Individual Effects of Stride Length and Frequency on Shock Attenuation during Running

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ABSTRACT

MERCER, J. A., P. DEVITA, T. R. DERRICK, and B. T. BATES. Individual Effects of Stride Length and Frequency on Shock Attenuation during Running. *Med. Sci. Sports Exerc.*, Vol. 35, No. 2, pp. 307–313, 2003. Shock attenuation during running is the process of absorbing impact energy due to the foot-ground collision, reducing shock wave amplitude between the foot and head. Shock attenuation is affected by changes in stride length and stride frequency, but it is not clear whether either parameter individually affects shock attenuation. **Purpose:** To identify the independent affects of stride length (SL) and stride frequency (SF) on shock attenuation. **Methods:** Subjects ($N = 10$) completed three experiments consisting of SL and SF manipulations relative to preferred stride length (PSL) and frequency (PSF). During experiment 1, stride length was manipulated (+15% PSL, PSL, −15% PSL) while stride frequency was always set to PSF. During experiment 2, stride frequency was manipulated (+15% PSF, PSF, −15% PSF) while stride length was always set to PSL. During experiment 3, stride length and stride frequency were manipulated concurrently (+10% PSL/−10% PSF, PSL/PSF, and −10% PSL/+10% PSF). Running velocity was always the product of stride length and stride frequency. Transfer functions were calculated using tibial and forehead surface mounted accelerometer data to represent shock attenuation. **Results:** Shock attenuation changed only when stride length changed ($P < 0.05$). Specifically, shock attenuation increased as stride length increased. **Conclusion:** It was concluded that changes in stride length not stride frequency affected shock attenuation. **Key Words:** IMPACT ABSORPTION, ACCELEROMETRY, STRIDE LENGTH, LOCOMOTION

The collision between the foot and ground during running results in a shock wave of energy that is transmitted throughout the body. Shock attenuation is the process of absorbing impact energy and reducing the amplitude of the shock wave (6,20). Understanding factors that affect shock attenuation is important because the magnitude and rate of the large impact forces during the stance phase of running are hypothesized to be related to overuse injuries (12,20,25).

Shock attenuation is accomplished through a complex interaction of many factors with some of these factors related to the kinematics of running. Shock attenuation is affected by lower-extremity geometry at impact because the magnitude of the impact is affected by the spatial orientation of the lower-extremity segments at the moment of impact (5,6,16,18). Derrick et al. (5) suggested that lower-extremity stiffness varies with geometry and changes in stiffness can alter impact magnitude. Running with a greater knee flexion angle at impact can reduce lower-extremity stiffness and increase shock attenuation (18). Running velocity also directly affects shock attenuation because it has been observed that shock attenuation increased as velocity increased (22). Stride length and stride frequency, the fundamental components of running velocity, have also been shown to affect shock attenuation, but the effect of these parameters have only been studied when both are manipulated simultaneously (6,10). Derrick et al. (6) showed that when running at a constant velocity, shock attenuation increased with concurrent increases in stride length and decreases in stride frequency. Thus, it is not clear whether one or both components of running velocity affect shock attenuation. Although the relationship between limb geometry and shock attenuation suggest that stride length is the component of running velocity that influences shock attenuation, it is certainly reasonable that altered muscle forces required to change stride frequency may also affect shock attenuation.

Understanding the independent affects of stride length and stride frequency on shock attenuation is important because this knowledge will provide insight into the consequences of gait adaptations in runners. The purpose of this study was to identify the independent affects of stride length and stride frequency on shock attenuation.

METHODS

Ten male recreational runners (mean ± SD: age, 24 ± 5.8 yr; mass, 78.4 ± 9.6 kg; height, 1.84 ± 0.10 m) completed

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TABLE 1. Stride frequency and stride length combinations for each experiment.

