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Effect of running speed on temporal and frequency indicators from wearable MEMS accelerometers

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ABSTRACT

Amplified by the development of new technologies, the interest in personal performance has been growing over the last years. Acceleration has proved to be an easy variable to collect, and was addressed in several works. However, few of them evaluate the effect of running speed on relevant indicators. The influence of the sensors location on the measurement is rarely studied as well. This study is dedicated to investigating the effect of running speed on acceleration measured at three different positions on 18 volunteers. All participants were equipped with three inertial measurement units: on the dorsal surface of the right foot (Fo), at the centre of gravity of the tibia (Ti), at the L4-L5 lumbar (Lu). The test was performed on a treadmill at nine randomised speeds between 8 and 18 km/h. Ten accelerometric variables were calculated. Linear regressions were used to calculate speed from the indicators calculated on (Lu), (Ti), (Fo). Indicators associated to signal energy were highly correlated with speed ($r^2 > 0.90$). Median frequency appears to be affected by the frequency resolution. Finally, the measurement points closest to the impact zone result in the most correlated indicators.

KEYWORDS

Biomechanics;
sports; velocity;
linear correlation

Introduction

Running has become a social phenomenon. Results published by the Outdoor Foundation (2018) indicate that 55.9 million people practised running in the USA in 2017. A similar study was conducted in England and shows that 7.0 million people run on a regular basis in 2017 (Sport England, 2018). In France, a survey by Odoxa (2018) highlights that 33% of the French population practices running. Furthermore, interest in the monitoring of individual performance is growing, amplified by the use of new technologies such as Micro-Electro-Mechanical Systems (MEMS). These systems provide enthusiasts as well as athletes with lightweight, autonomous and wireless sensors (Lowe & O'laighin, 2013). These new tools incorporate various sensors including accelerometers, which have been validated for the running activity (Buchheit, Gray, & Morin, 2015; Lee, Mellifont, & Burkett, 2010; Purcell, Channells, James, & Barrett, 2005). Acceleration patterns, derived from tri-axial accelerometers, provide important information, since they may be representative of the running cycle (Lee et al., 2010; Norris, Kenny, & Anderson, 2016; Purcell et al., 2005), stride type

(Eskofier, Musho, & Schlarb, 2013; Giandolini et al., 2014), shock attenuation and vibration transfer (Derrick, Dereu, & McLean, 2002; Shorten & Winslow, 1992), frequencies excitation (Friesenbichler, Stirling, Federolf, & Nigg, 2011; Giandolini, Gimenez, Millet, Morin, & Samozino, 2013), kinematics of the participant (Strohrmann, Harms, Kappeler-Setz, & Troster, 2012) or mechanical properties of the lower limbs (Buchheit et al., 2015). The studies cited above used parameters resulting from the acceleration to characterise the locomotion, the technique and the evolution of the athletes. These parameters have been assessed for different running conditions like speed (Lee et al., 2010; Sheerin, Besier, & Reid, 2018; Shorten & Winslow, 1992) or running surface conditions (Boey, Aeles, Schütte, & Vanwanseele, 2017; Giandolini et al., 2014; Purcell et al., 2005). Shorten and Winslow (1992) observed, in a population of twelve male volunteers, a linear increase in the spectral power of the signal with speed at the leg during foot contact. McGregor, Busa, Yaggie, and Bollt (2009) studied, for two populations of trained and untrained male volunteers, the evolution of the resultant value of the accelerometric signal on the posterior part of the body, in the lumbar region in order to approximate the centre of mass. The indicator displays a strong linear correlation ($r = 0.95$) with the oxygen consumption, which increases during an incremental speed test. This study also showed that linear correlation exists between this indicator and the running speed, yet without specifying its value. Other studies have focused on running parameters. Mercer, Vance, Hreljac, and Hamill (2002) observed a linear correlation between stride frequency and speed ($r = 0.89$) using an accelerometer placed at the leg, for eight male volunteers. Neville, Rowlands, Wixted, and James (2011) confirm these results with a strong correlation of stride frequency ($r^2 = 0.901$), using an accelerometer placed at various positions between the middle and upper thoracic vertebrae of an elite male athlete.

