Kinematic Adaptations during Running: Effects of Footwear, Surface, and Duration

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ABSTRACT

HARDIN, E. C., A. J. VAN DEN BOGERT, and J. HAMILL. Kinematic Adaptations during Running: Effects of Footwear, Surface, and Duration. *Med. Sci. Sports Exerc.*, Vol. 36, No. 5, pp. 838–844, 2004. Repetitive impacts encountered during locomotion may be modified by footwear and/or surface. Changes in kinematics may occur either as a direct response to altered mechanical conditions or over time as active adaptations. **Purpose:** To investigate how midsole hardness, surface stiffness, and running duration influence running kinematics. **Methods:** In the first of two experiments, 12 males ran at metabolic steady state under six conditions; combinations of midsole hardness (40 Shore A, 70 Shore A), and surface stiffness (100 $kN \cdot m^{-1}$, 200 $kN \cdot m^{-1}$, and 350 $kN \cdot m^{-1}$). In the second experiment, 10 males ran for 30 min on a 12% downhill grade. In both experiments, subjects ran at 3.4 $m \cdot s^{-1}$ on a treadmill while 2-D hip, knee, and ankle kinematics were determined using high-speed videography (200 Hz). Oxygen cost and heart rate data were also collected. Kinematic adaptations to midsole, surface, and running time were studied. **Results:** Stance time, stride cycle time, and maximal knee flexion were invariant across conditions in each experiment. Increased midsole hardness resulted in greater peak ankle dorsiflexion velocity ($P = 0.0005$). Increased surface stiffness resulted in decreased hip and knee flexion at contact, reduced maximal hip flexion, and increased peak angular velocities of the hip, knee, and ankle. Over time, hip flexion at contact decreased, plantarflexion at toe-off increased, and peak dorsiflexion and plantarflexion velocity increased. **Conclusion:** Lower-extremity kinematics adapted to increased midsole hardness, surface stiffness, and running duration. Changes in limb posture at impact were interpreted as active adaptations that compensate for passive mechanical effects. The adaptations appeared to have the goal of minimizing metabolic cost at the expense of increased exposure to impact shock. **Key Words:** LOWER-EXTREMITY GEOMETRY, MIDSOLES, SURFACE STIFFNESS, GRADE, SHOCK

Loading is necessary for maintenance of cartilage, bone, and muscle health (9,18,34,49). An optimal loading window for tissue health can be characterized by repeated impacts of certain magnitude, duration, and frequency, but these variables and their interaction are not well understood (42). Beyond the optimal loading window, repeated impacts can cause osteoarthritis (45,47) in animal models. Although no direct experimental evidence of this exists in humans, there is general agreement that repeated impact loading can lead to injury. For example, higher rates of injury have been associated with greater running mileage (28,31,35).

It is unclear how the human musculoskeletal system adapts during repetitive loading such as running, and how any adaptations within a session of running are influenced by footwear, surface or running duration. It is known, however, that changes in joint rotations affect the impact imposed on the body (8,21). The kinematic changes that may modify impact forces are foot inversion, ankle dorsiflexion, and knee flexion (2,10,21,22,37,40). Although experimental evidence of kinematic adaptations to impact is limited (19), this may be because some effective kinematic adaptations are too small to be measured or because of the limited conditions under which adaptations are examined.

Conditions such as footwear and surface are external influences on the foot/ground impact, of which footwear has received more attention from runners and researchers. Running shoe midsoles have commonly been designed to cushion impact, but their effectiveness has been debated (6,39). There is some evidence, however, that properly cushioned footwear can decrease injuries (30). Shoes with less cushioning, on the other hand, have been shown to cause increased knee flexion velocity (20,51) and increased energy cost (20). With respect to the running surface, harder surfaces have been associated with injuries (41) whereas a "tuned" surface can alleviate injures and enhance speed (36). Furthermore, surface stiffness modifications can cause leg stiffness to change in order to keep the same combined stiffness of the runner and surface (16,32), although this concept of leg stiffness is reflective of a whole body response rather than an adaptation at a particular joint.

There may be a cumulative injury risk with each running stride (31) because longer distances run per week have been associated with injury, both in the general population (33,35) and in military trainees where diet and activities are controlled (31). It seems then that over longer distances runners may function near the limit of healthy loading.

