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Kinematics and shock attenuation during a prolonged run on the athletic track as measured with inertial magnetic measurement units

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ABSTRACT

Background: Tibial stress fractures are common running related injury and their etiology may include biomechanical factors like impact forces, shock attenuation, lower limb kinematics and how these factors are influenced by intense or prolonged running. Inertial-magnetic measurement units (IMUs) have recently emerged as an alternative to motion capture but their use to date was mostly limited to segmental and joint motion.
Research question: The present study sought to examine the effects of a prolonged run on shock attenuation, peak tibial and sacral acceleration (PTA, PSA), and lower limb kinematics using IMUs.
Methods: Ten trained male runners (31 +/− 5 yr, 183 +/− 3 cm, 76 +/− 9 kg) performed a twenty-minute prolonged run on an athletic track at estimated lactate threshold speed. Eight IMUs, positioned over the feet, lower and uppers legs, sacrum and sternum, were used to calculate joint kinematics, impact parameters and shock attenuation in the time domain (1-(PSA/PTA)*100).
Results: PTA increased while PSA and shock attenuation did not change following the prolonged run. Hip and knee flexion at midstance decreased. Vertical lower leg angle at initial contact did not change.
Conclusion: By using IMUs, it was shown that a prolonged run at estimated lactate threshold speed had significant effects on kinematics and tibial acceleration parameters. By modifying hip and knee joint kinematics during stance, the body was able to maintain sacral acceleration possibly by shifting from active shock attenuation to more passive mechanisms.
Significance: The present study shows that inertial sensors can be used in outdoor running to measure joint kinematics and kinetic parameters like PTA, PSA and shock attenuation simultaneously. The results of this study show new insights into how the body copes with impact during prolonged running.

1. Introduction

Running is associated with a high incidence of injuries [1]. Tibial stress fractures are among the most common running related overuse injuries [1]. Although the etiology of tibial stress fractures is not yet fully understood, biomechanical factors like impact forces, vertical loading rates and shock attenuation are suggested to be factors in their development [2]. Prolonged running has a negative influence on these factors, which may further contribute to injury risk [3–11].

Peak tibial acceleration (PTA) has been used frequently as a surrogate measure of the load (impact) on the tibia during running [10,12,13]. PTA is influenced by running speed or any factor that alters segment position and velocity prior to impact [14]. Retrospectively, runners with tibial stress fractures had higher PTA values compared with runners without tibial stress fractures [2]. Elevated PTA values might therefore indicate an increased risk of tibial stress fractures or may be a consequence of the injury. PTA correlates with spatiotemporal and kinematic parameters that quantify running technique as well as kinetic factors like ground reaction forces and loading rates [15,16]. Changes in these factors over the course of a run or over the course of a training period may provide information regarding injury risk.

Shock attenuation is the decrease in the magnitude and frequency of the impact shock wave as it propagates from the foot through the body [17]. The body utilizes different shock attenuation strategies to cope with the impact imposed to it during running. Shock can be attenuated by passive (e.g. heel fat pad, skin, bone, ligaments and tendons) and active mechanisms (e.g. eccentric muscular contractions). The importance of knee flexion in attenuating impact during in-line skating on

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a treadmill was demonstrated by Edwards et al. [17]. The knee joint acted as a low pass filter, where flexing the knee more resulted in greater shock attenuation. Flexing the knee may shift the shock-attenuating responsibilities away from passive biological tissue toward active muscular contractions. Additionally, a less flexed knee during initial loading and mid-stance was related to higher PTA values and the degree of shock attenuation in those with a retrospective tibial stress fracture [12,18]. Therefore, examining PTA in relation to changes in kinematic parameters such as knee flexion and hip flexion might provide insight into the role of shock attenuation in running injury development.

Intense, prolonged running was observed to influence kinematics as such that it resulted in higher PTA and lower shock attenuation compared with the start of the run [4,5]. In other studies, peak head acceleration (PHA) significantly increased as a consequence of two consecutive 20 min exhaustive runs at lactate threshold speed [10]. Therefore, if PTA increases as a consequence of intense prolonged running, runners may be less capable of attenuating shock.

