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Does the cleat model interfere with ankle sprain risk factors in artificial grass?

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ABSTRACT

Background: The cleats-surface interaction has been described as a possible risk factor for lateral ankle sprain. However, their interaction is still unknown in individuals with chronic ankle instability. The purpose of this study was to determine the influence of different soccer cleats on kinematic, kinetic and neuromuscular ankle variables on artificial grass in soccer players with and without chronic ankle instability.

Methods: Eighty-two amateur athletes divided in two groups: 40 with chronic ankle instability and 42 without chronic ankle instability. All subjects performed 2 series of 6 consecutive crossover jumps with dominant foot, each one with one of the four models of cleats (Turf, Artificial grass, Hard and Firm ground). Cleat and group main effect and interactions of kinematic, kinetic and neuromuscular variables were analyzed according to factorial repeated measures ANOVA.

Findings: No statistically significant cleat and group main effect and interactions were identified in kinematic, kinetic and electromyographic magnitude of the peroneal muscles. A main effect of the group was observed for peroneus longus activation time for TF model ($p = 0.010$).

Interpretation: In soccer players, the contributor variables for ankle sprain were not influenced by the kind of soccer cleat used in a functional jump test on artificial grass. However, players with chronic ankle instability present delayed postural adjustments in peroneus longus with the TF model compared to players without chronic ankle instability.

1. Introduction

With > 265 million practitioners worldwide (Kunz, 2007), modern soccer become faster, unpredictable and extremely competitive (Sterzing et al., 2009). The development of the ultimate third generation artificial grass fields held the possibility for more hours of practice (FIFA, n.d.; FIFA, n.d.) and the several modifications done in the cleats, such as the distribution and the geometry of the studs (Fig. 1) (Lees and Nolan, 1998), have contributed to the fulfillment of the player's needs (Conenello, 2010; Sterzing, 2016). Currently there are four types of cleats used in artificial grass fields: Turf (TF), Artificial grass (AG), Hard Ground (HG) and Firm Ground (FG). The TF and AG models are suitable for artificial fields, the HG model for hard natural or dirt soccer fields and the FG model is indicated for natural grass in good conditions (Conenello, 2010; Queen et al., 2008). Despite the recommendations stated for each model, most players select the cleat model based on its stability (Hennig, 2011) against ankle sprains (Silva et al., 2017a).

Ankle sprain represents 10–30% of all musculoskeletal disorders (Fong et al., 2007) and about 76% in soccer (Garrick and Requa, 1988). A sudden and unexpected inversion/supination motion (Richie, 2001), with or without plantar flexion (Mok et al., 2011) is the most common injury mechanism (85% of cases) (Morrison and Kaminski, 2007). It has been estimated that from all athletes that suffered an ankle sprain, 40–75% may develop chronic ankle instability (CAI), characterized by

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	Studs/sole material	Studs: Number	Studs: Size	Studs: Geometry
	Rubber studs and compliant sole	> 55	6-7 mm	Prismatic
	Plastic studs and rigid plastic sole	22	8-10 mm	
		14	10-12 mm	
		11	10-12 mm	

Fig. 1. Cleat's characteristics
Legend: TF – Turf; AG – Artificial grass; HG – Hard ground; FG – Firm ground; mm – millimeters.

persistent residual pain, edema and reports of giving way and instability (Gribble et al., 2014; Hertel, 2002).

Despite the lack of consensus, several intrinsic risk factors for lateral ankle sprain (LAS) have been described: female gender (Doherty et al., 2014); taller and heavier athletes; ankle ligament instability; dominant limb (Beynnon et al., 2002); decreased dorsiflexion (Noronha et al., 2006); ankle alignment deformities (calcaneal varus); type of foot (cavus) (Morrison and Kaminski, 2007); increased center of pressure (COP) displacement (McKeon and Hertel, 2008; Munn et al., 2010); functional strength asymmetries of the ankle flexors (Fousekis et al., 2012); decreased evertor strength (Arnold et al., 2009), increased peroneal muscular activation time (Beynnon et al., 2002) and previous sprain history (Pourkazemi et al., 2014). However, only severity of initial sprain (grade II) is considered as a predictor for re-sprain (Pourkazemi et al., 2014). The fact that most individuals with a LAS episode will sustain at least one additional sprain, with many developing CAI (Gribble et al., 2016), and that this condition increase the risk of LAS, corroborates the need of studying athletes with CAI also (Silva et al., 2017a). On the other hand, extrinsic factors such as cleat-surface interaction have been studied (Silva et al., 2017b), highlighting the need of identifying an easily modifiable risk factor that help players to reduce different injuries risk. A systematic review demonstrated that different cleat models have been evaluated in terms of risk of injury in cases of calcaneal apophysitis, repeated impact injuries, and knee and ankle injuries related to increased joint loading (Silva et al., 2017b). However, only recently it has been demonstrated that in healthy athletes different cleat models aren't related to differences in ankle sprain risk factors (Silva et al., 2017a). Moreover, according to our knowledge, and despite deserving the attention of all sports and health professionals (Boer et al., 2014; Kemler et al., 2016), no study has evaluated the influence of different cleat models in ankle sprain risk factors in athletes with postural control deregulation such as CAI.