	Velocity ($m \cdot s^{-1}$)	Stride Frequency Target	Stride Length Target	Stride Frequency Achieved ($strides \cdot s^{-1}$)	Stride Length Achieved (m)	Stance time (s)	Shock Attenuation (dB)
Experiment 1	4.4	PSF	+15% PSL	1.44 ± 0.05	3.08 ± 0.12	0.237 ± 0.021	−10.6 ± 2.7
	3.8	PSF	PSL	1.40 ± 0.06	2.75 ± 0.12	0.243 ± 0.020	−9.5 ± 3.6
	3.3	PSF	−15% PSL	1.39 ± 0.07	2.35 ± 0.11	0.260 ± 0.022	−7.4 ± 3.6
Experiment 2	4.4	+15% PSF	PSL	1.58 ± 0.10	2.81 ± 0.18	0.221 ± 0.016	−10.0 ± 3.6
	3.8	PSF	PSL	1.40 ± 0.06	2.75 ± 0.12	0.243 ± 0.020	−9.5 ± 3.6
	3.3	−15% PSF	PSL	1.22 ± 0.07	2.67 ± 0.15	0.273 ± 0.024	−9.2 ± 3.2
Experiment 3	3.8	+10% PSF	−10% PSL	1.55 ± 0.10	2.48 ± 0.16	0.227 ± 0.016	−8.1 ± 3.3
	3.8	PSF	PSL	1.40 ± 0.06	2.75 ± 0.12	0.243 ± 0.020	−9.5 ± 3.6
	3.8	−10% PSF	+10% PSL	1.26 ± 0.07	3.05 ± 0.16	0.262 ± 0.021	−9.9 ± 3.1

Preferred stride frequency (PSF) and preferred stride length (PSL) were calculated while running at 3.8 $m \cdot s^{-1}$. During each condition, subjects matched their stride frequency to that of an electronic metronome set to result in a specific stride length and stride frequency combination for a given treadmill velocity. Each subject completed three experiments, with each experiment set to manipulate stride length and/or stride frequency. During Experiment 1, subjects ran at different stride lengths using the same stride frequency. During Experiment 2, subjects ran at different stride frequencies using the same stride length. During Experiment 3, both stride length and stride frequency were changed concurrently while running at the same velocity. Treadmill velocity was always set to the product of stride length and stride frequency. Group mean and SD values of actual stride frequencies and stride lengths achieved during each condition are presented as well as stance time and shock attenuation data. For shock attenuation, a greater negative value indicates a greater amount of shock attenuation.

all conditions after giving written informed consent. The study was approved by the university's institutional review board. There was a large range of running experience between subjects, with one subject having completed a marathon whereas other subjects ran only for fitness benefits. At the time of testing, all subjects were free from injury.

Head and leg accelerations were recorded (1000 Hz) over 10 treadmill running strides by using uniaxial piezoelectric accelerometers (model 8628B50, Kistler, Amherst, NY) that had a 0.5–5 kHz frequency response, ±50-g range, and 6.7 g of mass. This model was used because the majority of frequency spectral power of head and leg acceleration signals during running is below 50 Hz and impact accelerations during running are below 20 g. Before testing, accelerometers were mounted to an impact tester to confirm identical performance. During experimental conditions, an accelerometer was secured to the forehead surface to measure head accelerations while another was secured to the distal anterior-medial aspect of the tibia to measure leg accelerations. Accelerometers were secured such that the sensitive axes were vertically aligned with the subject standing. The sensitive axes of the two accelerometers may change during manipulation of stride length. However, previous research (6) reported that changes in alignment of the leg accelerometer during stride lengths that were 40% different affected acceleration magnitude by only 0.1 g (about 1–2% of impact peak magnitude).

To secure the leg accelerometer, the accelerometer was first super-glued to a small piece of balsa wood, which was used to provide a flat surface to press against the leg. The accelerometer was then wrapped using a combination of athletic tape and adhesive compression bandage (Adban™, Kimberly-Clark, Roswell, GA). The head accelerometer was first glued to a leather headband, which could be tightened about a subject's head. An adhesive compression bandage (Adban™) was then wrapped around the head to apply additional tension to the accelerometer. For both accelerometers, enough tension was applied through the wrapping to secure the accelerometer but not cause extreme discomfort to the subject.

