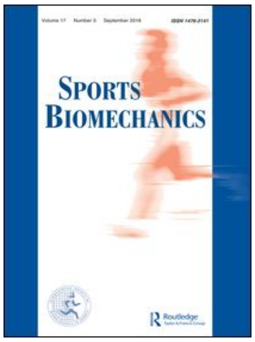

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Relationship between muscular extensibility, strength and stability and the transmission of impacts during fatigued running

Alberto Encarnación-Martínez [a], Roberto Sanchis-Sanchis [a], Pedro Pérez-Soriano [a] and Antonio García-Gallart [b]

[a]Research Group in Sports Biomechanics, Department of Physical Education and Sports, University of Valencia, Valencia, Spain; [b]Department of Sports Sciences, Catholic University of Murcia, Murcia, Spain

ABSTRACT

The aim was to analyse the relationship between isokinetic strength, dynamic stability, muscular extensibility and impacts transmission during fatigued running. Low- and high-frequency impacts—related to body movements and the severity of impacts, respectively—were assessed in 17 male recreational runners, before and after a treadmill running fatigue protocol, using a triaxial accelerometry system. High-frequency impacts in the tibia were negatively correlated to the knee angle at which the quadriceps peak torque was reached ($p = 0.014$), and also to the extensibility of the hamstrings and soleus ($p = 0.001$ and $p = 0.023$, respectively). The increases of high-frequency impacts in tibia caused by fatigue were positively related to the knee angle at which the hamstrings peak torque was reached ($p = 0.001$) and to stability after landing ($p = 0.007$). The attenuation of high-frequency impacts was positively related to hamstrings/quadriceps ratio of strength ($p = 0.010$) and to stability ($p = 0.006$). Limiting possible deficits in hamstring and soleus range of motion, improving stability after landing, developing hamstring and quadriceps strength in elongated muscle range, and maintaining a balanced ratio of hamstring/quadriceps strength could help to reduce the injury risk in running.

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KEYWORDS
Accelerometry; fatigue; range of motion; postural control; isokinetic strength

Introduction

The number of runners has increased continuously in recent decades, and 40–50% of runners get injured yearly (Fields et al., 2010). It has been suggested that strength programmes could be a crucial aspect in order to reduce sports injuries to less than a third (Lauersen et al., 2014). Moreover, strengthening has been shown to be beneficial in the treatment of certain running injuries and could be favourable to avoid injuries during running (Fields et al., 2010). It has also been described that lower limb stability training could be beneficial for injury prevention (Lauersen et al., 2014), as other authors have suggested that postural control deficits could increase the risk of injury (Brazen et al., 2010; Gribble et al., 2012). In contrast to strength and stability training, although stretching has been considered as a good strategy to reduce running-related injuries (Baxter et al., 2017), it has not been shown to be a beneficial effect

CONTACT Alberto Encarnación-Martínez Alberto.Encarnacion@uv.es

on injury prevention in runners (Baxter et al., 2017; Fields et al., 2010; Lauersen et al., 2014). Nonetheless, joints' range of motion values outside the normal range (i.e., low or high muscle extensibility) could be harmful (Baxter et al., 2017). So, joint range of motion can have a major effect on maintaining a healthy running pattern.

During running, an impact is generated with each foot contact with the ground. This impact produces vertical forces from 1.5 to 2.5 times the body weight (Derrick et al., 2002). As a product of that impact with the ground a shock wave is generated, which is transmitted through the whole body (Encarnación-Martínez et al., 2018; Gruber et al., 2014; Mercer et al., 2003), where musculoskeletal structures, active movements and external implements are responsible for absorbing the impact (Mercer et al., 2003). When this rapid deceleration produced by the foot contact with the ground is measured on the tibia, it is commonly referred to as tibial acceleration (Encarnación-Martínez et al., 2018).

Biomechanical changes associated with fatigue can increase injury risk during running (Hreljac, 2004). It has been shown that fatigued running can affect lower limb kinematics (Jewell et al., 2017). Significant increases in rearfoot eversion, knee adduction and internal rotation peak angles, and hip internal rotation are some of that changes observed during fatigued running (Benson & O'Connor, 2015). Moreover, kinematic changes produced during fatigued running can increase the stress and impact forces received by lower extremities (Milgrom et al., 2007). Sample entropy has been shown to be reduced in runners with medial tibial stress syndrome in fatigued running, meanwhile shock attenuation increased (Schütte et al., 2018). Furthermore, 2 days after a marathon, fatigue could be detected by a higher self-reported pain and higher peak mediolateral acceleration. (Clermont et al., 2019). It has also been shown that greater muscular activation can reduce these impacts (Potthast et al., 2010), and that poor stability after landing could increase the vertical forces received (Brazen et al., 2010). However, there are no studies that analyse these factors during fatigued running and how they could affect impact transmission.

Tibial accelerations have been traditionally evaluated in the time domain (García-Pérez et al., 2014; Lucas-Cuevas et al., 2015; Mizrahi, Verbitsky, Isakov et al., 2000). However, the analysis in the frequency domain allows for the analysis of the frequency contents of the tibial and head acceleration signals. Vertical acceleration of the centre of mass (COM) and voluntary motion is contained in the low-frequency band (3–8 Hz), while rapid deceleration related to foot contact with the ground, and with the severity of the impact, is contained in the high-frequency band (9–20 Hz) (Gruber et al., 2014; Shorten & Winslow, 1992). Unlike time domain analysis, frequency domain analysis allows us to directly determine the attenuation of the impact in the human body. Knowing that impact attenuation ability could be decreased during fatigued running and that less attenuation has been related to an increased injury risk, frequency domain analysis to assess impact attenuation might provide useful information to help understand running injury risks (Derrick et al., 2002; Lucas-Cuevas et al., 2015; Mizrahi, Verbitsky, Isakov et al., 2000).

