may simply be that the diet was different from the diet normally consumed by the subjects. The cyclists’ normal diet contained a mean of 264 g·d⁻¹ of carbohydrate, while the MCHO diet contained a mean of 258 g·d⁻¹. Thus, the MCHO diet was most like the subjects’ normal diet, whereas the LCHO diet was vastly different. Further support for this concept can be based on the fact that two subjects verbally reported that they did not like the HCHO diet. These two subjects also scored higher in tension, depression, and anger and lower in vigor while on the HCHO diet. The other subjects generally had similar mood scores while on the MCHO and HCHO diets. It may be that in some cases a diet that deviates from a subject’s “normal” diet may be perceived somewhat adversely. Support for this concept can be found with the work of Rosen et al. (9). These investigators reported that obese females placed on carbohydrate-containing and carbohydrate-restricted hypocaloric diets exhibited a tendency toward dysphoric moods and attitudes during the 1st wk of the diet, but after 6 wk on the diet the subjects’ moods and attitude were not different from baseline, predicted values. Thus, some adaptation to the different diets occurred over time as the subjects became more accustomed to the diets or the diets became more “normal.” In summary, the present study found that the consumption of a low-carbohydrate (13% kcal), high-protein, and high-fat diet caused significant changes in the mood state of female cyclists undergoing a training and exercise program. These changes were generally of an adverse nature and could be considered detrimental to training and performance. These changes in mood were improved with the addition of more dietary carbohydrate (54% kcal). Higher-carbohydrate diets (72% kcal) caused no further changes in mood as compared with the moderate-carbohydrate diet.

REFERENCE


Athletic footwear: unsafe due to perceptual illusions

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ABSTRACT

ROBBINS, S. E. and G. J. GOUW. Athletic footwear: unsafe due to perceptual illusions. Med. Sci. Sports Exerc., Vol. 23, No. 2, pp. 217-224, 1991. Modern athletic footwear provides remarkable plantar comfort when walking, running, or jumping. However, when injurious plantar loads act negligible perceived plantar discomfort, a perceptual illusion is created whereby perceived impact is lower than actual impact, which results in inadequate impact-modulating behavior and consequent injury. The objective of this study was to examine how plantar tactile (mechanical) events affect perceived plantar discomfort. Also, we evaluated the feasibility of a footwear safety standard we propose, which requires elimination of the above illusion. Twenty subjects gave numerical estimates of plantar discomfort produced by simulated locomotion (constant vertical (0.1-0.7 g·cm⁻²) and horizontal (0.1-0.9 g·cm⁻²) plantar loads), with the foot supported by either a smooth rigid surface or a rigid surface with 2 mm high rigid irregularities. Vertical or horizontal load alone evoked no discomfort (P > 0.05), whereas together, discomfort emanated from loads as low as 0.4 kg·cm⁻². Irregularities heightened discomfort by a factor of 1.89. This suggests that the proposed safety standard is feasible, since compliance could be achieved simply by adding surface irregularities to insoles and by other changes that heighten localized plantar load. However, until this standard is adhered to, it might be more appropriate to classify athletic footwear as "safety hazards" rather than "protective devices.

ATHLETIC INJURIES, PROTECTIVE DEVICES, INJURY PREVENTION, SHOCK ABSORPTION

Impact (shock, shock loading) is defined as "a collision between two bodies, which occurs in a very small interval of time, during which the two bodies exert on each other relatively large forces" (2). Acute overloading is injury following a single loading, e.g., falling from a high place. Chronic overloading is injury following a multitude of single loads applied over a period of time, each of which is incapable of causing acute overloading, e.g., running related injuries (47). During locomotion (walking, running) or activities where people repetitively jump (e.g., aerobics, gymnastics), the plantar surface (sole of foot) sustains repeated impact consisting of large rapidly applied vertical and horizontal plantar loads (1.1,11,43). The vertical component of plantar impact results in propagation of shock waves (20,31,38,52,62,63) that produce chronic overloading of bone and connective tissue in various mammals (18,31,42-45,52,53), and data suggest that it is equally destructive to humans (31,10,20,21,23,25-28,31,34,46,56-59,61).

