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Effects of Synthetic Turf Systems With and Without a Shock Pad on Lower Extremity Biomechanics During a 90° Cutting Movement With Differing Approach Velocities

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I am submitting herewith a thesis written by Thomas Kenneth Elvidge entitled "Effects of Synthetic Turf Systems With and Without a Shock Pad on Lower Extremity Biomechanics During a 90° Cutting Movement With Differing Approach Velocities." I have examined the final electronic copy of this thesis for form and content and recommend that it be accepted in partial fulfillment of the requirements for the degree of Master of Science, with a major in Kinesiology.

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Effects of Synthetic Turf Systems With and Without a Shock Pad on Lower Extremity Biomechanics During a 90° Cutting Movement With Differing Approach Velocities

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Abstract

The purpose of this study was to examine differences in lower extremity kinematics and kinetics on two different synthetic turf systems (turf only and turf with a shock pad) for two approach velocities (3.0 and 4.0 m/s) during a 90° cutting movement. Twelve recreational male American football and soccer players were recruited to perform five trials for each of the four conditions. A three-dimensional motion analysis system synchronized to a force platform was used to collect marker coordinate and ground reaction force (GRF) data respectively. A 2 x 2 (surface x approach velocity) ANOVA was used to analyze kinematic and kinetic variables. Across surface conditions, there was a general lack of significant differences. While there was a lack of differences for kinematics and kinetics, there might have been increased co-contractions to stabilize the lower extremity with the increased deformation on the shock pad condition, which was undetectable via the inverse dynamics. However, knee frontal-plane peak loading eccentric power was found to be greater on the shock pad condition ($p = 0.013$) while knee frontal-plane peak push-off eccentric power was reduced on the shock pad condition ($p = 0.020$). A surface x approach velocity interaction was detected for knee sagittal-plane peak eccentric power ($p = 0.018$). Post-hoc analysis found a significant difference for approach velocity on the turf only condition. As the protocol dictated a change in performance, the largest changes were seen in peak hip extension ($p = 0.007$) and knee extension ($p = 0.004$) moments, suggesting that these were the major factors for determining the performance improvement. There were also increases in ankle eversion moment ($p < 0.001$) and ankle inversion ROM ($p = 0.001$) as approach velocity increased. These increases potentially suggest that the risk of a lateral ankle

sprain injury increases as approach velocity increases. As approach velocity increased, it was found that peak push-off vertical GRF decreased (*p* = 0.011) as peak push-off medial GRF increased (*p* = 0.025). This suggests that as approach velocity increases, medial forces become more important than vertical forces during the push-off phase.

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Chapter 1

Introduction

Synthetic turf pitches have become increasingly common across the world due to reduced financial costs and all-weather ability (62). However, despite the approval by many sport governing bodies for their use (9, 10, 12), concerns exist from elite players particularly from an injury perspective (13, 15). The notion that synthetic turf increases the risk of injury has been well researched with conflicting findings. While some have found that the relative risk of suffering an injury is comparable between synthetic turf and natural grass (24, 51, 87), others have found that the risk of injury increases on synthetic turf (72). In particular, it has been shown that the risk of suffering an anterior cruciate ligament (ACL) or ankle sprain injuries is significantly higher on synthetic turf (51, 72).

Synthetic turf pitches have become more popular and widespread (88) and have been substantially developed since the first generation of synthetic turf (88, 105). Synthetic turf has been investigated regarding its impact attenuation ability compared to natural grass with contrasting findings (60, 139, 140). It is suggested that the thickness and compliance of a surface are related to the maximal displacement of a surface, which is hypothesized as being directly related to the ability of a surface to absorb impact forces (130). Therefore, hypothetically, increasing the thickness of a surface would increase the potential for impact attenuation, and this in turn increases the ability of the surface to absorb impact forces (130, 139). Furthermore, in comparing synthetic turf to natural grass, it has been observed that although overall loading doesn't change between surfaces, during cutting the medial forefoot experiences the highest plantar

pressure and subsequent loading at these regions of the foot is highest (60). This increased loading of the medial forefoot in combination with high frictional forces and torsional resistance associated with the shoe-surface interaction (when wearing cleats) for natural grass are the proposed causes which cause the ‘cleat-catch’ mechanisms, where the cleat gets caught in the surface (36, 68). These increased frictional and rotational forces may in turn result in excessive shear and rotational forces that propagate up the kinetic chain to affect joints such as the knee (36).

A recent method for improving impact attenuation and injury prevention of synthetic turf is the addition of an underlying shock pad (101). As shock pads are not common in synthetic turf installations, their effects on movement patterns have not been extensively studied and are not fully understood. In comparing three different synthetic turf configurations, Duraspine ULTRA 42 (professional level with underlying shock pad), Duraspine ULTRA 50 (recreational level), and Duraspine ULTRA 60 (professional level without underlying shock pad), it has been shown that the ULTRA 42 configuration had significantly reduced impact forces compared to the ULTRA 50 configuration during a stop sprint task (101). Given that the ULTRA 50 configuration had an increased infill thickness compared to the ULTRA 42 configuration but the total height (infill thickness + shock pad height) was similar between the two, it can be suggested that the inclusion of a shock pad has a greater influence on the impact absorption properties than infill thickness (101). The material that shock pads are constructed of has also been shown, from a mechanical perspective, to potentially effect the impact absorption properties of the surface (151).

The shoe-surface interface is an important factor when considering injury mechanisms (117); where, the role of footwear to is transfer force from the athlete to the surface safely and without performance deficits (19, 134). It has been argued that footwear has a larger influence than the surface for impact absorption (47). In terms of soccer and football cleats, differences in cleat configurations have been shown to influence the relative distribution of plantar pressure (19); where, increased frictional forces are associated with regions of higher pressure and subsequent increases in loading at the areas (36). Additionally, traction properties of footwear are often related to their loading characteristics. Increased rotational traction seen in cleats compared to running shoes were seen to be linearly correlated with increased frontal plane joint moments at the ankle and knee joints (138). It has also been seen that players wearing cleats with increased rotational traction significantly increased the risk of suffering an ACL injury (85).

Analysis of cutting movements has revealed that the vertical and horizontal GRFs during the stance phase are comprised of two peaks: an initial passive impact peak (42, 61) and a global active peak associated with propulsion force of the athlete. It is suggested that the risk of ACL injury is greatest during the loading phase (131). Cutting movements require an athlete to decelerate before accelerating away in a different direction (67) and is seen as a key performance attribute in football and soccer (63, 100). The cutting angle has been shown to have effects on kinematic and kinetic variables, although these changes are not linearly correlated with cutting angle (68). When comparing 90° to 45° cutting movements, peak knee extensor moments and peak posterior GRF were seen to increase, suggesting that the knee joint serves as the

primary mechanism for absorbing the increased posterior GRF (68). Analysis in the frontal plane has also revealed differences with cutting angle, with the peak internal knee adduction moment during 180° cutting movements being significantly higher than during 90° cutting movements (40). These differences in cutting angle and subsequent influence on joint kinetics and kinematics has been proposed to influence the risk of an ACL injury. Increased peak lateral GRF and internal knee adduction moments have been shown to be key risk factors for suffering a non-contact ACL injury (103, 132), with these values increasing as cutting angle increases. Similarly, increases in peak posterior GRF and internal knee extensor moments have been identified as fundamental contributors to anterior tibial shear force (129, 160), which has been shown to be an indicator of ACL loading (20, 97).

One key component that determines loading during a cutting movement is the approach velocity. Loading is related to deceleration, leading to the suggestion that increased deceleration results in increased loading (146). Therefore, as the amount of deceleration required to change direction is related to the input velocity, approach velocity of a cutting movement is an important factor for lower extremity loading. As approach velocity increases, it has been shown that both sagittal and frontal plane variables are altered. With approach velocity increasing from 2.0 to 5.0 $m \cdot s^{-1}$, peak internal knee adduction moment increased from 0.12 ± 0.17 to 1.14 ± 0.84 N·m/kg while peak posterior GRF increased from 5.0 N/kg to 12.9 N/kg (146). These results also suggest that increases in loading are non-linear so a minimum threshold may exist for an approach velocity to cause potentially hazardous loading under deceleration and that increasing approach velocity increases the variability of loading. Furthermore,

extrapolation of these data to higher approach velocities usually experienced in game situations could produce internal knee adduction moments with a magnitude that results in ACL injury (146). Intra-study comparisons of cutting movements using similar cutting angles but different approach velocities support the notion that increased loading occurs with increased approach velocity (86, 132). Although not directly related to cutting movements, changes in linear running velocity also reveal differences in loading with increases in vertical GRF (35, 78, 84, 111, 116), sagittal plane ankle and knee moments (3, 137), and knee joint stiffness (3, 46, 137).

Longitudinal studies have shown that the majority of ACL injuries result from non-contact mechanisms in dynamic sports such as American football and soccer (2, 30). The exact mechanisms behind non-contact ACL injuries has been researched extensively. Studies using questionnaire (27) and qualitative and quantitative analyses of injury videos (29, 83) have confirmed that non-contact situations appear to be the primary causation of injury with two injury mechanisms proposed: valgus collapse (increases in knee abduction angle) where valgus motion of the tibia causes the knee to collapse inwards, and anterior tibial shear when the knee is close to full extension during the initial loading phase when completing a sudden deceleration movement or landing (27, 28, 39, 83). Furthermore, cadaver studies have suggested that anterior shear force at the proximal tibia is the primary mechanism for loading the ACL (97), with peak posterior GRF related to this variable. However, *in vitro* analysis has suggested that sagittal plane mechanisms alone were unable to load the ACL and additional loading in the frontal plane from knee adduction moments is required to load the ACL to a magnitude where it could rupture (103). In all, there is a wealth of literature on the

mechanisms behind non-contact ACL injuries with the primary mechanism appearing to be related to anterior shear force, with frontal plane loading also contributing as a secondary mechanism. Therefore, changes in surface properties and its effects on impact attenuation could have a potential influence on the risk of ACL injury.

Statement of Problem

Currently, there is a limited amount of research investigating the difference in movement patterns from a biomechanical perspective that result from changes in synthetic turf properties. More specifically, there is a lack of research that has examined the effect of underlying shock pads on football- and soccer-specific movements that have previously been associated with lower extremity injury and the effect that different energy inputs have on lower extremity joints. Furthermore, the effects of increasing approach velocity on impact related biomechanical variables are currently unknown. Therefore, the purpose of this study was to examine the effects of different approach velocities for a 90° cutting movement on both impact attenuation related and performance related kinematics, kinetics, and energetics on synthetic turf systems with and without an underlayment shock pad.

Significance of Study

The overall aim of this study was to provide extensive details regarding the kinematics and kinetics of the lower limb during a football- and soccer-specific cutting movement with different input energy from varying the approach velocity. By analyzing kinematics and kinetics of the lower limb during a football- and soccer-specific cutting movement with different input energy from varying the approach velocity on synthetic turf systems, this research aimed to provide information on impact attenuation capacity

of the examined turf systems and how the turf system could influence biomechanical variables related to performance. Previous research has also revealed that impact peak GRFs and frontal plane movements during dynamic movements can result in injury. It has also been found that surface properties can influence the movement patterns, which in turn can influence injury risk and performance. The results of the current study may allow the development of an artificial turf system that can reduce the risk of injury and enhance performance, giving this study the potential to affect a large number of players of all abilities who use artificial turf pitches to play various dynamic sports including football and soccer.

Hypotheses

The following hypotheses were made and tested:

1) The synthetic turf system with an underlying shock pad would produce lower GRFs, and decreased lower extremity sagittal and frontal plane moments, powers, and range of motion (ROMs) compared to the synthetic turf system without an underlying shock pad.
2) An increased approach velocity would cause increases in GRFs and lower extremity sagittal and frontal plane moments, powers, and ROMs regardless of types of turf systems tested..

Limitations

1. All subjects were recruited from students at the University of Tennessee who play American football or soccer regularly.
2. The artificial turf that was used was only one type of artificial turf.

3. The artificial turf that was used was brand new, where most pitches have some level of wear.
4. The underlying shock pad that was used for testing was only one density and from one company.
5. The cleats that were used for testing were only one make and model.
6. Testing was conducted in a controlled laboratory setting and participants may have altered how they completed the cutting movements compared to how the movements would be completed if they were playing a game or participating in practice.
7. The motion analysis system lacks accuracy as the markers used to calculate 3-dimensional kinematic data are placed manually using the palpation method.

Delimitations

1. All subjects were free from injury, regularly participating in American football or soccer, and had no previous history of severe lower extremity injury.
2. Following a sufficient warm-up, each subject performed five successful cutting trials for each of the approach velocities on each of the two surface conditions with sufficient rest time between trials.
3. The installed area of artificial turf, size of the lab, and position of the cameras ensured subjects had sufficient space to both accelerate prior to the cutting movement and decelerate following the completion of the trial.
4. Kinematic data was collected at 240 Hz using Vicon 3D motion analysis system (Vicon MX, Oxford, Metrics, Oxford, UK) and kinetic data was collected at 1200 Hz using a AMTI force platform (American Mechanical Technology Inc., MA).

5. Nike cleats (Nike Vapor Untouchable Pro, Nike, Beaverton, OR) are sold in specific cleat configurations.
6. The most up-to-date generation of artificial turf was used during testing.

Chapter 2

Literature Review

The purpose of this investigation was to examine changes in kinetic and kinematic variables in different artificial turf conditions during 90° cutting movements with differing approach velocities designed to simulate gameplay movements. This literature review consists of six main sections: differences between synthetic turf and natural grass, preliminary work on the effects of shock pads, the role of different footwear characteristics on the shoe-surface interaction, biomechanical characteristics of cutting movements including the effects of changing the approach velocity, background information regarding the prevalence of injuries associated with change in surface, and the mechanisms behind anterior cruciate ligament injuries.

Synthetic Turf vs. Natural Grass

Synthetic turf was first developed to provide a space for recreation in urban areas, and the first synthetic turf surface was then produced by Monsanto in 1964 (88). Since then, synthetic turf surfaces have become more widespread across a variety of locations and for a variety of sports (88). This has led to the development of other types of synthetic turf surfaces, with three generations of synthetic turf surfaces being developed.

The first generation was comprised of 10 mm polyvinyl chloride and a foam mat, but it was often associated with excess traction and skin abrasions (114). First generation turf was characterized by the fibers acting as the playing surface, while the second generation synthetic turf differed as the playing surface was comprised of a mix between the fibers and a sand infill (88). This second generation synthetic turf was a

closer representation of natural turf in terms of both functionality and aesthetics (88). This was achieved through the increased length of nylon, polypropylene or polyethylene fibers to 22 to 25 mm, the inclusion of an underlying pad beneath the layer of fibers, and giving the turf a soil-like element via a sand and/or rubber infill that was applied between fibers (52, 105). The most current iteration of synthetic turf, third generation, uses the same materials for the fibers but with an increased length of 40 to 70 mm, in addition to a rubber and sand infill (up to 50 mm) and the entire system being installed on top an asphalt or crushed aggregate base (105). It has been well reported that there is an increase in the injury rate for the early generations of synthetic turf (1, 76, 88, 127, 133).

The idea that the properties of the playing surface can be a contributing factor to ACL injuries has been identified within specific sports. In a longitudinal study, 5.1% of soccer players who suffered an ACL injury identified the playing surface as the primary mechanism that caused the injury (126). The notion that the shoe-surface interface can be influenced by the surface to then have an effect on the risk of injury is highlighted by the disparity in injury rates between indoor and outdoor soccer (126). The increase in the proportion of ACL injuries occurring during indoor soccer is primarily due to an increased stiffness and frictional forces generated due to the shoe-surface interaction during change of direction movements (66). The combination of an increase in traction with known intrinsic risk factors has been suggested to increase the risk of suffering an injury (120).

The effect of surface on movement patterns and a potential effect on the risk of injury has been extensively investigated. Comparative studies between synthetic turf and natural grass while completing high-load movements such as cutting has revealed

differences in various biomechanical variables; in particular plantar pressures. It is thought that understanding the specific loading patterns for individual surfaces may help determine the mechanisms behind surface specific injury patterns (107). When completing an agility slalom on both surfaces, it was found although the overall loading did not change between surfaces, the distribution of the plantar pressures did differ (60). It was proposed by the authors that this alteration in loading patterns would affect the types of injuries an athlete is predisposed to, based upon the surface on which they are playing on (60). When comparing synthetic turf with a shock pad to natural grass, the compliance of synthetic turf condition was found to reduce the loading at the highest plantar pressure region, the medial forefoot, through a combination of foot motion and the cushioning effect of the shock pad (107). In contrast, there was increased medial forefoot loading for the natural grass condition and it was proposed that the reduced compliance of the surface caused this increase in loading (107).

When completing cutting movements, the medial forefoot has been identified as the region of the foot with the highest plantar pressure and subsequent loading (60). On natural grass, the ‘cleat-catch’ mechanism has been identified where the cleat gets caught in the surface (71). It is proposed that this occurs due to a combination of increased loading on the medial forefoot and high frictional forces and torsional resistance associated with the shoe-surface interface for natural grass (36). The increased frictional and rotational forces as a result of the shoe-surface interaction may in turn result in excessive shear and rotational forces that propagate up the kinetic chain to affect joints such as the knee. The ‘cleat-catch’ mechanism has been proposed as a

reason for the increased incidence of ACL injuries in the Australian Football League where the depth of thatch of the natural grass influences the rotation of the foot (122).

In contrast, more recent research has suggested that the properties of natural grass actually reduce the risk of injury in comparison to modern synthetic turf (79). This research suggests that the release mechanisms of the cleat differ between surfaces and that for synthetic turf, the properties of the surface prevent a ‘divoting’ mechanism (79). It was proposed that this mechanism is an important injury mitigating mechanism for natural grass where the properties of natural grass allow the surface to fail and ‘divoting’ to occur, limiting the loads placed upon the foot and subsequently loading of more proximal joints up the kinetic chain.

Much of the early research investigating the ability of synthetic turf to provide reduced impact absorption properties in comparison to natural grass yielded conflicting results (60, 139, 140). The ability of a surface to absorb impact is hypothesized to be related to the maximum possible displacement of the surface, meaning that the thickness and compliance of the surface are key factors in determining this maximal displacement (130). The thickness and compliance of the surface can be altered through mechanical factors including the amount and type of infill and the installation of an under-layer shock pad, thus influencing the overall impact properties of the surface. This is because an increase in displacement prolongs the contact time, resulting in the same force being applied over a longer period of time and reducing the amount of force transmitted to the lower extremity joints (101). Therefore, increasing the thickness of the surface increases the potential for impact attenuation capacity, and this in turn increases the ability of the surface to act effectively to absorb impact forces (130, 139).

Synthetic turf has been used to evaluate cleated footwear within a laboratory environment (18, 138). Comparing synthetic turf cleats to natural turf cleats and running shoes when completing 180° cutting movement on synthetic turf, it was found that increased loading response peak knee adduction moment and negative plantarflexion power were observed for the synthetic turf cleats compared to natural turf cleats (18). The authors suggested that given knee extensor moments and powers remained constant, that the synthetic turf cleats actually served to increase loading at the knee joint during the movement (18).

Additionally, the perception of muscle soreness in elite athletes following playing on synthetic turf compared to natural grass has been investigated. While this is not pure biomechanical data, elite rugby players reported increased muscle soreness over a 4-day period, suggesting underlying mechanisms between the two surfaces that could contribute to alterations in movement patterns, which may influence the risk of injury (153). This highlights how loading patterns when performing on synthetic turf compared to natural grass may be altered.

Shock Padding

As synthetic turf has been developed through multiple iterations, a recent addition has been the proposal of a layer of cushioning beneath the surface of the synthetic turf to act as additional shock padding (101). Effects of shock padding on movement patterns is yet to be fully understood, especially considering that shock pads are not common in synthetic turf installation and tend to be installed for professional sporting use (101). When completing a slalom agility test on natural grass and synthetic turf with a shock pad installed beneath the surface, the relative plantar pressure loading

was shown to differ between surface conditions (107). Specifically, there were significantly higher relative loads observed at the medial forefoot (9.8%) and lateral midfoot (15.5%) when comparing natural grass to the synthetic turf with shock pad system. It was shown that the shock pad was able to decrease the magnitude at the highest plantar pressure loading region, but that it was not the primary mechanism of change. It was hypothesized that if the shock pad was the primary mechanism, the change would have been constant across all areas of the foot. Therefore, alterations in loading of the foot appear to be in part due to the shock pad and in part due to the motion of the foot (107). This shows that there is an interaction effect between the shock pad and the movement patterns to alter overall loading of the foot which may have subsequent effects on loading of the lower extremity joints.

Synthetic turf configurations can vary in terms of the depth of the infill and the inclusion of a shock pad. In comparing three different configurations for FieldTurf synthetic turf, Duraspine ULTRA 42 (professional level with underlying shock pad), Duraspine ULTRA 50 (recreational level), and Duraspine ULTRA 60 (professional level without underlying shock pad), it was found that the configuration commonly used for recreational play provided the least impact absorption through reduced contact time and greater impact force during dynamic movements (90° cutting movement and a stop sprint task)(101). Given the high number of people who interact with the playing surface, this finding may be of some concern for athletes using these facilities. Furthermore, as it was found that impact force was significantly higher for the recreational configuration (ULTRA 50 infill thickness = 35 mm) compared to two professional configurations (ULTRA 42 infill thickness = 25 mm; ULTRA 60 infill thickness = 45 mm), it brings in

question the theory that the increased potential for impact absorption is directly correlated to increased potential for surface deformation is questioned. While the increased infill configuration supports this theory, the ULTRA 42 system had a similar total height (infill thickness + shock pad height) to the ULTRA 50 system, suggesting that the impact absorption is influenced more by the inclusion of a shock pad than by increased infill (101). Given that infill materials can move around during continued usage and a shock pad remains fixed beneath the surface, including a shock pad may serve as a more consistent option for maintaining the impact absorption properties of a synthetic turf installation. However, it would be interesting to know how the material of a shock pad changed over time with continued usage and the subsequent effects on its impact absorption properties. Overall, this study was able to conclude that synthetic turf systems can reduce the impact forces to improve the impact absorption properties of the surface through the inclusion of a shock pad.

However, it also has to be considered that for sports such as football, increasing infill may have a detrimental effect on factors such as ball roll and bounce, with the installation of shock pads shown to influence ball roll and bounce (88). The majority of shock pads are constructed using flexible foams, made from various plastics and rubbers to give the desired material properties. The development of shock pads must also consider environmental factors and these factors will affect the properties of the shock pad. For instance, shock pads constructed using open cell foams have the potential to fill with water where cold temperatures can cause this water to freeze thus altering the properties of the shock pad (88).

