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Can Trained Runners Effectively Attenuate Impact Acceleration During Repeated High-Intensity Running Bouts?

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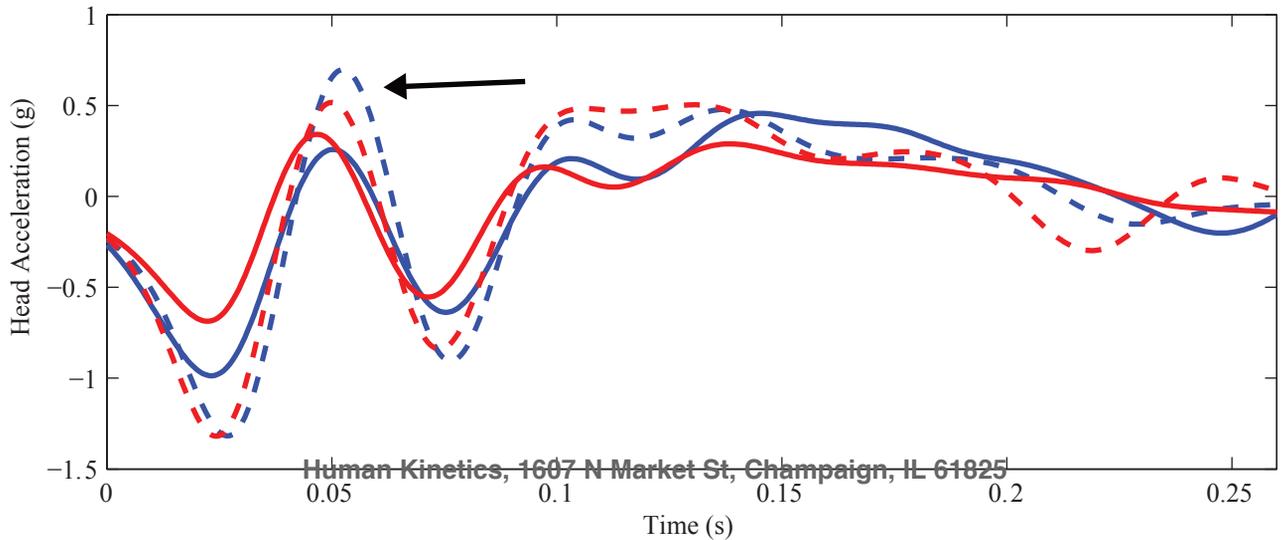
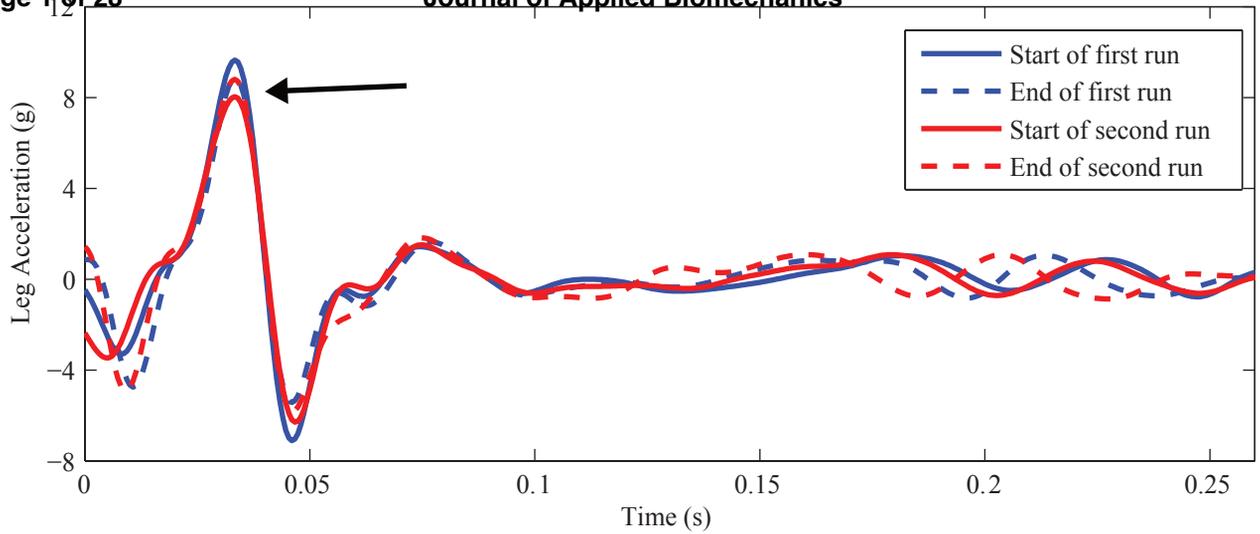


