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Effect of rocker shoes and running speed on lower limb mechanics and soft tissue vibrations

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ABSTRACT

Previous studies have shown a possible effect of running speed and the sole material of footwear on lower-limb mechanics and soft tissue vibrations, while little information has been offered concerning the influence of the shape of the footwear's sole. The purpose of this study is to assess the effect of running speed and rocker shoes on muscular activity, ground reaction force, and soft tissue vibrations. Twenty participants performed heel-toe running with two shoes, differentiated only by their sole shape (i.e. rocker and non-rocker), at four running speeds. Ground reaction force and electromyograms of the *gastrocnemius medialis* and *vastus lateralis* were measured, and soft tissue accelerations of the same muscles were recorded with tri-axial accelerometers. A continuous wavelet transform was applied to the accelerometer's signals to analyse them in the time-frequency domain. The rocker of the shoes did not change the muscular activations, ground reaction force, nor power of soft tissue vibrations. In opposite, increased running speed led to an augmentation of all of the measured parameters. Interestingly, running speed augmentation led to a greater increase in high frequencies component of soft tissue vibrations (25–50 Hz, 242%) than lower ones (8–25 Hz, 111%). Consequently, we indicated a 10% increase in the relative part of the high frequencies of the total power. In conclusion, although rocker shoes have shown an effect on lower-limb kinetics in some studies, no influence on soft tissue vibration is denoted. By contrast, soft tissue vibrations may be modulated by changing running speed.

1. Introduction

Running is one of the most commonly practiced sports, with hundreds of millions of runners worldwide. Despite running's positive effects on health (Lee et al., 2017), inappropriate management of the training load may lead to increased risk of injury. Indeed, the incidence of lower-limb injuries varied from 19% and 79% of runners (van Gent et al., 2007). One of the causes of such injuries is excessive mechanical stress (Zadpoor and Nikooyan, 2011), which is characterized by a succession of muscular contractions, impact with the ground, and vibration of soft tissues (i.e. muscles, fat, and skin) caused by the impact. Moreover, some authors showed that continuous vibrations above 30 Hz cause muscular damages (de Hoyo et al., 2013) and fatigue (Cardinale and Lim, 2003; Rittweger et al., 2003).

Running speed is one of the factors that influences lower-limb mechanics. Indeed, studies have shown that lower-limb muscular activity during the stance phase (Dorn et al., 2012; Hamner et al., 2010; KyrÖlÄinen et al., 1999), as well as leg muscle pre-activation (Boyer and Nigg, 2004), increased with higher running speeds. Concerning ground reaction force, it has been observed that increasing running velocity led to an increase in passive and active peaks, vertical loading rate (VLR), and impact frequency (Boyer and Nigg, 2004; Nilsson and Thorstensson, 1989). Finally, only one study has quantified the effect of running speed on soft tissue vibrations (Boyer and Nigg, 2004), and the results showed that soft tissue acceleration amplitude and frequency were higher when running velocity increased.

Footwear type may also modify lower-limb mechanics. Studies in which the hardness or stiffness of shoes were changed did not detect altered muscular activity (Azevedo et al., 2012; Kahle et al., 2016; Roy and Stefanyshyn, 2006) or consequent subject-specific changes (Nigg et al., 2003; Wakeling et al., 2002), but other authors reported decreased muscular activity at low running speeds with increased shoe stiffness (Chen et al., 2014). Concerning ground reaction force, it has been demonstrated that running with harder or less cushioned shoes increased the impact peak and VLR (Logan et al., 2010; Ly et al., 2010; Zadpoor and Nikooyan, 2010), while some studies indicated that wearing softer shoes raised the

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impact peak (Baltich et al., 2015). By contrast, increasing the thickness of the sole did not alter the ground reaction force during running (Chambon et al., 2014; Hamill et al., 2011). Soft tissue vibrations have also been shown to be sensitive to changes in the sole material. Indeed, elastic soles increased the magnitude of vertical accelerations of the *tibialis anterior* during running in some works (Giandolini et al., 2017), while some authors did not find any differences (Boyer and Nigg, 2004).

