



Short communication

Validity and reliability of peak tibial accelerations as real-time measure of impact loading during over-ground rearfoot running at different speeds

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ARTICLE INFO

Article history:

Accepted 22 January 2019

Keywords:

Running biomechanics

Tibial shock

Impact

Validation

Wearable

ABSTRACT

Studies seeking to determine the effects of gait retraining through biofeedback on peak tibial acceleration (PTA) assume that this biometric trait is a valid measure of impact loading that is reliable both within and between sessions. However, reliability and validity data were lacking for axial and resultant PTAs along the speed range of over-ground endurance running. A wearable system was developed to continuously measure 3D tibial acceleration and to detect PTAs in real-time. Thirteen rearfoot runners ran at 2.55, 3.20 and 5.10 m·s⁻¹ over an instrumented runway in two sessions with re-attachment of the system. Intraclass correlation coefficients (ICCs) were used to determine within-session reliability. Repeatability was evaluated by paired T-tests and ICCs. Concerning validity, axial and resultant PTAs were correlated to the peak vertical impact loading rate (LR) of the ground reaction force. Additionally, speed should affect impact loading magnitude. Hence, magnitudes were compared across speeds by RM-ANOVA. Within a session, ICCs were over 0.90 and reasonable for clinical measurements. Between sessions, the magnitudes remained statistically similar with ICCs ranging from 0.50 to 0.59 for axial PTA and from 0.53 to 0.81 for resultant PTA. Peak accelerations of the lower leg segment correlated to LR with larger coefficients for axial PTA (r range: 0.64–0.84) than for the resultant PTA per speed condition. The magnitude of each impact measure increased with speed. These data suggest that PTAs registered per stand-alone system can be useful during level, over-ground, rearfoot running to evaluate impact loading in the time domain when force platforms are unavailable in studies with repeated measurements.

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1. Introduction

Impact reduction by treadmill-based gait retraining led to less running-related injuries in novice runners (Chan et al., 2017). Treadmill running however can affect running characteristics (Chambon et al., 2014; Nigg et al., 1995), urging an over-ground approach. Over-ground gait retraining targeting impact reduction requires a wearable system that permits real-time detection of an impact measure in the time domain. Peak tibial acceleration (PTA) may be a good candidate there axial PTA best correlated to the instantaneous vertical loading rate (LR) of the ground reaction force (Greenhalgh et al., 2012; Hennig et al., 1993; Loughton et al., 2003). This LR variable has been proposed as the preferred method for calculating time-domain based impact loading in lab setting (Ueda et al., 2016) and is more sensitive to changes in the higher frequency component of the ver-

tical ground reaction force than the vertical impact peak (Shorten and Mientjes, 2011). Moreover, LR could distinguish groups of rearfoot runners with and without a history of tibial stress fracture whereas the vertical impact peak and the active peak could not (van der Worp et al., 2016; Zadpoor and Nikooyan, 2011). Axial PTA was also well above average in rearfoot runners with a previous tibial stress fracture (Milner et al., 2006).

Axial PTA has been used as biofeedback variable for treadmill-based gait retraining, successfully targeting impact reduction over various sessions (Clansey et al., 2014; Crowell and Davis, 2011). More recently, the resultant PTA was used as the biofeedback variable for lower impact running (Wood and Kipp, 2014). Resultant PTA is the peak acceleration of the norm of the tibial acceleration vector, and thus incorporates the three orthogonal components, which might be of greater importance in impact assessment than only axial (Giandolini et al., 2016). There namely occurs a sudden change in velocity of the lower leg segment in both vertical and horizontal directions (Lafortune, 1991), and both may contribute to the

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overall severity of lower leg impact. Resultant PTA seems reliable during treadmill running (Sheerin et al., 2017), although its reliability has to be tested when running over-ground at different speeds.