In a more mechanical approach to understanding the effects of shock pads, one study investigated the effect of different shock pad construction during linear running on synthetic turf (151). Although they used a human subject as opposed to mechanical tests to examine the different systems, the dependent variables in the study are related to the effects on the materials, not human movement. However, this study does give a good indication of the differences in shock pad construction and how this has the potential to effect human movement. Energy behaviors of two different shock pads (one rubber based and one foam based) were determined using a 'hysteresis energy ratio', which described the ratio of energy loss to input energy. Using this ratio, it was found that the rubber shock pad had less energy loss compared to the foam shock pad. It was postulated that this was due to the more elastic response of the rubber shred particles that made up the rubber shock pad. The increased intrinsic stiffness of the bonded rubber shreds, combined with lower volume of air voids, reduced the compressive strain of the surface and caused a stiffer response. This elastic buckling of the bonded rubber shred structure is the primary mechanisms by which the rubber shock pad absorbs impact energy, and then through the hysteresis properties of the material, energy is returned through unloading of the material. In comparison, the foam shock pad has open cells and a lower mass density due to the relative ratio between the volume of air and solid material. The generation of heat within the air pockets that are formed during the construction of compressed foam occurs due to the deformation of the cell walls within the material, which has been described as the mechanisms behind energy loss in compressed foam (99). Therefore, the initial compression of shock pads comprised of highly porous material is a consequence of air being forced from the open cells to

reduce the volume of the material with only a small amount of compression associated with compression of the solid particles. At this point, the intrinsic stiffness of the solid particles that make up the shock pad has a more prominent role in the resistance of further deformation. Rubber has the ability to deform via distortion as a result of the relatively higher Poisson’s ratio, which allows the regular void space to be filled. As the load increases, the void space is further reduced, giving the rubber shock pad stiffness properties associated with a solid block of rubber. However, the foam shock pad is constructed of flocculated particles to make up the solid particles making it more difficult to evaluate the further compression due to the non-linear nature of the deformation.

The conclusions that can be made, however, appear to suggest that foam based shock pads provide an increased potential for energy dissipation through the material structures that make up these type of shock pads. This could reduce the amount of force that propagates up the human body reducing the loading on the lower extremity joints. However, this does not consider effects on human performance when completing high load movements, where the loading of the surface may be different compared to material testing protocols. It also does not consider the comparative loads that will be placed on the surface by the athlete and how these differ between different aged athletes and athletes competing at different skill levels.

Footwear Mechanisms

The role of footwear during a dynamic movement is to transfer force from the surface to the athlete in a safe manner while maintaining performance (19, 134). The shoe-surface interface is an important consideration for injury mechanisms in terms of frequency and severity (117). In fact, given the importance of the shoe-surface interface

as a contributor to injury mechanisms, it has been argued that footwear can have a greater influence than the surface with regards to impact absorption (47).

In addition to alterations in the surface, it has also been shown that this interface can be influenced by changes in cleat configurations (109). Soccer cleats are primarily categorized to firm ground (FG), soft ground (SG) and artificial turf (AT) cleat configurations. However, for financial and perceptive reasons, players continue to wear cleat configurations for natural grass (FG and SG cleats) when playing soccer on artificial turf (109). These various cleat configurations have been investigated with regards to the shoe-surface interaction where the dependent variable examined are the traction characteristics (138). One such characteristic is the rotational resistance/friction of the shoe-surface interaction; where, it has been shown that cleats that increased rotational resistance significantly increased the risk of suffering an ACL injury (85). However, in comparing different cleat configurations in modern cleats, testing mechanical rotational traction revealed no significant differences in rotational traction while bladed FG cleats had the highest translational traction. This suggests that modern bladed cleats offer the potential to enhance performance without a concurrent increase in risk of injury (138). Similarly, no differences were found in joint moments at the ankle and knee joints. While there were no significant differences, there was a linear increase in joint moment in the frontal plane as the rotational traction moment increased when comparing soccer cleats to running shoes. This does suggest that if a shoe-surface interaction produced large enough rotational traction that this could increase the joint moment in the frontal plane, which could increase the risk of injury (138). In contrast, Müller found that SG and FG cleats had decreased horizontal foot translation and

increased ankle moment during a cutting movement, suggesting an increased loading on the body to potentially increase the risk of injury (110).

A fundamental component of this interface related to traction variables, that affects the risk of injury, is the frictional forces associated with the specific shoe-surface interface (117). Increases in frictional forces have been associated with increases in loading (36), suggesting that greater plantar pressure at a specific region of the foot could produce increased frictional force for the shoe-surface interface at that region, which could influence the risk of injury. The effect of cleat configuration on plantar pressure has been an area of investigation for soccer cleats. Comparing blade and stud cleat configurations, differences in plantar pressures were observed at medial aspects of the foot and at the heel. These increased plantar pressures at specific regions could precipitate injuries to that part of the foot and the transmission of force up the joint could be altered (19).

In terms of non-contact ACL injury, cleated footwear had been suggested to directly influence non-contact ACL injury mechanisms with shorter cleats being associated with a reduced risk of non-contact ACL injury (85). However, while this association can suggest that shorter cleats should be worn to reduce the risk of injury and that traction is a direct mechanism for non-contact ACL injury, movement pattern adaptations to changes in footwear construction and surfaces have previously been identified and therefore, may alter the biomechanical factors associated with non-contact ACL injuries (108).

An important note for many of the studies that have been cited in this section is that the differences in cleats used to compare the effects of cleat configuration may not

be solely restricted to outsole configuration. Many studies used different models of cleats to test differences in cleat configuration, but in doing this, other characteristics of the soccer cleats may have also differed such as the midsole density or upper material properties.

Biomechanics of Cutting Movements

Biomechanical analysis has been completed on a variety of cutting movements, with the main focus being on 45, 90, and 180° cutting movements (21, 40, 70, 103, 146). Ground reaction forces (GRF) that occur during the contact phase of human movements such as cutting and running are comprised of two peaks, an initial peak representing the initial contact between the athletes and the surface, which is passive in nature (42, 61, 118), and a global peak which contributes to the propulsion of the athlete and is regarded as an active force. It is this first peak, where the lack of active neuromuscular control may cause a large force to be transferred to the lower extremity joints increasing the risk of injury (42, 61, 118). In fact, biomechanical risk factors of ACL injuries were most closely associated with those during the first-peak during a cutting movement. This suggests that the risk of ACL injury is greatest during the loading phase (131), and thus altering the biomechanical markers at this time point may reduce the risk of injury (80).

Change of direction movement, commonly referred to as agility, is a performance attribute associated with success in sports such as soccer (63, 125) and American football (100). To change direction, an athlete must decelerate along the current direction before accelerating away in a different direction (67). As the cutting angle increases, changes in center of mass (COM) position and horizontal GRF suggest that

the deceleration and translation of the body also increase. However, many of these changes are not linearly correlated with cutting angle as the lower extremity and trunk joint patterns have been shown to differ between 45° and 90° cutting movements (68). Specifically, peak hip extensor and ankle plantarflexion moments were unchanged between 45° and 90° cutting movements; while, the peak knee extensor moments were increased, suggesting that the knee was the primary joint for absorbing the increased peak posterior GRF for the 90° cut. Furthermore, comparisons between 90° and 180° cuts have shown that internal knee adduction moment at initial contact during 180° cuts was significantly reduced compared to 90° cuts (40). In contrast, peak internal knee adduction moment was significantly higher in the 180° cutting movement (40). Approach velocity has also been shown to influence lower extremity biomechanics, with increases in peak internal knee adduction moment, knee flexion angle at contact, and both peak posterior and medial GRFs as approach velocity for a 45° cutting movement increased from 2.0 to 5.0 $m \cdot s^{-1}$ (146). Therefore, it can be suggested that both angle and velocity of the cutting movement will affect the joint pattern produced.

Cutting movements often warrant attention during analysis to the frontal plane due to frontal plane mechanisms being considered key risk factors for ACL injuries (70). During change of direction tasks, it is suggested that there is an increase in knee internal adduction loading in the frontal plane, which increases the risk of suffering a non-contact ACL injury (103). In addition, the increased knee adduction loading during cutting movement are also associated with increased peak lateral GRF (132), and increased knee abduction and internal rotation angles at contact (104, 132).

The biomechanical variables associated with successful performance of a cutting movement has also been shown to differ with the cutting angle (68, 157). One study found that with a shallower angle (45°), there were increases in mean hip power in the sagittal plane and peak ankle plantarflexor moment during the deceleration phase (70). These variables were significantly correlated to improved agility measured through the time to complete a standardized agility T-test. In comparison, for a larger cutting angle (90°), medial-lateral GRF impulse and mean hip power in the frontal plane were the key biomechanical markers correlated to the performance (70). The increase in performance due to medial-lateral GRF impulse can be explained by the increased redirection demands on the body, which requires an increase in medial GRF to be applied for a longer contact time when cutting at 90° compared to 45° (69). It has also been found that increases in hip frontal plane power cause the hip to adduct, which allows for increased trunk lean into the cut to enhance performance (68, 70). Overall, it can be summarized that 45° cutting movement performance can be attributed primarily to sagittal plane variables; while, 90° cutting movement performance can be more attributed to frontal plane variables.

In terms of predictors of peak knee adduction moment, the angle of the cutting movement again influences the biomechanical risk factors associated with peak knee adduction moment. For 45° cutting movements, it was found that the medial-lateral distance between the center of mass (COM) and center of pressure (COP) was a predictor of peak knee adduction moment (70). It was hypothesized that an increased medial body position relative to the foot results in the COP being positioned more laterally with respect to the overall COM and tibia COM. This in turn can increase the

internal knee adduction moment via an increased moment arm for the frontal-plane resultant GRF about the knee joint (70). For 90° cutting movements, hip internal rotation angle at initial contact was found to predict peak knee adductor moment where a reduced hip internal rotation angle resulted in increased peak knee adduction moment (70). It can therefore be suggested that an increase in hip internal rotation at initial contact could reduce knee loading and also improve performance (70). Furthermore, it has been found that during a 90° cutting movement, where there is increased deceleration and redirection requirements, changes in biomechanical variables across the lower extremity joints do not occur uniformly (68). It was found that there is an increased demand placed upon the knee joint in the sagittal plane during the deceleration phase (68). This has ramifications for the potential of suffering an ACL injury as peak knee extensor moments, peak posterior GRF, and increased quadriceps activations have been identified as contributors to anterior tibial shear force (129, 160), which has been an indicator of ACL loading (20, 97).

It has been reported that pivot tasks, where an athletes completes a 180° cutting movement, provide a realistic representation of the demands change of direction tasks during soccer matches (65). Replicating this movement during laboratory-based biomechanical studies, it has been hypothesized that this tasks presents a high risk of a non-contact ACL injury due to significant increases in knee abduction angle and loading at initial contact and at peak vertical GRF when compared to unanticipated 45° side-step cutting movement and drop-landing tasks (40). It has also been reported that during pivot and side-step cutting movements, knee alignment is constantly in an abducted position where there is an increased potential for strain placed on the ACL

(40, 59). This is particular relevant to the pivot task where subjects generated approximately 11° of knee abduction (40). When completing a pivot task, the foot at initial contact is placed perpendicular to the direction of motion on approach which may work to increase the knee abduction angle, especially considering the athlete must come to a complete stop before altering direction. This is in comparison to a 45° cutting task, where the athlete is only required to decelerate momentarily and not to a complete stop, thus altering the movement patterns and relative joint loading between tasks. Furthermore, dynamic change of direction movements have also been shown to increase the internal knee adduction moment, in particular during pivot tasks (40). Multi-directional movements produce increased loading in the frontal plane, highlighting their potential to increase the risk of injury. Specifically, dynamic valgus loading has been identified as a key risk factor for ACL injuries (58, 103), with increased knee abduction angle coupled with increased internal knee adduction moment shown via computer simulation to increase loading of the ACL (16). In fact, these variables have been identified as potential markers to predict athletes that will suffer an ACL injury (73). Additionally, decreases in knee flexion angle at initial contact and increased peak posterior GRF were also observed during the pivot task compared to the drop-landing task (40). A decreased knee flexion angle has been an associated risk factor for ACL injuries where increased quadriceps activation combined with straighter alignment in the sagittal plane has been proposed to increase the strain on the ACL (25, 155). Similarly, increases in posterior GRF have been highly correlated with proximal anterior tibia shear force, which in turn causes increased anterior displacement of the tibia and subsequent stress of the ACL (129, 160). A subsequent study attempting to identify the

effect of foot orientation during 90° and 180° cutting movements revealed that all subjects for all tasks produced a knee valgus orientation for both rear-foot and forefoot initial contact (41).

Effects of Cutting Velocity

During the deceleration phase of a cutting movement, the impact attenuation properties of the surface will influence lower extremity loading, which in turn will influence the proposed fundamental injury mechanisms for a non-contact ACL injury (81, 104, 123). Given that loading is related to deceleration, it can be suggested that an increased deceleration would result in increased loading of the lower extremity (146). Therefore, an increase in approach velocity would require greater deceleration during cutting movements, and therefore should result in increased loading. This notion has been supported by work that has directly compared changes in approach velocity during cutting movements and the effect of velocity on lower extremity loading (146). It has been found that peak internal knee adduction loading increased from 0.12 ± 0.17 $N \cdot m \cdot kg^{-1}$ at an approach velocity of $2.0\ m \cdot s^{-1}$ to $0.15 \pm 0.13\ N \cdot m \cdot kg^{-1}$ at $3.0\ m \cdot s^{-1}$. Significant increases are then seen as peak internal knee adduction loading increases to $0.58 \pm 0.55\ N \cdot m \cdot kg^{-1}$ at $4.0\ m \cdot s^{-1}$ and further significant increases to 1.14 ± 0.84 $N \cdot m \cdot kg^{-1}$ at $5.0\ m \cdot s^{-1}$ (146). These results highlight two concepts: firstly, that increases in loading are non-linear with increases in approach velocity suggesting that there may be a minimum threshold for approach velocity that causes loading under deceleration to be considered hazardous. Secondly, as approach velocity increases, so too does the variability in loading. In relation to injury risk, if the significantly increased knee adduction moments observed for an approach velocity of $5.0\ m \cdot s^{-1}$ for a 45° cutting

movement were to be extrapolated to higher approach velocities investigated in other works (132), it can be postulated to that such approach velocities could produce internal knee adduction loading at a magnitude that results in ACL injury (146). Furthermore, in a separate study examining a 45° cutting movement at approach velocity of 5.0 $m \cdot s^{-1}$ on synthetic turf, non-normalized peak knee abduction moments had a mean of 121.4 N·m, which is dangerously close to the previous cadaver studies which indicated that ACL damage can occur at 125 N·m (128, 138). In contrast, the increased loading may instead result in reduced performance (146). Further differences are seen in other lower extremity biomechanical markers for non-contact ACL injuries with varying approach velocity. Knee flexion angle at touchdown was one variable seen to differ, with a touchdown angle of 19.4° ± 4.21 at 5.0 $m \cdot s^{-1}$ seen to be significantly greater than the respective angles for 2.0 and 3.0 $m \cdot s^{-1}$ of 14.9° ± 4.11 and 17.0° ± 3.01 (146). Another variable seen to differ was peak posterior GRF, which significantly increased for approach velocities of 2.0 $m \cdot s^{-1}$, 3.0 $m \cdot s^{-1}$, 4.0 $m \cdot s^{-1}$, and 5.0 $m \cdot s^{-1}$ with respective values of 5.0 $N \cdot kg^{-1}$ ± 1.1, 7.2 $N \cdot kg^{-1}$ ± 1.3, 10.3 $N \cdot kg^{-1}$ ± 2.4m and 12.9 $N \cdot kg^{-1}$ ± 2.5.

Although there are a limited number of studies that have directly compared approach velocities, similar studies using different approach velocities can also be compared (86, 132). For example, peak valgus loading was found to be 0.23 $N \cdot m \cdot kg^{-1}$ using a 3.5 $m \cdot s^{-1}$ approach velocity with a cutting angle range of 35° to 60° (86) compared with a 5.5 – 7.0 $m \cdot s^{-1}$ approach velocity, with a similar cutting angle range of 35° to 55°, generated peak valgus loading of 1.2 $N \cdot m \cdot kg^{-1}$ (132). This again supports the notion that approach velocity can have a significant effect on peak knee valgus loading. However, it should be noted that even for a set approach velocity, variability of peak

valgus loading is high, highlighting that individual differences may exist with certain individuals more susceptible to increased valgus loading than others (132).

The effect of velocity has also been shown to alter lower extremity biomechanics during linear running. It is well researched that as running speed increases from walking and running into sprinting, the shape and magnitude of the vertical GRF changes (78, 84). Furthermore, it has been established that as locomotion velocity increases within a given task, there is an increase in the magnitude of the vertical GRF (35, 78, 84, 111, 116). As running speed increases, leg spring stiffness has also been shown to increase (3, 93, 106), with knee joint stiffness shown to be the key mechanism behind alterations in leg stiffness (3, 46, 137). In relation to leg stiffness, lower extremity joint kinetics are also seen to be influenced by running velocity. As running velocity is increased, increases in sagittal plane ankle and knee moments and mechanical power outputs have been observed (3, 137). When comparing slow running (2.61 ± 0.81 m·s^{-1}) to fast running (6.59 ± 0.24 m·s^{-1}), peak ankle extension moment significantly increased from 2.45 ± 0.46 N·m·kg^{-1} to 3.43 ± 0.49 N·m·kg^{-1}, while peak knee extension moment significantly increased from -1.97 ± 0.45 N·m·kg^{-1} to -2.98 ± 0.37 N·m·kg^{-1}. During the transition from running to sprinting, ankle mechanical power for energy generation increased from 61.7 ± 8.2 J to 106.2 ± 15.7 J (137). Furthermore, mechanical energy absorbed at the knee joint decreases from 43.2 ± 10.1 J when running at 4.0 ± 0.4 m·s^{-1} compared to 11.4 ± 6.9 J when sprinting between 7.4 – 8.1 m·s^{-1} (137), highlighting that altered knee stiffness with increase running speed. Although this research examining linear running is not directly related to cutting movements, what it does highlight is the ability of running velocity to influence lower extremity biomechanics related to impact

loading. If running velocity during linear motion has this significant of an effect on sagittal plane lower extremity biomechanics, it is reasonable to think that the same concepts can apply to change of direction movements.

Surface-related Injury Prevalence

In recent years, there has been an increase in the number of installations and the usage of synthetic pitches across the world. At a recreational level, the reduced financial cost of maintaining pitches in addition to the ability to use the pitches in most weather conditions have been the driving reasons behind the surge in their usage (62). However, at the professional level, despite various governing bodies approving the use of synthetic turf in professional competition (9, 10, 12), there has been increased skepticism among players for the use of synthetic turf with concerns regarding the risk of injury (13, 15) and negative effect on performance (11).

There have been many studies assessing the relative risk of suffering an injury on different playing surfaces in soccer. In examining professional soccer injuries in various leagues across Europe, differing conclusions have been drawn from epidemiological data based on injury prevalence in soccer. In professional soccer in Norway, although no significant differences were found for the injury rates between natural grass and synthetic turf, a strong trend was seen for an increased risk of injury at the knee and ankle joints during competition (24). This suggests that when players perform with maximal effort during competition, this could increase their risk of suffering an injury on synthetic turf. This is supported by Ekstrand et al. (51), who found that there is a significant increase in ankle sprain injuries on synthetic turf, but that the overall prevalence of injury remains constant. However, in a recent study of professional

soccer in Italy, it was found that the overall incidence rate of injury does not change when comparing third-generation synthetic turf and natural grass (87). It should be noted that for all of these studies, the authors give the breakdown of injuries as the type of injury suffered (e.g. sprain, fracture, etc.) and did not give the mechanisms behind each injury. This notion is further supported by research using non-elite athletes. Although the overall incidence rate of injury when comparing synthetic turf and natural grass was found to be comparable for high school athletes competing in a range of sports, the types of injuries were found to differ between the two surface conditions (107). Specifically, it was found that there was an increase in the proportion of non-contact injuries suffered when playing on synthetic turf. To gain a more comprehensive understanding of this topic, an in-depth longitudinal study assessing the specific types of injuries and how they occur in professional soccer with respect to the synthetic turf versus natural grass would give a better indication of the relative risk of suffering specific types of injures.

This type of study has been conducted in a comparable sport involving dynamic movements, using professional American football athletes competing in the National Football League (NFL). It was found by the NFL Injury Surveillance System that between the 2000 and 2009 seasons there was a 67% higher incidence rate of ACL injuries on third generation synthetic turf than natural grass (72) while the rates of knee and ankle sprains in general were 22% greater on synthetic turf, with ankle eversion sprains 31% higher (72). ACL injuries are devastating knee injuries that can force athletes to retire from their sport, or allow them to return but at a reduced level of play compared to their pre-injury abilities (34, 135). Suffering an ACL injury can also

increase the risk of suffering from a future ACL injury for elite performers (48), while for the general population there is a reduced risk of a future ACL injury on the same limb (94). This could be due to an increased exposure following an injury for this population that is required for training (77); whereas, following an ACL injury in the general population there is a reduction in activity levels thus reducing the risk of a subsequent injury (4). ACL injuries can also result is further knee injuries other than re-injury of the ACL (54, 149) and ultimately lead to knee osteoarthritis (92, 148, 150). This highlights that as synthetic turf has a significant effect on the rate of lower extremity injury for elite athletes performing dynamic movements, there is an increased potential to cause career-ending injuries and further movement impairments late in life. Although recent NFL epidemiological data contrasts a previous epidemiological study of the NFL, where it was found that there was no effect of synthetic turf on ACL injury incidence (30), it is worth noting that this study used injury data between the 1994 and 1998 seasons where first generation synthetic turf was the preferred synthetic turf, which has a different composition and construction to modern synthetic turf.

With soccer being a sport that is becoming increasing popular for females (14), increasing work has been done to understand injury rates within females. Within elite female soccer in Sweden, the overall rate of injury has been shown to be 24 injuries per 1000 hours of competition and 7 injuries per 1000 hours of training with 23% of injuries occurring at the knee and 15% being defined as major injuries requiring more than 30 days to return to play (53). Differences in ACL injury rates have also been observed between men and women. Over a five-year period between 1989 – 1993, it was found that in collegiate soccer, females had a significantly higher risk of suffering an ACL

injury (5). This notion was solidified by a more comprehensive longitudinal study where it was found that females were four times more likely to suffer a non-contact ACL injury (2). These two studies highlight that despite advancements in surfaces, footwear, and training regimes, the overall rate of non-contact ACL injuries is greater in females and that injury rates have remained constant, suggesting that female soccer players have a higher predisposed risk of suffering an injury.

There are different mechanisms that cause an ACL injury that are usually divided into two categories: contact injury and non-contact injury. Following a longitudinal study examining ACL injury rates in collegiate soccer players over 13 seasons, it was found that non-contact ACL injuries accounted for 67% of all female ACL injuries while accounting for 58% of all male ACL injuries (2). Furthermore, general non-contact injuries make up 21-61% of injuries to the lower extremity in dynamic sports such as soccer (2, 156) and American football (30, 102, 124). It has been hypothesized that non-contact injuries occur as a result of foot fixation to the surface, which causes potentially injurious forces at joints of the lower extremity (121). Specifically, rotational forces during a movement with the foot in a fixed position has been postulated as a mechanism for ACL injuries (85), while lateral forces have been associated with ankle inversion injuries (26).

