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The influence of different soccer cleat type on kinetic, kinematic and neuromuscular ankle variables in artificial turf

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Lateral ankle sprain is the most prevalent injury in soccer athletes. Enhanced by the variety of soccer cleats and by increased use of artificial turf, the interaction between the ground and the footwear has taken high importance as a lateral ankle sprain risk factor. The higher incidence of injuries in the second half of the match reflects the need of studying this interaction during tasks involving muscle fatigue. To evaluate the influence of different soccer cleats on kinetic, kinematic and neuromuscular ankle variables in artificial turf under two conditions: with and without fatigue of lateral ankle dynamic stabilizers. Study design: Experimental study within-subjects design. Twenty-four healthy athletes participated in this study. All subjects performed three sets of five medial-lateral unipodal jumps, each one with one of three models of cleats (Turf, Hard and Firm ground) on two conditions: with and without fatigue induced by the isokinetic dynamometer. The electromyographic activity of long and short peroneal heads, ground reaction forces and the movement of the rear-foot were collected and used to calculate kinematic (ankle eversion/inversion, centre of pressure displacement and velocity), kinetic (loading rate of the ground reaction forces) and neuromuscular variables (activation time of peroneal muscles). With the exception of decreased peroneal activation time with the Hard ground model (without fatigue vs. with fatigue), no statistically significant differences were identified in the ankle variables, between cleats, neither between the two evaluated conditions. In healthy soccer athletes, the contributor variables for ankle sprain were not influenced by the kind of soccer cleat in a functional test on a third generation artificial turf.

Keywords: lateral ankle sprain; cleats; soccer; artificial turf; fatigue

1. Introduction

With over 265 million athletes, soccer is the most played sport in the world (Kunz, 2007). Involving running with direction shifts, rotations, sudden stops and multiple jumps (Dubin, Comeau, McClelland, Dubin, & Ferrel, 2011), soccer is associated with a high prevalence of ankle sprain (Fong, Hong, Chan, Yung, & Chan, 2007; Waterman, Owens, Davey, Zacchilli, & Belmont, 2010). Ankle sprains account for 80%–85% of all ankle injuries and 77%–80% of them affect the lateral ligaments (lateral ankle sprain (LAS)) (Fong, Chan, Mok, Yung, & Chan, 2009; Garrick, 1977; Garrick & Requa, 1988). Amongst amateur players, its incidence is about 2.16/1000 h of exposure (Kofotolis, Kellis, & Vlachopoulos, 2007), being more frequent during match periods (11.68/1000 h of exposure) (Fong et al., 2007). The combination of excessive inversion or supination and plantar flexion while having an external rotation force applied on the leg has been traditionally described as the main lesion mechanism (Richie, 2001). More recent studies have demonstrated a trend of sudden ankle inversion but not plantar flexion (Mok et al., 2011).

Despite the lack of consensus in some variables, several intrinsic risk factors for LAS have been described: anthropometric data (taller and heavier athletes); ankle ligament instability; dominant limb (Beynnon, Murphy, & Alosa, 2002); decreased dorsiflexion (Noronha, Refshauge, Herbert, & Kilbreath, 2006); female gender (Doherty et al., 2014); ankle alignment deformities (calcaneal varus); type of foot (cavus) (Morrison & Kaminski, 2007); increased centre of pressure (COP) displacement

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(McKeon & Hertel, 2008; Munn, Sullivan, & Schneiders, 2010); decreased evertor strength (Arnold, Linens, Motte, & Ross, 2009); increased peroneal muscular activation time (Beynnon et al., 2002) and previous sprain history (Pourkazemi, Hiller, Raymond, Nightingale, & Refshauge, 2014). Although not normally included in the intrinsic risk factors, fatigue should be studied due to its influence on joint-position sense (Bisson, McEwen, Lajoie, & Bilodeau, 2011; Jackson, Gutierrez, & Kaminski, 2009; Lin, Li, Tsai, & Liing, 2008). In fact, several authors demonstrated that fatigue can lead to a predisposition for injury, affecting the joint-position sense that could reduce the capability to jump landing with the foot in an appropriate position (Lin et al., 2008; Sandrey & Kent, 2008). Changes on afferent muscles' discharge patterns, due to metabolic acidosis, may reduce the responsiveness of the Golgi tendon organs and muscle spindles, thus originating proprioceptive deficits, increased activation time, decreased muscular strength and postural stability (Hiemstra, Lo, & Fowler, 2001). In terms of extrinsic risk factors, the increased traction index between the footwear and the playing field has been described as the most relevant for the LAS (Hennig, 2011; Lake, 2000).

