Comparison of leg dominance and fatigue state on lower extremity kinematics during cutting manoeuvres in male soccer players

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Submitted in the fulfilment of the requirements for the degree of Master of Arts in Human Movement Science (Research) in the Faculty of Health Sciences at the NELSON MANDELA UNIVERSITY

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DECLARATION

Submitted in the fulfilment of the requirements for the degree of Master of Arts in Human Movement Science (Research) in the Faculty of Health Sciences at the NELSON MANDELA UNIVERSITY

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Qualification: Master of Arts (MA) degree in Human Movement Science

Title of treatise: Comparison of leg dominance and fatigue state on lower extremity kinematics during cutting manoeuvres in male soccer players

Declaration: In accordance with Rule G4.6.3, I hereby declare that the above-mentioned proposal is my own work and that it has not previously been submitted for assessment to another University or for another qualification.

Date: December 2018

Signature:

Acknowledgements

“Any fool can know. The point is to understand.” Albert Einstein

Firstly, I would like to acknowledge my Lord and savior Jesus Christ, whose all sufficient grace and mercy has and continues to carry me through this journey. I am forever grateful to Him who grants me the ability to learn and the knowledge and insight to grow even though I don’t deserve it.

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Abstract

Context. Soccer is one of the most popular sports played in South Africa and around the world. Soccer is a high intensity, semi-contact sport which is associated with an increased prevalence of injuries, especially to the lower extremities. Central and neuromuscular fatigue is believed to cause changes to kinetic and kinematic patterns of soccer players which may increase the risk for injuries, specifically related to the anterior cruciate ligament (ACL).

Purpose. To investigate the effects of fatigue on knee joint kinematics during the stance phase of a cutting manoeuvre of the dominant and non-dominant legs.

Design and method. A quantitative approach, more specifically, an experimental study design was adopte and a quasi-experimental study design was selected. A 'within-participants post-test only design' was used, which is also known as a 'repeated measures design' because all participants were 'repeatedly' measured under each experimental condition. Due to the non-randomization of the quasi-experimental design, non-probability sampling was utilized to sample the population group for the proposed study.

Results. A total of 13 male soccer players volunteered for the study. The participants had the following characteristics (mean ± SD): age 22.15 ± 2.77 years; height 169.64 ± 5.75 cm and weight 64.60 ± 7.04 kg. Non-significant differences within hip joint kinematics were observed between the dominant and non-dominant legs in a non-fatigued state in all three planes of motion ($F = 0.61$, $p = 0.55$). Similar kinematic characteristics were observed for the knee joint ($F = 1.25$, $p = 0.48$) and the ankle joint ($F = 3.33$, $p = 0.64$). Non-significant differences were also observed during the fatigued state in all three planes of motion for the hip joints ($F = 8.98$, $p = 0.72$), knee joints ($F = 5.21$, $p = 0.41$) and ankle joints ($F = 0.21$, $p = 0.12$). Peak vertical forces were however significantly different between the fatigued state compared to the non-fatigued state during the cutting manoeuvre ($F = 23.51$, $p = 0.035$), thereby indicating that neuromuscular fatigue may influence landing forces on impact during a directional change.

Conclusion. The effect of leg dominance did not have a statistically significant impact on any kinematic measures as well as the interactions between fatigue and non-fatigue trials were also not observed for any of the kinematic parameters. Several initial contact and peak stance–phase lower limb-joint rotations were influenced by fatigue during the execution of the sub-maximal 60° cutting manoeuvre. The main effect of fatigue produced an increase in knee internal rotation and hip abduction and a decrease in peak knee abduction angles compared to non-fatigue, but they were not statistically significant. Significant differences were found between dominant and non-dominant legs as well as between fatigue and non-fatigue with ankle pronation ($p=0.007$) and ankle external rotation ($p=0.033$). Knee abduction angle during cutting ($p=0.061$) also showed an effect even though not statistically significant. The purpose was to examine the combined effects of leg dominance and fatigue on lower-limb biomechanics during a sub-maximal 60° cutting manoeuvre. The conclusion of the present study related to limb dominance was that no statistically significant differences were evident for any of the dependent variables (limb dominance; fatigue state) related to the independent variables (i.e. joint [hip, knee, ankle], contact time, ground reaction however, between-subject fatigue variations that is large enough could negatively impact the biomechanical data comparisons. Future research should target specific locations of fatigue within a general fatigue paradigm and develop standardized tasks to achieve this.

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List of Abbreviations

ACL Anterior Cruciate Ligament

EMG Electromyogram

CP Creatine Phosphate

CNS Central Nervous System

3D Three-Dimensional

ROM Range of Motion

DoF Degrees of Freedom

BMI Body Mass Index

DNF Dominant Non-Fatigue

DF Dominant Fatigue

NDNF Non-Dominant Non-Fatigue

NDF Non-Dominant Fatigue

GRF Ground Reaction Force

COD Change of Direction

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Chapter 1: FORMULATION OF THE RESEARCH PROBLEM

1.1 Introduction

Soccer is a semi-contact sport comprised of participants of varying levels of performance ability (McLean, Walker & Van Den Bogert, 2005). In 2007 FIFA conducted a large-scale survey involving 207 member associations showing that football has strengthened its position as the world's number one sport since FIFA's last 'Big Count' in 2000. The popularity of soccer is vast, with approximately 265 million players being actively involved in soccer around the world, roughly about 4% of the world's population, and approximately 1. 486 million in a South African context (FIFA, 2007). The FIFA World Cup draws over 200 nations into the quest to qualify and 3.3 million spectators watch the month-long tournament (Giulianotti & Robertson, 2012). The nature of soccer play dictates a reliance on multiple directional changes (702 ± 267) at varying speeds for successful sporting performance (Bloomfield, Polman & O'Donoghue, 2007). The requirement of multiple directional changes is in turn reliant on good neuromuscular development, specifically in terms of agility, balance, coordination and flexibility of the lower extremity (Lambert, St. Clair Gibson & Noakes, 2005; Markolf, Burchfield, Shapiro, Shepard, Finerman & Slauterbeck, 1995). Lower extremity injuries typically account for 60% of all soccer-related injuries, of which non-contact injuries account for approximately 58% (Hawkins, Hulse, Wilkinson, Hawkins, Hulse, Wilkinson, Hodson, Gibson, Way & Care, 2001). Research has shown that players typically obtain at least one performance inhibiting injury per year (Chomiak, Junge, Peterson & Dvorak, 2000; Myer, Sugimoto, Thomas, Hewett, Schmikli, de Vries, Inklaar, Backx, Waldén, Atroshi, Magnusson, Wagner, Hägglund, Dvorak, Junge, Dvorak, Godolias, St Clair Gibson, Lambert, Noakes, Nagano, Ida, Akai & Fukubayashi, 2011) . Mechanisms of knee joint mechanics during directional changes in fatigued and non-fatigued situations therefore need to be investigated further in an attempt to mitigate the injury rate and severity.

1.2 Contextualization of the problem

The frequency of soccer injuries is estimated to be 10-35 injuries per 1000 playing hours, with the majority of injuries occurring in the lower extremities, mainly the knees and ankles (Alentorn-Geli, Myer, Silvers, Samitier, Romero, Lázaro-Haro & Cugat, 2009; Dvorak, Junge, Chomiak, Graf-Baumann, Peterson, Rösch & Hodgson, 2000) . Most injuries occur in the second half of a game thus researchers speculate that fatigue could play a role in this phenomenon (Gioftsidou, Malliou, Pafis, Beneka, Tsapralis, Sofokleous, Kouli and Godolias, 2012).

Neuromuscular training programs have successfully been applied by soccer players to reduce the rate of injuries, especially those associated with the anterior cruciate ligaments (ACL). Waldén, Atroshi, Magnusson, Wagner and Hägglund (2012) applied a cluster-randomized control design to a total of 4564 young soccer players (230 teams) and intervened with a 15-minute neuromuscular warm-up focused on the development of core stability, balance and jump landing with correct knee alignment. The intervention was led by a coach and was carried out twice a week throughout the seven-month competitive season in 2009. The clubs in the control group continued with training as per usual. Individual playing time and acute knee injuries causing loss of time from play were recorded during the season. Waldén *et al.* (2012), found that a neuromuscular warm-up programme reduced the rate of anterior cruciate ligament injury in adolescent female football players by 60%. Gioftsidou *et al.* (2012), designed balance training programmes that incorporated soccer skills with dynamic balance. Thirty eight professional soccer players were randomly assigned into 3 groups: Group A performed the balance training program with a frequency of 6 times per week, for 3 weeks, Group B performed the balance training program with a frequency of 3 times per week, for 6 weeks and Group C (control) did not follow a highly specific balance training program, but continued standard soccer training. Evaluation was performed with the use of an electronic stability system (indices-deviations) and a wooden balance board (time on balance) before (pre-test) and after the training period (post-test). The specific balance exercises demanded a combination of balance ability and certain soccer skills, such as kicking or headers. The results showed that both training groups improved their balance ability equally ($p<0.05$) despite the different frequency of the balance training program. It is thus suggested that that balance training program can be implemented in soccer players daily, or at least 3 times per week, to improve the players balance and proprioceptive ability, which in turn may reduce the potential for lower extremity injury. Research suggests that balance training programs that included static, dynamic and sport specific soccer skills can be effective in improving balance abilities for the lower limb in soccer players and that an increase in balance and proprioception in turn can lead to a decrease in injuries (Gioftsidou *et al.*, 2012; Myer *et al.*, 2011; Xia, Zhang, Wang, Sun, Fu, Myer, Ford, McLean, Hewett, Deutsch, Altchek, Veltri, Potter, Warren, Biomechanics, Adams, Bogduk, Krych, O'Malley, Johnson, Mohan, Hewett, Stuart, Dahm, Bates, Ford, Xu, Khoury, Myer, Ford, Paterno, Quatman, Manske, Prohaska, Lucas, McLean, Borotikar, Newcomer, Koppes & McLean, 2017). More importantly, there is an association between balance ability, proprioception, coordination, change of direction ability, agility and knee joint kinematics given that the knee joint is one of the major joints controlling such movements (Gioftsidou *et al.*, 2012).

The knee joint is one of the largest and most complex joints in the human body (McLean, Su, van den Bogert, den Bogert & McLean, 2003). The knee joint is a major load-bearing joint that consists of numerous structures including ligaments, menisci and the patella (Mclean, Foundation, Bogert, Su & Bogert, 2004; McLean, Walker, *et al.*, 2005). Muscles surrounding the knee have the ability to support the large external loads applied to the joint and reduce the potential for ligament loading, owing to their anatomical movement arms (Besier & Lloyd, 2003). Muscles also tend to contribute to joint stability (Alentorn-Geli *et al.*, 2009), implying therefore that muscular fatigue might be a risk factor for ligament injuries. Energy absorption is decreased in fatigued muscles whereby lowering the extent of the stretch (both active and passive) that may lead to injury (i.e. fatigue may lower the injury threshold) (Alentorn-Geli *et al.*, 2009). Improved neuromuscular control could assist soccer players to better absorb energy, leaving less energy to be absorbed by other structures such as ligaments. Alentorn-Geli *et al.* (2009) found that under fatigued conditions male and female athletes experienced lower knee flexion angles and increased proximal tibial anterior shear force as well as increased knee varus moments when performing stop-jump tasks. Nyland, Caborn, Shapiro and Johnson (1997) investigated the effects of hamstring fatigue on transverse plane knee control during a running crossover cut directional change (functional pivot shift) and the authors found that an eccentric work-induced hamstring fatigue created decreased dynamic transverse plane knee control and in turn showed increased knee internal rotation during impact-force absorption. Decreased knee internal rotation during hamstring fatigue was also indicated. Put more simply, fatigue may reduce the injury threshold and alter the knee kinematics; both factors interacting to ultimately increase the risk for injury.

Injury risk is dependent on the loading rate and the movement strategy employed during different game situations. According to McLean *et al.* (2003), the key to understanding the potential mechanisms of knee injury is to determine the joint loading characteristics associated with an injury-causing event. ACL injuries are often non-contact in nature (Wong & Hong, 2005) and occur during the landing or stance phase of ‘‘high-risk’’ sporting postures that incorporate sudden deceleration and/or rapid speed or directional changes, such as cutting or pivoting movements (McLean *et al.*, 2003; Paes, Fernandez, Hader, Mendez-Villanueva, Palazzi, Ahmaidi, Buchheit, Condello, Kernozek, Tessitore, Foster, Abt, Lovell, Page, Marrin, Brogden, Greig, Cortes, Quammen, Lucci, Greska, Onate, Paul, Bradley, Nassis, Nikolaidis, Dellal, Torres-Luque, Ingebrigtsen, Mohr, Krustrup, Bangsbo, McGovern, Dude, Munkley, Martin, Wallace, Feinn, Dione, Garbalosa, Strauss, Jacobs, Berg, Carling, Le Gall & Dupont, 2016). Some researchers have shown that the greater the number of cutting or change of direction (COD) manoeuvres an athlete engages in, the greater the fatigue experienced by that athlete (Besier & Lloyd, 2003; Paes *et al.*, 2016) . Greater insights into the effects of fatigue on the mechanics of turning and change of direction tasks are, therefore, warranted.

Fatigue is an intrinsic factor affecting the musculoskeletal and neurological systems and is associated with decreased knee proprioception and increased joint laxity compared to baseline values (Chappell, 2005). Altered biomechanical patterns and an associated decline in muscle performance and decision-making ability have been noted in soccer players post-fatigue (Cortes, Quammen, Lucci, Greska & Onate, 2012). The extent to which the biomechanical factors related to knee joint motion change in response to fatigue have been studied typically by the implementation of pre-fatiguing protocols (Cortes *et al.*, 2012) through the use of sophisticated laboratory-based 3D motion analysis systems. Unanticipated cutting tasks which do not allow for pre-planning of a movement have been reported to cause knee mechanics (i.e. increased knee internal rotation) which may increase the risk of anterior cruciate ligament injury and fatigue has also been shown to have similar effects. Soccer players often have to perform unanticipated tasks when they are fatigued (Collins, Almonroeder, Ebersole & O'Connor, 2016). One particular study collected three-dimensional knee joint kinematics and kinetics data from 13 female athletes while they performed a run-and-cut task, before and after completion of an fatigue protocol (intermittent shuttle run) and found that participants demonstrated a 68% increase in their peak knee abduction angles following completion of the fatigue protocol. There was also a 23% increase in peak knee abduction angles and a 33% increase in the peak internal knee adduction moments (Collins *et al.*, 2016).

The available evidence therefore suggests that cutting manoeuvres during soccer under fatigued conditions may be a major factor for increased injury risk, specifically to the lower extremity. Some questions remain unanswered however, such as how knee kinematics change during: (1) randomized cutting manoeuvres at specific angles at a controlled speed, (2) fatigued vs. non-fatigued trials under the same conditions proposed in (1), and (3) dominant vs. non-dominant cutting trials under conditions proposed in both (1) and (2).

1.3 Research question, aims, objectives

1.3.1 Research Question

The research question is posed as follows: “Do leg dominance and fatigue state affect the lower extremity kinematics during cutting manoeuvres in South African male soccer players?”

1.3.2 Research Aim and Objectives

The aim of the present research was to compare the effect of leg dominance and fatigue state on the lower extremity kinematics during cutting manoeuvres in South African male soccer players.

In order to achieve the aim of the research the primary objectives have been set as follows; to determine:

- the three-dimensional joint kinematics using 3D motion analysis (e.g. Vicon motion analysis system) to investigate the effects of leg dominance and fatigue on lower-limb joint rotations at initial contact and at peak stance-phase.
- the differences in stance phase duration related to leg dominance and neuromuscular fatigue as well as differences in lower limb joint rotations during controlled cutting manoeuvres.
- the ground reaction forces during the cutting manoeuvres to investigate whether differences existed in terms of leg dominance and neuromuscular fatigue trials.

1.4 Significance of the study

Fatigue is believed to be a contributing factor to injury in all sporting codes, which is especially true for soccer. Research has also been conducted to investigate the effects of specific training programs to prevent the effects of fatigue and subsequent injuries that may occur. However, very little research has investigated the effects of fatigue by including soccer skills during fatigue protocols and the lack of drills specific to the athletic event that incorporate multiple directions (e.g., cutting, side step, etc.), and deceleration and accelerations are yet to be applied. Furthermore, few studies have utilized functional agility protocols that can mimic skills commonly used by soccer players to assess lower extremity biomechanics during a fatigued status. In order to formulate effective training programs and preventative strategies to decrease the number of injuries caused by fatigue and the effects of fatigue and lower leg dominance on the kinematics of especially the knee joint was needed. Soccer-specific skills and change of direction movements were also incorporated to strengthen the findings. The research aims to strengthen the body of knowledge that already exists around the effects of fatigue on lower limb kinematics, but also aims to strengthen the notion that sport-specific skills should be added to give more, game-like, accurate results. The biomechanical analysis of human motion has broad applications for performance enhancement and injury prevention not only in soccer, but in almost all sporting codes. By investigating the effects of leg dominance and fatigue on lower extremity kinematics, possible injury mechanisms can be identified and injury prevention strategies can be developed to reduce lower limb injuries. It is envisaged that the findings of this study can be used to guide future studies on the effects of fatigue and leg dominance on lower limb kinematics especially in compromised population groups (i.e. setting normative ranges for prospective injury studies).

Chapter 2: LITERATURE REVIEW

2.1 Introduction

Although soccer is predominantly considered a semi-contact sport, there are a considerable number of annual injuries, with the rate, severity and location of such injuries varying substantially from country to country (Dvorak & Junge, 2000). A South African study (Bayne, Schwellnus, van Rensburg, Botha & Pillay, 2018) involving one professional team, in the Premier Soccer League (PSL), reported a much higher match injury incidence, 89 per 1000 playing hours compared to European data of 13.4 injuries per 1000 playing hours (Calligeris, Burgess & Lambert, 2015). The first finding of Bayne *et al.* (2018) was that the average training and match exposure times per soccer player during the season were 12 252 minutes (204 hours) and 1 834 minutes (31 hours), respectively. These findings can be compared to European data, which have shown that the average training and match exposure times per soccer player are much higher than the South African soccer player's times. European soccer players tend to average 15 720 minutes (262 hours) and 2 400 minutes (40 hours) training and playing matches, respectively (Calligeris *et al.*, 2015). Another study conducted on South African soccer players participating in the Coca Cola League found that in 23 matches played, a total of 15 injuries were sustained by 10 players. The incidence of injuries per 1 000 hours game time was 39.5 (Bailey, Erasmus, Lüttich, Theron & Joubert, 2009). The study also indicated that more injuries occurred at the beginning of the season rather than mid/end season (Bailey *et al.*, 2009). The finding of the former study suggests that player conditioning and running mechanics related to change of direction, agility, balance, coordination and lower extremity flexibility may need greater attention, specifically during pre-season conditioning as well as the beginning phases of the in-season conditioning program.

Most injuries tend to occur in the second half of a game, indicating that certain injuries may be exacerbated by fatigue (Gioftsidou, Ma lliou, Pafis, Beneka, Tsapralis, Sofokleous, Kouli & Godolias, 2012). To prevent such injuries, a two-pronged approach is typically proposed: (1) to incorporate specific exercise programs that integrate strengthening exercises to restore muscle imbalances, stretching exercises to increase muscle flexibility, and balance exercises to improve proprioception (Gioftsidou *et al.*, 2012); and (2) to assess mechanical factors during movements that may predispose players to injury, specifically cutting as well as change of direction manoeuvres during fatigued and non-fatigued states (Xia *et al.*, 2017).

The following chapters will delve deeper into soccer injuries in general and what joints they implicate. Subsequent sections will then focus on the mechanism of injuries to achieve a better understanding of the possible causes. Thereafter the neuromuscular system will be examined, followed by research related to how fatigue may influence the functionality of the neuromuscular system. Lastly, knee joint biomechanics and the working of a motion analysis system will be explored to identify the possible risk factors and causes of injury as well as determining the effect of leg dominance and fatigue on lower limb kinematics.

2.2 Soccer injuries

Mcgrath and Ozanne-smith (1997) described soccer as a vigorous, high intensity, intermittent ball and contact sport. The characteristics of soccer and the functional activities involved cause strain on the technical and physical skills of the individual players. The Harstad Injury Prevention Study, in Norway, found that soccer contributed to 44.8% of all sport injuries sustained, implying it has the highest population-based incidence of injury compared to other sports (Ytterstad, 1996). Similar findings have been reported in review articles (Inklaar, 1994; Naidoo, 2007) . These studies, however, based their findings on incidence of injuries in the total sport population, rather than taking into consideration the number of injuries in relation to the number of participants participating in each sport respectively.

From the elite/professional players representing their clubs and country to the amateur playing on a recreational level, all players are susceptible to injury (Inklaar, 1994). The nature of the game of soccer, in which players make sharp turns off a planted foot, and intense contact with the ball and other players, along with the essential underlying components of running and kicking, indicate the vulnerability of the lower extremities. Joints that are commonly affected are the ankle and knee joints, as well as the muscles and ligaments of the thigh and calf among others (Dvorak & Junge, 2000). The epidemiological soccer literature clearly indicates that the majority of soccer injuries occur to the lower extremities. Lower extremity injuries account for between 58% to 93% of all injuries (Naidoo, 2007). A literature review showed that 61% to 90% of all soccer-related injuries occurred in the lower extremities, of which the most common were contusions (34%), sprains (21%), and strains (15%) (Myer *et al.*, 2011). It is pertinent to note that injuries can be classified by rate/frequency (i.e. the number of total injuries per year) or by severity (i.e. time lost due to injury) (Inklaar, 1994). Hawkins *et al.* (2001), provided a breakdown of the location of injuries sustained during soccer competition and training and are as follows: thigh (23%), knee (17%), ankle (17%), lower leg (12%), groin (10%), neck/spine (6%), foot (5%), upper limb (3%), hip (2%), abdomen (1%), head (1%), chest (1%) and toe (1%). Although such findings indicate that the knee joint is not necessarily the most frequently injured joint, is often the most severe. Injury of the anterior cruciate ligament (ACL) is potentially the most traumatic sports-related knee injury and usually amounts to the most game-time lost (McLean *et al.*, 2003). Based on such classifications it can be noted that thigh contusions may be classified as the most common injury (23%), but that the most severe injuries relate to knee sprains (ACL [21%], meniscus [14%]) (Hewett et al. 2005).

Training/game time lost so injury is also a major concern for coaches. Literature shows to what extent time is lost due to injury and proves the effects it might have on a team. According to a study by Hawkins *et al.* (2001), on average 4.0 matches were missed per injury sustained, with 78% of the injuries leading to a minimum of one match missed. A mean of 24.2 days was lost per injury based on the incidence of injuries per month. The mean number of injuries per club per season was 39.1. These records confirm previous reports of the high risk of injury in professional football and the risk of injury having previously been identified as being greater than in many other team sports (Griffin, Agel, Albohm, Arendt, Dick, Garrett, Garrick, Hewett, Huston, Ireland, Johnson, Kibler, Lephart, Lewis, Lindenfeld, Mandelbaum, Marchak, Teitz & Wojtys, 2008; Hawkins *et al.*, 2001; McLean, Felin, Suedekum, Calabrese, Passerallo & Joy, 2007; Myer *et al.*, 2011).

Overuse injuries are also common amongst soccer players and are associated with running, jumping, pivoting, heading and kicking of the ball. With the most soccer injuries being traumatic in nature (Hewett, Myer, Ford, Heidt & AJ, 2005), the proportion of injuries being caused by overuse varying from 9% to 34% (Myer *et al.*, 2011).

Apart from the debilitating effects of traumatic injury to the knee joint, a musculoskeletal injury often increases the likelihood of significant long-term effects, including permanent knee instability, meniscal tears, cartilage injury and the development of osteoarthritis (McLean *et al.*, 2003). Additionally, post-ACL surgery evidence has revealed that less than 50% of patients will return to sports within one year, less than 65% will return within two years and approximately 11% will cease sports participation altogether (Bailey *et al.*, 2009). In light of these factors, there are two key aspects to consider: firstly, according to (McLean *et al.*, 2003), to understand the potential mechanisms of knee injury by determining the joint loading characteristics associated with an injury-causing event, and secondly, to either strengthen the associated musculoskeletal structures or to avoid the injury-causing event (the former being an external variable beyond the control of most individuals).

ACL injuries are often non-contact in nature (Wong & Hong, 2005) and occur during the landing or stance phase of 'high-risk' sporting postures that incorporate sudden deceleration and/or rapid speed or directional changes, such as cutting or pivoting movements (Condello, Kernozek, Tessitore & Foster, 2016; McGovern, Dude, Munkley, Martin, Wallace, Feinn, Dione & Garbalosa, 2015; McLean *et al.*, 2007, 2003). An increased number of cutting or change-of-direction (COD) manoeuvres experienced causes greater fatigue and in turn causes a decrease in speed which is associated with a higher post-test blood pH (blood lactate accumulation) (Besier & Lloyd, 2003; Gaitanos, 1990; McGovern *et al.*, 2015). The conclusion could then be made that an increase in blood lactate level is associated with an increase in the fatigue level experienced by an athlete and in turn causes a decrease in power output/speed. Buchheit, Bishop, Haidar, Nakamura and Ahmaidi (2010) investigated the physiological responses of shuttle repeated-sprint running and found higher values for $\dot{V}O_2$ and/or blood lactate accumulation during the shuttle protocol compared to linear strategies. He suggested that it could be due to the involvement of additional muscles during the 180° changes of directions, as upper-body (e.g., respiratory, back, abdominal and arms muscles) and bi-articulate leg (e.g., biceps femoris, rectus femoris, hip adductors, iliopsoas) muscles are more active during the deceleration, turning and acceleration phases, than during straight-line running.

