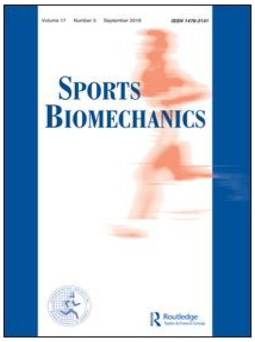

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Lack of impact moderating movement adaptation when soccer players perform game specific tasks on a third-generation artificial surface without a cushioning underlay

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ABSTRACT

The objective of this study was to investigate how the inclusion of a cushioning underlay in a third-generation artificial turf (3G) affects player biomechanics during soccer-specific tasks. Twelve soccer players (9 males/3 females; 22.6 ± 2.3 y) participated in this study. Mechanical impact testing of each 3G surface; without (3G-NCU) and with cushioning underlay (3G-CU) were conducted. Impact force characteristics, joint kinematics and joint kinetics variables were calculated on each surface condition during a sprint 90° cut (90CUT), a sprint 180° cut (180CUT), a drop jump (DROP) and a sprint with quick deceleration (STOP). For all tasks, greater peak resultant force, peak knee extensor moment and peak ankle dorsi-flexion moment were found in 3G-NCU than 3G-CU ($p < 0.05$). During 90CUT and STOP, loading rates were higher in 3G-NCU than 3G-CU ($p < 0.05$). During 180CUT, higher hip, knee and ankle ranges of motion were found in 3G-NCU ($p < 0.05$). These findings showed that the inclusion of cushioning underlay in 3G reduces impact loading forces and lower limb joint loading in soccer players across game-specific tasks. Overall, players were not attempting to reduce higher lower limb impact loading associated with a lack of surface cushioning underlay.

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3G Surface; force measurements; kinematics; kinetic; soccer

Introduction

Soccer is typically played on a natural grass surface, although this type of surface is limited by expensive maintenance costs during seasonal climate changes. Artificial turf was introduced to reduce these maintenance concerns and to help keep participation levels high

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Supplementary data can be accessed here

all year round (Fédération Internationale de Football Association, 2015). The evolution of the first-developed artificial turf to the present day has seen some dramatic advancements with the inclusion of synthetic fibres with rubber particle and/or sand in-fill and sometimes the addition of shock attenuating underlayers that have all helped to emulate natural grass surfaces (Fleming, 2011). Third-generation artificial turfs (3G), introduced in the 1990s, were aimed to simulate natural turf characteristics permitting, for instance, the use of studded soccer boots. The 3G systems are composed of monofilament or fibrillated long fibres (approximately a length of 40–65 mm) with a moderately low tuft density. Then, surfaces are 'filled' with a sand layer at the base and rubber or elastomeric particles (e.g., recycled tyre) to help with player performance and comfort levels (Fleming, 2011). Moreover, a cushioning underlay may be included if the 3G system does not provide enough shock absorption (Fleming, 2011).

Previous studies have provided evidence that the mechanical characteristics of 3G systems and natural turf can be very closely matched (Livesay, Reda, & Nauman, 2006; Martinez, Dura, Gamez, Zamora, & Alcantara, 2004; Villwock, Meyer, Powell, Fouty, & Haut, 2009); however, factors such as maintenance, its rate of use and its configuration are found to be important in modifying the mechanical characteristics of 3G systems (Burillo, Gallardo, Felipe, & Gallardo, 2012). Importantly, when different types of 3G systems and their constitute components are compared, diverse mechanical characteristics have been found (Burillo et al., 2012; Sánchez-Sánchez, Felipe, Burillo, Del Corral, & Gallardo, 2014; Sánchez-Sánchez et al., 2017). For example, one study showed that the inclusion of cushioning underlay in 3G systems increased mechanical shock absorption compared to a no underlay condition (Burillo et al., 2012). Consequently, these 3G systems with cushioning underlay would help to meet the Fédération Internationale de Football Association (FIFA) mechanical requirements (60–70% shock absorption) (Fédération Internationale de Football Association, 2015) and ensure their safety. In another study, Sánchez-Sánchez et al. (2017) also reported that cushioning underlays in 3G systems increased shock absorption. In addition to these mechanical differences between surfaces conditions, biomechanical data are required to fully understand the effects that each surface condition has on the loads experienced by the body during a game of football and to highlight potential for differences in injury risk between surface.

