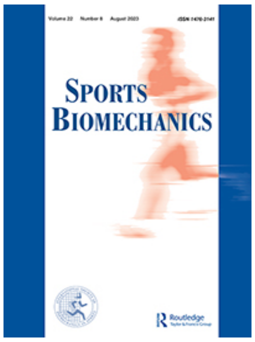

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Assessments performed on harder surfaces can misrepresent ACL injury risk

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ABSTRACT
Changes in surface hardness are likely to alter an athlete's movement strategy. Anterior cruciate ligament (ACL) injury risk assessments that are performed on a different surface to that used for training and competition may, therefore, not represent an athlete's on-field movement strategies. The aim of this study was to examine the influence of surface hardness on multidirectional field sport athletes' movement strategies in movements that are commonly used in ACL injury risk assessments (bilateral and unilateral drop jumps, and a cutting manoeuvre). Ground reaction forcesand three-dimensional lower limb kinematics were recorded from 19 healthy, male, multidirectional field sport athletes performing bilateral and unilateral drop jumps, and a 90° cutting task on Mondo track (harder surface) and artificial turf (softer surface). Continuous (statistical parametric mapping) and discrete analyses revealed alterations in vertical and horizontal braking forces and knee and hip moments between surfaces of different hardness in all three movements ($p \leq 0.05$, $d > 0.5$). Injury risk assessments performed on a harder surface (e.g. Mondo track) can misrepresent an athlete's risk of ACL injury compared to the same movements performed on a softer more cushioned surface that is typically used for training and/or matches (e.g. artificial turf).

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ACL injury risk; field-based assessments; surface hardness; lower limb biomechanics

Introduction

Anterior cruciate ligament (ACL) rupture is the leading injury to cause absence from playing and training in multidirectional field sports (e.g. Rugby Union, football, and Gaelic football) (Awwad et al., 2019; Murphy et al., 2012; Pulici et al., 2022; Yeomans et al., 2021). The majority of ACL ruptures are non-contact in nature (Alentorn-Geli et al., 2009; Johnston et al., 2018), occurring in the initial deceleration phase following ground contact during unilateral landings and change of direction manoeuvres (e.g. cutting) (N. Bates et al., 2020; B. Boden et al., 2000; Cochrane et al., 2007). To aid injury risk assessments, it is thus vital to monitor an athlete's lower limb biomechanics in movements which replicate common ACL injury mechanisms (e.g, landing from a jump or change of direction). Moreover, playing surfaces that are commonly used in multidirectional field sports (e.g. natural grass, artificial turf) have been shown to influence

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ACL injury risk (Dragoo et al., 2013; Howard et al., 2020; Loughran et al., 2019; Robertson et al., 2022), although findings are inconclusive on which surface is most likely to reduce or increase injury rates. The ability to assess an athlete's risk of injury may, therefore, also be hampered by changes in surface on which ACL injury risk assessments take place and the surface athletes train and/or play matches on.

In multidirectional field sports, the type of landing surface used during assessments of ACL injury risk (e.g. rubber gym surface, indoor wooden/concrete flooring) may differ to the surface athletes perform on during training and matches (e.g. artificial turf, natural grass). Surfaces can have different shock absorption properties (such as surface hardness), which are important for improving player comfort and diminishing the risk of injuries (Farhang et al., 2016; Silva et al., 2009). Surface hardness relates to the ability of a surface to absorb energy when an athlete impacts the surface (Brosnan et al., 2009; Rogers, 1988) and is directly related to the resistance of a surface to undergo vertical deformation (Shorten & Himmelsbach, 2002). Mechanical tests have shown that harder surfaces absorb a smaller proportion of energy upon impact during landing compared to softer more cushioned surfaces (Allgeuer et al., 2008; Barrett et al., 1997). Decreased energy absorption is often accompanied by larger peak vertical ground reaction forces (GRFs) that are also applied over a shorter time period (Benanti et al., 2013; Bishop, 2003). Considering that larger GRFs are a mechanical indicator of higher external knee moments (Dai et al., 2014), assessments that are performed on harder surfaces (e.g. rubber gym surface) may misrepresent biomechanical deficits associated with ACL injury compared to softer surfaces that athletes train and play matches on (e.g.artificial turf).

Previous research has shown larger peak vertical (Lozano-Berges et al., 2021; McGhie & Ettema, 2013) and horizontal braking (Stiles & Dixon, 2006) GRFs on harder surfaces. In contrast, others have reported lower peak vertical GRFs (McNitt-Gray et al., 1994; Stiles & Dixon, 2006, 2007) and, even, similar magnitudes of vertical GRF between harder and softer surfaces (Dixon et al., 2000, 2005). Additionally, higher loading rates (i.e. peak vertical GRFs over shorter time periods) have been reported on harder surfaces (Dixon et al., 2000; Lozano-Berges et al., 2021; Stiles & Dixon, 2007) and have been proposed to influence ACL injury (Paterno et al., 2007), potentially due to limited time to dissipate the impact forces (Huang et al., 2020). Whilst surface hardness clearly influences the magnitude and rate of impact forces on landing, athletes are able to attenuate these forces on harder surfaces through neuromuscular adjustments and by altering their kinematics (Qu et al., 2022; Wang et al., 2019). For example, increasing sagittal plane lower limb joint range of motion (RoM) may help effectively dissipate the forces and moments exerted on the body during running (Dixon et al., 2005; Gerritsen et al., 1995). Increased sagittal plane flexion RoM is also associated with increased energy absorption (i.e. negative work) to decelerate the body during landing (Skinner et al., 2015). Thus, safer landing strategies to help reduce ACL injury risk can be employed by increasing sagittal plane lower limb joint flexion RoM and negative work when landing on harder surfaces (Alonzo et al., 2020; Skinner et al., 2015).

In movements typically used to assess ACL injury risk (a bilateral drop jump and 90° cut), peak ankle, knee and hip (dorsi)flexion moments were larger on a harder surface (artificial turf without cushioning underlay), with no increases in lower limb joint flexion RoM (Lozano-Berges et al., 2021). Larger knee flexion moments have been associated with greater anterior translation of the tibia relative to the femur (Sell et al., 2007), which is a primary ACL loading mechanism (Markolf et al., 1990). Joint stiffness is also modified by changes in joint moments and RoM, which may contribute to ACL injury (Jones, Moore, King, Stiles, Verheul, et al., 2022). Furthermore, peak knee abduction moments were higher on a harder surface during a 90° cut (Lozano-Berges et al., 2021). Deficits in frontal plane mechanics, such as increased knee abduction moments, observed on a harder testing surface (e.g. rubber gym surface) may lead to athletes being incorrectly identified at increased ACL injury risk compared to a softer more cushioned surface which more closely replicates where athletes train and play matches (e.g. artificial turf).

