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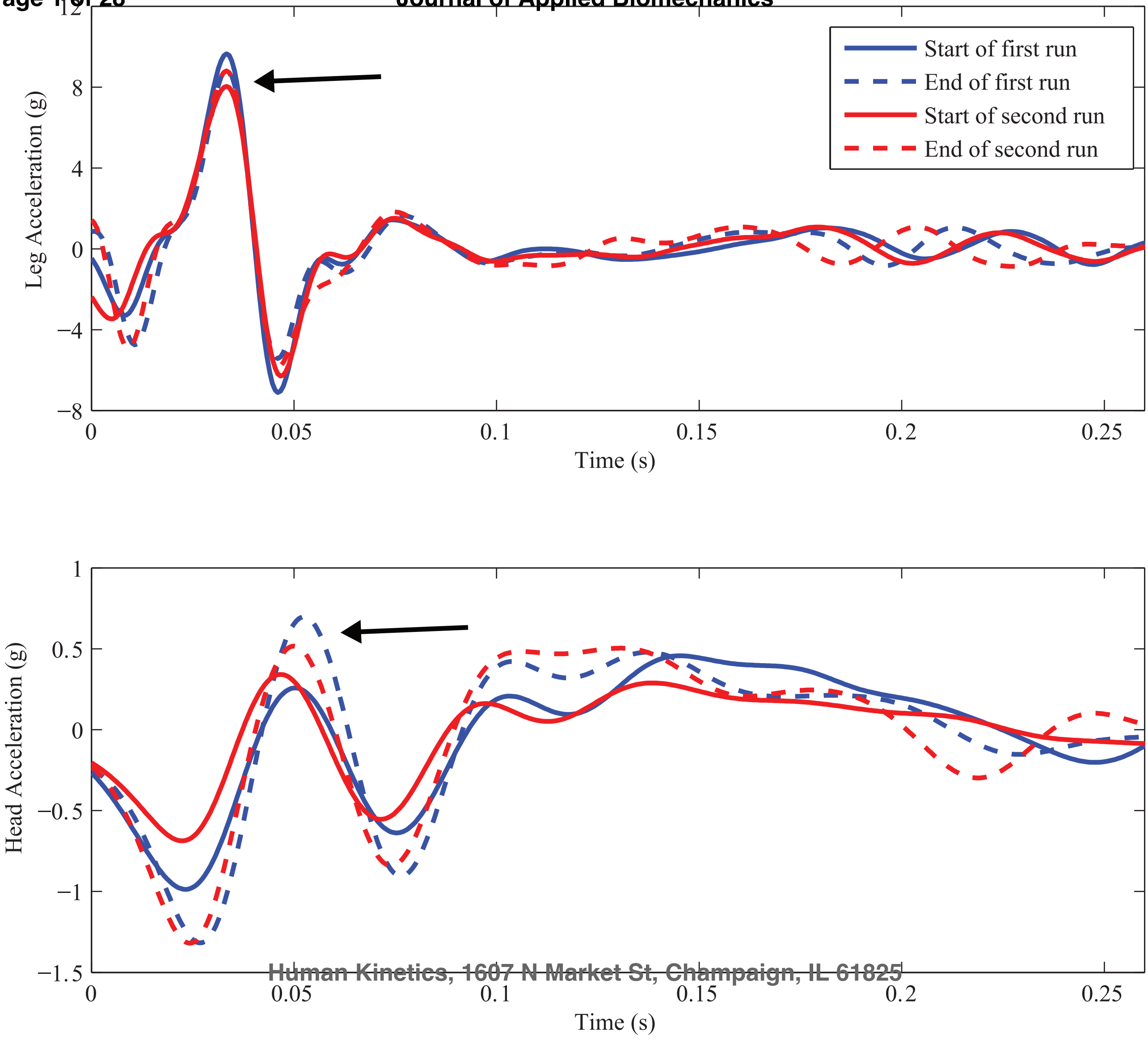
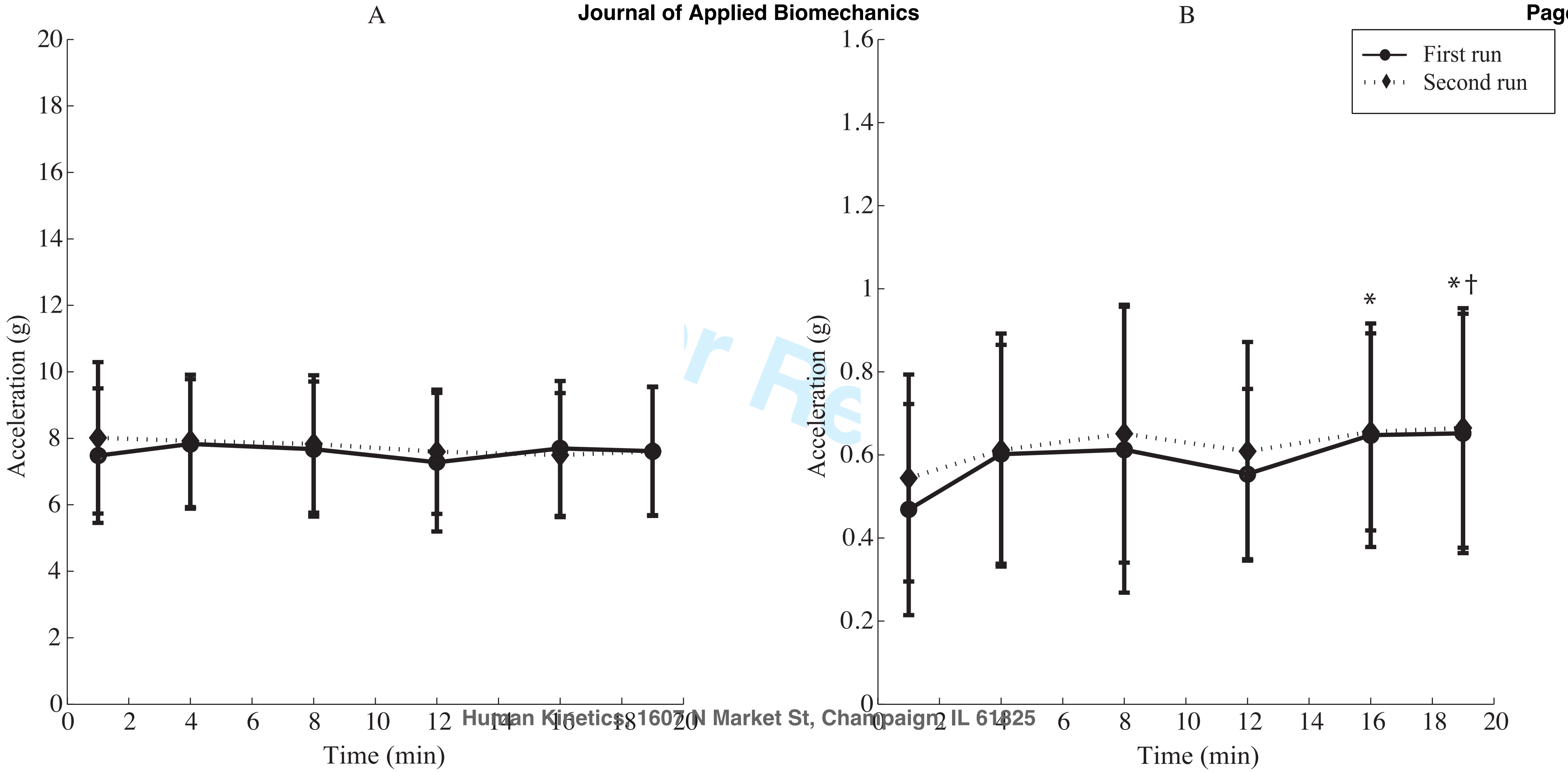