The high incidence of chronic overloading during locomotion suggested to footwear designers that the lower extremity is fragile. Accordingly, over the past 15 yr athletic footwear has been designed to shield the lower extremity from damage, as is delicate merchandise when injured. However, the use of expensive (compliant, soft) packaging materials. The recent model have the most packaging, hence the greatest compliance and comfort, which follows compliance (12,13,17,37). This footwear has been something less than successful protective devices. A comparison of earlier epidemiology-studies dealing with running-related injury incidence with recent reports suggests that there is presently a higher incidence of these injuries (e.g., Martin et al. (33), training injuries in year prior to running event: males 46%, females 40%; Caspersen et al. (10), training injuries in year prior to running event: males 35%, females 35%). Wearing of expensive running shoes that are promoted as having additional features that protect (e.g., more cushioning, "pronation correction") are injury significantly more frequently than runners employing inexpensive shoes (costing less than US $40), with no major manufacturer superior to others with respect to injury incidence (32). Moreover, runners who have a footwear brand preference are more likely (P < 0.05) to be injured than those who have brand loyalty (37). The increased injury incidence with modern running shoes can be attributed to greater impact when runners use footwear more of the current design, when compared with footwear in use a decade earlier (37). Furthermore, when runners unaccustomed to barefoot running run barefoot, mean impact is no higher than when shod and in some cases is lower (13,15,19,27,30,54). In addition, in barefoot populations running-related injuries are rare, which indicates that
humans adapted to barefoot running run with lower impact than the unadapted group referred to above (49). This also suggests that the lower extremity is inherently durable and is made susceptible to injury by footwear use (7,57,58). Based on the above data, notwithstanding unsupported claims by footwear manufacturers of improved protection with their products, it seems appropriate to consider expensive athletic footwear from major manufacturers (and perhaps less expensive shoes) as unsafe.

Our initial hypothesis (49), which attempted to explain the inability of athletic footwear to protect and the freedom from injury when barefoot-adapted, has progressed with the addition of recent data (47,48,50,51) (Fig. 1). Our present hypothesis is as follows:

In humans, avoidance of uncomfortable or painful but locally innocuous plantar cutaneous tactile stimuli moderates shock on subsequent impacts when humans walk, run, or jump repetitively. This feedback control circuit is optimized in terms of protection for mechanical interaction of the bare foot and natural surfaces. Eventually, learning allows anticipatory avoidance. Modern athletic footwear is unsafe because it attenuates plantar sensations that induce the behavior required to prevent injury. (Activity is behavior that moderates stimulus intensity or evades the stimulus entirely. Natural surfaces refer to naturally deposited ground, i.e., irregular surfaces.)

What support does this hypothesis have? It explains the difference in injury incidence between barefoot and shoe runners (26,33,49) via the requirement of plantar discomfort on impact for optimized shock absorption. This is strengthened by reports indicating that, when habitually barefoot humans walk (and probably when they run), they have greater knee flexion, which has been shown to reduce shock (35), compared with shod subjects. In addition, when running, the longitudinal foot arches deflect from highly arched to flat with each gait cycle, which likely has shock absorbing properties (6, 8).

The theory explains why material tests fail to predict actual impact when running (13,15,19,20,34). The more compliant shoe, which according to material tests should attenuate shock more effectively, fails to do this because it produces greater plantar comfort (29), hence less impact-modulating behavior.

The linkage between plantar perceptual processes and impact-modulating behavior is also clear. When the plantar surface is rapidly and heavily loaded to simulate vertical loading during running, avoidance by hip flexion increases in relation to surface characteristics producing discomfort, such as irregularity (50). Furthermore, we demonstrate that both activity (irregular, rigid surfaces, heightened plantar discomfort) was more effective than barefoot activity indoors (regular, compliant surfaces) in inducing raising of the medial longitudinal arch of the foot. This adaptation can be explained by local differences in tactile sensibility along the plantar surface (48).

Moreover, in a psychophysical study we found that, when subjects wear more athletic footwear and the plantar surface is loaded to simulate the impact of locomotion, a perceptual illusion is produced whereby perceived plantar impact is less than actual impact (51). An illusion is defined as “something that deceives or deludes by producing a false impression” (Oxford English Dictionary). We refer to this as the “discomfort-impact illusion”. When the plantar surface is similarly loaded but supported by a simulated natural surface (compacted gravel, irregular), it produces plantar discomfort load, load estimates are accurate; hence, this discomfort-impact illusion is eliminated.