Biomechanical assessments of ACL injury risk often take place on harder surfaces (e.g. rubber gym surface), as opposed to softer more cushioned surfaces that may more closely represent the surface that athletes train and/or play matches on (e.g. artificial turf). Differences in surface hardness are likely to alter an athlete's movement strategies, such as GRFs and lower limb joint moments, RoM, stiffness and work (Jones, Moore, King, Stiles, Laudani, et al., 2022; Jones, Moore, King, Stiles, Verheul, et al., 2022; Nigg & Yeadon, 1987). It is likely that movement strategies associated with ACL injury would be more pronounced on harder surfaces. Potential biomechanical deficits would therefore be highlighted to a greater extent when testing on harder surfaces. Further research is needed to establish the extent to which an athlete's movement strategy changes between surfaces of varying hardness in movements included in ACL injury risk assessments.

Current research examining biomechanical differences between surfaces has primarily focused on discrete data analysis during the braking phase. The braking phase is important since non-contact ACL ruptures typically occur within the first ~ 60 milliseconds after initial contact (N. Bates et al., 2020). However, this type of analysis presents several limitations. Firstly, biomechanical differences may occur over time periods of the braking phase that cannot be described by discrete variables. Secondly, the magnitude of discrete variables may not change between surfaces, but the timing of these variables may differ. Finally, discrete data variables may not be independent, therefore analysing them as such may produce biased results exhibiting type 1 and type 2 errors (Knudson, 2009; T. C. Pataky et al., 2013, 2015). Rather than analysing discrete data points during the braking phase alone, the use of continuous data analysis techniques, such as statistical parametric mapping (SPM) (T. C. Pataky et al., 2013), may provide novel insights into biomechanical differences between surfaces over the whole of the braking phase.

The aim of this study was to examine the influence of surface hardness on an athlete's movement strategy in movements that are commonly used to assess risks of ACL injury (bilateral and unilateral drop jumps, and a cutting manoeuvre), using both discrete variables and continuous data. It was hypothesised that (1) movement strategies associated with ACL injury would be more pronounced on a harder surface (e.g. vertical and braking impact forces and knee flexion and abduction moments), (2) athletes would increase lower limb joint RoM and eccentric joint work during landing to attenuate these increased forces and moments and that (3) joint stiffness would be similar between surfaces due to relative increases in both joint moments and RoM.

Method

Participants

Nineteen male, multidirectional field sport athletes who regularly trained and competed on artificial turf (football, rugby union, rugby league, American football) and aged between 18 and 35 years participated in this study (age: 24 ± 4 years; height: 1.82 ± 0.07 m; mass: 85.7 ± 9.4 kg). Participants were required to be free from lower limb injury in the 6 months prior to testing. Prior to data collection, all participants provided informed consent. Ethical approval was obtained from Cardiff Metropolitan University ethics committee, with reference number PGR-3539.

Experimental procedure

Participants completed a short warm-up consisting of slow running and stretching, and then performed the following three movements in order: a bilateral drop jump from 30 cm, a unilateral drop jump from 20 cm and a 90° pre-planned cut (hereafter referred to as the cut). Briefly, during the drop jumps participants placed their hands on their hips and were told to roll from the box (placed directly in front of the force plate) and upon hitting the ground, to jump as high as they could while spending as little time as possible on the force plate. For the bilateral drop jump, participants began with their feet approximately hip width apart and landed with one foot on each of the force plates (King, Richter, Franklyn-Miller, Daniels, Wadey, Moran, et al., 2018). For the cut, participants were required to start at a distance of 5 m from the force plates, run as fast as possible towards the force plates, cutting left or right while planting their contralateral foot on the force plate, and then to accelerate away after changing direction (King, Richter, Franklyn-Miller, Daniels, Wadey, Jackson, et al., 2018). Before test trials were captured, participants underwent two sub-maximal practice trials of each movement. Trials were performed on a harder Mondo athletic track surface (Mondo track) first, followed by the same battery of tests on a softer more cushioned third generation (3G) artificial turf surface with no infill (artificial turf) which more closely represents the surfaces that athletes typically train and play matches on. Participants wore their own trainers for trials performed on Mondo track, and for the artificial turf trials participants used the footwear they would usually wear on artificial turf (e.g. cleats) to enhance the ecological validity. The dominant leg for all participants was tested first (determined by the leg with which an athlete would prefer to kick a ball (Peters, 1988)). Participants were given a 30 s rest period between trials to avoid fatigue. Five valid attempts (determined by confirming a full foot contact on the force plate) were recorded for each limb.

Biomechanical data collection

All data collections took place in the National Indoor Athletics Centre, Cardiff Metropolitan University. A 12-camera three-dimensional motion capture system (250 Hz; Vicon Motion Systems Ltd., Oxford, UK) was used to collect marker trajectories. Two force platforms (1000 Hz; 9287CA, 90 × 60 cm, Kistler, Winterthur, Switzerland) were embedded in the ground to collect GRFs and were synchronised to the Vicon system. The force platforms were covered with the same surface as the surrounding