However, since these studies are carried out in many different conditions (performance, slope, gender), it is difficult to compare all data obtained through measurement of the acceleration in an attempt to highlight a general phenomenon. The same goes for the designation of an indicator related to the stride technique, vibratory amplitude or excitation frequencies, having the greatest sensitivity to the running speed. In addition, few studies have focused on the most sensitive measurement point in the lower limbs. Among the works cited, only the study proposed by Mercer et al. (2002) enables comparison between strike and shock indicators, although the latter are computed between two measurement points, making it difficult to conclude whether speed sensitivity occurs at one measurement point or another. The present paper aims at assessing the effect of running speed on several indicators derived from the acceleration measurement, based on three specific body locations in a population of 18 volunteers. It is first hypothesised that running speed influences temporal and frequency indicators for a non-professional population, and then that running speed affects specific indicators differently depending on the measurement points.

Methods

Participants

Eighteen volunteers (10 males and 8 females) participated in this study (age: 31.4 ± 8.9 years; height: 1.72 ± 0.09 m; mass: 64.9 ± 12.3 kg). All participants were at least recreational runners, with a minimum training frequency of two times per week and met at least one of the following criteria: 10 km official race time below 55 min, or half marathon

time below *1h50min*. Participants were selected in order to generate a broad panel encompassing the general behaviour of the human body.

Protocol

All participants were equipped with three inertial measurement units (IMU—Hikob, Villeurbanne, France—dimensions: $45 \times 36 \times 17$ mm, mass: 22g) featuring a high-frequency tri-axial accelerometer validated for running Provot, Chiementin, Oudin, Bolaers, and Murer (2017). The first one was located at the foot level, on the dorsal surface of the right shoe above the metatarsals (Fo). The second one was mounted at the centre of mass of the leg, according to anthropometric data described by Winter (2009), on the protruding part (midshaft) of the tibia (Ti). The last one was placed on the trunk near the L4-L5 space, on the line joining the two iliac crests, one axis of the IMU being aligned with the vertical axis of the body (Lu) (Figure 1). The shoe device was mounted on the lace, held in the eyelets of the housings. Tibia and lumbar devices were maintained by a Velcro strip specifically designed for the test and a medical elastic band to avoid slippage on the skin. Acquisition was performed using a sampling frequency of 1344Hz, with a maximum magnitude of ± 24 g for the foot and tibia devices, and ± 8 g for the lumbar device. All raw data were recorded on a memory card. All three IMUs were synchronised using radio frequency remote control. The running test was performed on a NordiTrack C300° treadmill (Logan, USA). In order to minimise side effects due to the equipment, all participants were asked to wear similar running shoes (Kalenji°, Ekiden One°, Villeneuve d'Ascq, France) and socks adapted to their size. Finally, heart rate was constantly monitored using a heart rate monitor (Polar°, Kempele, Finland).

Before the test, each participant was asked to warm up with two minutes of walking, followed by five minutes of running at 10 km/h to get used to the treadmill. Nine randomised tests at different speeds (8 – 9 – 10 – 11 – 12 – 13 – 14 – 16 – 18 km/h) were performed. All nine measurements were performed during a one-minute running session at the selected speed, followed by a cool-down of at least two minutes. This procedure was performed to reach the resting heart rate, as determined prior to the warm-up. All measurements were recorded over a period of 30s in a steady state within the one minute interval, removing the first and last 15 seconds of the trial, to compute the indicators mentioned below.

Data processing

Ten indicators were computed in this study, and divided in two groups and two subgroups (Figure 2). The first group was computed from the raw signal, for all three IMUs, based on resultant acceleration (Res) which is defined as the square root of the sum of the squared accelerations on each axis. Two statistical indicators were then also computed on the signal in the temporal and spectral domains. In the temporal domain, the Root Mean Square (RMS) value was computed. It is generally relevant for defining the magnitude of a varying signal, such as vibration and shock (Kobsar, Osis, Hettinga, & Ferber, 2014). In the spectral domain, the computed indicator was the spectral energy (SE), defined as the sum of the squared fast Fourier transformation (Eskofier et al., 2013).

