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ORIGINAL ARTICLE

Step length and grade effects on energy absorption and impact attenuation in running

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Abstract
We sought to examine the effect of step length manipulation on energy absorption and impact attenuation during graded running. Nineteen runners (10F, 9M) ran on an instrumented treadmill at three step lengths (preferred and ±10% preferred) at each of five grades (0°, ±5°, and ±10°) while 3D motion data were captured. Speed was held constant at 3.33 m/s and step length was manipulated by syncing cadence to a metronome. Manipulating step length altered energy absorption ($p \leq 0.002$) and impact attenuation ($p < 0.0001$) across all grades. Energy absorption at the knee joint was most responsive to step length manipulations [Δ range (±10%SL-PrefSL) = 0.076–0.126 J/kg, $p < 0.0001$], followed by the ankle (Δ range = 0.026–0.100 J/kg, $p = 0.001$) and hip (Δ range = 0.008–0.018 J/kg, $p < 0.006$). Shortening step length reduced knee joint energy absorption at all grades with the smallest effect observed during uphill running ($\Delta \geq -0.053$ J/kg), while large reductions occurred during level ($\Delta = -0.096$ J/kg) and downhill running ($\Delta \geq -0.108$ J/kg). Increasing step length resulted in greater knee joint energy absorption ($p \leq 0.037$) across all grades of running. Impact attenuation was greatest at long step lengths ($\Delta = 2.708$) and lowest at short step lengths ($\Delta = -2.061$), compared to preferred. Overall, Step length influenced the energy absorption and impact attenuation characteristics of the lower extremity during level and graded running. Adopting a shorter step length may be a useful intervention to reduce knee joint loading, particularly during downhill or level running. Elongating step length placed a greater demand on the lower extremity joints, which may expedite the development of neuromuscular fatigue.

Keywords: Gait modification, cadence, lower extremity loading, uphill and downhill running

Highlights

- Step length influenced the energy absorption and impact attenuation characteristics of the lower extremity during level and graded running; however, the grade of running moderated the effect of step length.
- Impact attenuation remained constant across grades of running but was altered via step length manipulation due to concomitant changes in joint energy absorption.
- Reducing step length may be a useful intervention to reduce knee joint loading, particularly during level and downhill running.
- Increasing step length may not be a beneficial gait modification, particularly during downhill running, due to the increased joint energy absorption that would reduce the capacity to run for prolonged periods of time.

Introduction

The impact between the foot and the ground during running causes a rapid acceleration of body segments that diminishes in magnitude, due to active and passive attenuation processes, as one travels proximally from the foot-ground interface (Derrick, 2004; Derrick, Hamill, & Caldwell, 1998). A causal relationship between segmental accelerations and running-related injury risk has been suggested, with greater accelerations reported to increase risk of

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injury (Chan et al., 2018; Milner, Ferber, Pollard, Hamill, & Davis, 2006). Given that running biomechanics change as a function of grade (Vernillo et al., 2017), it is possible that the magnitude to which impacts are attenuated and the relative contribution of active attenuation processes could also change with grade. Ultimately, these biomechanical changes may be an important factor in injury risk and running performance during graded running.

While it may be intuitive to use peak impact accelerations to estimate injury risk, alterations in effective mass can have large effects on segmental accelerations, that do not reflect changes in the applied force (Derrick, 2004). This makes it difficult to interpret changes in segmental accelerations when large alterations in contact geometry are expected, such as during graded running (Vernillo et al., 2017). Alternatively, impact attenuation may provide complimentary information, as it reflects the dissipation of impact energy through active and passive attenuation processes (Derrick, 2004). Impact attenuation is typically defined as the change in the high frequency power of segmental accelerations between distal and proximal segments (Hamill, Derrick, & Holt, 1995). When viewed alongside changes in joint energy absorption, changes in impact attenuation can provide insight into the relative contribution of active attenuation processes (Derrick et al., 1998). Muscles exhibit the greatest ability to attenuate impact acceleration through eccentric contractions (Derrick et al., 1998), but high magnitudes of eccentric work are associated with muscle damage (Eston, Mickleborough, & Baltzopoulos, 1995) and reduction in running performance (Braun & Dutto, 2003). Impact attenuation, impact magnitude, and joint energy absorption can be altered via step length manipulations (Derrick et al., 1998; Hamill et al., 1995; Mercer, Devita, Derrick, & Bates, 2003), and this may be a beneficial intervention during graded running to improve performance and reduce injury risk (Chan et al., 2018; Millet, Hoffman, & Morin, 2012; Rowlands, Eston, & Tilzey, 2001).

There exists a positive relationship between step length and impact attenuation during running, in which longer than preferred step lengths are associated with greater attenuation and *vice versa* (Hamill et al., 1995; Mercer et al., 2003). Since head acceleration is typically maintained at a constant magnitude when manipulating step length, the greater tibial acceleration associated with elongated step lengths is often observed alongside greater impact attenuation (Busa, Lim, van Emmerik, & Hamill, 2016; Mercer et al., 2003). Greater attenuation is accomplished by increased energy absorption at lower extremity joints when running with elongated step lengths (Derrick et al., 1998; Heiderscheit, Chumanov, Michalski, Wille, & Ryan, 2011). In contrast, running with a shortened step length is associated with lower tibial acceleration (Derrick et al., 1998) and a reduction in impact attenuation that is accomplished by absorbing less energy at the lower extremity joints (Derrick et al., 1998; Heiderscheit et al., 2011).

