The role of biomechanical markers of dynamic stability in the execution of highly dynamic tasks

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Related research dissemination

Publications

Sankey, S.P., Malfait, B., Raja Azidin, R. M. F., Deschamps, K., Robinson, M.A., Staes, F., Vershueren, S., and Vanrenterghem, J. (2015). How reliable are knee mechanics in side-cutting? *Gait and Posture*. 41(4), 905-911. (**STUDY 1**)

Raja Azidin, R. M. F., **Sankey, S.P.,** Robinson, M.A. and Vanrenterghem, J. (2015). Effects of treadmill versus overground soccer match simulations on the biomechanical markers of anterior cruciate ligament injury risk in side cutting. *Journal of Sport Sciences*. 33(13): 1332-1341.

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Manuscripts in preparation

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List of abbreviations

ACL – Anterior Cruciate Ligament

A-P – anterior-posterior

M-L – medio-lateral

CoM – centre of mass

CoP – centre of pressure

GRF – ground reaction force

IAA – Induced Acceleration Analysis

KAM – knee abduction moment

DK – Direct Kinematic – modelling with 6 degrees of freedom at each joint

IK – Inverse Kinematic – modelling with reduced degrees of freedom

Thesis abstract

The primary aim in human locomotion is to control the body's centre of mass sufficiently to perform the task as safely and as efficiently as possible. Control of the centre of mass is likely to involve the interaction of several movement strategies each deployed for a specific role. When the task becomes more dynamic involving movement in multiple planes, the task becomes more difficult and the movement strategies need to adapt. If those movement strategies begin to fail, or involve dangerous deviations, perhaps due to degradation of the physical and neuromuscular mechanisms required to execute them, then control of the centre of mass and consequently *whole-body dynamic stability* is compromised. When whole-body dynamic stability is compromised, this may lead to dynamic stability issues at a joint level, which may be a precursor to undesirable joint moments and an increase in injury risk. That said, research has yet to provide a holistic account of whole-body dynamic stability for highly dynamic tasks. Therefore, it was the intention in this doctoral thesis to outline and explore the interplay between movement strategies that can contribute significantly to whole-body dynamic stability and mechanisms that may indicate potential injury risk.

In this research project, biomechanical observation of side cutting was utilised for its relevance with regards to sports performance and association with common lower limb injuries and even injury screening. Initially, study one focused on methodological concerns with the reliability of the kinetic and kinematic data typically derived from side cutting. Our findings identified new insights into variability of kinematic and kinetic data in a detailed view across phases of ground contact. In study two we developed a novel, holistic approach to quantify the movement strategies that contribute to control of the centre of mass, or *whole-body dynamic stability*, in side cutting. This approach has allowed us to express original insights into the key mechanisms for medial acceleration of the centre of mass; the extent of

destabilisation that excessive ground reaction forces can generate; and the interaction of key movement strategies adopted to correct for destabilisation and retrieve control. Furthermore, in studies three and four we have been able to demonstrate the robustness of our measurement of movement strategies in quantifying responses to increasingly challenging scenarios. In addition, the final two studies allowed us to highlight the need for adaptability in movement strategies between tasks of varied complexity, and the transition between movement strategies within the side cutting task itself. Overall, our findings may provide valuable information for the performer and supporting practitioners to develop training strategies based on biomechanical markers of whole-body dynamic stability, which may preclude negative injurious consequences.

General Introduction

Background of the research

Stability can be defined as the capacity to control or minimise unwanted or unnecessary movements, where a more stable system exhibits fewer unnecessary movements and is able to tolerate the challenges of the task in hand. In static standing, vertical support of the centre of mass (CoM) is typically the primary aim, and minimal horizontal movement of the CoM may be part of static stability control. Whereas, in dynamic tasks, acceleration of the CoM becomes necessary in one plane, or multiple planes at the same time. Therefore, dynamic stability may be defined as the capacity to control and minimise unnecessary movements, whilst tolerating the changing demands of necessary CoM accelerations in a dynamic task (Patla, Adkin and Ballard, 1999; van Emmerik et al., 2016). In dynamic tasks, like walking, jumping or changing direction, generating forces to bring about such acceleration is essential. However, the first challenge, and one of the primary objectives of this thesis, is to outline what movement strategies are initially necessary to accelerate the CoM in the intended direction of travel and to mitigate unnecessary deviations. It is important to note, whilst it is clear that sensory systems and passive controllers may contribute to static and dynamic stability, this thesis is concerned with biomechanical factors that may help quantify those important movement strategies.

Some insights towards potential movement strategies at play in dynamic tasks exist in the research on balance in standing, walking and turning during locomotion (MacKinnon and Winter, 1993; Kuo, 1995; Winter, 1995; Patla, Adkin and Ballard, 1999; Blenkinsop, Pain and Hiley, 2017). However, there is scope for further exploration of the roles they fulfil, and research is limited on faster highly dynamic tasks (Houck, Duncan and De Haven, 2006). In standing, it has been reported that the primary movement strategies for static stability control come from the hip and ankle joints and are specifically adopted to adjust the origin of the GRF vector (CoP) relative to the CoM, often considering its velocity, to maintain stability (Winter, 1995; Hof et al., 2005; van Emmerik et al., 2016; Blenkinsop, Pain and Hiley, 2017). It is important to note that research suggests the body may typically engage one control strategy at a time, with larger stability corrections from the hip joint and smaller ones from the ankle, depending on the direction of movement required (Kuo, 1995; Winter, 1995; Blenkinsop, Pain and Hiley, 2017). In walking, an additional movement strategy that precedes all others is where to place the foot (MacKinnon and Winter, 1993; Patla, Adkin and Ballard, 1999). Thereafter, it appears that the hip and ankle movement strategies retain their important roles - now in dynamic stability control - helping to achieve an upright posture whilst maintaining economy of movement (van Emmerik et al., 2016). Existing research suggests this relationship is also observed in turning (Patla, Adkin and Ballard, 1999) and faster changes of direction (Houck, Duncan and De Haven, 2006), where the hip and ankle appear to work as a double inverted pendulum. Furthermore, research suggests that when such a change of direction is necessary, medio-lateral control of the CoM becomes the main priority (Patla, Adkin and Ballard, 1999). Thus, over the duration it takes to complete a dynamic task involving a change of direction, it is likely that movement strategies serve multiple roles, and the priorities are likely to be direct acceleration of the CoM and control of the movement deviations in the medio-lateral direction.

Dynamic tasks that involve fast progressive movements with medio-lateral force generation are essential to the success of performance in sports, particularly those that involve competing with opponents in match-play. If the performer cannot adequately control and tolerate often repetitive demands of dynamic tasks, their performance may fail, or they may risk potential injury. Appropriately directed biomechanical observation has the unique opportunity to quantify mechanisms that may increase injury risk, and also quantify movement strategies that may precede those mechanisms. Control of the CoM is challenging but imperative for efficient execution of dynamic tasks, and is likely to involve deployment of a series of movement strategies that may change within phases of the task. Knowledge of the interplay of the roles of important movement strategies would help inform the specificity of training intervention strategies. Furthermore, the extent to which one must deploy movement strategies to manage unnecessary deviations, rather than medial acceleration of the CoM, for example, may indicate the condition of *whole-body dynamic stability* in the individual. That is to say, an individual with better whole-body dynamic stability may demonstrate a greater economy of movement, with less need for corrective adjustment, and fewer undesirable joint dynamic stability issues as a result.

Over the course of the following literature review, firstly, an overview is presented of the well-established role of the side cutting task as a screening tool for anterior cruciate ligament (ACL). Subsequently, the focus moves to some of the current gaps in understanding of technique or postural adjustment, followed by development toward a holistic account of *whole-body dynamic stability* and the benefits that this approach may offer. Experimental reliability and time-series analysis is then briefly outlined, followed by a review of two specifically challenging scenarios relevant to the current body of side cutting research.

Literature Review

Side cutting as a dynamic task and a screening tool

Non-contact injury risk in the lower limb in sports that involve dynamic tasks, like football (soccer), remains a significant performance and financial burden to sport (Hawkins et al., 2001; Woods, 2004; Padua et al., 2018). In case series video analyses of anterior cruciate ligament (ACL) injuries in professional football, basketball and American football, 64%, 72% and 72.5% of ruptures were non-contact, respectively (Krosshaug et al., 2007; Waldén et al., 2015; Johnston et al., 2018). Attempts to reduce the incidence of ACL injury, have yet to make a significant impact, but, exploration of the potential injury mechanisms including dynamic control of the knee joint (dynamic joint stability) are well documented (McLean et al., 2004; McLean et al., 2005; Dempsey et al., 2007; Dempsey et al., 2009; Myers and Hawkins 2010; Hashemi et al., 2011; Sigward et al., 2012). The ACL injury, in particular, is a debilitating injury that usually results in a significant amount of time out of sport or physical activity, subsequently the challenges in returning to full fitness are substantial, and then chances of re-injury remain elevated for some time (Waldén et al., 2015; Johnston et al., 2018). Over the last 15-20 years the side cutting task that has emerged as the most prevalent tool in dynamic screening, based on the demand for medial force generation and control, inherent in the epidemiology and mechanisms of ACL injury, and the challenge it poses for dynamic knee joint stability.

Side cutting, also occasionally referred to as sidestepping (see figure LR1 below), is a dynamic task that involves single-leg ground contact and support, in addition to substantial multi-planar loading and movement (McLean et al., 2005; Donnelly et al., 2012; Xie et al., 2012). The turn or contact step involved in side cutting manoeuvres can be divided into three phases: the deceleration; the change of direction; and the acceleration. The deceleration is more commonly referred to as the *weight acceptance phase*; which Besier et al. (2001a) defined as the initial contact to the first 'trough' in the vertical ground reaction force (GRF). It is in this weight acceptance phase, or the first phase immediately following initial ground contact, that has been reported to be the most common instance for ACL injury (Krosshaug et al., 2007; Waldén et al., 2015; Johnston et al., 2018) and the most detrimental for known injury mechanisms (Houck and Yack, 2003; Hashemi et al., 2011; Sigward et al., 2012; Xie et al., 2012). Perhaps this is not surprising considering the rate and magnitude that GRFs can reach in the early phase of such sports related movements (Hewett et al., 2005; Kristianslund et al., 2014). Regarding the specific mechanisms, earlier in vitro studies suggested significant ACL load was observed with higher peak knee abduction moment, and the knee joint close to full extension (Seering et al., 1980; Markolf et al., 1995). Indeed, in specific side cutting research it is those injury mechanisms that are most prevalent in the weight acceptance phase, typically in conjunction with internal tibial rotation (McLean et al., 2004; McLean et al., 2005; Chaudhari and Andriacchi, 2006). Thus, it is clear the side cutting task may offer a useful biomechanical opportunity to explore the ACL injury research paradigm. However, the key issue may be that mechanisms that cause dangerous stress on the ACL in side cutting, like high peak knee abduction moment, are themselves likely to be a consequence, perhaps of deviations that occur in the body when attempting to control the knee joint or whole-body CoM.

Figure LR.1. Photo sequence of the phases of the side cutting task. From the left the images show: the approach, the turn, and the exit.

Current issues in posture or technique adjustment

In executing dynamic tasks, such as side cutting, posture or technique adjustments may provide the most accessible opportunities to improve performance, and more importantly, reduce injury risk. The impact of controlled technique refinements have been explored by several research groups specifically for side cutting (Dempsey et al., 2007; 2009; Donnelly et al., 2012; Kristianslund et al., 2014). However, it is difficult to determine what the consequences for control of the CoM may be if the results are not expressed in terms of the position or influence on the CoM. Donnelly et al. (2012) conducted a study with male amateur football players where planned and unplanned side cuts were performed. With predictive simulation analyses, they found that upper and lower body kinematic changes could be made and that would bring about a reduced peak knee abduction moment. They reported that redirecting the CoM medially and towards the direction of travel would achieve this aim, and therefore reduce potentially injurious knee loading. However, little is known about how such CoM control may be achieved in practice, and whether this would be possible in more challenging scenarios.

Studies have reported specific technique adjustments that influence characteristics of GRF, which may be a useful approach to understand control of the CoM (Kristianslund et al., 2014; Havens and Sigward, 2015a; 2015b; 2015c). Specifically, one study reported GRF magnitude, and the moment arm of that vector to the knee joint, showing the consequences for undesirable knee abduction moments (Kristianslund et al., 2014). In this case, the authors suggest that technical adjustments that reduce the moment arm of the GRF vector to the knee joint, such as stance width and knee valgus motion, are probably more important than reducing the magnitude of the same vector, when the aim is to reduce peak knee abduction moments. However, of course, technical adjustments that work for one characteristic of the GRF vector are likely to influence other characteristics. So, it would be important to represent GRF vector characteristics in an integrated manner, if possible, but this is not always clearly identified in the literature. Further research has expressed similar GRF characteristics, including impulses and a whole-body moment arm length through the separation distance between the CoM and the centre of pressure (CoP) (Havens and Sigward, 2015a; 2015b; 2015c). In their most recent study, they reported that only the separation distance between the CoM and CoP predicted peak knee abduction moment for 45° side cutting, but again, although this wasn't their main aim, the link between other GRF characteristics was unclear. In much of the relevant side cutting research, the aim seems to be to identify attributes that are detrimental to injury risk mechanisms, and recommend that they are reduced. Instead, it may be more important to identify why unfavourable movement strategies or technical adjustments may occur in the first place, and it is reasonable to suggest answers may reside with a more holistic account of *whole-body dynamic stability*. However, to the best of our knowledge, there has not yet been a direct quantification, even in preliminary analyses, of whole-body dynamic stability for side cutting, or the movement strategies that are required, whilst retaining the ability to quantify the implications within the injury and performance paradigms.

Whole-body dynamic stability

We propose that quantifying of the interplay between movement strategies used to control the CoM in dynamic tasks represents the status of whole-body dynamic stability, and fits as a working definition of the term. However, quantifying whole-body dynamic stability, specifically in side cutting, is certainly challenging as the task demands control in transition of the body to a new direction, with pronounced multi-planar loading, over a single foot contact lasting ~0.250s. Thus, due to the nature of the task, it is likely the performer will need to deploy several movement strategies, perhaps involving several mechanisms, to achieve whole-body dynamic stability. Even then, the observer may need some concession that several corrective strategies, with desirable or undesirable performance and injurious consequences, may be required within the task performances. Recently, David et al. (2017) have explored the idea of integrating single mechanisms of loading into defined whole-body movement strategies for side cutting, reporting that certain strategies may present undesirable loading for ACL injury risk. Specifically, their study showed that a movement strategy including rear-foot strike, with less body pre-orientation in the new direction, and greater rate of knee flexion may result in higher knee valgus moments. However, it is still not clear what this means for how GRFs were deployed to control of the CoM, and why differentiating between three movement strategy categories was necessary in the first place.

To investigate the role of whole-body dynamic stability in movement strategies we may consider the previous work by Hof and colleagues on (quasi) static stability (Hof, Gazendam, and Sinke, 2005; Hof, 2005; 2007; 2008). The authors presented a mechanical approach to the problem, outlining three additive mechanisms related to the control of the CoM: (1st) the movement of the CoP in relation to the projected location and velocity of the centre of mass; (2nd) counter-rotation of segments; and (3rd) application of an external force. It was suggested that these three mechanisms represent three (quasi) static stability mechanisms that may also be active in dynamic tasks (Hof, 2007). Indeed, Hof, Gazendam and Sinke (2005) suggested that these mechanisms could be applied to starting, stopping and turning, which are components of landing and side cutting tasks. It seems, if we are to develop a robust observation of side cutting performance, this additive and perhaps sequential approach may be useful. Furthermore, for side cutting in particular, we must focus on the acceleration of the CoM in the new direction of travel, which is, of course, substantially explained by the medio-lateral (M-L) component of the GRF vector. Whilst exploring the M-L characteristics of the vector, we can then express how this movement is achieved, and what movement strategies may be beneficial for performance, or, indeed, lead to undesirable joint loading.

Acceleration of the CoM is intimately linked to the GRFs that are generated during a given task, so characterisation of those forces may provide an opportunity to quantify whole-body dynamic stability. In characterisation of the M-L component of the GRF vector, firstly, we can consider the point of application, which can be represented, in part, by the initial foot placement. Typically, foot placement, or stance width, has been determined by the position of the foot, or CoP within the foot, in relation to the instantaneous position of the CoM (Dempsey et al., 2009; Kristianslund et al., 2014; Havens and Sigward, 2015a; 2015b; 2015c). However, in a task that involves translation of the CoM, research has suggested it may be important to consider where the CoM is heading, represented by also taking into account the dynamic nature of the CoM by factoring in its velocity (Hof, Gazendam and Sinke, 2005; Havens, Mukherjee and Finley, 2018). Furthermore, whilst foot placement may be fixed at initial contact, CoP position within the boundaries of the borders of the planted foot can change over contact time, so represents a separate characteristic of the origin of the GRF vector. Finally, we can attempt to evaluate the magnitude of this M-L component of the GRF vector, and for quantifying whole-body dynamic stability, attempt to reveal the mechanisms by which it is achieved. Whilst these last mechanisms are rather elusive and difficult to quantify independently, the relatively novel approach of Induced Acceleration Analysis (IAA) may offer an opportunity to interrogate the presence of certain mechanisms and even the amount of contribution. In IAA, as reported in Kepple, Siegel and Stanhope (1997), and applied more recently in João et al. (2014) and Moniz-Pereira et al. (2018), the relative contribution of each lower limb joint moment to the individual components of the GRF can be expressed. More specifically, they can be expressed in selected planes, where the components represent the vertical, anterior-posterior (A-P), and - the priority here – M-

L acceleration of the body's CoM. Thus, IAA may feasibly allow one to differentiate between joint movements that are desirable or undesirable in relation to performance (adding to the magnitude of M-L acceleration) and potentially injurious loading (adding to knee joint loading). Furthermore, this approach retains one's capacity to observe sources of variability that may be key in adaptable deployment of movement strategies to achieve whole-body dynamic stability in side cutting.

Experimental reliability and time-series analysis

To replicate and explore such task execution in the laboratory initially requires confidence in the observation techniques adopted and specific awareness of their reliability. Systematic approaches have been developed to evaluate intra and inter-researcher reliability of biomechanical observations, albeit outside of side cutting, that may be easily applicable to more dynamic tasks (Schwartz, Trost, and Wervey, 2004; Queen, Gross, and Liu, 2006; Ferrari, Cutti, and Cappello, 2010; Deschamps et al., 2012). Specifically, in exploration of the reliability of kinematic data in gait analysis, Schwartz, Trost and Wervey, (2004) calculated hip, knee and ankle variability from signal deviations in inter-trial, inter-session, and inter-researcher perspectives. This approach allowed for quantification of the reliability of data when collected over a large number of trials, in different testing sessions, and with different researchers - all of which are important factors for most biomechanical laboratory settings. In addition, the method proposed by Schwartz, Trost and Wervey, (2004) can be used to quantify how the variability changes over the task time-series, in their case, the gait cycle. As mentioned previously, the weight acceptance phase of side cutting is of specific interest for ACL injury, so the opportunity to quantify variability during this phase is particularly attractive. This is a potential advancement of existing methods where, reliability is reported for the collapsed time-series in common approaches such as average intra-class correlation coefficients (Houck, Duncan and De Haven, 2006; Ford et al., 2005), coefficients of multiple correlations (Sigward and Powers, 2006a; Sigward and Powers, 2006b), and coefficients of multiple determinations (Besier et al., 2001a; 2003).

Typically, similar to the investigation of reliability, the majority of side cutting research to date has also focused on analyses of variables as metrics, perhaps represented as a collapsed time-series view of a phase of the task. However, this is likely to be insufficient especially if we are concerned with additive or sequentially deployed movement strategies to control the CoM that may vary over the course of ground contact. Statistical Parametric Mapping (SPM) is a method for conducting statistical analyses on 1D continua that is possible with smoothed and event or temporal bound data that is inherent in biomechanical observation (Pataky, 2010). SPM is now well-established in time-series biomechanical analyses, and is particularly important when there is no specific grounds for hypotheses that are restricted to single instances or events within a given task (Pataky, Vanrenterghem and Robinson, 2015). Analysis of side cutting using SPM has recently been reported in expression of the different movement strategies over the weight acceptance phase (David et al., 2017), and to investigate the effects anticipation and physical exertion on trunk and lower limb biomechanics (Whyte et al., 2018). When exploring control of the CoM in side cutting increasingly challenging scenarios, it is possible that the movement strategies important to whole-body dynamic stability will respond in different ways over ground contact time, however, the extent of which is not yet know.

Challenging side cutting scenarios - anticipation

Within sports performance one frequent challenging for dynamic side cutting may be based on how much time the individuals have to react. The timing of the stimulus that triggers the execution of a dynamic change of direction may require adjustment or additional movement strategies to retain control of the CoM and complete the task successfully. However, if deployed inadequately the deficiencies in those movement strategies may have serious consequences for injury risk. Existing research on the comparison between anticipated and unanticipated side cutting has identified significant increases in frontal and transverse plane moments (Besier et al., 2001b), and increased hip abduction angles (4.0-7.6°) with unanticipated stimulus (Houck, Duncan and De Haven, 2006). Both studies considered foot placement to be a contributory factor to their observations, however, this was not measured and reported directly. More recent research by Weinhandl et al. (2013) identified that ACL loading significantly increased in unanticipated side cutting conditions compared to anticipated conditions. The authors used musculoskeletal modelling to estimate ACL load and were particularly clear that they attribute the loading to a combination of sagittal plane shear forces with frontal and transverse plane knee moments. A recent systematic review and meta-analysis was conducted on anticipated and unanticipated side cutting knee mechanics (Brown, Brughelli and Hume, 2014). The authors reported, overall, that higher knee joint angles and moments were found when the side cutting task stimulus was unanticipated, potentially increasing the load on the ACL, and therefore increasing the chance of injury. Brown, Brughelli and Hume (2014) also reported consistency in the kinematic and kinetic results between observations made in laboratory and field settings. The authors attributed their observations to temporal restrictions for postural adjustment, or whole-body dynamic stability, in the unanticipated side cutting condition, compared to the anticipated. The level of complexity of the unanticipated stimulus, may be dependent on both the available reaction time and the nature of the visual stimulus (Lee et al., 2013; Mornieux et al., 2014; Lee et al., 2018; Weir et al., 2019). Understanding the effect on movement strategies associated to whole-body dynamic stability, may provide a valuable insight into tasks that are increasingly more challenging.

Challenging side cutting scenarios - prolonged physical exertion

Athletes are required to perform the same intensity of side cutting task when they are fresh in the early stages of a game, or perhaps fatigued in the later stages of the game, or game phase (e.g. halves in soccer) in response to the prolonged nature of physical exertion. This poses a significant problem, as the same biomechanical stress, in scenarios where athletes are in neuromuscular fatigue, will increase the likelihood of onset of injury. Research from Tsai et al. (2009) showed that exertion resulted in clear changes in knee mechanics, particularly external knee valgus moments and peak knee internal rotational angles – which are known mechanisms of ACL injury. Caution must be applied as ‘fatigue’ as a term is often misused, which can lead to confusion of central, peripheral and mechanical ‘fatigue’. Instead, notification of the specific exercise exertion is probably more appropriate. In that regard, soccer match simulation has received considerable attention in the literature (Small, McNaughton, Greig, and Lovell, 2010). More recently, a modified version of this soccer match simulation protocol was adapted for assessing the implications for known injury risk mechanisms (Raja Azidin et al., 2015). Whilst other exertion protocols have been recently developed (Khalid et al., 2015; McGovern et al., 2015; Collins et al., 2016; Whyte et al., 2018), to the best of our knowledge, only one has reported negative consequences specifically for peak knee abduction moment (Tsai et al., 2009). This may suggest that for the majority of the exertion-induced studies, other than that of Tsai and colleagues, the movement strategies that were adopted, were successful at mitigating risky biomechanical loads, however, this has not yet been addressed specifically. Thus, understanding the response of whole-body dynamic stability movement strategies to anticipation and exertion, with or without negative consequences, may provide valuable information towards injury and performance screening.

Summary

The prevalence of potentially debilitating lower limb injuries like that of the anterior cruciate ligament is a very real concern, especially in non-contact circumstances where the mechanisms of the injury may be less obvious. The aim of a screening task like side cutting is to provide valuable and reliable information that can be utilised as part of training to prevent such injuries from happening or as part of progressive rehabilitation programmes to allow those individuals to return to play. That said, occasionally the information gained form injury screening can be difficult to interpret and develop into meaningful strategies. Establishing a holistic approach to quantifying whole-body dynamic stability may offer unique opportunities to address many of the common concerns with balancing the injury and performance paradigms, as control of the centre of mass will be the number one priority. If sufficiently robust, a holistic approach to quantifying whole-body dynamic stability should have application beyond the task and performance constraints set out in the present research project, or more broadly, beyond side cutting and anterior cruciate ligament injury risk research altogether.

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Objectives

The general objective of this doctoral thesis was to develop a reliable and robust observation of the mechanisms and movement strategies to control the centre of mass, which represent the status of whole-body dynamic stability, whilst exploring the implications for markers of injury risk and performance.

Initially, in **Study 1** the aim was to quantify the reliability of data derived from common dynamic task assessment relevant for injury screening, in this case focusing on knee joint kinetics and kinematics. Attention was given to the recent approaches to side cutting investigation from our own research group, which are typically consistent with other researchers in this field. Briefly, the approach involved calculation of signal variability for all trials (trial-to-trial), between-session (per observer), and between-observer, with a ratio expressed comparing the trial-to-trial and between-observer variability as the primary reliability measure. Furthermore, the investigation also considered how reliability and variability may differ between kinematic and kinetic data with different modelling constraints.

In **Study 2** the aim was to quantify movement strategies that represent whole-body dynamic stability, and explore the association with potentially injurious mechanics and side cutting performance. Based on the high medial acceleration of the CoM that is required in side cutting, mechanisms that represent characteristics of the medio-lateral ground reaction force vector were proposed to represent sub-components of whole-body dynamic stability. Medio-lateral foot placement and centre of pressure position were used to identify the point of application, whilst three further mechanisms were used to represent planar contribution to the magnitude of the ground reaction force vector: sagittal triple acceleration (combined hip, knee and ankle); frontal plane hip acceleration; and transverse plane hip acceleration. Each mechanism was regressed against injury risk, represented by peak knee abduction moment, and performance, represented by change of direction angle and average medial acceleration of the centre of mass.

The aims in **Study 3** were to quantify the anticipatory effects on movement strategies and their adaptability toward whole-body dynamic stability. A pre-planned task provides the ideal opportunity for the individual to organise their movement strategies, and should mean whole-body dynamic stability and measure of injury risk is 'optimal'. An unanticipated task may ideally involve, firstly, similar injury risk or performance outcome, and secondly, similar movement strategy to anticipated side cutting despite the challenges to reaction time. However, any differences found may represent reduced whole-body dynamic stability and the consequences for injury risk and performance. The specific mechanisms described in the previous study were compared between two side cutting conditions: one where the participants knew in advance which direction they should turn; and one where the stimulus provided only 0.5-0.65 seconds for them to react before the turn.

The aims of the final study, **Study 4**, were to investigate the effects of a 90-minute bout of soccer-specific exertion on whole-body dynamic stability in unanticipated side cutting, whilst exploring the implications for injury risk and aspects of performance. Understanding the response of whole-body dynamic stability movement strategies may indicate how tolerant and robust those movement strategies may be, whilst highlighting any potential exertion-induced weaknesses that may develop or lead to increased injury risk. Unanticipated side cutting task observation was embedded into a 90-minute overground match-simulation. Assessment of specific mechanisms of whole-body dynamic stability were taken in pre-simulation, then at intervals within six 15-minute blocks, with a 15-minute break for half-time.

General Methods

In this section further details are provided on common approaches used across studies 1-4 within the thesis that are particularly novel or relatively limited in the existing body of literature in this field.

Standard Lab-set up

An overview of the standard lab set-ups for studies 1-4 are provided in Figures GM.1 and GM.2. Studies 1 and 4 were conducted using the lab set-up shown in Figure GM.1. In Study 4, side cutting was unanticipated and embedded into a match-simulation protocol explained in detail in that study. To signal the unanticipated direction of the turn there was an additional computer screen placed 3 metres beyond the force platform, at 1 metre from the floor, facing the participants. In studies 2 and 3, side cutting tasks were conducted using the laboratory set-up seen in Figure GM.2. The light units beyond the force platform would signal the direction to turn, depending on the condition.

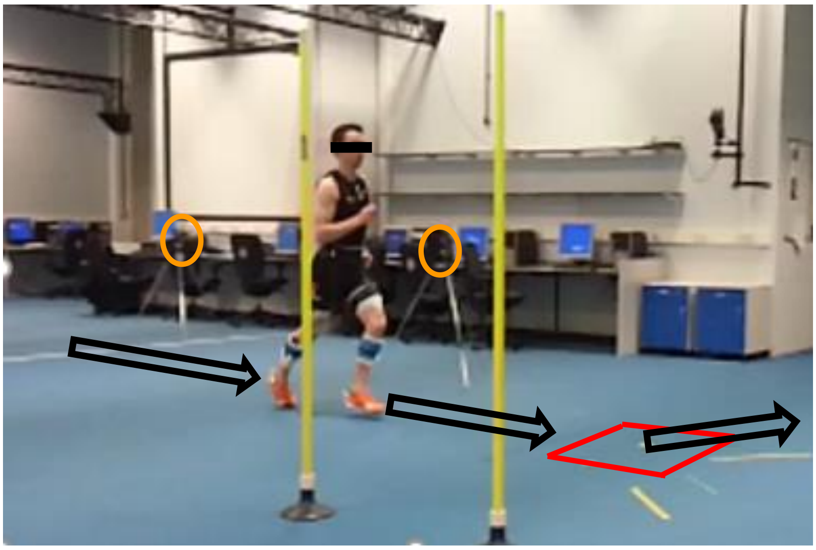

Figure GM.1. Example of the standard laboratory set-up for study 1 and study 4. Arrows show the approach and 45° exit for the side cutting task; orange ovals highlight the timing gates to control the approach speed; red rectangle highlights the force platform where the side cutting turn was completed. Motion Capture cameras were rail mounted and aimed at the area around the force platform.

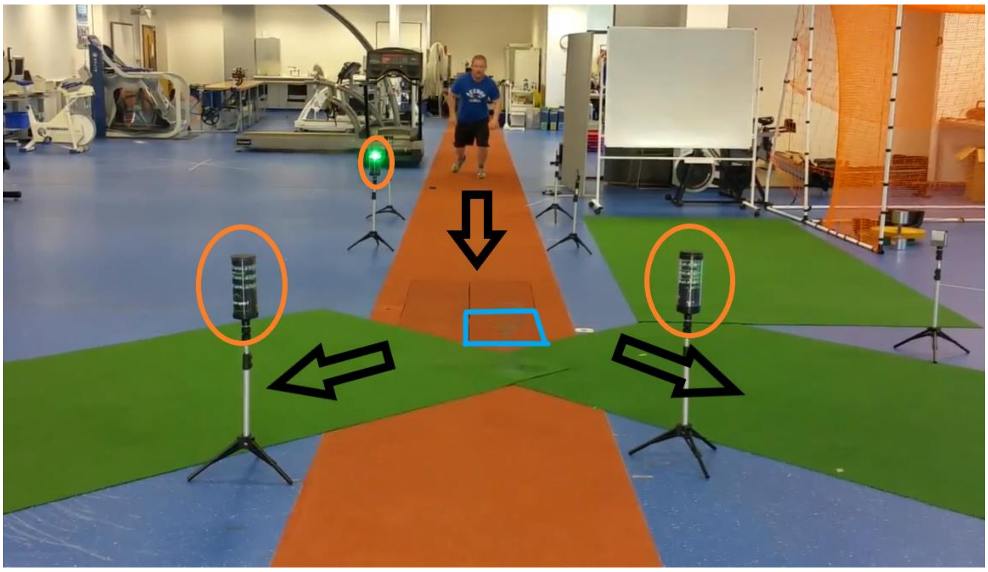

Figure GM.2. Example of the lab set-up for anticipated (ANT) and unanticipated (UNANT) 45° side cutting trials in studies 2 and 3. Approach and side cutting direction are indicated by the arrows; the force plate is highlighted by the blue rectangle; and the orange ovals highlight the trigger light unit and the UNANT cueing stimulus light units. Motion Capture cameras were rail mounted and aimed at the area around the force platform.

Lower Limb and Trunk Marker Model

The marker model used for studies 1-4 comprised eight segments including: thorax/abdomen; pelvis; thighs; shanks; and feet. All participants involved in studies 1-4 had 44 reflective markers captured based on the Liverpool John Moores University Lower Limb and Trunk eight segment model (Vanrenterghem et al., 2010). Additional details of the physically placed markers (anatomical markers and marker clusters – see Figure GM.3. below), virtual landmarks and segment definitions are found in the appendices (Appendix 1, page 163, from Malfait et al., 2014).

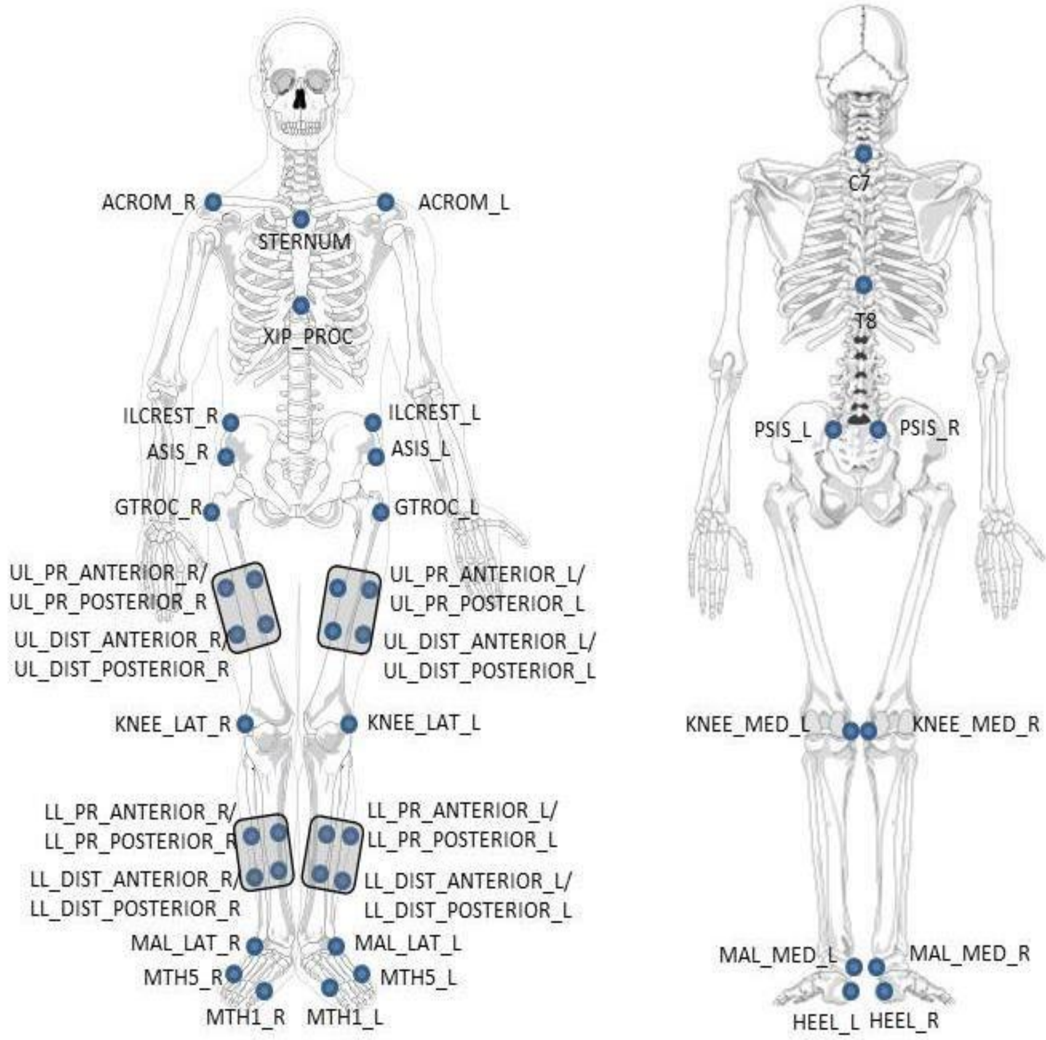

Figure GM.3. Example of the physically placed markers - anatomical markers and marker clusters – from the LJMU LLT model (Malfait et al., 2014).

