In Vivo Examination of the Dynamic Properties of the Human Heel Pad

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Abstract


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The shock-absorbing characteristics of the heel pad in vivo were examined in adults (N = 16) and 7-year-old children (N = 5) using a drop-impact tester (wt = 5 kg). Impact velocities were 0.72 m/s and 0.93 m/s. It was found that in adults the average peak deceleration was 11.6 G at an impact velocity of 0.93 m/s. The maximum deformation of the heel pad was 11.3 mm, and the computed energy absorption during impact amounted to 79% (range = 75%–89%). These mechanical characteristics remained nearly the same even after 6 min of repeated impacts by the impact tester and even after a 10 km run. The children had larger values of peak deceleration and maximum deformation and smaller energy loss than the adults. It was concluded that the heel pad was a fairly effective shock attenuator and high energy absorber, and that these characteristics remain nearly unchanged even after a relatively long period of repeated impacts. It was also concluded that the mechanical properties of the children’s heel pads were different from those of the adults.

Key words

Shock absorbency, heel pad, impact test, energy absorption, children

Introduction

The heel pad, which consists of fat and tough fibrous tissue, appears to be a fairly effective shock absorber. Although the locomotor system contains many muscle-tendon-joint regions that absorb kinetic energy during various forms of locomotion, shock absorbing function of the heel pad apparently plays a major role in protecting the lower extremity and foot in particular. It has been reported that insufficient heel pad function capacity can lead to the development of shock-dependent injuries, such as heel pain (14, 18, 19), plantar fasciitis (21) and Achilles tendinitis (9). An understanding of the normal function of the heel pad is therefore of great importance to practitioners in the areas of sports medicine and athletic coaching, and knowledge of the material property of biological tissue, particularly the heel pad, is also useful in the design of safe walking and running footwear. Cavanagh et al. (4) reported that in modelling shoe/foot interaction in running, a knowledge of various constants for the foot is essential.

The results of studies to determine the normal function of heel pad shock absorbency (2, 4, 7, 10, 11, 23) have not always been consistent. Bennett and Ker (2) studied the dynamic compression properties of human heel pads in vitro using an Instron material-testing machine. They found that the pads were highly resilient and could absorb only about 30% of the compressive energy. A similar result was reported by Andersen et al. (1). In contrast, Jørgensen and Bøjesen-Møller (11) conducted an in vitro experimental study of the mechanical properties of the human heel pad by dropping an accelerometer-mounted missile onto isolated subcalcaneal fat tissue and found that the heel pads absorbed 84% of the input energy.

Several in vivo experiments have been reported. Cavanagh et al. (4) used a ballistic pendulum to impact against the heels. At the impact velocity of 1.03 m/s and 1.44 m/s, the heel deformed around 9 mm and 11 mm, respectively. The energy absorption values were 85% and 90%, respectively. Denoth (7) used a force platform to measure the force acting on the heels during impact. An accelerometer was mounted on the epicondylus lateralis to obtain the acceleration information during impact, which was then integrated twice to obtain the vertical displacement of the heel. The deformation of the heel ranged from 4 mm to 10 mm for forces between single and double body weight. Thus, the values of energy dissipation by the heel pad reported by different researchers vary; those found in some in vivo experiments are nearly 3 times greater than those in in vitro experiments, and the deformation of the heel found in in vivo experiments shows a relatively large variation.

The purpose of this study was to determine the mechanical properties of the normally functioning heel pad in vivo in adults and children. In addition, the effect of successive impacts on the heel pad shock absorbency was investigated. Some methodological improvements in comparison to previous studies were used. A free-fall impact tester, which has often been used in testing the cushioning properties of running shoes (3, 5, 8, 12, 13), was introduced, because it could simulate the foot/ground impact conditions better than the pendulum method (4, 23). Displacement of the impactor was measured.

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directly using a position transducer, while a double integration method of acceleration data was used in the previous studies (4, 7, 23).

**Methods**

**Subjects**

The physical characteristics of the 9 females and 7 males (age range, 19 to 35 years old) and the five 7-year-old boys who volunteered as subjects for this study are shown in Table 1. Each subject signed an informed consent statement approved by the University. The percentage fat was estimated from measurement of skinfold thicknesses using a skinfold caliper (17). Although the subjects represented a relatively wide range of body composition characteristics in terms of the amount of body fat, all of them were considered to be within the normal range (16).

