Footwear Affects the Behavior of Low Back Muscles When Jogging

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Use of modified shoes and insole materials has been widely advocated to treat low back symptoms from running impacts, although considerable uncertainty remains regarding the effects of these devices on the rate of shock transmission to the spine. This study investigated the effects of shoes and insole materials on a) the rate of shock transmission to the spine, b) the temporal response of spinal musculature to impact loading, and c) the time interval between peak lumbar acceleration and peak lumbar muscle response. It was hypothesised that shoes and inserts a) decrease the rate of shock transmission, b) decrease the low back muscle response time, and c) shorten the time interval between peak lumbar acceleration and peak lumbar muscle response. Twelve healthy subjects were tested while jogging barefoot (unshod) or wearing identical athletic shoes (shod). Either no material, semi-rigid (34 Shore A), or soft (9.5 Shore A) insole material covered the force plate in the barefoot conditions and was placed as insole when running shod. Ground reaction forces, acceleration at the third lumbar level, and erector spinae myoelectric activity were recorded simultaneously. The rate of shock transmission to the spine was greater ($p < 0.0003$) unshod (acceleration rate: Means ± SD 127.35 ± 87.23 g/s) than shod (49.84 ± 33.98 g/s). The temporal response of spinal musculature following heel strike was significantly shorter ($p < 0.023$) unshod (0.038 ± 0.021 s) than shod (0.047 ± 0.036 s). The latency between acceleration peak (maximal external force) and muscle response peak (maximal internal force) was significantly ($p < 0.021$) longer unshod (0.0137 ± 0.022 s) than shod (0.004 ± 0.040 s). These results suggest that one of the benefits of running shoes and insoles is improved temporal synchronization between potentially destabilizing external forces and stabilizing internal forces around the lumbar spine.

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Introduction

The loading rate (the load amplitude divided by the time to reach the maximal amplitude) has been shown to be an important physical factor influencing injury to the musculoskeletal system. Repetitive, rapidly applied impulsive loading has been shown to produce joint degeneration, whereas slowly applied loads of equal or even greater magnitude often have no deleterious effects [15–17]. With every step, an impulsive shock wave is generated at heel strike that is transmitted from the lower extremities through the spine [7]. To avoid the jarring and potentially damaging effects of this shock wave, the human body has evolved complex mechanisms for dampening the shock wave.

The effectiveness of shock absorbing shoes on dampening this shock wave during gait has been discussed controversially [3–5,7,9,11,12]. Although there is a general agreement that insoles or shoe modifications may lower impact loading at heel strike [3,4,9,12], there is some concern that insoles may lower the shock absorbing behavior of the body at the same time [8,18,20].

Despite this controversy, the use of shock absorbing insoles has been successfully used to treat low back patients [26], and it has been shown that shock absorbers lower the shock wave at low back level [25].

On the other hand, an active shock absorbing behavior of the human body has been described that might be diminished by soft inlays [8,18,20]. In fact, it has been shown that the development of external forces [11], as well as the transmissibility of impact forces through the human body [7], are increased by wearing soft soles. So far, the mechanism of active shock absorption is unclear.

However, there is some evidence that the neurophysiologic system is involved [8,18–21], and it has been shown that gluteal muscle activation in walking is related to proprioception at the level of the sole [2]. It has also been shown that a reduced shock absorbing capacity of the human musculoskeletal system from the femoral condyle to the forehead correlates with the presence of low back pain [25]. It is interesting to note that in this study the low back patients had no X-ray findings. Thus, there is no evidence that a damaged disc or joint is the reason for the decreased shock absorbing capacity, but there might be an altered behavior of the active part of the musculosceletal system. In fact, it has been shown that the lumbar muscles of low back sufferers react slower to sudden loading than the spinal muscles of healthy individuals [1]. Nevertheless, the role of the spinal muscles in energy dissipation is as yet unclear.