Over-ground gait retraining by biofeedback on PTA preferably happens on a PTA component that is valid and reliable across a wide range of speeds. In this paper, we present validity-reliability data of a wearable system able to continuously measure three-dimensional tibial acceleration for real-time detection of axial or resultant PTAs during over-ground running. To assess the measurement quality and potential usability for real-time biofeedback on time-domain based impact loading in over-ground setting, the reliability within-session and repeatability between-sessions were determined for axial and resultant PTAs. Criterion validity was assessed by comparing axial PTA and resultant PTA to the vertical LR across speeds. Additionally, as a functional verification, the system should measure higher PTAs when speed increases since running speed influences PTAs and LR (Boey et al., 2016; Lafortune, 1991; Nigg et al., 1987). Reasonable agreement for clinical measurements was expected for within-session reliability of the impact measures with no substantial differences between-sessions at all speeds. We also hypothesized that axial and resultant PTAs would be correlated to the LR, and that PTAs measured by this wearable system would increase in magnitude as running speed increased.

2. Methods

2.1. Participants

Thirteen uninjured rearfoot runners (height: 1.75 ± 0.08 m, weight: 70.6 ± 10.8 kg, reported training pace: 3.2 ± 0.5 m·s⁻¹, mean \pm SD) completed the study (appendix A). Ethical approval was obtained from the local ethical committee (2015/0864). Written informed consent was acquired.

2.2. Wearable system supporting time-synchronized registration of shock

Two lightweight, low-power MEMS tri-axial accelerometers (LIS331, Sparkfun, Colorado, USA; 1000 Hz, ± 24 g, 0.002 gram) (STMicroelectronics, 2011) were fitted in a shrink socket (total mass less than 3 grams) and are part of a remote-controlled backpack system ($mas_{total} = 1.6$ kg, appendix A). We designed an algorithm to detect PTA in real-time and implemented this in a custom-built Java program. A footfall was assumed when registering a value above 3 g (threshold identical to Clansley et al., 2014) with no larger value appearing in the next 0.375 s.

2.3. Experimental protocol

The attachment site was similar to previous research (Clansley et al., 2014; Gruber et al., 2014; Laughton et al., 2003). Skin pre-tension was applied in order to counteract an overestimation of PTA due to unwanted oscillations and can improve the axial PTA - LR relationship (Clansley et al., 2014; Pearsall et al., 2002). Hence, skin was pre-stretched through non-elastic taping (appendix A). Each accelerometer was placed on a lower leg alongside the distal anteromedial aspect, ~ 8 cm above the medial malleolus (Clansley et al., 2014), and its axial axis was aligned with the long axis of the tibia by palpation. Tape was tightly fastened by one of the test leaders over the accelerometers and around the lower legs to the limit of subject tolerance. Standardized neutral distance running shoes (Li Ning Magne, ARHF041) as in Breine et al. (2014) and lycra clothing were provided. Insoles were not allowed.

Participants ran along a 32-m long running track with a build-in force platform (2.1*0.5-m, AMTI, Watertown, MA; 1000 Hz). They performed a self-selected 5 min warm-up; serving as familiariza-

tion to the experimental setup. A pressure plate (Footscan, RScan International, Olen, Belgium; 500 Hz) was mounted on top of the 2-m force platform. Dynamically calibrated plantar pressures of the foot-shoe system permitted a direct qualitative assessment of the footfall pattern in Footscan v7.105 software. The foot strike index was not calculated as we observed that this group of runners were heel strikers who made no categorical shift. The video shows pressure measurements of a representative subject across the speeds (supplementary video). Runners were instructed to run at three randomly assigned speeds: 2.55, 3.20 and 5.10 ± 0.2 m·s⁻¹; providing a continuum of endurance running (Bramble and Lieberman, 2004). The lowest speed represents slow jogging. The doubled, highest speed condition corresponds to an elite marathon pace. The 3.20 m·s⁻¹ condition stems with the participants' self-reported training speed and the self-selected running pace of healthy gait retrainers (Willy et al., 2015). Timing gates, situated 6.85 m apart, monitored speed. Verbal feedback on speed was given trial-by-trial. After 13-m along the track, motion capture cameras were triggered and ground reaction forces were simultaneously recorded for 2.5 s. Participants rested ~ 90 s between trials. Four left footfalls on the force platform were deemed sufficient per speed per subject there three footfalls per participant per speed has already resulted in ICC values higher than 0.80 for LR (Breine et al., 2014), indicating low inter-trial variability within-session. We used the largest number of footfalls available for synchronization in all conditions. Subjects performed a second test session where the protocol was repeated in the exact order multiple days (range: 3–72, mean and median: 23) later.