Three experiments were designed to systematically manipulate stride length and stride frequency (Table 1) relative to preferred stride length (PSL) and preferred stride frequency (PSF), which were determined while running at 3.8 $m \cdot s^{-1}$. Stride length and stride frequency combinations were calculated as percentages of PSL and PSF with treadmill velocity set to the product of stride length and stride frequency. During experiment 1, subjects ran at stride lengths ± 15% of PSL while stride frequency was maintained at PSF for each condition. During experiment 2, conditions consisted of running at stride frequencies ±15% of PSF with stride length maintained at PSL. During experiment 3, conditions included running at +10% of PSL and −10% PSF as well as −10% PSL and +10% PSF with velocity set at 3.8 $m \cdot s^{-1}$. The ± 10% range was selected in order to compare results with other published data (6). Subjects also ran a condition at the PSL-PSF combination at 3.8 $m \cdot s^{-1}$.

During testing, subjects matched their stride frequency to that of an electronic metronome set to the frequency required to result in the specific stride frequency-length combination for the given treadmill velocity. Data collection lasted 20 s, beginning after the subject indicated they matched the beat of the metronome and the tester confirmed that stride frequency matched the metronome beat. Condition order was balanced, with each subject randomly assigned a set order. Subjects were allowed (and encouraged) to rest between conditions to minimize the effect of fatigue.

For each subject, 10 consecutive stance phase head and leg acceleration profiles within the 20-s data set were extracted per condition and transformed to the frequency domain. Example head and leg acceleration profiles are illustrated in Figure 1 along with event markers identifying beginning and ending points of a data set extracted for analysis. This acceleration pattern was typical in all strides for all subjects and all conditions and has been observed using bone mounted (11,13,17,27) and surface mounted accelerometers (6,17,18,22,23,24). The beginning event (B, Fig. 1) was identified as the local minimum on the leg acceleration profile just before the distinct positive impact peak with the ending event (E, Fig. 1) as the local minimum

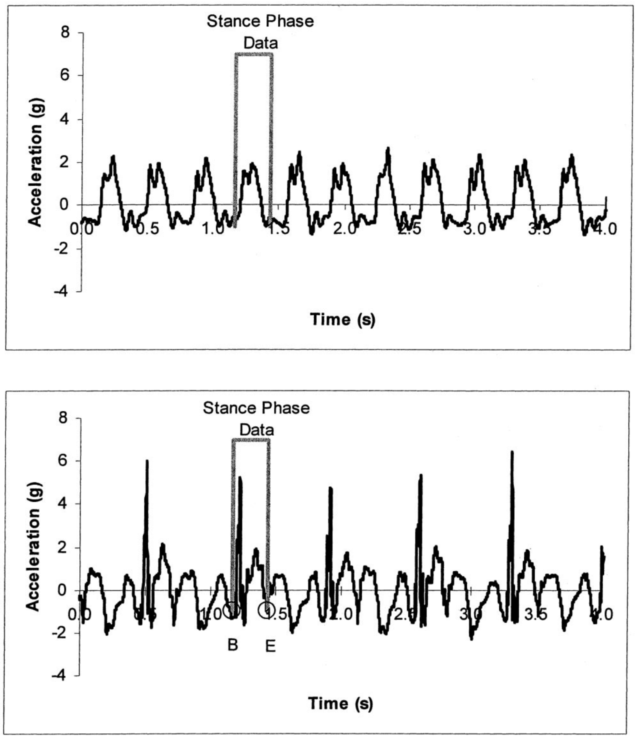

FIGURE 1—An illustration of typical head (*top*) and leg (*bottom*) acceleration profiles (4 s) during running for a single subject. Also illustrated are the discrete points used to extract a sample of data for frequency analysis, with B indicating the beginning point of a data set, E the ending point.

after a low magnitude peak. These selection criteria allowed for analysis of similar patterns between strides and conditions. Within the extracted data set, the section of interest was the impact phase, which consisted of primarily a large positive peak followed by a negative peak.