Studies investigating the interaction between step length, energy absorption, and impact attenuation have only been performed on level ground. Graded running places unique demands on the musculoskeletal system when compared to level ground running, resulting in differences in physiological demand and gait kinematics and kinetics (Gottschall & Kram, 2005; Vernillo et al., 2017). Downhill running is associated with greater impact magnitudes (Chu & Caldwell, 2004) and increased energy absorption when compared to level running (Buczek & Cavanagh, 1990; Vernillo et al., 2017). The increased eccentric muscular work required to absorb more energy during downhill running may also be associated with muscle damage and delayed onset muscle soreness (DOMS), which negatively affects running performance (Braun & Dutto, 2003; Chen, Nosaka, & Tu, 2007). In contrast, uphill running is more energetically costly than level or downhill running (Minetti, Moia, Roi, Susta, & Ferretti, 2002) but is associated with lower impact magnitudes and reduced lower extremity energy absorption, especially when compared to downhill running (Devita, Janshen, Rider, Solnik, & Hortobágyi, 2008). Step length and frequency are also known to change during graded running (Vernillo et al., 2017), and step length manipulation may aid in understanding the injury and performance implications of these natural changes to preferred step length.

The purpose of this study was to examine the effect of step length manipulations on impact attenuation and joint energy absorption during level, uphill, and downhill running. It was hypothesized that impact attenuation would remain constant across grades of running. It was also hypothesized that increasing step length would be associated with an increase in joint energy absorption and impact attenuation, while decreasing step length would be associated with a decrease in joint energy absorption and impact attenuation, during both level and graded running.

Methods

Participants

Nineteen participants [10 females (27 ± 10 years, 1.84 ± 0.04 m, 66.8 ± 6.9 kg), 9 males (28 ± 8 years, 1.79 ± 0.06 m, 73.7 ± 8.0 kg)] were recruited for this study following written informed consent and approval by the University of Calgary Conjoint Health Research Ethics Board (#REB14-1117). All participants were injury free within the last six months prior to participation, regularly performed physical activity including running, and were comfortable running on a treadmill. Each participant visited the lab on two separate occasions. During the first visit, participants were familiarized to the treadmill gradients and footwear to be used for experimental conditions (Salomon X-Scream 3d, mass = 303 g, heel height = 33.7 mm, heel-to-toe insole drop = 12.3 mm). Familiarization included running for 1 min at each of the five grades used during testing (0°, ±5°, and ±10°), and 1 min of running at each grade while practicing step length manipulations of ±10%. Participant height and mass were measured using a balance and stadiometer, and anthropometric measurements of the lower limbs and pelvis were used to estimate segment masses, moments of inertia, and centre of mass locations derived from Vaughan, Davis, and O'Connor (1999).

Experimental protocol

Upon arrival for the second visit, participants were prepared for motion analysis. Seventeen retro-reflective markers were placed on the participants' right lower limb to create three-dimensional segment coordinate systems for the pelvis, thigh, shank, and foot; a single researcher performed all marker placements. Markers on the anterior superior iliac spine (bilateral), greater trochanter (bilateral), and sacrum were used to create the pelvis coordinate system. The thigh, shank, and foot coordinate systems were defined using markers on the medial and lateral midthigh, lateral femoral condyle, proximal and distal anterior shank, posterior shank, lateral malleoli, heel, fifth metatarsal head, and dorsal foot. Static trials were recorded to establish virtual markers and segmental reference systems with additional markers placed on the medial condyle and malleolus to define the knee and ankle joint centres, respectively. These two markers were then removed during running trials. The marker coordinates were used to track the motion of the pelvis, femur, shank, and foot through space, each of which were modelled as a rigid segment. Tri-axial accelerometers were placed on the bony surface of the distal 1/3 of the participant's anteromedial tibia and the sacrum, with the vertical axis of the accelerometer placed in line with the long axis of the segment. A sacral accelerometer was chosen, in place of a head-mounted accelerometer, as most of the impact attenuation occurs due to lower extremity active attenuation processes (Derrick et al., 1998). Accelerometers were also placed on the right shoe of each participant: one at the heel just above the midsole and another on top of the shoe above the metatarsals. All accelerometers were tightly affixed to the skin or shoe using adhesive tape. Participant's performed a 5-min warm up and accommodation at a self-selected speed on an instrumented treadmill (Bertec, Columbus, Ohio, USA). For the study protocol, participants ran at each of five grades (0°, ±5°, and ±10°) in a random order. Within each graded condition, participants ran at three step frequencies, which were always completed in the same order (preferred, −10% preferred, and +10% preferred). Preferred step frequency, specific to each grade, was determined by counting the number of steps the participant took over a 20 s period. Running speed was held constant at 3.33 m/s and step frequency was manipulated by asking participants to match their cadence to the beat of a metronome, thereby manipulating step length. Foot strike pattern was not constrained between running conditions. After the participant had acclimated to the target speed and step rate, 15 s of data were captured at each condition. Participants were given as much rest as they desired between conditions to ensure they were comfortable and alert throughout the protocol. Both motion capture (Vicon Motion Systems Ltd., Oxford, England) and force platform data were collected concurrently using Vicon Nexus software (Vicon Motion Systems Ltd., Oxford, England) at 200 and 1000 Hz, respectively. The synchronized accelerometry data were collected at 2000Hz using LabChart 8 software (ADInstruments, Sydney, Australia).

Data analysis

Kinematics and kinetics. All data were processed using Matlab software (MathWorks, Natick, Massachusetts, U.S.A.). The stance phase of running was determined from the vertical ground reaction force and a 20 N threshold. For each participant and condition, the data from 10 stance phases were analyzed. Step length was calculated as the product of running speed and the time between contralateral foot strikes. Marker and ground reaction force data were used to calculate hip, knee, and ankle joint angles and moments during stance. Joint kinematics were calculated using a flexion-extension, adduction-abduction, and internal-external Cardan sequence of rotations, and internal joint moments were calculated using an inverse dynamics analysis with rigid body assumptions. Lower extremity joint power in the sagittal plane, normalized to participant mass, was calculated as the product of the net joint moment and angular velocity for the hip, knee, and ankle. For this study, inverse dynamics was performed on the raw data and data filtering was performed on the calculated joint powers. This has been suggested as a method to retain signal power, while reducing artifact associated with separate kinematic and kinetic processing methods (Edwards, Troy, & Derrick, 2011). Joint power data were filtered using a 4th order low-pass Butterworth filter with a cut-off frequency selected to maintain 95% of the signal power. Average cut off frequencies for the hip, knee, and ankle were 21, 22, and 22 Hz, respectively. Joint energy absorption was calculated from each joint power curve as the time integral of all negative data points during the absorption phase of running. The absorption phase was defined as the time between foot strike and the instant that the anterior-posterior ground reaction force crossed zero during midstance.