Due to the important implications of 3G surface type and associated mechanical characteristics on injury risk and performance, some investigators have begun to examine the effects of surfaces with different cushioning conditions (i.e., the inclusion of a shock pad or different amount of infill) on biomechanics of players during dynamic game-like activities (Low & Dixon, 2016; McGhie & Ettema, 2013; Meijer, Dethmers, Savelberg, Willems, & Wijers, 2006). For instance, Low and Dixon (2016) investigated the influence of different cushioning underlay densities on heel impact during running and turning movements. Their results showed that low shock-pad densities (more cushioning) reduced forces acting on the heel. In another study by McGhie and Ettema (2013), the 3G system that had the thinnest layer of infill and no shock-pad demonstrated higher peak impact forces and lower time of contact than two different 3G conditions (one with shock-pad and another one without shock-pad and with a thicker layer of infill).

Although, previous findings suggest that the severity of impacts experienced by soccer players are greater on artificial surfaces without a cushioning underlay some studies reported no ground reaction force (GRF) reductions with softer insoles (Nigg, Cole, & Bruggemann, 1995), higher loading rates in harder surfaces (Stiles, Guisasola, James, & Dixon, 2011) and, even, no correlations between the stiffness of the surface and peak impact force (Dixon, Batt, & Collop, 1999; Nigg, 1990). In addition, Dixon, Collop, and Batt (2000) observed kinematic modifications between surfaces of different stiffness during running suggesting that surface characteristics might encourage joint mechanical adaptations of the players. In contrast, Karamanidis, Arampatzis, and Bruggemann (2006) reported that differences in surface condition did not affect joint kinetics. Based on the previous evidence, there still appears to be ambiguity on how the player's lower limb joint kinetics and kinematics adapt during different dynamic, game-like movement tasks. For example, do players modify their movement strategies to keep lower limb loading levels to similar levels despite interacting with a surface with less cushioning properties (i.e., with no cushioning underlay)? Therefore, impact force characteristics and joint kinematic and kinetic adaptations as a whole might provide a greater insight into how the body might adapt biomechanically to reduced surface cushioning. In addition, with the association between greater impact forces and loading rates with overuse injury risk in athletes (Ferber, McClay-Davis, Hamill, Pollard, & McKeown, 2002; Hreljac, Marshall, & Hume, 2000), there is requirement for a further understanding how players adapt to different surfaces during different game-like movement tasks.

The objective of this study was to investigate the influence of a cushioning underlay on soccer player biomechanics (impact force, joint kinetics and movement characteristics) across a range of game-specific tasks. We hypothesised that greater peak resultant force, impulse during impact phase and resultant loading rate will be observed in a 3G surface without cushioning underlay (3G-NCU) than 3G with cushioning underlay (3G-CU) during all tasks. Therefore, players will increase mainly their ankle and knee ranges of motion (ROM) and moments to cope with the higher impact forces and keep them down to a similar manageable level (similar to when the cushioning layer is present).

Methods

Participants

Fifteen injury-free soccer players (10 males and 5 females) were recruited for this study. The inclusion criteria were that they had to be playing soccer for a period of at least a year before data collection and did not have any serious or major injuries to their lower limbs within the 6 months before testing. Before data collection, all participants completed and signed an informed consent form approved by the Liverpool John Moores University Research Ethics Committee. Due to the quality of their data, three participants were excluded from the study. Therefore, a total of 12 players were included (9 males and 3 females; mean age 22.6 ± 2.3 years; height 174.9 ± 0.1 cm; 69.4 ± 13.4 kg) in the study analysis.

