The Effect of Foot Orthotics and Gait Velocity on Lower Limb Kinematics and Temporal Events of Stance

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Darl Vander Linden, PhD, PT

The stance phase of gait is a closed chain lower extremity activity that requires coordinated movement between the proximal and distal joints (4, 10, 11, 21, 28, 32, 34, 39). The foot performs many essential dynamic functions during the stance phase that enable the body to progress forward in normal walking (27). Functional movement of the foot from a mobile adaptor to a rigid lever for propulsion depends on a balance between pronation and supination of the subtalar joint during the appropriate portions of stance. This ensures proper biomechanical functioning of the foot and normal associated movement at the ankle and knee (16). The pathomechanics of gait manifest clinically as a variety of musculoskeletal disorders, such as patellar pain syndrome, shin splints, planter fascitis, sinus tarsitis, and numerous foot and leg tendinitises (10).

Foot orthotics are intended to restore dynamic stability and reduce compensatory pronation of the subtalar joint during the initial stance phase of gait. Precise subtalar joint motion is essential to avoid stress to the ankle and proximal joints of the limb (31). Rapid pronation occurs immediately after heel contact, and then progressive supination continues until terminal stance (19). Pronation that continues beyond midstance prevents the conversion of the foot into a rigid lever for propulsion and imposes functional limitations upon the joint structures of the lower limb as the foot prepares to leave the ground (4, 39, 40). Orthotic devices are designed to control the amount, rate, and temporal sequence of subtalar joint movement and restore normal biomechanical relationships in the lower extremity during stance (3, 8, 12, 27, 33, 36, 39, 42). Thus, clinical symptoms that develop in the lower limb as a result of joint laxity and abnormal muscle activity are supposedly alleviated.

Several studies have shown that...
Foot orthotics are intended to restore dynamic stability and reduce compensatory pronation of the subtalar joint during the initial stance phase of gait.

Orthotic devices reduce abnormal pronation and clinical symptoms in the lower limb in groups of runners (3, 12, 33, 36, 42). For example, during treadmill running, orthotics have been shown to significantly modify selected aspects of lower extremity mechanics during stance and reduce the period and amount of pronation (3). Reductions in maximum pronation were coincident with decreased maximum ankle dorsiflexion and knee flexion during stance. In a group of orienteers and runners, orthotic foot support reduced excessive mobility of the rearfoot in stance and alleviated pain and discomfort in the leg (42). Orthotic inserts used by runners to control abnormal pronation were found to be similarly effective in decreasing the maximum pronation angle and the total period of pronation (33). Results of a study on the effects of soft and semirigid orthotics on rearfoot movement in a group of treadmill runners demonstrated significantly reduced rate and amount of calcaneal eversion with use of the orthotic, although both the rate and amount increased as speed of running increased (36). Similar results were documented in a study of shoe design and rearfoot movement (8). Shoe design that included additional medial support along the longitudinal arch of the foot reduced the mean pronation of subjects with painful lower extremity pathomechanics to levels below those shown by symptom-free subjects in unmodified shoes. These data reinforced the effectiveness of the medial support in repositioning the rearfoot and eliminating the need for excessive compensatory subtalar joint pronation. A recent motion analysis study documented the effectiveness of orthotic devices during the loading response of gait (27). Results concurred with the previous studies that used high-speed filming techniques to capture rearfoot motion in running. Reduced calcaneal eversion and total rearfoot movement and reduced calcaneal eversion velocity and force moments resulted from orthotic support during the initial loading response.

During the initial contact phase of stance, the subtalar joint moves from a supinated to a pronated position due to the ground reaction forces imposed upon the calcaneus (4, 32). Subsequent subtalar joint pronation, consisting of the triplanar motion of calcaneal eversion and talar adduction and plantarflexion, continues in response to the flexion moment at the knee (13, 26). Additionally, automatic internal rotation of the tibia occurs as the knee flexes during initial contact (40). The talus rotates anteromedially and caudally to absorb the transverse rotational movement of the tibia in the ankle mortise (12, 34). The talus, therefore, functions as a torque convertor for the lower extremity primarily due to the joint congruity between the trochlea and ankle mortise (34). The motions of subtalar joint pronation and tibial internal rotation and flexion are, thus, interdependent and necessary for normal kinematics of the lower extremity during gait. Excessive, prolonged subtalar joint pronation during the initial loading response necessitates additional internal rotation and inclination of the tibia and flexion of the knee to maintain talocrural joint congruency (4, 10, 13, 26, 28, 39, 40). During midstance to terminal stance position, extension of the tibiofemoral joint is delayed or reduced. Torsional stress is created as the lower extremity attempts extension while the tibia is prolonged in medial inclination. Tiberio (40) suggests that this biomechanical dilemma to the locomotor system during stance is solved in the form of compensation in the lower extremity kinetic chain. Theoretically, the tibiofemoral joint may compensate by producing increased femoral internal rotation, which will reduce abnormal shear force at the knee but may alter patellofemoral tracking. The disrupted patellar mechanism will increase compression between the lateral articular surface of the patella and the femoral condyle and contribute to patellofemoral dysfunction.