Therefore, the aim of the study is to evaluate the influence of different cleat models on variables related to the risk of ankle sprain in athletes with and without CAI. Specifically, postural alignment related variables, loading rate of vertical and lateral components of the ground's reaction forces; center of pressure (COP) displacement-related variables and neuromuscular variables (onset timing and magnitude of the peroneal muscle) were selected for analysis. Based on evidence that the cleat-surface traction must be enough to prevent slip and facilitate turning maneuvers (Conenello, 2010), but also that excessive fixation has been implicated in non-contact injuries during turning and cutting maneuvers associated to increased torque on lower extremity joint structures (Lambson et al., 1996), it can be hypothesized that the studs' number, distribution and height can significantly affect joint stability, expressed through kinetics, kinematics and muscle activity variables.

2. Methods

An experimental intra-subject study design was developed in a sample of federated amateur soccer players, with and without CAI.

2.1. Participants

Eighty-two male athletes aged between 18 and 30 years, from 32 of the 96 clubs of the Porto Football Association participated in the present study. The recruitment of the sample was conducted by electronic invitation to all clubs. The athletes who volunteered for the study have been included. The sample was divided in two groups based on the presence of CAI: 40 athletes were included in the group with CAI, and 42 in the without CAI group.

To participate in the present study athletes must had federated soccer practice in the last 5 seasons, as well a foot size of 41. Participants assigned to the CAI group met the criteria set by the International Ankle Consortium (Gribble et al., 2014). To be included in the CAI group, athletes should have history of ankle sprains in the dominant limb for less than one year and respond "yes" in 5 or more questions regarding their dominant limb in the Ankle Instability Instrument (AII) (Docherty et al., 2006; Gribble et al., 2014) and presented a positive drawer test (Docherty et al., 2006; Gribble et al., 2014; van Dijk, 2002; Vries et al., 2010). Athletes were excluded if they presented one or more of the following criteria: history of surgery in both lower limbs, pathologies that directly affect the balance, conditions that alter peripheral sensory afferents, any type of neuro-musculoskeletal injury beyond ankle sprain in both lower limbs in the last year, and occurrence of ankle sprain in the last 3 months, due to the possibility of still being in an acute or subacute stage (Caffrey et al., 2009; McKeon et al., 2010).

Healthy control participants were selected according to the same exclusion criteria applied to the CAI group and were also excluded if they had history of ankle sprain.

The characteristics of the participants are presented on Table 1. It should be noted that both groups were comparable in age, body mass, height and have similar experience of official soccer practice (11 years) and hours of training in the current season (7 h/week). The group with CAI presented on average 1.9 ankle sprains in the dominant foot, most of which (39.96%) occurred > 12 months ago (Table 1). Since decreased dorsiflexion could be a predictor of the occurrence of ankle sprain we also characterized participants regarding this variable (Noronha et al., 2006). The group with CAI presented decreased values comparing to the group without CAI.

The study was conducted according to the ethical norms of the Institutions involved and conformed to the Declaration of Helsinki, with informed consent from all participants.

2.2. Instruments

Anthropometric data were evaluated with a scale – Seca® 760 (1 kg accuracy), and a stadiometer - Seca® 222 (1 mm accuracy) (SECA, 2014). Dorsiflexion range of motion was assessed with a fluid-filled inclinometer with 1° increments (MIE Medical Research Ltd., Leeks, UK) (Rabin et al., 2015). To control the jump speed, an on-line digital metronome was used (www.metronomeonline.com).

Table 1
Sample characterization. Significant differences were identified with a asterisk (*).