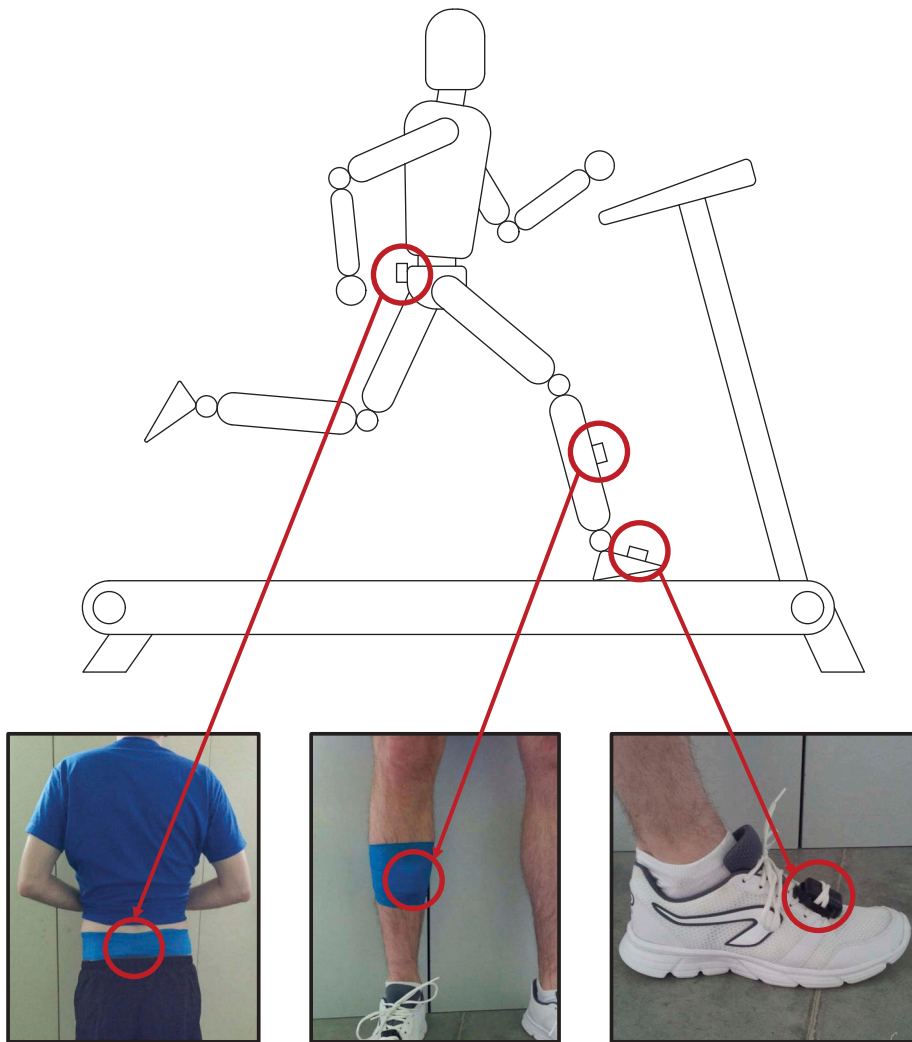


Figure 1. IMU locations on each participant's body.

Then, eight indicators were determined and averaged on accelerometric signals representing each running stride. In the temporal domain, seven of these eight stride signal indicators were computed following a specific methodology based on the longitudinal (Lon) axis of the lumbar (Lu sensor) (Gaudino, Gaudino, Alberti, & Minetti, 2013). In order to avoid discrepancies stemming from the use of different methodologies, these indicators were not computed for the tibia and foot. The first indicator was the mean stride frequency (MSF) (Purcell et al., 2005). The six last stride indicators were computed on decomposed stride signal during two phases: a contact phase where the foot is in contact with the ground, and a flight phase where the runner is suspended and has no contact with the ground. Based on the literature, the decomposition of the stride signal may be understood in two different ways. First, based on a physical (mechanical) theory, the contact phase is said to begin and end at the exact moment where the last point of the feet loses contact with the ground (Blickhan, 1989; Lee et al., 2010), which occurs when

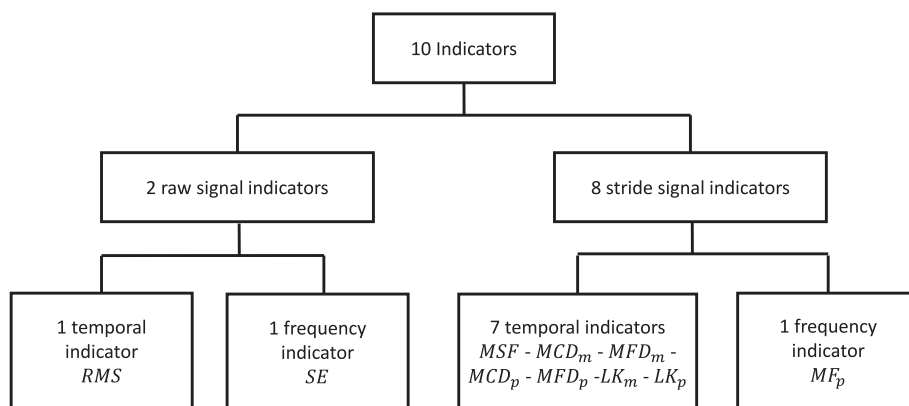


Figure 2. Different accelerometric indicators. Root mean square (RMS), Spectral energy (SE), Mean stride frequency (MSF), Mean Contact Duration (mechanical) (MCD_m), Mean Flight Duration (mechanical) (MFD_m), Mean Contact Duration (physiological) (MCD_p), Mean Flight Duration (physiological) (MFD_p), Mechanical leg stiffness (LK_m), Stable leg stiffness (LK_p), Median frequency for stable contact (MF_p).