3G systems

A 3G system (type: monofilament; material: polypropylene, height: 40 mm, weight: 2 kg/m^2; Direct Artificial Grass, United Kingdom) with styrene-butadiene rubber and

Figure 1. Experimental setup. Two 0.9 × 0.6 m force platforms (white rectangles) are situated in the centre of the three-dimensional recording volume. Dashed line represents a sprint 90° cut (90CUT), solid line represents a sprint 180° cut (180CUT) and dashed-dotted line represents a sprint with quick deceleration (STOP). 90CUT, 180CUT and STOP tasks were performed on force platform 1. For the STOP task participants started to decelerate on force platform 1.

quartz sand infill (infill characteristics were installed following the manufacturer guidelines) was used in this study. Two artificial grass surface conditions were examined with and without the inclusion of a cushioning underlay (type: felt; thickness: 11 mm; weight: 1.42 kg/m^2): 3G-NCU and 3G-CU. The cushioning underlay was placed under the entire 3G system (force platforms and its surroundings). Both 3G-NCU and 3G-CU systems were placed over and around two force platforms (90 × 60 cm, 9281B, Kistler Holding AG, Winterthur, Switzerland) and in a calibrated volume of a 10-camera opto-electronic motion capture system (Oqus 400, Qualisys AB, Gothenburg, Sweden) (see Figure 1).

Mechanical properties of the 3G systems

Peak impact acceleration (g) of each 3G surface condition was measured using a standard impact test proposed by the American Society for Testing and Materials (American Society for Testing and Materials, 2013). It has been shown that this standard test in a tennis shoe-surface combination supplied an accurate prediction of the impact attenuate ability (Dixon & Stiles, 2003). In brief, this procedure involved 10 preconditioning impact trials, followed by 20 impact trials on each surface system. Each trial consisted of dropping a weighted shaft (8 kg) 8 cm onto the 3G surface, with the impacting head of the shaft resembling the area of a human heel. For simulating running impacts, the impact velocity is typically around 1.0 m/s (Lafortune & Lake, 1995) but, to more closely resemble highly dynamic tasks in soccer, an impact velocity of 1.25 m/s was used in this study. This impact testing showed 28% greater peak acceleration in the 3G-NCU (17.0 ± 0.5 g) versus the 3G-CU surface condition (13.3 ± 0.3 g). Therefore, from a mechanical perspective, the addition of a cushioning underlay reduced impact severity by 28%.

Protocol

All participants wore standardised soccer shoes (Li-Ning Glory, Hong Kong, China), with a hard-ground stud design (30 studs) suitable for use on artificial surfaces. Data collections on the two types of 3G conditions were conducted on separate days (within a 2-week period) to reduce to effects of fatigue. Before testing, all participants performed a warm-up routine that consisted of 3 min of ankle, knee and hip articular mobilisation, 5 min of low intensity cycling (Wattbike cycle ergometer, Wattbike Ltd, Nottingham, United Kingdom) and five sub-maximal familiarisation trials of each task included in the experimental testing protocol. On each 3G condition, participants performed four different tasks; a sprint 90° cut (90CUT), a sprint 180° cut (180CUT), a drop jump (DROP) and a sprint followed by a quick deceleration (STOP). For each cutting task, participants were instructed to maximally sprint (4.5 m) and change direction when landing on the force platform (see Figure 1). Cutting and the stop tasks were performed after a 4.5 m approach runway. On average, seven acceptable trials of each task per participant-surface condition were used in this study. Trials were discarded if participants did not land with their foot fully on the force platform or they did not cut to the required cutting angle outlined by gate posts. All participants performed the 90CUT, 180CUT and STOP with their dominant leg. The dominant leg was determined by players preferred kicking leg. For the STOP task, participants had to start to decelerate with their dominant leg on the first force platform and they had to complete the stop within the next two steps. For the DROP task, participants dropped off a 0.3 m box and landed with their feet in the middle of each force platform (mounted side by side (Figure 1)) before jumping vertically as high as they could. The initial landing phase of the DROP was analysed for the dominant leg only.

Biomechanical analyses

Forty-three retro-reflective, spherical markers (12 mm diameter) were placed on participant's seventh cervical vertebrae, manubrium, xiphoid process, left and right acromion, left and right posterior superior iliac spine, left and right superior iliac crest, left and right anterior superior iliac spine, lateral and medial femoral knee epicondyles, lateral and medial ankle malleoli, proximal and distal posterior heel, lateral heel and the fifth metatarsal head. Four marker cluster sets were mounted on rigid, lightweight thermoplastic plates and attached on the lateral aspect of participants shanks and thighs. In addition, virtual landmarks were created on the first metatarsal head using a digitised pointer (C-Motion, Inc., Germantown, Maryland, USA). Motion data were collected at 500 Hz using a 10-camera motion capture system. In synchronisation with the motion capture system, two force platforms capturing at 3000 Hz was embedded in the floor and situated in the centre of the calibrated volume (see Figure 1).