<table>
<thead>
<tr>
<th>N</th>
<th>Height (cm)</th>
<th>Weight (kg)</th>
<th>Skinfold thickness (mm)</th>
<th>Estimated Percentage fat (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Adult Females</td>
<td>159.9 ± 4.0</td>
<td>59.6 ± 4.6</td>
<td>17.2 ± 3.1</td>
<td>23.7 ± 3.6</td>
</tr>
<tr>
<td>Adult Males</td>
<td>171.2 ± 6.2</td>
<td>65.4 ± 10.1</td>
<td>14.4 ± 4.7</td>
<td>17.8 ± 4.5</td>
</tr>
<tr>
<td>Boys</td>
<td>121.6 ± 2.8</td>
<td>22.1 ± 3.9</td>
<td>8.5 ± 2.4</td>
<td>16.2 ± 4.0</td>
</tr>
</tbody>
</table>

**Impact tester**

An impact tester was designed and constructed by the investigators based on the structure of a conventional free-fall impact testing machine used to test running shoes (Fig. 1). It consisted of an upper rigid frame constructed of steel bars (40 × 40 mm), a weighted stainless steel impact shaft (diameter: 10 mm; length: 70 mm), a lightweight uniaxial accelerometer (Kyowa Ltd.: Type AS50B), and a position transducer (Kyowa Ltd.: DLT 50A). The highest response frequency for the accelerometer was 1 kHz, and that for the position transducer 5 kHz. A steel heel contact head with a flat lower surface (diameter: 40 mm) and additional weight were connected to the bottom of the shaft. The mass of the shaft together with the weights totalled 5 kg. Two low-friction slide bearings were employed to guide the steel shaft so that it dropped vertically onto the heel. A mechanical releaser, which was also fixed to the upper part of the frame, was used by the experimenter to initiate the drop of the impact shaft. The heel was impacted by the shaft through a hole approximately 60 mm in diameter which had been drilled in a 30 mm-thick plywood board attached horizontally to the middle of the frame. The foot contact side of this hole was larger (approx. 90 mm in diameter) so that laterally deformed fat tissue did not lock up during impact between the borders of the hole and the impact head. A wooden block (50 × 50 × 40 mm) was affixed on the lower side of the board at the posterior portion of the heel. Nylon strap belts were used to fix the foot in position and to restrict medio-lateral motion. To adjust for the length of the leg of each subject, concrete blocks were used, and for fine height adjustment, steel plates were inserted in the space between the knee and the plywood board. After adjustment, the femoral condyles of the knee could be firmly pressed against the plate.

The output signals from both the accelerometer and the position transducer were sampled at 2 kHz by an A/D converter and stored in a computer (NEC LTD: PC 9801-RA). The signal from the position transducer was filtered at 300 Hz prior to A/D conversion.

**Impact velocity and mass selection**

To accurately simulate actual heel ground contact, the impact velocity and mass values of the impact shaft, which determine impacting energy, must be similar to those found during running. Nigg (20) reported that the touch-down velocity of the heel in 5 runners wearing 5 different types of running shoes ranged from 0.8 m/s and 1.2 m/s at 3.5 m/s running velocity. Theoretically, at the drop distance of 50 mm, the impact velocity is 1 m/s, and for this reason a drop height of 50 mm was chosen. The average vertical velocity of the shaft for the 5 ms prior to heel contact was computed from derivative of the displacement curve to be 0.93 m/s. This velocity is lower than the theoretical value due to the effect of the mechanical releaser, which slightly lowered the shaft at the moment of release, and the slight effect of friction between the shaft and slide bearings. The drop height of 30 mm (the impact velocity, 0.72 m/s) was also employed to investigate the effect of slower impact velocity as well as to compare children with adults. It is possible that touch-down velocity in barefoot condition is slower than that in shod condition.

The amount of mass at heel contact moment during running is difficult to estimate because it is affected by numerous factors (23). Using peak ground reaction forces measured during running and drop impact testing, researchers have estimated the mass at heel strike to be around 7 kg (3). Consequently, in the testing of shoe sole material cushioning capacity, the impact mass of 7 kg has been commonly used (5, 8, 12). The impact shaft used in this study weighed 5 kg. During preliminary testing, a 7.5 kg impact mass was tested. However, the majority of the subjects complained of pain in the heel and/or the area around the femoral epicondyles. Cavanagh et al. (4), who used a ballistic pendulum to strike the heel, also found that the upper limit of subject tolerance of the impact was at a lower force level than that during actual running. They considered this to indicate that the pressure distribution conditions with a flat plate were not equivalent to those with a shoe.
Experimental procedure

Each subject lay on his/her stomach with the right knee flexed at a right angle to the floor. The ankle was slightly dorsiflexed at 5 to 10 degrees more than a right angle. After the ankle and foot were secured, a small pillow was placed under the right greater trochanter so that there was slight flexion of the hip, which allowed the femoral condyles to directly contact the floor.