Therefore, the aim of this study was to analyse the relationship between isokinetic strength, dynamic postural control, and muscular extensibility with impact transmission during fatigued running. Based on the existing literature, we hypothesised that (a) lower quadriceps and hamstrings isokinetic strength values would be related to a greater power in the high-frequency tibial acceleration range; (b) lower hamstrings muscle range of motion values would not be related to a greater power in the high-frequency tibial acceleration range; (c) lower dynamic stability would be related to a greater power in the high-frequency acceleration range at tibial location.

Methods

Participants

Seventeen male recreational runners (age of 28.7 ± 8.3 years, height of 1.78 ± 0.07 m, body weight of 72.2 ± 8.2 kg), with a running experience of 7.6 ± 5.1 years and experienced in treadmill running, participated in this study. The inclusion criteria were: not competitive (i.e., recreational), ran a minimum of twice a week and more than 20 km/week in the last year, and had no injuries in the previous 6 months. Informed consent was provided to all participants before inclusion in the study, which was approved by the Catholic University of Murcia Ethics Committee (registry number: 6775).

Experimental protocol

Participants were evaluated over 2 days (Figure 1). On the first day, maximal aerobic speed was calculated using a maximal effort 5 min running test on a 400-m track (Berthon et al., 1997; García-Pérez et al., 2014; Lucas-Cuevas et al., 2015). On the second day, isokinetic strength, dynamic postural control, and muscular extensibility were assessed before the fatigue protocol. Muscle extensibility was evaluated before the warm-up exercises (López-Miñarro & Rodríguez-García, 2010). Once finished, the familiarisation with the treadmill (Excite®+ Run MD Inclusive, Technogym Trading S.A., Cesena, Italy) and a self-determined 10-min warm-up were carried out (García-Pérez et al., 2014; Lucas-Cuevas et al., 2015). Finally, postural control in landing and reaching tasks, and quadriceps and hamstrings isokinetic strength were assessed in a random order in the dominant limb.

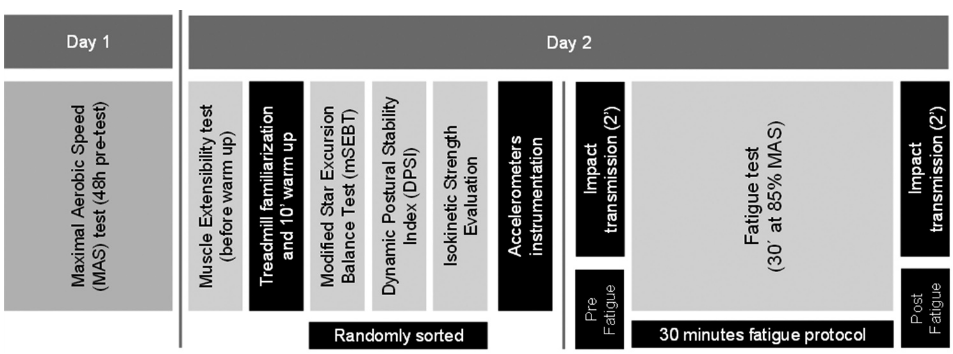

Figure 1. Experimental protocol flow diagram.

Impact transmission was evaluated before and after a treadmill running fatigue protocol. Running accelerations were recorded in a 2-min treadmill running period at

3.89 m/s (Lussiana et al., 2017) and 0% slope through 3 series of 10 s using a wireless triaxial accelerometry system (Blautic®, Valencia, Spain; sampling frequency 300 Hz, range ± 16 G, mass 2.5 g). Accelerations were analysed from the acceleration-signal data of the vertical axis. One accelerometer was placed on the distal and anteromedial portion of the tibia (Lucas-Cuevas et al., 2017) on the dominant leg, and the second one on the participant's forehead (Encarnación-Martínez et al., 2018; García-Pérez et al., 2014; Gruber et al., 2014; Mercer et al., 2003). Skin was prepared and accelerometers were fixed according to Encarnación-Martínez et al. (2018) recommendations.

Treadmill running fatigue protocol

Participants were instructed to run for 30 min on a treadmill (0% slope) at 85% of maximal aerobic speed (García-Pérez et al., 2014), which was calculated on the first day from the 5-min running field test (Lucas-Cuevas et al., 2014). Also, according to Hafer et al. (2017), a minimum rating of 17/20 ('Very Hard') of perceived exertion on the Borg´s Scale 6–20 (Borg, 1982) was also set.

Muscular extensibility evaluation

A manual goniometer and inclinometer were used to evaluate the range of motion of the hip flexor musculature, quadriceps, hamstrings, gastrocnemius (ankle dorsiflexion with extended knee) and soleus (ankle dorsiflexion with flexed knee) (Figure 2). In order to reduce the measurement variability, all the measures were taken by the same researcher. The modified Thomas test, proposed by Wakefield et al. (2015) (Figure 2(a)), was used to measure the hip flexor musculature extensibility following the recommendations to standardise the pelvic tilt and lumbar lordotic curve. In the same position, quadriceps extensibility was evaluated by measuring knee flexion angle (Cejudo et al., 2015) (Figure 2(b)).