Over the past few years, an increasing implementation of third generation synthetic turfs has been observed, mainly amongst amateur clubs. The *Fédération Internationale de Football Association* (FIFA) has encouraged such an implementation due to the turf's durability, the low maintenance cost and the ability to withstand a high number of playing hours, regardless of weather conditions (FIFA, 2009, 2012). Currently, sporting footwear manufacturers offer various options for synthetic turfs. Amongst all the characteristic properties of this type of footwear, the two considered most important by athletes are stability of the foot in the shoe and good traction on the field (Hennig, 2011; Kinchington, Ball, & Naughton, 2011). The component that has the major influence in fulfilling the last requirement is the sole of the footwear (Sterzing, Müller, & Milani, 2010). There are mainly three sole models allowed on synthetic fields: *Turf*; *Hard ground* and *Firm ground* (Figure 1). Despite all models can be used in synthetic fields, some manufacturers recommend the use of the first model for synthetic turf, the second model to either hard natural and synthetic turfs and the last one for normal natural turfs (Queen et al., 2008). However, the footwear selection is not usually

Figure 1. Cleat models used in the present study.

based on this recommendation, as the *Firm ground* and *Hard ground* models are frequently used on synthetic playing fields (Sterzing et al., 2010). This fact may be the reason behind a higher frequency of LAS in third generation synthetic turfs when compared to natural turfs (Ekstrand, Timpka, & Hägglund, 2006). However, it should be noted that there is not an absolute consensus about this epidemiological factor, as other studies have not identify significant differences in injury frequency between these two types of playing field (Aoki et al., 2010; Ekstrand, Hagglund, & Fuller, 2011).

Considering the exponential increase on the number of synthetic turfs, the different cleats options and the high frequency of LAS in football players, especially on the second half of matches (Hawkins & Fuller, 1999; Tsiganos, Sotiropoulos, & Baltopoulos, 2007), it becomes relevant to analyse the type of cleats that best suit the synthetic turf, as a means to minimize the high injury rates. This study aims to evaluate the influence of the soccer shoe models on kinetic, kinematic and neuromuscular ankle variables in artificial turf, with or without ankle of lateral ankle dynamic stabilizers. Specifically, this study aims to evaluate the influence of soccer shoe models in kinematic variables (ankle eversion/inversion range of movement); loading rate of vertical and lateral components of the ground's reaction forces (GRF); COP displacement-related variables and neuromuscular variables (activation time of the peroneals). Despite the lack of evidence, we thought that increased values in these variables could be associated to higher ankle instability and a consequent increased risk of ankle sprain.

2. Methods

2.1. Participants

Twenty-four male football players with at least five consecutive years of official competition and aged between 18 and 30 years participated in this study (age = 23.13 ± 1.90 years, height = 1.76 ± 0.06 m, weight = 68.36 ± 5.20 kg; mean ± SD). Only male athletes were included since the risk of ankle sprain is different in males and females (Doherty et al., 2014). The following exclusion criteria were defined: previous ankle sprain history, the presence of another neurological, muscular or skeleton injury (less than one year ago) of the lower limbs, history of lower limb surgery history, balance disturbances, neuropathies or other pathologies that affect posture control, as well as athletes that were under the influence of local anaesthetics (McKeon, Booi, Branam, Johnson, & Mattacola, 2010). To guarantee sample homogeneity as to foot type, six individuals with flat foot soles were excluded. All participants presented pes cavus.

Most participants have between 11 and 15 years of official football practice (67%) and a training period of 7–8 h per week in the current season (79%).

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

To characterize the sample as to the type of foot (cavus, normal and flat) a pressure platform *Emed*® model a-50 (*Novel*, 964 Grand Avenue) with an acquiring frequency of 50 Hz was used (St. Paul, MN, USA). This platform has a sensor area of 389 × 226 (mm) and a 2 sensors/cm^2 resolution and presents an excellent intra-observer reliability (ICC = 0.975) in the determination of the foot arch index (Akins, Keenan, Sell, Abt, & Lephart, 2012). The calculation of the *Cavanagh index,* as an indicator of the type of foot, was made using a foot characterization *software* (Oliveira, Sousa, Santos, & Tavares, 2012).