Although the previous studies have yielded relevant insights about the influence of footwear type and running velocity on lower-limb mechanics, some concerns remain. Firstly, only the sole material, thickness or hardness, has been tested, and very few studies have investigated the effect of the sole shape, especially rocker shoes (i.e. with a curved sole), on muscular activity, ground reaction force, and soft tissue vibrations. Despite that rocker shoes did not modify muscular activity (Sobhani et al., 2013), their possible effect on ground reaction force is still debated. Indeed, some authors found that wearing rocker shoes decreased impact and passive peaks during running (Lin et al., 2017), though others did not find any effect (Boyer and Andriacchi, 2009). In addition, no study has yet investigated the influence of rocker shoes on soft tissue vibrations, although a reduction of soft tissue vibrations could be one mechanism to explain the popularity of such running shoes. Secondly, soft tissue acceleration signals are often not stationary, and they therefore require a more appropriate treatment than fast Fourier transform (Gao and Yan, 2011). In this context, a wavelet-based approach was developed to obtain a relevant time-frequency analysis (Enders et al., 2012). In addition, such an analysis makes it possible to calculate the power and a damping coefficient for several bandwidth frequencies (Khassetarash et al., 2015b).

The purpose of this study is to assess the effect of running speed and rocker shoes on muscular activity, ground reaction force, and soft tissue vibrations. On one hand, we hypothesized that increasing running speed leads to an augmentation of muscular activity, ground reaction force, and soft tissue acceleration parameters. On the other hand, running with rocker shoes should not affect muscular activity, decrease impact peak and therefore, decrease the shock and its propagation.

2. Methods

2.1. Participants

In total, 20 healthy male recreational runners (age: 23.9 ± 2.1 years, height: 1.77 ± 0.05 m, mass: 73.6 ± 7.4 kg) volunteered to partake in this study. None of the participants presented any history of lower-limb injury for six months prior to the experiment. All participants were rear-foot strikers, which was monitored with video during the warm-up period (Enders et al., 2014; Rice et al., 2016). The participants signed an informed consent document, which was approved by the local university ethics committee according to the Declaration of Helsinki.

2.2. Experimental procedure

Before beginning the tests, each subject underwent a 20-minute warm-up phase during which time they ran at different velocities on the indoor running track used for the tests with two types of shoes: rocker and non-rocker running shoes (Fig. 1). The rocker shoes were specially designed and characterized by greater rear and front elevation of the sole and a smaller ground contact surface, compared to the non-rocker shoes (Table 1). The apexes were defined as the point where the sole was not in contact with the ground anymore. The mean thickness of the sole, corresponding to the mean sole height calculated at the rear and front apexes, were only 2 mm different (corresponding to less than 5%) between the two shoes (Table 1).

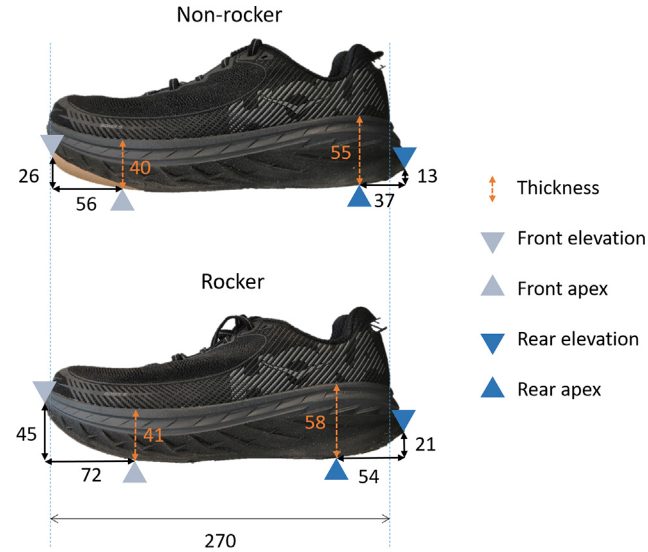

Fig. 1. Shoe without rock (top) and with a rock (bottom). All distances are in millimeters.

Table 1
Shoe characteristics for a man US size 9 (270 mm).

	Non-rocker shoe	Rocker shoe
Mass (g)	237	241
Sole material	EVA 40 - 45 C	EVA 40 - 45 C
Mean thickness (mm)	48	50
Front apex (%)	21	27
Rear apex (%)	14	20
Ground contact (%)	65	53

The tests consisted of running at 8, 10.5, 13, and 15.5 km/h on a running track with the rocker and non-rocker shoes in a counterbalanced order. Participants were instructed to land on the force plate with their right foot without modifying their natural gait. Running speed was controlled with photocells placed 1.5 m from each side of the centre of the force plate. Trials were excluded from the analysis if the running pattern was changed to ensure the runner landed on the force plate (visually controlled) or if the running speed was 5% above or below the required running speed. A condition was completed when five valid trials were documented.