In a recent report (36) relating to this illusion, impact was measured when 15 well-trained gymnasts walked off a platform 0.69 m high and landed on either yielding mats or a hard surface. With every subject, impact when landing on the hard surface was lower than on the yielding mats. The yielding surface was accounted for by “the landing strategy chosen by the gymnasts...” Actual impact measured was contrary to the subjects’ impression; hence, a perceptual illusion was produced by the yield-

**METHODS**

**Apparatus.** Similar to apparatus used in a previous report (50) (Fig. 2), the equipment used in this experiment was adjustable so that, when the subject was seated, the knee was flexed at 90 degrees. Impact was delivered by pneumatic actuators: vertical impact to the thigh near the knee and horizontal impact to the foot near the Achilles tendon attachment. The thigh and foot were conjoined to their respective impact application plates by several layers of elastic crepe bandages. With this attachment, when the loads were removed, pressure induced plantar cutaneous sensations ceased as the thigh was lifted, and the foot was passively moved so as to be repositioned for reaplication of impact. Uncomfortable sensations from the thigh and Achilles attachment were minimized by interfaces composed of elastomeric material.

A programmable controller and electronic air pressure regulators allowed vertical and horizontal impact to be selectable through a keypad. The retracted position of each pneumatic actuator was adjustable so as to allow positioning of the foot on the plantar contact surface in a geometry whereby plantar load was equalized with respect to the foot’s medial-lateral and anterior-posterior arcs. The travel of the actuators was maintained at 6 cm vertically and 12 cm horizontally. Vertical impact was programmed to reach 0.4 kg cm$^{-2}$ prior to application of the horizontal component (Fig. 3). This was found in pilot studies to optimize the rate of loading (loading was complete in 1 s) while preventing foot horizontal movement. The apparatus and testing procedure permitted seven impacts per minute. The left leg was used.

The foot was free to move across the plantar contact surface until limited by actuator travel; thus, when movement occurred, a steady state was not reached whereby horizontal impact programmed was actually delivered to the plantar surface.

**Subjects.** The 20 subjects were a sample of male volunteers from a university population (height range 163–188 cm (mean 176.8 ± 7.5); age range 20–28 yr
The relation between vertical impact and the perceived magnitude of these loads is linear to an amplitude of 2.0 kg·cm⁻², whereas the slope increases probably due to the onset of plantar pain, but only when the foot impact is an extremity of an irregular surface (51). Data recorded included:

1. Normalized impact (kg·cm⁻²).
2. Subjects' estimate of perceived discomfort (ordinal scale 0–100).
3. Whether the foot moved across the plantar contact surface when impact was applied (movement).

Plantar contact surfaces used:
1. Smooth rigid acrylic plastic—the foot support surface of the apparatus.
2. Smooth rigid ultra-high molecular weight polyethylene (molecular weight 5–6 million) with 2% silicone (UHMWP)—a custom product supplied by Solidor Canada Co. (Montreal, Canada).
3. Textured surface—poured molded urethane rubber compound (Devcon Flexane Liquid 94) featuring rigid irregularities (2 mm diameter, 2 mm height, spherical end) directed at the plantar surface in an offset pattern, at the density of 1.4 cm⁻².

Testing procedure:
1. Instructions to subjects: Subjects were told that the purpose of the experiment was to provide estimates of discomfort that would be produced by plantar impact. It was explained that the magnitude of impact reaches its maximum just prior to the removal of the loads, so that their estimates were to be based on the sensations experienced at that time.
2. Setting upper limits of rating scales for each plantar contact surface: The maximum impact (0.7 kg·cm⁻² vertical; 0.9 kg·cm⁻² horizontal) was applied once for each surface. Subjects were asked to select the surface that produced the greatest discomfort. Discomfort produced by the maximum impact against the most uncomfortable surface was assigned the discomfort rating of 100. The maximum impact was then reapplied with the most uncomfortable surface followed by the maximum impact with one of the two remaining surfaces. Subjects were asked to estimate this discomfort relative to the most uncomfortable surface. The upper limit for the third surface was similarly estimated. The maximum impact was then reapplied twice for each surface, and subjects were allowed to readjust their estimates.
3. Random series of load pairings: Subjects were given a chart which displayed three ranges. The upper limit was the estimate of maximum discomfort for each surface; the lower limit was zero. Subjects were instructed to estimate discomfort produced by impacts that would follow from within the specified ranges. Maximum and minimum impacts were given twice, followed by a series of 15 impacts within the range in random order. This was repeated three times, with impacts in the same range repeated in random order for the three surfaces chosen in random order.