To analyze data in the frequency domain, mean and linear trends were first removed from each data set. The length of data sets needed to be a power of two in order to calculate power spectral density (PSD) using a fast Fourier transformation function (26). Data sets were therefore padded with zeros in order to total 1024 data points per acceleration profile. Power spectral densities were calculated for padded data sets, with PSD adjusted to account for changes in power due to the zero-padding procedure by accounting for the number of zeros added to the data set (22). The sample rate and length of data set analyzed resulted in 0.98-Hz frequency bins because frequency bin size is determined as the ratio of sample rate to length of data set (22,26). These bins were adjusted so each frequency bin was 1 Hz with a transfer function calculated for each frequency bin using the equation:

$$\text{TransferFunction}_i = 10 \cdot \log_{10}\left(\frac{\text{PSD}_{\text{head}_i}}{\text{PSD}_{\text{leg}_i}}\right)$$

where *i* represents a frequency bin, and PSD_{head} and PSD_{leg} represent PSD of head and leg acceleration profiles, respectively. Shock attenuation was calculated as the average transfer function across the 10- to 20-Hz frequency range (6,22). Units for shock attenuation are decibels (dB), where positive values indicate a gain and negative values attenuation of PSD_{leg} relative to PSD_{head}. For further analysis of components of shock attenuation, PSD_{head} and PSD_{leg} profiles were each averaged across the 10- to 20-Hz frequency range for each data set and then averaged across strides for each subject. Therefore, PSD_{head} and PSD_{leg} represent the PSD across the 10- to 20-Hz frequency range. These procedures to transform time-domain data to the frequency domain have been used by other researchers (6,10,22).

Stride lengths and frequencies were calculated in order to confirm that subjects achieved the assigned stride length and stride frequency combinations. These parameters were calculated for each condition *post hoc* by first calculating stride time as the time between successive leg impact peaks. Stride frequency was calculated as the inverse of stride time and stride length by dividing running velocity ($m \cdot s^{-1}$) by stride frequency ($strides \cdot s^{-1}$).

For each experiment, shock attenuation comparisons were made using a repeated measures ANOVA using SPSS (version 7.0, Chicago, IL) with an alpha level of 0.05 used as the criteria for significance. For each experiment, the primary dependent variable was shock attenuation, with PSD_{head} and PSD_{leg} tested to explain shock attenuation results. All shock attenuation data were transformed to a linear scale before any mathematical treatment. In all statistical tests, the specific stride length-stride frequency combination was the independent variable.

RESULTS

Target stride length and stride frequency combinations were achieved (Table 1). Stride lengths were 31% different and stride frequencies only 3% different in experiment 1 ($P < 0.05$). In contrast, stride frequencies were 29% different and stride lengths only 5% different during the $\pm$ 15% PSF running conditions in experiment 2 ($P < 0.05$). During experiment 3, the −10% PSL stride length was 19% different than the +10% PSL stride length ($P < 0.05$).

PSD_{head} was consistent between all running conditions, with no statistical differences between conditions for any experiment (Fig. 2). In contrast, PSD_{leg} was significantly different between conditions for each experiment (Fig. 3, $P < 0.05$). Change in PSD_{leg} was over four times greater when stride length was manipulated (Fig. 3a, 144% difference between $\pm$ 15% PSL) compared with the change in PSD_{leg} when stride frequency was manipulated (Fig. 3b, 33% difference between $\pm$ 15% PSF).

There was a significant difference in shock attenuation when stride length was manipulated and stride frequency held constant (experiment 1, $P < 0.05$), with mean shock attenuation 43% greater during the +15% PSL compared with the −15% PSL conditions (Table 1 and Fig. 4a). Shock attenuation data for all conditions are presented in Table 1 and Fig. 4. In contrast, there was no change in shock attenuation when stride frequency was manipulated and stride length held constant (experiment 2, Fig. 4b). When both stride frequency and stride length were manipulated concurrently and speed was held constant (experiment 3), shock attenuation was different between stride length and