Each adult subject performed several practice impacts from both 30 and 50 mm drop heights prior to data collection. Data were collected for 5 impacts at 30 mm drop height followed by 5 impacts at 50 mm drop height. The children were tested at only the 30 mm drop height.

In three subjects, following several practice impact trials, the right heel was impacted at the rate of once every 10 s for 6 min at 50 mm drop height. For each parameter, the average value for each subject for the initial 5 trials was compared with that for the last 5 trials.

In a further investigation of the effect of repeated impacts, the mechanical properties of the heel were examined before and after long running distance in two male subjects who were long-distance heel-toe runners. Impact tests were performed a few minutes before and after each subject ran 10 km in 36 min while wearing their own jogging shoes.

Cushioning property test for an ordinary running shoe

Using the same impact tester, the mechanical properties of the heel portion in a commercially available unused running shoe were examined. It had a 25-mm-thick ethyl vinyl acetate (EVA) foam midsole, 7 mm-thick rubber sponge outer sole, and 4 mm-thick soft EVA foam inner sole. Impacting was performed at the center of the heel portion 5 times from the two different heights.

Parameters evaluated

Fig. 2A shows representative output signals from the accelerometer and position transducer. Prior to the impact moment, the deceleration signal is nearly constant at 0 G. The small fluctuations of the signal represent the slight vibration of the impact shaft. After impact, the signal increased in a curvilinear fashion, reached the peak, then decreased in a similar curvilinear fashion to 0 G. The displacement signal showed curvilinear increase from the moment of release to heel impact moment, while just prior to the impact it showed a nearly linear path. After impact, the linear increase persisted for about 5 ms, but with an increase of deformation, the increase became less marked, showing a curvilinear path. It reached the peak slightly after the deceleration peak moment. The displacement then decreased slowly. During each trial, the four parametric values defined in Fig. 2A (P1, T1, P2, and T2) were obtained. The deformation of the heel was computed from the displacement data during the impact between the impact moment (Fig. 2A: dotted line) and zero deceleration moment (Fig. A: dashed line). The computed force and deformation of the heel pad were used to plot the force-deformation curve (Fig. 2B). The force-deformation curve typically showed curvilinear increase from zero deformation at zero force level to peak deformation at near peak force level, and curvilinear decrease from the peak deformation to residual deformation at zero force level. The residual deformation even when the load is removed indicates the visco-elastic nature of the heel pad tissue.

Test calibration and reliability of testing method

In order to determine the effect of friction upon energy absorption, the apparatus was tested by impacting a steel spring from various drop heights. The energy loss values ranged from 2% to 3%, suggesting the effect of friction was fairly small. Some energy loss was also believed to be due to the lack of perfect elastic body of the spring.

In order to evaluate the reliability of the testing method, a test-retest method was used. The test-retest correlation values for the 5 parameters obtained in 11 subjects ranged from 0.86 to 0.93, confirming fairly high reproducibility of the method.

Statistical Analyses

The mean values for P1, T1, P2, T2 and energy absorption were computed from the average data for the 5 impact trials for each subject under each impact velocity condition. Separate means and standard deviations were computed for the adults and children. T-test was used to examine differences between the means. The level of significance used was 5%.
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Results

Results for adults

Peak deceleration and peak deformation

Figs. 3A–C show deceleration and displacement curves for three subjects at each of the 0.72 m/s (dashed line) and 0.93 m/s (solid line) impact velocities. The displacement curves showed a nearly linear increase for both drop velocities, but the slope for the latter was slightly steeper than that for the former. The deceleration curves for most subjects showed a unimodal curve pattern (Figs. 3A and B). In several subjects, however, a small secondary peak (Fig. 3C) was evident. Table 2 presents average maximum, and minimum and S.D. values in all adult subjects for each parameter. There was no significant difference between females and males for any parameter. As impact velocity increased, P1 increased and T1 decreased. Comparison of T1 and T2 values indicated that the heel continued to deform even after peak deceleration, as was shown in Fig. 3.