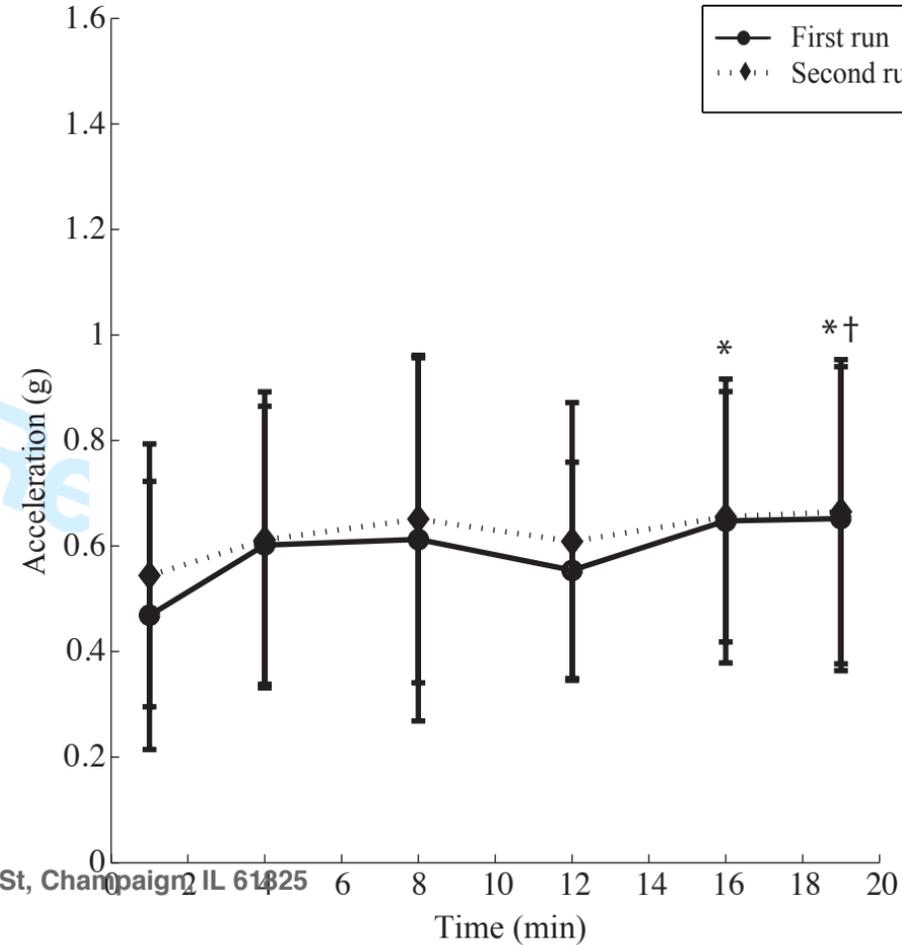
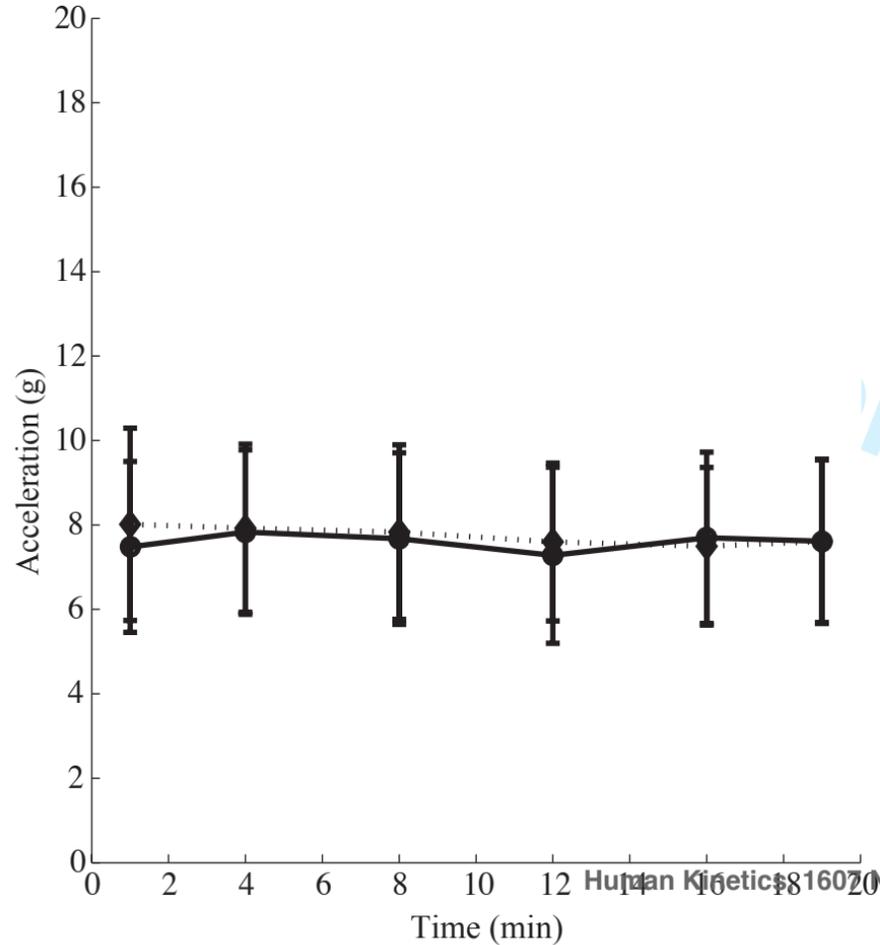
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