		With CAI	Without CAI	p Value
Age (years) – mean (SD)		21.4 (2.97)	21.3 (2.63)	0.853
Body mass (Kg) - mean (SD)		68.8 (4.91)	69.0 (7.19)	0.846
Height (m) - mean (SD)		1.7 (0.05)	1.8 (0.06)	0.734
Dorsiflexion ROM (degrees) - mean (SD)		35.8 (4.26)	41.1 (4.04)	< 0.001*
Official football practice (years) – mean (SD)		10.8 (3.63)	10.8 (2.60)	0.906
Training period (hours per week) – mean (SD)		6.8 (1.53)	7.2 (1.82)	0.271
Cleat model preference for play on artificial grass - relative frequencies (%)	TF model	3.7%	0.0%	0.222
	AG model	17.1%	22.0%	
	HG model	14.6%	11.0%	
	FG model	13.4%	18.3%	
Number of sprains in the dominant foot - mean (SD)		1.9 (1.64)	–	–
How long ago did the last sprain occur - relative frequencies (%)	3–6 months	7.58%	–	–
	6–12 months	20.08%	–	–
	12–24 months	32.58%	–	–
	> 24 months	39.96%	–	–

The "Ankle Instability Instrument" was used to identify athletes with CAI. This instrument presents high values of test-retest reliability (ICC = 0.95). Internal consistency reliability estimates (Cronbach alpha) for each factor and the total measure ranged from 0.74 to 0.83 (Docherty et al., 2006).

The ankle eversion/inversion range of motion was monitored with the *Qualisys motion capture*® system, with 4 cameras (*Oqus* 1) with an acquiring frequency of 100 Hz (*Qualisys AB*, Packhusgatan 6 S-411 13 *Gothenburg* Sweden) and 19 mm reflector markers. This instrument present an excellent intra-observer reliability (ICC = 0.90) (Sinclair et al., 2012).

The ground reaction forces (GRF) signal was collected with a sampling frequency of 100 Hz, with two *Bertec*® FP4060-10/8 force platforms connected to a AM 6300 amplifier (*Bertec Corporation*, 6171 Huntley Road Suite J Columbus, U.S.A.) and to the *Qualisys motion capture*® system. The instrument shows an excellent intra-observer reliability in jump assessments (ICC 0.92–0.98) (Hori et al., 2009). The platforms were covered with a 3rd generation artificial grass carpet (6 m^2), composed of polyethylene fibers (60–65 mm) and filled with purified silica and rubber.

The electromyographic signal (EMG) of the peroneal muscles (main lateral stabilizers of the ankle during inversion) was monitored using a bioPLUX research wireless signal acquisition system (Plux Ltd., Portugal). The signals were collected at a sampling frequency of 1000 Hz and were pre-amplified in each electrode and then fed into a differential amplifier with an adjustable gain setting (25–500 Hz; common-mode rejection ratio: 110 dB at 50 Hz, input impedance of 100 MΩ and gain of 1000). Self-adhesive silver chloride EMG electrodes were used in a bipolar configuration and with 20 mm between detection surface centers (Dahlhausen®, Köln, Germany). The skin impedance was measured with an Electrode Impedance Checker (Noraxon USA, Inc.). The determination of the peroneus longus (PL) and peroneus brevis (PB) muscles' activation time through EMG signal presents an excellent intra-observer reliability (0.82–0.91) (Hopper et al., 1998).

Finally, the data processing and analysis were made using the following software: Matlab R2012a (The MathWorks Inc., Boston, USA) and *Acqknowledge* 3.9 (BIOPAC Systems, Inc. Goleta, USA).

2.3. Procedures

2.3.1. Preparation of the participants

The muscle belly skin surface of selected muscles and patella of the dominant limb was prepared to reduce the electrical resistance to < 5kΩ. The dominant limb was determined asking the participant to kick a ball, the dominant limb was considered the one that kicked the ball. For PL and PB muscles the electrodes were placed according to the SENIAM recommendations. These locations were confirmed by palpation, during the voluntary contraction of those muscles, always by the same researcher (physiotherapist, expert with 16 years of practice). The ground electrode was placed in the patella (Hermens et al., 2000). Three reflector markers with 19 mm of diameter were placed in the posterior face of the leg and on the shoe: (1) 2 cm below the popliteal fold in the medium point between the lateral and medial face, (2) over the Achilles tendon in the alignment of the two malleolus and (3) in the center of the posterior face of the shoe (Beynnon et al., 2001; Norkin and White, 2009; Silva et al., 2017a). All participants wore a new cleat (size 41), ensuring a distance of 0.5cm between the longest toe and the front of the cleat.