With each step internal muscle forces at the low back level are generated to achieve equilibrium and stabilize the lumbar spine. Proper function of this system seems crucial, because external forces which act earlier than the human control system is able to respond might injure the spine. Muscle activity dissipates energy since the contracted muscle is a viscoelastic element. The standing impact studies of Pope et al. [14] clearly show the attenuation in the bent knee stance. Sudden loads have been noted in many epiodemiologic studies to be associated with reports of acute low back pain. If the load is applied to a spinal motion segment, then in the absence of muscular stability the displacement will exceed the range of the neutral zone and soft tissue structures (e.g. interspinous ligaments and intervertebral disc) will be loaded.

The purpose of this study was to explore the influence of shoes and associated materials on shock wave transmission and, simultaneously, the motor response in the lower back to heel strike impact.

Based on the positive experience in low back patients with shock absorbers [26] it was hypothesised that shoes would demonstrate more protective effects than bare feet (a shoe main effect), that soft materials would demonstrate more protective effects than hard materials or no materials (a material main effect) and that shoes with a soft insole would be more protective than other combinations of shoes and materials (a shoe by material interaction), where protective effects were defined as:
1. Decreases in the loading rate experienced at the spine.
2. Decrease of the response time of the spinal muscles to heel strike (occurrence time relative to heel strike).
3. Reduction of the time interval between peak lumbar acceleration and peak lumbar muscle response.

Materials and Methods

Twelve generally healthy volunteers were recruited by local advertising. Five were female and seven male. The mean age was 32.9 years with a standard deviation of 7.9 years. Ages ranged from 21 to 48. All subjects signed an informed consent statement that had been approved by the institution's review board.

The experiment was a 2wShoe × 3wMaterial × 3wRepetition factorial experiment where each subject ran at a self-paced slow jogging speed on a laboratory runway of 8 m length. As a completely within subjects design, each subject was exposed to all levels of each factor in the design (jogging both with shoes and barefoot [2wShoe], with either no material, a hard material or a soft material placed on the force plate or in the shoe as an insole [3wMaterial] and repeating each of these six combinations three times [3wRepetitions]). Each repetition was considered valid when the following criteria were met: 1) no change in the jogging style, 2) the right heel contacted the force plate (Type 4060A, Bertec Corporation, Worthington, OH, USA), and 3) the velocity was constant (+/- 15 percent).

Since a fundamental assumption in repeated measures designs is the independence of events across trials, a preliminary study was done to evaluate the independence of responses across trials. Once the three criteria for validity of repetitions were included, trial to trial variations in jogging speed and mechanics (acceleration amplitude and acceleration rate at low back level) were minimal. Further, the low demand and simple nature of the task made learning or fatigue effects extremely unlikely.

When subjects arrived for their scheduled participation, they were fitted with the correct size of New Balance 600 running shoe (New Balance Athletic Shoe, Inc. Boston, MA, USA), and had the EMG electrodes and accelerometer devices fitted and tested. Subjects were then informed about what was expected of them in terms of trying to run in a natural and consistent form across all repetitions of the short jogging distance. Because the pilot efforts suggested little learning or fatigue effects associated with the protocol, the ordering of the trials was fixed to minimize subject efforts in putting on and taking off shoes. Subject's first nine trials were barefoot (unshod) and their last nine trials were with shoes (shod). The nine unshod trials varied the material on the force plate from none to hard to soft with the material placed on the surface of the force plate. The nine shod trials varied the material insoles from none to hard to soft, with the materials used as shoe insoles. Greater detail concerning the experimental design is provided in Table **1**.

Ground reaction forces, acceleration at L3 level and erector spinae muscle activity were simultaneously monitored during jogging. Ground reaction forces were measured by a force plate. Acceleration was recorded by a single-axis, lightweight (0.4 g) accelerometer (Isoton™, PE Accelerometer Model 2250A-10, Endevco, San Juan Capistrano, CA, USA) attached to the skin at the L3 spinal process with double-sided adhesive tape [22,27]. Low back muscle activity was recorded 3 cm lateral to the midline at L3 level by surface bipolar EMG elec-

Table 1 Experimental design

Shoe Condition	Trials	Material	Combination of materials with shoe or force plate
Unshod	1 – 3	None	Uncovered force plate.
Unshod	4 – 6	Hard	Force plate covered with 10 mm thick, semi-rigid (35 Shore A) shock absorbing material[1].
Unshod	7 – 9	Soft	Force plate covered with 10 mm thick, soft (9.5 Shore A) shock absorbing material.
Shod	10 – 12	None	Shoe (80 Shore A) without insole.
Shod	13 – 15	Hard	A customized, 10 mm thick insole from semi-rigid shock absorbing material (PE Lite®).
Shod	16 – 18	Soft	A customized, 10 mm thick insole from soft shock absorbing material (PPT)[2].