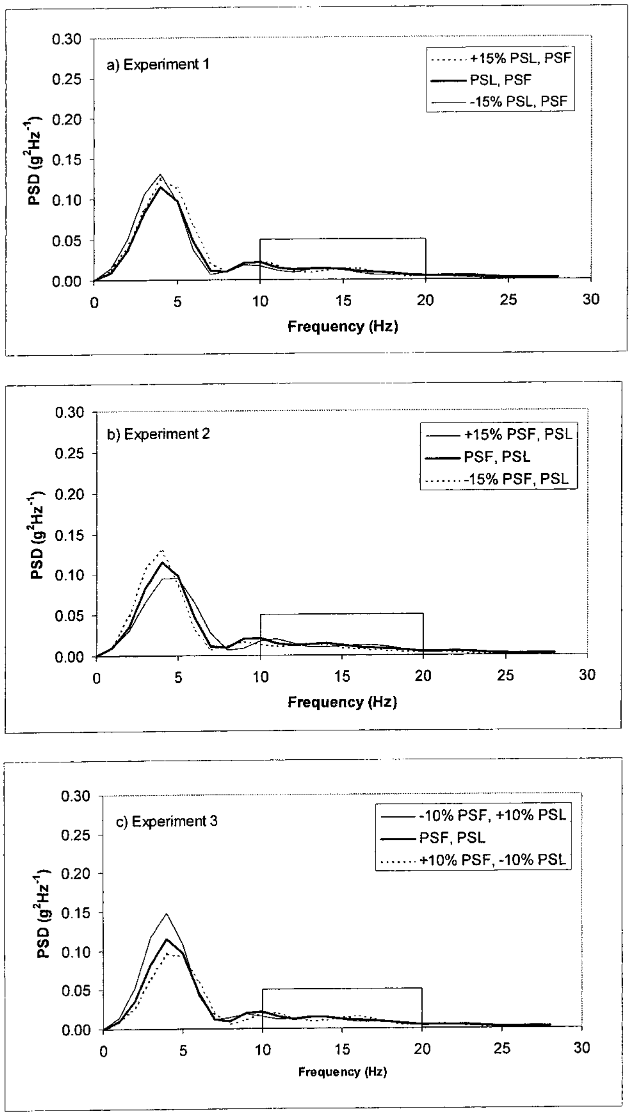

FIGURE 2—a, b, c. Group mean power spectral density for the head (PSD_{head}) while running at different stride lengths and stride frequencies. Subjects completed three experiments: a. stride length manipulated (±15% preferred stride length); b. stride frequency manipulated (±15% preferred stride frequency); c. stride length and stride frequency manipulated concurrently (±10% PSF and PSL). PSD_{head} data analyzed were averages across the 10- to 20-Hz frequency range as indicated by the box. There was no difference in PSD_{head} between any conditions within any experiment ($P > 0.05$).