It is believed that fatigue development during multiple sprint work is associated with a decreased power output (or speed) (Buchheit *et al.*, 2010). Such findings are relevant on the basis that neuromuscular and musculoskeletal fatigue tend to compromise the ability of the surrounding soft-tissue structures (i.e. muscles, tendons, ligaments) to withstand the high loading rates associated with changes of direction and cutting manoeuvres, thereby potentiating the likelihood for injury to those structures.

Discrepancies exist between the running speeds and turn frequencies of linear and shuttle running. Linear running typically exhibits higher average running speeds compared to shuttle running and would therefore result in higher $\dot{V}O_2$ values (i.e. higher physiological loading) due to the greater power output (speed) and contraction velocity requirements (Kramer, Watson, Du Randt & Pettitt, 2018). However, higher musculoskeletal loading is typically observed for shuttle running, thereby leading to differing fatigue mechanisms between linear- and shuttle-based running. Research has shown that $\dot{V}O_2$ requirements increase proportionally to running speeds and turning frequency during shuttle running although the average speed tends to be lower compared to linear running (Hatamoto, Yamada, Sagayama, Higaki, Kiyonaga & Tanaka, 2014). A contrasting view is offered by Buchheit *et al.* (2010), who have shown that no significant $\dot{V}O_2$ differences exist between linear and shuttle running, this could be attributed to the lower average speeds and increased energy requirements associated with the higher turning frequencies. Kramer *et al.* (2018) supports this concept and although significant differences are present these differences must be seen in context of the deceleration/acceleration phases of the shuttle running. According to Kramer *et al.* (2018) the higher amount of turning in addition to the all-out running may therefore cause a greater neuromuscular fatigue or higher muscle de-oxygenation compared to other shuttle- and field-based tests. Moreover, the energy cost of turning during shuttle running has been shown to be higher than that of linear running (Gaitanos, 1990). However, more complete insights into the effects of fatigue on the mechanics of turning and change-of-direction tasks are, therefore, warranted.

2.3 Mechanism of injury

Soccer injuries can be grouped into two categories namely (1) contact and (2) non-contact injuries. Soccer is considered a contact sport and players are often susceptible to direct blows to the body. These direct injuries could lead to a number of sub-injuries, such as contusion and disruption in blood vessels within the soft tissue leading to hematoma formation or bone fractures. Indirect injuries would result from forces generated within the musculoskeletal system during the activity. These types of injuries usually are encountered in the early and late stages of the game due to inflexibility, inadequate warm-up or fatigue. During a soccer match cutting, rotating, and pivoting movement occur often and it is during these types of motions that anatomically the leg, specifically the knee, can fall into a 'valgus' collapse (Loudon, Manske & Reiman, 2013). The closed-chain theory poses a possible reason for this, as significant knee valgus occurs when the thigh adducts and internaly rotates, while the knee (lower limb) moves into a position of abduction as the ankle and foot everts during weight-bearing motions (Loudon *et al.*, 2013). This position is possibly caused by faulty neuromuscular activation (Hewett *et al.*, 2005). The whole kinetic chain (trunk, hip, knee, ankle foot) is affected by the neuromuscular control of the lower limb, which is typically challenged during running motions that involve rapid directional changes.

Rahnama, Reilly and Lees (2002) identified 16 soccer-related playing activities as possible mechanisms to injury: dribbling the ball, goal catch, goal punch, throw, heading the ball, jumping to head, kicking the ball, making a tackle, making a charge, passing the ball, receiving a ball, receiving a tackle, receiving a charge, shot on goal, set kick, and throw in the ball. The researchers found an association between the injury incidence and playing action/mechanism. Injury incidence was found to be higher in receiving a tackle, making a tackle, and receiving a charge (Rahnama *et* *al.*, 2002). Goga and Gongal (2003) found that the most important factor contributing to severe injuries in South African soccer players was aggressive tackles. Hawkins *et al.* (2001) reported on soccer injury mechanisms, whereby 38% were classified as resulting from contact with another player or the ball and 58% related to a non-contact mechanism. In addition, running (19%), shooting (4%), turning (8%), and jumping (2%) accounted for 33% of all non-contact injuries. Contact injury mechanisms have previously been highlighted as an important area for consideration if the incidence of injury in professional soccer is going to be reduced (Griffin *et al.*, 2008). Similar findings were made by Twizere (2004), reporting 27.9% of injuries among Rwandan soccer players were as a result of contact with another player and that the mechanism responsible for injury during match play to be players colliding with another player (15.4%).

Sports injuries could be the result of both intrinsic and extrinsic factors. Inklaar (1994) identified intrinsic risk factors to be age, weight, gender and strength of the athlete as well as joint flexibility including pathological ligamentous laxity and muscle tightness, functional instability, previous injuries and inadequate rehabilitation. The researcher also indicated extrinsic risk factors to include the exercise load in soccer (matches and training), inadequate equipment (shin guards, taping, and shoes), playing field conditions and foul play (Hackney, 1994). Mcgrath and Ozanne-smith (1997) also include pre-season conditioning as an influencing factor for soccer injuries.

A few intrinsic and extrinsic factors are of higher importance according to Mcgrath and Ozanne-smith (1997) and will be elaborated on further, starting with the intrinsic factors. Muscle weakness and asymmetry of the concentric muscle and eccentric muscle ratio is an important factor which could lead to muscle failure due to intense load put on the muscle (Day & Duguet, 2003). Specific strengthening exercises and monitoring of this strength ratio is of great importance to prevent injuries, therefore attention should be given both to eccentric as well as concentric contractions in training; where concentric training tends to predominate (Gaitanos, 1990). Ekstrand and Gillquist (1982) indicate that joint stability, muscle tightness, inadequate rehabilitation and lack of training contribute to 42% of all injuries observed. They implemented a flexibility-training program over a complete season, and found a decrease of injury incidence, thus highlighting the importance of flexibility training. It is important to note that flexibility does not just refer to the length-tension relationship of muscles per se, but also that of the other soft tissue structures that are inherent to a given joint; specifically, ligamentous structures. Given that ligamentous structures function both as static and dynamic restraints, excessive joint rotations may predispose such structures to premature injury (i.e. when compared to 'normal' length-tension dynamics). Investigating the joint rotations that are characteristic of uninjured athletic populations during both non-fatigued and fatigued conditions may therefore guide future research related to compromised population groups. Stated differently, knowing the normal joint ranges during specific conditions may ease the identification of abnormal joint ranges by having pre-set criteria for what denotes 'normal'.

Secondly, exploring extrinsic factors, Rahnama *et al.* (2002) identified exercise and match load/intensity as an extrinsic factor responsible for high injury risk. Their results showed that injury incidence was highest in the first and last 15 minutes of a match, reflecting the intense engagement in the opening periods and possibly the effect of fatigue in the closing periods of a match (Rahnama *et al.*, 2002). Ekstrand and Gillquist (1982) suggested that twice as many injuries occur in soccer matches as in practice; this supports the idea that a higher intensity of play causes a higher injuries incidence compared to training. The skill level of a player is also listed as an intrinsic risk factor, but studies have found contradicting results. Inklaar (1994) assessed the correlation between the skill level of a player with the incidence of injury and found that lower skilled players had a higher injury incidence, but according to Goga and Gongal (2003) higher skilled players were more susceptible to injury.

Perhaps more importantly, the susceptibility of an athlete to injury seems to be dependent on several modifiable and/or non-modifiable risk factors, specifically related to ACL injuries (Cortes *et al.*, 2012), suggest that the combination of central (cognitive) and peripheral (muscle/nerve-inervation) fatigue may play a role in the biomechanical cutting patterns that tend to change post fatigue. Some of the *modifiable* factors include (Alentorn-Geli *et al.*, 2009):

- decreased hamstring to quadriceps strength ratio and recruitment,

- muscular fatigue resulting in altered neuromuscular control,
- decreased 'core' strength and proprioception,
- low trunk-, hip-, and knee flexion angles, as well as high dorsiflexion of the ankle when performing sport tasks,
- lateral trunk displacement and hip adduction combined with increased knee abduction movements (dynamic knee valgus), and
- increased hip internal rotation and tibial external rotation with or without foot pronation.

The *non-modifiable* risk factors, on the other hand, include:

- increased knee joint laxity,
- gender,
- type of soccer shoe worn by players (studded shoes),
- the condition of the playing surface (Wong & Hong, 2005).

Olsen, Myklebust, Engebretsen and Bahr (2004) observed in a detailed video analysis of ACL injuries, that 95% of the plant and cut injuries occurred while the athlete was moving in a lateral direction and attempting to change direction medially. They also found that single leg landing is another common ACL injury mechanism, with all the injuries occurring on the same leg used to take-off (Olsen *et al.*, 2004).

McLean, Walker, *et al.* (2005) compared hip, knee and ankle kinematics during a pre- and post-fatigue side-step manoeuvre. Their results indicated the peak rotation deviation for the hip joint; hip flexion 49.5°, hip abduction 24.3° and 19.3°. The present study show similar joint kinematic results with hip flexion to be, 59.91°, hip abduction 23.33° and hip internal rotation to be 14.83. The knee joint measurements of McLean, Walker, *et al.* (2005) also differ slightly. Knee flexion was 19.3°, knee adduction 3.8° and knee internal rotation was 19.0°. The results show that knee flexion peaks at 46.28°, knee abduction 19.05° and knee internal rotation 16.40°. A literature review conducted by Myer *et al.* (2009), indicated that a number of studies were in general agreement that injuries occurred in cutting (change of direction) or landing situations. During investigations, particularly of the knee joint, as well as the mechanism of injury, the knee was reported to be relatively straight at the point of injury and all the studies agreed that knee valgus occurred frequently. Olsen *et al.* (2004) found that the amount of internal/external knee rotation was 10° or less in 90% of the cases which implies that even slight knee internal/external rotation does contribute to knee injuries, particularly ACL injuries. Olsen *et al.* (2004) also stated that valgus loading in combination with external or internal knee rotation cause excessive ACL loading and in turn could cause a ligament rupture. Schreiber, Illingworth, Mejia and Fu (2012) added that a vigorous eccentric quadriceps contraction could also be a main cause of injury. Other likely components to the injury mechanism include anterior translation, dynamic valgus of the lower extremity with the joint near extension, low flexion probably due to increased quadriceps tention, most or all of the force on a single leg or foot with the foot displaced away from the body's centre of mass and increased trunk motion as experienced during a change of direction movement. There is a need for normative data for knee- and hip-joint rotation to help identify potential players with higher injury risk.

Neuromuscular control also plays an important role in the mechanism of injury and the lack thereof may have significant effects on injury incidence in soccer players. Neuromuscular control refers to unconscious activation of the dynamic restraints surrounding a joint in response to sensory stimuli (Griffin *et al.*, 2008). Proprioception is the sensory source best suited for providing the information necessary for mediating neuromuscular control, thereby enhancing functional joint stability. Sources of proprioceptive information include mechanoreceptors located in muscular, articular, and cutaneous tissues that are responsible for transducing mechanical events into neural signals (Griffin *et al.*, 2008). According to Chappell (2005), dynamic stabilization via the neuromuscular control system helps to protect the knee joint during dynamic, sport related tasks. Neuromuscular training programs have successfully been applied to reduce the rate of injuries, especially those associated with the anterior cruciate ligaments (ACL) in soccer players. Waldén *et al.* (2012) applied a cluster-randomized control design to a total of 4 564 young soccer players (230 teams) and intervened with a 15-minute neuromuscular warm-up focused on the development of core stability, balance and jump-landing with correct knee alignment, Waldén *et* *al.* (2012) found that a neuromuscular warm-up program significantly reduced the rate of anterior cruciate ligament injury in adolescent female football players. Gioftsidou *et al.* (2012), designed balance training programs that incorporated soccer skills with dynamic balance. The specific balance exercises demanded the combination of balance ability and certain soccer skills, such as kicking or headers. Gioftsidou *et al.* (2012) found that balance training programs that included static, dynamic and sport-specific soccer skills were effective in improving balance abilities for lower limbs in soccer players and that an increase in balance and proprioception in turn can lead to a decrease in injuries.

One aspect of agility that has not been adequately addressed in the literature is the influence of limb dominance on performance in agility tasks (Smykalski, 2016). Limb dominance is defined as an imbalance of muscle recruitment and muscular strength of the dominant limb over the non-dominant limb (Ross, Guskiewicz, Prentice, Schneider & Yu, 2004). The dominant limb is therefore typically capable of greater dynamic control (Hewett, Stroupe, Nance & Noyes, 2004). Even before a change of direction occurs during an agility task, a participant's running technique might contribute to the ability or inability to perform sprints with directional changes. It is important to note that agility performance is multi- faceted, requiring coordination, balance and flexibility of the lower extremity. The interplay of these factors (i.e. coordination, balance and flexibility) may be dependent on the type of agility task given that the extent of the directional change required would influence the magnitude of coordination, balance and flexibility needed for successful completion.

Brown, Zifchock and Hillstrom (2014) investigated the three-dimensional kinematic and kinetic differences between dominant and non-dominant leg while performing a maximal effort sprint of 25 m. There were no significant differences observed between the dominant and non-dominant limb in running gait patterns (Brown *et al.*, 2014). The lack of significant differences between limbs during various gait patterns is not surprising as this would indicate a degree of limb symmetry which is expected during gait tasks. However, whether such symmetry still holds true at different speeds and cutting tasks is not entirely clear. Velotta, Weyer, Ramirez, Winstead and Bahamonde

(2011) investigated the relationship between limb dominance and the type of agility task performed. When the task was manipulative in nature (i.e., when kicking a soccer ball), participants would use the dominant leg, but when the task involved stabilizing tasks, such as standing on one foot, participants would prefer to use their left or non-dominant leg (Velotta *et al.*, 2011). Although the dominant leg demonstrates earlier muscular activation during dynamic activities, it is possible that it may over compensate for the non-dominant leg and therefore not able to withstand the greater knee forces during dynamic activities and be at a greater risk for injury, Ford, Myer and Hewett (1997). In addition Ross *et al.* (2004) sought to compare biomechanical factors between the kicking and stance limbs and found differences between contralateral limbs in relation to isokinetic strength at the hip, knee, and foot; closed kinetic chain proprioception during a balance task as well as kinetic and kinematic differences during a jump-down manoeuvre. Finally, Ross *et al.* (2004) indicated that the kicking limb had superior thigh strength, better proprioception, and greater knee flexion range of motion over a longer period of time than did the stance limb. Brophy, Silvers, Gonzales and Mandelbaum (2010), found that non-dominant limb peak knee extension torque was greater than the dominant side and attributed it to the role of the non-dominant quadriceps supporting the swing of the kicking leg.

Coaches and players should therefore focus specifically on the modifiable risk factors to minimise the potential for injury. Since frequent changes of direction, pivoting and cutting are critical aspects of soccer, a focus should be placed on these mechanisms and the joint mechanics associated with such actions, and how such actions may be aggravated by fatigue, specifically around the knee joint.

2.4 Mechanisms of fatigue

For more than a century, scientists have attempted to understand the mechanisms responsible for the development of exercise-induced fatigue (Edwards, 1978; Hill, Long & Lupton, 1924; Hirasawa, Sledge & Woo, 2011). The majority of early research indicated that muscular, i.e. peripheral, fatigue may result from either metabolic limitations or the formation of 'fatigue bi-products'. In the early 1920's it was believed that 'fatigue is a chemical process' (Pierce, Enoka, Gandevia, McComas, Stuart & Thomas, 2013), and described this phenomenon as 'oxygen want' (Hill *et al.*, 1924). However, scientists also acknowledged the importance of central factors in the development of fatigue, Coelho (2015) showed that both central and peripheral fatigue contributed to the 'fatigue point'. Over the past century, our understanding of fatigue has significantly improved, however the precise mechanisms responsible for task failure/exhaustion (i.e. the point at which an individual voluntarily terminates exercise) remain unclear. Recent research examining exercise-induced fatigue has focused on alterations occurring within the neuromuscular system (Amann & Dempsey, 2008; Millet & Lepers, 2004; Perrey, Racinais, Saimouaa & Girard, 2010; Place, Lepers, Deley & Millet, 2004). Neuromuscular fatigue can be defined as an inability to maintain a given force or power output and has been found to have both central and peripheral origins (Allen, Lamb & Westerblad, 2008; Amann & Dempsey, 2008; Gandevia, 2001; Taylor & Gandevia, 2007). Central fatigue is defined as a progressive exercise-induced reduction in voluntary activation of muscle and occurs within the central nervous system (CNS) at cortical and spinal levels (Gandevia, 2001; Taylor & Gandevia, 2007). Peripheral fatigue, by contrast, occurs at or distal to the neuromuscular junction and is described as any decline in muscle performance (Allen *et al.*, 2008; Gandevia, 2001). To date, our ability to understand the development and relative contributions of central and peripheral fatigue mechanisms has been somewhat limited by the methodological designs used in research. Exercise-induced fatigue has primarily been investigated at task failure. Such methodology does not adequately capture the time course of fatigue development and inherently assumes that fatigue development is relatively constant across the entire exercise duration. However, a relatively linear increase in neuromuscular fatigue is unlikely to occur during high-intensity exercise and it is therefore important to examine the temporal changes in central and peripheral fatigue mechanisms during a repetitive, high-intensity exercise bout. Decorte, Lafaix, Millet, Wuyam and Verges (2012) have recently assessed the time course of central and peripheral fatigue development during intermittent bouts of constant load exercise. This is the only study that has shown that reduced efficacy of excitation-contraction (E-C) coupling was “compensated for” by an increase in central motor drive, as indicated by an increase in the electromyogram (EMG) amplitude (Decorte *et al.*, 2012). Although this study provided some insight into the temporal pattern of neuromuscular fatigue development and the magnitude of change in fatigue mechanisms, it is still unclear whether these changes are comparable during continuous exercise bouts. As neuromuscular fatigue develops, associated changes in joint rigidity, the relative activity of muscles and the timing of muscle activity are observed during dynamic multi-joint exercise (Gandevia, 2001). Such changes in muscle activation likely enforce changes in joint kinematics and increase the variability of movement patterns, which is commonly referred to as motor variability (Srinivasan & Mathiassen, 2012). Movement pattern changes have historically been considered as noise in the neuromuscular system and thus were viewed as a negative outcome of fatigue and therefore detrimental for exercise performance (Bartlett, Wheat & Robins, 2007). However, recent data have indicated that variations in movement patterns may in fact be a beneficial adaptation within the neuromuscular system that may assist in delaying further fatigue development, avoiding injury and maintaining task performance (Bartlett *et al.*, 2007; Srinivasan & Mathiassen, 2012). These adjustments occur presumably to increase the activation of less fatigued muscles or muscle groups and maintain task performance. To date few studies have documented kinematic alterations throughout a soccer match and the associated changes in lower limb muscle recruitment (EMG) strategies (Dingwell, Joubert, Diefenthaeler & Trinity, 1912). In addition, there are limited data describing the association between the site (i.e., mechanism) of neuromuscular fatigue and changes in muscle activation and/or joint kinematics, and thus alterations in movement variability during a cutting task (Decorte *et al.*, 2012). Therefore, the effects of modulating the level of movement variability within the system on skill execution, and thus running performance, are not known. Further work is needed to identify muscle activation and kinematic strategies utilised during fatiguing exercise, and to determine whether allowing changes in joint kinematics, and thus encouraging kinematic variability, is beneficial to performance. It has been well established that an individual's physiological profile largely dictates their exercise performance/capacity (Coelho, 2015). $\dot{V}O_{2max}$, metabolic thresholds (i.e. lactate or ventilation thresholds), maximal lactate steady-state (i.e. the highest exercise intensity at which blood lactate concentration is stable) and efficiency or economy have all been found to be significantly correlated with performance in sporting codes (Noakes, St. Clair Gibson & Lambert, 2004). In addition, such characteristics are also likely to be associated with both the timing and magnitude of muscle activation and kinematic changes during fatigue. However, few studies have examined the relationship between the physiological characteristics of soccer players and changes in biomechanics (i.e. kinetics and/or kinematics). It is also plausible that specific physiological characteristics may be related to an athlete's ability to resist fatigue and thus to maintain limb kinematics towards the end of a high intensity exercise bout.

Muscles that are used at a high intensity show a progressive decline in performance, but recover again after a period of rest; this reversible phenomenon is known as muscle fatigue (Zemková & Hamar, 2009). The phenomenon has been recognized for many years, but only recently have studies of the mechanisms been investigated (Porter & Whelan, 2009).First the discovery was made that exhausted muscles contained 'lactate'. Hill *et al.* (1924) suggested that during events lasting one to several minutes muscular fatigue was caused by lactate accumulation in the muscles. They proposed that during intense exercise an inadequate oxygen supply caused muscles to release energy anaerobically (without oxygen) and in doing so, they produced lactate which lowered their pH and caused them to fatigue (Hill *et al.*, 1924). This model was widely accepted and continues today, with only minor modifications, as the classic explanation for muscular fatigue in most athletic activities. However, doubt has been cast on this premise with the report in 1995 that high levels of lactate did not interfere with muscular contraction at near-normal body temperatures (Pate, Bhimani, Franks-Skiba & Cooke, 1995). This finding has led to a re-examination of the role of lactate in muscular fatigue. However, some early pieces of research that supported this contention suffered from a flaw in design, they were conducted with excised muscle fibres that were cooled below body temperature. More recent studies have demonstrated little or no effect of lactate accumulation and acidosis on muscle contraction force and velocity when the fibres were warmed near body temperature (Bangsbo, Iaia & Krustrup, 2007; Bangsbo, Madsen, Kiens & Richter, 1996). This finding has caused many in the scientific community to doubt acidosis as the cause of muscular fatigue. Some are now suggesting that fatigue results from creatine phosphate (CP) depletion and, specifically, the increases of inorganic phosphate and ADP that occur when the muscle supply of CP is reduced (Westerblad, Allen & Lännergren, 2002).

It was also suggested that failure of excitation-contraction (EC) coupling contributed to muscle fatigue by showing that a fatigued muscle could recover much of its force when perfused with caffeine, known to directly facilitate release of Ca^{2+} from the sarcoplasmic reticulum (SR) (Gaitanos, 1990). Thereafter an influential study showed that fast fibres fatigued extremely quickly, whereas slow fibres were essentially un-fatigable (Allen *et al.*, 2008). All these factors contribute to the mechanisms of fatigue a soccer player experiences during a soccer match (Allen *et al.*, 2008). This will further investigate the specific mechanism of fatigue and their effect on soccer players.

Fatigue can be defined as a decrease in force production, or an inability to regenerate the original force during the presence of an increased perception of effort (Strauss, Jacobs & Berg, 2012). Strauss *et al.* (2012) stated that fatigue either has a 'peripheral' or 'central' origin. Peripheral fatigue occurs when there is a decrease in the force generation capacity of the skeletal muscle because of action potential failure, or excitation-contraction coupling failure, or impairment of cross-bridge cycling, in the presence of unchanged or increasing neural drive (Strauss *et al.*, 2012). The peripheral model of fatigue in exercise generally concludes that fatigue during maximal intensity exercise is caused by substrate depletion or metabolite accumulation from energy utilization in the peripheral skeletal musculature (Lambert *et al.*, 2005). Evidence has shown that high-speed, high-intensity activities such as cutting, change of direction or jumping would more likely be limited by metabolite accumulation rather than substrate depletion (Allen, Lamb & Westerblad, 2008). Prolonged activities, such as continuous running, would be regulated more by energy supply factors such as glycogen availability (Allen *et al.*, 2008). Since soccer is dependent on both aspects and given that the intensities and durations of various activities fluctuate throughout the game, it might explain why soccer players need to develop fatigue resistance across the range of bio-energetic systems. Although Strauss *et al.* (2012) concluded that fatigue during submaximal exercise is caused by substrate energy compound depletion or excitation/contraction coupling failure, none of these metabolic changes were shown to directly cause fatigue during dynamic physical activity (St Clair Gibson, Lambert & Noakes, 2001).

Central fatigue occurs when there is a reduction in neural drive or motor command to the muscle resulting in a decline in force production or tension development (Strauss *et al.*, 2012). When destabilizing forces are anticipated, the central nervous system (CNS) is capable of adjusting muscle activation patterns to oppose these forces (Besier & Lloyd, 2003). This could explain the notion that central fatigue delays postural adjustments and cause insufficient planning of activation patterns. Cortes *et al.* (2012) explained that the combination of central (cognitive) and peripheral fatigue (muscles) may play a role in the biomechanical cutting pattern change post-fatigue. Fatigue studies have reported decreases in muscle strength, impaired joint position sense and delayed neuromuscular responses as being contributing factors to higher injury incidences during physically demanding tasks like the tasks executed during a soccer match (Nicholas, Nuttall, Williams, Sarro, Silvatti, Aliverti, Barros, Noakes, St Clair Gibson, Lambert, Ostenberg, Roos, Rozzi, Lephart, Gear & Fu, 2000; Nyland *et al.*, 1997). The purpose of the present study was to achieve neuromuscular fatigue in the lower extremities of the soccer players in order to evaluate the effects of fatigue on the joint kinematics. Even though EMG technology was not used to validate fatigue, a study by Mclean, Borotikar, Newcomer, Koppes and Mclean (2008) implemented a similar fatigue protocol and validated that fatigue was reached using EMG.