To gather and analyse the ankle eversion/inversion movement amplitude, the *Qualisys motion capture*® system was used, with four cameras (*Oqus* 1) with an acquiring frequency of 100 Hz (*Qualisys AB*, Packhusgatan 6 S-411 13 *Gothenburg*, Sweden) and 19 mm reflector markers. Although we use a 3D image capture system, the range of motion was analysed only in the frontal plane. The instrument presents an excellent intra-observer reliability (ICC = 0.90) for this plane (Sinclair et al., 2012).

The GRF signal was collected with a *Bertec*® FP4060-10 force platform connected to a AM 6300 amplifier (*Bertec Corporation*, 6171 Huntley Road Suite J, Columbus, OH, USA), connected to the *Qualisys motion capture*® system. In jump assessment, the instrument shows an excellent intra-observer reliability (ICC between 0.92 and 0.98) (Hori et al., 2009). The platform was covered with a third generation synthetic grass rug (2 m^2), composed of polyethylene fibres that are 60—65 mm high and filled with purified silica and rubber. Its installation was assured by specialized technicians, with the platform being later calibrated.

The electromyographic (EMG) signal of the peroneal muscles was monitored using a bioPLUX research wireless signal acquisition system (Plux Ltda, Lisbon, 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 a distance of 20 mm between detection surface centres (Dahlhausen®, Köln, Germany). The skin impedance was measured with an Electrode Impedance Checker (Noraxon USA, Inc., Scottsdale, AZ, USA). While determining the long peroneal (LP) and short peroneal (CP) muscles' activation time, the EMG signal shows an excellent intra-observer reliability (ICC between 0.82 and 0.91) (Hopper, Allison, Fernandes, O'Sullivan, & Wharton, 1998).

To control the jump speed, an online digital metronome was used (www.metronomeonline.com).

In the process of fatigue induction on the ankle evertors, the *Biodex*®*System 4* (*Biodex* Medical Systems, Inc. 20 Ramsey Road, Shirley, NY, USA) was used. This instrument has an excellent intra-observer reliability (ICC between 0.87 and 0.94) in the assessment of the evertors' isokinetic force (E. Aydog, S. Aydog, Çakei, & Doral, 2004).

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

2.3. Procedures

The procedures were divided in two moments: zero moment (M0 – condition without fatigue) and moment one (M1 – condition with fatigue). The M0 and M1 were made on the same day to assure the same location for the gathering of the EMG signal. The M0 always preceded M1.

2.3.1. Zero moment (M0)

2.3.1.1. Preparation of the participants. The midbelly skin surface of selected muscles and patella of the dominant limb were prepared (shaved and then the dead skin cells and non-conductor elements were removed with alcohol and with an abrasive pad) to reduce the electrical resistance to <5kΩ. For LP the electrodes were placed at 25% of the line between the head of the fibula and the lateral malleolus, for the CP the electrodes were placed at 25% of the line between the lateral malleolus and the head of the fibula. These locations were confirmed by palpation, during the voluntary contraction of those muscles. The ground electrode was placed in the patella (Hermens, Freriks, Disselhorst-Klug, & Rau, 2000). The dominant limb was determined asking the participant to kick a ball.

Additionally, 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 heel in the alignment of the two malleolus and (3) in the centre of the posterior face of the shoe (Beynnon, Renstrom, Alosa, Baumhauer, & Vacek, 2001; Norkin & White, 2009).

The size of the soccer shoe models was selected for each athlete according to the respective foot size ensuring the criteria of a distance of 0.5 cm between the longest toe and the front of the soccer shoe. It is also important to mention that soccer shoes were new for each participant.

2.3.1.2. Data collection. Cavanagh's index of each participant was calculated through plant pressure values, obtained during upright standing over 60 s. Participants were asked to remain as motionless as possible while looking at a visual reference, located at the eye-line level and at a distance of 3 m (Putti, Arnold, Cochrane, & Abboud, 2008).