2.4. Data processing

Tibial acceleration, ground reaction forces and detected PTAs were imported in MATLAB using custom scripts. Time series of ground reaction forces were low-pass filtered using a second order zero-lag Butterworth filter with a cut-off frequency of 60 Hz. For each left footfall on the force platform (unilateral analysis), initial contact and toe off were defined when the vertical ground reaction force deflected ≥ 5 N from baseline. The instantaneous vertical LR was calculated as the maximum slope of the vertical ground reaction force curve over an interval of 0.004 s within the first 0.050 s of stance (Breine et al., 2014).

Based on the infrared triggering (appendix A), a subsample of tibial acceleration was synchronized up to millisecond precision to the ground reaction force. This allowed to extract time series of tibial acceleration during stance (Fig. 1). Anticipating the use of PTAs for real-time biofeedback during running, 3D tibial acceleration remained unfiltered as this approach requires the least complex signal conditioning; permitting biofeedback on PTAs upon direct registration (Wood and Kipp, 2014). Individual tibial accelerations are provided (appendix A) showing that the curve of the axial tibial acceleration at 3.2 m·s⁻¹ resembled to that of Laughton et al. (2003), who also used a lightweight accelerometer with tight fixation to the skin, during rearfoot running at 3.7 m·s⁻¹. Resultant tibial acceleration was calculated as the vector sum of the three-dimensional accelerations. Axial and resultant PTAs were based on the positive peak value during stance. The magnitudes of these post-hoc determined axial and resultant PTAs and those detected real-time by the system's algorithm were inspected for coherence and showed full agreement.

2.5. Statistical analyses

The within-subject reliability of the impact measures was assessed for the trials at each speed in the first session by calculating the intraclass correlation coefficient (ICC_{2,4}) using a two-way random effects model (absolute agreement; average measures).

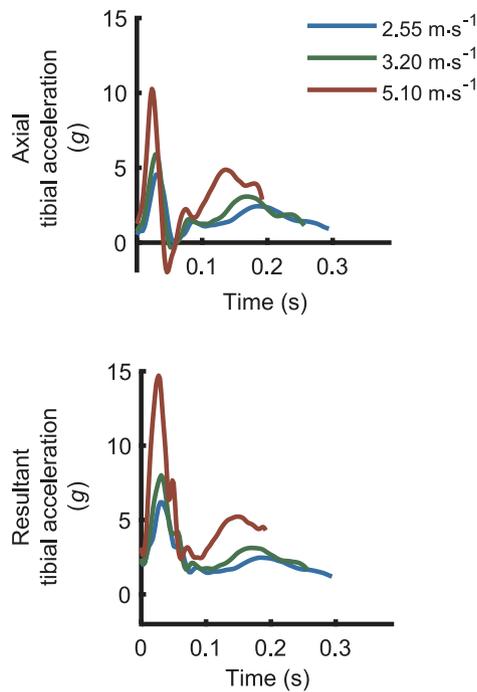


Fig. 1. Time series of (a) axial and (b) resultant tibial accelerations. Averaged signals of unfiltered tibial acceleration during the stance phase of running at 2.55 (blue), 3.20 (green) and 5.10 (red) $\text{m}\cdot\text{s}^{-1}$. Stance was normalized to mean contact time. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