The physical demands in soccer have been studied extensively, specifically the metabolic changes during a game and their relation to the development of fatigue (Bangsbo *et al.*, 2007). Soccer players regularly transition between brief bouts of high-intensity running and longer periods of low-intensity running during a soccer match. In addition to these activities, players frequently perform tackling, jumping and change of direction movements alongside complex technical skills

(Paul, Bradley & Nassis, 2015). According to heart-rate and body-temperature measurements elite soccer players have an average oxygen uptake of around 70% of maximum $\dot{V}O_{2max}$ during a match (Paul *et al.*, 2015). The rate of creatine-phosphate (CP) uptake and glycolysis during a soccer match is high and supported by findings of reduced muscle CP levels and increases in blood and muscle lactate levels, in addition muscle glycogen is decreased by 40% to 90% during a game and is probably the most important substrate for energy production (Bangsbo *et al.*, 2007). This phenomenon might be the cause of fatigue toward the end of a game. Elevation of blood glucose and a reduction of insulin levels are observed during a game and increased fat-oxidation is seen; probably because lowered muscle glycogen (Paul *et al.*, 2015). Thus, elite soccer players have high aerobic requirements throughout a game and extensive anaerobic demand during periods of a match leading to major metabolic changes, which might contribute to the observed development of fatigue during and toward the end of a game. Exactly which bio-energetic pathway would be most responsible for fatigue-induced changes that culminate in injury would be difficult to ascertain, given any of the three systems might be culpable. Furthermore, it is important not to neglect the role of the CNS (as previously mentioned) in that an injury may occur due to a lapse in concentration or an incorrect motor sequence command sent by the CNS to the motor unit. Much more research in this regard is presently still unfolding.

Given that soccer is arguably a submaximal sport with players likely to be working within their physical capacity it is very difficult to objectively identify fatigue and the resulting biomechanical factors associated with injury. The utilization of a functional agility fatigue protocol that can mimic skills commonly used by soccer players is needed to assess lower extremity biomechanics during a fatigued state (McLean *et al.*, 2007). Some researchers have incorporated soccer-based skills in the fatigue protocol to investigate whether fatigue would cause any difference to the skills assessed, but indicated that there was a lack of soccer-specific skills that incorporated multi-directional movements (e.g., cut-right, up-and-down, etc.), and that decelerations and accelerations were yet to be applied in fatiguing protocols (Chappell, 2005; McLean *et al.*, 2007). Hence, combining exercises with various muscular recruitment patterns, a deceleration and acceleration phase that can incorporate the entire lower extremity musculature that is normally involved during tasks utilized during a soccer match (e.g., sidestep, stop-jump) is of importance when investigating the effects of fatigue on lower limb mechanics (McLean *et al.*, 2007).

Cortes *et al.* (2012) performed a functional agility short term fatigue protocol that took approximately six minutes to complete. This short-time frame to induce fatigue had the same effects than other protocols of longer durations during which decreased knee and hip flexion, and increased knee internal rotation were observed (Page, Marrin, Brogden & Greig, 2016). Cortes *et al.* (2012) identified two points: (a) the neuromuscular adaptations were dependent on the intensity of the exercise and duration of the activity, therefore athletes may be injured earlier in the game/practice if played at a high level of intensity during a short period and (b) training programs should focus on developing strategies to accommodate the biomechanical changes during fatiguing activities (Cortes *et al.*, 2012). An alternative explanation arises that fatigue protocols are not stimulating the demands that are observed during an actual soccer match (Borotikar, Newcomer, Koppes & McLean, 2008; Chappell, 2005; Mclean *et al.*, 2005). Potentially, in order to develop a fatigue protocol that mimics the conditions of an actual game, it would be necessary to quantify the fatigue that athletes are experiencing during a soccer game or practice.

2.5 Knee joint biomechanics

The knee joint is the largest and arguably most complex joint in the human body (McLean, Su, & van den Bogert, 2003). The knee joint consists of numerous structures, including ligaments, menisci and the patella and is a major load-bearing joint (see Figure 1) (McLean *et al.*, 2003). Muscles surrounding the knee have the ability to support the large external loads applied to the joint and reduce the potential for ligament loading, owing to their anatomical moment arms (Besier & Lloyd, 2003). Given the anatomical structure of the knee, the predominant motion of the joint occurs on the sagittal plane, although in a number of sports activities, the knee is subjected to complex three-dimensional (3D) loading patterns which, if not correctly trained and progressively overloaded, could increase the susceptibility for injury to the joint and joint structures (McLean, Foundation & von den Bogert, 2004). Injury risk is dependent however on the loading rate and the movement strategy employed during any number of plays during a match situation.

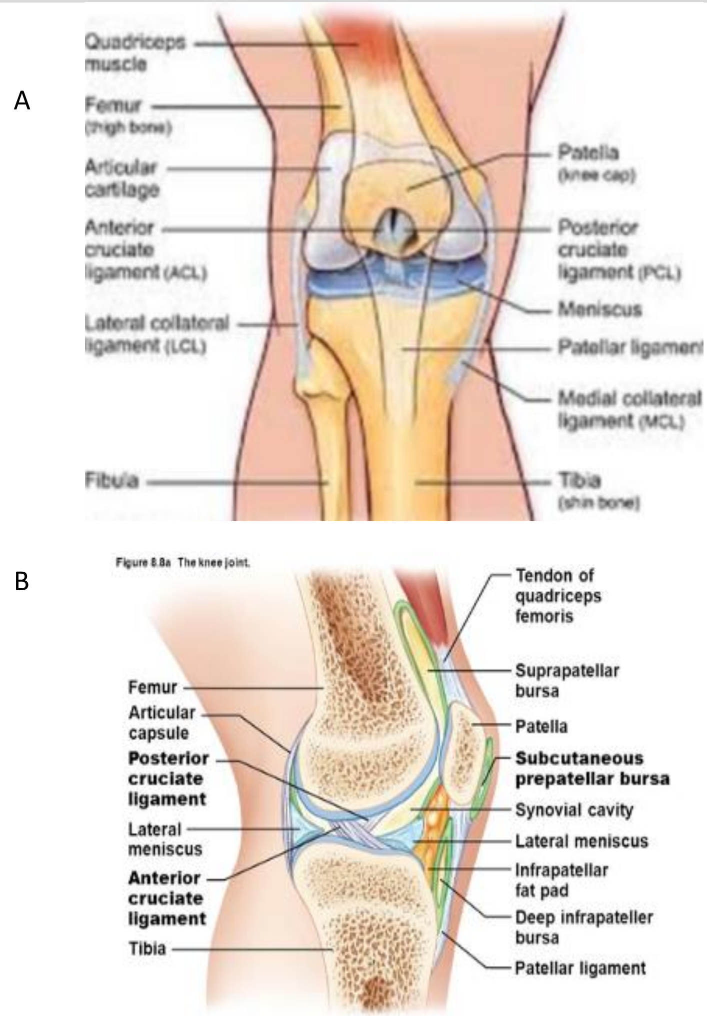

Figure 1: Anatomy of the knee joint. (A) Frontal and (B) sagital section through the right knee joint

The CNS may adopt a number of different neural strategies to counter the external loads during dynamic cutting tasks. However, two generalized strategies have been suggested by Besier and Lloyd (2003): the first involves 'selected activation' of muscles with moment arms that are best able to counter the external load. This would include increased activation of gracillis to counter an external valgus moment, or co-activation of the medial hamstrings and medial quadriceps to oppose external valgus loading. The second strategy suggested ‘generalized co-contraction’ which involves co-activation of hamstring and quadriceps muscles without any selectivity of specific muscles (Besier & Lloyd, 2003).

During cross-cutting tasks, quadriceps fatigue resulted in increased ankle dorsiflexion moments, decreased peak posterior breaking forces, decreased peak extension moments, and delayed peak knee flexion angles (Chappell, 2005). Hamstring fatigue resulted in decreased peak impact knee flexion moment, increased internal tibial rotation, and decreased peak ankle dorsiflexion (Nyland *et al.*, 1997). According to Chappell (2005), dynamic stabilization via the neuromuscular control system helps to protect the knee joint during dynamic sport-related tasks. Muscle antagonist–agonist relationships are vital for joint stabilization, thus muscle actions must be coordinated and co-activated in order to protect the knee joint (Chappell, 2005).

A free body diagram of forces acting on the knee joint (Figure 2) (Hewett *et al.*, 2005). More specifically, Figure 2. A shows the forces acting on the tibia in the sagittal plane equilibrium position between articular contact force, hamstrings force, quadriceps force, and ACL force. In this example, quadriceps force contributes to ACL force, whereas hamstrings and articular contact force protect the ACL. Figure 2. B Illustrates the forces acting on the tibia, showing the frontal plane equilibrium between external dynamic valgus load, articular contact force, quadriceps force, medial hamstrings force, and ACL force.

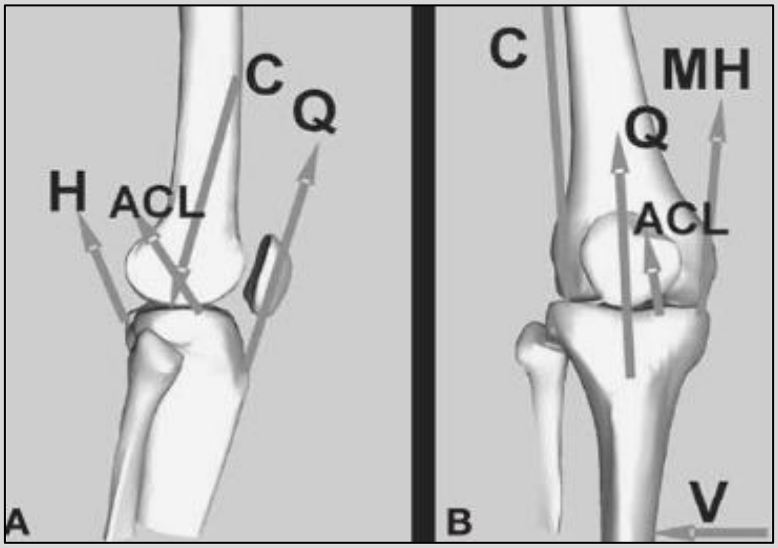

Figure 2: Forces acting on the knee joint. Panel A represents the sagittal plane; panel B represents the frontal plane (H = hamstring force vector; ACL = anterior cruciate ligament force vector; C = articular contact force vector; Q = quadriceps force vector; V = valgus force vector; MH = medial hamstring force vector)

Under external dynamic valgus loading, contact shifts to the lateral compartment. The moment balance with respect to the contact point shows that both quadriceps and medial hamstrings force help the ACL (and medial collateral ligament, not shown) stabilize the joint against dynamic valgus loading. Under a given dynamic valgus load (i.e. knee extension), any reduction in these muscle forces and any reduction in these muscle forces will cause an increase in ligament loading.

The knee and hip joint can move in all 3 plains and the normative ranges are important to note when evaluating lower limb kinematics. The range of motion of the knee and hip joint is illustrated by Figure 3 (PeerWell, 2013). The normative ranges are as follows; knee flexion 110°-130°, knee extension 30° and hip abduction 20°-30°, hip internal rotation 40°, hip external rotation 45°. It is important to link the knee joint mechanics to that of the hip and ankle since a change/no change in the knee joint kinematics might be compensated for by other joint e.g. ankle and/or hip. It has been postulated, for example, that if an athlete is not properly aligned or if an unusual foot placement at landing occurs, he or she may be at increased risk for injury (Hewett *et al.*, 2005). Rotation of the knee will result in adduction or abduction of the hip (coxal joint). If there is abduction (knee externally rotated) there will be more force required from gluteus medius, gluteus minimus, tensor fascia latae, gluteus maximus, psoas major, iliacus, and Sartorius. While if there is adduction (knee internally rotated) more force from pectineus, adductor brevis, adductor, longus, adductor magnus, and gracilis will be needed to stabilize the joint (Konrardy, 2017). Furthermore, landing in dynamic valgus could be proposed as potentially increasing the risk of knee injuries. Athletes are encouraged to avoid excessive valgus alignment at landing, cutting, or decelerating to minimize their risk of ankle/knee/hip injuries (Hewett *et al.*, 2005).

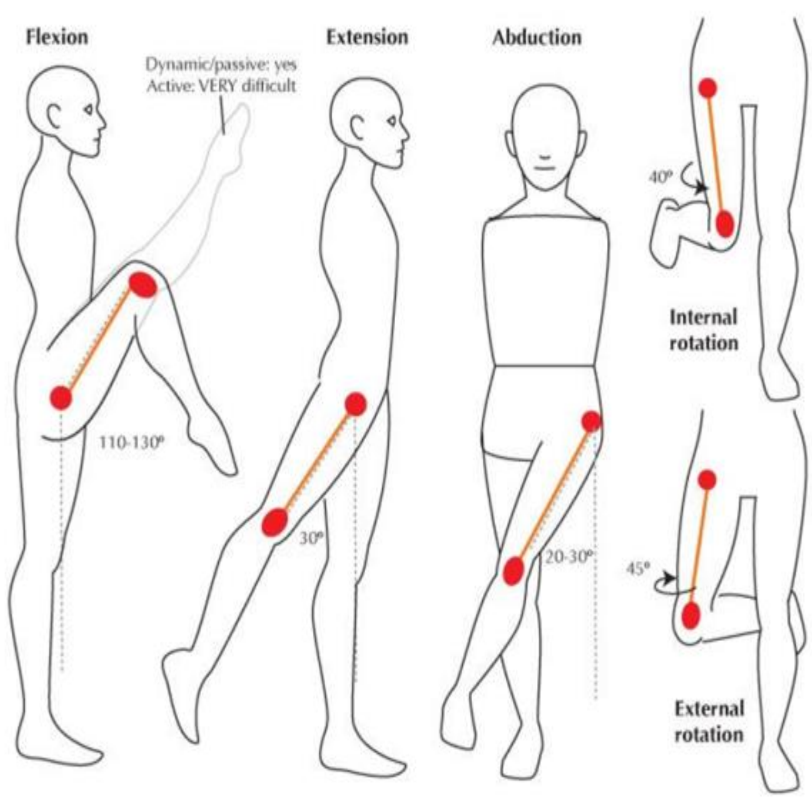

Figure 3: Dynamic knee and hip range of motion

Patla, Prentice, Robinson and Neufeld (1991), noted that frontal plane trunk orientation (thorax with respect to the global coordinate system), lateral trunk orientation and stride width decrease during cutting tasks. The decrease in step width was attributed to a decreased lateral foot placement of the planted foot. The above mentioned changes in step width, if not compensated for by alterations in trunk position may alter knee moments and potentially influencing ACL injury risk (Patla *et al.*, 1991). Additionally, peak knee valgus moment has been associated with initial contact hip flexion and internal rotation during sidestep movements and hip neuromuscular control is proven to influence knee valgus motions as well as loads experienced during single-leg squatting (Zeller, Mccrory, Kibler & Uhl, 2003). Ireland (1999), described a common mechanism by which non-contact ACL injuries occur in the female athlete and termed it as the position of no return. This is when a loss of control at the hip and pelvis occur in conjunction with internal rotation of the femur, valgus knee angulation and external tibial rotation on a pronated, externally rotated foot. Zeller *et al.* (2003) pose the question whether the loss of hip control demonstrated by increased hip adduction, flexion, and external rotation could be the cause of a decrease in knee varus/valgus control. Hip adduction normally causes the femur to internally rotate which results in the knee being forced into a degree of valgus (Ireland, 1999). Investigations into the risk factors involving the ankle and foot that may cause a predisposition for ACL injuries has been conducted and have indicated that factors such as hyper-pronation may increase the risk of injury. Although the current study does not specifically investigate or answer the question of whether the hip is an initiator of ACL injury, literature does suggest an association. Consistent with other biomechanical data, there is known to be a relationship between the coupling mechanical interrelationships of the hip and knee with muscle activation (Ireland, 1999). Considering the data supporting the importance of the hip muscle activation in maximizing knee and ankle muscle activation, the finding that dynamic ankle/foot pronation is created by proximal motion and position and that the interactive movements controlling distal body segments are generated by proximal segment activations. We do not believe the ankle motion differences are the primary causative factors but, in fact, accompany the more proximal changes (Dayakidis & Boudolos, 2006; Houck, Duncan & Haven, 2006).

The key in neuromuscular research involving knee joint mechanics therefore is to accurately assess changes in joint moments and ranges of motion (ROM) of the hip, knee and ankle joint during various dynamic activities. To do so, the use of optoelectronic motion capturing systems, amongst many other alternatives, are often used in biomechanics research.

Performance of motion capturing systems strongly depends on their setup and is highly sensitive against alterations. Marker properties, optical projections, video-digital conversion, camera configuration, lens distortion, calibration procedure, etc. influence the performance to various extents (Blake & Ferguson, in press). Producers state only rough estimates regarding accuracy and precision of their products.

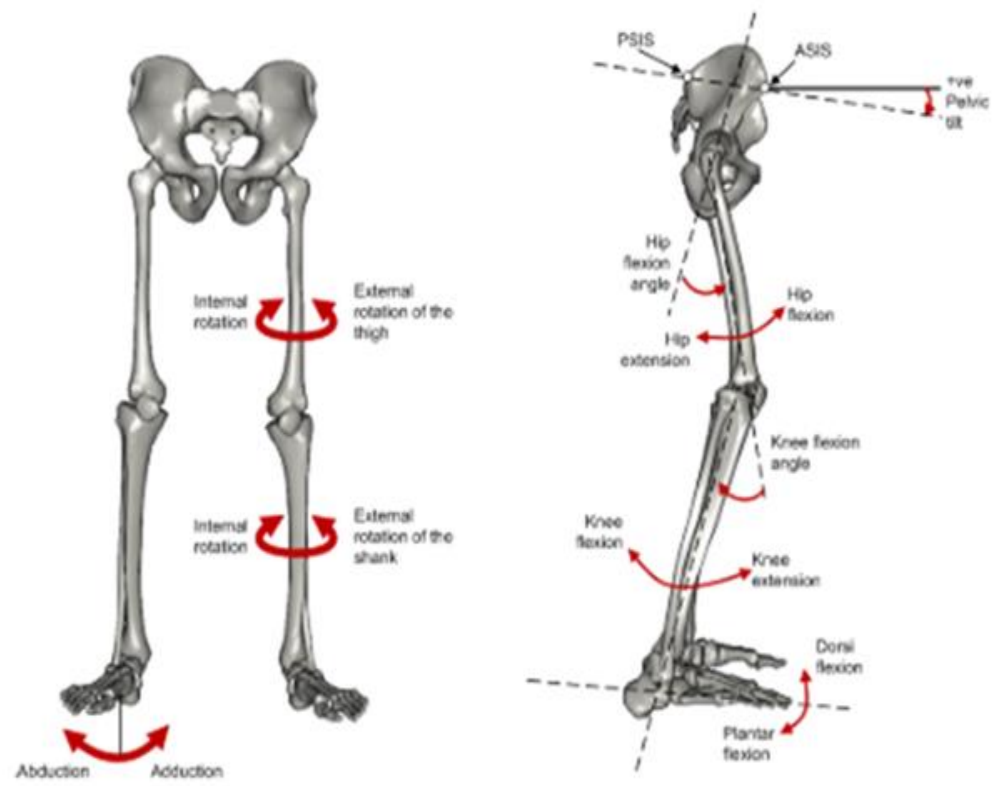

Figure 4: Joint rotation angles of the hip, knee and ankle

The kinetic model that is used by the motion analyses system is illustrated by Figure 4 (Hewett *et al.*, 2005). The pelvis (body) motion was described with respect to the global (lab) coordinate system via three translational and three rotational degrees of freedom (DoF). The hip, knee and ankle joints were defined locally and assigned three, three and two rotational DoF, respectively.

Motion analysis enables detection of subtle joint kinetic and kinematic changes that may influence the final clinical outcome such as in the case of ACL injuries, where an increases in pathological anterior tibial translation and internal tibial rotation increase the risk for injury and can be detected using systems such as the *Vicon-Nexus* (McLean, Huang, Bogert, Rigoldi, Galli, Cimolin, Camerota, Celletti, Tenore & Albertini, 2005).

As with any technology available there are potential limitations of 3D video analysis techniques are excessive financial resources, expertise and space. There is no argument about the ever-increasing use of optoelectronic technological systems in sports settings which must be matched by empirical investigation into its relevance and accuracy (Windolf, Götzen & Morlock, 2008).

2.6 Conclusion

Studies examining the effects of fatigue and leg dominance on lower limb kinematics were sourced to provide a theoretical framework from which to establish a suitable testing and fatigue protocol for the present study. Should this study indicate that fatigue or leg dominance does not have significant effects on lower leg kinematics during a cutting manoeuvre it would lead the way for further research on the major cause of lower limb injuries.

Chapter 3: RESEARCH METHODS AND PROCEDURES

3. 1 Introduction

The purpose of the present research is to compare dominant vs. non-dominant leg kinematics as well as the effect of fatigue during a cutting manoeuvre in South African male soccer players. The primary objective of the present study is to investigate the sagittal kinematics of the dominant vs. non-dominant leg during a 60° cutting manoeuvre and how these kinematics change during a fatigued state. A 3D opto-electric motion analysis system (*Vicon*) was used to determine the kinematic data of the lower limbs.

The research design, participant recruitment, sampling methods, measuring instruments, testing protocol, data collection and data analysis of the study are identified and explained in the following section to give an overview of the testing procedures and focus areas. The reasoning behind the research design, sampling method, and measuring instruments are also further elaborated on to provide a clearer overview as to why these variables were included.

3.2. Research design

The present study was explorative, quantitative and experimental in nature. A 'within-participants post-test only design' was used, which is also known as a 'repeated measures design' because all participants were 'repeatedly' measured under each experimental condition (i.e. use of their dominant vs. non-dominant leg; cutting in a fatigued vs. non-fatigued state) (Christensen, Johnson & Turner, 2014). By counterbalancing the order of the testing any carry-over effect was minimized. We randomized the direction in which athletes had to cut (which controlled for leg dominance) in addition to randomizing which leg would initiate the fatigue protocol to achieve this. The strength of this particular design is as follows: age, gender, and prior experience remain the same over the entire experiment and therefore participants serve as their own control. All participants were in all conditions and conditions cannot differ from one group/person to the other because the participants are compared to themselves. This makes it a powerful technique of control (Christensen *et al.*, 2014). Moreover, the sensitivity of the experiment is increased because each participant serve as their own control and are maximally sensitive to the effects of the independent variables if counterbalanced (Christensen *et al.*, 2014). Another advantage according to Christensen *et al.* (2014) is that the within-participant design needs typically requires a smaller sample because the number of participants needed for an entire experiment is equal to the number of participants needed for one experimental treatment condition. Christensen *et al.* (2014) state that this design also has a few disadvantages, namely, (1) the design can be taxing on participants as they have to participate in multiple treatment conditions, and (2) a sequencing effect can occur if not controlled for by a counterbalancing technique. It is important to note however, that counterbalancing only controls for linear sequencing effects; if the sequencing affects differential carryover effects, the confounding carryover sequencing effect will remain (Christensen *et al.*, 2014).

3.3 Participants

The population of interest for the proposed study pertains to South African male university soccer players based in Port Elizabeth. The soccer players utilized for the present study were selected from the local university (i.e. Nelson Mandela University in the Eastern Cape) who participated at Varsity Cup level. A total of 18 players (n = 18) met the inclusion criteria and were subsequently evaluated. However, due to data inconsistencies during data processing some data files could not be retained for further analysis; hence the final number of players that were included for analysis was 13 (n = 13). Participants had to be healthy individuals, with no current lower limb injuries. Anthropometric measures, such as height and weight as well as age, were captured before the start of testing.

The justification for the focus on soccer is two-fold: (a) ACL injuries commonly occur during rapid directional changes which are inherent to soccer, therefore more detailed biomechanical research is required; and (b) fatigue has been shown to be a predetermining factor for ACL injury and altered joint kinematics, therefore comparing pre-fatigued to post-fatigue sagittal joint kinematics would be important for this research, especially for soccer players.

3.4 Sampling methods

Sampling is the process of selecting a group of people, behaviour, or events to conduct a study on. A portion that best represents the whole population can be achieved through various sampling methods (De Vos, Strydom, Foche & Delport, 2011).

Due to the non-randomization of the quasi-experimental design, non-probability sampling, which is a type of sampling whereby subjects are selected non-randomly based on specific criteria, was utilized to sample the population group for the proposed study (Landreneau, 2010). Purposive sampling, a specific form of non-probability sampling, was used to acquire participants for the proposed study. Purposive sampling is also known as judgmental sampling. This type of sampling is based on the judgment of the researcher, as the population subset (i.e. South African male soccer players), is selected based on a specific characteristic or attributes relevant to the aim of the study (Delport & De Vos, 2011). The sample was selected based on the ease of access, cost effectiveness and convenience. The sampling criteria required participants to have played for the Nelson Mandela University Soccer Club for at least one year, at a competitive level (i.e. Varsity Cup). The sampling age range of the participants was based on the criteria set for players for Varsity Cup games; therefore, participants had to be of a university going age and fall within the age range of 18 to 25 years.

3.5 Inclusion and exclusion criteria

3.5.1 Inclusion criteria

The following criteria had to be met by participants to be included in the study:

- A male soccer player representing the Nelson Mandela University soccer club at a competitive level for at least one year.
- Be between the ages of 18 to 25 years (based on the selection criteria for inclusion in the varsity cup team).
- Players had to be free on injury for at least 6-12 months prior to testing (depending on the severity of any previous injury).
- Players were only allowed to participate if they have signed a voluntary informed consent form (Appendix C).

3.5.2 Exclusion criteria

Under the following specified circumstances, prospective participants were not permitted to take part in the proposed study:

- Soccer players with severe injuries that would limit movement.
- Players who were injured during the testing period.
- Soccer players who have not played at a university level for at least a year.
- A player that is not between the ages of 18 and 25 years.

3.6 Measuring instruments

Accurate results or the ability to capture the variables necessary for observation, are made possible by using the best-suited equipment and technology for that purpose (Delport & De Vos, 2011). The following measuring tools and equipment are valid and reliable in testing the respective variables for which they are intended:

- Vicon Motion Analysis System (Nexus 2.6, Oxford, UK) to measure whole body kinematics in a 3D space.
- Ten near infrared cameras (MX T20), and two high-speed digital cameras (Bonitas, DX3).
- Calibration wand.
- Kistler force plates (Kistler, Model 9290CD, Germany).
- Digital weighing scale (Jadever JWI-586 Weighing Indicator, UK) to measure weight to the nearest 0.01kg.
- Stadiometer (Holtain, Crymych UK) to measure height to the nearest 0.01m.
- Electronic timing gates (SmartSpeed Pro V1, Fusion Sport, USA).