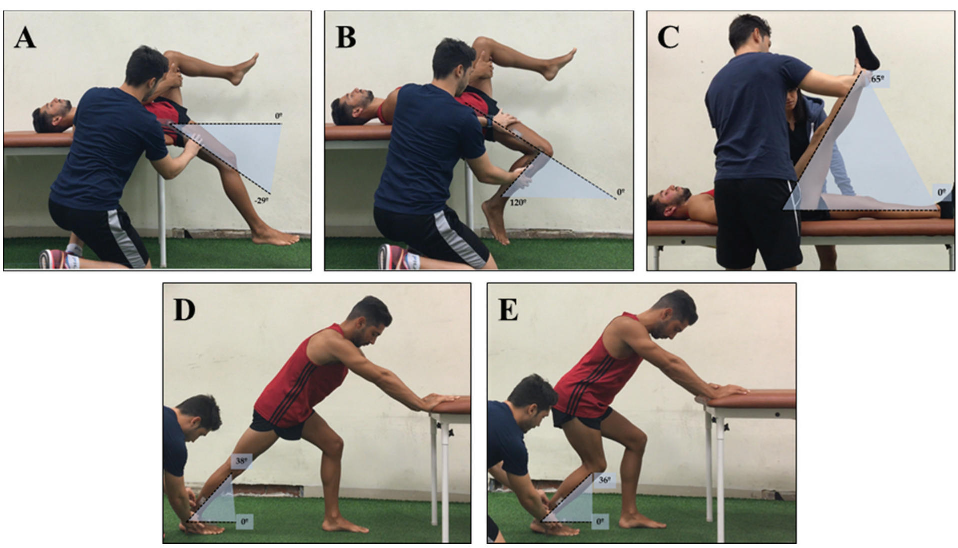

Figure 2. Hip flexor musculature (A), quadriceps (B), hamstrings (C), gastrocnemius (D) and soleus (E) extensibility measures.

To evaluate hamstrings extensibility, the passive straight leg raise test was used as described by López-Miñarro and Rodríguez-García (2010) (Figure 2(c)). The gastrocnemius muscle extensibility was evaluated in standing position with their feet parallel hip-width apart. Participants took a step forward, resting their hands on the wall or stretcher, and performing a dorsiflexion of the ankle slowly and progressively until the maximum sensation of stretching was reached or compensatory movements were detected (Cejudo et al., 2015) (Figure 2 (d)). Finally, an ankle dorsiflexion and knee flexion were performed slowly and progressively in this position to record the soleus extensibility (Phillips, 2007) (Figure 2(e)).

Dynamic postural control evaluation

The Modified Star Excursion Balance Test (mSEBT) and Dynamic Postural Stability Index (DPSI) were used to evaluate dynamic postural control in reach and landing tasks, respectively. According to Gribble et al. (2012), mSEBT evaluates the reach in anterior, posterolateral and posteromedial directions (Figure 3). Before the assessment, four familiarisation attempts in each direction were performed (Doherty et al., 2015; Van Lieshout et al., 2016). Three randomised attempts in each direction were then recorded (Doherty et al., 2015). The average reach in anterior, posterolateral and posteromedial directions and the summation of the three were saved. The reached distance was normalised to the limb length as previously reported (Whyte et al., 2015).

Figure 3. Anterior (A), posterolateral (B) and posteromedial (C) directions of Modified Star Excursion Balance Test (mSEBT) and Modified Dynamic Postural Stability Index (DPSI) test.

Dynamic postural control after landing was evaluated using an adaptation of the DPSI test proposed by Wikstrom et al. (2005) and Ross et al. (2005). Participants were placed 0.7 m from the centre of a force platform (Kistler 9286 BA, Kistler Group, Winterthur, Switzerland), and they were instructed to double-leg jump over an elastic band (set at 50% of their maximum jump height) with hands on hips and looking to the front, landing on their dominant limb, and stabilising as quickly as possible. On landing, participants remained in a single-leg stance for 20 s with the first 3 s from impact being used for further analysis (Wikstrom et al., 2010). This impact was identified as the instant when vertical ground reaction forces (VGRFs) exceeded 10 N (Meardon et al., 2016) (Figure 3).

Fifty percent of runners maximum jump height was calculated from the highest jump of three valid countermovement jumps (Wikstrom et al., 2010). A minimum of three practice attempts were allowed (Williams et al., 2016), and three attempts were performed to evaluate the mediolateral, anteroposterior, vertical, and global stability indices, understood as the dispersion of forces from the centre of pressure in each of the axes. Ground reaction force signals were recorded at a frequency of 1000 Hz using the formulas proposed by Wikstrom et al. (2010). It is important to note that these dynamic postural stability variables do not have specific units because they are dimensionless. Thus, higher values (i.e., 0.500) indicate worse stability and lower values (i.e., 0.200) indicate better stability.

Quadriceps and hamstrings isokinetic strength evaluation

Peak concentric torque values were recorded in the quadriceps and hamstring muscles of the dominant lower extremity using an isokinetic dynamometer (Biodex System Pro 3™, Biodex Medical Systems, Inc., New York, USA). Testing was performed in a seated position with a hip flexion angle of 85º; the trunk, waist and thigh were stabilised with straps (Kellis et al., 2011; Soleimanifar et al., 2012). The fulcrum of the dynamometer was aligned with the lateral femoral epicondyle, and the shin pad was placed 0.02 m above the medial malleoli.

