5-2010

Effects of Varying Amounts of Pronation on the Mediolateral Ground Reaction Forces During Barefoot Versus Shod Running

Joanna B. Morley
University of Nebraska at Omaha

Leslie M. Decker
University of Nebraska at Omaha

Tracy Dierks
Indiana University

Daniel Blanke
University of Nebraska at Omaha, dblanke@unomaha.edu

Jeffrey French
University of Nebraska at Omaha, jfrench@unomaha.edu

See next page for additional authors

Authors

Joanna B. Morley, Leslie M. Decker, Tracy Dierks, Daniel Blanke, Jeffrey French, and Nikolaos Stergiou

Journal of Applied Biomechanics, 2010, 2, 205-214

Effects of Varying Amounts of Pronation on the Mediolateral Ground Reaction Forces During Barefoot Versus Shod Running

Joanna B. Morley, Leslie M. Decker, Tracy Dierks, Daniel Blanke, Jeffrey A. French, and Nicholas Stergiou

Despite extensive research on running mechanics, there is still a knowledge gap with respect to the degree of relationship between mediolateral ground reaction forces (ML-GRF) and foot pronation. Our goal was to investigate whether differences exist in ML-GRF among runners that exhibit different degrees of pronation. Seventeen male and 13 female recreational runners ran with and without shoes while ML-GRF and frontal kinematics were collected simultaneously. Subjects were divided into groups based upon their peak eversion (low pronation, middle pronation, high pronation). Discrete parameters from the ML-GRF were peak forces, respective times of occurrence, and impulses. No significant differences were found between groups regarding the magnitude of ML-GRF. Based upon the relative times of occurrence, the peak medial GRF occurred closer to the peak eversion than the peak lateral GRF. Findings support the idea that the ML-GRF have less to do with pronation than previous research suggested.

Keywords: kinematics, kinetics, shoes, eversion, inversion, rearfoot

Over the past three decades running has become one of the most popular forms of exercise and research related to running mechanics has increased to reflect this trend (Cavanagh & Lafortune, 1980; Cavanagh, 1987; Lawrence, 1997; McClay, 2000; Albers & Hoke, 2003; Asplund & Tanner, 2004). However, the etiology of running injuries continues to elude scientists and clinicians alike. It is estimated that, in a given year, half of all runners will sustain a musculoskeletal injury and will subsequently be 50% more likely to become reinjured (Messier & Pittala, 1988; van Mechelen, 1992; Asplund & Tanner, 2004).

Despite the wealth of literature regarding running mechanics, the relationship between running mechanics and injuries is not well understood. Therefore, to prevent and treat incidence of injuries among runners it is crucial to understand the mechanisms that predispose and lead to injury. It has been suggested that a dynamic functional abnormality may be more important than a static misalignment in predisposing a runner to injury (Nigg, 1985; James & Jones, 1990; Van Mechelen, 1992). Most current research has focused on atypical subtalar (talocalcaneal) joint compensatory motion as a causal factor contributing to chronic injury in the lower extremities (James et al., 1978; Bates et al., 1982; Nigg & Morlock, 1987; Messier & Pittala, 1988; Holden & Cavanagh, 1991; Messier et al., 1991; Stacoff et al., 1991; Hamill et al., 1992; Messier et al., 1995; Freychat et al., 1996; Stergiou & Bates, 1997; Wen et al., 1997; Busseuil et al., 1998; McClay & Manal, 1998; Hreljac et al., 2000; Stacoff et al., 2000; Stergiou et al., 2003; Hreljac, 2004, 2005). During the stance phase of gait, the subtalar joint exhibits the triplanar, multiphasic motion commonly referred to as *pronation*. Foot pronation combines the movement of calcaneal (or rearfoot) eversion (frontal plane), forefoot abduction (horizontal plane) and dorsiflexion (sagittal plane) to aid in shock absorption during running (Buchbinder et al., 1979; Perry & Lafortune, 1995; Busseuil et al., 1998; Hintermann & Nigg, 1998). Foot pronation, however, is difficult to quantify because of its three-dimensional nature and there is substantial variation in the orientation of the subtalar axis across subjects and joint positions (Kirby, 2001). Rearfoot angle or foot eversion-inversion is often used to estimate foot pronation-supination because this angle is relatively independent from motions in other joints, thus less prone to errors (Stacoff et al., 1991; Perry & Lafortune, 1995). Furthermore, it has been demonstrated that

Morley and Decker are with Nebraska Biomechanics Core Facility, University of Nebraska at Omaha, Omaha, NE. Dierks is with the Department of Physical Therapy, Indiana University, Indianapolis, IN. Blanke is with Nebraska Biomechanics Core Facility, University of Nebraska at Omaha, Omaha, NE, and with Health Promotion, Social & Behavioral Health, College of Public Health, University of Nebraska Medical Center, Omaha, NE. French is with the Department of Psychology, University of Nebraska at Omaha, Omaha, NE. Stergiou is with Nebraska Biomechanics Core Facility, University of Nebraska at Omaha, Omaha, NE, and with the Department of Environmental, Agricultural, and Occupational Health Sciences, College of Public Health, University of Nebraska Medical Center, Omaha, NE.

rearfoot motion measured with either two-dimensional or three-dimensional analysis is essentially the same for the initial 80% of the stance phase (Areblad et al., 1990; Hamill et al., 1994). Differences are increased as the foot moves out of the plane with maximum differences occurring during toe-off that affect the end of the rearfoot angle curve (from 80% to the end of the stance phase; Areblad et al., 1990; Hamill et al., 1994). Because the key phenomena regarding the rearfoot angle occur between 15% and 65% of the stance phase, a two-dimensional analysis can be considered adequate (Stergiou et al., 1999).

Pronation during the stance phase of running is a natural movement and necessary to allow for the impact forces to be absorbed during a longer period by the supporting structures reducing the effective magnitudes of these forces. Without subtalar pronation, these forces would have to be abruptly and directly absorbed by the supporting structures, causing problems associated with excessive stress (Bates et al., 1978; Chu et al., 1986; Harris, 1991; Sangeorzan, 1991, Stergiou et al., 1999). Excessive pronation, on the other hand, may lead to injuries in diverse locations of the lower extremities (James et al., 1978; Viitasalo & Kvist, 1983; Clement et al., 1984; Messier & Pittala, 1988; Cook et al., 1990; James & Jones, 1990; van Mechelen, 1992; Hintermann & Nigg, 1998; Cheung et al., 2006). However, it is not clear if running injuries are directly or indirectly affected by excessive pronation. There are no results from prospective studies providing evidence for a direct relation between excessive pronation and an increased frequency of running injuries. The current knowledge suggests that excessive pronation per se may not be a sufficient factor for injury development, but that excessive pronation in combination with other anatomical or biomechanical factors, may lead to running injuries (Reinschmidt & Nigg, 2000). For instance, Stergiou & Bates (1997) suggested that lack of coordinative action between pronation of the subtalar joint and knee motion may have greater potential for predicting runners with susceptibility to injury. With increases in the vertical ground reaction forces (GRF) at impact, the actions of subtalar pronation/supination and knee flexion/extension become more asynchronous which leads to increased susceptibility to injury (Stergiou et al., 1999).