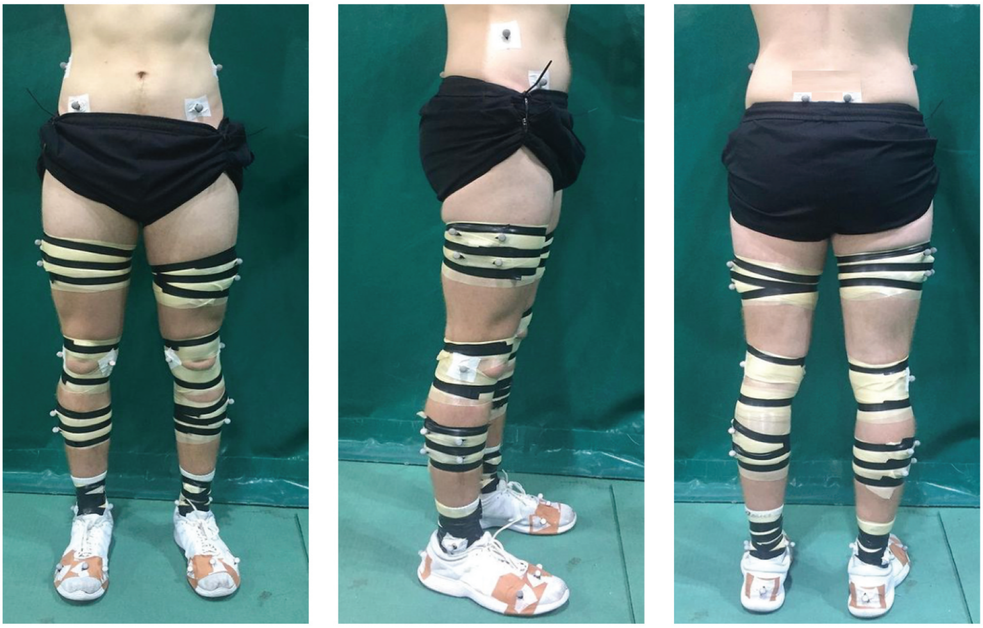

Figure 1. Example marker set-up.

testing area (Mondo track or artificial turf). This was particularly important for the cut to ensure that participants completed the whole trial on the same surface. During the trials, lower-body kinematic data were collected using a 38 reflective-marker set attached to the participants' skin bilaterally on the iliac crest, anterior and posterior superior iliac spine, lateral and medial femoral epicondyles, lateral and medial malleoli, first and fifth metatarsal heads, head of the second toe, and on the calcaneus, in addition to technical clusters of four markers attached in the middle of the thigh and shank segments (Figure 1). Reflective markers were secured to the skin or to the shoe using tape. All markers were applied by the same researcher to ensure accuracy and consistency throughout data collection. Following completion of trials on the Mondo track, the artificial turf was secured directly on top of the Mondo track (Figure 2).

Biomechanical data processing and analysis

Static and functional calibration trials were recorded of participants standing in the anatomical position and completing five body weight squats and five leg swings on each side, respectively. Initial marker labelling and gap filling took place in Vicon Nexus (v.2.12.1, Oxford Metrics Inc., Oxford, UK). Data were exported to Visual3D (v.6, C-motion, Rockville, MD, USA) where static calibration trials were used to create a seven segment (pelvis, thighs, shanks and feet) six degrees-of-freedom kinematic model. The local segment coordinate system for each segment was defined at the proximal joint centre. Segment inertial characteristics were estimated for each participant based on Dempster's regression equations (Dempster, 1955) and represented as geometric volumes (Hanavan, 1964). The ankle and knee joint centres were defined as the

Figure 2. Laboratory set-up with artificial turf.

mid-point between the medial and lateral markers, whilst the location of the hip joint centre was defined using a CODA pelvis orientation (Bell et al., 1989). Inverse kinematics and inverse dynamics were performed to determine lower limb joint angles, external moments and power. Kinematic and moment data were filtered using a fourth-order low pass Butterworth filter at 15 Hz (Kristianslund et al., 2012). The GRF data were filtered using a fourth-order low pass Butterworth filter at 35 Hz, which was determined using residual analysis, and GRF variables were expressed relative to body weight (BW). Ankle, knee, and hip kinematics were expressed relative to the proximal segments defined by an XYZ ordered Cardan angle sequence (Grood & Suntay, 1983). Positive sagittal plane angles and external joint moments relate to ankle dorsiflexion, and knee and hip flexion. Positive frontal plane angles and external joint moments relate to ankle eversion, and knee and hip abduction. Joint moments were normalised to body mass (Winter, 2009). All other biomechanical variables described below were calculated using these kinematic and kinetic data and exported to MATLAB (version R2022a, MathWorks Inc, Natick, MA, USA) and Excel (Microsoft Corporation, Redmond, WA, USA) for further processing and analysis.

The GRF data were reported in the vertical and horizontal anterior–posterior directions. The horizontal anterior–posterior direction represents propulsion (positive) and braking (negative), respectively. Kinetic and kinematic joint variables were reported in the sagittal and frontal planes. Increases in vertical and braking GRFs and alterations in sagittal and frontal plane variables have been suggested as important contributors to ACL injury (Bakker et al., 2016; Coventry et al., 2006; Peebles et al., 2020; Shimokochi et al., 2013; Yeow et al., 2011; Yu et al., 2006). The GRF, kinematic and kinetic analyses were carried out for the braking phase of the bilateral and unilateral drop jumps, and the cut. The braking phase was defined as the time between initial contact (determined as GRF > 20 N) to maximum knee flexion. Data from the braking phase of the first landing (i.e. landing from the box) of each trial for the bilateral and unilateral drop jump were analysed. Several discrete variables that have been linked to ACL injury risk were calculated. Peak force was identified as the maximum value during the braking phase, and time to peak force was also reported. Average vertical loading rate was calculated as the ratio of peak force to time to peak force. Instantaneous vertical loading rate was determined as the derivative of the vertical GRF with respect to time in the braking phase, and the maximum value represented the variable. Sagittal and frontal plane joint work was computed as the integral of joint power with respect to time, in which negative work represented energy absorption, and total lower extremity work was calculated as the sum of the ankle, knee and hip work in the respective planes. The relative contribution of the ankle, knee and hip work to total lower extremity joint work was calculated in both planes. Sagittal and frontal plane angle at initial contact, peak angle, joint RoM (angle at initial contact to angle at maximum knee flexion) and peak moment were reported. Sagittal plane change of joint moment was calculated as the magnitude of change from initial contact to maximum knee flexion, and subsequently sagittal plane joint stiffness was determined as the ratio of change in joint moment to joint RoM (*i.e.*, $k_j = \frac{\Delta M_j}{\Delta \theta_j}$, in which k represents joint stiffness, ΔM represents magnitude of change in joint moment, $\Delta\theta$ represents joint RoM and j represents the hip, knee or ankle joint).

Statistical analysis

Means and standard deviations (SDs) of five trials for each participant were computed for each discrete variable. Mean data from both the right and left leg were used, resulting in 38 samples being included in each analysis. For discrete data the Shapiro–Wilk test was used to test normality of the differences between surfaces for all variables in each movement. For each movement, a paired samples t-test was performed on variables that satisfied the assumptions of normality, while a Wilcoxon signed-rank test was performed on variables which violated the assumptions of normality.