Peak torque values were evaluated by performing two sets of concentric/concentric knee flexion-extension movements at 120º/s, in which the motion ranged from 0º (full extension) to 90º of knee flexion (Kellis et al., 2011; Soleimanifar et al., 2012). In the first set, three submaximal and three maximal contractions were performed as familiarisation (Soleimanifar et al., 2012). Quadriceps and hamstrings torque were evaluated in the second set, performing three repetitions of maximal effort and recording the highest value (Kellis et al., 2011; Soleimanifar et al., 2012). The angle at which the quadriceps and hamstrings concentric peak torque was reached and the hamstrings/quadriceps strength ratio were evaluated.

Data processing

A custom routine performed with MatLab R2013b program (Mathworks Inc, Natick, MA, USA) was used to analyse the acceleration data. To calculate power spectral density (PSD) and convert the unfiltered time domain data to frequency domain data the process used by Gruber et al. (2014) was followed. So, in the low (3–8 Hz) and high (9–20 Hz) frequency range, the maximum and total signal power magnitude in tibia and head were evaluated, as well as shock attenuation (Gruber et al., 2014) (Figure 4). The following equation was used to evaluate the shock attenuation (Derrick et al., 2002; Gruber et al., 2014; Mizrahi, Verbitsky, and Isakov 2000; Shorten & Winslow, 1992):

$$Shock\ Attenuation\ =\ 10\ x\ log10\ (PSD_{head}/\ PSD_{tibia}) \tag{1}$$

Finally, delta (Δ), or the variation between pre-fatigue and post-fatigue values, was calculated in the running impacts variables.

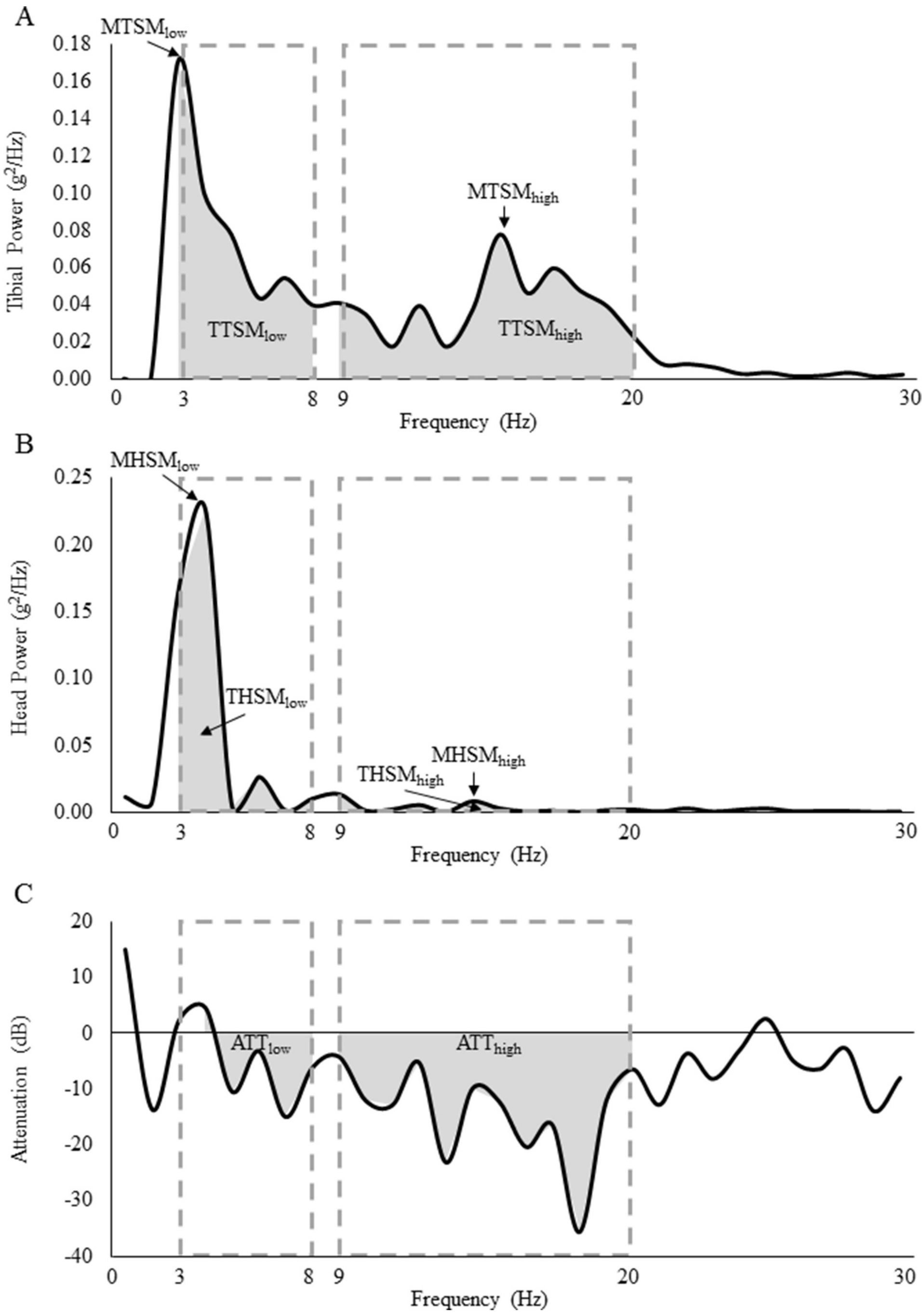

Figure 4. Power spectra of the tibia (A) and head (B) acceleration and shock attenuation (C) in the low (3–8 Hz) and high (9–20 Hz) frequency ranges of the frequency domain analysis. Low: Low-frequency range, High: High-frequency range, MTSM: Maximum Tibial Signal Magnitude, TTSM: Total Tibial Signal Magnitude, MHSM: Maximum Head Signal Magnitude, THSM: Total Head Signal Magnitude, ATT: Shock Attenuation. Inspired by figure of Gruber et al. (2014).