However, limited research has been performed in relating subtalar joint pronation measurements to another component of the ground reaction forces, the mediolateral (Hamill & Bates, 1988; Hamill et al., 1989). Analyses of vertical ground reaction forces (V-GRF) or anteroposterior ground reaction forces (AP-GRF) through force-time curves are now well defined because previous studies (e.g., Cavanagh & Lafortune, 1980; Miller, 1990) have specified various elements of these forces to describe running characteristics. The relationship of mediolateral ground reaction forces (ML-GRF) to running kinematics, however, has been limited due to the relatively small magnitude of these forces compared with the vertical and anterior-posterior forces and the lack of a consistent pattern for ML-GRF which has been attributed to intersubject variability (Miller, 1990). Some runners exhibit medially directed initial waves, while others show lateral ones. Force magnitudes and the number of zero line crossings are also variable among subjects, with differences existing both intrasubject and interlimb (Miller, 1990), although ML-GRF may be more sensitive than rearfoot motion per se to gain insight into functional injury mechanisms.

Giakas et al. (1996) used pronation and supination interchangeably with medial and lateral force excursions, with the premise that side-to-side forces are associated with side-to-side motion. This idea was supported by Bates et al. (1981) who reported that impulses associated with the peak medial ML-GRF corresponded to decreased maximum eversion. However, ML-GRF are not only determined by rearfoot motion, which is just one component of the movements that occur in the frontal plane, but also by other intrinsic (e.g., height and stiffness of the medial longitudinal arch of the runner's foot) and functional (e.g., subject specific movement coupling between foot and leg) factors (Reinschmidt & Nigg, 2000). Thus, it is crucial to gain a better understanding of the differences (and similarities) between ML-GRF and rearfoot motion. Identifying the extent of these differences (and similarities) may help to design better footwear solutions for different groups of runners, and ultimately reduce and prevent the incidence of running-related injuries.

Therefore, the purpose of this study was to investigate if differences exist in the force measures obtained from the mediolateral component of the GRF between runners that exhibit different degrees of pronation while running shod and barefoot. A barefoot condition was incorporated to eliminate the effects that shoes may have on altering foot mechanics, such as pronation. A secondary goal was to examine whether a particular ML-GRF measure was more related than others. Three hypotheses were tested: (1) since shoes are constructed to provide mediolateral stability, it was hypothesized that the amount of pronation would be larger while running barefoot, and that the ML-GRF incurred in the barefoot condition will be significantly greater than those obtained in the shod condition; (2) since subtalar pronation serves to absorb forces incurred during running, it was hypothesized that the ML-GRF would be significantly greater in a high pronation group compared with low pronation and middle pronation groups; (3) the time to peak lateral ML-GRF will occur about the same time that maximum eversion occurs indicating a closer linkage between the two. The dependent measures were variables derived from the rearfoot kinematics (maximum eversion and the time to maximum eversion) and the GRF (peak forces, their respective times of occurrence, and impulses) during the stance phase.

Methods

Subjects

Thirty healthy male ($N = 17$) and female ($N = 13$) recreational runners (age: 24.0 ± 1.84 years; body weight (BW): 73.5 ± 16.23 kg; height: 175.6 ± 9.01 cm) from the community volunteered as subjects for this study. All subjects were without injuries or physical impairments at the time of testing. Before the subjects were admitted to this research study, the investigators qualitatively analyzed their running style to ensure they used a heel strike pattern at their preferred pace. Before testing, each subject provided an informed consent and a health questionnaire approved by the University's Institutional Review Board.

Instrumentation

A Kistler force platform (Kistler Model 9281-B11, Amherst, NY) connected to a Kistler signal conditioner/amplifier (Kistler Model 9807) was used to record the GRF at a sampling rate of 960 Hz. The force platform was mounted flush with the floor in the middle of the runway. The mediolateral (F_x) GRF component was retained for further analysis.

A posterior view of the right lower extremity was obtained for all trials using a Panasonic WV-CL350 (Osaka, JA) video camera with a sampling frequency of 60 Hz. The video camera was located 10-m from the force platform and parallel to the walking pathway. A zoom lens (Cosmicar TV, 8–48 mm zoom lens, Cosmicar/Pentax Precision Co., Tokyo, Japan) was used in conjunction with the video camera to optimize image size and minimize perspective error. A light source (Pallite VIII using eight ELH 300-W tungsten-halogen projection lamps at 120 V AC) was mounted with the camera lens in the center of the ring to better illuminate the reflective markers. Reflective markers were positioned on the subject's right lower extremity to allow for path tracking and to provide reference points for determination of eversion angles. All positional markers were placed on the subjects by the same examiner. Marker placement was as follows: (a) center of the sole on the heel of the shoe, or heel, for the barefoot condition; (b) center of the heel tab, or calcaneus, in the barefoot condition; (c) center of the Achilles tendon, just above marker b; and (d) center of the calf, 20 cm above the Achilles tendon marker (Stergiou & Bates, 1997; Figure 1). The video images were stored on S-VHS video tapes via a Panasonic AG-1970P video camera recorder (VCR), which was interfaced with a Magnavox TV for an instant qualitative evaluation of the video recording. The video data were transformed to digital format and digitized via the Peak Motus video system (Peak Performance Technologies, Inc., Englewood, CO).

Video and force-plate data were synchronized via the Peak Event Synchronization Unit (ESU). Data synchronization was controlled by depression of a manual thumb switch that was connected to the ESU, thereby generating a voltage pulse square wave (VPSW) sent to the ADIU and VCR. The VPSW (3.9 V) initiated GRF data collection, and was recorded as the synchronization channel. A 16-line vertical digital bar code was simultaneously positioned in the upper-right quadrant of the video picture, which represented the frame in which GRF data collection started. To synchronize kinematic and GRF data, the frame with the bar code was matched with the initiation of the square wave in the GRF data.