ACL Injury Mechanisms

ACL injuries have been estimated to cost over $2 billion in the USA per year to treat approximately 175,000 ACL reconstruction surgeries (64). Complete ACL ruptures can also have repercussions to other structures at the knee joint including menisci damage and osteoarthritis (57, 75, 161). Specifically in soccer, it was found that 84% of

all ACL injuries were attributed to no contact with opposition players and change of direction movements accounted for 63% of ACL injuries (56). The act of pivoting or cutting has also been found to be the primary mechanism for 48% of all ACL injuries within soccer (126). The ability to prevent non-contact ACL injuries may be linked to our ability to understand injury mechanisms (8). There has been a greater focus on understanding these mechanisms as the concern regarding ACL is high due to the consequences of suffering an injury for both future performance and health, long recovery time, expensive treatment, and relatively large incidence rates (161).

To determine the mechanisms behind ACL injuries, a variety of methods have been used. One approach is to use questionnaires and athlete recall, given to athletes who suffered an ACL injury, to try and better understand athlete movement(s) that caused injury. One such study revealed that non-contact mechanisms accounted for 72% of ACL injuries (27). Another method employed is the analyses of videotapes of ACL injuries (29, 83). Although a partially subjective method, the use of videotapes enables researchers to view the exact movements that caused an injury compared to laboratory studies which use movement to replicate these movements. Through the usage of this method, two mechanisms have been proposed: valgus collapse (increases in knee abduction angle) where valgus motion of the tibia causes the knee to collapse inwards, and anterior tibial shear when the knee is close to full extension during the initial loading phase when completing a sudden deceleration movement or landing (27, 28, 39, 83).

Additionally, to further understand the effect of forces produced by the body on the ACL, cadaver studies have been conducted. It was found that the primary isolated

contributor for ACL strain was anterior shear force on the proximal end of the tibia (20, 97). However, the combination anterior shear force at the proximal end of the tibia with knee valgus moment increased the strain on the ACL further. This notion was supported by further research where ACL loading was found to increase as knee flexion decreased when ACL loading was due to anterior shear force in combination with knee valgus, knee varus or knee internal rotation moments (97). It appears that anterior shear force at the proximal tibia is the primary mechanism for ACL strain, where valgus/varus and internal rotation moments can cause further increases in loading. Based on this understanding of ACL loading, researchers have postulated that this can translate biomechanically to a lack of knee flexion, high quadriceps muscle activation, and high posterior GRF. The insertion site of the quadriceps muscles on anterior proximal tibia result in this muscle group producing anterior shear force at the proximal tibia. Subsequently, it has been found that ACL loading is increased by quadriceps muscle force with the knee at 0° to 45° flexion (6, 22, 49, 89) where further increases in knee flexion angle decrease ACL strain (6, 22, 50, 90). It has also been suggested that a knee flexion angle of 20° combined with a simulated quadriceps muscle force of 4500 N has the potential to cause ACL loading with six out of 13 cadaver knees showing rupture (43). This result suggests that individual differences between ligaments may exist, a concept supported by a correlation between ACL size and injury risk (144).

Research has also shown that decrease in knee flexion angle is related to increased loading of the ACL. Multiple mechanisms for why decreased knee flexion increases ACL loading have been proposed. One proposal is that decreased knee flexion increases the patella tendon-tibia shaft angle, which determines the anterior

shear force at the proximal tibia (119, 145). An increase in patella tendon-tibia shaft angle was related to increases in anterior shear force at the proximal tibia, highlighting the potential for this mechanism to influence ACL loading (119). Decreased knee flexion has also been associated with increases in ACL elevation and deviation angles (91). This elongates the ACL causing a larger resultant force along the longitudinal axis of the ligament suggesting the ACL plays an important role in stabilizing the knee joint at lower angles of knee flexion.

Peak ACL loading has been shown to occur at impact peak vertical GRF following initial contact during rapid deceleration tasks via *in vivo* testing of ACL loading (37). It has also been shown that impact peak vertical GRF and peak posterior GRF occur almost simultaneously in a stop-jump task (160). Consequently, peak posterior GRF may be a key mechanism for non-contact ACL injury. Conversely, another line of research using *in vitro* analysis suggests that such mechanisms may be unable to injure the ACL in isolation. When performing cutting movements, sagittal plane mechanisms were unable to load the ACL in isolation and loading in the frontal plane via knee valgus moments were required to load the ACL to a level where a rupture could occur (103). In vitro analysis has also revealed that ACL loading can occur when pure knee valgus moments are applied (154). Additionally, the application of knee valgus moments to a knee joint already having anterior tibial force applied increased the ACL loading with the knee flexion angle below 20° while the greatest loading occurred with internal rotation moments applied (97).

Gender differences in non-contact ACL mechanisms has been the focus of a large proportion of research for this injury. It has been consistently reported that female

athletes face a higher risk of suffering a non-contact ACL injury than male athletes (5, 23, 29, 136, 142, 143, 152). This increased risk has been linked to intrinsic factors such as a larger quadriceps angle in women and, when standardizing for body weight, a smaller ACL (141). Biomechanical studies have also observed differences between men and women during dynamic movements commonly associated with non-contact ACL injury. These differences include smaller knee flexion angles (38, 95), particularly after the age of 13 years (159), larger peak posterior GRF (38, 158), greater peak knee valgus angle (58, 59, 95), greater knee valgus motion (58) in dynamic movements including landing, cutting, and stopping tasks. Due to the increased rate of injury in female athletes, it is thought that these differences in biomechanical variables also act as mechanisms for sustaining a non-contact ACL injury. This notion is supported where it has been found that, following a prescreening of female athletes, those that went on to suffer ACL injuries had increased knee valgus angles at landing, increased knee adduction moments and increased GRF (73).

The role of leg dominance in relation to ACL injuries presents conflicting research. Some research has found that leg dominance does not influence the risk of a non-contact ACL injury (98, 115). However, research using a soccer-specific population found that, while for the entire sample (males and females) there was no effect of leg dominance, there was an increased risk for female athletes suffering a non-contact ACL injury on their non-dominant leg (32). Contrasting these works, it has been found that ACL injuries were significantly more frequency on the right leg during soccer, regardless of limb dominance (126).

Various forms of neuromuscular training have been proposed and implemented in an attempt to reduce the amount of movement in the frontal and transverse planes during cutting movements and try to ensure sagittal plane movement (44, 96, 113). However, ACL injury rates have remained constant despite these programs being developed (30). Specifically, various interventions and methods that have been proposed and implemented in an attempt to reduce the risk of ACL injuries in soccer have had little success on the overall rate of ACL injuries, as rates have remained unchanged (2). Furthermore, while there has been research reporting that these programs also enhance the performance of dynamic movements (112), conflicting research has also shown that they reduce an athletes' agility (147). This suggests that athletes may not be prepared to partake in such programs, where their focus is solely on performance gains, meaning that alternative methods of reducing the risk of injury must be found.

In all, there is a wealth of literature on the mechanisms behind non-contact ACL injuries with the primary mechanism appearing to be related to anterior shear force, with frontal plane loading also contributing as a secondary mechanism. What does appear to be apparent that a combination of mechanisms is required to injure the ACL with no single mechanism applied in isolation having the potential to result in injury.

Chapter 3

Methods

Participants

Twelve to fifteen active, healthy recreational American football and soccer players between the ages of 18 to 30 years, with at least three years of previous American football or soccer playing experiences at a competitive level (high school and upwards), volunteered to participate in the study. Participants also completed at least three bouts of physical activity per week. In-depth demographic information on subjects can be found in Appendix C. Exclusion criteria for the study were set as follows: if they had a history of serious lower extremity injury (including but not limited to ligament rupture, bone fractures, or bone dislocation), had a current lower extremity injury, or if a subject answered “yes” to any question of the Physical Activity Readiness Questionnaire (PAR-Q). Each subject completed two testing sessions, each lasting about 120 minutes. Before the commencement of each testing session, subjects provided written informed consent via the informed consent form approved by the University of Tennessee Institutional Review Board. Participants were recruited through discussions with coaches of the college American football and UT club soccer teams, flyers posted in locations across UT campus, announcements at undergraduate biomechanics and physical education and activity program classes, e-mail and word of mouth.

Power analysis using GPower (55) was conducted in order to determine the number of subjects required in order for the study to have sufficient power. Peak vertical GRF (31, 33), peak posterior GRF (33, 40, 146) and peak knee extension moment (17,

40, 138) were the variables that were used to complete the power analysis. This analysis resulted in a range of six to eighteen participants that were required for the study to achieve a power of 0.80 and an alpha level of 0.05.

Equipment

Shoe: Subjects wore a pair of American football cleats (Nike Vapor Untouchable Pro, Nike, Beaverton, OR) during testing sessions.

Turf: A 2" monofilament synthetic turf with 1/2" stitch gauge (PowerBlade Shaw Industries, Dalton, GA) was used as the turf for the study. For the turf surface condition, the turf was affixed to the lab surface or force platform using double-sided tape. For the turf system with the shock pad, the shock pad was affixed to the floor first with the double-sided tape, and the turf piece was secured on top of the shock pad with double-sided tape (Figure 1). The turf (with and without shock pad) mounted on top of the force

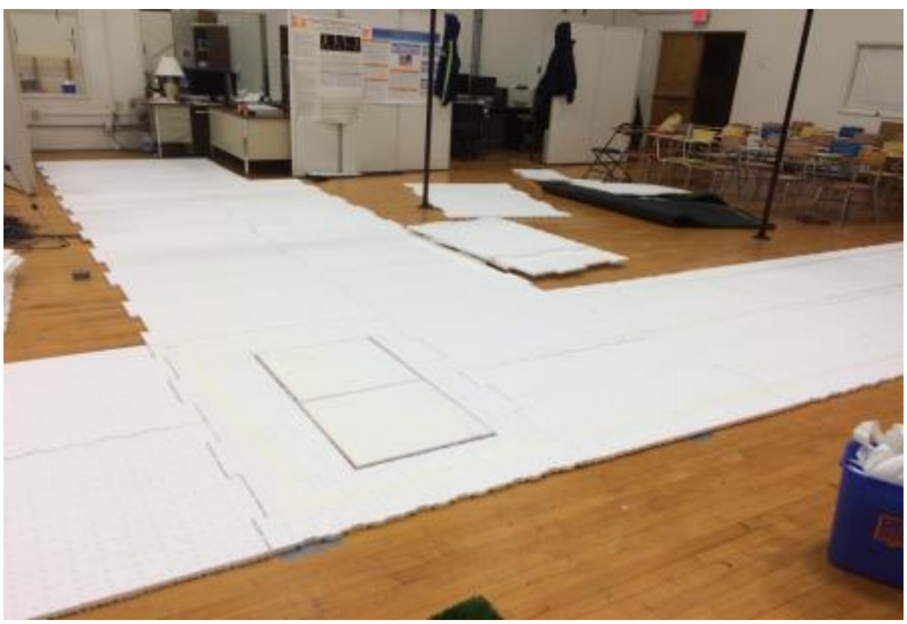

Figure 1: Shock pad set-up

platform was cut separate from the other pieces of turf. This allowed easier detection of whether the subject landed within the boundary of the force platform and to allow access of the calibration wand between the force platform and floor. The turf was first infilled with sand to a depth of 15.1 mm and then evenly distributed using a stiff brush. Crumb rubber was then added and evenly distributed using a stiff brush for a total depth of 32.1 mm. The turf piece on the approach for cutting movement was also infilled to ensure that subjects had a minimum of two to three steps before landing on the force platform. This ensured subjects had proprioceptive awareness of and adaptation to the surface conditions before landing on the force platform, reducing the chance of movement alterations on the approach. The exit turf pieces were also infilled for two to three steps. The infilled turf was checked to ensure the depth across a minimum of nine locations on the turf atop the force platform using a 3-pronge surface depth gauge (Canadian Playground Advisory Inc., Canada). If the infill height was not between 30 – 32 mm, then the surface was re-brushed to redistribute the infill material. In the shock pad condition, the shock pad was placed under the entire length of the approach and exit pieces.

Shock Pad: A foam based shock pad (POWERBASE/YSR, Brock International, Boulder, CO) was used to create a turf system with a shock pad condition. Double-sided carpet tape was used to attach the shock pad to the floor and to attach the turf to the shock pad. A separate piece of shock pad was attached to the top of the force platforms.

Biomechanical Equipment: A 12-camera infrared motion capture system (240 Hz, Vicon Motion Analysis, Inc., Oxford, UK) was used to collect three-dimensional (3-D)

marker data. Retroreflective anatomical markers were placed at anatomical landmarks bilaterally at the acromion process, iliac crest, greater trochanter, medial and lateral femoral epicondyle, medial and lateral malleoli, 1st and 5th metatarsal head, and the toe (defined as the most anterior aspect of the shoe). These landmarks were found via manual palpation and single markers were placed at these locations. Four tracking retroreflective markers were also mounted on a semi-rigid thermoplastic shell and attached to the trunk, pelvis, thigh, and shank segments. To track the motion of the foot, four discrete markers were affixed to the posterior and lateral aspects of the cleat (Figure 2). Subjects then completed a static trial, where they were required to maintain a pose whilst standing on the force platform, following which the anatomical markers

Figure 2: Marker set for static trial

were removed, leaving just the tracking markers for dynamic data collection. A single force platform (BP600600, 1200 Hz, Advanced Mechanical Technology, Inc., Watertown, MA 02472, USA) was used to measure GRFs. This force platform was integrated into the Vicon system to allow simultaneous collection of both 3-D kinematic and GRF data. Approach velocity during the cutting movements was measured and monitored using two pairs of photocells (Lafayette Instrument Co., Model 63501 1R) connected to an electronic timer (Model 54035A, Lafayette Instrument) and placed 1.5 m apart at the subject’s shoulder height with the final timing gate placed 0.7 m before the front edge of the force platform.

Protocol

Subjects were required to complete two testing sessions, following an identical protocol with the only difference being the change in the surface condition. Additionally, when subjects entered the laboratory for the first time, they filled out and signed the informed consent form, an information sheet and the PAR-Q. The information sheet (Appendix B) asked questions regarding age, American football or soccer experience, position, times played per week, times exercised per week other than football or soccer, and previous injury history. Subjects were asked to wear a tight fitting top and spandex shorts. If subjects arrived without this apparel, spandex shorts were provided by the laboratory.

Subjects began by completing a self-regulated warm-up for five minutes. Random assignment of the movement conditions determined the order in which subjects complete the protocol. The approach velocities for the 90° cut conditions were 3.0 ± 0.3 and 4.0 ± 0.4 m/s. Counterbalance randomization of the surface condition also

determined the order of the synthetic turf systems. For the 90° cutting movement, subjects were instructed to accelerate from a starting cone towards the force platform. Using their dominant foot (defined as dominant leg for kicking), they then decelerated and planted their dominant foot within the boundary of the force platform before changing direction 90° and performed a cutting movement to the contralateral side of the limb on the turf runway (Figure 3). The approach velocity was monitored to ensure that subjects' meet the desired range. For each approach velocity, subjects were given a minimum of three trials prior to data collection to adjust the starting position to allow subjects to complete the movement while ensuring movement patterns were consistent and to give verbal feedback regarding approach velocity to ensure they were within the approach velocity range. Additional practice trials were allowed if required to meet both

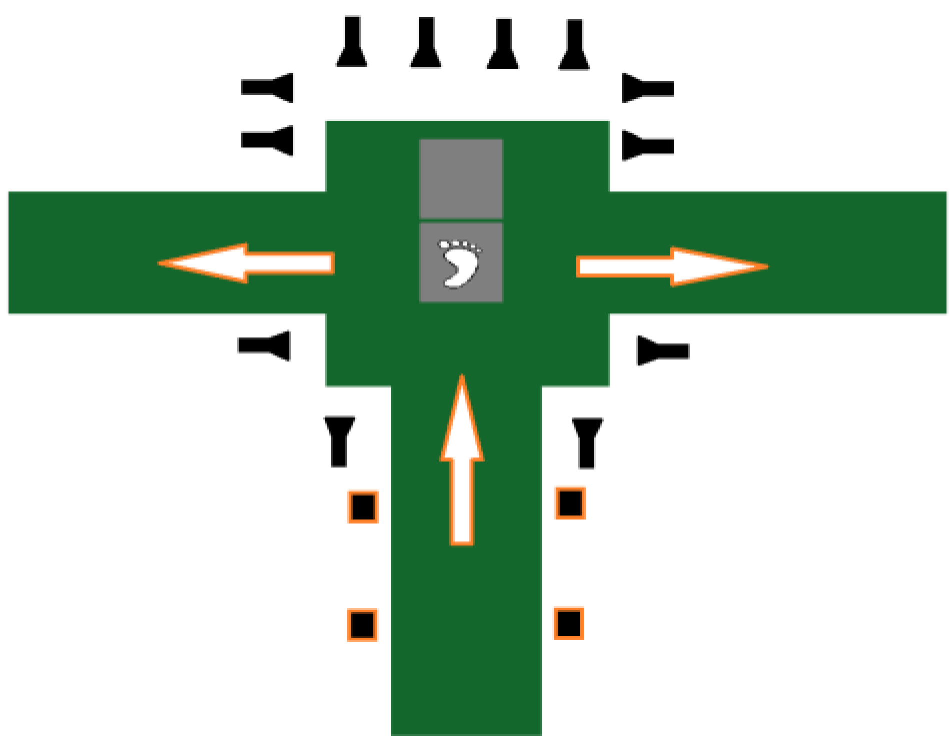

Figure 3: Laboratory set-up showing the protocol for the cutting movements

these requirements. When the subject felt comfortable, they completed five successful trials of the cutting movement on each surface condition. A successful trial was defined as having the foot land within the boundary of the force platform, a consistent movement pattern, and the approach velocity being within the set range. Between trials of each condition, participants were given a minimum of 60 seconds or as much rest as required.

Mechanical Testing

In addition to human testing, industry standard mechanical tests were completed on the two synthetic turf systems. The first test was the F355 A missile test (7), where the A missile was dropped three times on four locations across each the surface conditions and G_{MAX} and HIC values were measured. The last two drops for each location was recorded and averaged, with the average for each of the four locations then averaged.

The second test was the F355 E missile test, where the missile was initially dropped from a height of 1.3 m. The drop height was then altered until a HIC value of greater than 1000 was recorded. At this drop height, three drops were then performed across four locations.

Data Processing and Analysis

Synchronized marker coordinate and force platform data were imported to Visual3D to compute three-dimensional lower extremity joint kinematics and kinetics (version 5, C-Motion, Inc.). Marker coordinates and force platform data were filtered using a fourth order zero-lag Butterworth low-pass filter at a cutoff frequency of 12 Hz for both joint kinematics and kinetics calculations. Force platform data were filtered at

100 Hz for GRF variables. An X-Y-Z Cardan sequence was used to determine segment reference frames and the right-hand rule was used to determine positive joint rotations and moments. Ground reaction forces were normalized to body weight (BW). ROM was calculated at contact angle to peak angle. Joint moments were expressed as internal moments and normalized to body mass (Nm/kg). Joint powers were calculated as the product of joint moment and angular velocity. The kinematic, kinetic, and energetic variables of interest were ankle, knee, and hip joint ROMs, peak moments and power in the sagittal and frontal planes, and peak vertical and horizontal GRFs. Variables were defined by the phase of the stance phase cutting movement, with loading variables occurring during the deceleration phase of the cut and push-off variables occurring during the acceleration phase.

A 2 x 2 (Turf x Approach Velocity) repeated analysis of variance (ANOVA) was used to statistically analyze the effects of the two surface conditions and the two approach velocities on selected biomechanical variables. Post hoc comparisons were performed when a significant interaction of surface and movement or a surface and movement main effect were found using a pair-sample t-test using a Bonferroni-adjusted *p*-value of 0.0125.

Chapter 4

The Effects of Synthetic Turf Systems With and Without a Shock Pad on Lower Extremity Biomechanics During a 90° Cutting Movement with Differing Approach Velocities

Introduction

Synthetic turf pitches have become increasingly common across the world, largely due to their all-weather ability (19). However, despite the approval by many sport governing bodies for their use (3-5), concerns exist from elite players, particularly from an injury perspective (6, 7). The notion that synthetic turf increases the risk of injury has been well researched with conflicting findings. While some have found that the relative risk of an injury is comparable between synthetic turf and natural grass (9, 16, 27), others have found that the risk of injury increases on synthetic turf (21). In particular, it has been shown that the risk of suffering anterior cruciate ligament (ACL) or ankle sprain injuries is higher on synthetic turf (16, 21).

Synthetic turf has been investigated regarding its impact attenuation ability compared to natural grass with contrasting findings (18, 35, 36). It is suggested that the thickness and compliance of a surface are related to the maximal displacement of a surface, which is hypothesized as being directly related to a surface's ability to absorb impact forces (31). Therefore, increasing the thickness of a surface would increase the potential for impact attenuation, and this in turn increases the ability of the surface to absorb impact forces (31, 35).

A recent method for improving impact attenuation and injury prevention of synthetic turf is the addition of an underlying shock pad (28). Specifically, the original intent of shock pads was to reduce the Head Impact Criteria (HIC) but there were

concerns that they could potentially increase the risk of lower extremity injury. As shock pads are not common in synthetic turf installations, their effects on movement patterns have not been extensively studied and are not fully understood. In comparing three different synthetic turf configurations, it has been shown that the ULTRA 42 (turf with underlying shock pad) configuration had significantly reduced impact forces compared to the ULTRA 50 (turf only) configuration during a stop sprint task (28). Given that the ULTRA 50 configuration had an increased infill thickness compared to the ULTRA 42 configuration but the total height (infill thickness + shock pad height) was similar between the two, it can be suggested that the inclusion of a shock pad has a greater influence on the impact absorption properties than infill thickness (28). The material that shock pads are constructed of has also been shown, from a mechanical perspective, to potentially effect the impact absorption properties of the surface (38).

One key component that determines lower extremity loading during a cutting movement is the approach velocity. Increased deceleration as a result of increased approach velocity resulted in increased loading (37). As approach velocity increases, it has been shown that both sagittal and frontal plane variables are altered. With approach velocity increasing from 3.0 to 4.0 to 5.0 $m \cdot s^{-1}$, peak internal knee adduction moment increased from 0.15 $\pm$ 0.13 to 0.58 $\pm$ 0.55 to 1.14 $\pm$ 0.84 Nm/kg while peak posterior GRF increased from 7.2 to 10.3 to 12.9 N/kg (37). The results from this study also suggest that increases in as loading increases as approach velocity increase, a minimum threshold may exist for an approach velocity to causes potentially hazardous loading under deceleration and that increasing approach velocity increases the variability of loading. Furthermore, given that the higher approach velocities are

commonly experienced in game situations, high velocity cutting movements could produce internal knee adduction moments with a magnitude that results in ACL injury (37). Intra-study comparisons of cutting movements using similar cutting angles but different approach velocities support the notion that increased loading occurs with increased approach velocity (26, 32). Although not directly related to cutting movements, changes in linear running velocity also reveal differences in loading with increases in vertical GRF (12, 23, 25, 29, 30), sagittal plane ankle and knee moments (1, 33), and knee joint stiffness (1, 15, 33).

