Influence of Different Footwear on Force of Landing During Running

Roy TH Cheung, Gabriel YF Ng

**Background and Purpose.** Previous studies have demonstrated an increase in foot pronation with mileage in runners. Motion control footwear was designed to check excessive foot motions, but its clinical efficacy, especially in terms of pedographic analysis, has not been well reported. The purposes of this study were to investigate the changes in plantar force in people when running with motion control shoes and to compare pedographic measurements obtained in 2 footwear testing conditions (wearing motion control shoes and wearing neutral shoes) at the beginning and end of a 1.5-km running session.

**Subjects.** Twenty-five recreational runners who had ≥6 degrees of foot pronation participated in the study.

**Methods.** An insole sensor was used to register the plantar force of the subjects before and after running 1.5 km in different shoe conditions.

**Results.** There was no change in the magnitude and distribution pattern of plantar force with the motion control shoes after the 1.5-km run. With the neutral shoes, however, there was an increase in mean peak force under the medial midfoot (364–418 N, 15% increase) and first metatarsal head (524–565 N, 8% increase) toward the end of the running bout.

**Discussion and Conclusion.** The plantar force on the medial foot structures increased with mileage of running with neutral shoes but not with motion control shoes. This finding has implications for injury prevention with footwear selection for recreational runners who have more than 6 degrees of foot pronation.
Footwear and Plantar Forces During Running

It has been reported that running patterns changed when mileage was increased and, in particular, that foot pronation increased during landing.\(^1\)\(^-\)\(^4\) Some authors\(^5\) speculated that, with less active eccentric control of the foot invertor muscles, runners would be at higher risk for injuries due to decreased functioning of normal foot pronation to accommodate the impact loading.

According to a recent paradigm of muscle tuning,\(^6\) the impact loading in running is an input signal to the body that initiates vibrations of the soft tissue compartments of the leg. These vibrations are heavily damped, and this paradigm suggests that the muscles adapt to different input signals to minimize the vibrations. Because the foot invertors are the major rear-foot stabilizing muscles to check pronation, if they become less efficient in controlling foot pronation while running, this loss of efficiency would lead to higher plantar forces on the medial structure.

The “motion control” design is a common technology used in running shoes that aims to check excessive foot pronation so as to prevent pronation-related injuries. The most popular mechanism of motion control design is using duo-materials that have different deformation rates in the midsole of the shoes.\(^7\)\(^-\)\(^11\) At the touchdown phase of a gait cycle, the foot lands on the lateral aspect of the heel; a softer lateral flare of the shoes would decelerate the pronation movements of the rear foot by giving additional cushioning against the impact, whereas a firmer medial midsole could stop the foot from excessive pronation.\(^4\)\(^,\)\(^7\)

However, the efficacy of motion control technology is controversial. Various investigators\(^8\)\(^,\)\(^12\)\(^-\)\(^18\) have reported different results for rear-foot pronation with use of motion control shoes. Clarke et al\(^8\) reported successful control of excessive rear-foot movement by use of footwear with motion control features. Perry and Lafontune\(^1\)\(^4\) echoed this finding with use of a different wedging angle of the footwear. However, these studies adapted a 2-dimensional analysis of foot motion, which might induce projection error. The studies by Nigg and Morlock\(^13\) and Stacoff and colleagues,\(^15\)\(^-\)\(^17\) however, did not come to similar conclusion. Stacoff et al,\(^15\) using bone pin markers and 3-dimensional analysis, demonstrated that movements of the rear foot were not significantly affected by different heel-flare designs. Stacoff et al,\(^15\) however, tested only 5 subjects, and this small subject sample might be the main reason for their nonsignificant results.

These controversies also could be due to the incomparable methods among the studies. None of the previous studies provided information on the foot type screening of the subjects. Heterogeneity in foot type could result in various foot motion patterns, thus rendering the results inconclusive. Short running distances and a limited number of trials were other common limitations that may lead to a statistical type II error.\(^19\)

In a recent study,\(^1\) we found that foot pronation was effectively controlled during a 1.5-km running bout when runners used motion control footwear. However, the results of that study focused only on the kinematic data and not on the plantar force distribution. Planter force evaluation with pedography is used to determine specific loading characteristics at the sole of the foot, which reflects the amount of stress acting on the foot structures in real time, and this information may provide clues for identification of pathological running patterns that may be useful for injury prevention.\(^20\) Compared with traditional forceplate measurement, one significant advantage of planter force evaluation is that multiple steps can be collected more easily to simulate the actual sequence of locomotion with a less robust analysis method.\(^21\) Therefore, the application of planter force measurement is more popular among researchers for investigation of the shoe-foot interface.