Quantification of whole-body dynamic stability

In studies 2-4, factors that influence the medial GRF vector were calculated to represent whole-body dynamic stability. The first priority was to calculate foot placement, which would influence the origin of the GRF vector, and represents the dynamic relationship between the CoM and base of support. The first step in calculation of the proposed *whole-body dynamic stability* variables was to calculate the 'extrapolated' CoM (XCoM). The XCoM concept begins with the principle that in static stability the vertical projection of the CoM falls within the base of support, but expands this concept to include the velocity of the

CoM, suggesting that for understanding the conditions of stability one should take into account where the CoM is heading. Hof, Gazendam and Sinke (2005) propose that the XCoM concept can be adapted for dynamic scenarios still based on an inverted pendulum model that includes CoM positions, CoM velocity, leg length and acceleration (gravity). The first whole-body dynamic stability variable in this thesis - **(1) *foot placement*** – was calculated as the position of the XCoM relative to the borders of the foot, specifically the fifth metatarsal head. The XCoM was calculated according to Hof and collegues, and later described by Oberlander et al. (2012) - the equation can be written in the following way -

Equation 1:

$$\mathrm{XCoM} = \mathrm{pCoM} + \frac{\mathrm{vCoM}}{\sqrt{\mathrm{gl}^{-1}}}$$

- Where *pCoM* is the M-L position of the CoM, *vCoM* is the M-L velocity of the CoM, *g* is gravity, and *l* is the distance between the CoM and ankle in the frontal plane. Subsequently, the XCoM was calculated as a distance from metatarsal head 5 (MTH5) at the initial touchdown event, to represent M-L *foot placement*. This variable is similar to the measurement described for margin of stability, and may represent the initial condition of dynamic stability (Hof, Gazendam and Sinke, 2005). Specifically, for the purposes of these data, a positive value for *foot placement* indicated that the XCoM was medial to MTH5, whilst a negative value indicated that the XCoM had moved lateral to MTH5. The second whole-body dynamic stability variable - **(2) *M-L centre of pressure (CoP) position*** - was calculated, at every time point across ground contact, as the point of application of the ground reaction force vector, relative to previously mentioned MTH5 anatomical landmark.

IAA modelling was conducted in Visual 3D using the IAA Mambo supplementary software module. The original model is explained in detail elsewhere (Kepple, Siegel and Stanhope, 1997), and updated IAA modelling was most recently described and applied in João et al. (2014) and Moniz-Pereira et al. (2018). It was expected, due to the nature of the side cutting task, that there would be substantial rotational acceleration around the ankle joint, therefore the ‘free-foot’ IAA model was adopted as the most appropriate foot-floor interaction (Kepple, Siegel and Stanhope, 2002; João et al., 2014). In the ‘free-foot’ IAA model the ground contact connection is modelled as a pin joint through the CoP, with the axis parallel to the foot sagittal plane axis. IAA involved instantaneous angular acceleration calculation of the dynamic equations of motion, sequentially, for each joint moment that was available - after setting the Coriolis, Gravitational terms, and assuming all other joint frictions and torques were zero (João et al., 2014). Following this process, the relative contribution of each lower limb joint moment to the components of the GRF can be expressed, in selected planes, where the GRF components represent the vertical, anterior-posterior, and M-L acceleration of the body’s CoM. Following IAA, non-negligible contributions to the M-L ground reaction forces were found from sagittal plane hip, knee and ankle joints, and also in the frontal and transverse planes for the hip joint. Those contributions were then consolidated in their respective planes to represent the third, fourth, and fifth whole-body dynamic stability variables: **(3)** ***Sagittal triple acceleration*** (the sum of the sagittal plane hip, knee and ankle joint contributions); **(4)** ***frontal plane hip acceleration***; **(5)** ***transverse plane hip acceleration***. Following examples in previous research (Kepple, Siegel and Stanhope, 1997; João et al., 2014; Moniz-Pereira et al., 2018) the accuracy of IAA was determined by finding the absolute mean difference between the experimentally measured ground reaction forces and those estimated from IAA. The difference was then represented as a percentage of the maximum force obtained.

In summary, the five distinct whole-body dynamic stability variables in studies 2-4 were calculated to represent factors that influence the medio-lateral ground reaction force vector (see Figure GM.2. diagram below). The initial position of the base of support is defined by variable (1), then the movement of the origin of the ground reaction force vector (CoP) is defined by variable (2). The planar contribution to the magnitude of the medio-lateral ground reaction force vector are defined by variables (3), (4) and (5).

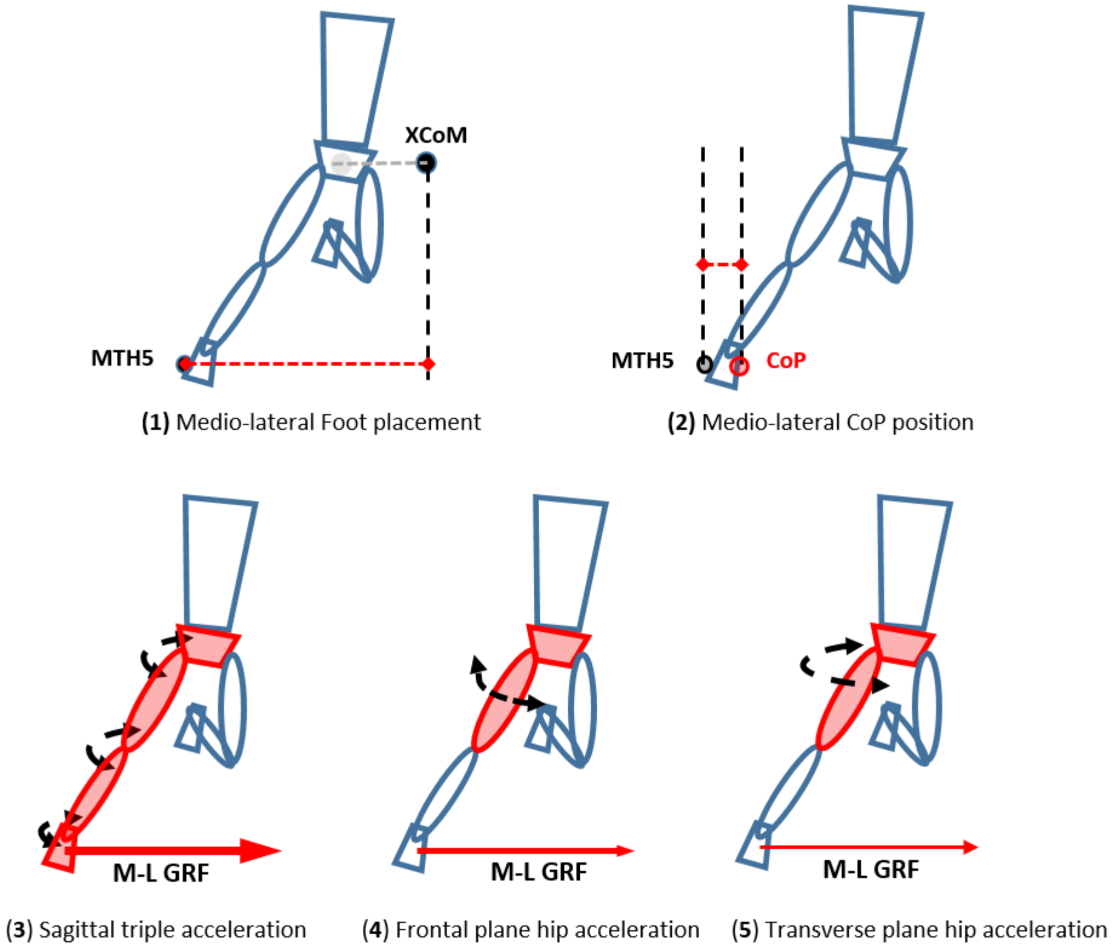

Figure GM.2. Diagram of the five distinct whole-body dynamic stability movement strategies for medio-lateral control of the CoM.

All time-series data were collected between the touchdown and toe-off events, representing the ground contact time boundaries for the side cutting tasks that were observed. Those signals were interpolated for 101 data points representing 0-100% of ground contact.

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Study 1

How reliable are knee kinematics and kinetics during side cutting manoeuvres?

Abstract

Side cutting tasks are commonly used in dynamic assessment of ACL injury risk, but only limited information is available concerning the reliability of knee loading parameters. The aim of this study was to investigate the reliability of side cutting data with additional focus on modelling approaches and task execution variables. Each subject (n=8) attended six testing sessions conducted by two observers. Kinematic and kinetic data of 45° side cutting tasks was collected. Inter-trial, inter-session, inter-observer variability and observer/trial ratios were calculated at every time-point of normalised stance, for data derived from two modelling approaches. Variation in task execution variables was regressed against that of temporal profiles of relevant knee data using one-dimensional statistical parametric mapping. Variability in knee kinematics was consistently low across the time-series waveform (≤5 °), but knee kinetic variability was high (31.8, 24.1 and 16.9 Nm for sagittal, frontal and transverse planes, respectively) in the weight acceptance phase of the side cutting task. Inverse kinematic modelling reduced the variability in sagittal (~6 Nm) and frontal planes (~10 Nm) compared to direct kinematic modelling. Variation in task execution variables did not explain any knee data variability. Side cutting data appears to be reliably measured, however high knee moment variability exhibited in all planes, particularly in the early stance phase, suggests cautious interpretation towards ACL injury mechanics. Such variability may be inherent to the dynamic nature of the side cutting task or experimental issues not yet known.

Introduction

The occurrence of non-contact lower-limb injury in sports that involve dynamic sporting tasks is a substantial burden on clubs and their players, both financially and in terms of playing time (Hawkins et al., 2001; Myers and Hawkins, 2012). Attempts to explore the mechanics of knee ligament injury, particularly of the anterior cruciate ligament (ACL), are well documented and frequently involve the estimation of knee kinematics and kinetics during side cutting tasks (Besier et al., 2001b; Pollard, Davis and Hamill, 2004; McLean, Huang and van den Bogert, 2005; Houck, Duncan and De Haven, 2006; Landry et al., 2009; Kristianslund and Krosshaug, 2013). Side cutting is commonly used as it challenges the knee in a manner that is consistent with the reported ACL injury mechanism (Markolf et al., 1995), and therefore could be important to assess ACL injury risk. Thus, it is important to know the reliability of side cutting data, as well as the variability within typical protocols so that appropriate limits for detectable differences can be established, and the correct interpretation of injury risk made.

Limited information concerning the reliability of side cutting data has been presented. The chosen analysis methods are varied and include average intra-class correlation coefficients (ICC) (Houck, Duncan and De Haven, 2006; Ford et al. 2005), coefficients of multiple correlations (CMC) (Sigward and Powers, 2006a; Sigward and Powers 2006b), and coefficients of multiple determinations (R^2) (Besier et al., 2001a; 2003). As well as different quantification methods, different components of reliability have been observed. Besier et al., (2001a; 2003) reported within and between session reliability for various tasks and found that, of their side cutting tasks (30° and 60°), transverse knee moments displayed the lowest reliability *within-session* (average $R^2 = 0.84 \pm 0.09$), and sagittal knee moments displayed the highest reliability *between-sessions* (average $R^2 = 0.89 \pm 0.04$). Sigward and Powers (2006a; 2006b) reported *between-session* reliability and found frontal and transverse plane kinematics (CMC = 0.63 and 0.61, respectively) to be less reliable than frontal and transverse plane kinetics (CMC = 0.90 and 0.93, respectively). Although this reliability evidence exists, they lack a number of facets that are important for clinical inference. Firstly, previous studies failed to consider *between-observer* reliability which is crucial to assess results across laboratories or in clinical practice. Secondly, these methods summarise reliability by either considering discrete time points (e.g. peak values) or collapsing the entire time-series (e.g. CMC calculates average reliability over time). Therefore information about whether reliability is evenly distributed across different phases of the side cutting manoeuvre is unknown. Thirdly, the summary reliability statistics are not presented in the context of the original data, making it difficult to interpret the magnitude of reliability (e.g. ICC of 0.6 versus 0.7) in the context of the magnitude of the actual data signals. A comprehensive observation of side cutting data reliability is therefore necessary.

We also take the opportunity to address i) the reliability of the modelling approach as this can affect knee kinematics and kinetics (Robinson, Tsao and Donnelly, 2014) and ii) the variability of the task itself. Firstly, different modelling approaches can be chosen to either allow or restrict joint rotations or translations and also attempt to reduce soft tissue artefact. In a recent comparison of the direct kinematic (DK) versus inverse kinematic (IK) modelling approaches (Robinson, Tsao and Donnelly, 2014), significantly larger peak knee abduction moments were found using the DK approach yet the reliability of two approaches are unknown. Secondly, as variability can also exist through variations in the execution of the side cutting task itself, we quantify whether knee kinematic and kinetic variability can be explained through inherent variations in task execution. Such information will help to standardise modelling approaches and evaluate the importance of task execution.

The purpose of this study was to investigate the reliability of side cutting data from an inter-trial, inter-session, and inter-observer perspective. This will be complemented by investigating the reliability of two modelling approaches (DK versus IK), and by examining the contribution of the side cutting task execution to the variability observed.

Methods

Participants

The participants for this study were eight recreationally active soccer players who had at least 6 years of playing experience and trained 1-2 times per week (four male; four female; age - 25.8 ± 4.4 years; mass - 64.8 ± 7.2 kg; height - 1.7 ± 0.1 m). All participants had no reported ACL injury and had been injury free for six months prior to data collection. All participants wore tight fitting shorts and standardised indoor footwear (Highroad). Females also wore a cropped vest, tight fitting base layer or sports bra. Ethical approval for this study was granted by the institutional ethics committee, and written consent was obtained from all participants.

Protocol

All participants engaged in a familiarisation session which included full replication of one session of the protocol. Prior to side cutting, all participants completed a ten minute general warm-up. This was followed immediately by a 5 minutes specific warm-up. Participants nominated their preferred leg for side cutting and this was standardised for the assessment. Approach speed was controlled using photocell timing gates (Brower Timing Systems, Utah, USA) which were placed 2 m apart, and 2 m from the force plates, where the side cutting was performed. Cones were also placed 3 m from the force plates to mark a target gate at the required 45°. Trials were excluded if approach speed was not between 4 and 5 $m \cdot s^{-1}$, targeting of the force plate was observed, or if the subjects did not achieve the angle of 45° determined by running between the cones.

Data were collected by two different observers using a repeated measures design over six separate sessions; four on day one, and two on day two (figure 1.1). The observers were both PhD students and had been working with this biomechanical model for approximately 4 months previous, in both application and processing. The two observers conducted three sessions each; two each on day one, and one each on day two, with 48 hours between day one and two. This allowed each participant to be tested by each observer, within and between days. A 10-minute cool down session was conducted before a 15-minute rest, and then the next session would start.

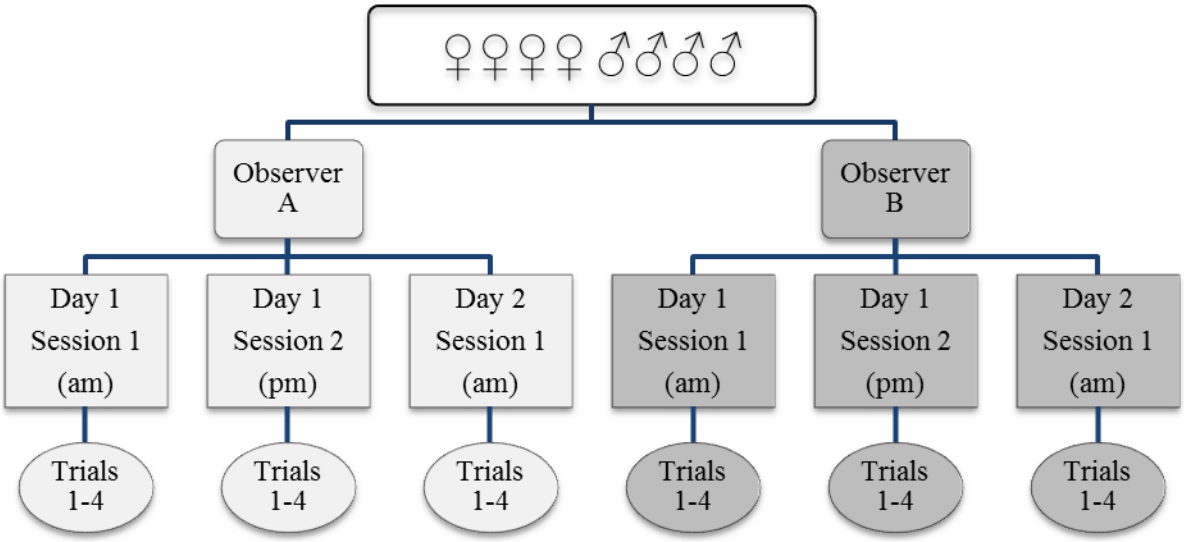

Figure 1.1. Schematic representation of the repeated-measures experimental design, showing eight participants; two observers; six sessions; and trials per side cutting direction.

Data Collection

All side cutting was performed over a 0.9 x 0.6 m Kistler force platform (9287C, Kistler Instruments Ltd., Winterthur, Switzerland) sampling at 1500 Hz for the measurement of ground reaction forces. Simultaneous kinematic data was recorded in Qualisys Track Manager (Qualisys AB, Gothenburg, Sweden) using 10 optoelectronic cameras (Oqus 3, Qualisys AB, Gothenburg, Sweden) sampling at 250 Hz.

Biomechanical model

A full description of the LJMU model utilised in the current study, based on direct kinematic (DK) calculations, is provided in supplementary material elsewhere (Malfait et al., 2014). Both observers were blind to the application of markers by the other observer. Each observer applied and removed the markers at the beginning and end of their testing sessions. Visual 3D (v.4.83, C-Motion, Germantown, MD, USA) was used for all modelling and analysis with segments being represented by geometric volumes. The inverse kinematic model (IK), processing was identical to recent study (Robinson, Tsao and Donnelly, 2014) where translational joint constraints were applied to the hip, knee, and ankle joints giving each segment three degrees-of-freedom each.

Data and statistical analysis

Marker coordinate and force data were filtered using a Butterworth 4th order low pass filter with a 20 Hz cut-off frequency (Kristianslund et al., 2012). Touch-down and toe-off events were identified using a threshold of 20 N. For the comparison of modelling techniques, DK and IK kinematics were used separately to estimate the net external moments using inverse dynamics. Knee angle and moment data (order of rotations – X, Y, Z) from sagittal, frontal and transverse planes were normalised to 101 data points, for the contact phase of side cutting. All mean peak knee angle and moment data, for three planes, were calculated during the weight acceptance phase of the side cutting. The weight acceptance phase was defined as 0-25% of normalised ground contact for this study.

The inter-trial, inter-session and inter-observer variability were estimated using the procedures outlined by Schwartz, Trost and Wervey (2004). As well as the point by point calculation over the entire contact phase, inter-observer variability was also expressed as a ratio to inter-trial variability. The same variability calculations (inter-trial, -session and –observer) were made for both modelling techniques, as well as calculation of overall average curves and standard deviations for angle and moment data, in all three planes.

One-dimensional statistical parametric mapping (SPM - Pataky, 2012) was used to examine the relationship between the DK knee angle and moment waveforms and selected task execution (TE) variables (resultant centre of mass (CoM) touchdown velocity; CoM toe-off velocity; CoM touchdown, and toe-off cutting angle; contact time; and both horizontal, and vertical impulses). This was similar to a recent investigation looking at the influence of approach speed on knee kinematics and kinetics during side cutting (Vanrenterghem et al., 2012). The following linear regression models were defined:

Knee angle $(t) = (\beta_1(t) \times \text{TE variable}) + \alpha_1(t) + \varepsilon(t)$

Knee moment $(t) = (\beta_2(t) \times \text{TE variable}) + \alpha_2(t) + \varepsilon(t)$

The slopes of the task execution variable-angle and -moment relations (β_1 and β_2) were computed at each time point (t) resulting in 101 β trajectories. These β trajectories were first computed for each subject and secondly, all subjects' β trajectories were submitted to a population-level one-sample t-test yielding a SPM{t} statistical curve. The significance of each SPM{t} was then determined topologically using random field theory (see Pataky, 2012).

Results

For all kinematics, inter-trial, -session and -observer variability was below 5.5° for the full waveforms, in all planes (figure 1.2 d-f). The inter-trial variability was consistently lowest and no part of the waveform provided consistently higher variability. Typically the waveforms of the inter-trial variability were similar but lower in magnitude than the inter-session and inter-observer variability.

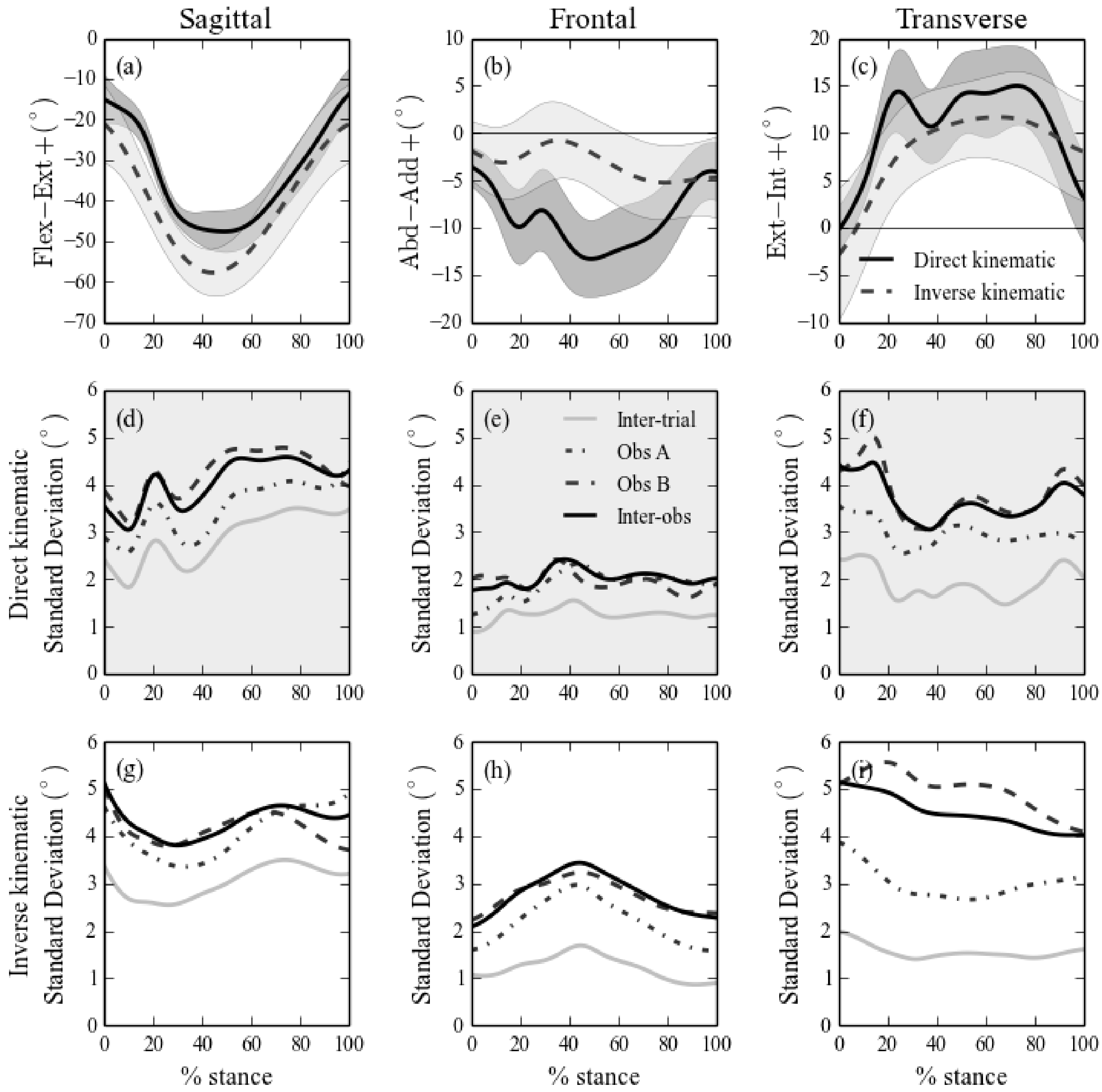

Figure 1.2. Kinematic data and error data for the knee in all three planes – planar data are in one column each. Row one (a-c) shows mean (±SD) knee kinematics for the Direct Kinematic (DK) versus Inverse Kinematic (IK) modelling approach; Row two (d-f) shows the standard deviation inter-trial, inter-session (Observer A = Obs A; Observer B = Obs B), and inter-observer error waveform observed for DK modelling; Row three (g-i) shows the standard deviation within-subject, inter-session (Obs A and Obs B), and inter-observer error waveform observed for IK modelling. [Shading in row two is to help distinguish between modelling approaches].

In the kinetic data, the weight acceptance phase of normalised ground contact (0-25%) provided the largest inter-trial, inter-session and –observer variability with peak magnitudes of all types of variability for the sagittal plane, frontal plane and transverse plane ranging between 32-42 Nm, 24-31 Nm and 17-20 Nm, respectively (figure 1.3, d-f). Inter-trial variability was lowest across all kinetic waveforms peaking at 32, 24 and 17 Nm for sagittal, frontal and transverse knee moments, respectively. Inter-session and –observer variability echoed the waveforms of inter-trial variability, but at a higher magnitude across the time-series. Differences between inter-trial variability and inter-session/–observer variability were highest in the sagittal plane and lowest in the transverse plane.

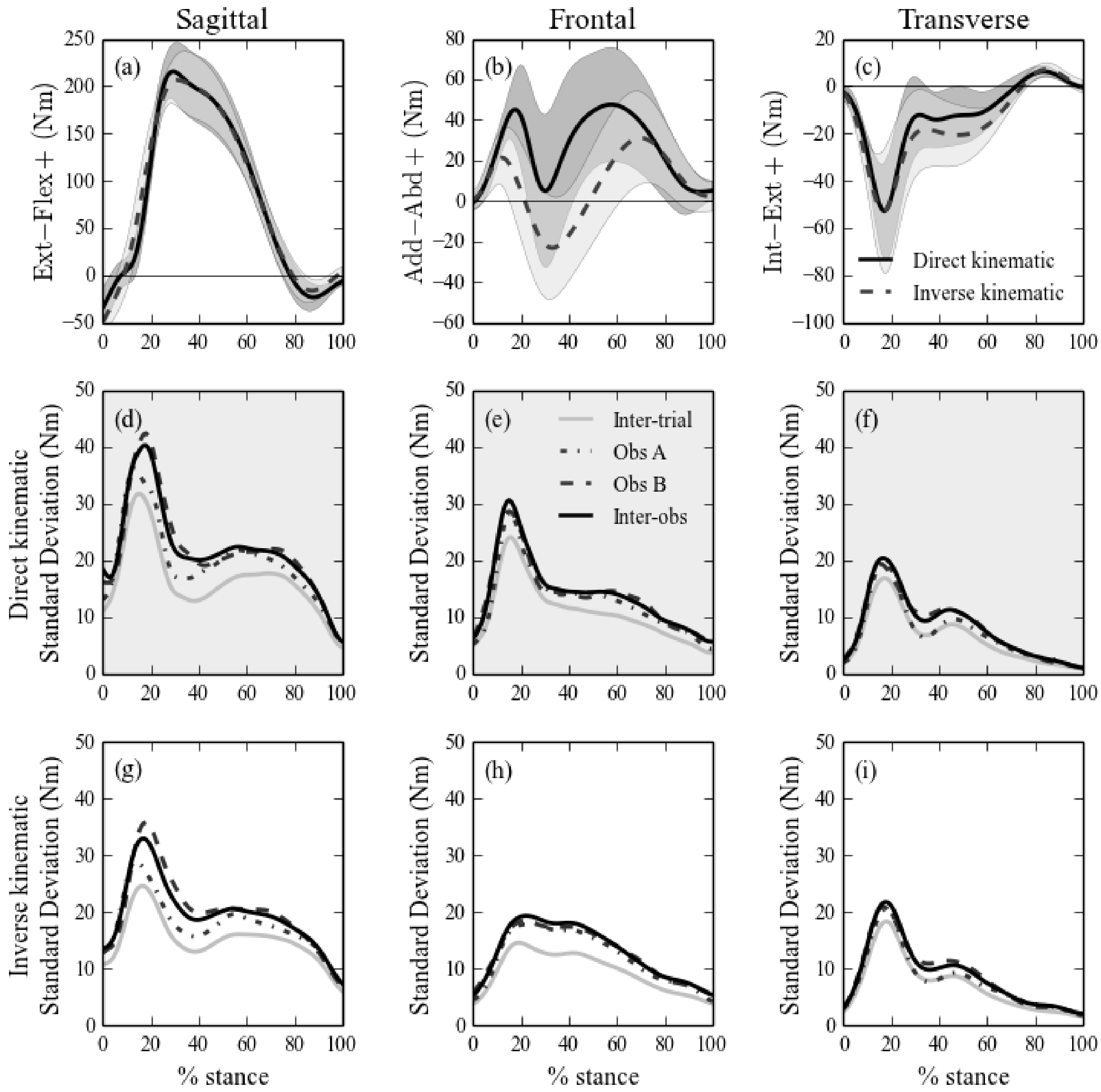

Figure 1.3. Kinetic data and error data for the knee in all three planes – planar data are in one column each. Row one (a-c) shows mean (±SD) knee kinetics for the Direct Kinematic (DK) versus Inverse Kinematic (IK) modelling approach; Row two (d-f) shows the standard deviation inter-trial, inter-session (Observer A = Obs A; Observer B = Obs B), and inter-observer error waveform observed for

DK modelling; Row three (g-i) shows the standard deviation within-subject, inter-session (Obs A and Obs B), and inter-observer error waveform observed for IK modelling. [Shading in row two is to help distinguish between modelling approaches].

Mean peak knee kinematics and kinetics (± standard deviation) from weight acceptance were presented for DK and IK, in all three planes, in addition to the mean inter-observer/inter-trial variability ratios for the same variables (Table 1.1). Where peaks were not clear in weight acceptance, the value at the upper threshold (25%) was used ('*' denotes this occurence in Table 1.1). Greater inter-observer/inter-trial ratios were found for IK in the frontal and transverse planes (2.3 and 2.9, respectively) versus DK (1.6 and 1.9, respectively).

The DK and IK derived kinematics and kinetics (figure 1.2 a-c and figure 1.3 a-c) were similar to those previously reported (Robinson, Tsao and Donnelly, 2014) where the frontal plane knee angles and moments differed most. IK kinematic variability appeared visually smoother in comparison to DK (figure 1.2 DK = d-f, IK = g-i). Where DK variability appeared to oscillate, particularly during weight acceptance, IK variability was more consistent. For the kinetic data, in weight acceptance, for DK modelling, inter-trial, inter-session and between–observer variability reduced from sagittal to frontal to transverse plane knee moments. In weight acceptance for IK modelling, in comparison to DK, there is a reduction in variability for sagittal plane (~ 6 Nm reduction) and frontal plane knee moment (~ 10 Nm reduction), but variability for the transverse plane knee moment remained similar. [1]

[1] Additional variability data for normalised knee abduction moment data using both modelling approaches, and normalised medial ground reaction force data is presented in Appendix 2, figure 1.5, page 168

Table 1.1. Direct kinematic (DK) and inverse kinematic (IK) derived peak mean (± SD) knee angle (deg) and knee moment (Nm) data from weight acceptance phase. Mean inter-observer/ inter-trial ratio, for DK and IK modelling, over full time series for knee angle and moment data for side cutting.

	Sagittal (FLEX/EXT)		**Frontal (ABD/ADD)**		**Transverse (IR/ER)**	
	DK	**IK**	**DK**	**IK**	**DK**	**IK**
Mean Peak Angles (deg)	**-36.41 ***	**-46.28 ***	**-9.93**	**-3.12**	**14.38**	**7.52 ***
SD	3.1	5.74	3.99	3.83	4.34	4.59
Mean Observer/trial ratio	1.4	1.4	1.6	2.3	1.9	2.9
Mean Peak Moments (Nm)	**197.6 ***	**187.6 ***	**45.0**	**21.4**	**-52.8**	**-52.9**
SD	23.8	18.0	19.62	19.8	20.3	26.3
Mean Observer/trial ratio	1.3	1.3	1.3	1.4	1.3	1.3

NB. '' denotes no clear peak was observed in weight acceptance of normalised ground contact.*

Variation in kinematic or kinetic profiles was not explained by variation in any of the task execution variables, as demonstrated in the SPM regression analysis by non-significant relationships. An example of SPM linear regression is also provided (figure 1.4). All SPM analyses are available in the Appendices (see Appendix 3, page 169).