PTA and shock attenuation have been measured frequently during running by means of uniaxial [2,4,5,12,16,19–27], biaxial [10] or triaxial [28–33] accelerometers. In the laboratory setting, these measurements can be synchronized with optical motion analysis systems to analyze accompanying joint kinematics. However, continuous measurement of impact and joint kinematics using motion capture, particularly over the course of a prolonged run, requires the use of treadmills that may alter gait compared with running over-ground.

The developments in wearable wireless sensor technology now enable continuous and simultaneous measurement of running kinematics and shock attenuation by using inertial-magnetic measurement units (IMUs). Previous research already showed that inertial sensors are a suitable tool for measuring PTA during running at different speeds [34]. The potential of using IMUs for continuous 3D kinematic analysis of running mechanics and impact parameters (peak sacral acceleration, PSA) has also been shown [35]. Since IMUs have the advantage of continuously and simultaneously measuring segmental accelerations and kinematics, they can present valuable real time information about possible changes in shock attenuation and running mechanics during a prolonged, over-ground level run. This real time information is also essential for monitoring gait in the clinical setting. For example, IMUs can be used by patients to monitor PTA and segment motion during a gait retraining session for primary or secondary prevention of tibial stress fractures. Therefore, the aim of the present study was to investigate shock attenuation and the kinematic mechanisms influencing shock attenuation over the course of a prolonged, over-ground level run using inertial-magnetic measurement units. Peak tibial and sacral acceleration, shock attenuation and lower limb kinematics were examined simultaneously. It was hypothesized that during a prolonged run, peak tibial acceleration will increase and peak sacral acceleration will increase accordingly due to a decreased shock attenuation.

2. Methods

2.1. Experimental design

Ten well-trained male runners (mean ± 1SD: Age = 31 ± 5 yr, height = 183 ± 3 cm, mass = 76 ± 9 kg) with a personal best 10,000 m performance time of 36.48 ± 3.81 min, who ran at least 40 km per week participated in this study (Table 1). All subjects were free from musculoskeletal injury for at least six months and wore their own preferred running shoes. The local Institutional Review Board approved the experimental protocol and all participants signed informed consent prior to participation.

After a ten-minute self-chosen warm-up, the runners individually performed a prolonged twenty-minute run on the athletic track. Running velocity corresponding to estimated lactate threshold speed was used to standardize the intensity of the run across participants. This velocity was estimated using each participants' 10 km seasonal best time [36]. Runners were instructed to follow a pacing cyclist who cycled at the calculated velocity for each runner. The cyclist rode a bicycle equipped with a validated cyclocomputer that presented velocity in increments of 0.5 km/h (BC 5.12, Sigma Sports, Neustadt, Germany). This pacing was used to prevent the runners from starting the run too quickly with the consequence of fading towards the end of the run and to rule out the effects of speed on the magnitude of PTA [14]. The cyclist traveled counter-clockwise in the second lane of a regular 400 m outdoor loop track at the runners' right-hand side in order not to intervene with the sport specific setting. The runner ran in the first lane. Prior to the running protocol, IMU sensor attachment, static and dynamic calibration procedures were performed to obtain sensor to segment calibration.

Table 1
Characteristics of the runners.

PP	Age	Height	Mass	10k Best
1	36	184	91	34.34
2	31	183	75	40.39
3	24	187	72	34.39
4	25	185	93	43.12
5	35	183	71	37.59
6	36	184	69	34.03
7	33	182	78	41.37
8	29	185	81	34.3
9	23	177	64	33.01
10	33	177	70	34.03
Mean	31	183	76	36.50
SD	5	3	9	3.43

2.2. Measurement device

Eight IMUs (MTx, Xsens Technologies B.V., Enschede, the Netherlands) were used to record segment orientation. Each IMU had a mass of 30 g and included a tri-axial accelerometer (range ± 18 g), a tri-axial gyroscope (range ± 1200°/s) and a tri-axial magnetometer (range ± 750 mGauss). The resolution of the 3D sensor orientation was 0.05 °; the resolution of the accelerometer was 2 mg while the resolution of the gyroscope was 0.6 deg/s. IMUs were integrated in a Lycra suit which positioned the sensors bilaterally over the following locations (Fig. 1): sternum (i.e. the manubrium); sacrum (between the left and right posterior superior iliac spine); upper leg (lateral at the iliotibial tract); lower leg (anteromedial part of the tibia) halfway between the knee and ankle joint [15,28]; and feet (i.e. fixed in customized clips between the shoe laces). The data collected from the IMUs positioned on the right lower leg was used for the joint kinematic and PTA analysis. Previous measurements of sacral acceleration is limited because it is difficult to limit skin artifact at this location. The Lycra suit, however, allowed for a secure attachment over the appropriate boney landmark to measure PSA.