After this task, the participant was asked to perform a brief warm-up of the lower limbs for 10 minutes in the cyclo-ergometer with 2% of the body weight, followed by self-directed stretching exercises (Brown, Bowser, & Simpson, 2012). Then, participants were informed that they should perform a three series of five consecutive lateral jumps with dominant foot, at a cadence of 120 beats per minute (controlled by metronome) while wearing one of the three models of cleats provided by ADIDAS (*Turf*, *Hard ground* and *Firm ground*). A two-minute resting period was set between each series (Caffrey, Docherty, Schrader, & Klossner, 2009; Docherty, Arnold, Gansneder, Hurwitz, & Gieck, 2005). The cadence adopted was based on the maximum cadence executed by healthy individuals in this kind of functional tests (Caffrey et al., 2009; Docherty et al., 2005). All participants were required to reach a minimum horizontal distance of 30 cm in their jumps, with a jump being considered the trajectory both forth and back. A trial was considered valid when the subject reached this distance in each jump with the defined cadence. Participants carried out a series of trials for familiarization with the task, in an attempt to memorize the execution speed and minimizing the effects of the learning process. To diminish the order effect, the sequence of the cleats used to carry out the protocol was previously randomized. The functional test adopted was selected with the intention of serving as a more demanding alternative than the single-footed static positioning tests, widely used in the posture control assessment (Delahunt, 2007) and was adapted from the *Side Hop Test*, previously used in the detection of functional deficits amongst individuals with and without ankle instability (Caffrey et al., 2009; Docherty et al., 2005). Figure 2 shows the sketch of the functional test, highlighting the individual's initial position according to the foot used for jumping.

2.3.2. *Moment one (M1)*

2.3.2.1. Application of the fatigue protocol. After a five-minute rest interval, all participants were subject to an evertor-oriented concentric/eccentric fatigue protocol in the isokinetic dynamometer, as they are the main ankle dynamic stabilizers in the LAS mechanism (Jackson et al., 2009; South & George, 2007). The eccentric component was included due to a higher production of force and because is more representative of the role of evertors muscles in ankle mediolateral stabilization (Gutierrez, Jackson, Dorr, Margiotta, & Kaminski, 2007; Sandrey & Kent, 2008).

The application of the fatigue protocol was carried out with individuals in a seated position, with their thigh, knee and ankle at 90°, 35° and 80°, respectively. The distal one-third of the thigh and the foot remained supported and stabilized by Velcro straps (Bisson et al., 2011). After 3–5 sub-maximal repetitions (that served as a warm-up and as a means of familiarization with the device), the *peak torque* was identified through three maximum contractions at 120°/s, in a movement amplitude of 30° (20° of inversion and 10° of eversion). It was determined that the individual reached a state of fatigue when he executed, consecutively, three eccentric contractions below 50% of the *peak torque* (Gutierrez et al., 2007).

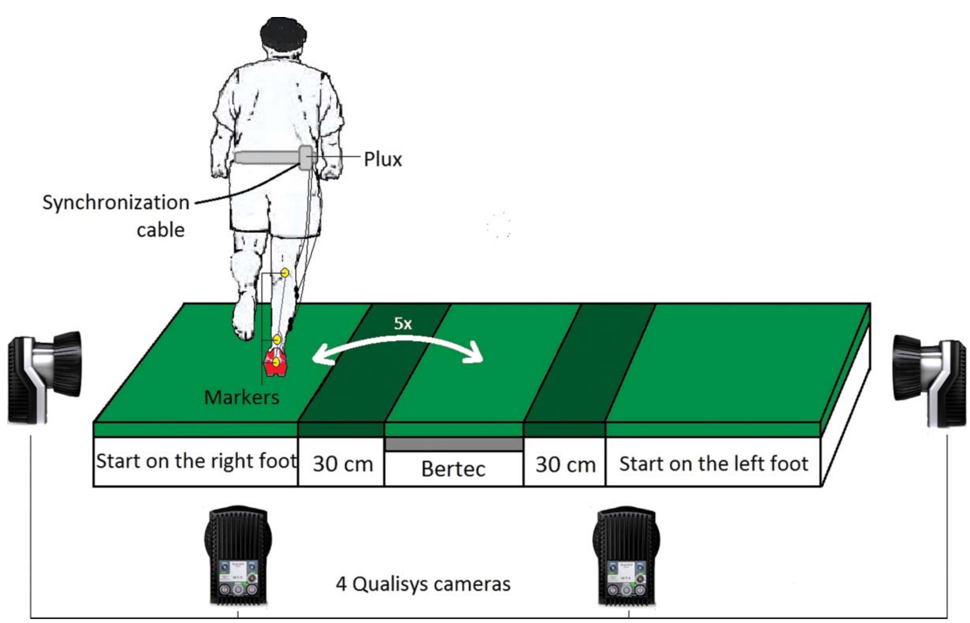

Figure 2. Experimental set up.

