THE EFFECTS OF HEADING MOTION AND GENDER ON LOWER EXTREMITY BIOMECHANICS IN SOCCER PLAYERS

A Thesis Presented for the
Master of Science
Degree
The University of Tennessee, Knoxville

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May 2018

ACKNOWLEDGEMENT

I would like to acknowledge and thank all the people who supported me and I worked with throughout my Master's degree. Firstly, I would like to express my gratitude to my advisor Dr. Joshua Weinhandl for his efforts to guide me, support me, and improve my knowledge either in classes or projects. For my study, I have been honored and grateful to work with him who has assisted me throughout the stages of my study until I have fulfilled it successfully. Also, I would like to thank my committee members, Dr. Songning Zhang and Dr. Eugene C. Fitzhugh, for their time to guide me and improve my study ideas and structure. Further, I want to thank all my lab colleagues, Master and PhD students, who have supported me during my study. Finally, I would not forget to thank my parents, siblings, and wife who have encouraged me in order to succeed and earn my Master degree.

ABSTRACT

Soccer continues to experience growth internationally in both men and women. Thus, it's reasonable to predict a high rate of injuries amongst those athletes since soccer includes strong physical contact. ACL injury is one of the most common injuries among athletes, and (70-84%) of ACL injuries occurs in non-contact situations, which often are associated with landing, deceleration, and sudden change of movement direction, and these situations might end with knee valgus collapse which's a common mechanism of ACL injury. Further, there are biomechanical differences of lower extremity joints in various tasks with both genders, which apparently means that the task utilized plays a crucial influence on joints dynamics. In soccer, the heading motion is one of the most common activities, which is more likely associated with landing. The purpose of this study was to investigate the effect of forward heading motion by comparing the kinetics and kinematics of lower extremity between genders during 1) a stop-jump task and 2) a jump-heading task. 10 male and 10 female soccer players performed 5 stop-jumps and 5 jumps with heading a soccer ball. Findings displayed a combination of reduced initial knee flexion, greater initial hip flexion, greater peak vertical GRF, and greater peak knee extension moments with both genders during the jump-heading task. Additionally, greater peak knee abduction angles were observed among female players during both tasks. These findings suggest that a higher risk of non-contact ACL injuries might occur during the jump-heading task compared with the stop-jump task, and this risk could be higher among female players than male players. These findings might lead soccer trainers to consider different strategies and techniques of landing after heading motions in game-like conditions. Also, PTs and rehabilitation designers who work with ACL-injured patients might consider the outcome of this study. Future studies should consider to examine ankle dynamics in addition to knee and hip in game-like conditions.

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LIST OF ABBREVIATIONS

ACL	Anterior cruciate ligament
PCL	Posterior cruciate ligament
MCL	Medial cruciate ligament
LCL	Lateral cruciate ligament
GRF	Ground reaction force
JH	Jump-heading task
SJ	Stop-jump task
N	Newton
BW	Body weight
kg	kilogram
m	Meter
º	Degree
yrs	Years

CHAPTER I:

INTRODUCTION

BACKGROUND

Soccer is considered one of the most popular sports in the world, if it is not the most popular one [1]. A survey conducted by the International Federation of Association Football (FIFA) found that the number of international participants of soccer has been growing every year. In 2006, there were 265 million active soccer players around the world, and 10% of this number were women [2]. From 2000 to 2006, in addition, it was declared that the number of registered male and female soccer players had increased by around 38.3 million players; female soccer players increased by 54% to 4.1 million players, and male soccer players increased as well by 21% to 34.2 million players. That means that soccer continues to experience growth internationally in both men and women [2]. For this reason, it is reasonable to predict a high rate of injuries amongst those athletes since soccer is a violent sport and includes physical contact which might lead to injuries. In 2014, The National Electronic Injury Surveillance System (NEISS) collected injury data from the emergency departments of 96 hospitals across the United States and reported more than 3.5 million injuries occurred in 25 sports [3]. Approximately 200,000 injuries were related to soccer between the ages 4 and 64 yrs old, and approximately 80% of those injuries occurred amongst players who were younger than 25 years old [3].

Many factors influence the incidence rate of injuries among athletes, such as environmental (changeable), neuromuscular (changeable), anatomical (unchangeable), and hormonal factors (unchangeable). Also, those factors tend to impact both genders differently [4-10]. According to the literature some of the kinematical and kinetical characteristics vary between male and female athletes during various sport activities [11-21]. During jump-landing,

stop-jump, and sidestepping tasks, greater hip flexion angle and knee flexion angle have been observed in males than females [16-19]. On the other hand, it has been reported that females had greater peak hip flexion than males during landing of jump-landing tasks [11]. Furthermore, Weinhandl et al. [18] reported that females land with smaller initial hip abduction compared to males, which might result in a greater knee abduction and higher ACL stress. Conversely, it has been reported that males show a greater hip abduction angle during the phase and landing phases of stop-jump tasks compared to females [22]. Other studies have found no significant differences between males and females in terms of hip abduction angles during peak anterior tibial shear force and peak hip abduction angle [13, 23]. Concerning the knee, there is an obvious agreement that females experience greater initial and peak knee abduction angles compared to males during landing of sidestepping and jump-landing tasks, which might lead to valgus collapse of the knee [12, 15, 16, 24]. In terms of the ankle, there is insufficient information in the literature; some studies have reported that females display a greater ankle pronation (eversion) than males during landing [15, 25], yet Mclean et al. [16] and Weinhandl et al. [18] did not observe any significant differences between males and females during landing. Therefore, it can be stated that some of those differences between male and female athletes might play an important role in the occurrence of injuries.

One of the major concerns amongst athletes is injuries, which might affect their athletic careers negatively, specifically anterior cruciate ligament injury (ACL). ACL injury has been recognized as one of the most common injuries amongst athletes and the most frequent acute ligament injury of the knee [26, 27] In the United States alone there are roughly 120,000 people who experience ACL injury annually, and there are around 211 ACL cases per each 100,000 people in Australia [26, 28]. Moreover, ACL injury has been prevalent in various sports, such as

soccer, volleyball, basketball, handball, and rugby [29]. According to the clinical files of 17,397 patients, ACL injury constituted the highest percentage of all internal knee injuries (20.3%). Moreover, Majewski et al. [30] reported that soccer accounted for 36.7% of all ACL injuries during a 10 year epidemiological study. When adjusted for athlete exposures, soccer still exhibits one of the highest ACL injury rates at 0.291 injuries per 1000 athlete exposures [31]. Importantly, around 70 to 84% of all ACL injuries in both female and male athletes happen in non-contact situations (without physical contact), and those situations often are characterized by landing, deceleration, and sudden change the direction of movement [32-34]. Due to the anatomical, hormonal, neuromuscular, and biomechanical differences between females and males, Hewett et al. [35] suggested that female athletes are at 4 to 6 times higher risk of ACL injury than male athletes during jumping, cutting, and pivoting tasks. Among female athletes, furthermore, the risk of ACL injury has reached 10 times among high schools' athletes and 5 times among colleges' athletes in the past 30 years [14]. Furthermore, Beynnon et al. [31] reported that women had an ACL injury rate of 0.391 compared to the injury rate of 0.198 in men. Conversely, it has been reported that the majority of ACL injuries occur among male athletes compared to female athletes [36], which might be due to the fact males are more exposed to athletic tasks that might put them at high risk of ACL injury. However, it seems that female athletes tend to experience ACL injuries more often than male athletes due to the biomechanical differences, which will be displayed thoroughly in the literature review.

Soccer players preform different dynamic tasks during playing, such as sprinting, cutting, jumping, zigzagging, and heading, which might place high stress on the ACL that could lead to rupture. To date, according to the literature, most of those tasks have been analyzed comprehensively to determine their association with known ACL injury risk factors in both

genders [12, 13, 15-18, 25, 37]. Nevertheless, the influence of heading motion on the biomechanical characteristics of lower extremity in soccer has not been investigated sufficiently. Butler et al. [11] conducted a study to investigate the mechanical differences between genders and two landings: a jump-landing task and a stop-jump task followed by heading a ball; It was found that peak hip flexion and peak knee flexion were significantly greater during the jump-landing task compared with the jump-heading task; However, the ground reaction force GRF was significantly greater during the jump-heading task [11]. Between genders, females showed significantly greater peak hip flexion and lower GRF compared to males in both tasks, which might be because men landed more stiffly than women. However, males were found to have greater peak knee extension moments and peak plantarflexion moments compared to females even though both male and female participants had similar ranges of motion at those joints, and these variables were normalized to the maximal jump height of each subject [11]. That might be due to the greater muscle strength and joint stiffness among men in comparison with women [11]. This study shows that the heading motion influences the mechanical characteristics of the lower extremity during landing, which might lead to increased injuries; also, these characteristics vary between both genders, which may increase the risk of injury in one gender more than the other. However, no one has studied the effects of heading motion on lower extremity mechanics in terms of the frontal plane.

STATEMENT OF THE PROBLEM

To date, there has been only one study which has investigated the effect of heading motion on the sagittal plane mechanics of the lower extremity in soccer players during landing. Specifically, this study did not investigate the effect of forward heading motion during landing on the mechanical features of the lower extremity in terms of the frontal plane, which is

absolutely important since many studies reported that increased knee abduction and hip adduction is considered one of the mechanisms of ACL injury [15, 16, 18, 23, 24, 38].

STATEMENT OF PURPOSE

Therefore, this study aims to compare the biomechanical characteristics of the hip and knee in the dominant leg of female and male soccer players during two jump tasks: 1) a stop-jump task and 2) a jump-heading task.

RESEARCH HYPOTHESES

1. It is hypothesized that both male and female subjects would have reduced peak/initial hip flexion angles, reduced peak/initial knee flexion angles, greater peak/initial hip adduction angles, and greater peak/initial knee abduction angles during the jump-heading task compared with the stop-jump task.
2. It is hypothesized that both male and female subjects would have a greater peak vertical GRF, peak hip extension moment, peak knee extension moment, peak hip abduction moment, and peak knee adduction moment during the jump-heading task compared with the stop-jump task.
3. It is hypothesized that males, compared to females, would have a greater vertical GRF, peak hip extension moment, peak knee extension moment, peak hip abduction moment, peak knee adduction moment, peak/initial hip flexion angle, peak/initial knee flexion angle; and reduced peak/initial hip adduction angle, and peak/initial knee abduction angle during both jump tasks.

**The moments are internal moments.

DELIMITATIONS

The exclusion criteria for both female and male subjects are:

- A history of any major lower extremity injuries (such as ACL injury) or surgeries.
- A history of any lower extremity injuries within the past 6 months, which might reduce their typical participation.
- Reporting any pain while playing soccer (a game or training) or while performing activities of daily living, such as walking or ascending/descending stairs.
- Reporting any cardiovascular disease or any risk factor that prohibits preforming aerobic exercises. For this reason, if a subject chooses "Yes" on any question of the Physical Activity Readiness Survey [39], he or she will be excluded from the study unless there is a written consent from the doctor verifying that the subject is healthy enough to participate in this study.

The inclusion criteria for the subjects:

- Men and women have to be between 18-30 years old.
- They have to be current field soccer players at a recreational level (average 1-2 hours per week).
- They have had at least 2 years of playing experience on a high school or college soccer team.
- They have to be able to perform specific soccer tasks: stop-jump task and jump-heading task.
- They have to be within the normal range of body mass index (BMI = 18-24.9 kg/m^2) [40].

- They have to score at least 71 out of 80 in Lower Extremity Functional Score (LEFS) [41].

LIMITATIONS

- The study will be conducted in a room (laboratory) not on a real soccer field.
- Skin marker artifacts might reduce the accuracy of the captured motion data.
- Subjects will use standardized running shoes which may impact lower extremity biomechanics during landing since soccer players use specific soccer shoes while playing.
- Landing on a force platform might affect the mechanics of landing since landing on a turf surface is softer and more cushioned.
- The isolation of a specific soccer task (jump-heading task) might limit the external validity since multiple tasks happen sequentially during real playing.

SIGNIFICANCE OF THE STUDY

The study could provide important information to the literature with regard to the influence of heading motion on the biomechanical characteristics of lower extremity joints during landing in soccer players. Further, since most of ACL injuries occur during landing in non-contact situations, and high knee abduction is considered one of ACL injury mechanisms, it is valuable to see how the sagittal and frontal kinetics and kinematics of hip, and knee are impacted after heading motion in both genders. The findings of this study could improve the fundamental knowledge, which is relative to the relation of heading motion in soccer and high load applied on knee, since there is insufficient information in this area. Also, this study might be beneficial to researchers who are interested in the differences between jump tasks and gender during landing in terms of the biomechanical features of the lower extremity. Moreover, this

study might provide insight for soccer trainers and physical therapists regarding the lower extremity of players at risk for injuries during landing after heading motion. This might help soccer trainers to think about different strategies and techniques of heading motion in games and practices. Also, rehabilitation designers might be able to create rehabilitation programs in order to reduce the risk of potential re-injury.

OPERATIONAL DEFINITION OF TERMS

- Kinetic variables: The variables that represent the forces acting on a body during motion, such as ground reaction force GRF, moment of force, impulse, work, and power.
- Kinematic variables: The variables that describe the motion of a body without regard to the forces that produce the motion, such as time, position, displacement, velocity, and acceleration.
- Non-contact ACL injury: The rupture of ACL that occurs without extrinsic contact by another player or object on the field.
- Moment of force: The force that tends to rotate an object about an axis. In this study internal joint moments of force are defined as the net rotational effect of agonist and antagonist muscle forces about a joint.
- Jump-landing task: It is a jump from the ground or a box followed by landing on force platforms using both legs.
- Stop-jump task: It is two steps or more followed by jumping using both legs and then landing on both legs.
- Sidestepping task: It is a jogging or running followed by sudden changing of direction once a leg strikes a force platform.

- Jump-heading task: It is a stop-jump followed by heading a ball then landing on both legs.
- Drop-landing task: It is a landing after dropping from a specific height.
- Drop-jump task: It is a dropping from a specific height followed by a vertical jump upon landing.

CHAPTER II:

LITERATURE REVIEW

INTRODUCTION

This study aims to investigate the effect of heading motion on the biomechanical characteristics of the hip, knee, and ankle in the dominant leg of female and male soccer players in terms of the sagittal and frontal plane. That is because only one study demonstrated the effect of heading motion on soccer players in terms of the sagittal plane even though the frontal plane biomechanical characteristics of the lower extremity joints are associated with the most common mechanisms of non-contact ACL injuries. This chapter, therefore, will provide an overview of knee joint anatomy including the joint capsule, cartilages, and ligaments. Also, ACL injury will be demonstrated thoroughly in terms of the mechanisms, the incidence rate amongst genders and sports, and the risk factors. Eventually, the kinematics and kinetics of the hip, knee, and ankle for both men and women will be displayed and compared in various jumping tasks in terms of the sagittal, frontal, and transverse plane.

KNEE JOINT ANATOMY

The knee is one of the most complex joints of body since it has diverse anatomical structures working together in order to achieve complex and specific functions [42]. These structures involve bony elements such as the femur, tibia, patella, and fibula; and soft tissue elements like ligaments, tendons, cartilages, muscles, and the joint capsule. Moreover, the knee is constructed of three joints which function as a unit [42]. These joints are the tibiofemoral, patellofemoral, and proximal tibiofibular joints [42]. Since the knee is responsible for carrying a large portion of our body weight during daily activities, it has 4 ligaments to provide passive stability in all directions [42]. Because of this, the knee allows a wide range of motion in flexion-

extension, adduction-abduction, and internal-external rotation, and allows 6 degrees of freedom [42].

KNEE JOINT CAPSULE

The medial compartment of knee has less flexion-extension and internal-external rotation excursion since it is strongly stable by the function of the two strongest ligaments, the posterior cruciate ligament (PCL) and the medial collateral ligament (MCL), and is thus the stable knee compartment [42, 43]. In addition, the lateral compartment is called the "mobile knee compartment due to the fact that the axis of rotation is based on the medial compartment," and the lateral compartment does not have distinct ligaments which directly connect the tibia to the femur. This allows more flexion-extension and internal-external rotation compared to the medial compartment [42].