The sensors were split in two wired roots of four sensors and connected to a central processing unit (XBus, Xsens Technologies B.V., Enschede, The Netherlands). The processing unit time-synchronizes the two roots after which the data are acquired and wirelessly transmitted (Bluetooth) to a receiver station at of the maximum available sampling rate, 100 Hz. The receiver station was connected, via USB, to a Windows 8 Pro tablet PC equipped with a dedicated software package, which stored the data (MT Manager 4.2.1, Xsens Technologies B.V., The Netherlands).

Xsens software (MT Manager *4.2.1*, Xsens the Netherlands) was used for data acquisition and MATLAB *R2013a* (MathWorks Inc., MA, USA) was used for data processing and analysis [35]. A Kalman filter (Xsens Kalman Filter, XKF) was used to fuse the data of accelerometers, gyroscopes and magnetometers to estimate the orientation of each sensor. Sensor orientations were converted to segment orientations by means of

Fig. 1. Measurement set-up. IMUs are visualized at the feet, medio-anterior part of the tibia, upper legs lateral at the iliotibial tract and sternum (the manubrium). Note that the IMU at the sacrum is not visible.

transformation matrices obtained from segment calibration procedures. A transformation matrix, based on the static and dynamic calibration, was defined to determine the time-invariant relation between each sensor frame and the corresponding anatomical segment frame. For the static calibration, the gravitational vector was measured while the subject was in quiet, upright stance. The gravitational vector defines the longitudinal axis of the segments (z-axis) trunk, upper leg and lower leg, while the longitudinal axis of the foot is perpendicular to the gravitational vector. For the dynamic calibration, the subject performed a set of flexion extension movements at the hip, knee and ankle joints. The average angular velocity vector during these movements, measured by the IMUs gyroscope, was assumed to correspond to the frontal/lateral segment axis. The sagittal segment axis was constructed using the vector cross product of the lateral and vertical axis (x-axis). Finally, a strictly orthogonal right-handed frame was obtained by replacing the lateral axis (y-axis) by the cross product of the sagittal and vertical axis. Joint angle trajectories of hip, knee and ankle where then determined following the Cardan convention with an YZX sequence. Data were checked for missing samples that could have occurred during the Bluetooth data transfer. For the analysis, only strides without missing samples were included.

Step detection was based on raw inertial data acquired from the foot sensors. A peak detection algorithm is used to identify local maxima in the accelerometer magnitude ($\boldsymbol{a} = \sqrt{a_x^2 + a_y^2 + a_z^2}$) evoked during foot strike [37]. The subsequent peak magnitude in the gyroscope signal, resulting from the fast plantar flexion during push off, was marked as toe off [38,39].

PTA and PSA were obtained by determining the maximum values after initial contact (IC) from the accelerometer data of the inertial sensors along the longitudinal axis at the tibia and sacrum after segment calibration (in m/s^2). Fig. 2 shows a typical example of the tibial acceleration and sacral acceleration values and their defined peaks.

Shock attenuation was calculated in the time domain [20,23] using sacral acceleration instead of the head acceleration [3,40]:

$$\left(1 - \frac{\mathit{Peak\ Sacral\ Acceleration}}{\mathit{Peak\ Tibial\ Acceleration}} *100\right)$$

Maximal values for hip and knee angle in the sagittal plane at IC and midstance (MS), were determined based on the joint angle trajectories. Vertical lower leg angle was determined at IC. Data were averaged over 20 strides on the straights of the track. The straights were identified using the heading of gyroscope. Normalized stance phase duration (expressed as a percentage of the total stride duration) was defined as contact duration ($t_{toe\text{-}off}$ - $t_{initial\text{-}contact}$) divided by stride duration.

2.3. Data analysis

At the beginning (3 min) and the end (18 min) of the run, 20 strides of the right leg were analyzed during the straights of the track for PTA, PSA, and shock attenuation and peak joint angle at IC and MS. Repeated measures one way ANOVAs, with Tukey post-hoc test, were used to statistically analyze mean differences of the defined spatiotemporal, kinetic and kinematic parameters between the two stages of the 20-minute run. Confidence interval was set to 95%.