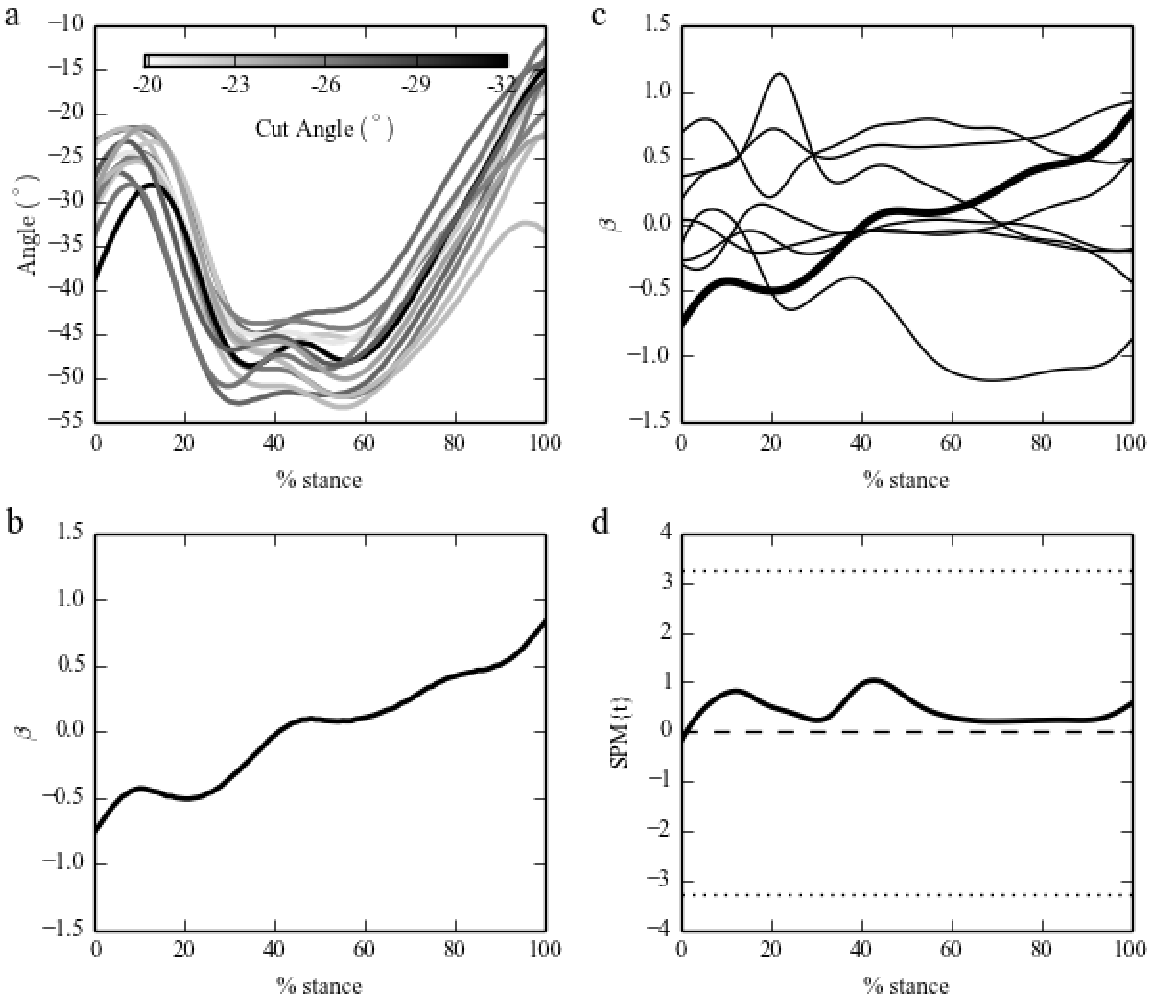

Figure 1.4. An example of the SPM analysis used to linearly regress task achievement variables and knee angles and moments across the entire stance phase. In (a) one subject's knee flexion angle waveforms are shown and shaded according to their cutting angle at take-off. In (b) the slope of the relationship between the knee flexion angles and the cutting angles at take-off is shown. The process in (a and b) is repeated for each subject to generate a *β* curve per subject (c), the *β* trajectory from (b) is shown in bold. All subjects' beta curves are then analysed using a one-sample t-test yielding the SPM{t} curve (d). As the critical t threshold of 3.26 was not exceeded, there was no significant relationship between subjects for knee flexion angle and cutting angle at take-off.

Discussion

The primary aim of this study was to investigate the reliability of side cutting data using inter-trial, inter-session and inter-observer observations. Whilst kinematic data variability was consistently low across the time-series, irrespective of plane, kinetic data variability was distinctly elevated to seemingly high magnitudes in the weight acceptance phase. Such observation is a concern when pursuing typical ACL injury markers, such as frontal plane knee moments, however, it is important to consider the source and proportionality of variability, to fully interpret the reliability of this data.

Previously, kinematic and kinetic data from side cutting has been suggested to be reliable, in inter-trial and inter-session observations (Besier et al., 2001a; Sigward and Powers, 2006a; Sigward and Powers 2006b). However, the current study is the first to investigate and present variability for every point across the time-series for side cutting data signals. Furthermore, the variability data suggests that the main issue lies with an inherently high inter-trial variability, and the addition of multiple sessions and observers has minimal impact. This is further supported by the observer/trial ratios, where the impact of multiple observers, and the experimental implications that introduces (e.g. marker placement), is less influential in kinetic data than kinematics. This is good news for studies using multiple sessions and observers, but requires further exploration of inter-trial variability

When exploring the source of the inter-trial variability, the dynamic nature of the side cutting task should be considered. For example, inconsistencies in technique, perhaps within-subject, such as horizontal forces, foot-placement or postural control, may elicit variable knee kinetics, whilst knee kinematics remains relatively unaffected. A similar study examining variability in drop vertical jumping (Malfait et al., 2014) found comparable peak magnitudes of kinematic variables to this study, but there is greater kinetic variability in side cutting. A proportional comparison of kinetic signal against observed variability may help to identify the impact of such variability on clinical inference. In the present study the knee kinetic trial-to-trial variability represented approximately 15, 56 and 34% of the average peak knee moment for sagittal, frontal and transverse planes, respectively. In Malfait et al. (2014), for drop vertical jumps, the knee moment trial-to-trial variability represented approximately 14, 26 and 29% of the average peak knee moment. Thus, although the side cutting task places a greater planar demand in execution compared to the drop vertical jump, the greatest variability may be considered proportionally similar at least in flexion/extension and internal/external rotation. The proportional variability in abduction/adduction is far greater for side cutting kinetics, compared to drop vertical jumps (Malfait et al., 2014), and is likely to be due to the larger horizontal forces required to execute the task.

Comparison of modelling approaches suggests a potential benefit of IK compared to DK as IK showed a reduction in variability reported in both the sagittal (~6 Nm) and frontal planes (~10 Nm). Therefore, the IK modelling approach could potentially offer an alternative when we are looking to reduce variability in observing knee sagittal and frontal plane loading. Increased variability in the DK approach could be due the soft tissue artefact which directly influences the calculated kinematics. DK modelling approaches would therefore require greater sample sizes to detect the same magnitude of effect as the IK approach. However, interpretation of the inter-observer/inter-trial ratio suggests that IK modelling may be more sensitive to multiple observers than DK modelling for kinematic data (see table 1). The specific causes of this discrepancy are unclear though. It may be that IK modelling "filters" true signal by fitting measured motion to the model and does not simply remove the effect of soft tissue artefact. This however requires further investigation.

Although the reporting of task execution variables during side cutting is limited, evidence has shown the importance of variables like approach velocity, in relation to known key loading variables (Vanrenterghem et al., 2012). The SPM regression analyses failed to find any significant relationship with the task execution variables and the joint kinematic or kinetic data. This suggests that the small variations in task execution, that occur over the narrow approach speed, and which are inherent to performing such a dynamic task, did not explain knee joint kinematic or kinetic variability. Researchers may expect that high magnitudes of variability could be reduced by more stringent task execution criteria, but our results indicate that this is unlikely.

High magnitudes of variability also have implications for the magnitudes of a detectable difference and therefore study design, in terms of sample recruitment. To illustrate this, sample size estimation was calculated for a one sample t-test. To observe a difference ≥10 Nm in the peak knee joint moment in the frontal plane (for DK only) a sample size of n≥48 is required (refer to Appendix 4, figure 1.6, page 173) based on our inter-trial variability of 24.1 Nm and a statistical power of 80 %. As the inter-session and inter-observer variability were greater than the inter-trial variability, additional participants would be required to detect the same 10 Nm difference (n=67 and n=76, respectively) in study designs requiring participants to be tested in different sessions or by different observers. Although 10 Nm was chosen as an arbitrary value, this indicates the relationship between the study design, the detectable difference, sample size and statistical power. The sample sizes calculated here are model and lab-specific therefore similar processes should be undertaken by other labs.

Limitations to this study were that no between-subject observation was made, which may potentially contribute to sources of reported variability. This would be an opportunity for further research, as would investigation of other potential ACL injury variables during side cutting that may not just be associated with the knee. It is possible that adjusting the dispersion or number of sessions, or the addition of further observers may have some impact on inter-session or inter-observer variability, however, the analyses was based on 192 trials of data using similar research design as published previously for relevant reliability studies (Schwartz, Trost and Wervey, 2004; Malfait et al., 2014). Thus, the main aim moving forward must be to explain the remaining inter-trial variability observed in the kinetic signal.

Indeed, inherent variability of the method derived from such experimental concerns as soft tissue artefact may reduce the inter-trial variability.

Conclusion

In conclusion, this is the first study that attempts to fully identify the reliability of kinematic and kinetic knee data from side cutting, using a method that provides a specific focus toward relevant phases of a highly dynamic task. Although the variability of the kinematic signals from side cutting does not pose a major cause for concern, the variability of the kinetic signals, specifically in the weight acceptance phase suggests that the use of these signals for diagnostic purposes may be challenging. An alternative approach may be to consider the variability itself as a predictor of ACL injury risk, as previously reported from different research perspectives (Heiderscheit, 2000; Preatoni et al., 2013). The relevance of signal variability as an ACL injury predictor requires further investigation.

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Study 2

Whole-body dynamic stability in side cutting: implications for markers of lower limb injury risk and change of direction performance

Abstract

Control of the centre of mass (CoM) whilst minimising the use of unnecessary movements is imperative for successful performance of dynamic sports tasks, and may indicate the condition of *whole-body dynamic stability*. The aims of this study were to express movement strategies that represent whole-body dynamic stability, and to explore their association with potentially injurious joint mechanics and side cutting performance. Twenty recreational soccer players completed 45° unanticipated side cutting. Five distinct whole-body dynamic stability movement strategies were identified, based on factors that influence the medial ground reaction force (GRF) vector during ground contact in the side cutting manoeuvre. Using Statistical Parametric Mapping, the movement strategies were linearly regressed against selected performance outcomes and peak knee abduction moment (peak KAM). Significant relationships were found between each movement strategy and at least one selected performance outcome or peak KAM. Our results suggest excessive medial GRFs were generated through sagittal plane movement strategies, and despite being beneficial for performance aspects, poor sagittal plane efficiency may destabilise control of the CoM. Frontal plane hip acceleration is the key non-sagittal plane movement strategy used in a corrective capacity to moderate excessive medial forces. However, whilst this movement strategy offered a way to retrieve control of the CoM, mitigating reduced whole-body dynamic stability, it also coincided with increased peak KAM. Overall, whole-body dynamic stability movement strategies helped explain the delicate interplay between the mechanics of changing direction and undesirable joint moments, providing insights that might support development of future intervention strategies.

Introduction

Control of the centre of mass (CoM) is prioritised above all other demands in dynamic movement (MacKinnon and Winter, 1993; Patla et al., 1999). When CoM control is lost, we may observe a fall, failure to execute the task, or a scenario where excessive stress is placed on the musculoskeletal system to prevent either of those from happening. To avoid a fall or a failure of the task one may exhibit undesirable deviations in technique that may be a precursor to dangerous joint loading. The influence of controlled technique changes in side cutting has been explored in the context of Anterior Cruciate Ligament (ACL) injury risk by several research groups (Dempsey et al., 2009; Donnelly et al., 2012; Kristianslund et al., 2014; Havens and Sigward, 2015a; 2015b; 2015c; Jones et al., 2015; David et al., 2017). However, it is often not clear how common kinematic and kinetic variables are associated with each other toward general control of the CoM, or even how their roles may change through phases of ground contact. Donnelly et al. (2012) used biomechanical simulations to suggest that redirecting the whole-body CoM medially and towards the direction of travel would bring about a reduced external peak knee abduction moment (KAM). Whilst this is important for ACL injury risk, it is unclear through which movement strategies such a redirection of the CoM could best be achieved without causing task failure or increased stresses elsewhere in the musculoskeletal system. Currently, this makes it difficult for a practitioner to interpret findings towards meaningful intervention strategies. Therefore, a more holistic view of the movement strategies that are necessary in control of the CoM may highlight the condition of whole-body dynamic stability, and the intricate interplay between task performance and injury risk.

Side cutting involves generating an impulse against the ground to decelerate then accelerate the CoM. In addition to the approaching velocity, it is specifically the accelerative impulse in the medial direction that determines the actual change of direction of the CoM and acceleration in the new direction of travel – i.e. task performance. Detailed expression of factors that influence the medial ground reaction force (GRF) vector may therefore quantify how these important impulses are generated, and thus, the movement strategies that are important for medial control of the CoM. To begin to quantify the medial GRF vector, we must start with where the foot is placed. It is possible to quantify foot placement by the dynamic association between the CoM and base of support, similar to the margin of stability previously reported (Hof et al., 2005; Havens et al., 2018). In this case, foot placement also represents an initial condition of whole-body dynamic stability within the task. Once the foot is placed, the origin of the GRF vector is the centre of pressure (CoP) under the foot. Although the CoP is limited to the boundaries of the base of support, the CoP position may change over ground contact time, perhaps in response to ankle movement. Subsequently, we can attempt to express the magnitude of the medial GRF vector, but more importantly, the contribution of the individual joint moments. Whilst joint contributions are rather elusive and difficult to quantify independently, the application of Induced Acceleration Analysis (IAA) may offer a useful approach.

Previous research has demonstrated the possibility of using IAA modelling to estimate the relative contribution of lower limb joint moments to GRF components (Kepple et al., 1997; João et al., 2014; Moniz-Pereira et al., 2018). Following calculation of the joint positions and net joint moments of the lower limb, the equations of motion can be solved, computing the relative contribution of each moment to accelerate the CoM (João et al., 2014). Thus, due to the direct relationship with acceleration of the CoM and GRF, one can express the relative contribution of each joint moment to the medial component of the GRF, for example. Once the factors that influence GRF vector are quantified, it is possible to determine their association with selected performance outcome and undesirable joint loading variables.

Therefore, key movement strategies for medial control of the CoM in side cutting are quantified through factors that influence the medial GRF vector, and this is likely to offer an integrated account of performance and injury risk. The aims of this study were to outline key mechanical movement strategies, specifically exploring their role in enhancing CoM change of direction angle and acceleration, whilst minimising peak KAM. As fulfilling these roles are likely to be disparate, this provides a unique challenge to the key movement strategies. Thus, it is hypothesised that movement strategies necessary to increase change of direction angle and acceleration, will also increase peak KAM.

Methods

Participants

The participants in this study were twenty healthy male recreational soccer players, with at least 6 years playing experience. The participants had a mean (± SD) age of 23 ± 3 years; mean height of 1.8 ± 0.1 m; and mean mass of 76.7 ± 10.4 kg. All participants were free from injury for at least 6 months, and written consent was retrieved from every participant. All participant recruitment processes were conducted in line with the university research ethics committee guidelines, which comply with the principles of the Declaration of Helsinki.

Protocol - Side cutting Assessment

Mock testing conditions were simulated in a familiarisation session no more than one week before the testing session. In the testing session, participants first completed a dynamic warm-up, as well as specific side cutting practice. Participants then completed a static trial and functional hip joint centre and knee joint axis tasks in the centre of the capture volume. Following calibration trials the motion trials were collected. The unanticipated side cutting task was controlled with a 4-5 $m.s^{-1}$ approach velocity, using timing gates (Smartspeed^{TM}, Fusion Sports, Australia) set 2 m apart, at 5 m and 3 m away from the force plate. The 45°

change with respect to forward progression was marked out to the left and right from the force platform with the use of cones. The preferred leg for change of direction was used for all trials meaning participants completed either side cutting or cross-over cutting depending on the light stimulus they received. To trigger the direction of onward progression, the light stimulus appeared on either the left or right, and participants were told to cut in the direction of the light (see Figure GM.2 or Appendix 5 page 174). The cueing light units to indicate the direction of the two unanticipated conditions were set up 3 m beyond the force plate, 1 m in height from the ground, and 2 m apart. If participants failed to adhere to the path or velocity constraints set for the side cutting task, that trial was discarded, and an additional trial was added to the trial count. On average the participants completed 24 trials in total, 12 side cutting and 12 cross-over cutting trials, subsequently, the participants completed a 10-15 minute cool-down protocol.

Biomechanical model

All participants had 44 reflective markers captured based on the Liverpool John Moores University Lower Limb and Trunk eight segment model (Vanrenterghem et al., 2010). Markers were applied to participants before a 15-minute dynamic warm-up, and bandages and strapping used to attach cluster plates on lower limb segments were adjusted for comfort, without compromising a secure fitting. 3D marker trajectory data were recorded using a 7-camera Vicon MX system (Vicon, Oxford Metrics, Oxford, UK) at 250 Hz for the side cutting motion trials. Joint centres, axes and local segment coordinate systems were defined as reported previously (Vanrenterghem et al., 2010). The side cutting tasks were executed on a 0.6 x 0.4 m force platform (Kistler, Winterthur, Switzerland), and force data were sampled at 1500 Hz and synchronised with the Vicon system. Calibration, modelling, and all kinematic and kinetic analyses were completed in Visual 3D Professional (v.5.00.16, C-Motion, Germantown, MD, USA), and were based on segmental data from Dempster's regression equations (Dempster, 1955), moment of inertia properties from Hanavan (1964), and the use of geometric volumes to represent each of the eight segments. Inverse kinematic (IK) modelling was used in Visual 3D as a pre-requisite for IAA.

Data processing

Only the side cutting trials were analysed. Marker coordinate data and analogue signals from the force plate channels were filtered using a Butterworth 4th order recursive low pass filter, with a 20 Hz cut-off frequency (Kristianslund et al., 2012). Following IK modelling, inverse dynamics calculations were used to estimate the net external joint moments (cardan sequence – X-Y-Z). Touchdown (TD) and toe-off (TO) events were calculated as reported previously (Vanrenterghem et al., 2012; Sankey et al., 2015) to determine the ground contact phase. The CoM transverse plane trajectory angle and velocity were calculated. *Change of direction angle* was calculated as the change in CoM trajectory angle between TD and TO. Changes in M-L CoM velocity, once divided by ground *contact time*, represented the *average medial CoM acceleration* from TD to TO. Change of direction angle and average medial CoM acceleration were used to represent two selected performance outcomes. *Peak KAM* relative to body mass was calculated over the *weight acceptance phase* according to previous research (Besier et al., 2001), and was found to be 0-23% ground contact, on average, in the present study.

Quantification of whole-body dynamic stability

Factors that influence the medial GRF vector were calculated to represent whole-body dynamic stability. The first priority was to calculate foot placement, which would influence the origin of the GRF vector, and represents the dynamic relationship between the CoM and base of support. Firstly, the 'extrapolated' CoM (XCoM) was calculated according to Hof (2008) (see Equation 1).

Equation 1:

$$\mathrm{XCoM} = \mathrm{pCoM} + \frac{\mathrm{vCoM}}{\sqrt{\mathrm{gl}^{-1}}}$$

Where *pCoM* is the M-L position of the CoM, *vCoM* is the M-L velocity of the CoM, *g* is gravity, and *l* is the distance between the CoM and ankle in the frontal plane. The first whole-body dynamic stability variable - (1) M-L *foot placement* – was calculated as the position of the XCoM relative to the fifth metatarsal head (MTH5) which was indicative of the lateral border of the foot. In this case, a positive value for foot placement would indicate the XCoM is medial to the planted foot, whilst a negative value would indicate the XCoM is lateral and considered outside of the base of support. The second variable - (2) *position of the CoP* - was calculated as the origin of the GRF vector under the planted foot – again, measured relative to MTH5, but unlike foot placement position of the CoP was measured across ground contact.

IAA modelling, explained in detail elsewhere (Kepple et al., 1997; João et al., 2014; Moniz-Pereira et al., 2018), was conducted in Visual 3D to determine all non-negligible (>10N) contributions to the medial GRF. Non-negligible contributions to the M-L GRFs were found in sagittal plane hip, knee and ankle joints, and also in the frontal and transverse planes for the hip joint. Those contributions were then consolidated in their respective planes to represent the third, fourth, and fifth whole-body dynamic stability movement strategies: (3) *Sagittal triple acceleration* (the sum of the sagittal plane hip, knee and ankle joint contributions); (4) *frontal plane hip acceleration*; (5) *transverse plane hip acceleration* (see Figure 2.1 diagram). Following previous research (Kepple et al., 1997; João et al., 2014; Moniz-Pereira et al., 2018) the error of IAA was determined by finding the absolute mean difference of CoM acceleration from the force plate ground reaction forces and those derived from the sum of all joint contributions in IAA. The difference was then represented as a percentage of the maximum force obtained from the force plate - in this case the mean error for medio-lateral IAA was found to be 7%. The mean error in the current study is comparable to the 4.8% and 5.4% mean vertical error reported for stair ambulation and hopping tasks, respectively.

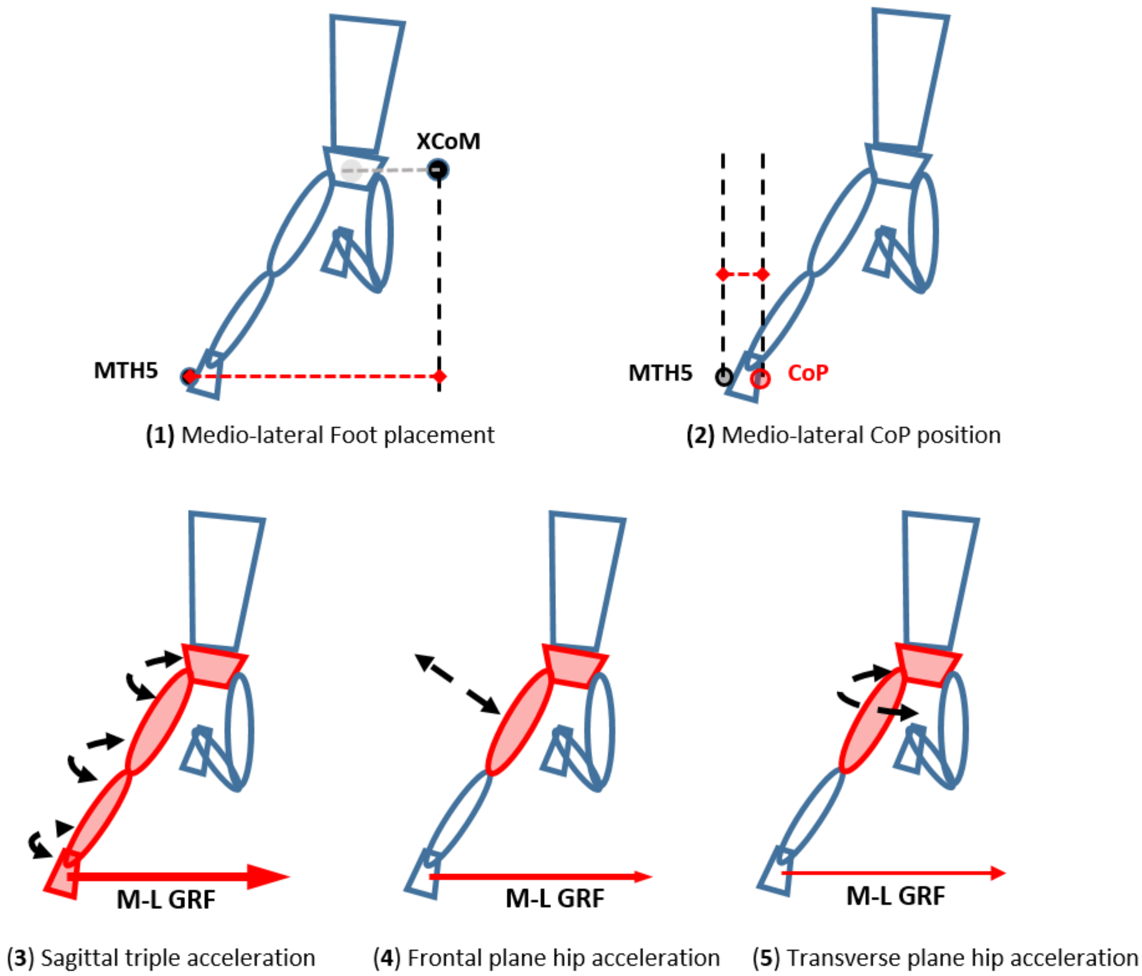

Figure 2.1. Diagram of the five distinct whole-body dynamic stability movement strategies for medio-lateral control of the CoM.

Statistical Analysis

Following normality tests, we ran Pearson's correlations for participant mass, height, and touchdown speed against our selected performance outcomes and peak KAM (IBM SPSS

Statistics, v23, Chicago, USA). This allowed us to investigate the impact of typical sources of between-individual (inter-individual) variability that may remain within the boundaries of the side cutting task exclusion criteria. Subsequently, all further statistical analyses were computed in SPM1D (v0.4, www.spm1d.org) using Python (Python v2.7.1 Enthought Canopy, v1.6.2, Enthought Python Distribution, Austin, TX, USA), and using Statistical Parametric Mapping (SPM) (Pataky, 2012) for regressions involving 1D time-series data.

Using SPM1D, non-parametric linear regression analyses were computed to investigate within-individual (intra-individual) variations in task execution. The regression analyses were similar to that previously reported (Vanrenterghem et al., 2012). Fifteen linear regression analyses were conducted for each combination of the three 0D independent variables - change of direction angle, average M-L CoM acceleration and peak KAM - regressed against the five whole-body dynamic stability dependent variables. Alpha was adjusted *a priori* from $\alpha = 0.05$ to $\alpha = 0.003$, using a Bonferroni correction based on the number of variables.

In further analysis, we noted that it may be possible to represent the extent of the sagittal and non-sagittal contributions to change of direction in a single metric, which may be a useful reference for practitioners when monitoring effectiveness of intervention strategies. To quantify this observation specifically, we calculated the sagittal triple acceleration impulse and total M-L force impulse, and expressed the former as a percentage of the latter, representing a *Sagittal Efficiency Ratio*. A *Sagittal Efficiency Ratio* of 100% would mean the impulses were equal and the medial CoM acceleration was entirely sagittal. However, lower or higher than 100% would mean non-sagittal movements were involved in generating (increasing) or moderating (reducing) medial CoM acceleration, respectively.

Results

On average, the resultant CoM velocity at touchdown was slightly lower than the required 4-5 ms^{-1} threshold, followed by a small increase in velocity by toe-off (see Table 2.1). The change of direction angle was also below the intended 45° at 20.6°, on average (see Table 2.1). No significant correlations were found between participant mass, height, and approach speed, and the selected performance outcomes or peak KAM.

Table 2.1. Side cutting performance outcome variables and peak knee abduction moment – means are presented with standard deviations (*SD*).

Performance outcome/ joint loading variable	**Unanticipated side cutting**
Touchdown Velocity (ms^{-1})	3.95
±*SD*	*0.30*
Toe-off Velocity (ms^{-1})	4.00
±*SD*	*0.24*
Change of direction angle (°)	20.6
±*SD*	*3.2*
Average medial CoM acceleration (ms^{-2})	4.91
±*SD*	*0.91*
Contact time (s)	0.28
±*SD*	*0.03*
Peak KAM (Nm/kg)	0.44
±*SD*	*0.25*

* 'CoM' denotes centre of mass; 'KAM' denotes knee abduction moment

In the 15 regression outputs, negative betas indicated negative relationships, whilst positive betas indicate positive relationships between the independent and dependent variables. The average foot placement was found to be 0.428 ± 0.059 m representing the medial distance from the MTH5, on the lateral border of the foot, to the XCoM. A narrower foot placement (see Table 2.2), followed by a more lateral position of the CoP during contact, seems to increase average medial CoM acceleration and change of direction angle to a lesser extent (see Figure 2). However, our findings suggest a narrower foot placement in particular, may also lead to greater peak KAM (see Table 2.2).

Table 2.2. Summary of general findings of the multiple linear regression analyses conducted in SPM1D. The significance of each regression between pairs of variables is presented, and when $p<0.003$ the direction of the relationship is also presented in parenthesis ('-ve' = variables have a negative relationship; '+ve' = variables have a positive relationship).

	Selected side cutting performance outcome variables		
Whole-body dynamic Stability variables	Average medial CoM acceleration (TD-TO) ^{0D}	Change of direction angle (TD-TO) ^{0D}	Peak KAM (weight acceptance phase) ^{0D}
(1) Foot Placement ^{0D}	p = 0.008	p = 0.001* (-ve)	p = 0.001* (-ve)
(2) M-L Centre of Pressure (CoP) position ^{1D}	p < 0.003* (-ve)	p = 0.001* (-ve)	p > 0.003
(3) Sagittal triple acceleration ^{1D}	p < 0.003* (+ve)	p < 0.003* (+ve)	p < 0.003*(+ve)
(4) Frontal plane hip acceleration ^{1D}	p > 0.003	p < 0.003* (+ve)	p < 0.003* (+ve)
(5) Transverse plane hip acceleration ^{1D}	p > 0.003	p < 0.003* (+ve)	p < 0.003* (+ve)

'*' = significance ($\alpha = 0.003$); '^{0D}' = 0-dimensional data; '^{1D}' = 1-dimensional (time-series) continua. Beta regression data are presented in Appendix 6 page 175 for variable 1, and single subject examples are presented in the second column in figures 2-5 for mechanisms 2-5.

Increases in *sagittal triple acceleration, frontal plane hip acceleration* and *transverse plane hip acceleration* are all related to a greater change of direction angle, however, this is at the expense of increased undesirable joint moments (see Table 2.2 and Figures 2.3-2.5). An increase in *sagittal triple acceleration* aligns with increases in average medial CoM acceleration, which was more pronounced later in ground contact. The positive relationship between frontal plane hip acceleration and change of direction angle was observed despite the fact that this strategy appears to be almost exclusively for creating lateral, or unloading, ground reaction forces (see Figure 2.4). Transverse plane hip acceleration appears to alternate between a loading and unloading role through medial then lateral ground reaction forces over ground contact (see Figure 2.5). The contribution to medio-lateral forces from sagittal triple acceleration were typically in excess of total medio-lateral forces over ground contact (see Figure 2.6). The *Sagittal Efficiency Ratio* was 131.6 ± 30.3 %, indicating that impulses from sagittal triple acceleration were excessive, on average, nearly 32 % greater than the total medio-lateral impulses.

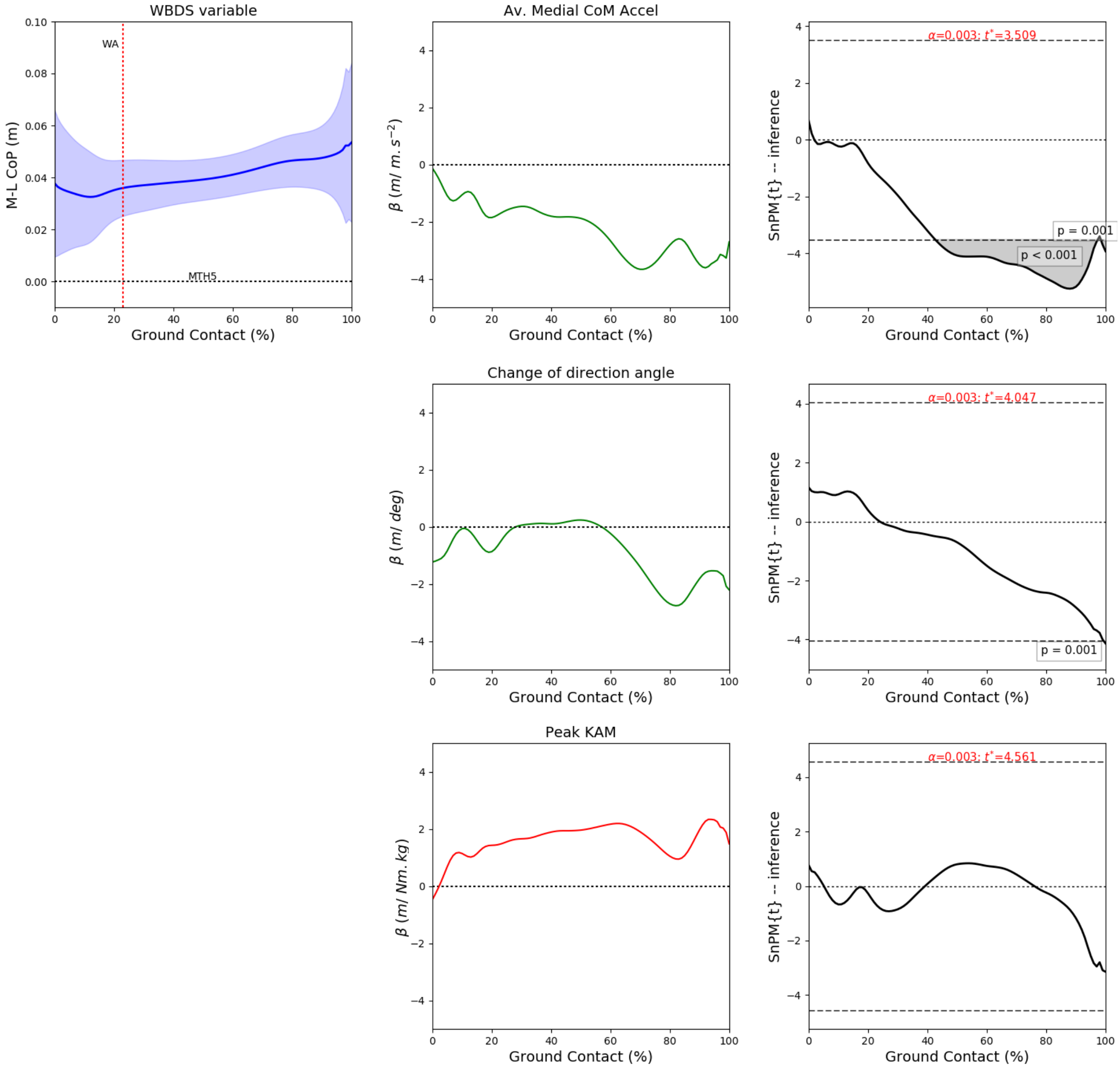

Figure 2.2. Characterisation of the relationship between whole-body dynamic stability (WBDS) variable *M-L CoP position* and average medial CoM acceleration, change of direction angle, and peak knee abduction moment (Peak KAM). Row 1, column 1 shows the mean and standard deviation of the time-series M-L *CoP position* signals, lateral border of the foot is represented by dotted line and label at position '0.00' on the y-axis highlighting the position of metatarsal head 5 (MTH5); weight acceptance (WA) is indicated by the vertical line at 23% ground contact. Row 1 column 2 shows the beta curves in regression against average medial CoM acceleration; then the row 1 column 3 shows the one sample t-test statistical curve (SnPM{t}), where α = 0.003, with inference boundaries and p values for significance clusters, where applicable. Columns 2 and 3 are repeated for change of direction angle and Peak KAM on rows 2 and 3, respectively. Example beta regression curves are presented in column 2: green for selected performance outcomes, and red for peak KAM.