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TABLE 1. Subject characteristics for both experiments; subjects were similar in body mass, height, and average running distance per week.

Subjects	Body Mass (kg)	Height (m)	Average Running Distance per Week (km)
Experiment 1			
Mean ± SD (N = 12)	69.4 ± 4.44	1.75 ± 0.053	20.0 ± 7.4
Experiment 2			
Mean ± SD (N = 10)	74.0 ± 6.97	1.76 ± 0.058	22.7 ± 11.9

Furthermore, when distance running occurs on a downhill grade, runners encounter greater impact magnitude (23,27) and require greater negative work from the lower extremities (4). In fact, prolonged downhill running has been used to study the influence of duration on the attenuation of impact shock (38). The kinematic mechanisms that are responsible for such adaptations are not well understood.

The primary purpose of this paper was to investigate adaptations in sagittal plane kinematics to midsole hardness, surface stiffness and running duration.

METHODS

Subjects

The subjects for both experiments were volunteers from a university population and had similar body mass, height, and weekly mileage characteristics (Table 1). In both experiments, an informed consent document and Physical Activity Readiness Questionnaire were read and signed by the subjects in accordance with University of Massachusetts Amherst policy. Each subject was a heel-toe runner, was free from lower-extremity injury, and had previous treadmill running experience. Heel-toe runners were used because footstrike pattern influences ankle stiffness (48) and if not controlled can confound lower-extremity measurements. The experimental sample size was estimated from previous running kinematic data (25) using a power analysis (7). For experiment 1, 12 subjects were used to keep the condition order balanced, but 10 were deemed necessary to provide the statistical power (0.80) to detect parameter differences of 20%. Similarly, 10 subjects were deemed sufficient for experiment 2.

Treadmill

Subjects ran on a treadmill with modifiable bed compliance in experiment 1 (Precor, M9.3s, Bothell, WA). Three compliance settings were chosen to simulate different surface hardnesses which corresponded to stiffnesses of 100 $kN \cdot m^{-1}$ (soft), 200 $kN \cdot m^{-1}$ (medium), and 350 $kN \cdot m^{-1}$ (hard). Treadmill stiffness was calculated by measuring maximal vertical deflection of the treadmill bed (Motion Analysis Corp., Santa Rosa, CA) at each of the three compliance settings and assuming that a 75-kg runner would exert 2.5 times body weight at midstance (43).

Shoes

All subjects wore shoes designed specifically for these experiments. These shoes were identical in mass and construction. For experiment 1, they differed only in midsole hardness, 40 Shore A, and 70 Shore A (soft and hard, respectively). For experiment 2, all subjects wore the hard shoes (70 Shore A).

Protocol

Experiment 1. After a standard treadmill warm-up, subjects ran at 3.4 $m \cdot s^{-1}$ for 6 min in each of six conditions with the treadmill at level grade. Condition order was a combination of midsole hardness and surface stiffness (Table 2), and order was balanced across the 12 subjects using a Latin square. This duration was chosen to assure that subjects had reached metabolic steady state. Kinematic and metabolic data were collected from the third to sixth min during each of the six conditions. Between conditions, subjects rested until their heart rate was less than 120 bpm and they reported readiness for the following condition. Subjects were assured to be running at metabolic steady state from the third to sixth minute, as there were no statistically significant differences in oxygen uptake between these minutes.

Experiment 2. Subjects ran on a hard-surfaced treadmill for 30 min at 3.4 $m \cdot s^{-1}$ with the treadmill positioned to a −12% grade while kinematic and metabolic data were collected. This grade and duration was chosen because it represented a heightened impact exposure compared with the subjects' normal daily run, both in impact number and magnitude, yet the grade was not so steep that the nature of the task changed. In addition, this duration is a recommended daily exercise duration for these individuals to develop and maintain fitness (1,17). Indeed, we were specifically investigating adaptations to multiple impacts, not those to cardiorespiratory fatigue. This type of running does not elicit cardiorespiratory fatigue. Kinematic and metabolic data were collected during six 5-min intervals (0–5, 5–10, 10–15, 15–20, 20–25, and 25–30 min).