2.3. Instrumentation and data collection

In line with previous studies (Boyer and Nigg, 2004; Enders et al., 2014; Khassetarash et al., 2015a; Wakeling et al., 2003; Wakeling and Nigg, 2001), electromyograms of each participant's *gastrocnemius medialis* (GAS) and *vastus lateralis* (VL) of the right leg were recorded with pairs of surface electrodes (Trigno Wireless EMG, Delsys, MA, USA, 1000 Hz). After having shaved and cleaned the skin with alcohol, electrodes were placed on GAS and VL according to SENIAM recommendations. Ground reaction force was recorded with a force plate (AMTI, OR6-7-2000, Watertown, MA, USA, 1000 Hz). Soft tissue accelerations of the right leg were recorded with two tri-axial accelerometers (Mega Electronics, Kuopio, Finland, ±100 g, 1000 Hz). In line with the work of Boyer (2006), the accelerometers were firmly tapped on the muscle belly of the GAS and VL. The y-axis was oriented in order to be collinear to the longitudinal axis of the thigh and shank respectively. All signals were synchronized with LabChart 8.0 (ADinstrument, Sydney, Australia).

2.4. Data treatment

The stance phase corresponded to the time when the vertical ground reaction force was superior to 3 N. Raw electromyograms were firstly filtered with a band-pass Butterworth filter at 10–499 Hz (De Luca, 1997). Then, signals were rectified, and a low-pass Butterworth filter with a cut-off frequency at 8 Hz was applied to obtain signal envelops. The signals were normalized for each participant with the maximum values obtained at 15.5 km/h with the non-rocker shoes during the stance phase (Albertus-Kajee et al., 2011). According to Boyer's (2006) methodology, the mean activity of GAS and VL was calculated for the pre-activation phase, during the 50 ms before impact (EMG_{pre}), and during the stance phase (EMG_{stance}). Ground reaction force was normalized relative to the participants' body weight. In accordance with Boyer and Nigg's (2007a) work, only the vertical ground reaction force was analysed. The impact peak, which corresponds to the first peak after the impact, and the active peak, which corresponds to the second peak, were automatically detected. The VLR, as corresponds to the average slope between 20% and 80% of the impact peak, and the impact frequency (Eq. (1)), were then computed with the following equation (Boyer and Nigg, 2004).

$$\text{Impact frequency} = \frac{1}{2(\text{Impact peak}/\text{VLR})} \quad (1)$$

In line with the recommendations from Khassetarash et al. (2015a), only the longitudinal axis of soft tissue accelerations was analysed. In the temporal domain, peak accelerations (Acc_{peak}) were calculated as the maximal absolute value of the acceleration signal during the stance phase (Fig. 2). For the frequency analysis, a continuous wavelet transform was applied to the raw accelerometer's signals in order to analyse the time-frequency characteristics of these signals (Fig. 2). The method described by Phinyomark et al. (2009) was used to select the mother wavelet, which was the Complex Gaussian 56. Then, the modulus of the coefficients obtained from the continuous wavelet transform were split into two frequency components: low frequencies (LF, between 8 Hz and 25 Hz) and high frequencies (HF, between 25 Hz and 50 Hz) (Fig. 2) (Boyer and Nigg, 2006). The very low frequencies under 8 Hz, which represent movement (Winter, 2009), and the very-high frequencies above 50 Hz, which represent the noise of the attachment system (Shorten and Winslow, 1992), were not analysed. The main frequency, which corresponds to the mean of the median frequencies measured at each sample time, was computed for each muscle (i.e. GAS and VL). The mean power (P_{mean}) and the maximal power (P_{max}) were calculated for each frequency band (Fig. 2). The relative part of HF was determined by calculating the ratio between P_{mean} of HF and the total mean power (HF_{ratio}). A 500 ms window after impact was considered to compute the damping coefficient (DAMP) of GAS and VL for the two frequency bands (LF and HF). To avoid side effects, DAMP corresponded to the slope of the curve between 80% and 20% of the power range computed between the maximal and minimal power (Fig. 2).

