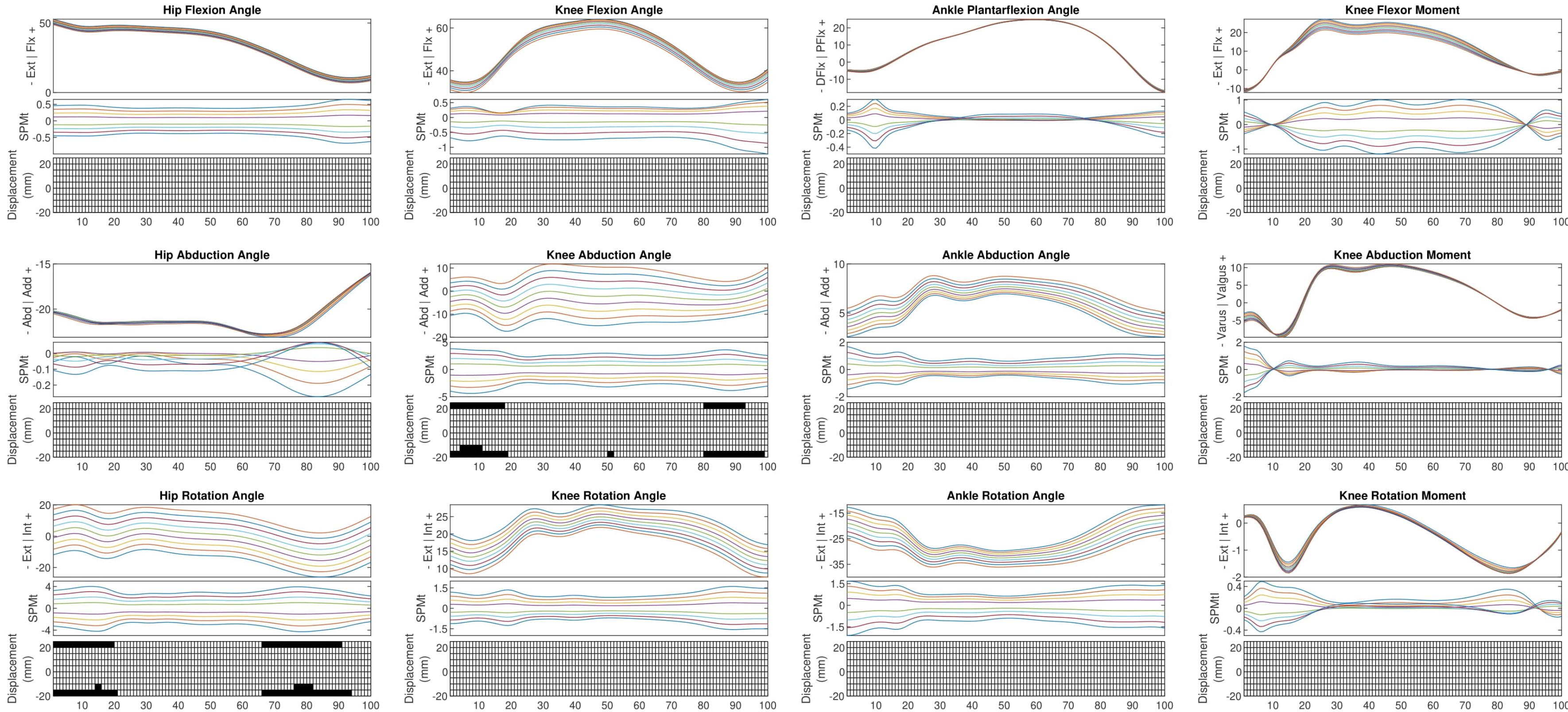


Figure 7.1: Sensitivity analysis for the THI marker at sample size of n = 10. Top panel of each graph depicts kinematic and kinetic signals under each THI marker displacement. The middle panel shows the SPMt curve produced following 1S independent samples SPM t-test between the unaltered condition and each of the displacement conditions. Lastly, the bottom panel depicts each SPMt as a function of both time and THI marker displacement. Image inference depicting phases of each signal (x-axis) significantly affected (black) by the corresponding marker displacement (y-axis).