Continuous waveform data for vertical and horizontal anterior-posterior GRFs, and sagittal and frontal plane joint angles and joint moments were temporally normalised from 0 to 100% of the braking phase (101 data points). The open-source spm1d MATLAB package (www.spm1d.org v0.4.7; run on MATLAB vR2022a, MathWorks Inc, MA, USA) was then used to identify regions of differences between surfaces using one-dimensional paired t-test SPM analysis (T. Pataky, 2012). A Cohen's d effect size was calculated in a point-by-point manner over the braking phase to determine the magnitude of significant differences, and the mean effect size over phases where $p \leq 0.05$ was calculated. Cohen's d effect sizes were interpreted as small ($0.2 \geq d < 0.5$), medium ($0.5 \geq d < 0.8$), and large ($d \geq 0.8$) (Cohen, 2013). If differences were found across >5% of the braking phase, the start and end points (% of the braking phase) of the period over which the significant difference was observed were reported. For this study data were only reported if $p \leq 0.05$ and Cohen's $d > 0.5$.

Results

Ground reaction forces

During the cut on Mondo track, there were higher vertical peak forces ($Z = -4.13$, $p < 0.001$, $d = -1.08$), and lower instantaneous loading rates ($t = -3.25$, $p = 0.002$,

$d = -0.53$) compared to artificial turf. On Mondo track, peak posterior GRFs were larger in the bilateral drop jump ($t = -4.06$, $p < 0.001$, $d = -0.67$) and the cut ($t = -6.53$, $p <$ 0.001, $d = -1.06$) compared to artificial turf. During the bilateral ($Z = -3.73$, $p = 0.001$, $d = -0.81$) and unilateral ($Z = -3.88$, $p < 0.001$, $d = -0.99$) drop jumps, a longer time to peak posterior GRF was found on artificial turf compared to Mondo track.

The GRF continuous waveform during landing was affected by the ground surface in all three movements. There were significant differences over a duration >5% of the braking phases, in both the vertical and anterior–posterior GRFs (Figure 3).

Sagittal plane

Knee joint

Peak knee flexion moments ($t = 3.98$, $p < 0.001$, $d = 0.65$) and eccentric knee work ($t = -3.54$, $p = 0.001$, $d = -0.57$) during the cut were higher on Mondo track compared to artificial turf.

Knee moments were affected by the ground surfaces in all movements, as revealed by the SPM analysis (Figure 4(a–c)). During the cut, smaller knee flexion angles were observed on Mondo track compared to artificial turf (0–42%, $p = 0.033$, $d = 0.52$).

Distal and proximal joints

Peak hip flexion moments were larger in the bilateral drop jump on Mondo track compared to artificial turf ($t = 3.16$, $p = 0.003$, $d = 0.51$). During the unilateral drop jump, hip flexion angles at initial contact were larger on Mondo track compared to artificial turf ($t = 3.22$, $p = 0.003$, $d = 0.52$), which supports the findings of the SPM analysis. Hip RoM was larger on artificial turf compared to on Mondo track in the unilateral drop jump ($t = -3.14$, $p = 0.003$, $d = -0.51$). On artificial turf, magnitudes of change in ankle moment were greater in the bilateral ($Z = -2.60$, $p = 0.009$, $d = -0.60$) and unilateral ($t = -3.14$, $p = 0.003$, $d = -0.51$) drop jumps compared to on Mondo track. During the cut, ankle stiffness was lower on artificial turf compared to on Mondo track ($Z = -2.97$, $p = 0.003$, $d = -0.72$).

Hip moments were affected by the ground surfaces in all movements, as revealed by the SPM analysis (Figure 4(d–f)). Larger ankle dorsiflexion moments were observed on Mondo track compared to artificial turf for the bilateral (0–34%, $p = 0.001$, $d = 0.76$) and unilateral (0–22%, $p = 0.008$, $d = 0.83$) drop jumps, especially during the earliest stages of landing. Likewise, for the unilateral drop jump, smaller ankle plantarflexion angles and larger hip flexion angles were displayed on Mondo track compared to artificial turf (6–36%, $p = 0.025$, $d = 0.52$ and 0–68%, $p = 0.015$, $d = 0.51$, respectively).

Frontal plane

Knee joint

Knee abduction moments were larger in the unilateral drop jump during the middle portion of the braking phase on artificial turf compared to Mondo track, as revealed by the SPM analysis (33–58%, $p = 0.001$, $d = 0.59$).