Kinematics

Right sagittal view kinematic data were recorded at 200 Hz with retroreflective markers placed on the skin over the humeral head, greater trochanter, femoral condyle, lateral malleolus, lateral aspect of the calcaneus, and the fifth

TABLE 2. Midsole hardness and surface stiffness for each of the six conditions in Experiment 1 (S40 = soft surface and 40 Shore A midsole; M40 = medium surface and 40 Shore A midsole; H40 = hard surface and 40 Shore A midsole; S70 = soft surface and 70 Shore A midsole; M70 = medium surface and 70 Shore A midsole; H70 = hard surface and 70 Shore A midsole).

Independent Variable	S40	M40	H40	S70	M70	H70
Midsole hardness	40[a]	40[a]	40[a]	70[a]	70[a]	70[a]
Surface stiffness	100[b] (soft)	200[b] (medium)	350[b] (hard)	100[b] (soft)	200[b] (medium)	350[b] (hard)

[a] Shore A scale; [b] $kN \cdot m^{-1}$.
Condition order was balanced across the 12 subjects using a Latin square.

metatarsal. Markers were digitized using a video processor (VP-110, Motion Analysis Corp.) and marker paths were edited and then digitally filtered with a low pass Butterworth filter at 12 Hz. This frequency was determined from a residual analysis (29). Hip, knee, and ankle angles and angular velocities were calculated from the filtered marker paths. The hip angle that is calculated with this method reflects the angle between the upper body and the thigh, rather than the angle between the pelvis and the femur. In experiment 1, mean values of the kinematic variables were obtained from ten random footfalls between minutes 3 and 5 for each subject during each condition. In experiment 2, mean values were obtained for each subject from 10 stance periods collected during each of six 5-min intervals (0–5, 5–10, 10–15, 15–20, 20–25, and 25–30 min). Kinematic variables from the hip, knee and ankle which were used in the statistical analysis were: 1) angle at contact, 2) maximum angle during stance, 3) angle at toe-off, 4) maximum angular velocity, and 5) minimum angular velocity.

Footstrike and toe-off were determined with a pressure sensor under the treadmill. This signal was interfaced to a microcomputer and sampled with a 12-bit A/D converter at 1 kHz. Mean stride frequency and stance time were calculated from the 10 stride cycles for the time periods within each condition in both experiments.

Oxygen Consumption and Heart Rate

Metabolic data were obtained in both experiments from an AeroSport TEEM 100 Metabolic Analysis System that was calibrated according to manufacturer specifications before each data collection session. The oxygen (O_2) and carbon dioxide (CO_2) analyzers were calibrated with known concentrations of gas. A high-flow pneumotach was used for all subjects and was calibrated with a calibrated 3-L syringe. Heart rate was measured with a Vantage Performance telemetry system. Metabolic and heart rate data were sampled every 20 s. Minute values of oxygen consumption ($\dot{V}O_2$) and heart rate were calculated using the average of three, 20 s samples. In experiment 1, the per-minute values at 3, 4, and 5 min were averaged to obtain a grand mean for each subject in each condition and used for statistical analysis. In experiment 2, the per-minute values within each 5-min interval (0–5, 5–10, 10–15, 15–20, 20–25, and 25–30 min) were averaged to obtain six mean values representing each 5-min interval per subject, and these were used in the statistical analysis.

Statistical Analysis

In experiment 1, midsole hardness and surface stiffness were the main effects tested with a two-way repeated-measures ANOVA. In experiment 2, time was the main effect tested with a one-way repeated-measures ANOVA. For both experiments, simple effects and contrasts were tested for kinematic, stride cycle, and oxygen cost variables with a significant effect ($P < 0.05$) using the Tukey pairwise comparisons of means and Bonferroni adjustment procedure in order to control the type 1 error ($P < 0.0033$).