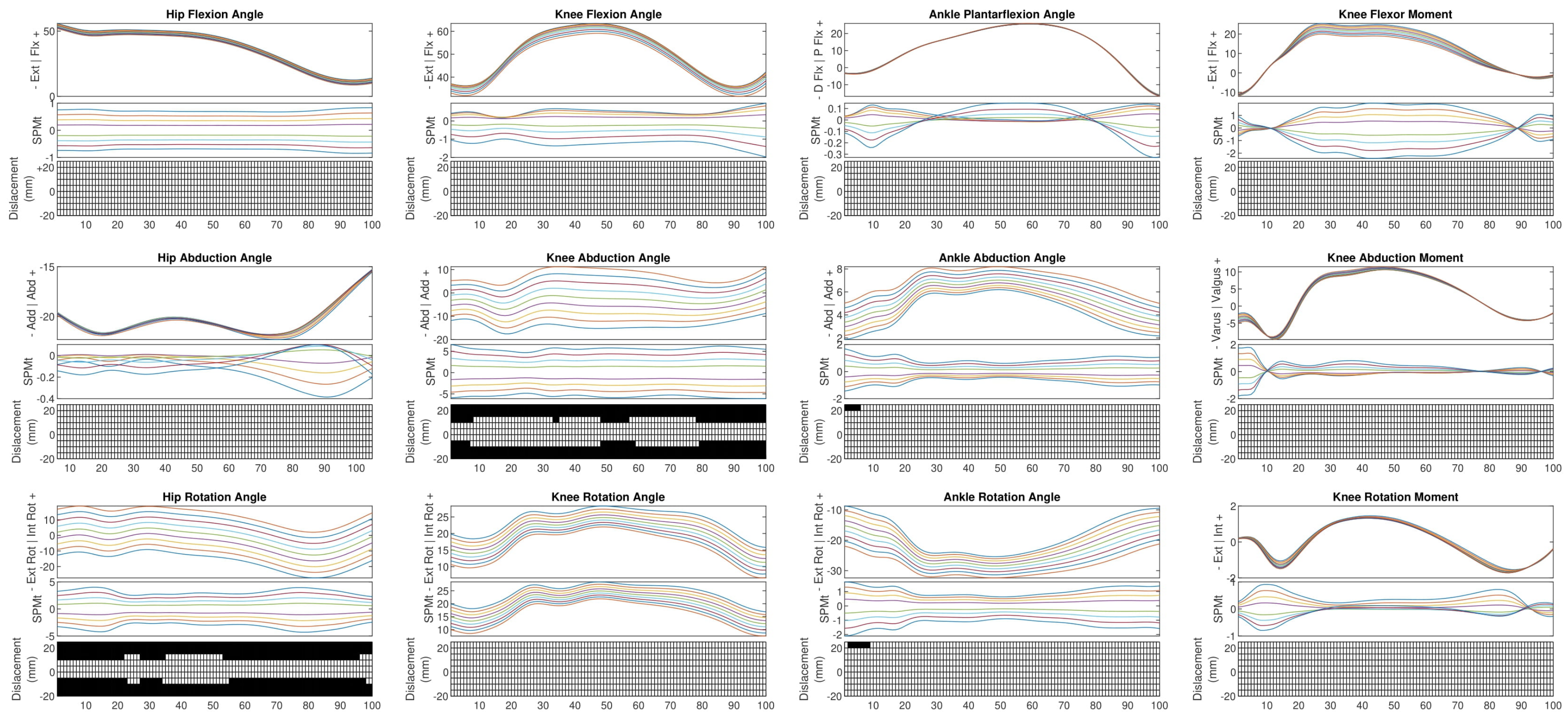


Figure 7.2: Sensitivity analysis for the THI marker at sample size of n = 25. Top panel of each graph depicts kinematic and kinetic signals under each THI marker displacement. The middle panel shows the SPMt curve produced following 1S independent samples SPM t-test between the unaltered condition and each of the displacement conditions. Lastly, the bottom panel depicts each SPMt as a function of both time and THI marker displacement. Image inference depicting phases of each signal (x-axis) significantly affected (black) by the corresponding marker displacement (y-axis).