3.7 Testing procedures

3.7.1 Capture volume set-up and calibration

Prior to testing the measuring instruments (Vicon system, force plates and weighing scale) were set up and calibrated. The calibration of the Vicon system consisted of making sure all the cameras were in working order, setting the volume and calibrating the force plates. The capture volume was set up according to a precise configuration such that both the speed and angle of the cutting manoeuvre could be recorded (see Figure 5). The capture volume incorporated ten MX T20-S near-infrared cameras (Vicon, Oxford, UK) sampling at a rate of 120 Hz as well as the use of two DVI Bonita 720c HD cameras (Vicon, Oxford, UK) sampling at a rate of 120 Hz. Every time the athlete entered the capture volume in preparation for one of the cutting manoeuvres, the system would capture the relevant kinetic and kinematic data. Six cameras were set up at approximately every 60°, mounted on to a railing approximately three meters off the ground. The remaining four cameras were set at ground positions on tripods that would maximise marker counts to facilitate data acquisition and processing (refer to Figure 6). Thereafter setting up of the running path and timing gates were performed as illustrated by Figure 5. Partici[ants were required to perform the cutting manoeuvre at a 60º angle relative to the line of travel. The angle was measured with a goniometer relative to the central axis of the imbedded force plate. The starting gates were placed three meters from the central axis of the force plate, with three finishing gates being placed on the left- and right-hand side as well as at neutral (i.e. without a directional deviation).

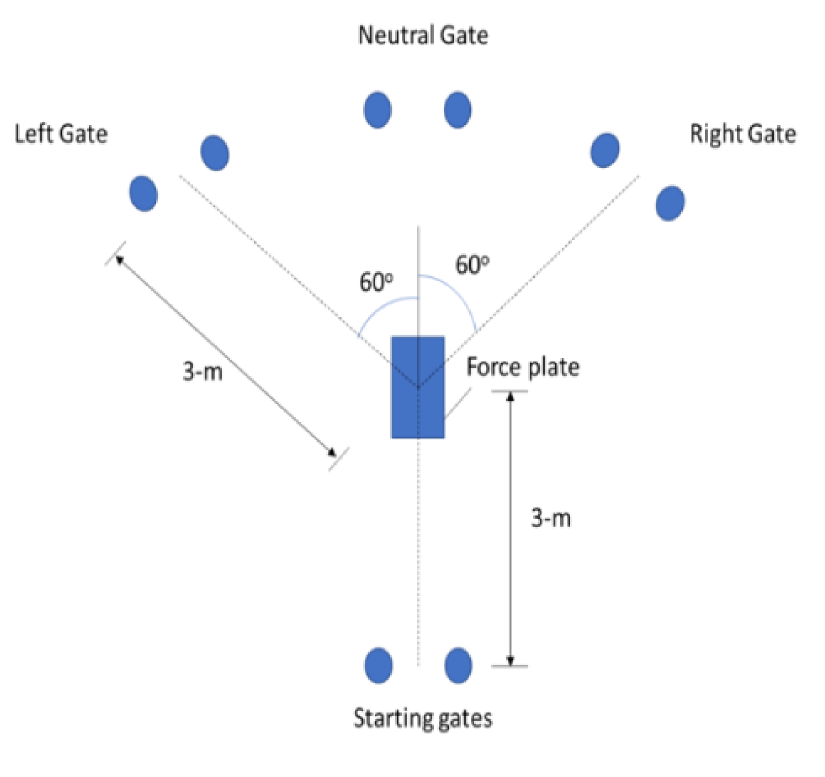

Figure 5: Experimental set-up for cutting manoeuvres

3.7.2 Arrival of participant, consent form, anthropometric measures, marker placement and warm-up

Upon arrival, consent forms were issued to the participants for signing, if they agreed to do so. Subsequently the testing procedures were explained verbally to the participants in detail. The participants were given sufficient time to ask questions and clarify information. The relevant anthropometric measurements required by the Nexus 2.6 software were taken in an adjacent laboratory prior to the testing and system calibration. These measures are clearly defined in Appendix C. A total of 43 retro-reflective markers and specific joint markers were placed on the participants to allow for data capturing by the Vicon motion analysis system (Figure 6). Before the commencement of testing, participants were allotted time to warm-up, consisting of cycling for five minutes on a stationary cycle ergometer and self-directed dynamic stretching.

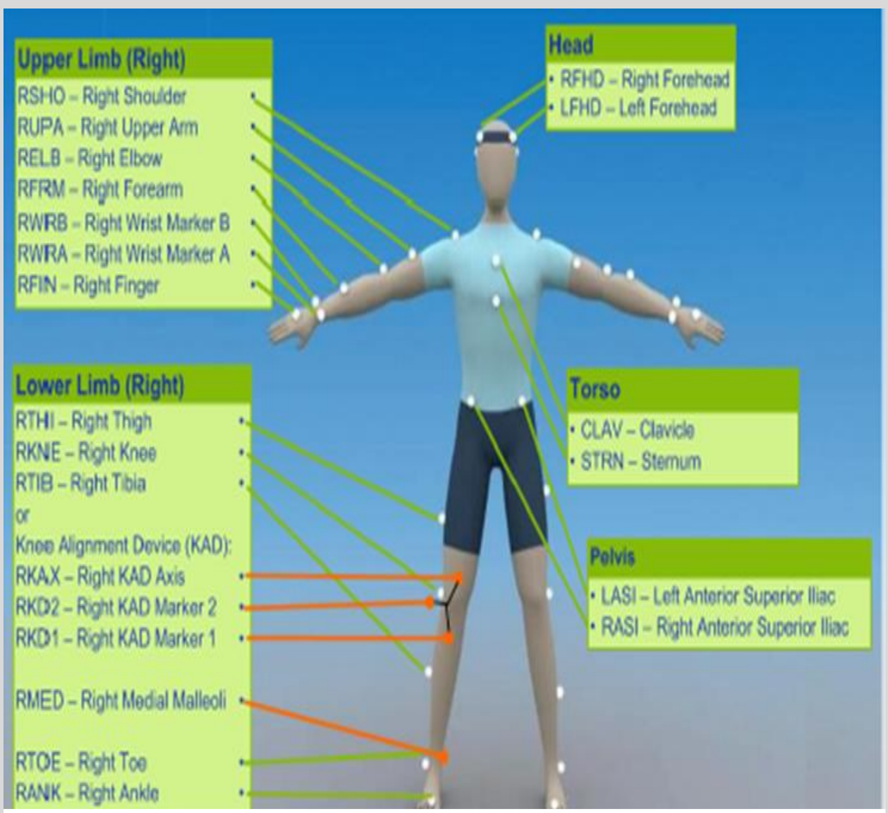

Figure 6: Marker placement

3.7.3 Trials

Participants had three-dimensional knee joint kinematic and ground reaction force data recorded during the execution of a 60° sub-maximal cutting manoeuvre (Figure 7), prior to and immediately after being exposed to a protocol designed to induce maximal neuromuscular fatigue of the muscles associated with the knee joint, with a specific focus on the knee extensor muscle group (Mclean *et al.*, 2008).

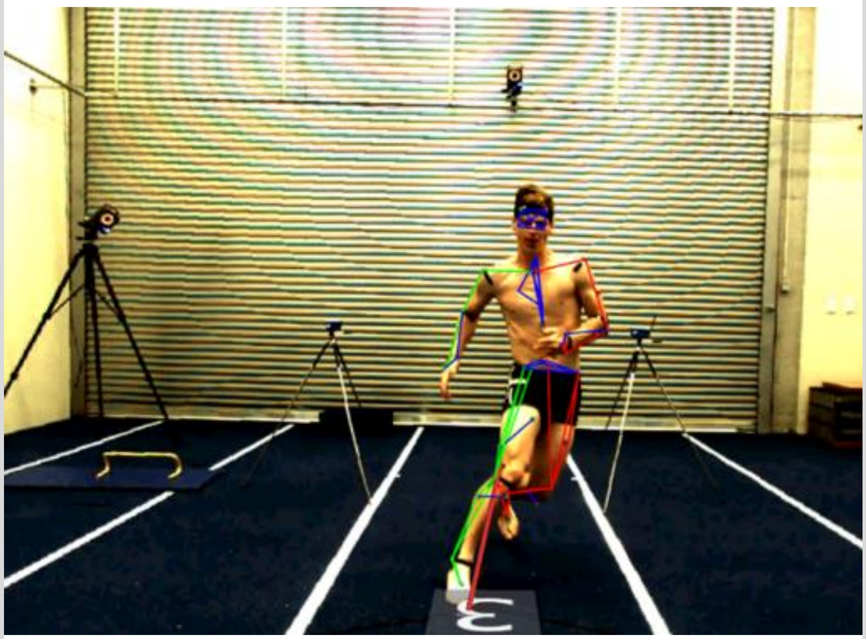

Figure 7: Cutting trail

The fatigue protocol consisted of continuous sets of eight stationary single-leg Bulgarian split squats followed by eight single-leg jumping Bulgarian split squats (Mclean *et al.*, 2008). Intra- and inter-participant variations in the squat tasks were minimized by having participants squat at a consistent frequency (1 Hz; measured by an electronic metronome [Pro Metronome, EUMLab, Xanin Tech GmBH, China]) and by ensuring that the position of the thighs was parallel to the ground at the completion of each squat. To ensure that a minimum of 90^o of knee flexion was obtained during the squatting tasks, a goniometer was used to measure the requisite angle as a baseline measure to which all future squats could be compared. Simultaneously over-laying of information was done on a portable computer screen to 'live-stream' the data, thereby allowing visually comparable live trials to the baseline measure, thus more objectively measuring the required squatting depth during the fatigue trials. Participants were instructed to continue to alternate between squat sequences and jump landing tasks until maximal fatigue was attained, as judged by at least two qualified Biokineticists. For the present study fatigue was defined as the point where participants: (1) could no longer complete at least five squats , (2) could no longer meet the 90^o criteria and/or (3) could not sustain the 1 Hz pace set by the metronome. Even thought muscle electromyography (EMG) or blood lactate levels was not measured we are confident that muscle fatigue was reached due to the task failure observed. To control for running speed, we set a tolerable range such that athletes would be completing the run at between 10-12 $km \cdot hr^{-1}$ (i.e. jogging speed), such that the time between gates should range approximately 1.80-2.16 seconds. Only trials that fell within this range were retained for analysis since running speed has been shown to affect both kinetic and kinematic data (Vicon, 2016).

3.8 Biomechanical analysis

Lower limb joint rotations were quantified for each trial based on the 3D coordinates of 43 precisely attached reflective skin markers (Mclean *et al.*, 2008) (See Appendix C). Markers were secured to pre-determined anatomical landmarks through the use of double-sided, adhesive tape. All attachment sites were shaved before the tape was applied and attachment over areas of large muscle mass were avoided to minimize excessive marker movement during ground contact. Additional tape was used to secure the markers to minimize the potential for marker movement or loss, particularly when participants began to sweat as maximal neuromuscular fatigue was reached. Following the placement of the markers, a static calibration trial was captured with the participant standing in a stationary (neutral) position (McLean *et al.*, 2007). A kinematic model was then defined, based on the data, consisting of nine skeletal segments (foot, talus, tibia and femer of each limb, and the pelvis). Hip, knee and ankle joint centres were defined in accordance with the Plug-in Gait model (Mclean *et al.*, 2008). Note that throughout this section, axis are described as follows; transverse axis (side to side or horizontal axis) relative to some defined "forward" direction) which pass from one side of the body to the other; sagittal axis is the line around which rotations in the frontal plane occur; and frontal axis is perpendicular to the ground and divides the body into dorsal (posterior or back) and ventral (anterior or front) portions. The 3D marker trajectories recorded during each trial were processed by the Vicon motion analysis system (Nexus 2.6) to solve for the 3D lower limb joint rotations in each time frame. The camera set-up is illustrated in Figure 8. Joint rotations were expressed relative to each participant's standing (neutral) position. This data was time normalized to 100% of stance phase during the cutting manoeuvre, being resampled at 1% increments (N = 101), with heel-strike and toe-off defined as the instant when the vertical GRF first exceeded or fell below 10N respectively (Mclean *et al.*, 2008). A more detailed explanation of how the joint kinematics is determined will follow.

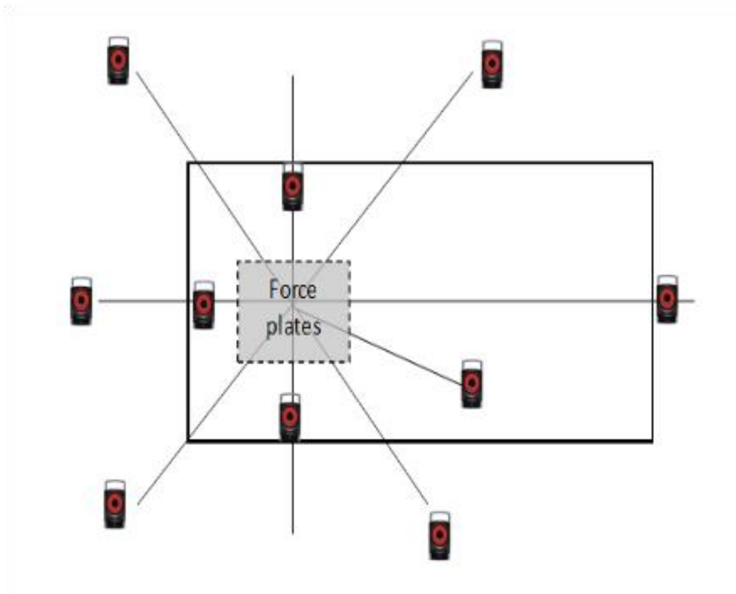

Figure 8: Vicon camera set-up in relation to the force plate

Systems such as the *Vicon-Nexus* make it possible to obtain 3D kinematic motion data and a rough explanation of the collection process is as follows. Forty-three reflective markers of 25 mm in diameter were secured with strapping tape to pre-defined anatomical locations. Attachment sites were first shaved if necessary and attachment over areas of large muscle mass were avoided in order to reduce excessive marker movement. The capture volume incorporated ten MX T20-S near-infrared cameras (Vicon, Oxford, UK) sampling at a rate of 120 Hz as well as the use of two DVI Bonita 720c HD cameras (Vicon, Oxford, UK) sampling at a rate of 120 Hz. Every time the athlete entered the capture volume in preparation for one of the cutting manoeuvres, the system would capture the relevant kinetic and kinematic data. Six cameras were set up at approximately every 60°, mounted on to a railing approximately three metres off the ground. The remaining four cameras were set at ground positions on tripods that would maximise marker counts to facilitate data acquisition and processing. Vicon recommends an ''umbrella'' camera configuration with uniform distribution and vertical grading around the observation centre, but our study had to compromise this ideal arrangement. Calibration of the volume was performed and according to Windolf *et al*. (2008) manual wand motion is uncontrolled and standardization of the calibration is preferred compared to the manual method by hand due to the more consistent calibration path. However, he also stated that a standardized (robotically performed) calibration method will optimize the wand pattern, the relevance for the end-user is low (Windolf *et al.*, 2008). Furthermore, investigation into different calibration volumes were performed by Windolf *et al.* (2008) to quantify the impact when capturing data in an 'uncalibrated' regions. A considerable drop of accuracy indicates spatial distortion in border areas and suggests to carefully including all relevant regions when calibrating by hand. Next a static trial was recorded via the high-speed video system with the participants standing in the neutral position as demonstrated by Mclean et al. (2004) and thereafter a dynamic trial was recorded. From the standing trial, a kinematic model (Figure 4) was generated by defining five skeletal segments (foot, talus, tibia and femer of the support limb, and the pelvis).

The following section lists all the kinematic variables calculated by the Vicon system's 'Plug-in Gait' feature. It includes information describing each variable in terms of ordered rotations or goniometric definitions. Note that absolute angles are measured relative to laboratory axes with the sagittal and transverse axes automatically selected according to the direction of walking. In Plug-in Gait, the laboratory axis closest to the subject's direction of progression is labelled the laboratory sagittal axis (Vicon, 2017).

Hip flexion/extension: hip flexion was calculated around an axis parallel to the pelvic transverse axis and passes through the hip joint centre. The sagittal femer axis is perpendicular to the plane of the hip flexion axis. Hip flexion was then the angle between the sagittal thigh axis and the sagittal pelvic axis. A positive (flexion) angle value was when the knee is in front of the body (Vicon, 2017). Hip adduction/abduction: hip adduction was measured in the plane of the hip flexion axis and the knee joint centre. The angle was measured between the long axis of the thigh and the frontal axis of the pelvis projected into this plane. A positive value was when the leg adducts (moves inwardly)

(Vicon, 2017). Hip rotation: hip rotation was measured at the long axis of the thigh segment and is calculated between the sagittal axis of the thigh and the sagittal axis of the pelvis projected into the plane perpendicular to the long axis of the thigh. An internally rotated thigh was assigned a positive hip rotation value (Vicon, 2017).

Knee flexion/extension: the sagittal shank axis is projected into the plane perpendicular to the knee flexion axis. Knee flexion is the angle in that plane between this projection and the sagittal thigh axis. Knee flexion corresponds with a positive angle (Vicon, 2017). Knee valgus/varus: this was measured in the plane of the knee flexion axis and the ankle centre and is the angle between the long axis of the shank and the long axis of the thigh projected into this plane. Based on this coordinate system, knee varus occurs when a positive number was seen at the distal aspect of the femer (outward bend of the knee) (Vicon, 2017). Knee rotation: knee rotation was measured at the long axis of the tibia. It was measured as the angle between the sagittal axis of the shank and the sagittal axis of the thigh, projected into a plane perpendicular to the long axis of the tibia. Internal tibial rotation was assigned a positive angle. If a tibial torsion value was present it was subtracted from the calculated knee rotation value, therefore a positive tibial torsion value provided a constant external offset to knee rotation (Vicon, 2017).

Ankle plantar/dorsi-flexion: the foot vector was projected into the foot sagittal plane. The angle between the foot vector and the sagittal axis of the tibia is labelled as foot dorsi/plantar-flexion. Dorsiflexion is assigned a positive number (Vicon, 2017). Foot rotation: this is measured at an axis perpendicular to the foot vector and the ankle flexion axis. Foot rotation is the angle between the foot vector and the sagittal axis of the shank, projected into the foot transverse plane. Internal rotation corresponds to a positive number (pronation) (Vicon, 2017).

3.9 Calibration

The accuracy of any measurement system or instrument is highly dependent on the accuracy of the set-up and the accuracy with which the data can be recorded and captured. All cameras, as well as the capture volume, were calibrated in two stages: firstly, a calibration wand was used whereby the examiner moves it around the capture volume; secondly the calibration wand was placed at the origin of the capture volume to create the reference frame for data capturing (Vicon, 2016). The goal of the first stage is two-fold: (1) to maximize the 'wand count' which relates to the number of times each of the five imbedded LED's (light emitting diodes) are seen by each camera (the target for good calibration was set to 1500 for each camera), and (2) to orientate each camera with respect to all other cameras. The latter goal provided the purpose of the second stage, which was to orientate all cameras relative to the capture volume. The force plate recording speed was set to 1200 Hz to coincide with the scaled sampling speed of the camera data (i.e. by a scaled factor of 10).

3.10 Statistical analysis

Measures of central tendency such as mean, median, standard deviation and standard error were used to describe the data as well as to assess the normality of the data. Descriptive statistics is also provided in the form of tables in the case of anthropometric data, and graphs such as mean plots with 95% confidence intervals (Gravetter & Wallnau, 2016). Inferential statistical analyses were conducted using a two-way analysis of variance (ANOVA), with repeated measures. A two-way ANOVA was used to determine whether there was an interaction between two independent variables (fatigue and leg dominance) on the dependent variables, such as knee flexion and extension angles, during a given task (Daniel & Cross, 2013). If any main effects were found, a Tukey's post-hoc test was conducted to determine which of the dependent variables were found to have a statistically significant effect. Practical significance was assessed using Cohen's d, and was interpreted as follows: trivial, < 0.2 = trivial effect; 0.2-0.5 = small effect; 0.5-0.8 = medium effect; < 0.8 = large effect (Cohen, 1988).

The latest versions of Statistica and OriginPro were used for statistical analyses. A qualified statistician based at the Nelson Mandela University, was consulted to ensure valid and accurate interpretation of the data.

3.11 Ethical considerations

In the current study, the following ethical issues were considered: informed consent, violation of privacy, competence of the researcher, harm or injury to participants and dissemination of results (Delport & De Vos, 2011). Anonymity of participants was maintained by the use of randomly assigned numbers to represent each participant. All participants were required to complete an informed consent form to clearly indicate that consent has been given by the participants for the upcoming testing, including comprehensive content in order to avoid dishonesty or confusion. In addition to this, the contact details of the researcher were provided to the participants to allow for open communication between the researcher and the participants should any queries or issues arise after the contact session. The study details were submitted to the Department of Human Movement Science at the Nelson Mandela University for approval. Upon acceptance, the proposal was forwarded to the Faculty of Health Sciences Postgraduate Studies Committee and Rec-H Committee of the Nelson Mandela University for final approval and ethical clearance. (Ethical clearance reference number: H17-HEA-HMS-006).

3.12 Limitations

Purposive sampling was used to identify and obtain participants from a specific soccer team (Nelson Mandela University), chosen geographical setting and a small number of participants who met the inclusion criteria were targeted. Data obtained from this small sample size is thus not representative of the larger population of male soccer players and, thus, no definitive findings could be generated; trends could however be observed.

3.13 Summary

This chapter focused on the methods and procedures followed in order to address the objectives set out in chapter one. The target population consisted of 13 South African male soccer players, between the ages of 18 and 25 from the Nelson Mandela University soccer team in Port Elizabeth. Three-dimensional joint kinematic and ground reaction force data were recorded during the execution of a 60°cutting manoeuvre, prior to and immediately after being exposed to a protocol designed to induce general neuromuscular fatigue of the lower extremity.

Chapter 4: RESULTS

4.1 Introduction

Chapter four presents the results of data analysis for participants in this study and aims to compare the three-dimensional joint kinematic of the ankle, knee and hip joint as well as the ground reaction force data during the execution of a sub- maximal sixty degree cutting manoeuvre, prior to and immediately after being exposed to a protocol designed to induce general neuromuscular fatigue. Three-dimensional joint kinematics were obtained by using 3D opto-electric motion analysis software in South African male soccer players during a sub-maximal 60° cutting manoeuvre. Male participants were between the ages of 18 and 28 years and are members of the Nelson Mandela University soccer team in Port Elizabeth. Additionally, anthropometric data were also obtained during initial data collection, specifically age, height and weight. In total, 13 participants met the inclusion criteria and were retained for analysis. A narrative synthesis of the findings is presented in the discussion that follows.

4.2 Results

Results of the selected variable of three-dimensional joint kinematics and ground reaction force data are presented in terms of statistical analysis.

4.2.1 Anthropometric measurements

The body mass index (BMI) is the metric used for defining anthropometric height/weight characteristics in adults and for classifying (categorizing) them into groups. The common interpretation is that it represents an index of an individual’s fatness (Mardolkar, 2017). However, it is increasingly clear that BMI is a rather poor indicator of the presence of body fat. Importantly, the BMI also does not capture information on the mass of fat in different body sites as well as muscle mass a person might have (Nuttal, 2015). The average BMI score for the soccer players was 22.52 ± 2.77 kg/m², which classify them as normal weight (BMI between 18.5 and 24.9 kg/m²). The anthropometric variables can be seen in Table 1.

Table 1: Means and Standard Deviations for Age and Anthropometric Data of Total Participant Group

Variable	Mean	SD
Age (years)	22.15	2.77
Height (cm)	169.64	5.75
Weight (kg)	64.60	7.04
BMI (kg/m²)	22.52	2.77

4.2.2 Three-dimensional joint kinematics

Three-dimensional joint kinematics is commonly used to determine motion of points/bodies/objects and systems of bodies (groups of objects) using three-dimensional imaging. Different variables (fatigue/leg dominance/environment/equipment etc.) can also be used and compared to determine if a significant effect is present. These values are then analysed and possible risks for injury are determined based on the finding.

4.2.2.1 Three-dimensional joint kinematics at initial contact

Throughout a movement there are different phases such as initial contact, peak stance-phase and push/toe off. The initial contact phases are the first to be investigated, to determine possible three-dimensional joint kinematic changes for the different variables. Table 4.2 illustrates the mean and standard deviation of the effects of leg dominance and fatigue on lower-limb joint rotation at initial contact. Hip flexion-extension rotation angle in the dominant leg (pre-fatigue) at initial contact was 59.94° ± 16.24° and post-fatigue 48.47° ± 37.38°; which show a decrease in flexion-extension angel of after fatigue. A small effect size of fatigue is noticeable in the dominant leg (*d*=0.398) as illustrated by Table 4.4*.* The same pattern is noticeable in the non-dominant leg as well, but to a smaller extent with an effect size of *d=0.313* (small). A decrease in joint rotation is also seen in knee adduction-abduction (12.34° ± 14.43°) to (7.04° ± 9.57°), knee internal-external rotation (5.92° ± -14.49°) to (-3.61° ± 13.05°), ankle dorsiflexion-plantar flexion (6.22° ± 15.72°) to (4.22° ± 13.06°) and ankle internal-external (4.40° ± 4.81°) to (3.34° ± 3.28°) rotation angles. This phenomenon could be attributed to altered muscle activation or change in muscle biomechanics caused by fatigue. Knee adduction-abduction angles at initial contact were similar between dominant (12.34° ± 14.34°) and non-dominant leg (9.92° ± 8.01°) pre-fatigue, but in the case of the dominant leg the knee adduction-abduction angle decreased to 7.04 ± 9.57° after fatigue while the non-dominant leg increased slightly 10.95° ± 8.70°.