RESULTS

Experiment 1. Stance time and stride cycle time were invariant across conditions. Oxygen consumption was greatest for the soft surface ($P < 0.001$) and decreased with increasing surface stiffness (Fig. 1). There were no significant differences in the heart rate (surface: $P = 0.007$; midsole: $P = 0.681$). Kinematic adaptations to the surface occurred at the hip and knee, whereas adaptations to the midsole were found only at the ankle (Fig. 2). The most striking kinematic differences were seen at the knee. On the hard surface, the hip and knee were at greater extension at foot contact than on the medium or soft surface ($P = 0.0001$ for each joint). Maximum hip flexion was significantly less on the hard surface ($P = 0.0001$), whereas maximum knee flexion did not change ($P = 0.936$). Joint velocity differences were found between the surfaces. Peak angular velocity of the hip, knee, and ankle were greatest on the hard

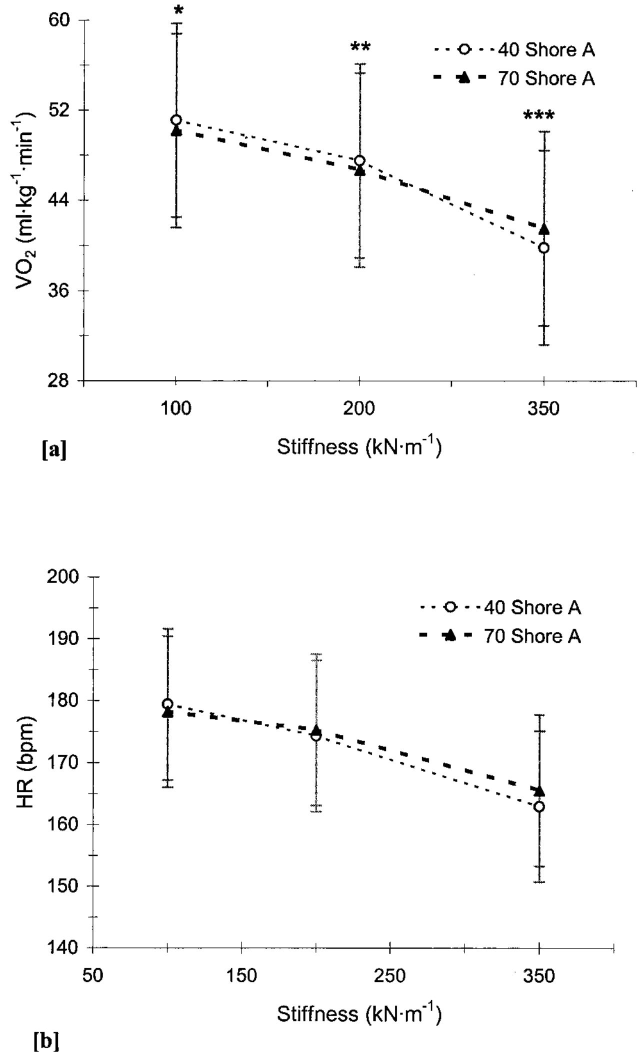

FIGURE 1—Oxygen consumption (a) and heart rate (b) values for the six conditions of midsole hardness and surface stiffness averaged over all subjects and trials in experiment 1 (mean ±SD). Stars (*, **, *) denote significant main effect of surface condition ($P < 0.0033$) with the number of stars corresponding to distinct groups. Oxygen consumption was greatest for the soft surface and decreased with increasing surface stiffness. There were no significant differences in the heart rate.**