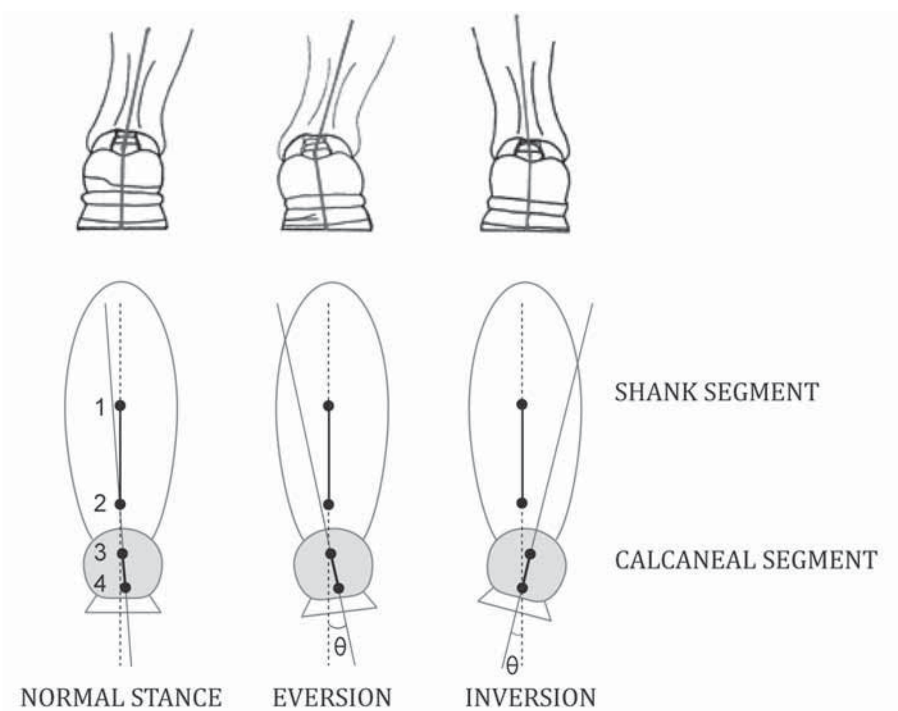

Figure 1 — Rear foot inversion/eversion during running relative to the subtalar joint reference position (dotted line). Shank and calcaneal segments are shown as defined by the (1) midcalf, (2) Achilles tendon, (3) heel tab, and (4) midsole markers. When these segments are extended, the resulting angle (θ) will represent eversion or inversion of the ankle-subtalar-joint complex.

Experimental Protocol

Subjects wore their own running shoes to ensure the most typical performance. Subjects wore shorts to allow for unrestricted movement, and tracking of the reflective markers. After the markers were affixed to the lower right extremity and before each condition, subjects were filmed standing stationary on the force platform to identify the reference position of the subtalar joint in a standing weight bearing position. This position served as the zero point for the processing of dynamic rearfoot eversion angles.

Subjects were given time to accommodate to the experimental setup and to adequately warm up before testing. Warm-up consisted of running through the testing area without concern for stepping on the force platform. The testing area was a 10-m runway with a 0.6-m-wide lane. During warm-up, the subjects established a comfortable self-selected running pace that was recorded using a photocell timing system. This system used two infrared timing lights connected to a digital timer. Based upon the subject's average running speed, a range that allowed ±5% deviation of this speed was used for the subsequent testing and a trial was considered acceptable only when the running speed was within this predetermined range. The investigator also asked the participants not to look at the floor to locate the force platform for proper right foot placement, as this could influence the subject's natural running kinematics and GRF. To ensure consistent right foot placement on the force platform, a foot placement marker was located approximately 5 m before the platform to designate the point in which to initiate running. This distance was determined through trial and error during the practice trials. Each trial was visually monitored to ensure that the stride was normal with a heel strike running pattern, and the foot was completely on the force platform. Visual inspection of the force curves allowed for an intertrial rest interval of 1 min. Every subject ran at the previously established comfortable self-selected pace with and without their athletic shoes. Each experimental condition (shod vs. barefoot) consisted of ten acceptable trials for a total of twenty acceptable trials per subject. Subjects were allowed as many trials as needed to achieve an acceptable trial.

Data Reduction and Analysis

All kinematic coordinates were scaled and smoothed using a low-pass Butterworth filter with a selective cutoff algorithm based on Jackson (1979). The cutoff frequencies used were between 16 and 22 Hz for the rear view coordinates (Stergiou et al., 1999). The smoothed data were visually compared with the raw data to verify the appropriateness of the processing. All data were smoothed by the same investigator to assure consistency of results. Subsequently, from the frontal plane coordinates, eversion was measured as the angle subtended by the bisection of the calcaneal and shank segments. Following data analysis, the subjects were divided into three equal groups based upon their peak eversion values obtained from the shod condition: the low pronation (3–8.9 deg), the middle pronation (9–12.9 deg), and the high pronation (13–18 deg) groups. The limits for the groups were based on Clarke et al. (1984). The kinematic parameters analyzed were the maximum eversion and the time to maximum eversion. The ML-GRF parameters analyzed were the peak medial ML-GRF, the peak lateral ML-GRF, their respective times of occurrence, the absolute difference between PM and PL, the impulses associated with the PM and the PL and the total medial and lateral impulses. Subsequently, ML-GRF values were normalized to body weight, whereas the impulses in newton seconds were normalized by dividing them with the impulse of the individual's body weight over the stance time generating units of body weight impulse, (Miller, 1990; Figure 2).

Statistical Analysis

Descriptive statistics were calculated for each of the kinematic and GRF variables for each subject in each condition. The group values were entered into a 2 × 3 mixed factor ANOVA (within-subjects factor: shod condition; between-subjects factor: pronation group). A one-way ANOVA was also performed to investigate if differences existed between groups with respect to speed of running. In tests that resulted in significant F-ratios ($p < .05$), a post hoc Tukey test was performed. All statistical measures were conducted at $\alpha = .05$.

Results

We identified that the mean running speed for all subjects was 3.41 $m \cdot s^{-1}$. Importantly, no significant differences were found in the running speed among the three pronation groups ($p = .88$). This excluded speed as a confounding factor.