[1] PE Lite®, Medium Density, Knite-Ride Inc., Kansas City, MO, USA
[2] PPT, Blue, Langer Biomechanics Group Inc., Deer Park, NY, USA

trodes with buildup preamplifiers to reduce the artifacts fixed at the right side. The same location of the bipolar surface electrodes had already been proven to be the best [6,13].

Data collection for all three systems (EMG, force plate, and accelerometer readings) was triggered synchronously by a switch, built into a ground based platform, located in the runway ahead of the force plate. To assess the rate of shock transmission to the spine, acceleration rate was calculated as acceleration amplitude divided by the time to acceleration peak. The time to acceleration peak was defined as acceleration duration (latency between acceleration onset and acceleration peak) (see Fig. **1**). To assess the low back muscle response time to heel strike impact, the latency between heel strike and muscle response onset (L1) was determined (see Fig. **1**). To assess the latency between maximal external and maximal internal force at low back level, the latency between acceleration peak and muscle response peak (L2) was calculated (see Fig. **1**). The erector spinae muscle response following heel strike was analyzed by using inspection of digitally-magnified raw EMG signals with resolution of 1 ms (Origin 3.5, Microcall Software Inc., Northampton, MA, USA), because all available time domain processing methods (average, integration, RMS, etc.) lead to an error equal to at least the length of its time constant. Onset of muscle response was defined as the first increase in the EMG activity following touchdown. Since there is also muscle activity due to gait, only an increase of at least twice the magnitude of back ground activity after touch down was considered as muscle response. To avoid a bias, the data were mixed in a random order by one of the investigators and analyzed by another one.

A graphic representation of these measurements is summarized in Fig. **1**.

A preliminary 3-way ANOVA was performed to evaluate the effects of the experimental conditions on jogging speed as an initial screen for order effects. To control overall experimentwise error rate and to evaluate the overall effects of the shoe, materials and repeated assessments on spine related outcomes two 3-way MANOVAs were done to simultaneously evaluate the acceleration (amplitude and duration) and (muscle onset latency [L1] and latency from acceleration peak to muscle maximum response [L2]) criteria. A critical p-value of 0.10 was used to identify significance for the overall MANOVA procedures. Follow-up univariate 3-way ANOVAs for the individual outcomes, and for the acceleration rate criteria were planned for any of the significant MANOVA results, and post hoc multiple comparisons were performed using Tukey's highest significant difference (HSD) follow-up procedures. Significant interactions among the experimental factors in the ANOVA procedures were followed up using tests of simple effects under the assumption that all of the factors in the models represented fixed effects. Critical p-values of 0.05 were used to identify significant results from follow-up univariate ANOVAs and follow-up tests of main and simple effects.

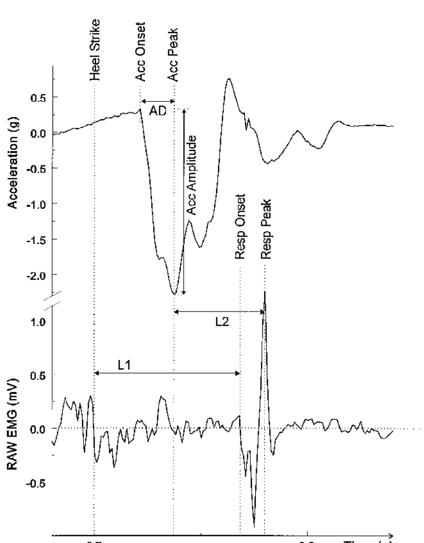

Fig. 1 Accelerometer and EMG signal from one trial, recorded simultaneously with the force plate measurement. Heel strike was determined by the force plate. Acc = acceleration, Resp = erector spine muscle response, AD = acceleration duration, L1 = latency between heel strike and muscle response onset, L2 = latency between acceleration peak and muscle response peak.