The dependent variables were then averaged across the footfalls of each participant for each test session. A paired sample *t*-test evaluated any systematic difference for each speed condition between both sessions (Milner et al., 2011). Test-retest reliability was evaluated by $\text{ICC}_{2,1}$ (absolute agreement; single measures). Each ICC value was interpreted as <0.75 poor-to-moderate, ≥ 0.75 good, and >0.90 reasonable agreement for clinical measurements (Portney and Watkins, 2009). The minimum detectable change was calculated using the equation $\text{MDC}_{95} = 1.96 \cdot \sqrt{2} \cdot [\text{SD} \cdot \sqrt{1 - \text{ICC}}]$. SD is the mean within-subject standard deviation of an impact measure at session 1 and ICC stands for the measure's reliability coefficient between a test and retest condition. Bland–Altman plots were created with 95% limits of agreement to visualize agreement for PTAs between test-retest. Validity of PTAs was assessed by Pearson correlation (e.g. Hennig et al., 1993; Laughton et al., 2003), and this between LR and axial as resultant PTAs for all conditions in the first session. These metrics are available in the accompanying online dataset (Mendeley Data, datasets/73wjmpf9kr/1). Its coefficient was categorized using the modified scale of Hopkins (2000) with $r \geq 0.30$ moderate, ≥ 0.50 large, ≥ 0.70 very large and ≥ 0.90 nearly perfect, =1 perfect. The effect of speed on PTAs was examined for the first session by performing repeated measures ANOVA with speed (3-levels) as within-subject factor and Bonferroni-corrected post-hoc comparison. The alpha level and confidence level were respectively set at 0.05 and 95% for analyses executed in SPSS (IBM Corp., Armonk, NY, USA). Results are reported as mean \pm SD.

3. Results

3.1. Reliability of impact measures

Within a session PTAs and LR showed reasonable agreement at different speeds (ICC range: 0.92–0.95) (Table 1A). At 3.20 $\text{m}\cdot\text{s}^{-1}$,

Table 1

Intraclass correlation coefficient (ICC) and minimum detectable change (MDC_{95}) values of impact measures across a continuum of running speeds. PTA: Peak Tibial Acceleration; LR: peak Loading Rate of the vertical ground reaction force.

Variable	2.55 $\text{m}\cdot\text{s}^{-1}$	3.20 $\text{m}\cdot\text{s}^{-1}$	5.10 $\text{m}\cdot\text{s}^{-1}$
(A) Within-session			
	$\text{ICC}_{2,4}$		
Axial PTA	0.94	0.95	0.92
	Reasonable	Reasonable	Reasonable
Resultant PTA	0.95	0.92	0.94
	Reasonable	Reasonable	Reasonable
LR	0.95	0.92	0.92
	Reasonable	Reasonable	Reasonable
	MDC_{95}		
Axial PTA	0.4	0.5	1.2
Resultant PTA	0.6	1.1	1.4
LR	4.8	9.4	17.1
(B) Between-sessions			
	$\text{ICC}_{2,1}$		
Axial PTA	0.56	0.59	0.50
	Poor-to-moderate	Poor-to-moderate	Poor-to-moderate
Resultant PTA	0.81	0.81	0.53
	Good	Good	Poor-to-moderate
LR	0.87	0.69	0.87
	Good	Poor-to-moderate	Good
	MDC_{95}		
Axial PTA	1.1	1.4	2.7
Resultant PTA	1.1	1.8	4.1
LR	8.0	18.2	22.6

the within-subject axial PTA ranged from 4.5 ± 0.8 g to 11.4 ± 1.2 g and the resultant PTA ranged from 6.3 ± 0.6 g to 16.1 ± 4.7 g. Test-retest magnitudes of the impact measures remained statistically unchanged (*p* range: 0.102–0.648) (Table 2). Table 1B shows the test-retest reliability of 4 analysed contacts per speed per session. Only resultant PTA achieved good repeatability scores but not at the highest speed. Fig. 2 provides Bland–Altman plots.

3.2. Validity of peak tibial accelerations as surrogate measure of impact loading

Both axial and resultant PTAs correlated ($p < 0.001$) to LR within each running condition (Table 3). Correlations coefficients were generally stronger for axial (large to nearly perfect) than resultant PTA (large). There was a significant main effect of running speed on PTAs (axial PTA: $F = 69.211$, $p < 0.001$, partial $\eta^2 = 0.85$; resultant PTA: $F = 76.693$, $p < 0.001$, partial $\eta^2 = 0.87$). From the lowest to the highest speed condition ($p \leq 0.001$ – 0.002), axial PTA and resultant PTA respectively increased 25% and 30% at 3.2 $\text{m}\cdot\text{s}^{-1}$, 123% and 134% at 5.1 $\text{m}\cdot\text{s}^{-1}$. LR was also affected by speed ($F = 82.248$, $p < 0.001$, partial $\eta^2 = 0.87$) with increments of 23% at 3.2 $\text{m}\cdot\text{s}^{-1}$ and 115% at 5.1 $\text{m}\cdot\text{s}^{-1}$.