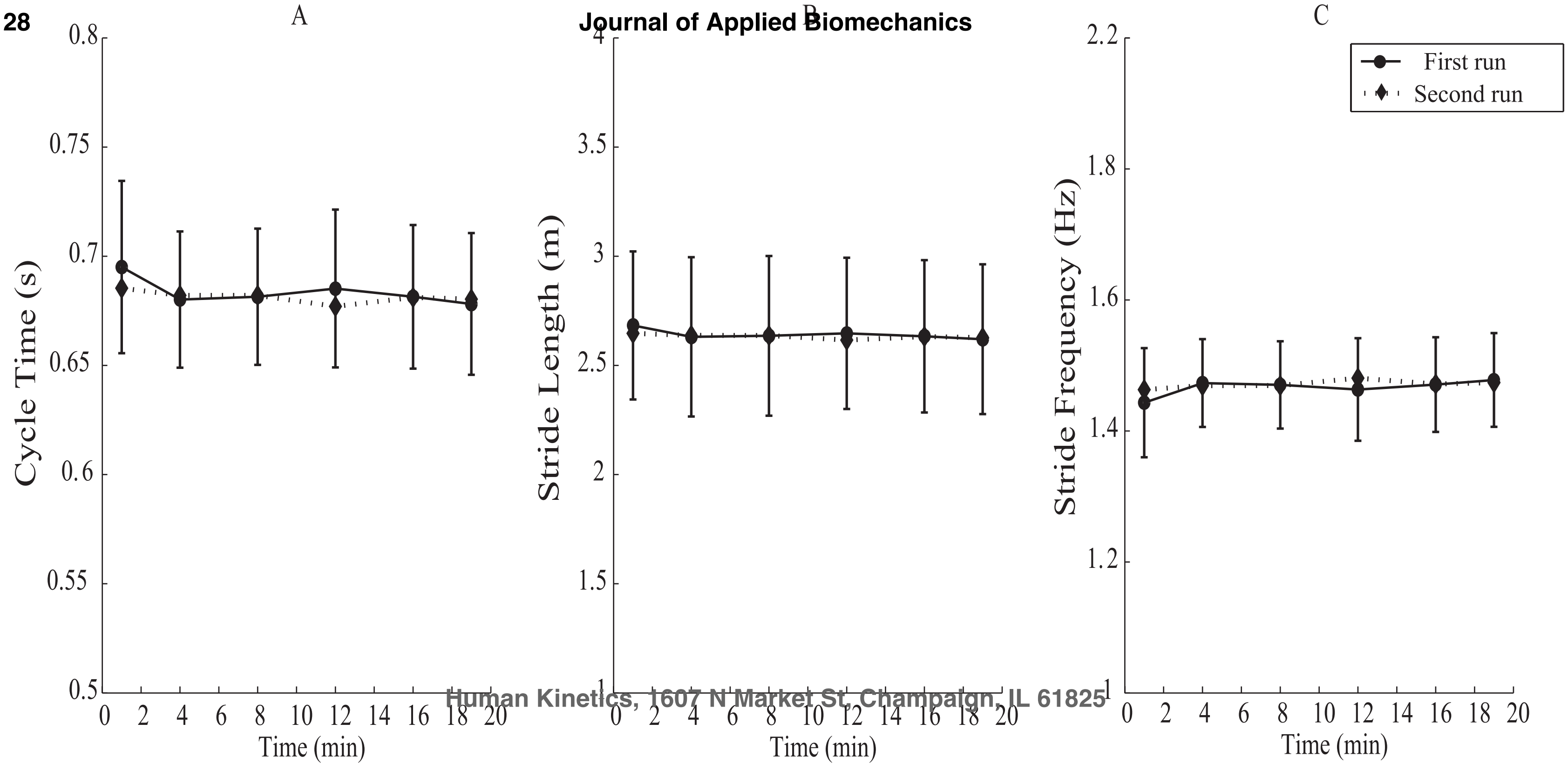
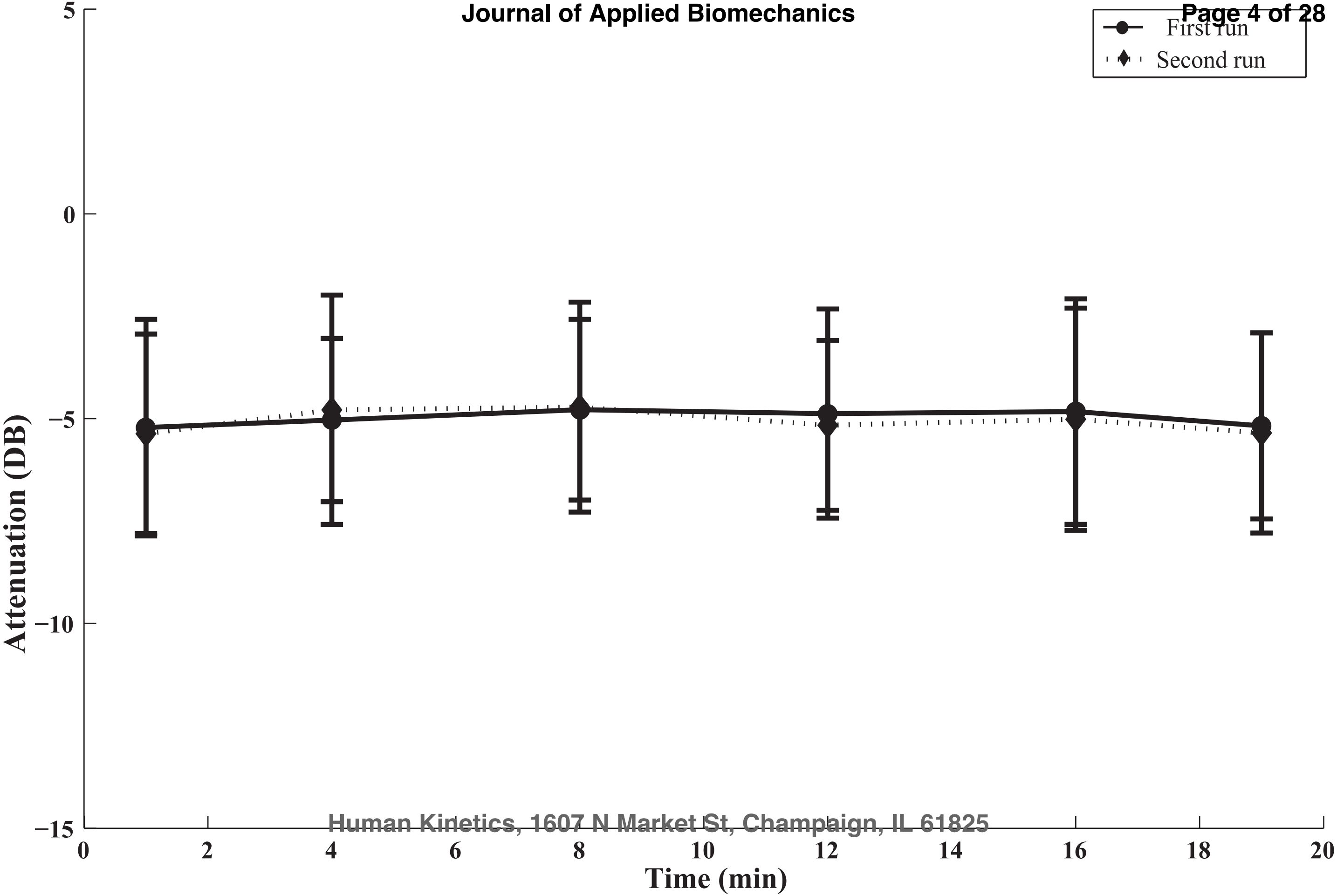