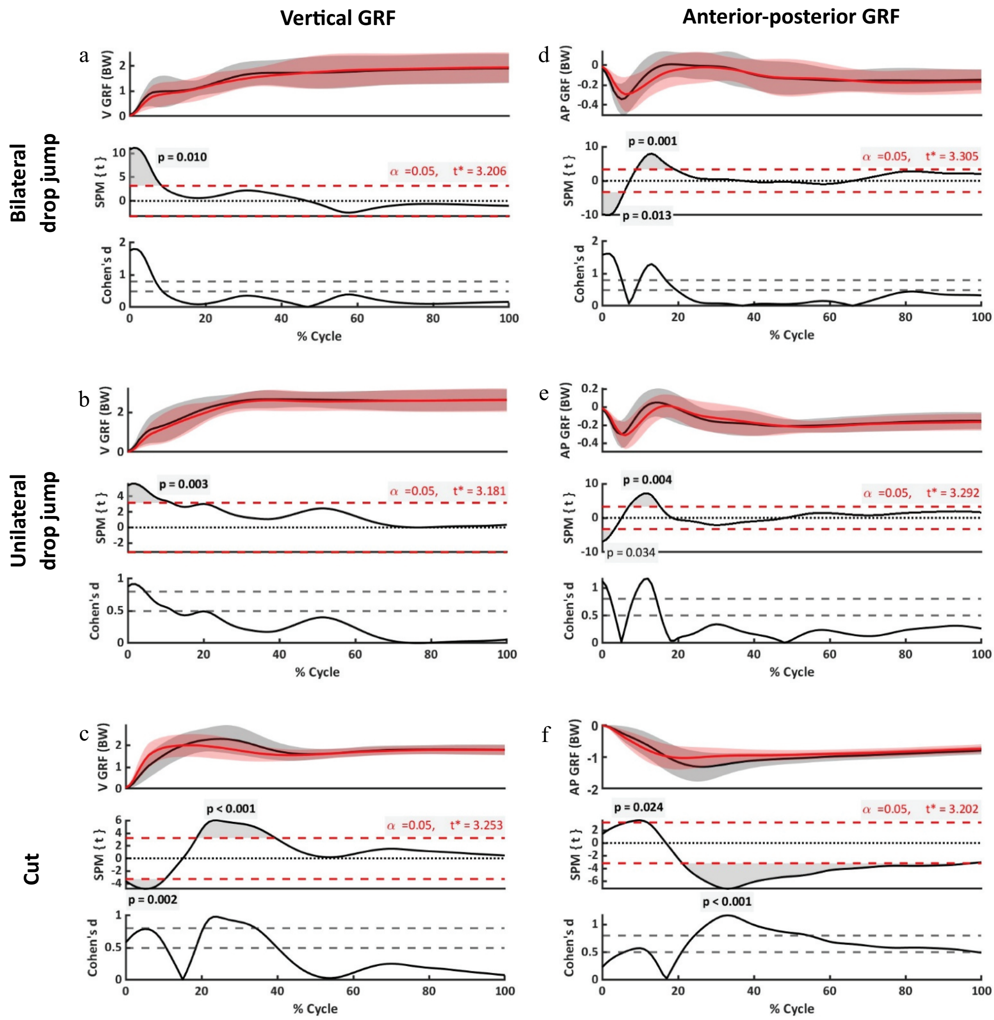

Figure 3. Differences in vertical and anterior-posterior GRF between surfaces for the bilateral drop jump (a and d, respectively), the unilateral drop jump (b and e, respectively), and the cut (c and f, respectively). The top panel of each graph represents the mean and SD clouds for the different surfaces with the black representing Mondo track and red representing artificial turf. The middle panel shows the SPM{t}– the *t*-statistic as a function of time describing the differences between the two surfaces, with the dashed red lines representing the critical threshold, and the shaded portion of the curve indicating $p < 0.05$. The bottom panel shows the ES as a function of time describing the magnitude of the effect, with the dashed grey and thick dashed grey lines representing the thresholds for Cohen's $d > 0.5$ and $d > 0.8$, respectively. Bold *p*-values indicate $p \leq 0.05$ and Cohen's $d > 0.5$ over a duration >5%. GRF = ground reaction force; SD = standard deviation; SPM = statistical parametric mapping; ES = effect size; V = vertical; AP = anterior–posterior.

Distal and proximal joints

Peak ankle eversion moments were higher during the unilateral drop jump on artificial turf compared to Mondo track ($t = -3.63$, $p = 0.001$, $d = -0.59$). In the cut, peak inversion moments were larger on Mondo track compared to artificial turf ($t = -5.13$, $p < 0.001$, $d = -0.53$). Additionally, during the cut, peak ankle inversion angles ($t = -4.02$, $p < 0.001$,

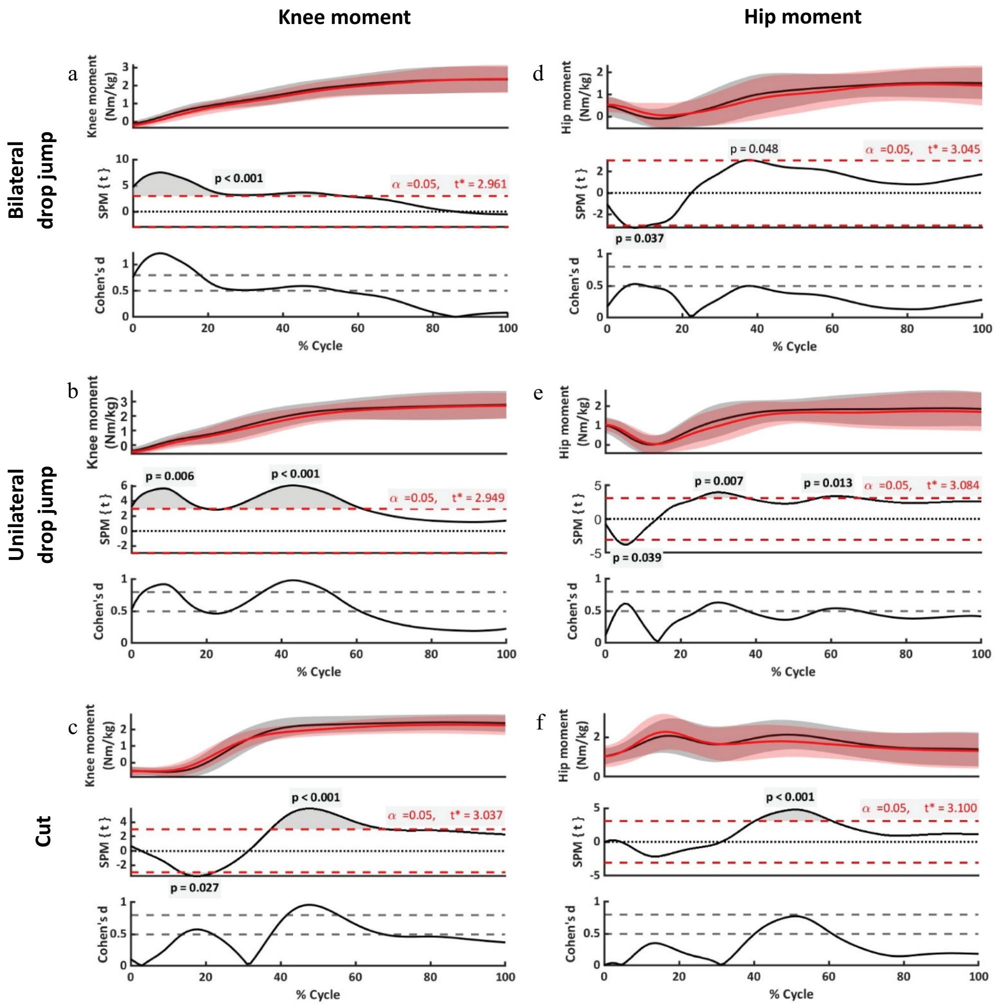

Figure 4. Differences in sagittal plane knee and hip moments between surfaces for the (a and d) bilateral drop jump, (b and e) unilateral drop jump, and (c and f) cut. The top panel of each graph represents the mean and SD clouds for the different surfaces with the black representing Mondo track and red representing artificial turf. The middle panel shows the SPM{t}– the *t*-statistic as a function of time describing the differences between the two surfaces, with the dashed red lines representing the critical threshold, and the shaded portion of the curve indicating $p < 0.05$. The bottom panel shows the ES as a function of time describing the magnitude of the effect, with the dashed grey and thick dashed grey lines representing the thresholds for Cohen's $d > 0.5$ and $d > 0.8$, respectively. Bold *p*-values indicate $p \leq 0.05$ and Cohen's $d > 0.5$ over a duration >5%. Positive values represent external flexion moments. SD = standard deviation; SPM = statistical parametric mapping; ES = effect size.