Data analysis

All kinematic and kinetic data were processed and analysed in Visual 3D (C-Motion, Inc., Germantown, Maryland, USA). A 6-degrees-of-freedom eight segment model including feet, upper and lower legs, pelvis and trunk was constructed for each participant. The local segment coordinate systems of the pelvis, thigh, leg and foot were derived from a standing calibration trial. Hip joint centres were defined based on Bell and colleagues equation (Bell, Pedersen, & Brand, 1990). A cardan rotation sequence of x (flexion/extension), y (abduction/adduction) and z (axial rotation) was used to calculate joint angles for the hip, knee and ankle (Cole, Nigg, Ronsky, & Yeadon, 1993). Lower limb joint kinematics and kinetics were positive for flexion/dorsi-flexion, abduction/eversion and internal rotation and extension/plantarflexion, adduction/inversion and external rotation were negative. All angles were normalised to an anatomical standing posture with feet comfortably apart, legs fully extended and pelvis and torso in a neutral position. Joint angles were referenced to coordinate systems embedded in the distal segment. Lower extremity 3D joint internal moments were calculated using a Newton–Euler inverse dynamics approach within the Visual 3D software. Internal joint moments at the ankle, knee and hip were calculated and reported in the coordinate system of the leg segment and were normalised to participant's body mass. The segmental data was based on Dempster's data (1955) and using geometrical volumes to represent each segment as cylinders or cones (Hanavan, 1964).

Initial foot contact and toe-off events were detected from the vertical GRF using an ascending and descending threshold of 15 N. Motion data were filtered using a fourth order Butterworth (BW) low-pass filter at 15 Hz, while the force data were filtered at 60 Hz. For lower extremity joint moments, we filtered both the force and motion data at the same cut-off frequency of 15 Hz using a fourth-order BW low-pass filter (Kristianslund, Krosshaug, & van den Bogert, 2012).

Instantaneous loading rates were determined from the resultant GRF for each cutting and stop task. The resultant GRF was calculated using the following equation:

$$Resultant\ GRF = \sqrt{x^2 + y^2 + z^2}$$

Where, x, y and z are the medial/lateral, anterior/posterior and vertical GRF components respectively. The derivative of the resultant GRF was used to calculate instantaneous loading rate. The instantaneous loading rate was calculated between 20% and 80% of initial contact to the first impact peak due to this period being the steepest part of the resultant GRF curve on landing (Milner, Ferber, Pollard, Hamill, & Davis, 2006). Impact peak and impulse of the resultant GRF was also determined. Impulse was calculated (trapezium rule) during the impact phase. Impact phase was determined from the period of initial contact to first initial impact peak. Given that transient impact force characteristics are a measure of landing severity and the association of greater impact forces and loading rates with overuse injury risk in athletes (Ferber et al., 2002;

Hreljac et al., 2000; Milner et al., 2006) the above impact force characteristics were deemed appropriate to be studied in this paper.

Sagittal plane hip, knee and ankle joint ROM were calculated from peak extension/plantar-flexion to peak flexion/dorsiflexion during the stance phase. Peak extension/plantar-flexion of the hip, knee and ankle was calculated during the first half of the stance. Peak lower extremity joint moments of the hip, knee and ankle were calculated in the sagittal and the frontal planes during the stance phase of each task. In addition, peak joint angular velocities were determined during the stance phase across each task. The reason for including lower extremity joint ROM, joint angular velocities and peak joint moment variables were due to them being important mechanisms for absorbing impact forces during the landing phase of ground contact (Clansey, Hanlon, Wallace, & Lake, 2012; Dufek & Bates, 1991; McNitt-Gray, 1993).

Following the methodology used in McGhie and Ettema (2013) study, approach velocity during 90CUT and STOP tasks was calculated (the first and the second pairs of photo cells were placed 1.5 and 0.5 m before the force platform, respectively).