Statistical analysis

Statistical analysis was performed using SPSS 19.0 (IBM Armonk, New York, USA). Descriptive statistics were described as mean ± standard deviation (SD). Normality and homoscedasticity were checked with the Shapiro–Wilk test and Levene Test, respectively. As inferential analysis, impact transmission characteristics between pre- and post-fatigue conditions were compared using a one-way repeated measure analysis of variance or the non-parametric alternative (Friedman Test). Besides, the effect size (ES) was assessed using Cohen's d (Cohen, 1992) through the formula proposed by Hunter and Schmidt (2004), and was interpreted as 0.0–0.2 = very small, 0.2–0.5 = small, 0.5–0.8 = medium, 0.8–1.2 = large, 1.2–2.0 = very large, and >2.0 = huge (Sawilowsky, 2009). To estimate the a priori sample size and desirable effect size for the primary outcome of each condition, G*Power 3.1.2 software was used (Erdfelder et al., 1996). Significance was defined as $p < 0.05$.

Finally, a correlational analysis through Pearson's Correlation Coefficient was made to evaluate the relationship between research factors and post-fatigue impact characteristics. The magnitude was interpreted as <0.1 = trivial, 0.1–0.29 = small, 0.3–0.49 = moderate, 0.5–0.69 = large, 0.7–0.89 = very large, ≥0.9 = extremely large (Hopkins et al., 2009). Moreover, the coefficient of determination (r^2), or the percentage of the variance in the dependent variable that can be explained by variations in independent variables, was calculated by squaring r and multiplying it by 100 (Congelosi et al., 1983).

Results

All participants (N = 17) completed the 30-min treadmill running fatigue protocol. Average speed during the fatigue protocol was 4.2 ± 0.3 m/s, and perceived effort was 17.6 ± 0.5 on the Borg's Scale. Descriptive results of extensibility, dynamic postural control and isokinetic concentric strength are presented in Table 1. Runners demonstrated an increase in the maximum and total high-frequency tibial acceleration signal power magnitude after the treadmill running fatigue protocol (Table 2).

Table 1. Descriptive results of extensibility, dynamic postural control and isokinetic concentric strength.

	Mean	SD	Min	Max	Range
Hip flexor musulature $_{ROM}$ (°)	−22.8	7.1	−38.0	−12.0	26.0
Quadriceps $_{ROM}$ (°)	118.6	8.7	102.0	136.0	34.0
Hamstrings $_{ROM}$ (°)	68.8	5.7	60.0	81.0	21.0
Gastrocnemius $_{ROM}$ (°)	34.1	7.6	22.0	50.0	28.0
Soleus $_{ROM}$ (°)	35.9	7.3	23.0	46.0	23.0
Anterior $_{mSEBT}$ (%LL)	74.0	6.1	65.3	87.6	22.4
Posterolateral $_{mSEBT}$ (%LL)	94.2	6.2	84.4	111.6	27.2
Posteromedial $_{mSEBT}$ (%LL)	96.8	5.9	83.9	110.1	26.2
Σ $_{mSEBT}$ (%LL)	264.9	14.3	240.0	309.3	69.3
Vertical Stability Index *	0.325	0.056	0.236	0.443	0.207
Mediolateral Stability Index *	0.114	0.010	0.093	0.133	0.040
Anteroposterior Stability Index *	0.031	0.005	0.023	0.042	0.019
Dynamic Postural Stability Index *	0.346	0.055	0.261	0.465	0.203
Quadriceps $_{TORQ}$ (Nm/BW)	245.28	39.60	175.66	313.08	137.42
Hamstrings $_{TORQ}$ (Nm/BW)	124.77	31.26	70.76	188.87	118.12
Quadriceps $_{ANG\text{-}TORQ}$ (°)	56.71	5.08	46.00	65.00	19.00
Hamstrings $_{ANG\text{-}TORQ}$ (°)	40.65	11.28	23.00	70.00	47.00
Hams./Quad. ratio (%)	50.6	8.2	33.8	61.4	27.6

SD: Standard Deviation, Min: Minimum, Max: Maximum, *: Dimensionless, ROM: Range of Movement, %LL: Percentage of Lower Limb, BW: Body Weight, TORQ: Peak Torque, ANG-TORQ: Peak Torque Angle, mSEBT: modified Star Excursion Balance Test, Σ: summation.

Table 2. Results of the frequency domain acceleration changes after fatigue.