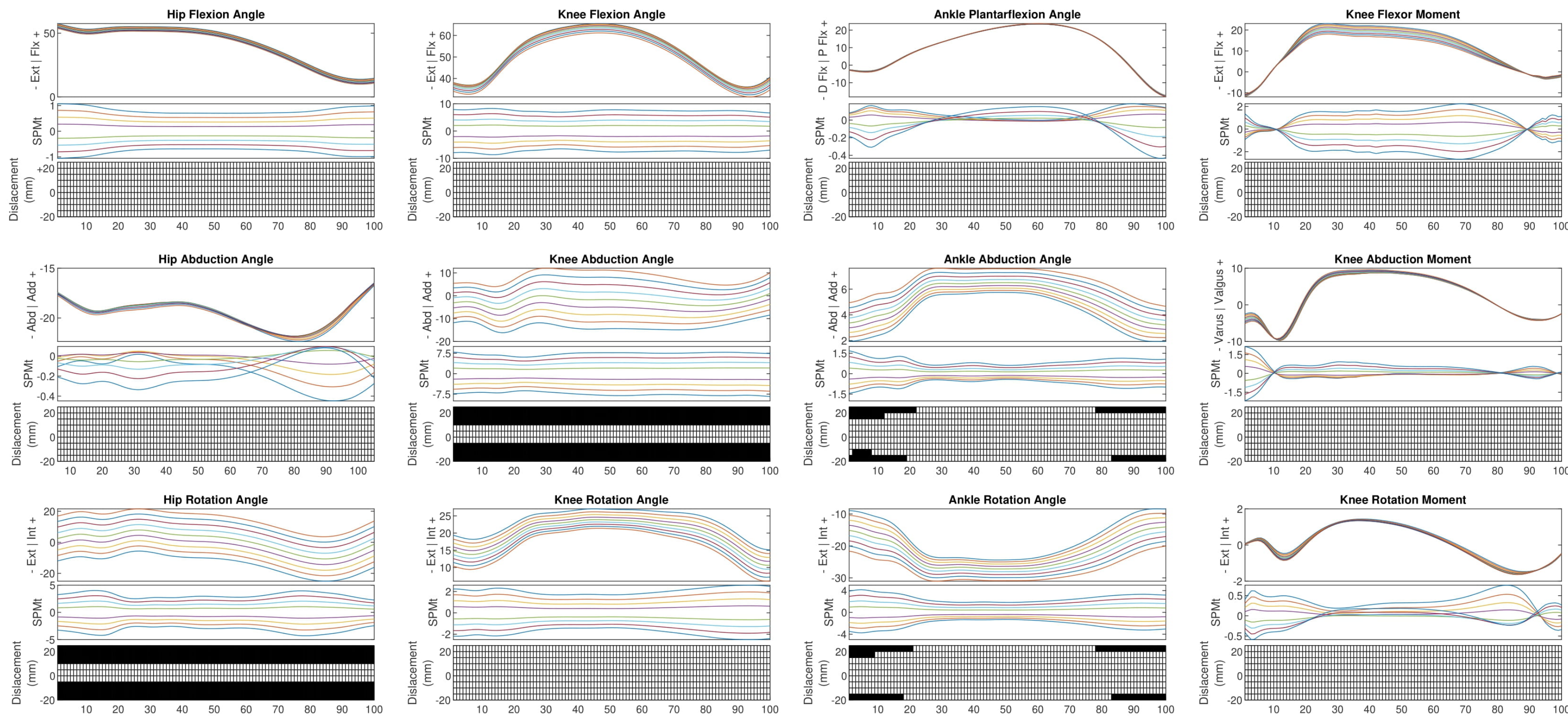


Figure 7.3: Sensitivity analysis for the THI marker at sample size of n = 50. Top panel of each graph depicts kinematic and kinetic signals under each THI marker displacement. The middle panel shows the SPMt curve produced following 1S independent samples SPM t-test between the unaltered condition and each of the displacement conditions. Lastly, the bottom panel depicts each SPMt as a function of both time and THI marker displacement. Image inference depicting phases of each signal (x-axis) significantly affected (black) by the corresponding marker displacement (y-axis).

