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 $g \cdot d^{-1}$ of carbohydrate, while the MCHO diet contained a mean of 258 $g \cdot d^{-1}$. 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' mood 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.

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REFERENCES

1. BERGSTROM, J. and E. HULTMAN. Nutrition for maximal sports performance. *J.A.M.A.* 221:999–1006, 1972.
2. KARLSSON, J. and B. SALTIN. Diet, muscle glycogen and endurance performance. *J. Appl. Physiol.* 31:203–206, 1972.
3. LIEBERMAN, H. R., J. J. WURTMAN, and B. CHEW. Changes in mood after carbohydrate consumption among obese individuals. *Am. J. Clin. Nutr.* 44:772–778, 1986.
4. MCNAIR, D. M., M. LORR, and L. F. DROPPLEMAN. *Manual: Profile of Mood States.* San Diego: Educational and Industrial Testing Service, 1971.
5. MORGAN, W. P., D. L. COSTILL, J. P. KIRWAN, J. S. RAGLIN, and P. J. O'CONNER. Influence of increased training and diet on mood states. *Med. Sci. Sports Exerc.* 20:S95, 1988.
6. O'KEEFFE, K. A., R. E. KEITH, G. D. WILSON, and D. L. BLESSING. Dietary carbohydrate intake and endurance exercise performance of trained female cyclists. *Nutr. Res.* 9:819–830, 1989.
7. PATE, T. D. and J. C. BRUNN. Fundamentals of carbohydrate metabolism. In: *Nutrition in Exercise and Sport*, J. F. Hickson and I. Wolinsky (Eds.). Boca Raton, FL: CRC Press, 1989, pp. 37–49.
8. RAABO, E. and T. C. TERKILDSEN. On the enzymatic determination of blood glucose. *Scand. J. Clin. Invest.* 12:402–408, 1960.
9. ROSEN, J. C., D. A. HUNT, E. A. H. SIMS, and E. A. H. BOGARDUS. Composition of carbohydrate-containing and carbohydrate-restricted hypocaloric diets in the treatment of obesity: effects on appetite and mood. *Am. J. Clin. Nutr.* 36:463–469, 1982.
10. SHERMAN, W. M. Carbohydrates, muscle glycogen and muscle glycogen supercompensation. In: *Ergogenic Aids in Sports*, M. H. Williams (Ed.). Champaign, IL: Human Kinetics Pub., 1983, pp. 3–26.
11. SHERMAN, W. M. Carbohydrate, muscle glycogen and improved performance. *Physician Sportsmed.* 15:157–164, 1987.
12. SPRING, B., D. MALLER, J. WURTMAN, L. DIGMAN, and L. COZOLINO. Effects of protein and carbohydrate meals on mood and performance: interactions with sex and age. *J. Psychiatr. Res.* 17:155–167, 1983.
13. SPRING, B., J. CHIODO, M. HARDEN, et al. Effects of noon meals varying in nutrient composition on plasma amino acids, glucose, insulin and behavior. *Psychopharmacol. Bull.* 22:1026–1029, 1986.
14. ZAR, J. H. *Biostatistical Analysis.* Englewood Cliffs, NJ: Prentice-Hall, Inc., 1984, p. 185.

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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 elicit negligible perceived plantar discomfort, a perceptual illusion is created whereby perceived impact is lower than actual impact, which results in inadequate impact-moderating 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 (concurrent vertical (0.1–0.7 $kg \cdot cm^{-2}$) and horizontal (0.1–0.9 $kg \cdot cm^{-2}$) 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 $kg \cdot cm^{-2}$. 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 loads. 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, and 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 (1,11,43). The vertical component of plantar impact results in propagation of shock waves (20,31,38,55,62,63) 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 (3,10,20,21,23,25–28,31–34,46,56,59–61).

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 shipped—with the use of yielding (compliant, soft) packaging materials. The more recent models 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 epidemiologic studies dealing with running-related injury incidence with recent reports suggests that there is presently a higher incidence of these injuries (e.g., Marti 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%). Wearers of expensive running shoes that are promoted as having additional features that protect (e.g., more cushioning, "pronation correction") are injured 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 no brand loyalty (32). 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 (13–15,19,22,30,54). In addition, in barefoot populations running-related injuries are rare, which indicates that

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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.

(Avoidance is behavior that moderates stimulus intensity or evades the stimulus entirely. Natural surfaces refer to naturally deposited ground, i.e., irregular surfaces.)

