Researchers and clinicians have speculated that a possible cause of foot and ankle injuries during running is the body's inability to attenuate ground reaction forces experienced during the stance period. As a result, modifications in shoe hardness have been investigated as a mechanism to prevent such injuries. However, studies of footwear have not provided significant in vivo evidence that discrete changes in shoe hardness result in a decrease in ground reaction forces. It has been suggested that this lack of evidence is due to different ankle kinematic strategies as shoe hardness is modified. However, few studies have attempted to quantify changes in these kinematic strategies as shoe hardness is changed. The altered kinematic strategies noted in the literature may represent changes in coordination at the ankle joint. Previously, it has been suggested that imbalances in lower-extremity coordination may be related to foot and ankle injuries during running. Furthermore, Payne reported that the mechanics that underlie functional relationships between the lower-extremity structural components have not yet been elucidated.

Thus additional investigations are necessary to understand the effects of footwear on ankle coordinative strategies.

Dynamic systems theory has provided tools for the assessment of coordinative strategies during locomotion. Coordination is quantified on the basis of the phasing relationship of the lower-extremity segments. For example, evaluation of the phasing relationship between the foot and the shank segments can reveal the coordination strategies used at the ankle during the stance period. From a theoretical perspective, dynamic systems theory proposes that change from one coordination strategy to another occurs when a variable to which the neuromuscular system is sensitive is scaled up or down through a critical threshold. This variable is referred to as a control parameter. For example, Stergiou et al showed that obstacle height is a control parameter that can cause a change from a heel strike to a forefoot strike running pattern when the obstacle height reaches a specific critical threshold (15% of the runner's standing height). Shoe hardness may be another control parameter that promotes new coordinative strategies at the ankle during running.

The purpose of this investigation was to gain a better understanding of the relationship between footwear and ankle coordination strategies.
dynamic systems theory perspective, we proposed that shoe hardness is a control parameter that promotes changes in ankle coordination. By altering shoe hardness and footwear, we hypothesized that new ankle coordinate strategies will be observed during the running stance period. Investigating how footwear affects ankle coordination will provide new directions for podiatric medicine.

Materials and Methods

Eight healthy men who ran a mean ± SD of 44.5 ± 29.5 km/week for the preceding 4 months volunteered as subjects (mean ± SD age, 27.1 ± 4.9 years; mean ± SD body mass, 71.9 ± 9.1 kg; and mean ± SD height, 1.76 ± 0.07 m). All of the subjects exhibited a heel-toe foot-strike pattern while running at a self-selected comfortable pace on a treadmill. Each subject had previous treadmill running experience. Before testing, each subject read and signed an informed consent form that was approved by the institutional review board of the University of Nebraska at Omaha.

The subjects ran on a treadmill under three conditions—wearing hard shoes, wearing soft shoes, and barefoot—while kinematic data of the right lower extremity were collected using two high-speed (180-Hz) cameras (JC Labs, Mountain View, California) interfaced to high-speed videotape recorders. The cameras were positioned perpendicular to the plane of motion so that the sagittal and frontal (rear) views of the subject could be recorded. Both cameras were synchronized with the Peak Event and Video Control Unit (Peak Performance Technologies Inc, Englewood, Colorado). The Peak Event and Video Control Unit allowed for the two videotape images to be time-synchronized by a manual transistor/transistor/logic switch that emitted a change in the event-synchronization diode located in the upper left portion of the field of view of each camera. Before videotaping, reflective markers were positioned on the subject’s right lower extremity. All positional markers were placed on the subjects by the same examiner (M.J.K.). Sagittal plane marker placement was as follows: 1) greater trochanter, 2) axis of the knee joint as defined by the alignment of the lateral condyles of the femur, 3) lateral malleolus, 4) outsole of the shoe approximately at the bottom of the calcaneus, and 5) outsole of the shoe approximately at the fifth metatarsal head. Frontal plane marker placement was based on an absolute approach as follows: 1) midline of the Achilles tendon between the lateral and medial malleoli, 2) below the belly of the gastrocnemius muscle on a line joining the previous marker with the bisection of the leg at the level of the popliteal fossa, and 3) and 4) on the heel counter of the shoe on a line that approximates the bisection of the posterior aspect of the calcaneus. When the shoes were removed for the barefoot condition, the shoe-placed sagittal and frontal markers were relocated directly on the skin and at the same anatomical locations. To reduce the probability of markers not being placed anatomically correctly between subjects and conditions, a standing calibration was used for each condition.