In the shoe condition, the maximum eversion showed significant differences between all groups (Table 1). In the barefoot condition, the underpronation group exhibited significantly less maximum eversion (6.3 ± 2.6°) than the overpronation group (9.2 ± 3.2°). There were no significant differences between either the underpronation group or the overpronation group and the normal pronation group (6.7 ± 1.7°) (Table 1). Figure 3 provides a representative curve of eversion/pronation during the stance phase of running.

The barefoot condition resulted in decreased eversion angle across all groups, although not significant for the low pronation group. Both the middle pronation and high pronation groups exhibited significantly lower maximum eversion angles in the barefoot condition compared with the shod condition. The high pronation group exhibited decreased maximum eversion from 14.8 (± 1.5°) in the shod condition to 9.2 (± 3.2°) in the barefoot condition, and the middle pronation group showed a similar reduction from 10.3 (± 0.9°) in the shod condition to 6.7 (± 1.7°) in the barefoot condition. The maximum eversion in the low pronation group was not significantly influenced by the barefoot condition (6.7 ± 2.1° in the shod

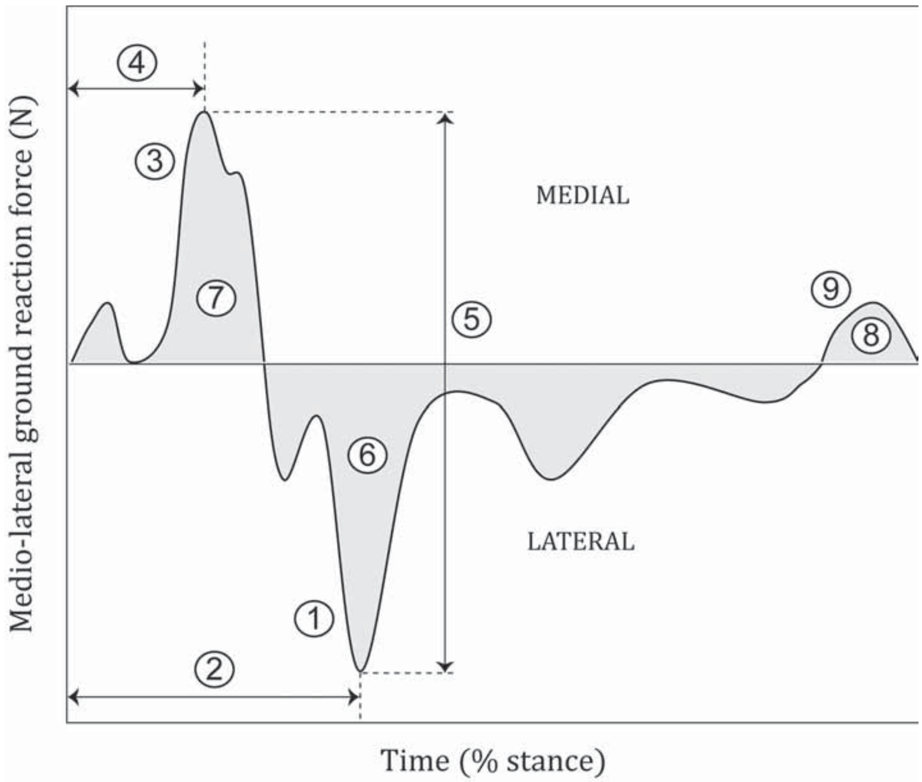

Figure 2 — A representative ML-GRF pattern exhibited during the stance phase of running. ML-GRF values were normalized to body weight, whereas the impulses in newtons seconds were normalized by dividing them with the impulse of the individual's body weight over the stance time generating units of body weight impulse (Miller, 1990). ML-GRF parameters: (1) PL: peak lateral ML-GRF, (2) TPL: time to PL, (3) PM: peak medial ML-GRF, (4) TPM: time to PM, (5) AD: absolute difference between PM and PL, (6) IPL: impulse associated with PL (area under PL), (7) IPM: impulse associated with PM (area under PM), (8) TLI: total lateral impulse (total area above the zero line), (9) TMI: total medial impulse (total area below the zero line). Note: Several deviations from this pattern may exist for different types of runners. This ML-GRF pattern was constructed based upon several patterns found in the literature (Bates et al., 1981, 1983; Cavanagh & Lafortune, 1980; Cavanagh, 1987; Freychat et al., 1996; Giakas et al., 1996; Mann et al., 1981; Munro & Miller, 1987).

condition compared with 6.3 ± 2.6° for barefoot condition) (Table 1).

The laterally directed peak forces incurred in the barefoot condition were significantly greater than those obtained in the shod condition in the middle pronation group, but not in the low pronation and high pronation groups. The peak lateral ML-GRF in the middle pronation group significantly increased from 9 ± 0.02% of body weight in the shod condition to 13 ± 0.04% of body weight in the barefoot condition. The middle pronation group also showed a significant increase of absolute difference between the peak medial ML-GRF and the peak lateral ML-GRF in the barefoot condition (27 ± 0.08% of body weight) compared with the shod condition (23 ± 0.06% of body weight). The peak lateral ML-GRF was not significantly different between conditions in both the low pronation and high pronation groups. The peak medial ML-GRF was not significantly different between conditions in all groups. Only the middle pronation group showed a significant decrease of absolute difference between the peak medial ML-GRF and the peak lateral ML-GRF in the shod condition (23 ± 0.06% of body weight) compared with the barefoot condition (27 ± 0.08% of body weight). The impulse associated with the peak medial ML-GRF was significantly lower in the barefoot condition compared with the shod condition in the low pronation group. The impulse associated with the peak lateral ML-GRF was significantly lower in the barefoot condition compared with the shod condition in the high pronation group. The total medial and lateral impulses showed no significant differences between conditions in all groups (Table 1).