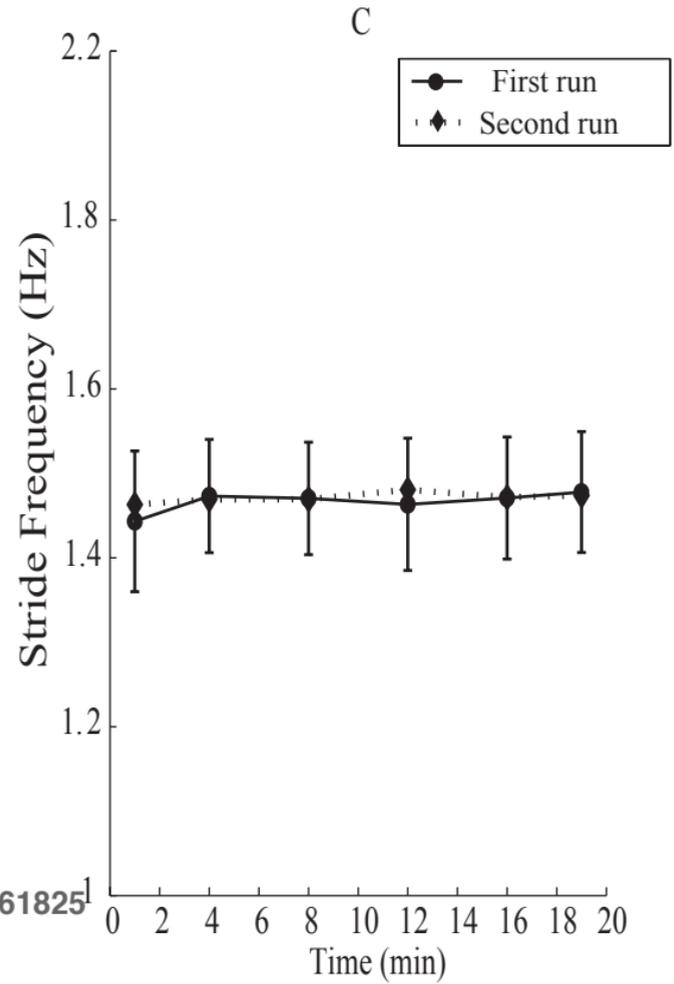
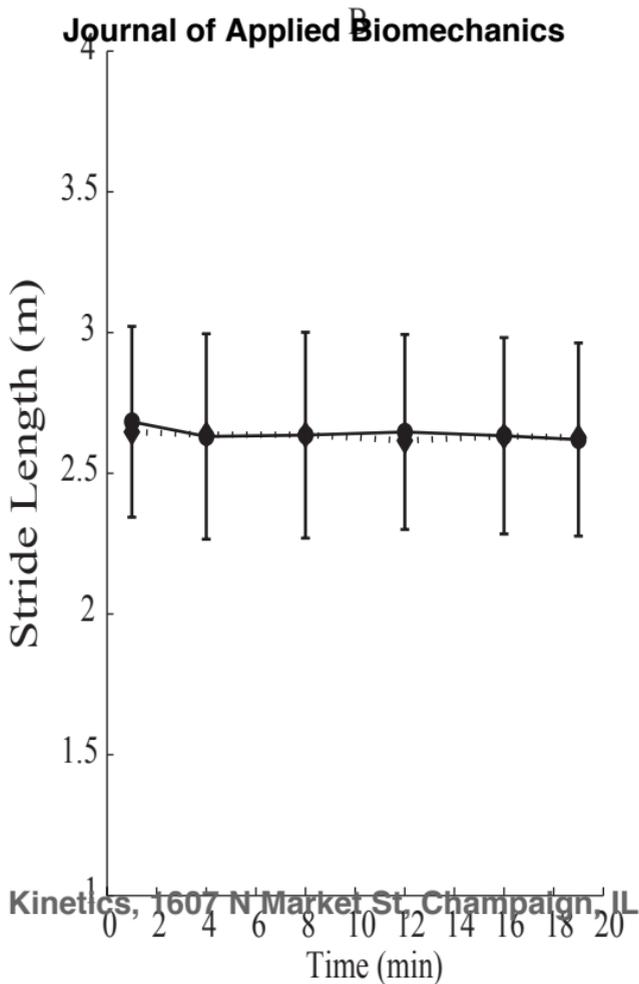
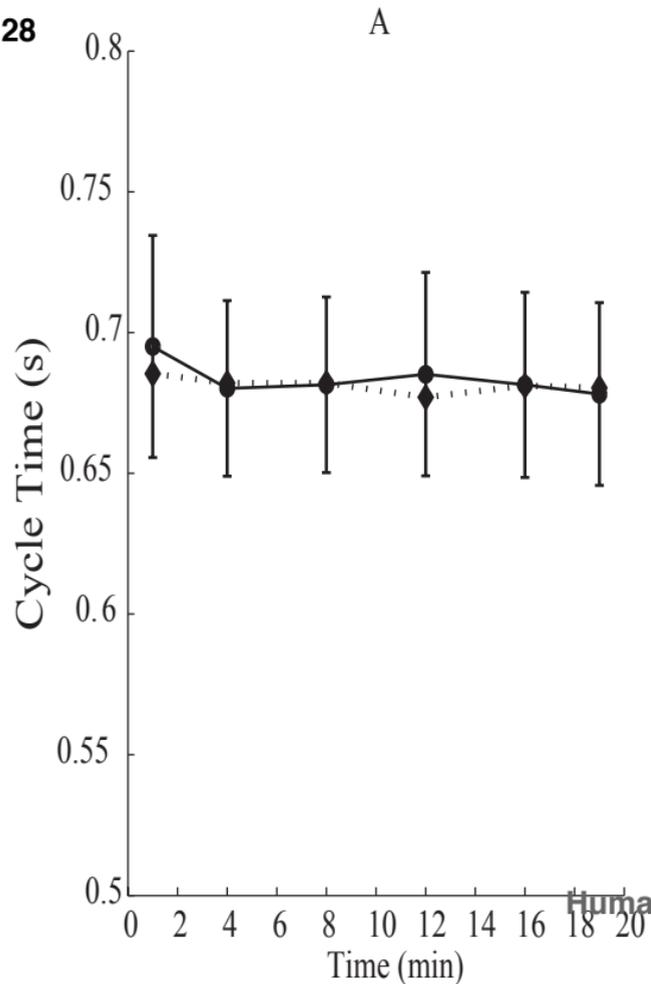