$d = 0.65$) and ankle RoM ($t = 4.65$, $p < 0.001$, $d = 0.75$) were greater on Mondo track compared to artificial turf. In the bilateral drop jump on artificial turf, there were higher magnitudes of hip work compared to on Mondo track ($t = 4.23$, $p < 0.001$, $d = 0.66$).

During the unilateral drop jump, hip adduction moments were higher on Mondo track compared to artificial turf (5–14%, $p = 0.017$, $d = 0.58$). In the unilateral drop jump, ankle eversion moments were greater between 0–11% ($p = 0.027$, $d = 0.65$) and 27–98%

($p < 0.001$, $d = 0.68$) of the braking phase on artificial turf compared to Mondo track. Towards the end of the braking phase during the cut, ankle inversion moments and angles were larger on Mondo track compared to artificial turf (35–100%, $p < 0.001$, $d = 0.78$ and 26–100%, $p < 0.001$, $d = 0.58$, respectively).

Discussion and implication

Changes in surface hardness are likely to alter an athlete's movement strategy. Therefore, movements included in injury risk assessments performed on a harder surface may misrepresent an athlete's ACL injury risk, compared to the surface that athletes typically use for training and/or matches. The aim of this study was to examine the influence of surface hardness on an athlete's movement strategies in movements performed in ACL injury risk assessments (bilateral and unilateral drop jumps, and a cutting manoeuvre). It was hypothesised that (1) movement strategies associated with ACL injury would be more pronounced on a harder surface (e.g. vertical and braking impact forces and knee flexion and abduction moments), (2) athletes would increase lower limb joint RoM and eccentric joint work during landing to attenuate these increased forces and moments and that (3) joint stiffness would be similar between surfaces due to relative increases in both joint moments and RoM.

In support of the first study hypothesis, knee flexion moments were larger on the harder Mondo track as opposed to the softer artificial turf during the bilateral (0–57%) and unilateral (0–20% and 25–62%) drop jumps. In addition, during the cut higher peak knee flexion moments were found on Mondo track, which aligned with larger knee flexion moments during the middle of the braking phase. However, contrary to the first hypothesis, in the early stages of the braking phase during the cut, higher knee flexion moments were observed on the softer artificial turf surface. This may be attributed to the larger knee flexion angles observed during the first half of the braking phase in the cut on artificial turf. A more flexed knee enables better GRF attenuation during landing (as seen in the cut on artificial turf with lower vertical and posterior GRFs between 19–39% and 21–97%, respectively). However, a flexed knee increases the distance of the GRF vector to the knee joint centre (Zavatsky et al., 1994), which contributes to an increased knee flexion moment. The higher knee flexion moments with increased knee flexion during landing may increase ACL strain. This is potentially due to rapid translational joint forces that propagate up the kinetic chain, as opposed to excessive quadriceps activation induced anterior tibial translation (Podraza & White, 2010). Increased external knee flexion moments may contribute to a heightened risk of ACL injury (Chappell et al., 2002; Yu & Garrett, 2007). Therefore, during the drop jumps, it would be inaccurate to use knee flexion moments to represent an athlete's ACL injury risk on a harder surface. Conversely, since higher knee flexion moments are reported between 14–22% of the braking phase on artificial turf compared to Mondo track during the cut, athletes who may be at increased risk of ACL injury may not be identified if performing the cut on harder surfaces compared to surfaces typically used in training or matches.

In partial support of the first hypothesis, during both drop jumps the vertical GRFs in the early stages of the braking phase were larger on Mondo track compared to artificial turf. During the cut, however, vertical GRFs were smaller immediately following initial contact before becoming larger later on during the braking phase on Mondo track. The lower vertical GRFs observed first following initial contact on the harder Mondo track is similar to previous findings, where lower vertical GRFs have been observed on a harder surface compared to a more cushioned surface (McNitt-Gray et al., 1994; Stiles & Dixon, 2006). Larger GRFs have been associated with higher peak ACL strain (Cerulli et al., 2003), which likely increases ACL injury risk (Decker et al., 2003; Hewett et al., 2005; Podraza & White, 2010; Yu et al., 2006). Grund et al. (2010) analysed real situations where ACL ruptures occurred in soccer players and found that vertical GRFs larger than 2.5 times body weight were associated with an injury risk for ACL. However, the magnitudes of vertical GRFs during the cut on artificial turf in the first 10% of the braking phase following ground contact did not exceed this value. Therefore, whilst increased vertical GRFs were observed during the first 10% of the braking phase on artificial turf in the cut, magnitudes during this phase are unlikely to increase the risk of ACL injury compared to testing on Mondo track. Furthermore, peak vertical GRFs (>2.5 BW) during the cut (which occurred after the initial 10% of the braking phase) were larger on Mondo track compared to artificial turf (<2.5 BW). Larger vertical GRFs during cutting manoeuvres have been associated with increased ACL strain (Weinhandl & O'Connor, 2017). In support of the second hypothesis, magnitudes of negative knee work were also greater on Mondo track. A negative relationship exists between peak vertical GRFs and negative knee work (Norcross et al., 2010). Considering peak vertical GRFs were higher on Mondo track during the cut, increasing negative knee work may be a strategy employed by athletes to help attenuate the larger peak vertical GRFs experienced on harder surfaces. Furthermore, although peak vertical GRFs were higher during the cut on Mondo track, instantaneous loading rates were lower. This may be due to the inability of a harder surface to undergo vertical deformation. Therefore, the impact forces must be attenuated over a longer time period, which would be beneficial to reduce ACL injury risk. Conversely, no differences in peak vertical GRFs were found between surfaces for either drop jump. On each surface however, peak vertical GRFs during the unilateral drop jump exceeded a minimum value of 2.5 BW, which has been reported when an ACL has ruptured (Grund et al., 2010). Thus, whilst differences were observed in vertical GRFs in phases of the braking phase between surfaces for all movements, only magnitudes of peak vertical GRFs that may contribute to ACL injury were displayed on both surfaces during the unilateral drop jump and on Mondo track in the cut. This suggests that peak vertical GRFs may be used to more accurately identify ACL injury risk on either surface during the drop jumps, but on Mondo track during the cut, using peak vertical GRF may misrepresent athletes risk of ACL injury.