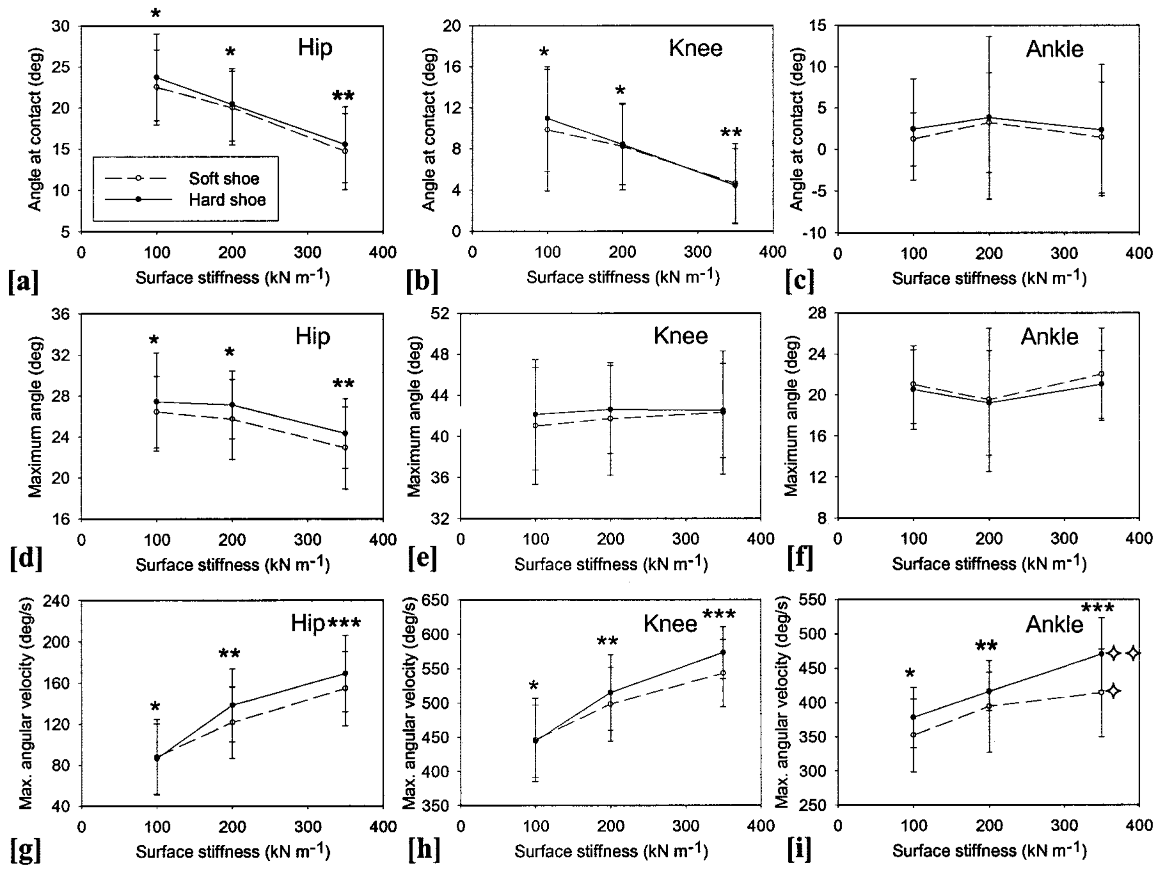

FIGURE 2—Hip, knee, and ankle kinematics at contact ([a], [b], [c]), and maximal flexion and dorsiflexion during stance of the hip, knee, and ankle ([d], [e], [f]) and peak angular flexion and dorsiflexion velocity of the hip, knee, and ankle ([g], [h], [i]) averaged over all subjects and trials in experiment 1 (mean ± SD). Greater flexion (hip and knee) or dorsiflexion (ankle) is represented by increasing positive angles. The ankle was neutral at 0°. Stars (*, **, *) denote significant main effect of surface condition ($P < 0.0033$) with the number of stars corresponding to distinct groups. Diamonds (◇, ◇◇) denote the significant main effect of midsole condition ($P < 0.0033$) with the number of diamonds corresponding to distinct groups. Increased surface stiffness caused greater hip and knee extension at contact ([a], [b]), decreased maximal hip flexion ([d]), and greater peak angular flexion and dorsiflexion velocity of the hip, knee, and ankle ([g], [h], [i]). Maximal knee flexion and ankle dorsiflexion remained the same. Increased midsole hardness independently resulted in greater peak ankle velocity ([i]).**

surface and increased with surface stiffness ($P < 0.001$ for all joints). Increased midsole hardness independently resulted in greater peak ankle velocity ($P = 0.0005$).

Experiment 2. Kinematic adaptations over time occurred at the ankle and hip joints. The hip joint was in a more extended position at contact in the last 15 min versus the first 5 min of running ($P = 0.0017$) (Fig. 3). Ankle adaptations consisted of increased plantarflexion at toe-off after 5 min of running ($P = 0.0000$), and increased peak dorsiflexion and peak plantarflexion velocity ($P = 0.0008$ and $P = 0.0002$) during the last 15 min of running versus the first 5 min. Ankle angle at contact and maximum dorsiflexion during stance were invariant ($P = 0.0121$ and $P = 0.5816$). There were no adaptations in knee kinematics. Stance time and stride cycle time were invariant over the running duration. Oxygen consumption and heart rate gradually increased over the 30 min of running ($P = 0.0000$) (Fig. 4).