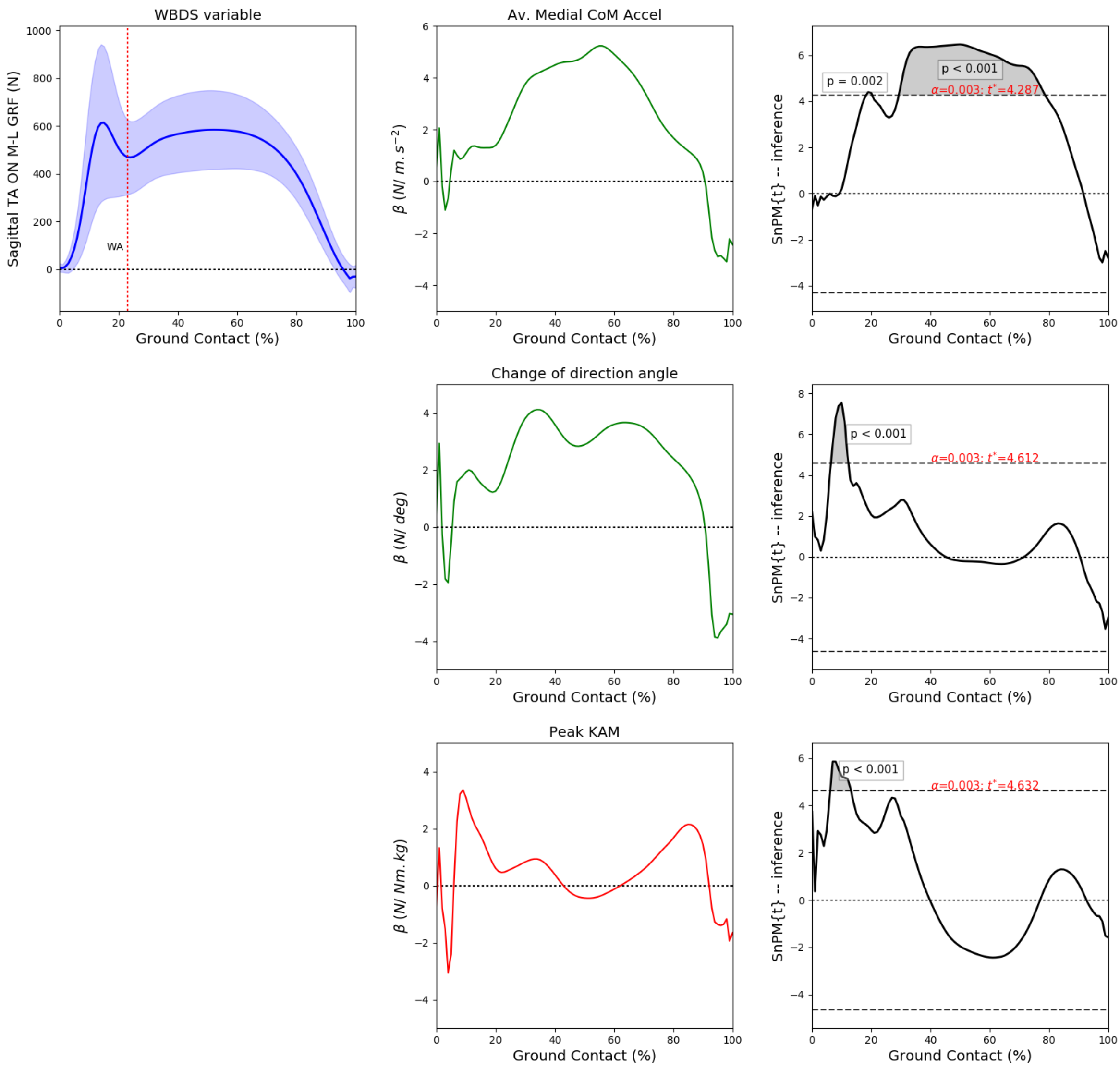

Figure 2.3. Characterisation of the relationship between whole-body dynamic stability (WBDS) variable *sagittal triple acceleration (TA)* contribution to M-L GRF and average medial CoM acceleration, change of direction angle, and peak knee abduction moment (Peak KAM). Row 1, column 1 shows the mean and standard deviation of the time-series Sagittal TA signals; weight acceptance (WA) is indicated by the vertical line at 23% ground contact. Row 1 column 2 shows the beta curves in regression against average medial CoM acceleration; then the row 1 column 3 shows the one sample t-test statistical curve (SnPM{t}), where α = 0.003, with inference boundaries and p values for significance clusters, where applicable. Columns 2 and 3 are repeated for change of direction angle, then Peak KAM on rows 2 and 3, respectively. Example beta regression curves are presented in column 2: green for selected performance outcomes, and red for peak KAM.

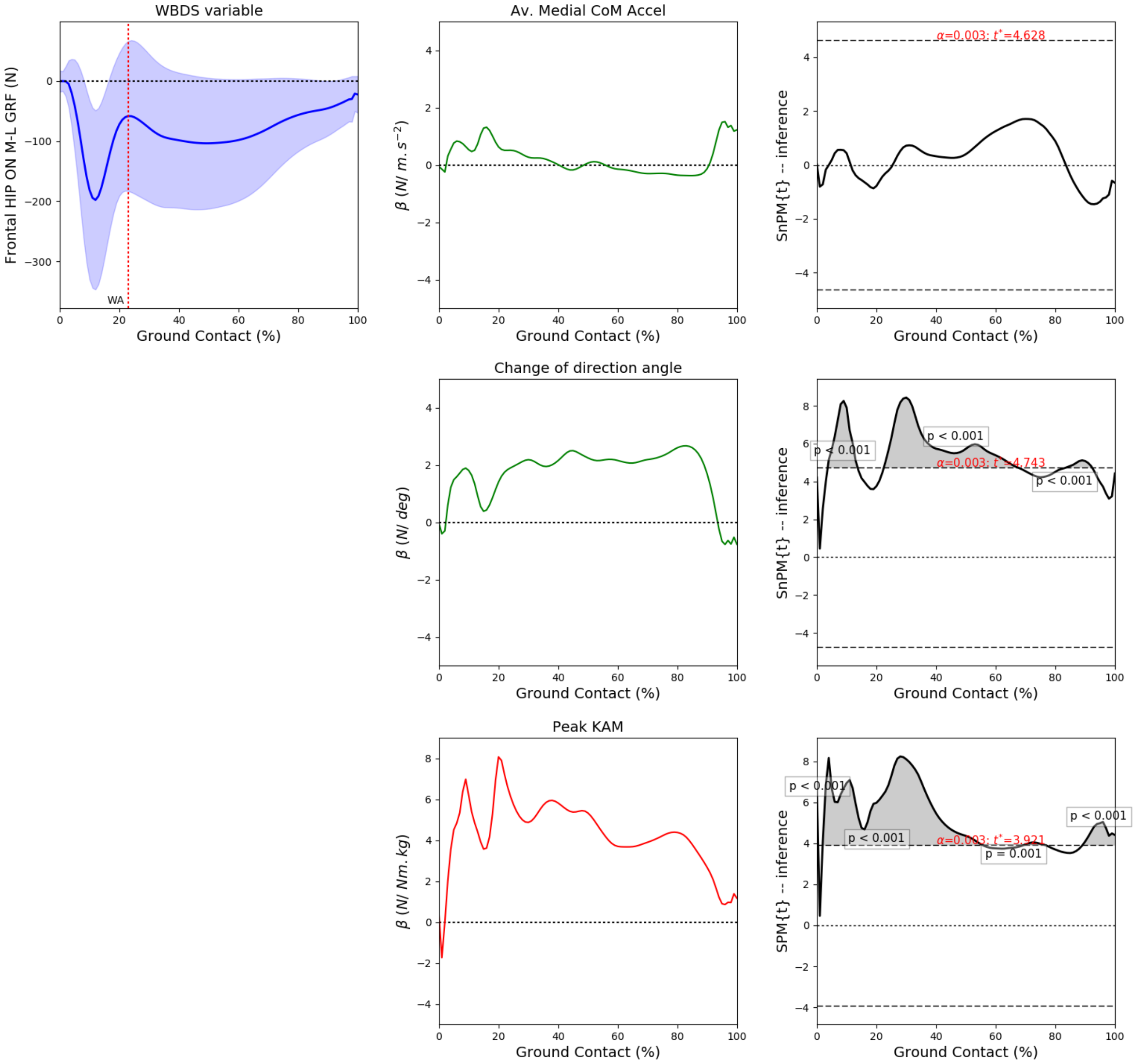

Figure 2.4. Characterisation of the relationship between whole-body dynamic stability (WBDS) variable *frontal plane hip acceleration* contribution to M-L GRF and average medial CoM acceleration, change of direction angle, and peak knee abduction moment (Peak KAM). Row 1, column 1 shows the mean and standard deviation of the time-series frontal plane hip acceleration signals; weight acceptance (WA) is indicated by the vertical line at 23% ground contact. Row 1 column 2 shows the beta curves in regression against average medial CoM acceleration; then the row 1 column 3 shows the one sample t-test statistical curve (SnPM{t}), where $\alpha = 0.003$, with inference boundaries and p values for significance clusters, where applicable. Columns 2 and 3 are repeated for change of direction angle, then Peak KAM on rows 2 and 3, respectively. Example beta regression curves are presented in column 2: green for selected performance outcomes, and red for peak KAM.

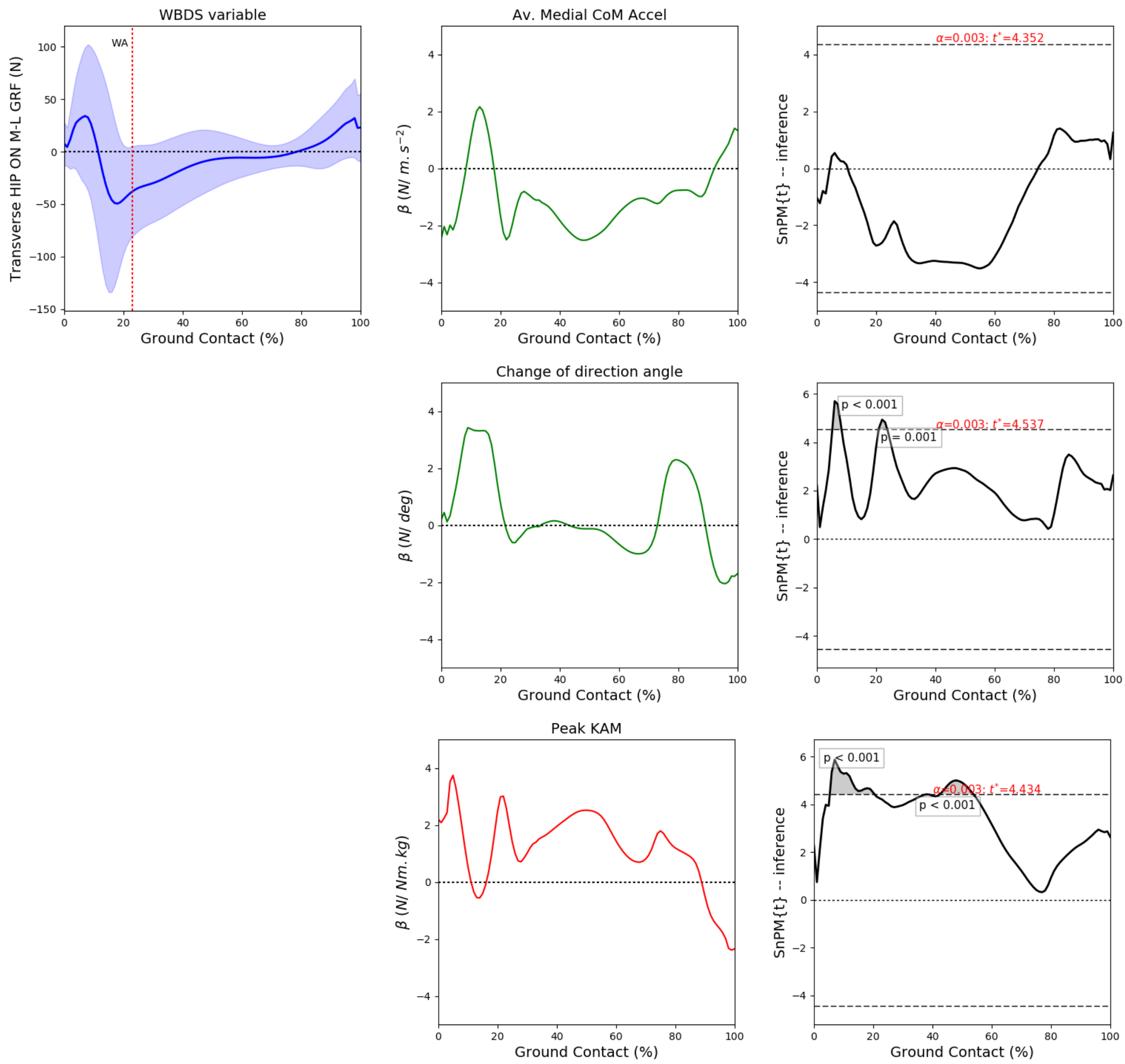

Figure 2.5. Characterisation of the relationship between whole-body dynamic stability (WBDS) variable *transverse plane hip acceleration* contribution to M-L GRF and average medial CoM acceleration, change of direction angle, and peak knee abduction moment (Peak KAM). Row 1, column 1 shows the mean and standard deviation of the time-series transverse hip acceleration signals; weight acceptance (WA) is indicated by the vertical line at 23% ground contact. Row 1 column 2 shows the beta curves in regression against average medial CoM acceleration; then the row 1 column 3 shows the one sample t-test statistical curve (SnPM{t}), where $\alpha = 0.003$, with inference boundaries and p values for significance clusters, where applicable. Columns 2 and 3 are repeated for change of direction angle, then Peak KAM on rows 2 and 3, respectively. Example beta regression curves are presented in column 2: green for selected performance outcomes, and red for peak KAM.

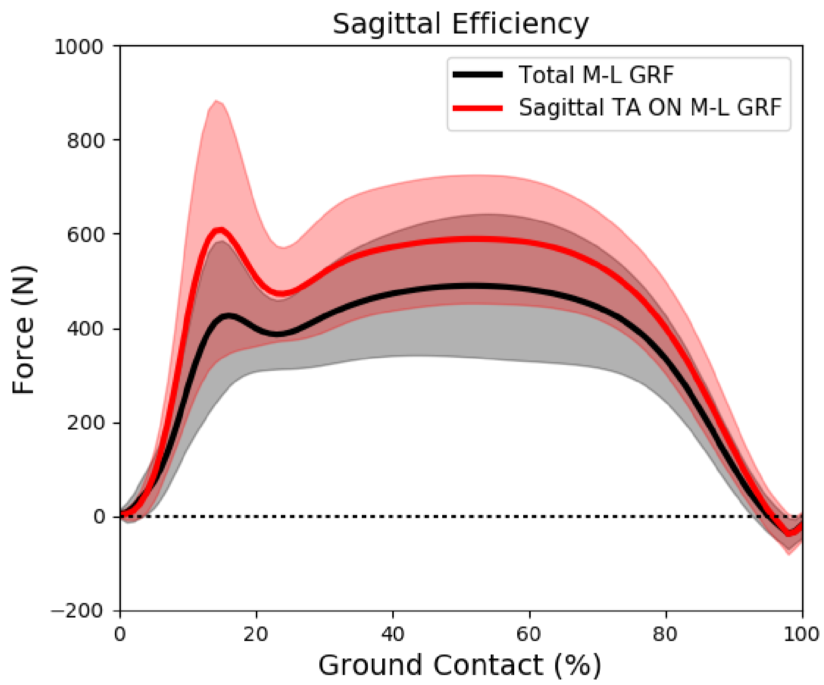

Figure 2.6. Comparison of the total medio-lateral ground reaction forces (Total M-L GRF) and the sagittal triple acceleration contribution to medio-lateral ground reaction forces, estimated by induced acceleration analysis, representing Sagittal Efficiency in whole-body dynamic stability.

Discussion

The aims of this investigation were to outline mechanical movement strategies that are integral to the medial control of the CoM in side cutting, and thereby represent the condition of whole-body dynamic stability. Furthermore, we aimed to explore the influence of those specific movement strategies on redirecting and accelerating the CoM, and undesirable, potentially injurious, joint moments. Our investigation has allowed us to express systematic movement strategies that each fulfil different roles to achieve whole-body dynamic stability in unanticipated side cutting. Our findings have confirmed the hypothesis that movement strategies to increase change of direction angle and acceleration are also associated with increased peak KAM. Therefore, our findings may offer new understanding of the performance-injury trade-off in unanticipated side cutting. Specifically, we have found a narrower foot placement, along with high sagittal plane loading, are beneficial for performance aspects. However, sagittal plane strategies for generating medial forces are often excessive and inefficient, as expressed in the Sagittal Efficiency Ratio, which likely leads to destabilisation of the body. Such destabilisation requires corrective non-sagittal movement strategies and may result in higher peak KAM, and frontal plane hip acceleration appears to fulfil this important role.

The status of whole-body dynamic stability is initially determined by foot placement, and in this study a narrower foot placement may represent a more unstable initial condition (reduced medial distance from the foot to the XCoM). It has been suggested that a wider foot placement may be better for change of direction (Jones et al., 2015); however, this may not be possible in unanticipated side cutting. Furthermore, although reducing stance width may be a way to reduce harmful peak KAMs in various side cutting tasks (Dempsey et al., 2009; Kristianslund et al., 2014; Havens and Sigward, 2015a; Jones et al., 2015), our findings suggest this movement strategy is insufficient on its own for control of the CoM in unanticipated side cutting.

Following foot placement, excessive sagittal forces, as evidence by the 132% Sagittal Efficiency Ratio, risk destabilisation of the CoM, jeopardising whole-body dynamic stability and failure of the task. However, those excessive forces were moderated by frontal plane hip acceleration, which acts in countermovement to the medial forces, and more prominently so in weight acceptance. Whilst previous studies have reported the negative effects on peak KAM of a laterally flexed trunk (Dempsey et al., 2009; Jamison et al., 2012; Kristianslund et al., 2014; Jones et al., 2015), we have been able demonstrate that frontal plane hip acceleration may be the direct movement strategy at work here. Moreover, the countermovement may be necessary to control the CoM, and sufficient enough to engage the hip in transverse plane hip acceleration, which may explain a re-orientation of the pelvis. Later in ground contact, the role of hip movement strategy diminishes, and this appears to make way for an ankle movement strategy to take over. Specifically, our results suggest from ~43% ground contact a more lateral CoP position, which is likely due to inversion of the subtalar joint, increases the ability to accelerate the CoM medially. Perhaps this is evidence of a double pendulum interaction between hip and ankle movement strategies at work in the frontal plane, as previously reported for a range of tasks (MacKinnon and Winter, 1993; Winter, 1995; Houck et al., 2006).

On average participants' final approach velocity and change of direction angle was lower than expected when the CoM trajectory was analysed directly, despite participants apparently meeting the predetermined constraints at the time of data collection in the lab. However, this limitation is a frequent observation in the literature for change of direction angle (Dos'Santos et al., 2018). Observation of the preceding steps may clarify other braking characteristics; nonetheless, our findings should be interpreted in light of the mean approach velocity we reported, which was around 4 m.s^{-1}. The use of IAA in decomposition of movement dynamics continues to be a source of some debate (Chen, 2006; João et al., 2014), which is beyond the scope of this study. However, we feel this approach offers important insights into mechanics in the kinetic chain, and we have provided a calculation of the error of the induced acceleration model found in this study, which is comparable to previous examples mentioned earlier.

Conclusion

This study provides insights into the movement strategies used to achieve whole-body dynamic stability in unanticipated side cutting. Our findings reveal important non-sagittal corrective hip movements are essential to retrieve control of the CoM in the presence of otherwise excessive destabilising sagittal forces. Whilst a purely sagittal pogo stick like movement strategy may be the most efficient aim for dynamic changes of direction, this may not be possible in practice. However, practitioners looking to improve change of direction performance of their athletes may focus on the sagittal efficiency as their first priority. Reducing corrective frontal plane movement strategies may only be possible when this first priority is addressed. More holistic intervention strategies should consider an integrated approach to training and monitoring of foot placement, sagittal plane loading, and frontal plane hip engagement. In this study, we have been able to demonstrate a direct method for monitoring the necessary interlinked movement strategies and the status of whole-body dynamic stability in side cutting, which may also be applicable to other dynamic tasks.

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Study 3

Anticipatory effects on whole-body dynamic stability and adaptability in side cutting

Abstract

Side cutting efficiency is determined by deployment of movement strategies that allow control of the Centre of Mass (CoM) and therefore *whole-body dynamic stability*. However, whole-body dynamic stability may become jeopardised when an external trigger in the last steps prior to the manoeuvre limits the available time to deploy appropriate movement strategies. This likely leads to poorer task performance and potentially undesirable joint loading compared to anticipated side cutting. Observation of the anticipatory effects on movement strategies and their adaptability may improve our understanding of whole-body dynamic stability and better inform injury prevention and performance enhancement interventions. Twenty recreational soccer players completed 45° anticipated (ANT) and unanticipated (UNANT) side cutting with a 4-5 $m.s^{-1}$ approach speed whilst 3D motion capture and ground reaction force data were collected. Kinematics and kinetics, and task execution variables were calculated using a lower limb and trunk model. Performance outcomes, peak knee abduction moments, and whole-body dynamic stability movement strategies were calculated for each side cutting task with an average of 12 trials per participant, per condition, which were randomised and counterbalanced. Five distinct whole-body dynamic stability movement strategies were identified, based on factors that influence the medial ground reaction force (GRF) vector during ground contact in the side cutting manoeuvre. Multiple t-tests were conducted using Statistical Parametric Mapping to investigate the differences between ANT and UNANT side cutting whole-body dynamic stability, task performance outcomes, and joint loading variables. UNANT side cutting was performed significantly slower, with a significantly longer contact time, but with a sharper change of direction angle than ANT side cutting. There was no significant difference in peak knee abduction moment between UNANT and ANT conditions, but there were significant differences for all whole-body dynamic stability movement strategies between UNANT and ANT side cutting conditions. Peak knee abduction moment and average medial acceleration

of the centre of mass did not change significantly in side cutting tasks with different anticipatory demands. Although, participants were able to make a sharper change of direction with limited anticipation, which suggests participants were able to adapt their movement strategies to the anticipatory demands. However, there are signs of anticipatory postural adjustments which indicate the unanticipated task is becoming more challenging, and these adjustments seem to form part of a corrective movement strategy to maintain control of the CoM. With limited anticipation participants are force into a narrower foot placement, and subsequently the sagittal triple acceleration forces become more excessive and inefficient, possibly leading to destabilisation of the body. Consequently, greater frontal plane hip acceleration was required to counteract the destabilisation. This adjustment appears to be beneficial by mitigating the transference of high ground reaction forces to undesirable knee moments that may lead to increased injury risk.

Introduction

Side cutting efficiency is determined by deployment of movement strategies that allow control of the centre of mass (CoM), whilst mitigating unnecessary deviations and therefore providing an indication of the condition of *whole-body dynamic stability*. Typically side cutting tasks in sport are triggered by external stimuli such as movements of other players, and this can influence the time the performer has to deploy the appropriate movement strategies (Besier et al., 2001b; Houck, Duncan and De Haven, 2006; Mornieux et al., 2014). If external stimuli become challenging, one may see failures in those movement strategies, and thus, failures in *whole-body dynamic stability*. This can in turn lead to potentially dangerous movement deviations like those reported for Anterior Cruciate Ligament (ACL) injury (Hewett et al., 2005; Weinhandl et al., 2013; Brown, Brughelli and Hume, 2014; Almonroeder, Garcia and Kurt, 2015). However, the extent to which the various mechanisms that contribute to whole-body dynamic stability are affected by the level of anticipation remains unknown.

Quantification of whole-body dynamic stability in side cutting is challenging, as control of the CoM involves several interactive movement strategies, to correctly perform the task in a fraction of a second (David et al., 2017). Research involving visually cued changes of direction in walking suggests medio-lateral (M-L) control of the CoM is the first mechanical priority of the performer (Patla, Adkin and Ballard, 1999). The M-L demand in side cutting is greater than in walking (Houck, Duncan and De Haven, 2006), and accelerating the CoM in the new direction of travel requires high M-L force generation and control (Donnelly et al., 2012). The characteristics of the M-L force vector can be expressed in terms of the moment arm and magnitude, and both have been expressed in summative terms previously, typically using measurements of stance width and peak force or impulse, respectively (Kristianslund et al., 2014; Havens and Sigward, 2015; Jones et al., 2015). However, in summative observations one misses the time-varying aspect of such characteristics, which may be necessary for a full account of anticipatory differences considering the dynamic nature of the side cutting task. Furthermore, whilst the general extent of the M-L ground reaction force (GRF) may have been reported (Kristianslund et al., 2014), the important movements that contribute to its magnitude have not. To the best of our knowledge, this is certainly the case in comparative analysis between tasks of different anticipatory demands. The relatively novel approach of Induced Acceleration Analysis may be a useful way to calculate the joint and planar contribution to M-L GRF (Kepple, Siegel and Stanhope, 1997; João et al., 2014; Moniz-Pereira et al., 2018), and thus, quantify the medio-lateral control of the CoM.

Therefore, the aim of this study was to quantify the anticipatory effects on whole-body dynamic stability movement strategies in side cutting, whilst expressing the consequences for performance and undesirable knee joint moments. It is hypothesised that reduced preparation time, represented by an unanticipated external stimulus, will: 1, Result in significant increases in undesirable knee joint moments and significantly poorer performance of the side cutting task; and 2, Demonstrate significant differences in the deployment of whole-body dynamic stability movement strategies.

Methods

Participants

The participants in this study were twenty healthy male recreational soccer players, with at least 6 years playing experience, consisting of between one and two sessions a week, for one to two hours per session. Participants had a mean (± SD) age of 23 ± 3 years; mean height of 1.8 ± 0.1 m; and mean mass of 76.7 ± 10.4 kg. All participants were free from injury for at least 6 months, and written consent was provided. All participant recruitment processes were conducted in line with the university research ethics committee guidelines, which comply with the principles of the Declaration of Helsinki.

Experimental design

A repeated-measures design was used in which participants initially completed a one hour familiarisation session, where they were introduced to the laboratory environment and the *unanticipated* (UNANT) and *anticipated* (ANT) side cutting tasks. Participants were also introduced to the marker model that would be used for 3D motion capture and were allowed to practice the task in mock testing conditions. Due to the requirement to randomised trial data collection, and the available time with each participant, collection of trials continued until a minimum of 12 valid trials per condition were recorded (see details later). ANT (pre-planned) side cutting trials were randomised and counterbalanced with trials from two UNANT conditions – *open* side cutting and *crossover* side cutting - using an external light stimulus. Each participant nominated their preferred limb, which they later used to complete all three conditions for all side cutting trials. The experimental testing session took place at the same time of day as the familiarisation session, and no more than seven days after that initial session.

Protocol – Side cutting assessment

In the experimental testing session participants performed a dynamic warm-up - comprising body weight squats, over-ground shuttle runs, as well as specific side cutting practice. They then completed a static trial and functional hip joint centre and knee joint axis tasks in the centre of the capture volume. Following calibration trials the motion trials were then collected. The one ANT and two UNANT 45° side cutting conditions began with a 10 m approach to the force plate, and the approach speed was controlled to 4-5 $m.s^{-1}$, with timing gates (Smartspeed™, Fusion Sports, Australia) set 2 m apart at 5 m and 3 m away from the force plate. The preferred leg was used for all side cutting in both directions – for example, if right leg preferred, the participant would make an *open* side cutting manoeuvre to the left, and a *crossover* side cutting manoeuvre to the right, and vice versa if left leg preferred. ANT side cutting was always *open* and not dependent on a cueing light, and UNANT side cutting was either *open* or *crossover*, depending on the light stimulus flashing on the right or left. The cueing light units used to indicate the direction of the two UNANT conditions were set up 3 m beyond the force plate, 1 m in height from the ground, and 2 m apart. To promote consistency the left and right cutting directions were identified by 2 m wide tracks, set-up 45° in each direction, ahead of the approach direction, beyond the force plate. An image of the lab set-up is presented in figure 3.1. The light stimulus was cued by the timing gate set-up 3 m from the force plate and included a 0.1 s consistent delay between triggering and onset. Considering the 4-5 ms^{-1} approach speed, this gave the participants 0.5-0.65 seconds to react and execute the side cutting task. The Smartspeed™ system allowed manual programming of the corresponding directions of the cueing light stimulus, and was used in conjunction with a 6x6 Latin Square, so 36 trials and three conditions were randomised and counterbalanced. When participants failed to meet the criteria set for any side cutting task, that trial was discarded, and an additional trial was added to the trial count. Once 12 good side cutting trials were collected, per condition, the participants completed a 10-15 minute cool-down protocol.

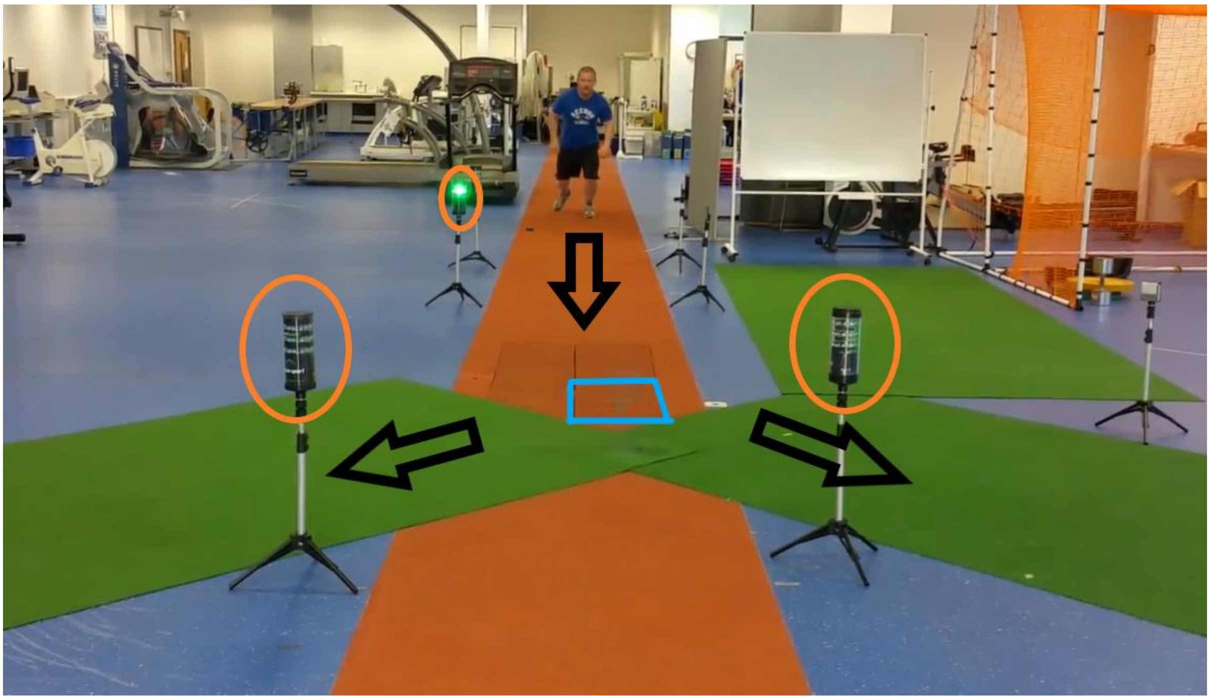

Figure 3.1. Example of the lab set-up for anticipated (ANT) and unanticipated (UNANT) side cutting trials. Approach and side cutting direction are indicated by the arrows; the force plate is highlighted by the blue box; and the orange ovals highlight the trigger light unit and the UNANT cueing stimulus light units.

Biomechanical model

All participants had 44 reflective markers captured based on the Liverpool John Moores University (LJMU) Lower Limb and Trunk (LLT) eight segment model (Vanrenterghem et al., 2010; Malfait et al., 2014). Markers were applied to participants before a 15-minute dynamic warm-up, and bandages and strapping used to attach cluster plates on lower limb segments were adjusted for comfort, without compromising a secure fitting. 3D Marker trajectories were recorded using a 7-camera Vicon T40S and Vicon MX system, controlled by Vicon Nexus version 1.8.5 software (Vicon, Oxford Metrics, Oxford, UK). 3D marker trajectory data were recorded at 250 Hz for the side cutting trials, then later processed and labelled using the same software. The side cutting tasks were executed on a 0.6 x 0.4 m force platform (Kistler, Winterthur, Switzerland), and data were sampled at 1500 Hz and synchronised with the Vicon system. Calibration, modelling, and all kinematic and kinetic analyses were completed in Visual 3D Professional (v.5.00.16, C-Motion, Germantown, MD, USA). Inverse kinematic (IK) modelling was used to constrain all translational motion of the hip, knee and ankle, as well as some rotational constraints. Specifically, this left hips with three; the ankle with two (sagittal, transverse); and the knee with one (sagittal) rotational degrees of freedom. The IK modelling restrictions were matched to the requirements of Induced Acceleration Analysis, described briefly later.

Data processing

Only the *open* side cutting trials from both the ANT and UNANT were analysed. Not all 12 trials in each condition were always useable, due to some marker trajectory losses beyond recovery, leaving 230 out of 240 trials for the ANT condition; 231 out of 240 trials for the UNANT condition. Analogue force plate signals and 3D marker trajectories were filtered with a Butterworth 4th order recursive low pass filter, with a 20 Hz cut-off frequency (Kristianslund, Krosshaug and van den Bogert, 2012). Net joint moments were estimated through inverse dynamics (cardan sequence, XYZ). Initial foot contact with the ground, or *touchdown* (TD), was represented as the minima prior to an ascending vertical GRF gradient; and the *toe-off* (TO) event was represented by a minima following a descending gradient of the same vertical component of GRF. Centre of Mass (CoM) was calculated for every instant across the side cutting task, then CoM velocity was calculated as the first derivative. The following performance outcomes were calculated using trigonometry: CoM trajectory angle (transverse plane); CoM trajectory velocity (transverse plane); in addition to separated anterior and lateral components of CoM velocity. These performance outcomes were specified discretely at the two side cutting events, TD and TO, and the total change between TD and TO calculated. *Change of direction angle* was calculated as the change in CoM trajectory angle. Changes in M-L CoM velocity, once divide by ground *contact time*, represented the *average medial CoM acceleration*. Peak knee abduction moment (*peak*

KAM) was calculated over the 'weight acceptance phase', which is categorised in previous research (Besier et al., 2001a).

Quantification of whole-body dynamic stability

The expression of five *whole-body dynamic stability* mechanisms are described in detail in the previous chapters (General Methods and Study 2), and represent the control of the CoM through factors that influence the medio-lateral component of the GRF vector. Briefly, five distinct movement strategies were calculated: (1) M-L foot placement and (2) M-L CoP position, and (3) sagittal triple acceleration, (4) frontal plane hip acceleration and (5) transverse plane hip acceleration. Variables 3-5 were determined by non-negligible (>10 N) contribution to the M-L GRF using IAA in Visual 3D software (Kepple, Siegel and Stanhope, 1997; João et al., 2014; Moniz-Pereira et al., 2018), and consolidated into the respective planes. Following examples in previous research (Kepple, Siegel and Stanhope, 1997; João et al., 2014; Moniz-Pereira et al., 2018) the accuracy of IAA was determined by finding the absolute mean difference between the force platform ground reaction forces and those derived from IAA. The difference was then represented as a percentage of the maximum force obtained - in this case the mean error for medio-lateral IAA was found to be 7%. The extent of the excessive medial forces from sagittal plane contributions we determined by calculating the sagittal triple acceleration impulse as a proportion of the total medio-lateral force impulse, and thus representing a *Sagittal Efficiency Ratio*. A *Sagittal Efficiency Ratio* of 100% would mean the impulses were equal; lower than 100% would mean the total impulses are greater; whilst a ratio greater than 100% would indicate the sagittal triple acceleration impulses are greater than the total medio-lateral impulses.

Statistical Analysis

Initially, in SPSS version 23 (IBM SPSS Statistics, Chicago, USA) we ran Shapiro Wilks normality test, then either parametric or non-parametric paired samples t-tests between the selected performance outcome and knee joint loading variables comparing ANT with UNANT side cutting. Alpha was adjusted from $\alpha = 0.05$ to $\alpha = 0.008$, with Bonferroni correction for multiple comparisons set to three decimal places. Later the same process was followed comparing the Sagittal Efficiency Ratio between ANT and UNANT conditions.

All subsequent statistical analyses were computed using Statistical Parametric Mapping (SPM) (Pataky, 2012) in SPM1D (version 0.4) using Python (Python version 2.7.1 Enthought Canopy, version 1.6.2, Enthought Python Distribution, Austin, TX, USA). Normal distribution of all 0D and 1D signals were calculated with D'Agostino-Pearson's K2 test. Subsequently, either nonparametric or parametric paired t-tests were conducted for the five whole-body dynamic stability movement strategies to compare ANT and UNANT side cutting conditions. For five paired t-tests in SPM, alpha was adjusted from $\alpha = 0.05$ to $\alpha = 0.01$, with Bonferroni correction for multiple comparisons.