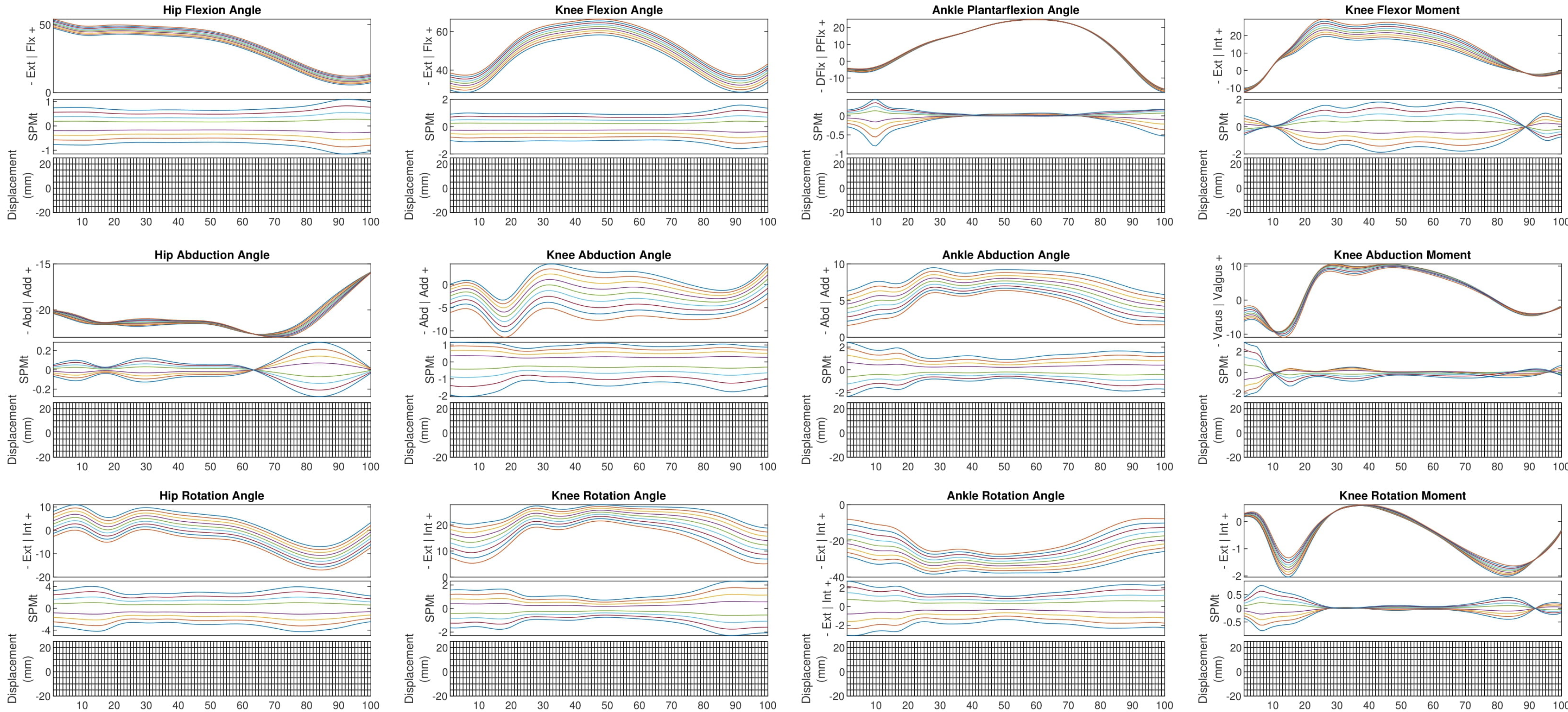


Figure 7.4: Sensitivity analysis for the KNEE marker at sample size of n = 10. Top panel of each graph depicts kinematic and kinetic signals under each KNEE marker displacement. The middle panel shows the SPMt curve produced following 1S independent samples SPM t-test between the unaltered condition and each of the displacement conditions. Lastly, the bottom panel depicts each SPMt as a function of both time and KNEE marker displacement. Image inference depicting phases of each signal (x-axis) significantly affected (black) by the corresponding marker displacement (y-axis).