DISCUSSION

The primary purpose of this paper was to investigate adaptations in sagittal plane kinematics induced by changes in shoe and surface hardness as well as by prolonged downhill running. The accompanying changes in heart rate and oxygen uptake were also quantified. With increased surface stiffness (experiment 1), the lower extremity was in greater extension in the hip and knee at contact, maximal hip flexion angle decreased, and peak angular velocity increased in all joints (Fig. 2). At the same time, there was a decrease in oxygen uptake with increasing surface hardness (Fig. 1). Shoe hardness only affected the kinematics at the ankle (Fig. 2). With increased duration of downhill running (experiment 2), kinematic adaptations occurred only at the hip and ankle (Fig. 3), whereas oxygen uptake and heart rate both increased significantly over time (Fig. 4). Our intention in presenting results from these two experiments was to find common principles that govern kinematic adaptations during running. We will start with a comparison to other results and theories presented in the current literature, and then proceed to develop a comprehensive understanding based on our own results.

A recent study on the effect of surface compliance on running showed that the metabolic cost was lowest for the most compliant surface (32). This is contrary to our findings. Earlier studies found no effect of surface on energy cost (3,44). These inconsistencies may be caused

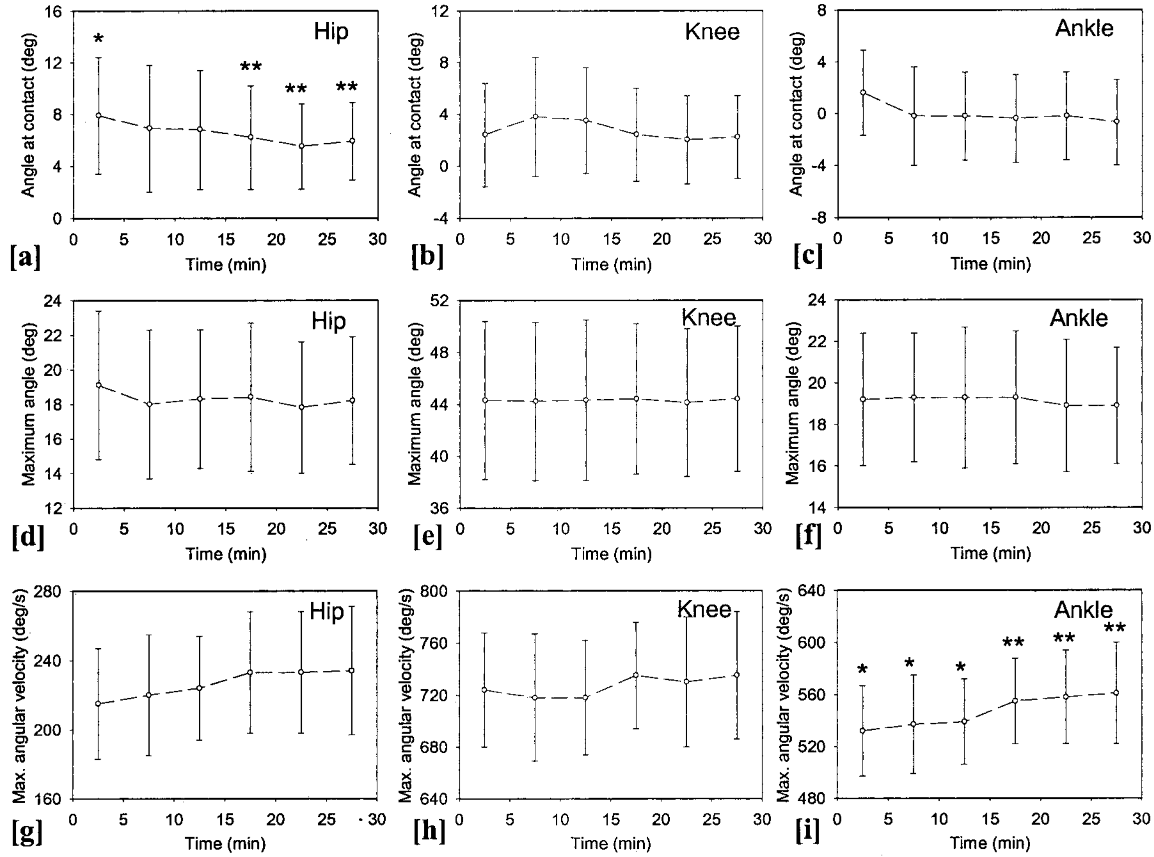

FIGURE 3—Hip, knee and ankle kinematics at contact ([a], [b], [c]), and maximal flexion and dorsiflexion during stance of the hip, knee, and ankle ([d], [e], [f]) and peak angular flexion and dorsiflexion velocity of the hip, knee, and ankle ([g], [h], [i]) averaged over all subjects and trials in experiment 2 (mean ± SD). Stars (*, **) denote significant main effect of surface condition ($P < 0.0033$) with the number of stars corresponding to distinct groups. Greater flexion (hip and knee) or dorsiflexion (ankle) is represented by increasing positive angles. The ankle was neutral at 0°. Hip extension at contact increased during the last 15 min, as did maximal dorsiflexion velocity. Maximal plantarflexion velocity and plantarflexion angle at toe-off also increased significantly (not shown). Knee kinematics remained the same over time.