4. Discussion

The purpose of this study was to determine the reliability and validity of PTAs measured by a wearable system during over-ground running across a wide range of speeds. Supporting the hypothesis regarding reliability, reasonable inter-trial reliability of axial and resultant PTAs was confirmed within-session for clinical measurements. No statistically significant difference in PTAs was found between-sessions, although the repeatability of resultant PTA was superior to axial PTA. During treadmill running, Sheerin et al. (2016) also found better repeatability for resultant PTA up to the tested speed of 3.7 $\text{m}\cdot\text{s}^{-1}$. Despite the poor-to-moderate reliability for axial PTA between-sessions, the minimum detectable difference in axial PTA at the speed resembling training

Table 2
Test-retest values of axial and resultant PTAs and the LR at the tested running speeds. Mean ± SD.

Variable	2.55 m·s ⁻¹		3.20 m·s ⁻¹		5.10 m·s ⁻¹	
	Test	Retest	Test	Retest	Test	Retest
Axial PTA (g)	5.6 ± 1.3	6.2 ± 1.3	7.0 ± 2.0	7.7 ± 1.4	12.5 ± 3.0	13.3 ± 3.4
Resultant PTA (g)	7.7 ± 2.4	8.1 ± 2.3	10.0 ± 3.2	10.2 ± 2.4	18.0 ± 4.8	19.4 ± 4.3
LR (BW/s)	84.9 ± 19.3	88.4 ± 20.3	104.5 ± 4.7	107.0 ± 1.3	182.2 ± 7.2	192.2 ± 6.1