A study by Willems et al\(^22\) showed that runners with exercise-related lower-leg pain had higher plantar pressure over the medial foot structures than over the lateral foot structures. Another study\(^23\) also demonstrated a positive relationship between metatarsal stress fractures in runners with increased plantar loading on the toes. Planter pressure measurement is commonly used to evaluate the efficacy of orthoses in patients with diabetes mellitus or rheumatoid arthritis.\(^24,\)\(^25\) However, its application for evaluation of sports footwear is less common. If there are differences in pedographic measurements when runners use different footwear, these differences will shed light on footwear selection for preventing running-related injuries among runners who are at high risk for such injuries.

The literature contains little information on the efficacy of motion control shoes and particularly on their effects on the plantar force distribution pattern during landing. Therefore, the objective of this study was to compare pedographic measurements before and after a moderate-distance running session in 2 footwear testing conditions—wearing motion control shoes and wearing neutral shoes—in recreational runners.

Method

Subjects

Twenty-eight female recreational runners with clinical signs of overpronating feet were invited to participate in the study. After the first as-

---

May 2008  Volume 88  Number 5  Physical Therapy  621
Footwear and Plantar Forces During Running

Table 1. Subject Demographic Data

<table>
<thead>
<tr>
<th>Variable</th>
<th>$\bar{x}$</th>
<th>SD</th>
</tr>
</thead>
<tbody>
<tr>
<td>Age (y)</td>
<td>23.6</td>
<td>6.8</td>
</tr>
<tr>
<td>Running experience (y)</td>
<td>3.6</td>
<td>3.2</td>
</tr>
<tr>
<td>Average running distance per week (km)</td>
<td>2.1</td>
<td>1.2</td>
</tr>
<tr>
<td>Average frequency of running per week</td>
<td>2.6</td>
<td>1.4</td>
</tr>
<tr>
<td>Height (m)</td>
<td>1.55</td>
<td>0.07</td>
</tr>
<tr>
<td>Weight (kg)</td>
<td>46.3</td>
<td>4.0</td>
</tr>
<tr>
<td>Body mass index</td>
<td>19.3</td>
<td>1.4</td>
</tr>
</tbody>
</table>

Assessment, 3 subjects were excluded because their rear-foot pronation fell short of the required range. The mean age of the remaining 25 subjects was 23.5 years (SD=6.8). Other demographic data about the subjects are presented in Table 1. Each subject had regularly run at least once per week for no less than 1 year prior to the study. None of the subjects were supported by any running project. They also did not participate in other recreational sports. Their average running distance per week was only 2.1 km (SD=1.2), which indicated that they were at most recreational runners.

Professional or higher-level runners were excluded because they are highly trained and usually run much longer distances than the test distance of this study. In addition, a previous epidemiological study revealed that these runners were less prone to foot pronation-associated injuries than amateur runners. Only female subjects were involved in this study because of potential gender differences in the movement pattern of the lower extremity during running, which can be used in running shoe design to control the effects of differences in shoe construction between running shoes for men and women. Subjects with injuries to their lower limbs that required medical treatment within 6 months of the beginning of the study were excluded. The subject pool comprised only runners who were free from any known musculoskeletal or cardiopulmonary problems and recreational runners with clinical signs of potential foot pronation problems, as evaluated by kinematic measurement (VICON hardware model, V-370, and Workstation 4.0 software®). Furthermore, by revealing the subjects’ running pattern using an in-sole sensor, “non-heel-strikers” (those who land on their midfoot or forefoot rather than their heel) were excluded.