by differences in surface construction. Our subjects mentioned a sensation of "running on sand" indicating that the surface may have had too much damping or inertia to effectively produce a "rebound" effect as in other surfaces used (32). Two previous studies found that a harder surface resulted in lower leg stiffness (16,32) with leg stiffness being defined as the peak force divided by the length change of the leg. This is consistent with our observation that posture at time of initial contact was more extended on the hard surface, whereas the maximum knee flexion angle remained unchanged.

Harder midsoles have been found to cause an increase in knee flexion velocity (5). We found this result as well, but it was not statistically significant ($P = 0.099$). We did find an increase in dorsiflexion velocity that is similar to observations in subjects with knee pain (46). This increase in flexion velocities can be explained as a passive mechanical response to the earlier rise of the ground reaction force that occurs with increased shoe or surface hardness or, possibly, during knee pain. This mechanism may help regulate impact force magnitude (51). Earlier studies found increased metabolic cost with hard shoes (20). Our findings were similar but once again not statistically significant. This may be because the effect of shoe properties on metabolic cost seems to be more pronounced on harder surfaces (Fig. 1) and our hardest surface was less hard than that generally used for overground laboratory running.

With increased duration of downhill running, we found increased peak dorsiflexion velocity, and no change in knee kinematics as has been previously observed (38). During level running under cardiovascular fatigue, some have observed that the knee angle at contact is more flexed (12), whereas others have found lower-extremity segments to be more vertical at contact (14,50). We found increased extension at the hip joint that, as measured, was reflective of a more vertical position of upper body mass relative to the hip rather than a change in the angle of the pelvis. This upper body position should require less muscle activation for support, which would be a desirable adaptation considering that this type of running produces muscle damage (27). Cardiovascular fatigue did not occur during this downhill run as was evident from the oxygen uptake and heart rate responses at the end of 30 min. These levels were still below the steady-state values found in experiment 1.

Increased ankle, knee, and hip flexion velocities occurred on the harder surfaces. We agree with others that these changes in joint angular velocities are a passive, mechanical response of the system, an uncontrollable response to the impact forces,

