EFFECTS OF ONE-HOUR ROAD RUNNING AND SHOE ON TIBIAL ACCELERATIONS IN RECREATIONAL RUNNERS

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The purpose of this study was twofold. Firstly, we evaluated the effect of one-hour running on tibial acceleration parameters. Secondly, we determined whether the shoe fatigue after one-hour running would influence these tibial acceleration parameters. Ten runners ran at a constant preferred speed with a standard running shoe. They also ran one bout of 5 min with another identical but fresh running shoe before and after the one-hour run. Tibial acceleration amplitudes and wavelet analyses demonstrated a significant reduction of the tibial impact parameters after one-hour running, but no significant shoe effect. These reductions could be attributable to the slight increase in stride frequency from the beginning to the end of the one-hour running. Noteworthy, the tibial acceleration reduction was observable along the anteroposterior axis, not along the vertical axis. Considering the tibial acceleration as a potential risk factor for developing running-related injuries, these parameters did not change in a harmful way after one-hour of running at the preferred running speed. The shoe modification during the one-hour run did not seem to influence running biomechanics.

KEYWORDS: Running, Footwear, Fatigue, Tibial acceleration, Time-frequency analysis

INTRODUCTION: Tibial acceleration is a common measurement of shocks during sports activity, especially during running. It has been previously proposed as a surrogate measurement of the loading applied to a runner's skeleton (Sheerin et al. 2019). In this line, it could be an interesting indicator to investigate injury risk in running. Running-related injury incidence rate ranges from 2.5 to 33 injuries per 1000 hours of running (Videbaek et al. 2015), and most of these injuries are overuse injury occurring progressively. This emphasized the importance of time course analyses of running mechanics and tibial acceleration.

Previous studies investigating the time-course of tibial acceleration parameters over time found significant increases in the peak-to-peak amplitudes (PPA) with the runners' fatigue (Mizrahi et al. 2000; Reenalda et al 2019). These studies were performed in a laboratory or on athletic track for a limited duration (20-30 min) at high constant speed, which was not representative of the runners' habits. A recent study investigated tibial acceleration during the Boston Marathon (Ruder et al. 2019). Authors found reduced vertical tibial acceleration amplitude at the end of the marathon which was no more present while correcting for the reduction of speed at the 40th kilometre relatively to the 10th kilometre. One limitation of both recent studies was the use of ±16g IMU-based accelerometers (Reenalda et al. 2019; Ruder et al. 2019). Such a range of measurement might be not sufficient to record either simple spatiotemporal running parameters or peak tibial accelerations (Mitschke et al. 2018).

In the study of the variation of tibial acceleration during a prolonged running session, it seems also important to take into account the influence of footwear. Indeed, shoe cushioning properties have been shown to have an influence on tibial acceleration parameters (Morio et al., 2009; Horvais et al. 2019). Moreover, these cushioning properties have also been shown to evolve during a running session (Divert et al. 2005; Chambon et al. 2014).

The purpose of the present study was twofold. Firstly, we investigated the effect of one-hour running on tibial acceleration parameters with embedded ±50g accelerometer on a road running-like environment. It was hypothesized that while running at a constant speed, tibial acceleration would increase at the end of the one-hour running session. Secondly, we determined whether the changes of shoe cushioning properties after one-hour running would influence these tibial acceleration parameters. It was hypothesized that interaction would appear with a significant increase of the tibial acceleration due to the stiffening of the worn shoes after one-hour of running compared to equivalent fresh running shoes.