Data analysis. Data were sorted by subject, plantar contact surface, impact components, and movement and were evaluated by analysis of variance. Least squares linear regressions were obtained for groups that contained greater than three readings at a minimum of two levels of applied load. Slopes were grouped by plantar contact surface, applied loads, and movement and were evaluated by analysis of variance. Hypotheses were tested using post hoc t-tests.

RESULTS

Figure 4 relates plantar discomfort to vertical impact, horizontal impact, surface texture, friction, and movement.

Discomfort as a function of vertical impact. When vertical impact was below 0.4 kg·cm⁻² (groups 1–6), no relation was present between discomfort and horizontal impact (min. slope -1.79; max. slope 3.12; mean slope 0.41). When vertical impact was above 0.4 kg·cm⁻² (groups 7–18), there was a significant relation between these variables (min. slope 1.59; max. slope 13.63; P < 0.05). In groups 13–18, a change in the vertical impact from 0.4 to 0.7 kg·cm⁻² increased discomfort by a mean factor of 2.26 (min. 1.18; max. 3.43).

Discomfort as a function of horizontal impact. Similarly to the above, when horizontal impact was below 0.4 kg·cm⁻², there was no relation between discomfort and vertical impact, whereas, when horizontal impact was at or above 0.4 kg·cm⁻², a significant relation was present between these variables (groups 13–18; min. slope 8.47; max. slope 56.49; mean slope 23.00; P < 0.05).

Discomfort as a function of plantar contact surface texture. When vertical impact was below 0.4 kg·cm⁻² (groups 1–6), there was no significant difference in the discomfort produced from the foot impacting the irregular or smooth surfaces (mean slope 56.49; P = 0.05). When vertical impact was 0.4 kg·cm⁻² or greater (groups 13–18), a significant difference was present between these variables (mean slope comfort vs horizontal impact: plastic 0.86; irregular 0.73; UHMWP -0.36; P < 0.05), whereas, when vertical impact was 0.4 kg·cm⁻² or greater (groups 13–18), a significant difference was present between these variables (mean slope comfort vs horizontal impact: plastic 19.10; irregular 36.49; UHMWP 19.51; P < 0.05). When considering the relation between discomfort and horizontal impact, the smooth surfaces (plastic and UHMWP) differed significantly from the irregular surface but not from each other (plastic mean slope 19.02; UHMWP mean slope 13.42; irregular mean slope 36.49; plastic vs UHMWP P < 0.05; plastic vs irregular P < 0.05; UHMWP vs irregular P < 0.05, from groups 13–18).

Discomfort as a function of friction and movement. The irregular surface has higher friction than the smooth surfaces. The irregular surface caused greater discomfort (P < 0.05) discomfort than the smooth surfaces. When comparing groups with the same programmed impact but differing in movement, estimates of discomfort were always higher in the non-movement groups (movement groups 4–9), mean 4.10; non-movement groups (13–18), mean 23.00; P < 0.05).

Plantar pain. Subjects considered the discomfort rating of 100 to be consistent with moderate pain, and 70 indicated mild pain.

DISCUSSION

This experiment relates plantar loading during locomotion and jumping plantar discomfort when shod and underfoot. Both horizontal and vertical impact have persuasive importance to plantar discomfort inasmuch as no discomfort resulted from impact regardless of surface if one element of impact was below 0.4 kg·cm⁻².
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Due to concerns about subject safety, the rate of loading (complete in 1 s) was slower than in actual running; hence, the "sting" that one experiences when the plantar surfaces slap against a rigid surface was not reproduced. Also, the lower frequency when compared with actual running. Other possible limitations of this technique are dealt with elsewhere (50). Future simulations will address these problems while retaining safety precautions.