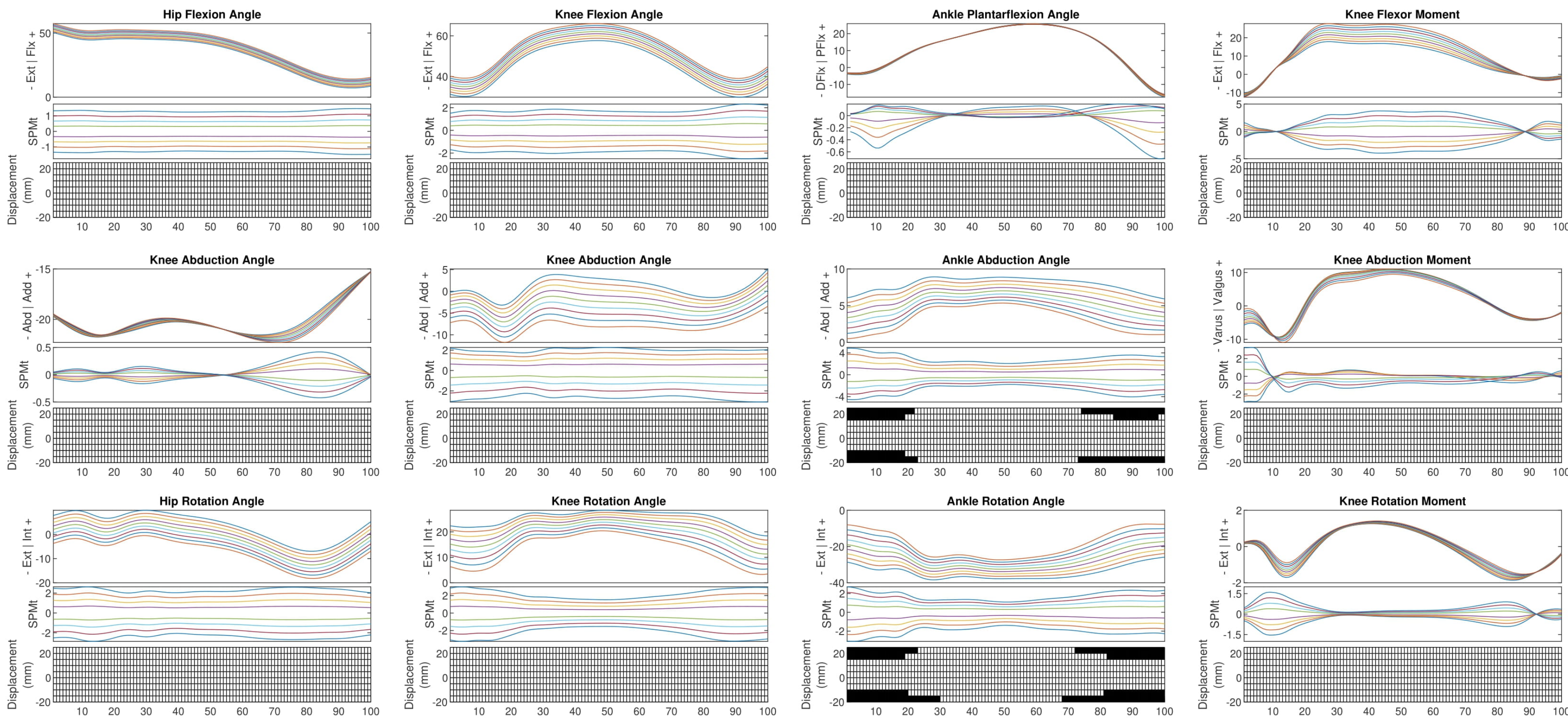


Figure 7.5: Sensitivity analysis for the KNEE marker at sample size of n = 25. Top panel of each graph depicts kinematic and kinetic signals under each KNEE marker displacement. The middle panel shows the SPMt curve produced following 1S independent samples SPM t-test between the unaltered condition and each of the displacement conditions. Lastly, the bottom panel depicts each SPMt as a function of both time and KNEE marker displacement. Image inference depicting phases of each signal (x-axis) significantly affected (black) by the corresponding marker displacement (y-axis).