2.3.2. Data collection

All the participants were submitted to a functional test adapted from

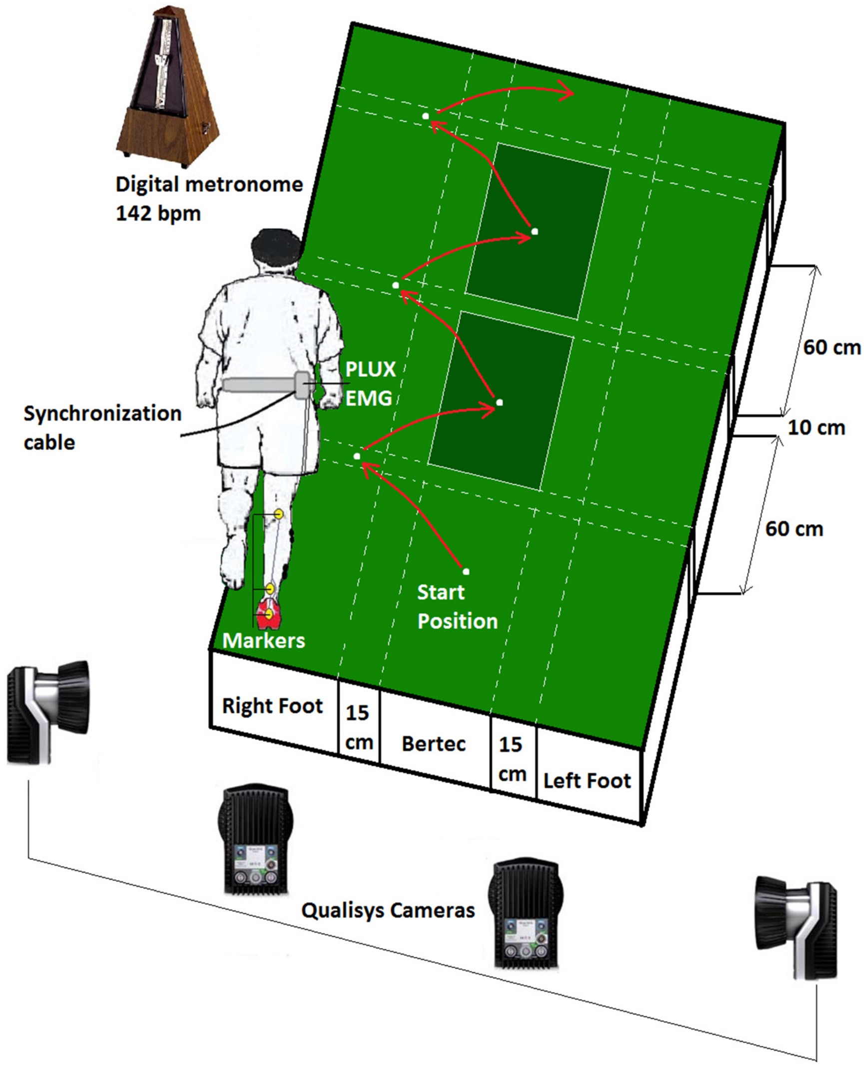

Fig. 2. Functional test
Legends: bpm – beats per minute; EMG - Electromyography.

the 6-m crossover test after 10 min of warm-up in the cycloergometer with 2% of the body weight and self-directed stretching exercises (Brown et al., 2012; Silva et al., 2017a). The participants performed 2 series of 6 consecutive crossover jumps with the dominant foot, at a cadence of 142 beats per minute (controlled by metronome) while wearing one of the four models of cleats (*Turf*, Artificial grass, *Hard ground* and *Firm ground*). A 2-minute resting period was set between each series (Caffrey et al., 2009; Docherty et al., 2005). The cadence adopted was based on the maximum cadence executed by active individuals in this kind of functional tests (Caffrey et al., 2009; Docherty et al., 2005). Participants carried out a series of trials for familiarization with the task, to memorize the execution speed and minimizing the effects of the learning process. To diminish the order effect, the sequence of the cleats was randomized. All possible sequences were written on several papers, which the participant himself selected from a bag. The paper selected was removed until there were no more papers left in the bag. All participants had to respect the distances indicated in Fig. 2 in their jumps. A trial was considered valid when the subject reached this distance in each jump with the defined cadence and land inside the force plate landmarks (Caffrey et al., 2009; Docherty et al., 2005).