Deals with public health concerns raised by footwear designed for high impact environments that produce perceptual illusions. Bare-foot activity when practiced (no need for thermal insulation; no risk of injury; social acceptability) deserves consideration since plantar sensory mediated protective adaptations seem optimized for this condi-
tion. Although this may run counter to notions prevalent in economically advanced countries regarding dangers of barefoot activity and necessity of footwear even when barefoot activity is feasible, supporting data are lacking, and many have concluded that footwear design is guided by fashion rather than by health considerations (16,57).

Since this experiment indicates that protective plan-
tars stimulate the body to run more efficiently. This new level, an inadequate understanding of the physiology of human impact control has resulted in footwear which makes chronic overloading inevitable by providing excessive shock to the wearer even when enormous vertical impact is experienced. Informing the public of this hazard seems to be a responsible first step, since care to moderate impact even with existing products is somehow unneeded. It is necessary when improved safety standards for footwear result in products that take into account the importance of planar sensory feedback in high impact environments.

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REFereNCES


Segmental contributions to total body moment to stand

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ABSTRACT

PAI, Y.-C. and M. W. ROGERS. Segmental contributions to total body moment in sit-to-stand. Med. Sci. Sports Exerc., Vol. 23, No. 2, p. 225-230, 1991. In a previous investigation, we reported that the maximum linear momentum of the body center of mass (CM) during a sit-to-stand task showed a relative invariance in the horizontal vs. vertical direction of motion in the speed of ascent increased from natural to fast. The present study investigated the segmental contributions to the vertical component of this difference by examining the linear momentum of the shank, thigh, and head-ankle-trunk segments separately. Segmental forces and moments for the relative invariance in its magnitude. In contrast, the thigh was the major contributor to the horizontal linear momentum of the CM. The head-ankle-trunk was responsible for the progressive increase in its magnitude across the range of speeds. Moreover, this study emphasizes the power of the head-ankle-trunk and of the shank in their general profile and peak magnitudes further suggested that a simplifying strategy may have been employed to reduce the overall number of degrees of freedom associated with the sit-to-stand movement.

IMPLUSE-MOMENTUM PRINCIPLE: BALANCE CONTROL, SEGMENTAL CONTRIBUTIONS, DEGREES OF FREEDOM

Standing from a seated position is one of the most commonly executed functional activities. Although the ability to effectively execute the sit-to-stand (STS) movement is a vital prerequisite for upright mobility, quantitative information pertaining to the STS task has been scarce until recently. In one of the earliest reports, Kelley et al. (1971) described the joint torque-time history at the lower limb in conjunction with electromyographic recordings of selected lower limb muscles. Recent, attention has been directed at investigating the dynamic effects of sitting posture among healthy (8, 18) and neurologically impaired (12 subjects) the influence of seat height (6, 16) and the use of arm support (3, 17), as well as computer simulation of the control processes (10) under conditions. Since the ability to rise from a seated position is frequently impaired among a variety of clinical populations, efforts to examine the functional outcome of surgical intervention (5), functional electrical stimulation (4), or rehabilitation training (7) have focused on the performance of this task.

To characterize the motion of the total body during STS, we previously have shown (15) that the horizontal and vertical components of the mean maximum linear momentum of the body center of mass (CM) differed markedly across a range of self-selected speeds. When the speed of ascent increased progressively from slow to natural to fast, the increase in the mean maximum linear momentum of the CM in the horizontal direction was disproportionately smaller than its vertical counterpart. Moreover, the speed increase from slow to fast was primarily achieved through the increase in the maximum vertical linear momentum. Based on these observations it was proposed that the directionally specific contributions of body motion were attributable to the differences in the mechanical and anatomical constraints on movement that exist in the two directions (15). It was also suggested that the relative invariant features of the horizontal motion of the CM may have reflected a simplifying strategy for the neuromuscular control of balance during the STS movement.

Since the motion of the total body is dependent on the moment of the individual segments of the body, the differences in the maximum linear momentum of the CM in the horizontal vs. vertical direction also must be reflected at the segmental level. Thus, the purpose of this study was to investigate the segmental origin of the directionally specific differences in the maximum linear momentum of the CM during the STS movement.

METHODS

The general methods have been described previously in detail (15) and are briefly presented below.

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