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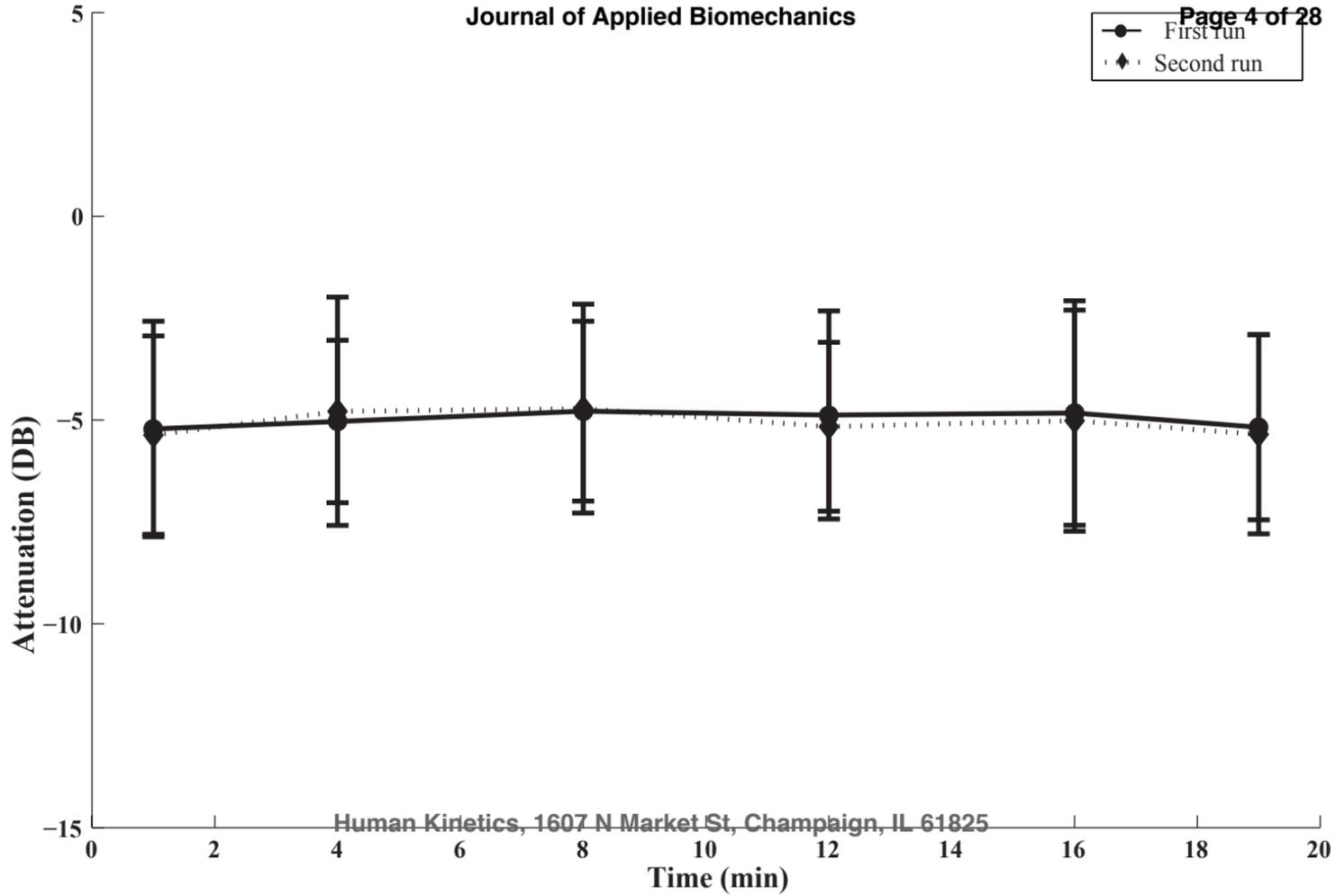
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Review