the ground reaction force reaches zero. Second, based on a physiological theory, the contact phase is considered to begin and end at the exact moment where body weight is supported or not by the ground (Cavagna, 1970; Gaudino et al., 2013), which occurs when the ground reaction force equals the body weight. These parameters are computed using the longitudinal acceleration signal of the lumbar, following the matching of accelerometric signals and force plate proposed by Gaudino et al. (2013). The six indicators were the mean contact duration, computed for the mechanical (MCD_m) and physiological (MCD_p) theories; the mean flight durations (MFD_m and MFD_p); the leg stiffness described by a bi-axial spring mass model (Buchheit et al., 2015; Morin, Dalleau, KyröLäinen, Jeannin, & Belli, 2005) as a function of the contact and flight durations (LK_m and LK_p). In the spectral domain, the last stride indicators were computed using the fast Fourier transformation of the stride signal, relying on the resultant acceleration of each sensor (Res) and following the decomposition performed at the lumbar and extended to the other points as the sensors were synchronised. This last indicator was the median frequency of the stable contact (MF_p), as described by Giandolini, Pavailler, Samozino, Morin, and Horvais (2015). All ten indicators are summarised in Table 1. To ensure that test conditions have no significant influence on any indicator, the intra and inter test repeatabilities of all ten indicators were validated in a previous study for one participant, over two series of ten consecutive measurements at constant conditions, with and without removal of the sensors (Provot, Munera, Bollaers, Vitry, & Chiementin, 2016).

Statistical analysis

Two numerical tests were performed using Matlab R2017b (MathWorks, Natick, USA) in an attempt to assess the general behaviour of all eighteen participants with respect to speed. First, a Friedman test was carried out on each indicator computed for all eighteen participants at the nine different running speeds mentioned earlier. A p -value below 0.05 corresponds to a significant effect of the speed on the measured indicator for all participants. Then, based on the works of McGregor et al. (2009), which conclude that

Table 1. Details of the ten indicators computed for three different locations: Foot (Fo), Tibia (Ti) and Lumbar (Lu) computed for the longitudinal (Lon) axis of the human segment or the resultant acceleration (Res).

Indicator	Notation	Point	Axis	Reference
Root mean square [m/s^2]	RMS	Fo/Ti/Lu	Res	Kobsar et al. (2014)
Spectral energy [$(\text{m/s}^2)^2$]	SE	Fo/Ti/Lu	Res	Eskofier et al. (2013)
Mean stride frequency [Hz]	MSF	Lu	Lon	Mercer et al. (2002)
Mean Contact Duration (mechanical) [s]	MCD_m	Lu	Lon	Lee et al. (2010)
Mean Flight Duration (mechanical) [s]	MFD_m	Lu	Lon	Lee et al. (2010)
Mean Contact Duration (physiological) [s]	MCD_p	Lu	Lon	Cavagna (1970)
Mean Flight Duration (physiological) [s]	MFD_p	Lu	Lon	Cavagna (1970)
Mechanical leg stiffness [N/mm]	LK_m	Lu	Lon	Morin et al. (2005)
Stable leg stiffness [N/mm]	LK_p	Lu	Lon	Morin et al. (2005)
Median frequency for stable contact [Hz]	MF_p	Fo/Ti/Lu	Res	Giandolini et al. (2015)

a linear correlation study is sufficient, a Bravais-Pearson test was performed for each indicator and each participant compared to speed, defining eighteen correlation coefficients r_i ($i = 1$ to 18) for each indicator. For each indicator, a squared mean correlation coefficient, r_m^2 (Eq. 1) was calculated in order to estimate the overall linear trend of the indicators. Finally, an accuracy rate (Acc) was computed as the ratio between the number of linear participants (Bravais-Pearson p -value < 0.05) and the total number of participants, in an attempt to investigate the linearity correlation for all participants. The accuracy rate was presented as a percentage.

$$r_m^2 = \left(\frac{\sum_{i=1}^{18} r_i}{18} \right)^2 \quad (1)$$

Results

Results of the study are present in Table 2, and showed that raw signal indicators related to the energy of the signal (RMS and SE, Figure 3) presented an accuracy of 100% for each case. In all participants, indicators were considered linear compared to speed with $r_m^2 > 0.90$. These indicators displayed good linearity, especially for Ti and Fo.