KNEE JOINT CARTILAGE

The medial and lateral menisci, with their capsular attachments, play an important role in force control and load absorption [42]. Both of the menisci are located around the intercondylar eminence, and are fixed mediolaterally to the peripheral capsule and anterioposteriorly to the tibial plateau [42, 43]. Also, they have some fiber connections with the ACL and PCL; and some other connections with transverse ligaments [42]. The posterior horn of medial meniscus provides co-restraint to the ACL and PCL as well [42]. Both medial and lateral menisci move base on the femoral condyle and tibial plateau movement whether it is anteriorly or posteriorly, and they move during all the knee motions [42]. Importantly, the most movable meniscus is the lateral meniscus compared with the lateral one, and that is the reason behind the increased prevalence of medial meniscus tears compared to the lateral meniscus. On the other hand, the lateral meniscectomy requires longer recovery time than the medial meniscectomy [42].

KNEE JOINT LIGAMENTS

The ligaments are anterior cruciate ligament ACL, posterior cruciate ligament PCL, medial lateral ligament MCL, and lateral collateral ligament (LCL) [43, 44]. The ACL is inserted in the posterior part of the inner surface of the lateral femoral condyle of femur in order to control the anterior translation of tibia against femur [42, 43]. Its attachment length is around 13-25 mm with a width of 6-13 mm, it is much wider at the femoral and tibial insertions than the midpoint, and it runs from the femur anteriorly, medially, and distally to the tibia [42, 43]. In addition, the ACL has two functional main bundles: the anteromedial (AM) and posterolateral (PL) [42, 43]. Looking to a cross-section of the ACL, it can be seen that the ACL has irregular shape rather than circular shape since the shape changes based on the angle of flexion [42, 43]. Also, the orientation of the AM and PL bundles change once the knee flexes, and that is due to the fact that the AM bundle tends to be tighter during knee flexion, and the PL bundle tends to be tighter during knee extension and internal rotation [42-44].

The PCL is larger in length and width than the ACL[42, 44]. Moreover, the PCL is considered the strongest ligament of the knee, and is responsible of controlling the posterior tibial translation and external tibial rotation [42-44]. It is constructed of multiple bundles, similar to the ACL. Most of the previous studies confirmed that the PCL has two bundles: the anterolateral (AL) and the posteromedial (PM) bundle which provide stability to the knee through the full range of flexion [43].

The LCL begins from the lateral epicondyle of femur and ends to the head of fibula; the proximal tibiofibular joint plays an important role to adapt the length of the ligament during flexion –extension [42, 43]. Since the biceps femoris tendon and the LCL run closely at the insertion place (fibular head), once the biceps femoris contracts, the LCL is actively tightened

[42, 43]. With regard to the MCL, it runs from the medial epicondyle of femur to the lateral aspect of tibial tuberosity, which plays an important role to constrain knee abduction and external rotation, as well as medial and lateral translation of the tibia [43, 44]. The MCL is tightened in extension and external rotation; and loose in flexion and internal rotation [42, 44].

THE ANTERIOR CRUCIATE LIGAMENT INJURY

ACL rupture has been known as one of the most frequent injuries amongst athletes [45, 46]. In the United States, around 120,000 people experience ACL injury annually, and around 211 cases per 100,000 people occur every year in Australia [26, 28]. In addition, ACL rupture has been widespread in different sports, such as soccer, volleyball, basketball, and handball [29]. ACL injury amongst athletes are more common in non-contact situations during competition than perturbed situations by other opponents, and these non-contact situations are often characterized by landing, rapid deceleration, and sudden change in movement direction [37, 47, 48].

In order to enable athletes after ACL injury to return to sports at competitive levels, they are highly recommended to undergo the surgical intervention, ACL reconstruction, which requires a long period of recovery to reestablish the muscle strength around the knee, the knee mechanics, and the knee stability [45, 46]. However, for some reason almost 80% of athletes who underwent ACL reconstruction could not return to the same level of intensity and competition of sports [49]. In addition, it was reported that from 50% to 80% of female soccer players and 41% of male soccer players would be at risk of developing degenerative changes, pain, and abnormal laxity in the injured knee after having ACL reconstruction [50]. However, the reconstruction surgery is the only solution to preserve the knee mechanics and stability [46, 51, 52].

ACL INJURY EPIDEMIOLOGY

It is common for athletes to suffer an ACL injury in non-contact situations or without physical contact to other players, and previous studies showed that 70 to 84% of all ACL tears are non-contact ACL injuries in both female and male athletes [4, 8]. Further, these non-contact situations ACL injuries often are associated with change in direction or cutting with deceleration, landing in or near full knee extension, or pivoting with knee close to full extension [4, 8, 53, 54]. Also, around 80% of ACL injuries in team handball occurred during cutting and landing activities [55]. Anterior shear force at the knee, which is particularly generated at knee flexion angles around 20-30º, might be the most critical force associated with ACL injury. The risk of ACL injury will also increase if there is a combination of forces in addition to the anterior shear force so that the ACL might be strained if it is subjected to an integration of anterior shear force and external abduction or internal rotation torques [4, 8, 53]. Even though researchers have not agreed on whether female athletes have greater knee flexion than male athletes during landing and cutting tasks [14], some recent studies confirmed that valgus collapse, in addition to high knee external rotation and foot plantarflexion, during deceleration tasks is the common mechanism of ACL injury amongst female athletes since this mechanism produces greater anterior tibial translation and knee joint loads [8, 14, 53]. Also, decreased hip extension and greater hip adduction in women during landing tasks have been identified as important predictors of high ACL injury risk [14].

GENDER DIFFERENCES IN ACL INJURY RATES

Due to the anatomical, hormonal, neuromuscular, and biomechanical variations between males and females, it is logical to state that there are differences in the risk of ACL injury between males and females [4, 8, 14, 53]. It was demonstrated that female athletes are at 4 to 6

times higher risk of ACL injury than male athletes participating in jumping, cutting, and pivoting [35, 56, 57]. Also, the risk of ACL injury has increased 10-fold in high schools and 5-fold in colleges among female athletes in the past 30 years, and this difference of injury risk is basically caused by the mechanical differences of lower extremity among males and females [14]. Moreover, previous studies reported a 2-3 female-to-male ACL injury ratio [56, 58, 59]. However, another study found a 7-9 female-to-male ACL injury ratio [60]. Also, a recent study reported that female players were four years younger than male players when experiencing the ACL injury [61]. Further, according two studies from the Nation Collegiate Athletic Association Injury Surveillance System (NCAA-ISS) in America, female athletes experienced 6% of all match injuries and 2% of all training injuries, but male athletes had less than 1% of all injuries [62]. On the other hand, it was demonstrated that males account for the plurality of ACL injuries compared to females due to the greater exposure of males to athletic tasks[36]. Also, a national population-based study, which included ACL injury data of 238,488 claims from 2000-2006, found that males had a greater rate of the ACL injury in most all ages compared to females [36], which might be due to the larger sample size of males in the study.

ACL INJURY RATES BY SPORT

Being successful and professional is a compulsion which is pervading amongst athletes of all sports over the world. So much so that they often strive to attain their goals and obtain the peak no matter the obstacles they encounter. The obstacles which have drawn widespread anxiety among all athletes are injuries. According to an analysis study done by investigating the clinical documents of 17,397 patients, the internal knee injuries accounted 44.8%, and ACL injury had the highest percentage (20.3%) among the other injuries [30]. Interestingly, even though females participation in athletics has increased prominently in recent years, this study

showed that males had more ACL cases than females (66.4% and 33.6% respectively) [30], which might be due to the fact that males are more exposed to athletic tasks that predispose them to the high risk of ACL injury. Also, it was found that the most ACL injuries occurred in soccer (36.7%) skiing (33.7%) handball (8.1%) volleyball (3%) and basketball (2.1%) [30]. The previous percentages could be logical since soccer is the most widespread sport in the world based on the number of the international participants that reached 265 million active players in 2006 and has been increasing every year. Furthermore, the range of ACL injuries amongst soccer players during official games and training time was from 0.06 to 3.7 injuries in each 1,000 hours of active soccer competition [4].

ACL INJURY RISK FACTORS

ENVIRONMENTAL RISK FACTORS

The most important environmental factors are the weather and the playing surface since they directly influence the traction or friction coefficient of the shoe-surface interface in many athletic events [5]. Since players must be able to start moving rapidly, and change direction suddenly and sharply, their feet must be strong and stable enough on the surface. However, too firm of a foot fixation to the surface might lead to "footlock," which contributes to knee and ankle trauma, and poor foot fixation causes slipping, which results in player-surface contact trauma [5]. Moreover, greater traction leads to faster running and sharper cutting angles, which produces greater stress on the lower extremity joints, especially the knee [5, 63].

With regard to the playing surface, it was found that the surface with high density of grass coverage, such as artificial turf, leads to a higher friction coefficient and a greater ground reaction force, and these forces increase while the artificial turf ages, which subsequently increases the risk of getting ACL injury [4, 8]. In addition, previous studies reported that high

shoe-surface friction led to around two-thirds of the non-contact soccer injuries [63]. Epidemiologically, Powell and Schootman [64] investigated the injury incidence on AstroTurf (artificial) and natural grass, and they found that playing on AstroTurf shows significant increase in the number of knee sprains, especially ACL and MCL sprains compared with playing on the natural grass [64]. Furthermore, it was demonstrated that playing soccer indoor places the players at risk of knee and ankle injuries 6 times more often than playing soccer outdoor because both the artificial turf and indoor courts have high coefficient of friction compared with the natural grass and indoor courts [65].

Since the dry and warm weather increases the friction coefficient and torsional resistance during playing, high rate of ACL tears was observed on natural grass [4, 8]. Further, Torg et al. [66] explained that an increase in turf temperature increases the shoe-surface interface friction, which might place the knee and ankle at risk of injury. Similarly, Orchard et al. [67] confirmed that the risk of knee and ankle injury would be lower in cold weather, including ACL tears on both natural and artificial turf. In addition, footwear modulates and support foot fixation while playing so it is also considered a potential risk factor for ACL tears since it was found that the number, length, and cleat placement were related to the probability of suffering an ACL injury [4].

ANATOMICAL RISK FACTORS

Since the anatomy is not changeable, it is so difficult to find some preventive ways to reduce the risk of anatomical factors [4]. Also, there is no prominent evidence confirming that the high rate of non-contact ACL injury is caused by anatomical differences with regard to age and gender [7]. However, there are some anatomical considerations that should be discussed thoroughly since they may lead to ACL injury [4].

Body mass index (BMI) has been shown in some studies as a crucial factor in ACL injuries, particularly amongst female adolescent soccer players and collegiate recreational athletes [7, 8] . Also, it was demonstrated that a high BMI is associated with a more extended lower extremity position and less knee flexion during landing of a stop-jump task [7, 8]. On the other hand, some other studies found no impact of the BMI on ACL injures in female athletes, which directly conflicts with the results mentioned previously [4, 68].

Joint laxity is considered a risk factor that might lead to high risk of ACL injury [7]. Based on a soccer study, the risk of leg injuries is significantly related to the joint laxity and knee hyperextension which had been noticed with female soccer players [69], and another study stated that military cadets with increased joint laxity are at 2.8 times greater risk of non-contact ACL injury compared with normal cadets [70]. Moreover, it was found that knee joint laxity has been associated with increased knee adduction-abduction and internal-external rotation, which might lead to valgus collapse, one of the most common ACL injury mechanisms [24, 71]. Also, healthy females showed greater knee joint laxity compared with healthy males, and that might contribute to the higher risk of ACL injury in females [70, 72].

It was recognized that some of the structural properties of the knee anatomy vary between males and females [7, 8]. Precisely, females have a smaller intercondylar notch than males, which predisposes the ACL to be hit by the notch during knee rotation, and may explain why female athletes are more likely to suffer an ACL injury than male athletes [7, 53, 70]. Further, it was found that the length, volume, and mass of ACLs in females were smaller compared with males. Also, the percentage of collagen fibrils is lower in the ligaments of females compared with males, which reduces the stiffness of the ligaments [4, 73].

Q-angle is one of the anatomical factors that has been suggested to be related to a high risk of ACL injury [4]. The Q-angle is known as the angle produced by drawing a line form the anterior-superior iliac spine (ASIS) to the central patella and a second line from the central patella to tibial tubercle [4, 7]. A great Q-angle may influence the lower extremity biomechanics and then leads to a high stress on the knee [74, 75]. Also, it was found that female basketball players who suffered knee injuries showed greater Q-angle compared with non-injured players [76]. However, other studies confirmed that Q-angle is not a prominent risk factor of either knee abduction angle or ACL injury risk [77, 78].

Since the knee is not an isolated joint, and is related to other joints and segments during movement including the trunk, pelvis, hip, and ankle, the mechanics of these joints might increase the risk of ACL injury [4, 7]. With regard to the pelvis, anterior pelvic tilt changes the position of hip to be internally rotated and flexed, which in turn causes weakness and excessive stretching of the hamstrings and that affects the moment arms of the muscles around hip, especially gluteal muscles [79]. One of the hamstring muscles functions is to prevent genu recurvatum (hyperextension) and anterior tibial translation, and the gluteal muscles assist hip flexion and prevent valgus collapse, which means that the weakness of these muscles might place high stress on the ACL during activates [79]. Also, it is common among ACL injured subjects to have genu recurvatum and ankle pronation compared to non-ACL injured subjects [80].

HORMONAL RISK FACTORS

The relation of ACL injury and hormones remains inconclusive [4, 8]. In any case, it was found that ACL cells have estrogen and progesterone receptors, and these hormones impact the tensile properties of ligaments, as well as the entire strength, aerobic capacity, anaerobic

capacity, and endurance of female athletes [81-83]. On the other hand, other studies confirmed that there is no significant difference in the maximum force, stiffness, and energy of the ACLs as hormone level changes [84]. Based on a review paper of 9 studies, similarly, 6 studies found that the anterior knee laxity of females was not significantly affected by the menstrual cycle, while 3 studies found that there is a significant relation between the menstrual cycle and anterior knee laxity [85]. Based on the literature, the menstrual cycle has three phases: follicular (day 0-9), ovulatory (day 10-14), and luteal (day 15-28), and these phases have different percentages of estrogen production [4, 8]. For this reason, it was reported that the period of menstrual cycle having the highest estrogen surge, which is called pre-ovulatory phase, showed more ACL injuries among female athletes [86]. Hicks-Little et al. [87], in addition, found that the ovulation and luteal phases of the menstrual cycle showed a significant increase of knee anterior translation, which places high stress on the ACL.

NEUROMUSCULAR RISK FACTORS

Neuromuscular control refers to the unconscious contraction of the muscles that are located around joints in response to a sensory stimulation, so the muscles must work coordinately and be co-activated to protect the joint [6, 10, 53]. This technique of unconscious activation of the muscles is essential during daily activities and sports, especially when people suddenly change direction, and perform cutting or pivoting movement [8, 9, 53]. With regard to the knee, the co-contraction of hamstrings and quadriceps plays an important role in reducing knee motion and loads, which might elevate the risk of ACL injury [10]. Hence, efficient hamstrings activation decreases the ACL loads produced by the quadriceps and provides more stability to the knee joint [10, 88, 89]. Colby et al. [90] found that the maximal activation of quadriceps muscle occurred at the moment before foot strike during multiple activities: sidestep

cutting, cross-cutting, stopping, and landing with healthy collegiate male and female athletes. Also, the same study reported that there was a submaximal hamstring activation at and just after the heel strike. This delay in hamstrings activation, along with the decreased hamstrings-to-quadriceps activation ratio might cause a significant increase of the knee anterior translation. Further, it was observed that female soccer, basketball, and volleyball players showed greater quadriceps and soleus activation and decreased hamstrings activation during hopping and stop-jump landing task compared with male players [91, 92]. Moreover, it was recognized that there is an imbalance of agonist-antagonist muscles strength, flexibility, and coordination within the lower extremity of women [93, 94], which decreases stability and places high stress on the joints, particularly the knee joint.