Knee flexion-extension angle of the dominant leg was 26.68° ± 13.77° pre-fatigue compared to the non- dominant leg's 35.58° ± 15.12°. A noticeable difference was seen, which indicates that the non-dominant leg had a bigger flexion-extension angle than the dominant leg (pre-fatigue) even though not statistically significant (*p=0.161)*. A medium effect size (0.615) was, thus seen in knee flexion-extension angle (pre-fatigue) between dominant and non-dominant leg as seen is Table 4.

There were no significant differences between pre- and post-fatigue states in any of the joint rotations at initial contact, which indicate that fatigue did not influence the joint rotation angles (see Table 3). However, the most noticeable difference was seen for knee flexion-extension (-11.85°) which is interpreted as the knee being 11.85° more flexed pre-fatigue than post fatigue. This supports the notion that knee flexion decreased in a fatigued state and that many injuries occur because of the knee being in a less flexed position (Olsen *et al.*, 2004). Other noticeable difference occurred during hip adduction-abduction (6.33°) and ankle internal-external rotation (-10.47°).

Table 2: Effects of leg dominance and fatigue (mean ± SD) on lower-limb joint rotations at initial contact

	Dominant		Non-dominant	
Rotation	Pre	Post	Pre	Post
Hip flexion-extension	59.94 ± 16.24	48.47 ± 37.38	60.15 ± 8.41	56.94 ± 11.84
Hip adduction-abduction	-15.28 ± 24.03	-25.26 ± 24.82	-13.62 ± 6.47	-13.38 ± 7.54
Hip internal-external rotation	-2.07 ± 41.66	5.30 ± 46.54	11.81 ± 16.20	19.28 ± 26.89
Knee flexion-extension	26.68 ± 13.77	27.15 ± 14.09	35.58 ± 15.12	24.20 ± 16.81
Knee adduction-abduction	12.34 ± 14.34	7.04 ± 9.57	9.92 ± 8.01	10.95 ± 8.70
Knee internal-external rotation	5.92 ± 14.49	-3.61 ± 13.05	6.04 ± 7.47	3.17 ± 11.10
Ankle dorsiflexion-plantarflexion	6.22 ± 15.72	4.22 ± 13.06	5.01 ± 11.06	3.73 ± 12.68
Ankle internal-external rotation	4.40 ± 4.81	3.34 ± 3.28	2.23 ± 2.71	3.08 ± 2.94
Ankle supination-pronation	-18.35 ± 15.01	-15.92 ± 14.32	-13.35 ± 13.28	-21.39 ± 15.35

The effect of leg dominance did not have a statistically significant impact on any kinematic parameters as well as the interactions between fatigue and non-fatigue trails were also not observed. The parameters however, that did show the most significant effect during the initial contact was the knee and ankle joint, specifically knee flexion-extension (*p=0.161*), knee adduction-abduction (*p=0.280*) and ankle supination-pronation (*p=0.200*) as shown in table 3.

Table 3: Initial contact analyses of differences (mean with 95% confidence bounds) between leg dominance and fatigue state for lower-limb joint rotations

Rotation	Difference of differences	95% CI		p-value
		Lower bound	Upper bound	
Hip flexion-extension	8.26	-15.87	32.39	0.495
Hip adduction-abduction	10.22	-9.83	30.27	0.310
Hip internal-external rotation	0.10	-38.88	39.08	0.996
Knee flexion-extension	-11.85	-28.57	4.87	0.161
Knee adduction-abduction	6.33	-5.33	17.99	0.280
Knee internal-external rotation	6.66	-6.53	19.85	0.315
Ankle dorsiflexion-plantarflexion	0.72	-14.04	15.48	0.922
Ankle internal-external rotation	1.91	-2.03	5.85	0.334
Ankle supination-pronation	-10.47	-26.65	5.71	0.200

Effect size is a quantitative measure of the magnitude of a phenomenon and a larger absolute value always indicates a stronger effect. A medium effect size of fatigue is noted for knee adduction-abduction ($d=0.435$), knee internal-external rotation ($d=0.691$) and knee flexion-extension angles as seen in Table 4.4. In other words, the effect size of fatigue on the joint rotation angle of the dominant/non-dominant leg was of medium effect practically. For example, the non-dominant leg's knee flexion-extension angle pre-fatigue was 35.58° ± 15.12°, while the angle after fatigue was 24.20° ± 16.81°, this indicates a decrease in joint rotation angle and is classified as a medium effect size (*d=0.712*) according to Cohen's d. The same observation can be made for the effect the variable leg dominance has on the kinematic parameters. Leg dominance has a medium effect size on knee adduction-abduction (*d=0.428*), knee flexion-extension (*d=0.615*) and hip adduction-abduction (*d=0.648*). This means that for the later, hip adduction-abduction angles of the dominant leg had a value of 25.26° ± 24.82° and the non-dominant an angle of 13.38° ± 7.54°. There is thus a decrease in joint rotation, which indicated a medium effect size of fatigue.

Table 4: Effect size (Cohen's d) for each dependent experimental condition at initial contact

Rotation	Dominant vs. Non-dominant		Pre vs. Post Fatigue	
	Pre	Post	Dominant	Non-dominant
Hip flexion-extension	-0.016	-0.305	0.398	0.313
Hip adduction-abduction	-0.094	-0.648	0.409	-0.034
Hip internal-external rotation	-0.439	-0.368	-0.167	-0.337
Knee flexion-extension	-0.615	0.19	-0.034	0.712
Knee adduction-abduction	0.208	-0.428	0.435	-0.123
Knee internal-external rotation	-0.01	-0.56	0.691	0.303
Ankle dorsiflexion-plantarflexion	0.089	0.038	0.138	0.108
Ankle internal-external rotation	0.556	0.083	0.257	-0.301
Ankle supination-pronation	-0.353	0.369	-0.166	0.560

4.2.2.2 Three-dimensional joint kinematics at peak stance-phase

The second phase investigatesthe peak stance-phase. This phase indicates what the maximum kinematic parameter of each joint was (e.g. peak knee flexion defined as the point at which the ground reaction force is at its greatest throughout the movement). It is important to evaluate lower limb joint rotation angles during peak stance-phase, as this is when the greatest force is experienced by the joint. Abnormal joint angle changes during this phase could cause exercise shear, torsion or compression forces on the structures surrounding the joints and may increase the risk of injury substantially. The mean and standard deviation of the effects of leg dominance and fatigue on lower-limb joint rotation at peak stance-phase is illustrated in Table 5. The positive or negative angles indicate what motion is being performed e.g. a positive angle in the sagittal plane means the lower limb is in front of the body and thus flexion in being performed. This has been explained in the previous chapter under '3.8 Biomechanical analysis'. Hip flexion, abduction and internal rotation angles are similar pre and post-fatigue as well as in the dominant and non-dominant leg. The hip orientation at peak stance-phase are as follows; 56°-60° of hip flexion, 21°-24° abduction and 14°-22° internal rotation which is consistent with the hip's biomechanical movement during a cutting task (Hewett *et al.*, 2005). Hip internal rotation angle of the dominant leg, pre-fatigue (14.83° ± 16.77) is slightly lower than post-fatigue (22.55° ± 28.08°). An increase is thus seen in internal rotation of the hip. A similar increase in hip internal rotation was also noted in the non-dominant leg. These finding correlate with findings made by Alentorn-Geli *et al.*, 2009; Lucci, Corte, Van Lunen, Ringleb and Onate, (2009) that hip internal rotation increases with fatigue.

Within the body a kinetic chain exists, the concept was introduced by Franz Reuleaux, a mechanical engineer, in 1875. He proposed that rigid, overlapping segments were connected via joints and this created a system whereby movement at one joint produced or affected movement at another joint in the kinetic link (Deschamps, Eerdekens, Geentjens, Santermans, Steurs, Dingenen, Thysen & Staes, 2018). According to this theory the increased hip internal rotation angle should affect the knee and ankle joint rotations as well. This is indeed evident in knee joint rotation angles, where knee adduction angle increases by 7.88° and the ankle internal rotation increased by 1.05° Internal rotation of the knee (4.93° ± 12.03°) at peak stance-phase during fatigue is noticeably less than before fatigue (16.40° ± 18.55°) as illustrated by Table 5. This shows a decrease of 11.47° which is consistent with literature and a possible risk factor to injury (Dvorak *et al.*, 2000). It is important to note that the different activation patterns at the knee may also have been due to moment requirements at the ankle and hip joints. The bi-articular muscles of hip and knee, and ankle and knee, contribute to the moments generated at the hip and ankle, respectively. Thus, if changes occurred in the external moments at the hip and ankle during a cutting manoeuvre, these may have resulted in changed activation patterns of the bi-articular muscles.

Knee joint rotation angles were within the normal ranges of knee joint movement during a cutting manoeuvre, but it is interesting to note that the knee flexion at peak stance-phase in the dominant leg pre-fatigue is 46.28° ± 18.14° while the knee flexion angle of the non-dominant leg was 54.96° ± 7.77°. This indicates that the dominant leg reaches a smaller knee flexion at peak stance-phase than the non-dominant leg. Knee abduction angle (19.05° ± 14.31°) was also noticeable smaller in the dominant than in the non-dominant leg (14.28° ± 11.00°). The non-dominant leg is thus more adducted than the dominant leg (pre-fatigue).

McLean, Walker, *et al.* (2005) compared hip, knee and ankle kinematics during a pre and post-fatigue sidestep manoeuvre. Their results indicated the peak rotation deviation for the hip joint; hip flexion 49.5°, hip abduction -24.3° and hip internal rotation 19.3° Our joint kinematic shows similar results with hip flexion to be, 59.91°, hip abduction 23.33 and hip internal rotation to be 14.83°. The knee joint measurements of McLean, Walker, *et al.*, (2005) also differ slightly. Knee flexion was 19.3°, knee adduction 3.8° and knee internal rotation was 19.0°. Our results show that knee flexion peaks at 46.28°, knee abduction 19.05° and knee internal rotation 16.40°. Normative values for all joint angles during a cutting task are needed so researchers and coaches can compare the athlete to the norms and not only each athlete's own baseline values.

Table 5: Effects of leg dominance and fatigue (mean ± SD) on peak stance-phase lower limb joint rotations

	Dominant		Non-dominant	
Rotation	Pre	Post	Pre	Post
Hip flexion	59.91 ± 11.68	57.67 ± 4.02	60.67 ± 7.26	56.94 ± 11.84
Hip abduction	-23.33 ± 5.96	-24.97 ± 5.48	-21.70 ± 9.59	-21.72 ± 10.74
Hip internal rotation	14.83 ± 16.77	22.55 ± 28.08	13.20 ± 16.01	19.51 ± 26.81
Knee flexion	46.28 ± 18.14	49.53 ± 13.04	54.96 ± 7.77	44.46 ± 21.65
Knee abduction	19.05 ± 14.31	11.17 ± 12.96	14.28 ± 11.00	14.74 ± 17.87
Knee internal rotation	16.40 ± 18.55	4.93 ± 12.03	16.61 ± 9.15	14.20 ± 12.90
Ankle dorsiflexion	27.43 ± 16.44	28.01 ± 10.70	25.79 ± 13.79	24.59 ± 16.39
Ankle pronation	7.31 ± 10.40	4.00 ± 2.74	4.13 ± 5.46	3.22 ± 2.47
Ankle external rotation	-21.97 ± 18.55	-20.92 ± 13.35	-20.87 ± 10.06	-24.39 ± 11.28

No significant differences between pre and post-fatigue trails were found in any of the joint angles at peak stance, which indicate that fatigue or leg dominance did not have a significant effect on the joint rotation angles (see Table 6). In terms of the difference of differences the knee flexion angle was the most noticeable from pre to post-fatigue (13.75°) which indicates that knee flexion decreased with 13.75°. This supports the findings by (Cortes *et al.*, (2012) and Olsen *et al.*, (2004) that knee flexion angles decreases during a fatigued state and that a greater number of injuries occur to the knee joint when the it is in a less flexed position. Knee abduction (8.34°) and internal rotation (9.06°) however, showed noticeable increases in joint rotation angles at peak stance-phase. Therefore, during peak-stance-phase the parameters that showed the biggest difference in terms of significance were knee flexion ($p=0.129$), knee abduction ($p=0.297$) and knee internal rotation ($p=0.235$).

Table 6: Peak stance-phase difference of differences (mean with 95% confidence bounds) between limb dominance and fatigue state (mean ± SD) for lower limb joint rotations

Rotation	Difference of differences	95% CI		p-value
		Lower bound	Upper bound	
Hip flexion	-1.49	-11.86	8.88	0.774
Hip abduction	1.62	-7.6	10.83	0.725
Hip internal rotation	-1.41	-26.63	23.81	0.911
Knee flexion	-13.75	-31.63	4.13	0.129
Knee abduction	8.34	-7.56	24.24	0.297
Knee internal rotation	9.06	-6.1	24.22	0.235
Ankle dorsiflexion	-1.78	-17.98	14.42	0.826
Ankle pronation	2.4	-4.47	9.27	0.486
Ankle external rotation	-4.57	-19.85	10.71	0.550

During a pre-fatigued state knee flexion angle for the dominant leg was 46.28° ± 18.14° and non-dominant leg 54.98° ± 44.46°. According to Cohen's d, this indicates a medium effects size of *d=0.622*, pre-fatigue (see Table 7*)*. Practically this means that leg dominance had an effect on the joint rotation angles in a pre-fatigue state. Knee internal rotation angle also show a medium effect size (*d=0.743*) post-fatigue.

In the case of fatigue vs. non-fatigue, a medium effect size was noted in the dominant leg during ankle pronation (*d=0.435*), knee abduction (*d=0.577*) and knee internal rotation (*d=0.734*). A medium effect size was also noted for knee flexion (*d=0.646*), in the non-dominant leg.

Table 7: Effect size (Cohen's d) for each dependent experimental condition for peak stance kinematics

	Dominant vs Non-dominant		Pre vs Post Fatigue	
Rotation	Pre	Post	Dominant	Non-dominant
Hip flexion	-0.078	0.083	0.256	0.380
Hip abduction	-0.204	-0.381	0.286	0.002
Hip internal rotation	0.099	0.111	-0.334	-0.286
Knee flexion	-0.622	0.284	-0.206	0.646
Knee abduction	0.374	-0.229	0.577	-0.031
Knee internal rotation	-0.014	-0.743	0.734	0.215
Ankle dorsiflexion	0.108	0.247	-0.042	0.079
Ankle pronation	0.383	0.299	0.435	0.215
Ankle external rotation	-0.074	0.281	-0.065	0.329

4.2.2.3 Time to peak stance phase lower limb joint rotation

The angular displacement of the respective joint throughout the cutting task for each plane of movement is illustrated by Figure 9, 10 and 11. Frontal plane=abduction/adduction, sagittal plane=flexion/extension, transverse plane=internal/external rotation. The figures make it possible to establish the time it takes for each joint (and each variable) to reach peak stance phase.

The angular displacement of the ankle joint throughout the cutting manoeuvre is illustrated by Figure 4.1 Pronounced ankle pronation in the dominant non-fatigue leg is seen throughout, reaching a climax midway through the movement. It is also noticeable that peak ankle dorsiflexion occurs at about 55%-60% of the way through the movement for all variables.

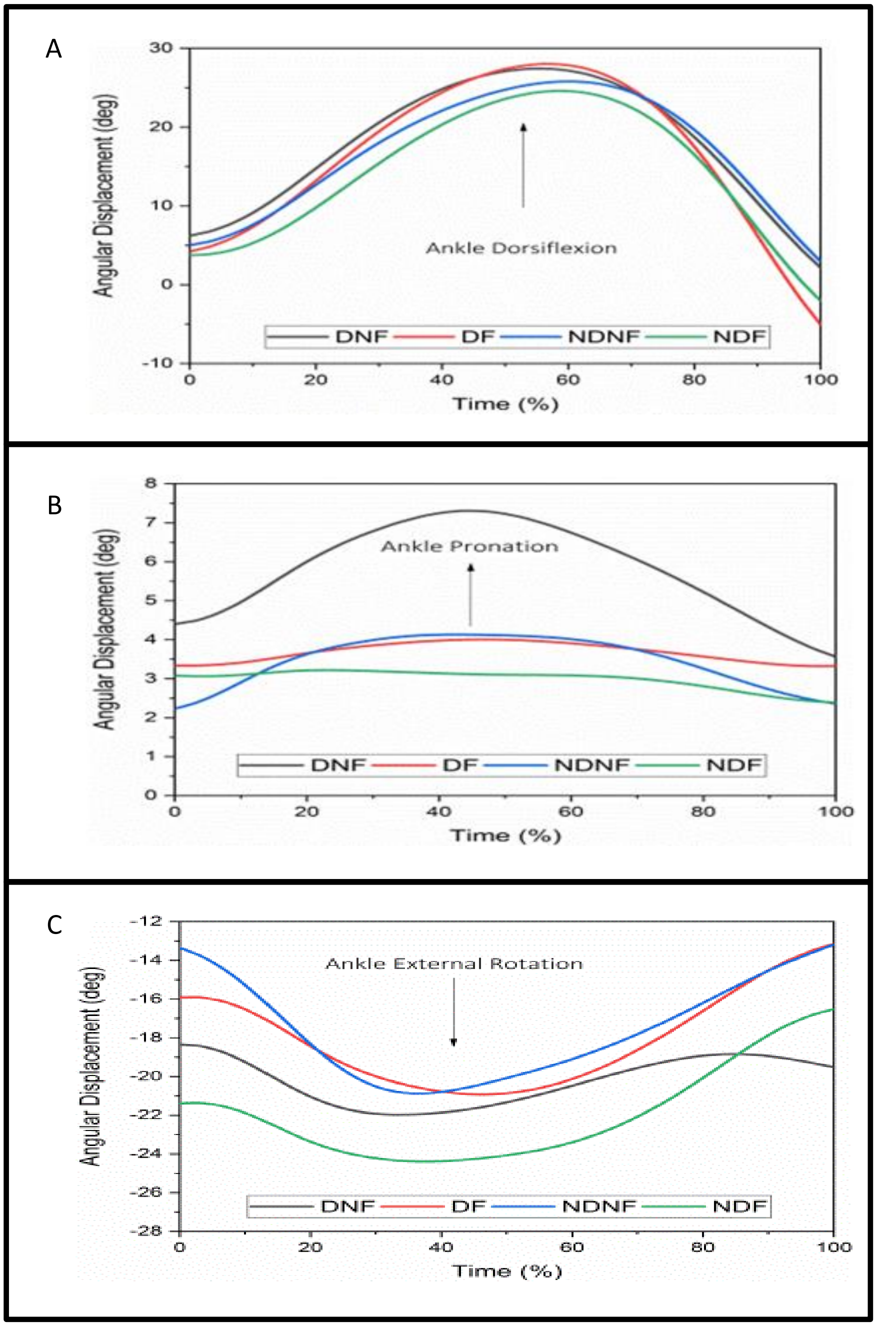

Figure 9: Effects of leg dominance and fatigue conditions on mean normalized kinematic parameters during a 60° sub- maximal cutting manoeuvre (ankle). For panels A, B and C, positive values denote ankle dorsiflexion, internal rotation and pronation respectively. DNF = dominant non-fatigued; DF = dominant fatigued; NDNF = non-dominant non-fatigued; NDF = non-dominant fatigued.

Angular displacement of the knee joint over time is illustrated by Figure 10. It is interesting to note that the angular displacement of the knee is marginal for the dominant leg post-fatigue compared to the other variables during knee internal rotation and knee abduction. There is also a noticeable difference between the dominant non-fatigued limb's angular displacement/knee abduction angles compared to the dominant fatigued leg. All of the peak angular displacements of the joints occur at about 40%-50% of the movement.

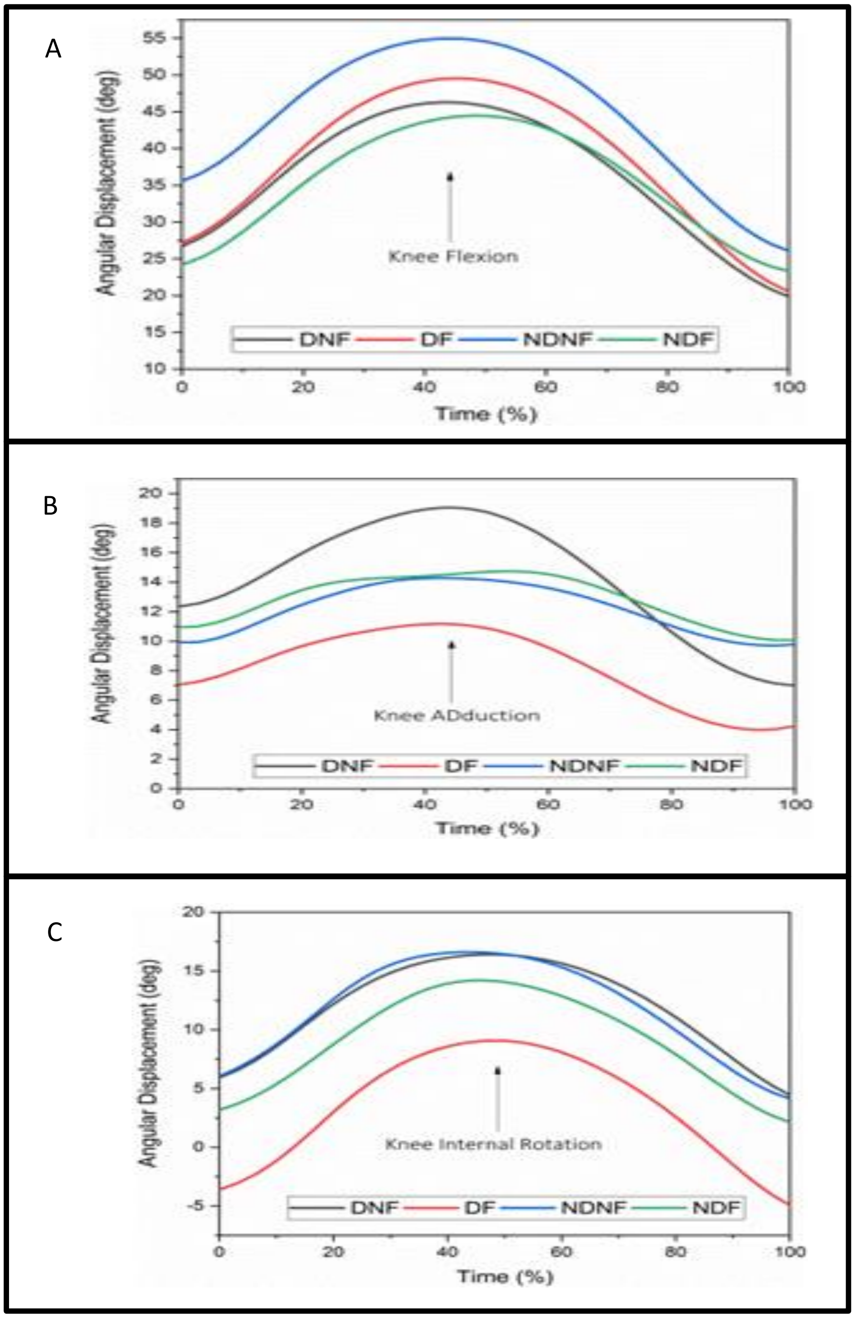

Figure 10: Effects of leg dominance and fatigue conditions on mean normalized kinematic parameters during a 60° sub- maximal cutting manoeuvre (Knee). For panels A, B and C, positive values denote knee flexion, abduction and internal rotation respectively. DNF = dominant non-fatigued; DF = dominant fatigued; NDNF = non-dominant non-fatigued; NDF = non-dominant fatigued.

The hip's angular displacement's curve/shape looks different to the previous figures. Peak hip flexion (positive displacement) occurs quite early in the stance phase (0%-20%) and then moves into hip extensions as the movement progresses. Similarly, with hip internal/external rotation, the hip starts in slight external rotation, but moves into internal rotation at about 20% of time and the back into external rotation. At the end of time the hip position is similar to that of the beginning. There is a clear difference between dominant and non-dominant leg hip angular displacement as shown by the red and black (dominant), and blue and green (non-dominant).