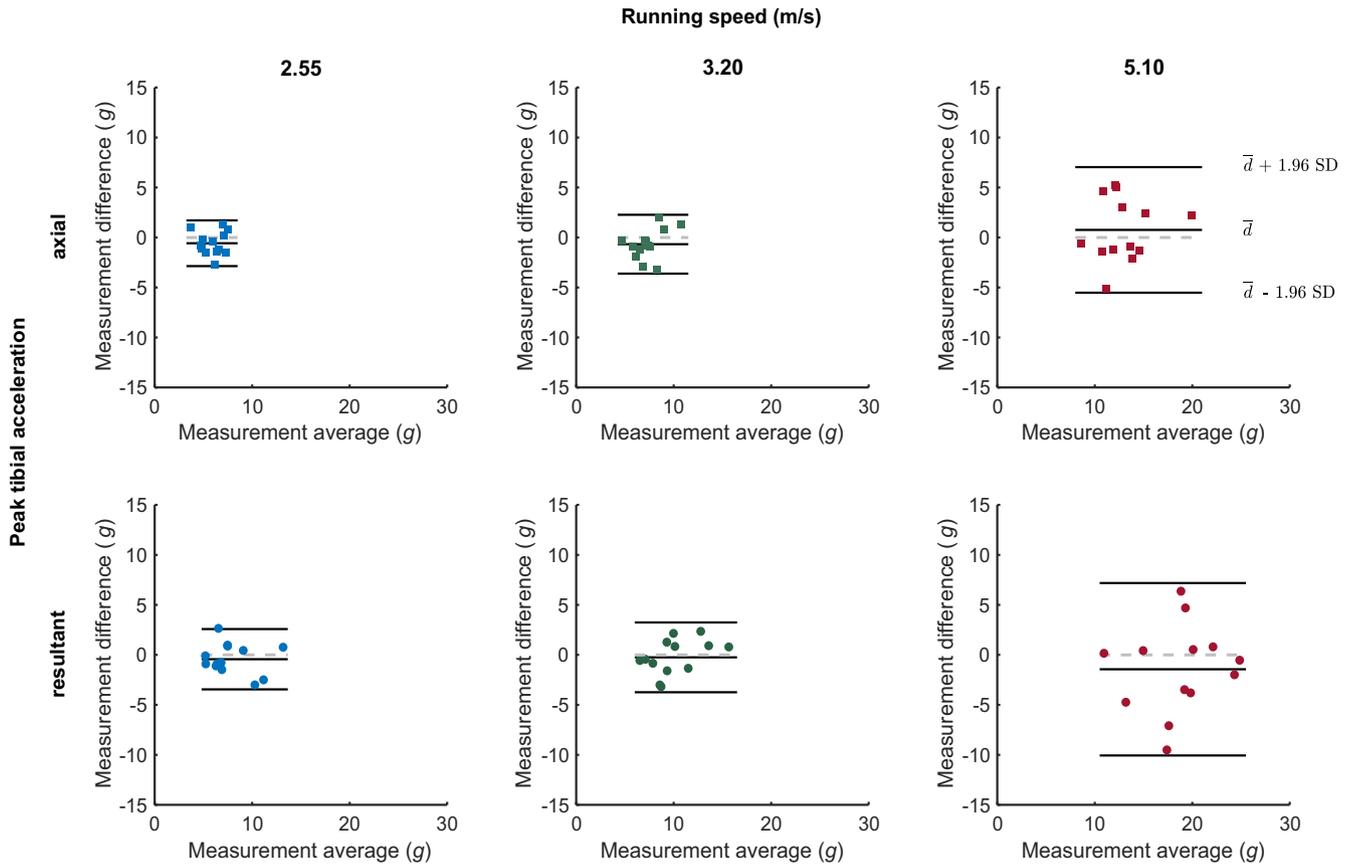


Fig. 2. Bland–Altman plots showing for one data point per participant, the bias \bar{d} as mean difference and the variability as 95% limits of agreement (bias ± 1.96 SD horizontal lines). The axial (upper part) and resultant (lower part) PTAs are shown for each speed condition.

Table 3
Significant correlation coefficient (r) for the relationship between axial or resultant PTA and LR per speed condition. The combined condition provides the agreement between these impact measures for all speeds combined.

Relationship	2.55 m*s ⁻¹	3.20 m*s ⁻¹	5.10 m*s ⁻¹	Combined
Axial PTA – LR	0.80 Very large	0.64 Large	0.84 Very large	0.92 Nearly perfect
Resultant PTA – LR	0.59 Large	0.57 Large	0.61 Large	0.85 Very large

pace (1.4 g) was smaller than the change in axial PTA (3.3 g) after completing a biofeedback-based running retraining program (Clansey et al., 2014) meaning that the changes induced by biofeedback on axial PTA outweigh measurement variability. Notably, the repeatability of axial PTA might be improved by adding more footfalls to the analysis there the interpretation shifted from poor-to-moderate (0.59, $n_{\text{synchronized contacts}} = 4$, $MDC_{95} = 1.4$ g) to good (0.75, $n_{\text{synchronized contacts}} = 6$, $MDC_{95} = 1.0$ g) in a post-hoc analysis. Strictly controlling for tightness of the skin pre-stretch may further improve test-retest reliability. When repeatability is prioritized and sensor alignment is not strictly controllable, there

is a slight preference for resultant PTA in protocols were PTAs of rearfoot runners are compared between sessions. Resultant PTA is namely an orientation-independent value, whereas axial PTA depends on the sensor's orientation.

Supporting the hypothesis regarding validity, axial and resultant PTAs largely correlated with LR though the correlation was stronger for the axial PTA. The correlation coefficient for axial PTA ($r_{\text{mean/speed}} = 0.76$) is at the upper end of what has been reported in literature on group level (range: 0.47–0.70) during over-ground running at submaximal speed with a skin-mounted accelerometer situated at the distal part of the lower leg

(Greenhalgh et al., 2012; Laughton et al., 2003). This relationship may be fortified by e.g. further increasing the mechanical skin-to-bone coupling (e.g. Pearsall et al., 2002) or by mounting the sensor more distally (e.g. Zhang et al., 2016). A certain amount of variance in loading rate remains unexplained by tibial shock, which is unsurprisingly because LR is derived from full-body segmental accelerations (Bobbert et al., 1991). As expected, different PTAs were registered when running speed was altered. The increments in axial and resultant PTAs at the lower speeds in this over-ground setting were rather analogous to those on treadmill (Sheerin et al., 2016). When speeding from 3.2 to 5.1 m·s⁻¹ the relative increase in resultant PTA stems with that of axial PTA (approximately 80%). The absolute increase in resultant PTA was however more pronounced compared to axial PTA due to increased but more variable accelerations in the horizontal plane during the impact phase of running gait. When validity is prioritized, our data suggest to prefer axial PTA to resultant PTA as surrogate measure for impact loading due to the generally stronger relationship with LR and its link with stress fracture susceptibility in rearfoot runners (Milner et al., 2006).