2.3.3. *Data processing*

All variables were analyzed during the foot contact periods on force plates and the average values were used for analysis. The signal from the force platform was low pass filtered through a 4th order *Butterworth* filter of 15 Hz and was normalized to the body weight. The initial contact with the ground was defined as the instant where the value of the vertical component of the GRF was > 10 N (Brown et al., 2012). The loading rate of the vertical (Fz) and medio-lateral (Fx) components of the GRF was obtained by calculating the difference between the maximum and minimum values, divided by the time interval. The medio-lateral (COPx) and anteroposterior (COPy) displacements of the COP were calculated for each contact period. The medio-lateral (V_COPx) and antero-posterior (V_COPy) average speeds for the COP displacement were assessed by dividing the COP displacement by the time interval (Duarte and Freitas, 2010).

A 2nd order *Butterworth* low-pass filter of 6 Hz was applied to kinematic data. The total ankle eversion/inversion range of movement (ROM) was obtained through the difference between the maximum eversion and inversion the angle formed between the 'leg' segment and the 'hind foot' segment (Whatman et al., 2012).

The electromyographic signals were filtered using a zero-lag, second-order Butterworth filter with an effective band pass of 10 to 500 Hz. The root mean square was calculated using a moving window of 20 samples (Schmid et al., 2010). The temporal analysis was made in relation to the instant of foot contact to the ground (T0). The timing of EMG onset was defined for each muscle as the beginning of the interval

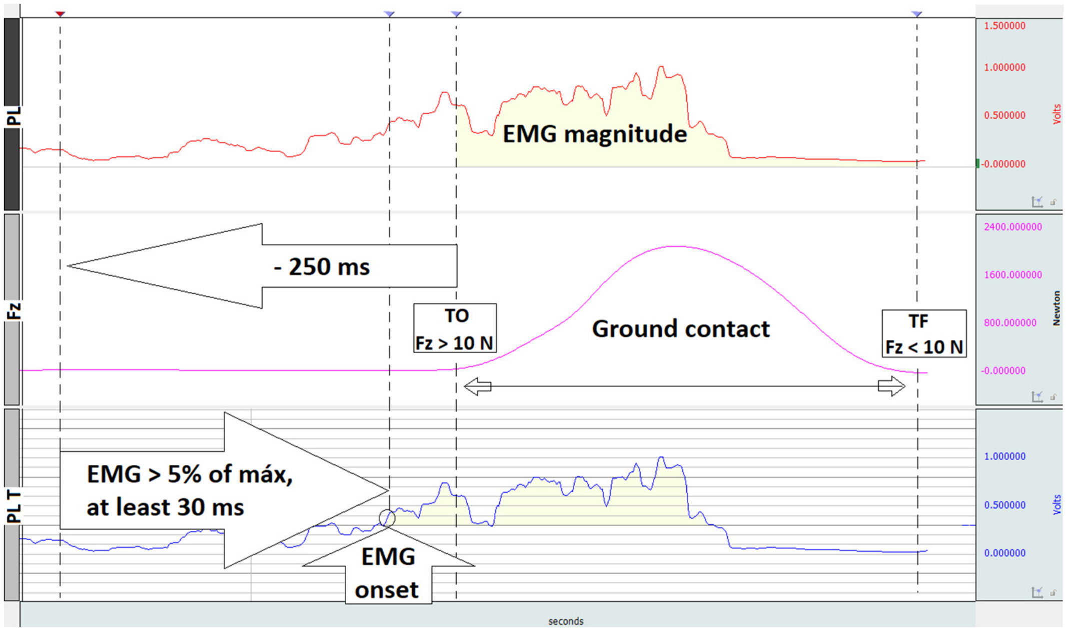

Fig. 3. EMG analysis procedure
Legends: TO – instant of foot contact to the ground; TF – instant when the foot leaves the ground; PL – peroneus longus; Fz – Vertical component of the ground reaction forces; PL T – timing analysis of peroneus longus; ms – milliseconds; EMG - Electromyography.

Table 2
Group and cleat model main effects and interactions of kinematic variables.