Joint markers were digitized using the Peak Motus system (Peak Performance Technologies Inc). The obtained kinematic positional coordinates of the sagittal and frontal markers were scaled and smoothed using a Butterworth low-pass filter with a selective cutoff algorithm based on the work of Jackson. The cutoff frequency values used were 13 to 16 Hz for the sagittal view coordinates and 16 to 20 Hz for the frontal view coordinates.

Shoe hardness was determined from rearfoot impact characteristics. Two running shoe models from two respectable manufacturers that had similar characteristics for all shoe features except midsole hardness—Response Cushion (Adidas, Portland, Oregon) and Beast (Brooks Sports, Inc, Bothell, Washington)—were evaluated using an impact testing system (Exeter Research Inc, Brentwood, New Hampshire). The testing procedure included 25 pre-impact trials using a mass of 8.5 kg dropped from a height of 0.05 m, followed by 20 impact trials. The American Society for Testing and Materials recommendations were followed for the testing procedure, except the number of trials was increased (from 10 to 20) to improve data reliability and validity. On the basis of the impact testing results regarding peak acceleration, the two running shoes were then classified as hard (mean ± SD, 15.05 ± 0.32 g) and soft (mean ± SD, 10.53 ± 1.01 g).

The subjects were allowed to warm up for a minimum of 8 min. This warm-up duration has been considered sufficient for individuals to achieve a proficient treadmill movement pattern. During the warm-up session, each subject established a self-selected comfortable running pace. Subjects were instructed to select a pace that would be similar to a pace that they would use when performing continuous aerobic running. This self-selected pace was used for all conditions. The mean ± SD pace was 3.24 ± 0.85 m/sec. After warming up, the subjects ran under three different conditions: wearing a soft shoe, wearing a hard shoe, and barefoot. Barefoot running was considered the baseline for the control parameter. The two shoe conditions were the increases (scaling) in the control parameter. The order of presentation of the conditions was randomly selected. All subjects
eased into their self-selected running pace before
data collection, which did not begin until the subject
stated that he felt comfortable and could maintain the
pace for a long duration. Once the subject felt com-
fortable running on the treadmill, data from ten con-
secutive footfalls (trials) were collected for further
analysis. Between conditions, the subjects were al-
lowed a minimum of 5 min of rest.

From the plane coordinates obtained from the
sagittal foot and shank and the frontal foot and leg,
angular displacements were calculated relative to the
right horizontal axis. From these angular displace-
ments, the angular velocities were calculated using a
finite difference approach. All kinematic angular dis-
placements and velocities were normalized to 100
points for the stance period using a cubic spline rou-
tine to enable mean ensemble curves to be derived
for each subject condition.

For every footfall, the relative knee angle was cal-
culated, and the maximum knee flexion was identified
using interactive laboratory software. Subsequently,
the time of occurrence of maximum knee flexion
was used to evaluate each footfall for two periods: 1)
heel contact to maximum knee flexion (absorption)
and 2) maximum knee flexion to toe-off (propul-
sion). It was decided to divide the stance period at
maximum knee flexion because this event separates
the absorption and propulsion periods, during which
different kinematic strategies may exist.

Dynamic systems theory tools were used to exam-
ine changes in coordination.5–9 Phase portraits for the
sagittal foot and shank and the frontal foot and leg
segments were generated. A phase portrait is a plot
of a segment’s angular position versus its angular
velocity. The angular displacements and velocities
were normalized to their maximum absolute values.15
The resulting phase plane trajectories were then
transformed from Cartesian (z, ˙z) to polar coordi-
nates, with a radius and phase angle Φ = tan⁻¹[ ˙z/z].
Phase angles calculated from these trajectories had a
range of 0° to 180°. Figure 1 depicts an exemplar nor-
malized phase portrait and the calculated phase
angle. Phase angles allow for the incorporation of an-
gular displacements and velocities to examine coor-
dinative strategies.