Condition	Pre-Fatigue	Post-Fatigue			
Variable	Mean ± SD	Mean ± SD	Δ Mean ± SD	ES	95% CI
Maximum Head Signal Magnitude (g^2/Hz)	0.26 ± 0.02	0.25 ± 0.02	−0.01 ± 0.01	-	-
Maximum Tibial Signal Magnitude (g^2/Hz) §	0.18 ± 0.02	0.19 ± 0.02	0.01 ± 0.01	-	-
Total Head Signal Magnitude (g^2/Hz)	0.97 ± 0.07	0.93 ± 0.07	−0.03 ± 0.03	-	-
Total Tibial Signal Magnitude (g^2/Hz) §	2.15 ± 0.21	2.37 ± 0.21	0.01 ± 0.01	-	-
Maximum Head Signal Magnitude $_{low}$ (g^2/Hz)	0.18 ± 0.01	0.17 ± 0.01	−0.02 ± 0.01	-	-
Maximum Tibial Signal Magnitude $_{low}$ (g^2/Hz) §	0.13 ± 0.01	0.13 ± 0.01	0.00 ± 0.00	-	-
Total Head Signal Magnitude $_{low}$ (g^2/Hz)	1.75 ± 0.13	1.63 ± 0.14	−0.12 ± 0.09	-	-
Total Tibial Signal Magnitude $_{low}$ (g^2/Hz) §	4.14 ± 0.45	4.20 ± 0.37	0.06 ± 0.12	-	-
Maximum Head Signal Magnitude $_{high}$ (g^2/Hz) §	0.01 ± 0.00	0.01 ± 0.00	0.00 ± 0.00	-	-
Maximum Tibial Signal Magnitude $_{high}$ (g^2/Hz)	0.06 ± 0.01	0.08 ± 0.01**	0.02 ± 0.01	0.595	−0.027/-0.013
Total Head Signal Magnitude $_{high}$ (g^2/Hz) §	0.40 ± 0.04	0.41 ± 0.05	0.01 ± 0.05	-	-
Total Tibial Signal Magnitude $_{high}$ (g^2/Hz) §	3.56 ± 0.36	4.37 ± 0.47**	0.81 ± 0.32	0.488	−1.094/-0.526
Shock Attenuation $_{low}$ (dB) §	−54.73 ± 15.81	−59.25 ± 16.12*	7.61 ± 2.39	−0.283	−6.295/15.335
Shock Attenuation $_{high}$ (dB)	−128.40 ± 14.95	−147.41 ± 11.98**	22.21 ± 8.88	−1.396	9.833/28.187

ES: Effect Size, §: Non-parametric variables (Friedman´s Test), SD: Standard Deviation, Δ: Delta, low: lower frequency range, high: higher frequency range, *: $p < 0.05$ pre-fatigue vs post-fatigue, **: $p < 0.01$ pre-fatigue vs post-fatigue.

As shown in Table 3, the knee angle at which the quadriceps concentric peak torque was reached was negatively correlated to the maximum and total high-frequency tibial acceleration signal power magnitude ($r = -0.646$ and $r = -0.584$, respectively). In addition, the angle at which the hamstrings concentric peak toque was reached was positively correlated to the delta of the maximum high-frequency tibial acceleration signal power magnitude ($r = 0.713$), while the hamstrings/quadriceps strength ratio was positively correlated to the delta of high-frequency attenuation ($r = 0.606$).

Table 3. Relationship between isokinetic strength, muscular extensibility and dynamic stability with acceleration transmission modifications.

Research Factors	Variable	*r*	*P*	r^2	Intensity
Hamstrings $_{ROM}$	Total Tibial Signal Magnitude	−0.620	0.008**	38.4	Large
Hamstrings $_{ROM}$	Maximum Tibial Signal Magnitude $_{high}$	−0.565	0.018*	31.9	Large
Hamstrings $_{ROM}$	Total Tibial Signal Magnitude $_{high}$	−0.711	0.001**	50.5	Very Large
Soleus $_{ROM}$	Maximum Tibial Signal Magnitude $_{high}$	−0.584	0.014*	34.1	Large
Soleus $_{ROM}$	Total Tibial Signal Magnitude $_{high}$	−0.548	0.023*	30	Large
Posterolateral $_{mSEBT}$	Δ Total Tibial Signal Magnitude $_{low}$	−0.540	0.025*	29.2	Large
Posteromedial $_{mSEBT}$	Δ Total Tibial Signal Magnitude $_{low}$	−0.515	0.035*	26.5	Large
Vertical Stability Index	Maximum Tibial Signal Magnitude $_{low}$	0.496	0.043*	24.6	Moderate
Vertical Stability Index	Maximum Tibial Signal Magnitude $_{high}$	0.550	0.022*	30.3	Large
Vertical Stability Index	Δ Maximum Tibial Signal Magnitude	0.634	0.006**	40.2	Large
Dynamic Postural Stability Index	Maximum Tibial Signal Magnitude $_{low}$	0.497	0.042*	24.7	Moderate
Dynamic Postural Stability Index	Maximum Tibial Signal Magnitude $_{high}$	0.555	0.021*	30.8	Large
Dynamic Postural Stability Index	Δ Maximum Tibial Signal Magnitude	0.631	0.007**	40.2	Large
Quadriceps $_{ANG\text{-}TORQ}$	Maximum Tibial Signal Magnitude $_{high}$	−0.646	0.005**	41.7	Large
Quadriceps $_{ANG\text{-}TORQ}$	Total Tibial Signal Magnitude $_{high}$	−0.584	0.014*	34.1	Large
Hamstrings $_{ANG\text{-}TORQ}$	Maximum Tibial Signal Magnitude	−0.579	0.015*	33.5	Large
Hamstrings $_{ANG\text{-}TORQ}$	Maximum Tibial Signal Magnitude $_{low}$	−0.483	0.049*	23.3	Moderate
Hamstrings $_{ANG\text{-}TORQ}$	Δ Maximum Tibial Signal Magnitude $_{high}$	0.713	0.001**	50.8	Very Large
Hams./Quad. ratio	Δ Shock Attenuation $_{high}$	0.606	0.010**	36.7	Large

r^2: Coefficient of Determination. Δ: Delta changes = post—pre values, ROM: Range of Movement, ANG-TORQ: Peak Torque Angle, low: lower frequency range, high: higher frequency range, mSEBT: modified Star Excursion Balance, *: $p < 0.05$, **: $p < 0.01$.

