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Interaction between body composition and impact-related parameters in male and female heel-toe runners

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A R T I C L E I N F O

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A B S T R A C T

Background: Bone fatigue resistance and more generally the ability to dissipate the stress sustained in dynamic tasks are partly affected by tissue properties. Men and women demonstrate substantial differences in body composition.
Research question: To assess whether gender, as a function of body composition, affects impact-related parameters in running.
Methods: A qualitative study has been conducted. Twelve females and eighteen males performed four 2-min running trials at 2.8 m·s^{-1}, 3.3 m·s^{-1}, 3.9 m·s^{-1}, and 4.4 m·s^{-1} while recording axial and transverse tibial acceleration. Peak acceleration and power spectral density within the impact-related frequency range (vibration content) were measured. Bone mineral content, fat mass, lean mass, and muscle mass were assessed using an impedance meter. Two-way (gender × speed) ANOVAs were computed. Multiple linear regressions were then used to assess the magnitude of the effect of body composition indicators on impact-related parameters.
Results: Significant gender and speed effects were observed. Females and high running speeds were associated with greater peak acceleration and vibration content at the tibia. Small interactions were observed between muscle mass and axial peak acceleration and vibration content, and between bone mineral content and transverse peak acceleration and vibration content, and axial vibration content.
Significance: Women demonstrated greater mechanical stress than men during running. High mechanical stress was associated with low bone mineral content and muscle mass. These findings may have implications in the prevention and management of bone overuse injuries in runners.

1. Introduction

About 50–75% of all running injuries are reported to be overuse injuries, i.e. due to the constant repetition of the same movement [1]. Among them, bone stress injuries are very common in the long-distance running population [2]. Age, genetics, diet and also gender are thought to be intrinsic risks factors in osteo-articular degenerative changes [3]. Besides, the strain applied to the skeleton may be recognized as an extrinsic risk factors in osteo-articular injuries.

The strain applied to a joint is essentially comprised by compression and shear forces, resulting from the contraction of muscles surrounding the joint, and also the impact-induced mechanical stress, such as shock and transmitted vibrations [4–7]. Many ex vitro studies observed that subjecting specimens of animals cartilage to successive impacts lead to cartilage fibrillation and then degeneration [6,8]. Low-magnitude high-frequency vibrations has been shown to accelerate cartilage degradation as well in rats [9]. In humans, conclusions are more heterogeneous. While studies have found no relationship between running and joint degradation [10,11], some authors did establish a link [12].

Running is unique in that it imparts mechanical strain in a repetitive and cumulative manner. Compression force [13], high loading rate [14], shock acceleration [15] and impact-induced vibrations from the heel to the head [16] have all been identified as repetitive and cumulative loads associated with running. Shock and vibrations are partly attenuated by the fatty heel pad, bone deformation, and through muscle contraction [17–20]. Indeed, the 'muscle tuning' paradigm proposed that muscle activity is tuned in response to impact force characteristics to dampen the soft-tissue vibrations [21–24]. A potential reason for this damping may be that the locomotor system tries to minimize detrimental effects of repetitive shocks and vibrations. Besides, bone density and geometry were found to affect bone fatigue resistance and the risk of sustaining stress fractures [25,26]. Using a human pendulum, Schinkel-Ivy et al. noticed that tibial peak acceleration was negatively associated with both leg lean mass and bone mineral content [27].

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Their results tend therefore to demonstrate that the ability to attenuate impact-induced stress is in some way related to the musculoskeletal system properties.

Based on these findings, one could hypothesize that body composition (especially muscle mass and bone mineral content) influences stress magnitude experienced in running. If this assumption is correct, one could expect differences in mechanical stress intensity between males and females since body composition differs [27]. The purposes of this study are (i) to observe the effect of gender on shock acceleration and vibration content across different running speeds, and (ii) to assess whether shock acceleration and vibration content measured at the tibia are affected by body composition. We hypothesized that women would demonstrate greater shock intensity and vibration content because of low bone mineral content and muscle mass. We also expected to observe lower shock intensity and vibration content in high muscle mass runners.