Since effective neuromuscular control might help athletes protect their joints by absorbing the ground reaction force through active muscular restraints [14], muscular fatigue and weakness surrounding the knee might increase the risk of non-contact ACL injury [95]. During landing, muscular fatigue causes significant increases in hip extension, hip internal rotation, peak knee abduction, knee internal rotation, and ankle supination angle at initial contact, and that might be more obvious during single-leg landing task compared with double-leg landing task [96]. During sport competitions, some studies reported that soccer and rugby players are at increased risk of injury in the second half of games [24, 97]. During soccer matches, specifically, it was observed that more non-contact injuries occurred during the last third (15 min) of the first half and the last two thirds (30 min) of the second half [98, 99], which suggests that the muscular fatigue induces some kinematic and kinetic alterations of the lower extremity joints, which might increase the risk of ACL injury [100].

LOWER EXTREMITY JOINTS BIOMECHANICS AND ACL INJURY MECHANISMS

Since contact ACL injuries configure only 30% of all ACL injuries, it is more beneficial to focus on the non-contact injuries, which are considered the most dangerous injuries in terms of ACL injuries [4, 14]. These injuries often include the following conditions: landing with near to or full knee extension, sudden change in the movement direction, and deceleration [4, 14] since these conditions cause some mechanical alterations in the lower extremity joints, which might raise the strain of the ACL and lead to rupture. Thus, it is important to demonstrate the biomechanical risk factors of ACL injury in term of the kinetics and kinematics of each joint of the lower extremity in all the planes in order to give a comprehensive insight into the mechanical differences between males and females during particular tasks and help rehabilitation specialists to offer effective prevention programs of the ACL injury.

KINEMATICS

SAGITTAL PLANE

Many studies conducted reported kinematical differences in the sagittal plane between males and females in regard to trunk, hip, knee, and ankle during various sport tasks. The rule of thumb is that the more joints are flexed during landing, the more energy is absorbed by the muscles of a lower limb, the less anterior tibia shear force will be generated, and lower loads will be applied to the ACL [4, 16]. Concerning the trunk, Blackburn and Padua [101, 102] demonstrated that active trunk flexion during landing assists the hips and knees attain more flexion compared to more a erect/extended posture, which potentially reduces the ACL loading and injury risk. Conversely, the same study found that landing with leaning forward posture (flexed trunk) has no significant impact on the frontal and transverse plane kinematics of the knee and hip. Furthermore, a previous study investigated the effect of single-leg landing with

various sagittal plane body positions on the dynamics of lower extremity [48]. They found that landing with upright body position increases quadriceps muscle activation and decrease the knee flexion angle. On the other hand, landing with leaning forward position showed smaller vertical GRF, smaller quadriceps muscle activation, and greater knee flexion angle, but these variables were not significantly different between males and females [48]. For this reason, landing with a leaning forward strategy is considered more ACL protective, and landing with upright position is considered more ACL harmful for both genders [48, 101, 102].

In term of the hip, forty-four participants who play volleyball and basketball for recreation were examined in a study to determine the biomechanical differences between males and females during a jump-landing task [18]. It was found that both males and females landed with similar hip flexion angles regardless of landing height [18]. On the contrary, another study investigating volleyball players in simulated spike and block landings demonstrated that male players showed significantly greater hip flexion angle than female players at 40 cm block landing task [17]. Conversely, the peak hip flexion was greater with females than males during jump-landing task [11]. A comparison of three jump landing tasks confirmed that forward vertical jump task is the most demanding task compared to drop-landing and drop-jump task since it requires greater energy in order to effectively absorb the increased forces and complete the task successfully [13]. Further, hip flexion was significantly greater during forward vertical jump task compared to the other tasks. In term of a stop-jump task, Yin et al. [19] and Yu et al. [20] found that females showed significantly greater hip extension at the initial contact of landing. Moreover, McLean et al. [16] reported that females showed greater peak and initial hip extension than males during three tasks: sidestepping, jump-landing, and shuttle running, and these angles

were based upon the task, which means that the task applied might play an important role in the sagittal plane motion of the lower extremity joints.

The kinematical influences of the knee are often similar to the hip in many tasks. Thus, it was found that the peak knee extension during landing of volleyball players was significantly greater with females compared to males, and the peak ankle dorsiflexion was significantly greater with females as well compared to males [18]. Whereas, males relied on the knee musculature as the primary energy absorbers, while females used the ankle musculature as the primary energy absorbers [15, 18]. Also, Salci et al. [17] demonstrated that volleyball female players showed lower knee flexion than male players in landing from a 40 cm spike-landing task. However, the 60 cm landing height for both spike and block task displayed similar knee flexion angles among males and females [17]. This study further reported that both males and females landed with similar ankle plantarflexion. During a stop-jump task, it was found that males landed with greater initial knee flexion than females, which significantly decreases the load placed on the knee and reduces the risk of the ACL injury [19, 20]. Similarly, a study investigated different tasks and reported that the peak knee flexion was greater with males than females during sidestepping, jump-landing, and shuttle running, but it did not observe any differences with regard to the ankle [16]. Further, females during landing showed smaller knee flexion angular velocity, which may increase the load on the ligament and does not allow enough time for adjacent joints and muscles to absorb the high impact forces [4]. In addition, it was found that subjects with a history of ACL injury landed with smaller knee flexion angle compared to healthy subjects, which puts ACL reconstructed patients at risk of re-injury or injuring the contralateral ACL [4, 38]. Furthermore, youth female recreational soccer players tend to land

during stop-jump task with decreased initial knee and hip flexion angles compared to male players [21].

FRONTAL PLANE

Frontal plane hip angles might be critical predictors of the impact force at the knee [38]. It was reported that healthy active males and females respond differently during landing from various heights [18]. Females showed significantly greater peak and initial hip adduction during landing compared to males [18], which results in increased frontal plane knee loading and then high potential for ACL injury [38]. In addition, Cruz et al. [13] found that hip adduction angle has no significant difference between three landing tasks: drop-landing, drop-jump, and forward vertical jump. During a stop jump task, females prepared for the landing with greater hip abduction compared to males [22], which is opposite to the findings of Jacobs et al. [23], who demonstrated that healthy females landed with lower hip abduction compared to healthy males.

The biomechanical characteristics of the knee in the frontal plane are considered the most important predictors for ACL injury since it was found that knee valgus collapse is usually associated with rupture of the ACL [4, 38]. Kernozek et al. [15] and Ford et al. [24] reported that females had significantly greater peak knee abduction angles compared to males during the first 30-50% of the landing phase. This females place greater strain on the ACL during landing since it was found that ACL load might be six times greater at 5º of knee abduction [103]. In addition, it was demonstrated that female athletes had significant side-to-side differnces during landing compared to male athletes in term of the peak knee abduction angle [24]. Specifically, female athletes showed greater peak knee abduction angles with the dominant leg compared to the non-dominant leg, which may predispose female athletes to higher risk of non-contcat ACL injuries [24]. Moreover, it was reported that females had greater peak and initial knee abduction angle

during landing of three different tasks: sidestepping, jump-landing, and shuttle running compared to males [16]. Chappell et al. [12], further, found that female athletes had greater knee abduction angles during landing phases of stop-jump tasks compared to male athletes.

Concerning the ankle, there is a limited number of studies investigating frontal plane ankle biomechanics [4]. Kernozek et al. [15] found that female athletes displayed greater peak ankle eversion during landing compared to male athletes. During running and cutting task elite female soccer players had greater ankle eversion angle during the stance phase compared to male players [25] which might increase internal tibial rotation, knee abduction angle, anterior tibial translation, and the ACL strain [4]. On the other hand, other studies did not observed any significant differences between males and females with regard to frontal plane ankle kinematics during landing, sidestepping, and running [16, 18].

TRANSVERSE PLANE

The biomechanical differences on the transverse plane between males and females have been demonstrated in the literature in term of the hip and knee [4, 8]. While performing landing, female soccer players showed greater peak and initial hip external rotation compared to male players during a stop jump task [21]. Likewise, female athletes had greater hip external rotation compared to male athletes in preparation for landing of a stop jump task, which happened in conjunction with increased quadriceps and hamstring activation, and might increase the stress on the ACL during landing [22]. Also, during sidestepping and running greater hip external rotation was observed with female athletes compared to male athletes [16, 25].

While preparing for landing, significantly greater knee internal rotation was seen with female athletes compared to male athletes [22], which might increase ACL loading especially if greater knee internal rotation is associated with knee extension and abduction during the landing.

On the other hand, there were similar values of knee internal/external rotation angles among both genders during three different tasks: sidestepping, jump-landing, and shuttle running [16].

KINETICS

SAGITTAL PLANE

One of the most dangerous risk factors of ACL injury is the anterior tibial translation, which is a result of excessive proximal tibial anterior shear force and applies a direct strain on the ligament [4, 12]. Also, the high vertical GRF during landing is considered a risk factor of the ACL injury since this force is transmitted to the knee [4, 48]. Regarding the trunk, it was found that changing the body position on the sagittal plane during landing significantly affects the lower extremity dynamics and lower extremity muscle activation, which increases the risk of the ACL injury [48]. In the previous study, particularly, the landing with upright position displayed greater vertical GRF, peak knee extensor moment, quadriceps muscle activation, and less hip extension moment compared to the landing with leaning forward position. Further, it was reported that landing with an upright position produced 1.4-fold greater peak vertical GRF compared to landing with a forward leaning position, which generates a high compressive force on the tibiofemoral joint and high strain on the ACL [48]. In addition, a study investigated the impact of adding 10% of body mass to the trunk and kinematic trunk adaptations on the lower extremity dynamics and muscle activation [104]. The author demonstrated that the added trunk mass increased peak (17%) and average (35%) knee anterior shear forces when landing with extended trunk. However, added trunk mass when landing with a more flexed trunk displayed no increase in either the peak or average knee anterior shear force [104]. Also, increased trunk extension decreased hamstring muscle activation and increased quadriceps muscle activation [104].

In terms of the hip, a study examining the effect of spike and block landings from different heights found that there is no significant differences in hip extension moment between males and females, but it was clear that female players applied greater vertical GRF than male players during landing [17]. Similarly, a comparison between male and female athletes landing found no significant differences in hip extension moments between the groups [15]; however, the vertical and anterior GRF were significantly greater in female athletes than male athletes [15]. In addition, no difference were observed between three tasks: drop-landing, drop vertical jump, and forward vertical jump with regard to hip extension moment [13]. During jump-heading task, likewise, there was no significant difference in term of peak hip extension moment between males and females, and surprisingly males showed greater vertical GRF than females, which is because males might land stiffer than females [11]. On the other hand, Landry et al. [25] found that females had greater overall hip extension moment compared to males during running and cutting task [25].This difference was observed at the first 10% to 20% of the gait cycle, where ACL injuries are most likely to happen.

Sagittal plane knee kinetics have been examined in many studies, which showed significant gender differences [4, 11, 15, 17, 105]. A study looking at the landing of volleyball players found that the knee extension moment was greater with female players than male players during the block landing from 60 cm [17]. During a jump-heading task, conversely, it was found that males landed with greater peak knee extension moment and peak vertical GRF compared to females [11]. Moreover, it was reported knee extension moment was lower during landing of forward vertical jump task compared to drop-landing and drop vertical jump task [13]. However, no significant differences between males and females in term of knee extension moment during drop landing task [15]. During a stop-jump task, females landed with greater peak knee extension

moment and tibia anterior shear force compared to males, which causes high stress on the ACL [12]. This high tibia anterior shear force in females may be due to increased quadriceps muscle activation, decreased hamstrings muscle activation, and decreased knee flexion angle [12, 101]. Moreover, Weinhandl et al. [18] reported that females landed with greater peak knee extension moments compared to males during drop-landing task from different heights, which may also be due to the greater activation of quadriceps muscle in females compared to males [18]. Even though some previous studies stated that female athletes experience greater vertical ground reaction force [4, 15, 17], other studies found that the vertical ground reaction force was not significantly different between females and males during a stop-jump task [14, 19].

During a volleyball spike and block landing task the ankle did not show any significant difference between males and females even though the females showed greater vertical GRF compared to the males during landing of various heights [17]. During drop-landing, likewise, the peak ankle plantarflexion moment was not different between males and females during landing [15]. Conversely, a study investigating a jump-heading task found that males showed greater peak plantarflexion moment during landing compared to females [11]. Furthermore, it was demonstrated that females had greater peak ankle plantarflexion moment during landing with a leaning forward position compared to landing with an upright position [48]. Also, it was observed that the peak GRF occurred after the peak plantarflexion moment during landing with a leaning forward position, which allows the participants to prepare themselves to absorb the impact forces via the ankle plantarflexors [48].

FRONTAL PLANE

Frontal plane knee mechanics are related to non-contact ACL injuries [4]. With regards to the hip, peak hip adduction moment was not different between males and females during the

landing of drop-land task [15]. Also, the landing of a forward vertical jump task displayed greater hip adduction moment compared to drop-landing task, which is due to the fact that the forward vertical jump task is a more demanding task [13]. During a running and cutting tasks, females showed a greater hip abduction moment in the early stance phase in comparison with males, which increases the valgus knee stress and which elevates the risk of ACL injury [25]. Moreover, the peak hip abduction moment was significantly greater in females during unilateral drop-landing compared to males, which is considered a risk factor of non-contact ACL injury [14, 18].

At the knee, landing technique had a significant influence on all knee moment variables [105]. Thus, the peak knee adduction moment was significantly greater during step-back landing compared to stick-landing in a volleyball block task [105]. Also, the same study reported that the vertical GRF was significantly higher during stick-landing than during step-back landing, which is due to the absence of a secondary motion that occurs after the step-back landing. Further, peak knee adduction moment and vertical GRF of the right and left leg were significantly different, which increases the risk of non-contact ACL injury [105]. Moreover, it was found that a drop-landing task leads to a greater peak knee adduction moment in females compared to males [15]. Also, the landing of forward vertical jump resulted in a greater peak knee adduction moment compared to a drop-landing and a drop vertical jump [13]. Weinhandl et al. [18] found that females had greater peak knee adduction moment during unilateral landing regardless of landing height. Conversely, during a stop-jump task it was demonstrated that there was no significant difference between female and male athletes regard to knee adduction moment during the landing of three different phases: forward, vertical, and backward stop-jump [12].

In term of the ankle, there was no significant difference in peak ankle inversion moment between males and females during a double leg drop landing task [15]. Similarly, no significant difference was observed in peak internal ankle eversion moment during unilateral landings between males and females [18]. While performing running and cutting task, conversely, it was found that male athletes showed greater external eversion moment during the first 20% of stance phase followed by larger inversion moment for the rest of the stance phase compared to females [25].

TRANSVERSE PLANE

Transverse plane, lower extremity kinetics have not been demonstrated comprehensively in the literature, and there are only a few studies mentioning these variables. A study investigating the differences between both genders of soccer players in terms of the biomechanical features during an unanticipated side-cutting task found that males showed greater hip external rotation moment over the whole stance phase compared to females [25]. According to some of the previous studies female soccer, basketball, and volleyball players showed smaller hip external rotation moments and greater knee internal rotation moments compared to male players during landing [4, 14] even though the hip external rotation moment is considered a risk factor of ACL injury[14]. Conversely, it was found that females had more hip internal rotation moment and knee internal rotation moment during unanticipated cutting tasks at 90º angle [4, 14].

CONCLUSION

After reviewing numerous studies which are relative to the biomechanical differences of lower extremity joints in various tasks of both genders, it is evident that the task utilized plays a crucial influence on joints dynamics. In sports, those influences among players have been

meaningfully monitored and investigated in order to improve the performance and intensity of players, and simultaneously reduce the risk of potential injuries. Since soccer continues to experience growth internationally in both men and women [2], it is rational to predict a high incidence rate of injuries amongst soccer players since soccer is a strong sport and includes physical contact. ACL injury is one of the most common injuries amongst athletes [45, 46] since there are around 120,000 ACL injury cases annually in the United States alone [26, 28]. Furthermore, soccer accounted for 36.7% of all ACL injuries during a 10 year epidemiological study compared with other sports [30]. When adjusted for athlete exposures, soccer still exhibits one of the highest ACL injury rates at 0.291 injuries per 1000 athlete exposures [31]. Of those, women had an injury rate of 0.391 compared to the injury rate of 0.198 in men. Most ACL injuries (70-84%) occurred in non-contact situations, which often are associated with landing, deceleration, and sudden change the direction of movement [32-34].