Statistical analyses

The Statistical Package for the Social Sciences (SPSS) version 22.0 for Mac OS X (SPSS Inc., Chicago, IL, USA) was used for all statistical analyses. Data were presented as mean ± standard deviation. Kolmogorov–Smirnov tests were used to test normality in the distribution of the data for all outcome variables.

Paired *t* tests were performed to examine biomechanical differences between surface conditions using the mean result of all tasks performed by each participant. Effect size statistics using Cohen's *d* were calculated. Taking into account cut-off points established by Hopkins et al. (Hopkins, Marshall, Batterham, & Hanin, 2009), the effect size for Cohen's *d* can be trivial (0.0–0.2), small (0.2–0.6), moderate (0.6–1.2), large (1.2–2.0) or very large (>2.0). Paired *t* test results and effect sizes of all comparisons are shown in Supplementary Table 1. Statistical significance was set at $p < 0.05$.

Results

Approach velocities, stance times and impact force characteristics

No significant approach velocity differences were found between 3G conditions during 90CUT (3G-NCU: 3.8 m/s vs 3G-CU: 4.0 m/s) and STOP tasks (3G-NCU: 4.6 m/s vs 3G-CU: 4.9 m/s). Stance times during all tasks were greater on 3G-NCU compared to 3G-CU (Table 1; percentage changes ranged from 8.9% to 11.9%; *p* values ranged from 0.001 to 0.022; Cohen's *d* ranged from 0.37 to 0.95). Also, peak resultant impact force was significantly higher on 3G-NCU than 3G-CU during all tasks (Figure 2; percentage changes ranged from 37.4% to 59.8%; *p* values ranged from 0.010 to 0.017; Cohen's *d* ranged from 1.02 to 1.18). Higher impulse during impact phase of stance was found in the 3G-NCU compared to 3G-CU surface condition during 90CUT, DROP and STOP (Figure 2; percentage changes ranged from 30.1% to 58.3%; *p* values ranged from 0.007 to 0.031; Cohen's *d* ranged from 0.87 to 1.26). Instantaneous loading rates during 90CUT and STOP were also greater on 3G-NCU than 3G-CU surface condition

Table 1. Mean (SD) lower limb joint kinematics and kinetics for 90CUT, 180CUT, DROP and STOP across cushioning underlay conditions.

	90CUT		180CUT		DROP		STOP	
	3G-NCU	3G-CU	3G-NCU	3G-CU	3G-NCU	3G-CU	3G-NCU	3G-CU
Stance time (s)	0.34 (0.04)*	0.30 (0.04)	0.53 (0.10)*	0.48 (0.06)	0.53 (0.10)*	0.49 (0.09)	0.25 (0.05)*	0.23 (0.06)
Joint kinematics								
Hip ROM (deg)	9.6 (1.2)	9.7 (0.9)	31.0 (2.0)*	26.1 (1.7)	45.6 (3.6)	42.3 (3.9)	6.4 (0.8)	7.6 (0.9)
Knee ROM (deg)	40.4 (1.8)	37.8 (2.2)	46.8 (2.1)*	43.4 (1.9)	66.3 (3.0)	63.3 (4.3)	52.9 (3.6)	58.2 (4.3)
Ankle ROM (deg)	28.9 (1.2)	25.6 (2.3)	30.2 (2.6)*	22.7 (3.0)	44.7 (2.3)	43.7 (2.8)	30.4 (1.8)	28.4 (2.1)
Peak knee flexion velocity (deg·s)	605.3 (53.9)	587.0 (48.7)	311.7 (54.9)	306.0 (54.3)	618.5 (26.6)	599.7 (29.6)	237.6 (32.8)	216.0 (26.2)
Peak ankle dorsiflexion velocity (deg·s)	312.4 (30.2)	345.5 (59.4)	445.4 (71.1)*	371.8 (57.2)	702.8 (39.2)	720.1 (52.5)	411.7 (45.1)	424.3 (37.6)
Joint kinetics								
Peak hip extensor moment (Nm·kg)	−5.7 (0.9)	−4.0 (0.3)	−2.9 (0.5)	−2.9 (0.3)	−3.5 (0.4)*	−2.0 (0.1)	−8.3 (1.8)	−5.2 (0.4)
Peak knee extensor moment (Nm·kg)	−2.3 (0.4)*	−1.6 (0.2)	−2.4 (0.3)*	−1.2 (0.2)	−2.3 (0.3)*	−1.5 (0.1)	−2.1 (0.2)*	−1.3 (0.1)
Peak ankle plantarflexion moment (Nm·kg)	−3.4 (0.3)*	−2.4 (0.1)	−2.5 (0.2)*	−1.9 (0.1)	−2.7 (0.4)*	−1.8 (0.1)	−3.4 (0.4)*	−1.7 (0.1)
Peak hip abduction moment (Nm·kg)	6.0 (1.5)	2.9 (0.4)	6.1 (1.5)	3.1 (0.4)	1.2 (0.2)	1.0 (0.1)	0.7 (0.1)	0.5 (0.1)
Peak knee abduction moment (Nm·kg)	3.7 (0.9)*	1.7 (0.2)	4.0 (1.0)	1.9 (0.3)	0.5 (0.1)	0.5 (0.1)	0.3 (0.1)	0.3 (0.1)
Peak Ankle abduction moment (Nm·kg)	0.4 (0.1)	0.5 (0.1)	0.5 (0.2)	0.5 (0.1)	0.3 (0.1)*	0.1 (0.0)	0.7 (0.1)	0.6 (0.1)