2. Materials and methods

2.1. Participants

Thirty recreational and experienced runners were recruited (12 women: 29 ± 6 years, 59.9 ± 7.7 kg, 168 ± 6 cm; 18 men: 41 ± 12 years, 74.6 ± 10.8 kg, 177 ± 6 cm). Once the entire procedure had been fully explained to the participants, all gave their written informed consent to participate in this study. All subjects were visually classified as heel strikers during treadmill running and wore the same shoe with a full ethylene vinyl acetate midsole (Salomon X-Scream 3D: mass = 300 g, heel height = 23 mm, drop = 10 mm, hardness = 55 Shore C).

2.2. Experimental design

All the measurements performed have been approved by the local ethical committee, and complied with the principles laid down by the Declaration of Helskinki. The treadmill used in this study was the Precor 954i (Precor, Woodinville, USA). This treadmill was modified by a Precor technician in order to remove the impact absorbing function integrated in it. As a treadmill habituation and a warm-up, subjects first ran for 5 min at their preferred running speed. Then, they performed four 2-min running trials at 2.8 $m{\cdot}s^{-1}$, 3.3 $m{\cdot}s^{-1}$, 3.9 $m{\cdot}s^{-1}$, and 4.4 $m{\cdot}s^{-1}$ in a randomized order. During the last 30 s of each trial, three-dimensional tibia accelerations were recorded at 1344 Hz using a tri-dimensional wireless accelerometer (Agile Fox, Hikob, Villeurbanne, France; ± 24 g, mass = 22 g) placed on the anteromedial aspect of the left tibia at one third of its length from its distal end. Acceleration data were collected on a micro-SD card within the accelerometer. A follow up visit to the laboratory aimed at assessing their body composition by bioimpedance. Accordingly, a wrist-to-ankle multifrequency impedance meter (Z-Métrix®, BioparHom, La Motte Servolex, France) was used. Two Al/AlCl gel electrodes were then placed at each upper limb (onto the hand and at the wrist), and two others were placed at each lower limb (above the lateral malleolus and 5 cm above this latter) [28,29]. The bioimpedance meter delivers a low intensity sinusoïdal current (77 μA) at frequencies ranging from 1 kHz to 1000 kHz with the subject standing up. It permits the collection of resistances of different body compartments which allows the calculation of tissue indicators using validated equations: lean mass, muscle mass, fat mass and bone mineral content. The variables obtained were expressed in kilograms and in percentage of total body mass. The impedancemeter has been validated by a clinical study (n°2008-A01373-52) conducted on 48 women and 47 men. It was compared to biphotonic densitometer X-ray (reference method). The repeatability of the device was 0.53%. The error of measurement compared to the reference method ranged from 0.8% to 3.9% with an average error of 1.6%.

2.3. Data analysis

Acceleration data reduction was performed using Scilab 5.4.1 software (Scilab Enterprises, Orsay, France). All steps over the 30-second period were analyzed. For axial (corresponding to the tibia axial axis, hereinafter referred to as *axial acceleration*) and transverse (corresponding to the axis of the tibia antero-posterior medial aspect, hereinafter referred to as *transverse acceleration*) accelerations, 0.3-s subsamples roughly corresponding to stance phases were extracted considering the deflection before peaks of axial acceleration (i.e. the negative minimum before the positive acceleration peak) as the start of the impact phenomenon [16]. From the axial and transverse acceleration subsamples, both a time analysis and a frequency analysis were computed. For the time analysis, acceleration signals were 50 Hz low-pass filtered to limit the resonance frequency of the attachment in the quantification of impact shock magnitude – typically from 60 to 90 Hz [16]. Then, maximal peaks of acceleration (PA) were measured from the axial and transverse acceleration signals. For the frequency analysis, the procedure initially described by Shorten and Winslow [16] was used. Acceleration subsamples were padded with zero to obtain a total of 512 points per subsample. For each axial and transverse acceleration subsample, the power spectral density (PSD) was calculated using the Fast Fourier Transform. PSD curves were interpolated so each frequency bin was 2 Hz. The PSD within the impact frequency range (iPSD), i.e. from 10 to 30 Hz [16], was calculated for each subsample.