The phase angles were used to determine the phas-
ing relationships between the segments. Continuous
relative phase represents the phasing relationships
between the actions of the two interacting segments
at every point during a specific period. Continuous
relative phase provided a measure of coordination
for the two segments during the stance period, and it was
calculated by subtracting the phase angles of the cor-
responding segments throughout the stance period:

ΦFRONTAL FOOT-LEG RELATIVE PHASE = ΦFOOT – ΦLEG

and

ΦSAGITTAL FOOT-SHANK RELATIVE PHASE = ΦFOOT – ΦSHANK

Values close to 0° indicate that the two segments are
in-phase. Values close to 180° indicate that the two seg-
ments are out-of-phase. The continuous relative phase
curves for each segmental relationship were averaged
across footfalls, and mean ensemble curves were gen-
erated for all subject conditions. To test differences
between the curves, it was necessary to characterize
the curves by single numbers.5, 6, 10 Therefore, the
mean absolute value of the ensemble continuous rel-
ative phase curve values (MARP) was calculated by
averaging the absolute values of the ensemble curve
points for the designated periods (absorption and
propulsion):

\[
\text{MARP} = \frac{1}{N_p} \sum_{i=1}^{p} |\Phi_{\text{Relative Phase}}|
\]

where \( p \) is the number of points in this period. A low
MARP value indicated that there was a more coordi-
nated relationship (in-phase) between the interacting
segments, whereas a high MARP value indicated that
there was a less coordinated (out-of-phase) relation-
ship. Group means were calculated for the MARP of
each segmental relationship for each period and for
each condition.

One-way repeated measures analyses of variance
(shoe condition, with subject as the repeated factor)
were performed on the subject mean MARP values.
Statistical analysis was performed for each coordina-
tive relationship (frontal foot and leg and sagittal foot and shank) and for each period (absorption and propulsion). In tests that resulted in significant F ratios ($P < .05$), a post hoc Tukey multiple comparison test was performed to identify the location of significant differences. All statistical measures were conducted at $\alpha = .05$.

**Results**

Graphically, differences among the conditions were evident during absorption for the sagittal foot–shank relationship (Fig. 2A). The barefoot condition was more out-of-phase, whereas the two shoe conditions were more in-phase. A more out-of-phase relationship for the barefoot condition during absorption may be related to the fact that all of the subjects changed to a forefoot posture during the stance period when footwear was removed. Thus, while running barefoot, the subjects positioned the ankle in such a way that only the forefoot made contact with the ground during the stance period. During propulsion, all three conditions had a similar curve configuration. However, the barefoot foot–shank continuous relative phase was more in-phase. This is evident because the barefoot continuous relative phase was closer to $0^\circ$ during propulsion.

The frontal foot–leg relationship further suggested that the coordinative strategies were different for the barefoot condition compared with the two shoe conditions (Fig. 2B). The coordinative strategy for the foot–leg relationship while running barefoot was to maintain a more in-phase, coordinated relationship throughout the entire stance period. This is evident because the barefoot curve is much closer to $0^\circ$, which numerically indicates that the coordinative strategy was to maintain an in-phase relationship.

Post hoc analysis revealed that significant differences were found for all MARP value comparisons between the shoe conditions and the barefoot condition ($P < .05$) (Table 1). No significant differences were found between the two shoe conditions. The MARP values for the frontal foot–leg relationship were significantly lower while running barefoot, indicating a tendency toward a more coordinative relationship. During the absorption period, the MARP values for the sagittal foot–shank relationship were significantly higher, indicating a tendency toward an out-of-phase coordinative relationship.