Results

Performance outcomes including key joint loading data are presented in table 3.1 comparing ANT and UNANT side cutting conditions. Touchdown and toe-off velocity identify how well the primary pre-set task constraint was met. On average UNANT side cutting was performed slower than the pre-set task constraints (4-5 $m.s^{-1}$) and significantly slower than ANT side cutting. Change of direction angle, average medial CoM acceleration, and contact time provide general information on how the side cutting performance may have been achieved. In both conditions the participants struggled to meet the change of direction angle

(45°), but made a significantly sharper change of direction in UNANT compared to ANT conditions. Contact time was significantly greater for UNANT compared to ANT side cutting, however, average medial CoM acceleration was not significantly different between conditions. There was no significant difference in peak KAM between UNANT and ANT conditions.

Table 3.1. Comparison of side cutting performance outcome variables over ground contact for ANT and UNANT side cutting. Means are presented with standard deviation [SD]. Parametric or non-parametric paired t-test results ($\alpha = 0.008$) are also identified.

Performance outcome variable	**ANT side cutting**	**UNANT side cutting**	**Statistical difference**
Touchdown Velocity (ms^{-1}) ±SD	4.33 [0.33]	3.95 [0.30]	*$p<0.001$
Toe-off Velocity (ms^{-1}) ±SD	4.38 [0.28]	4.00 [0.24]	*$p<0.001$
Change of direction angle (°) ±SD	17.26 [3.45]	20.64 [3.20]	*$p<0.001$
Av. medial CoM acceleration (ms^{-2}) ±SD	5.29 [1.16]	4.91 [0.91]	^{np} $p=0.012$
Contact time (s) ±SD	0.235 [0.02]	0.28 [0.03]	*$p<0.001$
Peak KAM (Nm/kg) ±SD	0.35 [0.29]	0.44 [0.25]	$p=0.062$

'*' denotes a significant difference in comparison between conditions; '^{np}' denotes a non-parametric Wilcoxon Signed Rank test

For the comparison of whole-body dynamic stability movement strategies between UNANT and ANT side cutting, firstly, we observed a significantly narrower stance (-0.11 m, $p < 0.001$), followed by a significantly more lateral position of the CoP, later on in ground contact ($p < 0.001$), in UNANT side cutting (see figure 3.2). Sagittal triple acceleration was significantly lower in UNANT side cutting compared to ANT (figure 3.3), specifically in the propulsive phase of ground contact ($p = 0.002$). However, later we found that the sagittal impulses were significantly less efficient in UNANT side cutting compared to ANT ($p < 0.001$), where the Sagittal Efficiency Ratio was 131.6 ± 30.3% and 113.4 ± 4.6% for UNANT and ANT side cutting, respectively (see figure 3.4). The most prominent difference between UNANT and ANT side cutting was observed with the frontal plane hip acceleration movement strategy. Here we see that acceleration of the hip in the frontal plane was used significantly more to unload medial GRF in UNANT compared to ANT side cutting ($p < 0.001$), across the majority of ground contact (figure 3.3). In fact this mechanism was used almost exclusively by all participants for unloading medial ground reaction forces in UNANT side cutting. Transverse plane hip acceleration may be used more for accelerating the CoM medially early in ground contact in UNANT side cutting compared to ANT, and significantly less for unloading in the propulsive phase ($p < 0.001$).

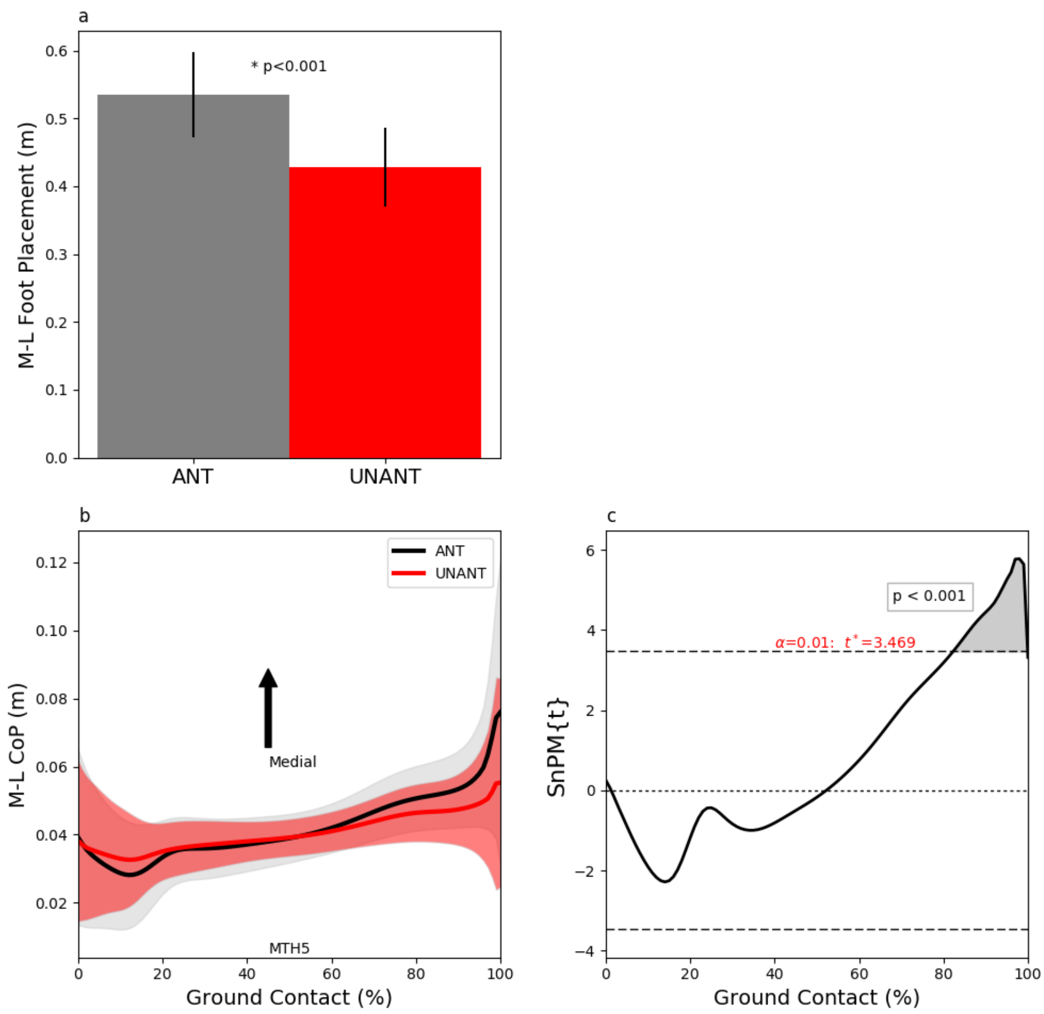

Figure 3.2. Comparison of (a) medio-lateral (M-L) foot placement, and (b) M-L centre of pressure (CoP) position between ANT and UNANT side cutting conditions. Means and standard deviations are presented for each variable discretely (a) or over ground contact (b). In image a and b '0.00' on the y-axis represents metatarsal head 5 (MTH5). Statistical differences are presented above the bar chart for M-L foot placement. For M-L CoP position, non-parametric (SnPM{t}) paired t-test results are presented in image c. All statistical comparisons were made in SPM1D and based on Bonferroni correction of alpha for multiple comparisons, $\alpha = 0.01$.

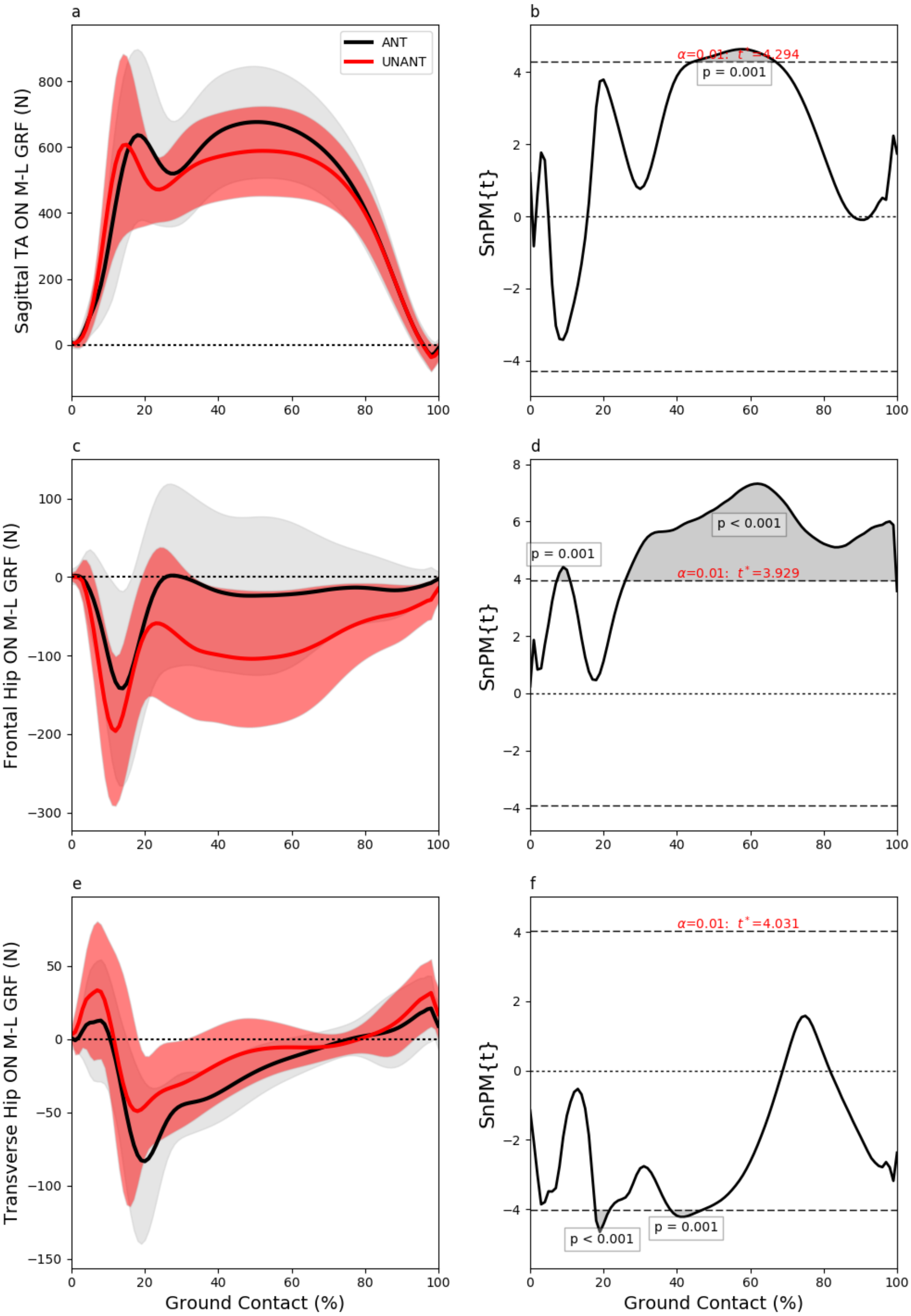

Figure 3.3. Comparison of sagittal triple acceleration, frontal plane hip acceleration and transverse plane hip acceleration contributions to medio-lateral (M-L) ground reaction force (GRF) across the side cutting task between *anticipated* (ANT) and *unanticipated* (UNANT) side cutting conditions are presented per row. Means and standard deviations are presented in column 1 and non-parametric (SnPM{t}) paired t-test results are presented in column 2. All statistical comparisons were made in SPM1D and based on Bonferroni correction of alpha for multiple comparisons, $\alpha = 0.01$.

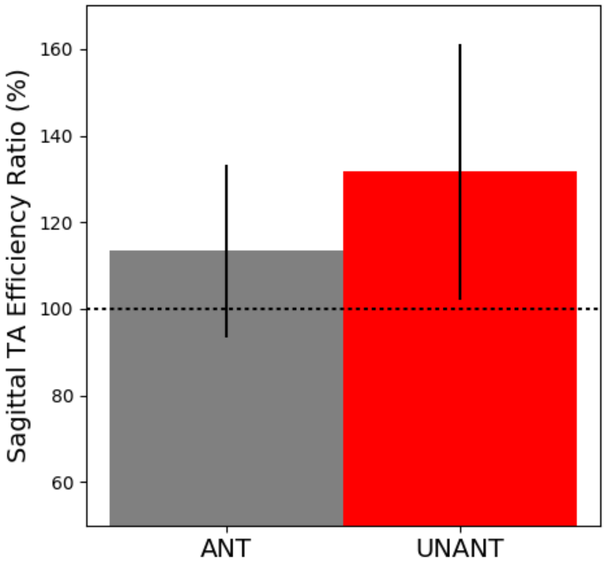

Figure 3.4. Comparison of Sagittal triple acceleration (TA) efficiency ratio between anticipated (ANT) and unanticipated (UNANT) side cutting conditions. A Sagittal Efficiency Ratio of 100% would mean the impulses were equal and the medial CoM acceleration was entirely sagittal. Lower or higher than 100% would mean non-sagittal movements were involved in generating (increasing) or moderating (reducing) medial CoM acceleration, respectively.

Discussion

The purpose of this study was to quantify the anticipatory effects on *whole-body dynamic stability* movement strategies, specifically for medio-lateral control of the CoM, in side cutting, whilst expressing the consequences for performance and undesirable knee joint moments. Our first hypothesis cannot be fully accepted as we found that limiting anticipation does not necessarily compromise known ACL injury risk mechanics, however, components of side cutting performance may be affected. In particular, it seems that with limited anticipation participants were able to redirect their CoM more medially, but with less acceleration on average over ground contact, so the impact on performance was not clear cut. This may be partially explained by slowing their approach into the side cutting task, in the final two metres before the turn, and increasing subsequent ground contact time. Regarding our second hypothesis, the differences in whole-body dynamic stability mechanisms with limited anticipation are perhaps sufficient enough for us to describe two anticipatory movement strategies, so this hypothesis can be accepted. Specifically, an unanticipated movement strategy would involve a narrower foot placement, inefficient sagittal plane loading, resulting in substantial unloading of the medial ground reaction forces from frontal plane hip acceleration, and a more lateral position of the CoP later in ground contact. Conversely, a movement strategy to describe how control of the CoM is achieved in anticipated side cutting would involve a wider foot placement and greater medial propulsion from more efficient sagittal triple acceleration.

In differentiating between side cutting tasks of contrasting anticipatory demands, the most important movement strategy for medio-lateral control of the CoM appears to be foot placement, sagittal triple acceleration and frontal plane hip acceleration. It seems that, with reduced anticipation, participants are forced to make a narrower foot placement, perhaps as the preparatory postural adjustments required for a wider foot placement were inadequate (Besier et al., 2001b; Xu, Carlton and Rosengren, 2004). Probably as a result of excessive sagittal plane impulses, a pronounced frontal plane hip strategy was deployed immediately following initial ground contact, which suggests this was a key corrective strategy adopted to control the CoM. Similar relationships between foot placement and the hip joint or trunk in the frontal plane have been reported previously for unanticipated changes of direction in walking (Patla, Adkin and Ballard, 1999; Xu, Carlton and Rosengren, 2004) and side cutting (Houck, Duncan and De Haven, 2006; Brown, Brughelli and Hume, 2014; Mornieux et al., 2014). The majority of studies have favoured lateral trunk flexion with foot placement to explain such anticipatory postural adjustments (Brown, Brughelli and Hume, 2014; Mornieux et al., 2014) and more extensively when not directly comparing anticipatory effects (Dempsey et al., 2009; Jamison, Pan and Chaudhari, 2012; Jones, Herrington and Graham-Smith, 2015). However, MacKinnon and Winter (1993) proposed an interaction between two strategies, a hip and ankle movement strategy, or a double inverted pendulum model in the frontal plane. In their approach, foot placement is important, but more importantly for the ankle movement strategy perhaps, is subtalar inversion/eversion. In our research medio-lateral CoP position may represent the ankle movement strategy, and our results show a more lateral position of the CoP with limited anticipation later in ground contact. The interaction between movement strategies have been highlighted during walking, where the hip is responsible for balancing larger destabilising effects on the trunk, through the pelvis, in the frontal plane, and the ankle may offer smaller refinements. More recent research has reported the application of a similar movement strategy model as the key anticipatory postural adjustment for walking and side cutting, respectively (Patla, Adkin and Ballard, 1999; Houck, Duncan and De Haven, 2006). In this regard, our results appear to show that both the hip and ankle movement strategies may interact in different ways with limited anticipation. Specifically, the transition from the hip to an ankle movement strategy, in the frontal plane, may occur later with limited anticipation, but the extent of subtalar inversion may be more pronounced later on in that condition. Therefore, our findings demonstrated larger destabilising effects with the body as a result of less efficient sagittal plane loading with limited anticipation. The difference in deployment of frontal plane hip acceleration is an indication of how much more whole-body dynamic stability is challenged with limited anticipation time.

There are some limitations with the current study. Firstly, we observed differences in the way some of the spatial-temporal task constraints we met between anticipatory conditions. In particular, the approach and exit velocity of the unanticipated side cutting was significantly slower compared to the anticipated side cutting. Furthermore, participants also performed a significantly sharper side cut in the unanticipated condition compared to the anticipated, this is potentially a concern considering approach velocity and side cutting angle have been shown to influence injury risk and performance (Vanrenterghem et al., 2012;

Dos'Santos et al., 2018). That said, those specific performance differences were not detectable at the time of data collection. Whilst more detailed observation of the preceding steps may help explain the differences found in approach velocity, as some research groups are investigating (Havens and Sigward, 2015; Jones, Herrington and Graham-Smith, 2016; Dos'Santos et al., 2018), the poor adherence to side cutting angle has been reported before (Vanrenterghem et al., 2012; Dos'Santos et al., 2018). As we were concerned with medio-lateral control of the CoM, average medio-lateral CoM acceleration may be considered equally important, and in this performance outcome there was no significant difference between anticipatory conditions. A second limitation may be the choice of spatial-temporal task constraints like approach velocity or angle, or perhaps more importantly for this comparison, timing of the unanticipated stimulus. In this study we estimated our participants had 0.50-0.65 s to react to the unanticipated stimulus which is based on the critical time conditions reported previously (Brown, Palmieri-Smith and McLean, 2009; Mornieux et al., 2014). To address this limitation in more detail would require further research on stimulus timing, or unrestricted performance constraints which are not typically feasible in a laboratory setting. One final limitation may be the relatively conservative Bonferroni correction of alpha for multiple comparisons. That said, for the number of dependent variables reported here, it is unlikely using an alternative approach would influence the key findings substantially.

Conclusion

In conclusion, our findings provide evidence that anticipation affects the way movement strategies are deployed to control the CoM in medio-lateral direction and, therefore, *whole-body dynamic stability*. To the best of our knowledge, this is the first occasion where movement strategies related to *whole-body dynamic stability* have been compared between side cutting tasks of different complexity, or more broadly, tasks that demand high medio-lateral force generation control with different preparatory challenges. In this investigation we have shown foot placement, sagittal plane loading efficiency and frontal plane hip acceleration are the key components of an anticipatory postural adjustment movement strategy. Specifically, with limited anticipation, participants are forced to make a narrower foot placement and, following excessive sagittal plane loads, subsequently, deploy frontal plane hip acceleration as a corrective movement strategy. This anticipatory adjustment is sufficient to retrieve control of the CoM and any possible destabilisation effects of the body, without significant detrimental effects on markers of injury risk or side cutting performance. Our findings have also allowed us to express how a corrective double inverted pendulum model involving hip and ankle movement strategies may differ in interaction with limited anticipation. Any challenges on top of limited anticipation, perhaps in a more sport-specific scenario (e.g. external stimuli or fatigue), may push the movement strategies beyond their adaptive capacity to control the CoM, and we may observe detrimental consequences, but further research is required here.

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Study 4

Effects of 90-minute match simulation on whole-body dynamic stability in unanticipated side cutting

Abstract

Control of the Centre of Mass (CoM) is essential for safe and efficient execution of a highly dynamic task like side cutting. However, performing such tasks whilst mitigating unnecessary deviations, and therefore maintaining whole-body dynamic stability, is challenging due to the extent of medial force generation required. Several movement strategies must be adopted for successful medio-lateral (M-L) control of the CoM, however, the extent to which those movement strategies are affected by soccer match-specific physical exertion is currently unknown. The aim of our study was to investigate how specific movement strategies to control the CoM are affected by physical exertion in match-simulation, and how side cutting performance and undesirable knee joint moments are affected as a result. Twenty one recreational soccer players completed a 90-minute over-ground soccer match-simulation. Integrated to the match-simulation were 45° unanticipated side cutting tasks using a 4-5 $m.s^{-1}$ approach speed. 3D motion capture and ground reaction force data were collected. Kinematics and kinetics, and task execution variables were calculated using a lower limb and trunk model. Performance outcomes, peak knee abduction moments (peak KAM), and whole-body dynamic stability variables were calculated for the side cutting tasks, with an average of 4 trials per participant, per 15-minute block of the simulation, inclusive of a pre-simulation condition and a 15-minute break for half-time. In total there were seven within-group levels. Five distinct whole-body dynamic stability movement strategies were identified, based on factors that influence the medial ground reaction force (GRF) vector during ground contact in the side cutting manoeuvre. To investigate the effect of the match-simulation, multiple ANOVAs were conducted using SPSS and SPM1D. Separate analyses were conducted on four participants who demonstrated high peak knee abduction moments, two SDs above the mean of the rest of the sample. Side cutting performance reduced significantly over the course of the match-simulation. Peak knee abduction moment reduced, but not significantly with elapsed match-simulation time.

Foot placement and frontal plane hip acceleration did not change significantly in response to the demands of the prolonged bout of physical exertion, however, other whole-body dynamic stability mechanisms did. Furthermore, in separate observation, individuals with high peak KAM displayed excessive medial ground reaction forces from sagittal triple acceleration, and greater lateral forces from frontal hip acceleration that reduced as time elapsed, compared to individuals with lower peak KAM. Whole-body dynamic stability movement strategies adapted to the demands of soccer-specific match simulation, and were typically successful in mitigating excessive frontal plane knee moments associated with injury risk. That said, several individuals within this sample, who showed considerably greater knee abduction moments, also demonstrated excessive sagittal plane impulses and subsequently, extensive use of the hip movement strategy in the frontal plane. Thus, a single match-simulation may be insufficient to increase injury risk markers, unless the individuals are already showing signs of a reduced ability to mitigate unnecessary movement deviations, emphasised by reduced whole-body dynamic stability.

Introduction

Each time a dynamic task is executed success is dependent on the deployment of movement strategies that allow for control of the centre of mass (CoM) and therefore *whole-body dynamic stability*. When dynamic tasks like side cutting are repeated, perhaps over the course of a prolonged bout of physical exertion, it is possible that the development of fatigue may lead to reduced efficiency in the deployment of those movement strategies. If those movement strategies start to fail, the control of the CoM is compromised to the extent that movement deviations become detrimental to task performance or even dangerous. This may result in undesirable knee joint mechanics like those reported previously for Anterior Cruciate Ligament (ACL) injury (Hewett et al., 2005). The extent to which deployment of whole-body dynamic stability movement strategies and dangerous deviations are affected by repetition, like that observed in soccer match-play, remains unknown.

During bouts of prolonged physical exertion there are many reasons that dynamic tasks like unanticipated side cutting could become increasingly challenging. The performer faces continual demands on their ability to process environmental information in the time available following external triggers; to execute the appropriate movement strategies for the situation; and to tolerate repetitive and often high impulses. Studies have typically characterised such demands as central, peripheral (neuromuscular), or mechanical fatigue (Collins et al. 2016; Edwards, 2018), respectively. It is likely that each aspect may contribute in some way to the higher incidence of injury observed later on in bouts of physical exertion (Hawkins et al., 2001). Many studies have reported the negative effects of physical exertion or fatigue on different performance outcomes or mechanical attributes, and there are several studies that have focused specifically on multi-directional side cutting tasks. Of those studies, to the best of the authors' knowledge, only one study has reported significant increases in peak knee abduction moments in response to fatigue (Tsai et al., 2009). More common responses to physical exertion appear to be kinematic postural adjustments including, a more extended knee (Greig, 2009; Lucci, et al., 2011; Cortes et al., 2013; McGovern et al., 2015; Raja Azidin et al., 2015; Whyte et al., 2018); increases in knee abduction angles (Borotikar et al., 2008; Collins et al., 2016); and increases in internal rotation angles of the knee (Borotikar et al., 2008; Sanna and O'Connor, 2008; Tsai et al., 2009). That said, it is possible that mechanics that are typically reported to contribute to increased ACL injury risk are themselves a consequence of poor control of the CoM in the new direction of travel. Medio-lateral control of the CoM is a priority when changing direction (Patla, Adkin and Ballard, 1999; Donnelly et al., 2012), thus, observing the movement strategies that contribute to this challenge, is expected to provide unique insight.

Quantifying the mechanical change following match-specific bouts of physical exertion may provide valuable information for injury screening or training intervention, whether preventative, or as a method of tracking athletic status when aiming to return-to-play. Therefore, the aim of this investigation was to identify the effects of a match-specific bout of exertion on deployment of movement strategies to provide medio-lateral control the CoM during unanticipated side cutting. It was hypothesised that over the course of the match simulation there will be a significantly reduced side cutting performance and increase in undesirable knee loading. It was also hypothesised that there will be significant differences in the way performance and corrective movement strategies will be deployed to control the CoM in the medio-lateral direction in response to match simulation.

Methods

Participants

The participants in this study were twenty one healthy male recreational soccer players, with at least 6 years playing experience, consisting of between one and two sessions a week, for one to two hours per session. The participants had a mean (± SD) age of 26 ± 7 years; mean height of 1.8 ± 0.1 m; and mean mass of 79.2 ± 11.2 kg. All participants were free from injury for at least 6 months, and written consent was retrieved from every participant. All participant recruitment processes were conducted in line with the university research ethics committee guidelines, which comply with the principles of the Declaration of Helsinki.

Experimental Design

A repeated-measures design was used in which participants initially completed a one and a half hour familiarisation session that involved practice of the procedure for data collection in mock testing conditions. During the familiarisation session participants were introduced to the laboratory environment, the unanticipated side cutting task, the match-simulation, and the marker model that would be used for 3D motion capture. Each participant then nominated their preferred limb, which they later used to complete all side cutting trials. Participants completed a full 15-minute sample of the over-ground match simulation as part of their familiarisation. The experimental testing session took place at the same time of day as the familiarisation session, and no more than 7 days after that initial session. In the experimental session participants initially completed eight unanticipated side cutting trials before the simulation started. Then each participant completed a full 90-minute match simulation organised into two 45-minute periods, with a 15-minute break for half-time. At least eight unanticipated side cutting trials were included as an integral part of each 15-minute block, four for each of the two conditions – *open* and *crossover* side cutting.

Protocol – Side cutting assessment

Following application of the marker model to the participant, body weight squats, over-ground shuttle runs, and specific side cutting practice were conducted for the warm-up. The marker model was then checked and adjusted for comfort and stability, if required. Participants then engaged in static calibration and functional joint tasks for the hip and knee. Immediately following the removal of calibration only markers, participants completed the pre-simulation condition. The 45° unanticipated side cutting task began with a 10 m approach to the force plate, and the approach speed was controlled to 4-5 m.s^{-1}, with timing gates (Brower Timing Systems, Utah, USA) set at 2 m apart and 2 m away from the force plate. At the first timing gate, 4 m from the force plate, the participant triggered the custom cueing programme (Matlab, MathWorks, Natick, MA. USA) on a PC. The screen of the PC was placed 3 m beyond the force plate, centred, facing the participant, and at 1 m in height from the floor. At 0.150 s after the trigger, the participants were presented with a full screen arrow pointing left (blue) or right (red). Considering the approach speed exclusion criteria (4-5 m.s^{-1}), this gave the participants 0.65-0.85 seconds to react and contact the force plate to execute the task with their nominated preferred side cutting limb. In earlier pilot work the triggering gate was placed 1 m closer, but participants struggled to reliably respond in time as the simulation progressed. One of two conditions were then executed within a single foot contact in response to the stimulus, either *open* side cutting (45° turn to the opposite direction of the contact foot used) or a *crossover* side cutting (45° turn to the same direction as the contact foot used). When participants incorrectly responded to the stimulus, or otherwise failed to meet the criteria set for the side cutting task, that trial was discarded, and an additional trial was added to the trial count and randomisation sequence in the Matlab software. For the pre-simulation condition, each participant completed eight trials - four trials per condition. Within two minutes after pre-simulation, participants then started the

90-minute match-simulation completing eight side cutting trials embedded into each 15-minute block, this is explained further below.

Protocol - Match-simulation

The over ground match simulation adopted in this investigation has been explained in detail elsewhere (Raja Azidin et al., 2015). The match simulation was an adapted version of that outlined in Small et al. (2010). Briefly, the simulation involved a 15-minute playback of an audio recording verbally cueing shuttle running at a range of intensities from walking to sprinting, along with short utility side step and backtracking manoeuvres. If completed successfully, for a 45-minute half (3 x 15-minute bouts) each participant would cover 5.39 km, or 10.78 km total distance for the full 90-minute match simulation. The side cutting task was embedded utilising the verbal cue '…stride' that occurred eight times per 15-minute block. If the participant failed to meet the side cutting inclusion criteria, the playback was briefly paused to add a trial to the randomisation programme and immediately repeat the task until a successful repetition was achieved, then the simulation recording was continued. Typically the successful trial was achieved at the first repeated attempt, but on rare occasions a second repeat was required, each repeated trial delayed the simulation progress by ~20 seconds. If the total delay exceeded 5 minutes for the 90-minute match simulation the results were not submitted as part of the subsequent analyses.

Biomechanical model

All participants had 44 reflective markers represented in the Liverpool John Moores University (LJMU) Lower Limb and Trunk (LLT) eight segment model previously explained in detail elsewhere (Vanrenterghem et al., 2010; Malfait et al., 2014). Single markers were attached with double-sided tape to base layer clothing or skin, and cluster plates were used on the lower limb segments and secured with Velcro, tape and bandages to ensure secure fitting through the simulation. Starting reference points of cluster plates were marked on the skin to help identify if any cluster plate movement occurred. For single markers on standardised clothing and football shoes, additional glue was used to ensure the secure fit. Single markers on skin typically required some skin preparation – shaving and cleaning – and application before warm-up which reduced the chance for any prior sweating to occur. All participants went through re-calibration for the marker model in the 15-minute half-time break. 3D Marker trajectories were recorded using a 10-camera Oqus system (Qualisys AB, Gothenberg, Sweden) at 100 Hz for the calibration trials and 250 Hz for the side cutting motion trials. Joint centres, axes and local segment coordinate systems were defined as reported previously (Robinson and Vanrenterghem, 2012; Malfait et al., 2014). The side cutting tasks were executed on a 0.9 x 0.6 m force platform (Kistler, Winterthur, Switzerland), and data were sampled at 1500 Hz and synchronised with the Qualysis system. Calibration, modelling, and all kinematic and kinetic analyses were completed in Visual 3D Professional (v.5.00.16, C-Motion, Germantown, MD, USA). Inverse kinematic (IK) modelling was used to constrain all translational motion of the hip, knee and ankle, as well as some rotational constraints. Specifically, this left hips with all three; the ankle with two (sagittal, transverse); and the knee with one (sagittal) rotational degrees of freedom. The IK modelling restrictions were matched to the requirements of Induced Acceleration Analysis, described briefly later.

Data processing

Only the *open* side cutting data were selected for further analyses, represented by four trials per 15-minute block in addition to the pre-simulation data, therefore separated into seven repeated measures levels: pre-simulation – '00'; 0-15 minutes – '15'; 15-30 minutes – '30'; 30-45 minutes – '45'; 60-75 minutes – '75'; 75-90 minutes – '90'; and 90-105 minutes –

‘105’. The match simulation was recorded for cumulative time from the beginning, so 45-60 minutes represented the half-time break where no motion trials were collected. With four trials per participant, each repeated measures condition was represented by 84 trials, or 588 trials in total for the investigation. Marker coordinate data and analogue signals from the force plate channels were filtered using a Butterworth 4th order recursive low pass filter, with a 20 Hz cut-off frequency, based on recommendations of Kristianslund, Krosshaug and van den Bogert (2012). Net joint moments were estimated through inverse dynamics (cardan sequence, XYZ). Initial foot contact with the ground, or *touchdown* (TD), was represented as the minima prior to an ascending vertical GRF gradient; and the *toe-off* (TO) event was represented by a minima following a descending gradient of the same vertical component of GRF. Centre of Mass (CoM) was calculated for every instant across the side cutting task, then CoM velocity was calculated as the first derivative. The following performance outcomes were calculated using trigonometry: CoM trajectory angle (transverse plane); CoM trajectory velocity (transverse plane); in addition to separated anterior and lateral components of CoM velocity. These performance outcomes were specified discretely at the two side cutting events, TD and TO, and the total change between TD and TO calculated. *Change of direction angle* was calculated as the change in CoM trajectory angle. Changes in M-L CoM velocity, once divide by ground *contact time*, represented the *average medial CoM acceleration*. *Knee joint angle at TD,* and peak knee abduction moment (*peak KAM*) were also calculated, the latter was measured over the ‘weight acceptance phase’ - from TD to the first trough of the vertical GRF (Besier et al., 2001).

Quantification of whole-body dynamic stability

The expression of five distinct *whole-body dynamic stability* variables are described in detail in a previous chapter (General Methods and Study 2), and represent the control of the CoM through factors that influence the medio-lateral component of the GRF vector. Briefly, five distinct movement strategies were calculated: (1) M-L foot placement and (2) M-L CoP position, and three GRF magnitude variables (3) sagittal triple acceleration, (4) frontal plane hip acceleration and (5) transverse plane hip acceleration. The GRF magnitude variables were determined by non-negligible (>10 N) contribution to the M-L GRF using IAA in Visual 3D software (Kepple, Siegel and Stanhope, 1997; João et al., 2014; Moniz-Pereira et al., 2018), and consolidated into the respective planes. Following examples in previous research (Kepple, Siegel and Stanhope, 1997; João et al., 2014; Moniz-Pereira et al., 2018) the accuracy of IAA was determined by finding the absolute mean difference between the force platform ground reaction forces and those derived from IAA. The difference was then represented as a percentage of the maximum force obtained - in this case the mean error for medio-lateral IAA was found to be 6%. The extent of the excessive medial forces from sagittal plane contributions we determined by calculating the sagittal triple acceleration impulse as a proportion of the total medio-lateral force impulse, and thus representing a *Sagittal Efficiency Ratio*. A Sagittal Efficiency Ratio of 100% would mean the impulses were equal and the medial CoM acceleration was entirely sagittal. Lower or higher than 100% would mean non-sagittal movements were involved in generating (increasing) or moderating (reducing) medial CoM acceleration, respectively.

Statistical Analyses

Statistical comparison of performance outcome and key joint loading data over the match simulation were calculated in SPSS version 23 (IBM SPSS Statistics, Chicago, USA). Following Shapiro Wilks test for normal distribution either Repeated Measures (RM) ANOVA's or Friedman's test were used to establish statistical main effects. Alpha was adjusted from $\alpha = 0.05$ to $\alpha = 0.007$, with Bonferroni correction for multiple comparisons - seven performance outcome including key joint loading variables - set to three decimal places. Later, a Shapiro Wilks test followed by a RM ANOVA was used to establish main effects for the Sagittal Efficiency Ratio over the course of the match-simulation.