Accelerometry. Time- and frequency-domain analyses were performed on the axial acceleration signals from the tibia- and sacrum-mounted accelerometers. It is well documented that high frequency components (>10 Hz) of segmental accelerations are produced by foot-ground impact (Shorten & Winslow, 1992). For time-domain analysis, a high-pass Butterworth filter with a cutoff frequency of 5-Hz was first applied to isolate the impact component of acceleration. Peak acceleration magnitudes were then quantified. For frequency-domain analyses, the time-series acceleration signals were separated into individual segments from the instant of peak negative tibial acceleration prior to foot contact through the subsequent 0.3-s (Figure 1(A)). The data were then zero padded to a total of 1024 points. The power spectral density was then calculated, with a frequency resolution of 1.95 Hz, using a Fourier transform (Figure 1(B)) (Shorten & Winslow, 1992). Impact attenuation between the tibia and sacrum at each frequency within the 10–30 Hz frequency range (Bediz, Nevzat Özgüven, & Korkusuz, 2010; Shorten & Winslow, 1992) was quantified (Figure 1(C)) with a transfer function:

$$\text{Transfer Function}_i = 10\,\log_{10}(\mathrm{PSD}_{\mathrm{sacrum},i}/\mathrm{PSD}_{\mathrm{tibia},i})$$

where TF_i is the attenuation (in dB) between the tibia ($\mathrm{PSD}_{\mathrm{tibia},i}$) and sacrum ($\mathrm{PSD}_{\mathrm{sacrum},i}$) power spectral densities for the i-th frequency bin within the 10–

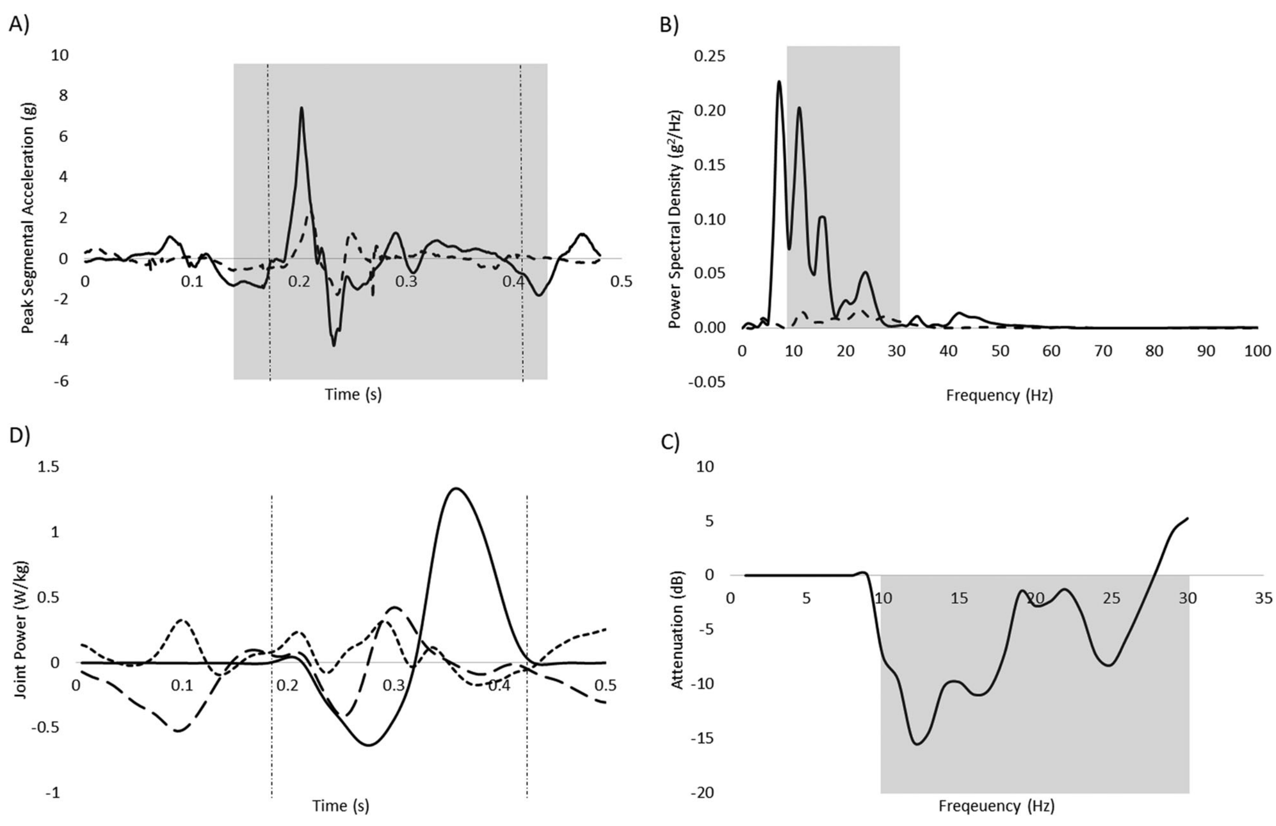

Figure 1. Graphical representation of accelerometer processing technique from a representative subject during level running. (a) Time-series acceleration data for the tibia (solid black) and sacrum (dashed black) with the dashed vertical lines representing foot-contact and foot-off, respectively. (b) Power spectral density vs. frequency for the tibia (solid black) and sacrum (dashed black). (c) Attenuation between the sacrum and tibia. The grey windows isolate the region of interest in each plot. (d) Time-series data for ankle (solid line), knee (wide dashed line), and hip (narrow dashed line) joint power for the same representative subject and trial.