The indicator associated to the complete stride (MSF, Figure 4), which was computed only based on the longitudinal axis of Lu, presented a good linearity ($r_m^2 > 0.93$) with perfect accuracy (100%).

The four indicators related to the flight and contact duration (MCD_m , MFD_m , MCD_p and MFD_p ; Figure 5) presented very different results. On the one hand, the two indicators related to the decomposition of mechanical events yield opposite findings: MCD_m had acceptable linearity ($r_m^2 = 0.89$) and excellent accuracy (100%); while MFD_m seemed to be independent on speed (p -value = 0.172). On the other hand, indicators based on the decomposition of physiological events presented, for both indicators, poor linearity ($r_m^2 = 0.55$) and accuracy ($> 70\%$).

Two indicators were associated with leg stiffness (LK_m and LK_p ; Figure 6). Both indicators presented average accuracy (77.8% for LK_m , 72.2% for LK_p). However, stiffness based on physiological contact events (LK_p) resulted in poor linearity ($r_m^2 = 0.30$) compared to that based on mechanical contact events (LK_m) with average linearity ($r_m^2 = 0.70$).

Table 2. Results of the speed influence protocol for the indicators computed at three different locations: Foot (Fo), Tibia (Ti) and Lumbar (Lu); following the longitudinal (Lon) axis of the human segment or the resultant acceleration (Res). Missing values (-) indicate a Friedmann p -value above 0.05, i.e., no influence of the running speed.

Indicators	Point	Axis	r_m^2	Acc (%)
RMS [m/s^2]	Fo	Res	0.98	100
RMS [m/s^2]	Ti	Res	0.98	100
SE [$(m/s^2)^2$]	Ti	Res	0.94	100
SE [$(m/s^2)^2$]	Fo	Res	0.94	100
RMS [m/s^2]	Lu	Res	0.94	100
SE [$(m/s^2)^2$]	Lu	Res	0.93	100
MSF [Hz]	Lu	Lon	0.93	100
MCD _m [s]	Lu	Lon	0.89	100
LK _m [N/mm]	Lu	Lon	0.70	77.8
MFD _p [s]	Lu	Lon	0.55	72.2
MCD _p [s]	Lu	Lon	0.55	77.8
LK _p [N/mm]	Lu	Lon	0.30	72.2
MF _p [Hz]	Ti	Res	0.03	11.1
MF _p [Hz]	Lu	Res	0.03	11.1
MF _p [Hz]	Fo	Res	—	—
MFD _m [s]	Lu	Lon	—	—