3. Results

Table 2 presents average values and standard deviation for the measured parameters at the beginning and end of the prolonged intense run. PTA increased significantly between the beginning and end of the run ($p < 0.05$) while PSA and shock attenuation did not change significantly. Running velocity did not change between the beginning and end of the run ($p < 0.05$).

Typical examples for the hip and knee joint angle trajectories over the normalized stride cycle for the beginning and end of the run are shown in Fig. 3. Hip flexion at IC did not change but decreased significantly at MS between the beginning and end of the run ($p < 0.05$). Knee flexion increased at IC and decreased significantly at MS between the beginning and end of the run ($p < 0.05$). Vertical lower leg angle at IC did not change.

4. Discussion

This study aimed at simultaneously measuring the effects of a prolonged, over-ground level run on shock attenuation parameters and lower limb kinematics by using inertial-magnetic measurement units (IMUs). During the run of 20-minutes, PTA increased significantly while PSA and shock attenuation did not change. In addition, there was a significant increase in knee flexion at initial contact and significant decreases in hip and knee flexion at midstance. Contrary to our hypothesis, PSA and shock attenuation did not change likely as a result of the differences in knee and hip kinematics observed between the start and end of the run. This result may not be consistent with previous prolonged running studies [4,5,10] because the run in this study was not meant to be maximally exhausting.

These findings indicate that given the higher tibial shock occurring during the run, the body was able to maintain sacral acceleration possibly by modulating knee joint flexion at initial contact and midstance and hip flexion at midstance between the beginning and end of the prolonged run. Impact shock can be attenuated by passive deformation of the body tissues and by active mechanisms such as eccentric muscle contractions, changes to joint angles, and modulating limb stiffness [41–44]. It is possible that sacral acceleration was maintained between the beginning and end of the prolonged run by increasing the role of passive shock attenuation mechanisms given that on a kinematic level, less flexion was observed in the hip and knee at midstance. A more extended knee will increase joint stiffness and impact forces and therefore the amount of shock that must be attenuated [45]. The decrease in knee flexion range of motion during the first half of stance indicates a reduction in the contribution of active muscular contractions to attenuate shock [17]. In the present study, PSA was maintained despite a significant increase in PTA, a more extended knee at midstance, and likely a less compliant limb at the end of the run. That is, the combination of greater peak tibial acceleration and a more extended limb suggest that shock attenuation was maintained by increasing the role of passive attenuation mechanisms. Therefore, a

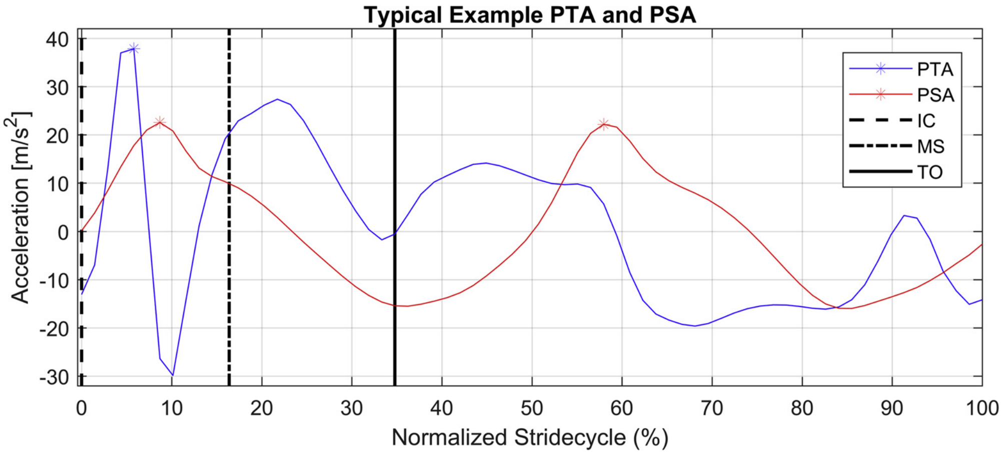

Fig. 2. Typical examples of the tibial (in blue) and sacral (in red) acceleration values of one typical stride where PTA and PSA are marked with an asterisk. Note that during one stride the sacral acceleration has a second peak when the other leg touches ground. Initial Contact (IC), midstance (MS) and toe-off (TO) are marked.