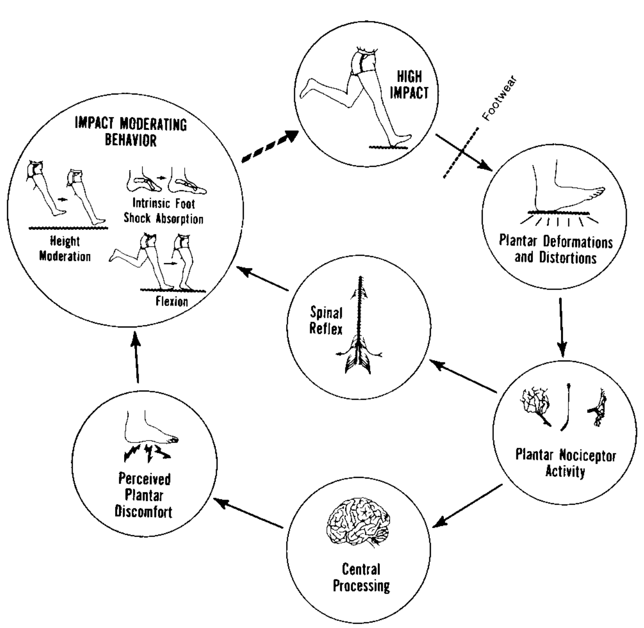


Figure 1—The hypothesis of innate behavioral impact moderation in graphic form. The *dashed arrow* between impact moderating behavior and high impact indicates control exerted over high impact on subsequent impacts as a desire to minimize uncomfortable plantar sensations. Footwear is seen as interfering with the link between high impact and adequate stimuli of plantar deformations and distortions. According to current thinking, impact moderating behavior is composed of three elements, but we believe that a more complex model will evolve.

What support does this hypothesis have? It explains the difference in injury incidence between barefoot and shod 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, their longitudinal foot arches deflect from highly arched to flat with each gait cycle, which likely has shock absorbing properties (6,58).

The theory explains why material tests fail to predict actual impact when running (13–15,19,30,54). 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-moderating behavior.

The linkage between plantar perceptual processes and impact-moderating 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 demonstrated that barefoot activity outdoors (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 modern 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) that produces plantar discomfort, 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 surface. The diminished impact vis-à-vis the hard 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 yielding layers. To appreciate the depth of this perceptual illusion, consider that, based on visual anticipation and previous learning of consequences of landing of mats and hard surfaces, these subjects reduced impact-moderating behavior so as to compress a 10.8-cm-thick mat and to further deliver peak impact to the surface below that was 20–25% greater than when they landed directly on the hard surface. Yet subjects retained the impression of lower impact when landing on the mats! We believe that the discomfort-perceptual illusion when jumping may be related to the low pain threshold at the metatarsal-phalangeal joint area of the plantar surface, which we previously reported (48). The perceptual illusion when landing on mats seems more profound than that produced by athletic footwear during locomotion, and this may be accounted for by differences in thickness of the yielding layers (training type running shoe sole thickness is about 2.8 cm). Similarly, the enhanced illusion created by thicker soles of more expensive footwear may account for more injuries than inexpensive shoes, which usually have thinner soles (32). It may also explain the paradoxical sharp increase in impact when subjects run with unusually compliant shoes (below Shore 30) (37).

Clearly, when plantar sensory consequences of impact are attenuated, humans underestimate impact and reduce impact-moderating behavior, which may elevate impact sufficiently to cause chronic overloading. For purposes of reducing the risk of injury, it does not seem unreasonable to propose a safety standard for footwear, particularly for footwear promoted for use in a high impact environment, which requires elimination of the discomfort-impact illusion. But is it possible to heighten plantar sensory feedback sufficiently to eliminate this illusion while wearing footwear; i.e., is this proposed safety standard feasible? To address this question and to understand more clearly the plantar perceptual consequences of locomotion and jumping, we performed a psychophysical experiment in which subjects estimated plantar discomfort produced by simulated locomotion consisting of simultaneous application of plantar vertical and horizontal loads. We measured plantar discomfort as a function of vertical load, horizontal load, surface irregularity differences, friction (plantar surface against plantar contact surface), and movement (horizontal foot movement produced by impact).

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 reapplication of impact. Uncomfortable sensations from the thigh and Achilles attachment were minimized by interfaces composed of elastomeric material.

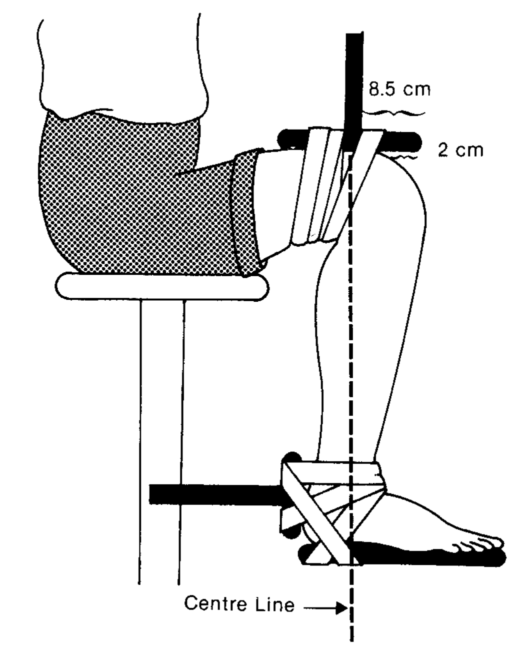


Figure 2—Positioning of subject's leg in apparatus.