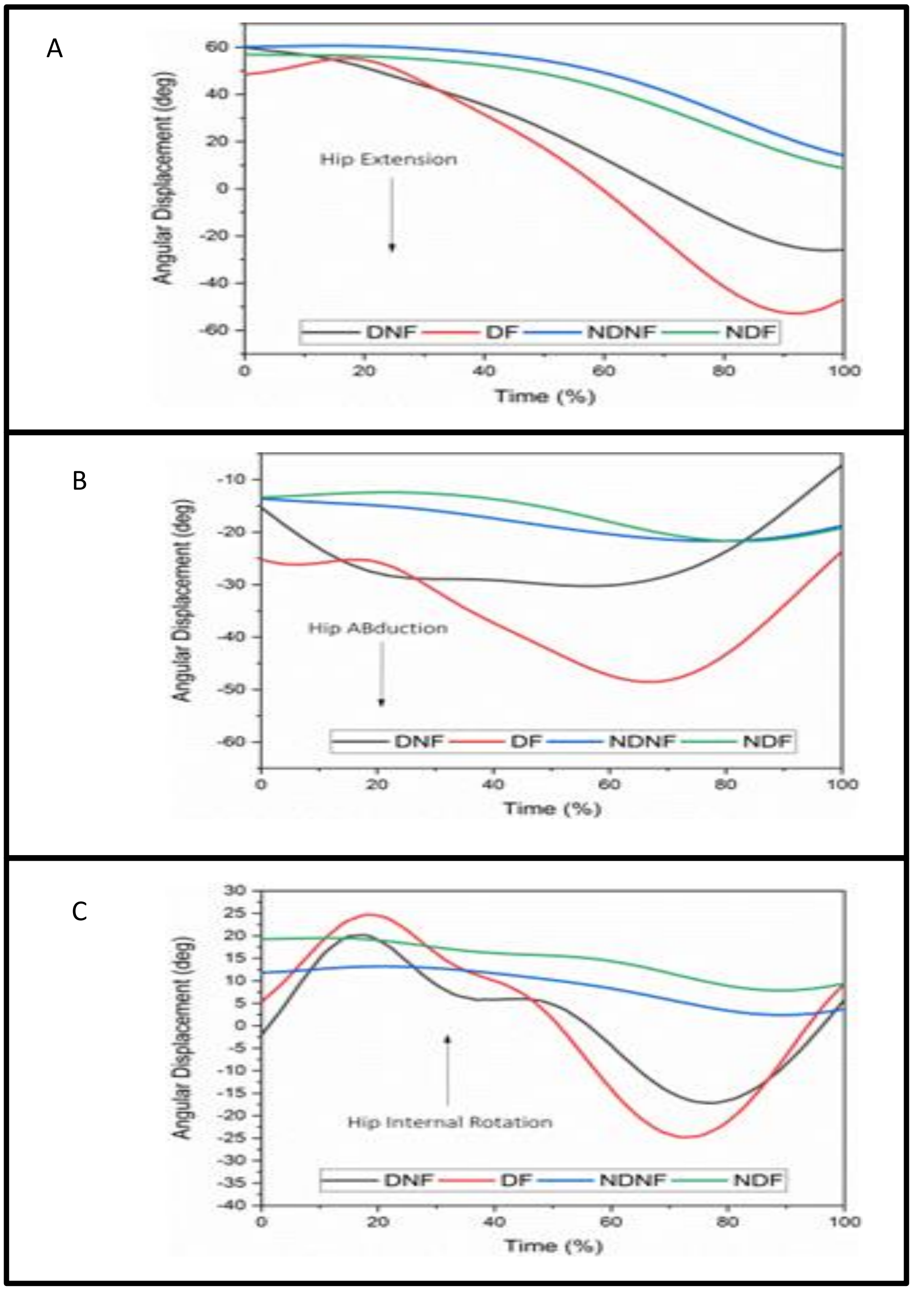

Figure 11: Effects of leg dominance and fatigue conditions on mean normalized kinematic parameters during a 60° sub- maximal cutting manoeuvre (Hip). For panels A, B and C, positive values denote hip flexion, adduction and internal rotation respectively. DNF = dominant non-fatigued; DF = dominant fatigued; NDNF = non-dominant non-fatigued; NDF = non-dominant fatigued.

Combined, figure 9-11 illustrate the effect of leg dominance and fatigue on kinetic parameters during the peak stance-phase of a 60° sub-maximal cutting manoeuvre. A number of initial contact and peak stance–phase lower limb-joint rotation were influenced by fatigue during the execution of the sub-maximal 60° cutting manoeuvre. The main effect of fatigue produced an increase in knee internal rotation and hip abduction and a decrease in peak knee abduction angles compared to non-fatigue but was not statistically significant. Difference between dominant and non-dominant was primarily seen during ankle pronation, knee adduction, hip external rotation and hip abduction. The dominant leg showed increased ankle pronation and knee adduction while it indicated significant decrease is hip external rotation and hip abduction compared to the non-dominant leg.

The percentage of time the lower limb took to reach peak stance is illustrated by Table 8. It took less time to reach peak stance phase during hip abduction (77%), knee flexion (43%), knee internal rotation (47%), ankle dorsiflexion (55%), ankle pronation (44%) and ankle external rotation (33%). In a pre-fatigued state compared to post fatigue. On the other hand, hip flexion (0%), hip internal rotation (19%) and knee abduction (42%) reached peak stance phase quicker post-fatigue. In addition similar finding was observed for the non-dominant leg, with the exception of knee abduction (pre=43%; post=53%), ankle dorsiflexion (pre=60%; post=59%) and ankle pronation (pre=42%; post=23%) where the opposite was noted.

Table 8: Time (% of stance phase) to peak stance-phase lower-limb joint rotations

	Dominant		Non-dominant	
Rotation	Pre	Post	Pre	Post
Hip flexion	15 ± 4	0 ± 7	15 ± 6	0 ± 8
Hip abduction	77 ± 8	78 ± 13	78 ± 5	83 ± 14
Hip internal rotation	25 ± 15	19 ± 10	21 ± 15	11 ± 10
Knee flexion	43 ± 11	45 ± 4	44 ± 7	49 ± 3
Knee abduction	44 ± 6	42 ± 14	43 ± 14	53 ± 9
Knee internal rotation	47 ± 4	48 ± 5	44 ± 12	45 ± 14
Ankle dorsiflexion	55 ± 10	57 ± 16	60 ± 9	59 ± 3
Ankle pronation	44 ± 11	47 ± 15	42 ± 14	23 ± 16
Ankle external rotation	33 ± 13	46 ± 5	37 ± 7	38 ± 12

In the sagittal plane, flexion of the hip, knee and ankle joint is associated with force absorption. Impulse force can be defined as the product of average force and the time it is exerted The definition of impulse force makes it possible to minimize the impact force a joint experiences on collision/initial contact (Von Ossietzky, 2011). $F=\nabla p/\nabla t$, therefore if 't' is increased, for a constant change in momentum, the force on the body is reduced. In other words, if an impact stops a moving object, then the change in momentum is a fixed quantity, and extending the time of the collision will decrease the time average of the impact force by the same factor.

The difference of differences for time to peak stance shows an increase for hip abduction (4%), knee flexion (3%) and knee abduction (12%) between limb dominance and fatigue state as seen in Table 9. These increases indicate that the peak stance phase was reached after a slightly longer period. The difference, however, were not statistically significance; *p=0.502*, *p=0.442* and *p=0.061*. In contras the difference of differences also show a decrease of 4%, 3%, 22% and 12% in hip internal rotation, ankle dorsiflexion, ankle pronation, and ankle external rotation respectively. Significant differences were found between dominant and non-dominant leg as well as between fatigue and non-fatigue with ankle pronation (*p=0.007*) and ankle external rotation (*p=0.033*). Knee abduction (*p=0.061*) showed an effect even though not statistically significant.

Table 9: Time to peak stance-phase difference of differences (mean with 95% confidence bounds) between limb dominance and fatigue state for lower-limb joint rotations

Rotation	Difference of differences	95% CI		p-value
		Lower bound	Upper bound	
Hip flexion	0	-7.16	7.16	0.998
Hip abduction	4	-7.88	15.88	0.502
Hip internal rotation	-4	-18.22	10.22	0.574
Knee flexion	3	-4.8	10.79	0.442
Knee abduction	12	-0.58	24.58	0.061
Knee internal rotation	0	-10.88	10.88	0.998
Ankle dorsiflexion	-3	-14.78	8.78	0.611
Ankle pronation	-22	-37.75	-6.25	0.007
Ankle external rotation	-12	-22.97	-1.03	0.033

As mentioned previously Cohen's d is an effect size used to indicate the standardised difference between two means. A small (0.2-0.49), medium (0.5-0.79) and large (>0.8) effect size can be calculated. Due to the statistically significant p-value (see Table 9) found for ankle pronation (*p=0.007*) and external rotation (*p=0.033)*, these joints are expected to show a large effect size. In the case of fatigue, a large effect size was seen for ankle external rotation (*d=1.32*) in the dominant leg as well as for ankle external rotation (*d=1.264*) in the non-dominant leg as illustrated in Table 10). Other large effect sizes are also noted for hip flexion [dominant leg (*d=2.31*), non-dominant leg (*d=2.121*)], knee flexion [non-dominant leg (*d=0.935*)] and knee abduction [non-dominant leg (*d=0.85*)].

In the case of dominant vs. non-dominant leg, a large effect size is also expected for ankle pronation and ankle external rotation because of their significant p-values. This is indeed the case and a large effects size is found for ankle pronation (*d=1.548*) post fatigue and ankle external rotation (*d=0.87*) post fatigue. Knee flexion (*d=1.131*) and knee abduction (*d=0.935*) *also* indicates large effect sizes in the fatigued states.

Table 10: Effect size (Cohen's d) for each dependent experimental condition for time to peak stance phase

	Dominant vs. Non-dominant		Pre vs. Post Fatigue	
Rotation	Pre	Post	Dominant	Non-dominant
Hip flexion	0.00	0.00	2.631	2.121
Hip abduction	-0.150	-0.370	-0.093	-0.476
Hip internal rotation	0.267	0.8	0.471	0.784
Knee flexion	-0.108	-1.131	-0.242	-0.928
Knee abduction	0.093	-0.935	0.186	-0.85
Knee internal rotation	0.335	0.285	-0.221	-0.077
Ankle dorsiflexion	-0.526	-0.174	-0.15	0.149
Ankle pronation	0.159	1.548	-0.228	1.264
Ankle external rotation	-0.383	0.87	-1.32	-0.102

4.2.3 Ground reaction force

Ground reaction force (GRF) is the force exerted by the ground on a body in contact with it (Thompson, 2002). Lateral posterior and vertical GRF is illustrated by Figure 12. For the lateral component it is evident the dominant leg exerts a greater GRF irrespective of fatigue. Posterior GRF shows that the dominant leg still applies a greater GRF, but only during a fatigued state. These findings can be attributed to a decrease in force absorption ability by especially the knee joint. It is proposed that the quadriceps ability to eccentrically control knee flexion decreased due to fatigue. This phenomenon was also present when investigating the time to peak stance phase. In the vertical aspect the curve look similar, except the dominant no fatigue and non-dominant fatigue leg show a greater GRF compared to the other two variables.

Peak vertical forces were significantly different between the fatigued state compared to the non-fatigued state during the cutting manoeuvre (F= 23.51 N.kg^{-1}, *p*= 0.035). This might indicate that neuromuscular fatigue influence landing forces on impact during a directional change and thus correlates with McGovern *et al.* (2015), McLean *et al.* (2003), McLean *et al.* (2007) and Condello *et al.* (2016) findings. Patterns and magnitudes of initial-contact and peak kinematic and kinetic data corresponded with data previously reported for similar movements (Condello *et al.*, 2016; Decker, Torry, Wyland, Sterett & Steadman, 2003).

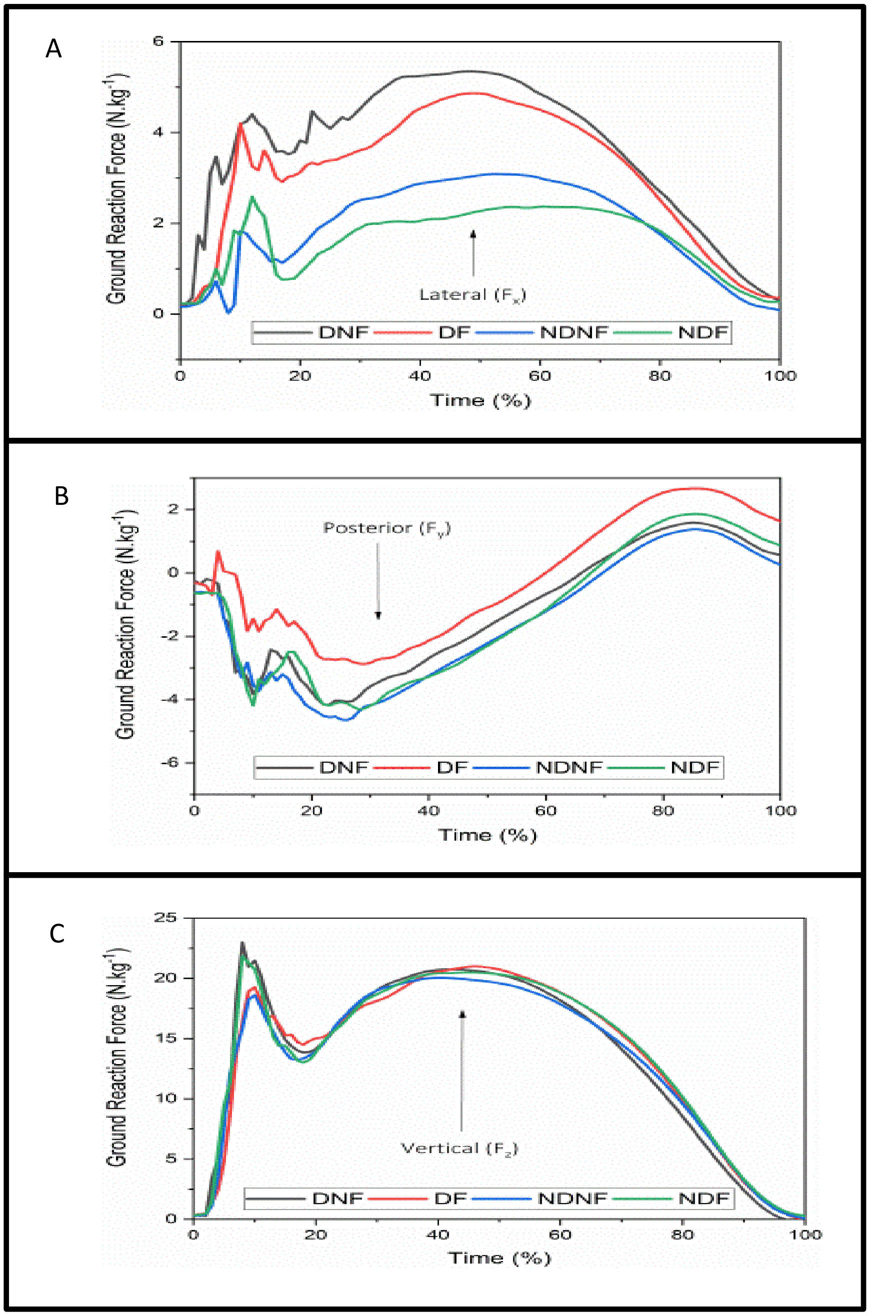

Figure 12: Effects of leg dominance and fatigue conditions on mean normalized ground reaction force data during a 60° sub- maximal cutting manoeuvre. For panels A, B and C, values denote lateral, posterior and vertical forces respectively. DNF = dominant non-fatigued; DF = dominant fatigued; NDNF = non-dominant non-fatigued; NDF = non-dominant fatigued.

4.3 Summary

The effect of leg dominance did not have a statistically significant impact on any kinematic measures as well as the interactions between fatigue and non-fatigue trials were also not observed for any of the kinematic parameters. However, significant differences in time to peak stance phase lower limb joint rotations were found between dominant and non-dominant legs as well as between fatigued and non-fatigued states with ankle pronation ($p=0.007$) as well as ankle external rotation ($p=0.033$). Knee abduction ($p=0.061$) also showed an effect even though not statistically significant. In addition, ground reaction forces indicated that peak vertical forces were also significantly increase or decrease between the fatigued state compared to the non-fatigued state during the cutting manoeuvre ($F = 23.51$, $p = 0.035$).

The main effect of fatigue produced an increase in knee internal rotation and hip abduction and a decrease in peak knee abduction angles compared to non-fatigue, but was not statistically significant. Differences between dominant and non-dominant legs were primarily seen during ankle pronation, knee adduction, hip external rotation and hip abduction. The dominant leg had increased ankle pronation and knee adduction while it had significant decrease is hip external rotation and hip abduction compared to the non-dominant leg.

Chapter 5: DISCUSSION

5.1 Introduction

Soccer is predominantly a semi-contact sport with a considerable annual injury count, with the rate, severity and location of such injuries varying substantially (Dvorak & Junge, 2000). The majority of injuries tend to occur in the second half of a soccer game, potentially indicating that fatigue, whether muscular, neurological, localized or general, may be one of the causing factors (Gioftsidou, Malliou, Pafis, Beneka, Tsapralis, Sofokleous, Kouli & Godolias, 2012). Soccer requires proficientcy of dribbling and shooting with both lower limbs, even though the players may have a preferred/dominant limb. Controversy exist among researchers with regards to the effect that leg-dominance has on in lower limb kinematic data and the risk for injury (Brophy *et al.*, 2010; Brown *et al.*, 2014; Smykalski, 2016). This current study suggests that there is little to no effect of leg dominance on three-dimensional joint kinematics.

The effect of fatigue on lower limb kinematics is also a highly debated topic among researchers and athletes. Literature supports both notions and provides valid arguments for each, but as we discover the true effects of fatigue (that it is likely linked to inorganic phosphates and hydrogen ion accumulation) the argument for fatigue having no significant effect on limb kinematics is becoming more plausible (Bangsbo *et al.*, 1996; Chappell, 2005; Cortes *et al.*, 2012; McLean *et al.*, 2007).

Anterior cruciate ligament (ACL) injuries, although not the most common, tend to be one of the most traumatic sports-related knee injuries in soccer. Acute injury to the knee joint can be debilitating and injury to the musculoskeletal system can often increase the likelihood of significant long-term effects, including permanent knee instability, meniscus tears, cartilage injury and the development of osteoarthritis (McLean *et al.*, 2003). According to (McLean *et al.*, 2003), the key to understanding the potential mechanisms of particularly knee injuries, is to determine the joint loading characteristics associated with an injury-causing event. More complete insights into the effects of fatigue on the mechanics of turning and change-of-direction tasks are therefore warranted.

5.2 Effects of leg dominance on three-dimensional joint kinematics

One of the key findings of the present study related to limb dominance was that no statistically significant differences were evident for any of the dependent variables (limb dominance; fatigue state) related to the independent variables (i.e. joint [hip, knee, ankle], contact time, ground reaction force). Non-significant differences were found in the hip joint (dominant vs. non-dominant legs in a non-fatigued state) in all three planes of motion ($F=0.61$, $p=0.55$). Statistical significance is the probability that the observed difference between two groups is due to chance. If the *P* value is larger than the alpha level chosen (e.g. 0.05), any observed difference is assumed to be explained by sampling variability. With a sufficiently large sample, a statistical test will almost always demonstrate a significant difference, unless there is no effect whatsoever, that is, when the effect size is exactly zero; yet very small differences, even if significant, are often meaningless. Thus, reporting only the significant p-value for an analysis is not adequate for readers to fully understand the results. Similar kinematic characteristics were observed for the knee joint ($F=1.25$, $p=0.48$) and the ankle joint ($F=3.33$, $p=0.64$).

Non-significant differences were also observed during the fatigued state in all three planes of motion for the hip joints ($F = 8.98$, $p=0.72$), knee joints ($F=5.21$, $p=0.41$) and ankle joints ($F = 0.21$, $p = 0.12$). Such a finding is not surprising based on the available literature. Soccer does not put exclusive strain on only one lower extremity, however most soccer players definitely have a preferred kicking limb (Brophy *et al.*, 2010). This is likely to put differential demands on the lower extremities given the differences in muscle activation seen in the kicking limb compared with the supporting limb (Calligeris *et al.*, 2015). A number of studies have investigated the relationship between leg dominance and injury, but did not find any significant results (Brophy *et al.*, 2010; Smykalski, 2016; Velotta *et al.*, 2011). A relatively recent retrospective study of just over 300 participants with non-contact ACL tears, reported no significant correlation between the side of injury and the dominant limb for kicking (Velotta *et al.*, 2011).

Researchers defined the kicking limb as the preferred/dominant limb to use with movements involving coordination and this might be as a result of an innate cerebral dominance that lateralizes a person's dominance to one side of the body (Ross *et al.*, 2004). However, individuals may simply prefer using the kicking limb with other activities and might lead to better strength development of knee extensors and flexors of that particular limb (Smykalski, 2016). Even though players may have a preferred kicking leg, the game of soccer requires the players to dribble and shoot with both legs. This could also contribute to the cause of our findings. Ross *et al.* (2004), also found similar results where no significant differences were found between limbs with hip abduction/adduction and foot dorsiflexion/plantar flexion and foot eversion/inversion. In addition Ross *et al.* (2004) found that during a balance test, both limbs took just as long to stabilize, but each limb used different strategies when landing on a single limb. The difference in muscle activation strategies and utilization of different muscle groups to support joint position could explain why no significant difference between dominant and non-dominant leg was found.

5.3 Effects of fatigue on three-dimensional joint kinematics

The correlation between neuromuscular fatigue and athletic performance is well documented in the available literature (Alentorn-Geli *et al.*, 2009; Coelho, 2015; McLean *et al.*, 2007; Naidoo, 2007; Sanna & O'Connor, 2009). Neuromuscular fatigue can be defined as an inability to maintain a given force or power output and has been found to have both central and peripheral origins. Some muscle properties change during fatigue including the action potential, extracellular and intracellular ion concentrations, and many intracellular metabolites (Allen et al., 2008). A range of mechanisms have been identified that contribute to the decline of performance (Noakes *et al.*, 2004). Many muscle properties change during fatigue including the action potential, extracellular and intracellular ions, and many intracellular metabolites. Noakes et al. (2004) explain that the mechanisms considered are the effects of ionic changes on the action potential, failure of SR Ca2+

release by various mechanisms, and the effects of reactive oxygen species. (Decorte *et al.*, 2012) have recently assessed the time course of central and peripheral fatigue development during intermittent bouts of constant load exercise. This is the only study that has shown that reduced efficacy of excitation-contraction (E-C) coupling was “compensated for” by an increase in central motor drive, as indicated by an increase in the electromyogram (EMG) amplitude. As mentioned previously as neuromuscular fatigue develops, associated changes in muscle activity and activation occur (Gandevia, 2001). Joint kinematics changes and increases in the variability of movement patterns (joint variability) is often noticed (Srinivasan & Mathiassen, 2012). This phenomena indicated that variations in movement patterns may in fact be a beneficial adaptation within the neuromuscular system that may assist in delaying further fatigue development, avoiding injury and maintaining task performance (Bartlett *et al.*, 2007; Srinivasan & Mathiassen, 2012). These adjustments occur presumably to increase the activation of less fatigued muscles or muscle groups and maintain task performance. Increases in knee internal rotation (Nyland et al., 1997) and abduction loading (Chappell, 2005), have previously been observed in the presence of fatigue during crossover cutting manoeuvres. The fact that no significant differences were observed in the present study for both of the aforementioned loading parameters may be attributable to variations in the movement tasks and/or test populations (Chappell, 2005). Differences in subject skill levels have also been one of the proposed explanations for the differences in joint biomechanical profiles for similar movement tasks (Chappell, 2005; Lattanzio, Petrella, Sproule & Fowler, 1997). Specific factors were controlled for in the present study (e.g. using only University soccer players competing at Varsity Cup level) to make comparisons to the literature more robust, yet despite this no differences in lower limb kinematics were evident. However, knee flexion angle differences were the most noticeable from pre- to post-fatigue, decreasing by 13.75°. The reduced knee flexion angle at peak posterior ground reaction force has been theorized to be associated with increased anterior shear force (Dai, Mao, Garrett & Yu, 2014). This anterior shear force is thought to increase the load and strain in the anterior cruciate ligament and thus the probability of rupture. Hip flexion also showed a decrease (1.49°), while knee abduction increased by 8.34° and internal rotation increased by 9.06°, thus indicating that the knee and hip joint were more extended post-fatigue. These altered knee and hip positions have been theorized to increase the likelihood of injury. We found that during post-fatigue, the participants were in a more extended position at the knee and hip joint. Landing in an extended position has been associated with increased proximal tibia anterior shear force due to increased patellar tendon-tibia shaft angle (Condello *et al.*, 2016). This increased force is thought to increase the load placed on the ACL during cutting/landing activities (Chappell, 2005). We speculate that our participants had increased load on the ACL due to their erect posture at post-fatigue. This is most likely due to the mechanical disadvantage experienced by the hamstring muscles that cannot co-contract strongly enough at smaller angles to produce a large posterior force (Ford *et al.*, 1997). It could also be due to the lack in eccentric hamstring control and quadriceps dominance.

The muscle activation patterns during the landing phase of jumping and cutting movements in particular are considered to be quadriceps dominant (Hewett et al., 2005) and has been shown that the quadriceps are more effectively recruited at or near exhaustion (Clark, Bryant, Culgan & Hartley, 2005). It is possible to successfully perform sagittal-plane landing movements, which are controlled and driven primarily by the knee extensors (Chappell, 2005), despite other major muscle groups such as the hamstrings and external hip rotators being fatigued. In this instance knee abduction is likely to increase because external knee-abduction moment is known to be sensitive to both hamstring and hip rotator muscle control (Besier & Lloyd, 2003). However, a lack of effective muscle synchrony can potentially cause hazardous anterior knee loads (Cowling & Steele, 2001). However, a relatively linear increase in neuromuscular fatigue is unlikely to occur during high-intensity exercise; therefore, it is important to examine the temporal changes in central and peripheral fatigue mechanisms during a repetitive, high-intensity exercise bout. Decorte *et al.* (2012) has recently assessed the time course of central and peripheral fatigue development during intermittent bouts of constant load exercise. This is the only study that has shown that reduced efficacy of excitation-contraction (E-C) coupling was “compensated for” by an increase in central motor drive, as indicated by an increase in the electromyogram (EMG) amplitude (Decorte *et al.*, 2012).