Table 2
Average values for the defined parameters for the beginning and end of the run and their significance. An * denotes a significant difference with $p < 0.05$ while NS indicates the difference was not significant. IC means initial contact while MS means midstance.

20 Minute run on the athletic track

	Begin		**End**			
Parameter	Average	SD	Average	SD		
Velocity (m/s)	**4.4**	0.4	**4.5**	0.3	NS	0.202
	Right leg					
	Begin		**End**			
	Average	SD	Average	SD		
PTA (m/s2)	**48.7**	15.4	**52.3**	21.1	*	< 0.05
PSA (m/s2)	**24.6**	7.1	**24.9**	6.1	NS	0.338
Shock Attenuation	**51.9**	16.2	**53.5**	15.0	NS	0.209
Hip @ IC (degrees)	**33.5**	6.7	**33.4**	7.3	NS	0.924
Hip @ MS (degrees)	**22.5**	5.7	**20.2**	6.5	*	< 0.05
Knee @ IC (degrees)	**15.8**	7.6	**16.7**	7.6	*	< 0.05
Knee @ MS (degrees)	**45.8**	5.2	**45.3**	5.1	*	< 0.05
Vertical lower leg angle (Degrees)	**−9.4**	2.8	**−9.4**	2.6	NS	0.993

prolonged run may result in a greater reliance of passive shock attenuation via deformation of the heel fat pad, the running shoe, ligaments, bone and articular cartilage [41] than active mechanisms.

The addition of ground reaction force measurements is needed to understand whether the peak tibial acceleration increased in the present study because of a change in effective mass rather than an increase in impact forces applied to the foot. Effective mass will be reduced with a more flexed knee at initial contact, in turn causing an increase in peak tibial acceleration without increasing the forces applied to the foot [46]. If there is no change in impact force, then there is likely no increased injury potential associated with increased peak tibial acceleration [6]. We found increased knee joint flexion at initial contact following the prolonged run, therefore the increase in peak tibial acceleration may be the result of a decrease in effective mass. An decrease in effective mass, rather than an increase in impact forces, is supported by findings in a recent review article that impact forces do not change following prolonged running [47]. Regardless of the cause, effective mass or impact forces, increased tibial accelerations must be attenuated before they are transmitted to the head. The more extended hip and knee at midstance found in the present study suggest a decreased role of active attenuation mechanisms and an increased role in passive attenuation mechanisms in order for sacral acceleration to remain the same.

It has been suggested that the body acts like a low pass filter by reducing the signal power of high frequency components of tibial shock [17]. The degree of shock attenuation increases as the magnitude of the input signal, tibial shock, increases in several conditions including: healthy runners in an unfatigued state [6,7], those experiencing lower extremity or low back pain [3,40], and healthy runners in a moderately fatigued state [3]. Additionally, head acceleration remains statistically similar across endurance and maximal sprint speeds [7]. Therefore, the body is able to attenuate the shock imposed to it under a variety of conditions. Despite the less flexed hip and knee observed at midstance during the end of the run in the present study, it might be that a prolonged, over-ground level run at estimated lactate threshold speed was not sufficiently intense to affect the ability to attenuate shock negatively. Conversely during a marathon, which can without any doubt be considered an intense event, PSA increased significantly indicating that the runners may have become incapable of maintaining shock attenuation via mechanisms of the lower extremity [35].

The shift from active to passive shock attenuation mechanisms, demonstrated here and by others [10,41,48,49], might result in elevated mechanical forces transmitted to the bone and other non-elastic structures. An accumulation of these mechanical forces could eventually exceed the repairing and remodeling process of the bone structure over time and as such might lead to overuse injuries like tibial stress fractures [10]. It is therefore possible that changes to the musculoskeletal system as observed by changes in running mechanics during a prolonged run could alter the ability to attenuate shock [4,10]. Considerable further research is needed to support these claims.