2.5. Statistics

The interaction effect between the shoes (rocker vs non-rocker shoes; fixed effect) and running speed (8 vs 10.5 vs 13 vs 15.5 km/h; fixed effect) on the parameters cited in the previous section were tested using linear mixed models (Bates et al., 2015) with the participants entered as a random intercept. Likewise, since the effect of the shoes or running speed may differ among participants, the fixed effects were also set as random slopes. We obtained the p-values from likelihood ratio tests, which were conducted by testing the full model against the model without the effect tested. The level of significance was set at $p < 0.0017$ (Bonferroni correction: 0.05/30). If an interaction effect was found, main effects were calculated for each specific running speed and shoes. When main effect was found, post-hoc pairwise comparisons were performed using Tukey's honestly significant difference. The normality, homoscedasticity, and linearity of the linear-mixed

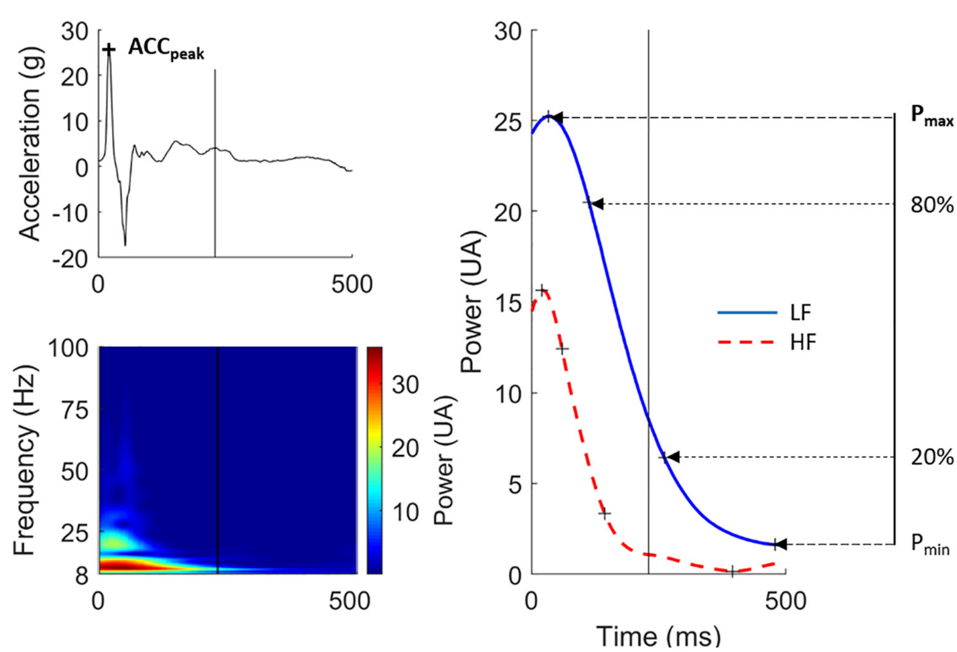

Fig. 2. Top left: acceleration signal of the *gastrocnemius medialis* at 15.5 km/h; bottom left: The map represents the power at each time and frequency obtained by the wavelet analysis; right: Low (LF, blue solid line) and high (HF, red dashed line) frequency components of the signal. The damping coefficient corresponds to the mean slope between 80% and 20%. The black vertical line represents the end of the stance phase. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

model residuals were graphically controlled. All models were performed using the package lme4 (Bates et al., 2014, p. 4) of R software (R 3.3.3, RCore Team 2014, Vienna, Austria).

3. Results

3.1. Muscular activity

The linear mixed models revealed an interaction effect of the shoes and running speed on EMG_{pre} of GAS ($p < 0.0017$) and EMG_{stance} for GAS and VL, while only a running speed effect was observed for EMG_{pre} of VL ($p < 0.0017$). However, for both muscles, there was no effect of the shoe condition on EMG_{pre} and EMG_{stance} at a given running velocity.