Data are means (standard error). 90CUT: sprint 90° cut; 180CUT: sprint 180° cut; DROP: drop jump; STOP: sprint with quick deceleration; 3G-NCU: 3G turf system without cushioning underlay; 3G-NCU: 3G turf system with cushioning underlay; ROM: range of motion.
Significance: $^*p < 0.05$ between surface conditions.

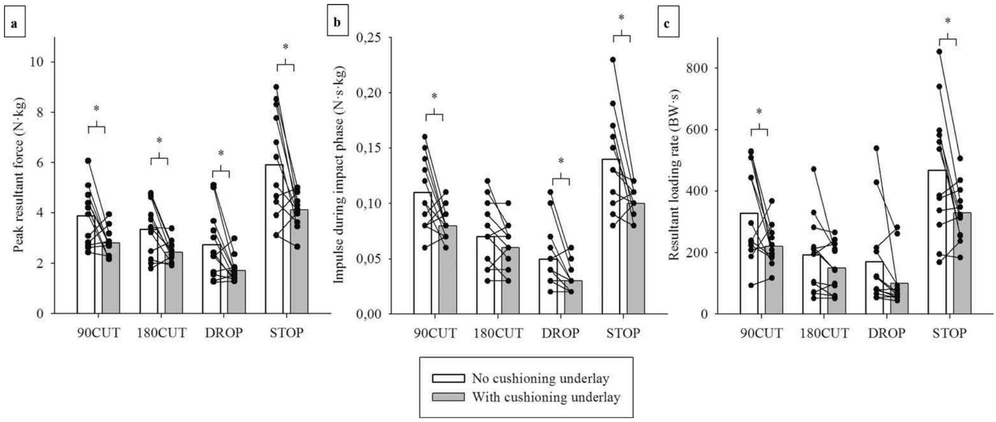

Figure 2. Comparisons of peak resultant force (a), impulse stance during impact phase (b) and resultant loading rate (c) between with and without cushioning underlay conditions in each task separately. Each point represents the mean of each participant in each task and surface condition. 90CUT: a sprint 90° cut; 180CUT: a sprint 180° cut; DROP: a drop jump (DROP); STOP: a sprint with quick deceleration.

*: Significant differences were set at $p < 0.05$.

(Figure 2; percentage changes were 48.3–42.5%; p values were 0.042 and 0.022; Cohen's d were 0.91 and 0.86, respectively). Moreover, STOP seems to be the most demanding task in terms of impact force characteristics because it showed the highest absolute magnitude of load and differences between 3G conditions in peak resultant force (1.79 N·kg), impulse during impact phase (0.04 N·s·kg) and resultant loading rate (Supplementary Table 1; 139.77 BW·s). Although the overall finding demonstrated that there were moderate increases in impact force variables on 3G-NCU condition, a large interparticipant variability in response to the different surface conditions was found (Figure 2).