30 Hz frequency range. An average TF within the 10–30 Hz range was subsequently determined.

The two accelerometers affixed to the participant's shoe were used to identify foot strike pattern during running conditions using a previously validated method (Giandolini et al., 2014). Briefly, the difference in time between the metatarsal and heel acceleration peak was used to classify foot strike pattern (FSP) with, fore-foot strike <−5.49 ms <mid-foot strike <15.2 ms <rear-foot strike (Giandolini et al., 2014).

Statistical analysis

Dependent variables of interest included hip, knee, and ankle sagittal plane energy absorption, peak tibial acceleration, peak sacral acceleration, impact attenuation, and step length. Variables were calculated for each of the ten analyzed stance phases per condition, and then averaged to calculate the mean of each condition. A two-way (3 step lengths × 5 grades) repeated measures ANOVA was used to examine differences between conditions ($\alpha = 0.05$). The assumption of sphericity was verified using the Mauchly's test. If violations to this assumption were observed, the Greenhouse-Geisser or Huynh-Feldt method was used to overcome the effects of violation. In the case of a significant interaction or condition effect, *post-hoc* testing was performed using a Bonferroni correction. All statistical analyses were performed in SPSS version 24 (IBM, Houston, Texas, USA).

Results

Mean, standard deviation, and results of statistical analyses for discrete variables of interest are presented in Table I.

In general, participants transitioned to a more anterior FSP across uphill running conditions, whereas they maintained a similar FSP during downhill conditions as to their level running FSP (Figure 2).

Subjects were able to successfully modify their gait to meet step frequency targets. On average, subjects increased their step frequency by 9.5% during short step length conditions and reduced their step frequency by 9.7% during long step length conditions. The greatest errors from step frequency targets were observed during uphill running for short step lengths (+8.8 and 8.9% for +5° and +10°, respectively), and during downhill running for long step lengths (−9.7 and −9.2% for −5° and −10°, respectively).

Energy absorption

A significant step length × grade interaction was observed for hip ($p = 0.001$) (Figure 3(A)) and knee energy absorption ($p < 0.001$) (Figure 3(B)). At the hip, decreasing step length reduced hip joint energy absorption at −10° only ($p = 0.013$), while increasing step length resulted in greater hip joint energy absorption during level ground only ($p = 0.024$). At the knee, decreasing step length reduced knee joint energy absorption at all grades ($p \leq 0.01$), but the magnitude of the effect was not consistent across grades. Decreasing step length resulted in the greatest reduction in knee joint energy absorption compared to preferred step length during level ($\Delta = -0.096$ J/kg) and −5° ($\Delta = -0.108$ J/kg) running (Figure 3(B)), and the lowest effect during uphill running ($\Delta \geq -0.053$ J/kg). Increasing step length resulted in greater knee joint energy absorption at all grades ($p \leq 0.003$), with the greatest increase occurring during −5° and −10° conditions.

At the ankle, a main effect of step length was observed for ankle joint energy absorption ($p < 0.001$) (Figure 3(C)). Running with a shortened step length resulted in no change in ankle joint energy absorption ($p = 0.164$), while increasing step length resulted in an average 0.1 J/kg increase in ankle joint energy absorption across all grades ($p = 0.001$). A main effect of grade was also observed ($p < 0.001$) (Figure 3(D)), with energy absorption increasing across grades (+10°, +5°, 0°, −5°, −10°).

Accelerometry

No interaction was observed for any of the accelerometry-derived measures. However, a main effect of step length ($p < 0.001$) was observed for impact attenuation, where attenuation was greatest during long step lengths ($p < 0.001$) and lowest during short step lengths ($p < 0.001$) (Figure 4(A)). The effect of step length manipulation on impact attenuation was the result of changes in segmental accelerations. Peak tibial acceleration was greatest during long step lengths when compared to both preferred and short step lengths ($p \leq 0.001$) (Figure 4(B)), which were not significantly different from each other ($p = 1.000$). In contrast, peak sacral acceleration was greatest during short step lengths when compared to both preferred and long step lengths ($p < 0.001$) (Figure 4(C)), which were not significantly different from each other ($p = 0.083$).

No main effect of grade was observed for impact attenuation ($p = 0.265$) (Figure 4(A)); however, a main effect of grade was observed for peak tibial acceleration ($p < 0.001$) and peak sacral acceleration ($p < 0.001$). Peak tibial acceleration increased during downhill running ($p \leq 0.024$) (Figure 4(B)) relative to level ground, but no difference in peak

Table I. Mean (SD) for discrete variables of interest in each grade and step length condition.