It is important for lower extremity joints to attain greater flexion during landing in order to give muscles enough time to absorb a high GRF [46]. This technique would reduce anterior shear force at the knee and load on ACL [46]. Based on some studies, hip and knee flexion angle were smaller in females compared with males during jump-landing, stop-jump, and sidestepping [16, 17, 19-21]. On the other hand, females showed a greater hip flexion during jump-heading task, which might be due to the effect of jump height variability between males and females [11]. In terms of hip extension moment, no difference was observed during jump-landing, and the vertical GRF was greater in females [13, 15, 17]. However, Butler et al. [11] reported a greater vertical GRF in males compared with females during jump-heading task, which might be because males landed stiffer than females. Also, landing with upright positon might result in a greater vertical GRF compared with landing with leaning forward [48]. Compared with males,

further, females showed a greater knee extension moment during drop-landing and stop-jump task, which might be due to the high activation of quadriceps [12, 17, 18]. Nevertheless, a greater knee extension moment was reported in males during jump-landing task [11].

With regard to frontal plane, a greater hip adduction and knee abduction angle were observed in females during landing of various tasks compared with males, which might increase the knee loading and cause high strain on ACL [12, 16, 18, 23, 24]. Moreover, peak hip abduction moment was greater in females during unilateral drop-landing task compared with males [14, 18], but another study found no difference between genders during drop-landing task [15]. Compared with males, females showed a greater knee adduction moment during drop-landing task and a greater knee adduction moment during unilateral drop-landing [15, 18]. Also, the peak knee adduction moment was greater during step-back landing compared to stick-landing in a volleyball block task [105].

The heading motion is one of the common activities among soccer players during either games or practices. It is usually associated with landing, which might predispose soccer players to injuries, specifically ACL injury. However, there is no epidemiological study calculating the ACL injuries caused by heading motion, and only one study investigating the effect of heading motion on the biomechanical characteristics of lower extremity joints. This study displayed many significant differences between genders and landing tasks in terms of sagittal plane [11]. However, the biomechanical characteristics of lower extremity joints in frontal plane are considered the most important predictors for ACL injury since knee valgus collapse is a common mechanism of ACL injury [4, 38]. For this, reason, I believe that conducting this study is valuable to improve our knowledge and insight in how the kinetics and kinematics of hip and knee for both genders would be affected by the heading motion.

CHAPTER III:

MATERIALS AND METHODS

PARTICIPANTS

In this study, email, flyers, and word of mouth were used to recruit the male and female participants between 18 to 30 years of age who meet the following inclusion criteria.

- Be current field soccer players at a recreational level (average 1-2 hours per week).
- Had at least 2 years of playing experience on a high school or college soccer team.
- Able to perform stop-jump task and jump-heading tasks.
- Be within the normal range of body mass index (BMI = 18-24.9 kg/m^2) [40].
- Score at least 71 out of 80 in Lower Extremity Functional Score (LEFS) [41].
- Be free from any history of ACL injury or reconstruction.
- Be free from any lower extremity injuries within the past 6 months that might affect the typical participation.
- Be free from pain while playing soccer (a game or training) or performing activities of daily living.
- Be and free from any cardiovascular diseases or primary risk factors that prohibit participation in daily exercises or sports.

The Physical Activity Readiness Questionnaire (PAR-Q) and LEFS were completed by the participants [39, 41]. If a subject chose "Yes" on any question of PARS or the score of LEFS was 70 or lower, he or she was excluded from the study. With regard in PAR-Q, however, if there was a written consent from their doctor verifying that the subject was healthy enough to participate in the study, the subject was included in the study. An a priori power analysis, using G*Power [106] and data from the study by Butler et al. [11] and the study by Ford et al [24], was

conducted to determine the number of participants required. The statistical test run in the power analysis was a two-way mixed ANOVA, and the variables utilized included: vertical ground reaction force (VGRF), peak knee flexion angle, peak knee extension moment, peak knee abduction angle, peak hip flexion angle, and peak hip extension moment. The total sample sizes needed were 9, 27, 22, 11, 22, and 22, with an average of 19. Therefore, it was estimated that a minimum of 10 participants in each group for a total sample of 20 with an alpha of 0.05 and a beta 0.80 ($P< 0.05$ and the power = 80%). Before beginning the experiment, all the participants were asked to sign an informed consent letter which was approved by the Institutional Review Board IRB at the University of Tennessee, Knoxville.

INSTRUMENTS

A twelve-camera motion analysis system (200 Hz, Vicon Motion Analysis Inc., Oxford, UK) was used during testing to collect three-dimensional (3D) marker coordinate data. The participants were instructed to wear tight-fitting shorts and t-shirts; they also were provided with standardized running shoes (Noveto, Adidas, USA). Two kinds of retroreflective markers were placed on the participants: anatomical markers and tracking markers (four reflective markers placed on a semi-rigid thermoplastic shell). The anatomical markers were placed bilaterally on the 1^{st} and 5^{th} metatarsal heads, medial and lateral malleoli, medial and lateral femoral epicondyles, greater trochanters, and iliac crests. The tracking marker clusters were placed on the lateral aspects of each shank and thigh and the posterior aspect of the pelvis. Four individual tracking markers were placed on the posterior and lateral aspect of the heel counter of the shoes making a trapezoidal shape. After the collection of a static trial, the anatomical markers were removed before beginning movement trials.

An AMTI force platform (2000 Hz, BP600600, American Mechanical Technology Inc., Watertown, MA, USA) was used to measure the ground reaction force (GRF) of the dominant leg during landing. A movable and adaptable hoist (Exploder The System, Innovative Fitness Products, Fayetteville, AR) holding a soccer ball through a rope was used as a target during performing jump-heading task (Figure 1). The hoist was placed lateral to the force platforms and subjects in order to positon the ball over the line between the two platforms and over the anterior edge of the platforms toward the beginning point of the task. The height of the ball was adjusted for each participant based on the maximum height he or she could jump to head the ball; this height was calculated while the subjects were practicing the tasks of this study by using Matlab (MathWorks, Natick, MA, USA). Soft plastic sport cones were used to guide the participants to perform the task from the same point for each trial (Figure 1, 2). A physician scale was used to measure the mass and height of the participants.

EXPERIMENTAL PROCEDURES

Prior to data collection, the hoist was set up in its position, and the examiner explained the required tasks to the participants. The participants were asked to complete the PAR-Q and LEFS and sign the informed consent. After putting on the specific shorts and running shoes, they were instructed to warm up for 5 minutes of jogging using a treadmill, and preform self-selected stretching exercises for the lower extremities. The height and weight were measured for each participant using a physician scale (with no shoes), and the dominant leg was determined by asking them which leg is most commonly used to kick a soccer ball. Afterwards, the previously described marker sets were placed and taped, and a static trial was collected while the subject was standing on a force platform with feet apart and arms crossed on the chest. After the static trial, the anatomical markers were removed; a ROM trial then was collected, which assists Vicon

software used by the examiner to track the markers, fill the gaps, and clean the data efficiently and quickly. In this trial, while a subject was standing on a force platform, he or she was asked to perform hip flexion, hip 45º flexion, hip abduction, hip 45º extension, hip extension, hip circumduction, knee flexion/extension 3 times, ankle plantar/dorsiflexion 3 times, and ankle circumduction in each leg separately. After the completion of the ROM trial, the movement trials began.

There were two jump tasks: a stop-jump task and a jump-heading task. For the stop-jump task, the subjects were instructed to perform a 2-step approach ending with one foot on each force platform, immediately perform a vertical jump based on the maximal jump height, and land on the same two force platforms (Figure 2). Five successful trials of this task were performed by each subject, and those successful trials required 2-step approach, maximal jump height, maximal double-leg jumping and landing with one foot on each force platform, and a balanced and controlled landing. For the jump-heading task, the subjects performed a 2-step approach prior to performing a maximal double-leg jump (25 cm before the force platforms), heading the soccer ball with a maximum force using the forehead, and landing on both feet, one foot on each force platform (Figure 1). Five successful trials of this task were performed by each subject. In order to consider the trials successful, they required 2-step approach, double-leg jumping (25 cm before the force platforms), maximal jump height, heading the soccer ball maximally using the forehead, landing in a balanced and controlled manner with one foot on each force platform. Any unsuccessful trial was repeated until each subject completed 5 successful trials of each task so that the total number of trials for each subject was 10 trails (5 for each task). The subjects were encouraged to perform the same jump height in both tasks. The order of each task was counterbalanced between subjects.

The subjects were given time prior to data collection to practice until they were comfortable with the two tasks. Throughout the practice period, the investigator set up the height of the soccer ball and beginning point of the tasks. The height of the ball was based on the average of maximal jump height of three practice trials of stop-jump task for each subject, which was calculated by using a customized Matlab script (MathWorks, Natick, MA, USA). Founded on impulse-momentum theorem, the jump height was calculated by dividing the squared takeoff velocity by the acceleration of gravity times two. After practicing the tasks 3-5 times, the subject was then asked from where he or she felt comfortable and confidant to start the 2-step approach while being able to meet all the criteria of successful trials for each task. Once the practicing trials were performed successfully using the determined beginning positions, these beginning points were identified for each task by using cones and tape. The beginning points of the two tasks were not controlled objectively in order to make the tasks more similar to reality during soccer games and practice.

DATA ANALYSIS

A Visual3D biomechanical analysis software suite (C-Motion, Inc., Germantown, MD, USA) was used to analyze and compute 3D kinematic and kinetic data. Kinematic and GRF data were filtered using a fourth-order Butterworth low-pass filter at 10 Hz via Visual3D before calculating 3D kinematics and kinetics of the lower extremity [107]. A Cardan rotational sequence (X-Y-Z) was applied for 3D angular computations using a joint coordinate system for each joint [108]. Hip joint centers were determined at 25% of the distance between the right and left greater trochanter markers, knee joint centers were at the midpoint of the distance between the medial and lateral femoral epicondyle markers, and ankle joint centers were at the midpoint of the distance between the medial and lateral malleolus markers [108-110]. The definition of

angular kinetic and kinematic variable conventions was depending on the right-hand rule. Knee extension and adduction, and hip flexion and adduction angles and moments were expressed by positive values. The joint moments were calculated through an inverse dynamic approach and expressed as internal moments in the distal reference frame. Body segment parameters were estimated from Dempster [111]. GRF variables were first normalized to body weight (BW) and then to body weight times square root of subject jump height yielding a unit of BW/m. Joint moments were first normalized to subject mass (kg) and then to body mass times square root of subject jump height yielding a unit of Nm/(kg.m). The square root of subject jump height was used to normalized GRF and joint moments because the average GRF during landing is proportionate to the square root of jump height based on the impulse-momentum relationship and the properties of uniformly accelerated motion [113]. The variables of interest for this study were peak vertical GRF, peak and initial knee flexion angle, peak and initial knee abduction angle, peak knee extension moment, peak knee adduction moment, peak and initial hip flexion angle, peak and initial hip adduction angle, peak hip extension moment, and peak hip abduction moment. Those variables were investigated during the second landing of stop-jump task and the landing of jump-heading task. Kinematic and kinetic data were analyzed from the initial ground contact to the peak knee flexion angle of the dominant leg.

STATISTICAL ANALYSIS

The mean and standard deviation of the dependent variables of all the groups (gender and jump task) were computed. A two-way (gender × task) within/between analysis of variance (ANOVA) was performed to determine the main effect of gender and jump task, as well as to examine the gender × jump task interactions. For each dependent variable showing significant interaction, *post hoc* simple main effect analysis (independent t-test and paired t-test*)* were

performed. The statistical tests were done via Statistical Package for the Social Sciences software (22.0 IBM SPSS, Chicago, IL) with an alpha of 0.05.

CHAPTER IV:

THE EFFECTS OF HEADING MOTION AND GENDER ON LOWER EXTREMITY BIOMECHANICS IN SOCCER PLAYERS

ABSTRACT

Soccer continues to experience growth internationally in both men and women. Thus, it's reasonable to predict a high rate of injuries amongst those athletes since soccer includes strong physical contact. ACL injury is one of the most common injuries among athletes, and (70-84%) of ACL injuries occurs in non-contact situations, which often are associated with landing, deceleration, and sudden change of movement direction, and these situations might end with knee valgus collapse which's a common mechanism of ACL injury. Further, there are biomechanical differences of lower extremity joints in various tasks with both genders, which apparently means that the task utilized plays a crucial influence on joints dynamics. In soccer, the heading motion is one of the common activities, which is more likely associated with landing. The purpose of this study was to investigate the effect of forward heading motion by comparing the kinetics and kinematics of lower extremity between genders during 1) a stop-jump task and 2) a jump-heading task. 10 male and 10 female soccer players performed 5 stop-jumps and 5 jumps with heading a soccer ball. Findings displayed a combination of reduced initial knee flexion, greater initial hip flexion, greater peak vertical GRF, and greater peak knee extension moments with both genders during the jump-heading task. Additionally, greater peak knee abduction angles were observed among female players during both tasks. These findings suggest that a higher risk of non-contact ACL injuries might occur during the jump-heading task compared with the stop-jump task, and this risk could be higher among female players than male players. These findings might lead soccer trainers to consider different strategies and techniques of landing after heading motions in game-like conditions. Also, PTs and rehabilitation designers who work with ACL-injured patients might consider the outcome of this study. Future studies should consider to examine ankle dynamics in addition to knee and hip in game-like conditions.

INTRODUCTION

Soccer is considered one of the most popular sports in the world, if not the most popular one [1], and continues to experience growth internationally in both men and women [2]. For this reason, it is reasonable to expect a high rate of injuries amongst those athletes since soccer is a violent sport and includes physical contact which might lead to injuries. In 2014, The National Electronic Injury Surveillance System (NEISS) collected injury data across the United States and reported that more than 3.5 million injuries occurred in 25 sports [3]. Approximately 200,000 injuries were related to soccer between the ages of 4 and 64 yrs old, and 80% of these injuries occurred amongst players younger than 25 yrs old [3]. For this reason, these injuries are deemed a part of the major concerns amongst soccer players.

ACL injury has been recognized as one of the common injuries amongst athletes and the most frequent ligament injury of the knee [26, 27] since there are around 120,000 ACL injury cases annually in the United States alone [26, 28]. Majewski et al. [30] reported that soccer accounted for 36.7% of all ACL injuries during a 10 year epidemiological study. When adjusted for athlete exposures, soccer still exhibits one of the highest ACL injury rates at 0.291 injuries per 1000 athlete exposures [31]. Of those, women had an injury rate of 0.391 compared to the injury rate of 0.198 in men. Hewett et al. [35], however, suggested that female athletes are at 4 to 6 times higher risk of ACL injury than male athletes during jumping, cutting, and pivoting tasks. Most ACL injuries (70-84%) occur in non-contact situations, which often are associated with landing, deceleration, and sudden change the direction of movements [32-34]. During landing, a combination of a reduced knee flexion, greater knee abduction, greater knee external rotation, and greater ankle plantarflexion is a common mechanism of ACL injury amongst athletes, especially females, because this commination might maximize the loads placed on the ACL [53,

54]. It also has been reported that landing with greater knee extension moments and knee adduction moments is considered a potential risk factor of non-contact ACL injuries because these moments might increase anterior tibial translation, which produces more strain on the ACL [12, 112].

Based on the literature, some kinematic and kinetic characteristics vary between male and female athletes during various sport activities [11-21], which suggests that the sport-specific task utilized is vitally important when assessing joints dynamics. In terms of kinetics, there appears to be no gender difference in hip extension moments during jump-landing [13, 15]; however, vertical GRF was greater in females than males [17].Compared with males, females show a greater knee extension moment during drop-landing and stop-jump tasks, which might be due to the high activation of quadriceps [12, 17, 18]. Also, peak hip abduction moments and knee adduction moments were greater in females during unilateral drop-landing task compared to males [14, 15, 18], but no difference in peak hip abduction moment was found between genders during drop-landing tasks [15]. During jump-heading tasks, on the contrary, Butler et al. [11] reported a greater vertical GRF in males than females, which might be attributed to the jump height differences between genders. The previous study also showed a greater knee extension moment in males compared to females during jump-landing tasks, which might be because males land more stiffly than females [11]. Regarding kinematics, attaining greater joint flexion during landing gives the muscles enough time to absorb a high GRF, which would reduce stress placed on the knee and ACL [16]. During jump-landing, stop-jump, and sidestepping, hip and knee flexion were smaller in females compared to males [16, 17, 19-21]. Furthermore, a greater hip adduction and knee abduction angle were observed in females during the landing of various tasks compared to males, which might increase the knee loading placing higher strain on the ACL [12,

16, 18, 23, 24]. Also, females showed a greater hip flexion during jump-landing tasks [11], which might refer to discrepancies between genders regarding hip joint mechanics and hip gluteal muscles activation.