**Discussion**

The results of this investigation indicate that ankle coordinative strategies are affected by footwear. Ex-
amination of the absorption period revealed that both MARP segmental relationships (sagittal foot and shank and frontal foot and leg) were significantly different between the barefoot condition and the footwear conditions. Analysis of the continuous relative phase graph (Fig. 2A) and the MARP values (Table 1) of the sagittal foot–shank relationship suggested that the behavior of the ankle was different during absorption and propulsion. The MARP values indicated that while running barefoot there was a more out-of-phase foot–shank relationship during absorption. During the investigation, we noted that while running barefoot all of the subjects positioned the ankle in such a way that only the forefoot made contact with the ground during the stance period. Making contact with only the forefoot during the stance period seems to affect the coordinative strategy at the ankle. It is possible that the surrounding ankle musculature plays a different mechanical role while running barefoot. Such mechanical changes may be responsible for the new coordinative strategies observed in this investigation.

An alternative explanation for the coordination differences noted while running barefoot is that the lower extremities' perception of impact forces may affect the selection of a coordinative strategy. Cole et al suggested that the human body makes kinematic adjustments according to the severity of the perceived impact. Such perceptual information about the amount of impact may be gained through the mechanoreceptors of the foot. De Clercq et al indicated that during barefoot heel-toe running, the heel pad becomes deformed to its physiologic maximum. De Clercq et al stated that a flatter foot placement results in less deformation of the fatty heel tissue and prevents overloading of the heel. It is possible that the altered coordinative strategies observed during barefoot running may be related to a perceived high-impact situation by the mechanoreceptors of the foot. Further investigations are necessary to support the notion that perceptual information is related to changes in ankle coordinative strategies.

Analysis of the frontal coupling of the foot and leg also revealed significant differences in the coordinative strategies while running barefoot and with shoes. Examination of the MARP values during absorption and propulsion indicated that the foot–leg relationship exhibited a more in-phase relationship while running barefoot than when running with shoes (Table 1). Bates et al determined that rearfoot motion decreased while running barefoot because the foot was placed in a more inverted position. Thus the in-phase relationship seen in this investigation may be related to the same functional adaptation of decreasing rearfoot motion. The coordinative strategy observed while running barefoot may be related to the ankle joint being less dependent on the subtalar joint to attenuate impact forces. The ankle was more planatarflexed and more inverted. Thus there is greater dependence on the large gastrocnemius musculature to attenuate impact forces. Therefore, we suggest that during running with footwear, the ankle joint complex adopts a coordinative strategy that uses the subtalar joint as a force-attenuating mechanism. However, as impact forces exceed a perceived critical injurious threshold, the ankle joint complex transitions to a coordinative strategy that depends more on the large gastrocnemius musculature to attenuate forces.

A limitation of this investigation is that we did not account for movement of the foot in the shoe. It was assumed that the frontal plane motion of the shoe mapped onto the frontal plane motion of the foot inside the shoe. Therefore, it is possible that the significant differences observed in the foot–leg coordination strategy are due to differences in using shoe- versus skin-mounted markers to define the motion of the foot segment in the frontal plane. Future investigations of leg–foot coordination in the frontal plane should consider addressing movement of the foot in the shoe.

No statistically significant differences were found...
in the ankle coordinative strategies used between the two shoe conditions. Lake and Lafortune determined that individuals could not perceive differences in impact between small incremental changes in material densities. This study may have yielded similar results in that the change in shoe hardness may not have been sufficient for subjects to exhibit a change in ankle coordinative strategies. An alternative explanation is that the subjects’ daily footwear may have matched the footwear densities used in this investigation. Perhaps if the subjects used footwear that had a stiffness factor similar to that of the shoes used in this investigation, the ankle may not have sensed a need to change ankle coordinative strategies. Future investigations of the relationship between ankle coordination and footwear should take into consideration the type of daily footwear used by the subjects.

Conclusion

Our approach to understanding the effects of footwear on coordinative strategies at the ankle was based on the principles of dynamic systems theory, which quantifies coordination on the basis of the phasing relationship of the lower-extremity segments. It is possible that imbalances in coordination are related to foot and ankle injuries during running. This investigation provides initial insight into how footwear affects ankle coordination. The results indicate that the ankle adopts new coordinative strategies while running barefoot versus with footwear. These coordinative strategies may be related to different mechanisms to attenuate impact forces while running. By using the information presented in this study, a future approach that applies dynamic systems theory measures may provide further insight into the relationship between coordinative strategies and footwear. Such investigations will provide additional information on how footwear affects ankle coordination and will guide future research in podiatric medicine.

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