	+10°	+5°	0°	−5°	−10°	*P*-Value
Step Length (m)						
−10% SL	0.98 (0.07)$^{bcde,\ *\wedge}$	1.03 (0.06)$^{acd,\ *\wedge}$	1.08 (0.07)$^{ab,\ *\wedge}$	1.09 (0.07)$^{ab,\ *\wedge}$	1.04 (0.09)$^{a,\ *\wedge}$	<0.001$_{\text{step length}}$
Preferred SL	1.06 (0.06)$^{bcde,\ \#\wedge}$	1.12 (0.07)$^{acd,\ \#\wedge}$	1.18 (0.08)$^{ab,\ \#\wedge}$	1.20 (0.08)$^{ab,\ \#\wedge}$	1.15 (0.09)$^{a,\ \#\wedge}$	<0.001$_{\text{grade}}$
+10% SL	1.18 (0.08)$^{bcde,\ *\#}$	1.25 (0.07)$^{acd,\ *\#}$	1.30 (0.09)$^{abd,\ *\#}$	1.33 (0.09)$^{abce,\ *\#}$	1.27 (0.10)$^{ad,\ *\#}$	0.009$_{\text{grade × step length}}$
Negative Hip Work (J/Kg)						
−10% SL	−0.024 (0.023)	−0.012 (0.010)^{de}	−0.010 (0.006)$^{de,\ \wedge}$	−0.038 (0.024)$^{bc,\ \wedge}$	−0.046 (0.040)$^{bc,\ \wedge}$	0.002$_{\text{step length}}$
Preferred SL	−0.016 (0.025)^{de}	−0.010 (0.009)^{de}	−0.015 (0.017)$^{de,\ \wedge}$	−0.055 (0.038)^{abc}	−0.076 (0.056)$^{abc,\ \#}$	<0.001$_{\text{grade}}$
+10% SL	−0.020 (0.032)^{de}	−0.013 (0.014)^{cde}	−0.039 (0.039)$^{bde,\ *\#}$	−0.077 (0.062)$^{abce,\ \#}$	−0.110 (0.090)$^{abcd,\ \#}$	0.001$_{\text{grade × step length}}$
Negative Knee Work (J/Kg)						
−10% SL	−0.045 (0.035)$^{bcde,\ *\wedge}$	−0.105 (0.064)$^{acde,\ *\wedge}$	−0.216 (0.118)$^{abde,\ *\wedge}$	−0.431 (0.177)$^{abc,\ *\wedge}$	−0.497 (0.217)$^{abc,\ *\wedge}$	<0.001$_{\text{step length}}$
Preferred SL	−0.080 (0.046)$^{bcde,\ \#\wedge}$	−0.158 (0.087)$^{acde,\ \#\wedge}$	−0.313 (0.154)$^{abde,\ \#\wedge}$	−0.539 (0.233)$^{abc,\ \#\wedge}$	−0.586 (0.184)$^{abc,\ \#\wedge}$	<0.001$_{\text{grade}}$
+10% SL	−0.127 (0.061)$^{bcde,\ *\#}$	−0.269 (0.136)$^{acde,\ *\#}$	−0.425 (0.185)$^{abde,\ *\#}$	−0.718 (0.279)$^{abc,\ *\#}$	−0.767 (0.302)$^{abc,\ *\#}$	<0.001$_{\text{grade × step length}}$
Negative Ankle Work (J/Kg)						
−10% SL	−0.477 (0.098)$^{bcde,\ \wedge}$	−0.568 (0.134)$^{ace,\ *\wedge}$	−0.654 (0.189)$^{abe,\ *\wedge}$	−0.699 (0.274)$^{ae,\ *\wedge}$	−0.985 (0.384)^{abcd}	<0.001$_{\text{step length}}$
Preferred SL	−0.489 (0.102)$^{bcde,\ \wedge}$	−0.605 (0.134)$^{acde,\ \#\wedge}$	−0.707 (0.215)$^{abe,\ \#\wedge}$	−0.762 (0.284)$^{abe,\ \#\wedge}$	−0.951 (0.334)^{abcd}	<0.001$_{\text{grade}}$
+10% SL	−0.559 (0.142)$^{bcde,\ *\#}$	−0.678 (0.165)$^{acde,\ *\#}$	−0.805 (0.268)$^{abe,\ *\#}$	−0.901 (0.347)$^{abe,\ *\#}$	−1.074 (0.387)^{abcd}	0.198$_{\text{grade × step length}}$
Impact Attenuation (dB)						
−10% SL	−0.825 (6.025)$^{*\wedge}$	−0.244 (7.929)$^{*\wedge}$	−1.760 (7.335)$^{*\wedge}$	−1.626 (5.436)$^{*\wedge}$	−0.187 (5.126)$^{*\wedge}$	<0.001$_{\text{step length}}$
Preferred SL	−2.578 (5.720)$^{\#\wedge}$	−2.364 (7.128)$^{\#\wedge}$	−3.796 (5.134)$^{\wedge}$	−4.068 (4.579)$^{\#\wedge}$	−2.142 (4.325)$^{\#\wedge}$	0.265$_{\text{grade}}$
+10% SL	−4.933 (5.397)$^{*\#}$	−6.102 (5.450)$^{*\#}$	−6.384 (4.560)$^{*\#}$	−6.399 (4.524)$^{*\#}$	−4.669 (4.158)$^{*\#}$	0.510$_{\text{grade × step length}}$
Peak Tibial Acceleration (g)						
−10% SL	6.986 (4.094)^{e}	7.071 (3.928)^{de}	8.148 (4.927)^{e}	10.236 (5.238)$^{b,\ \wedge}$	12.194 (5.270)^{abc}	<0.001$_{\text{step length}}$
Preferred SL	6.731 (3.994)$^{e,\ \wedge}$	6.111 (3.405)^{de}	7.184 (3.884)$^{de,\ \wedge}$	10.650 (4.412)$^{bc,\ \wedge}$	13.382 (5.619)^{abc}	<0.001$_{\text{grade}}$
+10% SL	9.232 (6.012)$^{e,\ *}$	8.146 (5.153)^{e}	9.374 (4.804)$^{e,\ *}$	13.083 (5.539)$^{*\#}$	14.928 (5.432)^{abc}	0.488$_{\text{grade × step length}}$
Peak Sacral Acceleration (g)						
−10% SL	6.095 (4.035)$^{e,\ *\wedge}$	6.695 (3.976)$^{e,\ *\wedge}$	6.346 (3.669)$^{e,\ *\wedge}$	8.088 (4.635)$^{*\wedge}$	10.042 (5.335)$^{abc,\ *\wedge}$	<0.001$_{\text{step length}}$
Preferred SL	3.959 (2.327)$^{de,\ \#\wedge}$	4.522 (2.689)$^{e,\ \#\wedge}$	4.328 (2.466)$^{de,\ \#}$	6.124 (3.550)$^{ac,\ \#}$	8.557 (4.368)$^{abc,\ \#}$	<0.001$_{\text{grade}}$
+10% SL	2.976 (1.622)$^{de,\ *\#}$	3.567 (2.040)$^{e,\ *\#}$	4.201 (2.826)$^{e,\ \#}$	5.626 (3.503)$^{a,\ \#}$	7.780 (4.184)$^{abc,\ \#}$	0.518$_{\text{grade × step length}}$