2.4. Statistics

All data are presented as mean ± standard deviation. Normal distribution was assessed by the Shapiro-Wilk normality test and variance homogeneity by Fisher *F*-tests. To study the effect of both gender and speed on impact-related parameters, two-way within subject ANOVAs (gender × speed) were computed for impact-related variables (statistical power greater than 70%). Newman-Keuls post-hoc tests were used to determine significant differences if the ANOVA revealed a significant main effect. In view of the results obtained from ANOVAs, a Bravais-Pearson correlation was computed *a posteriori* from the impact-related variables ad mass. Then, to investigate the relationship between impact-related parameters and body composition indicators, block entry multiple regression analysis were computed for each speed. Axial and transverse PA and iPSD were considered as the dependent variables and body composition indices (expressed in percentage of body mass) as the independent variables. Block entry multiple regressions were preferred to consider the magnitude and significance of the effect of each body composition indicator. Outcomes of multiple regressions were interpreted using the qualitative approach of Batterham and Hopkins [30]. For each association between dependent and independent variables, the weighted β coefficient was extracted, and 95% confidence intervals were calculated for β values. Then, the Cohen's scale of magnitude was used to quantify the magnitude of interactions (large interaction for $\beta > 0.8$ or $\beta < -0.8$; moderate interaction for β between 0.5 and 0.79 or between -0.5 and -0.79; small interaction for β between 0.2 and 0.49 or between -0.2 and -0.49; trivial interaction for β between -0.19 and 0.19).

3. Results

A main speed effect was observed for both axial and transverse PA and iPSD ($p < 0.001$, Fig. 1). Post-hoc tests revealed that as running speed increased from 2.8 to 4.4 $m{\cdot}s^{-1}$, axial PA and iPSD significantly increased as well by 15–51% per speed increment ($p < 0.001$). Regarding the transverse component of acceleration, as running speed increases from 3.3–4.4 $m{\cdot}s^{-1}$, transverse PA and iPSD significantly increase by 22–91% per speed increment ($p < 0.01$). A main gender effect was observed only for axial PA and iPSD ($p < 0.001$). Females demonstrated greater axial PA than males at 4.4 $m{\cdot}s^{-1}$ (Δ 30%,