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 arches. The travel of the actuators was maintained at 6 cm vertically and 12 cm horizontally. Vertical impact was programmed to reach 0.4 $kg \cdot cm^{-2}$ 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

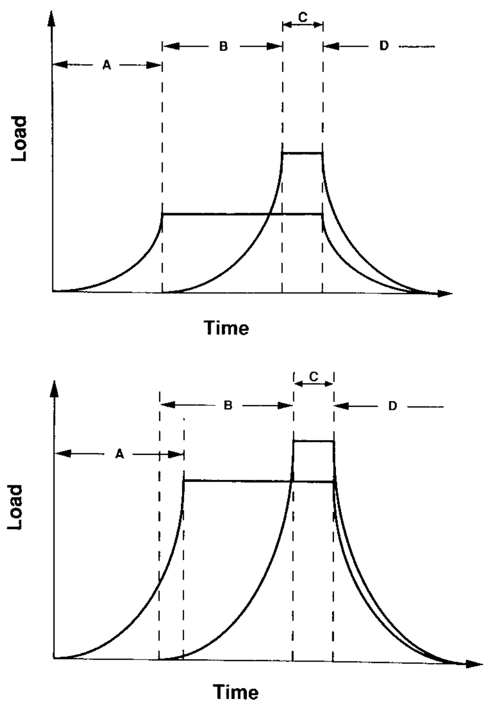


Figure 3—Schematic diagram of load application by the testing apparatus. Region A: vertical load increases with time until a preset level is reached; horizontal load is zero. Region B: horizontal load increases with time until a present level is reached; vertical load remains constant. Region C: horizontal and vertical loads remain constant. Region D: horizontal and vertical loads drop to zero. The *top panel* represents the case whereby the maximum vertical load is 0.4 kg·cm^{-2} or less, in which case the horizontal load is applied immediately after the maximum vertical load has been reached. The *bottom panel* represents the case whereby the maximum vertical load is larger than 0.4 kg·cm^{-2}, in which case the horizontal load is applied when the vertical load reaches 0.4 kg·cm^{-2}.

(mean 25.4 ± 1.9)). Their consent was obtained according to the Declaration of Helsinki.

Normalization of applied impact. Vertical and horizontal impact were normalized with respect to foot contact area using a previously reported equation which relates weight-bearing (at 50 kg vertical load) plantar contact area, on a flat rigid plantar contact surface, with subject height (50).

Impact applied. Impact consisted of all possible combinations of three vertical impacts (0.1, 0.4, and 0.7 kg·cm^{-2}) and five horizontal impacts (0.1, 0.3, 0.5, 0.7, and 0.9 kg·cm^{-2}). The values of applied impact chosen were related to our goal of focusing on horizontal impact (two previous reports have dealt with vertical impact (50,51)) while avoiding constraints due to plantar pain, since a pilot study indicated that vertical impact of 0.7 kg·cm^{-2} and horizontal impact of 0.9 kg·cm^{-2} produced moderate pain in most subjects when the irregular plantar contact surface was used. Despite the limited range of vertical impact, these results can presumably be extrapolated to higher impact levels (when running, peak vertical impact often exceeds 2.0 kg·cm^{-2}), since we have shown that the relation between vertical impact and the perceived magnitude of these loads is linear to an amplitude of 2.0 kg·cm^{-2}, whereafter the slope increases probably due to the onset of plantar pain, but only when the foot impacts an extremely irregular surface (51).

Data recorded. 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 Solidur Canada Co. (Montreal, Canada).

3) Textured surface—pour 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.