Contrary to our expectations, the ML-GRF were not significantly greater in the high pronation group compared with the low pronation and middle pronation groups. The peak medial ML-GRF and the peak lateral ML-GRF were not significantly different between groups in both conditions. The absolute difference between the peak medial ML-GRF and the peak lateral ML-GRF was statistically different between groups in the shod condition: both low pronation and high pronation groups revealed significantly lower absolute difference between

Table 1 Group means (*M*) and standard deviations (*SD*) for all measured parameters for each experimental condition (shod versus barefoot)

	Shod Condition			Barefoot Condition		
Variable	Low	Middle	High	Low	Middle	High
ME	$6.7^{N,O} \pm 2.1$	$10.3^{O,*} \pm 0.9$	14.8* ± 1.5	$6.3^{O} \pm 2.6$	6.7* ± 1.7	9.2* ± 3.2
TME	38.6* ± 7.1	40.6* ± 10.9	36.9* ± 8.3	25.0* ± 11.5	23.8* ± 10.2	27.2* ± 8.4
PM	0.10 ± 0.03	0.13 ± 0.05	0.10 ± 0.04	0.10 ± 0.03	0.14 ± 0.05	0.11 ± 0.03
TPM	32.9* ± 10.4	28.7 ± 11.2	23.9 ± 13.7	24.0* ± 13.1	22.4 ± 7.9	21.9 ± 10.8
PL	0.11 ± 0.06	0.09* ± 0.02	0.11 ± 0.05	0.12 ± 0.03	0.13* ± 0.04	0.12 ± 0.03
TPL	22.5* ± 20.3	17.7* ± 6.2	20.0* ± 14.9	12.9* ± 12.2	7.23* ± 3.4	9.72* ± 9.0
AD	$0.20^{N} \pm 0.05$	$0.23^{O,*} \pm 0.06$	0.21 ± 0.05	0.22 ± 0.04	0.27* ± 0.08	0.24 ± 0.05
IPM	4.75* ± 3.5	5.61 ± 3.7	3.67 ± 3.2	3.36* ± 1.9	4.81 ± 3.3	3.47 ± 2.3
IPL	1.25 ± 0.8	1.11 ± 0.5	2.85* ± 3.8	1.35 ± 0.8	1.04 ± 0.7	1.43* ± 0.7
TMI	5.52 ± 3.5	6.65 ± 3.8	4.25 ± 3.4	4.29 ± 2.3	6.16 ± 3.6	4.37 ± 2.1
TLI	2.27 ± 1.7	1.86 ± 1.0	4.96 ± 3.8	2.54 ± 1.47	2.07 ± 1.0	3.71 ± 2.6

Note. Subjects were divided into three equal groups ($N = 10$) based upon their peak eversion values: the low pronation (3–8.9 deg), the middle pronation (9–12.9 deg), and the high pronation (13–18 deg) groups. The kinematic parameters are the maximum eversion (ME) and the time to maximum eversion (TME). The ML-GRF parameters are the peak medial ML-GRF (PM), the peak lateral ML-GRF (PL), their respective times of occurrence (TPM and TPL), the absolute difference between PM and PL (AD), the impulses associated with the PM and the PL (IPM and IPL) and the total medial and lateral impulses (TMI and TLI). Timing parameters are expressed in percentage of stance, ME in degrees, impulses in newton seconds, PM and PL in body weight.

*Significantly different between conditions for the same group ($p < 0.05$).

^{N,O,U}Significantly different between groups for the same condition ($p < 0.05$).

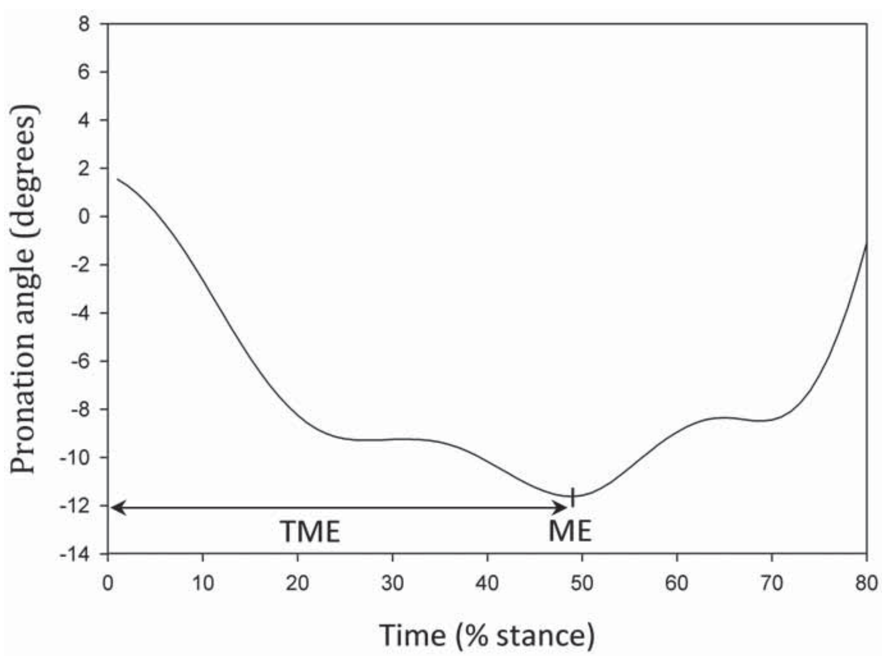

Figure 3 — A representative curve of eversion/pronation exhibited during the initial 80% of the stance phase of running. ME: maximum eversion, TME: time to maximum eversion.

the peak medial ML-GRF and the peak lateral ML-GRF than the middle pronation group. The impulses associated with the peak medial ML-GRF and the peak lateral ML-GRF as well as the total medial and lateral impulses showed no significant differences between groups in both conditions (Table 1).

Based upon relative times of occurrence, the peak medial ML-GRF always occurred after the peak lateral ML-GRF, and the maximum eversion always occurred after the peak medial ML-GRF. The time to maximum eversion occurred significantly earlier in the barefoot condition compared with the shod condition in all groups.

In the low pronation group, the time to peak medial ML-GRF occurred significantly earlier in the barefoot condition (32.9 ± 10.4% of stance) compared with the shod condition (24.0 ± 13.1% of stance). The time to peak lateral ML-GRF occurred significantly earlier in the barefoot condition compared with the shod condition in all groups (Table 1).

All groups demonstrated greater variability of the maximum eversion values in the barefoot condition compared with the shod condition, as indicated by increased standard deviations between the two conditions. In all groups and in both conditions, variability of the time to peak lateral ML-GRF values was noticeably high. The standard deviation values approached mean values in both conditions for the low pronation group and in the barefoot condition for the high pronation group. Calculations of all impulse parameters presented large amounts of between-subjects variability: in some cases, the standard deviation values approached or even exceeded the mean values (e.g., impulses associated with the peak lateral ML-GRF of the shod condition in the high pronation group; Table 1).

Discussion

The purpose of this study was to investigate if differences exist in ML-GRF between runners that exhibit different degrees of pronation while running shod and barefoot. The eversion range of motion in the study group runners was between –3.18° (*SD* 0.949) and –17.69° (*SD* 1.119). Based upon these values, subjects were assigned to one of three different pronation groups (Clarke et al., 1984). Peak rearfoot eversion values were found to be significantly different between groups in the shod condition, which was expected due to the group placement criteria.