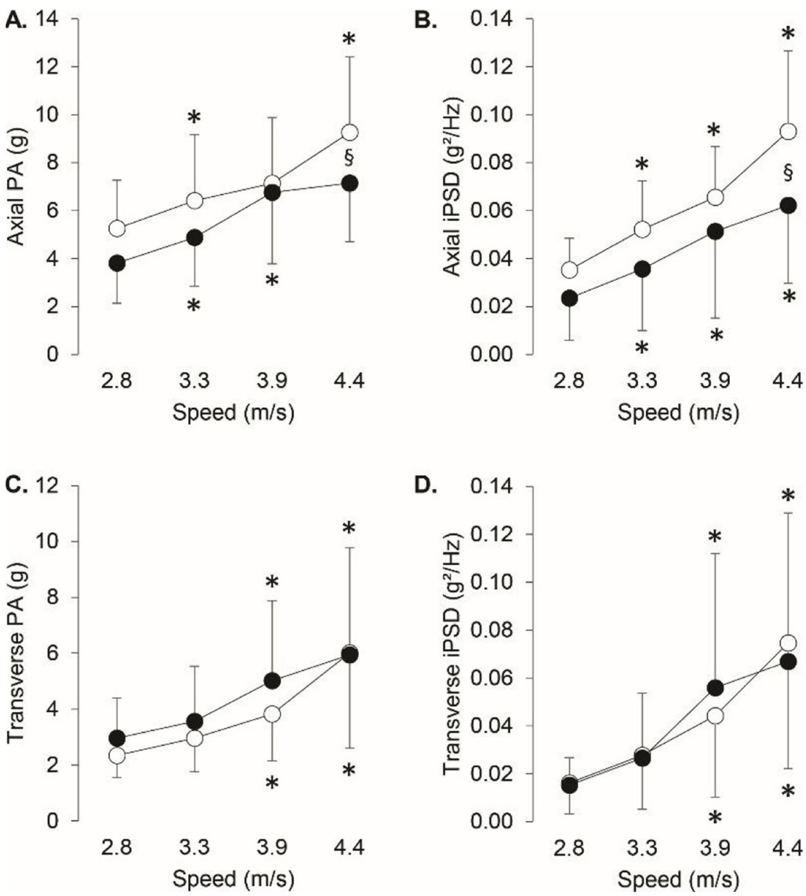

Fig. 1. Axial shock intensity (panel A) and vibration content (panel B), and transverse shock intensity (panel C) and vibration content (panel D) as a function of running speed for males (black dots) and females (white dots). * indicate significant differences in-between two running speed increments. § indicate significant differences between males and females.

p = 0.008). At 3.3 m·s^{-1}, axial PA tended to be greater in females than in males (Δ 31%, p = 0.06). Similarly, axial iPSD was significantly greater in females than in males at 4.4 m·s^{-1} (Δ 49%, p < 0.001), and tended to be higher in females at 3.3 and 3.9 m·s^{-1} (Δ 47%, p = 0.06, and Δ 28%, p = 0.092, respectively). No difference between males and females were observed for transverse iPSD as well as no interaction effect.

Body-related variables were presented in Table 1 for females and males. Compared to females, males demonstrated greater body mass, absolute lean mass, muscle mass and bone mineral content, but lower relative fat mass. Body mass was significantly related to axial PA at all speeds (2.8 m·s^{-1}: r = -0.375; 3.3 m·s^{-1}: r = -0.393; 3.9 m·s^{-1}: r = -0.391; 4.4 m·s^{-1}: r = -0.406), and to axial PSD at 3.3, 3.9 and 4.4 m·s^{-1} (3.3 m·s^{-1}: r = -0.378; 3.9 m·s^{-1}: r = -0.387; 4.4 m·s^{-1}: r = -0.419). Regarding transverse impact-related variables, only PA at 2.8 m·s^{-1} was positively correlated to body mass (r = 0.422). Considering that body mass influences impact characteristics, relative body composition indices were used for multiple linear regressions in order to determine the specific effect of lean mass, muscle mass, fat mass and bone mineral content on axial and transverse PA and PSD.

Results for multiple regressions were summarized in Figs. 2 and 3. For axial PA (Fig. 2), small interactions were observed with muscle mass, especially at 2.8, 3.9 and 4.4 m·s^{-1}. A negative small interaction with bone mineral content was also noticed but only at the lowest speed. For axial iPSD (Fig. 2), negative small interactions with muscle mass were also observed at all speeds. Negative small interactions were similarly noted with bone mineral content at all speeds. These results tend to indicate that low relative muscle mass and low bone mineral content are associated with high shock intensity and vibration content along the axial axis of the tibia. The contribution of lean mass in axial stress magnitude was, while small to moderate, highly variable as indicated by the 95% intervals.

For transverse PA (Fig. 3), negative large and small interactions with bone mineral content were observed at 2.8 and 3.3 m·s^{-1},

Table 1
Mean ± standard deviation for body-related parameters expressed in centimeters, in kilograms and in percentage of body mass for females and males. Significant differences between females and males are highlighted by bold p values.