Concerning pre-activations, the EMG_{pre} of GAS was not influenced by running speed when the non-rocker shoes were used ($p = 0.003$; 12.7 ± 2.3%). However, when running with rocker shoes, EMG_{pre} was superior (up to 143%) at 15.5 km/h compared to two other speeds (8 and 10.5 km/h), and it was superior at 13 km/h compared to 8 km/h (Fig. 3). For the EMG_{pre} of VL, a significant increase (up to 74%) was observed between 8 and 15.5 km/h (Fig. 3). Concerning activations during the stance phase, EMG_{stance} of GAS when running with the non-rocker shoes was significantly smaller (up to −59%) at 8 km/h compared to the other running velocities. When running with the rocker shoes, EMG_{stance} at 8 km/h was inferior (up to −54%) to those measured at 13 and 15.5 km/h (Fig. 4). EMG_{stance} of VL was superior (up to 106 and 172, for the non-rocker and rocker shoes respectively) at 15.5 km/h compared to two other speeds (8 and 10.5 km/h), and it was superior at 13 km/h compared to 8 km/h (Fig. 4).

3.2. Ground reaction force

The linear mixed models did not reveal any interaction effect of the shoes and running speed on the impact peak, active peak, VLR, nor impact frequency. However, there was a running speed effect on these four variables ($p < 0.0017$).

Significant increases of the impact peak, active peak, and VLR (up to 29%, 53%, and 115% respectively) were observed with the augmentation of running velocity (Table 2). A significant increase

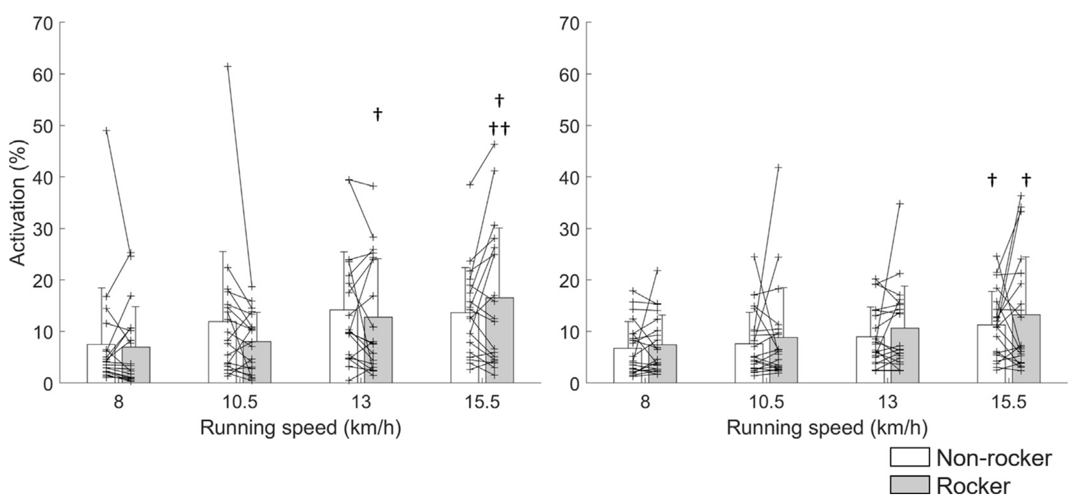

Fig. 3. Mean ± standard deviation (bars) and individual responses (thin cross) of EMG_{pre} for the *gastrocnemius medialis* (left) and *vastus lateralis* (right) at different running speeds with non-rocker (white) and rocker shoes (grey). † different from 8 km/h for the similar shoe condition ($p < 0.0017$). †† different from 10.5 km/h for the similar shoe condition ($p < 0.0017$).