As running shoes lend stability to foot motion and cushioning in the midsole area of the foot to control foot pronation (or eversion), it was hypothesized that the amount of pronation would be larger while running barefoot, and that the ML-GRF incurred in the barefoot condition will be significantly greater than those obtained in the shod condition. The first part of the hypothesis was rejected, while the second part was supported by our results. The barefoot condition resulted in decreased eversion values across all groups. A possible explanation for this phenomenon is modifications to running technique and increased plantar flexion in early stance while running barefoot which is associated with subtalar joint supination (which consists of talar dorsiflexion-abduction and calcaneal inversion) (Kurz & Stergiou, 2004). In addition, it is important to consider the effect of the shoe as demonstrated by Reinschmidt et al. (1997) using bone pins inserted into the tibia and calcaneus of five subjects. Essentially they showed that the actual movement of the foot within the shoe is less than what the shoe would indicate. Furthermore, the barefoot condition resulted in larger ML-GRF peaks. The time to peak medial ML-GRF and the time to peak lateral ML-GRF were also found to occur significantly earlier in the barefoot condition compared with the shod condition in all groups. These findings were consistent with those reported by Hamill et al. (1996). Messier et al. (1991) found that the time to peak lateral ML-GRF was a significant discriminator between controls and a group that exhibited patellofemoral pain and increased pronation. Our results did not support such a claim.

As subtalar pronation serves as a mechanism to transmit and dampen impact forces to the lower extremity during ambulation (Neely, 1998), it was hypothesized that ML-GRF would be significantly greater in the high pronation group compared with the low pronation and middle pronation groups. This hypothesis was not supported by our results. We found that the peak medial ML-GRF and the peak lateral ML-GRF were not significantly different between groups. This is a very important finding because it shows that pronation and supination cannot be used interchangeably with medial and lateral excursions, as suggested by previous findings (Bates et al., 1981; Giakas et al., 1996). Increased amount of pronation does not result in increased medial or lateral excursion. In support of this idea, our results also showed that the absolute difference between the peak medial ML-GRF and the peak lateral ML-GRF was less in the high pronation and low pronation groups compared with the middle pronation group in the shod condition (i.e., there was no increase as a function of the increased amount of pronation). However, these results suggest that the absolute difference between the peak medial ML-GRF and the peak lateral ML-GRF is sensitive to any deviation from neutral pronation (i.e., either high pronation or low pronation), and not exclusively to an excessive amount of eversion. Furthermore, the fact that the middle pronation group showed lower absolute difference between the peak medial ML-GRF and the peak lateral ML-GRF in the shod condition compared with the barefoot condition supports that the ML-GRF are considerably influenced by footwear.

Lastly, it was hypothesized that the time to peak lateral ML-GRF will occur about the same time that maximum eversion occurs indicating a closer linkage between the two. This hypothesis was not supported by our results. Based upon relative times of occurrence, the peak medial ML-GRF always occurred after the peak lateral ML-GRF, and the maximum eversion always occurred after the peak medial ML-GRF. From these findings, it can be concluded that the peak medial ML-GRF was more closely time linked to maximum eversion than peak lateral ML-GRF. This was found to be more evident in the barefoot condition. Interestingly, Messier et al. (1991) found that the time to peak lateral ML-GRF was a significant discriminator between controls and a group that exhibited patellofemoral pain (i.e., exhibiting higher pronation). The time to peak lateral ML-GRF occurred significantly later during stance for the injured group compared with a control group. Therefore, our results and Messier et al.'s results underscore the importance of maximum eversion as an important clinical descriptor.

Impulses associated with the peak medial ML-GRF and the peak lateral ML-GRF were measured to further explain patterns of change in the mediolateral forces with runners who pronate either more or less than middle pronation strikers. Based on the findings of Bates et al. (1982), it was expected that the impulses associated with medially directed forces would be higher in the low pronation group compared with the middle pronation and high pronation groups. Bates et al. (1981) found that impulses associated with the peak medial ML-GRF corresponded to decreased maximum eversion. This finding was not supported by our results since neither impulses associated with the peak medial ML-GRF nor impulses associated with the peak lateral ML-GRF differed between groups. However, wearing shoes had a significant increasing effect on both impulses associated with the peak medial ML-GRF in the low pronation group and impulses associated with the peak lateral ML-GRF in the high pronation group. These results support that shoes function to allow for forces to be absorbed over a longer period of time.

There were some limitations in our study that need to be considered when interpreting the findings. A first limitation was the use of a two-dimensional versus three-dimensional analysis: the rearfoot motion was captured in a frontal plane analysis manner. However, Hamill et al. (1992, 1994) reported, based on the Areblad et al. (1990) study, that differences between the two types of analysis are minimal at midstance but increase as the foot moves out of plane especially during the latter portion of the stance phase (from 80% to the end of the stance phase). Therefore, Hamill et al. (1994) suggested that variables such as maximum eversion, heel and leg angles and times to these events are valid for reporting rearfoot motion. Results from the current study are similar in value to those reported by others (Clarke et al., 1984; Nigg, 1986) and the critical events occurred between 15% and 65% of the stance phase. Based on these observations, the authors do not feel that a two-dimensional analysis presents a serious limitation. A second limitation of this study was the use of a 60-Hz camera to collect running data: the accuracy of determining eversion was limited to the amount of data collected per second. However, our eversion values were comparable with other studies reported in the literature (Bates et al., 1978, 1979; Clarke et al., 1984; Edington et al., 1990; Hamill et al., 1992; Nigg, 1986; Nigg & Morlock, 1987) and thus the equipment available did not hinder our conclusions. Another possible limitation of the study was that a more sensitive grouping variable may have been the maximum velocity of eversion/pronation. However, we decided to use the maximum eversion value due to the existence of previous literature that has established clinically relevant limits of pronation (Clarke et al., 1984). Future studies should be conducted to verify our findings with both higher speed cameras and a three-dimensional analysis. Furthermore, although our findings provide the basis for further studies with respect to excessive rearfoot eversion motion, barefoot running, and ML-GRF, further research must be performed in this area before knowledge gained can be used for practical applications (e.g., injury prevention and gait rehabilitation).