A limitation to this study is the sample frequency of the IMUs. Due to the bandwidth of the Bluetooth connection, the maximal update range was limited at 100 Hz. Recent research suggests that a sampling frequency of 200 Hz is required to accurately determine the absolute values of tibial acceleration and kinematic parameters [50]. Looking at a typical tibial acceleration signal for an individual stride as presented in Fig. 2, it can be seen that the PTA might indeed be underestimated due to the relatively low sample frequency. However, spectral analysis of the tibial and sacral acceleration signals collected for this study revealed that 99.5% of the signal power was contained within frequencies 1–35 Hz. This finding is supported by previous work demonstrating that the primary signal content of tibial impact resided under 50 Hz [14]. A separate spectral analysis of the signal from a high-resolution accelerometer (1200 Hz) mounted over the tibia and performed by our laboratory further confirmed that 96.3% of the signal content resides under 50 Hz. As an additional test, we upsampled the data from the present study to 1000 Hz following published methods [51] and compared several peaks from the start and the end of the run with the original data. Although the magnitude of the peaks was different, the overall results remained consistent with the original, 100 Hz data. Given that intra-individual changes were of interest and that PTA is

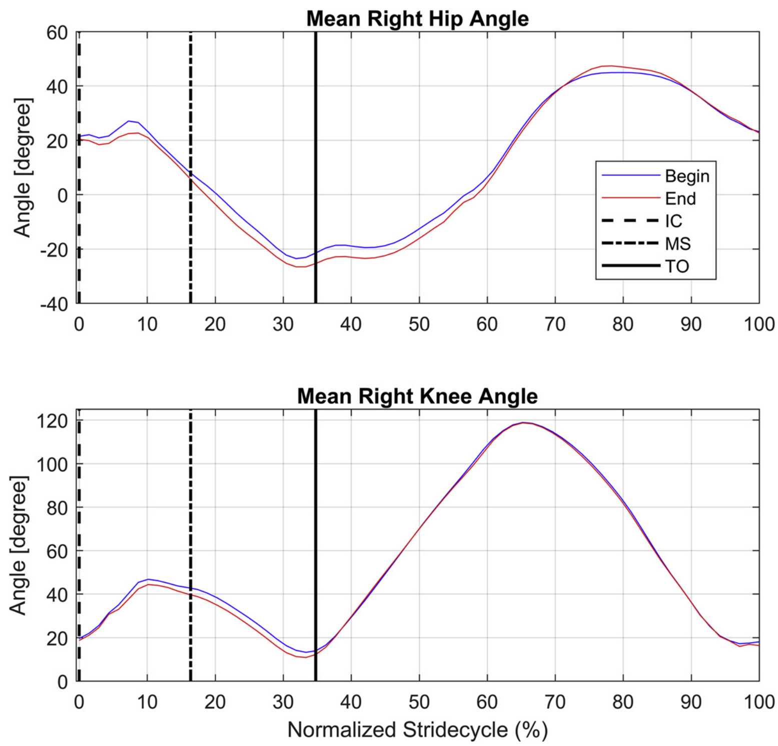

Fig. 3. Typical joint angle trajectories for the right hip and right knee over the normalized stride cycle, averaged over 20 strides for the beginning (blue line) and end (red line) of the run. Initial Contact (IC), midstance (MS) and toe-off (TO) are marked.

averaged over multiple (n = 20) strides, we expect the error to be systematic and of minor influence on the results. Nevertheless, we suggest future studies to use a higher sample frequency so that absolute magnitude of PTA and PSA can be determined more appropriately. This study is a first step towards simultaneously measuring kinematics and shock attenuation in the real world. As such, it provides very relevant information about shock attenuation in relation to kinematic parameters.

Another limitation to this study is that the results are only generalizable to well-trained endurance athletes running at high intensity. Changes in kinematics or shock attenuation strategies might have been different in other groups of runners, for instance novice runners, or at other intensities or durations. Furthermore, runners used their own preferred footwear. Although this contributed to the ecological validity of this real-world study, the individual differences in running shoe properties could have influenced the shock attenuation strategies. Nevertheless, the present measurement setup allows for larger scale measurement in real world settings as well as in more and more heterogeneous groups of runners.

5. Conclusion

By using inertial-magnetic measurement units, it was shown that a prolonged, over-ground level run had significant effects on lower limb kinematics and peak tibial acceleration. No change in peak sacral acceleration, despite a significant increase in peak tibial acceleration, indicates that the body was still able to maintain shock attenuation at the end of the run. It was assumed that shock attenuation was maintained by shifting from less active to more passive attenuation mechanisms given that on a kinematic level, less flexion was observed in the hip and knee at midstance.

Conflict of interest statement

None.

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