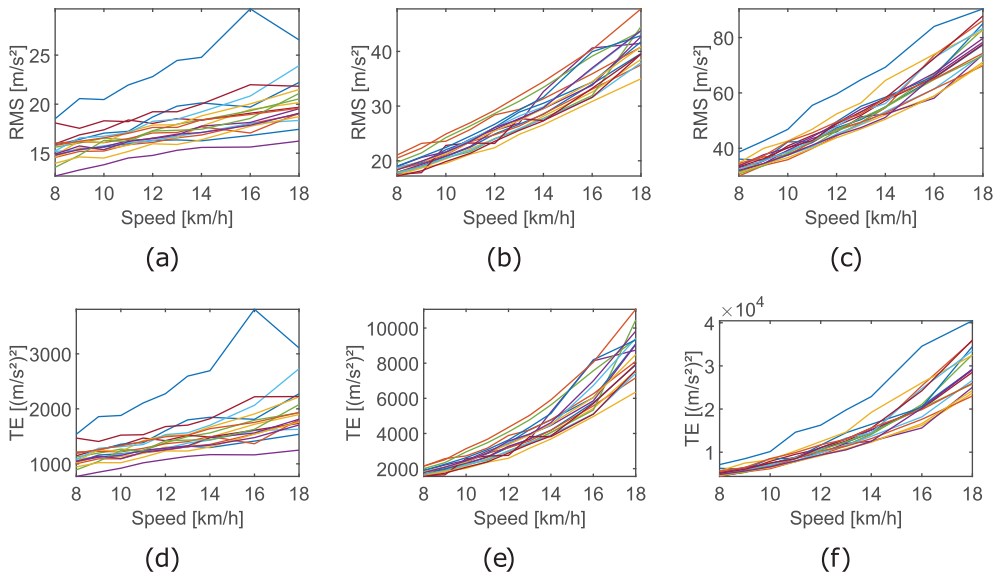


Figure 3. Top—RMS versus running speed for the 18 participants (a) for Lu ($r_m^2 = 0.94$; Acc = 100%), (b) for Ti ($r_m^2 = 0.98$; Acc = 100%), (c) for Fo ($r_m^2 = 0.98$; Acc = 100%). Bottom—SE versus running speed for the 18 participants (e) for Lu ($r_m^2 = 0.93$; Acc = 100%), (f) for Ti ($r_m^2 = 0.94$; Acc = 100%), (g) for Fo ($r_m^2 = 0.94$; Acc = 100%).

The last indicator of interest was MF_p (Figure 7). The associated accuracy was below 15%, and a linearity is also poor ($r_m^2 < 0.05$). In one case (Fo sensor), p -value was greater than 0.05 (p -value = 0.401), reflecting an insignificant effect of speed on the indicators.

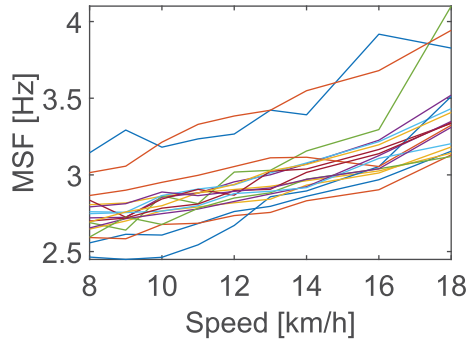


Figure 4. MSF versus running speed for the 18 participants, Lu sensor ($r_m^2 = 0.93$; $Acc = 100\%$).

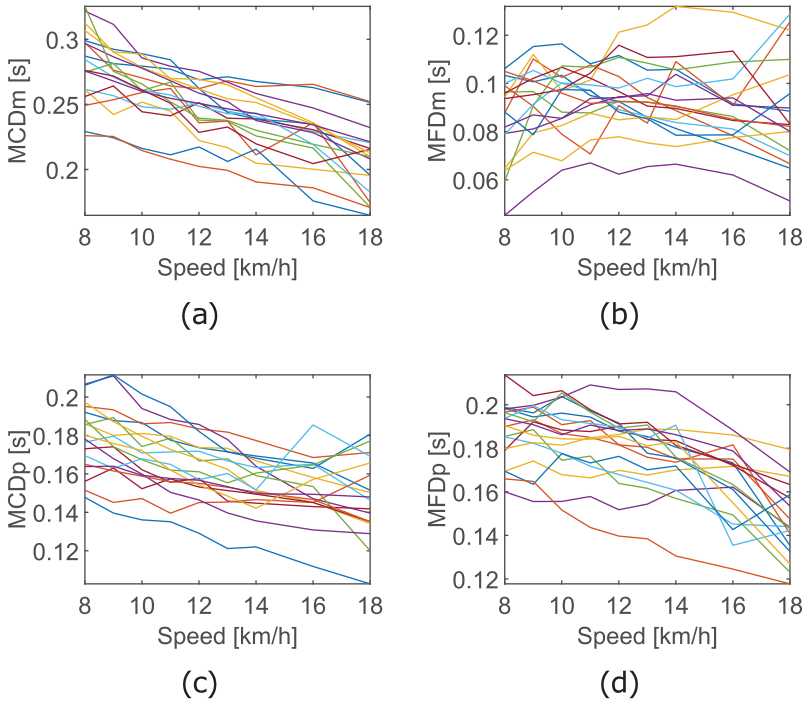


Figure 5. Decomposed stride indicator versus running speed for the 18 participants (a) MCD_m ($r_m^2 = 0.89$; $Acc = 100\%$), (b) MFD_m (not correlated), (c) MCD_p ($r_m^2 = 0.55$; $Acc = 77.8\%$), (d) MFD_p ($r_m^2 = 0.55$; $Acc = 72.2\%$).