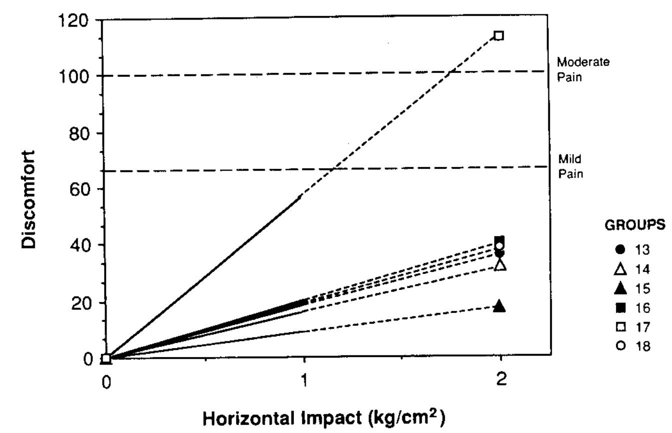
3) Random series of load pairings: Subjects were given a chart which displayed three ranges. The upper limit was the estimate of maximum discomfort for each surface; the lower limit was zero. Subjects were instructed to estimate discomfort produced by impacts that would follow from within the specified 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 different randomly obtained order for the three surfaces chose 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; $P > 0.05$), though, when vertical impact equaled or exceeded 0.4 kg·cm^{-2} (groups 13–18), there was a significant relation between these variables (min. slope 1.99; max. slope 56.49; mean slope 18.42; $P < 0.05$). 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).



Groups	N	Surface	I_V(kg/cm^2)	Move	Slope	p<0.05
1	20	PLASTIC	.1	+	3.12 ± 1.59	
2	20	IRREG	.1	+	1.4 ± 2.85	i
3	20	UHMWP	.1	+	1.07 ± 1.51	
4	20	PLASTIC	.4	+	-1.40 ± 2.83	e
5	19	IRREG	.4	+	0.02 ± 5.74	j
6	20	UHMWP	.4	+	-1.79 ± 2.97	
7	18	PLASTIC	.7	+	4.92 ± 3.27	af
8	20	IRREG	.7	+	20.87 ± 6.92	afgij
9	18	UHMWP	.7	+	1.99 ± 2.21	bh
10	< 3	PLASTIC	.1	-	NA	
11	< 3	IRREG	.1	-	NA	
12	< 3	UHMWP	.1	-	NA	
13	14	PLASTIC	.4	-	17.54 ± 8.37	e
14	17	IRREG	.4	-	16.48 ± 13.61	k
15	11	UHMWP	.4	-	8.47 ± 5.90	
16	19	PLASTIC	.7	-	20.65 ± 5.72	cf
17	20	IRREG	.7	-	56.49 ± 8.64	cdgk
18	17	UHMWP	.7	-	18.37 ± 5.69	dh

Figure 4—Results. Table: N, number of slopes; I_v, vertical impact programmed to be delivered (in movement groups actual horizontal load was less than the value in this column); slope, slope of horizontal impact vs discomfort; $P < 0.05$, probability from *t*-tests; pairings with significant difference can be obtained by matching lower case letters. NA, paucity of data in these groups did not allow further calculations. Graph: plot of discomfort as a function of horizontal impact for non-movement groups. *Dashed parts of lines* are extrapolations into load ranges seen in shod runners.

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 discomfort 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 discomfort 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 ($P < 0.05$) 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 and plantar discomfort when shod and unshod. Both horizontal and vertical impact have permissive 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}

(groups 1–6; Fig. 4). Further, these impact components act synergistically in producing discomfort, since discomfort was always greater when these elements were equal rather than divergent. For example, with vertical at 0.4 $kg \cdot cm^{-2}$ and horizontal at 1.0 $kg \cdot cm^{-2}$, discomfort was 14.16, whereas, with vertical at 0.7 $kg \cdot cm^{-2}$ and horizontal at 0.7 $kg \cdot cm^{-2}$, discomfort was 22.28 ($P < 0.05$).

Surface irregularities strongly influenced discomfort, not permissively, but rather as an adjuvant, since smooth surfaces produced substantial discomfort (vertical impact 0.7 $kg \cdot cm^{-2}$; slope of discomfort-horizontal impact for smooth surfaces 19.51). Heightened discomfort with the irregular surface was not a function of vertical deformation via irregularities, for no subject reported discomfort when standing on this surface. Nor is it a property of urethane rubber material, seeing that when smooth it is similar to acrylic plastic and UHMWP in both hardness and friction. We favor an explanation whereby vertical deformation by way of irregularities anchor epidermis in the horizontal plane, which causes shear-distortion of subcutaneous tissues when horizontal load is applied. Subcutaneous shear-distortion is thought to be an adequate stimulus for certain nociceptors (4,8).

With horizontal foot movement across contact surface, estimates of discomfort were lower than without movement, which is explained by programmed horizontal load being higher than actual load applied. Movement of the foot across irregularities in and of itself did not heightened discomfort.

Since our interest centers on the control of vertical impact during locomotion and jumping, the results can be restated: horizontal impact and small surface irregularities, thought not producing chronic overloading by themselves, cause this indirectly through their influence on vertical impact-moderating avoidance.