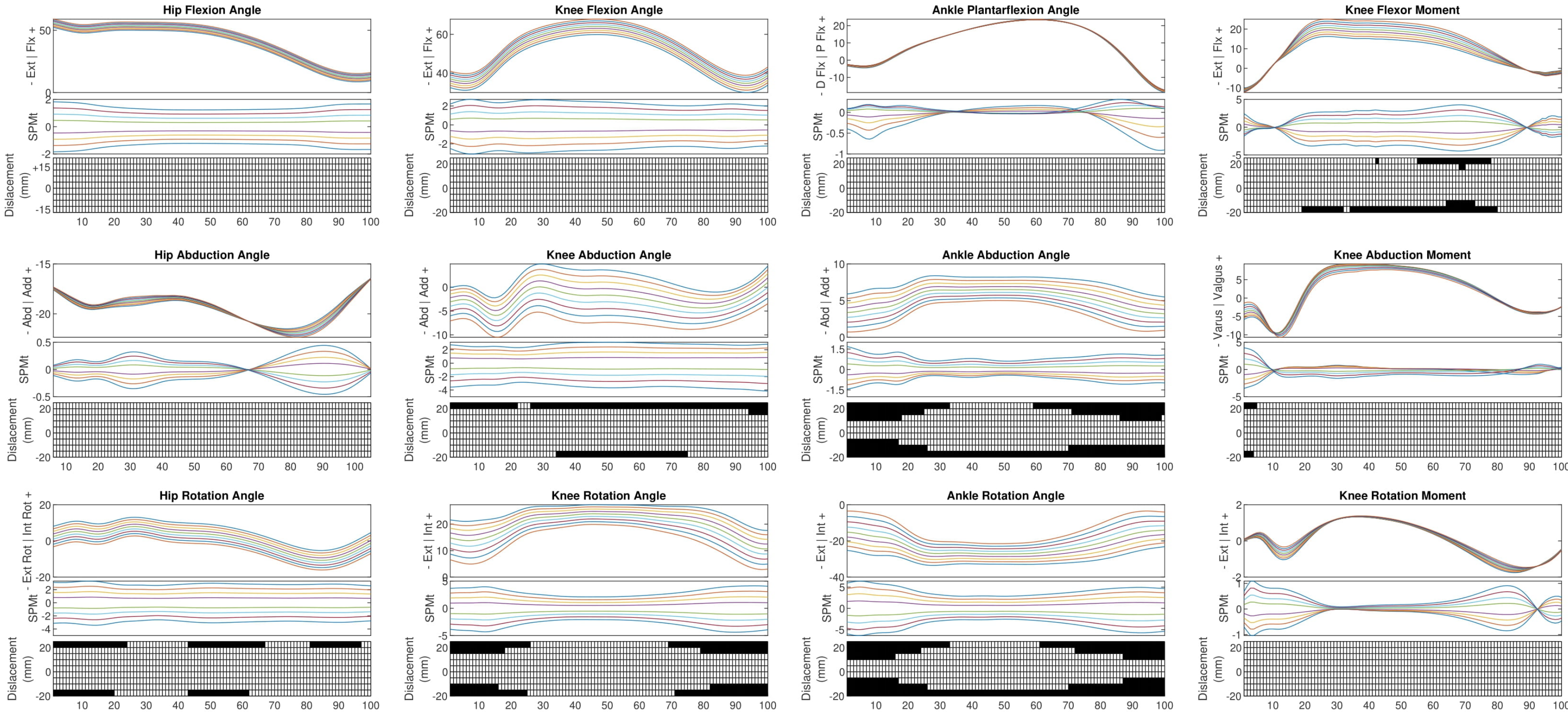


Figure 7.6: Sensitivity analysis for the KNEE marker at sample size of n = 50. Top panel of each graph depicts kinematic and kinetic signals under each KNEE marker displacement. The middle panel shows the SPMt curve produced following 1S independent samples SPM t-test between the unaltered condition and each of the displacement conditions. Lastly, the bottom panel depicts each SPMt as a function of both time and KNEE marker displacement. Image inference depicting phases of each signal (x-axis) significantly affected (black) by the corresponding marker displacement (y-axis).