Results

Regarding the shock transmission to the spine, the overall 3-way MANOVA demonstrated significant shoe and material effects by Wilks' Lambda, $F_{2,10} = 57.03$, $p < 0.0001$ and $F_{4,42} = 6.69$, $p < 0.0003$, respectively. Based on the 3-way ANOVA follow-ups, acceleration amplitude demonstrated no significant shoe ($p = 0.18$), material ($p = 0.35$), or repetition ($p = 0.42$) main effects, nor any interaction effects involving these factors. However, acceleration duration demonstrated a significant shoe main effect ($p < 0.0001$), with shorter durations in the unshod (0.021 ± 0.009 s) compared to the shod conditions (0.039 ± 0.011 s). A significant material main effect ($p < 0.0006$), with shorter durations in the no material condition (0.027 ± 0.014 s) compared to the hard and soft material conditions (0.030 ± 0.013 s and 0.033 ± 0.014 s, respectively) was found. The materials themselves were not significantly different from each other based on Tukey's HSD follow-ups of the significant materials main effect. The pattern of results for acceleration rate were consistent with those observed for acceleration duration. There was a significant shoe main effect ($p < 0.0003$), where acceleration rate was significantly higher unshod (127.35 ± 87.23 g/s) compared to shod (49.84 ± 33.98 g/s). The significant materials effect for acceleration rate ($p < 0.02$) again demonstrated a similar grouping pattern. Significantly higher deceleration rates in the no material conditions (106.79 ± 91.42 g/s) compared to the hard and soft material conditions (78.07 ± 61.56 g/s and 80.27 ± 71.09 g/s, respectively) were found, which were not themselves significantly different from each other based on Tukey's HSD follow-ups of the significant materials main effect. Acceleration rate measured in the different material conditions, barefoot and shod, are presented in Fig. **2**.

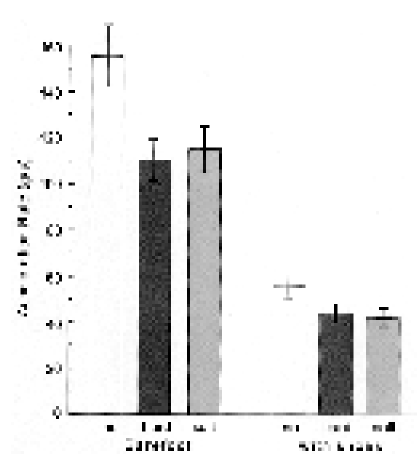

Fig. 2 Acceleration rate at low back level following heel strike in shod and unshod conditions. No material (no), hard shock absorbing material (hard), and soft shock absorbing material (soft) was placed on the ground and in shoes, respectively. Standard error bars are indicated.

The muscle onset latency (L1) and the latency between acceleration peak and muscle peak response (L2) were considered simultaneously within a 3-way MANOVA, which simultaneously considered the effects of shoe, materials, repetitions and their interactions. Results revealed a significant shoe main effect by Wilks' Lambda, $F_{5,7} = 10.81$, $p < 0.0034$. The materials and repetition main effects were non-significant, nor were any of the interaction effects significant.

Delay in the temporal response of the spine musculature to heel strike (L1) showed a significant shoe main effect ($p < 0.023$.) The low back muscle response onset following heel strike occurred significantly earlier in the unshod (0.038 ± 0.021 s) than shod condition (0.047 ± 0.036 s). There was a trend toward a material main effect ($p < 0.064$) with the pattern of means in the expected order with fastest onset of muscle response with no material (0.036 ± 0.021 s), a slower response with hard material (0.041 ± 0.020 s), and the slowest response with soft material (0.050 ± 0.042 s) as shown in Fig. **3**.

The latency between peak acceleration and peak muscle response at the lower back showed a significant shoe main effect ($p < 0.021$) with the pattern of means demonstrating significantly shorter latencies in the shod conditions (0.0040 ± 0.040 s) compared with the unshod conditions (0.0137 ± 0.022 s) (Fig. **4**). No significant material ($p = 0.62$) or repetition ($p = 0.39$) main effects or interaction effects were observed.