		Females			Males			p value
Height	(cm)	168	±	6	177	±	6	**0.0003**
Mass	(kg)	56.9	±	7.7	74.6	±	10.8	< **0.0001**
Lean mass	(kg)	41.1	±	4.3	51.9	±	5.3	< **0.0001**
	(% of body mass)	71.3	±	6.6	70.6	±	6.7	0.800
Muscle mass	(kg)	21.6	±	4.6	36.8	±	4.0	< **0.0001**
	(% of body mass)	36.0	±	3.1	49.7	±	2.7	< **0.0001**
Fat mass	(kg)	12.5	±	4.7	12.9	±	3.9	0.816
	(% of body mass)	21.6	±	6.0	17.1	±	3.6	**0.014**
Bone mineral content	(kg)	1.75	±	0.18	2.22	±	0.11	< **0.0001**
	(% of body mass)	3.91	±	0.46	3.69	±	0.53	0.259

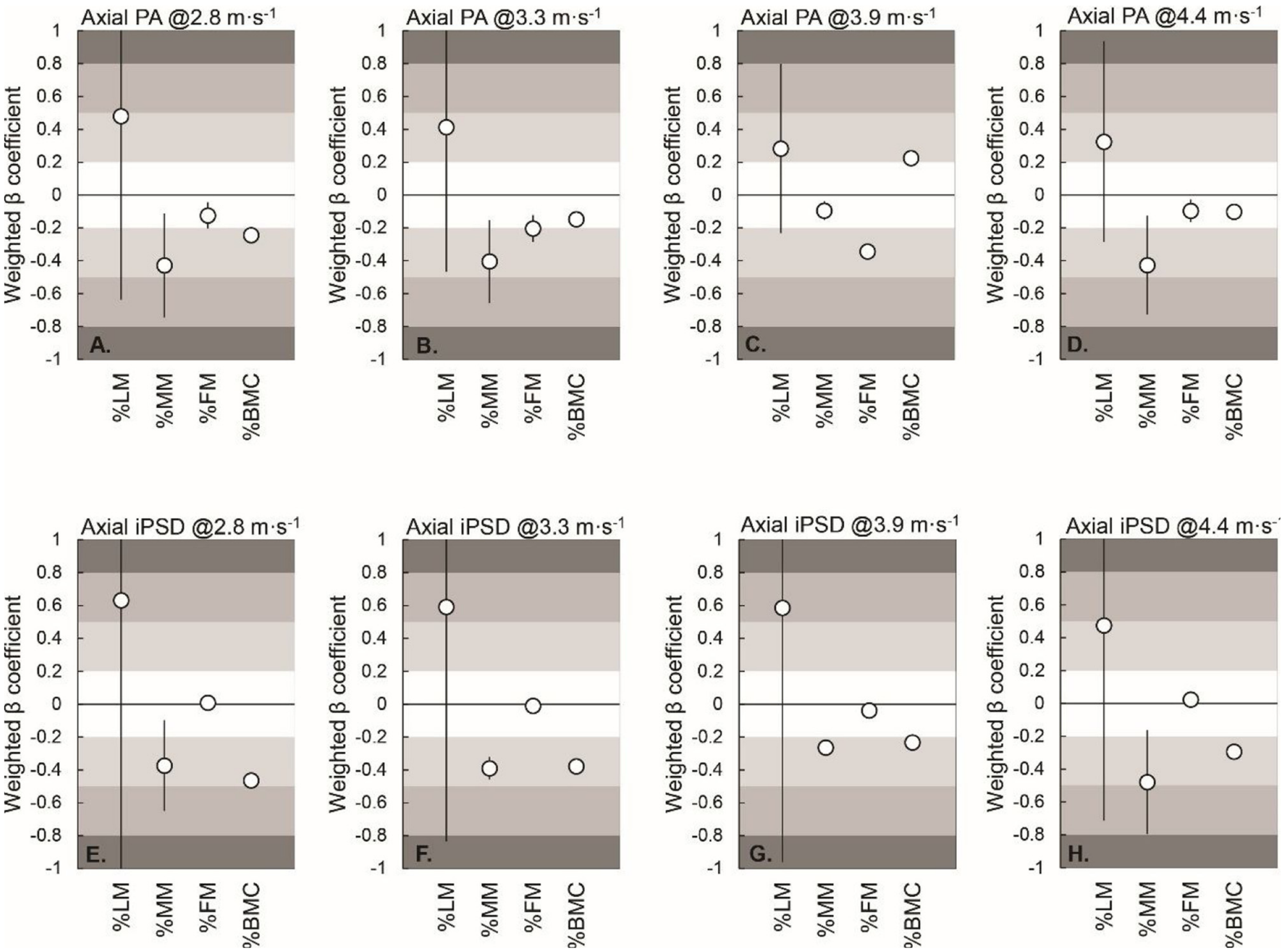

Fig. 2. Weighted β coefficients for interactions between axial shock intensity (panels A–D) or vibration content (panels E–H) and body composition indices: lean mass (%LM), muscle mass (%MM), fat mass (%FM), and bone mineral content (%BMC). Bar errors represent 95% confidence intervals of β coefficients. Dark gray areas denote large interactions, gray areas denote moderate interactions, light gray areas denote small interactions, and white areas denote trivial interactions.

respectively. For transverse iPSD (Fig. 3), a small negative interaction with bone mineral content was noticed at the lowest speed, whereas small positive interactions were observed with bone content at 3.9 and 4.4 $m{\cdot}s^{-1}$. Besides, a small negative interaction was observed with lean mass at 3.9 $m{\cdot}s^{-1}$ only. Other interactions were either trivial or highly variables.

4. Discussion

Our purposes were (i) to observe the effect of gender on tibial shock acceleration and vibration content across different running speeds, and (ii) to assess whether mechanical stress is affected by body composition. We confirmed our hypothesis that women demonstrate greater shock intensity and vibration content at tibia than men along its longitudinal axis. Overall, we observed that the magnitude of mechanical stress, mainly in its axial dimension, tended to be affected by body mass. More precisely, indicators of mechanical stress were correlated to bone mineral content and muscle mass.