5. Conclusion

The wearable system was able to continuously detect PTAs in real-time and can be used for applications aiming at monitoring (e.g. before, during and after an in-field intervention) the impact loading experienced in the time domain by a runner during real world locomotion. The desired PTA component depends on the weight given to validity over repeatability. Lightweight accelerometers connected to a wearable system that is able to detect PTAs in real-time offers research possibilities for over-ground gait monitoring outside the lab where force platforms cannot be utilized.

Acknowledgements

This study was funded by the Research Foundation – Flanders (FWO.3F0.2015.0048.01 and the Methusalem project titled ‘Expressive music interaction’) and interreg EU (Nano4Sports project). The authors would like to thank PhD Bastiaan Breine and MSc Rud Derie for data analysis assistance, Ing. Davy Spiessens and Ivan Schepers for the technical support, and MSc Sander De Bock, Pieter Van Dam, Laura Rosseeuw, Tijn Vercauteren and Valentine Van-deputte for their help with data collection. Layla Aerts provided pictures of the measurement system in use.

Conflict of interest

No conflict of interest.

Appendix A. Supplementary material

Supplementary data to this article can be found online at <https://doi.org/10.1016/j.jbiomech.2019.01.039>.

References

- Bobbert, M.F., Schamhardt, H.C., Nigg, B.M., 1991. Calculation of vertical ground reaction force estimates during running from positional data. *J. Biomech.* 24, 1095–1105.
- Boey, H., Aeles, J., Schütte, K., Vanwanseele, B., 2016. The effect of three surface conditions, speed and running experience on vertical acceleration of the tibia during running. *Sport. Biomech.*, 1–11.
- Bramble, D.M., Lieberman, D.E., 2004. Endurance running and the evolution of Homo. *Nature* 432, 345–352.
- Breine, B., Malcolp, P., Frederick, E.C., De Clercq, D., 2014. Relationship between running speed and initial foot contact patterns. *Med. Sci. Sports Exerc.* 46, 1595–1603.
- Chambon, N., Delattre, N., Guéguen, N., Berton, E., Rao, G., 2014. Shoe drop has opposite influence on running pattern when running overground or on a treadmill. *Eur. J. Appl. Physiol.* 115, 911–918.
- Chan, Z.Y., Zhang, J.H., Au, I.P., An, W.W., Shum, G.L., Ng, G.Y., Cheung, R.T., 2017. Gait retraining for the reduction of injury occurrence in novice distance runners: 1-year follow-up of a randomized controlled trial. *Am. J. Sports Med.* 5, 036354651773627.
- Clansey, A., Hanlon, M., Wallace, E., Nevill, A., Lake, M., 2014. Influence of Tibial shock feedback training on impact loading and running economy. *Med. Sci. Sport. Exerc.* 46, 973–981.
- Crowell, H.P., Davis, I.S., 2011. Gait retraining to reduce lower extremity loading in runners. *Clin. Biomech.* 26, 78–83.
- Giandolini, M., Horvais, N., Rossi, J., Millet, G.Y., Samozino, P., Morin, J.B., 2016. Foot strike pattern differently affects the axial and transverse components of shock acceleration and attenuation in downhill trail running. *J. Biomech.* 49, 1765–1771.
- Greenhalgh, A., Sinclair, J., Protheroe, L., Chockalingam, N., 2012. Predicting Impact shock magnitude: which ground reaction force variable should we use? *ISSN Int. J. Sport. Sci. Eng.* 06, 1750–9823.
- Gruber, A.H., Boyer, K.A., Derrick, T.R., Hamill, J., 2014. Impact shock frequency components and attenuation in rearfoot and forefoot running. *J. Sport Health Sci.* 3, 113–121.