To prevent evertors fatigue recovery, participants performed the jump sequence immediately after the application of the fatigue protocol (40 s, in average). Also, between each model, the participant carried out, once more, the fatigue protocol.

2.3.4. *Data processing*

All variables were analysed during the three middle foot contact periods and the average values were used for analysis. The signal coming from the force platform was filtered through a low pass fourth-order *Butterworth* filter of 15 Hz and normalized to the body weight. The initial contact with the ground was defined as the moment where the value of the vertical component of the GRF was greater than 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, and it represents the relation between the GRF maximum and the time needed to reach it. The medio-lateral (COPx) and anteroposterior (COPy) displacements of the COP were obtained by calculating the difference between the maximum and minimum value of the COP. It was also calculated the medio-lateral (V_COPx) and antero-posterior (V_COPy) average speeds for the displacement of the COP, by dividing the displacement by the time interval (Duarte & Freitas, 2010).

A second-order *Butterworth* low-pass filter of 6 Hz was applied to cinematic data. The ankle eversion/inversion range of movement was obtained through the angle between the 'leg' segment and the 'hind foot' segment. In this analysis, the amplitude variation between the maximum eversion and inversion angles during the supporting period was used (Whatman, Hume, & Hing, 2012).

The EMG signals were filtered using a zero-lag, second-order Butterworth filter with an effective band pass of 10–500 Hz. The root mean square was calculated using a moving average of 20 samples (Schmid, Moffat, & Gutierrez, 2010). The temporal analysis was made in relation to the supporting moment (T0), being defined for each muscle as the time when a value equal to or higher than 5% of maximum obtained in each trial was observed, for at least 30 ms (Hodges & Bui, 1996; Nieuwenhuijzen, Gruneberg, & Duysens, 2002). The start of the muscular activation was searched in a temporal window starting at −250 ms in relation to T0 (Shiratori & Latash, 2001). Each of the previously described variables was analysed before and after the fatigue protocol.

2.5. *Statistics*

PASW® Statistics 20 (*Predictive Analytics Software*) software was used with a significance level of 0.05. The mean and median were used as measures of the central tendency, and the standard and interquartile deviations as dispersion measures (Marôco, 2010).

ANOVA repeated measures and Friedman tests were used to compare the different models of cleats without and with fatigue (Marôco, 2010). *T*-test for paired samples or the *Wilcoxon* test was used to compare the same cleats between conditions (non-fatigue and fatigue) (Marôco, 2010). Power analysis indicated that for the majority of variables analysed, 1-β was higher than 60% (Dupont & Plummer Jr, 1990).

3. Results

3.1. *Comparison of kinetic, kinematic and neuromuscular variables between cleats, with and without fatigue of the main evertors*

No statistically significant differences were found in kinetic, kinematic and neuromuscular factors between the different cleats with and without fatigue of evertors muscle (Table 1).

3.2. *Comparison of kinetic, kinematic and neuromuscular aspects obtained with each cleat model between non-fatigue and fatigue conditions*

When comparing the obtained parameters in each model of cleats in the two conditions evaluated (with and without fatigue of the peroneal muscles), no statistically significant differences were found in most analysed variables. The only variable showing statistically significant differences ($T = 2.074$; $P = 0.05$) was the activation time of the short peroneal (TA_CP) in the *Hard ground* model. In this model, the activation time was earlier in the condition with fatigue (-0.102 ± 0.093 s), when compared to the condition without fatigue (-0.079 ± 0.095 s). The results are summarized in Table 1.

4. Discussion

Although the main concern of the football player is related to performance over safety and comfort (Hennig, 2011; Lake, 2000), it is absolutely relevant to carry out comparative studies between cleats using variables that could be related to injury risk. It should be noted that despite in the present study the selected variables have been related to ankle sprain prediction, evidence supporting this assumption is lacking. Future studies are demanded to support this assumption. The high costs associated with the treatment of LAS, the risk of subsequent instability (Morrison & Kaminski, 2007) and the reduced volume of bibliography available on the relation between traction imposed by footwear and the LAS (Hennig, 2011) support, thus, this line of investigation focused on health promotion.

Table 1. Comparison of cleats and conditions without and with fatigue.