The average jogging velocity was 1.49 m/s (range, 1.45 m/s to 1.51 m/s). Jogging velocity was not significantly related to shoe, material, or repetition.

Discussion

This study was conducted to address the hypotheses that wearing shoes and insert materials 1) decreases the rate of shock transmission to the lower back, 2) decreases the response time of the spinal muscles to heel strike, and 3) reduces the time interval between acceleration peak and muscle response peak at the lower back in jogging. The results support hypothesis 1 and 3. Hypothesis 2 was not supported. On the contrary, the muscle response was significantly later in running shod than in running barefoot and, furthermore, increased with increasing softness of the sole material.

The study hypotheses were based on the positive experience with shock absorbing materials in low back patients [15]. From this report it was assumed that shock absorbers shorten the latency between external (passive shock, potentially destructive) and internal (active muscular, protective) forces experienced at the low back. The observed neuromuscular delay caused by shoes and insert materials suggests, on the first view, that this assumption was wrong. However, at the same time, the latency between heel strike and acceleration peak at low back level increased by wearing shoes, due to an increased time interval between acceleration onset and acceleration peak. Thus, with shoes, the lower back experienced the impact force peak later. In fact, this mechanical delay predominates the muscle response delay so that, after all, wearing shoes decreases the time interval between maximum external and maximum internal forces experienced in the lower back during running. Since the shock wave tends to destabilize equilibrium and the muscle response may help regain equilibrium, it seems that shortening of this time interval may have clear benefits.

The results raised the question whether the proprioception afferent from the heel, or low back muscle stretch reflex triggers the neuromuscular shock-absorbing behavior at lumbar level. The muscle responses occurred with a mean latency between 36 and 50 ms after touchdown. This is long enough for the monosynaptic stretch reflex (M1), which takes 30 to 50 ms [8,23]. The onset of acceleration at the low back occurred about 18 ms after touchdown, and the latency between acceleration onset and muscle response onset was under 20 ms in the barefoot situation, which is too short even for an M1 reflex. Based on these temporal data it appears that the lumbar muscle activity was not triggered by a local shock wave at the lumbar level, but by proprioception at the heel level. The proprioceptors appear sensitive enough to discriminate differences of the magnitude of the standard deviation of the impact duration (about 0.01 s). This indicates that heel afferent proprioception could be sensitive to heel loading rate.

Perhaps the single biggest limitation of this study is the lack of multiple shoe conditions, which kept this study from considering the possible differential effects of various shoe designs relative to various introductions of insole materials. The inability

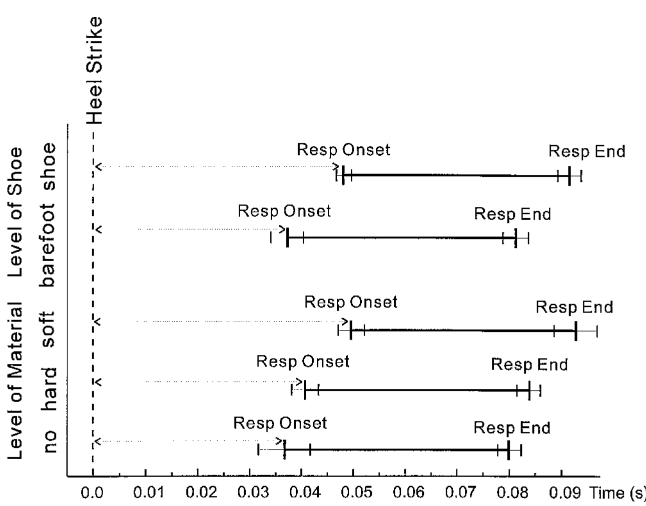

Fig. 3 Latency between heel strike and erector spinae muscle response onset (L1). Resp Onset = muscle response onset, Resp End = muscle response end. Standard error bars are indicated.