Joint kinematics

During 180CUT task, hip, knee and ankle ROM and peak ankle dorsi-flexion velocity on 3G-NCU were significantly higher than on 3G-CU condition (Table 1; percentage changes ranged from 7.8% to 33.1%; p values ranged from 0.001 to 0.024; Cohen's d ranged from 0.49 to 0.76). There were no significant differences in joint kinematics variables between surface conditions during 90CUT, DROP and STOP (Table 1; p values ranged from 0.069 to 0.929; Cohen's d ranged from 0.03 to 0.76).

Joint kinetics

Peak knee and ankle extensor moments were significantly higher in all tasks for 3G-NCU compared to 3G-CU turf condition (Table 1; percentage changes ranged from 32.4% to 104.1%; p values ranged from 0.001 to 0.035; Cohen's d ranged from 0.70 to

1.68). During DROP task, peak hip extensor and peak ankle abduction moments were significantly greater for 3G-NCU surface condition compared to the 3G-CU (Table 1; percentage changes were 77.5–155.1%; *p*-value were 0.004 and 0.042; Cohen's *d* were 1.50 and 0.83, respectively). Additionally, higher peak knee abduction moment during 90CUT was found on 3G-NCU compared to 3G-CU (Table 1; percentage change was 119.6%; p = 0.039; Cohen's *d* 0.91).

Discussion and implications

The main finding of this study is that the inclusion of cushioning underlay increased cushioning of 3G-CU and, consequently, impact force characteristics (mainly peak resultant force and impulse during stance phase), joint kinematics (mainly knee and ankle ROM) and kinetics data (mainly peak knee and ankle flexion moment) during turning, jumping or stopping were reduced in comparison with those performed in 3G-NCU. Despite these greater impact force values observed in 3G-NCU compared to 3G-CU during all tasks, a large interparticipant variability has been found between different surface conditions. Overall, the general hypothesis that cushioning underlays reduce impact force characteristics and kinetics data is confirmed. On the other hand, the hypothesis that players adapt their movement patterns to cope with higher impact forces is partially supported because of the lack of kinematics differences in DROP and STOP tasks. These greater impact forces found in 3G-NCU compared to 3G-CU could be associated with the incidence of injury. In fact, Woods, Hawkins, Hulse, and Hodson (2002) compared the number of soccer injuries during preseason and season and found a higher number of injuries during preseason and 70% of these injuries occurred when players played predominantly on a dry and hard surface. Although it has been demonstrated that runners who have experienced a stress fracture show no impact force differences compared to those without stress fractures (Zadpoor & Nikooyan, 2011), a study performed by Hreljac (2004) reported that high impact forces may be an important factor to have a greater risk of developing an injury during running. Therefore, based on the previous literature, and our present findings of greater impact forces across all tasks when playing without a cushioning underlay, it may be that playing soccer on hard surfaces regularly (e.g., without a cushioning underlay), may put players at an increased risk of developing an overuse impact related injury.

Different from previous published studies that did not show the greatest impact force with the hardest surface (Stiles & Dixon, 2006; Stiles et al., 2011), impact force data of this study support the obtained mechanical results (the highest impact forces were found on the surface with less shock absorption). Due to these confusing results obtained by these authors between mechanical data of the surfaces and impact forces received by the players, they suggested the assessment of dynamic stiffness as well as the measures of whole leg and/or joint torsional stiffness to understand the mechanical behaviour of the surface (Dixon & Stiles, 2003).

The rate of increase in the impact force is considered a more reliable and better indicator of the impact severity than the peak impact force (Stiles & Dixon, 2007). Although resultant loading rates were only higher on 3G-NCU than 3G-CU during 90CUT and STOP, greater but not significant resultant loading rate were found between these 3G conditions during the other tasks. The lack of differences found may be explained by participants increasing their lower limb joint ROM at the hip, knee and ankle to cushion the impact during the 180CUT and DROP tasks. Therefore, these adaptations could be considered as an attempt to moderate the impact severity.