The heading motion is one of the common activities in soccer during either games or practices. It is usually associated with landing, which might predispose soccer players to injuries, specifically ACL injury. Furthermore, although the frontal kinetics and kinematics of the hip and knee are considered ones of the primary predictors of ACL injuries [15, 16, 18, 23, 24, 38], only one study investigating the effect of heading motion on biomechanical characteristics of lower extremity joints in the sagittal plane [11]. Therefore, the purpose of this study was to investigate the effect of forward heading motion on the sagittal and frontal biomechanical characteristics of the hip and knee in the dominant leg of female and male soccer players during two jump tasks: 1) a stop-jump task and 2) a jump-heading task. We hypothesized that 1) both male and female players would have reduced peak/initial hip flexion angles, reduced peak/initial knee flexion angles, greater peak/initial hip adduction angles, and greater peak/initial knee abduction angles during the jump-heading task compared to the stop-jump task, 2) both male and female players would have a greater peak vertical GRF, peak hip extension moments, peak knee extension moments, peak hip abduction moments, and peak knee adduction moments during the jump-heading task compared to the stop-jump task, and 3) male players would have greater kinetic variables and sagittal kinematic variables, as well as reduced frontal kinematic variables, compared to female players during both jump tasks.

MATERIALS AND METHODS

PARTICIPANTS

Ten healthy young males (age: 23.1±1.6 yrs, height: 1.79±0.06 m, mass: 67.7±7.7 kg) and ten healthy young females (age: 20.7±1.1 yrs, height: 1.64±0.05 m, mass: 57.3±4.0 kg) were recruited from a university campus (Table 1). The eligibility criteria included current field soccer players between 18 to 30 yrs old who participated at a recreational level (1-2 hours per week), pain-free while playing soccer or performing activities of daily living, at least 2 years of playing experience on a high school or college soccer team, the ability to perform stop-jump and jump-heading tasks, having a normal range of body mass index (BMI = 18-24.9 kg/m^2) [40], and scoring at least 71 out of 80 in Lower Extremity Functional Score (LEFS) [41]. Participants were excluded if they had a history of any ligamentous knee injury, ACL reconstruction, lower extremity injuries within the past 6 months, or cardiovascular diseases using the Physical Activity Readiness Questionnaire (PAR-Q) [39]. An a priori power analysis, using data from Butler et al. [11] and Ford et al [24], was conducted to determine the number of participants required in G*Power [106]. 10 participants in each group (sample = 20) was required to detect differences between groups and jump tasks with an alpha of 0.05 and a beta 0.80. Before beginning the experiment, all the participants were asked to sign an informed consent form which was approved by the Institutional Review Board IRB at the University of Tennessee, Knoxville.

INSTRUMENTS

A twelve-camera motion analysis system (200 Hz, Vicon Motion Analysis Inc., Oxford, UK) was used during testing to collect three-dimensional (3D) marker coordinate data. The participants were instructed to wear tight-fitting shorts and t-shirts; they were also provided with

standardized running shoes (Noveto, Adidas, USA). Two kinds of retroreflective markers were placed on the participants: anatomical markers and tracking markers (four reflective markers placed on a semi-rigid thermoplastic shell). The anatomical markers were placed bilaterally on the 1st and 5th metatarsal heads, medial and lateral malleoli, medial and lateral femoral epicondyles, greater trochanters, and iliac crests. The tracking marker clusters were placed on the lateral aspects of each shank, thigh, and the posterior aspect of the pelvis. Four individual tracking markers were placed on the posterior and lateral aspect of the heel counter of the shoes making a trapezoidal shape. An AMTI force platform (2000 Hz, BP600600, American Mechanical Technology Inc., Watertown, MA, USA) was used to measure the GRF of the dominant leg during landing. A movable and adaptable hoist (Exploder The System, Innovative Fitness Products, Fayetteville, AR) holding a soccer ball through a rope was used as a target during performance of the jump-heading task (Figure 1). The height of the ball was adjusted for each participant based on their maximum jump height, which was calculated from the top of a participant's head using Matlab (MathWorks, Natick, MA, USA). Plastic cones were used to guide the participants to perform the tasks from the same point each trial (Figure 1, 2). A physician scale was used to measure the mass and height of the participants.

EXPERIMENTAL PROCEDURES

After completing the PAR-Q and LEFS, and signing the informed consent, participants were instructed to warm up for 5 minutes of jogging using a treadmill, after which they performed self-selected stretching exercises for the lower extremities. The dominant leg was determined by asking them which leg is preferred to kick a soccer ball. Afterwards, the previously described marker sets were placed and taped, which led to a static trial. Then, the anatomical markers were removed.

There were two jump tasks requiring five trials each: stop-jump task and jump-heading task. For the stop-jump task, the participants were instructed to perform a 2-step approach ending with one foot on each force platform, immediately perform a maximal effort vertical jump, and then land on the same two force platforms (Figure 2). The successful trials of this task required a 2-step approach, maximal jump height, double-leg jumping and landing with one foot on each force platform, while landing in a balanced and controlled fashion. For the jump-heading task, the participants performed a 2-step approach prior to performing a maximal effort double-leg jump (25 cm before the force platforms), heading the soccer ball with a maximum force using the forehead, and landing on both feet, one foot on each force platform (Figure 1). The successful trials of this task required a 2-step approach, double-leg jumping (25 cm before the force platforms), maximal jump height, heading the soccer ball maximally using the forehead, while landing in a balanced and controlled manner with one foot on each force platform. Any unsuccessful trial was repeated, and the participants were encouraged to perform the same jump height in both tasks. The order of each task was counterbalanced between participants.

The participants were given time prior to data collection to practice until they were comfortable with the two tasks. Throughout the practice period, the investigator set up the height of the soccer ball and beginning point of the tasks. The height of the ball was based on the average of maximal jump height of three trials of a stop-jump task for each subject, which was calculated by using a customized Matlab script (MathWorks, Natick, MA, USA) based on impulse-momentum theorem.

DATA ANALYSIS

A Visual3D biomechanical analysis software suite (C-Motion, Inc., Germantown, MD, USA) was used to analyze and compute 3D kinematic and kinetic data. Kinematic and GRF data

were filtered using a fourth-order Butterworth low-pass filter at 10 Hz via Visual3D before calculating 3D kinematics and kinetics of the lower extremity [107]. A Cardan rotational sequence (X-Y-Z) was applied for 3D angular computations using a joint coordinate system for each joint [108]. Hip joint centers were determined at 25% of the distance between the right and left greater trochanter markers, knee joint centers were at the midpoint of the distance between the medial and lateral femoral epicondyle markers, and ankle joint centers were at the midpoint of the distance between the medial and lateral malleolus markers [108-110]. A right-hand rule was used to define angular kinematic and kinetic variable conventions. Knee extension and adduction, and hip flexion and adduction angles and moments were expressed by positive values. The joint moments were calculated through an inverse dynamic approach and expressed as internal moments in the distal reference frame. Body segment parameters were estimated from Dempster et al. [111]. GRF variables were first normalized to body weight (BW) and then to body weight times square root of subject jump height yielding a unit of BW/m. Further, joint moments were first normalized to subject mass (kg) and then to body mass times square root of subject jump height yielding a unit of Nm/(kg.m). The square root of subject jump height was used to normalized GRF and joint moments because the average GRF during landing is proportionate to the square root of jump height based on the impulse-momentum relationship and the properties of uniformly accelerated motion [113]. The variables of interest for this study were investigated during the second landing of stop-jump task and the landing of jump-heading task from the initial ground contact to the peak knee flexion angle of the dominant leg.

STATISTICAL ANALYSIS

The mean and standard deviation of the dependent variables were computed by gender and task. A two-way (gender × task) mixed design analysis of variance (ANOVA) was

performed to examine the gender × jump task interactions and determine the main effect of gender and jump task. For each dependent variable showing significant interaction, *post hoc* simple main effect analysis (independent t-test and paired t-test*)* were performed. The statistical tests were done via Statistical Package for the Social Sciences software (22.0 IBM SPSS, Chicago, IL) with an alpha of 0.05.

RESULTS

Male players showed significantly greater maximum jump height in both jump tasks compared with female players ($P < 0.001$), whereas there was no significant difference of the maximum jump height within each gender between the jump-heading task and the stop-jump task ($P = 0.13$) (Table 3).

With regard to joint kinematics, both male and female players showed greater initial knee flexion during the stop-jump task compared with the jump-heading task ($P = 0.003$) (Table 4). Greater peak knee abduction angles were observed with female players in both jump tasks compared with male players ($P = 0.02$) (Table 4). In terms of the hip, both male and female players exhibited greater initial hip flexion angles during the jump-heading task compared with the stop-jump task ($P = 0.02$) (Table 4).

A significant interaction between gender and jump task in peak vertical GRF was observed ($P = 0.008$) (Table 5). *Post hoc* simple main effect analysis demonstrated that male players had 0.36 BW increase in the peak vertical GRF during the jump-heading task compared with the stop-jump task ($P = 0.001$) (Table 6). During the jump-heading task, moreover, male players exhibited about 0.6 BW increase in the peak vertical GRF compared with female players ($P < 0.001$) (Table 6). For peak knee extension moments, a similar relation was observed in terms of a significant interaction ($P = 0.04$) (Table 5). Peak knee extension moments were

significantly greater in male players during the jump-heading task than the stop-jump task (P = 0.004) (Table 6). Compared with female players, male players exhibited greater peak knee extension moments during the jump-heading task and the stop-jump task (P <0.001 and P = 0.03, respectively) (Table 6). Peak knee adduction moments also showed a significant interaction between gender and jump task (P = 0.005) (Table 5). Unlike female players, male players exhibited a 0.15 Nm/kg increase in peak knee adduction moments during the stop-jump task compared with the jump-heading task (P = 0.03) (Table 6).

After normalizing joint kinetic data to the maximum jump height of players, there was still a significant interaction between the gender and the jump task in the peak vertical GRF (P = 0.02) (Table 7). Also, male players exhibited a 0.5 BW/m increase in the peak vertical GRF during the jump-heading task compared with the stop-jump task (P = 0.001) (Table 8). With regard to peak knee extension moments, a significant main effect of the jump task was observed (P = 0.001) (Table 7). For peak knee adduction moments, on the contrary, a significant interaction between the gender and the jump task was still observed (P = 0.004) (Table 7). Compared with the jump-heading task, male players showed 0.22 BW/m greater peak knee adduction moments during the stop-jump task (P = 0.03) (Table 8).

DISCUSSION

This study aimed to compare dominant leg kinetics and kinematics of the hip and knee in the sagittal and frontal planes of male and female soccer players during a stop-jump task and a jump-heading task. It was hypothesized that both male and female players would show reduced sagittal kinematics and greater frontal kinematics of the hip and knee during the jump-heading task than the stop-jump task. Moreover, it was hypothesized that both male and female players would show greater sagittal and frontal kinetics of the hip and knee, and a greater peak vertical

GRF during the jump-heading task than the stop-jump task. We also hypothesized that male players, when compared to female players, would show greater kinetics, greater sagittal kinematics, and reduced frontal kinematics of the hip and knee during the two jump tasks. It is important to state that our findings partially supported these hypotheses.

Regarding kinematics, our findings showed that both male and female players demonstrated significantly greater initial knee flexion during the stop-jump task than the jump-heading task, which might increase the load placed on the knee and increase the risk of non-contact ACL injuries during the latter task [19, 20]. This result supported the hypothesis in terms of jump task differences, yet no differences between genders were observed, which contradicted our hypothesis as well as most of the literature. During a stop-jump task, previous studies comparing the two groups reported that males showed greater knee flexion at initial foot contact, which would significantly decrease the load placed on the knee and reduce the load on the ACL during landing [19-21]. Unlike initial knee angles, our findings showed a greater initial hip flexion for both male and female players during the jump-heading task than that during the stop-jump task. This increased hip flexion at initial foot contact might be attributed to the flexion momentum of the trunk during the jump-heading task. This concept was supported by Blackburn et al. [101, 102], who demonstrated that active trunk flexion during landing promotes a greater flexion of the hip and knee. Additionally, it was reported that initial hip flexion angles were significantly greater during forward vertical jump task because this task is the most challenging and demanding task compared with drop-landing and drop-jump task [13]. Regarding gender differences, our findings were consistent with previous studies which stated that both males and females demonstrated a similar initial hip flexion during drop-landing from different heights [15, 18]. However, McLean et al. [16] reported that males showed greater initial hip flexion than

females during three tasks: sidestepping, jump-landing, and shuttle running. For peak flexion angles, our findings did not support the first and third hypotheses because neither hip angles nor knee angles differed significantly between the genders or jump tasks. Similarly, Butler et al. [11] did not observe differences between genders regarding peak knee flexion angles during jump-landing and jump-heading tasks. During jump-landing task, nevertheless, the same study stated that peak hip flexion angles were greater with females than males compared with jump-heading task [11]. On the contrary, previous studies reported that males showed greater peak hip flexion angles and peak knee flexion angles during landing compared with females [16, 17].

In terms of hip and knee frontal plane kinematics, which might be considered ones of the primary predictors of knee injuries [38], most of our findings did not demonstrate any significant difference between the genders or jump tasks. Specifically, neither initial and peak hip adduction angles, nor initial knee abduction angles supported our hypotheses. Likewise, Cruz et al. [13] found no differences between three different landing tasks with respect to initial hip adduction angles. However, two studies comparing males to females confirmed that the latter group showed greater initial hip adduction during landing [18, 23]. This result was later contradicted by Bodor et al. [22] who demonstrated the opposite. On the other hand, our findings relating to peak knee abduction angles were in agreement with the hypotheses for gender differences only; female players showed greater peak knee abduction angles during both tasks than male players. This finding is highly significant because the frontal plane kinematics of the knee are considered one of the most crucial predictors of ACL injuries, whose common mechanism is knee valgus collapse [4, 38]. Similarly, females showed greater peak abduction angles during 30-50% of the landing phase compared with males [12, 15, 24], and that would place a 6 times greater load on the ACL at 5º of knee abduction [103].

Regarding kinetics, a high vertical GRF during landing is considered a risk factor for ACL injuries because this force is transmitted to the soft tissue of the knee, especially ligaments and muscles [4, 48]. Our GRF findings supported the hypotheses partially, there was only a significant difference between male and female players during the jump-heading task (2.44 BW and 1.83 BW, respectively, Table 6), and between the jump-heading task and the stop-jump task among male players (2.44 BW and 2.08 BW, respectively, Table 6). The increased peak vertical GRF exhibited by male players might be due to the jump height differences between male and female players (Table 3). Furthermore, changing the trunk position on the sagittal plane during the jump-heading task might contribute in the increased peak vertical GRF because it was reported that the lower extremity dynamics and muscle activation are influenced by the trunk position during landing, which might increase the risk of non-contact ACL injuries [48]. Similarly, a previous study showed a greater peak vertical GRF in male players than female players during jump-landing and jump-heading task, and the peak was greater during the jump-heading task than the stop-jump task with both genders [11]. However, the researchers observed a greater peak vertical GRF in females than males during landing with similar jump heights, which might be caused by the reduced hip and knee flexion with females compared to males [15, 17]. With respect to peak knee extension moments, our findings supported our hypothesis because these moments were greater with males than females during the jump-heading task and the stop-jump task. Additionally, these moments were greater during the jump-heading task than the stop-jump task among male players (Table 6), which might lead to a greater anterior tibial translation during the jump-heading task. Butler et al. [11] reported the same knee extension moments in their study with an exception: when compared to the jump-heading task, the jump-landing task showed greater peak knee extension moments. However, previous studies stated that

greater peak knee extension moments were observed with females than males during bilateral landing [12, 17] and unilateral landing [18]. Regarding frontal knee moments, our findings did not support our hypotheses because it was believed that increased knee frontal kinematics during jump-heading task would result in increased knee frontal kinetics. However, our knee frontal kinematic findings showed no differences between the two jump tasks. Our findings showed that peak knee adduction moments were significantly greater during the stop-jump task than the jump-heading task among males, but there were no differences between genders in each jump task (Table 6). Similarly, in terms of peak knee adduction moments during three different phases of the stop-jump task, no difference between males and females was observed [12]. However, Kernozek et al. [15] found greater peak knee adduction moments in females than males during a drop-landing task. Concerning sagittal and frontal plane hip moments, our findings did not display any differences between genders or jump tasks. Observing no difference, these findings were consistent with previous studies that examined peak hip extension moments between males and females during landing [11, 13, 15, 17]. For peak hip abduction moments, it was reported that these moments were greater during the landing of a vertical forward jump compared with a drop-landing task [13]. Also, these moments were greater in females than males during a unilateral drop-landing task, which might place high stress on the knee joint [18].