Note: [a]different from +10°, [b]different from +5°, [c]different from 0°, [d] different from −5°, and [e]different from −10°. * different from preferred step length (SL), # different from −10% SL, and ^ different from +10% SL.

tibial acceleration was observed when comparing level ground to uphill running conditions ($p \geq 0.845$). A similar pattern was observed for peak sacral acceleration, which was greater during the −10° condition ($p < 0.001$) (Figure 4(C)), but no differences were observed in uphill running ($p \geq 0.139$) when compared to level ground.

Discussion

Running outdoors, particularly trail running, is often characterized by frequent changes in graded terrain. Manipulating step length has been reported to influence lower extremity loading and impact attenuation during level ground running (Derrick et al., 1998), and this may have important

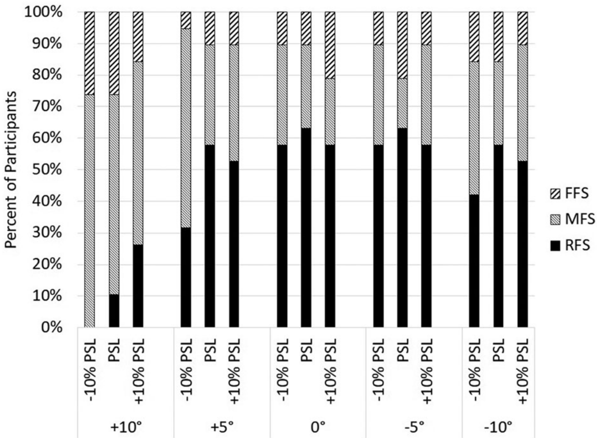

Figure 2. Percent of participants who utilized a fore-foot strike, mid-foot strike, or rear-foot strike during the experimental conditions.

implications for both injury prevention and performance. In this study, we sought to determine the effect of step length manipulations on impact attenuation and joint energy absorption during downhill, level, and uphill running. Overall, step length manipulations altered energy absorption and impact attenuation during graded and level ground running; however, the grade of running moderated the effect of step length.

Energy absorption

While the hip, knee, and ankle were all responsive to step length manipulations, the largest changes in energy absorption were observed at the knee joint. Shortening step length resulted in reductions in knee joint energy absorption that formed a pseudo u-shaped curve across grades (+10°, +5°, 0°, −5°, −10°), peaking during level and −5° conditions. The ability to reduce energy absorption demands during downhill running may be of great importance given that knee joint energy absorption increased by 77% and 94% during −5° and −10°, respectively, when compared to level ground. Since muscle damage caused by eccentric exercise is the main contributor of DOMS (Friden & Lieber, 2001), running with a shortened step length during downhill running