METHODS: Ten regular runners (7 males and 3 females; 35.5 ± 8.2 years; 1.75 ± 0.10 cm; 72.4 ± 10.9 kg) volunteered to participate in this study. They were free from lower-limb injury at the time of the study since at least one year. All the participants gave informed consent, in compliance with the ethical rules and laws that regulate human experimentations in France.
Participants preferred running speed was defined as their average training speed for a one-hour session (11.1 ± 0.8 km/h). The running speed was then imposed with recorded beeps and 20 m marks on the asphalt track and it was also controlled with a GPS watch (Fenix 5, Garmin™). The rate of perceived exertion (RPE) was reported at the end of the run.
The participants ran with a provided standard running shoe (Kiprun Fast 2019, Kalenji™) during the one-hour running session, thereafter called 'worn shoes'. Before and immediately after the one-hour run, they also ran 5 min at the same speed with another new pair of shoes of identical brand, size, thereafter called 'fresh shoes'. In order to compare the beginning of the run (PRE) to the end of the run (POST), the 5th and the 60th minutes of the one-hour run were compared for the worn shoes, and the last minutes of both 5 min runs before and after were compared for the fresh shoes. The first 4 minutes of run were not considered for the analyses due to the warm-up of the shoes and the optimisation of the participants running pattern (Divert et al. 2005; Morio et al. 2011).
A lightweight triaxial accelerometer (6 grams, ±50 g, Dytran™) was firmly strapped on the skin of the right leg onto the anteromedial aspect of the lower leg 12 cm proximal to the medial malleolus (Morio et al. 2011). The accelerometer signal was recorded at 2000 Hz with a homemade embedded A/D board in a small backpack of 1.8 kg.
First, a low-pass critically damped filter with a cut-off frequency of 100 Hz was applied to the tibial acceleration signals. Then, the PPA was calculated for each tibial impact on the anteroposterior (AP) and vertical axes (V). Mediolateral axis was excluded from the analyses because of technical issues with the embedded system for several participants. The stride frequency of the right leg was calculated as the peak of vertical tibial acceleration frequency.
Time-frequency content of the tibial acceleration was extracted from the raw signals using the wavelet analysis technique (Figure 1). The window duration for the time-frequency analyses was 200 ms, taken 50 ms before and 150 ms after the peak of vertical tibial acceleration. The Mexican Hat mother wavelet was used for these analyses (Morio et al. 2019). Wavelet scalogram was expressed in pseudo-frequency (*pf*) following the Equation (1), where *Fw* was the centre frequency of the 'mexh' wavelet, *dt* was the time period of the signal and *s* was the scale numbers of the wavelet. *Fw* value of the 'mexh' wavelet was 0.25 Hz and the 76 pseudo-frequency scales were range from 1.39 Hz to 251.65 Hz. The mean power frequency (MPF) was calculated regarding the Equation (2), where *W* was the wavelet decomposition spectrum, *n* the length of the signal and *ns* the number of scales. The MPF corresponded to the frequency weighted mean of *W*. The peak frequency (PF) was calculated regarding the Equation (3) as the frequency scale with the maximum of *W* content in it. Data analyses were performed with Matlab R2019A and the Wavelet Toolbox (The MathWorks™).

$$pf = \frac{Fw}{dt \times s} \text{ Eq. (1)} \qquad \mathrm{MPF} = \frac{\sum_{\mathrm{i}=1}^{n}\sum_{\mathrm{j}=1}^{ns}|W(\mathrm{i,j})| \times pf(\mathrm{j})}{\sum_{\mathrm{i}=1}^{n}\sum_{\mathrm{j}=1}^{ns}|W(\mathrm{i,j})|} \text{ Eq. (2)} \qquad \mathrm{PF} = pf\left(\max\left(\sum_{\mathrm{i}=1}^{n}\frac{|W_{\mathrm{ij}}|}{ns_{\mathrm{j}}}\right)\right) \text{ Eq. (3)}$$

Two-way ANOVA (time x shoe effects) with repeated measures was applied on every dependent variable (PPA_{AP}, PPA_{V}, MPF_{AP}, MPF_{V}, PF_{AP}, PF_{V}, stride frequency). The effect sizes were reported through the calculation of the partial omega square, ω_p^2. After a Bonferroni correction for the family test, time main effect plus interaction, α was set at 0.025.

RESULTS: At the end of the run, the RPE rating was 12.5 ± 1.4 [10–15]. The stride frequency was greater at the 60th minute of running (POST) compared to the PRE measurement with a large effect size of 0.54. The PPA_{AP} demonstrated a significant reduction at the POST measurement compared to PRE with a large effect size of 0.48 (Table 1). There was no interaction between the shoe and the time effects.