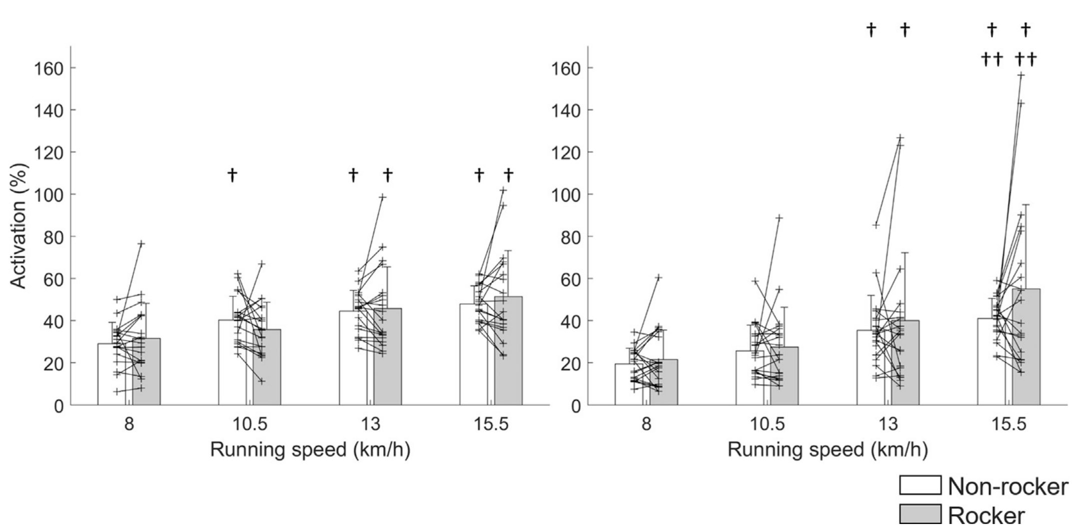

Fig. 4. Mean ± standard deviation (bars) and individual responses (thin cross) of EMG_{stance} for the *gastrocnemius medialis* (left) and *vastus lateralis* (right) at different running speeds with non-rocker (white) and rocker shoes (grey). † different from 8 km/h for the similar shoe condition ($p < 0.0017$). †† different from 10.5 km/h for the similar shoe condition ($p < 0.0017$).

Table 2
Mean ± SD values of ground reaction force parameters with respect to running speed.

	8 km/h	10.5 km/h	13 km/h	15.5 km/h
Impact peak (bw)	$1.56 \pm 0.36_a$	$1.81 \pm 0.35_b$	$2.07 \pm 0.54_c$	$2.40 \pm 0.57_d$
Active peak (bw)	$2.16 \pm 0.17_a$	$2.43 \pm 0.18_b$	$2.63 \pm 0.23_c$	$2.79 \pm 0.23_d$
VLR ($bw \cdot s^{-1}$)	$49.47 \pm 19.13_a$	$64.73 \pm 18.65_b$	$83.69 \pm 26.87_c$	$106.34 \pm 29.37_d$
Impact frequency (Hz)	$16.16 \pm 5.14_a$	$18.26 \pm 4.88_{a,b}$	$21.63 \pm 9.71_{b,c}$	$23.02 \pm 7.22_c$

For a given parameter, means that does not share the same letter are significantly different ($p < 0.0017$). bw: body weight, VLR: Vertical Loading Rate.

(up to 42%) of the impact frequency was revealed with running speed augmentation, except between 10.5 km/h and 8 km/h or 13 km/h, and between 13 and 15.5 km/h (Table 2).

3.3. Soft tissue vibrations

The linear mixed models did not reveal any interaction effect of the shoes and running speed, but there was a main effect of the running speed ($p < 0.0017$) on the soft tissue acceleration parameters (Table 3).

The Acc_{peak} increased significantly with running speed augmentation, regardless of the muscle (up to 134% and 144% for GAS and VL, respectively). The main frequency increased with running velocity augmentation (up to 22% and 27% for GAS and VL, respectively), with the following exceptions: between 10.5 km/h and 8 km/h or 13 km/h, and between 13 and 15.5 km/h for GAS, and between 8 and 10.5 km/h, and between 13 and 15.5 km/h for VL. Regardless of the muscle, P_{mean} raised significantly with increased running speed in LF (up to 119% and 104% for GAS and VL, respectively) and HF bands (up to 257% and 228% for GAS and VL, respectively). In addition, HF_{ratio} increased with running velocity augmentation (up to 46% and 45% for GAS and VL. respectively) except between 8 and 10.5 km/f for GAS. Furthermore, for both GAS and VL, P_{max} raised significantly with increased running speed in LF (up to 91% and 80% for GAS and VL, respectively) and HF (up to 167% and 153% for GAS and VL, respectively). Finally, DAMP significantly increased with running speed augmentation in LF (up to 100% and 80% for GAS and VL, respectively) and HF bands (up to 159% and 150% for GAS and VL, respectively).

4. Discussion

The purpose of this study was to assess the effect of running speed and rocker shoes on muscular activity, ground reaction force, and soft tissue accelerations. As expected, increasing running speed led to an augmentation of muscular activity, ground reaction force, and soft tissue vibrations. By contrast, running with rocker shoes did not alter these parameters.