Regarding muscular extensibility (Table 3), hamstrings extensibility was negatively correlated to the total power magnitude in all the tibial signal (r = −0.620), as well as the maximum and total high-frequency tibial acceleration signal power magnitude (r = −0.565 and r = −0.711, respectively). Moreover, the soleus extensibility was negatively correlated to the maximum and total high-frequency tibial acceleration signal power magnitude (r = −0.584 and r = −0.548, respectively).

Finally, concerning the stability variables (Table 3), the vertical stability index and the dynamic postural stability index were positively correlated to the maximum high (r = 0.550 and r = 0.555, respectively) and low-frequency tibial acceleration signal power magnitude (r = 0.496 and r = 0.497, respectively). Furthermore, the vertical stability index and the dynamic postural stability index were positively related to the delta of the maximum power magnitude in all the tibial signal (r = 0.634 and r = 0.631, respectively). Posterolateral and posteromedial reaches of mSEBT were negatively correlated to the delta of the total low-frequency tibial acceleration signal power magnitude (r = −0.540 and r = −0.515, respectively).

Discussion and implications

The aim of this study was to analyse the relationship between isokinetic strength, dynamic postural control, and muscular extensibility with impact transmission during fatigued running. The first hypothesis was that lower quadriceps and hamstrings isokinetic strength values would be related to a greater power in the high-frequency acceleration range. The first hypothesis was confirmed.

High-frequency range in the impact analysis is directly related to the severity and transmission of the impact (Shorten & Winslow, 1992). Our results showed an increased magnitude of tibial high-frequency acceleration and a greater high-frequency attenuation during fatigued running. These results are in line with those reported by with Mizrahi, Verbitsky, Isakov et al. (2000) in level running. Attenuation in the low-frequency zone increased with fatigue, but with a low effect size. We interpreted it as the optimisation of the intrinsic damping systems of the human body in order to prevent the disruption of the vestibular and visual systems that is produced by excessive accelerations in the head caused by fatigue (Gruber et al., 2014). Fatigue tends to increase knee flexion during the foot-strike (Mizrahi, Verbitsky, Isakov et al., 2000), increasing the displacement of the lower extremities. In our study, the accelerations in the head were constant, as previously reported (Gruber et al., 2014; Lucas-Cuevas et al., 2015), so this hypothetical increase in the athlete's own movements would explain the increase in low-frequency attenuation.

To our knowledge, there are no studies that directly analysed the relationship between strength and impact transmission during running, with or without fatigue. Only one study has analysed the relationship between impacts and different muscular activation levels. Potthast et al. (2010) used a pneumatically driven impactor under the heels in supine position to analyse tibial acceleration impacts with different muscular preactivation levels (0%, 30%, and 60%) of gastrocnemius, hamstrings, and quadriceps. In their study, tibial acceleration impacts decreased as muscle activation increased, with different interface hardness and knee angle. Hamner et al. (2010) analysed running accelerations during the initial contact phase. They reported that during the early part of the stance phase the main contributor to braking and support is the quadriceps muscle group. Moreover, the muscles that showed the earliest signs of fatigue during running are biceps femoris and rectus femoris, are both required during the contact phase (Hanon et al., 2005).

It has been suggested that local muscular endurance during concentric action of hip extensors, and during eccentric action of knee flexors is important to maintain a stable running style or stride mechanics, preventing or delaying the kinematic changes associated with fatigued running (Hayes et al., 2004). Our results support this idea, because modifications in high-frequency impact during fatigued running were smaller in those participants that achieved greater hamstring peak torque with the leg less extended. Therefore, our results suggest that the development of knee flexors and hip extensor strength, as well as eccentric muscle actions of the hamstrings, should be introduced into a training programme for runners, as previous studies indicated (Hayes et al., 2004). Furthermore, a greater quadriceps muscle activation in a more flexed knee position was related to lower impact acceleration (Potthast et al., 2010), similar to our results in which we found an association between lower high-frequency impact components and quadriceps peak torque in a more flexed position, revealing a protective mechanism against high-frequency impacts during fatigued running. So, improving the strength of this muscle group and maintaining a balanced hamstrings/quadriceps ratio should also be considered in complementary training programmes.

The second hypothesis, that lower hamstrings muscle range of motion values would not be related to a greater power in the high-frequency acceleration range at the tibia, was not accepted. As far as we know, there are no investigations that analyse the relationship of muscular range of motion on the impact transmission during running. According to Baxter et al. (2017), the scientific literature has not shown positive effects of stretching on performance, recovery or injury prevention for endurance runners. Other investigations reported similar results, excluding flexibility training as an injury prevention measure (Fields et al., 2010; Lauersen et al., 2014; Thacker et al., 2004).