12
8
4
0
−4
−8
Leg Acceleration (g)
0
0.05
0.1
0.15
0.2
0.25
Time (s)
Start of first run
End of first run
Start of second run
End of second run
1
0.5
0
−0.5
−1
−1.5
Head Acceleration (g)
0
0.05
0.1
0.15
0.2
0.25
Time (s)


A
B
First run
Second run
Acceleration (g)
Time (min)
*
* †


A
B
C
Cycle Time (s)
Stride Length (m)
Stride Frequency (Hz)
Time (min)
Time (min)
Time (min)
First run
Second run


First run
Second run
Attenuation (DB)
5
0
−5
−10
−15
0
2
4
6
8
10
12
14
16
18
20
Time (min)

Can trained runners effectively attenuate impact acceleration during repeated high-intensity running bouts?

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Abstract

The purpose of this study was to investigate the effects of prolonged high intensity running on impact accelerations in trained runners. Thirteen male distance runners completed two 20-minute treadmill runs at speeds corresponding to 95% of onset of blood lactate accumulation. Leg and head accelerations were collected for 20 s every 4^{th} minute. Rating of perceived exertion (RPE) scores were recorded during the 3^{rd} and last minute of each run. RPE responses increased ($p < .001$) from the start (11.8 ± 0.9, moderate intensity) of the first run to the end (17.7 ±1.5; very hard) of the second run. Runners maintained their leg impact acceleration, impact attenuation, stride length and stride frequency characteristics with prolonged run duration. However, a small (0.11-0.14g) but significant increase ($p < .001$) in head impact accelerations were observed at the end of both first and second runs. It was concluded that trained runners are able to control leg impact accelerations during sustained high-intensity running. Alongside the substantial increases in perceived exertion levels, running mechanics and frequency domain impact attenuation levels remained constant. This suggests that the present trained runners are able to cope from a mechanical perspective despite an increased physiological demand.

Keyword: High-intensity, impact acceleration, running, treadmill.

Word count: 3606

Introduction

It is estimated that a runner covering 20 miles per week will on average collide with the ground over 1.3 million times in a one-year period [1].These repetitive high impact loads during running transmit impact shock waves up through the musculoskeletal system [2,3]. The body is known to act as a low-pass filter, whereby the high, transient acceleration of the lower leg (leg impact acceleration (input)) is severely dampened before it reaches the head (head impact acceleration (output)) [2-4]. The process of dissipating these impact energies during running is termed impact attenuation, which is a function of the output relative to the input [4]. Between the foot and the head these impact energies are attenuated passively by components such as muscle, heel fatpad, bone, cartilage, synovial fluid and other structural components [4,5] and actively by eccentric muscular contractions controlling lower limb joint motion during landing [4-6].

The associated reduction in neuromuscular functionality with prolonged running [7-9], has led researchers to consider that this impairment may decrease the impact absorbing capacity of the body and therefore lead to a greater risk of overuse injury development or degenerative disease [10-13]. Nevertheless, despite these claims, the influence of sustained high-intensity effort running on impact accelerations (peak positive axial component) still appears to be conflicting and unclear. Several studies have reported significant increases in leg impact acceleration with prolonged running [1,11,13-16], while others have reported no changes [17-19]. Researchers believe that these discrepancies are attributed to the type of fatigue protocols implemented and also the trained status of participants [18,20]. Mizrahi, Verbitsky, Isakov [12] who ran active individuals at their anaerobic threshold speed for 30 min found significant increases in leg impact accelerations 15 minutes into the run. This increase in leg impact acceleration was also observed in other studies who conducted similar protocols and with comparable subject cohort groups [11,14]. In contrast, Mercer, Bates, Dufek, Hreljac [18] found no changes in leg and head accelerations while running before and after a graded exhaustive treadmill run. Similarly, Abt, Sell, Chu, Lovalekar, Burdett, Lephart [17] who recruited experienced runners reported both consistent leg and head impact accelerations before and after a 17.8 minute exhaustive treadmill run. Another possible explanation for the discrepancies in results could be related to differences between treadmill and overground running. For instance, individuals are constrained to run at a constant speed during treadmill running and the fluctuations of impact accelerations may be less pronounced than in overground running where speed is self-regulated and more variable. .