Running mechanics can be described as a chain of consequences involving muscular activity, ground reaction force, and soft tissue vibrations. A running cycle may be seen as the activation of the prime mover muscles (for example, GAS and VL), which leads to a flying phase that ends with ground impact at landing, thus creating a shockwave that propagates along the lower-limbs and causes soft tissue vibrations (Nigg and Wakeling, 2001). These soft tissue vibrations may also be damped by appropriate activation of the leg muscles (Wakeling and Nigg, 2001). Before the impact, muscle pre-activation is necessary in order to increase joint stiffness and adapt the landing angle of the joint (Kyröläinen et al., 1999). With the exception of GAS with the non-rocker shoes, we observed that increasing running speed requires higher muscle pre-activation, which confirms the results of Boyer and Nigg (2004). Changes in muscular pre-activations with changed footwear could have been expected as rocker shoes may modify running kinematics (Boyer and Andriacchi, 2009; Sobhani et al., 2016). However, rocker shoes did not alter pre-activations of GAS and VL in our study. It can be assumed that kinematic changes between the two shoes were too small for rear-foot strikers to experience modifications in muscular pre-activations. However, running kinematics were not controlled in the current study, and therefore the link between muscular pre-activation and kinematics should be investigated. Once the foot strikes the ground, muscular activations are necessary to brake and then propel the body in a forward direction. As expected, we observed that increasing GAS and VL activities during the stance phase is necessary to run faster (Nilsson and Thorstensson, 1989; Tsuji et al., 2015). As running speeds were identical between our two shoe conditions, footwear did not alter muscular activations during the stance phase, which confirms the ideas in the study of Sobhani et al. (2013).

Table 3
Mean ± SD values of soft tissue acceleration parameters with respect to running speed.

		8 km/h	10.5 km/h	13 km/h	15.5 km/h
GAS	Acc_{peak} (g)	$9.36 \pm 3.77_a$	$12.23 \pm 4.39_b$	$16.53 \pm 5.26_c$	$21.86 \pm 7.09_d$
	Main frequency (Hz)	$16.87 \pm 3.31_a$	$17.95 \pm 3.05_a$	$19.34 \pm 3.57_{a,b}$	$20.58 \pm 4.13_b$
	HF_{ratio} (%)	$22.07 \pm 7.44_a$	$24.48 \pm 7.02_b$	$28.79 \pm 7.51_c$	$32.25 \pm 7.07_d$
	P_{mean} LF (UA)	$7.79 \pm 2.02_a$	$10.68 \pm 2.30_b$	$13.83 \pm 3.07_c$	$17.06 \pm 3.56_d$
	P_{max} LF (UA)	$10.93 \pm 3.28_a$	$14.01 \pm 3.43_b$	$17.22 \pm 4.12_c$	$20.86 \pm 4.90_d$
	DAMP LF (UA/s)	$37.57 \pm 12.59_a$	$48.75 \pm 13.70_b$	$60.55 \pm 17.25_c$	$75.01 \pm 21.33_d$
	P_{mean} HF (UA)	$2.36 \pm 1.35_a$	$3.63 \pm 1.75_b$	$5.79 \pm 2.54_c$	$8.45 \pm 3.47_d$
	P_{max} HF (UA)	$5.16 \pm 3.04_a$	$7.10 \pm 3.60_b$	$10.11 \pm 4.57_c$	$13.80 \pm 6.15_d$
	DAMP HF (UA/s)	$35.59 \pm 23.96_a$	$48.44 \pm 29.56_b$	$68.11 \pm 38.77_c$	$92.33 \pm 51.71_d$
VL	Acc_{peak} (g)	$10.77 \pm 7.65_a$	$13.66 \pm 8.01_b$	$20.12 \pm 11.36_c$	$26.40 \pm 13.01_d$
	Main frequency (Hz)	$15.90 \pm 3.23_a$	$17.18 \pm 2.87_a$	$19.04 \pm 4.13_b$	$20.26 \pm 5.00_b$
	HF_{ratio} (%)	$25.12 \pm 6.62_a$	$29.59 \pm 6.60_b$	$34.38 \pm 6.71_c$	$36.51 \pm 5.91_d$
	P_{mean} LF (UA)	$8.51 \pm 4.01_a$	$10.72 \pm 4.13_b$	$14.02 \pm 5.07_c$	$17.35 \pm 6.01_d$
	P_{max} LF (UA)	$11.06 \pm 5.48_a$	$13.03 \pm 5.17_b$	$16.30 \pm 6.18_c$	$19.91 \pm 7.18_d$
	DAMP LF (UA/s)	$39.28 \pm 18.30_a$	$48.21 \pm 18.52_b$	$60.17 \pm 22.65_c$	$70.70 \pm 25.33_d$
	P_{mean} HF (UA)	$3.22 \pm 2.25_a$	$4.93 \pm 3.10_b$	$7.94 \pm 4.21_c$	$10.56 \pm 4.99_d$
	P_{max} HF (UA)	$6.08 \pm 4.85_a$	$8.07 \pm 5.19_b$	$11.93 \pm 6.67_c$	$15.39 \pm 7.76_d$
	DAMP HF (UA/s)	$39.00 \pm 30.34_a$	$52.08 \pm 33.38_b$	$76.94 \pm 43.88_c$	$97.55 \pm 48.82_d$