In conclusion, results obtained from this study were analyzed with regard to proposed hypotheses to formulate the following conclusions. (1) The barefoot condition resulted in decreased maximum eversion and time to maximum eversion across all groups and increased peak lateral ML-GRF (as well as larger absolute difference between the peak medial ML-GRF and the peak lateral ML-GRF) in the middle pronation group. These results would imply that shoe functions to decrease ML-GRF to improve side-to-side mobility. (2) Increased amount of pronation does not result in increased medial or lateral excursion. Therefore, pronation and supination should not be used interchangeably with medial and lateral. (3) The peak medial ML-GRF occurred closer to maximum eversion than peak lateral ML-GRF. This was found to be more evident in the barefoot condition.

Acknowledgments

The study was supported by the University Committee on Research of the University of Nebraska at Omaha and the Nebraska Research Initiative. We would also like to thank Paul DeVita and Steve McCaw for their critical reading of earlier drafts of this paper.

References

Albers, D., & Hoke, B.R. (2003). Techniques in Achilles tendon rehabilitation. *Techniques in Foot & Ankle Surgery, 2,* 208–219.

Areblad, M., Nigg, B.M., Ekstrand, J., Olsson, K.O., & Ekstrom, H. (1990). Three-dimensional measurement of rearfoot motion during running. *Journal of Biomechanics, 23,* 933–940.

Asplund, C., & Tanner, S.M. (2004). The runner. In R.B. Birrer & F.G. O'Connor (Eds.), *Sports Medicine for the Primary Care Physician* (pp. 223–231). Boca Raton, FL: CRC Press.

Bates, B.T., James, S.L., & Osternig, L.R. (1978). Foot function during the support phase of running. *Running, 3,* 24–30.

Bates, B.T., James, S.L., Osternig, L.R., & Sawhill, J. (1981). On the effects of running shoes on ground reaction forces. In A. Morecki, K. Fidelus, K. Kedzior, & A. Wit (Eds.), *Biomechanics VII-B* (pp. 226–234). Baltimore, MD: University Park Press.

Bates, B.T., James, S.L., Osternig, L.R., Sawhill, J., & Hamill, J. (1982). Effects of running shoes on ground reaction forces. In A. Morecki & K. Fidelus (Eds.), *Biomechanics VII*. Baltimore, MD: University Park Press.

Bates, B.T., Osternig, L.R., Mason, B., & James, L.S. (1979). Foot orthotic devices to modify selected aspects of lower extremity mechanics. *American Journal of Sports Medicine, 7,* 338–342.

Bates, B.T., Osternig, L.R., Sawhill, J.A., & James, S.L. (1983). An assessment of subject variability, subject-shoe interaction, and the evaluation of running shoes using ground reaction force data. *Journal of Biomechanics, 16,* 181–191.

Buchbinder, M.R., Napora, N.J., & Biggs, E.W. (1979). The relationship of abnormal pronation to chondromalacia of the patella in distance runners. *Journal of the American Podiatry Association, 69,* 159–162.

Busseuil, C., Freychat, P., Guedj, E.B., & Lacour, J.R. (1998). Rearfoot-forefoot orientation and traumatic risk for runners. *Foot & Ankle International, 19,* 32–37.

Cavanagh, P.R. (1987). The biomechanics of lower extremity action in distance running. *Foot & Ankle International, 7,* 197–217.

Cavanagh, P.R., & Lafortune, M.A. (1980). Ground reaction forces in distance running. *Journal of Biomechanics, 13,* 397–406.

Cheung, R.T.H., Ng, G.Y.F., & Chen, B.F.C. (2006). Association of footwear with patellofemoral pain syndrome in runners. *Sports Medicine (Auckland, N.Z.), 36,* 199–205.

Chu, M.L., Yazdani-Ardakani, S., Gradisar, I.A., & Askew, M.J. (1986). An in vitro simulation study of impulsive force transmission along the lower skeletal extremity. *Journal of Biomechanics, 19,* 979–987.

Clarke, T.E., Frederick, E.C., & Hamill, C.L. (1984). The study of rearfoot movement in running. In E.C. Frederick (Ed.), *Sport Shoes and Playing Surfaces* (pp. 166–189). Champaign, IL: Human Kinetics Publishers.

Clement, D.B., Taunton, J.E., & Smart, G.W. (1984). Achilles tendinitis and peritendinitis: etiology and treatment. *American Journal of Sports Medicine, 12,* 179–184.

Cook, S.D., Brinker, M.R., & Poche, M. (1990). Running shoes: Their relationship to running injuries. *Sports Medicine (Auckland, N.Z.), 10,* 1–8.

Edington, C.J., Frederick, E.C., & Cavanagh, P.R. (1990). Rearfoot motion in distance running. In P.R. Cavanagh (Ed.), *Biomechanics of Distance Running* (pp. 135–164). Champaign, IL: Human Kinetics Publishers.

Freychat, P., Belli, A., Carret, J.P., & Lacour, J. (1996). Relationship between rearfoot and forefoot orientation and ground reaction forces during running. *Medicine and Science in Sports and Exercise, 28,* 225–232.

Giakas, G., Baltzopoulos, V., Dangerfield, P., Dorgan, J.C., & Dalmira, S. (1996). Comparison of gait patterns between healthy and scoliotic patients using time and frequency domain analysis of ground reaction forces. *Spine (Phila Pa 1976), 21,* 2235–2242.

Hamill, J., & Bates, B.T. (1988). A kinetic evaluation of the effects of in-vivo loading on running shoes. *The Journal of Orthopaedic and Sports Physical Therapy, 10,* 47–53.

Hamill, J., Bates, B.T., & Holt, K.G. (1992). Timing of lower extremity joints actions during treadmill running. *Medicine and Science in Sports and Exercise, 24,* 807–813.

Hamill, J., Bates, B.T., Knutzen, K.M., & Kirkpatrick, G.M. (1989). Relationship between selected static and dynamic lower extremity measures. *Clinical Biomechanics (Bristol, Avon), 4,* 217–225.

Hamill, J., Knutzen, M., Bates, B.T., & Kirkpatrick, G. (1996). Evaluation of two ankle appliances using ground reaction force data. *The Journal of Orthopaedic and Sports Physical Therapy, 7,* 244–249.

Hamill, J., Milliron, M.J., & Healy, J.A. (1994). Stability and rearfoot motion testing: a proposed standard. *Proceedings of the Eighth Biennial Conference of the Canadian Society for Biomechanics*, 324–325.

Harris, G.F. (1991). Analysis of ankle and subtalar motion during human locomotion. In J.B. Stiehl (Ed.), *Inman's Joints of the Ankle* (pp. 75–84). Baltimore, MD: Williams and Wilkins.