All statistical analyses for the *whole-body dynamic stability* variables were computed using Statistical Parametric Mapping (SPM) (Pataky, 2012) in SPM1D (version 0.4) using Python (Python version 2.7.1 Enthought Canopy, version 1.6.2, Enthought Python Distribution, Austin, TX, USA). Normal distribution of all 0D and 1D signals were calculated with D'Agostino-Pearsons K2 test. Subsequently, either parametric or non-parametric repeated measures ANOVA's were completed for each of the five *whole-body dynamic stability* variables. Alpha was adjusted from $\alpha = 0.05$ to $\alpha = 0.01$, with Bonferroni correction for multiple comparisons.

It was noted that four of the twenty one participants exhibited high peak KAM. To explore the role of whole-body dynamic stability for those participants further analysis was conducted. Each whole-body dynamic stability variable was averaged for the four participants with high peak KAM (*high knee loading group*), and compared to the remaining 17 with lower peak KAM (*low knee loading group*). The *high knee loading* group exhibited an average peak KAM more than two SDs above the average for the *low knee loading* group.

Results

Comparisons of side cutting performance outcome variables are presented in table 4.1. Touchdown and toe-off velocity of the CoM did not change significantly over the match simulation, and participants were able to adhere to a key task constraint by not slowing down over ground contact. Participants were also able to maintain contact time on the ground when performing the task. However, participants were not able to maintain either change of direction angle, or average medial CoM acceleration in response to the physical exertion ($p < 0.001$). Whilst participants performed the task with a more extended knee at initial ground contact, as the match simulation progressed ($p = 0.005$), they were able to control undesirable knee loading and in fact demonstrated a trend for reduction of peak knee abduction moment.

Table 4.1. Comparison of the performance outcome variables (mean ± *SD*) for 45° unanticipated side cutting tasks from pre-simulation (Pre-Sim) and over the course of six 15-minute blocks of a 90-minute match simulation. Significance of main effect from within-group analyses is also presented.

Performance Outcome		Pre-Sim	First half (minutes)			Second half (minutes)			significance
			0-15	15-30	30-45	60-75	75-90	90-105	
Touchdown Velocity ($m \cdot s^{-1}$)	mean	4.74	4.80	4.78	4.75	4.68	4.71	4.67	p=0.056 [np]
	SD	*0.36*	*0.30*	*0.24*	*0.31*	*0.29*	*0.32*	*0.38*	
Toe-off Velocity ($m \cdot s^{-1}$)	mean	4.57	4.64	4.63	4.60	4.57	4.53	4.50	p=0.270 [np]
	SD	*0.37*	*0.36*	*0.32*	*0.39*	*0.41*	*0.44*	*0.49*	
Av. Medial CoM accel. ($m \cdot s^{-2}$)	mean	6.46	7.00	6.76	6.40	6.26	6.04	5.88	*p<0.001 [np]
	SD	*1.19*	*1.41*	*1.64*	*1.50*	*1.22*	*1.62*	*1.47*	
Change of direction angle (°)	mean	19.52	20.42	19.67	18.60	18.68	17.98	17.93	*p<0.001[GG]
	SD	*4.31*	*4.34*	*4.97*	*4.36*	*4.73*	*4.73*	*4.76*	
Contact Time (s)	mean	0.224	0.220	0.219	0.217	0.220	0.218	0.222	p=0.893 [np]
	SD	*0.021*	*0.021*	*0.021*	*0.021*	*0.020*	*0.022*	*0.024*	
Knee Angle @ TD (°)	mean	23.24	25.09	23.93	22.10	21.34	19.75	19.41	*p=0.005[GG]
	SD	*9.25*	*8.12*	*9.32*	*9.93*	*9.67*	*10.02*	*10.24*	
Peak KAM (Nm·kg)	mean	0.72	0.83	0.78	0.71	0.57	0.59	0.57	p=0.05 [np]
	SD	*0.71*	*0.65*	*0.79*	*0.72*	*0.50*	*0.61*	*0.52*	

NB: '' denotes significance with a Bonferroni corrected alpha level (α = 0.007) for multiple comparisons; '[np]' denotes Friedman's Non-Parametric repeated measures test; '[GG]' denotes Greenhouse-Geisser correction for violation of Sphericity assumption.*

The responses of the whole-body dynamic stability variables to the match-simulation are presented in figures 4.1 – 4.5. Initially we see that the moment arm of the medio-lateral ground reaction force vector did not change by adjustment of the foot placement ($p > 0.01$, see figure 4.1), however, the CoP was positioned more laterally, for the majority of ground contact, as the simulation progressed ($p < 0.001$, see figure 4.2). Although, there was a notable rapid medial shift of the CoP position in the final ~15% of ground contact (*p = 0.001*). In figure 4.3, we observe sagittal triple acceleration was used significantly less (*p = 0.005*) to direct the CoM medially, in response to the demands of the match simulation, particularly in the weight acceptance phase of side cutting. On average, the Sagittal Efficiency Ratio improved slightly over the course of the match-simulation, from pre-simulation; to the end of the first half; then the end of the second half (115.6 ± 18.7 %; 111.3 ± 14.1 %; and 109.5 ± 16.9 %, respectively) albeit not significantly (*p = 0.072*). Frontal plane hip acceleration did not change significantly (*p > 0.01*, see figure 4.4), however, the contribution to medial ground reaction forces form transverse hip acceleration reduced just before toe-off, in response to exertion (*p = 0*.001, see figure 4.5).

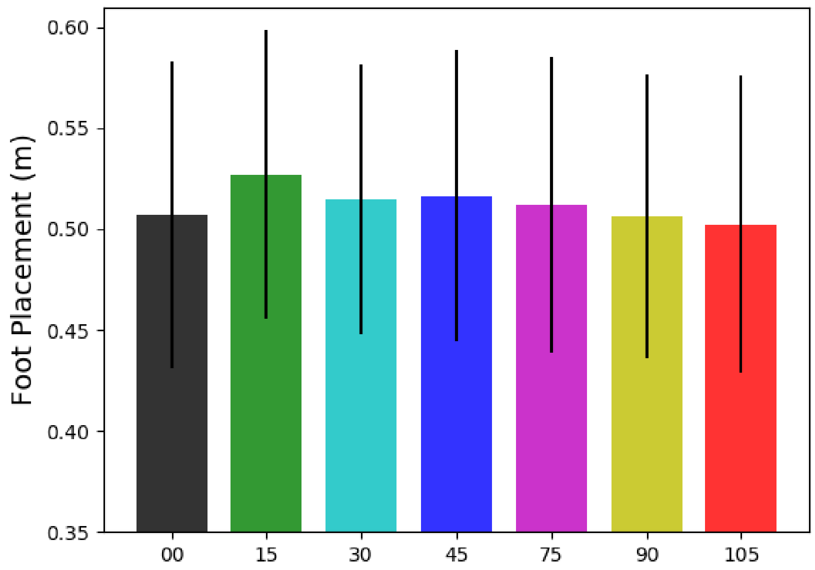

Figure 4.1. Within-group comparison of the first *whole-body dynamic stability* mechanism – medio-lateral (M-L) foot placement (mean ± SD) - metatarsal head 5 (‘0.0’ on y-axis) to extrapolated CoM (MTH5 – XCoM) at touchdown (TD) – over the course of a 90-minute match simulation including 15-minute half-time break. Pre-simulation (‘00’); 0-15 minutes (‘15’); 15-30 minutes (‘30’); 30-45 minutes (‘45’); 45-60 minutes = half-time; 60-75 minutes (’75’); 75-90 minutes (‘90’); 90-105 minutes (‘105’).

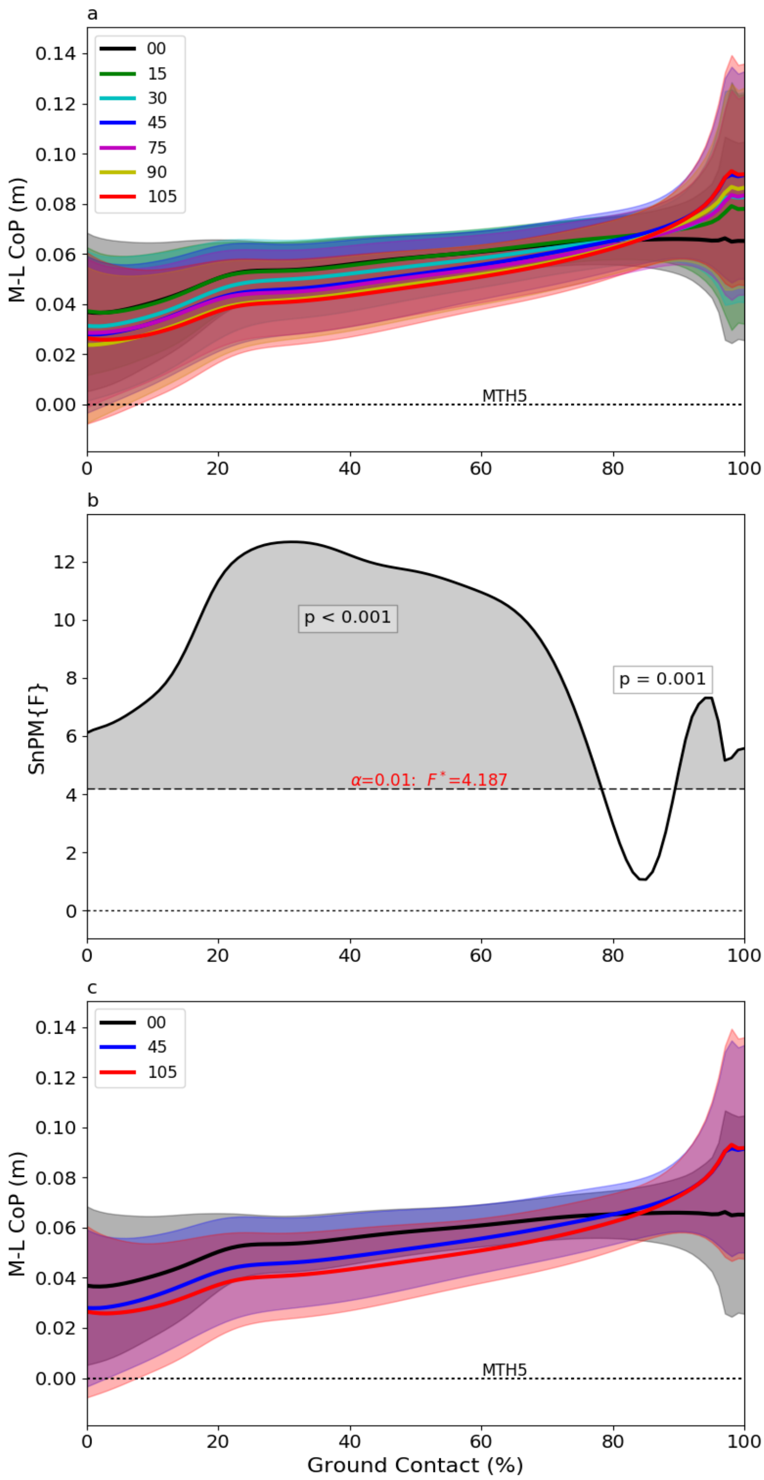

Figure 4.2. Within-group comparison of the second *whole-body dynamic stability* variable – medio-lateral (M-L) centre of pressure (CoP) position (mean ± SD) - metatarsal head 5 to CoP (MTH5 – CoP) – for the entire ground contact of the side cutting task. Comparison is displayed for the course of a 90-minute match simulation and factors in a 15-minute half-time break. Image 'a' includes the comparison of seven test times; image 'b' shows the repeated measures ANOVA statistical main effect (*SnPM{F}*) where alpha was adjusted for multiple comparisons (Bonferroni – $\alpha = 0.01$); and image 'c' compares the Pre-simulation '00' with the last 15-minute blocks of the first half ('45') and second half ('105').

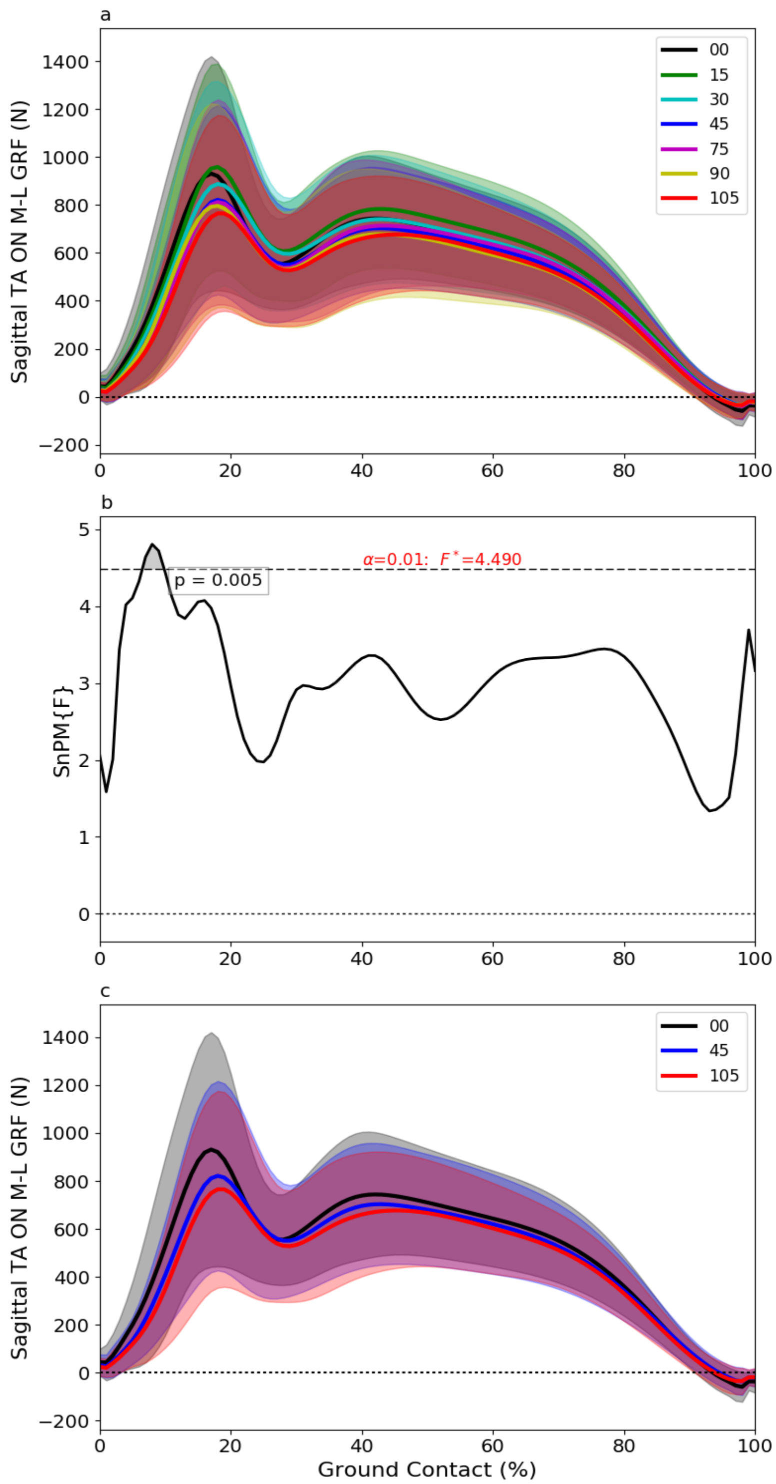

Figure 4.3. Within-group comparison of the third *whole-body dynamic stability* variable – Sagittal triple acceleration (TA) (combined hip, knee and ankle) contribution to medio-lateral (M-L) ground reaction force (GRF) – for the entire ground contact of the side cutting task. Comparison is displayed for the course of a 90-minute match simulation and factors in a 15-minute half-time break. Image 'a' includes the comparison of seven test times; image 'b' shows the repeated measures ANOVA statistical main effect (*SnPM{F}*) where alpha was adjusted for multiple comparisons (Bonferroni – $\alpha = 0.01$); and image 'c' compares the Pre-simulation '00' with the last 15-minute blocks of the first half ('45'), and second half ('105').

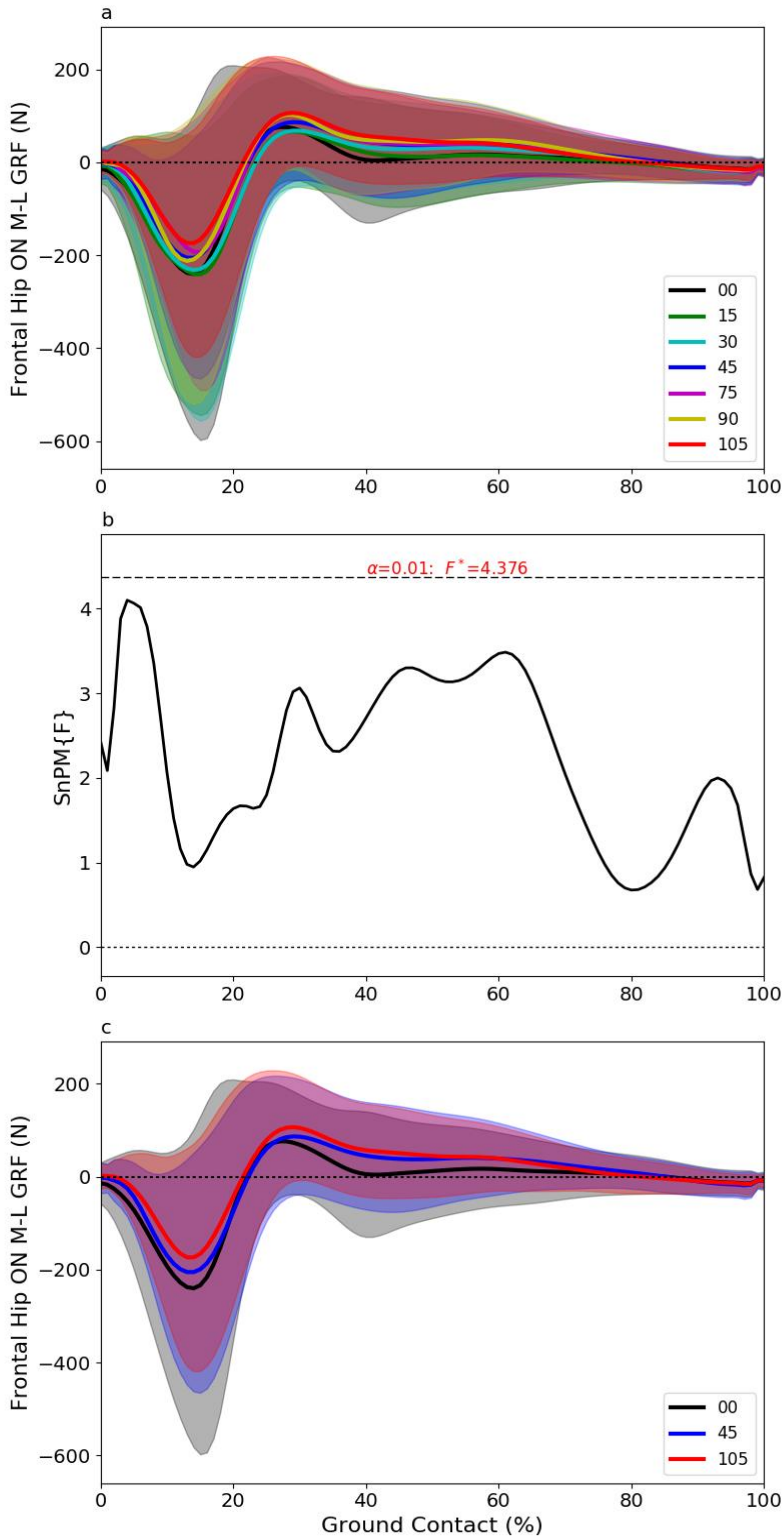

Figure 4.4. Within-group comparison of the fourth *whole-body dynamic stability* variable – frontal plane hip acceleration contribution to medio-lateral (M-L) ground reaction force (GRF) – for the entire ground contact of the side cutting task. Comparison is displayed for the course of a 90-minute match simulation and factors in a 15-minute half-time break. Image 'a' includes the comparison of seven test times; image 'b' shows the repeated measures ANOVA statistical main effect (*SnPM{F}*) where alpha was adjusted for multiple comparisons (Bonferroni – $\alpha = 0.01$); and image 'c' compares the Pre-simulation '00' with the last 15-minute blocks of the first half ('45'), and second half ('105').

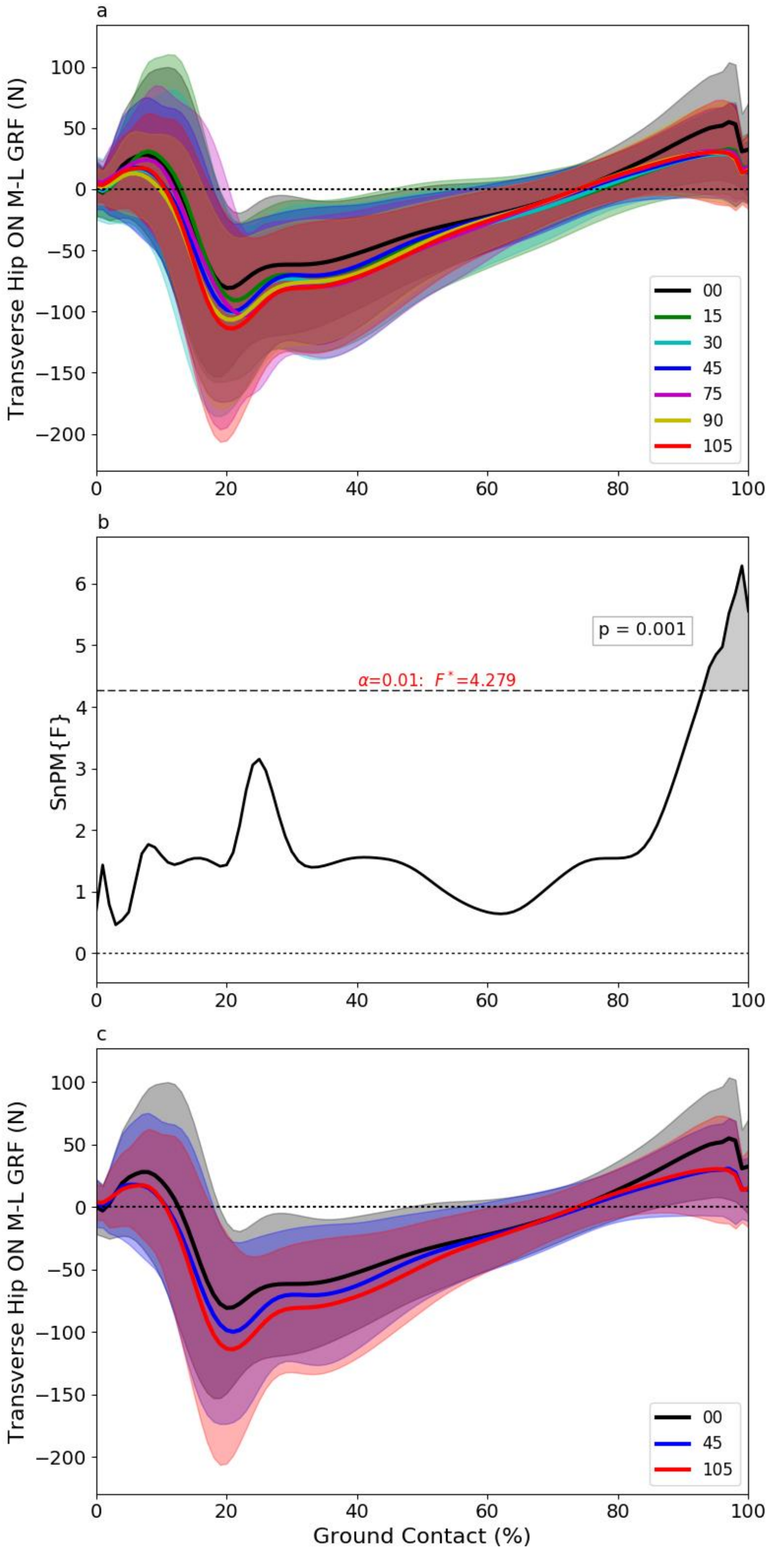

Figure 4.5. Within-group comparison of the fifth *whole-body dynamic stability* variable – transverse plane hip acceleration contribution to medio-lateral (M-L) ground reaction force (GRF) – for the entire ground contact of the side cutting task. Comparison is displayed for the course of a 90-minute match simulation and factors in a 15-minute half-time break. Image 'a' includes the comparison of seven test times; image 'b' shows the repeated measures ANOVA statistical main effect (*SnPM{F}*) where alpha was adjusted for multiple comparisons (Bonferroni – $\alpha = 0.01$); and image 'c' compares the Pre-simulation '00' with the last 15-minute blocks of the first half ('45'), and second half ('105').

In separate comparison of the *high* and *low knee loading* groups, we see foot placement and medially directed forces through sagittal triple acceleration, and laterally directed forces (or unloading) through frontal plane hip acceleration, were each greater for the *high knee loading* group (see figure 4.6). More importantly, in response to the match simulation, there was a reduction in the extent of sagittal triple acceleration over ground contact for both the *high* and *low knee loading* groups equating to -11.6 and -13.5 Ns, respectively. However, a pronounced difference was observed in the role of frontal plane hip acceleration between loading groups in response to prolonged physical exertion. Specifically, the *low knee loading* group presented a small increase in frontal plane hip derived medial impulse of 6 Ns, whereas the *high knee loading* group demonstrated reduced lateral, or unloading impulse of -14 Ns over ground contact from the same movement strategy. The high knee loading group also exhibited a notable reduction in foot placement variability in comparison to the low knee loading group, especially at the end of the second half.

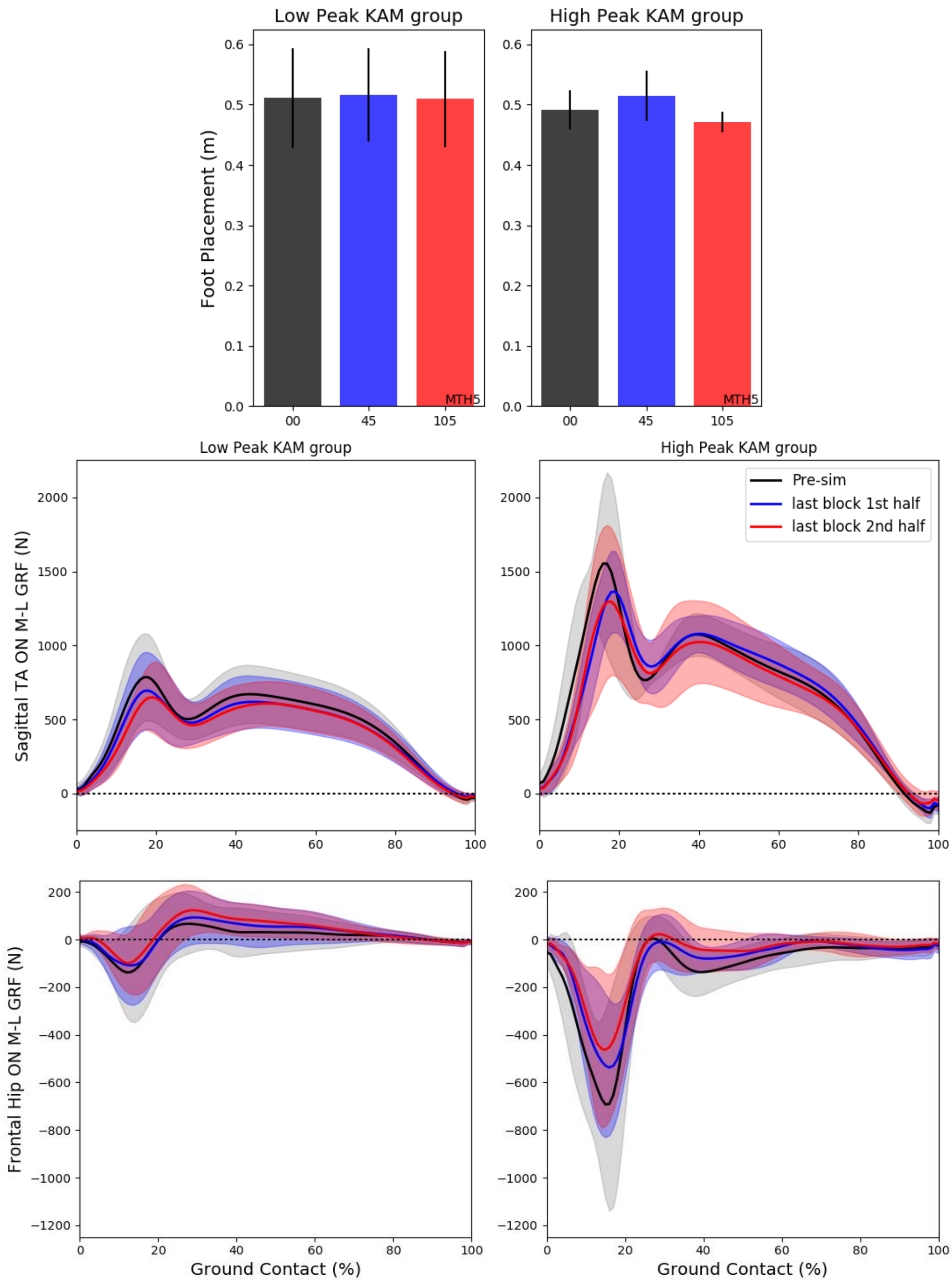

Figure 4.6. Comparison of selected whole-body dynamic stability variables over the match simulation and between *high knee loading* (n=4) and *low knee loading* groups (n=17). The pre-simulation and the last 15-minute blocks of the first and second halves of the match simulation are presented. Foot placement is presented on row one. Sagittal triple acceleration of combined hip, knee and ankle (Sagittal TA) is presented on row two, and frontal plane hip acceleration variable is represented on row three. The first column represent the *low knee loading group* - low peak knee abduction moment (Peak KAM); and the second column represents the *high knee loading group* – high Peak KAM, respectively.

Discussion

The purpose of this investigation was to identify the effects of soccer match-specific physical exertion on deployment of movement strategies to control the CoM in the medio-lateral direction during unanticipated side cutting. Our findings suggest that, injury risk was unaffected by the match simulation, showing a marginal reduction in peak KAM. However, participants were not able to maintain average medial acceleration of the CoM and change of direction angle, despite consistency with approach and exit velocity of performance. Thus, in reference to our first hypothesis, our prediction for undesirable knee joint loading was incorrect, however, our predictions for performance were partially confirmed. This means participants may have been prioritising injury risk over performance, which is understandable as the task constraints mean the side cutting task was likely submaximal for the duration. More importantly, our observation of whole-body dynamic stability may offer an explanation of how mitigation of injury risk was possible, partially supporting our second hypothesis. Broadly, we observed that, whilst foot placement was unaffected, sagittal triple acceleration reduced and sagittal impulses tended to be less excessive as the simulation progressed. This is likely to explain why, on average, whilst the countermovement role of the frontal plane hip movement strategy is still important, the extent of its involvement was unaffected in response to the physical exertion. Instead, the more significant adjustments were seen in the more lateral position of the CoP, perhaps indicating the prevalence of an ankle strategy, or subtalar inversion were possible to increase the medial moment arm to mitigate any further drops in performance. That said, there were some individuals in the sample exhibiting particularly high peak knee abduction moments. These individuals presented excessive and inefficient sagittal plane contribution to medial GRF, which probably explained the extensive demand on the frontal plane hip movement strategy, even in pre-simulation, to moderate those forces. Furthermore, in the high loading group, there was a notable reduction in the capacity of frontal plane hip acceleration as time elapsed, along with a distinct reduction in the variability of foot placement, which may both be signs of a dangerous reduction in whole-body dynamic stability.

In our investigation we found that exertion-induced changes in peak knee abduction moments may not put the performer at greater risk of ACL injury, despite a more extended knee posture, and we are not the first to report this finding (Sanna and O'Connor, 2008; Greig, 2009; Lucci, et al., 2011; Cortes et al., 2013; Khalid et al., 2015; McGovern et al., 2015; Raja Azidin et al., 2015; Whyte et al., 2018). In fact, to the best of our knowledge, as mentioned previously, only one study reported significant increases in peak external knee abduction moments (internal peak knee adductor moments) specifically for side cutting (Tsai et al., 2009). Although our findings differ in terms of knee abduction moments, several differences in our approaches may explain the responses observed, including anticipation, the sex of the participants, and the specificity of the physical exertion protocol. Furthermore, one may expect that the increase in peak knee abduction moments that Tsai and colleagues observed, may be explained by failings in movement strategies their participants adopted, however, that was not reported specifically in their investigation. Studies that reported similar findings to our own regarding peak knee abduction moment and physical exertion, suggest that adopting a hip movement strategy may explain why changes in trunk mechanics (e.g. ipsilateral flexion) may not pose an increased injury risk (Whyte et al., 2018). We have been able to demonstrate evidence that, frontal plane hip acceleration is the dominant hip movement strategy, which is most likely to be associated with mitigating destabilising movement of the trunk. That said, the present findings suggest it is possible to deploy a sufficient hip movement strategy that is tolerant to the demands of exertion. Subsequently, this may mean an ankle movement strategy can be deployed to mitigate further reduction in whole-body dynamic stability and side cutting performance. Thus, this transition between movement strategies, in what previous research has described as a double inverted pendulum model (MacKinnon and Winter, 1993; Winter, 1995; Patla, Adkin and Ballard, 1999; Houck, Duncan and De Haven, 2006), may be indicative of a safer condition of whole-body dynamic stability, which exists despite the challenges of exertion. Furthermore, and importantly for injury screening, this emphasises that physical exertion, or match-simulation fatigue, may not be the real cause of increased injury risk. Instead, exertion-induced screening may only serve to highlight existing inadequacies in movement strategies to control the CoM that are present in already higher risk individuals.

It is important to note that our simulated environment does not maximally challenge participants in terms of side cutting performance, allowing them to prioritise safety as a compensation to potentially impaired neuromuscular control. Indeed recent research proposed reduced neuromuscular control is not definitive with exertion (Barber-Westin and Noyes, 2017), and that fatigued athletes may prioritise safety by just moving more slowly (Doyle et al., 2018). The benefit of performing isolated side cutting tasks more slowly has been reported previously (Vanrenterghem et al., 2012), where injury risk reduces with slower approach velocities and sharper changes of direction can be achieved. Although we did not specifically find our participants slowed down for the task, per se, they were not able to maintain cutting angle and medial acceleration of their CoM, during ground contact, as the simulation progressed. Interestingly, we appear to observe a drop in the ability to change direction over the course of the simulation that was similar to what we may expect if we had asked the participants to approach the task faster. However, of course, submaximal changes of direction performance may not be a luxury in actual match-play, at any point in time. In a real match situation one would be forced to perform maximally repeatedly, i.e. better than the opponent, and one may not have that safety margin, possibly overreaching the capacity of a corrective movement strategy to control injury risk. In those instances, we may see a deterioration of whole-body dynamic stability towards what we observed in the individuals with higher peak knee abduction moments. In addition to continual exposure to high forces, less controlled circumstances may mean external opponent perturbations and further challenges influencing the task are more common. So, if reduced corrective capacity is combined with an external impact, or a desperate attempt to perform beyond one's ability to control the movement, this could be the trigger for injury, or present the worst case scenario for the athlete.