How much horizontal impact is experienced during locomotion when shod and unshod? Whereas vertical impact during locomotion in footwear has received considerable attention, reports about horizontal impact are rare. In one report in which minute plantar attached shear transducers were used, horizontal impact when walking barefoot approximated vertical impact but fell by two-thirds by adding walking shoes and still further by various footwear modifications (41). Although horizontal impact during running has not been measured, yielding contoured layers of modern athletic footwear probably diminish horizontal impact at least as effectively as leather shoes. This is also suggested by the ability of modern athletic shoes to control plantar blistering, a consequence of horizontal loads applied to the unadapted plantar skin.

We can now calculate plantar discomfort experienced by shod and unshod runners and estimate vertical impact constraints imposed by these sensations. Consider the irregular surface used in this experiment, which is actually just mildly irregular when compared with actual natural surfaces, as a natural surface. Think of the smooth surfaces as the regular interior of modern athletic footwear. Conservatively (41), assume that horizontal load with the barefoot runner is 1.0 $kg \cdot cm^{-2}$ and is half this value with modern athletic footwear. Maximum discomfort tolerated will be considered moderate pain (100) on the ordinal scale. The maximum vertical impact tolerated by the barefoot runner would be 1.03 $kg \cdot cm^{-2}$ (190% of body weight; from groups 14 and 17), whereas the maximum vertical impact tolerated by the shod runner would be 5.76 $kg \cdot cm^{-2}$ (820% of body weight; groups 13, 15, 16 and 18; Fig. 4). Since a multiple of 190% of body weight is lower than vertical impact often measured when subjects run in modern footwear, and 820% of body weight greatly exceeds the vertical impact measured when shod subjects run at maximum velocity, we conclude that plantar sensations induce the barefoot runner to mitigate vertical impact considerably, whereas the shod runner using currently available footwear is not persuaded by plantar sensations to lessen vertical impact at all. This analysis allows room for sensory attenuation from increased plantar rigidity through hyperkeritinization of the barefoot runner's sole; otherwise, without the conservative assumptions made above, it is difficult to understand how the barefoot runner could endure running.

As previously noted (48), these experiments suggest how at the receptor and primary afferent level impact is sensed. Polymodal nociceptors with C-fiber afferents seem well equipped to satisfy all of the requirements for impact sensing (4,5,8,9,24,39,40). They are the predominant nociceptor with C-fiber afferents at the distal portion of the extremity in higher mammals. Also, their threshold, response to deformation and horizontal skin displacement, ease at being sensitized, and temporal response pattern to sustained stimuli conform well to the dynamics of locomotion.

Is a safety standard for footwear which would require the elimination of the plantar discomfort-impact illusion feasible? Adding small rigid irregularities to flat, fairly rigid insoles would reduce vertical impact below levels commonly recorded in shod runners. Sensory feedback could be augmented further by creating firm and uncontoured interiors (features typically present in inexpensive footwear), since contouring diminishes peak horizontal impact delivered to the plantar surface by distributing it over a larger foot area. Considering the above, this proposed safety standard seems viable.

Would locomotion in shoes with the above modifications be uncomfortable? We do not think so. Whereas the comfort of current athletic footwear may (perhaps should) not be obtainable, behavioral adaptations that reduce plantar impact would allow reasonable comfort during locomotion.

The limitations of this study relate to the quality of the simulation. 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 surface slaps against a rigid surface was not reproduced. Also, loads were cycled at a low 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 safeguards.

The following discussion deals with public health concerns raised by footwear designed for high impact environments that produce perceptual illusions. Barefoot activity when practical (no need for thermal insulation; no risk of crush injuries; social acceptability) deserves consideration since plantar sensory mediated protective adaptations seem optimized for this condition. Although this may run counter to notions prevalent in economically advanced countries recounting 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 health considerations (16,57).

Since this experiment indicates that protective plantar sensations require horizontal impact, and this occurs only when there is locomotion (horizontal displacement of the body), continuous stationary jumping may be particularly dangerous. This may explain the unusually high injury incidence in those who participate in "aerobic dance", which relies on stationary repetitive jumping to obtain an aerobic training effect (21,23,46,61).

Attention needs to be paid to claims made by manufacturers of athletic footwear. They cannot be faulted for their products' poor sensory feedback, as this is new information; advertising suggesting that current footwear protects better than earlier models or reduces injury risk is spurious in the light of reports indicating otherwise (1,13–15,30,37,54) and is irresponsible inasmuch as this contributes to morbidity by giving the user a false sense of security.

Safety standards for athletic footwear are inadequate because they do not account for the discomfort-impact illusion. Until footwear is available that can meet this standard, it should be dealt with as are other commonly used products that create hazardous perceptual illusions. For example, the automobile wide-angle rearview mirror has contributed to accidents by making objects appear farther away than they actually are. To moderate injury risk, a message is etched on the mirror describing the illusion so that drivers can compensate for it. It does not seem unreasonable to require a similar message to be affixed to footwear that produce injurious perceptual illusions.