Currently, there is a limited amount of research investigating the difference in movement patterns from a biomechanical perspective that result from changes in synthetic turf properties. Specifically, there is a lack of research that has examined the effect of underlying shock pads on dynamic sporting movements that have previously been associated with lower extremity injury and the effect that different approach velocities have on lower extremity joints. Furthermore, the effects of increasing approach velocity on both impact related and performance biomechanical variables during 90° cutting movement are currently unknown. Therefore, the purpose of this study was to examine the effects of synthetic turf systems with and without an underlayment shock pad and different approach velocities for a 90° cutting movement on impact attenuation related and performance related kinematics, kinetics, and energetics.

It was hypothesized that: 1) The synthetic turf system with an underlying shock pad would produce lower GRFs, and decreased lower extremity sagittal and frontal plane moments, powers, and range of motion (ROMs) compared to the synthetic turf system

without an underlying shock pad; 2) An increased approach velocity would cause increases in GRFs and lower extremity sagittal and frontal plane moments, powers, and ROMs regardless of types of turf systems tested.

Methods

Participants

Twelve active and healthy recreational male American football and soccer players (mean ± SD age: 21.9 ± 2.7 years, height: 185.1 ± 6.3 cm, mass: 78.5 ± 9.4 kg) with at least three years of previous American football or soccer playing experiences at a competitive level (high school and upwards), volunteered to participate in the study. Participants were also required to have completed at least three bouts of physical activity per week. The required number of participants was estimated via a power analysis using GPower (17). Peak vertical GRF (10, 11), peak posterior GRF (11, 13, 37) and peak knee extension moment (8, 13, 34) were the variables that were used to complete the power analysis, resulted in a minimum of six to eighteen participants that were required for the study to achieve a power of 0.80 and an alpha level of 0.05. Participants were excluded from the study if they had a history of ACL injury or had a current lower extremity injury. Participants provided written informed consent, which was approved prior to testing by the Institutional Review Board at the University of Tennessee, Knoxville.

Instrumentation

A 12-camera infrared motion capture system (240 Hz, Vicon Motion Analysis, Inc., Oxford, UK) was used to collect three-dimensional (3-D) marker data. Retroreflective anatomical markers were placed at anatomical landmarks bilaterally at

the acromion process, iliac crest, greater trochanter, medial and lateral femoral epicondyle, medial and lateral malleoli, 1^{st} and 5^{th} metatarsal head, and the toe (defined as the most anterior aspect of the shoe). These landmarks were found via manual palpation and single markers were placed at these locations. Four tracking retroreflective markers mounted on a semi-rigid thermoplastic shell and were also attached to the trunk, pelvis, thigh, and shank segments. To track the motion of the foot, four discrete markers were affixed to the posterior and lateral aspects of the cleat (Figure 2). Participants then completed a static trial following which the anatomical markers were removed, leaving just the tracking markers for dynamic data collection. A force platform (BP600600, 1200 Hz, Advanced Mechanical Technology, Inc., Watertown, MA 02472, USA) was used to measure GRFs. This force platform was integrated into the Vicon system to allow simultaneous collection of both 3-D kinematic and GRF data. Approach velocity during the cutting movements was measured and monitored using two pairs of photocells (Lafayette Instrument Co., Model 63501 1R) connected to an electronic timer (Model 54035A, Lafayette Instrument) and placed 1.5 m apart at the participant's shoulder height with the final timing gate placed 0.7 m before the front edge of the force platform.

Participants wore American football cleats (Nike Vapor Untouchable Pro, Nike, Beaverton, OR) during testing. A 2" monofilament synthetic turf with 1/2" stitch gauge (PowerBlade Shaw Industries, Dalton, GA) was installed in the laboratory for testing. For the turf only condition (TURF), turf was affixed to the laboratory floor via double-sided carpet tape. For the turf on the force platform, a square piece of turf was cut separately and affixed to the force platform using double-sided carpet tape. The turf was

first infilled with sand to a depth of 15.1 mm then with crumb rubber to a total depth of 32.1 mm. For the turf condition with a shock pad (PAD), double-sided carpet tape was used to attach the shock pad to the floor and to attach the turf to the shock pad. A separate piece of shock pad was attached to the top of the force platform. A foam based shock pad (POWERBASE/YSR, Brock International, Boulder, CO) was used in turf with a shock pad condition (Figure 1).

Testing Protocol

Participants attended two testing sessions, with an identical testing protocol with the only difference being the change in the turf condition. All participants wore a tight-fitting top and spandex shorts. Participants began by completing a self-regulated warm-up for five minutes on a treadmill followed by body height measurements.

Two turf conditions were first randomized to determine their testing order (testing session; seven with TURF first and five with PAD first). Approach velocity was then randomized during each testing session. The approach velocities for the 90° cut conditions were 3.0 ± 0.30 (SLOW) and 4.0 ± 0.40 m/s (FAST). These approach velocities were based on both pilot work and previous work, which found no difference in peak knee adduction between 2.0 and 3.0 m/s while task achievement decreased at 5.0 m/s suggesting it was too fast for consistent movement patterns (37). Participants were instructed to accelerate from a starting cone towards the force platform. Using their dominant foot (defined as dominant leg for kicking), they then decelerated and planted their dominant foot within the boundary of the force platform before turning 90° and performed a cutting movement in the direction of the contralateral limb on the turf runway (Figure 3). For each approach velocity, participants were given a minimum of

three practice trials prior to actual data collection to ensure participants could complete the movement with consistent movement patterns. Verbal feedback regarding approach velocity was provided to ensure they were within the approach velocity range. Additional practice trials were allowed if required to meet both these requirements. When the participant felt comfortable and consistently replicate the movement, they completed five successful trials of the cutting movement on each surface condition. A successful trial was defined as having the foot land within the boundary of the force platform and the approach velocity being within the respective set range. Between trials of each condition, participants were given a minimum of 60 seconds or as much rest as required.

Mechanical Testing

In addition to human testing, industry standard mechanical tests were completed on the two synthetic turf systems. The first test was the F355 A missile test (2), where the A missile was dropped three times on four locations across each the surface conditions and G_{MAX} and HIC values were measured. The last two drops for each location was recorded and averaged, with the average for each of the four locations then averaged.

The second test was the F355 E missile test, where the missile was initially dropped from a height of 1.3 m. The drop height was then altered until a HIC value of greater than 1000 was recorded. At this drop height, three drops were then performed across four locations.

Data Processing and Analysis

Synchronized marker coordinate and force platform data were imported to Visual3D (C-Motion, Inc.). Marker coordinates and force platform data were filtered using a fourth order zero-lag Butterworth low-pass filter at a cutoff frequency of 12 and 12 Hz respectively for joint kinematics and kinetics calculations. Force platform data filtered at 100 Hz for GRF variables. Three-dimensional lower extremity joint kinematics and kinetics were calculated using Visual3D software suite (C-Motion, Inc.). An X-Y-Z Cardan sequence was used to determine segment reference frames and the right-hand rule was used to determine positive joint rotations and moments. Segment inertial characteristics were determined using data from Dempster (14) and joint moments were reported in the distal reference frame. Ground reaction forces were normalized to body weight (BW). ROM was calculated at contact angle to peak angle. Joint moments were expressed as internal moments and normalized to body mass (Nm/kg). Joint powers were calculated as the product of joint moment and angular velocity. The kinematic, kinetic, and energetic variables of interest were ankle, knee, and hip joint ROMs, peak moments and power in the sagittal and frontal planes, and peak vertical and horizontal GRFs. Variables were extracted and analyzes during the phase of the stance phase, with loading variables occurring during the deceleration phase of the cut and push-off variables occurring during the acceleration phase.

A 2 x 2 (Turf x Approach Velocity) repeated analysis of variance (ANOVA) was used to statistically analyze the effects of the two surface conditions and the two approach velocities on selected biomechanical variables. Post hoc comparisons were

performed with Bonferroni adjustments using a paired-sample t-test when a significant interaction was found with an adjusted *p*-value of 0.0125.

Results

In terms of differences in the surface conditions, the only detected differences were in knee frontal-plane power. During early stance, there was an increase in frontal-plane peak loading eccentric power and during late stance, there was a decrease in frontal-plane peak push-off eccentric power when cutting on the PAD condition compared to cutting on the TURF condition (Table 3).

When comparing approach velocities, there were significant increases in initial peak posterior (p = 0.001) and lateral (p = 0.002) GRFs as the approach velocity increased (Table 1). For peak vertical GRF, there was a significant increase in 1st peak (F = 7.475, p = 0.023) but a significant decrease in 2nd peak (F = 9.294, p = 0.011) with increases in velocity.

Both ankle loading inversion ROM (F = 19.142, p = 0.001) and peak ankle eversion moment (F = 47.198, p < 0.001) increased as approach velocity increased (Table 2). Furthermore, sagittal-plane peak ankle concentric power (F = 6.729, p = 0.025), frontal-plane peak ankle eccentric power (F = 25.621, p < 0.001), and frontal-plane peak ankle concentric power (F = 48.986, p < 0.001) were significantly greater at the faster cutting approach velocity.

At the knee joint, there were significant increases in the knee abduction ROM (F = 10.954, p = 0.008), peak knee extension moment (F = 13.439, p = 0.004), peak knee loading adduction moment (F = 18.806, p = 0.001), and peak push-off knee adduction moment (F = 12.404, p = 0.006) as approach velocity increased (Table 3). A significant

Table 1. Peak GRF (BW): Mean ± STD.

	Slow Cut TURF	Fast Cut TURF	Slow Cut PAD	Fast Cut PAD	Interaction *p* value	Surface *p* value	Approach Velocity *p* value
Peak Loading Posterior GRF	0.69±0.34	0.96±0.44	0.78±0.30	1.05±0.43	0.933	0.239	**0.001**
Peak Push-off Posterior GRF	0.96±0.18	1.13±0.19	0.99±0.21	1.19±0.23	0.576	0.393	**<0.001**
Peak Loading Medial GRF	-0.77±0.40	-1.02±0.38	-0.80±0.56	-1.19±0.38	0.376	0.141	**0.002**
Peak Push-off Medial GRF	-1.00±0.13	-1.18±0.19	-0.82±0.63	-1.21±0.21	0.317	0.489	**0.025**
Peak Loading Vertical GRF	1.98±0.76	2.24±0.63	2.10±0.69	2.37±0.66	0.516	0.178	**0.023**
Peak Push-off Vertical GRF	2.23±0.24	2.10±0.18	2.25±0.30	2.11±0.24	0.927	0.715	**0.011**

Table 2. Ankle ROM (°), peak joint moments (Nm/kg), and peak powers (W/kg): Mean ± STD.

	Slow Cut TURF	Fast Cut TURF	Slow Cut PAD	Fast Cut PAD	Interaction *p* value	Surface *p* value	Approach Velocity *p* value
Dorsi-flexion ROM	33.11±12.32	30.24±11.76	31.36±14.77	31.60±11.76	0.555	0.928	0.666
Inversion Loading ROM	13.99±5.25	19.04±7.63	13.32±3.99	18.24±6.21	0.928	0.535	**0.001**
Peak Plantar-Flexion Moment	-3.09±0.68	-2.97±0.55	-3.03±0.66	-2.88±0.45	0.866	0.403	0.133
Peak Eversion Moment	-0.65±0.31	-0.83±0.34	-0.64±0.21	-0.85±0.22	0.628	0.912	**<0.001**
Sagittal-plane Peak Eccentric Power	-9.04±4.78	-8.89±4.87	-10.13±6.24	-9.94±5.69	0.961	0.266	0.776
Sagittal-plane Peak Concentric Power	12.25±3.57	13.96±3.89	12.31±3.30	13.29±4.05	0.425	0.584	**0.025**
Frontal-plane Peak Eccentric Power	-2.38±2.11	-5.52±4.08	-1.97±1.55	-4.58±2.88	0.531	0.328	**<0.001**
Frontal-plane Peak Concentric Power	1.31±1.37	2.19±1.70	1.30±1.12	2.27±1.15	0.746	0.886	**<0.001**

Table 3. Knee ROM (°), peak joint moments (Nm/kg), and peak powers (W/kg): Mean ± STD.

	Slow Cut TURF	Fast Cut TURF	Slow Cut PAD	Fast Cut PAD	Interaction *p* value	Surface *p* value	Approach Velocity *p* value
Flexion ROM	-38.20±6.31	-37.43±8.93	-37.65±6.65	-35.70±9.47	0.720	0.504	0.486
Abduction ROM	-4.94±2.81	-6.20±3.74	-4.96±4.51	-7.60±3.76	0.446	0.601	**0.008**
Peak Extension Moment	3.45±0.50	3.73±0.60	3.48±0.77	3.71±0.74	0.602	0.973	**0.004**
Peak Loading Adduction Moment	0.27±0.22	0.76±0.49	0.30±0.17	0.81±0.53	0.833	0.460	**0.001**
Peak Push-off Adduction Moment	0.30±0.16	0.58±0.30	0.35±0.20	0.58±0.28	0.177	0.704	**0.006**
Sagittal-plane Peak Eccentric Power	-18.92±9.73[A]	-26.18±12.47[A]	-20.70±8.70	-24.56±12.28	**0.018**	0.951	0.001
Sagittal-plane Peak Concentric Power	9.81±1.31	12.19±2.40	10.19±2.39	11.18±3.13	0.168	0.604	**0.005**
Frontal-plane Peak Loading Eccentric Power	-0.60±0.48	-1.75±1.75	-1.07±1.40	-2.42±2.19	0.417	**0.013**	0.080
Frontal-plane Peak Loading Concentric Power	0.61±0.58	1.94±1.62	1.09±1.05	2.00±1.89	0.515	0.208	**0.028**
Frontal-plane Peak Push-off Eccentric Power	-0.52±0.26	-1.05±0.56	-0.42±0.28	-0.63±0.43	0.089	**0.020**	**0.024**

[A] Significantly different between approach velocity conditions for the TURF condition.

interaction was found for sagittal-plane peak eccentric power (Table 3). Post hoc comparisons revealed that sagittal-plane peak eccentric power increased for the fast velocity compared to the slow condition on the PAD condition (t = 5.993, p > 0.001). In terms of other power variables, the fast velocity also caused significant increases in sagittal-plane peak concentric power (F = 12.322, p = 0.001), frontal-plane peak loading concentric power (F = 6.567, p = 0.028), and frontal-plane peak push-off eccentric power (F = 7.693, p = 0.024).

In terms of hip moments, there were significant increases in peak loading extension moment (F = 10.703, p = 0.007), peak push-off extension moment (F = 7.399, p = 0.020), and peak loading adduction moment (F = 25.034, p = 0.001) as the approach velocity increased (Table 4). There were also increases in sagittal-plane peak eccentric power (F = 9.594, p = 0.013) and sagittal-plane peak concentric power (F = 14.103, p = 0.003).

Synthetic turf mechanical tests of the F355 A missile test showed that there was a decrease in the G_{MAX} value for the PAD condition (Table 5), while the F355 E missile test showed that there was an increase in the critical fall height for the PAD condition (Figure 4).

Discussion

The purpose of this investigation was to investigate the effects of inclusion of a shock pad within a synthetic turf system and changes in approach velocity on lower extremity kinematics, kinetics, and energetics during a 90° cutting movement. The first hypothesis was that peak GRFs and lower extremity moments, powers, and ROM in the sagittal and frontal planes would decrease when completing the cutting task on PAD

Table 4. Hip ROM (°), peak joint moments (Nm/kg), and peak powers (W/kg): Mean ± STD.

	Slow Cut TURF	Fast Cut TURF	Slow Cut PAD	Fast Cut PAD	Interaction *p* value	Surface *p* value	Approach Velocity *p* value
Hip Flexion ROM	1.73±6.95	-0.184±5.51	1.516±7.01	-1.753±6.51	0.678	0.790	0.114
Hip Adduction ROM[A]	1.79±2.16	2.34±2.11	1.85±2.26	5.02±7.03	0.420	0.677	**0.026**
Peak Loading Extension Moment	-1.15±0.27	-1.62±0.42	-1.27±0.30	-1.62±0.78	0.543	0.605	**0.007**
Peak Push-off Extension Moment	-1.18±0.37	-1.55±0.76	-1.28±0.41	-1.70±0.70	0.699	0.313	**0.020**
Peak Loading Adduction Moment	0.26±0.25	0.95±0.40	0.33±0.33	0.77±0.56	0.092	0.477	**0.001**
Sagittal-plane Peak Eccentric Power	-2.86±2.03	-4.55±2.07	-3.37±2.23	-4.11±2.25	0.270	0.837	**0.013**
Sagittal-plane Peak Concentric Power	2.59±1.53	4.15±2.55	2.85±1.48	4.38±3.28	0.806	0.706	**0.003**

[A] Although approach velocity had a significant main effect, Hip Adduction ROM had a sample size of 6

Table 5. F355 A missile G_{MAX} results (g): Mean ± STD

	TURF	PAD
G_{MAX}	153.13±18.03	72.63±5.89

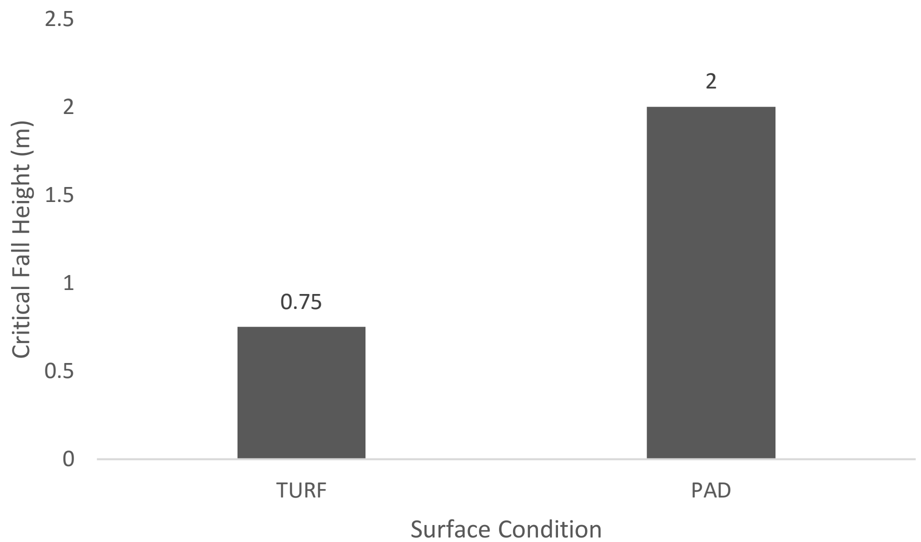

Figure 4: Critical Fall Heights from F355 E missile testing

condition compared to the TURF condition. However, this hypothesis is mostly not supported by our results. Only two variables were found to show significant differences between the surface conditions. A reason for the lack of significant differences may be potentially due to the limitations of inverse dynamics. For the TURF condition with less surface compliance, this could have resulted in increased muscle co-contraction to help stabilize the joint, which cannot be determined by inverse dynamics. Furthermore, the lack of reductions in peak vertical and horizontal GRFs may have been due to subjects adjusting their efforts, resulting in decreased co-contractions for the PAD condition rather than kinematic or kinetic differences. Although there was a lack of general significant differences between the surface conditions, the decreased G_{MAX} values for the PAD condition (Table 5) suggest that synthetic turf systems with a shock pad has the potential to improve field safety without negatively influencing human movement perfromance. These findings could have substantial influence for future synthetic turf installations. The design of the shock pads is primarily to reduce HIC and G_{MAX} values and increase the critical fall height to improve field safety. However, there were concerns that the increased displacement in the surface with a shock pad would increase the risk of lower extremity injury. The findings of this study suggest that for the one shock pad model tested, there is little evidence to suggest that the inclusion of a shock pad would increase the risk of lower extremity injury, while reduced HIC values suggest that the inclusion of a shock pad may improve overall field safety. However, the discrepancy between the mechanical turf testing and human biomechanical testing warrants further investigations into how these two factors interact.

Nevertheless, knee frontal-plane peak loading eccentric power and knee frontal-plane peak push-off eccentric power differed between surface conditions. It was found that knee frontal-plane peak loading eccentric power increased for the PAD compared to the TURF condition, while this peak was also associated with negative power for the both sagittal and transverse planes. This suggests that during the initial loading phase of stance, there were increased eccentric contractions of the muscles crossing the knee joint to stabilize the joint. This stabilization is due to this eccentric power being associated with a knee abduction moment while the knee is adducting, thus indicating that the eccentric muscle contractions are allowing the knee to remain in a more stable position to support the rest of the body. With no significant difference in the peak knee loading abduction moment, it suggests that the cause of this significant difference comes from either small differences in both peak knee loading abduction moment and the associated knee adduction velocity or from a significant change in knee abduction velocity. This could suggest that the inclusion of a shock pad doesn't change the magnitude of loading but the rate at which this loading occurs. Furthermore, with no significant difference in the knee abduction ROM for the loading phase, it can be suggested that this increase in power could be due to the increased compliance of the shock pad, which may be associated with increase knee muscular demands to stabilize the knee. If these increased muscular demands were excessive and repeated, it could result in a muscular injury. The other significant difference detected was a decrease in the frontal-plane peak push-off eccentric power for the PAD condition. At this time point, it was found that the knee had an abduction angular velocity with an associated adduction moment, which was found to be non-significant between the surface

conditions. Again, this suggests the difference in power is either from small, non-significant changes in both the adduction moment and abduction angular velocity or from a large significant difference in the abduction angular velocity. This decrease could reduce the stress of the knee ligaments that are needed to stabilize the knee in the frontal plane. However, before these conclusions can be made, further research is warranted in this area.

The results related to this hypothesis raised a number of questions as to why more differences were not seen between the turf conditions. It was expected that the shock pad would allow for greater deformation during the loading phase of the cutting movement when GRFs were at their highest. This increased deformation would have allowed the shock pad to absorb some of this force, thus resulting in lower GRFs and reducing the force the body had to absorb. However, the lack of significant differences between the turf conditions suggests that this hypothesized mechanism did not occur entirely or did not occur with a large enough effect to cause observable differences at the lower extremity joints. It is also worth considering that the shock pad was designed primarily to reduce the HIC values, where the focus was mainly on vertical loads. As there is a substantial horizontal loading component during the 90° cutting movement, a large portion of the loading would have occurred as a shear force and not a compressive force. Therefore, it was not known how the shock pad would interact with large shear forces as this had previously not been tested. This could warrant further investigation into how the shock pad would behave with a large horizontal force from a mechanical perspective which may provide greater insight into why the shock pad did not influence the anticipated lower extremity loading as predicted.