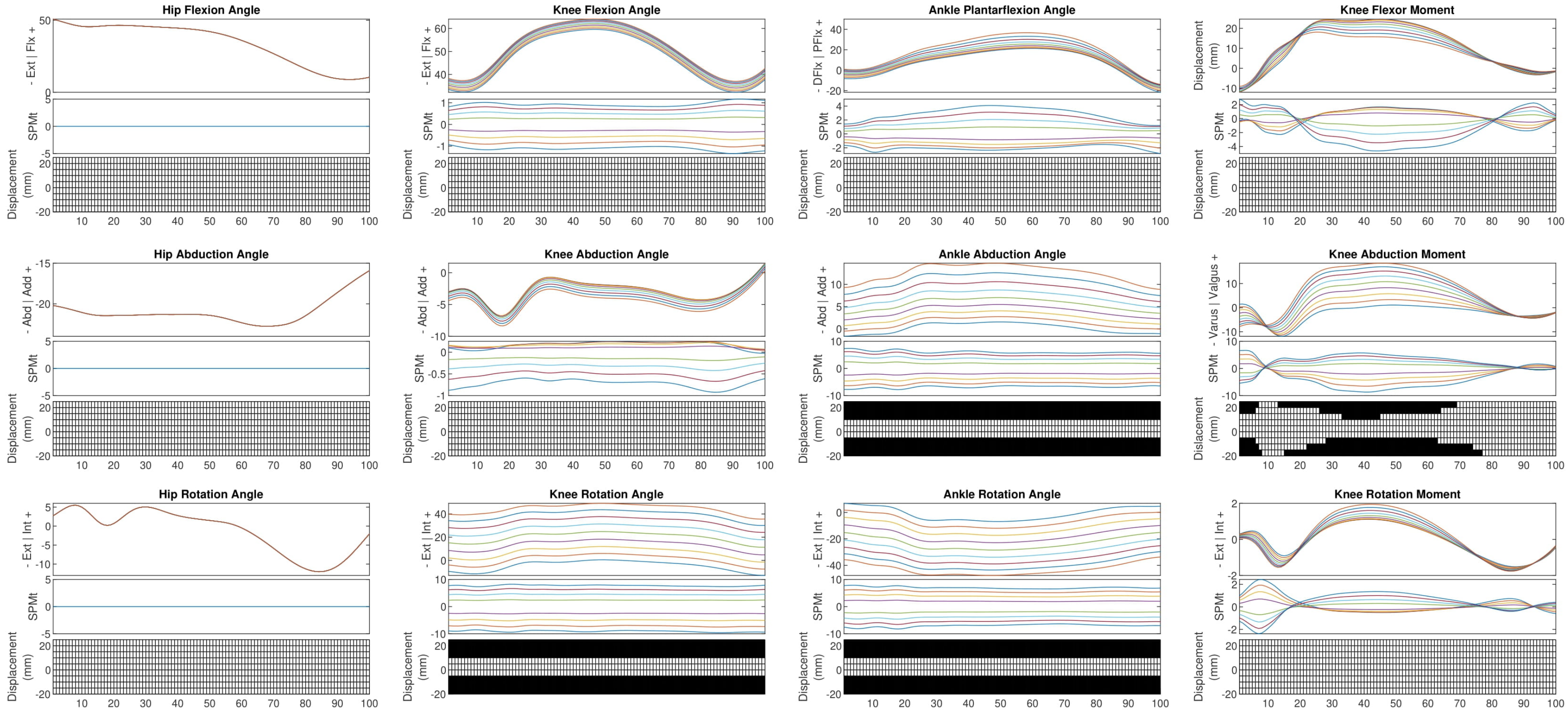


Figure 7.7: Sensitivity analysis for the TIB marker at sample size of n = 25. Top panel of each graph depicts kinematic and kinetic signals under each TIB marker displacement. The middle panel shows the SPMt curve produced following 1S independent samples SPM t-test between the unaltered condition and each of the displacement conditions. Lastly, the bottom panel depicts each SPMt as a function of both time and TIB marker displacement. Image inference depicting phases of each signal (x-axis) significantly affected (black) by the corresponding marker displacement (y-axis).