It is proven that previous experience in performing a task lays down a neurological movement pattern required to perform that skill (Besier & Lloyd, 2003). Perhaps our specific sidestepping tasks were less familiar to the soccer players; therefore, the muscle activation patterns required to counter the valgus and internal rotation loads during this manoeuvre may not be as finely tuned as those chosen to perform a sidestep with a ball. This raises the question; can people be trained to improve muscle activation patterns to counter the external loads applied to the knee joint during cutting manoeuvres to reduce the risk of ligament injury? Perhaps with further training or practice performing the cutting tasks, muscle activation patterns might be redirected to counter the applied varus and external rotation load. Assessing muscle activation strategies during fatigued motions would add significant insight to this theory and should be considered in future work. The increased knee internal rotation that was noted could contribute to increased valgus load on the knee joint and its associated risk for injury. The fatigue protocol mainly targeted the gluteus maximus for hip flexion and extension and gluteus medius for stabilize. The increased internal rotation might have occurred due to the decrease in muscle activation and contractility of the gluteus muscle group especially gluteus medius, how's function it is to abduct and medially rotate the hip. The decreased hip flexion angle is associated with the fact that fatigue decreases the ability to produce muscular activity as in a non-fatigued status, as well as decreased neuromuscular control and accumulation of lactic acid in muscles (Hill *et al.*, 1924). The present study associates the decrease in the muscles' ability to perform with the decision process to successfully complete the task presented by the simulation software (e.g., unanticipated), it may be plausible to assume that the combination of the central (decision process) and peripheral fatigue (muscles) may have contributed to the altered biomechanical patterns seen post-fatigue. McLean and associates have proposed that the central process is diminished with such fatigue models (McLean & Samorezov, 2009).

Previous researchers have reported altered movement patterns during post-fatigue; specifically increased knee abduction and internal rotation angles, increased hip rotation angles, hip internal rotation moments, and decreased knee flexion angles (Chappell, 2005; Mclean *et al.*, 2008; McLean *et al.*, 2007). The present study made similar findings. The reduced knee flexion angle at peak posterior ground reaction force has been theorized to be associated with increased anterior shear force (Yu Lin, & Garrett, 2006). This anterior shear force is thought to increase the load and strain in the anterior cruciate ligament and thus the probability of rupture.

There are four main systems that allow muscles to be fuelled and continue working; the ATP-CP or Phosphate system, anaerobic or fast glycolysis, aerobic or slow glycolysis, and beta oxidation (St Clair Gibson *et al.*, 2001). Short-term explosive movements, such as jumping, are predominantly governed by anaerobic bioenergetics, specifically the ATP-PC system. This system uses an ATP (Adenosine Triphosphate) molecule that undergoes hydrolysis (reaction using water) to form ADP (Adenosine Diphosphate), an inorganic phosphate molecule, a hydrogen ion, and energy. This breakdown of ATP is the driving force behind muscular contraction. While about 50-60% of the ATP stored in the muscle can be used up within 10 seconds (especially during high intensity exercise), there is now a massive concentration of hydrogen ions within the muscle, which is called exercise-induced metabolic acidosis (Fitts, 1994). When that specific pH level is exceeded, the muscle begins to malfunction, a burning sensation can be felt, fatigue is thus reached. Intracellular acidosis due mainly to lactic acid accumulation has been regarded as the most important cause of skeletal muscle fatigue, but studies have shown little direct effect of acidosis on muscle function at physiological temperatures (Bruton, Lännergren & Westerblad, 2013; Xia *et al.*, 2017). The role of reduced pH as an important cause of fatigue is now being challenged, and studies (Bruton *et al.*, 2013; Pate *et al.*, 1995) show that reduced pH may have little effect on contraction in mammalian muscle at physiological temperatures. Acidification has been considered to be an important factor behind the reduced shortening speed in fatigue however, using skinned rabbit muscle fibres, Pate *et al.* (1991) showed that acidification has little effect on the shortening speed at 30°C.

The fatigue protocol adopted may also be an important contributing factor because local (Nyland *et al.*, 1997) and general (Chappell, 2005) fatigue models have varying effects on proprioceptive control (Lattanzio *et al.*, 1997). The fatigue protocol implemented in the present study was aimed at inducing localized muscular fatigue, specifically of the extensor muscle group, and therefore only focused on the effects of neuromuscular fatigue on joint kinematic changes rather than general fatigue. No significant differences between pre- and post-fatigue states in any of the joint rotations were noted which indicate that fatigue did not influence the joint rotation angles. The most noticeable differences were seen in knee flexion-extension; dominant leg (26.68° ± 13.77°) compared to the non- dominant leg (35.58° ± 15.12°). There is thus a decrease of 11.85°. This supports the notion that knee flexion decreases in a fatigued state and that more injuries occur because of the knee in a less flexed position (Olsen *et al.*, 2004). Other noticeable difference occurred during hip adduction-abduction (6.33°) and ankle supination-pronation (-10.47°). However, hip internal rotation angle of the dominant leg, pre-fatigue (14.83° ± 16.77) show an increase post-fatigue (22.55° ± 28.08°). A similar increase in hip internal rotation is also noted in the non-dominant leg. These finding correlate with findings made by Alentorn-Geli *et al.* (2009) and Lucci, Corte, Van Lunen, Ringleb and Onate (2012).

Different muscle-activation strategies may be one of the reasons why no significant differences were found in the present study, specifically related to knee joint kinematics following a fatigued state. It is important to take cognizance of the fact that some muscles might have been more fatigued than others, whereby less fatigued muscles could compensate to maintain relatively normal joint kinematics (St Clair Gibson *et al.*, 2001). Further work is needed to identify muscle activation and kinematic strategies utilised during fatiguing exercise, and to determine whether allowing changes in joint kinematics, and thus encouraging kinematic variability, is beneficial to performance.

Given that the cutting speed was controlled for (e.g. 10-12 km·hr^{-1}) it may be true that moderate directional changes (i.e. cutting at 60^{o}) at such speeds were not enough of a challenge to the musculoskeletal system to induce noticeable effects. Conversely, in light of the small sample available for the present study one cannot neglect the fact that a type II error might have occurred (i.e. a false negative when indeed significant differences may present in the true population). This short-time frame to induce fatigue had the same effects than other long lasting protocols (e.g., decreased knee and hip flexion, and increased knee internal rotation) (Borotikar *et al.*, 2008). This raises two points: (i) the neuromuscular adaptations are dependent on the intensity of the exercise and duration of the activity; thus athletes may be injured earlier in the game/practice if played at a high level of intensity during short period, and (ii) training programs should focus on developing strategies to accommodate the biomechanical changes during fatiguing activities. An alternative explanation, perhaps our fatigue protocol, and other protocols are not soliciting the demands that are observed during an actual game (Chappell, 2005; Mclean *et al.*, 2008). Potentially, in order to develop a fatigue protocol that mimics the conditions of an actual game, it would be necessary to quantify the fatigue that athletes are experiencing when ACL tears occur. Even though both the anaerobic and aerobic energy system are taxed during a soccer game, we focused specifically on the anaerobic system and concluded that fatigue did not have a significant effect on hip (flexion-extension $p=0.495$, adduction-abduction $p=0.310$, internal-external rotation $p=0.996$) knee (flexion-extension $p=0.161$, adduction-abduction $p=0.280$, internal-external rotation $p=0.315$) or ankle (dorsiflexion-plantar flexion $p=0.922$, internal-external rotation $p=0.334$, supination-pronation $p=0.200$) rotation at initial contact. This study thus supports the notion that lactate does not have a significant effect on contractility of muscle fibres.

Research by Chidnok, DiMenna, Fulford, Bailey, Skiba, Vanhatalo and Jones (2013) has shown that the recovery rate of [PCr] is exponential, replenishing approximately 13-27% within 18-48 seconds. We needed to take this into account when planning the cutting trials. To account for such rapid [PCr] recovery kinetics two strategies were implemented in the present study; firstly, we minimized the 'rest' periods between cutting trials, and secondly, although we performed a total of five cutting trials per athlete, for the present study, only the first two cutting trials per leg were retained for complete analysis. The issue of speed may also be important since it is possible that a fatigued athlete may be moving slower when performing a cut. The decision in the current study to control for speed was made in order to eliminate this as a confounding factor in comparing pre- and post-fatigue mechanics. The average running speed was controlled for and set at between 10-12 $km \cdot hr^{-1}$ (i.e. jogging speed), and the average distance run for each trial would be approximately 6m from the starting gate to exiting through the finish gate. The time between gates for each trail ranged between 1.80-2.16 seconds, which is still within the domain of the anaerobic energy system.

It is also important to note that the cutting direction was always randomized to prevent changes in cutting mechanics due to directional anticipation. Studies have reported that unanticipated tasks, which do not allow an athlete to develop a motor plan prior to performing a movement, does indeed, alter joint mechanics differently to when the task is pre-planned (Besier & Lloyd, 2003; Collins *et al.*, 2016). The harmful effects associated with unanticipated tasks are thought to be due to the temporal constraints imposed on the components of the central nervous system which control movement. It appears that an athlete's risk of ACL injury is dependent on their ability to process information within the central nervous system and execute an effective motor response. During out cutting task, a participant had 0.2 seconds to react from breaking through the starting gates beam. Unanticipated tasks, which do not allow an athlete to develop a motor plan prior to initiating a movement, promote mechanics of the knee which could increase the risk of injury. These effects are thought to be due to the temporal constraints imposed on the components of the central nervous system which control movement, and it appears that an athlete's risk for injury is dependent on their ability to process information within the central nervous system and execute an effective motor response (Besier & Lloyd, 2003). Athletes must often perform unanticipated tasks when they are fatigued and thus randomization of the cutting direction (i.e. to the dominant or non-dominant side) was an important component of the protocol in the present study.

Proper foot contact with the force plate was also important to ensure and on some instances, athletes had to repeat certain trials if the foot placement on the force plate was not satisfactory. When trial repetition was necessary the fatigue protocol was re-initiated using the same criteria of (1) could no longer complete at least five squats unassisted, (2) could no longer meet the 90^o criteria or (3) could not sustain the 1 Hz pace set by the metronome to ensure that complete neuromuscular fatigue was indeed reached prior to commencing data collection. This is a pertinent point in that a trial was considered optimal if an athlete did not purposefully target the force plate (which might alter their cutting approach), and secondly, by repeating the fatigue protocol after some time it was ensured that subsequent trials would still occur in a fatigued state given that [PCr] recovery kinetics were exponential as previously stated.

It is important to note that muscle activation was not quantified during our fatigue protocol and therefore we cannot comment on the precise fatigue-induced changes that would have occurred. The present study made use of a similar fatigue protocol to Mclean 2008. They used EMG technology to determine fatigue in addition to the measures we implemented. Considering the fact that Mclean's study validated fatigue by EMG, the intensity and nature of our fatigue protocol, coupled with the strict criteria used to qualify fatigue, we are confident that such changes did occur and were an important contributor to our observed kinematic adaptations or lack there off. Nyland *et al.* (1997) showed that during the landing phase of a cutting manoeuvre selective medial lower-limb muscle groups are activated to counter external abduction loads that are simultaneously applied to the knee joint. Our fatigue model, involving continuous sets of unilateral Bulgarian split squats (BSS) coupled with unilateral Bulgarian jump trials, likely impacted directly on the contractility of specifically the knee extensor muscle group and therefore reducing their capacity to oppose such loads. The (BSS) is completed with one foot elevated, typically the back leg, and most of the body weight being carried by the front leg. The athlete descends by flexing the front leg at the knee and lowers the elevated knee towards the ground until the front leg is parallel with the ground. The athlete pushes through the heel and returns to the starting position. Muscle recruitment during the BSS is as follows; gluteus maximus, semitendinosus, semimembranosus, biceps femoris long head, and the adductor magnus is activated in the rear leg, while the front leg recruits the psoas major, iliacus, pectineus, rectus femoris, adductor brevis, adductor brevis, tensor fascia latae, sartorius, and the gracilis. Lower body movements such as running, jumping etc. are predominantly unilateral or single leg bearing (Konrardy, 2017). The theory then arises that to most effectively mimic game-like muscular fatigue, the fatigue protocol should mirror the mechanics and forces of the skills being performed (Phillips, 1997). A BSS elicits a force equal to about 85% of the total load on their lead foot, while in a conventional squatting, the total load is absorbed across both feet, meaning that both feet absorb around 50% of the total load (Konrardy, 2017).

In the presence of fatigue, knee motions and loads are present during cutting manoeuvres and may increase non-contact ACL injury risk. However, we found that fatigue had the above mentioned effect and do not have a notably significant effect on joint kinematics of healthy, uninjured male soccer players. Further research will need to be done however, before the precise mechanism of how fatigue manifests can be identified. Several aspects are still unknown, for example whether the fatigue protocol used in the present study caused changes in the central processing, the afferent sensory pathways, the efferent motor pathways, or a combination of the two. Differentiation of fatigue effects within each of these facets of control, although challenging, is not only crucial in terms of understanding its implications within the injury mechanism, but also with regard to injury prevention for the future.

This last point is particularly of interest because there is uncertainty regarding the fact that learned neuromuscular adaptations are maintained during fatigue (Lambert *et al.*, 2005). It may be possible however to counter the impact of fatigue on central processing, decision-making and the resultant movement response by including more complex, challenging decision-making tasks within neuromuscular training programs (Allen *et al.*, 2008). A differentiation should, however, be made between an increase in task complexity and that of an injury-causing event. Exposing individuals to movement tasks incorporating a combination of decision-making tasks and inducing fatigue will in theory then enhance the player's potential to perform safe and effective movement responses when exposed to such tasks within the game environment. Thought should be given to methods in which tasks could be integrated within an already challenging training schedule. These programs are designed to prevent the above injuries from occurring and should not be the cause thereof. Regardless, further exploration of this topic is needed.

In summary, kinematic and kinetic methods enable underlying rigid body motion to be successfully quantified by placing markers on the skin. The potential for error arises however from excessive skin-movement/perspiration on skin (dynamic sport posture). Several steps were taken to prevent such errors from occurring. First, a single qualified biokineticist placed markers on anatomic landmarks for all subjects, decreasing the potential for significant inter-tester errors that have been mentioned previously (Nikolaidis, Dellal, Torres-Luque & Ingebrigtsen, 2015). Although it is acknowledged that errors arising from excessive skin movement are unavoidable, especially in movements with a significant impact phase. It is for this reason that numerous trials were conducted in order to maximise the signal-to-noise ratio of the underlying kinematic data.

5.4 Time to peak stance lower limb joint rotation

Since the amount of knee flexion is linked to force absorption (which must be controlled by eccentric contraction of the quadriceps), then an increase in time might reflect an inability to adequately control force loading. Interestingly, the difference of differences for time to peak stance shows an increase of 3% between limb dominance and fatigue state for knee flexion, therefore hinting at a slightly longer time frame, although the difference did not reach statistical significance. This is interesting since both legs reached a similar amount of knee flexion (dominant=49.54^o ± 13.04^o vs. non-dominant=44.46^o ± 21.65^o, *p=0.129*), which implies that the increase in time is independent of the amount of knee flexion (i.e. greater peak flexion would require more time to achieve), and must therefore be related to the fatigue state. The difference of differences for time to peak stance also shows an increase for hip abduction (4%) and knee abduction (12%) between limb dominance and fatigue state. The increase in hip and knee abduction can also be attributed to fatigue, because the angles for the dominant and non-dominant leg were similar (hip abduction; dominant=-24.97^o ± 5.48^o vs. non-dominant=-21.72^o ± 10.74^o and knee abduction; dominant=11.17^o ± 12.96^o vs. non dominant=14.74^o ± 17.87^o). The differences, however, were not statistically significance; *p=0.502, p=0.442* and p=0.061.

Opposing the work of Condello *et al.* (2016), noticeable increases in peak knee abduction were evident after fatigue and as mentioned previously, this could be due to hip external rotator fatigue. We also found that peak knee flexion, abduction and internal rotation consistently occurred at the middle of stance, both pre and post fatigue which differ from the findings of (Condello *et al.*, 2016). The reason for especially peak knee flexion, occurring during middle of the stance, is because this is where the greatest amount of force is absorbed to minimize compression, torsion or shear forces on the knee, hip and ankle joint. The force absorption is achieved by eccentric contraction of the quadriceps muscle to slow down knee flexion. Recent research suggests that non-contact ACL injury occurs relatively early in stance (within the first 50 milliseconds) (Griffin *et al.*, 2008). If this is true, then it seems that the temporary peak abduction in the presence of fatigue/non-fatigue may be an important contributing factor in ACL injury risk. Other primary and secondary load peaks should be considered to gain more insight into potential injury mechanisms, especially those evident early in stance. The fatigue-induced coupling of the peak knee abduction moment and the relatively large internal rotation moment peak evident within the first 40% of stance, which are both known to induce reasonable ACL loading (Griffin *et al.*, 2008; Markolf *et al.*, 1995), and may present as a combination within the injury mechanism. Further investigation into the impact of neuromuscular fatigue on the temporal aspects of knee-joint loading and the risk of injury thus seems well warranted. In particular, a detailed focus on knee loading characteristics occurring in early stance may present with substantial benefit.

Significant differences were found between dominant and non-dominant leg as well as between fatigue and non-fatigue trials with regards to ankle pronation ($p=0.007$) and ankle external rotation ($p=0.033$). The increase in time to reach peak pronation (difference of differences 22%) and external rotation (difference of differences 12%) found during the present research study corresponds with the findings by Condello *et al.* (2016). They proposed that increased frontal-plane ankle rotations minimize energy transfer onto the knee joint and therefore it would decrease loading of the knee. Increased pronation of the foot may be due to decreased eccentric control of the tibialis anterior, tibialis posterior, but can also be due to forefoot, ligamentous laxity or tightness in the gastrocnemius and soleus muscles. During the fatigue protocol static BBS were alternated with BBS jump trails. This could cause the gastrocnemius and soleus muscle group to become overactive and because of the increased muscle stimuli they tighten decreasing ankle ROM. Ankle external rotation on the other hand, can be caused by a number of factors including:

a posterior tilt of the pelvis will orientate the hip joint outwards which may lead to externally rotated hips, knees and feet. Externally rotated ankles can also be caused by overactive muscles or tight muscles (e.g. hamstring, abdominals and gluteus maximus, posterior portion of gluteus medius, piriformis, obturator internus/externus, gemellus superior/inferior, gastrocnemius), weak/inhibited muscles (e.g. lumbar erectors and hip flexors, pectineus, anterior glute medius tensor fascia latae, adductor magnus and ankle pronation).

5.5 Ground reaction force data

Newton's second law of motion (law of acceleration) states that, if the upward push of the ground is equal to the earth's downward attraction (i.e., weight), the net force on the body equals zero and the resulting acceleration is zero. If the ground reaction force is greater than body weight, there is an net positive force acting on the body and the acceleration is positive. Finally, if the ground reaction force is less than body weight, the net force on the body is negative and the acceleration is negative. In addition Newton's third law of motion (law of reaction) states that for every action, there is an equal and opposite reaction (Thompson, 2002). Due to the gravity, we constantly maintain contact with the ground, and in this process, there occur interactions between the body and the ground. The reaction force supplied by the ground is specifically called the ground reaction force (GRF), which is basically the reaction to the force the body exerts on the ground (Ball, Stock & Scurr, 2010). However, in a more general case, the GRF will also have a component parallel to the ground (medio-lateral force/ horizontal (frictional) forces (Kwon, 1998).

Peak vertical forces were significantly different between the fatigued state compared to the non-fatigued state during the cutting manoeuvre (F= 23.51 N.kg^{-1}, *p= 0.035*). This might indicate that neuromuscular fatigue influence landing forces on impact during a directional change and thus correlates with McGovern *et al.* (2015); McLean *et al.* (2007) and Condello *et al.* (2016) findings. Prolonged exercise can lead to muscle fatigue, which will reduce the ability of posture control to counter for the impact with the ground at the touchdown phase. ACL injury during the cutting tasks is usually caused by the lack of proper management of a ground contact because of neuromuscular fatigue (McGovern *et al.*, 2015). One of the plausible explanations from the findings is that the body ulters its posture and makes adjustments to manage ground impact as muscle fatigue develops, but cannot sustain it in an exhausted state. Under pre-fatigue condition appropriate control of the landing posture is required as a protective behaviour in terms of maintaining the impact force, but this is not possible in a fatigued state.

Human and computer-generated models have shown that an increase in knee abduction and/or adduction loading contributes to an increase in ACL loading (Hewett *et al.*, 2005). Furthermore, Hewett *et al.* (2005) indicated that abduction motions combined with loading due to landing tasks may indeed predict ACL injury risk. A vitro study demonstrated that, knee valgus, varus, and internal rotation moments increase ACL loading, but their effects were significant only when an anterior shear force was present at the knee (Olsen *et al.*, 2004). Increased anterior shear forces at the knee due to a small knee flexion angle and increased compression forces on a posteriorly tilted tibial plateau are primary causes of anterior translation of the tibia relative to the femur. Although knee valgus/varus and internal rotation moments affect ACL loading when combined with significant anterior shear forces at the knee, current literature does not support them as primary ACL loading mechanisms relevant to ACL injuries (Schreiber *et al.*, 2012). However, further insight into the effects of frontal-plane knee and ACL loading during dynamic sports postures is warranted. Research by Smykalski (2016) also suggest that greater forces and joint moments are produced with greater speeds, however we cannot comment on this as we kept the speed at which the players performed the cutting task constant.

5.6 Limitations

As with every study performed, certain limitations arose. Firstly, a limited number of male participants who met the inclusion criteria to take part in the study were sourced, which ultimately resulted in a small representation of elite male soccer players.

Secondly, the fatigue protocol implemented in this study only produced neuromuscular fatigue and did not induce general/central fatigue. During a soccer game a player does not only experience one type of fatigue and all of the factors together might have an effect on lower limb joint kinematics.

Thirdly, even though we did not make use of an EMG or measure the lactic acid levels in the blood to determine fatigue, we are confident that muscle fatigue was reached due to the task failure achieved by following the validated protocol by Borotikar *et al.* (2008).

Lastly, designated running speed of 10-12 km·hr^{-1} (i.e. jogging speed) were controlled for, which may or may not be representative of the speed at which ligament injury occurs in soccer. Unfortunately, no data were available to represent the actual running speeds at which ligament injury occurs in sport and experimental data is unavailable, so we can only speculate. More importantly, however, athletes perform running and cutting manoeuvres from slow to fast speeds with or without ligament injury.

5.7 Practical applications

- Coaching staff and researchers are encouraged to use the study as a template for future research. Coaches will have comparative values to evaluate their players and researchers would have a standardized fatigue protocol that has been validated.
- The current study can be used as a baseline in future research to investigate the effects of fatigue and leg dominance on lower limb joint kinematics. The results can be compared to participants with a previously injured limb to identify the effects thereof.
- Coaches will be able to use the data as a baseline value and evaluate players' risk for injury, as well as evaluating players post-rehabilitation to determine if the player is fit to return to the sport.

- Differences in the execution of a COD task are related to the performance demands in soccer and the ability to perform sharp COD movements could provide benefits in terms of spatial and temporal advantages over an opponent. Training should consider addressing the enhancement of the ability to perform CODs and technical skills with both legs, to increase the unpredictability of the athletes during their actions.

5.8 Future directions

Injury rate and severity of injuries as well as the mechanisms thereof have been investigated extensively, but more insight is needed into the effect of fatigue on players on the turning mechanism and change of direction tasks. Several other questions remain unanswered, such as how knee kinematics change during high-intensity cutting manoeuvres, whether fatigue or leg dominance plays a significant role and whether changes in joint kinematics can be effectively detected using 3D video analysis. We propose the fatigue that soccer players experience during a game should be quantified and a fatigue protocol should be developed, that mimics the conditions of an actual soccer game. This may produce more accurate results in terms of the effect of fatigue and leg dominance on the joint kinematics of a player.

5.9 Conclusion

The present study set out to examine the combined effects of leg dominance and fatigue on lower-limb biomechanics during a sub-maximal 60° cutting manoeuvre. ACL loading is primarily caused by the loading experienced at the knee joint and failure to include the combined dynamic lower-limb system within analyses may compromise the understanding of how fatigue occurs within the complete injury mechanism, thus limiting successful application of screening and prevention strategies. Within the literature there are a variety of fatigue models that have been implemented to investigate the impact of fatigue on both sensory and neuromuscular function. General loading fatigue models employ drills that may more effectively replicate actual game play and unlike localized models, directly affect limb proprioception. However, the most effective way to study the exact manifestations of fatigue remains unclear. Our protocol impacted largely the neuromuscular system, defining fatigue based on volitional exhaustion, among other parameters. One of the key findings of the present study related to limb dominance was that no statistically significant differences were evident for any of the dependent variables (limb dominance; fatigue state) related to the independent variables (i.e. joint [hip, knee, ankle], contact time, ground reaction however, between-subject fatigue variations that was large enough could negatively impact the biomechanical data comparisons. Subjects of similar (University soccer players) fitness levels where chosen, even though it was a small sample size and they were submitted to a standardized fatigue protocol. They were objectively assessed for fatigue according to specific criteria. The fact that post-fatigue three dimensional joint measures were similar between dominant and non-dominant legs also suggests that consistency was achieved in peripheral neuromuscular fatigue, at least as it relates to performance. Currently isolated fatigue models and general fatigue models do not show a significant change in joint biomechanics, thus our assumption of comparable fatigue effects at a muscular level remains largely speculative. Future work should address this limitation by developing standardized tasks that potentially targets specific locations of fatigue within a general fatigue paradigm. It is also possible that participants experienced varying levels of central and/or sensory fatigue, which may also have extensive effects on the results attained. Further assessment of the potential contributions of central fatigue to the ACL injury mechanism is also suggested. Finally, although we are confident that participants were fatigued to a reasonable level, the rapid deterioration observed in the effects of fatigue suggests that a more effective protocol of maintaining levels of fatigue for each repetition should be investigated. Others have examined changes in landing biomechanics cumulatively by quantifying these parameters in parallel with the progression of fatigue. Such an approach largely negates the above concern and seems to provide a viable basis for future research.