A systematic review found that women are more exposed to such overuse injuries [31]. In the present study, it has been observed that females demonstrated greater shock intensity and vibration content along the tibia axial axis at high running speeds. This result is in agreement with previous findings highlighting a 25% greater tibia axial peak acceleration in females during a pendulum task mimicking the impact force sustained in running [27]. Intrinsic characteristics can be responsible for greater stress magnitude in women. Indeed, Schinkel-Ivy et al. (2012) found that tibia peak acceleration is affected by leg fat mass, lean mass, and bone mineral content.

In the present study, small to large interactions were observed between bone mineral content and either axial or transverse shock intensity and vibration content. Low bone mineral content was previously identified as a risk factor for stress fractures in female distance runners [32]. Besides, an experiment conducted on rats demonstrated that high bone density was associated with a greater fatigue resistance [26]. We hypothesize then that greater fatigue resistance in high density bones could be partly attributed to the fact that the impact magnitude sustained by such bones is lower in dynamic tasks, as demonstrated by the present experiment and the study of Schinkel-Ivy et al. [27]. Interestingly, the outer diameter of tibia bone was identified as a significant predictor of tensile stress acting at the tibia [25]. In this sense, it has been demonstrated that runners with a history of stress fractures have lower tibia cross-sectional area [33,34] as well as thinner tibia cortex [35]. These results are also in agreement with the findings of Burkhart and collaborators who observed after a 12-month follow-up that a high ratio between fat mass and bone mineral content increased by a factor of 1.6 the risk of sustaining an injury to the distal lower extremity [36]. In other words, more bone mass than fat mass would reduce the risk of lower leg injuries. The present findings confirm that the stress magnitude applied to bone is related to skeletal properties.