In terms of joint kinematics, the present study demonstrated greater hip, knee and ankle joint ROM, and peak ankle dorsi-flexion angular velocity on the less cushioned surface (3G-NCU) during 180CUT. In support of our results, previous studies showed joint modifications (e.g., increments of knee ROM) to reduce peak impact forces in less cushioned surface (Dixon, Collop, & Batt, 2005; Gerritsen, van den Bogert, & Nigg, 1995). Importantly, knee ROM increased in less cushioned surfaces because of its important role to absorb and dissipate the load (Gerritsen et al., 1995). In fact, the increase in knee ROM during the less cushioned surface loads the knee extensors (Damm et al., 2013) and may increase the risk of suffering from patellofemoral pain syndrome (Gecha & Torg, 1988). To reinforce these observations, our present study demonstrated that both knee ROM and extensor moment were higher in less cushioned surfaces. On the other hand, although not significant, the hip, knee and ankle during most of the other soccer tasks showed greater ROM on 3G-NCU compared to 3G-CU. Overall, players adapted their lower extremity joint to cushion ground impacts during 180CUT; nevertheless, these lower extremity joint adjustments were not found during 90CUT, DROP and STOP. Therefore, it seems that during 180CUT, players conscious attempt to land differently and/or to moderate the impact received.

Turning tasks such as 90CUT and 180CUT are highly influenced by the traction between the footwear and surface. Damm et al. (2013) reported that the higher knee flexion results obtained in the less cushioned surface are mainly explained by the traction instead of the cushioning of the surface. In contrast, this study found greater knee ROM in 3G-NCU compared to 3G-CU with the same level of traction (same footwear in both surfaces conditions). Thus, surface cushioning as well as its traction with the footwear should be considered due to their demonstrated influence in biomechanical parameters and the possible occurrence of musculoskeletal injuries.

In addition to the mentioned impact and kinematic findings, greater peak knee flexion and peak ankle dorsi-flexion moments were found in 3G-NCU than 3G-CU during all tasks. It has been demonstrated that high peak knee extensor moment during vertical drop jump landings was associated with the increase of the risk of anterior cruciate ligament injury (Leppänen et al., 2017). In this study, knee extensor moment was clearly reduced after the inclusion of the cushioning underlay and, therefore, the risk of anterior cruciate ligament injuries might be reduced. Moreover, greater peak ankle dorsi-flexion moments were found in less cushioned surface condition during all tasks performed. Thus, soccer practice in hard surfaces increases ankle moments and therefore, the loading of this joint, triceps surae and Achilles tendon might be also increased. This provokes strain and stress in this tendon and, as Paavola (2001) suggested, if the tendon repeatedly experienced a tension between 4% and 8% of the maximum tension, Achilles tendon could be damaged. Although the above-mentioned references related joint moments and injury risk, the direct association between greater joint moments and injury risk still remains to be established.

The main limitation of this study is that the measurements were performed in a lab setting as opposed to a more ecological valid scenario (e.g., outside on a soccer pitch). However, the 3G surface was well replicated in the lab and the soccer manoeuvres were performed with no movement restrictions (e.g., large testing area). Moreover, FIFA tests could not be performed to assess mechanical properties of the 3G systems used in this study and, consequently, no FIFA qualification were obtained. Although a power analysis revealed that the study had adequate statistical power (80%) to detect significant differences ($p < 0.05$) between surface cushioning conditions, another limitation of this study is the relatively low sample size ($n = 12$). On the other hand, the main strength is that the analysis of different biomechanical variables such as impact, joint kinematics and joint kinetics data provide evidence about how a cushioning underlay potentially influences impact moderating behaviour in soccer players.

Conclusion

In summary, the results obtained in this study demonstrated that the inclusion of cushioning underlay in 3G systems modified player and surface interaction across a range of soccer specific tasks. Overall, the cushioning underlay reduced resultant impact force characteristics and lower limb joint loading across most of the soccer-specific tasks. Importantly, it can be observed that findings of this study were predominantly explained by the mechanical cushioning of each surface condition instead of any impact moderating behaviour of the player (small ROM adaptations only). Considering the strong association of greater impact forces and joint loading for overuse injury risk, the inclusion of a cushioning underlay within a game scenario may reduce the incidence of impact related injuries in soccer player populations. It is therefore recommended that future studies are warranted to clarify how the inclusion of cushioning underlay in real-world outside soccer pitches as opposed to a lab setting influences impact moderating behaviour in soccer players.

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