Braking forces were higher in the early stages (within the first 20%) of the braking phase on artificial turf for all movements, which rejects our first hypothesis. Larger braking forces require an increase in net internal knee extensor moment that is generated by the quadricep muscles to counteract the force. This could contribute to higher proximal tibial anterior shear forces and subsequent increased ACL loading (Yu & Garrett, 2007). Magnitudes of posterior GRFs between −2.6 and −0.5 BW have been observed when the ACL has ruptured (Grund et al., 2010). Our study shows that posterior GRFs did not exceed the minimum −0.5 BW associated with ACL rupture on either surface for the drop jumps (between −0.45 and −0.39 BW), which compares well to previous research of drop-jump landings on artificial turf (−0.402 ± 0.097 BW (Claudino et al., 2017)). Thus, it is unlikely these values are large enough in magnitude to influence

ACL injury risk on either of the surfaces included in this study. Conversely, during the cut posterior GRFs for the majority of the braking phase (after 21%) and peak braking forces were larger on Mondo track. Unlike the drop jumps, these magnitudes exceeded the minimum posterior GRF magnitude indicative of ACL rupture on both surfaces. This is likely due to task differences, as during the cut athletes started 5 m from the force plates and were required to run as fast as possible before planting their contralateral foot on the force plate and cutting in the desired direction. The running motion of the cut results in athletes having a much higher horizontal velocity than in the drop jumps (predominantly vertical motion) at initial contact. Subsequently, much larger horizontal decelerations are required to reduce the horizontal velocity of the athlete prior to changing direction in the cut. Although there are slight differences in posterior GRFs between surfaces, the magnitudes displayed on either Mondo track or artificial turf during all movements suggests that it may be possible to accurately identify athletes risk of ACL injury regardless of surface.

In a prospective study by Hewett et al. (2005), peak hip flexion moments were higher in athletes who went on to sustain an ACL injury compared to those that did not when performing a vertical drop jump (~2.4 Nm/kg and ~1.8 Nm/kg, respectively). It could therefore be postulated that hip flexion moments larger than 2.4 Nm/kg in a vertical drop jump may place an athlete at an increased risk of ACL injury. During the earliest stages of the braking phase, hip flexion moments were larger on artificial turf in both the drop jumps, whilst larger hip flexion moments were observed later in the braking phase during the unilateral drop jump on Mondo track. The magnitude of hip flexion moment during these phases however, did not exceed 2.4 Nm/kg, which implies that differences between surfaces in these phases of the drop jumps are unlikely to influence ACL injury risk. Peak hip flexion moments were larger on Mondo track during both drop jumps but only magnitudes of peak hip flexion moments in the unilateral drop jump on Mondo track were sufficiently high to indicate increased risk of ACL injury. This suggests that during the unilateral drop jump, the use of hip flexion moments likely misrepresents an athletes ACL injury risk when performed on a harder surface. However, it should be noted that other researchers have found external hip flexion moments not to be associated with ACL injury risk (Leppänen et al., 2017). Therefore, hip flexion moments as an outcome measure for ACL injury should be used with caution.

Frontal plane loading, such as knee abduction moments above 0.4 Nm/kg, can influence ACL injury risk (B. Boden et al., 2000; Hewett et al., 2005; Krosshaug et al., 2016; Myer et al., 2015; Tran et al., 2016; Weiss & Whatman, 2015). During jump landing and cutting manoeuvres, larger knee abduction moments place greater tensile stress on the ACL (N. A. Bates, Schilaty, Krych, et al., 2019; N. A. Bates, Schilaty, Nagelli, et al., 2019; Garrett & Yu, 2007; Weiss & Whatman, 2015). No differences were observed in peak knee abduction moment between surfaces for any movement. In the unilateral drop jump, higher knee abduction moments were displayed in the continuous data on artificial turf compared to Mondo track, which rejects the first hypothesis. However, in both the unilateral drop jump and cut the magnitudes of the peak knee abduction moment exceeded 0.4 Nm/kg on both surfaces. Furthermore, landing with increased hip adduction moments have also been associated with greater risk of ACL injury (B. P. Boden et al., 2009; Tran et al., 2016). Larger hip adduction moments may place an increased valgus stress over the knee (Hewett et al., 2005). During the unilateral drop jump however, larger knee abduction moments and smaller hip adduction moments were observed on artificial turf. Since magnitudes of knee abduction moments were greater than 0.4 Nm/kg on both surfaces during the unilateral drop jump and cut, this variable could be used to correctly identify athletes who may be at increased risk of ACL injury on either surface.

Previous research has suggested that athletes increase the RoM of all the lower limb joints when landing on harder surfaces to attenuate the forces exerted on the body during landing (Dixon et al., 2005; Gerritsen et al., 1995). Increased sagittal plane RoM helps dissipate the kinetic energy of landing via eccentric muscle action (Coventry et al., 2006), resulting in increased energy absorption (i.e. negative work). During the unilateral drop jump, no differences were found in ankle RoM or negative ankle work between surfaces, yet athletes displayed smaller ankle plantarflexion angles in the early stages of the braking phase on Mondo track. The plantarflexor muscles have been proposed to be less effective at attenuating GRFs when landing with reduced ankle plantarflexion (Shimokochi et al., 2013). Furthermore, for the majority of the braking phase hip flexion angles were larger on Mondo track in the unilateral drop jump. Whilst larger vertical GRFs were observed during the first 12% of the braking phase on Mondo track, there was no difference in peak vertical GRFs between surfaces in the unilateral drop jump. This may suggest that increasing proximal (hip) joint flexion helped attenuate the larger vertical GRFs produced in the initial stages of the braking phase on a harder surface in a unilateral drop jump. In contrast, there were no changes in distal or proximal joint flexion during the bilateral drop jump and cut between surfaces. During the cut however, knee flexion angles were higher on Mondo track during most of the braking phase. Furthermore, on both surfaces during the cut and unilateral drop jump knee flexion angles were between ~ 20–40°. At smaller knee flexion angles (<50°), large quadriceps muscles contractions have been found to induce anterior tibial shear force (Shelburne et al., 2005) and anterior tibial translation (DeMorat et al., 2004; Victor et al., 2010), which have both been proposed to strain the ACL. Lower knee flexion reduces the flexibility and adaptability in the knee joint, subsequently increasing the force applied to the ligaments by the muscles (Benjaminse et al., 2017) and increasing ACL injury risk. These findings partially support the second hypothesis, as during the cut and unilateral drop jump athletes increased their lower limb joint RoM on Mondo track, potentially to help attenuate the higher forces and moments exerted on the body on a harder surface.