From a methodological standpoint, most of previous studies recording tibial acceleration in a context of injury prevention only measured axial acceleration. The present study demonstrates that the magnitude of mechanical stress acting along the transverse axis of the tibia is lower than the stress acting along its axial axis by only 25% on

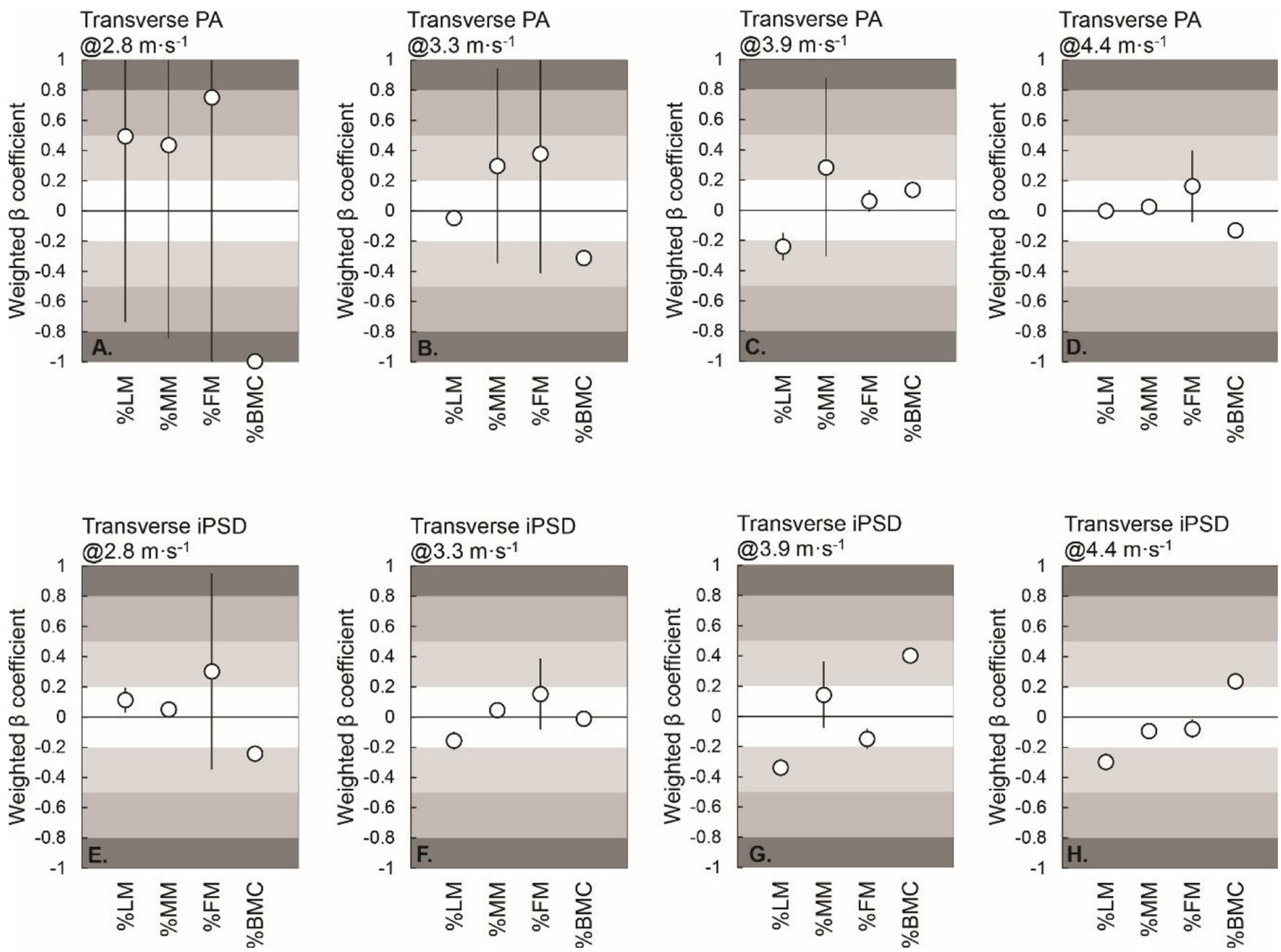

Fig. 3. Weighted β coefficients for interactions between transverse shock intensity (panels A–D) or vibration content (panels E–H) and body composition indices: lean mass (%LM), muscle mass (%MM), fat mass (%FM), and bone mineral content (%BMC). Bar errors represent 95% confidence intervals of β coefficients. Dark gray areas denote large interactions, gray areas denote moderate interactions, light gray areas denote small interactions, and white areas denote trivial interactions.