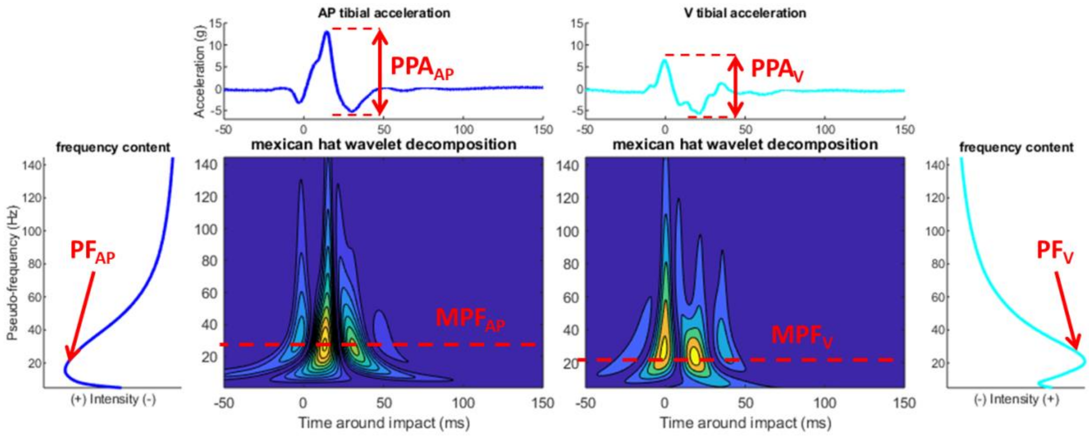

Figure 1: Representation of a typical anteroposterior (AP) and vertical (V) tibial acceleration, and the associated wavelet decomposition and the frequency content analyses.

Table 1: Dependent variables (mean ± sd) and ANOVA results for the time (PRE vs POST) and shoe (Fresh vs Worn) conditions. * indicates a significant effect p<0.025.

	PRE		POST		time effect			interaction		
	Fresh	Worn	Fresh	Worn	$F_{(1,9)}$	p	ω_p^2	$F_{(1,9)}$	p	ω_p^2
Stride frequency (Hz)	1.37 ± 0.08	1.38 ± 0.07	1.40 ± 0.06	1.40 ± 0.06	12.89	0.006 *	0.54	0.00	0.972	0.00
PPA_{AP} (g)	10.7 ± 3.6	11.2 ± 4.3	10.0 ± 3.6	10.2 ± 3.4	10.42	0.010 *	0.48	0.15	0.709	0.00
PPA_{V} (g)	6.4 ± 1.9	6.7 ± 2.3	6.9 ± 2.0	6.8 ± 2.4	3.44	0.097	0.20	3.43	0.097	0.20
PF_{AP} (Hz)	19.2 ± 6.1	19.3 ± 6.8	16.4 ± 6.6	17.1 ± 6.4	3.42	0.097	0.20	0.07	0.803	0.00
PF_{V} (Hz)	7.6 ± 0.6	7.6 ± 0.6	8.6 ± 1.4	9.4 ± 4.6	3.31	0.102	0.19	0.26	0.623	0.00
MPF_{AP} (Hz)	26.5 ± 2.3	26.2 ± 2.1	25.2 ± 2.1	24.9 ± 2.6	4.96	0.053	0.28	0.00	0.972	0.00
MPF_{V} (Hz)	17.2 ± 1.3	17.2 ± 1.3	17.3 ± 1.3	17.8 ± 3.2	1.15	0.312	0.01	0.5	0.498	0.00

DISCUSSION: The present results did not confirm any of our hypotheses. There was no increase in tibial acceleration parameters after a one-hour prolonged run and there was no interaction effect between the worn and the fresh shoes.

The POST measurements even presented a reduction of PPA_{AP} associated with an increase of stride frequency. It could be questioned whether these results were an adaptation to prolonged effort during the run. Another possibility would be that the increased stride frequency would be the response to the repeated impacts. Similar protective adaptation of increased cadence to exhaustive exercise has been previously reported in the literature (Giandolini et al. 2016), but this was reported after an ultramarathon race of 110 km. The present result was closer to the previous ones during shorter running duration. The prolonged running at a 10 km preferred speed is neither an exhaustive exercise (Morio et al. 2011; Giandolini et al. 2016) nor a short running session at high running speed (Mizarhi et al. 2000; Reenalda et al. 2019). The different durations and the exertion levels might explain the differences between our results and the previous literature. The relatively low RPE at the end of the one-hour running might also explain the absence of increased frequency of the tibial acceleration compared to previous results during an exhaustive exercise (Morio et al. 2011).

The mediolateral axis was not reported here because of recording difficulties for some participants. Nevertheless, the results observed on the sole anteroposterior axis confirmed the interest of measuring 3D tibial acceleration when studying running impact (van den Berghe et al. 2019), contrary to lot of previous study which used only the vertical tibial acceleration (Mizrahi et al. 2000; Giandolini et al. 2016; Ruder et al. 2019). The absence of interaction indicated that the shoe properties modifications during one-hour running were not sufficient to induce a modification in the tibial acceleration parameters.

CONCLUSION: This study was the first to investigate tibial acceleration parameters during a one-hour running session at a constant preferred speed, which corresponded to a typical recreational running session. Tibial acceleration parameters did not increase at the end of this submaximal running session nor were they influenced by the modification of the shoe properties. The stride frequency increased at the end of the session which might be a protective mechanism to reduce the impact.

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