Joint stiffness is characterised by the interaction between joint moments and RoM (Baltich et al., 2015). We hypothesised that joint stiffness would be similar between surfaces due to relative increases in both joint moments and RoM. In support of our third hypothesis, sagittal plane ankle, knee and hip stiffness was similar between surfaces for both the drop jumps. Conversely, during the cut, whilst no differences were found between surfaces for knee and hip stiffness, ankle stiffness was higher on Mondo track. Although not significant, there was a trend towards larger changes in ankle moment and reduced ankle RoM on Mondo track in the cut, which resulted in the observed increase in ankle stiffness. Restricted sagittal plane ankle RoM, especially, has been associated with increased frontal plane knee motion (Lima et al., 2018; Wyndow et al., 2016). Furthermore, a strong positive relationship has been found between frontal plane knee motion and knee abduction moments (Myer et al., 2010), which have been proposed to increase ACL strain (P. Boden et al., 2010; Ford et al., 2003; Hewett et al., 2005; Quatman

& Hewett, 2009). Greater ankle stiffness may therefore be associated with higher ACL injury risk due to poor frontal plane mechanics (e.g. increased knee abduction angles and moments). However, during the cut no differences were observed in knee abduction angles or moments, suggesting further work is needed to understand the effect of changes in sagittal plane ankle stiffness on ACL injury risk between surfaces.

From a practitioner's perspective, biomechanical deficits associated with ACL injury were predominantly more pronounced on the harder surface (e.g. magnitudes indicative of ACL injury were displayed on the harder Mondo track but not on the softer more cushioned artificial turf). Consequently, if testing on a harder surface compared to the surface athletes train and play matches on, practitioners may incorrectly identify and target deficits which are not true biomechanical deficits according to the movement strategies employed on surfaces more relevant to training and matches. However, while some biomechanical variables were more pronounced on one surface than the other, magnitudes indicative of ACL injury risk were displayed on both surfaces. Thus, using these variables, athletes displaying biomechanical deficits associated with ACL injury would be correctly identified regardless of the testing surface. In contrast, during the cut, knee flexion moments that may contribute to ACL injury were lower on Mondo track. Therefore, if using knee flexion moments as an outcome measure of ACL injury risk, performing the cut on a harder surface to that of which athletes typically use in training and/or matches may misrepresent an athlete's ACL injury risk.

A limitation of this study was that footwear were not standardised between surfaces or participants. However, wearing cleats on artificial turf and standard running shoes on Mondo track provides greater ecological validity as opposed to wearing one type of footwear on both surfaces. Cleats have been reported to have higher translational and rotational traction compared to a standard running shoe (Stefanyshyn et al., 2010). Whilst increased traction is beneficial for performance during cutting manoeuvres, it has also been found to alter an athlete's movement strategy, which may influence ACL injury risk. For example, larger braking forces and frontal plane knee moments have been observed during a cutting manoeuvre using cleats compared to no cleats (Schrier et al., 2014; Wannop et al., 2019), which may increase an athlete's ACL injury risk. However, in this study, when performing movements included in ACL injury risk assessments, the magnitudes of biomechanical variables that may be associated with ACL injury risk were displayed on both Mondo track and on artificial turf. This may suggest that regardless of the shoe used, it is possible to correctly identify athletes displaying movement strategies associated with ACL injury risk in biomechanical assessments performed on harder surfaces – as long as the shoe is appropriate to the surface. Additionally, standardising footwear between participants may improve internal validity by preventing differences in footwear introducing variance in the biomechanical variables. However, standardising footwear reduces ecological validity and may also promote unnatural movement strategies due to lack of long term familiarity in new footwear, subsequently limiting the relevance of the findings to realistic applications (Hunter et al., 2020). Considering multidirectional field sport teams do not standardise footwear, it was deemed appropriate to assess athletes in the footwear they typically wear during sport.

Athletic track has ~30% reduced peak impact force and FIFA standard artificial turf has ~ 60% reduced peak impact force compared to a surface with negligible cushioning, such as concrete (Colino et al., 2017, 2020). Biomechanical assessments in a laboratory environment are typically performed on a concrete surface which is considered to have negligible cushioning. However, in applied field-based settings ACL injury risk assessments may be performed on a rubber gym surface. Therefore, Mondo track (e.g. a rubber-based surface) provides greater ecological validity with the type of test surface as opposed to a concrete laboratory surface. Infill on artificial turf also acts as another layer to further improve the shock absorption abilities of the surface (Alcántara et al., 2009; McGhie & Ettema, 2013). In this current study, however, 3G artificial turf without infill was used. This could further influence the impact forces and an athlete's movement strategies during landing. However, surface hardness is affected by the compaction (density) of the rubber infill following repeated impacts (Fleming et al., 2015). Testing on artificial turf with no infill was therefore used to overcome this. Future studies could consider the influence of artificial turf with different infill properties on an athlete's movement strategy when performing movements in assessments of ACL injury risk. Additionally, only males were recruited for the study, meaning findings may not be generalisable to female populations. Although female athletes are at increased relative risk of non-contact ACL injuries, males account for the highest absolute number of ACL injuries in multidirectional field sports due to the greater overall number of male athletes (Eime et al., 2021; Montalvo et al., 2019). Future research could carry out similar analyses in female athletic populations to examine the influence of surface hardness on female athletes movement strategies.

Conclusion

Continuous and discrete data analyses revealed alterations in movement strategies between surfaces of different hardness when performing movements used to assess risks of ACL injury. This highlights the importance of including both types of analyses in assessments of ACL injury risk. Although clear differences were found between surfaces, magnitudes of biomechanical variables that may place an athlete at an increased risk of ACL were mostly observed on Mondo track but not artificial turf. A key takeaway from this study is that injury risk assessments performed on a harder surface (e.g. Mondo track) may misrepresent an athlete's risk of ACL injury, compared to the same movements performed on a softer more cushioned surface that is typically used for training and/or matches (e.g. artificial turf). Nevertheless, some biomechanical variables suggested to contribute to ACL injury (e.g. knee flexion moments) may not be identified during a cutting manoeuvre on harder surfaces. Performing the cut on more cushioned surfaces, closer to those that are used in the field, is more likely to help identify those athletes who could be at greater risk of sustaining an ACL injury.

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