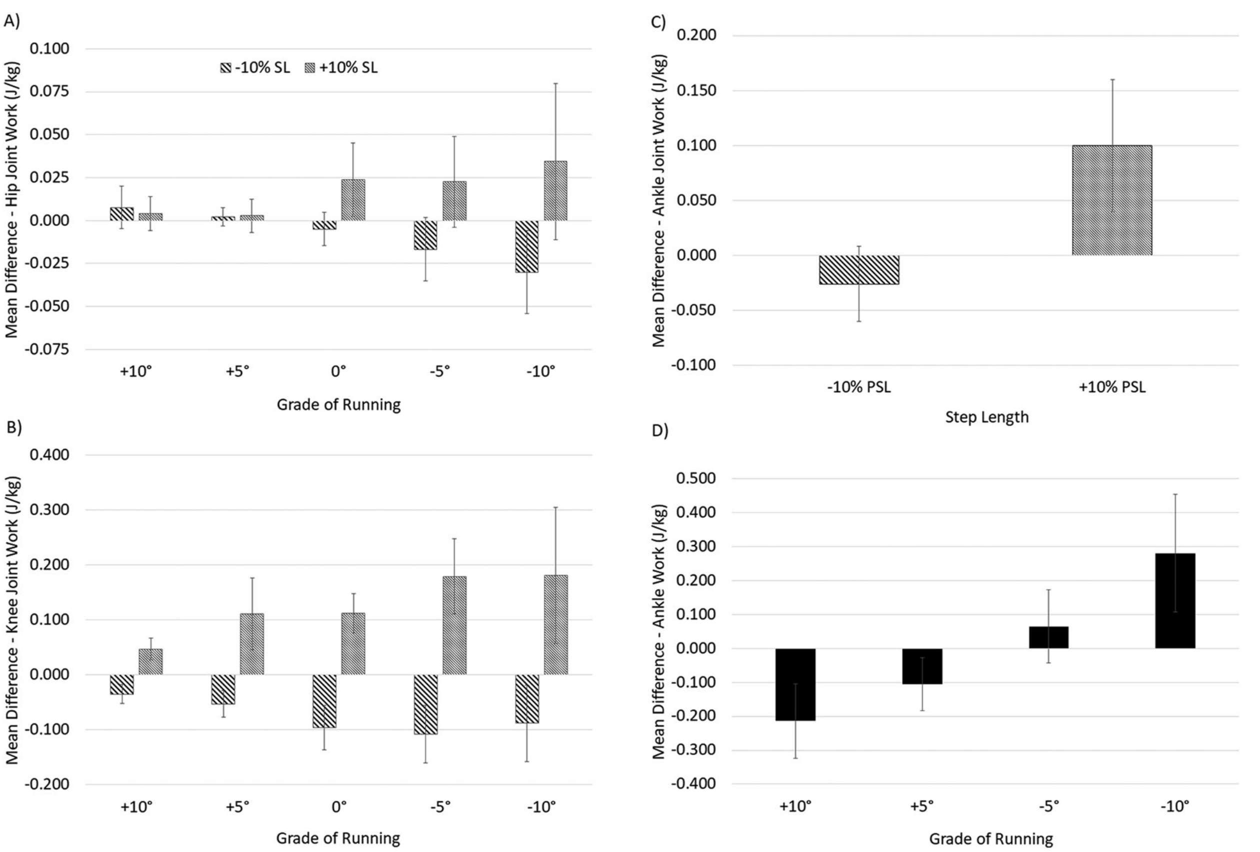

Figure 3. Mean difference and 95% confidence interval (CI) in (a) negative hip joint work, (b) negative knee joint work, and (c) negative ankle joint work during −10% step length (SL) vs. preferred SL (heavy patterned bars) and +10% SL and preferred (light patterned bars). (d) presents mean difference and 95% CI in negative ankle joint work between uphill and downhill running grades and 0°. A positive difference indicates an increase in energy absorption and a negative number indicates a reduction in energy absorption. A 95% confidence interval that crosses zero indicates a non-significant difference.

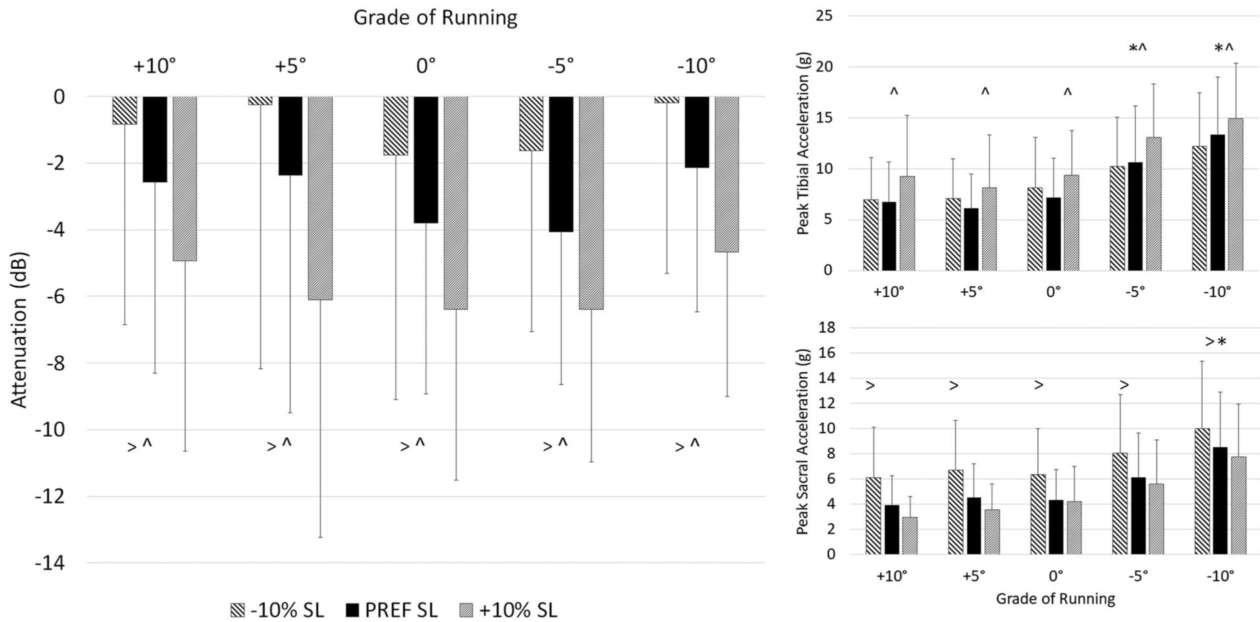

Figure 4. Mean ± one standard deviation for (a) impact attenuation, (b) peak tibial acceleration, and (c) peak sacral acceleration during each grade and step length condition. * indicates a significant ($p < 0.05$) difference from 0° across all step lengths. ^indicates a significant difference ($p < 0.05$) between +10% SL and preferred SL. >indicates a significant difference ($p < 0.05$) between −10% SL and preferred SL.

may reduce the performance deficits associated with DOMS (Braun & Dutto, 2003). In fact, Rowlands et al. (2001) observed that running with a shortened step length reduced muscle strength losses following a downhill running bout when compared to running with a preferred step length, suggesting that eccentric muscle damage encountered during downhill running is mitigated when running with a shorter step length. While Rowlands et al. (2001) only examined the knee extensors, the hip extensors are also prone to DOMS following downhill running (Jafariyan, Monazzami, Nikosefat, Nobahar, & Yari, 2017), and the present findings demonstrate that shortening step length can reduce hip energy absorption during steep downhill running. Collectively, these findings suggest that reducing step length may be an effective intervention to mitigate eccentric muscle damage of the knee extensors during both shallow and steep downhill grades and in the hip extensors during steep downhill grades.

While running with a short step length during uphill running was also associated with reductions in energy absorption at the knee, the benefit this provides is likely small, as energy absorption demands during uphill running were substantially less than during level or downhill running. Furthermore, given the greater energy cost of uphill running compared to level and downhill running (Minetti et al., 2002), further increases in energy cost due to manipulating step length (Snyder & Farley, 2011) would be expected to have performance detriments.