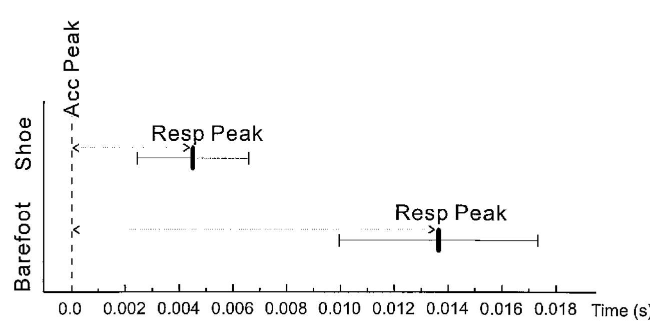

Fig. 4 Latency between acceleration peak (Acc Peak) and muscle response peak (Resp Peak) following heel strike in shod and unshod condition (L2). Standard error bars are indicated.

of the current study to demonstrate the anticipated interaction between the shoe (shod, unshod) and material (none, hard, soft) factors was, on first consideration, a surprise. The most obvious explanation, lack of power owing to the relatively small sample size of twelve subjects, although plausable and almost obligatory, was unsatisfying as the pattern of results across the material conditions in the shod and unshod conditions did not hint toward any anticipated effects. Similarly, confounding effects due to lack of randomization of the ordering of the shoe/material combinations was not judged a likely explanation for either the observed shoe and material main effects or for the lack of the anticipated shoe x material interaction effects. The jogging task was simply too similarly executed across trials and shoe/material combinations to suggest either learning or fatigue effects.

Another limitation is the fact that it is extremely difficult to measure the force that acts *in vivo* on the lumbar spine directly. However, accelerometers have been proven useful to estimate impact loading. To reduce the bias in the acceleration outcomes, we followed the fundamental requirements to properly measure bone vibration *in vivo* by skin attached accelerometers: a thin layer of soft tissue between accelerometer and bone [22], use of a light weight accelerometer [10], and a tight attachment to the skin [22,27].

Even if there is latency difference in the other mutually orthogonal directions at the lumbar level, it would not significantly influence the results and conclusions of the study, because of the too brief latency of the axial component of the acceleration. The few significant differences in external forces between the hard and the soft insole materials should also be interpreted with some caution. In this study we investigated two of the most commonly used inshoe orthotic materials - PeLite and PPT. These materials are US orthotic industry standards, and represent the range of softness-hardness that can be used comfortably without adversely affecting gait. It might be expected that testing harder and softer insole material than these we used might provide a more robust test of the hypotheses. In this initial study, however, scientifically well characterized materials were used for the sake of the clinical significance, practical application and future reproduction of the study design. Robbins and Gouw [19], in a review of the literature and their own results on athletic footwear and chronic overloading, summarized that soft shoes with thick yielding midsoles are probably dangerous because they attenuate tactile plantar sensations required for protective impact moderating behavior. They developed the theory of a plantar surface, sensory-mediated feedback system for neuromuscular control of the shock absorbing behavior [18,20,21]. Their theory has been supported by the experimental observation that stereotypic ipsilateral hip flexion and contralateral hip extension following a rapid, heavy loading of the leg increases in amplitude as the irregularity of the plantar surface support increases [21]. The low back muscle response to running was dramatically affected by use of shoes or soft insole shoeing materials. The latency from heel strike to muscle response onset was prolonged. This is consistent with the observation of Robbins and Gouw who found that the active damping of the external force energy started later and progressed less efficiently with use of shoes or soft insole materials [18,20]. Thus, too soft shoes might be worse, especially for low back patients since their ability to react to sudden load is diminished anyway [1].

Our results help to explain other observations of shock absorbing behavior [7,24]. Forner et al. [7] recently examined the properties of shoe insert materials as they effect shock wave transmission between tibia and forehead. They studied the difference between materials with lower rigidity and loss tangent (low energy absorbing) and higher rigidity with high loss tangent. They found more transmission of acceleration from the tibia to the forehead with the least rigid material. Our study suggests that this decrease in shock absorbing behavior is due to an increased latency of spinal muscle response when wearing very soft shoes.

In summary, we found that shoes and insert materials not only reduce the loading rate, but affect the low back muscles. They can protect the lower spine from heel strike impact in two ways: by reducing the impact loading rate and by minimizing the latency between maximum external and internal force.

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