average. As a matter of fact, transverse strains should not be neglected. This last point should be taken into account in future experiments aimed at characterizing strains in running.

Besides, muscles are well known to influence impact attenuation through the 'muscle tuning' mechanism (Wakeling and Nigg [20]). This paradigm relies on the fact that muscle activity is tuned in response to ground reaction forces in order to dampen tissue vibrations. This response has been observed during static whole body continuous or pulsed vibration tasks [37,38], as well as during dynamic tasks such as walking [39] and running [21]. In the present study, vibration content along the axial axis was negatively related to muscle mass at 4.4 $m{\cdot}s^{-1}$, i.e. high vibration power is associated with low muscle mass. This relationship tends to confirm that the muscular system plays a role in vibration dampening. It also demonstrates that muscles not only participate in soft-tissues vibrations but could also influence vibration dissipation within the overall body. As previously observed, an increase in running speed is accompanied by both an increase in muscle activity, and an increase in impact and soft-tissues vibrations [21]. One can then reasonably assume that the greater muscle activity is related to a necessity to increase the active propulsive force needed to run at higher speeds, but also to an increase in the energy to be dissipated within soft-tissues [21]. This may partially explain why the association between muscle mass and vibration content has been noticed at high running speeds. From an injury perspective, this result tends to be in agreement with the study of Beck et al. (2000) highlighting that the occurrence of stress fractures is associated with a low muscles cross-sectional area. The authors concluded that weaker muscles may contribute to bone fatigue damage, and that the muscular system may serve a protective role in resisting mechanical stresses.

In a cross-sectional study performed on 93 adolescent female runners, longer duration of running participation, menstrual irregularities, lower body mass index, and lower lean mass were associated with a greater risk of bone health impairment [40]. Besides, Schinkel-Ivy et al. [27] found that high tibia peak acceleration was correlated to low leg lean mass. One would then assume that low lean mass is associated with high mechanical stress. However, in the present study, results relative to lean mass are highly inconsistent. We hypothesize that this discrepancy may be due to the fact that lean mass is a combination of several factors including, among others, bone mass and muscle mass, two body components having opposite effects on impact-related stress.

One limitation of this study is that the full body composition was assessed. Since impact properties at tibia are measured, further experiments should assess the composition of lower limb tissues. Moreover, it is worth mentioning that the present study aimed at focusing on gender differences from a body composition perspective. The present study highlights that body composition matters in impact magnitude in heel-toe runners. It is nevertheless worth mentioning that many other factors could affect impact properties as well (e.g. anthropometrics, running experience, running kinematics). Other factors should be investigated in future experiments. Regarding running kinematics, future researches should especially focus on step frequency and leg geometry at touchdown. Indeed, step frequency affects knee flexion angle prior contact [41]. Derrick demonstrated then that over-striding and greater knee contact angle were associated with greater peak leg acceleration [41].

5. Conclusion

This study showed that heel-toe female runners sustain greater shock intensity and vibration content along the tibia longitudinal axis at high running speed than their male counterparts. This gender effect could be associated to low bone mineral content and muscle mass. While the influence of bone mineral content was highlighted for low running speeds, the association between stress magnitude and muscle mass was observed for high running speeds. This may indicate that the role of the muscular system in vibration dampening increases as the running speed does, likely as a consequence of an increase in impact magnitude.

Conflict of interest statement

This study was funded by Amer Sports Footwear as part of its research work. All authors disclose any financial and personal relationships with other people or organizations.

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