For a given parameter, means that does not share the same letter are significantly different ($p < 0.0017$). Acc_{peak}: peak acceleration, P_{mean}: mean power, P_{max}: maximum power, DAMP: damping coefficient, HF_{ratio}: relative part of HF, $g = 9.81\ m \cdot s^{-1}$.

Human strategies for increasing running speed consist of increasing step frequency and amplitude (Dorn et al., 2012), which involves an increase in the vertical displacement of the centre of gravity (Hamner et al., 2010). Consequently, as observed in a previous study (Boyer and Nigg, 2004), we indicated that greater running velocities are associated with greater ground impact magnitude and frequency. Rocker shoes were designed with a curved sole in order to smooth the roll of the foot, especially during the braking phase. However, contrary to our hypotheses, we did not observe a decrease in ground impact peak and VLR in comparison to the non-rocker shoes. Our results differ from those of Lin et al. (2017); however, an explanation for these discrepancies may be that Lin et al. (2017) did not test exactly same sole material and thickness in their study, and these parameters are known to influence ground reaction force (Nikooyan and Zadpoor, 2011).

As previously depicted, ground impact leads to the propagation of a shockwave, which creates soft tissue vibrations (Boyer and Nigg, 2007b). Therefore, in line with Boyer and Nigg (2004), we observed that increasing running speed induced an augmentation of all of the parameters linked to soft tissue accelerations. By contrast, as the rocker shoes did not modify muscular activations nor ground reaction force, soft tissue accelerations parameters were not modified by the rocker shoes. Running faster required higher muscular activity and induced greater impact frequency, which led to an increased main frequency of soft tissue vibrations (Boyer and Nigg, 2004). In addition, an increase in peak acceleration with running speed explained the greater mean and maximal powers regardless of the frequency bandwidth and muscle. More precisely, the proportion of the high frequencies increased with running velocity, which indicates that the increase in the total power is mainly due to an increase in the high frequencies power. This result may confirm that the frequency component of soft tissue vibration signals is influenced by impact frequency and muscular activity (Wakeling and Nigg, 2001). Furthermore, vibrations may be damped by controlling the muscular activity (Boyer and Nigg, 2007b), which may influence the apparition of muscular fatigue caused by high vibration frequencies (Cardinale and Lim, 2003; Rittweger et al., 2003). In our study, increasing running speed led to an increased damping coefficient, mainly for high frequencies, which may be partly explained by the notion that the neuromuscular system tends to adopt a protective mechanism to avoid excessive vibrations (Nigg and Wakeling, 2001).

Our study suffers from some limitations, including a short adaptation time for running with rocker shoes. We have only assessed the short-term adaptations to rocker shoes, and thus a longer period of adaptation may lead to different results. The second limitation is that all runners were rear-foot strikers, and therefore our results cannot be applied to fore-foot strikers.

5. Conclusion

Although rocker shoes could have an effect on joint kinematics and kinetics during running, they did not modify GAS or VL activities, ground reaction force, nor soft tissue vibrations. This finding can be explained by the fact that, contrary to other studies, the sole material and thickness of the two shoes were identical. In common use, rocker shoes are "maximalists", with thicker and softer soles than non-rocker ones. This combination of characteristics, compared to the only change in the shape of the sole, could modify lower-limb mechanics and soft tissue vibrations. By contrast, running speed augmentation increased muscular activity, ground reaction force, and soft tissue vibrations. Interestingly, running speed augmentation led to a greater increase in soft tissue acceleration parameters in high frequencies, which are known to induce muscular fatigue or damages. Nevertheless, further studies are needed to better understanding the link between soft tissue vibrations in running and fatigue or damages.

Conflict of interest statement

None of the authors are in conflict of interest with regards to this research.

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