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Appendix A: Terminology

Fatigue: Fatigue can be defined as a decrease in force production, or an inability to regenerate the original force during the presence of an increased perception of effort (Strauss, Jacobs & Berg, 2012). Strauss *et al.* (2012) stated that fatigue either has a 'peripheral' or 'central' origin.

Opto-electronic: The use of video or digital cameras to capture images of an individual performing a Function Movement Screen or any movement (Nordin and Frankel 2012:428). For the purpose of this study, it refers to the Vicon and Biomechanics of Human Movement Video analysis program collectively.

Soccer: This sport, also called football, is a game played between two teams consisting of eleven players. The objective of the game is to propel a ball into the opposition's goal post by any means possible, except by the use of the player's arms (Rahnama *et al.*, 2002).

Degrees of freedom (DoF): A number of independent variable factors affecting the range of states in which a system may exist, in particular any of the directions in which independent motion can occur (Keselman, Algina & Kowalchuk, 2001).

Range of motion (ROM): The full movement potential of a joint, measured in degrees (Ross *et al.*, 2004).

Kinematics: The branch of mechanics concerned with the motion of objects without reference to the forces which cause the motion (Patla *et al.*, 1991).

Kinetics: The branch of classical mechanics that is concerned with the relationship between motion and its causes, specifically, forces and torques (Patla *et al.*, 1991).

Appendix B: Anthropometric measures

Leg length: measured from the anterior superior iliac spine to the medial malleolus in a supine position.

Knee width: the distance between the medial and lateral femoral epicondyles which is measured using a large sliding calliper.

Ankle width: the distance between the medial and lateral malleoli.

Shoulder offset: measured from the base of the acromio-clavicular joint to the mid-point of the articulation of the humeral head with the glenoid cavity.

Elbow width: the distance between the medial and lateral humeral epicondyles.

Wrist width: the distance between the anterior and posterior aspects of the wrist at the level of the distal radial head and ulnar styloid process.

Hand thickness: the distance between the dorsum and palmar surface of the midpoint of the hand.

Appendix C: Marker Placement

Head markers

LFHD	Left front head	Approximately over the left temple
RFHD	Right front head	Approximately over the right temple
LBHD	Left back head	Placed on the back of the head, roughly in the horizontal plane of the front head markers
RBHD	Right back head	Placed on the back of the head, roughly in the horizontal plane of the front head markers

Torso markers

C7	7th cervical vertebra	Spinous process of the 7th cervical vertebra
T10	10th thoracic vertebra	Spinous process of the 10th thoracic vertebra
CLAV	Clavicle	Jugular notch where the clavicles meet the sternum
STRN	Sternum	Xiphoid process of the sternum
RBAK	Right back	Placed in the middle of the right scapula. This marker has no symmetrical marker on the left side. This asymmetry helps the auto labelling routine determine right from left on the subject.

Arm markers

LSHO	Left shoulder marker	Placed on the left acromio-clavicular joint

LUPA	Left upper arm marker	Placed on the upper arm between the elbow and shoulder markers. Placed one marker width lower than the midpoint between the LSHO and LELB markers
LELB	Left elbow	Placed on the lateral epicondyle approximating elbow joint axis
LFRA	Left forearm marker	Placed on the lower arm between the wrist and elbow markers. Should be placed one marker width lower than the midpoint between the LWRA and LELB markers
LWRA	Left wrist marker A	Left wrist bar, thumb side
LWRB	Left wrist marker B	Left wrist bar, pinkie side
LFIN	Left fingers	On the dorsum of the hand just below the head of the second metacarpal

The above markers are placed on corresponding sites on the right arm. Markers LUPA and LFAM are placed one marker width above the middle of their points, equidistant from the reference markers used to locate the mid-point of the arm and forearm. For the corresponding right-hand side, the markers are placed one marker space below the midpoint between the points of reference.

Lower body

LASI	Left ASIS	Placed directly over the left anterior superior iliac spine
RASI	Right ASIS	Placed directly over the right anterior superior iliac spine
LPSI	Left PSIS	Placed directly over the left posterior superior iliac spine
RPSI	Right PSIS	Placed directly over the right posterior superior iliac spine
LKNE	Left Knee	Placed on lateral epicondyle of the left knee

LTHI	Left thigh	Midway between the LKNE and LASI markers
LANK	Left ankle	Placed on the lateral malleolus along the imaginary line that passes through the trans-malleolar axis
LTIB	Left tibia marker	Placed midway between the LKNE and LANK markers
LTOE	Left toe	Placed over the second metatarsal head, on the mid-foot side of the equinus break between the fore-foot and mid-foot
LHEE	Left heel	Placed on the calcaneus at the same height above the plantar surface of the foot as the toe marker

(Vicon, 2016:1).

For all markers on the left side of the body, a corresponding marker is placed on the right hand side. Whenever an asymmetrical placement is required, the right is always placed higher relative to left.

Appendix D: Information and informed consent form

RESEARCHER'S DETAILS	
Title of the research project	Comparison of leg dominance and fatigue state on lower extremity kinematics during cutting manoeuvres in male soccer players
Reference number	
Principal investigator	**Madeleine Nienaber**
Address	**14 Cornelia Avenue, Framesby Port Elizabeth**
Postal Code	**6045**
Contact telephone number (private numbers not advisable)	**076 194 1764**

A. DECLARATION BY OR ON BEHALF OF PARTICIPANT		**Initial**
I, the participant and the undersigned		
ID number		
OR		
Address (of participant)		

A.1 HEREBY CONFIRM AS FOLLOWS:		**Initial**
I, the participant, was invited to participate in the above-mentioned research project		
that is being undertaken by	Madeleine Nienaber	
From the	Human Movement Science Department	
of the Nelson Mandela University.		

THE FOLLOWING ASPECTS HAVE BEEN EXPLAINED TO ME, THE PARTICIPANT:			**Initial**
2.1	**Aim:**	The investigators are studying the comparison of the kinematics of the knee joint derived from 3D video analysis to that of 3D opto-electric motion analysis in South African male soccer players during 60° cutting tasks.	

		The information will be applied to firstly, use 3D video analysis to determine the kinematics of the knee joint during all-out running and how these kinematics change as the athlete begins to fatigue. Secondly, this information will be compared to the 3D kinematic data obtained from a 3D motion analysis system that will be recording simultaneously to determine whether differences between 3D video analysis and 3D motion analysis are present and the extent of these differences.		
2.2	**Procedures:**	I understand that the participants will have three-dimensional knee joint kinematic and ground reaction force data recorded throughout the execution of 60^{o} cutting manoeuvers, prior to and immediately after being exposed to a protocol designed to induce general neuromuscular fatigue (Mclean *et al.*, 2008). Prior to testing, participants will be allotted time to warm-up, consisting of cycling and self-directed stretching. The fatigue protocol will consist of continuous sets of five double leg squats between eight jump trials. Subjects will continue to alternate between the squat sequence and jump landing tasks until maximal fatigue is attained, which is defined as the point where they can no longer complete three complete squats unassisted.		
2.3	**Risks:**	The possibility of injury exists, but will be minimised as far possible by ensuring that the participant warms up and performs a number of practice trials before the testing commences.		
2.4	**Possible benefits:**	As a result of my participation in this study, an alternative method to motion analysis might be validated. This will serve to create a more feasible alternative for joint kinematic analyses in a private practice setting. The participant will be taught as much as possible in the process, and get exposure to research in a university setting.		
2.5	**Confidentiality:**	My identity will not be revealed in any discussion, description or scientific publications by the investigators. Results from this study will be coded and only the researcher will have access to the names of the participants. Full anonymity and confidentiality will be maintained at all times.		
2.6	**Access to findings:**	Should you be interested in finding out on advancements in this research, you may contact the principal researcher. Please see contact details above.		

2.7	**Future use of information:**	I hereby consent that my data/results may be used for future research.			
2.8	**Voluntary participation / refusal / discontinuation:**	My participation is voluntary.	**YES**	**NO**	
		My decision whether or not to participate will in no way affect my present or future care/employment/lifestyle.	**TRUE**	**FALSE**	

3.	**THE INFORMATION ABOVE WAS EXPLAINED TO ME/THE PARTICIPANT BY:**							**Initial**
Madeleine Nienaber								
in	**Afrikaans**		**English**		**Xhosa**		**Other**	
and I am in command of this language, **or** it was satisfactorily translated to me by								
(name of translator)								
I was given the opportunity to ask questions and all these questions were answered satisfactorily.								

4.	No pressure was exerted on me to consent to participation and I understand that I may withdraw at any stage without penalisation.	

5.	Participation in this study will not result in any additional cost to myself.	

A.2 I HEREBY VOLUNTARILY CONSENT TO PARTICIPATE IN THE ABOVE-MENTIONED PROJECT:			
Signed/confirmed at		on	20
Signature or right thumb print of participant	Signature of witness:		
	Full name of witness:		

<table>
<tr><th colspan="8">STATEMENT BY OR ON BEHALF OF INVESTIGATOR(S)</th></tr>
<tr><td>I,</td><td colspan="3">(name of interviewer)</td><td colspan="4">declare that:</td></tr>
<tr><td rowspan="2">1.</td><td colspan="3">I have explained the information given in this document to</td><td colspan="4">(name of patient/participant)</td></tr>
<tr><td colspan="3">and/or his representative</td><td colspan="4">(name of representative)</td></tr>
<tr><td>2.</td><td colspan="7">He was encouraged and given ample time to ask me any questions;</td></tr>
<tr><td rowspan="3">3.</td><td>This conversation was conducted in</td><td>Afrikaans</td><td></td><td>English</td><td></td><td>Xhosa</td><td></td></tr>
<tr><td colspan="7">And no translator was used <u>OR</u> this conversation was translated into Other</td></tr>
<tr><td>(language)</td><td colspan="2">by</td><td colspan="4">(name of translator)</td></tr>
<tr><td>4.</td><td colspan="3">I have detached Section D and handed it to the participant</td><td colspan="2">YES</td><td colspan="2">NO</td></tr>
<tr><td colspan="4">Signed/confirmed at</td><td colspan="4">on 20</td></tr>
<tr><td colspan="3" rowspan="2">Signature of interviewer</td><td colspan="5">Signature of witness:</td></tr>
<tr><td colspan="5">Full name of witness:</td></tr>
</table>

Appendix E: Application for approval NMU research ethics committee (Human)

APPLICATION FOR APPROVAL
NMMU RESEARCH ETHICS COMMITTEE (HUMAN)

<u>SECTION A:</u> (To be filled in by a representative from the Faculty RTI Committee)					
Application reference code:	**H** **HUMAN**	.... **YEAR**	.... **FACULTY**	.... **DEPARTMENT**	.... **NUMBER**
Resolution of FRTI Committee:	☐ **Ethics approval given (for noting by the REC-H)** ☐ **Referred to REC-H for consideration** (if referred to REC-H, electronic copy of application documents to be emailed to Imtiaz.Khan@nmmu.ac.za)				
Resolution date:					
Faculty RTI representative signature:					

1. GENERAL PARTICULARS	
TITLE OF STUDY	
a) Concise descriptive title of study (must contain key words that best describe the study): **Comparison of leg dominance and fatigue state on lower extremity kinematics during cutting manoeuvres in male soccer players**	
PRIMARY RESPONSIBLE PERSON (PRP)	
b) Name of PRP (must be member of permanent staff. Usually the supervisor in the case of students): **Mark Kramer Building 125, Room 0119, HMS Department, NMU**	
c) Contact number/s of PRP: **041 504 4630**	
d) Affiliation of PRP: Faculty **Health Sciences Specify here, if "other"** Department (or equivalent): Human Movement Science	
PRINCIPLE INVESTIGATORS AND CO-WORKERS	
e) Name and affiliation of principal investigator (PI) / researcher (may be same as PRP): **Madeleine Nienaber** Gender: **Female**	
f) Name(s) and affiliation(s) of all co workers (e.g. co-investigator/assistant researchers/supervisor/co-supervisor/promoter/co-promoter). If names are not yet known, state the affiliations of the groups they will be drawn from, e.g. Interns/M-students, etc. and the number of persons involved: **N/A**	
STUDY DETAILS	
g) Scope of study: **Local**	h) If for degree purposes: **Master's**
i) Funding : **NMU Research Grant** Additional information (e.g. source of funds or how combined funding is split) **Not applicable**	
j) Are there any restrictions or conditions attached to publication and/or presentation of the study results? **No**	

If YES, elaborate (Any restrictions or conditions contained in contracts must be made available to the Committee):
Not applicable

k) Date of commencement of data collection: **2017/08/01**
Anticipated date of completion of study: **2 years**

l) Objectives of the study (the major objective(s) / Grand Tour questions are to be stated briefly and clearly):
Firstly, to use 3D video analysis to determine the sagittal kinematics of the knee joint during 60° cutting and how these kinematics change during a fatigued state. Secondly, the sagittal 3D kinematic data obtained from a 3D motion analysis system that will be recorded simultaneously will be compared to determine whether differences between 3D video analysis and 3D motion analysis are present.

m) Rationale for this study: briefly (300 words or less) describe the background to this study i.e. why are you doing this particular piece of work. A few (no more than 5) key scientific references may be included:
The frequency of soccer injuries is estimated to be approximately 10 to 35 per 1000 playing hours with the majority of injuries occurring in the lower extremities; mainly the knees and ankles (Dvorak & Junge, 2000). Most injuries occur in the second half of a game indicating that they may be mediated by fatigue (Gioftsidou, Malliou, Pafis, Beneka, Tsapralis, Sofokleous, Kouli, & Godolias, 2012). To prevent such injuries, rehabilitation specialists propose specific exercise programmes that incorporate strengthening exercises to restore muscle imbalances, stretching exercises to increase muscle flexibility, and balance exercises to improve proprioception (Gioftsidou et al., 2012). Soccer is one of the most popular sports and attracts many participants all over the world; the high number of participants may account of the substantial quantity of musculoskeletal injuries (Dvorak & Junge, 2000; Junge & Dvorak, 2004). Furthermore, soccer is a semi-contact sport associated with an increased prevalence of predominantly lower extremity injuries, with estimates that most players will obtain at least one performance inhibiting injury per year (Dvorak & Junge, 2000). Fatigue is an extrinsic factor affecting the musculoskeletal and neurologic systems and is associated with decreased knee proprioception and increased joint laxity compared to baseline values (Chappell, 2005). Altered biomechanical patterns and an associated decline in muscle performance and decision making ability have been noted in soccer players post-fatigue (Cortes, Quammen, Lucci, Greska, & Onate, 2012). The extent to which the biomechanical factors related to knee joint motion change in response to fatigue has been studied typically by the implementation of pre-fatiguing protocols (Cortes et al., 2012) through the use of sophisticated lab-based 3D motion analysis systems. Several questions however remain unanswered, such as how sagittal plane knee kinematics change during 60 degree cutting tasks, whether fatigue plays a predominant role, and whether changes in knee kinematics can be as effectively detected using 3D video analysis. Motion analysis systems such as Vicon, are regarded as the gold standard of motion analysis, but require significant financial resources, maintenance and space. If more feasible alternatives such as 3D video analysis can be used in to acquire similar data with an acceptable level of accuracy at a fraction of the cost, the barrier to motion analysis in a variety of settings can be substantially decreased. The biomechanical analysis of human motion has broad applications for performance enhancement and injury prevention not only in soccer, but in almost all sporting codes; therefore it is hoped that the findings of the present study can be used to guide future studies in the methods of motion analysis and making such analyses available to a broader audience.

METHODOLOGY

n) Briefly state the methodology (specifically the procedure in which human subjects will be participating) (the full protocol is to be included as *Appendix 1*):
The study will be conducted under a quantitative approach and therefore a quasi-experimental study design will be utilised to sample the soccer population for this study. The population of interest for the present study pertains to South African university soccer players based in Port Elizabeth. Due to the non-randomisation of the quasi-experimental design, non-probability sampling, which is a type of sampling whereby subjects are selected non-randomly based on specific criteria (Landreneau, 2010), will be

utilised. Purposive sampling, a specific form of non-probability sampling, will be used to obtain the required participants for the present study. Participants will have three-dimensional knee joint kinematic and ground reaction force data recorded throughout the execution of 60 degree cutting manoeuvers, prior to and immediately after being exposed to a protocol designed to induce general neuromuscular fatigue (Mclean, Borotikar, Newcomer, Koppes, & Mclean, 2008) . Prior to testing, participants will be allotted time to warm-up, consisting of cycling for five minutes on a stationary cycle ergometer followed by self-directed stretching. The fatigue protocol will consist of continuous sets of five double leg squats between eight jump trials. Intra and inter subject variations in the squat tasks will be minimized by having subjects squat at a consistent frequency (1 Hz) and enforcing that the thighs finish parallel with the ground at the end of each squat. Subjects will continue to alternate between squat sequence and jump landing tasks until maximal fatigue is attained, which is defined as the point where they can no longer complete three complete squats unassisted (McLean, et al., 2008).

o) State the minimum and maximum number of participants involved (Minimum number should reflect the number of participants necessary to make the study viable)
Min: **10** Max: **20**

2. RISKS AND BENEFITS OF THIS STUDY

a) Is there any risk of harm, embarrassment or offence, however slight or temporary, to the participant, third parties or to the community at large? **Yes**
If YES, state each risk, and for each risk state i) whether the risk is reversible, ii) whether there are alternative procedures available and iii) whether there are remedial measures available.
There may be a risk for delayed onset of muscle soreness (DOMS) due to the fatigue protocol to which the participant may be unaccustomed. Delayed onset of muscle soreness of completely reversible within 24-36 hours. Unfortunately the purpose of the protocol is to enduce fatigue, and the stipulated protocol is the least arduous of the available neuromuscular fatigue protocols. In terms of remedial measures, participants will be encourage to engage in dynamic stretching prior to the activity, and static stretching once the protocol has been completed.

b) Has the person administering the project previous experience with the particular risk factors involved? **Yes**
If YES, please specify: **The primary researcher is a intern biokineticist and has experience with injuries and the treatment thereof**

c) Are any benefits expected to accrue to the participant (e.g. improved health, mental state, financial etc.)? **Yes**
If YES, please specify the benefits: As a result of participation in this study an alternative method to motion analysis might be validated. This will serve to create a more feasible alternative for sagittal knee joint kinematic analyses in a private practice setting. Participants will also gain insights into potential injury mechanics due to muscular fatigue.

d) Will you be using equipment of any sort? **Yes**
If YES, please specify: Weighing scale, Stadiometer, GeneActive tri-axial accelerometers, Vicon Motion Analysis System, Biomechanics of Human Movement Video analysis program, and Kistler force plates

e) Will any article of property, personal or cultural be collected in the course of the project? **No**
If YES, please specify: **Not applicable**

3. TARGET PARTICIPANT GROUP

a) If particular characteristics of any kind are required in the target group (e.g. age, cultural derivation, background, physical characteristics, disease status etc.) please specify: All male soccer players representing the NMU soccer club at a competitive level for at least one year will be able to participate. Players will only be allowed to participate if they have signed a voluntary informed consent form. Players must be between the ages of 18-25 years and free of injury.

b) Are participants drawn from NMMU students? **Yes**

c) If participants are drawn from specific groups of NMMU students, please specify: **NMU soccer club players**
d) Are participants drawn from a school population? **No** If YES, please specify: **Not applicable**
e) If participants are drawn from an institutional population (e.g. hospital, prison, mental institution), please specify: **Not applicable**
f) If any records will be consulted for information, please specify the source of records: **Not applicable**
g) Will each individual participant know his/her records are being consulted? **Not applicable** If YES, state how these records will be obtained: **Not applicable**
h) Are all participants over 18 years of age? **Yes** If NO, state justification for inclusion of minors in study: **Not applicable**

4. CONSENT OF PARTICIPANTS
a) Is consent to be given in writing? **Yes** If YES, include the consent form with this application [Appendix 2]. If NO, state reasons why written consent is not appropriate in this study. **Not applicable**
b) Are any participant(s) subject to legal restrictions preventing them from giving effective informed consent? **No** If YES, please justify: **Not applicable**
c) Do any participant(s) operate in an institutional environment, which may cast doubt on the voluntary aspect of consent? **No** If YES, state what special precautions will be taken to obtain a legally effective informed consent: **Not applicable**
d) Will participants receive remuneration for their participation? **No** If YES, justify and state on what basis the remuneration is calculated, and how the veracity of the information can be guaranteed. **Not applicable**
e) Which gatekeeper will be approached for initial permission to gain access to the target group? (e.g. principal, nursing manager, chairperson of school governing body) **NMU Soccer Club coach**
f) Do you require consent of an institutional authority for this study? (e.g. Department of Education, Department of Health) **No** If YES, specify: **Not applicable**

5. INFORMATION TO PARTICIPANTS
a) What information will be offered to the participant before he/she consents to participate? (Attach written information given as [*Appendix 3*] and any oral information given as [*Appendix 4*])
b) Who will provide this information to the participant? (Give name and role) **Madeleine Nienaber** **Primary researcher**
c) Will the information provided be complete and accurate? **Yes** If NO, describe the nature and extent of the deception involved and explain the rationale for the necessity of this deception: **Not applicable**

6. PRIVACY, ANONYMITY AND CONFIDENTIALITY OF DATA
a) Will the participant be identified by name in your research? **No** If YES, justify: **Not applicable**
b) Are provisions made to protect participant's rights to privacy and anonymity and to preserve confidentiality with respect to data? **Yes** If NO, justify. If YES, specify: **All participant-specific information will be coded; the codes will be known only to the primary research; all information will be stored in a secure location to which only the primary**

researcher has access.

c) If mechanical methods of observation be are to be used (e.g. one-way mirrors, recordings, videos etc.), will participant's consent to such methods be obtained? **Yes**
If NO, justify: **Not applicable**

d) Will data collected be stored in any way? **Yes**
If YES, please specify: (i) By whom? (ii) How many copies? (iii) For how long? (iv) For what reasons? (v) How will participant's anonymity be protected? **(i) by the primary researcher, (ii) 3 copies, (iii) maximum of 5 years, (iv) to allow for follow up studies or should the participant wish to gain access for future reference – consent will be sought from the participant for this, (v) all information relevant to the participant will be coded and stored in a secture location to which only the primary researcher has access.**

e) Will stored data be made available for re-use? **Yes**
If YES, how will participant's consent be obtained for such re-usage? **Provision will be made in the informed consent form as well as verbally.**

f) Will any part of the project be conducted on private property (including shopping centres)? **No**
If YES, specify and state how consent of property owner is to be obtained: **Not applicable**

g) Are there any contractual secrecy or confidentiality constraints on this data? **No**
If YES, specify: **Not applicable**

7. FEEDBACK

a) Will feedback be given to participants? **Yes**
If YES, specify whether feedback will be written, oral or by other means and describe how this is to be given (e.g. to each individual immediately after participation, to each participant after the entire project is completed, to all participants in a group setting, etc.): **Feedback will be given in both verbal and written format once the project is completed (due to the data analysis process being involved and complex).**

b) If you are working in a school or other institutional setting, will you be providing teachers, school authorities or equivalent a copy of your results? **Yes**
If YES, specify, if NO, motivate: **The athletes and coaches will receive a copy of the results which may benefit the coaching and training of the athletes.**

8. ETHICAL AND LEGAL ASPECTS

The Declaration of Helsinki (2000) or the Belmont Report will be included in the references: **Yes**
If NO, motivate: **Not applicable**
(A copy of the Belmont Report is available at the following link for reference purposes: http://www.nmmu.ac.za/documents/rcd/The%20Belmont%20Report.pdf)

a) I would like the REC-H to take note of the following additional information:
None

9. DECLARATION

If any changes are made to the above arrangements or procedures, I will bring these to the attention of the Research Ethics Committee (Human). I have read, understood and will comply with the *Guidelines for Ethical Conduct in Research and Education at the Nelson Mandela Metropolitan University* and have taken cognisance of the availability (on-line) of the Medical Research Council Guidelines on Ethics for Research (http://www.sahealthinfo.org/ethics/).
All participants are aware of any potential health hazards or risks associated with this study.
I am not aware of potential conflict(s) of interest which should be considered by the Committee.
If affirmative, specify: **Not applicable**

	03 March 2019
SIGNATURE: **Madeleine Nienaber** (Primary Responsible Person)	Date
	03 March 2019
SIGNATURE: **Mark Kramer** (Principal Investigator/Researcher)	Date

10. SCRUTINY BY FACULTY AND INTRA-FACULTY ACADEMIC UNIT

This study has been discussed, and is supported, at Faculty and Departmental (or equivalent) level. This is attested to by the signature below of a Faculty (e.g. RTI) and Departmental (e.g. HoD) representative, neither of whom may be a previous signator.

NAME and CAPACITY (e.g. HoD)	SIGNATURE	Date
NAME and CAPACITY (e.g. Chair:FacRTI)	SIGNATURE	Date

11. APPENDICES

In order to expedite the processing of this application, please ensure that all the required information, as specified below, is attached to your application. Examples of some of these documents can be found on the Research Ethics webpage (http://www.nmmu.ac.za/default.asp?id=4619&bhcp=1). You are not compelled to use the documents which have been provided as examples – they are made available as a convenience to those who do not already have them available.

APPENDIX 1: Research methodology

Attach the full protocol and methodology to this application, as "Appendix 1" and include the data collection instrument e.g. questionnaire if applicable.

APPENDIX 2: Informed consent form

If no written consent is required, motivate at 4a). The intention is that you make sure you have covered all the aspects of informed consent as applicable to your work.

APPENDIX 3: Written information given to participant prior to participation

Attach as "Appendix 3". The intention is that you make sure you have covered all the aspects of written information to be supplied to participants, as applicable to your work.

APPENDIX 4: Oral information given to participant prior to participation

If applicable, attach the required information to your application, as "Appendix 4".

APPENDIX 5, 6, 7: Institutional permissions

Attach any institutional permissions required to carry out the research e.g. Department of Education permission for research carried out in schools.