While the previous reports provide an insight into the effects on high-intensity running impact accelerations, little attention has been focused on examining how trained runners modulate impact accelerations during repeated bouts of prolonged high intensity running. The majority of studies that recruited trained runners implemented an exhausting, single bout, short duration run protocol [1,17,18,21], however, these types of situations are not representative of what runners perform during their weekly training schedule. Generally, trained runners are known to perform repeated bouts of high intensity running during single training sessions [7,22]. Therefore, the purpose of this investigation was to examine the effects of two bouts of high-intensity running on impact acceleration and attenuation. It was hypothesized that leg impact acceleration would increase progressively over the two runs, whereas, head impact acceleration would remain consistent, and be indicative of an improvement in impact attenuation with prolonged running. Given that stride characteristics are known to play an important role in the modulation of impact accelerations during running [2,23,24], the present study also aimed to investigate runners stride characteristic during these repeated bouts of high-intensity running. It was hypothesized that runners would increase their stride length and decrease stride frequency in order to maintain velocity when they were towards the end of the prolonged running bouts.

Method

Participants

Thirteen male trained distance runners (age 35.1 ± 10.2 years, height 1.8 ± 0.1 m, mass 73.1 ± 11.1 kg, weekly mileage 70 ± 21 km per week) participated in this study. Each participant signed an informed consent form approved by the University Research Ethics Review Board. All participants were free from musculoskeletal injuries at the time of testing and had no reported cardiorespiratory conditions or problems.

Determination of 95% of OLBA

A week prior to the high intensity run protocol, participants underwent a standardized incremental lactate threshold running test [25]. This test was used to determine each participant's speed for the high intensity run protocol at 95% of OBLA, which equates to a blood lactate concentration of 3.5 Mm. All participants were familiarized with treadmill running and were instructed not run or train 24 h prior to testing. The test consisted of participants running on a treadmill (T170 DE, HP Cosmed, UK) at a 1% gradient for 3-minute stages. Before and after the test all participants performed a 5-min warm-up and cool-down jog. The test started at a speed that was perceived 'comfortable/easy pace' for participants to run. Between each stage a 30-s time period was allocated to allow the collection of blood from each participant's fingertip. Participant blood lactate levels were measured and recorded using a lactate analyser (Lactate Pro; UK). The treadmill speed was increased 1 $km \cdot hr^{-1}$ every 3-minute stage. The test was terminated once participants' blood lactate concentration exceeded 4.0 Mm. Each participant's velocity was based on their OBLA of 3.5 Mm [26]. This 95% OBLA threshold marker was chosen based on previous reports that showed reductions in muscular strength after runners completed on average 40 minutes of treadmill running [27,28]. The 95% OBLA marker was determined by polynomial regression model outlined by Newell, Higgins, Madden, Cruickshank, Einbeck, McMillan, McDonald [25]. The average 95% of OBLA speed was 14.0 ± 2.4 $km \cdot hr^{-1}$.

Procedures

A week after the determination of OBLA participants returned to the laboratory for the second time to perform the fatiguing testing. After this, participants completed two bouts of 20-minute treadmill runs at 95% of their OBLA. Between the first and second 20-minute running bouts participants performed six discontinuous overground running trials at 4.5 $m \cdot s^{-1}$ over a 15-m runway. This was part of a previous related study [20]. The total duration of these trials lasted for 3-5 minutes in duration.

Two lightweight (2.8 g) biaxial (± 10g; frequency range of 5-5 kHz) accelerometers (Noraxon, Scottsdale, AZ) were attached to participant's distal anteromedial aspect of the tibia and anterior aspect of the forehead before the running [20]. The accelerometer had a built in DC filter (5 Hz High-pass filter) that removed the acceleration offset related to the orientation or position. To minimize the influence of angular motion of the shank on the leg acceleration profiles, the accelerometer was placed as close as possible to the ankle joint (i.e. approx. 7cm-10cm above the malleoli). These sites were selected to minimize the effects of soft tissue oscillations during impact. As a precaution to reduce any unwanted skin oscillations, the skin around the accelerometers sites was stretched using adhesive kinesiology tape (Vivomed, UK). At the site of leg (tibia) accelerometer attachment, the skin was shaved using sterilized razors and then cleaned. The axial axis of the leg accelerometer was aligned with the longitudinal axes of the tibia bone. Once the accelerometers were attached, they were securely tightened using self-gripping bandage. In addition, participants wore a headband to further secure the accelerometer to the head. The axial component of both sets of accelerometry data were recorded at 1500 Hz for 20 s and were captured during the first, 4th, 8th, 12th, 16th and last minute of each run. Rating of perceived exertion (RPE) scores were measured on a 6-20 point scale and were collected during the third and last minute of each run. RPE is used as a non-invasive physiological valid tool for prescribing exercise intensity [29] and has been previously used to determine if physiological fatigue was likely to have occurred while running at a given running speed [30].