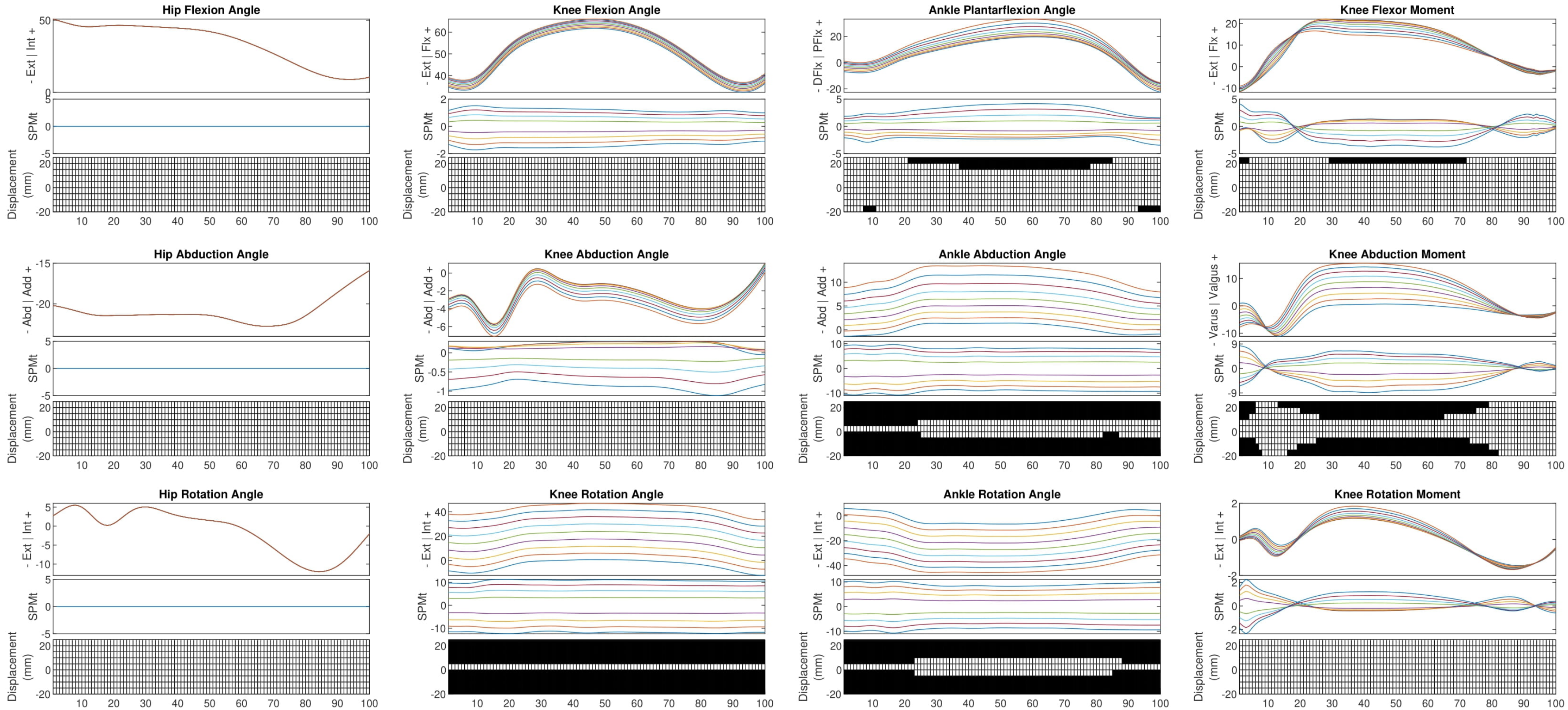


Figure 7.8: Sensitivity analysis for the TIB marker at sample size of n = 50. Top panel of each graph depicts kinematic and kinetic signals under each TIB marker displacement. The middle panel shows the SPMt curve produced following 1S independent samples SPM t-test between the unaltered condition and each of the displacement conditions. Lastly, the bottom panel depicts each SPMt as a function of both time and TIB marker displacement. Image inference depicting phases of each signal (x-axis) significantly affected (black) by the corresponding marker displacement (y-axis).

7.4 Appendix D - Chapter 3 Supplementary Material

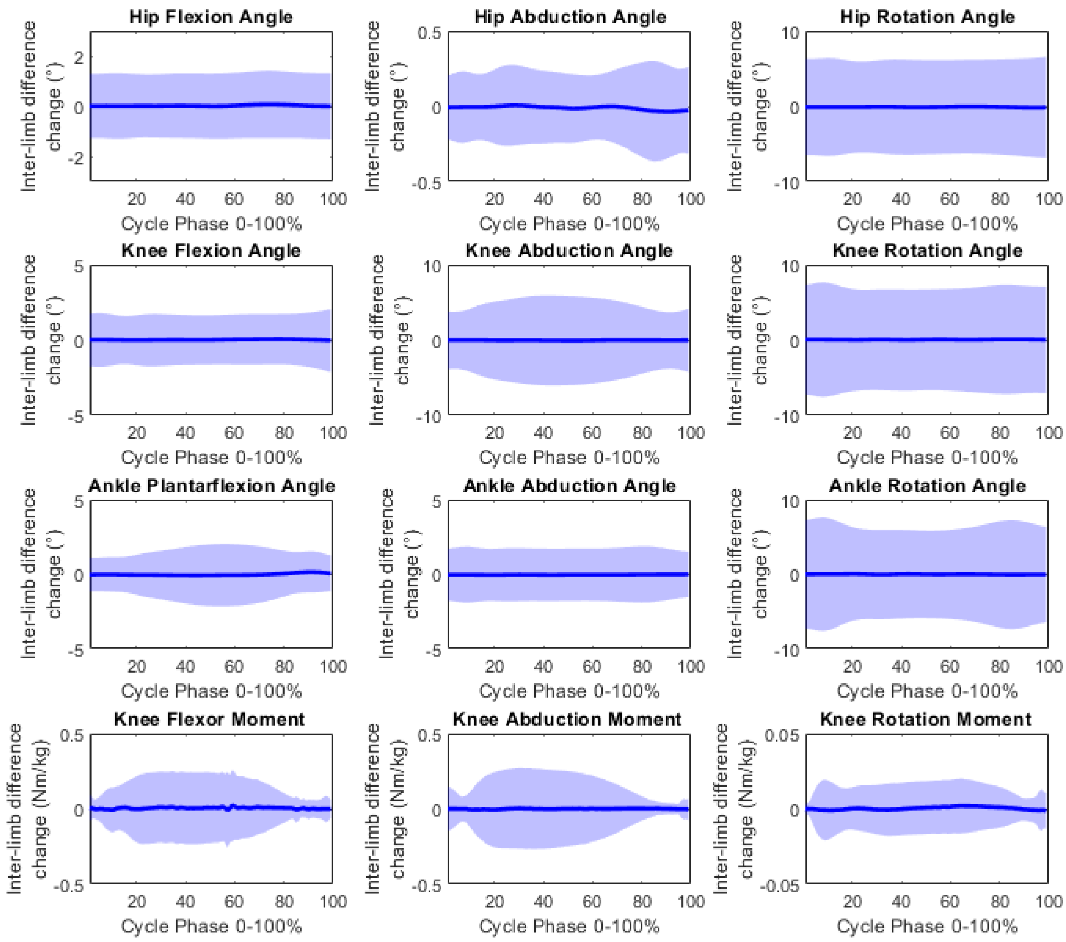


Figure 7.9: Change in inter-limb difference measures for kinematic and kinetic variables from marker placement error across the entire stance phase.

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