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

3
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39 **Abstract**

40

41 The purpose of this study was to investigate the effects of prolonged high intensity
42 running on impact accelerations in trained runners. Thirteen male distance runners
43 completed two 20-minute treadmill runs at speeds corresponding to 95% of onset of
44 blood lactate accumulation. Leg and head accelerations were collected for 20 s every 4th
45 minute. Rating of perceived exertion (RPE) scores were recorded during the 3rd and last
46 minute of each run. RPE responses increased ($p < .001$) from the start (11.8 ± 0.9 ,
47 moderate intensity) of the first run to the end (17.7 ± 1.5 ; very hard) of the second run.
48 Runners maintained their leg impact acceleration, impact attenuation, stride length and
49 stride frequency characteristics with prolonged run duration. However, a small (0.11 -
50 $0.14g$) but significant increase ($p < .001$) in head impact accelerations were observed at
51 the end of both first and second runs. It was concluded that trained runners are able to
52 control leg impact accelerations during sustained high-intensity running. Alongside the
53 substantial increases in perceived exertion levels, running mechanics and frequency
54 domain impact attenuation levels remained constant. This suggests that the present
55 trained runners are able to cope from a mechanical perspective despite an increased
56 physiological demand.

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

58

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60 **Word count: 3606**

61

62 Introduction

63

64 It is estimated that a runner covering 20 miles per week will on average collide with
65 the ground over 1.3 million times in a one-year period ¹. These repetitive high impact
66 loads during running transmit impact shock waves up through the musculoskeletal system
67 ^{2,3}. The body is known to act as a low-pass filter, whereby the high, transient acceleration
68 of the lower leg (leg impact acceleration (input)) is severely dampened before it reaches
69 the head (head impact acceleration (output)) ²⁻⁴. The process of dissipating these impact
70 energies during running is termed impact attenuation, which is a function of the output
71 relative to the input ⁴. Between the foot and the head these impact energies are attenuated
72 passively by components such as muscle, heel fatpad, bone, cartilage, synovial fluid and
73 other structural components ^{4,5} and actively by eccentric muscular contractions
74 controlling lower limb joint motion during landing ⁴⁻⁶.

75

76 The associated reduction in neuromuscular functionality with prolonged running ⁷⁻⁹, has
77 led researchers to consider that this impairment may decrease the impact absorbing
78 capacity of the body and therefore lead to a greater risk of overuse injury development or
79 degenerative disease ¹⁰⁻¹³. Nevertheless, despite these claims, the influence of sustained
80 high-intensity effort running on impact accelerations (peak positive axial component) still
81 appears to be conflicting and unclear. Several studies have reported significant increases
82 in leg impact acceleration with prolonged running ^{1,11,13-16}, while others have reported no
83 changes ¹⁷⁻¹⁹. Researchers believe that these discrepancies are attributed to the type of
84 fatigue protocols implemented and also the trained status of participants ^{18,20}. Mizrahi,

85 Verbitsky, Isakov¹² who ran active individuals at their anaerobic threshold speed for 30
86 min found significant increases in leg impact accelerations 15 minutes into the run. This
87 increase in leg impact acceleration was also observed in other studies who conducted
88 similar protocols and with comparable subject cohort groups^{11,14}. In contrast, Mercer,
89 Bates, Dufek, Hreljac¹⁸ found no changes in leg and head accelerations while running
90 before and after a graded exhaustive treadmill run. Similarly, Abt, Sell, Chu, Lovalekar,
91 Burdett, Lephart¹⁷ who recruited experienced runners reported both consistent leg and
92 head impact accelerations before and after a 17.8 minute exhaustive treadmill run.
93 Another possible explanation for the discrepancies in results could be related to
94 differences between treadmill and overground running. For instance, individuals are
95 constrained to run at a constant speed during treadmill running and the fluctuations of
96 impact accelerations may be less pronounced than in overground running where speed is
97 self-regulated and more variable. .

98
99 While the previous reports provide an insight into the effects on high-intensity
100 running impact accelerations, little attention has been focused on examining how trained
101 runners modulate impact accelerations during repeated bouts of prolonged high intensity
102 running. The majority of studies that recruited trained runners implemented an
103 exhausting, single bout, short duration run protocol^{1,17,18,21}, however, these types of
104 situations are not representative of what runners perform during their weekly training
105 schedule. Generally, trained runners are known to perform repeated bouts of high
106 intensity running during single training sessions^{7,22}. Therefore, the purpose of this
107 investigation was to examine the effects of two bouts of high-intensity running on impact

108 acceleration and attenuation. It was hypothesized that leg impact acceleration would
109 increase progressively over the two runs, whereas, head impact acceleration would
110 remain consistent, and be indicative of an improvement in impact attenuation with
111 prolonged running. Given that stride characteristics are known to play an important role
112 in the modulation of impact accelerations during running^{2,23,24}, the present study also
113 aimed to investigate runners stride characteristic during these repeated bouts of high-
114 intensity running. It was hypothesized that runners would increase their stride length and
115 decrease stride frequency in order to maintain velocity when they were towards the end
116 of the prolonged running bouts.