Variables/cleat models		Groups		Group cleat interaction	Main effects of the group	Main effects of the cleat models
		With CAI mean (SD)	Without CAI mean (SD)	p Value (1- β)	p Value (1- β)	p Value (1- β)
ROM (degrees)	TF	7.4 (4.90)	5.9 (3.24)	0.257	0.255	0.263
	AG	6.7 (3.30)	5.2 (2.96)	(0.359)	(0.205)	(0.354)
	HG	6.5 (3.36)	6.7 (4.61)			
	FG	6.2 (4.60)	5.7 (3.43)			
COPx (mm)	TF	212.2 (150.57)	174.3 (133.30)	0.144	0.542	0.526
	AG	195.8 (123.33)	197.8 (148.03)	(0.470)	(0.093)	(0.209)
	HG	199.9 (132.91)	175.8 (126.86)			
	FG	186.5 (109.82)	181.0 (125.00)			
COPy (mm)	TF	143.6 (48.39)	158.9 (76.67)	0.480	0.451	0.380
	AG	147.0 (43.89)	155.0 (58.23)	(0.229)	(0.116)	(0.278)
	HG	141.6 (38.42)	149.8 (59.70)			
	FG	144.8 (41.40)	145.7 (57.78)			
V_COPx (mm/s)	TF	773.8 (531.61)	657.6 (503.22)	0.224	0.714	0.763
	AG	717.5 (467.24)	742.7 (558.14)	(0.386)	(0.065)	(0.126)
	HG	730.1 (464.15)	686.8 (492.53)			
	FG	698.8 (416.86)	687.3 (463.95)			
V_COPy (mm/s)	TF	519.1 (164.21)	593.4 (263.14)	0.444	0.242	0.302
	AG	548.9 (163.46)	591.4 (243.16)	(0.245)	(0.214)	(0.326)
	HG	525.52 (140.41)	580.9 (249.33)			
	FG	527.40 (181.46)	547.0 (212.92)			

of at least 30 ms where a value equal to or higher than 5% of maximum obtained in each trial was observed in a time window starting at −250 ms in relation to T0 (Fig. 3) (Hodges and Bui, 1996; Nieuwenhuijzen et al., 2002; Shiratori and Latash, 2001). To allow us to look at the net effect of the stimulus in dynamic and cyclic tasks, averaged EMG magnitude of the active period was subtracted from the raw signal for the EMG onset. The timing of muscle onset was confirmed by visual inspection (Nieuwenhuijzen et al., 2002). The analysis of the magnitude of activation of the peroneal muscles was performed through the mean RMS of the EMG signal during the periods of contact with the platforms and was normalized to the signal obtained during maximal voluntary isometric contractions (MVIC) (Akhbari et al., 2007). MVIC were assessed with the participant in a supine position, knee extended, heel off the table and the foot in a slight plantar flexion. The resistance was manually applied always by the same researcher while the participant was encouraged verbally to execute their maximum eversion contraction (Kendall et al., 2005). For signal normalization, mean RMS of the interval between the second 2 and second 4 during each MVIC were used for analysis. The final values of the analysis (time and magnitude of activation) for each cleat were calculated considering the average of the 4 contacts of the participant on the platforms during the 2 jump repetitions.

2.4. Statistics

PASW® Statistics 20 software was used with a significance level of

Table 3
Group and cleat model main effects and interactions of kinetic variables.

Variables/cleat models		Groups		Group cleat interaction	Main effects of the group	Main effects of the cleat models
		With CAI mean (SD)	Without CAI mean (SD)	p Value (1- β)	p Value (1- β)	p Value (1- β)
LRVz (BW/s)	TF	2.37 (0.47)	2.57 (0.78)	0.575	0.241	0.050
	AG	2.32 (0.42)	2.33 (0.47)	(0.189)	(0.215)	(0.641)
	HG	2.44 (0.66)	2.56 (0.73)			
	FG	2.35 (0.33)	2.43 (0.37)			
LRVx (BW/s)	TF	0.42 (0.10)	0.46 (0.09)	0.582	0.230	0.517
	AG	0.44 (0.10)	0.46 (0.13)	(0.186)	(0.223)	(0.212)
	HG	0.43 (0.10)	0.45 (0.08)			
	FG	0.43 (0.11)	0.44 (0.09)			

0.05. The mean and median were used as measures of the central tendency, and the standard deviation and interquartile range as dispersion measures (Marôco, 2010).

The chi-square test was used to test an association between a cleat model preference and the groups with and without CAI. Regarding the kinematic, kinetic and neuromuscular variables, the main effect of the group and the cleat and related interactions were analyzed according to the factorial repeated measures ANOVA. The magnitude of the effects was accessed through the Cohen's *d*.

The Bonferroni correction was used for the post hoc analysis. T-Student test was used to compared sociodemographic data and the same cleat in different groups. Power analysis (1-β) was performed to give an indication of the power of hypothesis tests (Marôco, 2010).

3. Results

No statistical significant cleat and group main effects and interaction were observed in kinematic variables (Table 2).