Testing Procedure

All subjects gave their written informed consent before being tested. Two types of standardized footwear—motion control shoes (Adidas Supernova Control) and neutral shoes (Adidas Supernova Cushion)—were provided to all subjects. The neutral shoes were designed to reduce the impact loading rate, but they have no known effects on foot motion control, and there was no reported difference in the kinematic profiles between barefoot running and running with neutral shoes. According to the manufacturer’s information, the construction of these 2 models was similar except for the midsole material. For the Supernova Control shoes, 2 levels of hardness were introduced on the lateral and medial parts of the heel to control rear-foot motion. In the Supernova Cushion shoes, the midsole material consists of a single level of hardness. Otherwise, the 2 models are comparable, with similar designs for the heel counter and upper sole.

To ensure that data were collected under similar shoe fitting conditions, shoes of the same size were provided to each subject during both testing conditions. The same pair of shoes, which had color markers to guide the positions of knots, were used in the 2 shoe models. This helped to standardize the shoe fitting in both testing conditions. We used treadmill running in this study because it was more stable and easy to control than running over ground. Even though minor differences such as shorter stride length, higher cadence, and a shorter nonsupport phase existed in treadmill running, these differences could be minimized with practice. A speed of 2.78 m/s (10 km/hr) was chosen for this study to simulate the speed at which the subjects would normally run. As the subjects recruited in this study used to run on treadmills in fitness clubs and the testing speed was similar to their usual running speed, the adaptation period would be relatively short. Because there was no reported literature describing the adaptation time required, self-reports from the subjects and hand-off from the treadmill railing were regarded as signs of adaptation. Although there was no actual measurement for the length of this period, it was typically well within 1 minute according to our observations of the subjects.

Kinematic Measurements

The detailed procedure for obtaining kinematic measurements has been reported by Cheung and Ng. In brief, a VICON 3-dimensional motion analysis system with 3 cameras...
was used to capture the subjects’ rear-foot movements during running. After standard calibration procedures, the motion was filmed at 60 Hz. Although a higher sampling frequency was preferred, this rate was regarded as acceptable in studies of running at slow to moderate speeds.\textsuperscript{32} Four light-reflective spherical markers were attached on the posterior border of shoe and calf in accordance with the methods of previous studies.\textsuperscript{13–15} Two vectors were formed by 4 markers, and the rear-foot angle was defined as the acute intercept angle between these 2 vectors. Although this kinematic measurement may include an error of translation movement between the shoe and the foot, this measurement demonstrated a similar motion pattern when compared with skeletal motion measured by bone markers.\textsuperscript{33} To ensure the repeatability of the study, the positions of the reflective markers were recorded manually with the subject at rest before every running trial. The cameras were positioned within 2 m from the subjects to maximize the resolution of the recorded image. The test-retest reliability of data obtained with this setup was shown to be very good (intraclass correlation coefficient \[\text{ICC}(2,1) = .89\]) in an unpublished pilot test of 5 subjects with 2 testing sessions conducted 1 week apart.

The mean pronation range of motion (average of 25 running steps) was used to differentiate subjects with different foot postures. Subjects with a maximum pronation range of motion of less than 6 degrees in the neutral shoe condition were excluded, and subjects with $\geq 6$ degrees of pronation were included in the study. Several studies\textsuperscript{5,34–36} have suggested different cutoff values for classifying overpronation, and the selected cutoff value of 6 degrees in this study was determined by taking an average value among the previous studies. Because there was no reported criterion including sensitivity, specificity, or agreement between raters for this cutoff value, subjects in this study were regarded only as people with a potential foot pronation problem.

**Plantar Force Measurement**

Plantar force measurement was performed with a sensor insole in the left shoe. Each insole comprised 99 sensors covering the whole plantar surface, with each sensor having a sampling frequency of 99 Hz. The thickness of the insole was less than 1 mm, thus ensuring minimal disturbance to the normal gait pattern. This equipment has been reported to have very good test-retest reliability (\[\text{ICC} = .84–.99\]) in both walking and running, and it was regarded as one of the most accurate instruments for in-shoe kinetic assessments.\textsuperscript{37–40} comparable to forceplate and F-scan measurements.

To quantify the loading in different areas of the foot, the insole was subdivided into 10 anatomical regions (Fig. 1) according to a previous study.\textsuperscript{21} In each area, the mean maximum plantar force derived from 25 steps was taken as the primary outcome measure of this study.