117

118 **Method**

119

120 **Participants**

121

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

127

128 **Determination of 95% of OLBA**

129

130 A week prior to the high intensity run protocol, participants underwent a standardized
131 incremental lactate threshold running test²⁵. This test was used to determine each
132 participant's speed for the high intensity run protocol at 95% of OBLA, which equates to
133 a blood lactate concentration of 3.5 Mm. All participants were familiarized with
134 treadmill running and were instructed not run or train 24 h prior to testing. The test
135 consisted of participants running on a treadmill (T170 DE, HP Cosmed, UK) at a 1%
136 gradient for 3-minute stages. Before and after the test all participants performed a 5-min
137 warm-up and cool-down jog. The test started at a speed that was perceived
138 'comfortable/easy pace' for participants to run. Between each stage a 30-s time period
139 was allocated to allow the collection of blood from each participant's fingertip.
140 Participant blood lactate levels were measured and recorded using a lactate analyser
141 (Lactate Pro; UK). The treadmill speed was increased 1 km·hr⁻¹ every 3-minute stage.
142 The test was terminated once participants' blood lactate concentration exceeded 4.0 Mm.
143 Each participant's velocity was based on their OBLA of 3.5 Mm²⁶. This 95% OBLA
144 threshold marker was chosen based on previous reports that showed reductions in
145 muscular strength after runners completed on average 40 minutes of treadmill running
146^{27,28}. The 95% OBLA marker was determined by polynomial regression model outlined
147 by Newell, Higgins, Madden, Cruickshank, Einbeck, McMillan, McDonald²⁵. The
148 average 95% of OBLA speed was 14.0 ± 2.4 km·hr⁻¹.

149

150 **Procedures**

151

152 A week after the determination of OBLA participants returned to the laboratory for
153 the second time to perform the fatigue testing. After this, participants completed two
154 bouts of 20-minute treadmill runs at 95% of their OBLA. Between the first and second
155 20-minute running bouts participants performed six discontinuous overground running
156 trials at $4.5 \text{ m}\cdot\text{s}^{-1}$ over a 15-m runway. This was part of a previous related study²⁰. The
157 total duration of these trials lasted for 3-5 minutes in duration.

158
159 Two lightweight (2.8 g) biaxial ($\pm 10\text{g}$; frequency range of 5-5 kHz) accelerometers
160 (Noraxon, Scottsdale, AZ) were attached to participant's distal anteromedial aspect of the
161 tibia and anterior aspect of the forehead before the running²⁰. The accelerometer had a
162 built in DC filter (5 Hz High-pass filter) that removed the acceleration offset related to
163 the orientation or position. To minimize the influence of angular motion of the shank on
164 the leg acceleration profiles, the accelerometer was placed as close as possible to the
165 ankle joint (i.e. approx. 7cm-10cm above the malleoli). These sites were selected to
166 minimize the effects of soft tissue oscillations during impact. As a precaution to reduce
167 any unwanted skin oscillations, the skin around the accelerometers sites was stretched
168 using adhesive kinesiology tape (Vivomed, UK). At the site of leg (tibia) accelerometer
169 attachment, the skin was shaved using sterilized razors and then cleaned. The axial axis
170 of the leg accelerometer was aligned with the longitudinal axes of the tibia bone. Once
171 the accelerometers were attached, they were securely tightened using self-gripping
172 bandage. In addition, participants wore a headband to further secure the accelerometer to
173 the head. The axial component of both sets of accelerometry data were recorded at 1500
174 Hz for 20 s and were captured during the first, 4th, 8th, 12th, 16th and last minute of each

175 run. Rating of perceived exertion (RPE) scores were measured on a 6-20 point scale and
176 were collected during the third and last minute of each run. RPE is used as a non-invasive
177 physiological valid tool for prescribing exercise intensity²⁹ and has been previously used
178 to determine if physiological fatigue was likely to have occurred while running at a given
179 running speed³⁰.

180

181 **Data analysis**

182

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

198

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

199

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

202

203 The transfer function values at impact frequencies of 10-20 Hz were averaged to
204 obtain a measure of the impact attenuation in the body ¹. The reason for selecting this
205 frequency portion was due to its association with transient impact phase of the foot
206 contacting the ground ². For example, a greater absolute value in the 10-20 Hz range
207 indicated a greater impact attenuation. The peak axial accelerations of the head and leg
208 were extracted during the early impact phase of stance and averaged over each 20-s trial
209 (Figure 1). The number of ground contacts analyzed varied between runners depending
210 individual running speed and stride length, but typically there was around 25 contacts in a
211 20-s trial. Stride characteristics were calculated based on the peaks of the leg impact
212 accelerations for each trial ³². Cycle time was the average time between each consecutive
213 leg impact acceleration (stride) and stride frequency was calculated as the inverse of this
214 time. Once stride frequency was calculated, stride length was computed from the
215 treadmill velocity and stride frequency.