There were several limitations with the current investigation, not least the scope for developing the match-simulation protocol to replicate actual on field match-play more closely. In this study the match-simulation involved no opponents or contact with a ball, however, the simulation was a useful tool to replicate some of the match-specific demands, whilst allowing more control over otherwise external variables. Another limitation may be the visual cueing of the unanticipated side cutting direction, which was a series of arrows on a computer monitor. More advanced and realistic methods are possible, perhaps using a physical opponent or simulated opponent on a larger screen (Lee et al., 2013; 2018), however, this is an experimental consideration that may require further investigation itself. Furthermore, we settled at a cueing response time that was achievable when embedded in the match simulation for our participants based on earlier pilot work, however, a faster response time or more variable approach may be more sport-specific. Finally, the Bonferroni correction for multiple comparisons is seen as a relatively conservative adjustment which may lead to some type II errors. However, the Bonferroni correction is a logical and simple calculation to apply to the different statistical methods used in our study, and any alternatives are unlikely to affect the main findings for the number of dependent variables we reported.

Conclusion

In this study we have shown the effects of soccer-specific match simulation on whole-body dynamic stability movement strategies in unanticipated side cutting, and the possible implications for injury risk and aspects of performance. To the best of our knowledge this is the first study that has reported the roles of movement strategies to control the CoM in side cutting, or more broadly, tasks that involve medio-lateral force generation control, in response to a prolonged bout of physical exertion. We provide evidence that most participants preferred to prioritise controlling injury risk, at some expense to side cutting performance as time elapsed. The specific exertion-induced reduction in peak knee abduction moments and aspects of side cutting performance are likely attributable to the reduction in sagittal triple acceleration, with improved efficiency, and therefore reduction in potentially excessive, destabilising medial ground reaction forces. Perhaps as a result, the frontal plane hip movement strategy appears to be unaffected over time. This implies that perhaps whole-body dynamic stability is not compromised by this match-simulation extensively enough, and an ankle movement strategy was free to act in a role to mitigate further performance reduction as time elapsed. That said, certain individuals with higher peak knee abduction moments did demonstrate excessive movement deviations and poor whole-body dynamic stability from the outset, which may even diminish with prolonged bouts of exertion. Less constrained task performance, opponents and perturbations, as observed in field match-play, may change whole-body dynamic stability, or push movement strategies past their corrective capacity to control the CoM, with detrimental effects beyond what we observed here. The balance with performance and injury risk will always be difficult to achieve, but we have shown observation of whole-body dynamic stability offers a unique perspective to contribute to those paradigms, and the current body of side cutting research.

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General Discussion

The aim of this doctoral thesis was to explore the role of whole-body dynamic stability in the execution of highly dynamic tasks, and identify the implications for injury and performance screening. Following investigation of the reliability and variability of our general approach, an original method for quantifying whole-body dynamic stability was outlined. This method involved expressing factors that influence the medio-lateral ground reaction force vector as movement strategies used to control and accelerate the centre of mass in the new direction of travel. The approach was then applied to scenarios that challenged the adaptability of the whole-body dynamic stability mechanisms, with two specific environmental constraints that are progressively relevant to sport-specific scenarios. The key findings from each study are summarised below, followed by discussion of critical interpretation, practical relevance, and future directions for this field of sports biomechanics research.

Brief of key findings

The initial findings in study 1 suggested that the general motion capture approach was reliable providing consistent biomechanical data that was sufficiently robust for common laboratory testing issues. However, kinetic variability of knee data was distinctly elevated, particularly in the weight acceptance phase in early ground contact, and later similar findings were observed in the ground reaction force data. With trial-to-trial variability of knee moments and ground reaction forces both peaking in the weight acceptance phase, this suggests that adaptability, or the ability to control and moderate internal and external biomechanical loads, may be more challenging immediately following initial ground contact. The results of study 2 showed that all whole-body dynamic stability movement strategies are directly related to peak knee abduction moments and performance outcome measures. Once the foot is placed, two further key movement strategies emerged, the first was sagittal triple acceleration, which was key to develop the necessary medial ground reaction forces required for the task. The second key movement strategy was frontal plane hip acceleration, which provides a strategy to moderate often excessive forces developed by sagittal triple acceleration and mitigate the resulting destabilisation effect on the body. In study 3 the results showed that increased task complexity, represented by limited anticipation time, forces a narrower foot placement. Subsequently, lower yet less efficient forces generated by sagittal triple acceleration were also found, requiring a greater and a more prolonged moderation from frontal plane hip acceleration. The results in the final study, study 4, showed that peak knee abduction moments appeared to marginally reduce as match-exertion progressed, but so too did aspects of side cutting performance. Both are likely to be attributed to the drop in sagittal triple acceleration despite the fact that sagittal efficiency was maintained over the course of the match-exertion. As a result, no surplus exertion-induced demand was placed on frontal plane hip acceleration to re-stabilise the body, perhaps allowing the ankle movement strategy to make performance refinements. That said, there were some individuals who demonstrated particularly high peak knee abduction moments in this study, for whom it was clear that this was a consequence of substantially reduced whole-body dynamic stability.

Critical Interpretation

Previous research has identified that control of the centre of mass is the fundamental priority in human movement whether walking straight (MacKinnon and Winter, 1993) or with a change of direction (Patla, Adkin and Ballard, 1999). When completing a highly dynamic task, like side cutting, the priority to focus on the centre of mass remains the same, and could provide the safest option towards mitigating prominent injury risk markers (Donnelly et al., 2012). That said, up until this point, there has been very little research on quantifying the movement strategies required to redirect and control the centre of mass in the new direction of travel, or medio-lateral whole-body dynamic stability. In this doctoral thesis a novel method to express a holistic approach to quantifying whole-body dynamic stability was introduced, considering how it is achieved and may be compromised in highly dynamic tasks, particularly those that involve significant medio-lateral force generation and control. The work in this thesis has allowed us to develop a novel framework outlining relevant movement strategies in the context of whole-body dynamic stability during highly dynamic tasks (see figure GD.1). The key findings of this thesis are explored in this framework, with consideration of future avenues for further research in this area, which may strengthen the understanding of the interaction between the performance and injury paradigms.

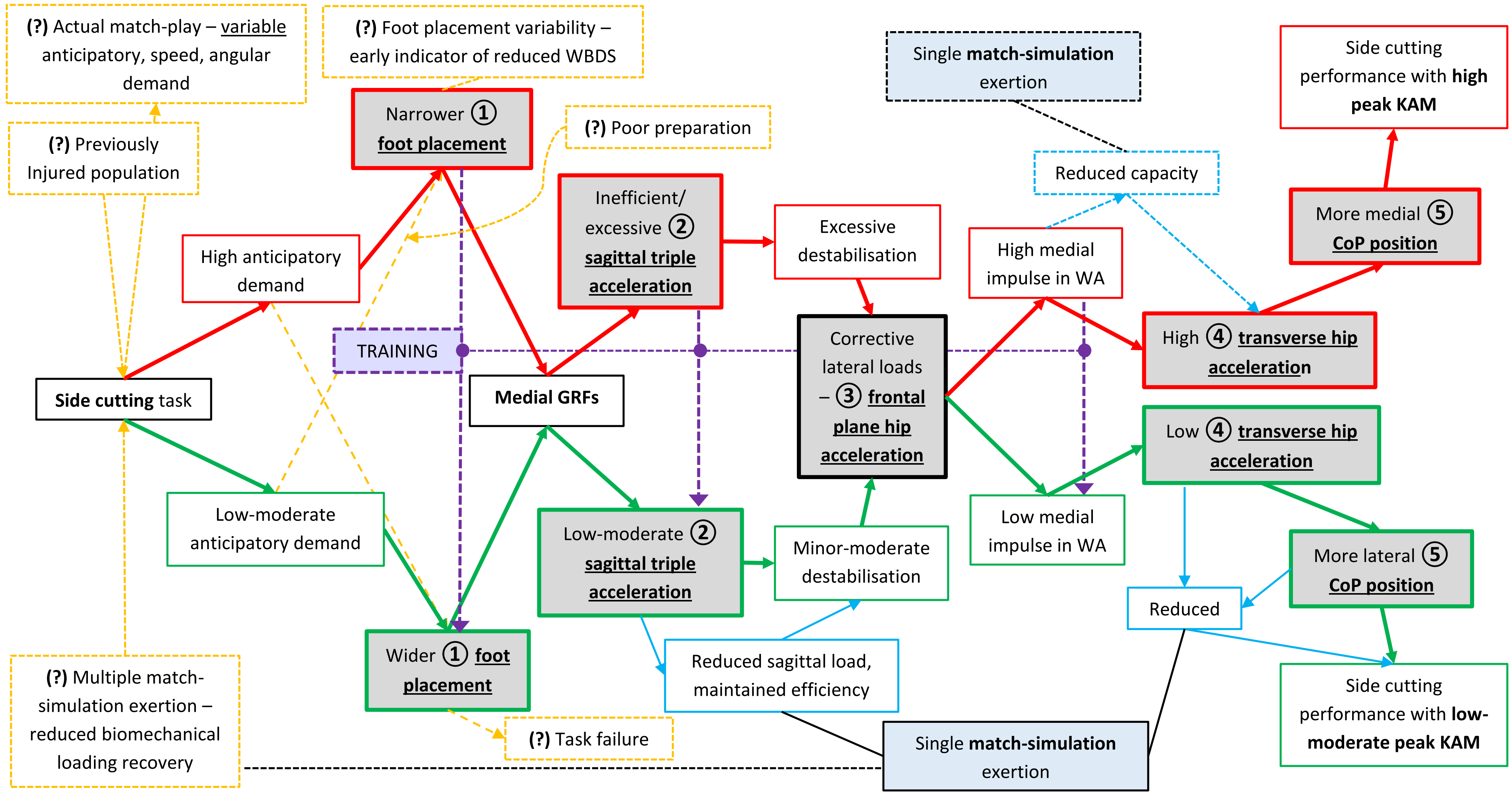

Figure GD.1. Framework for the interaction of internal and external biomechanical loading in the context of whole-body dynamic stability (WBDS) mechanisms (indicated by circled numbers). [GRFs = ground reaction forces; WA = weight acceptance; CoP = centre of pressure; KAM = knee abduction moment. Dotted lines = requires further evidence/ possible future directions for research.]

Foot placement

When completing a change of direction in a single foot contact, the foot placement is imperative and is the first mechanism of whole-body dynamic stability. In figure GD.1 we highlight how the anticipatory demand of the side cutting task is essential to determine the relative width of foot placement. This means that with high anticipatory demand, or limited time to respond to the cueing stimulus, a narrower foot placement is much more likely, and the performer triggers the first part of a movement strategy that may result in undesirable knee loads. However, if the relative anticipatory demand is low-moderate, where the performer has sufficient time to prepare for the turn, then a wider foot placement is possible. This alternative scenario may be safer and ultimately better for side cutting performance. Of course, it is possible that a narrow foot placement is made despite sufficient time to respond, however, this is probably indicative of poor preparation preceding the initial ground contact. Similarly, a wider foot placement in circumstances with limited anticipation, on its own, is likely to result in failure of the task, but further research may be required here. Much of the existing research suggests a wider foot placement may be more dangerous yet better for performance (Dempsey et al., 2009; Kristianslund et al., 2014; Havens and Sigward, 2015; Jones, Herrington and Graham-Smith, 2015), however, this was typically in reference to anticipated, or pre-planned, side cutting. The research in this thesis may support the notion that anticipated and unanticipated side cutting could be considered as different tasks, as proposed previously (Weir et al., 2017). However, the common interaction of biomechanical whole-body dynamic stability mechanisms suggests it may be more accurate to express them as a progressive extension from one another. For example, it may be possible with progressive training to make a wider foot placement in successful task execution with high anticipatory demand. However, this may be dangerous without subsequent adjustments in other whole-body dynamic stability movement strategies. Thus, considering the need to address sports specific demands in biomechanical testing, it is probably more useful to explore the high anticipatory demand pathway, as this is more likely to lead to clearer understanding of injury mechanisms in those scenarios. Indeed, individuals who we found displaying high peak knee abduction moments (see study 4), did so when anticipation was limited, and had even slightly narrower foot placement than other individuals. Furthermore, it was interesting to note that the high knee loading group also displayed lower variability in foot placement, which was lowest at the end of the match-simulation. Thus, it is possible that reduced foot placement variability may be an early indicator of an underlying mechanical issue, or reduced whole-body dynamic stability downstream, which may lead to joint dynamic stability issues, but further research may be required here.

Sagittal triple acceleration - the performance movement strategy

In a side cutting task, following foot placement, the performer must generate substantial medial ground reaction forces (see figure GD.1). In this doctoral thesis it has been demonstrated that the dominant movement strategy redirecting the centre of mass is sagittal triple acceleration, or the combined hip, knee and ankle joint contribution to medio-lateral ground reaction forces in the sagittal plane. Whilst previous research has identified the importance of the hip in the sagittal plane for side cutting performance (Havens and Sigward, 2015), this thesis was the first to express how the hip contributes to the medial ground reaction force as part of a broader sagittal plane strategy. Furthermore, research has suggested that encouraging sagittal plane loading may decrease undesirable loading in the frontal plane (Sigward and Powers, 2007), but we do not find the situation to be this simple. The problem is, sagittal plane loading is almost always in excess of the total medial ground reaction forces generated, and this was a consistent finding across studies 2-4. This highlights the fact that deploying efficient sagittal triple acceleration impulse is actually quite challenging, and on its own may increase loading associated with injury risk.

Therefore, as one cannot execute a side cutting task with purely sagittal mechanisms, some kind of non-sagittal moderation is required. The extent of moderation required is then directly related to the efficiency of sagittal plane loads. So, excessive loads, cause excessive destabilisation of the body, requiring greater moderation, and are more likely to result in undesirable joint loading (see figure GD.1). We found that higher anticipatory demand is more likely to result in excessive sagittal plane loading, and puts the individuals on a less favourable pathway. However, in figure GD.1 we also point out that it is possible to generate medial ground reaction forces through sagittal plane loading relatively efficiently and maintain this efficiency over the course of single match-simulation exertion. Thus, perhaps through training, one can adapt their performance movement strategy, moving away from excessive and inefficient loading, and this would certainly have positive implications for side cutting performance. Nevertheless, the extent of the final detrimental consequences may be dependent on how the corrective movement strategy is deployed to moderate contribution of sagittal triple acceleration to those forces.

Frontal plane hip acceleration - the corrective movement strategy

When medial forces generated by sagittal triple acceleration are excessive, then a corrective movement strategy is essential for task performance, but the key question is, to what extent is correction required? This doctoral thesis has demonstrated the prominent role of frontal plane hip acceleration to moderate medial ground reaction forces, which implies that much of the necessary correction is actually derived from a frontal plane hip movement strategy. Previous research has eluded to the importance of a hip strategy in the frontal plane (Sigward and Powers, 2007; Whyte et al., 2018), however, there is some inconsistency in the specified variable to represent this strategy. In fact, much of the available research seeks to attribute this correction to the amount of lateral trunk flexion (Dempsey et al., 2009; Jamison, Pan and Chaudhari, 2012; Brown, Brughelli and Hume, 2014; Mornieux et al., 2014; Jones, Herrington and Graham-Smith, 2015). However, research in walking (MacKinnon and Winter, 1993; Winter, 1995); with changes of direction (Patla, Adkin and Ballard, 1999); and later, specifically with side cutting (Houck, Duncan and De Haven, 2007), suggested that it is the role of the hip to control the trunk. This is in agreement with what we find for side cutting, and based on the mechanical principles of whole-body dynamic stability, frontal plane hip acceleration is more likely to be the true mechanism of moderation, providing more direct observation of the corrective movement strategy. Whereas lateral trunk flexion, and hip abduction are valuable observations, they can be considered indirect mechanical consequences.

Whilst there is a need for moderation from frontal plane hip acceleration in all scenarios presented as part of the current research, the nature of the deployment of the corrective movement strategy may also be important. In figure GD.1 (mechanism 3) two emerging options for frontal plane hip acceleration are suggested. From the studies in this thesis, it appears that whilst it is possible for individuals to be on a more dangerous pathway with mechanisms 1 and 2, resulting in excessive destabilisation of the body, it remains possible to mitigate negative consequences by providing more corrective control throughout ground contact. More specifically, deploying frontal plane hip acceleration with a more conservative medial impulse in the weight acceptance phase appears to lead to low-to-moderate peak knee abduction moments, despite previous excessive destabilisation. Whereas, high medial impulse, and therefore more substantial frontal plane hip acceleration in weight acceptance, is indicative of higher peak knee abduction moments. Furthermore, this less favourable pathway may lead to a reduced capacity of the corrective movement strategy in response to match-exertion. Therefore, the benefits of adopting a controlled corrective movement strategy throughout ground contact may be considerable when aiming to address issues around joint dynamic stability. However, further investigation is required on training strategy, especially with individuals who exhibit higher undesirable knee loads. That said, our findings here are in some agreement with earlier research from Sigward and Powers (2007) comparing females with normal and excessive frontal plane knee moments. They reported that higher risk individuals exhibited greater lateral ground reaction forces and increased hip abduction. Through our research we have now been able to explain the relationship between these variables within the context of a more direct measure of whole-body dynamic stability.

Transverse hip acceleration and CoP position - performance refinements

Following the more pronounced activity of the corrective frontal plane hip movement strategy, once the body has been re-stabilised, it may be possible to re-orientate the pelvis to the new direction of travel and extend the moment arm of the ground reaction force vector. It is thought that transverse hip acceleration and centre of pressure positioning, respectively, are direct indication of the capacity of these movement strategies in the context of whole-body dynamic stability. For the sake of clarity, it may be initially useful to consider them separately. Firstly, the ability to rotate the pelvis in the new direction of travel may be key for better side cutting performance, and this has been expressed in recent research (Byrne et al., 2017; Staynor, Donnelly and Alderson, 2018). Our findings highlight that the extent of transverse hip acceleration is probably associated with the extent of frontal plane hip acceleration, as has been suggested in the interaction of the hip in the frontal and transverse planes previously (MacKinnon and Winter, 1993; Houck, Duncan and De Haven, 2006). It appears that greater transverse hip acceleration results from excessive and rapid frontal plane hip acceleration, and may mean that it is harder to then moderate re-orientation and control of the pelvis in the new direction of travel (see figure GD.1).

Subsequently, we move to the final whole-body dynamic stability movement strategy, centre of pressure positioning. Similar to transverse plane hip acceleration, centre of pressure position seems to be related to the extent of frontal plane hip acceleration. As the centre of pressure position itself is likely to be an indication of the extent of subtalar inversion, observation of this mechanism demonstrates an ankle movement strategy in the context of whole-body dynamic stability. It appears to be easier in side cutting to engage the ankle movement strategy only once the hip movement strategy is less active, or has already re-stabilised the body, meaning the centre of pressure is then free to move more laterally. This hip-ankle movement strategy interaction likely represents the double inverted pendulum method for frontal plane balance expressed in earlier research (MacKinnon and Winter, 1993; Patla, Adkin and Ballard, 1999; and Houck, Duncan and De Haven, 2006). However, we have been able to represent this interaction as part of a broader series of movement strategies for whole-body dynamic stability. In this regard, the transition from hip to ankle movement strategies was most apparent in response to the match-simulation exertion, where with consistent corrective loading from the hip, the ankle movement strategy played a more prominent role for stabilising the body. Overall, whilst observation of transverse plane hip acceleration and centre of pressure position may only be useful once re-stabilisation is established, they provide a clear indication of performance refinements for side cutting in the context of whole-body dynamic stability.

Dynamic stability in different tasks

In this thesis, whilst the focus has been on highly dynamic tasks, it is clear that there are some similarities with the movement strategies adopted in other tasks like standing, walking and turning whilst walking. The findings of this thesis are in agreement with the importance of the roles of the hip, and, to a lesser extent, the ankle, in achieving stability and control of the CoM (Kuo, 1995; Winter, 1995; Patla, Adkin and Ballard, 1999; van Emmerik et al., 2016; Blenkinsop, Pain and Hiley, 2017). Specifically, the findings are consistent with reports from standing research that suggest the hip plays the largest role in mitigating unnecessary deviations (Kuo, 1995; Winter, 1995; Blenkinsop, Pain and Hiley, 2017). Whilst the current findings demonstrate that transition from hip to ankle movement strategy may exist, for dynamic tasks it seems unlikely that the body can limit itself to one movement strategy at a time. Indeed, this thesis demonstrates that there may be as many as five integrated movement strategies working to accelerate the body and mitigate unnecessary movement, at any one time. Furthermore, the studies in this thesis only focus on the medio-lateral control of the CoM, so the roles of movement strategies are likely to be more complex, once the factors that influence the coordination of the 3D nature of the GRF vector are considered.

The key findings throughout this thesis have demonstrated a clear association between the foot placement and sagittal plane impulses as important movement strategies to accelerate the body in the intended direction of travel. Although the importance of foot placement is in agreement with the literature on walking with and without a turn (MacKinnon and Winter, 1993; Patla, Adkin and Ballard, 1999), to the best of our knowledge the current findings are the first to show the direct interplay of these movement strategies in this role. Indeed, if one were to consider the order of priority of important movement strategies for dynamic tasks, foot placement is undoubtedly first, followed by sagittal plane impulses (see Figures GD.1 earlier and GD.2 below). Subsequently, the frontal plane hip movement strategy serves to mitigate unnecessary deviations, followed by the transverse plane hip and ankle movement strategies working to mitigate detrimental effects on side cutting performance. That said, in spite of some similarities in movement strategies, it is unlikely that this specific approach would work for standing tasks, due to absence of foot placement as part of the task, and the negligible requirement for CoM acceleration. However, for walking and other dynamic tasks, with some refinement toward the intended direction of CoM acceleration, the approach outlined in the thesis may offer some insights into important movement strategies and their role in whole-body dynamic stability.

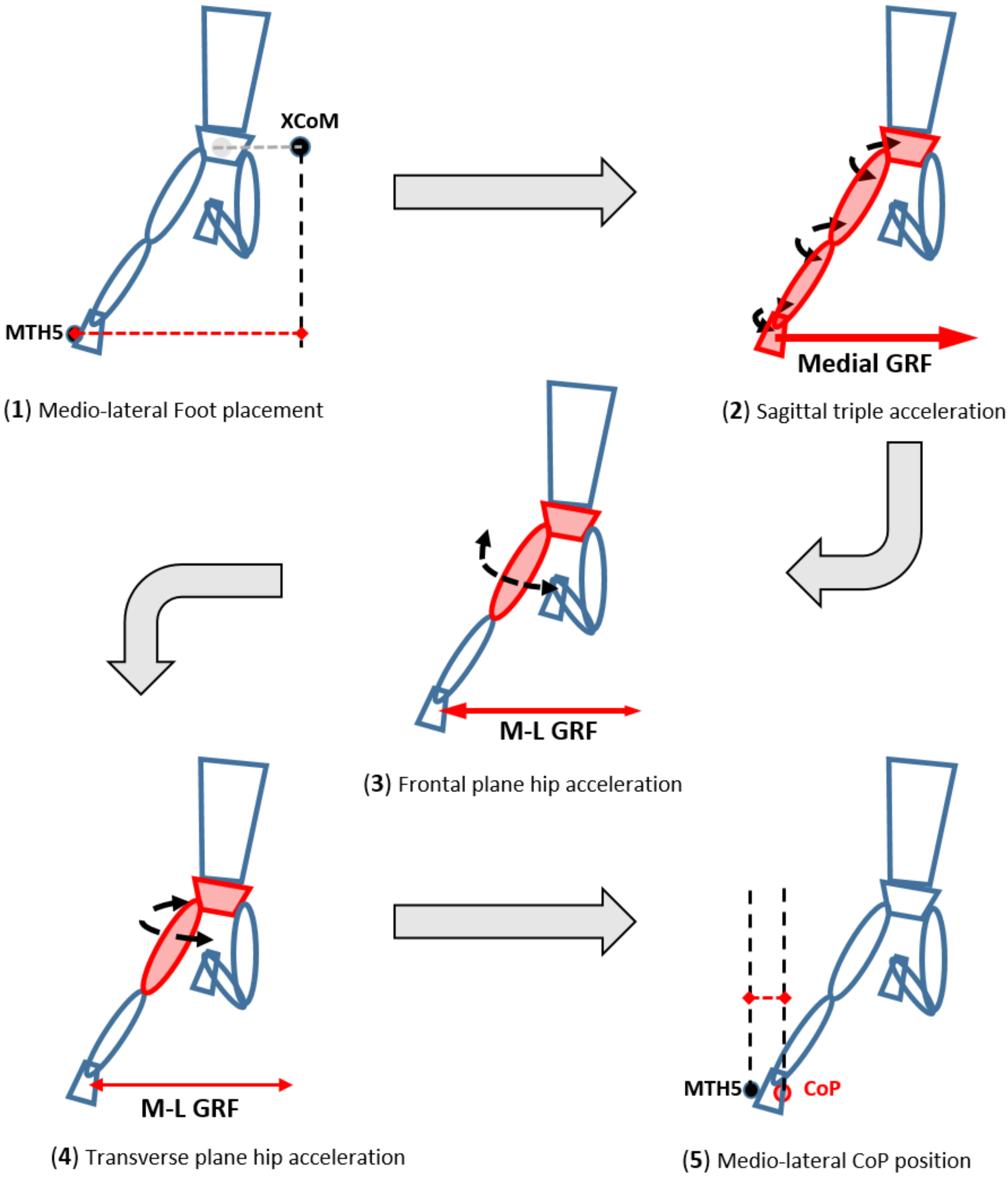

Figure GD.2. Diagram of the priority order of the five distinct whole-body dynamic stability movement strategies.

Practical Implications

When observing dynamic tasks from a biomechanical perspective, typically the priority is to advance injury prevention strategies whilst maintaining maximal performance. The findings of this thesis offer a unique perspective to contribute to this in an integrated manner. Previous studies have failed to provide meaningful training interventions to reduce injury risk in side cutting (Donnelly et al., 2015; Whyte et al., 2018). Specifically, research which reported the effects of a substantial training programme for trunk or 'core strength' and various intensities of balance training did not find any reduction in peak knee abduction moment (Donnelly et al., 2015; Whyte et al., 2018). Although some studies presented favourable findings with some technical corrections (Dempsey et al., 2009) it is not clear whether those adaptations would remain, or whether they would change under circumstances of limited anticipation or match-exertion. In figure GD.1 the key findings of this thesis are represented along with a suggestion for an alternative approach for developing a training strategy in consideration of whole-body dynamic stability. Specifically, it may be useful when side cutting involves a high anticipatory demand to train individuals to be able to deploy a wider foot placement, if required, as this may have performance benefits. Such training may initially involve strategies to improve reaction time to a relevant unanticipated cueing stimulus, and postural preparation for the ground contact and turn. However, as is suggested in figure GD.1, unless this is accompanied by training and observation of whole-body dynamic stability mechanisms downstream, this is likely to result in task failure or side cutting performed with undesirable joint loads. Specifically, wider foot placement following limited anticipation must be completed with more efficient sagittal plane loading and the ability to moderate frontal plane loads with lower medial impulse in the weight acceptance phase of the task. The Sagittal Efficiency Ratio was presented as a reference of how excessive the medial impulses from sagittal triple acceleration were above the total ground reaction force impulses derived from induced acceleration analysis. In this thesis the most excessive sagittal plane forces were observed in the high peak knee abduction group in study 4. This group had an average Sagittal Efficiency Ratio of 142%, whereas the most efficient forces were observed in the low knee loading group at the end of the match-simulation, averaging 104%.

In applied settings, practitioners could utilise the Sagittal Efficiency Ratio to understand the status of performance or potential for dynamic destabilisation in their athletes which may help with developing training intervention or making decisions about rehabilitation of athletes and returning to play. For example, one may expect that compensatory mechanisms following injury would mean the Sagittal Efficiency Ratio is high, indicating excessive sagittal plane loading, and excessive destabilisation. However, further research may be required with a previously injured population. Theoretically, the ratio should be tolerant to a variety of task intensities and performance constraints. Alternatively, an in-depth observation of whole-body dynamic stability, as this thesis demonstrates, may allow for a more detailed account of the effectiveness of any training intervention. Perhaps, initially focusing on foot placement, with the performance and corrective movement strategies, and their interaction, as a priority. More broadly, practitioners could develop programmes for generating medial ground reaction forces; improving sagittal efficiency; training the hip movement strategy and faster transition to the ankle movement strategy in ground contact; improving reaction time and foot placement in response to perturbations or progressively challenging demands. In any of these scenarios, and perhaps those that have more clinical relevance, our novel method for quantifying whole-body dynamic stability could be a valuable screening tool.

Future directions

Whilst this thesis has served to outline and highlight potential uses and the robustness of the measurement of whole-body dynamic stability, there are potentially three key areas to develop the applied context of this approach. Firstly, it would be useful to understand more about the clinical relevance of this observation, which may be possible by comparing injured and healthy populations, whether that be prospectively or retrospectively. For retrospective observations (post-injury), the research design may have to consider a reduction in task intensity, or the sample may need to be controlled for those only in later stage rehabilitation before returning to match-play. However, it would be important to know how whole-body dynamic stability may be affected by the injury to clarify aspects of the framework presented earlier, and accurately inform screening observation and training intervention following rehabilitation. Indeed, one dangerous outcome to avoid is returning to match-play too soon, where re-injury is then a common problem. Observation of whole-body dynamic stability may help safeguard against this, allowing the biomechanist to identify potentially dangerous movement strategies that are not clear when measuring peak knee abduction moments alone, let alone when qualitatively observing movement in clinical settings. Secondly, although not necessarily in a clinical scenario, there is scope for refining the laboratory based observation of side cutting, as consistent limitations across the studies of this thesis may be attributable to the constraints of the task. Whilst it is often necessary to constrain tasks that are observed in a laboratory setting, it may be possible to analyse variable anticipatory, approach speed, and angular demands of the side cutting task, perhaps even developing this approach towards maximal side cutting performance. Alternatively, a more natural progression to the complexities of the dynamic tasks in sport, would be to develop observation within actual match-play, or small-sided games for starters. Whilst this may be possible with wearable technologies or player tracking and machine learning in future, their reliability and complexity are still inadequate. To date, the value of well-directed laboratory based observation is retained. Finally, when addressing our match-simulation findings it appears that match-exertion in a singular form may not offer further evidence of injury incidence. However, the reason for this may exist within the growing body of research on player load and recovery (Vanrenterghem et al., 2017; Windt and Gabbett, 2017). Reduced biomechanical load recovery between games may be the real issue leading to higher injury incidence. In this regard, it may be useful to observe the response of whole-body dynamic stability to scenarios where accumulations of biomechanical loading is high and recovery is low, perhaps in the form of multiple match-simulations within a short timeframe. Such research may have important implications for the understanding of biomechanical recovery for in-season training programmes, particularly in the case of high match congestion situations.

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Summary

The general objective of this doctoral thesis was to develop a reliable and robust observation of the mechanisms and movement strategies to control the centre of mass, which represent the status of whole-body dynamic stability, whilst exploring the implications for markers of injury risk and performance.

Initially, it was important to establish the reliability of the general approach used to collect and analyse biomechanical data from the side cutting task. In this regard, it was determined that was tolerant to some common data collection issues, and did not pose any specific cause for concern, beyond what may be expected for this type of biomechanical investigation. However, it was observed that knee joint loading and ground reaction force data were distinctly more variable in the early phase of ground contact, in comparison to later on in the task. This novel finding may mean that the early protective phases of ground contact are harder to control with repeated performance, and suggests any subsequent movement strategies for achieving control may need to adapt in often challenging scenarios.

To quantify whole-body dynamic stability holistically in side cutting, and therefore address the primary aim of this doctoral thesis, advanced and relatively novel biomechanical methods were used to express characteristics of the medio-lateral ground reaction force vector. Within the five mechanisms quantified, three key mechanisms emerged as the priority for explaining the status of whole-body dynamic stability. Firstly, our findings have highlighted how important the foot placement can be to dictate the nature of the subsequent movement strategies, and although a narrower foot placement was necessary with limited anticipation time, this may ultimately lead to undesirable knee loading. Secondly, our findings highlighted the primary movement strategy for developing medial ground reaction

forces was through sagittal triple acceleration, or the combined contribution of the hip, knee and ankle in the sagittal plane. Whilst this movement strategy is essential, the medial ground reaction forces generated by sagittal plane loading are often excessive, to the extent that excessive destabilisation of the body occurs and a corrective movement strategy is required to moderate and re-stabilise the body. Thirdly, the extent of destabilisation is highlighted in the use of a hip movement strategy in the frontal plane that is activated to retrieve control of the centre of mass and re-align the lower and upper body and reducing the dangerous consequences as soon as possible. Once this correction is achieved we observe a transition to an ankle movement strategy that allows a more lateral position of the centre of pressure, which coincides with the ability to also redirect the pelvis in the new direction of travel. Furthermore, and promisingly, increases in the extent of the hip movement strategy may provide advanced warning of increases in peak knee abduction moment, and therefore, potentially injury risk.

In addition, in the framework outlined for the interaction of biomechanical loading in the context of whole-body dynamic stability, we have been able to identify particular pathways that may lead to undesirable consequences, as well as safer alternatives. This framework has important implications for understanding how whole-body dynamic stability can inform injury and performance objectives in research and practice in this field. Typically movement strategies adopted by participants tested in the current studies were mostly successful in controlling the centre of mass in a manner that did not compromise unwanted loading, and perhaps, joint stability. However, we were able to establish some insight in certain higher risk individuals, observing that greater undesirable knee loading appears to stem from deficiencies in mechanisms of, and therefore reduced, whole-body dynamic stability.

In conclusion, this doctoral thesis has outlined a novel and robust observation of the role of whole-body dynamic stability movement strategies that has the potential to inform the injury and performance paradigms in an integrated manner, advancing this field of research. Quantifying whole-body dynamic stability may have broader application than the scope outlined in this thesis, perhaps including change of direction performance in more sport-specific scenarios or clinical observation of previously injured individuals.

Appendices

Appendix 1. General Methods – Details of Liverpool John Moores University (LJMU) Lower Limb and Trunk (LLT) 3D motion capture model (Malfait et al., 2014).