The second hypothesis stated that as approach velocity increases, there would be increases in GRFs and lower extremity sagittal and frontal plane moments, powers, and ROMs regardless of types of turf systems tested. This hypothesis can be accepted after a number of significant differences were detected between the two approach velocity conditions. In terms of GRFs, there were increases in 1^{st} and 2^{nd} peak GRF in the posterior and medial directions. Posterior GRF peaks were comparable across approach velocities (0.73 BW for 3 m/s, 1.05 BW for 4 m/s) to a previous investigation (37) but medial GRF peaks were greater in the current study. This is most likely due to the increased cutting angle (90° vs. 45°) used in the current study compared to previous study (37) and it has been shown that changes in cutting angle will influence GRF magnitudes (13). These increases in posterior and medial GRFs peaks are most likely due to the increased deceleration requirements with the increased approach velocity.

Vertical GRF peaks exhibited interesting but conflicting patterns as approach velocity increased. Peak loading vertical GRF during the loading phase was significantly greater for the fast condition (Table 1). The values from this study were substantially higher than reported values from a 45° cutting movement using a similar approach velocity (13). The increased loading vertical GRF peak follows previous research which found that vertical GRF increases as running velocity increases (23). Given that the approach velocity increased, this meant that there was an increase in acceleration in the vertical direction of the body when coming into contact with the force platform during the loading phase, causing the increase in vertical GRF. However, the peak push-off vertical GRF was found to be significantly lower for the fast velocity (Table 1). This result was surprising as it was expected that all GRFs would increase. For the faster

approach velocity, the increased demands to decelerate and then change direction potentially resulted in increased trunk lean, which has been suggested to increase medial GRF during a cutting movement (22). This increased demand for medial GRF could be the reason for the decreased peak push-off vertical GRF observed.

As the experimental protocol dictated that performance had to increase in the form of a faster approach velocity, the mechanisms behind the increased performance can be identified. Although the cutting task required a change of direction which would suggest that frontal plane variables would have an increased role compared to linear running, the largest moments produced were still peak plantarflexion moment at the ankle and peak extension moment at the knee joint, suggesting that these two components are the biggest determinants for performance during a 90° cutting movement regardless of approach velocity. The magnitudes of these moments are comparable to previous research using a 90° cutting movement with an approach velocity similar to the fast velocity (4.15 ± 0.32 m/s) (20). It was found that peak ankle plantar-flexion moment did not differ (Table 2) while the peak knee extension moment increases as approach velocity increases (Table 3). This suggests that the knee has a more prominent role in performance than the ankle joint as approach velocity increases during the cutting movement. The role of the hip remains unclear. In contrast to the present study, the peak hip extension moment has been found to be the largest moment (3.11 ± 1.10 Nm/kg vs. 1.618 ± Nm/kg) (20). Nevertheless, it was found in the current study that peak hip extension moment significantly increases as approach velocity increases. These results suggest that the primary mechanisms for the increase in

approach velocity are increased knee and hip extension while there is a reduced role for ankle plantarflexion.

Further significant differences between approach velocities were detected in frontal plane ankle loading. With an increased approach velocity, there was an increase in peak ankle eversion moment (Table 2). This increase in ankle eversion moment coincided with an increase in the ankle inversion ROM (Table 2). It is this increase in ankle inversion ROM which potentially increases the risk of a lateral ankle sprain injury (24, 39). When analyzing a trial which resulted in an ankle sprain, it was found that there was a significant increase in ankle inversion angle coupled with an ankle inversion moment (24), which conflicted the control trials and the findings from this study of the ankle exhibiting an eversion moment during the stance phase of a cutting movement. This suggests that with an increased ankle inversion ROM as approach velocity increases, the risk of suffering a non-contact ankle sprain is increased when completing a 90° cutting movement as there is greater potential for the ankle to undergo excessive inversion which could results in the ankle frontal-plane moment becoming an inversion moment as the ankle ‘rolls’. This notion is further supported by the findings related to ankle joint frontal plane power during the loading phase, where the primary role of the ankle appears to be energy absorption given the large negative power values which significantly increase as approach velocity increases. If the ankle has to absorb too much energy, a larger eccentric contraction will occur and combined with the potentially hazardous kinematic position. This increases the risk of lateral ankle sprain due to the possibility that the eccentric muscle contractions are unable to provide sufficient resistance to excessive ankle inversion.

There were a number of limitations within this study. Firstly, given that the protocol required participants to meet an approach velocity with a range ±10% (e.g. 2.7 – 3.3 m/s for the slow velocity), this gave the opportunity for greater variability in approach velocity and subsequently loading variables compared to previous studies that have used a tighter range of ±5%. In addition, the exit velocity was not controlled meaning that, although participants were instructed to accelerate out of the cutting movement with the velocity they approached it with, there was the potential for this to influence loading variables. In terms of the surface conditions, when installing a synthetic turf system for a full-size pitch, shock pads will be placed onto a lining layer and then turf placed on top with the weight of the turf keeping all the layers connected. The use of tape is different from the actual field installation and therefore, it may have some effects which may or may not have direct impacts on the results of this study. In terms of movement patterns, there were observational differences in the techniques employed by different participants to complete the cutting tasks. In an attempt to minimize these differences and to best ensure that participants completed the cutting tasks with sufficient task-achievement, the selected approach speeds were based on previous work that found the optimal balance between task-achievement and generating sufficient loading at the knee joint (37). Finally, given the testing was conducted in the laboratory, it is possible that the movement patterns displayed by participants were not consistent with those shown during actual game-play.

Conclusion

The findings from this study have shown that the effects of a shock pad on lower extremity loading during a 90° cutting movement are limited. Although there was some

evidence of changes in loading to the knee from significant differences in frontal plane power, in general, there were a lack of differences between the turfs with and without shock pad. In terms of approach velocity, there were many differences detected at the hip, knee, and ankle joint in addition to increases in GRFs. As the protocol dictated a change in performance, the largest changes were seen in peak hip and knee extension moments, suggesting that these were the major factors for determining the performance improvement.

Bibliography

1. Arampatzis A, Bruggemann GP, Metzler V. The effect of speed on leg stiffness and joint kinetics in human running. Journal of biomechanics. 1999;32(12):1349-53.
2. ASTM I. ASTM F355-16e1, Standard Test Method for Impact Attenuation of Playing Surface Systems, Other Protective Sport Systems, and Materials Used for Athletics, Recreation and Play. West Conshohocken, PA; 2016.
3. BBCSport. Uefa approves artificial pitches 2004 [Available from: http://news.bbc.co.uk/sport2/hi/football/europe/4000885.stm.
4. BBCSport. SFA vote backs artificial pitches 2005 [Available from: http://news.bbc.co.uk/sport2/hi/football/scot_div_1/4601513.stm.
5. BBCSport. Artificial pitches set to return to Football League 2014 [Available from: http://www.bbc.com/sport/football/29376418.
6. BBCSport. Artificial pitches: PFA voices injury and integrity concerns 2014 [Available from: http://www.bbc.com/sport/football/29549466.
7. BBCSport. Women's Super League: Casey Stoney does not like plastic pitches 2016 [Available from: http://www.bbc.com/sport/football/36233256.
8. Bennett HJ, Brock E, Brosnan JT, Sorochan JC, Zhang S. Effects of Two Football Stud Types on Knee and Ankle Kinetics of Single-Leg Land-Cut and 180 degrees Cut Movements on Infilled Synthetic Turf. Journal of applied biomechanics. 2015;31(5):309-17.
9. Bjorneboe J, Bahr R, Andersen TE. Risk of injury on third-generation artificial turf in Norwegian professional football. British journal of sports medicine. 2010;44(11):794-8.
10. Brock E, Zhang S, Milner C, Liu X, Brosnan JT, Sorochan JC. Effects of two football stud configurations on biomechanical characteristics of single-leg landing and cutting movements on infilled synthetic turf. Sports biomechanics / International Society of Biomechanics in Sports. 2014;13(4):362-79.
11. Brughelli M, Cronin J, Chaouachi A. Effects of running velocity on running kinetics and kinematics. Journal of strength and conditioning research / National Strength & Conditioning Association. 2011;25(4):933-9.
12. Cavanagh PR, Lafortune MA. Ground reaction forces in distance running. Journal of biomechanics. 1980;13(5):397-406.
13. Cortes N, Onate J, Van Lunen B. Pivot task increases knee frontal plane loading compared with sidestep and drop-jump. Journal of sports sciences. 2011;29(1):83-92.
14. Dempster WT, Gabel WC, Felts WJ. The anthropometry of the manual work space for the seated subject. American journal of physical anthropology. 1959;17:289-317.
15. Derrick TR, Hamill J, Caldwell GE. Energy absorption of impacts during running at various stride lengths. Medicine and science in sports and exercise. 1998;30(1):128-35.
16. Ekstrand J, Timpka T, Hagglund M. Risk of injury in elite football played on artificial turf versus natural grass: a prospective two-cohort study. British journal of sports medicine. 2006;40(12):975-80.
17. Faul F, Erdfelder E, Lang AG, Buchner A. G*Power 3: a flexible statistical power analysis program for the social, behavioral, and biomedical sciences. Behavior research methods. 2007;39(2):175-91.
18. Ford KR, Manson NA, Evans BJ, Myer GD, Gwin RC, Heidt RS, Jr., et al. Comparison of in-shoe foot loading patterns on natural grass and synthetic turf. Journal of science and medicine in sport / Sports Medicine Australia. 2006;9(6):433-40.

19. Fuller CW, Dick RW, Corlette J, Schmalz R. Comparison of the incidence, nature and cause of injuries sustained on grass and new generation artificial turf by male and female football players. Part 1: match injuries. British journal of sports medicine. 2007;41:I20-I6.
20. Havens KL, Sigward SM. Joint and segmental mechanics differ between cutting maneuvers in skilled athletes. Gait & posture. 2015;41(1):33-8.
21. Hershman EB, Anderson R, Bergfeld JA, Bradley JP, Coughlin MJ, Johnson RJ, et al. An analysis of specific lower extremity injury rates on grass and FieldTurf playing surfaces in National Football League Games: 2000-2009 seasons. The American journal of sports medicine. 2012;40(10):2200-5.
22. Hewett TE, Torg JS, Boden BP. Video analysis of trunk and knee motion during non-contact anterior cruciate ligament injury in female athletes: lateral trunk and knee abduction motion are combined components of the injury mechanism. British journal of sports medicine. 2009;43(6):417-22.
23. Keller TS, Weisberger AM, Ray JL, Hasan SS, Shiavi RG, Spengler DM. Relationship between vertical ground reaction force and speed during walking, slow jogging, and running. Clinical biomechanics (Bristol, Avon). 1996;11(5):253-9.
24. Kristianslund E, Bahr R, Krosshaug T. Kinematics and kinetics of an accidental lateral ankle sprain. Journal of biomechanics. 2011;44(14):2576-8.
25. Kyröläinen H, Komi PV, Belli A. Changes in Muscle Activity Patterns and Kinetics With Increasing Running Speed. The Journal of Strength & Conditioning Research. 1999;13(4):400-6.
26. Landry SC, McKean KA, Hubley-Kozey CL, Stanish WD, Deluzio KJ. Neuromuscular and lower limb biomechanical differences exist between male and female elite adolescent soccer players during an unanticipated side-cut maneuver. The American journal of sports medicine. 2007;35(11):1888-900.
27. Lanzetti RM, Ciompi A, Lupariello D, Guzzini M, De Carli A, Ferretti A. Safety of third-generation artificial turf in male elite professional soccer players in Italian major league. Scand J Med Sci Sports. 2016.
28. McGhie D, Ettema G. Biomechanical analysis of surface-athlete impacts on third-generation artificial turf. The American journal of sports medicine. 2013;41(1):177-85.
29. Munro CF, Miller DI, Fuglevand AJ. Ground reaction forces in running: a reexamination. Journal of biomechanics. 1987;20(2):147-55.
30. Nigg BM, Bahlsen HA, Luethi SM, Stokes S. The influence of running velocity and midsole hardness on external impact forces in heel-toe running. Journal of biomechanics. 1987;20(10):951-9.
31. Shorten M, Himmelsbach J. Shock attenuation of sports surfaces. In: Ujihashi S, Haake SJ, editors. The Engineering of Sport. 4. Oxford, UK: Blackwell Science; 2002. p. 152-9.
32. Sigward SM, Powers CM. Loading characteristics of females exhibiting excessive valgus moments during cutting. Clinical biomechanics (Bristol, Avon). 2007;22(7):827-33.
33. Stefanyshyn DJ, Nigg BM. Mechanical energy contribution of the metatarsophalangeal joint to running and sprinting. Journal of biomechanics. 1997;30(11-12):1081-5.
34. Stefanyshyn DJ, Lee J-S, Park S-K. The influence of soccer cleat design on resultant joint moments. Footwear Science. 2010;2(1):13-9.

35. Stiles V, Dixon S. Biomechanical response to systematic changes in impact interface cushioning properties while performing a tennis-specific movement. Journal of sports sciences. 2007;25(11):1229-39.
36. Stiles VH, Dixon SJ. The influence of different playing surfaces on the biomechanics of a tennis running forehand foot plant. Journal of applied biomechanics. 2006;22(1):14-24.
37. Vanrenterghem J, Venables E, Pataky T, Robinson MA. The effect of running speed on knee mechanical loading in females during side cutting. Journal of biomechanics. 2012;45(14):2444-9.
38. Wang X, Fleming PR, Forrester S. Advanced Measurement of Sports Surface System Behaviour under Player Loading. Procedia Engineering. 2014;72:865-70.
39. Zhang S, Wortley M, Chen Q, Freedman J. Efficacy of an ankle brace with a subtalar locking system in inversion control in dynamic movements. The Journal of orthopaedic and sports physical therapy. 2009;39(12):875-83.

References

1. Adkison JW, Requa RK, Garrick JG. Injury rates in high school football. A comparison of synthetic surfaces and grass fields. Clinical orthopaedics and related research. 1974(99):131-6.
2. Agel J, Arendt EA, Bershadsky B. Anterior cruciate ligament injury in national collegiate athletic association basketball and soccer: a 13-year review. The American journal of sports medicine. 2005;33(4):524-30.
3. Arampatzis A, Bruggemann GP, Metzler V. The effect of speed on leg stiffness and joint kinetics in human running. Journal of biomechanics. 1999;32(12):1349-53.
4. Ardern CL, Webster KE, Taylor NF, Feller JA. Return to sport following anterior cruciate ligament reconstruction surgery: a systematic review and meta-analysis of the state of play. British journal of sports medicine. 2011;45(7):596-606.
5. Arendt E, Dick R. Knee injury patterns among men and women in collegiate basketball and soccer. NCAA data and review of literature. The American journal of sports medicine. 1995;23(6):694-701.
6. Arms SW, Pope MH, Johnson RJ, Fischer RA, Arvidsson I, Eriksson E. The biomechanics of anterior cruciate ligament rehabilitation and reconstruction. The American journal of sports medicine. 1984;12(1):8-18.
7. ASTM I. ASTM F355-16e1, Standard Test Method for Impact Attenuation of Playing Surface Systems, Other Protective Sport Systems, and Materials Used for Athletics, Recreation and Play. West Conshohocken, PA; 2016.
8. Bahr R, Krosshaug T. Understanding injury mechanisms: a key component of preventing injuries in sport. British journal of sports medicine. 2005;39(6):324-9.
9. BBCSport. Uefa approves artificial pitches 2004 [Available from: http://news.bbc.co.uk/sport2/hi/football/europe/4000885.stm.
10. BBCSport. SFA vote backs artificial pitches 2005 [Available from: http://news.bbc.co.uk/sport2/hi/football/scot_div_1/4601513.stm.
11. BBCSport. Artificial pitches: 'One goal-kick bounced over the crossbar' 2014 [Available from: http://www.bbc.com/sport/football/29894478.
12. BBCSport. Artificial pitches set to return to Football League 2014 [Available from: http://www.bbc.com/sport/football/29376418.
13. BBCSport. Artificial pitches: PFA voices injury and integrity concerns 2014 [Available from: http://www.bbc.com/sport/football/29549466.
14. BBCSport. Women's football funding increased after targets met 2015 [Available from: http://www.bbc.com/sport/football/33743255.
15. BBCSport. Women's Super League: Casey Stoney does not like plastic pitches 2016 [Available from: http://www.bbc.com/sport/football/36233256.
16. Bendjaballah MZ, Shirazi-Adl A, Zukor DJ. Finite element analysis of human knee joint in varus-valgus. Clinical biomechanics (Bristol, Avon). 1997;12(3):139-48.
17. Bennett HJ, Brock E, Brosnan JT, Sorochan JC, Zhang S. Effects of Two Football Stud Types on Knee and Ankle Kinetics of Single-Leg Land-Cut and 180 degrees Cut Movements on Infilled Synthetic Turf. Journal of applied biomechanics. 2015;31(5):309-17.
18. Bennett HJ, Brock E, Brosnan JT, Sorochan JC, Zhang S. Effects of Two Football Stud Types on Knee and Ankle Kinetics of Single-Leg Land-Cut and 180° Cut Movements on Infilled Synthetic Turf. Journal of applied biomechanics. 2015;31(5).

19. Bentley JA, Ramanathan AK, Arnold GP, Wang W, Abboud RJ. Harmful cleats of football boots: a biomechanical evaluation. Foot and ankle surgery : official journal of the European Society of Foot and Ankle Surgeons. 2011;17(3):140-4.

20. Berns GS, Hull ML, Patterson HA. Strain in the anteromedial bundle of the anterior cruciate ligament under combination loading. Journal of orthopaedic research : official publication of the Orthopaedic Research Society. 1992;10(2):167-76.

21. Besier TF, Lloyd DG, Ackland TR, Cochrane JL. Anticipatory effects on knee joint loading during running and cutting maneuvers. Medicine and science in sports and exercise. 2001;33(7):1176-81.

22. Beynnon BD, Fleming BC, Johnson RJ, Nichols CE, Renstrom PA, Pope MH. Anterior cruciate ligament strain behavior during rehabilitation exercises in vivo. The American journal of sports medicine. 1995;23(1):24-34.

23. Beynnon BD, Vacek PM, Newell MK, Tourville TW, Smith HC, Shultz SJ, et al. The Effects of Level of Competition, Sport, and Sex on the Incidence of First-Time Noncontact Anterior Cruciate Ligament Injury. The American journal of sports medicine. 2014;42(8):1806-12.

24. Bjorneboe J, Bahr R, Andersen TE. Risk of injury on third-generation artificial turf in Norwegian professional football. British journal of sports medicine. 2010;44(11):794-8.

25. Blackburn JT, Padua DA. Sagittal-Plane Trunk Position, Landing Forces, and Quadriceps Electromyographic Activity. Journal of Athletic Training. 2009;44(2):174-9.

26. Bloemers FW, Bakker FC. Acute Ankle Syndesmosis Injury In Athletes. European Journal of Trauma. 2006;32(4):350-6.

27. Boden BP, Dean GS, Feagin JA, Jr., Garrett WE, Jr. Mechanisms of anterior cruciate ligament injury. Orthopedics. 2000;23(6):573-8.

28. Boden BP, Torg JS, Knowles SB, Hewett TE. Video analysis of anterior cruciate ligament injury: abnormalities in hip and ankle kinematics. The American journal of sports medicine. 2009;37(2):252-9.

29. Boden BP, Sheehan FT, Torg JS, Hewett TE. Noncontact anterior cruciate ligament injuries: mechanisms and risk factors. The Journal of the American Academy of Orthopaedic Surgeons. 2010;18(9):520-7.

30. Bradley JP, Klimkiewicz JJ, Rytel MJ, Powell JW. Anterior cruciate ligament injuries in the National Football League: epidemiology and current treatment trends among team physicians. Arthroscopy : the journal of arthroscopic & related surgery : official publication of the Arthroscopy Association of North America and the International Arthroscopy Association. 2002;18(5):502-9.

31. Brock E, Zhang S, Milner C, Liu X, Brosnan JT, Sorochan JC. Effects of two football stud configurations on biomechanical characteristics of single-leg landing and cutting movements on infilled synthetic turf. Sports biomechanics / International Society of Biomechanics in Sports. 2014;13(4):362-79.

32. Brophy R, Silvers HJ, Gonzales T, Mandelbaum BR. Gender influences: the role of leg dominance in ACL injury among soccer players. British journal of sports medicine. 2010;44(10):694-7.

33. Brughelli M, Cronin J, Chaouachi A. Effects of running velocity on running kinetics and kinematics. Journal of strength and conditioning research / National Strength & Conditioning Association. 2011;25(4):933-9.

34. Carey JL, Huffman GR, Parekh SG, Sennett BJ. Outcomes of anterior cruciate ligament injuries to running backs and wide receivers in the National Football League. The American journal of sports medicine. 2006;34(12):1911-7.
35. Cavanagh PR, Lafortune MA. Ground reaction forces in distance running. Journal of biomechanics. 1980;13(5):397-406.
36. Cawley PW, Heidt RS, Jr., Scranton PE, Jr., Losse GM, Howard ME. Physiologic axial load, frictional resistance, and the football shoe-surface interface. Foot & ankle international. 2003;24(7):551-6.
37. Cerulli G, Benoit DL, Lamontagne M, Caraffa A, Liti A. In vivo anterior cruciate ligament strain behaviour during a rapid deceleration movement: case report. Knee surgery, sports traumatology, arthroscopy : official journal of the ESSKA. 2003;11(5):307-11.
38. Chappell JD, Yu B, Kirkendall DT, Garrett WE. A comparison of knee kinetics between male and female recreational athletes in stop-jump tasks. The American journal of sports medicine. 2002;30(2):261-7.
39. Cochrane JL, Lloyd DG, Buttfield A, Seward H, McGivern J. Characteristics of anterior cruciate ligament injuries in Australian football. Journal of science and medicine in sport / Sports Medicine Australia. 2007;10(2):96-104.
40. Cortes N, Onate J, Van Lunen B. Pivot task increases knee frontal plane loading compared with sidestep and drop-jump. Journal of sports sciences. 2011;29(1):83-92.
41. Cortes N, Morrison S, Van Lunen BL, Onate JA. Landing technique affects knee loading and position during athletic tasks. Journal of science and medicine in sport / Sports Medicine Australia. 2012;15(2):175-81.
42. Dayakidis MK, Boudolos K. Ground reaction force data in functional ankle instability during two cutting movements. Clinical biomechanics (Bristol, Avon). 2006;21(4):405-11.
43. DeMorat G, Weinhold P, Blackburn T, Chudik S, Garrett W. Aggressive quadriceps loading can induce noncontact anterior cruciate ligament injury. The American journal of sports medicine. 2004;32(2):477-83.
44. Dempsey AR, Lloyd DG, Elliott BC, Steele JR, Munro BJ, Russo KA. The effect of technique change on knee loads during sidestep cutting. Medicine and science in sports and exercise. 2007;39(10):1765-73.
45. Dempster WT, Gabel WC, Felts WJ. The anthropometry of the manual work space for the seated subject. American journal of physical anthropology. 1959;17:289-317.
46. Derrick TR, Hamill J, Caldwell GE. Energy absorption of impacts during running at various stride lengths. Medicine and science in sports and exercise. 1998;30(1):128-35.
47. Dixon SJ, Stiles VH. Impact absorption of tennis shoe-surface combinations. Sports Engineering. 2003;6(1):1-9.
48. Dodson CC, Secrist ES, Bhat SB, Woods DP, Deluca PF. Anterior Cruciate Ligament Injuries in National Football League Athletes From 2010 to 2013: A Descriptive Epidemiology Study. Orthopaedic journal of sports medicine. 2016;4(3):2325967116631949.
49. Draganich LF, Vahey JW. An in vitro study of anterior cruciate ligament strain induced by quadriceps and hamstrings forces. Journal of orthopaedic research : official publication of the Orthopaedic Research Society. 1990;8(1):57-63.