Data analysis

The accelerometry data were collected in Qualisys track manager software (Qualisys, Gothenburg, Sweden) and exported into MatLab (R2013a, Mathworks, Natick, MA) for processing and analysis. Stance phases were extracted from the head and leg acceleration profile data and transformed into the frequency domain using a Fast Fourier Transform using methods previously outlined[4,31] . Before acceleration data were transformed to the frequency domain, the mean and linear trends were removed. The length of the data needed to be a power of two for the power spectral density (PSD) so the acceleration data were padded with zeros in order to total 512 data points. Power spectral density (PSD) profiles were generated from frequencies 0 Hz to the Nyquist frequency (F_N) using a square window. The resulting PSD profiles were normalized to 1 Hz frequency bins with power adjustments made to reflect padding of zeros. After binning, the PSD was normalized to the sum of the powers from 0 to F to be equal to the mean squared amplitude of the data in the time domain. Transfer functions (TF) were calculated from the power spectral densities at the head (PSD_{head}) and the leg (PSD_{leg}) using the following formula :

$$TF\ (DB) = 10log_{10}(\frac{PSD_{head}}{PSD_{leg}})$$

where the TF is the gain and attenuation in decibels and the PSD_{head} and PSD_{leg} are the power spectral densities of the head and leg at each 1 Hz frequency interval.

The transfer function values at impact frequencies of 10-20 Hz were averaged to obtain a measure of the impact attenuation in the body [1]. The reason for selecting this frequency portion was due to its association with transient impact phase of the foot contacting the ground [2]. For example, a greater absolute value in the 10-20 Hz range indicated a greater impact attenuation. The peak axial accelerations of the head and leg were extracted during the early impact phase of stance and averaged over each 20-s trial (Figure 1). The number of ground contacts analyzed varied between runners depending individual running speed and stride length, but typically there was around 25 contacts in a 20-s trial. Stride characteristics were calculated based on the peaks of the leg impact accelerations for each trial [32]. Cycle time was the average time between each consecutive leg impact acceleration (stride) and stride frequency was calculated as the inverse of this time. Once stride frequency was calculated, stride length was computed from the treadmill velocity and stride frequency.

Statistical analysis

Dependent variables of head and leg impact accelerations (time domain), impact acceleration (frequency domain), stride length; stride frequency, cycle time and RPE scores were tested for normality using Mauchly's test. Statistical tests were performed using SPSS, version 22 (SPSS, Chicago, IL). The results were presented as means ±S.D. A two-way (first and second run) within group, repeated measures ANOVA was used to detect differences across 6 time points for each dependent variable. For the RPE scores, a two-way (first and second run) repeated measures (2 levels) ANOVA was used to compare the start and end time points between runs. A critical value of $p < 0.05$ was assumed for significance. When either a significant interaction or a main effect was observed for a dependent variable, Bonferroni adjusted *post hoc* analyses were used to determine where the differences rested.

Results

Runners were able to maintain their leg accelerations throughout each high-intensity run (Figure 2). At the start of the first run, leg impact accelerations were 7.67 (2.1) g and remained constant at 8.03 (2.3) g for the last minute of the run (Figure 2A). Similarly, leg impact accelerations were 7.43 (2.1) g at the first minute of the second run and did not change at 8.04 (2.3) g by the last min of the second run. A significant main effect of run ($p < .035$) and time ($p < .001$) in head impact acceleration was observed. In the first run, head impact acceleration increased significantly from the start at 0.47 (0.25) g to the 16th at 0.61 (0.31) g and last minute of the run at 0.65 (0.31) g. Likewise, head impact accelerations also significant increased in the last minute (0.66 (0.28) g) of the second run compared to the start of the run (0.55 (0.20) g). It was apparent that there was a general increased offset in head impact acceleration values in the second run as compared to the first run.

RPE responses had a significant run by time interaction ($p < .001$) whereby, greater changes were observed in the second run as compared to first run (4.6 Δ versus 2.5 Δ). The pairwise comparisons revealed that between the start and end of the first run, RPE responses significantly increased ($p < .001$) from 11.8 (0.9) to 14.3 (1.2). Similarly, RPE responses progressively increased from 13.1 (1.2) to 17.7 (1.5) (very hard) at the start of the second run compared to the end. In addition, between the end of the first run and the start of the second, runners RPE responses significantly decreased ($p = .02$). All runners maintained a consistent cycle time (Figure 3A), stride length (Figure 3B) and stride frequency (Figure 3C) with increased run duration. Similarly, no changes in impact attenuation were found across any time point of both runs (Figure 4). Although not reported, the PSD profiles for the head and leg were analyzed but no differences were found.