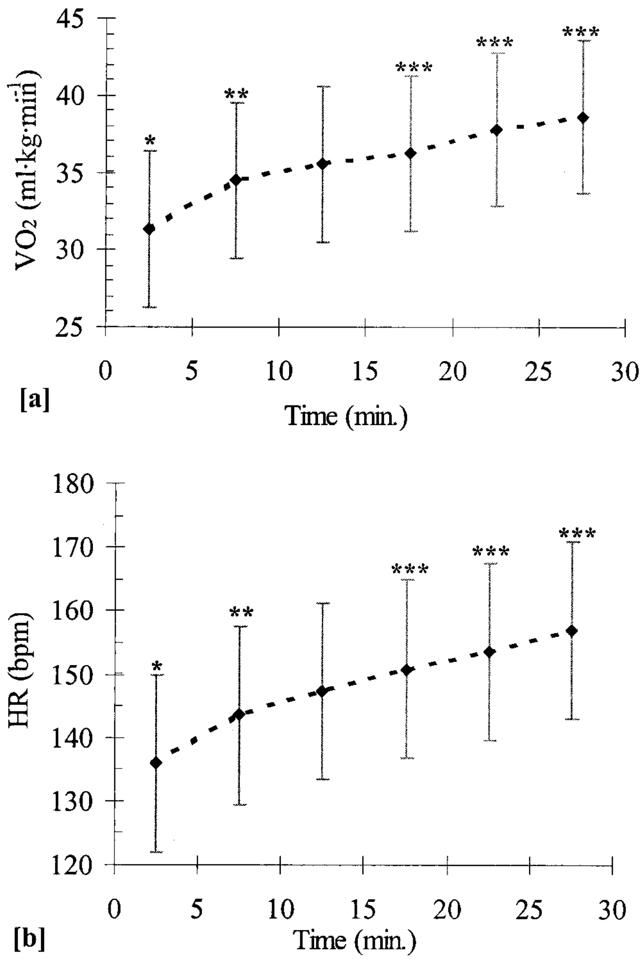

FIGURE 4—Oxygen consumption (a) and heart rate (b) during the 30 min of running of experiment 2 (mean ± SD). Stars (*, **, *) denote significant main effect of surface condition ($P < 0.0033$) with the number of stars corresponding to distinct groups. Oxygen consumption and heart rate gradually increased over the running duration and was statistically less from 5 to 10 min vs 10 to 30 min. These differences represented the typical oxygen cost and heart rate drift for this type of running.**

and the chosen posture at time of contact geometry (13,51). If no adaptation of the initial angles or muscle activation would occur, this would quickly drive the knee to a greater amount of knee flexion. This would cause metabolic cost to rise due to the increase in the knee extensor moment required for weight support and push-off from a more flexed posture (37). We observed that runners avoided this scenario by landing on the hard surface with a more extended knee and hip. As a result, maximum knee flexion was invariant; this variable did not change by more than one degree for any of the conditions studied (Figs. 2[e] and 3[e]).

When considering the implications for impact loading, it was surprising that runners adopted a more extended posture at impact on the harder surface. Such a postural adaptation is thought to increase landing stiffness, or increase effective mass (10). This scenario would increase impact shock (26) as well as vertical ground reaction force upon contact (22). In this particular situation, regulation of metabolic cost therefore appeared to be more important to the runners than regulation of impact shock. This is similar to the finding that runners choose a stride length that minimizes metabolic cost rather than impact shock (24). The kinematic adaptations that occurred during prolonged downhill running (Fig. 3) were less pronounced than those due to surface but those that were found (more extended hip at impact, faster dorsiflexion velocity) were similar to those caused by increased shoe or surface hardness.

Alternatively, kinematic adaptations may be interpreted in terms of leg stiffness using a mass-spring model (11,16,32). If subjects land with a more extended hip and/or knee, this gives them an increased amount of flexion before they reach the same invariant maximum knee angle at midstance. This would then be interpreted as a greater change in leg length and hence a decrease of average leg stiffness. Such a decrease in leg stiffness has been interpreted as a desire to maintain the same combined leg-surface stiffness as surface stiffness increases and has also been observed for single steps and hopping (15,16,36). Kinematic adaptations to achieve this in hopping occur at the ankle (15). Similarly, our results from experiment 2 could indicate that the runners want to maintain the same leg stiffness as running duration increases. Although lower-extremity stiffness was not directly measured in our experiments, our kinematic results are consistent with the idea that leg stiffness and surface stiffness are invariant when combined serially. It is interesting to note that a decrease in *average* leg stiffness, as defined using a mass-spring model (16,32) can occur at the same time as an increase in *instantaneous* leg stiffness when the knee is more extended at impact. The term "leg stiffness" must therefore be carefully defined and not loosely used without a specific definition.

CONCLUSIONS

Based on the results of our two studies, we can conclude that:

1) Kinematic adaptations occurred with changes in midsole hardness, surface stiffness, and over time.

2) The changes in limb posture at the time of impact were interpreted as active adaptations that compensate for passive mechanical effects.

3) The goal of these adaptations appears to be the minimization of metabolic cost, at the expense of increased exposure to impact shock.

Sport and clinical relevance. These observations may be some of the natural adaptations to surface and duration. Although they may be desirable from an energetics point of view, they may contribute to the high risk of overuse injury in distance runners, especially those who run on a hard surface.

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