In each testing session, the system was calibrated before use. The left insole was connected by leads to an 8-bit analog-to-digital converter box, which was suspended on the railing of the treadmill. The converter box was connected to a laptop computer for data collection. Another insole sensor without connection was put on the right shoe to minimize the running pattern disturbance.

Immediately before and after the running trial, a handheld dynamometer (MicroFet2 force gauge)\textsuperscript{5} was used to measure the isometric strength (force-generating capacity) of the left foot invertors. Each subject was asked to stand next to a wall-mounted dynamometer with the left first metatarsophalangeal joint comfortably positioned on the dynamometer pad and the ankle-foot complex in a neutral position. A stick marker was used to ensure the same foot position for the retest measurement. The right leg was positioned exactly one step behind the left leg, and both feet were a shoulder width apart (Fig. 2). The subject was asked to maximally

---

\textsuperscript{1} Hoggan Health Industries, 8020 South 1300 West, West Jordan, UT 84088

---
invert the left foot during testing for 3 seconds. Three measurements of maximal voluntary contraction (MVC) were taken, and the mean value was used for analysis. An unpublished pilot study involving 10 subjects who were healthy demonstrated good intrarater reliability (ICC[3,1] = .77), and the upper and lower bounds of the 95% confidence interval for the ICC were .938 and .324, respectively. The subject with the lowest MVC had measurements of 4.1 to 13.3 kgf, and the subject with the highest MVC had measurements of 8.1 to 16.5 kgf. Instruction and demonstration were given to the subjects to avoid accessory movements such as hip adduction, hip medial (internal) rotation, and trunk rotation. The whole data collection process was monitored by an experienced physical therapist who gave immediate feedback to the subjects if accessory movements were noticed. The drop in inversion force after running was monitored. This drop in force output might reflect some degree of fatigue of the muscle group responsible for controlling foot pronation instead of being an indication of general exhaustion.

This study comprised 2 testing sessions conducted at 1 week apart. Before each session, a standardized warm-up stretching exercise of the quadriceps femoris, hamstring, and calf muscles was performed. Subjects were allowed some treadmill exercise practice to adapt to the speed of the treadmill. The first session was with neutral shoes assessment, which aimed to identify the natural running pattern of the subjects. Only subjects with maximum rear-foot pronation of \( \geq 6 \) degrees\(^{32,34,35} \) in the neutral shoes assessment proceeded to the next session. The second testing session was with the motion control shoes. Data were recorded before and after the 1.5-km run in each testing session.

Subjects ran on a leveled treadmill at a speed of 10 km/hr. The plantar force data were collected for 25 steps of the left foot starting from the moment that the subject had caught up with the pace of the treadmill without holding on to the railing. The subject continued to run for 9 minutes (1.5 km) at that pace, and the plantar force of another 25 steps toward the end of the run was measured for analysis. Immediately after running, the inversion strength of the left foot was measured to monitor the level of drop in MVC of that muscle group.

**Data Analysis**

A \( 2 \times 2 \) repeated-measures analysis of variance (ANOVA) (general linear model) was used to test the effects of timing and shoe type on the regional mean peak force. Both shoe and timing were treated as the within factors. Because the tests were done repeatedly for the 10 anatomical regions, a Bonferroni adjustment was applied, and alpha was adjusted to .005. A simple paired \( t \) test was done to detect differences between subjects’ MVC of foot inversion before and after the 1.5-km running bout.
Results
Initially, 28 subjects signed up for this study, but 3 subjects were excluded after the kinematic assessment because their rear-foot pronation was less than 6 degrees. No subject was excluded due to musculoskeletal injury or non-heel-strike landing pattern.

Data of the remaining 25 subjects were used in the analysis. Their kinematic response to motion control is illustrated in Figure 3. It suggested that the rear-foot motion was controlled by the motion control shoes even after the 1.5-km run. The rear-foot pronation measurements are shown in Table 2. Other detailed data sets are reported elsewhere.4

The MVC produced by the foot invertors was reduced by an average of 30% to 40% in both shoe conditions (P<.01) after the run (Tab. 3). The repeated-measures ANOVA revealed a significant interaction for the factors of time and shoe type at the area under the medial midfoot (P=.001). Therefore, these 2 factors were analyzed separately. Pedographic analysis (Tab. 4) revealed that, for the neutral shoe condition, there was a 15% increase in peak force over the first metatarsal head region (364 – 418 N, P=.001) after the run. There was an 8% increase in peak force over the first metatarsal head region (524 – 565 N, P=.001). In the motion control shoe testing condition, the plantar force remained similar (P=.572) among the different anatomical regions.