Hintermann, B., & Nigg, B.M. (1998). Pronation in runners: Implications for injuries. *Sports Medicine (Auckland, N.Z.), 26,* 169–176.

Holden, J.P., & Cavanagh, P.R. (1991). The free moment of ground reaction in distance running and its changes with pronation. *Journal of Biomechanics, 24,* 887–897.

Hreljac, A. (2004). Impact and overuse injuries in runners. *Medicine and Science in Sports and Exercise, 36,* 845–849.

Hreljac, A. (2005). Etiology, prevention, and early intervention of overuse injuries in runners: a biomechanical perspective. *Physical Medicine and Rehabilitation Clinics of North America, 16,* 651–667.

Hreljac, A., Marshall, R.N., & Hume, P.A. (2000). Evaluation of lower extremity overuse injury potential in runners. *Medicine and Science in Sports and Exercise, 32,* 1635–1641.

Jackson, K.M. (1979). Fitting of mathematical functions to biomechanical data. *IEEE Transactions on Bio-Medical Engineering, 26,* 515–521.

James, S.L., Bates, B.T., & Osternig, L.R. (1978). Injuries to runners. *American Journal of Sports Medicine, 6,* 40–50.

James, S.L., & Jones, D.C. (1990). Biomechanical aspects of distance running injuries. In P.R. Cavanagh (Ed.), *Biomechanics of Distance Running*. Champaign, IL: Human Kinetics.

Kirby, K.A. (2001). Subtalar joint axis location and rotational equilibrium theory of foot function. *Journal of the American Podiatric Medical Association, 91,* 465–487.

Kurz, M.J., & Stergiou, N. (2004). Does footwear affect ankle coordination strategies? *Journal of the American Podiatric Medical Association, 94,* 53–58.

Lawrence, R.M. (1997). Why run? In G.N. Guten (Ed.), *Running injuries* (pp. 1–3). Philadelphia, PA: W.B. Saunders.

Mann, R.W., Baxter, D.E., & Lutter, L.D. (1981). Running symposium. *Foot & Ankle, 1,* 190–224.

McClay, I. (2000). The evolution of the study of the mechanics of running. Relationship to injury. *Journal of the American Podiatric Medical Association, 90,* 133–148.

McClay, I., & Manal, K. (1998). A comparison of three-dimensional lower extremity kinematics during running between excessive pronators and normals. *Clinical Biomechanics (Bristol, Avon), 13,* 195–203.

Messier, S.P., Davis, S.E., Curl, W.W., Lowery, R.B., & Pack, R.J. (1991). Etiologic factors associated with patellofemoral pain in runners. *Medicine and Science in Sports and Exercise, 23,* 1008–1015.

Messier, S.P., Edwards, D.G., Martin, D.F., Lowery, R.B., Cannon, D.W., James, M.K., et al. (1995). Etiology of iliotibial band friction syndrome in distance runners. *Medicine and Science in Sports and Exercise, 27,* 951–960.

Messier, S.P., & Pittala, K.A. (1988). Etiologic factors associated with selected running injuries. *Medicine and Science in Sports and Exercise, 20,* 501–505.

Miller, D.I. (1990). Ground reactions forces in distance running. In P. Cavanagh (Ed.), *Biomechanics of Distance Running* (pp. 203–224). Champaign, IL: Human Kinetics Publishers.

Munro, C.F., Miller, D.I., & Fuglevand, A.J. (1987). Ground reaction forces in running: a reexamination. *Journal of Biomechanics, 20,* 147–155.

Neely, F.G. (1998). Biomechanical risk factors for exercise-related lower limb injuries. *Sports Medicine (Auckland, N.Z.), 26,* 395–413.

Nigg, B.M. (1985). Biomechanics, load analysis and sports injuries in the lower extremities. *Sports Medicine (Auckland, N.Z.), 2,* 367–379.

Nigg, B.M. (1986). *Biomechanics of running shoes*. Champaign, IL: Human Kinetics Publishers.

Nigg, B.M., & Morlock, M. (1987). The influence of lateral heel flare of running shoes on pronation and impact forces. *Medicine and Science in Sports and Exercise, 19,* 294–302.

Perry, S.D., & Lafortune, M.A. (1995). Influences of inversion/eversion of the foot upon impact loading during locomotion. *Clinical Biomechanics (Bristol, Avon), 10,* 253–257.

Reinschmidt, C., & Nigg, B.M. (2000). Current issues in the design of running and court shoes. *Sportverletzung Sportschaden, 14,* 71–81.

Reinschmidt, C., van Den Bogert, A.J., Murphy, N., Lundberg, A., & Nigg, B.M. (1997). Tibiocalcaneal motion during running, measured with external and bone markers. *Clinical Biomechanics (Bristol, Avon), 12,* 8–16.

Sangeorzan, B.J. (1991). Biomechanics of the subtalar joint. In J.B. Stiehl (Ed.), *Inman's Joints of the Ankle* (pp. 65–73). Baltimore, MD: Williams and Wilkins.

Stacoff, A., Kalin, X., & Stussi, E. (1991). The effects of shoes on the torsion and rearfoot motion in running. *Medicine and Science in Sports and Exercise, 23,* 482–490.

Stacoff, A., Nigg, B.M., Reinschmidt, C., van den Bogert, A.J., & Lundberg, A. (2000). Tibiocalcaneal kinematics of barefoot versus shod running. *Journal of Biomechanics, 33,* 1387–1395.

Stergiou, N., & Bates, B.T. (1997). The relationship between subtalar and knee function as a possible mechanism for running injuries. *Gait & Posture, 6,* 177–185.

Stergiou, N., Bates, B.T., & James, S.L. (1999). Asynchrony between subtalar and knee joint function during running. *Medicine and Science in Sports and Exercise, 31,* 1645–1655.

Stergiou, N., Bates, B.T., & Kurz, M.J. (2003). Subtalar and knee joint interaction during running at various stride lengths. *Journal of Sports, Medicine, and Physical Fitness, 43,* 319–326.

Van Mechelen, W. (1992). Running injuries. A review of the epidemiological literature. *Sports Medicine (Auckland, N.Z.), 14,* 320–335.

Viitasalo, J.T., & Kvist, M. (1983). Some biomechanical aspects of the foot and ankle in athletes with and without shin splints. *American Journal of Sports Medicine, 11,* 125–130.

Wen, D.Y., Puffer, J.C., & Schmalzried, T.P. (1997). Lower extremity alignment and risk of overuse injuries in runners. *Medicine and Science in Sports and Exercise, 20,* 1291–1298.