50. Durselen L, Claes L, Kiefer H. The influence of muscle forces and external loads on cruciate ligament strain. The American journal of sports medicine. 1995;23(1):129-36.
51. Ekstrand J, Timpka T, Hagglund M. Risk of injury in elite football played on artificial turf versus natural grass: a prospective two-cohort study. British journal of sports medicine. 2006;40(12):975-80.
52. Ekstrand J, Hagglund M, Fuller CW. Comparison of injuries sustained on artificial turf and grass by male and female elite football players. Scand J Med Sci Sports. 2011;21(6):824-32.
53. Engstrom B, Johansson C, Tornkvist H. Soccer injuries among elite female players. The American journal of sports medicine. 1991;19(4):372-5.
54. Faude O, Junge A, Kindermann W, Dvorak J. Risk factors for injuries in elite female soccer players. British journal of sports medicine. 2006;40(9):785-90.
55. Faul F, Erdfelder E, Lang AG, Buchner A. G*Power 3: a flexible statistical power analysis program for the social, behavioral, and biomedical sciences. Behavior research methods. 2007;39(2):175-91.
56. Fauno P, Wulff Jakobsen B. Mechanism of anterior cruciate ligament injuries in soccer. International journal of sports medicine. 2006;27(1):75-9.
57. Finsterbush A, Frankl U, Matan Y, Mann G. Secondary damage to the knee after isolated injury of the anterior cruciate ligament. The American journal of sports medicine. 1990;18(5):475-9.
58. Ford KR, Myer GD, Hewett TE. Valgus knee motion during landing in high school female and male basketball players. Medicine and science in sports and exercise. 2003;35(10):1745-50.
59. Ford KR, Myer GD, Toms HE, Hewett TE. Gender differences in the kinematics of unanticipated cutting in young athletes. Medicine and science in sports and exercise. 2005;37(1):124-9.
60. Ford KR, Manson NA, Evans BJ, Myer GD, Gwin RC, Heidt RS, Jr., et al. Comparison of in-shoe foot loading patterns on natural grass and synthetic turf. Journal of science and medicine in sport / Sports Medicine Australia. 2006;9(6):433-40.
61. Frederick EC, Hagy JL, Mann RA. The prediction of vertical impact force during running. Journal of biomechanics. 1981;14(7):498.
62. Fuller CW, Dick RW, Corlette J, Schmalz R. Comparison of the incidence, nature and cause of injuries sustained on grass and new generation artificial turf by male and female football players. Part 1: match injuries. British journal of sports medicine. 2007;41:I20-I6.
63. Gil S, Ruiz F, Irazusta A, Gil J, Irazusta J. Selection of young soccer players in terms of anthropometric and physiological factors. The Journal of sports medicine and physical fitness. 2007;47(1):25-32.
64. Gottlob CA, Baker CL, Jr., Pellissier JM, Colvin L. Cost effectiveness of anterior cruciate ligament reconstruction in young adults. Clinical orthopaedics and related research. 1999(367):272-82.
65. Greig M. The influence of soccer-specific activity on the kinematics of an agility sprint. Eur J Sport Sci. 2009;9(1):23-33.
66. Griffin LY, Albohm MJ, Arendt EA, Bahr R, Beynnon BD, Demaio M, et al. Understanding and preventing noncontact anterior cruciate ligament injuries: a review of

the Hunt Valley II meeting, January 2005. The American journal of sports medicine. 2006;34(9):1512-32.
67. Hase K, Stein RB. Turning strategies during human walking. Journal of neurophysiology. 1999;81(6):2914-22.
68. Havens KL, Sigward SM. Joint and segmental mechanics differ between cutting maneuvers in skilled athletes. Gait & posture. 2015;41(1):33-8.
69. Havens KL, Sigward SM. Whole body mechanics differ among running and cutting maneuvers in skilled athletes. Gait & posture. 2015;42(3):240-5.
70. Havens KL, Sigward SM. Cutting mechanics: relation to performance and anterior cruciate ligament injury risk. Medicine and science in sports and exercise. 2015;47(4):818-24.
71. Heidt RS, Jr., Dormer SG, Cawley PW, Scranton PE, Jr., Losse G, Howard M. Differences in friction and torsional resistance in athletic shoe-turf surface interfaces. The American journal of sports medicine. 1996;24(6):834-42.
72. Hershman EB, Anderson R, Bergfeld JA, Bradley JP, Coughlin MJ, Johnson RJ, et al. An analysis of specific lower extremity injury rates on grass and FieldTurf playing surfaces in National Football League Games: 2000-2009 seasons. The American journal of sports medicine. 2012;40(10):2200-5.
73. Hewett TE, Myer GD, Ford KR, Heidt RS, Jr., Colosimo AJ, McLean SG, et al. Biomechanical measures of neuromuscular control and valgus loading of the knee predict anterior cruciate ligament injury risk in female athletes: a prospective study. The American journal of sports medicine. 2005;33(4):492-501.
74. Hewett TE, Torg JS, Boden BP. Video analysis of trunk and knee motion during non-contact anterior cruciate ligament injury in female athletes: lateral trunk and knee abduction motion are combined components of the injury mechanism. British journal of sports medicine. 2009;43(6):417-22.
75. Irvine GB, Glasgow MM. The natural history of the meniscus in anterior cruciate insufficiency. Arthroscopic analysis. The Journal of bone and joint surgery British volume. 1992;74(3):403-5.
76. Jamison S, Lee C. The incidence of female hockey injuries on grass and synthetic playing surfaces. Australian Journal of Science and Medicine in Sport. 1989;21(2):15-7.
77. Kamath GV, Murphy T, Creighton RA, Viradia N, Taft TN, Spang JT. Anterior Cruciate Ligament Injury, Return to Play, and Reinjury in the Elite Collegiate Athlete: Analysis of an NCAA Division I Cohort. The American journal of sports medicine. 2014;42(7):1638-43.
78. Keller TS, Weisberger AM, Ray JL, Hasan SS, Shiavi RG, Spengler DM. Relationship between vertical ground reaction force and speed during walking, slow jogging, and running. Clinical biomechanics (Bristol, Avon). 1996;11(5):253-9.
79. Kent R, Forman JL, Lessley D, Crandall J. The mechanics of American football cleats on natural grass and infill-type artificial playing surfaces with loads relevant to elite athletes. Sports biomechanics / International Society of Biomechanics in Sports. 2015;14(2):246-57.
80. Kim JH, Lee KK, Ahn KO, Kong SJ, Park SC, Lee YS. Evaluation of the interaction between contact force and decision making on lower extremity biomechanics during a side-cutting maneuver. Archives of orthopaedic and trauma surgery. 2016;136(6):821-8.

81. Koga H, Nakamae A, Shima Y, Iwasa J, Myklebust G, Engebretsen L, et al. Mechanisms for noncontact anterior cruciate ligament injuries: knee joint kinematics in 10 injury situations from female team handball and basketball. The American journal of sports medicine. 2010;38(11):2218-25.
82. Kristianslund E, Bahr R, Krosshaug T. Kinematics and kinetics of an accidental lateral ankle sprain. Journal of biomechanics. 2011;44(14):2576-8.
83. Krosshaug T, Nakamae A, Boden BP, Engebretsen L, Smith G, Slauterbeck JR, et al. Mechanisms of anterior cruciate ligament injury in basketball: video analysis of 39 cases. The American journal of sports medicine. 2007;35(3):359-67.
84. Kyröläinen H, Komi PV, Belli A. Changes in Muscle Activity Patterns and Kinetics With Increasing Running Speed. The Journal of Strength & Conditioning Research. 1999;13(4):400-6.
85. Lambson RB, Barnhill BS, Higgins RW. Football cleat design and its effect on anterior cruciate ligament injuries. A three-year prospective study. The American journal of sports medicine. 1996;24(2):155-9.
86. Landry SC, McKean KA, Hubley-Kozey CL, Stanish WD, Deluzio KJ. Neuromuscular and lower limb biomechanical differences exist between male and female elite adolescent soccer players during an unanticipated side-cut maneuver. The American journal of sports medicine. 2007;35(11):1888-900.
87. Lanzetti RM, Ciompi A, Lupariello D, Guzzini M, De Carli A, Ferretti A. Safety of third-generation artificial turf in male elite professional soccer players in Italian major league. Scand J Med Sci Sports. 2016.
88. Levy IM, Skovron ML, Agel J. Living with artificial grass: a knowledge update. Part 1: Basic science. The American journal of sports medicine. 1990;18(4):406-12.
89. Li G, Rudy TW, Sakane M, Kanamori A, Ma CB, Woo SL. The importance of quadriceps and hamstring muscle loading on knee kinematics and in-situ forces in the ACL. Journal of biomechanics. 1999;32(4):395-400.
90. Li G, Zayontz S, Most E, DeFrate LE, Suggs JF, Rubash HE. In situ forces of the anterior and posterior cruciate ligaments in high knee flexion: an in vitro investigation. Journal of orthopaedic research : official publication of the Orthopaedic Research Society. 2004;22(2):293-7.
91. Li G, Defrate LE, Rubash HE, Gill TJ. In vivo kinematics of the ACL during weight-bearing knee flexion. Journal of orthopaedic research : official publication of the Orthopaedic Research Society. 2005;23(2):340-4.
92. Lohmander LS, Östenberg A, Englund M, Roos H. High prevalence of knee osteoarthritis, pain, and functional limitations in female soccer players twelve years after anterior cruciate ligament injury. Arthritis & Rheumatism. 2004;50(10):3145-52.
93. Luhtanen P, Komi PV. Force-, power-, and elasticity-velocity relationships in walking, running, and jumping. European journal of applied physiology and occupational physiology. 1980;44(3):279-89.
94. Maletis GB, Inacio MC, Funahashi TT. Risk factors associated with revision and contralateral anterior cruciate ligament reconstructions in the Kaiser Permanente ACLR registry. The American journal of sports medicine. 2015;43(3):641-7.
95. Malinzak RA, Colby SM, Kirkendall DT, Yu B, Garrett WE. A comparison of knee joint motion patterns between men and women in selected athletic tasks. Clinical biomechanics (Bristol, Avon). 2001;16(5):438-45.

96. Mandelbaum BR, Silvers HJ, Watanabe DS, Knarr JF, Thomas SD, Griffin LY, et al. Effectiveness of a neuromuscular and proprioceptive training program in preventing anterior cruciate ligament injuries in female athletes: 2-year follow-up. The American journal of sports medicine. 2005;33(7):1003-10.
97. Markolf KL, Burchfield DM, Shapiro MM, Shepard MF, Finerman GA, Slauterbeck JL. Combined knee loading states that generate high anterior cruciate ligament forces. Journal of orthopaedic research : official publication of the Orthopaedic Research Society. 1995;13(6):930-5.
98. Matava MJ, Freehill AK, Grutzner S, Shannon W. Limb dominance as a potential etiologic factor in noncontact anterior cruciate ligament tears. The journal of knee surgery. 2002;15(1):11-6.
99. McCullagh PJ, Graham ID. A preliminary investigation into the nature of shock absorbency in synthetic sports materials. Journal of sports sciences. 1985;3(2):103-14.
100. McGee KJ, Burkett LN. The National Football League combine: a reliable predictor of draft status? Journal of strength and conditioning research / National Strength & Conditioning Association. 2003;17(1):6-11.
101. McGhie D, Ettema G. Biomechanical analysis of surface-athlete impacts on third-generation artificial turf. The American journal of sports medicine. 2013;41(1):177-85.
102. McHugh MP, Tyler TF, Mirabella MR, Mullaney MJ, Nicholas SJ. The effectiveness of a balance training intervention in reducing the incidence of noncontact ankle sprains in high school football players. The American journal of sports medicine. 2007;35(8):1289-94.
103. McLean SG, Huang X, Su A, Van Den Bogert AJ. Sagittal plane biomechanics cannot injure the ACL during sidestep cutting. Clinical biomechanics (Bristol, Avon). 2004;19(8):828-38.
104. McLean SG, Huang X, van den Bogert AJ. Association between lower extremity posture at contact and peak knee valgus moment during sidestepping: implications for ACL injury. Clinical biomechanics (Bristol, Avon). 2005;20(8):863-70.
105. McLeod A. An assessment of the benefits of artificial turf over natural turf playing surfaces within the independent school environment. Cranfield University, Cranfield. 2003.
106. Mero A, Komi PV. Force-, EMG-, and elasticity-velocity relationships at submaximal, maximal and supramaximal running speeds in sprinters. European journal of applied physiology and occupational physiology. 1986;55(5):553-61.
107. Meyers MC, Barnhill BS. Incidence, causes, and severity of high school football injuries on FieldTurf versus natural grass: a 5-year prospective study. The American journal of sports medicine. 2004;32(7):1626-38.
108. Milburn PD, Barry EB. Shoe-surface interaction and the reduction of injury in rugby union. Sports medicine (Auckland, NZ). 1998;25(5):319-27.
109. Muller C, Sterzing T, Lange J, Milani TL. Comprehensive evaluation of player-surface interaction on artificial soccer turf. Sports biomechanics / International Society of Biomechanics in Sports. 2010;9(3):193-205.
110. Müller C, Sterzing T, Lake M, Milani TL. Different stud configurations cause movement adaptations during a soccer turning movement. Footwear Science. 2010;2(1):21-8.

111. Munro CF, Miller DI, Fuglevand AJ. Ground reaction forces in running: a reexamination. Journal of biomechanics. 1987;20(2):147-55.
112. Myer GD, Ford KR, Palumbo JP, Hewett TE. Neuromuscular training improves performance and lower-extremity biomechanics in female athletes. Journal of strength and conditioning research / National Strength & Conditioning Association. 2005;19(1):51-60.
113. Myer GD, Ford KR, McLean SG, Hewett TE. The effects of plyometric versus dynamic stabilization and balance training on lower extremity biomechanics. The American journal of sports medicine. 2006;34(3):445-55.
114. Naunheim R, Parrott H, Standeven J. A comparison of artificial turf. The Journal of trauma. 2004;57(6):1311-4.
115. Negrete RJ, Schick EA, Cooper JP. Lower-limb dominance as a possible etiologic factor in noncontact anterior cruciate ligament tears. Journal of strength and conditioning research / National Strength & Conditioning Association. 2007;21(1):270-3.
116. Nigg BM, Bahlsen HA, Luethi SM, Stokes S. The influence of running velocity and midsole hardness on external impact forces in heel-toe running. Journal of biomechanics. 1987;20(10):951-9.
117. Nigg BM, Segesser B. The influence of playing surfaces on the load on the locomotor system and on football and tennis injuries. Sports medicine (Auckland, NZ). 1988;5(6):375-85.
118. Nigg BM. Impact forces in running. Current Opinion in Orthopaedics. 1997;8(6):43-7.
119. Nunley RM, Wright D, Renner JB, Yu B, Garrett WE. Gender Comparison of Patellar Tendon Tibial Shaft Angle with Weight Bearing. Research in Sports Medicine. 2003;11(3):173-85.
120. Olsen OE, Myklebust G, Engebretsen L, Holme I, Bahr R. Relationship between floor type and risk of ACL injury in team handball. Scand J Med Sci Sports. 2003;13(5):299-304.
121. Orchard J, Seward H, McGivern J, Hood S. Intrinsic and extrinsic risk factors for anterior cruciate ligament injury in Australian footballers. The American journal of sports medicine. 2001;29(2):196-200.
122. Orchard JW, Chivers I, Aldous D, Bennell K, Seward H. Rye grass is associated with fewer non-contact anterior cruciate ligament injuries than bermuda grass. British journal of sports medicine. 2005;39(10):704-9.
123. Quatman CE, Hewett TE. The anterior cruciate ligament injury controversy: is "valgus collapse" a sex-specific mechanism? British journal of sports medicine. 2009;43(5):328-35.
124. Ramirez M, Schaffer KB, Shen H, Kashani S, Kraus JF. Injuries to high school football athletes in California. The American journal of sports medicine. 2006;34(7):1147-58.
125. Reilly T, Williams AM, Nevill A, Franks A. A multidisciplinary approach to talent identification in soccer. Journal of sports sciences. 2000;18(9):695-702.
126. Rochcongar P, Laboute E, Jan J, Carling C. Ruptures of the anterior cruciate ligament in soccer. International journal of sports medicine. 2009;30(5):372-8.

127. Rodeo SA, O'Brien S, Warren RF, Barnes R, Wickiewicz TL, Dillingham MF. Turf-toe: an analysis of metatarsophalangeal joint sprains in professional football players. The American journal of sports medicine. 1990;18(3):280-5.
128. Seering WP, Piziali RL, Nagel DA, Schurman DJ. The function of the primary ligaments of the knee in varus-valgus and axial rotation. Journal of biomechanics. 1980;13(9):785-94.
129. Sell TC, Ferris CM, Abt JP, Tsai YS, Myers JB, Fu FH, et al. Predictors of proximal tibia anterior shear force during a vertical stop-jump. Journal of orthopaedic research : official publication of the Orthopaedic Research Society. 2007;25(12):1589-97.
130. Shorten M, Himmelsbach J. Shock attenuation of sports surfaces. In: Ujihashi S, Haake SJ, editors. The Engineering of Sport. 4. Oxford, UK: Blackwell Science; 2002. p. 152-9.
131. Sigward SM, Powers CM. The influence of gender on knee kinematics, kinetics and muscle activation patterns during side-step cutting. Clinical biomechanics (Bristol, Avon). 2006;21(1):41-8.
132. Sigward SM, Powers CM. Loading characteristics of females exhibiting excessive valgus moments during cutting. Clinical biomechanics (Bristol, Avon). 2007;22(7):827-33.
133. Skovron ML, Levy IM, Agel J. Living with artificial grass: a knowledge update. Part 2: Epidemiology. The American journal of sports medicine. 1990;18(5):510-3.
134. Smith N, Dyson R, Janaway L. Ground reaction force measures when running in soccer boots and soccer training shoes on a natural turf surface. Sports Engineering. 2004;7(3):159-67.
135. Söderman K, Pietilä T, Alfredson H, Werner S. Anterior cruciate ligament injuries in young females playing soccer at senior levels. Scandinavian Journal of Medicine & Science in Sports. 2002;12(2):65-8.
136. Stanley LE, Kerr ZY, Dompier TP, Padua DA. Sex Differences in the Incidence of Anterior Cruciate Ligament, Medial Collateral Ligament, and Meniscal Injuries in Collegiate and High School Sports: 2009-2010 Through 2013-2014. The American journal of sports medicine. 2016;44(6):1565-72.
137. Stefanyshyn DJ, Nigg BM. Mechanical energy contribution of the metatarsophalangeal joint to running and sprinting. Journal of biomechanics. 1997;30(11-12):1081-5.
138. Stefanyshyn DJ, Lee J-S, Park S-K. The influence of soccer cleat design on resultant joint moments. Footwear Science. 2010;2(1):13-9.
139. Stiles V, Dixon S. Biomechanical response to systematic changes in impact interface cushioning properties while performing a tennis-specific movement. Journal of sports sciences. 2007;25(11):1229-39.
140. Stiles VH, Dixon SJ. The influence of different playing surfaces on the biomechanics of a tennis running forehand foot plant. Journal of applied biomechanics. 2006;22(1):14-24.
141. Sutton KM, Bullock JM. Anterior cruciate ligament rupture: differences between males and females. The Journal of the American Academy of Orthopaedic Surgeons. 2013;21(1):41-50.
142. Swenson DM, Collins CL, Best TM, Flanigan DC, Fields SK, Comstock RD. Epidemiology of knee injuries among U.S. high school athletes, 2005/2006-2010/2011. Medicine and science in sports and exercise. 2013;45(3):462-9.

143. Tranaeus U, Gotesson E, Werner S. Injury Profile in Swedish Elite Floorball: A Prospective Cohort Study of 12 Teams. Sports health. 2016.
144. Uhorchak JM, Scoville CR, Williams GN, Arciero RA, St Pierre P, Taylor DC. Risk factors associated with noncontact injury of the anterior cruciate ligament: a prospective four-year evaluation of 859 West Point cadets. The American journal of sports medicine. 2003;31(6):831-42.
145. van Eijden TM, de Boer W, Weijs WA. The orientation of the distal part of the quadriceps femoris muscle as a function of the knee flexion-extension angle. Journal of biomechanics. 1985;18(10):803-9.
146. Vanrenterghem J, Venables E, Pataky T, Robinson MA. The effect of running speed on knee mechanical loading in females during side cutting. Journal of biomechanics. 2012;45(14):2444-9.
147. Vescovi JD, VanHeest JL. Effects of an anterior cruciate ligament injury prevention program on performance in adolescent female soccer players. Scand J Med Sci Sports. 2010;20(3):394-402.
148. von Porat A, Roos EM, Roos H. High prevalence of osteoarthritis 14 years after an anterior cruciate ligament tear in male soccer players: a study of radiographic and patient relevant outcomes. Annals of the rheumatic diseases. 2004;63(3):269-73.
149. Walden M, Hagglund M, Ekstrand J. High risk of new knee injury in elite footballers with previous anterior cruciate ligament injury. British journal of sports medicine. 2006;40(2):158-62; discussion -62.
150. Walden M, Hagglund M, Magnusson H, Ekstrand J. Anterior cruciate ligament injury in elite football: a prospective three-cohort study. Knee surgery, sports traumatology, arthroscopy : official journal of the ESSKA. 2011;19(1):11-9.
151. Wang X, Fleming PR, Forrester S. Advanced Measurement of Sports Surface System Behaviour under Player Loading. Procedia Engineering. 2014;72:865-70.
152. Wild CY, Steele JR, Munro BJ. Why do girls sustain more anterior cruciate ligament injuries than boys?: a review of the changes in estrogen and musculoskeletal structure and function during puberty. Sports medicine (Auckland, NZ). 2012;42(9):733-49.
153. Williams S, Trewartha G, Kemp SP, Michell R, Stokes KA. The influence of an artificial playing surface on injury risk and perceptions of muscle soreness in elite Rugby Union. Scand J Med Sci Sports. 2016;26(1):101-8.
154. Withrow TJ, Huston LJ, Wojtys EM, Ashton-Miller JA. The effect of an impulsive knee valgus moment on in vitro relative ACL strain during a simulated jump landing. Clinical biomechanics (Bristol, Avon). 2006;21(9):977-83.
155. Wojtys EM, Ashton-Miller JA, Huston LJ. A gender-related difference in the contribution of the knee musculature to sagittal-plane shear stiffness in subjects with similar knee laxity. The Journal of bone and joint surgery American volume. 2002;84-a(1):10-6.
156. Woods C, Hawkins R, Hulse M, Hodson A. The Football Association Medical Research Programme: an audit of injuries in professional football: an analysis of ankle sprains. British journal of sports medicine. 2003;37(3):233-8.
157. Xu D, Carlton LG, Rosengren KS. Anticipatory postural adjustments for altering direction during walking. Journal of motor behavior. 2004;36(3):316-26.
158. Yu B, Herman D, Preston J, Lu W, Kirkendall DT, Garrett WE. Immediate effects of a knee brace with a constraint to knee extension on knee kinematics and ground

reaction forces in a stop-jump task. The American journal of sports medicine. 2004;32(5):1136-43.
159. Yu B, McClure SB, Onate JA, Guskiewicz KM, Kirkendall DT, Garrett WE. Age and gender effects on lower extremity kinematics of youth soccer players in a stop-jump task. The American journal of sports medicine. 2005;33(9):1356-64.
160. Yu B, Lin CF, Garrett WE. Lower extremity biomechanics during the landing of a stop-jump task. Clinical biomechanics (Bristol, Avon). 2006;21(3):297-305.
161. Yu B, Garrett WE. Mechanisms of non-contact ACL injuries. British journal of sports medicine. 2007;41 Suppl 1:i47-51.
162. Zhang S, Wortley M, Chen Q, Freedman J. Efficacy of an ankle brace with a subtalar locking system in inversion control in dynamic movements. The Journal of orthopaedic and sports physical therapy. 2009;39(12):875-83.