In summary, people who perform activities involving high impact while wearing footwear currently promoted as offering protection in this environment are at high risk for injury. Unlike the natural state (barefoot and natural surfaces), where impact is sensed and, through impact-moderating behavior, is maintained at a safe level, an inadequate understanding of the physiology of human impact control has resulted in footwear which makes chronic overloading inevitable by providing plantar comfort 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 may reduce morbidity. However, this will become unnecessary when improved safety standards for footwear result in products that take into account the importance of plantar sensory feedback in high impact environments.

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REFERENCES

1. Bates, B. T., L. R. Osternig, J. A. Sawhill, and S. L. James. An assessment of subject variability, subject-shoe interaction and the evaluation of running shoes using ground reaction force data. *J. Biomech.* 16:181–192, 1983.
2. Beer, F. P. and E. R. Johnston Jr. *Vector Mechanics For Engineers: Dynamics,* 4th Ed. New York: McGraw-Hill, 1984, p. 572.
3. Bertram, J. J. Stress fractures in Royal Marine recruits. *Roy. Natl. Med. Serv.* 56:78–91, 1970.
4. Bessou, P. and E. R. Perl. Response of cutaneous sensory units with unmyelinated fibers to noxious stimuli. *J. Neurophysiol.* 32:1025–1043, 1969.
5. Bessou, P., R. P. Bessou, E. R. Perl, and C. B. Taylor. Dynamic properties of mechanoreceptors with unmyelinated (C) fibers. *J. Neurophysiol.* 34:116–131, 1971.
6. Bradford, E. H. An examination of human gait. *Trans. Am. Orthop. Assoc.* 10:137, 1897.
7. Bremmer, J. M., J. S. Lawrence, and W. E. Maill. Degenerative arthritis in a Jamaican rural population. *Ann. Rheum. Dis.* 27:326–332, 1968.
8. Burgess, P. R. and E. R. Perl. Cutaneous mechanoreceptors and nociceptors. In: *Handbook of Sensory Physiology, Vol. 2, Somatosensory System,* A. Iggo (Ed.). Berlin: Springer, 1973, pp. 59–78.
9. Campbell, J. N. and R. H. LaMotte. Latency to detection of first pain. *Brain Res.* 266:203–208, 1983.
10. Caspersen, C. J., K. E. Powell, J. P. Koplan, R. W. Shirley, C. C. Campbell, and R. K. Sikes. The incidence of injuries and hazards in recreational and fitness runners (Abstract). *Med. Sci. Sports Exerc.* 16:113, 1984.
11. Cavanagh, P. R. and M. A. Lafortune. Ground reaction forces in distance running. *J. Biomech.* 13:397–406, 1980.
12. Cavanagh, P. R., G. A. Valiant, and K. W. Misevich. Biological aspects of modelling shoe/foot interaction during running. In: *Sports Shoes and Playing Surfaces,* E. C. Frederick (Ed.). Champaign, IL: Human Kinetics, 1984, pp. 24–46.
13. Clarke, T. E. and E. C. Frederick. Dynamic load displacement characteristics of various midsole materials. *J. Biomech.* 15:340, 1982.
14. Clarke, T. E., E. C. Frederick, and L. B. Cooper. The effects of shoe cushioning upon selected force and temporal parameters in running. *Med. Sci. Sports Exerc.* 14:144, 1982.