Because male and female players differed significantly in terms of maximum jump height, which might influence the kinetic findings (Table 3), the peak vertical GRF, hip moments, and knee moments were normalized to the maximum jump height of each participant (Table 7, 8). Even after the data normalization, significant differences between the tasks remained with regard to peak vertical GRF, peak knee extension moments, and peak knee adduction moments. However, the significant differences between genders of these variables

were no longer present. Therefore, caution should be taken when interpreting gender differences in GRF and joint moments normalized using traditional methods if landing height varies between individuals. The present study displayed some significant differences between the two jump tasks with respect to hip and knee kinetics and kinematics, which might indicate to a high risk of non-contact ACL injuries during the jump-heading task.

This study had limitations that may have impacted the findings, so it's necessary to state them. First, even though our male and female participants were field soccer players who had at least 2 years' experience on either a high school or college team and who played soccer recreationally 1-2 hours a week, the skill level of heading a ball was somewhat different between genders. Secondly, though the participants were encouraged to head the ball as forcefully as they could, this force was not controlled objectively in order to minimize the potential variations. Also, it is rare to perform an isolated specific soccer task (i.e., a jump-heading task) without following movements such as a jumping, a running, or a sidestepping after the heading during a game or practice, which might make the present findings not entirely applicable during real play. Moreover, the present study was performed in a room (laboratory) with a wooded floor, and participants wore standardized running shoes. These nongame-like conditions might impact lower extremity dynamics during landing because soccer players in real situations would use cleats and play on a turf ground. Finally, the findings of peak GRF might have been underestimated because they were low-pass filtered at 10 Hz.

CONCLUSION

Male and female soccer players demonstrate alternative landing mechanics in each jump task: jump-heading and stop-jump. Our findings displayed a combination of reduced initial knee flexion, greater initial hip flexion, greater peak vertical GRF, and greater peak knee moments

with both genders during the jump-heading task. Additionally, greater peak knee abduction angles were observed among female players during both tasks. These findings suggest that a higher risk of non-contact ACL injuries might occur during the jump-heading task compared with the stop-jump task, and this risk could be more prominent among female players than male players. This might benefit researchers who are interested in the biomechanical differences of the lower extremity between tasks and gender. Our findings also might lead soccer trainers to consider different strategies and techniques of landing after heading motions in game-like conditions, such as landing more softly to reduce the knee loads. Moreover, our findings might be helpful for the clinicians and rehabilitation designers who work with ACL-injured patients so that they might stay away from the jump-heading task during the rehabilitation period. To provide more applicable findings, future studies could implement tests similar to those of the present study using a turf and cleats in game-like conditions, such as the effect of lateral heading motion using the side of the head during landing with some physical contact. Also, it would be beneficial to include ankle kinetics and kinematics in an investigation. Lastly, comparing the mechanics of dominant and non-dominant legs during the jump-heading task would be a valuable study.

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APPENDICES

APPENDIX A. PARTICIPANTS' DEMOGRAPHICS

TABLE 1. The mean and standard deviation (SD) of participants' demographics.

	Max Jump Height (m)		**Height (m)**	**Weight (kg)**	**Age (yrs)**	**Soccer Experience in High School/College (yrs)**
	JH	SJ				
Male	0.48 (0.05)	0.47 (0.06)	1.79 (0.06)	67.72 (7.69)	23.1 (1.59)	3.2 (0.79)
Female	0.32 (0.03)	0.32 (0.03)	1.64 (0.05)	57.32 (4.03)	20.7 (1.06)	3.4 (0.69)

JH Jump-Heading Task.
SJ Stop-Jump Task.

TABLE 2. Individual participant demographics.

Subject		Max Jump Height (m)		Height (m)	Weight (kg)	Age (yrs)	Dominant Leg	Soccer Experience in High School/College (yrs)
		JH	SJ					
Male	**1**	0.56	0.56	1.86	71.67	26	Rt	3
	2	0.46	0.42	1.76	57.15	24	Rt	3
	3	0.45	0.39	1.83	81.19	24	Rt	4
	4	0.51	0.53	1.81	68.04	22	Lt	4
	5	0.54	0.56	1.71	61.23	22	Rt	3
	6	0.41	0.42	1.76	75.75	22	Rt	3
	7	0.48	0.46	1.71	58.97	25	Rt	4
	8	0.48	0.44	1.86	70.31	23	Rt	2
	9	0.44	0.46	1.74	62.59	22	Lt	2
	10	0.5	0.48	1.79	70.31	21	Rt	4
Female	**1**	0.32	0.29	1.66	50.01	19	Lt	3
	2	0.34	0.31	1.56	58.97	20	Rt	3
	3	0.3	0.33	1.61	51.71	20	Rt	4
	4	0.32	0.33	1.68	56.69	22	Rt	4
	5	0.37	0.29	1.65	61.23	22	Rt	4
	6	0.3	0.27	1.71	61.01	21	Rt	4
	7	0.37	0.33	1.71	61.23	22	Rt	2
	8	0.28	0.31	1.58	58.97	20	Rt	3
	9	0.3	0.28	1.58	58.97	21	Rt	4
	10	0.32	0.28	1.62	54.43	20	Rt	3

JH Jump-Heading Task.
SJ Stop-Jump Task.
Rt Right Leg.
Lt Left Leg.

APPENDIX B. CHAPTER IV TABLES

TABLE 3. The descriptive statistics (mean, standard deviation SD, and p-value) of maximum jump height.

	Jump-Heading Task	**Stop-Jump Task**	***p*-value**		
	Mean (SD)	Mean (SD)	**Gender ME**	**Jump Task ME**	**Interaction**
Max Jump Height (m)					
Male	0.48 (0.05)	0.47 (0.06)	**0.000***	0.13	0.94
Female	0.32 (0.03)	0.31 (0.01)			

ME Main effect.
***** Statistically significant difference at $p<0.05$.

TABLE 4. The descriptive statistics (mean, standard deviation SD, and p-value) of knee and hip angles.

	Jump-Heading Task	**Stop-Jump Task**	***p*-value**		
			Gender ME	**Jump Task ME**	**Interaction**
	Mean (SD)	Mean (SD)			
Initial Knee Flexion Angle (˚)			0.07	**0.003***	0.52
Male	-21.27 (4.76)	-27.19 (10.35)			
Female	-16.72 (5.62)	-20.71 (7.20)			
Peak Knee Flexion Angle (˚)			0.56	0.13	0.6
Male	-88.23 (12.22)	-92.99 (17.76)			
Female	-85.79 (13.57)	-88.13 (15.65)			
Initial Knee Frontal Angle (˚)			0.055	0.37	0.65
Male	5.53 (3.32)	6.59 (4.76)			
Female	2.87 (3.44)	3.23 (3.13)			
Peak Knee Frontal Angle (˚)			**0.02***	0.33	0.92
Male	2.18 (5.35)	2.87 (6.03)			
Female	-3.43 (4.83)	-2.59 (4.46)			
Initial Hip Flexion Angle (˚)			0.43	**0.02***	0.11
Male	24.78 (6.78)	22.43 (9.52)			
Female	26.13 (11.23)	14.59 (14.62)			
Peak Hip Flexion Angle (˚)			0.97	0.79	0.14
Male	62.12 (16.01)	68.76 (21.05)			
Female	67.38 (20.84)	62.77 (24.98)			
Initial Hip Frontal Angle (˚)			0.11	0.61	0.19
Male	-13.39 (5.29)	-14.58 (6.51)			
Female	-10.37 (4.38)	-9.84 (4.93)			
Peak Hip Frontal Angle (˚)			0.17	0.64	0.13
Male	-10.60 (7.3)	-12.16 (8.49)			
Female	-7.33 (5.69)	-6.49 (7.09)			

ME Main effect.

***** Statistically significant difference at $p<0.05$.

TABLE 5. The descriptive statistics (mean, standard deviation SD, and p-value) of peak vertical GRF and joint moments.

	Jump-Heading Task	**Stop-Jump Task**	***p*-value**		
	Mean (SD)	Mean (SD)	**Gender ME**	**Jump Task ME**	**Interaction**
Peak Vertical Ground Reaction Force (BW)			**0.005***	**0.000***	**0.008***
Male	2.44 (0.27)	2.08 (0.34)			
Female	1.83(0.29)	1.75 (0.46)			
Peak Knee Extension Moment (Nm/kg)			**0.001***	**0.001***	**0.04***
Male	3.17 (0.28)	2.64 (0.39)			
Female	2.29 (0.53)	2.13 (0.54)			
Peak Knee Adduction Moment (Nm/kg)			0.22	0.16	**0.005***
Male	0.32 (0.26)	0.47 (0.38)			
Female	0.28 (0.18)	0.23 (0.1)			
Peak Hip Extension Moment (Nm/kg)			0.37	0.89	0.86
Male	-1.46 (0.49)	-1.50 (0.82)			
Female	-1.32 (0.25)	-1.31 (0.29)			
Peak Hip Abduction Moment (Nm/kg)			0.37	0.47	0.89
Male	- 0.36 (0.38)	- 0.43 (0.51)			
Female	- 0.26 (0.12)	- 0.31 (0.18)			

ME Main effect.

***** Statistically significant difference at $p<0.05$.

TABLE 6. Dependent *t*-test and paired *t*-test of the kinetic variables.

	Jump-Heading Task		**Stop-Jump Task**		**Gender**
	Mean (SD)	*p*-value	Mean (SD)	*p*-value	*p*-value
Peak Vertical Ground Reaction Force [a, c] (BW)		**0.000***		0.09	
Male	2.44 (0.27)		2.08 (0.34)		**0.001***
Female	1.83(0.29)		1.75 (0.46)		0.29
Peak Knee Extension Moment [a,b,c] (Nm/kg)		**0.000***		**0.03***	
Male	3.17 (0.28)		2.64 (0.39)		**0.004***
Female	2.29 (0.53)		2.13 (0.54)		0.12
Peak Knee Adduction Moment [c] (Nm/kg)		0.74		0.07	
Male	0.32 (0.26)		0.47 (0.38)		**0.03***
Female	0.28 (0.18)		0.23 (0.1)		0.09

***** Statistically significant difference at $p<0.05$.

[a] Significant difference between genders during jump-heading task.

[b] Significant difference between genders during stop-jump task.

[c] Significant difference between jump tasks for males.

TABLE 7. The normalized descriptive statistics (mean, standard deviation SD, and p-value) of peak vertical GRF and joint moments.

	Jump-Heading Task	**Stop-Jump task**	***p*-value**		
	Mean (SD)	Mean (SD)	**Gender ME**	**Jump Task ME**	**Interaction**
Peak Vertical Ground Reaction Force (BW/m)			0.71	**0.001***	**0.02***
Male	34.53 (4.76)	29.58 (5.84)			
Female	31.64 (5.31)	30.43 (8.37)			
Peak Knee Extension Moment (Nm/(kg.m))			0.41	**0.001***	0.08
Male	4.58 (0.49)	3.82 (0.69)			
Female	4.06 (0.94)	3.78 (0.98)			
Peak Knee Adduction Moment (Nm/(kg.m))			0.49	0.25	**0.004***
Male	0.45 (0.36)	0.67 (0.52)			
Female	0.50 (0.33)	0.40 (0.18)			
Peak Hip Extension Moment (Nm/(kg.m))			0.43	0.93	0.74
Male	- 2.09 (0.66)	- 2.14 (1.12)			
Female	- 2.36 (0.42)	- 2.27 (0.49)			
Peak Hip Abduction Moment (Nm/(kg.m))			0.94	0.39	0.98
Male	- 0.51 (0.52)	- 0.61 (0.69)			
Female	- 0.49 (0.25)	- 0.60 (0.41)			

ME Main effect.

***** Statistically significant difference at $p<0.05$.

TABLE 8. Dependent *t*-test and paired *t*-test of the normalized kinetic variables.

	Jump-Heading Task		**Stop-Jump Task**		**Gender**
	Mean (SD)	*p*-value	Mean (SD)	*p*-value	*p*-value
Peak Vertical Ground Reaction Force [c] (BW/m)		0.22		0.79	
Male	34.53 (4.76)		29.58 (5.84)		**0.001***
Female	31.64 (5.31)		30.43 (8.37)		0.31
Peak Knee Adduction Moment [c] (Nm/(kg.m))		0.75		0.14	
Male	0.45 (0.36)		0.67 (0.52)		**0.03***
Female	0.50 (0.33)		0.40 (0.18)		0.08

***** Statistically significant difference at $p<0.05$.

[a] Significant difference between genders during jump-heading task.

[b] Significant difference between genders during stop-jump task.

[c] Significant difference between jump tasks for males.

APPENDIX C. CHAPTER IV FIGURES



FIGURE 1. The hoist of soccer ball was placed lateral to the force platforms and participants, and the ball was placed over the line in between the two platforms and over the anterior edge of the platforms toward the beginning point of the task. Cone A shows the point at which the participants begin the two-step approach, and cone B shows the point where the participants jump to head the ball during the jump-heading task.

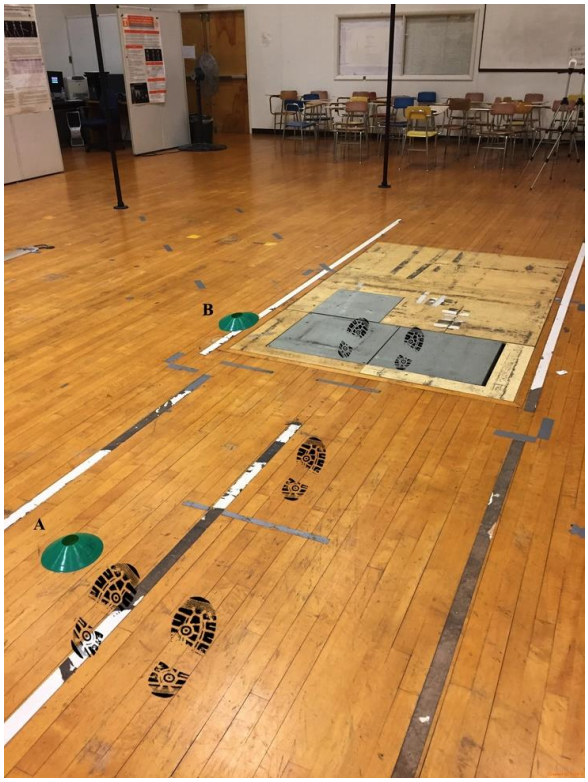


FIGURE 2. Cone A shows the point at which the participants begin the two-step approach, and cone B shows the point where the participants perform a maximal vertical jump during the stop-jump task.