Force-deformation relationship curve and energy loss

Figs. 4A and B show average force-deformation relationship curves for 2 adults. Fig. 4A shows the results for a subject who had average energy values of 76.8% and 78.0% at the impact velocities of 0.72 m/s and 0.93 m/s, respectively, and Fig. 4B shows that for the subject who had the largest energy absorption values (87.6% and 89.4% at 0.72 m/s and 0.93 m/s, respectively). Regardless of the value of energy loss, observation of the force-deformation paths in all subjects suggested that they generally shared some common features. The change of force in relation to the magnitude of heel deformation was best described by a curvilinear path. The force increase after impact was characterized by a small increase in force with a large deformation. After the heel was deformed by approximately 5 mm, the rate of increase in force accelerated.
Impact velocity = 0.72 m/s

<table>
<thead>
<tr>
<th>P1: Peak deceleration (G)</th>
<th>T1: Time to peak deceleration (ms)</th>
<th>T2: Time to peak deformation (ms)</th>
<th>Energy loss (absorption) (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>AV</td>
<td>9.9</td>
<td>13.5</td>
<td>8.3</td>
</tr>
<tr>
<td>SD</td>
<td>0.8</td>
<td>0.6</td>
<td>0.8</td>
</tr>
<tr>
<td>Max</td>
<td>10.6</td>
<td>14.1</td>
<td>9.2</td>
</tr>
<tr>
<td>Min</td>
<td>8.9</td>
<td>11.0</td>
<td>7.6</td>
</tr>
</tbody>
</table>

Table 3 Summary of the results for children.

<table>
<thead>
<tr>
<th>Drop height (mm)</th>
<th>Impact velocity (m/s)</th>
<th>P1: Peak deceleration (G)</th>
<th>T1: Time to peak deceleration (ms)</th>
<th>P2: Peak deformation (mm)</th>
<th>Energy loss (Absorption) (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>30</td>
<td>0.72</td>
<td>10.9</td>
<td>15.6</td>
<td>9.3</td>
<td>53.3</td>
</tr>
<tr>
<td>50</td>
<td>0.98</td>
<td>13.8</td>
<td>14.0</td>
<td>11.5</td>
<td>53.7</td>
</tr>
</tbody>
</table>

Table 4 Summary of the testing results for the running shoe sole.

and the slope of the force-deformation path was steeper than the initial slope. The force peak followed this acceleration. The heel continued to deform even after the force peak. The pattern of force decrease in relation to the changes in deformation was approximately the reverse of that during the force increase phase; during the initial force decrease phase, the force decrease was relatively large compared with that of the deformation, but as the force magnitude decreased, the deformation decrease accelerated. At the moment of zero force, the heel was still deformed by around 6 mm.

The amount of energy absorbed by the heel pad was computed from the data for each trial. The average energy absorption for all adults at impact velocities of 0.72 m/s and 0.93 m/s were 77.4% and 78.8%, respectively (Table 2). Therefore, the value of energy absorption at 0.93 m/s impact velocity was 1.4% larger than that at 0.72 m/s impact velocity. There was relatively large inter-subject variability (range) in energy absorption values.

Comparison of results for children and adults

A summary of the results for children is presented in Table 3. The average peak deceleration (9.9±0.8 G) was significantly larger than that of the adult subjects (8.7±1.2 G) at the same impact velocity (0.72 m/s). Although T1 for the (13.5±0.6 ms) was shorter than that for the adults (14.8±2.0 ms), this difference was not significant. The average P2 (8.3±0.8 mm) and T2 (16.0±1.8 ms) in the children were significantly lower than those for the adults (9.5±1.0 ms and 19.0±2.1 ms, respectively). The energy absorption in the children (73.9±2.9%) was significantly smaller than that in the adults (77.4±3.5%).

Results for repeated impacts

Fig. 5A shows P1, P2, and energy absorption values for repeated impacts for the three subjects using the impact tester. Although the average values during the initial 5 and final 5 impacts were slightly different, this difference was not significant in any subject. P1 and energy absorption values before and after the 10-km run are shown in Fig. 5B. Neither of the two subjects showed a significant pre- and post-run difference for either parameter.
Dynamic properties of the running shoe sole

Table 4 is a summary of the results for running shoe sole tested using the same apparatus. The P1 values at 0.72 and 0.93 m/s impact velocity (10.9 G and 13.8 G, respectively) were larger than those for the adult heel pad (8.7 G and 11.6 G, respectively), while P2 at each velocity was almost the same. T1 values (15.6 ms and 14.0 ms) were longer than those of the heel pad (14.8 ms and 12.5 ms). The amount of energy absorbed was around 53%, which was apparently smaller than that for the heel pad.