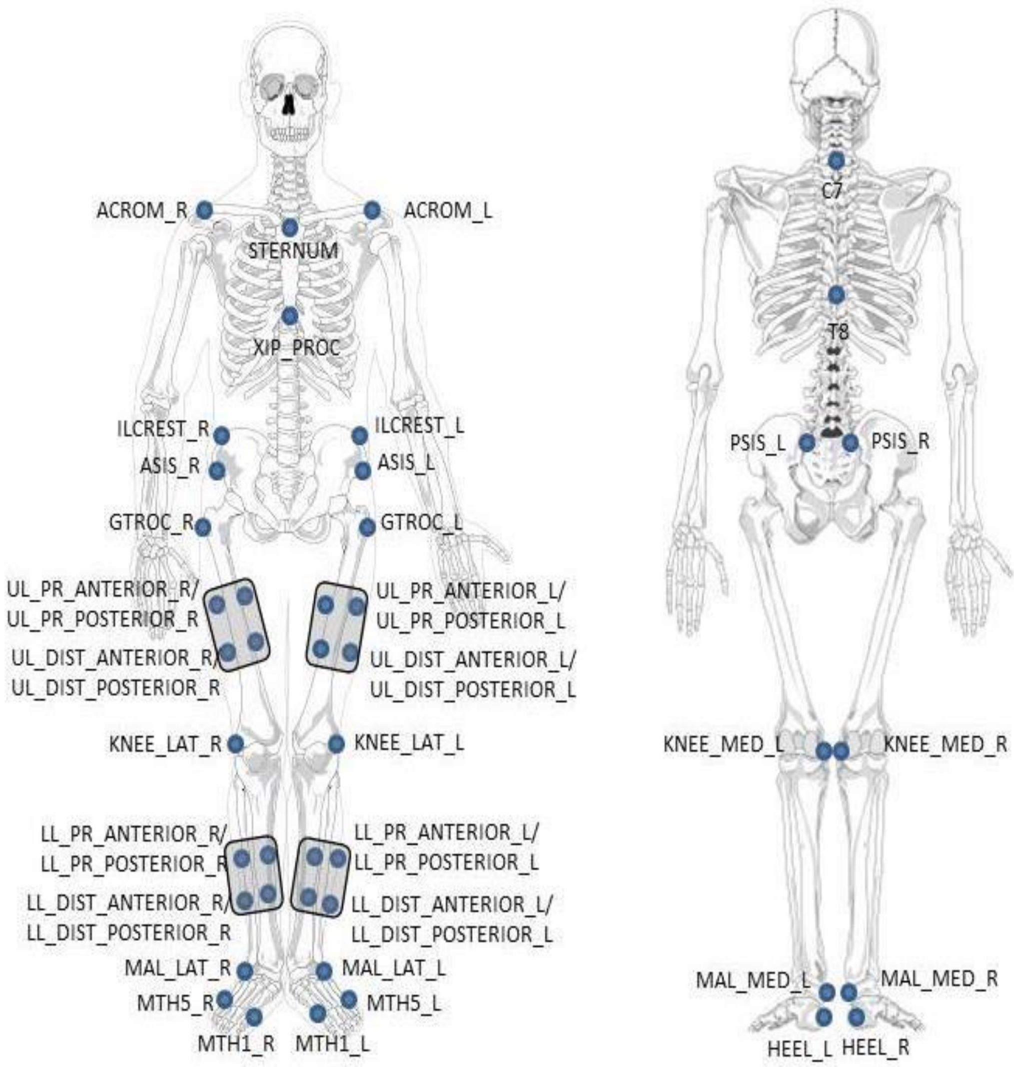
1.1. Anatomical Markers

Trunk	
C7	Processus spinosus vertebra C7
STERNUM	Sternum
XIP_PROC	Xiphoid process
T8	Processus spinous vertebra T8
ACROM_L	Acromion left (acromioclavicular joint)
ACROM_R	Acromion right (acromioclavicular joint)
Pelvis	
ASIS_L	Anterior sacral iliac spine left
PSIS_L	Posterior sacral iliac spine left
ILCREST_L	Iliac crest left
ASIS_R	Anterior sacral iliac spine right
PSIS_R	Posterior sacral iliac spine right
ILCREST_R	Iliac crest right
Lower limbs	
GTROC_L	Greater trochanter left
KNEE_MED_L	Knee medial femoral epicondyle left
KNEE_LAT_L	Knee lateral femoral epicondyle left
MAL_MED_L	Malleolus medial left
MAL_LAT_L	Malleolus lateral left
HEEL_L	Heel left
MTH1_L	Metatarsal head 1 left
MTH5_L	Metatarsal head 5 left
GTROC_R	Greater trochanter right
KNEE_MED_R	Knee medial femoral epicondyle right
KNEE_LAT_R	Knee lateral femoral epicondyle right
MAL_MED_R	Malleolus medial right
MAL_LAT_R	Malleolus lateral right
HEEL_R	Heel right
MTH1_R	Metatarsal head 1 right
MTH5_R	Metatarsal head 5 right

1.2 Marker Clusters

UL_PR_ANT_L	Upper leg proximal anterior left
UL_PR_POST_L	Upper leg proximal posterior left
UL_DI_ANT_L	Upper leg distal anterior left
UL_DI_POST_L	Upper leg distal posterior left
LL_PR_ANT_L	Lower leg proximal anterior left
LL_PR_POST_L	Lower leg proximal posterior left
LL_DI_ANT_L	Lower leg distal anterior left
LL_DI_POST_L	Lower leg distal posterior left
UL_PR_ANT_R	Upper leg proximal anterior right
UL_PR_POST_R	Upper leg proximal posterior right
UL_DI_ANT_R	Upper leg distal anterior right
UL_DI_POST_R	Upper leg distal posterior right
LL_PR_ANT_R	Lower leg proximal anterior right
LL_PR_POST_R	Lower leg proximal posterior right
LL_DI_ANT_R	Lower leg distal anterior right
LL_DI_POST_R	Lower leg distal posterior right

1.3 Physically placed markers



1.4 Virtual landmarks

THORAX_PROX	Midpoint between C7 and STERNUM
THORAX_DIST	Midpoint between T8 and XIP_PROC
F_L(R)HIP	Functional hip joint
F_L(R)KNEE	Functional knee joint
F_L(R)KNEE_X	Projected landmark offset along functional knee axis
L(R)LK	Lateral knee joint marker projected onto functional knee axis
L(R)MK	Medial knee joint marker projected onto functional knee axis
L(R)ANKLE	Midpoint between MAL_MED_L(R) and MAL_LAT_L(R)
L(R)TOE	Midpoint between MTH1 and MTH5

1.5 Segment definitions (anatomical and technical frames)

Thorax/Abdomen:

Origin: Midpoint of the line connecting the ACROM_R and ACROM_L
Z-axis: Line connecting the Origin and the midpoint of ILCREST_R and ILCREST_L, pointing vertically
Y-axis: Line perpendicular to the Z-axis and a least-squares plane fit to the
ACROM_L, ACROM_R, ASIS_L and ASIS_R, pointing anteriorly
X-axis: Cross-product of the plane formed by the Z and Y axes, pointing laterally **Tracking Markers**: C7, STERNUM, T8, XIP_PROC

Pelvis:

Origin: Midpoint of the line connecting ILCREST_R and ILCREST_L
Z-axis: Line connecting the Origin to the midpoint of the line connecting the GTROC_R and GTROC_L, pointing vertically
Y-axis: Line perpendicular to the Z-axis and a least-squares plane fit to the
ILCREST_R, ILCREST_L, GTROC_L and GTROC_L, pointing anteriorly
X-axis: Cross-product of the plane formed by the Z and Y-axes, pointing laterally **Tracking Markers**: From ASIS, PSIS, ILCREST

Thighs:

Origin: Coincident with F_L(R)HJC
Z-axis: Line connecting F_L(R)HJC to midpoint of the line connecting L(R)LK and L(R)MK, pointing upwards
Y-axis: Line perpendicular to the Z-axis and the plane formed by L(R)LK and
L(R)MK, pointing anteriorly
X-axis: Cross-product of the plane formed by the Z- and Y-axes, pointing laterally
Tracking Markers: Upper Leg marker cluster

Shanks:

Origin: Midpoint of the line connecting L(R)LK and L(R)MK
Z-axis: Line connecting midpoint of the L(R)LK and L(R)MK and L(R)ANKLE, pointing vertically
Y-axis: Line perpendicular to the Z-axis and the plane formed by the L(R)MK,
L(R)LK and L(R)ANKLE, pointing anteriorly
X-axis: Cross-product of the plane formed by the Z and Y-axes, pointing laterally **Tracking Markers**: Lower Leg marker cluster

Feet:

Origin: Coincident with L(R)ANKLE
Z-axis: Line connecting L(R)ANKLE and the midpoint of the line between MTH5_L(R) and MTH1_ L(R), pointing posteriorly
Y-**axis**: Line perpendicular to the Z-axis and plane formed by the L(R)ANKLE,
MTH5_L(R) and MTH1_L(R), projecting vertically
X-**axis**: Cross-product of the plane formed by the Z and Y-axes, pointing right **Tracking Markers**: From HEEL, MTH5, MTH1, MAL_LAT

Virtual Feet:

Origin: Coincident with HEEL_L(R)
Z-axis: Line connecting HEEL_L(R) and L(R)TOE, pointing vertically
Y-**axis**: Line perpendicular to the Z-axis and plane formed by the HEEL_L(R),
L(R)TOE & RANKLE, pointing anteriorly
X-**axis**: Cross-product of the plane formed by the Z and Y-axes, pointing laterally **Tracking Markers**: HEEL, MTH5, MTH1

Appendix 2. Study 1 – Variability plots for normalised knee abduction moments and medial ground reaction forces.

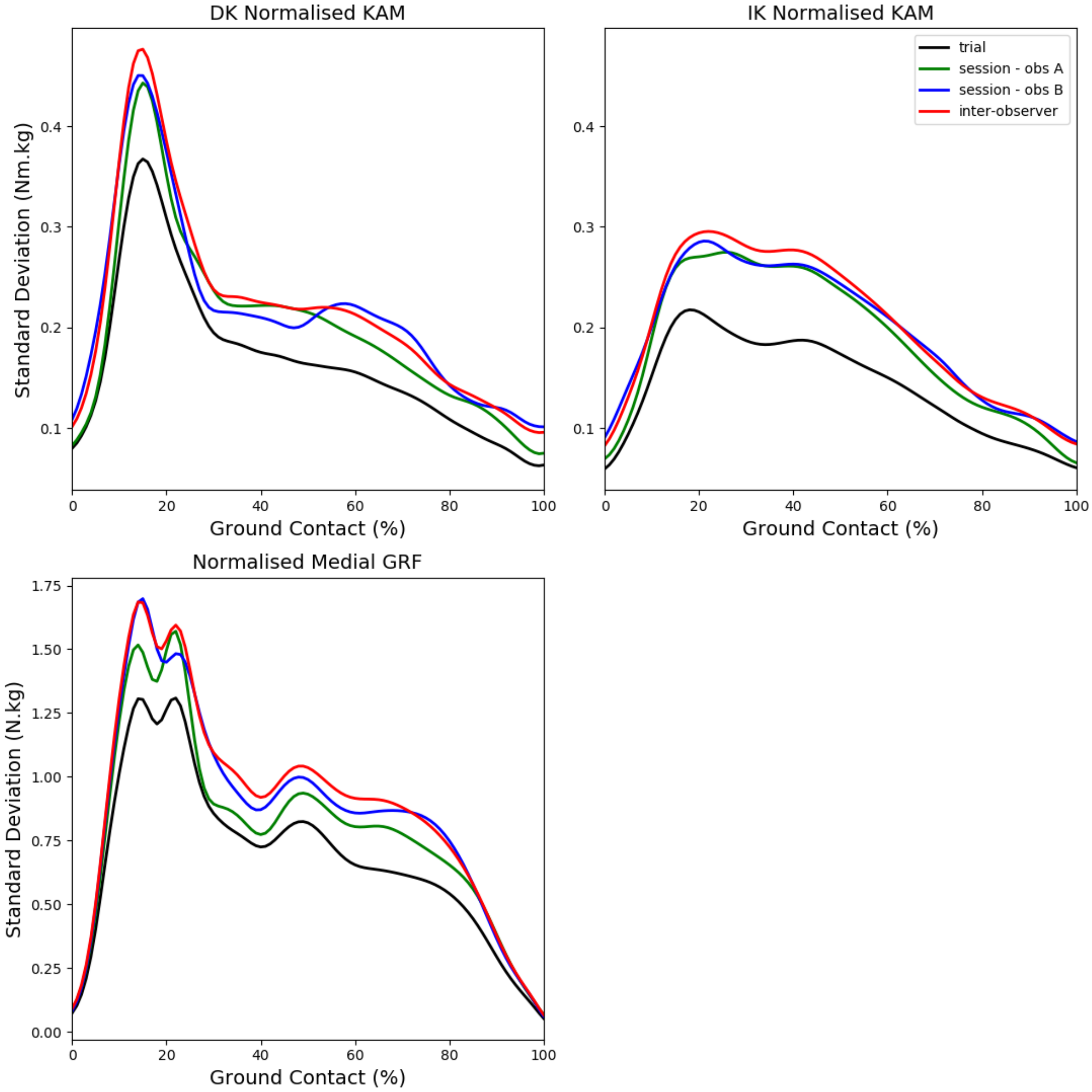


Figure 1.5. Additional variability plots over ground contact for average normalised knee abduction moment (KAM) data and medial ground reaction forces (GRF) from side cutting. Row one shows the knee abduction moment data from Direct Kinematic (DK) followed by the Inverse Kinematic (IK) modelling approaches. Row two shows the Normalised medial GRF data.

Appendix 3. Study 1 – additional regression data.

This appendix contains the linear regression analysis for the task execution variables and the knee angle and moment components. There are no significant relationships between the task execution variables and the angle and moment components that coincide with the times of high variability during weight acceptance.

(a) Knee angle sagittal plane vs. task execution variables

(b) Knee angle frontal plane vs. task execution variables

(c) Knee angle transverse plane vs. task execution variables

(d) Knee moment sagittal plane vs. task execution variables

(e) Knee moment frontal plane vs. task execution variables

(f) Knee moment transverse plane vs. task execution variables

Abbreviations:

CoM Vel TD: Centre of mass velocity at touch down

CoM Vel TO: Centre of mass velocity at take off

CoM Ang TD: Approach angle

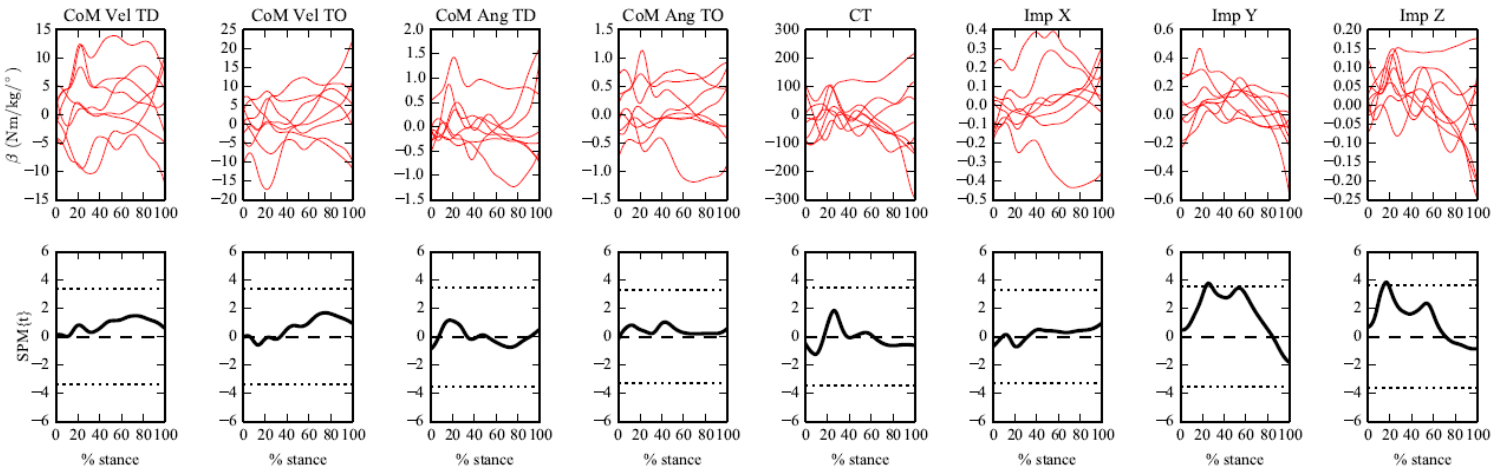
CoM Ang TO: Exit (cut) angle

CT: Contact time

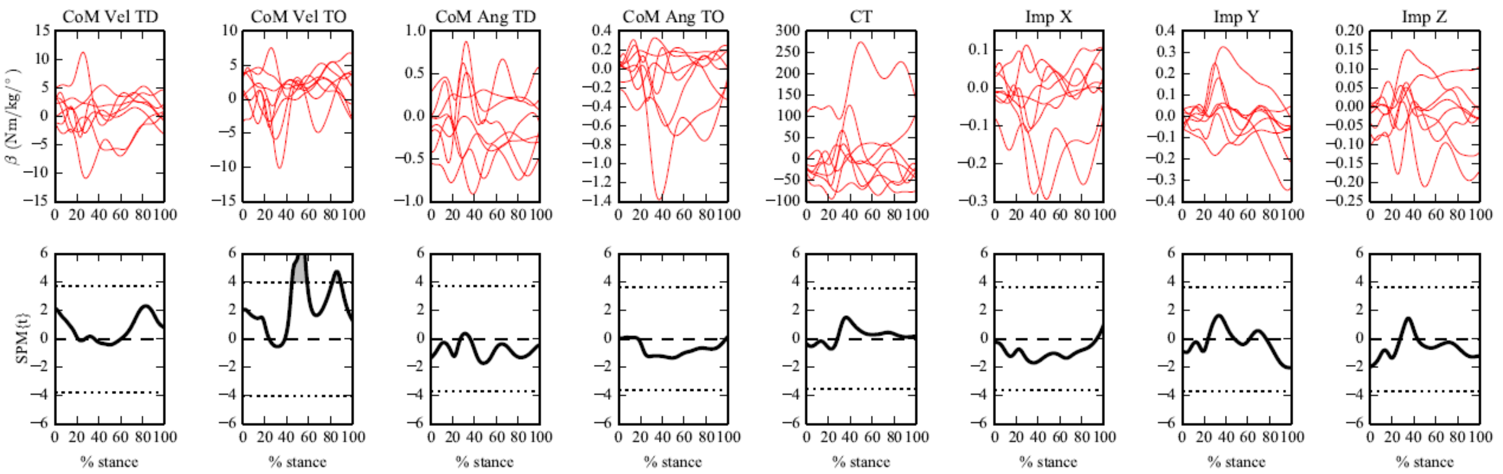
Imp X: Medio-lateral impulse

Imp Y: Anterior-posterior impulse

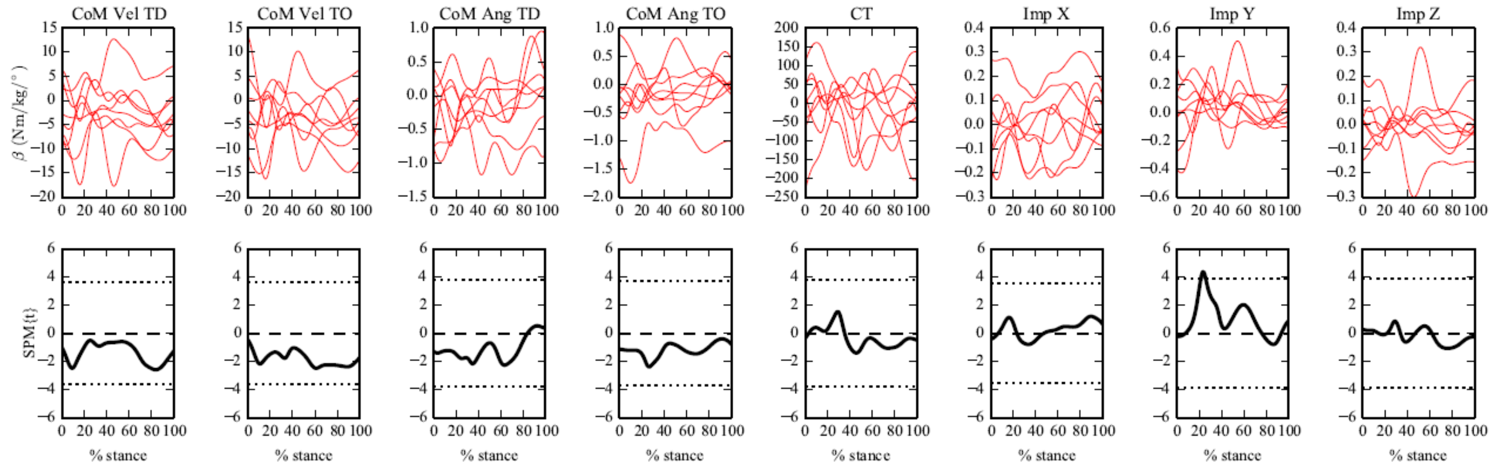
Imp Z: Vertical impulse


CoM Vel TD
CoM Vel TO
CoM Ang TD
CoM Ang TO
CT
Imp X
Imp Y
Imp Z
β (Nm/kg/°)
SPM{t}
% stance

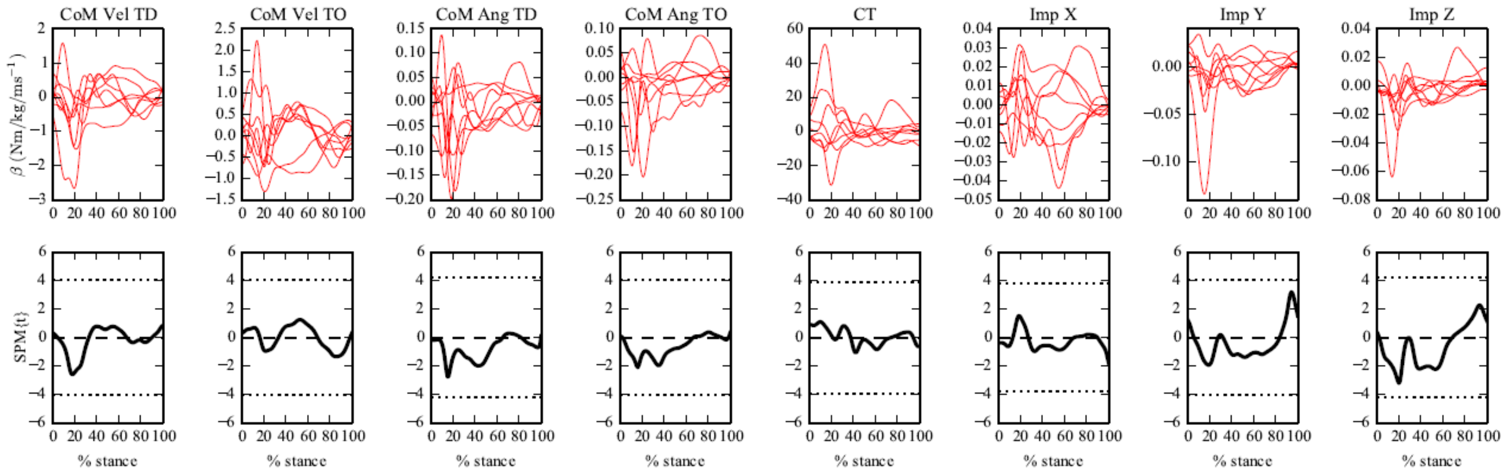
(a)


CoM Vel TD
CoM Vel TO
CoM Ang TD
CoM Ang TO
CT
Imp X
Imp Y
Imp Z
β (Nm/kg/°)
SPM{t}
% stance

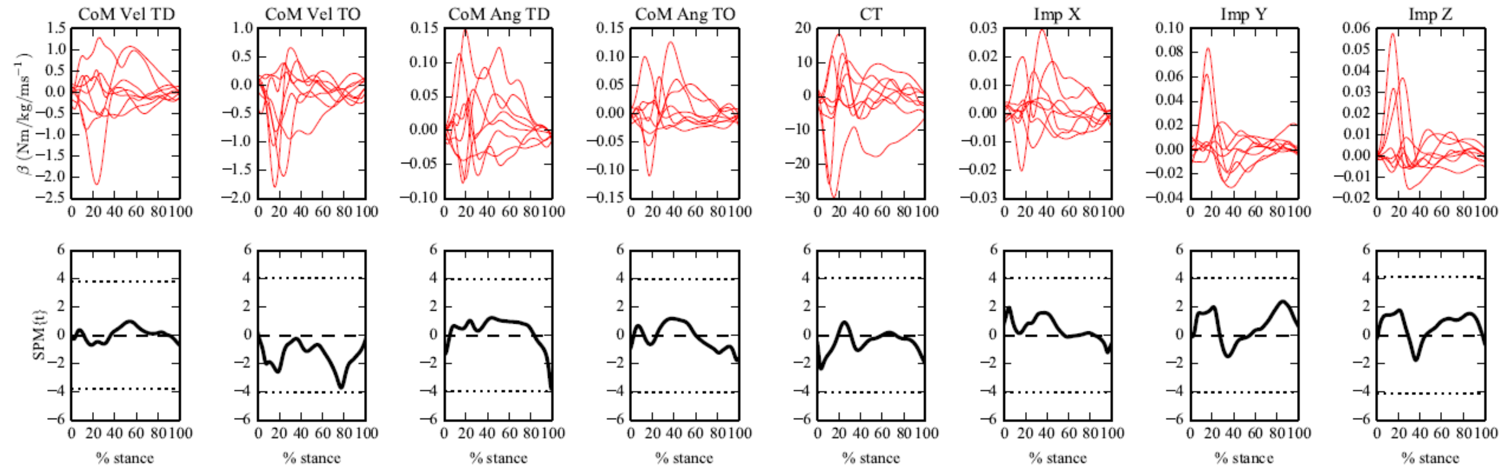
(b)


CoM Vel TD
CoM Vel TO
CoM Ang TD
CoM Ang TO
CT
Imp X
Imp Y
Imp Z
β (Nm/kg/°)
SPM{t}
% stance

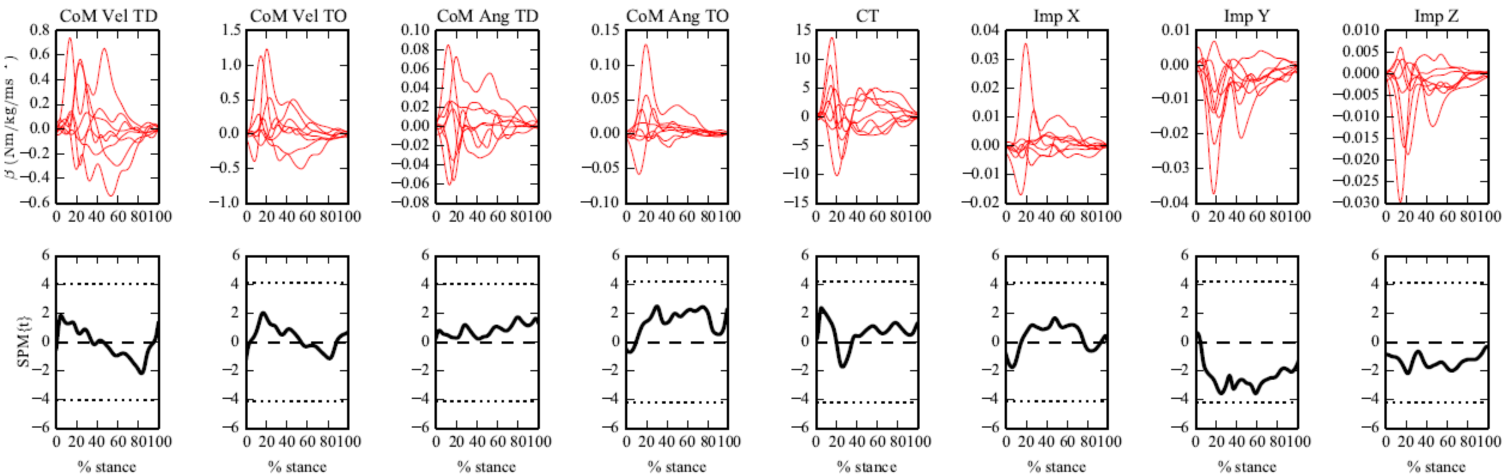
(c)


CoM Vel TD
CoM Vel TO
CoM Ang TD
CoM Ang TO
CT
Imp X
Imp Y
Imp Z
β (Nm/kg/ms^{-1})
SPM{t}
% stance

(d)


CoM Vel TD
CoM Vel TO
CoM Ang TD
CoM Ang TO
CT
Imp X
Imp Y
Imp Z
β (Nm/kg/ms⁻¹)
SPM{t}
% stance

(e)


CoM Vel TD
CoM Vel TO
CoM Ang TD
CoM Ang TO
CT
Imp X
Imp Y
Imp Z
SPM{t}
% stance

(f)

Appendix 4. Study 1 - Sample Size Estimation

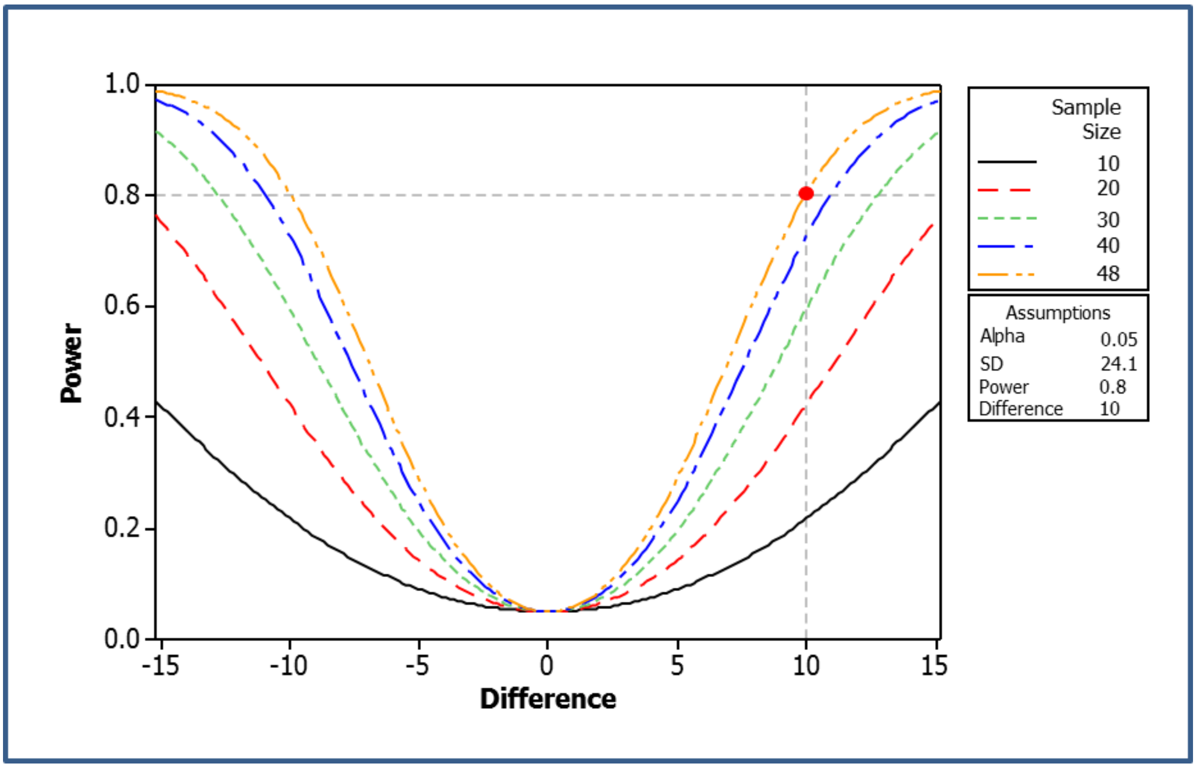
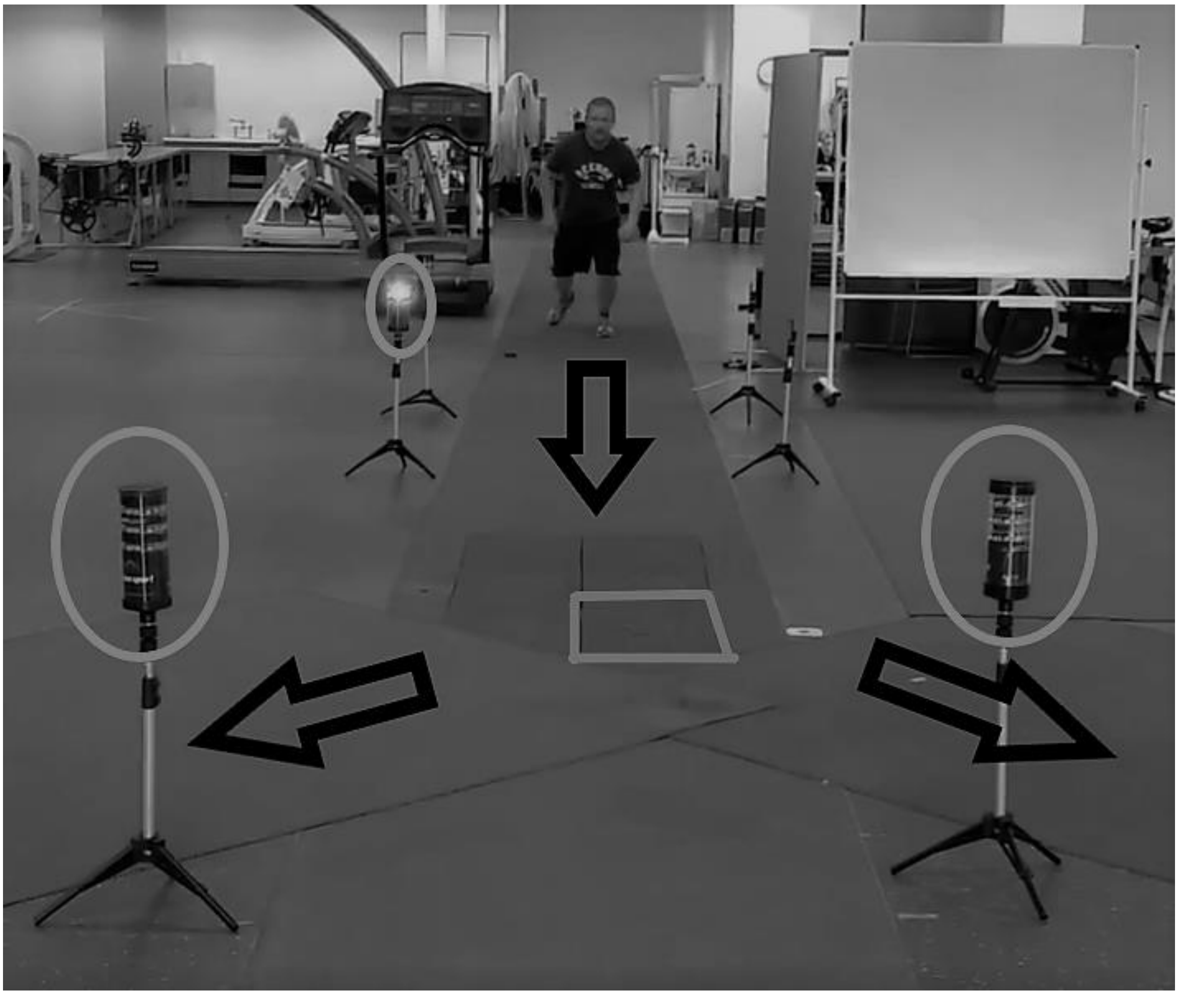


Figure 1.6. An illustration of the sample size estimation based on the peak knee joint moment in the frontal plane. Sample sizes are plotted at intervals of 10 participants until the 10 Nm difference intersects a statistical power of 80 % (n≥48). The alpha level was set to 0.05. The SD was taken from the inter-trial calculations (24.1 Nm).

Appendix 5. Study 2 - Diagram of the laboratory set-up for the 45° unanticipated side cutting task. Cueing lights and the force plate are highlighted, the approach and change of direction paths are marked with arrows.



Appendix 6. Study 2 - 0D betas for multiple regression of foot placement against selected side cutting performance outcomes (one subject per row).

Average M-L CoM Acceleration (m/ ms^{-2})	Change of Direction angle (m/ degrees)	Peak KAM (m/ Nm.kg)
1.34	-1.11	-1.94
0.25	-2.43	-2.90
-0.82	-1.07	-1.79
1.05	0.16	0.16
2.33	-1.07	-5.40
-0.72	-3.71	-3.54
1.14	-0.68	-0.22
0.31	-2.20	-0.76
-1.62	-3.23	-2.84
2.05	-2.46	-3.06
-1.13	-4.22	-0.87
1.67	-1.77	-0.37
3.49	-2.33	-0.54
2.75	-3.77	-2.85
0.50	-0.83	-1.44
1.41	-2.38	-0.23
2.36	-4.28	-1.46
0.25	-1.90	0.30
1.24	-1.39	-4.26
-0.14	-1.01	-0.05