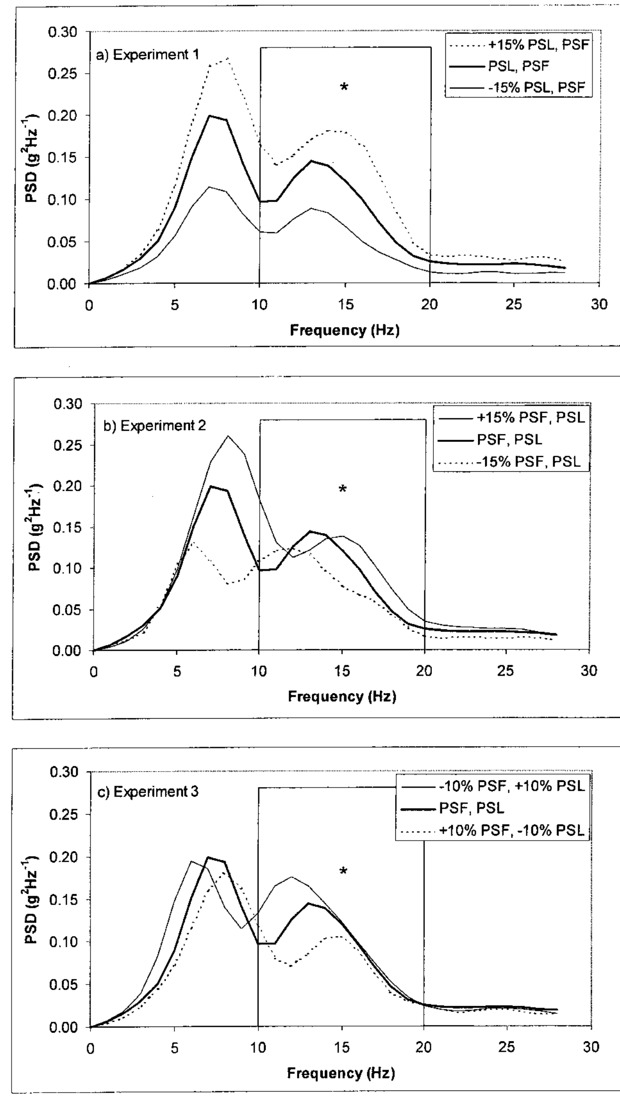

FIGURE 3—a, b, c. Group mean power spectral density for the leg (PSD_{leg}) during three experiments: a. stride length manipulated (±15% preferred stride length); b. stride frequency manipulated (±15% preferred stride frequency); c. stride length and stride frequency manipulated concurrently (±10% PSF and PSL). PSD_{leg} data analyzed were averages across the 10- to 20-Hz frequency range as indicated by the box. PSD_{leg} was different between conditions for all experiments ($P < 0.05$). PSD_{leg} was 144% different between the longest and shortest stride lengths (experiment 1), 33% different between the highest and lowest stride frequencies (experiment 2), and 57% different between longest and shortest stride lengths used in experiment 3.

stride frequency combinations ($P < 0.05$), with shock attenuation about 18% greater during the +10% PSL/−10% PSF compared with −10% PSL/+10% PSF conditions (Fig. 4c). Group statistical results appeared to be valid indicators of single subject responses as 9 of 10, 8 of 10, and 9 of 10 subjects responded similarly to the group for experiments 1, 2, and 3, respectively.

DISCUSSION

Previous studies have investigated shock attenuation by experimentally varying stride length and stride frequency (6,10) or by varying speed and allowing subjects to freely select a stride length and frequency combination (22). Because stride length and frequency were varied concurrently in these studies (6,10,22), it was not clear whether stride length or frequency had individual effects on shock attenuation. By independently manipulating stride length and frequency at different velocities, our observations indicated that stride length has a major influence on determining magnitude of shock attenuation. This conclusion was based on the observation that shock attenuation changed only when stride length changed (Table 1 and Fig. 4).

The shock attenuation patterns illustrated in Fig. 4 were similar with other studies (6,10,16,22). The change in shock attenuation between the +10% PSL/−10%PSF and −10%

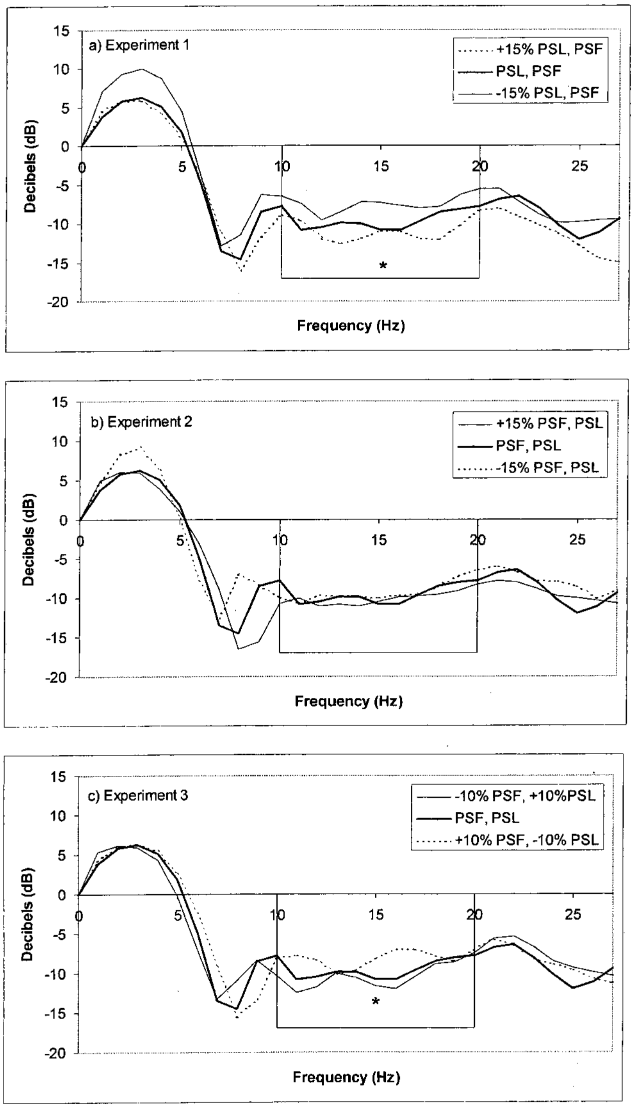

FIGURE 4—a, b, c. Group mean shock attenuation patterns while running at different stride lengths and stride frequencies. Subjects completed three experiments: 1) stride length manipulated (a); 2) stride frequency manipulated (b); 3) stride length and stride frequency manipulated concurrently (c). Shock attenuation changed only when stride length changed (experiments 1 and 3, $P < 0.05$) in the frequency range analyzed (*box*). As stride length increased, shock attenuation increased.