Discussion

Shock absorbency of heel pads in adults

At 0.93 m/s impact velocity, the peak deceleration (11.6 G) for the adult subjects was 84% of the value for the shoe sole (13.8 G), suggesting that the heel pad is an even better shock attenuator than the running shoe sole. This finding is consistent with that of Jørgensen and Bojsen-Moller (11), who reported that the peak force value for the heel pad was 89% of that for shoe sole material.

From the product of peak deceleration and the mass of the impact shaft, the peak force was estimated to be 569 N, which is equivalent to the average body weight of the subjects. For this body weight force, the average maximum deformation in all subjects was 11.3 mm. Cavanagh et al. (4) reported maximum deformations of 8.8 and 10.9 mm when the applied forces were 338 and 676 N, respectively. Denoth (7) reported that the deformation ranged from 4 to 10 mm for forces between single and double body weight. Recently, using X-ray film, De Clercq and Kunn (6) found that the deformation of the heel pad during running was 0.4 ± 0.7 mm under barefoot conditions. Compared with these results, our subjects showed slightly larger average deformation, despite the smaller magnitude of the applied force. One reason for this discrepancy may have been the difference in the area of the impact surface. The ballistic pendulum used by Cavanagh et al. (5) had a surface area of 60.8 cm² as compared to the surface area (diameter, 4 cm) used in this study. It is conceivable that an impact head with a smaller surface area would penetrate more deeply. In addition, the impact site used in this study was a central region of the heel pad, where the fat tissue was thickest. This could also have permitted a deeper penetration. Another reason for the larger deformation may be the deformation of the tissue above the femoral epicondyles. All of our subjects reported that they felt some compressive force between the femoral epicondyles and the floor after impact. In fact, deformation reported by Cavanagh et al. (5) using the pendulum method, where the knee was also placed against a rigid wall and the applied force was relatively small, was larger than that reported by Denoth (7), in which study the heelstrike was performed by movement of the leg against a rigid floor.

The energy absorbed by the heels during impact in the adult subjects in this study was 79% (range: 75%–89%) at the higher impact velocity and 77% (range: 73%–88%) at the lower velocity. The findings that the heel pad has a relatively large energy-absorbing capacity and that the impact velocity changes the magnitude of energy absorbed are consistent with those of other researchers. For instance, Cavanagh et al. (4) found that the value of energy absorption changed from 85% to 90% when the velocity of impact was changed from 1.0 m/s to 1.44 m/s. Valiant and Cavanagh (23) reported that the energy absorption ranged from 84% to 99% at impact velocities between 0.8 and 1.2 m/s. However, Bennett and Ker (2), who examined isolated cadaver heel pads using an Instron machine, found that the pads absorbed only 30% of the compressive energy. With respect to the large discrepancy, they commented that their tests represented the properties, energy dissipation in particular, of only the fat pads, whereas the pendulum impact results might reflect the properties of the entire lower leg. This assumption may not be correct. In a recent study by Jørgensen and Bojsen-Moller (11), who used a drop-impact apparatus in 10 isolated heel pads, the average energy loss was estimated to be 84% for the measured peak force of 772 N. In their study, an accelerometer-mounted 1.6 kg missile with an impact area of 9 cm² was dropped on the specimens at 1.4 m/s impact velocity. The major reason for the large discrepancy in the above results seems to be methodological. In the Instron method, the compression head is displaced in a predetermined sinusoidal mode, and thus the head displacement trajectories for the loading (compression) and unloading (decompression) phases are nearly the same. However, in the free-fall drop-impacting method, as shown in our Fig 3, the slope of the displacement trajectory during the loading phase is much steeper than that during the unloading phase. De Clercq et al. (6), who recently observed the deformation of the heel using X-ray film during running, found a displacement trajectory similar to that in the present study. Indeed, the peak deformation, which occurred around 16 ms after the onset of impact in this study, was also demonstrated at nearly the same point in the deformation time curve of De Clercq et al. In summary, our results support the idea that the heel pad is a highly effective energy absorber, through which a considerable proportion of the kinetic energy at heelstrike is dissipated. This would protect the heel bones and lower extremity at the initiation of the support phase during locomotion. Patients who lose the calcaneal fat pad complain of pain and burning under the heel after prolonged walking or standing, and bony irregularity or periostitis about the calcaneal tuberosity is commonly seen on radiographs (19). According to Jørgensen and Bojsen-Moller (11), the low recoiling capacity of the heel pad is also an important mechanism which allows the foot to stay on the ground and secure grip during locomotion.