		Average ± SD		Comparison of cleats without fatigue		Comparison of cleats with fatigue		Comparison of conditions without and with fatigue	
Variable	Cleats	Without fatigue	With fatigue	Test value	Sample value	Test value	Sample value	Test value	Sample value
AEIROM (degrees)	T	8.90 ± 1.95	9.49 ± 2.27	$F = 1.038$	0.366	$X^2 = 1.714$	0.424	$T = -0.853$	0.409
	HG	8.50 ± 1.44	8.99 ± 3.31					$Z = -0.220$	0.826
	FG	8.16 ± 1.63	8.66 ± 2.62					$T = -1.314$	0.211
LRVz (N/s)	T	14.10 ± 3.91	13.53 ± 1.67	$X^2 = 0.348$	0.840	$F = 1.716$	0.193	$Z = -1.373$	0.170
	HG	14.15 ± 3.17	12.84 ± 1.86					$Z = -1.060$	0.289
	FG	13.37 ± 2.36	12.73 ± 1.90					$Z = 0.675$	0.507
LRLx (N/s)	T	2.74 ± 0.78	2.69 ± 0.76	$F = 0.514$	0.559	$X^2 = 2.000$	0.368	$Z = -0.698$	0.485
	HG	2.73 ± 0.75	2.43 ± 0.56					$T = 1.605$	0.123
	FG	2.63 ± 0.81	2.39 ± 0.61					$T = 1.273$	0.217
COPx (mm)	T	82.65 ± 30.65	75.12 ± 29.52	$X^2 = 1.778$	0.411	$X^2 = 0.471$	0.790	$T = 0.387$	0.704
	HG	68.58 ± 28.28	75.18 ± 20.72					$Z = -1.207$	0.227
	FG	68.87 ± 21.79	74.65 ± 30.25					$Z = -0.213$	0.831
COPy (mm)	T	100.86 ± 26.70	103.31 ± 20.43	$F = 0.530$	0.593	$F = 1.302$	0.283	$T = -0.445$	0.661
	HG	102.04 ± 26.55	99.05 ± 25.34					$T = 0.413$	0.684
	FG	94.31 ± 14.39	94.93 ± 27.23					$T = -0.097$	0.924
V_COPx (mm/s)	T	276.50 ± 98.81	258.67 ± 102.86	$X^2 = 0.111$	0.946	$X^2 = 0.471$	0.790	$T = 0.134$	0.895
	HG	240.36 ± 97.26	252.61 ± 64.78					$Z = -0.923$	0.356
	FG	254.16 ± 106.53	243.24 ± 96.06					$Z = -0.065$	0.948
V_COPy (mm/s)	T	336.14 ± 76.17	356.44 ± 74.42	$F = 0.501$	0.609	$F = 4.228$	0.022* *post hoc* 0.079	$T = -1.036$	0.312
	HG	349.4 ± 83.33	328.71 ± 75.13					$T = 0.975$	0.341
	FG	332.33 ± 101.17	305.86 ± 87.78					$T = 1.076$	0.295
AT_LP (s)	T	−0.0551 ± 0.07144	−0.0512 ± 0.06637	$X^2 = 1.000$	0.607	$F = 1.429$	0.250	$T = -0.550$	0.588
	HG	−0.0494 ± 0.07106	−0.0361 ± 0.08409					$Z = -1.460$	0.144
	FG	−0.0496± 0.07229	−0.0550 ± 0.06686					$T = 0.158$	0.876
AT_CP (s)	T	−0,0991 ± 0.08182	−0.1041 ± 0.05551	$F = 1.172$	0.319	$F = 0.278$	0.758	$T = 0.486$	0.632
	HG	**−0.0791 ± 0.09545**	**−0.1024 ± 0.09277**					$\boldsymbol{T = 2.074}$	**0.050**
	FG	−0.1067 ± 0.08805	−0.1154 ± 0.08555					$T = 0.237$	0.815

Note: (SD – standard deviation); (T – turf); (HG – hard ground); (FG – firm ground) (AEIROM – ankle eversion/inversion range of movement); (LRVz – loading rate of the vertical force); (LRLx – loading rate of the lateral force); (COPx – lateral displacement of the COP); (COPy – rearward displacement of the COP); (V_COPx – lateral displacement speed of the COP); (V_COPy – rearward displacement speed of the COP); (AT_LP – activation time of the long peroneal); (AT_CP – activation time of the short peroneal). Bold values represent significant statistical differences.