PSL/+10% PSF conditions in our experiment was 18%, which is comparable to the mean shock attenuation change of 12% between identical manipulations of stride length and frequency reported by Derrick and colleagues (6).

The use of surface mounted accelerometers to record head and leg segmental accelerations during running is limited by accelerometer mass and the soft tissue between the accelerometer and bone (21,23,27). Using low mass accelerometers combined with high tension and analyzing acceleration data in the frequency domain independent of resonant frequencies constitutes an acceptable approach to using surface mounted accelerometers (22,23,27). Estimations of resonant frequencies of surface mounted accelerometers have ranged from 27.3 to 40.6 Hz (9,15) to 60 to 100 Hz (22). In our study, accelerometer mass was 6.7 g, which falls within the lower range of masses used in other studies (range: 1.5–181.4 g), tension was applied to subject tolerance level, and frequency range analyzed was 10–20 Hz. The 10- to 20-Hz frequency range should capture the frequencies associated with the impact phenomenon (6,20,22). Nevertheless, shock attenuation was compared *post hoc* over several different possible frequency ranges, such as 11–20, 10–15, 11–18, and 16–20 Hz, to determine whether the chosen frequency range (i.e., 10–20 Hz) led to a unique interpretation of results. The results were independent of the frequency range analyzed. No analysis was completed above 20 Hz because minimal power existed in both PSD_{leg} and PSD_{head} profiles above this level.

Power within the frequency domain analysis is related to the time domain signal power. Stance times represent the length of the extracted data set before zero padding. Because stance times decreased with increased running speed (Table 1), it was considered that extracted data set length (i.e., the length of data set before zero padding) could have affected time domain signal power. However, time domain signal powers for either head or leg acceleration profile were not correlated with stance time (leg: $r = 0.05$, $P > 0.05$; head: $r = 0.10$, $P > 0.05$). It would be expected that changes in stance time would affect the low frequency range of the signals, but only the frequencies representing the impact phase were analyzed. Thus, it was concluded that any changes in shock attenuation were due to experimental manipulation of stride length and/or stride frequency.

The results of our three experiments were 1) concurrent changes in stride length and stride frequency affects shock attenuation, 2) stride length changes affects shock attenuation, and 3) stride frequency changes do not affect shock attenuation. These results have lead us to conclude that the hypothesis that shock attenuation would change only when stride length changed is tenable since shock attenuation changed only when stride length changed and did not change with independent changes in stride frequency.

A possible explanation for why shock attenuation changed only when stride length changed is that impact magnitude changed as stride length changed. Considering that PSD_{head} did not significantly change across conditions, a change in impact magnitude should affect shock attenuation. There is evidence that stride length changes affect impact magnitude. For example, Challis (2) reported that ground reaction force impact peak was affected by concurrent changes in stride length and frequency, whereas Mercer et al. (19) reported that impact magnitude across speeds was affected by stride length constraints. The reason for a relationship between impact magnitude and stride length may be that lower-extremity geometry changes as stride length changes. Simulation results support this hypothesis because it has been determined that impact magnitude is partly explained by lower-extremity geometry at impact (3,8,14). For example, Gerritsen and colleagues (8) reported that simulated GRF impact peaks increased by 68 N for every degree change in leg angle. The simulated impact peak was also sensitive to foot angle at impact, as evident by an 85-N increase in impact peak for every degree change in angle.

Although running kinematics were not analyzed in our study, it is possible that a particular geometry was used per stride length condition. For example, magnitude of knee flexion angle at contact and/or foot-strike pattern may have changed between stride length conditions and affected impact magnitude. Changes in shock attenuation may be related to changes in lower-extremity geometry, which may change as stride length changes.