Discussion
The aim of this study was to compare pedographic measurements taken before and after a 1.5-km running session in 2 different footwear conditions. Our findings indicated that the peak plantar force varied according to different footwear and timing of the test.

The drop in MVC after running was monitored only for the invertor muscle group. This drop should not be interpreted as an indicator of general muscle fatigue or cardiovascular exhaustion. Furthermore, the reduction of MVC, which is a static measurement, may not directly reflect the change in dynamic activities such as running.

The skill levels and endurance of the recreational runners recruited for this study were not very high; thus, a relatively short running distance of only 1.5 km effectively led

<table>
<thead>
<tr>
<th>Variable</th>
<th>Rear-Foot Angle (°) in Neutral Shoe Condition</th>
<th>Rear-Foot Angle (°) in Motion Control Shoe Condition</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>X</td>
<td>SD</td>
</tr>
<tr>
<td>Before 1.5-km run</td>
<td>13.9</td>
<td>1.34</td>
</tr>
<tr>
<td>After 1.5-km run</td>
<td>17.7</td>
<td>1.30</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Variable</th>
<th>MVC Output (kgf) for Neutral Shoe Condition</th>
<th>MVC Output (kgf) for Motion Control Shoe Condition</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>X</td>
<td>SD</td>
</tr>
<tr>
<td>Before 1.5-km run</td>
<td>16.5</td>
<td>4.7</td>
</tr>
<tr>
<td>After 1.5-km run</td>
<td>10.7</td>
<td>3.9</td>
</tr>
</tbody>
</table>
to a 30% to 40% drop in MVC of their foot invertor muscles. Furthermore, recreational runners are more susceptible to running-related injuries; thus, the findings of this study would be more clinically relevant to this population.

The pedographic data without zonal analysis did not detect any significant difference in plantar force measurement when rear-foot pronation was increased. Similar findings also had been reported by different researchers. However, when the data were analyzed on an anatomical zonal basis, the results for the neutral shoe condition suggested that the medial foot structure sustained a higher loading at the end of the running bout. Although there was an 8% increase in loading under the first metatarsal area, the $P$ value of .021 fell between the conventional alpha of .05 and the Bonferroni-adjusted alpha of .005; therefore, judgment on its significance is suspended.

Neutral shoes, with the midsole material consisting of a single level of hardness, were not able to control excessive rear-foot motion. We hypothesize that, after 1.5-km run, the foot invertors, which stabilize the rear foot, had become less efficient in controlling foot pronation and that the increased pronation movement could have led to an increase in plantar force on the medial structure. Such an increase in plantar force on the medial side of the foot could result in an extra moment to the foot structure so that additional muscle work is required. This phenomenon could be regarded as muscle tuning with respect to a different surface and environment. This paradigm was first proposed by Nigg. The impact forces at heel-strike can be regarded as an input signal to soft tissue vibrations. The muscle tuning paradigm refers to the adaptation of muscles to different input signals so as to minimize these vibrations. When the muscles are fatigued, they are less efficient; thus, the muscle tuning system becomes less responsive to the impact force, which may alter the movement pattern. Furthermore, according to 2 recent prospective cohort studies, such an increased plantar force pattern could be associated with exercise related lower-leg pain in runners and metatarsal stress fracture.

The pattern of plantar force was not different in the motion control shoes testing condition before and after the 1.5-km running bout. This finding suggested that not only were the motion control shoes able to control rear-foot motion after a moderate running distance, but they also checked the build-up of plantar force.

Motion control shoes usually contain duo-material in the midsole, with a firmer medial aspect to stop the foot from excessive pronation. In order to control excessive rear-foot movements, motion control shoes usually have firmer midsole materials, which could result in a relatively high average plantar force. In future studies, testing shoes with similar firmness is warranted.