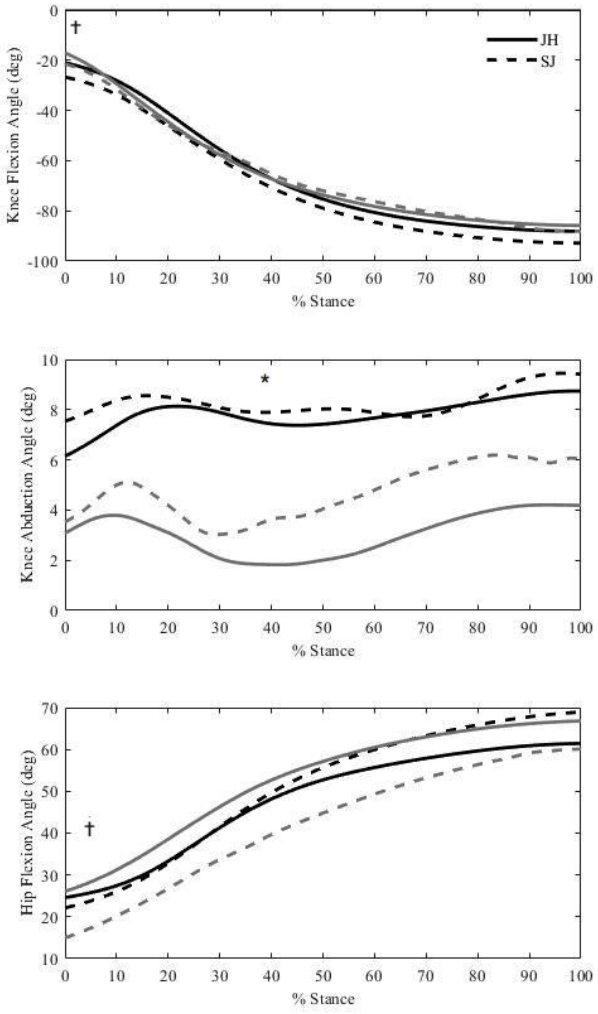


FIGURE 3. Hip and knee kinematic curves. Comparison of hip and knee joint angles between the jump-heading task (JH) and stop-jump task (SJ) in males (black line) and females (gray line). * indicates a significant difference between genders ($p < 0.05$), and † indicates a significant difference between jump tasks ($p < 0.05$).

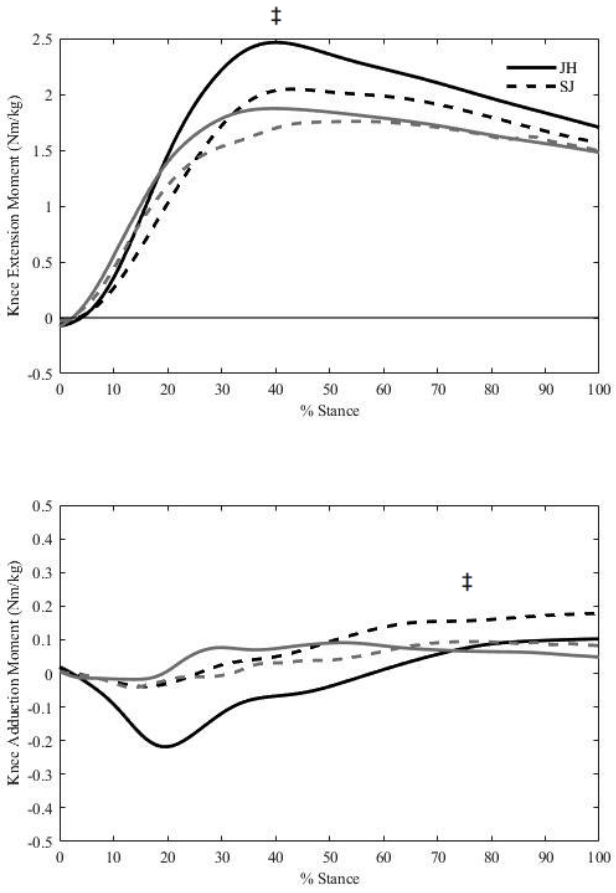


FIGURE 4. Knee kinematic curves. Comparison knee joint moments between the jump-heading task (JH) and stop-jump task (SJ) in males (black line) and females (gray line). ‡ indicates a significant interaction between gender and jump task ($p < 0.05$).

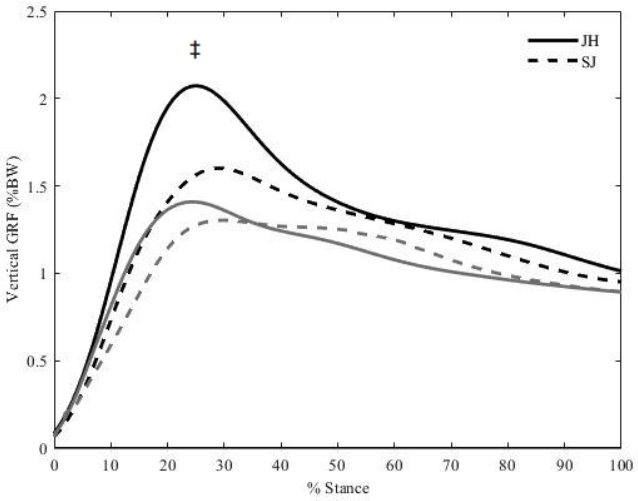


FIGURE 5. Peak vertical GRF curves. Comparison of peak vertical GRF between the jump-heading task (JH) and stop-jump task (SJ) in males (black line) and females (gray line). ‡ indicates a significant interaction between gender and jump task ($p < 0.05$).

APPENDIX D. INFORMED CONSENT

Informed Consent

The Effects of Heading Motion and Gender on Lower Extremity Biomechanics in Soccer Players.

INTRODUCTION

Participants are invited to participate in a research study conducted in the University of Tennessee Biomechanics Lab (HPER 136). The purpose of this study is to compare the biomechanics of the hip, knee, and ankle in the dominant leg of female and male soccer players during two landing tasks: 1) stop-jump task and 2) jump-heading task

ELIGIBILITY

To participate in this study, you must be a current field soccer player between 18 and 30 years of age. Also, you must play soccer at a recreational level (1-2 hours per week) and have two years of playing experience in a high school or college soccer team. You must have NOT had a history of ACL injury or reconstruction, any lower extremity injuries within the past 6 months, pain while playing soccer or performing activities of daily living, and cardiovascular diseases.

INFORMATION ABOUT PARTICIPANTS' INVOLVEMENT IN THE STUDY

You will come into the Biomechanics Lab for one session, which will last approximately one hour. You will also complete the Physical Activity Readiness Questionnaire PAR-Q and Lower Extremity Functional Scale LEFS to ensure that they are qualified to participate. You will change into Spandex shorts and running shoes, which will be provided. Height and weight will be taken, followed by a 5-minute warmup jog and stretching of the lower extremities. The dominant leg will be determined by asking you which leg is used in kicking a ball. Reflective markers (plastic spheres) will be placed on anatomical landmarks on both legs and pelvis using double sided tape. Additionally, thermoplastic shells with reflective markers will be attached via Velcro to neoprene straps wrapped around the calves, thighs, and pelvis of the participant. Both of these will be used to track foot, leg, and pelvis movement throughout the session.

Data collection will consist of two jumping conditions: stop-jump task and jump-heading task. A movable hoist that holds a soccer ball through a rope and the height of the ball will be set up while the participant is practicing the tasks. The stop-jump task consists of performing a 2 step approach ending with one foot on each force platform, immediate performing vertical jump based on the maximal jump height, and landing on the same two force platforms. The jump-heading task consists of performing a 2 step approach, a 2 leg jumping 10 cm before the force platforms based on the maximal jump height, heading the soccer ball, and landing on both feet, each foot on a different force platform. Five successful trials for each condition will then be completed. The session will conclude once these trials are collected.

RISKS

Because the participant is being asked to jump with and without heading a soccer ball, then land on a flat surface below, there is a possibility of lower extremity injury. You will be required to

warm up with a 5-minute jog and stretching of the lower extremity to ensure your muscles are ready for jumping tasks. For jump-heading task, the height of the ball will be set up based on your maximal jump height; also, the researcher will be nearby, ensuring that you will have assistance present at all times. 3-5 practicing trials will be before data collection until you feel comfortable and confident with the jumping tasks. _____ Initials A loss of confidentiality is also a possible risk related to the research. Such a disclosure might link you to your data or your association with the study.

BENEFITS

You will not earn direct benefits from your participation in this experiment. The data collected from you may help provide a better understanding of how heading motion affects the biomechanical characteristics of lower extremity amongst soccer players. The data collected may also provide soccer players, coaches, and researchers a better insight into how heading motion might be related to non-contact ACL injuries.

COMPENSATION

After completing the consent form and screening questionnaires and determining that you are eligible to participate in the study, you will receive a Starbucks gift card ($10) to express our appreciation to your participation. However, if you are ineligible for the study, you will **NOT** receive the gift card.

CONFIDENTIALITY

The information collected in this study will be kept secured and private. Each participant will be identified by a given number. For data security, it will be saved confidentially using a password-protected computer and a locked cabinet in the Biomechanics lab. The only people who conduct this study can view your information, unless you write a permission letter for someone else.

CONTACT INFORMATION

If you have any question about the study or the experimental procedures, (or you have some adverse effects due to the experiment,) you may contact the researcher, Abdulmajeed Alfayyadh, at majeed@vol.utk.edu or (302) 898-5467 (phone number), or Dr. Joshua Weinhandl, PhD at jweinhan@utk.edu. To make sure about your rights as a participant, you may contact the University of Tennessee IRB Compliance Officer at utkirb@utk.edu or (865) 974-7697.

PARTICIPATION

You are participating in this study voluntarily, and you can deny to participate without any sanction. After your decision to participate, you may withdraw from the study at any time without any sanction. Your participation may be ended by the investigator without regard to your consent, such as if you unable to comply with the study procedures or do not meet the eligibility criteria.

CONSENT

I have read the above information. I have received a copy of this form. I agree to participate in this study.

Participant's Name (printed) _____

Participant's Signature _____ Date _____

APPENDIX E. LOWER EXTREMITY FUNCTIONAL SCALE

Lower Extremity Functional Scale

We are interested in knowing whether or not you are having any difficulty at all with the activities listed below. Please provide an honest answer for each activity.

KEY
0 - Extreme difficulty or unable to perform activity
1 - Quite a bit of difficulty
2 - Moderate difficulty
3 - A little bit of difficulty
4 - No difficulty

Today, ***do you*** or ***would you*** have any difficulty at all with:	Extreme 0	Quite a bit 1	Moderate 2	Minimal 3	None 4
1. Any of your usual work, housework or school activities	☐	☐	☐	☐	☐
2. Your usual hobbies, recreational or sporting activities	☐	☐	☐	☐	☐
3. Getting into or out of the bath	☐	☐	☐	☐	☐
4. Walking between rooms	☐	☐	☐	☐	☐
5. Putting on your shoes or socks	☐	☐	☐	☐	☐
6. Squatting	☐	☐	☐	☐	☐
7. Lifting an object, like a bag of groceries from the floor	☐	☐	☐	☐	☐
8. Performing light activities around your home	☐	☐	☐	☐	☐
9. Performing heavy activities around your home	☐	☐	☐	☐	☐
10. Getting into or out of a car	☐	☐	☐	☐	☐
11. Walking 2 blocks	☐	☐	☐	☐	☐
12. Walking a mile	☐	☐	☐	☐	☐
13. Going up or down 10 stairs (about 1 flight)	☐	☐	☐	☐	☐
14. Standing for 1 hour	☐	☐	☐	☐	☐
15. Sitting for 1 hour	☐	☐	☐	☐	☐
16. Running on even ground	☐	☐	☐	☐	☐
17. Running on uneven ground	☐	☐	☐	☐	☐
18. Making sharp turns while running fast	☐	☐	☐	☐	☐
19. Hopping	☐	☐	☐	☐	☐
20. Rolling over in bed	☐	☐	☐	☐	☐

APPENDIX F. PHYSICAL ACTIVITY READINESS QUESTIONNAIRE (PAR-Q)

Physical Activity Readiness Questionnaire (PAR-Q)

Name #: ……

Date: ……..

PAR-Q is designed to help you help yourself. Many health benefits are associated with regular exercise, and the completion of PAR-Q is a sensible first step to take if you are planning to increase the amount of physical activity in your life.

For most people, physical activity should not pose any problem or hazard. PAR-Q has been designed to identify the small number of adults for whom physical activity might be inappropriate or those who should have medical advice concerning the type of activity most suitable for them.

Common sense is your best guide in answering these few questions. Please read them carefully and check the "yes" or "no" box opposite the question if it applies to you.

	Yes	No
1- Has your doctor ever said you have heart trouble?	☐	☐
2- Do you frequently have pains in your heart and chest?	☐	☐
3- Do you often feel faint or have spells of severe dizziness?	☐	☐
4- Has a doctor ever said your blood pressure is too high?	☐	☐
5- Has your doctor ever told you that you have a bone or joint such as arthritis, that has been aggravated by exercise or might be made worse with exercise?	☐	☐
6- Is there a good physical reason not mentioned here why you should not follow an activity program even if you wanted to?	☐	☐
7- Are you over the age of 65 and not accustomed to vigorous exercise?	☐	☐

APPENDIX G. IRB APPROVAL LETTER



THE UNIVERSITY OF TENNESSEE KNOXVILLE

April 05, 2017

Abdulmajeed Barakat Alfayyadh,
UTK - Coll of Education, Hlth, & Human - Kinesiology

Re: UTK IRB-17-03644-XP
Study Title: The Effects of Heading Motion and Gender on Lower Extremity Biomechanics in Soccer Players Dear

Abdulmajeed Barakat Alfayyadh:

The UTK Institutional Review Board (IRB) reviewed your application for the above referenced project. It determined that your application is eligible for expedited review under 45 CFR 46.110(b)(1), categories (4) and (7). The IRB has reviewed these materials and determined that they do comply with proper consideration for the rights and welfare of human subjects and the regulatory requirements for the protection of human subjects.

Therefore, this letter constitutes full approval by the IRB of your application (version 1.3) as submitted, including Informed Consent (v1.3), Majeed Study Flyer-2 (v1.1), Majeed Study Email (v1.0), Physical Activity Readiness Questionnaire (v1.0), and the LEFS Questionnaire (v1.0). The listed documents have been dated and stamped IRB approved. Approval of this study will be valid from April 05, 2017 to April 04, 2018.

In the event that subjects are to be recruited using solicitation materials, such as brochures, posters, web-based advertisements, etc., these materials must receive prior approval of the IRB. Any revisions in the approved application must also be submitted to and approved by the IRB prior to implementation. In addition, you are responsible for reporting any unanticipated serious adverse events or other problems involving risks to subjects or others in the manner required by the local IRB policy.

Finally, re-approval of your project is required by the IRB in accord with the conditions specified above. You may not continue the research study beyond the time or other limits specified unless you obtain prior written approval of the IRB.

Sincerely,



Colleen P. Gilrane,
Ph.D.
Chair

Institutional Review Board | Office of Research & Engagement 1534 White Avenue Knoxville, TN 37996-1529
865-974-7697 865-974-7400 fax irb.utk.edu

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Flagship Campus of the University of Tennessee System

APPENDIX H. INDIVIDUAL RESULTS FOR VARIABLES OF INTEREST

TABLE 9. The mean and standard deviation (SD) of individual sagittal hip kinematics.