Discussion and implications

Firstly, this study focused on the effect of the running speed using indicators computed from acceleration on a population of non-professional runners. The most correlated indicators are those directly related to the energy contained in the accelerometer signal, such as RMS and SE. Indicator SE could however be improved using quadratic correlation, according to the curve trend. These results are consistent with other studies, particularly those by McGregor et al. (2009), highlighting a correlation between the

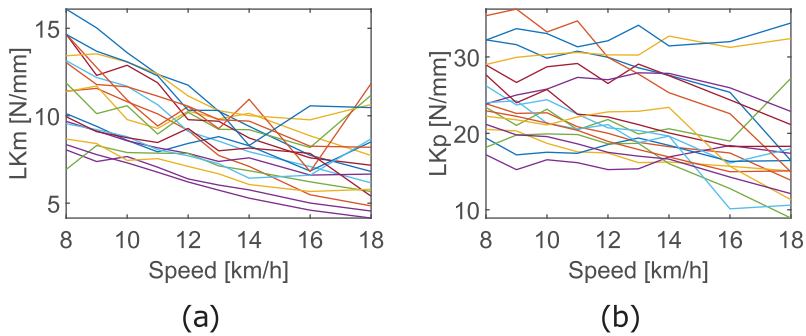


Figure 6. Leg stiffness versus running speed for the 18 participants (a) LK_m ($r_m^2 = 0.70$; $Acc = 77.8\%$), (b) LK_p ($r_m^2 = 0.30$; $Acc = 72.2\%$).

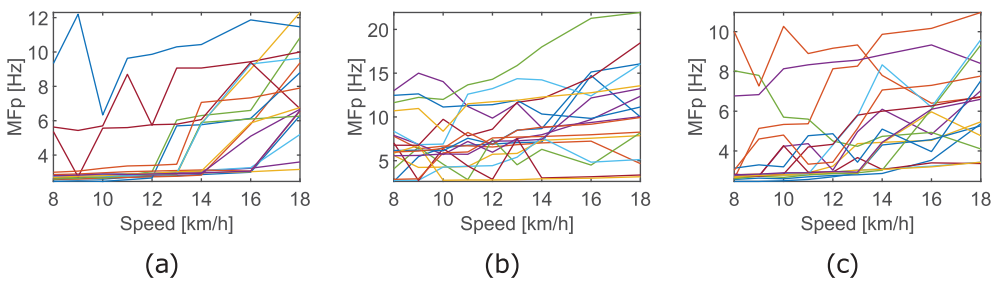


Figure 7. MF_p versus running speed for the 18 participants (a) for Lu ($r_m^2 = 0.03$; $Acc = 11.1\%$), (b) for Ti ($r_m^2 = 0.03$; $Acc = 11.1\%$), (c) for Fo (not correlated).

resultant acceleration at the lumbar and V_{O_2} , which increased linearly with speed. The trend is very similar to the results of RMS versus speed in the present study ($r^2 > 0.90$). The running speed linearly affects MSF with a correlation of $r_m^2 = 0.93$ as well. It confirms the results of other studies on similar population sizes (Mercer et al., 2002; Wixted et al., 2005). If studies have shown that stride frequency differs between treadmill and overground studies (Riley et al., 2008; Schache et al., 2001), the linear trend of this indicator with speed has already been observed in outdoor studies, yet with a lower correlation ($r = 0.80$) (Giandolini et al., 2015).

The study described herein also shows that other indicators present reduced linear correlation. Among these indicators, it is possible to observe two behaviours. The first one is that the two signal decompositions lead to different results. With regards to the mechanical decomposition, a strong influence of the running speed on MCD_m is actually observed, but not for MFD_m (Figure 5). For physiological decomposition, both MCD_p and MFD_p indicators display average correlation and accuracy (Figure 5). It is noteworthy that depending on the stride decomposition method, two parts of the signal may be differentiated: one which varies linearly with speed, an other which is not impacted by running speed and seems constant. One possible explanation may be that as speed increases, the ground contact time decreases while the flight time remains constant. Similar results for the decrease in contact time are presented by Novacheck (1998). Separation of the signal based on the decomposition method may be explained by the