Increasing step length was associated with greater energy absorption at the knee and ankle across all grades of running, while hip joint energy absorption was greater during level running only. This suggests that individuals match the greater energy absorption demands of long step lengths by increasing hip contributions during level ground running, but likely rely on different joints to compensate during uphill and downhill running. This re-distribution of joint energy absorption may partially explain the large increases in knee joint energy absorption observed with longer step lengths during downhill running. While the knee may be most affected by increasing step length, the plantar flexors are also prone to DOMS following downhill running (Maeo, Ando, Kanehisa, & Kawakami, 2017). Increasing step length may be particularly deleterious to the knee and ankle musculature during downhill running, as increased joint energy absorption could lead to greater eccentric muscle damage and consequently greater DOMS symptoms (Eston, Lemmey, McHugh, Byrne, & Walsh, 2000). In fact, it has been observed that adopting a longer step length during downhill running is associated with greater DOMS symptoms (Rowlands et al., 2001). Given that DOMS is associated with changes in running biomechanics and increased energy cost of movement (Braun & Dutto, 2003; Chen et al., 2007), it is possible that adopting a strategy that increases DOMS symptoms (i.e. increasing step length during downhill running) would elicit further changes to an individual's gait pattern, thereby amplifying the increase in energetic cost of running.

While the effects of increased step length were much less pronounced during uphill running, step length elongation still resulted in greater joint energy absorption at the ankle and knee. While performance is unlikely to suffer due to a small increase in energy absorption, manipulating step length during uphill running will still increase the energy cost of running (Snyder & Farley, 2011) and this is in addition to the greater energy cost of running uphill (Minetti et al., 2002). Thus, increasing step length during uphill running should be avoided as it is likely to be detrimental to running performance.

Accelerometry

Step length manipulation was also associated with changes in segmental accelerations and impact attenuation. Increasing step length by 10% from preferred resulted in an average 90% increase in impact attenuation (PSL = −2.99 dB vs. +10% PSL = −5.70 dB) and shortening step length resulted in an average 68% reduction in attenuation (PSL = −2.99 dB vs. −0.929 dB) across all grades of running. It has been suggested that the increased energy absorption performed with long step lengths is a mechanism to counteract the increased tibial acceleration, in order to maintain constant head accelerations (Hamill et al., 1995). This effect was also observed in the present study, where peak tibial acceleration was greater when running with long step lengths, while peak sacral acceleration remained constant between long and preferred step lengths. However, during short step length conditions, peak sacral accelerations increased, which contradicts the observations of others (Busa et al., 2016; Hamill et al., 1995; Mercer et al., 2003). While some studies have observed that runners maintain a constant head acceleration when manipulating step length, Derrick et al. (1998) observed a linear trend in head acceleration across step lengths, where it was greatest during +20% step length and lowest during −20% step length. This suggests that proximal segment accelerations (i.e. head or sacrum) may not always be a priority in running. This idea is supported by the observation that sacral accelerations also increased during downhill running conditions, which has been observed previously (Mizrahi, Verbitsky, & Isakov, 2000). While proximal stability has been suggested to be a criterion for gait pattern selection, it seems possible that there is a point where the complexity of the task is too great to allow individuals to select a running gait that is metabolically efficient and minimizes proximal segment accelerations (Busa et al., 2016).

Alterations in tibial accelerations should also be considered within the framework of effective mass (Derrick, 2004). As the knee becomes more extended at foot contact, the effective mass of the tibia increases, resulting in reduced tibial acceleration if touchdown velocity remains constant (Derrick, 2004). While changes in knee contact angle as a function of step length were minimal ($\Delta = \pm 2°$), knee contact angle was greatly affected by grade (40° at +10° and 12° at −10°). Thus, effective mass would be greatest during downhill running and lowest during uphill running. The large effective mass during downhill running would reduce the measured acceleration; whereas, the small effective mass during uphill running would increase the measured acceleration, for a given touchdown velocity.

Overall, much work is needed to understand the link between accelerometry-derived measures, musculoskeletal injury risk, and running performance. Large alterations in effective mass during graded running make it difficult to infer information regarding the forces acting on the segments from peak accelerations. Thus, impact attenuation may be a more useful measure as it can provide information about how impact energy is being transferred through the musculoskeletal system, as observed by the concomitant changes in joint energy absorption and impact attenuation when manipulating step length.

Limitations

There are a few limitations associated with this study. First, running speed was kept constant across graded running conditions, and this may not reflect how individuals prefer to interact with graded terrain. Foot strike pattern was also not constrained between conditions; however, we believe this adds to the ecological validity of the study as it demonstrates that a mix of foot strike patterns are used by participants when navigating graded and level running terrain. Lastly, it is important to note that all previous studies, to our knowledge, examining impact attenuation and step length have been performed using head-mounted accelerometers. Sacral accelerometers will provide different information than head accelerometers (in both the time- and frequency-domain), which can be observed in the large changes in sacral accelerations observed during step length manipulations. For instance, the present study observed a 53%–69% change in impact attenuation (tibia-sacrum) during ±10% step length conditions, whereas, studies using head-mounted accelerometers observed a 4%–15% change in attenuation during ±10% step length conditions (Derrick et al., 1998; Mercer et al., 2003). While this makes it difficult to compare our findings to the literature, we believe that measuring attenuation between the tibia and sacrum may be more representative of the adaptability of lower extremity active attenuation processes.

Conclusion

The findings of the present study demonstrate that acute step length manipulations are effective at altering lower extremity energy absorption and impact attenuation during level, uphill, and downhill running. Impact attenuation appears to remain constant across running grades but can be changed via step length due to alterations in joint energy absorption. Elongating step length may not be a beneficial gait modification, particularly during downhill running, due to the increased energy absorption that would reduce the capacity to run for prolonged periods of time. In contrast, adopting a short step length during downhill running appears to be a useful intervention for reducing the demand placed on the knee and hip. This running strategy may aid in reducing muscle damage and prolonging running performance.

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