Subject		Initial Flexion Angle (˚)		Peak Flexion Angle (˚)	
		JH	**SJ**	**JH**	**SJ**
Male	**1**	35.38 (3.11)	25.74 (2.16)	93.36 (4.14)	90.47 (3.42)
	2	28.83 (8.91)	8.80 (6.15)	72.08 (13.33)	53.39 (2.92)
	3	19.94 (3.62)	17.55 (6.21)	57.15 (4.20)	51.86 (7.39)
	4	19.48 (4.96)	30.78 (8.21)	51.66 (2.59)	78.57 (6.07)
	5	28.36 (3.83)	34.87 (4.79)	72.69 (8.11)	98.64 (5.74)
	6	32.14 (2.82)	19.61 (8.11)	65.18 (8.90)	64.67 (14.79)
	7	16.43 (6.37)	18.88 (4.11)	48.34 (8.24)	76.03 (2.43)
	8	28.36 (7.85)	29.41 (6.19)	69.62 (18.25)	82.92 (4.57)
	9	15.76 (7.66)	7.44 (9.12)	36.09 (7.31)	27.43 (10.67)
	10	23.12 (6.19)	31.21 (14.14)	54.91 (7.52)	63.58 (16.71)
Female	**1**	34.17 (4.78)	28.59 (4.75)	94.38 (6.69)	92.95 (4.76)
	2	18.49 (1.19)	7.67 (4.84)	65.79 (5.82)	54.18 (6.91)
	3	24.06 (7.91)	20.03 (2.83)	47.89 (6.34)	57.16 (4.95)
	4	14.78 (4.70)	7.34 (4.24)	36.22 (10.54)	31.99 (5.11)
	5	23.45 (4.89)	20.81 (8.17)	86.46 (8.55)	95.78 (2.03)
	6	38.09 (6.04)	-12.85 (58.13)	74.57 (11.23)	26.84 (92.99)
	7	20.89 (5.17)	9.24 (3.27)	72.56 (5.66)	72.11 (7.01)
	8	18.83 (6.12)	11.17 (6.45)	65.04 (5.21)	70.05 (2.89)
	9	18.20 (3.45)	11.82 (5.56)	39.26 (2.97)	39.56 (4.87)
	10	50.33 (3.41)	42.12 (10.15)	91.61 (2.52)	87.09 (3.13)

JH Jump-Heading Task.
SJ Stop-Jump Task.

TABLE 10. The mean and standard deviation (SD) of individual frontal hip kinematics.

Subject		Initial Adduction Angle (˚)		Peak Adduction Angle (˚)	
		JH	**SJ**	**JH**	**SJ**
Male	**1**	-5.47 (2.11)	-2.87 (4.15)	0.17 (0.66)	2.47 (2.84)
	2	-20.35 (6.57)	-20.09 (2.61)	-19.51 (5.53)	-20.09 (2.61)
	3	-10.03 (3.94)	-9.76 (3.31)	-8.15 (2.28)	-9.20 (2.89)
	4	-13.63 (5.27)	-11.64 (2.12)	-12.64 (5.16)	-11.28 (2.22)
	5	-10.97 (2.79)	-19.56 (2.32)	-8.29 (1.24)	-15.14 (2.81)
	6	-11.90 (3.04)	-11.19 (1.69)	-5.83 (1.29)	-9.04 (0.95)
	7	-6.94 (0.76)	-9.03 (3.56)	-0.32 (2.02)	0.39 (2.21)
	8	-18.49 (3.28)	-22.20 (4.02)	-15.45 (1.47)	-20.42 (3.68)
	9	-15.88 (2.55)	-18.19 (3.08)	-15.85 (2.54)	-18.07 (2.91)
	10	-20.32 (3.02)	-21.25 (6.04)	-20.12 (3.06)	-21.25 (6.04)
Female	**1**	-11.05 (1.01)	-13.19 (2.92)	-7.69 (1.84)	-7.86 (2.84)
	2	-2.49 (1.42)	-2.68 (0.92)	0.67 (1.74)	2.48 (2.26)
	3	-12.95 (2.59)	-14.97 (1.61)	-10.53 (1.61)	-13.41 (2.46)
	4	-6.39 (1.01)	-6.75 (2.25)	0.99 (1.54)	-2.29 (1.30)
	5	-12.01 (4.72)	-14.37 (2.93)	-11.19 (3.29)	-13.64 (2.66)
	6	-9.75 (2.63)	-5.45 (6.38)	-6.09 (2.03)	3.37 (15.81)
	7	-17.75 (2.76)	-16.88 (1.00)	-16.19 (3.00)	-16.88 (1.00)
	8	-12.84 (1.62)	-9.49 (3.74)	-10.19 (1.57)	-5.63 (2.89)
	9	-5.98 (2.53)	-4.28 (0.45)	-1.85 (1.48)	-0.77 (1.12)
	10	-12.49 (1.42)	-10.36 (2.55)	-11.22 (0.88)	-10.33 (2.53)

JH Jump-Heading Task.
SJ Stop-Jump Task.

TABLE 11. The mean and standard deviation (SD) of individual sagittal knee kinematics.

Subject		Initial Flexion Angle (˚)		Peak Flexion Angle (˚)	
		JH	**SJ**	**JH**	**SJ**
Male	**1**	-22.64 (2.69)	-21.48 (2.77)	-102.76 (4.25)	-94.59 (3.31)
	2	-21.18 (5.49)	-20.31 (4.33)	-100.13 (13.11)	-101.95 (3.89)
	3	-21.20 (2.92)	-18.41 (2.64)	-83.22 (3.72)	-74.03 (5.59)
	4	-24.07 (8.49)	-44.66 (5.32)	-90.17 (6.07)	-113.75 (4.52)
	5	-21.99 (1.08)	-26.81 (2.73)	-104.00 (4.49)	-124.03 (4.41)
	6	-20.71 (3.64)	-22.26 (3.49)	-82.59 (12.67)	-81.64 (5.05)
	7	-12.10 (1.66)	-18.99 (1.11)	-71.82 (6.69)	-85.27 (3.35)
	8	-28.01 (2.39)	-31.78 (4.11)	-94.17 (5.54)	-98.39 (2.16)
	9	-14.81 (5.41)	-21.25 (4.56)	-69.19 (5.45)	-65.07 (2.95)
	10	-26.03 (5.21)	-45.98 (18.73)	-84.24 (8.95)	-91.19 (16.76)
Female	**1**	-20.38 (1.75)	-24.32 (4.07)	-104.29 (4.98)	-100.62 (5.07)
	2	-17.71 (1.32)	-17.42 (3.43)	-76.65 (5.22)	-69.33 (2.87)
	3	-18.27 (3.34)	-21.84 (4.25)	-78.02 (4.55)	-86.27 (5.91)
	4	-15.91 (3.94)	-18.92 (2.53)	-70.34 (2.41)	-70.61 (2.47)
	5	-22.22 (2.31)	-22.56 (4.40)	-101.16 (4.78)	-98.18 (6.03)
	6	-20.93 (2.05)	-31.87 (5.61)	-84.65 (5.85)	-108.52 (6.29)
	7	-15.61 (3.85)	-13.19 (2.38)	-90.25 (3.38)	-90.18 (7.65)
	8	-6.14 (2.68)	-11.15 (4.82)	-97.91 (7.58)	-96.19 (3.05)
	9	-7.93 (1.22)	-14.22 (4.01)	-63.45 (3.08)	-62.16 (2.78)
	10	-22.05 (3.82)	-31.69 (6.66)	-91.15 (3.35)	-99.22 (5.14)

JH Jump-Heading Task.
SJ Stop-Jump Task.

TABLE 12. The mean and standard deviation (SD) of individual frontal knee kinematics.

Subject		Initial Abduction Angle (˚)		Peak Abduction Angle (˚)	
		JH	**SJ**	**JH**	**SJ**
Male	**1**	6.27 (1.55)	4.29 (1.71)	5.81 (1.24)	3.85 (1.48)
	2	8.78 (1.22)	4.89 (2.02)	0.45 (3.73)	-7.02 (4.79)
	3	2.32 (1.38)	1.71 (1.69)	-0.66 (2.42)	-1.32 (2.59)
	4	-0.59 (2.43)	5.02 (3.19)	-8.02 (5.07)	-1.96 (2.84)
	5	8.77 (1.51)	12.81 (4.16)	7.77 (1.69)	8.15 (1.75)
	6	3.22 (1.63)	2.69 (1.39)	-1.11 (2.09)	0.99 (2.36)
	7	5.75 (0.54)	6.29 (1.29)	5.75 (0.54)	6.29 (1.29)
	8	8.21 (0.78)	10.99 (1.64)	7.13 (1.31)	10.23 (1.67)
	9	3.46 (1.04)	1.99 (1.69)	-2.63 (0.82)	-1.74 (2.69)
	10	9.12 (1.57)	15.25 (5.06)	7.34 (1.98)	11.21 (1.12)
Female	**1**	1.60 (0.54)	2.92 (1.86)	-8.52 (1.86)	-6.02 (1.69)
	2	0.04 (1.01)	-1.79 (0.93)	-5.39 (1.09)	-9.66 (2.29)
	3	3.10 (1.73)	4.69 (2.47)	-3.79 (1.88)	-1.95 (2.95)
	4	2.07 (0.76)	0.47 (0.73)	-8.56 (2.77)	-8.00 (1.41)
	5	4.53 (2.55)	4.89 (5.29)	4.33 (2.17)	3.87 (4.63)
	6	-0.04 (2.26)	4.28 (3.65)	-9.01 (3.82)	-1.39 (1.71)
	7	9.38 (3.00)	7.48 (1.14)	2.44 (3.96)	2.45 (3.37)
	8	0.19 (2.26)	0.89 (2.79)	-0.33 (2.19)	0.71 (2.94)
	9	1.65 (0.78)	0.82 (0.99)	-5.69 (1.11)	-4.94 (1,89)
	10	5.46 (0.39)	7.69 (1.49)	0.22 (1.57)	-1.06 (1.63)

JH Jump-Heading Task.
SJ Stop-Jump Task.

TABLE 13. The mean and standard deviation (SD) of individual hip kinetics.

Subject		Peak Extension Moment (Nm/kg)		Peak Abduction Moment (Nm/kg)	
		JH	**SJ**	**JH**	**SJ**
Male	**1**	-2.49 (0.19)	-1.81 (0.33)	-1.22 (0.14)	-1.01 (0.18)
	2	-1.66 (0.35)	-0.39 (0.19)	-0.11 (0.14)	-0.08 (0.11)
	3	-1.14 (0.16)	-1.25 (0.63)	-0.52 (0.29)	-0.35 (0.18)
	4	-1.47 (0.42)	-3.55 (2.87)	-0.04 (0.14)	-1.63 (3.16)
	5	-1.62 (0.44)	-1.78 (0.27)	-0.02 (0.09)	-0.05 (0.04)
	6	-1.59 (0.43)	-1.22 (0.48)	-0.46 (0.18)	-0.39 (0.29)
	7	-0.56 (0.42)	-1.24 (0.39)	-0.65 (0.28)	-0.39 (0.21)
	8	-1.53 (0.27)	-1.43 (0.30)	-0.29 (0.09)	-0.19 (0.08)
	9	-1.40 (0.54)	-1.08 (0.55)	-0.23 (0.21)	-0.18 (0.13)
	10	-1.09 (0.56)	-1.27 (0.64)	-0.01 (0.09)	-0.05 (0.20)
Female	**1**	-1.31 (0.16)	-1.15 (0.38)	-0.26 (0.05)	-0.23 (0.13)
	2	-1.59 (0.42)	-1.19 (0.36)	-0.19 (0.09)	-0.31 (0.21)
	3	-1.18 (0.42)	-1.33 (0.38)	-0.19 (0.15)	-0.22 (0.06)
	4	-0.71 (0.49)	-0.56 (0.41)	-0.42 (0.23)	-0.41 (0.18)
	5	-1.46 (0.26)	-1.62 (0.23)	-0.24 (0.10)	-0.19 (0.14)
	6	-1.15 (0.16)	-1.42 (0.58)	-0.28 (0.14)	-0.45 (0.13)
	7	-1.49 (0.33)	-1.63 (0.19)	-0.27 (0.28)	-0.14 (0.07)
	8	-1.40 (0.18)	-1.58 (0.33)	-0.11 (0.06)	-0.25 (0.09)
	9	-1.39 (0.20)	-1.21 (0.51)	-0.48 (0.12)	-0.73 (0.19)
	10	-1.46 (0.16)	-1.32 (0.46)	-0.12 (0.03)	-0.16 (0.05)

JH Jump-Heading Task.
SJ Stop-Jump Task.

TABLE 14. The mean and standard deviation (SD) of individual knee kinetics.

Subject		Peak Extension Moment (Nm/kg)		Peak Adduction Moment (Nm/kg)	
		JH	SJ	JH	SJ
Male	1	2.81 (0.32)	2.35 (0.39)	0.07 (0.05)	0.07 (0.08)
	2	2.73 (0.73)	2.73 (0.27)	0.32 (0.23)	0.59 (0.08)
	3	3.32 (0.31)	2.60 (0.20)	0.09 (0.09)	0.14 (0.13)
	4	3.86 (0.62)	3.52 (0.99)	0.95 (0.58)	1.34 (1.73)
	5	3.17 (0.32)	2.52 (0.14)	0.39 (0.15)	0.78 (0.11)
	6	3.25 (0.21)	3.15 (0.55)	0.11 (0.14)	0.14 (0.23)
	7	2.86 (0.46)	2.25 (0.33)	0.01 (0.03)	0.14 (0.04)
	8	3.33 (0.62)	2.31 (0.66)	0.48 (0.07)	0.51 (0.21)
	9	3.31 (0.53)	3.16 (0.44)	0.28 (0.23)	0.54 (0.15)
	10	3.54 (0.33)	2.15 (0.52)	0.58 (0.27)	0.50 (0.34)
Female	1	2.05 (0.23)	1.82 (0.48)	0.73 (0.10)	0.50 (0.16)
	2	2.41 (0.19)	2.54 (0.22)	0.09 (0.07)	0.16 (0.12)
	3	2.83 (0.53)	2.38 (0.24)	0.24 (0.12)	0.19 (0.24)
	4	3.42 (0.78)	3.33 (0.42)	0.25 (0.15)	0.25 (0.11)
	5	2.51 (0.24)	1.88 (0.60)	0.17 (0.09)	0.22 (0.07)
	6	2.03 (0.51)	1.53 (0.87)	0.32 (0.18)	0.24 (0.12)
	7	1.92 (0.51)	1.69 (0.27)	0.22 (0.30)	0.16 (0.07)
	8	1.66 (0.24)	1.93 (0.36)	0.15 (0.06)	0.13 (0.08)
	9	2.36 (0.25)	2.45 (0.33)	0.24 (0.09)	0.20 (0.11)
	10	1.81 (0.34)	1.81 (0.30)	0.31 (0.19)	0.22 (0.05)

JH Jump-Heading Task.
SJ Stop-Jump Task.

TABLE 15. The mean and standard deviation (SD) of individual peak vertical GRF.

Subject		Peak Vertical GRF (BW)	
		JH	SJ
Male	1	2.17 (0.25)	1.86 (0.27)
	2	2.32 (0.54)	1.76 (0.15)
	3	2.60 (0.33)	2.22 (0.18)
	4	2.33 (0.26)	1.89 (0.23)
	5	2.07 (0.21)	1.80 (0.12)
	6	2.30 (0.04)	2.19 (0.21)
	7	2.35 (0.49)	2.17 (0.31)
	8	2.65 (0.41)	2.01 (0.28)
	9	2.96 (0.21)	2.93 (0.28)
	10	2.61 (0.27)	1.99 (0.21)
Female	1	1.63 (0.17)	1.31 (0.39)
	2	1.95 (0.06)	2.07 (0.22)
	3	1.89 (0.36)	1.77 (0.25)
	4	2.42(0.23)	2.53 (0.29)
	5	1.74 (0.29)	1.44 (0.30)
	6	1.59 (0.24)	1.34 (0.34)
	7	1.61 (0.44)	1.37 (0.22)
	8	1.54 (0.15)	1.52 (0.25)
	9	2.19 (0.11)	2.49 (0.26)
	10	1.69 (0.22)	1.69 (0.27)

JH Jump-Heading Task.
SJ Stop-Jump Task.

VITA

Abdulmajeed Alfayyadh was born in Arar city in 1989, and his family moved to Aljouf city, where he and his two brothers and two sisters grew up; he is the second oldest child of his siblings. He graduated from Aljouf University in 2013 with a Bachelor degree in Physiotherapy and Health Rehabilitation. After graduation, he was accepted to be a teaching assistant at the College of Applied Medical Sciences, Aljouf University. Consequently, he has been granted a scholarship to pursue his graduate studies in the United States. After he took an intensive English course at the University of Delaware for two years, he earned a M.S. degree in Kinesiology with concentration in Biomechanics at the University of Tennessee, Knoxville in Spring 2018. He has been admitted at two schools to pursue his PhD in Kinesiology with emphasis in Biomechanics in Fall 2018: Oregon State University and University of Kentucky.