4.1. Comparison of kinetic, kinematic and neuromuscular aspects between cleats, without fatigue of the main evertors

No significant differences were found in any of variables analysed in a non-fatigue condition. This result contradicts the conclusions of previous studies supporting the relation between higher traction indexes (inherent to higher studs) and a higher risk of injury (Lake, 2000; Queen et al., 2008; Sterzing, Müller, Hennig, & Milani, 2009). The fact that increased mechanical traction does not always lead to high biomechanical forces during direction change movements is a possible justification. Sterzing, Müller, Schwanitz, Odenwald, and Milani (2008) assessed the traction indexes of four models of cleats through a mechanical instrument (mechanical traction) and had recorded differences between them. However, when comparing different models of cleats in real conditions, those differences disappeared, showing the extreme need of considering the human structure biology during the analysis of traction properties (biomechanical forces) (Sterzing e al., 2008). Another justification may be related to the conclusions of a study that compared 10 models of cleats in four different types of ground (two of which were synthetic turf), through a mechanical instrument that simulated direction change movements, describing a greater influence of the ground in the traction properties rather than of the models of cleats (Villwock, Meyer, Powell, Fouty, & Haut, 2009).

The biomechanical (ankle inversion/eversion range of movement, loading rate of the GRF, speed and displacement of the COP) and neuromuscular (activation time of the peroneals) variables show very similar values between cleats. These results clearly decrease the relevance of the phenomenon of neuromotor adaptation to the cleats and its properties in healthy subjects. Otherwise the most used model (*Firm ground* $n = 15$; 62.5%) could have different values in the variables evaluated, because this phenomenon allows players to adjust their movements to the traction properties of the shoe to ground interface conditions (Hennig, 2011).

In healthy individuals, the amplitude of eversion and inversion varies between 5° and 10°, and 25° and 30°, respectively (Dubin et al., 2011). In this study, the average variations remain between 8.16° and 8.90°, which may suggest a good dynamic control regarding this demanding task. Although the foot was stabilized in the shoe, it is possible that this limited range may be due to the markers' placement on the shoe and therefore less likely to measure a change. In the future, the study of the type of sole support through plantar pressure insoles should be associated to this one, to allow for the description of whether the recorded amplitude occurred with greater or lesser support of the side of the foot – a possible injury risk factor.

In direction change manoeuvers, the combination of greater inversion amplitudes and high lateral GRF has been described as a potential injury source (Dayakidis & Boudolos, 2006). In the ground reception after the vertical jump, the increasingly fast growth registry of the vertical forces seems to be a neuromuscular response making the ankle more stable avoiding excessive inversion forces (Dayakidis & Boudolos, 2006). Additionally, the vertical and lateral loading rate is also a comfort indicator inherent to the imposed loads on the articular surfaces, expressing the tissue's ability to accommodate the load (Puddle & Maulder, 2013; Smith, Dyson, & Janaway, 2004). The fact that no significant differences between the cleats were recorded may indicate that the cleats are very similar. At first glance, comparing the rubber sole of the *Turf* with the plastic one and high studs of the *Hard* and *Firm ground*, one could expect that there were differences. Such a hypothesis would be consistent with the results obtained in a previous study that compared a *Turf* model with a *Soft ground* one (plastic sole and only six high aluminium studs), during a run at two velocities (5.4 and 4.4m/s), carried out by two male football players (Smith et al., 2004). This study showed that the second model imposed higher GRF than those imposed by the first one (Smith et al., 2004). The explanation for the lack of differences in the present study may be related the fact that the grass was in optimal conditions, which allowed for a similar absorbing of the load between models. For optimal conditions, we consider the vertical filaments without the rubber being compacted, allowing for a complete penetration of the higher studs (FIFA, 2009, 2012).