Our results showed that runners with lower hamstrings and soleus range of motion showed higher high-frequency impacts in tibia during fatigued running. For Hardin et al. (2004), the kinematic changes that may modify impact force during running are ankle dorsiflexion and knee flexion. Soleus range of motion is directly related to the ankle dorsiflexion that occurs during the initial contact in order to absorb the impact (Hardin et al., 2004). So, gastrocnemius and/or soleus tightness limits ankle dorsiflexion and could decrease the body's ability to attenuate the impact (Leardini et al., 2001; Macrum et al., 2012; Piva et al., 2005; Witvrouw et al., 2000). It has been shown that this limited dorsiflexion may cause an excessive subtalar joint eversion and tibial internal rotation during stance as compensation to improve the range of motion (Leardini et al., 2001; Macrum et al., 2012; Piva et al., 2005; Witvrouw et al., 2000). Despite the possible compensatory movements performed by athletes with limitations in the ankle range of motion to maintain adequate running technique described in previous studies, our results show that high-frequency impacts were higher during fatigued running in athletes with soleus tightness.

Athletes with hamstring tightness have been shown to exhibit the typical gait patterns during the middle and late swing phases during running (Davis Hammonds et al., 2012). Athletes could maintain the same range of hip flexion movement during swing phase, but with an increased knee flexion in the late swing (Davis Hammonds et al., 2012). This limitation in range of motion results in a greater knee flexion during the initial contact phase (Whitehead et al., 2007). Potthast et al. (2010) suggested that the knee angle during the ground contact explains around 25–29% changes of the variance of the tibia accelerations, being higher as knee angle increases. Thus, the greater tibial impact shown by runners with hamstrings tightness in our study could be related to a greater knee flexion during initial contact phase. Moreover, some studies have shown that fatigue increases knee flexion in the initial contact (Derrick et al., 2002; Mizrahi, Verbitsky, Isakov et al., 2000). This would further increase the severity of the impact transmission in runners with hamstring shortening. Therefore, coinciding with another study that describes hamstring and ankle dorsiflexion range of motion deficits as independently associated with the injury risk (Van Dyk et al., 2018), we suggest that athletes and coaches pay more attention to improving the extensibility of soleus and hamstrings because it could reduce impact severity and injury risk.

The third hypothesis was that lower dynamic stability would be related to a greater power in the high-frequency acceleration range at tibial location. According to our results, the hypothesis was accepted. Running is a dynamic task that demands an adequate control of the centre of mass over a narrow and changing base of support (Meardon et al., 2016). So, the mechanism of landing and energy absorption is similar to anteroposterior jumps and unilateral landing tasks. In our study, runners with greater vertical and global stability indexes in landing task generated lower values of high-frequency impacts and minor modifications in maximum acceleration of the whole tibia frequency signal during initial contact in running. Meardon et al. (2016) observed that injured runners reached a greater VGRF variability in vertical and global stability after landing than non-injured runners.

In addition, fatigue increases VGRFs and time to stabilisation (Wikstrom et al., 2004; Brazen et al., 2010) as well as knee and ankle flexion (Brazen et al., 2010) after landing, increasing the injury risk. These detrimental mechanisms manifested during landing could be similar to those that occur during running. Therefore, athletes with limited postural control would have a diminished neuromuscular control. It could make them more susceptible to injuries because the increase of the stabilisation time would represent a decrease in response or reaction time to high impact forces (Brazen et al., 2010). Thus, an adequate dynamic postural control should be maintained to control the severity of impacts received in the initial contact during fatigued running. Also, it will avoid excessive loads on the passive structures responsible to absorb the impact and could reduce the injury risk.

The study is not without limitations. Although our sample size was similar to other studies that analysed impacts transmission during fatigued running (Derrick et al., 2002; Encarnación-Martínez et al., 2018; Gruber et al., 2014), perhaps a higher sample size could explain more precisely the relationship between strength, extensibility, and stability and the impact transmission during fatigued running. Furthermore, although we used a validated method to estimate the maximal aerobic speed, it may be less accurate than a direct calculation with a gas analyser. Similarly, even though Borg's scale has been shown as an effective method to measure subjective effort perception, other direct techniques, like lactate analysis, could be more accurate to evaluate the fatigue level. Also, although the tests employed to evaluate range of motion are reliable, these assessments could be influenced by examiner's or participants' subjectivity, as well as stretch tolerance. Likewise, only a single joint strength assessment was performed, so more muscle groups could have been analysed, or using a more functional protocol (i.e., quadriceps concentric and hamstrings eccentric contraction). Therefore, we think it would be interesting to carry out more research in the future with a higher number of runners and using direct methods to register and monitor the maximal aerobic speed or VO_2max and fatigue level.

Conclusion

Greater hamstring and soleus range of motion, greater dynamic postural control after a landing task, greater hamstring and quadriceps strength with the lower leg less extended and an adequate quadriceps/hamstring strength compensation may protect against high-frequency impacts in tibia during fatigued running. Considering these observations, flexibility, postural stability, and strength training should be considered in training programmes for recreational runners because they could reduce the increase of the severity of high-frequency impacts as fatigue increases during running. High-frequency accelerations are directly related to the reception and absorption of impacts and represents one of the parameters most related to running injuries. Therefore, reducing deficits in hamstring and soleus range of motion, improving stability after landing, developing hamstring and quadriceps strength in elongated muscle range, and maintaining a balanced ratio of hamstring/quadriceps strength could help to reduce the injury risk in running.

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ORCID

Alberto Encarnación-Martínez http://orcid.org/0000-0003-2149-134X
Roberto Sanchis-Sanchis http://orcid.org/0000-0003-3392-8764
Pedro Pérez-Soriano http://orcid.org/0000-0002-9825-3801
Antonio García-Gallart http://orcid.org/0000-0002-1518-3829

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