15. Clarke, T. E., E. C. Frederick, and L. B. Cooper. Effects of shoe cushioning upon ground reaction forces in running. *Int. J. Sports Med.* 4:247–251, 1983.
16. Collazzo, C. A study of consumer problems in shopping for footwear, with emphasis on size and fit. *J. Test. Eval.* 16:421–424, 1988.
17. Cook, S. D., M. A. Kester, and M. E. Brunet. Shock absorption characteristics of running shoes. *Am. J. Sports Med.* 13:248–253, 1985.
18. Dekel, S. and S. L. Weissman. Joint changes after overuse and peak overloading of rabbit knees in vivo. *Acta Orthop. Scand.* 49:519–528, 1978.
19. Dickenson, J. A., S. D. Cook, and T. M. Leinhardt. The measurement of shock waves following heel strike while running. *J. Biomech.* 18:1–8, 1985.
20. Dyck, P. J., S. M. Classen, J. C. Stevens, and P. C. O'Brien. Assessment of nerve damage in the feet of long-distance runners. *Mayo Clin. Proc.* 62:568–572, 1987.
21. Francis, L. L., P. R. Francis, and K. Welshons-Smith. Aerobic dance injuries: a survey of instructors. *Physician Sportsmed.* 13:105–111, 1985.
22. Frederick, E. C., T. E. Clarke, and C. L. Hamill. The study of rearfoot movement in running. In: *Sports Shoes and Playing Surfaces,* E. C. Frederick (Ed.). Champaign, IL: Human Kinetics Publishers, 1984, pp. 90–198.
23. Garrick, J. G., D. M. Gillien, and P. Whiteside. The epidemiology of aerobic dance injuries. *Am. J. Sports Med.* 14:67–72, 1985.
24. Georgopulos, A. P. Functional properties of primary afferent units probably related to pain mechanisms in primate glabrous skin. *J. Neurophysiol.* 39:71–83, 1976.
25. Gilbert, R. S. and H. A. Johnson. Stress fractures in military recruits: a review of twelve years' experience. *Milit. Med.* 131:716–721, 1966.
26. Goodyear, N. N. and S. N. Blair. Incidence rates and risk of orthopedic injuries in healthy middle aged runners and nonrunners. *Med. Sci. Sports Exerc.* 16:114, 1984.
27. Gudas, C. J. Patterns of lower-extremity injury in 224 runners. *Compr. Ther.* 6:50–59, 1980.
28. Jacobs, S. J. and B. L. Berson. Injuries to runners: a study of entrants to a 10,000 meter race. *Am. J. Sports Med.* 14:151–155, 1986.
29. Light, L. H., G. E. McLellan, and L. Klenerman. Skeletal transients on heel strike in normal walking with different footwear. *J. Biomech.* 13:477–480, 1980.
30. Luethi, S. M., J. Denoth, X. Kaelin, et al. The influence of the shoe on foot movement and shock attenuation in running. In: *Biomechanics X-B,* B. Jonsson (Ed.). Champaign, IL: Human Kinetics Publishers, 1987, pp. 931–935.
31. Lysholm, J. and J. Wiklander. Injuries in runners. *Am. J. Sports Med.* 15:168–171, 1987.
32. Marti, B. Relationships between running injuries and running shoes: In: *The Shoe in Sport,* W. Pforringer and B. Segesser (Eds.). Chicago: Yearbook Medical Publishers, 1989, pp. 256–265.
33. Marti, B., J. P. Vater, C. E. Minder, and T. Abelin. On the epidemiology of running injuries: the 1984 Bern Grand-Prix study. *Am. J. Sports Med.* 16:285–294, 1988.
34. McDermott, M. and P. Freyne. Osteoarthrosis in runners with pain. *Br. J. Sports Med.* 17:84–87, 1983.
35. McMahon, T. A., G. Valiant, and E. C. Frederick. Groucho running. *J. Appl. Physiol.* 62:2326–2337, 1987.
36. McNitt-Gray, J. L., and T. Yokoi. The influence of surface characteristics on the impulse characteristics of drop landings. Proceedings of the 13th Annual Meeting of the American Society on Biomechanics, Aug. 23–25, 1989. Burlington, VT: 1989, pp. 92–93.
37. Nigg, B. M., A. H. Bahlson, J. Denoth, S. M. Luethi, and A. Stacoff. Factors influencing kinetic and kinematic variables in running. In: *Biomechanics of Running Shoes,* B. M. Nigg. (Ed.). Champaign, IL: Human Kinetics Publishers, 1986, pp. 139–159.
38. Paul, I. L., M. B. Munro, P. J. Abernathy, S. R. Simon, E. L. Radin, and R. M. Rose. Musculo-skeletal shock absorption: relative contribution of bone and soft tissues at various frequencies. *J. Biomech.* 11:237–239, 1978.
39. Perl, E. R. Mode of action of nociceptors. In: *Cervical Pain,* C. Hirsch and Y. Zotterman, (Eds.). Oxford: Pergamon Press, 1972, pp. 157–164.
40. Perl, E. R., T. Kumazawa, B. Lynn, and P. Kenins. Sensitization of high threshold receptors with unmyelinated (C) afferent fibers. *Prog. Brain Res.* 43:263–277, 1976.
41. Pollard, J. P., L. P. Le Quesne, and J. W. Tappin. Forces under the foot. *J. Biomed. Eng.* 5:37–40, 1983.
42. Radin, E. L., D. Eyre, J. L. Kelman, and A. L. Schiller. Effect of prolonged walking on concrete on the joints of sheep. *Arthritis Rheum.* 22:649, 1980.
43. Radin, E. L., R. B. Orr, J. L. Kelman, I. L. Paul, and R. M. Rose. Effect of prolonged walking on concrete on the knees of sheep. *J. Biomech.* 15:487–492, 1982.
44. Radin, E. L., H. G. Parker, G. V. Pugh, R. S. Steinberg, I. L. Paul, and R. M. Rose. Response of joints to impact loading. *J. Biomech.* 6:51–57, 1973.