No statistical significant cleat and group main effects and interaction were observed in the slope of Fz e Fx (Table 3).

While no statistical significant cleat and group main effects and interaction were observed in the EMG magnitudes and in the PB activation time (Table 4), a main effect of the group was observed for PL activation time for TF model ($p = 0.010$) with a moderate effect (Cohen's $d = 0.60$). For this variable, the group with CAI presented a delayed activation time. No main effects of the cleat and no significant cleat-group interaction were observed.

4. Discussion

The results of the present study demonstrate delayed PL activation with the TF model in the group with CAI compared to the group without CAI. However, no interaction was observed between the cleat and the group demonstrating that the effect of the cleat seems to be independent from the effect of the group. Also, no statistically significant main effects and interactions were observed for the other kinematic, kinetic and neuromuscular variables. The results of the present study should be discussed under the following assumptions: 1) the low degree of unpredictability of the functional test adopted on the present study decreased the test difficulty (Borotikar et al., 2008); and 2) although we can't access to the particularities of intervention, all athletes participated in the present study underwent physical therapy after injury and were competing for at least 3 months without restriction. The effectiveness of proprioceptive training programs in reducing the rate of ankle sprains and improving motor control are well established (Schiftan et al., 2015) and may have been a key factor for the few differences between groups.

The absence of significant main effects and cleat-group interactions in ankle ROM as well COP related variables and vertical and mediolateral loading rates of the ground reaction forces could be explained by the fact that our test was performed with cleats, while the original crossover test was described barefoot (Caffrey et al., 2009; Docherty et al., 2005). The use of cleats may have provided greater comfort and ankle stability minimizing the functional deficits expected in the group with CAI (Rabello et al., 2014), even when this group presents decreased ankle dorsiflexion. In fact, restricted ankle dorsiflexion could

Table 4
Group and cleat model main effects and interactions of neuromuscular variables. Significant differences were identified with a asterisk (*).

Variables/cleat models		Groups		Group cleat interaction	Main effects of the group	Main effects of the cleat models
		With CAI mean (SD)	Without CAI mean (SD)	p Value (1- β)	p Value (1- β)	p Value (1- β)
EMG_PL (%)	TF	95.6 (41.89)	104.9 (50.86)	0.138	0.674	0.306
	AG	103.0 (48.14)	95.4 (32.71)	(0.479)	(0.070)	(0.323)
	HG	102.6 (54.24)	94.5 (33.86)			
	FG	94.1 (40.81)	99.7 (45.95)			
EMG_PB (%)	TF	93.2 (39.77)	95.3 (31.33)	0.722	0.472	0.162
	AG	90.4 (31.67)	93.8 (29.07)	(0.138)	(0.110)	(0.449)
	HG	89.7 (30.41)	93.0 (27.69)			
	FG	86.8 (29.11)	91.9 (27.18)			
AT_PL (ms)	TF	−5.3 (74.48)	−64.3 (118.76)	0.089	0.031*	0.598
	AG	−45.7 (74.08)	−55.8 (84.89)	(0.552)	(0.582)	(0.180)
	HG	−28.9 (85.16)	−66.1 (85.03)		TF With > Without p = 0.010*	
	FG	−38.6 (79.56)	−41.4 (121.53)			
AT_PB (ms)	TF	−68.3 (81.38)	−71.6 (109.90)	0.488	0.757	0.814
	AG	−84.5 (86.56)	−67.0 (88.14)	(0.053)	(0.061)	(0.111)
	HG	−64.9 (81.88)	−76.8 (81.26)			
	FG	−69.1 (95.14)	−70.7 (88.48)			

lead to an abnormal lower extremity biomechanics during closed chain exercise (Dill et al., 2014), however our kinematic results didn't confirm this idea. Since we didn't evaluate the passive full ROM in other directions we can't assure that they are not in the origin of the non-existence differences in kinematic and kinetic variables. Future studies should evaluate these variables. The non-existence of a main effect of the cleat for the above mention variables together with the fact that during the stance phase the players achieved in average a functional ROM around 13% to 25% of the total ROM (Dubin et al., 2011), may suggest an appropriate dynamic control for the task. However, it should be noted that in the present study the degree of ankle inversion/eversion was calculated as the total difference, being not possible to identify if the movement occurred in the last degrees of inversion and thus be considered as more dangerous (Morrison and Kaminski, 2007).