Discussion

The primary purpose of this study was to investigate the effects of high intensity running on both head and leg impact accelerations in trained runners. The prescribed run protocol consisting of two consecutive bouts of 20 minute runs at 95% of OBLA was shown to be successful in progressively and substantially increasing runners perceived exercise exertion levels. Although no measures of fatigue were collected in the present study, it is plausible that the current protocol elicited a level of fatigue similar to previous protocols that implemented a 95% of OBLA running intensity [27,28]. The results of this study indicated that runners were able to effectively maintain their leg accelerations across both prolonged 20-minute runs at 95% of their OBLA. This finding rejects our hypothesis in which we expected greater leg impact accelerations with increased run duration. Contrary to our findings, significant increases in leg impact accelerations were found [11,12,14-16]. A possible explanation for the conflicting findings may be attributed to the trained status of the runners. In the present study we recruited trained distance runners, whereas the previous studies had healthy participants without an endurance background [11,12,14]. With trained distance runners being frequently exposed to prolonged running, it is likely that they have better mechanical coping strategies when placed under an increased physiological demand as compared to the less experienced non-runner counterpart. In support, others reported consistent leg impact accelerations after experienced runners completed an exhaustive running protocols [17,18]. Whilst these results are in support of the present study's, it is recognized that the shorter duration and exhaustive incremental protocol designs may be enough to induce a high level of neuromuscular fatigue that has been associated with a reduction in the muscles ability to effectively attenuate impact acceleration during running. Accordingly, it is probable that the incremental short duration exhaustive protocols may have impaired more of the central mechanisms (such as heart and lung function) as opposed to the peripheral mechanisms (neuromuscular function and neural transmission), that are responsible for controlling impacts pre-landing during running [33,34]. Evidence has shown that longer duration running protocols [9,35] elicit greater impairments in muscular activation and strength as compared to shorter and high intensity protocols [27,28] and these neuromuscular impairments have been associated with central fatigue [7]. With this being the case, it may be that the consistent leg impact accelerations are due to the current protocol not eliciting sufficient impairments to the peripheral and central mechanisms that control for landing phase during running. Evidently, given the differences in results between studies, it is acknowledged that numerous factors such as subjects training status, exercise duration, exercise intensity and exercise type, can all play an important role on the outcome of impact acceleration results.

Since leg impact accelerations are positively correlated with speed [1], another important consideration for the conflicting findings may be related to the experimental designs that controlled speed and those that allowed it to vary. Studies that controlled speed observed subtle changes in leg impact acceleration and running technique with prolonged running [19,31,36], whereas, when speed was allowed to vary during running leg impact accelerations changes were clearly larger [37]. Therefore, we acknowledge that controlling speed in the present study may have lead to only subtle changes in running mechanics, however we speculate this may not be the case with more severe fatigue levels.

Studies have found that regardless of magnitude of input acceleration at the leg during running (un-fatigued state) the output acceleration at the head remains consistent [2,23]. The authors believe that this maintenance may be related to the system's goal of wanting to optimize the stability of the head for allowing clear and consistent information to the vestibular and visual systems [38,39]. Although leg impact accelerations remained consistent with prolonged high-intensity running, our results showed a 30.5% and 20% increase in head impact acceleration between the first and last minute of the first and second run. While several studies reported no changes in head impact accelerations with high-intensity running [1,15-19], one study did report significant increases in impact accelerations at the sacrum site 20 minutes into a high intensity run [11]. The authors of this study claimed that the increase in impact accelerations more proximal up the system are related to induced neuromuscular fatigue reducing the musculoskeletal system's ability to effectively dissipate impact energy. Based on this claim, the present result would seem to indicate that the musculoskeletal system has a diminished capacity to attenuate and dissipate the foot strike initiated transient acceleration with increasing fatigue levels, however, it was apparent that the frequency components associated with impact phase of stance (impact attenuation) were not modified. In support, others reported a similar paradox in that they observed changes in impact accelerations in the time domain but not in the frequency domain[1]. Although, it remains unclear in the present study as to why the impact attenuation did not change despite the modifications in impact accelerations at the head. We speculate that this paradox in results may be due to the peaks identified in the time domain acceleration data still containing high and low frequency signals that are not attributed to transient impacts after ground contact. Our frequency domain measure of impact attenuation (which remained consistent) is likely to be a more reliable measure of the biomechanical responses to impact in the current protocol. This is because this measure is focused on frequencies associated with impact (10-20 Hz), with the low and high frequency portions of the leg and head acceleration profiles that are not associated with impact are removed from the analysis. Moreover, the current authors realize that the magnitudes of peak accelerations at the head are low in comparison to other previous running studies [1,18]. The possible reason for the low magnitudes observed in head accelerations may be attributed to the soft compliance (although not directly measured) of the treadmill surface further assisting with dissipation of impact energy during the stance phase of running. On the other hand, it is likely that the observed lower head accelerations in the present study may be due to DC filter removing the acceleration positional/orientation offset during running. In addition, another plausible explanation may be related to how the accelerometer was attached to the head. For instance, it could be that frequency response between the accelerometer and head/skull was poor as compared to others who used a bite bar accelerometer with better mechanical coupling (and thus a higher resonance frequency) [40].