45. Radin, E. L. and I. L. Paul. Response of joints to impact loading. *Arthritis Rheum.* 14:356–362, 1971.
46. Richie, D. L., S. F. Kelso, and P. A. Bellucci. Aerobic dance injuries: a retrospective study of instructors and participants. *Physician Sportsmed.* 13:130–140, 1985.
47. Robbins, C. E. and G. J. Gouw. Athletic footwear and chronic overloading: a brief review. *Sports Med.* 9:76–85, 1990.
48. Robbins, S. E., G. J. Gouw, and A. M. Hanna. Running-related injury prevention through innate impact-moderating behavior. *Med. Sci. Sports Exerc.* 21:130–139, 1989.
49. Robbins, S. E. and A. M. Hanna. Running related injury prevention through barefoot adaptations. *Med. Sci. Sports Exerc.* 19:148–156, 1987.
50. Robbins, S. E., A. M. Hanna, and G. Gouw. Overload protection: avoidance response to heavy plantar surface loading. *Med. Sci. Sports Exerc.* 20:85–92, 1988.
51. Robbins, S. E., A. M. Hanna, and L. Jones. Sensory attenuation induced by modern athletic footwear. *J. Test. Eval.* 16:412–416, 1988.
52. Serink, M. T., A. Nachemson, and G. Hansson. The effect of impact loading on rabbit knee joints. *Acta Orthop. Scand.* 48:250–262, 1977.
53. Simon, S. R., E. L. Radin, I. L. Paul, and R. M. Rose. The response of joints to impact loading. II. In vivo behavior of subchondral bone. *J. Biomech.* 5:267–272, 1972.
54. Snel, J. G., N. J. Dellmann, Y. F. Heerkens, and G. J. Van Ingen Schenau. In: *Biomechanics IX-A,* D. A. Winter, R. W. Norman, R. P. Wells, et al. (Eds.). Champaign, IL: Human Kinetics Publishers, 1985, pp. 133–138.
55. Soames, R. W. and R. P. S. Richardson. Stride length and cadence: their influence on ground reaction forces during gait. In: *Biomechanics IX-A,* D. A. Winter, R. W. Norman, R. P. Wells, et al. (Eds.). Champaign, IL: Human Kinetics Publishers, 1985, pp. 406–410.
56. Stanitski, C. L., J. H. McMaster, and P. E. Scranton. On the nature of stress fractures. *Am. J. Sports Med.* 6:391–396, 1978.
57. Stewart, S. F. Physiology of the unshod and shod foot with an evolutionary history of footgear. *Am. J. Surg.* 68:127–138, 1945.
58. Stewart, S. F. Human gait and the human foot. *Clinical Orthop.* 70:111–123, 1970.
59. Stokes, I., I. B. Faris, and W. C. Hutton. The neuropathic ulcer and loads under the foot in diabetic patients. *Acta Orthop. Scand.* 46:839–847, 1975.
60. Sullivan, D., R. F. Warren, and H. Pavlov. Stress fractures in 51 runners. *Clin. Orthop.* 187:188–192, 1984.
61. Vetter, W. L., D. L. Helfet, K. Spear, and L. S. Matthews. Aerobic dance injuries. *Physician Sportsmed.* 13:114–120, 1985.
62. Voloshin, A. and J. Wosk. An in-vivo study of lowback pain and shock absorption in the human locomotor system. *J. Biomech.* 15:21–27, 1982.
63. Voloshin, A., J. Wosk, and M. Brull. Force wave transmission through the human locomotor system. *J. Biomech. Eng.* 103:48–50, 1981.

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BIODYNAMICS

Segmental contributions to total body momentum in sit-to-stand

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ABSTRACT

PAI, Y.-C. and M. W. ROGERS. Segmental contributions to total body momentum in sit-to-stand. *Med. Sci. Sports Exerc.,* Vol. 23, No. 2, pp. 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 the vertical direction of motion as the speed of ascent increased from natural to fast. The present study investigated the segmental origin of this directionally specific difference by examining the linear momentum of the shank, thigh, and head-arm-trunk segments for ten healthy young adults at slow, natural, and fast self-selected speeds. Findings indicated that the head-arm-trunk was the major contributor to the horizontal maximum linear momentum of the CM and accounted for the relative invariance in its magnitude. In contrast, the thigh was the major contributor to the vertical maximum linear momentum of the CM and was responsible for the progressive increase in its magnitude across the range of speeds. Moreover, the compatibility between the motions of the head-arm-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.

IMPULSE-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. (13) described the joint torque-time history at the lower limb in conjunction with electromyographic recordings of selected lower limb muscles. Recently, 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) underlying the STS movement. 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 natural 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 differences in the total 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 relatively 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 motion 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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