COP displacement-related variable, widely used as postural control indicators (Palmieri, Ingersoll, Stone, & Krause, 2002) in LAS studies (Morrison & Kaminski, 2007), shows the proximal and distal body adjustments for the task. In most studies, this variable was evaluated in single-footed static positioning (Delahunt, 2007; Richie, 2001) or during gait (Willems, Witvrouw, Delbaere, De Cock, & De Clercq, 2005). Globally, the findings obtained in these studies demonstrate increased COP oscillations in individuals with functional ankle instability compared with subjects without ankle instability. As such, one would expect that, if a cleats model was more unstable (for example, for having excessively high studs, which would not fully penetrate in the synthetic turf), it would end up in significant differences between models, which was not the case. The justification for this phenomenon may be related to the turf conditions. In the present study, the grass was not subject to irrigation and being dry, conferred to the *Turf* model a more similar traction/stability in relation to the other models. In fact, in more realistic conditions (wet grass) the stability provided by this model (with lowest studs) would possibly be different from the others. In another perspective, if the infill of the synthetic turf was compacted due to an excessive usage, it may not allow the penetration of the studs in the *Hard* and *Firm ground* models and, thus, generate a possible difference between models. However, the penetration of the studs, in this case, does not appear to pose any difficulties. It should also be considered that cleats impose different stability levels and the athlete may have compensated those differences through proximal body strategies. However, future studies are required to confirm our hypothesis. Finally, the divergent characteristics when it comes to the sole and vamp may not have been sufficient to induce different levels of stability during the chosen functional test and also explain the obtained results.

In the Delahunt (2007) revision, several studies advocate that the muscular strength of the evertors does not seem compromised while comparing individuals with and without functional instability. More important than evertor's strength, the study of the evertors' muscular reflexes seems to be more important as an injury contributor variable (Delahunt, 2007; Rosenbaum, Becker, Gerngro, & Claes, 2000). Konradsen, Voigt, and Højsgaard (1997) advocate that the mioelectric activity has a delay that can go up to 90 ms due to neural latencies, also associated to an additional 90 ms delay required for the production of half of its maximum force (Konradsen et al., 1997). On the other hand, in just 100 ms, the lateral ligament system may be in injury risk, for which it is accepted a muscular pre-activation in dynamic activities, such as jumping (Richie, 2001). In the present study, muscular pre-activation values (long and short peroneal) were recorded with all the cleats, with no statistically significant differences being recorded among them. This fact may be explained considering the results obtained in the variables previously studied, which show an identical degree of instability between the cleats. As such, these will have produced a similar stimulation in the muscular receptors, culminating in an equivalent pre-activation with the different models.

4.2. Comparison of kinetic, kinematic and neuromuscular aspects between cleats, with fatigue of the main evertors

It is currently accepted that in a fatigue condition the muscle's mechanical properties and proprioceptive system, necessary for postural control, are affected (Bisson et al., 2011; Sandrey & Kent, 2008). When comparing fatigued muscles with non-fatigued ones, the first reveals a decreased production of force and an increase in muscular latency (Hiemstra et al., 2001). Of all the afferent information from the mechanical receptors of the skin, joints and muscles, it seems to be a higher relative contribution of the one coming from the muscle spindles (Konradsen, Ravn, & Sorensen, 1993). That premise is supported by the fact that anaesthesia in the ligaments and joint capsule does not appear to induce alterations in the activation time of the peroneals (Konradsen et al., 1993). As such, one would expect that eventual differences between the cleats in the zero moment were identified/amplified by the induction of fatigue. However, the fatigued condition did not reveal statistically significant changes between the cleats in any of the studied variables. One of the possible explanations is the fact that the isolated fatigue of the peroneal muscles (main dynamic stabilizers that oppose the inversion mechanism) was imposed, with some studies advocating that some muscular groups of proximal action (thigh or knee) appear to have more influence in the posture control than others of a more distal action (ankle) (Bisson et al., 2011). On the other hand, inducing fatigue only in the peroneal muscles may have been insufficient to find differences, as the isolated fatigue of a muscular group of the ankle appears to have a lesser impact than the simultaneous fatiguing of several muscular groups in that articulation (Boyas et al., 2011). Future studies using more global fatigue protocols are required. Furthermore, it would be important to explore the use of multidirectional jumps – closer to reality game (Caffrey et al., 2009). Finally, because in the present study the *post hoc* analysis indicated that the ideal 1-β values (80%) were not achieved (Dupont & Plummer Jr, 1990), future studies with a higher sample are required to confirm our results.

5. Conclusion

The findings obtained in the present study indicate that in healthy athletes LAS contributors variables are not influenced by different models of soccer shoe. This conclusion is inevitably associated to the particularities of the study, that is, to the fact that the assessment was made through a dynamic functional test adapted from the *Side Hop Test*, performed by healthy athletes and on a third generation synthetic turf.

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