Related to the lower-extremity geometry at impact, another possible explanation for the relationship between stride length and shock attenuation is that the amount of energy absorbed by the lower extremity is dependent on stride length. Derrick et al. (6) analyzed hip, knee, and ankle joint moments and powers during running with different stride length and frequency combinations at a set speed. It was reported that shock attenuation increased as stride length increased (and stride frequency decreased), and it was suggested that the mechanism used to accomplish attenuation of the shock wave was primarily through energy absorption about the ankle, knee, and hip. Generally, during longer stride length running conditions, the knee joint was primarily responsible for changes in overall energy absorption. The authors suggested that changes in magnitude of energy absorbed about the knee joint were partially due to the perpendicular distance (d_k) between the line of action of the resultant GRF and knee joint center of rotation. A greater d_k for a given impact force must be countered with a greater knee extensor moment to prevent collapse of the lower extremity. A greater knee extensor moment during the impact phase could result in a greater amount of energy absorbed about the knee. Derrick et al. (6) reported that d_k increased as stride length increased and suggested that lower-extremity stiffness at impact influenced shock attenuation (5). It may be that d_k is related to stride length during running at different velocities, thus explaining why shock attenuation changed only when stride length changed. This hypothesis is supported by the observations of other studies that have reported a relationship between magnitude of knee flexion at the time of impact and the impact response (4,8,16,18), which would affect shock attenuation. Knee flexion and d_k during the impact phase would be related to the lower-extremity stiffness and a more compliant lower extremity would be more effective at attenuating PSD_{leg} (5). In our study, neither knee angle nor lower-extremity compliance were recorded. Nevertheless, it would be expected that knee angle and d_k at impact were factors that contributed to lower-extremity stiffness during the impact phase for a given stride length independent of running velocity. Evidence supporting this hypothesis is that during running, lower-extremity stiffness is affected by changes in stride length (7). Therefore, it is expected that magnitude of shock attenuation is related to lower-extremity stiffness during the impact phase, which would change for different stride lengths.

We observed that running with a longer stride length relative to preferred stride length had the effect of increasing PSD_{leg} but had no affect on PSD_{head}. The change in PSD_{leg} was 4 times greater when stride length was varied compared with changes in PSD_{leg} when stride frequency was varied. Furthermore, the frequency at which peak power was observed in PSD_{leg} within the 10- to 20-Hz range seemed to shift between conditions. However, despite the changes in PSD_{leg}, the body responded to different levels of leg impact by absorbing sufficient energy that resulted in a narrow range of magnitudes of head shock and shock attenuation changed only when stride length changed. The observation that PSD_{head} remained within a narrow range of magnitude has also been observed during running with different stride lengths (6,10) and at different velocities (22). These observations support the hypothesis that the body acts as a low-pass filter, functioning to attenuate impact frequencies regardless of the impact magnitude (5,6,10,22).

Although impact energy can be absorbed through passive structures such as bone, cartilage, and tendon as well as shoes and running surfaces, a great potential source of energy absorption is through muscle contraction. Our results indicate that shortening stride length reduces impact energy during running, suggesting a lower demand for shock attenuation. However, the shorter stride length would also have the effect of increasing the number of impacts over a given running distance due to a greater stride frequency if velocity is maintained. In contrast, if stride length were increased, the number of impacts during running a given distance (velocity maintained) would decrease but impact energy would increase. Increasing stride length would seem to have the effect of increasing the energy-absorbing role of muscle. Given this understanding of the relationship between stride length and shock attenuation, there may be some benefit to prescribing changes in running style (i.e., stride length and frequency) for certain runners. However, the choice of specific stride length and frequency combination a runner selects is based on many factors. For example, Hamill et al. (10) showed that runners optimize running style on oxygen consumption not shock attenuation. Small changes in stride length, however, may not have a detrimental effect on oxygen consumption (1,10) but may have a substantial effect on shock attenuation.

In conclusion, this study was designed to better understand the effect of changes in gait on shock attenuation. Through systematic variations of stride length and stride frequency, it was concluded that shock attenuation was affected by changes in stride length but not stride frequency. Furthermore, despite PSD_{leg} changes, PSD_{head} was maintained within a narrow range of magnitude. A possible explanation for the relationship between stride length and shock attenuation is that shock attenuation is related to lower-extremity geometry and compliance at impact and these factors are directly related to stride length.

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