fact that physiological decomposition determines the duration of contact over a range narrower than mechanical decomposition. Indeed, the latter considers contact as the total duration of a ground reaction force. In this case, the entire impact is taken into account, therefore all the energy contained in the accelerometer signal. Thus the mechanical decomposition isolates the part of the stride signal presenting the highest energy content and the physiological decomposition shares this content between the contact and flight phases. These results affect the stiffness correlations LK_m and LK_p (Figure 6) which are calculated using the squared contact time (Morin et al., 2005). However, the correlations for LK_m and LK_p display a negative trend with increasing running speed. These observations contradict the conclusions of Arampatzis, Brüggemann, and Metzler (1999), who implement the same model but where the ground reaction force is measured and not computed by the model. The increase in the measured force shows a greater influence on the leg stiffness than the increase in speed itself. This trend is not observed in the present study, where the force estimation is carried out using a numerical model in conjunction with an accelerometer. These tools are then to be used carefully for estimating contact efforts and require further validation. The second behaviour is associated to MF_p (Figure 7), the plots of which levelling out for given speed ranges, especially at the lumbar level. This is accounted for in part by the strong sensitivity of the indicator to spectral resolution and harmonic distribution. Indeed, the spectrum is represented by a sequence of peaks associated with the amplitude of each excitation frequency. When resolution is too low, it is not possible to compute the median frequency with sufficient accuracy: it is therefore associated with the nearest measured frequency. Consequently, this indicator turns out to be very difficult to implement and requires optimal parametrisation of the measurement and processing. Excessive sensitivity to spectral resolution has already been observed on this indicator in a previous study (Provot et al., 2016). One possible means of improving the accuracy of this indicator is to compute it on signals of longer duration, with the aim of achieving higher spectral resolution.

Secondly, this article focuses on the varying influence of running speed for different measurement points. Examination of the two indicators computed for the three locations and having an important influence of the running speed (RMS and SE) indicates that correlations are higher for the foot and tibia than for the lumbar. The explanation is twofold. On the one hand, the sensor is located at the lumbar, thus far from the impact zone. Shock is necessarily attenuated, along with the energy of the accelerometric signal. Several studies have shown an increase in shock attenuation across the body with speed (Mercer et al., 2002; Shorten & Winslow, 1992). This attenuation of energy in the lower limbs therefore explains the lower correlation observed at the lumbar. On the other hand, the mounting of the sensor may also be involved. Indeed, the fitting to the lumbar by means of an elastic band is very sensitive to the fat mass of the participant. Several parameters can influence the acceleration measurement, such as the thickness of soft tissues, the mass of the sensors and the pre-loading (Nokes, Fairclough, Mintowt-Czyz, Mackie, & Williams, 1984; Ziegert & Lewis, 1979). If these first results show that indicators issued from the lumbar are less influenced by the running speed, this measurement point provides valuable information in terms of time stride parameters. As already mentioned, MSF presents excellent correlation ($r_m^2 > 0.90$) which is confirmed by the work of Neville et al. (2011) with a correlation of ($r^2 = 0.901$).

This study provides better understanding in how running speed influences indicators for a population of non-professional runners. First, the results related to the influence of running on temporal and frequency indicators could bring additional insight into in situ studies when running speed is hard to control. Future studies could be performed in an attempt to better understand the influence of sport shoes or field, particularly through indicators describing the transmission of shock energy into the body like RMS and SE. Then, results show that running speed affects specific indicators differently for three different measurement points, which may guide future studies. Whether focused on running kinematics or shock dynamics, it will be necessary to choose the most adapted measuring point, but also to select an accelerometer with the most adequate characteristics.

Yet the works detailed herein present some limitations, mostly because the influence of speed influence was estimated on a treadmill which offers controlled running speed, but also alters the running pattern. Results may slightly vary in real running conditions. Moreover, this study focused on a limited group of indicators, with results revealing an important dependency on the accuracy of the algorithm deployed. If accelerometers show potential to extract an important number of features, as presented in the study of Benson, Clermont, Osis, Kobsar, and Ferber (2018), there remains important to work on the quality and the significance of the data. Finally, the effect of speed is only investigated by means of a linear trend which extends our knowledge in the field for non-professional runners, but could be limited for indicators linked to the mechanical energy and displaying quadratic evolution.

Conclusions

This paper has underlined the excellent correlation of parameters resulting from acceleration measurements with the running speed. It also validated the hypothesis that running speed modifies both temporal and frequency indicators. These indicators could be used in the frame of a study without controlled speed conditions, and are not specific to a professional athletes panel. Despite promising preliminary results, future studies should target real motor conditions (e.g., athletics race, mountain trails) and overground running, in an attempt to minimise biases stemming from laboratory measurements. This study also confirmed that running speed influences specific indicators in various ways depending on the measurement points. Although the measurement points closest to the impact zone have the most correlated indicators, a more in-depth study is required, each measurement point yielding different information. The indicators validated herein have a major advantage, in the sense that they allow an overall study of the behaviour of the human body during the practice of running. One potential application would be to use them in the frame of fatigue, equipment or field investigations.

Disclosure statement

No potential conflict of interest was reported by the authors.

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