Appendices

Appendix A

INFORMED CONSENT FORM

Effects of shock padding on lower extremity biomechanics during 90° cutting movement, drop landing, and drop jump on synthetic turf

Principal Investigators: Songning Zhang, PhD and Thomas Elvidge, BSc
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Introduction

You are invited to participate in this research study because you are either an active football or soccer male recreational player between 18 and 30 years old. This research investigates the effects of an underlying shock pad beneath synthetic turf during dynamic movements typically associated with these sports. Specifically, cutting movements at 2 speeds and jumping and landing movements from 2 - 3 different heights. Please ask the study staff to explain any words or information that you do not clearly understand. Before agreeing to participate in this study, it is important that you read and understand the following explanation of the procedures, risks, and benefits.

Testing Protocol

If you agree to participate, you will attend two data collection sessions at the Biomechanics/Sports Medicine Lab on the UT campus. You will need to complete the demographic questionnaire and Physical Activity Readiness Questionnaire (PAR-Q) during the first session, which will be used for this study. Each data collection session will take approximately 2 – 2.5 hours. You will need to wear tight-fitting clothing appropriate for exercise which includes spandex shorts and t-shirt. If you do not have spandex type of clothing, spandex shorts will be provided.

We will measure your weight and height. We will also place reflective markers on your feet, ankles, legs, knees, thighs, pelvis and trunk. This will allow motion cameras to capture your body movements when performing the exercises. The motion cameras will not record images of you but simply track the motion of the markers placed upon your body. If you have any questions, interests, or concerns about any equipment to be used in this test, please feel free to ask the investigator or other research personnel.

During each data collection session, you will complete the same series of movements on the same synthetic turf surface with either a shock pad installed or no shock pad installed underneath. For each session, the following movements will be completed:

- 90° cutting movement with the following approach speeds:
  - 2.5 $m \cdot s^{-1}$
  - 3.5 $m \cdot s^{-1}$
- Drop landing tasks from the following heights:
  - 20 cm
  - 40 cm
  - 60 cm

- Drop jump tasks from the following heights:
  - 20 cm
  - 40 cm

For the cutting trials, you will be given practice trials to give you feedback on your approach speed and your starting position to make sure that your foot will landing within the data collection capture space. This will also help allow you to run as naturally as possible during the movements. Between trials and conditions, you will be given enough as much time as needed to rest and recover.

For the drop landing task, trials need to be completed with a normal landing style. We will determine your maximum knee flexion during several practice trials in each of the trial landing height and the mean of the maximum flexion angle and its range (± 9 degrees) will be used to monitor your knee flexion angle during the drop landing trials. If you are not within the knee flexion range, you will be asked to repeat the trial. For the drop jump task, you will be asked to perform the vertical jump as quickly as possible after the initial landing from the drop height. Again, you will have opportunity to practice trials to become familiar with the testing procedures as staff explain the movement to you and you will be given enough time between trials to rest. You can end any exercise early and do not have to complete the study visit.

Potential Risks

Risks associated with this study are minimal. There are minimum risks of a knee sprain during the cutting movements and an ankle sprain during the drop landing tasks but it is no greater than the risk you would experience when playing your sports. In order to prevent potential muscle strains and ligament sprains, you will be asked to perform a standardized warm-up and stretching all major muscle groups prior to the practice trials and actual testing. The turf surface is infilled with the sand and rubber particles evenly to prevent any possibility of injury due to unevenness. You are asked to practice the movements before the testing and take breaks as needed. In the unlikely event you are injured during the study, we will provide standard first aid. However, the University of Tennessee does not automatically provide reimbursement for medical care or other compensation and you will be responsible for any medical expenses. If you are injured, please notify Thomas Elvidge or Dr. Songning Zhang (974-2091).

Every research study involves some risk to your confidentiality. It is possible that other people could find out you were in the study or see your study information. But we will do our best to keep your information confidential to minimize this risk and keep all of your data on password-protected computers.

Benefits of Participation

Potential benefits for you is that you may learn your cutting movement techniques, landing control techniques, and experience performing these movements on the two different surfaces. If you wish, you can receive an individual report of your cutting, landing, and drop jump biomechanics to share with your athletic trainer and/or coaches in the case it might be helpful to your sport performance and injury prevention. Results from this study may help the understanding and improvement of synthetic turf and shock pad combinations which help force absorption during human cutting and landing performance.

Confidentiality

All information you provide will be kept confidential. Your research data and records will be stored securely and will be made available only to researchers who work on this study. The motion cameras will not record images of you. Your name will not be in any research data. Instead, a code number will replace your name on your data. Your name will not appear with the study results that will be presented at conferences and published in journals. Your data will be stored using password protected hard drives. Your data may be used for future research purposes after the completion of this study. If you decide to withdraw from the study, data collected up to that point may be used for research purposes, unless you request that it be destroyed.

Compensation

If you participate in both data collection sessions, you will receive a $70 gift card for your time and participation. If you withdraw from the study or do not complete the second data collection session, you will not be eligible to receive a gift card.

Contact Information

If you have any questions about the study at any time or if you experience any problems as a result of participating in this study you can contact Tom Elvidge or Dr. Songning Zhang at 1914 Andy Holt Ave. 136 HPER Bldg., the University of Tennessee and/or (865) 974-2091. Questions about your rights as a participant can be addressed to Compliance Officer in the Office of Research at the University of Tennessee at (865) 974-7697.

Voluntary Participation and Withdrawal

Your participation is entirely voluntary and your refusal to participate will involve no penalty to yourself. You may withdraw from the study at any time without penalty. In terms of benefits, a biomechanical report can be given to you if you do not complete both study sessions but the gift card will only be given following the conclusion of your second session. Your participation in this study may be stopped by if you fail to follow the study procedures or if the principal investigator believes it is in your best interest to stop participation.

Consent Statement

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

Subject's Name: ____________________
Subject's Signature: _________________________ Date: _________
Investigator's Signature: _____________________________ Date: __________

Appendix B

Subject Information Questionnaire

Subject # _____ Date:_______________

Age: ______ Height: ________ (inch) ________ (m)

Weight: ________(lb) _________ (kg)

Leg Dominance: R / L

- Select the major sport you played (check all applied):

American Football: _______ Soccer: _______ Other: _______

- If Other, please tell us the sports in the space below:

____

- Number of years played at the following levels:

Pre-High School: ______ High school: ______ College: ______
Professional: ______

- Current sport activity frequency (check only one choice below):

1 time/week ______2 times/week ______ 3 or more times /week ______

- The average duration of each time you play your sport(s) in a typical week (Check only one choice below):

30 min or less ___ 60 min or less ___ 90 min or less ___ 120 min or less ___ more than 120 min ___

- Previous major Injuries:

____
____
____

Appendix C

Subject Demographics

Table 6: Subject Demographics

	Mean	Standard Deviation
Age	21.92	2.68
Height (m)	1.85	0.06
Weight (kg)	78.46	9.40
Right Foot Dominant	11	-
Left Foot Dominant	1	-

Appendix D

Individual Subject Results

Table 7: Loading Peak GRF X (BW)

	TURF		PAD	
Subject	SLOW	FAST	SLOW	FAST
1	0.638±0.124	1.042±0.173	0.844±0.232	1.727±0.196
2	0.509±0.107	0.626±0.055	0.449±0.091	0.571±0.111
3	0.625±0.094	0.637±0.148	0.665±0.117	0.576±0.040
4	0.352±0.027	0.552±0.137	0.764±0.115	1.244±0.223
5	0.334±0.035	0.350±0.049	0.339±0.043	0.478±0.074
6	0.744±0.072	1.209±0.130	0.702±0.161	1.056±0.066
7	1.134±0.070	1.488±0.115	0.876±0.054	1.508±0.117
8	0.613±0.075	1.064±0.169	0.667±0.078	1.015±0.090
9	0.305±0.075	0.497±0.066	0.615±0.142	0.665±0.105
10	1.382±0.234	1.779±0.099	1.358±0.161	1.686±0.092
11	1.041±0.093	1.225±0.120	0.815±0.153	0.929±0.066
12	0.577±0.050	1.100±0.119	1.311±0.188	1.180±0.152
Mean	0.688±0.337	0.964±0.435	0.784±0.301	1.053±0.434

Table 8: Push-off Peak GRF X (BW)

	TURF		PAD	
Subject	SLOW	FAST	SLOW	FAST
1	1.066±0.099	1.487±0.221	0.968±0.105	1.264±0.074
2	0.843±0.085	0.910±0.125	0.791±0.037	1.015±0.081
3	0.949±0.045	1.095±0.105	1.102±0.100	1.116±0.029
4	0.973±0.047	1.248±0.109	1.347±0.180	1.415±0.122
5	0.780±0.080	0.924±0.033	0.751±0.068	0.937±0.071
6	0.837±0.035	0.999±0.101	0.782±0.089	0.950±0.042
7	1.023±0.014	1.189±0.069	0.951±0.056	1.271±0.043
8	1.354±0.073	1.248±0.229	1.207±0.033	1.511±0.050
9	0.723±0.112	1.127±0.161	1.136±0.045	1.453±0.162
10	1.148±0.142	1.392±0.157	1.108±0.099	1.319±0.033
11	1.038±0.050	1.112±0.029	1.069±0.067	1.208±0.031
12	0.801±0.090	0.872±0.052	0.627±0.064	0.780±0.066
Mean	0.961±0.179	1.134±0.191	0.987±0.214	1.187±0.229

Table 9: Loading Peak GRF Y (BW)

	TURF		PAD	
Subject	SLOW	FAST	SLOW	FAST
1	-1.113±0.078	-1.422±0.205	-1.071±0.137	-1.455±0.068
2	-0.457±0.104	-0.795±0.240	-0.671±0.055	-0.895±0.131
3	-0.706±0.087	-1.104±0.064	-0.906±0.108	-1.206±0.078
4	-0.360±0.055	-0.583±0.131	-0.520±0.059	-
5	-0.925±0.176	-0.951±0.051	-0.932±0.248	-1.019±0.145
6	-0.521±0.076	-0.658±0.105	-0.457±0.044	-0.609±0.098
7	-0.942±0.073	-1.211±0.045	-0.829±0.049	-1.390±0.183
8	-0.547±0.035	-0.601±0.047	-	-0.611±0.113
9	-0.367±0.184	-1.093±0.169	-1.272±0.078	-1.531±0.090
10	-1.762±0.167	-1.881±0.001	-1.867±0.029	-1.898±0.006
11	-0.633±0.055	-0.776±0.052	-0.880±0.049	-1.166±0.073
12	-0.880±0.112	-1.131±0.056	-0.681±0.105	-1.063±0.092
Mean	-0.768±0.396	-1.017±0.379	-0.917±0.375	-1.185±0.377

Table 10: Push-off Peak GRF Y (BW)

	TURF		PAD	
Subject	SLOW	FAST	SLOW	FAST
1	-1.061±0.093	-1.116±0.015	-0.980±0.055	-1.019±0.017
2	-0.914±0.060	-0.824±0.082	-0.858±0.040	-0.977±0.069
3	-0.988±0.110	-1.136±0.089	-1.007±0.033	-1.242±0.040
4	-1.156±0.075	-1.392±0.062	-1.098±0.157	-1.563±0.118
5	-0.932±0.047	-1.193±0.026	-1.079±0.074	-1.245±0.041
6	-0.962±0.099	-1.042±0.015	-0.855±0.073	-1.032±0.082
7	-0.911±0.048	-1.084±0.080	-0.903±0.032	-1.285±0.121
8	-1.091±0.118	-1.265±0.095	-1.026±0.106	-1.179±0.078
9	-0.931±0.122	-1.117±0.027	-1.160±0.106	-1.277±0.096
10	-1.283±0.108	-1.593±0.172	-1.330±0.123	-1.389±0.044
11	-0.922±0.037	-1.251±0.075	-1.154±0.072	-1.501±0.059
12	-0.843±0.060	-1.093±0.041	-0.628±0.035	-0.856±0.052
Mean	-0.999±0.126	-1.176±0.191	-1.007±0.182	-1.214±0.214

Table 11: Loading Peak GRF Z (BW)

	TURF		PAD	
Subject	SLOW	FAST	SLOW	FAST
1	2.490±0.254	3.191±0.435	2.549±0.218	3.284±0.129
2	0.667±0.084	1.868±0.454	2.038±0.082	1.998±0.123
3	2.087±0.169	2.092±0.110	2.236±0.200	2.326±0.125
4	-	-	0.900±0.075	2.326±0.214
5	1.881±0.120	1.784±0.073	1.752±0.221	2.167±0.174
6	1.735±0.224	1.786±0.142	1.889±0.137	1.584±0.138
7	2.915±0.168	2.900±0.184	2.529±0.110	2.763±0.264
8	0.832±0.076	1.060±0.045	0.917±0.035	1.098±0.064
9	-	2.266±0.324	2.689±0.174	2.816±0.257
10	2.866±0.320	2.986±0.146	3.160±0.228	3.276±0.186
11	1.984±0.101	2.251±0.227	1.861±0.184	1.924±0.213
12	2.291±0.246	2.482±0.110	2.682±0.157	2.851±0.182
Mean	1.975±0.756	2.242±0.626	2.100±0.693	2.368±0.664

Table 12: Push-off Peak GRF Z (BW)

	TURF		PAD	
Subject	SLOW	FAST	SLOW	FAST
1	2.290±0.067	2.062±0.064	2.168±0.098	2.055±0.012
2	2.149±0.166	1.754±0.184	2.114±0.075	1.937±0.053
3	2.144±0.086	2.124±0.094	2.181±0.065	2.249±0.051
4	2.098±0.152	2.269±0.079	2.417±0.196	1.993±0.092
5	1.886±0.094	1.961±0.049	1.867±0.073	1.849±0.043
6	2.240±0.134	1.955±0.116	2.170±0.164	1.816±0.118
7	2.396±0.090	2.169±0.111	2.414±0.033	2.375±0.096
8	2.792±0.080	2.322±0.081	2.704±0.106	2.427±0.132
9	2.138±0.125	2.095±0.049	2.448±0.083	2.299±0.117
10	2.346±0.069	2.350±0.064	2.370±0.142	2.056±0.062
11	2.359±0.161	2.223±0.026	2.475±0.130	2.446±0.098
12	1.887±0.029	1.919±0.078	1.607±0.075	1.831±0.042
Mean	2.227±0.244	2.100±0.180	2.245±0.296	2.111±0.237

Table 13: Ankle Dorsiflexion ROM (°)

	TURF		PAD	
Subject	SLOW	FAST	SLOW	FAST
1	34.474±15.318	39.624±7.101	36.075±6.311	30.975±4.270
2	44.921±4.342	45.429±4.524	39.875±2.923	39.143±4.442
3	38.794±3.661	28.856±6.382	46.159±2.199	37.579±6.410
4	41.038±6.055	40.576±7.160	46.774±2.016	28.698±8.634
5	39.245±9.836	15.142±3.701	45.230±4.064	19.163±1.166
6	10.803±0.587	20.069±11.615	10.283±5.152	14.036±2.659
7	25.074±3.165	15.587±1.839	27.180±2.674	33.458±3.739
8	33.480±3.664	40.325±6.770	29.289±3.283	51.505±3.212
9	38.053±2.019	39.610±5.680	45.271±1.876	41.557±3.522
10	39.030±2.822	35.195±5.451	30.609±2.952	30.636±1.486
11	7.795±4.426	11.987±2.643	15.408±2.176	12.100±4.226
12	44.577±2.191	30.449±4.193	4.136±3.481	40.328±3.238
Mean	33.107±12.315	30.237±11.758	31.357±14.769	31.598±11.758

Table 14: Ankle Inversion ROM (°)

	TURF		PAD	
Subject	SLOW	FAST	SLOW	FAST
1	15.167±4.555	21.699±3.958	11.216±1.562	13.279±1.295
2	11.259±1.485	14.011±2.200	6.092±1.813	13.462±3.646
3	17.431±3.212	23.416±4.237	11.848±2.362	16.654±3.367
4	13.727±0.601	16.553±1.974	14.084±4.458	23.516±3.587
5	16.588±3.851	23.404±1.291	12.597±2.406	27.243±2.809
6	15.351±1.686	21.899±8.191	15.080±3.446	20.792±4.197
7	13.304±2.103	25.040±2.251	12.774±2.010	19.825±2.037
8	10.819±1.265	5.370±1.534	12.055±1.069	7.660±5.482
9	4.571±1.673	10.586±4.308	8.829±2.605	11.964±3.767
10	25.630±5.951	34.581±3.984	21.159±4.474	27.379±1.505
11	15.918±4.021	16.086±2.805	17.584±2.848	21.771±5.246
12	8.071±3.242	15.851±0.985	16.489±2.127	15.301±2.128
Mean	13.986±5.247	19.041±7.626	13.317±3.986	18.237±6.207

Table 15: Peak Ankle Plantarflexion Moment (N·m/kg)

	TURF		PAD	
Subject	SLOW	FAST	SLOW	FAST
1	-3.115±0.390	-2.762±0.229	-2.732±0.238	-2.518±0.217
2	-3.074±0.248	-2.606±0.208	-2.900±0.092	-
3	-3.070±0.297	-2.918±0.159	-3.007±0.178	-2.924±0.130
4	-3.751±0.293	-3.847±0.271	-3.770±0.413	-3.665±0.246
5	-2.932±0.446	-2.918±0.148	-3.070±0.134	-2.718±0.133
6	-2.747±0.235	-2.576±0.211	-2.707±0.240	-2.283±0.235
7	-2.162±0.137	-2.237±0.107	-2.430±0.122	-2.733±0.094
8	-4.874±0.242	-3.869±0.385	-4.522±0.288	-3.761±0.281
9	-2.764±0.187	-2.579±0.387	-3.286±0.286	-3.061±0.197
10	-3.181±0.351	-3.728±0.096	-3.340±0.225	-2.937±0.172
11	-2.499±0.190	-2.594±0.104	-2.534±0.042	-2.481±0.130
12	-2.956±0.070	-2.943±0.027	-2.007±0.045	-2.538±0.043
Mean	-3.094±0.681	-2.965±0.549	-3.025±0.660	-2.876±0.451

Table 16: Peak Ankle Eversion Moment (N·m/kg)

	TURF		PAD	
Subject	SLOW	FAST	SLOW	FAST
1	-0.807±0.119	-1.127±0.144	-0.785±0.113	-0.945±0.067
2	-0.287±0.027	-0.367±0.156	-0.374±0.103	-0.587±0.145
3	-0.277±0.143	-0.607±0.115	-0.674±0.119	-0.749±0.131
4	-0.676±0.072	-0.816±0.089	-0.579±0.202	-0.730±0.181
5	-0.684±0.057	-0.848±0.053	-0.601±0.058	-0.844±0.144
6	-0.610±0.032	-0.835±0.149	-0.613±0.097	-0.759±0.102
7	-0.944±0.058	-1.126±0.137	-0.667±0.093	-1.119±0.077
8	-0.669±0.093	-0.628±0.092	-0.693±0.088	-0.701±0.281
9	-0.300±0.056	-0.656±0.055	-0.328±0.138	-0.498±0.135
10	-1.379±0.219	-1.668±0.074	-1.151±0.145	-1.238±0.087
11	-0.591±0.089	-0.556±0.050	-0.772±0.130	-1.001±0.193
12	-0.609±0.096	-0.729±0.066	-0.476±0.014	-1.054±0.093
Mean	-0.653±0.308	-0.830±0.344	-0.643±0.214	-0.852±0.223

Table 17: Ankle Sagittal-plane Peak Eccentric Power (W/kg)

	TURF		PAD	
Subject	SLOW	FAST	SLOW	FAST
1	-8.238±3.205	-5.978±1.269	-6.092±1.360	-6.103±0.016
2	-10.499±1.879	-11.383±1.741	-10.415±1.136	-12.364±1.795
3	-10.917±3.222	-5.523±0.554	-12.603±1.571	-8.545±2.511
4	-12.514±2.315	-12.988±2.272	-19.965±1.690	-
5	-7.185±1.987	-7.282±0.670	-6.812±0.605	-6.118±0.334
6	-4.791±0.537	-4.609±0.378	-3.979±0.487	-3.950±0.812
7	-4.663±0.335	-4.910±0.645	-5.451±0.617	-8.386±0.570
8	-21.710±1.260	-20.549±1.703	-	-23.204±0.648
9	-7.293±0.906	-10.101±2.799	-19.381±1.652	-17.040±0.883
10	-9.953±0.628	-12.972±2.021	-9.672±0.400	-9.765±1.468
11	-5.147±1.143	-4.752±0.036	-5.468±0.502	-5.749±0.910
12	-5.509±0.134	-5.660±0.121	-2.821±0.576	-4.823±0.485
Mean	-9.035±4.777	-8.892±4.873	-10.132±6.240	-9.939±5.692

Table 18: Ankle Sagittal-plane Peak Concentric Power (W/kg)