216

217 **Statistical analysis**

218

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

230

231 Results

232

233 Runners were able to maintain their leg accelerations throughout each high-intensity
234 run (Figure 2). At the start of the first run, leg impact accelerations were 7.67 (2.1) g and
235 remained constant at 8.03 (2.3) g for the last minute of the run (Figure 2A). Similarly, leg
236 impact accelerations were 7.43 (2.1) g at the first minute of the second run and did not
237 change at 8.04 (2.3) g by the last min of the second run. A significant main effect of run
238 ($p < .035$) and time ($p < .001$) in head impact acceleration was observed. In the first run,
239 head impact acceleration increased significantly from the start at 0.47 (0.25) g to the 16th
240 at 0.61 (0.31) g and last minute of the run at 0.65 (0.31) g. Likewise, head impact
241 accelerations also significant increased in the last minute (0.66 (0.28) g) of the second run

242 compared to the start of the run (0.55 (0.20) g). It was apparent that there was a general
243 increased offset in head impact acceleration values in the second run as compared to the
244 first run.

245

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

258

259 Discussion

260

261 The primary purpose of this study was to investigate the effects of high intensity running
262 on both head and leg impact accelerations in trained runners. The prescribed run protocol
263 consisting of two consecutive bouts of 20 minute runs at 95% of OBLA was shown to be
264 successful in progressively and substantially increasing runners perceived exercise

265 exertion levels. Although no measures of fatigue were collected in the present study, it is
266 plausible that the current protocol elicited a level of fatigue similar to previous protocols
267 that implemented a 95% of OBLA running intensity^{27,28}. The results of this study
268 indicated that runners were able to effectively maintain their leg accelerations across both
269 prolonged 20-minute runs at 95% of their OBLA. This finding rejects our hypothesis in
270 which we expected greater leg impact accelerations with increased run duration. Contrary
271 to our findings, significant increases in leg impact accelerations were found^{11,12,14-16}. A
272 possible explanation for the conflicting findings may be attributed to the trained status of
273 the runners. In the present study we recruited trained distance runners, whereas the
274 previous studies had healthy participants without an endurance background^{11,12,14}. With
275 trained distance runners being frequently exposed to prolonged running, it is likely that
276 they have better mechanical coping strategies when placed under an increased
277 physiological demand as compared to the less experienced non-runner counterpart. In
278 support, others reported consistent leg impact accelerations after experienced runners
279 completed an exhaustive running protocols^{17,18}. Whilst these results are in support of the
280 present study's, it is recognized that the shorter duration and exhaustive incremental
281 protocol designs may be enough to induce a high level of neuromuscular fatigue that has
282 been associated with a reduction in the muscles ability to effectively attenuate impact
283 acceleration during running. Accordingly, it is probable that the incremental short
284 duration exhaustive protocols may have impaired more of the central mechanisms (such
285 as heart and lung function) as opposed to the peripheral mechanisms (neuromuscular
286 function and neural transmission), that are responsible for controlling impacts pre-landing
287 during running^{33,34}. Evidence has shown that longer duration running protocols^{9,35} elicit

288 greater impairments in muscular activation and strength as compared to shorter and high
289 intensity protocols^{27,28} and these neuromuscular impairments have been associated with
290 central fatigue⁷. With this being the case, it may be that the consistent leg impact
291 accelerations are due to the current protocol not eliciting sufficient impairments to the
292 peripheral and central mechanisms that control for landing phase during running.
293 Evidently, given the differences in results between studies, it is acknowledged that
294 numerous factors such as subjects training status, exercise duration, exercise intensity and
295 exercise type, can all play an important role on the outcome of impact acceleration
296 results.

297
298 Since leg impact accelerations are positively correlated with speed¹, another important
299 consideration for the conflicting findings may be related to the experimental designs that
300 controlled speed and those that allowed it to vary. Studies that controlled speed observed
301 subtle changes in leg impact acceleration and running technique with prolonged running
302^{19,31,36}, whereas, when speed was allowed to vary during running leg impact accelerations
303 changes were clearly larger³⁷. Therefore, we acknowledge that controlling speed in the
304 present study may have lead to only subtle changes in running mechanics, however we
305 speculate this may not be the case with more severe fatigue levels.