Based on the evidence that cleats with studs that do not fully penetrate the artificial grass lead to increased instability (Clarke and Carré, 2010), increased values of COP related variables were expected in the present study in cleats with this feature. However, there wasn't observed a main effect of the cleat over these variables, revealing that athletes with and without CAI are able to deal with the postural challenges that different cleats can impose (Buchanan et al., 2008; Caffrey et al., 2009; Docherty et al., 2005). While similar results have been obtained by a recent study which used a similar sample of players without CAI (Silva et al., 2017a), the results in CAI are more surprising, since wasn't found significant interaction between cleats and the group. This could be due to the dry condition and the perfect state of the artificial grass. These particularities could have uniformized the cleat models' behaviour, allowing full penetration to the cleat with higher studs (HG and FG) and sufficient traction to the models with lower studs (TF and AG) (Silva et al., 2017a). Also, considering the evidence demonstrating postural control deficits in joints proximal to ankles with CAI (Bullock-Saxton, 1994; Caulfield and Garrett, 2002; Hertel and Olmsted-Kramer, 2007), future studies dedicated to the kinematic analysis of whole body are required to verify if the absence of differences in COP related variables are related to hip or trunk compensatory strategies.

Although some authors believe that increased vertical impact loads compared to mediolateral direction may allow the ankle joint to remain more stable avoiding excessive inversion forces (Dayakidis and Boudolos, 2006), no statistical significant main effects and interactions were observed, which agrees with the results obtained in the evaluation of healthy soccer players, that used the side hop test alternatively to the cross over in a very similar methodology (Silva et al., 2017a). In fact, it is possible that structurally different cleat models imposed different mechanical traction when assessed with mechanical instruments, but similar biomechanical traction when assessed in real conditions with soccer players in the same playing field (Sterzing et al., 2008). It has been argued that the playing fields are more determinant that the cleat models for identifying differences between models' mechanical properties (Villwock et al., 2009). These results contrast with studies associating higher studs with higher traction indices and consequently an increased risk of injury (Lake, 2000; Queen et al., 2008; Sterzing et al., 2009).

In the present study, no statistical main effects and interactions were observed in EMG magnitude of peroneal muscles. Despite the divergence of these results regarding peroneal strength deficits in individuals with CAI (Hartsell and Spaulding, 1999; Mattacola and Dwyer, 2002; McKnight and Armstrong, 1997; Pontaga, 2004; Wilkerson et al., 1997), the results of the present study indicate that athletes with CAI present no impairments in the magnitude of compensatory and accompanied postural adjustments. Despite both groups presented peroneal onset timings related to feedforward mechanisms, delayed activation of PL was observed in the TF model in CAI group (almost 60 ms of delay). Interestingly, this difference is only noticed with the cleats (TF model) that practically no athlete is accustomed to play with. According to Hennig (2011) neuromuscular adaptations to the cleats lead athletes to achieve a better motor performance with the models they are familiarized. In the present study the effect of the non-familiarization phenomenon was only observed in the CAI group (Hennig, 2011), revealing an increased difficulty to deal with new conditions. In another perspective, it should be considered that TF model presents the most compliant sole of all models studied, which could lead to higher foot segments mobility and interfere with ankle stability, particularly in the more vulnerably group (CAI). Future studies are required to confirm this hypothesis.

It should be noted that despite a main effect of the group for PL onset timing, this wasn't accompanied by a main effect of the group in kinematic and kinetic variables for the functional test performed, probably in more demanding tasks the differences observed in muscle activation timings could interfere with kinematic and kinetic variables increasing the risk of injury. Future studies involving more demanding postural tasks and assessing also other ankle muscles (e.g. tibialis anterior and soleus) are needed to confirm this hypothesis. It should be also considered the low observed power (1-β) obtained in the present study for the kinematic and kinetic variables future studies with higher sample are required to confirm our results.

Also, considering the limitations previously stated, future studies should analyse the influence of cleats on wet artificial grass fields, possibly with a few years of use, using full body kinematics, and evaluating dynamic tasks with some degree of unpredictability. It should be also considered that the interpretation of our results was based on the assumption that mechanical characteristics of the artificial grass maintained constant through the entire experiments, confirmed by an expert technician. However, because some variability may have occurred, future studies are required to assess this issue.

5. Conclusion

The findings obtained in the present study indicate that in soccer players, the contributor variables for ankle sprain were not influenced by the kind of soccer cleat in a functional test on a third-generation artificial grass. However, players with CAI present delayed postural adjustments in PL muscle with the TF model compared to players without CAI. The authors thank Adidas Portugal and Relvados e Equipamentos Desportivos Lda for their support.

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