- Hennig, E.M., Milani, T.L., Lafortune, M.A., 1993. Use of ground reaction force parameters in predicting peak Tibial accelerations in running. *J. Appl. Biomech.* 9, 306–314.
- Lafortune, M.A., 1991. Three dimensional acceleration of the tibia during walking and running. *J. Biomech.* 24, 877–886.
- Laughton, C.A., Davis, I.M., Hamill, J., 2003. Effect of strike pattern and orthotic intervention on Tibial shock during running. *J. Appl. Biomech.* 19, 153–168.
- Milner, C.E., Ferber, R., Pollard, C.D., Hamill, J., Davis, I.S., 2006. Biomechanical factors associated with tibial stress fracture in female runners. *Med. Sci. Sports Exerc.* 38, 323–328.
- Milner, C.E., Westlake, C.G., Tate, J.J., 2011. Test – retest reliability of knee biomechanics during stop jump landings. *J. Biomech.* 44, 1814–1816.
- Nigg, B.M., Bahlsen, H.A., Luethi, S.M., Stokes, S., 1987. The influence of running velocity and midsole hardness on external impact forces in heel-toe running [published erratum appears in *J. Biomech.* 1988;21(3):267]. *J. Biomech.* 20, 951–959.
- Nigg, B.M., De Boer, R.W., Fisher, V., 1995. A kinematic comparison of overground and treadmill running. *Med. Sci. Sports Exerc.* 27, 98–105.
- Pearsall, D., Henning, E., Sterzing, T., 2002. The Use of Skin Pretension To Modify Tibial Bone Acceleration Estimates. *World Congr. Biomech.*, p. 1.
- Portney, L.G., Watkins, M.P., 2009. Foundations of clinical research: application to practice. *Crit. Care Med.*
- Sheerin, K.R., Besier, T.F., Reid, D., Hume, P.A., 2017. The one-week and six-month reliability and variability of three-dimensional tibial acceleration in runners. *Sport. Biomech.* 3141, 1–10.
- Sheerin, K.R., Besier, T.F., Reid, D., Hume, P.A., 2016. The reliability and variability of three-dimensional tibial acceleration during running. In: 34th International Conference of Biomechanics in Sport. Tsukuba
- Shorten, M., Mientjes, M.I.V., 2011. The ‘heel impact’ force peak during running is neither ‘heel’ nor ‘impact’ and does not quantify shoe cushioning effects. *Footwear Sci.* 3, 41–58.
- STMicroelectronics, 2011. LIS331HH MEMS digital output motion sensor ultra low-power high performance three-axis “nano” accelerometer [WWW Document]. Data Sheet. <<http://www.st.com/st-web-ui/static/active/en/resource/technical/document/datasheet/DM00040962.pdf>>.
- Ueda, T., Hobara, H., Kobayashi, Y., Heldoorn, T., Mochimaru, M., Mizoguchi, H., 2016. Comparison of 3 methods for computing loading rate during running. *Int. J. Sports Med.* 37, 1087–1090.
- van der Worp, H., Vrieling, J.W., Bredeweg, S.W., 2016. Do runners who suffer injuries have higher vertical ground reaction forces than those who remain injury-free? A systematic review and meta-analysis. *Br. J. Sports Med.* 50, 450–457.
- Willy, R.W., Buchenic, L., Rogacki, K., Ackerman, J., Schmidt, A., Willson, J.D., 2015. In-field gait retraining and mobile monitoring to address running biomechanics associated with tibial stress fracture. *Scand. J. Med. Sci. Sports*, 1–9.
- Wood, C.M., Kipp, K., 2014. Use of audio biofeedback to reduce tibial impact accelerations during running. *J. Biomech.* 47, 1739–1741.
- Zadpoor, A.A., Nikooyan, A.A., 2011. The relationship between lower-extremity stress fractures and the ground reaction force: a systematic review. *Clin. Biomech.* 26, 23–28.
- Zhang, J.H., An, W.W., Au, I.P.H., Chen, T.L., Cheung, R.T.H., 2016. Comparison of the correlations between impact loading rates and peak accelerations measured at two different body sites: Intra- and inter-subject analysis. *Gait Post.* 46, 53–56.