	TURF		PAD	
Subject	SLOW	FAST	SLOW	FAST
1	10.073±1.887	10.155±1.744	9.028±0.336	8.781±0.434
2	13.148±2.483	11.646±1.923	13.867±0.708	15.810±3.089
3	11.065±1.492	10.680±0.203	10.304±0.583	11.773±1.352
4	21.285±1.127	23.594±3.479	19.669±1.194	23.648±0.878
5	13.305±2.345	15.780±1.312	14.754±1.556	14.550±1.119
6	11.954±1.352	11.360±1.710	11.014±1.318	8.877±0.687
7	8.177±0.874	12.432±0.742	10.234±0.453	15.162±1.137
8	16.456±2.777	15.604±1.465	14.718±1.334	14.529±1.477
9	9.461±1.072	10.630±2.247	12.225±1.916	11.713±0.993
10	11.504±1.796	17.605±0.733	12.334±1.170	11.040±1.728
11	10.027±1.247	15.382±0.558	12.722±0.265	13.453±1.015
12	10.535±0.826	12.630±0.709	6.798±0.253	10.079±0.557
Mean	12.249±3.571	13.958±3.893	12.306±3.299	13.285±4.048

Table 19: Ankle Frontal-plane Peak Eccentric Power (W/kg)

	TURF		PAD	
Subject	SLOW	FAST	SLOW	FAST
1	-2.798±1.164	-7.485±2.175	-1.654±0.124	-3.698±0.148
2	-0.424±0.060	-1.157±0.895	-0.511±0.345	-1.910±0.874
3	-0.476±0.370	-4.093±1.556	-1.515±0.489	-4.010±0.866
4	-	-	-0.500±0.385	-0.979±0.562
5	-2.190±1.154	-5.873±0.637	-1.325±0.366	-7.781±1.826
6	-2.966±0.365	-6.380±0.605	-2.788±0.923	-4.711±1.125
7	-3.485±0.686	-9.981±1.923	-2.090±0.636	-6.901±1.381
8	-1.794±0.340	-1.231±0.755	-1.595±0.185	-1.910±2.329
9	-0.114±0.041	-1.923±0.696	-0.590±0.633	-1.800±1.056
10	-7.820±4.256	-14.667±2.180	-6.053±1.466	-10.723±1.039
11	-2.151±0.440	-3.398±0.668	-3.353±0.406	-4.716±1.712
12	-1.998±0.858	-4.530±0.685	-1.680±0.428	-5.845±0.339
Mean	-2.383±2.109	-5.520±4.077	-1.971±1.549	-4.582±2.877

Table 20: Ankle Frontal-plane Peak Concentric Power (W/kg)

	TURF		PAD	
Subject	SLOW	FAST	SLOW	FAST
1	1.637±0.161	2.997±0.695	1.064±0.180	2.092±0.186
2	0.451±0.115	0.714±0.417	0.317±0.176	1.063±0.584
3	0.457±0.310	1.713±0.667	1.233±0.393	1.423±0.621
4	0.908±0.199	1.630±0.429	1.384±1.126	4.244±2.023
5	1.446±0.680	2.226±0.334	1.101±0.276	2.013±0.988
6	0.925±0.055	2.857±0.259	1.228±0.302	2.450±0.666
7	1.353±0.125	1.877±0.777	0.427±0.274	2.002±0.337
8	0.740±0.154	0.824±0.320	0.797±0.160	1.151±1.018
9	0.330±0.120	1.578±0.227	0.695±0.403	1.442±0.474
10	5.477±1.576	7.105±0.742	4.581±0.784	4.801±0.493
11	0.987±0.165	1.274±0.387	1.904±0.441	2.369±1.487
12	1.031±0.393	1.452±0.286	0.845±0.343	2.168±0.567
Mean	1.312±1.373	2.187±1.697	1.298±1.120	2.268±1.154

Table 21: Knee Flexion ROM (°)

	TURF		PAD	
Subject	SLOW	FAST	SLOW	FAST
1	-39.153±4.430	-46.055±4.208	-44.591±4.893	-45.723±1.124
2	-41.496±4.939	-45.728±6.083	-35.485±5.029	-33.468±5.545
3	-37.993±5.140	-31.045±7.153	-39.462±3.951	-29.368±5.871
4	-28.139±10.670	-30.657±8.155	-42.583±7.318	-16.301±4.843
5	-42.274±6.057	-40.861±2.788	-33.493±7.499	-38.093±5.920
6	-46.488±2.902	-50.128±5.728	-49.276±4.230	-54.263±6.150
7	-43.307±2.199	-36.799±4.676	-38.845±2.750	-33.209±3.443
8	-28.512±4.457	-33.287±8.724	-24.976±3.007	-40.757±5.969
9	-29.476±9.089	-29.445±2.543	-28.669±2.957	-27.163±3.623
10	-40.664±1.518	-43.999±5.408	-38.081±2.952	-36.182±4.126
11	-44.114±5.789	-41.837±6.828	-35.838±4.346	-35.172±6.814
12	-36.751±2.530	-19.252±5.598	-40.437±3.283	-38.698±1.954
Mean	-38.197±6.307	-37.425±8.930	-37.645±6.653	-35.700±9.469

Table 22: Knee Abduction ROM (°)

	TURF		PAD	
Subject	SLOW	FAST	SLOW	FAST
1	-10.286±1.544	-11.669±1.428	-8.817±0.840	-7.867±0.664
2	-	-1.400±0.836	-3.468±2.083	-4.786±0.959
3	-9.332±2.239	-8.743±1.960	-13.014±2.368	-10.885±2.959
4	-1.872±1.365	-6.810±1.891	-0.275±0.787	-3.127±2.076
5	-4.090±2.101	-3.033±1.230	-5.851±3.462	-7.956±2.765
6	-3.673±0.813	-3.338±1.063	-3.944±2.177	-5.006±1.956
7	-5.320±0.620	-10.603±2.750	-3.176±0.331	-7.424±1.315
8	-3.983±1.150	-3.359±1.516	2.275±1.535	-7.579±1.921
9	-4.599±2.454	-5.706±2.193	-0.899±0.664	-2.117±1.732
10	-6.711±1.915	-7.828±1.213	-3.499±2.009	-6.655±1.529
11	-2.743±0.982	-10.714±1.991	-10.760±1.409	-13.613±0.870
12	-1.722±1.418	-1.182±1.959	-8.104±1.410	-14.157±0.896
Mean	-4.939±2.813	-6.199±3.741	-4.961±4.507	-7.597±3.763

Table 23: Peak Knee Extension Moment (N·m/kg)

	TURF		PAD	
Subject	SLOW	FAST	SLOW	FAST
1	3.774±0.168	4.066±0.048	4.117±0.203	4.429±0.132
2	3.606±0.229	3.328±0.503	3.692±0.133	-
3	3.525±0.226	4.033±0.406	4.152±0.205	4.222±0.294
4	2.297±0.317	2.727±0.557	2.470±0.807	2.544±0.126
5	3.122±0.254	3.683±0.185	2.752±0.266	3.146±0.169
6	3.396±0.140	3.027±0.275	2.790±0.204	2.623±0.221
7	3.651±0.305	3.813±0.193	3.523±0.145	3.911±0.104
8	3.276±0.173	3.886±0.379	3.537±0.323	4.277±0.140
9	3.536±0.403	3.933±0.302	4.401±0.163	4.290±0.321
10	4.031±0.426	4.558±0.158	3.975±0.080	4.152±0.148
11	4.200±0.194	4.658±0.133	4.239±0.259	4.407±0.282
12	2.982±0.177	3.092±0.313	2.115±0.160	2.722±0.309
Mean	3.450±0.502	3.734±0.595	3.480±0.767	3.711±0.739

Table 24: Peak Knee Loading Adduction Moment (N·m/kg)

	TURF		PAD	
Subject	SLOW	FAST	SLOW	FAST
1	0.181±0.196	0.424±0.190	0.180±0.124	0.330±0.162
2	0.281±0.069	0.289±0.171	0.269±0.070	0.316±0.158
3	0.248±0.145	0.867±0.176	0.120±0.056	1.263±0.079
4	0.225±0.028	0.256±0.041	0.339±0.114	0.564±0.160
5	0.068±0.073	0.648±0.028	0.492±0.092	1.053±0.068
6	0.136±0.059	0.405±0.067	0.100±0.043	0.251±0.125
7	0.121±0.051	1.010±0.225	0.208±0.021	0.833±0.097
8	0.218±0.068	0.864±0.129	0.478±0.074	0.708±0.131
9	0.486±0.076	1.168±0.081	0.682±0.063	1.308±0.286
10	0.872±0.225	2.026±0.209	0.314±0.032	2.044±0.092
11	0.175±0.097	0.715±0.199	0.257±0.182	0.605±0.210
12	0.236±0.060	0.446±0.052	0.162±0.050	0.417±0.076
Mean	0.271±0.216	0.760±0.494	0.300±0.174	0.808±0.530

Table 25: Peak Knee Push-off Adduction Moment (N·m/kg)

	TURF		PAD	
Subject	SLOW	FAST	SLOW	FAST
1	0.193±0.080	0.644±0.067	0.302±0.045	0.455±0.033
2	0.322±0.068	0.377±0.150	0.230±0.039	0.254±0.099
3	0.386±0.094	0.847±0.158	0.776±0.073	0.836±0.074
4	-0.081±0.029	-0.070±0.010	0.102±0.022	0.227±0.090
5	0.424±0.152	0.575±0.046	0.659±0.113	1.004±0.086
6	0.386±0.042	0.477±0.081	0.370±0.067	0.311±0.084
7	0.205±0.065	0.634±0.182	0.201±0.056	0.671±0.123
8	0.138±0.047	-	0.414±0.141	0.446±0.158
9	0.377±0.025	1.103±0.179	0.134±0.072	0.942±0.133
10	0.554±0.198	-	0.248±0.149	0.887±0.280
11	0.283±0.143	0.655±0.114	0.382±0.064	0.552±0.140
12	0.353±0.109	0.600±0.046	0.376±0.074	0.340±0.057
Mean	0.295±0.164	0.584±0.304	0.349±0.200	0.577±0.282

Table 26: Knee Sagittal-plane Peak Eccentric Power (W/kg)

	TURF		PAD	
Subject	SLOW	FAST	SLOW	FAST
1	-27.606±8.641	-42.875±6.991	-33.172±4.965	-51.013±1.199
2	-13.935±2.775	-15.734±3.989	-13.655±2.204	-13.114±1.912
3	-16.004±0.846	-23.602±4.242	-22.024±2.355	-23.883±5.148
4	-4.432±2.090	-8.384±6.266	-11.370±1.462	-
5	-17.161±3.045	-24.136±3.024	-13.376±1.623	-20.010±4.709
6	-18.620±1.820	-19.855±3.044	-16.149±2.465	-17.317±1.769
7	-26.267±1.652	-31.596±3.040	-22.340±2.860	-26.712±3.919
8	-8.423±1.638	-12.495±3.651	-	-14.355±1.381
9	-13.185±1.569	-23.748±6.027	-25.875±5.033	-26.573±7.647
10	-38.899±7.334	-50.431±5.498	-36.607±2.999	-43.964±3.872
11	-28.725±1.794	-37.889±5.570	-26.258±5.166	-29.108±2.117
12	-13.733±1.851	-23.411±2.012	-19.098±2.547	-18.529±1.412
Mean	-18.916±9.729	-26.180±12.471	-20.701±8.704	-24.561±12.278

Table 27: Knee Sagittal-plane Peak Concentric Power (W/kg)

	TURF		PAD	
Subject	SLOW	FAST	SLOW	FAST
1	9.893±1.472	8.996±1.690	11.330±0.450	10.787±0.699
2	10.734±0.803	11.540±1.314	11.868±0.846	13.619±1.742
3	10.867±1.142	12.777±2.221	11.411±0.857	13.159±1.523
4	8.029±2.226	10.112±2.287	9.149±1.520	7.522±0.736
5	9.985±1.712	12.670±0.881	8.759±1.830	11.606±0.751
6	11.507±0.921	10.508±0.640	8.504±0.515	8.982±0.867
7	11.472±0.664	13.140±1.108	11.901±0.892	15.436±0.384
8	9.325±0.873	11.412±0.714	10.128±1.531	11.893±0.608
9	8.910±0.644	10.555±1.632	12.186±0.549	11.264±0.554
10	8.360±1.419	15.228±0.392	10.518±1.153	7.400±1.534
11	10.794±1.116	17.783±1.413	12.554±1.125	16.069±1.569
12	7.849±2.057	11.544±1.657	3.989±0.867	6.411±0.845
Mean	9.811±1.307	12.189±2.402	10.192±2.386	11.179±3.133

Table 28: Knee Frontal-plane Peak Loading Eccentric Power (W/kg)

	TURF		PAD	
Subject	SLOW	FAST	SLOW	FAST
1	-0.329±0.324	-0.487±0.386	-0.577±0.235	-1.786±0.273
2	-0.537±0.295	-0.431±0.172	-0.114±0.080	-0.396±0.221
3	-0.309±0.229	-1.268±0.549	-0.435±0.344	-2.701±1.270
4	-0.014±0.020	-0.056±0.074	-0.567±0.284	-0.803±0.547
5	-0.450±0.365	-1.308±0.433	-0.335±0.186	-3.065±0.305
6	-0.769±0.324	-1.337±0.962	-1.641±0.454	-2.833±0.574
7	-1.279±0.559	-0.892±0.481	-0.758±0.175	-0.277±0.411
8	-0.155±0.074	-	-	-1.781±0.878
9	-	-3.860±2.160	-0.265±0.116	-1.854±0.807
10	-1.581±1.294	-5.902±3.439	-1.181±1.078	-8.684±2.256
11	-0.328±0.218	-1.084±0.157	-0.745±0.236	-2.001±1.090
12	-0.879±0.238	-2.632±1.788	-5.092±1.821	-2.815±0.633
Mean	-0.603±0.483	-1.751±1.745	-1.065±1.404	-2.416±2.188

Table 29: Knee Frontal-plane Peak Loading Concentric Power (W/kg)

	TURF		PAD	
Subject	SLOW	FAST	SLOW	FAST
1	0.318±0.131	0.990±0.609	0.248±0.136	0.906±0.071
2	0.462±0.184	0.327±0.517	0.267±0.131	0.246±0.090
3	0.165±0.150	1.156±0.466	0.605±0.435	0.752±0.350
4	0.255±0.073	0.382±0.145	1.215±0.722	1.708±0.834
5	0.341±0.215	1.290±0.665	0.628±0.113	2.183±1.013
6	1.146±0.326	1.969±1.462	1.335±0.534	2.837±0.739
7	0.932±0.239	2.869±0.525	0.900±0.278	1.527±0.664
8	0.329±0.217	-	1.050±0.182	0.710±0.555
9	0.197±0.119	3.770±0.025	0.320±0.114	2.468±0.864
10	2.169±0.957	5.742±2.828	1.532±1.322	7.493±2.298
11	0.289±0.159	1.535±0.632	0.778±0.347	1.276±0.462
12	0.652±0.300	1.291±1.114	4.144±0.668	1.851±0.264
Mean	0.605±0.578	1.938±1.619	1.085±1.052	1.996±1.894

Table 30: Knee Frontal-plane Peak Push-off Power (W/kg)

	TURF		PAD	
Subject	SLOW	FAST	SLOW	FAST
1	-0.613±0.241	-1.867±0.589	-0.666±0.129	-1.199±0.297
2	-0.780±0.371	-0.847±0.695	-0.249±0.045	-0.281±0.146
3	-0.649±0.258	-1.587±0.362	-0.491±0.139	-1.524±0.358
4	-0.286±0.091	-0.382±0.141	-0.635±0.317	-0.736±0.298
5	-0.855±0.249	-0.642±0.227	-0.372±0.182	-0.360±0.080
6	-0.583±0.069	-0.800±0.168	-0.475±0.095	-0.656±0.340
7	-0.316±0.173	-1.446±0.780	-0.218±0.068	-0.258±0.144
8	-0.025±0.011	-	-0.079±0.053	-0.264±0.162
9	-	-0.464±0.070	-0.106±0.088	-0.401±0.195
10	-0.277±0.201	-0.386±0.204	-0.282±0.152	-
11	-0.756±0.503	-1.672±0.199	-0.423±0.173	-0.937±0.342
12	-0.594±0.200	-1.423±0.173	-1.072±0.282	-0.304±0.136
Mean	-0.521±0.259	-1.047±0.560	-0.422±0.277	-0.629±0.431

Table 31: Peak Hip Loading Extension Moment (N·m/kg)

	TURF		PAD	
Subject	SLOW	FAST	SLOW	FAST
1	-0.244±0.516	-1.188±0.187	-1.098±0.204	-1.150±0.276
2	-0.532±0.124	-2.067±0.454	-1.584±0.140	-2.275±0.359
3	-0.265±0.443	-1.069±0.107	-0.827±0.100	-0.658±0.106
4	-0.749±0.199	-1.777±0.334	-1.438±0.260	-3.554±0.564
5	-0.205±0.127	-1.549±0.136	-1.029±0.115	-1.950±0.198
6	-1.492±0.223	-2.238±0.412	-1.609±0.158	-2.150±0.255
7	-0.703±0.149	-2.191±0.345	-0.926±0.062	-1.505±0.195
8	-0.711±0.095	-1.628±0.216	-1.312±0.093	-1.209±0.270
9	-0.566±0.135	-1.656±0.206	-1.770±0.289	-1.557±0.214
10	-0.312±0.265	-0.910±0.166	-1.298±0.039	-1.411±0.205
11	-0.685±0.227	-1.565±0.435	-1.001±0.094	-0.925±0.275
12	0.124±0.199	-1.589±0.327	-1.332±0.290	-1.072±0.208
Mean	-1.148±0.272	-1.619±0.420	-1.269±0.297	-1.618±0.782

Table 32: Peak Hip Push-off Extension Moment (N·m/kg)

	TURF		PAD	
Subject	SLOW	FAST	SLOW	FAST
1	-0.861±0.391	-1.147±0.139	-1.647±0.334	-1.997±0.157
2	-1.662±0.441	-2.522±0.137	-1.256±0.267	-1.881±0.368
3	-0.873±0.453	-1.015±0.281	-0.638±0.100	-1.432±0.282
4	-1.032±0.228	-1.640±0.516	-1.663±0.203	-2.661±0.598
5	-1.174±0.398	-1.001±0.030	-1.158±0.110	-1.051±0.272
6	-1.877±0.090	-3.091±0.305	-1.585±0.148	-3.060±0.194
7	-1.165±0.128	-1.656±0.429	-0.989±0.201	-1.540±0.214
8	-1.127±0.124	-0.694±0.107	-1.074±0.113	-1.111±0.209
9	-0.910±0.159	-2.045±0.336	-1.786±0.261	-2.124±0.323
10	-1.721±0.121	-1.517±0.204	-1.682±0.202	-1.456±0.152
11	-0.954±0.175	-0.468±0.152	-0.556±0.151	-0.560±0.211
12	-0.817±0.250	-1.777±0.178	-1.302±0.245	-1.500±0.133
Mean	-1.181±0.368	-1.548±0.756	-1.278±0.412	-1.698±0.697

Table 33: Peak Hip Loading Adduction Moment (N·m/kg)

	TURF		PAD	
Subject	SLOW	FAST	SLOW	FAST
1	-0.194±0.162	0.895±0.279	-0.195±0.192	0.104±0.314
2	0.087±0.287	0.898±0.203	0.258±0.114	-
3	0.185±0.191	1.061±0.266	0.114±0.105	0.983±0.385
4	0.554±0.226	0.673±0.039	0.681±0.243	0.728±0.211
5	0.138±0.069	0.467±0.207	0.313±0.161	0.641±0.066
6	0.013±0.061	0.532±0.178	0.085±0.195	0.452±0.157
7	0.162±0.162	0.706±0.147	0.175±0.082	0.074±0.081
8	0.541±0.084	1.696±0.202	0.862±0.168	1.438±0.129
9	0.564±0.074	0.660±0.074	0.913±0.056	1.329±0.155
10	0.468±0.272	1.653±0.340	0.195±0.153	1.699±0.100
11	0.107±0.172	0.961±0.154	0.245±0.032	0.156±0.076
12	0.469±0.136	1.164±0.249	0.308±0.146	0.886±0.209
Mean	0.258±0.251	0.947±0.398	0.329±0.328	0.772±0.557

Table 34: Peak Hip Sagittal-plane Eccentric Power (W/kg)

	TURF		PAD	
Subject	SLOW	FAST	SLOW	FAST
1	-2.504±1.640	-3.698±0.791	-6.049±1.583	-4.323±1.164
2	-2.611±2.003	-5.481±3.502	-0.579±0.739	-1.304±2.127
3	-2.707±1.587	-2.895±1.639	-1.150±0.751	-2.288±1.659
4	-0.904±0.691	-0.822±1.854	-1.693±1.091	-
5	-2.588±1.296	-4.494±0.475	-1.684±0.509	-3.562±1.646
6	-6.189±0.771	-7.842±2.991	-7.129±2.099	-7.831±2.736
7	-4.806±0.945	-7.617±3.212	-2.549±1.187	-3.513±1.313
8	-0.188±0.567	-	-	-0.879±0.782
9	-0.519±1.344	-4.929±0.785	-3.864±2.137	-5.160±2.354
10	-5.916±1.119	-5.452±1.764	-6.321±1.323	-5.338±2.055
11	-4.191±2.015	-4.139±1.176	-3.112±0.824	-3.482±1.101
12	-1.220±0.791	-2.720±1.151	-2.941±0.633	-7.499±1.551
Mean	-2.862±2.032	-4.554±2.073	-3.370±2.226	-4.107±2.251

Table 35: Peak Hip Sagittal-plane Concentric Power (W/kg)

	TURF		PAD	
Subject	SLOW	FAST	SLOW	FAST
1	1.464±0.626	2.929±0.436	3.934±0.595	5.606±1.082
2	4.315±2.028	4.738±1.498	1.846±0.655	2.362±1.069
3	2.198±0.811	2.357±0.991	1.579±0.152	3.874±0.728
4	2.896±1.027	4.596±1.534	5.032±1.026	9.954±3.135
5	1.215±0.688	3.709±0.386	2.067±0.715	2.992±0.869
6	6.586±1.204	10.944±1.950	6.114±0.593	11.932±1.439
7	2.423±0.991	3.071±0.522	1.525±0.497	2.930±0.917
8	2.613±0.262	-	1.801±0.524	1.802±1.099
9	1.778±0.743	5.456±1.072	3.169±0.678	3.890±0.594
10	2.744±1.095	2.539±1.283	2.774±1.170	2.503±0.549
11	1.306±0.481	1.298±0.318	1.874±0.581	1.456±1.009
12	1.503±0.518	3.955±0.326	2.434±0.560	3.200±1.313
Mean	2.587±1.534	4.145±2.554	2.846±1.475	4.375±3.280

Vita

Thomas K. Elvidge was born in Southend-on-Sea, Essex, United Kingdom on May 19th 1993 to Mark and Colleen Elvidge. He attended Southend High School for Boys and went on to earn a Bachelors of Science with Honors degree in Sport and Exercise Science at the University of Bath, United Kingdom. As a part of his undergraduate degree, he spent a year working as an intern for Southampton Football Club within the Sports Medicine Department. He enrolled at the University of Tennessee, Knoxville to study for a Masters of Science degree in Biomechanics and will graduate in May 2017. Upon graduation, he will be going to work for Adidas in Portland, OR.