306

307 Studies have found that regardless of magnitude of input acceleration at the leg
308 during running (un-fatigued state) the output acceleration at the head remains consistent
309^{2,23}. The authors believe that this maintenance may be related to the system's goal of
310 wanting to optimize the stability of the head for allowing clear and consistent information

311 to the vestibular and visual systems^{38,39}. Although leg impact accelerations remained
312 consistent with prolonged high-intensity running, our results showed a 30.5% and 20%
313 increase in head impact acceleration between the first and last minute of the first and
314 second run. While several studies reported no changes in head impact accelerations with
315 high-intensity running^{1,15-19}, one study did report significant increases in impact
316 accelerations at the sacrum site 20 minutes into a high intensity run¹¹. The authors of this
317 study claimed that the increase in impact accelerations more proximal up the system are
318 related to induced neuromuscular fatigue reducing the musculoskeletal system's ability to
319 effectively dissipate impact energy. Based on this claim, the present result would seem to
320 indicate that the musculoskeletal system has a diminished capacity to attenuate and
321 dissipate the foot strike initiated transient acceleration with increasing fatigue levels,
322 however, it was apparent that the frequency components associated with impact phase of
323 stance (impact attenuation) were not modified. In support, others reported a similar
324 paradox in that they observed changes in impact accelerations in the time domain but not
325 in the frequency domain¹. Although, it remains unclear in the present study as to why the
326 impact attenuation did not change despite the modifications in impact accelerations at the
327 head. We speculate that this paradox in results may be due to the peaks identified in the
328 time domain acceleration data still containing high and low frequency signals that are not
329 attributed to transient impacts after ground contact. Our frequency domain measure of
330 impact attenuation (which remained consistent) is likely to be a more reliable measure of
331 the biomechanical responses to impact in the current protocol. This is because this
332 measure is focused on frequencies associated with impact (10-20 Hz), with the low and
333 high frequency portions of the leg and head acceleration profiles that are not associated

334 with impact are removed from the analysis. Moreover, the current authors realize that the
335 magnitudes of peak accelerations at the head are low in comparison to other previous
336 running studies^{1,18}. The possible reason for the low magnitudes observed in head
337 accelerations may be attributed to the soft compliance (although not directly measured) of
338 the treadmill surface further assisting with dissipation of impact energy during the stance
339 phase of running. On the other hand, it is likely that the observed lower head
340 accelerations in the present study may be due to DC filter removing the acceleration
341 positional/orientation offset during running. In addition, another plausible explanation
342 may be related to how the accelerometer was attached to the head. For instance, it could
343 be that frequency response between the accelerometer and head/skull was poor as
344 compared to others who used a bite bar accelerometer with better mechanical coupling
345 (and thus a higher resonance frequency)⁴⁰.

346

347 It has been well established that stride characteristics and active joint motion during
348 the impact phase play an important role in the modulation of leg impact accelerations
349 during running^{2,4,6,31}. For example, research has shown significantly greater leg impact
350 accelerations with a 20% increase in stride length from runners preferred⁴. In the present
351 study, runners were able to maintain their stride characteristics irrespective of increased
352 run duration and high physiological demands (increased RPE responses). Similarly,
353 others found no changes in stride length after a fatiguing run³¹. Evidently, it is suspected
354 that this lack of change in stride characteristics with prolonged run duration may be
355 accountable for the control of leg impact accelerations. These findings rejected our
356 hypothesis of an increase in stride length and a decrease in stride frequency with

357 prolonged high-intensity running. Based on previous reports ^{14,16}, we believed that
358 runners would be forced (during treadmill running) to adopt a strategy to conserve energy
359 through which they would decrease their stride frequency and increase stride length in
360 order to maintain running velocity and subsequently this change would result in an
361 increase in leg impact accelerations with fatigue.

362

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

380

381 One limitation of this study was that lower-extremity joint kinematics or kinetics
382 were not collected during the treadmill runs. Given that joint mechanics at initial contact
383 are considered to play an important role in the modification of impact accelerations
384 during running^{1,6,43}, it would have been beneficial to have assessed those parameters in
385 the present study. Another limitation of the current study is that no quantitative measures
386 of the fatigue were collected. More objective measures to quantify fatigue levels such as
387 EMG, isokinetic strength, twitch contractile stimulation, or oxygen consumption
388 measurements would have offered greater insight into the type and levels of fatigue that
389 were induced by the treadmill runs. Finally, in the majority of laboratory-based running
390 fatiguing studies, including the present study, runners are usually forced to run at a
391 controlled speed set by the treadmill, as opposed to real-world training and racing
392 scenarios in which runners typically regulate their speeds based on sensory inputs
393 outlined by the 'central governor model'⁴⁴. Nevertheless, although the present findings
394 provide an insight into the coping strategies for impact acceleration during high intensity
395 treadmill running, there is still a need for future studies to investigate impact
396 accelerations during real-world outdoor training and racing environments - by the use of
397 portable outdoor inertial sensor systems³⁷. Findings from such studies would not only
398 help with the understanding of runners impact acceleration patterns but may provide a
399 greater insight into the mechanisms which cause impact-related injuries in runners.

400

401 In conclusion, this present study found that trained runners are able to effectively
402 control leg impact accelerations and impact attenuation during sustained high intensity

403 running. It was apparent that despite the dramatic increases in perceived exertion levels,
404 running mechanics such as stride length and frequency remained consistent. This
405 indicates that trained runners are able to cope mechanically whilst being under a high
406 physiological demand.

407

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409

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414

415

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531

Figure Captions

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

535

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

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

541

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

544

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

547