It has been well established that stride characteristics and active joint motion during the impact phase play an important role in the modulation of leg impact accelerations during running [2,4,6,31]. For example, research has shown significantly greater leg impact accelerations with a 20% increase in stride length from runners preferred [4]. In the present study, runners were able to maintain their stride characteristics irrespective of increased run duration and high physiological demands (increased RPE responses). Similarly, others found no changes in stride length after a fatiguing run [31]. Evidently, it is suspected that this lack of change in stride characteristics with prolonged run duration may be accountable for the control of leg impact accelerations. These findings rejected our hypothesis of an increase in stride length and a decrease in stride frequency with prolonged high-intensity running. Based on previous reports [14,16], we believed that runners would be forced (during treadmill running) to adopt a strategy to conserve energy through which they would decrease their stride frequency and increase stride length in order to maintain running velocity and subsequently this change would result in an increase in leg impact accelerations with fatigue.

With the present study showing consistent stride characteristics throughout each run it is plausible that the trained distance runner again may have more effective mechanical coping strategies with prolonged running at a high physiological stress as compared to not so well trained counterparts [11,12,14,16]. It's possible that inexperienced runners who undertake prolonged, intense, physiologically demanding runs may not be able to maintain their running mechanics and impact attenuation during those runs and perhaps place themselves at an greater risk of injury due to them not coping from a mechanical perspective. Moreover, considering the important associations with the manipulation of stride characteristics and energy costs during running [2], a recent study by Vernillo, Savoldelli, Zignoli, Skafidas, Fornasiero, La Torre, Bortolan, Pellegrini, Schena [41] showed that despite a significant increase (3.9%) in step frequency after runners completed a ultramarathon, only a significant increase in energy costs was observed during downhill running and not in the level and uphill running. In contrast, others reported significant increases in energy cost with a 4.2% increase in stride frequency after a marathon [42]. Considering these discrepancies, it is apparent that there is still ambiguity within the literature on gait responses to fatigue on subsequent energy demands during running.

One limitation of this study was that lower-extremity joint kinematics or kinetics were not collected during the treadmill runs. Given that joint mechanics at initial contact are considered to play an important role in the modification of impact accelerations during running [1,6,43], it would have been beneficial to have assessed those parameters in the present study. Another limitation of the current study is that no quantitative measures of the fatigue were collected. More objective measures to quantify fatigue levels such as EMG, isokinetic strength, twitch contractile stimulation, or oxygen consumption measurements would have offered greater insight into the type and levels of fatigue that were induced by the treadmill runs. Finally, in the majority of laboratory-based running fatiguing studies, including the present study, runners are usually forced to run at a controlled speed set by the treadmill, as opposed to real-world training and racing scenarios in which runners typically regulate their speeds based on sensory inputs outlined by the 'central governor model' [44]. Nevertheless, although the present findings provide an insight into the coping strategies for impact acceleration during high intensity treadmill running, there is still a need for future studies to investigate impact accelerations during real-world outdoor training and racing environments - by the use of portable outdoor inertial sensor systems [37]. Findings from such studies would not only help with the understanding of runners impact acceleration patterns but may provide a greater insight into the mechanisms which cause impact-related injuries in runners.

In conclusion, this present study found that trained runners are able to effectively control leg impact accelerations and impact attenuation during sustained high intensity running. It was apparent that despite the dramatic increases in perceived exertion levels, running mechanics such as stride length and frequency remained consistent. This indicates that trained runners are able to cope mechanically whilst being under a high physiological demand.

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Figure Captions

Figure 1 – A representative subject acceleration profiles of the leg and head during running at the start and end of each run. Arrows indicate the impact acceleration peak during early stance.

Figure 2 - Leg (A) and head (B) impact acceleration values (mean and SD error bars) during both 20-minute runs. Solid line = First run, Dashed line = Second run. *Denotes post hoc differences ($p < .05$) compared to the time point at the start of first run and †

denotes post hoc differences ($p < .05$) compared to the time point at the start of second run

Figure 3 - Cycle time (A), stride length (B) and stride frequency (C) values (mean and SD error bars) during both runs. Solid line = First run, Dashed line = Second run.

Figure 4 – Impact attenuation values (mean and SD error bars) during both runs. Solid line = First run, Dashed line = Second run.