There are some limitations in this study that need to be considered when using the findings. Only female runners were tested; thus, the findings may not be applicable to male runners because it has been reported that female runners demonstrate more diverse lower-extremity movement patterns than male runners.

All subjects in this study had more than 6 degrees of rear-foot pronation. However, there was no consensus in the literature for classifying overpronators. The cutoff value in this study was taken as the mean of the cutoff values used in previous studies. The clinical applicability of this cutoff value has not been tested, and the results may not be generalized to people with different rear-foot pronation.

Moreover, only the foot invertors were examined in this study. Whether other muscles of the leg also contribute to rear-foot stabilization or the involvement of other

### Table 4.

<table>
<thead>
<tr>
<th>Variable</th>
<th>Mean Peak Force (N)</th>
<th>Before</th>
<th>After</th>
<th><strong>P</strong></th>
</tr>
</thead>
<tbody>
<tr>
<td>Medial heel</td>
<td>387</td>
<td>389</td>
<td>NS</td>
<td></td>
</tr>
<tr>
<td>Lateral heel</td>
<td>407</td>
<td>405</td>
<td>NS</td>
<td></td>
</tr>
<tr>
<td>Medial midfoot</td>
<td>364</td>
<td>418</td>
<td>.001b</td>
<td></td>
</tr>
<tr>
<td>Lateral midfoot</td>
<td>419</td>
<td>412</td>
<td>NS</td>
<td></td>
</tr>
<tr>
<td>First metatarsal</td>
<td>524</td>
<td>565</td>
<td>.021c</td>
<td></td>
</tr>
<tr>
<td>Second and third metatarsals</td>
<td>503</td>
<td>511</td>
<td>NS</td>
<td></td>
</tr>
<tr>
<td>Fourth and fifth metatarsals</td>
<td>491</td>
<td>505</td>
<td>NS</td>
<td></td>
</tr>
<tr>
<td>Hallux</td>
<td>340</td>
<td>353</td>
<td>NS</td>
<td></td>
</tr>
<tr>
<td>Second and third toes</td>
<td>99</td>
<td>104</td>
<td>NS</td>
<td></td>
</tr>
<tr>
<td>Fourth and fifth toes</td>
<td>76</td>
<td>80</td>
<td>NS</td>
<td></td>
</tr>
</tbody>
</table>

* NS = not significant ($P > .05$).

b Statistically significant only before Bonferroni adjustment.

c Statistically significant after Bonferroni adjustment.
muscles during MVC testing is unknown. Further study about leg muscle response in different footwear during running may shed light on that question.

Finally, the running distance of 1.5 km is considered to be a short distance for people who run for fitness. It is not known whether the motion control shoe design would have the same beneficial effects in longer running distances. A longer distance was not tested due to the low exercise tolerance of the subjects. The subjects in this study usually ran 2.6 times per week (SD=1.4 times per week) for an average distance of about 2 km, which fell short of the American College of Sports Medicine guidelines for the minimum amount of physical activity required for health and well-being. Therefore, caution should be exercised when applying the present findings to more serious runners.

Conclusion
People with foot pronation larger than 6 degrees develop higher plantar force over the medial foot structures with normal footwear than with motion control footwear after running for 1.5 km. The plantar force measured in motion control shoes was not different before and after the running bout. This finding has implications for injury prevention in recreational runners with appropriate footwear selection.

Both authors provided concept/idea/research design, writing, data analysis, project management, subjects, and consultation (including review of manuscript before submission). Mr Cheung provided data collection. Dr Ng provided fund procurement, facilities/equipment, institutional liaisons, and clerical support.

Ethical approval was obtained from the Human Subjects Ethics Subcommittee of The Hong Kong Polytechnic University.

References
1 Cheung RTH, Ng GYF. Efficacy of motion control shoes for reducing excessive rearfoot motion in fatigued runners. Phys Ther Sports. 2007;8:75–81.
30 Olympics Series Sports Medicine Workshop II: The Science Behind Modern-Day Athletic Footwear and Their Role in Contemporary Sports Medicine (organized by The Hong Kong Polytechnic University); November 7–9, 2003; Hong Kong, China.
Footwear and Plantar Forces During Running