Some differences in magnitude were observed even among the results obtained using the impact method. The mean values of energy absorption in this study were 77 and 79% for peak forces of 437 and 569 N. These values are approximately 10% smaller than those obtained by Cavanagh et al. (4) using the pendulum method. Jørgensen and Bojsen-Moller (11) reported a mean value of energy loss of 84% for a force of 772 N. The difference between our results and those of the two studies cited above might be a consequence of methodological differences. For instance, the area of contact in both the present study and that of Jørgensen and Bojsen-Moller was about one-fourth of that in the study by Cavanagh et al. With the larger contact area, a larger amount of fat tissue may have contributed to the process of energy dissipation, resulting in a larger energy loss. In addition, the mass of the impact shaft and impacting velocity used differed. Although the calculated force may be approximately equivalent in these studies, both the impact mass and its deceleration values were apparently different. To examine whether different combinations of mass and impact velocity which produce the same input energy change the energy absorption value, we tested the effect of a 7.5 kg mass impacting...
at velocities below 0.93 m/s. Impact velocity of around 0.75 m/s produced a peak force value similar to that for 5 kg mass at 0.93 m/s velocity. A comparison of the two conditions revealed that the energy loss values were about the same. Cavanagh et al. (4) also earlier reported that various combinations of mass and velocity which produced the same input energy did not change the force deformation curves.

Another factor contributing to discrepancy may be racial differences in the thickness of subcutaneous elastic adipose tissue. It is known that a thinner heel pad has lower energy absorption capacity (10), and that average body weight and height as well as percentage fat in the Japanese population are about 5% less than those in the Western population (15, 16). Our subjects may have had a thinner heel pad than those in the previous studies. Variation in fat and connective tissue thickness may also partially account for the wide range of measured energy absorption (75%–89%) in this study as well as that (84%–99%) in the study by Vallant and Cavanagh (23).

Shock absorbency of heel pads in children

The shock-absorbing capacity of the heel pads was found to be lower in children than in adults. The peak deceleration value for the children was 114% of the adult value. Thus, a larger amount of force would result from the same amount of impulsive force applied to the heel in children. The amount of energy absorbed was 95% of the value in adults, indicating that the juvenile heel pad is a less efficient shock absorber. This could be related to the thickness or underdevelopment of the heel pad in children. Skinfold thickness measurements suggested that the children had less fat than the adults. Jorgensen et al. (10), who investigated the relationship between heel pad thickness and shock absorbency in isolated cadaver heel pads, reported that the thinnest pad had the lowest shock absorption and that shock absorbency was decreased in older individuals. These results apparently reflect biological changes in the material properties of the unique fibrous structure and elastic adipose tissue of the heel, due to both development and degeneration.

Although differences were found between the heel pad function of adults and children in terms of shock absorbency, this finding was based solely upon the results for the 30 mm drop height condition. It is possible that the path of normal loading in children differs from that of adults since parameters such as body weight, effective mass involved at touchdown, muscle function, and muscle use differ. Thus, the shock absorbency of the heel pad of children may not necessarily be inferior to that of adults; the juvenile heel pad may rather conform to the motion of the juvenile.

Shock absorbency after repeated impacts

Jorgensen and Bojsen-Moller (11) reported that after in vitro heel pads had been subjected to pounding, which presumably ruptured some of the connective tissue septa and opened their fat chambers, the shock force attenuation capacity was significantly decreased, while no significant changes were found in energy loss. Although they did not indicate the specific duration of pounding, their results do show that the mechanical properties of the heel pad are changed after repeated impacts. In the present study, the mechanical properties were found to be nearly unchanged after 60 successive impacts (6 min) and after running 10 km (36 min). Cavanagh et al. (4), after 25 successive impacts (75 s) on the heel in vivo, found no significant difference between the 1st and 25th impacts in mechanical properties. The present results indicate that the heel pad in vivo is functionally durable after at least half an hour of impacting during distance running.

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