A 4:4:1 biomechanical relationship explains the relative contributions of the tibiofemoral, subtalar, and talocrural joints during closed chain activity (26). For every 4° of calcaneal eversion, there are 4° of talar adduction and tibial internal rotation and 1° of talar plantarflexion. This 1:1 relationship between subtalar adduction and tibial internal rotation occurs about a normal subtalar joint axis (13) and necessitates forward movement of the fixed leg in association with increased knee flexion (34). Additionally, the tibia internally rotates by one half the amount of knee flexion during the first 15° of stance phase knee motion (5). Dorsiflexion at the talocrural joint has been shown to produce substantial internal rotation about the vertical axis of the leg and pronation of the foot (18). Ten degrees of dorsiflexion produced an equal amount of rotation of the leg during loaded testing. Thus, from initial contact to midstance, foot pronation and tibial internal rotation of equivalent amounts occur simultaneously with a predictable amount of stance phase knee flexion (26, 40).

The mechanics of prolonged, excessive subtalar joint pronation dur-
ing the stance phase of gait require additional internal rotation of the tibia and resulting increased knee flexion due to the osseous configuration and axes of motion at the subtalar and talocrural joints (40). The theoretical argument proposed by Tiberio (40) suggests a compensatory reaction at the tibiofemoral joint in the form of femoral internal rotation to match the movement of the tibia. The compensated model of femoral internal rotation maintains the normal biomechanical relationship of the tibiofemoral joint during midstance. This compensatory movement enables the lower extremity to complete the extension phase of stance in preparation for terminal stance and continued forward progression, thus, reducing torsion and shear stress at the knee. Orthotic control of the subtalar joint has been shown to reduce abnormal pronation during the stance phase of running and walking, and, theoretically, subsequent movement at the tibiofemoral joint should be affected.

Each of the previous studies has provided useful biomechanical and quantitative information on lower extremity joint movement during human locomotion. Kinetic and kinematic parameters of walking and running have been investigated, and detailed joint movements have been described. These studies have documented selected arthrokinematic variables of the subtalar and tibiofemoral joints during stance, gait, and running. The effects of orthotic intervention on the articulating dynamics of the subtalar joint during running and, more recently, walking have been studied by many authors (3, 12, 27, 33, 36, 42). The normal functional biomechanics of the tibiofemoral joint have also been described in both running and walking (3, 6, 7, 9, 22, 25, 30, 36). Fast walking, comparable to slow running, has been shown to produce increased knee flexion as a shock absorption mechanism due to the more forceful onset of weight bearing, increased joint loading forces, and muscular demands (22, 25). Orthotic devices that control excessive pronation should, therefore, reduce the amount of knee flexion during stance and eliminate the need for compensatory internal rotation of the femur. The interactive effect of orthotics on the knee, however, has not been studied in normal walking. In addition, the studies that documented the effect of orthotics on rearfoot motion (3, 27) did not study the effects of speed of walking and running on the control of foot or knee motion. Empirical data suggest that orthotics mitigate symptoms of lower extremity pain during gait; yet, actual data that quantify combined subtalar and tibiofemoral joint kinematics are unavailable. The combined effects of speed and rearfoot orthotic control on joint movement at the ankle and knee are not detailed in any of the previous studies (3, 12, 27, 33, 36, 42).

The purpose of this study was to investigate the effects of speed and functional orthotic control of the subtalar joint on movement at the ankle and knee during the stance phase of gait. It was expected that, with the use of foot orthotics, there would be a proportional decrease in the angular movement of the rearfoot, ankle, and knee as well as rate and duration of pronation during stance. It was also hypothesized that orthotic control of the foot and changes in rate and amount of rearfoot, ankle, and knee movement would be proportional to changes in speed of gait.

METHODS

Subjects

Ten subjects, five males and five females, between the ages of 20 and 45 years participated in the study. The individuals were recruited from local orthopaedic specialists, podiatrists, and physical therapists. The selected group sought professional treatment for activity-related foot or knee pain secondary to forefoot varus and excessive subtalar joint pronation. All subjects had functional foot orthoses prescribed to correct the biomechanical dysfunction. Seven subjects wore rigid orthotic inserts, and three used semirigid devices. Each participant received a lower extremity musculoskeletal examination, as described by McPoil and Brocato (20). Criteria for inclusion in the study were normal and symmetrical muscle strength, flexibility, and range of motion in the lower quarter; a minimum of 3° of calcaneal eversion of the right foot in relaxed stance; a minimum of 6° forefoot varus of the right foot in prone subtalar neutral; and participation in moderate aerobic exercise 3–5 days per week. Subject data are shown in Table 1.

Equipment

The Motion Analysis™ Expert Vision™ system (Motion Analysis Corporation, 3650 North Laughlin Road, Santa Rosa, CA 95403) and four NEC TI-22P CCD (NEC America, Inc., 1255 Michael Drive, Wood Dale, IL 60191, 312860-7600) video
Reduced calcaneal eversion was documented with orthotic use throughout the stance phase of gait.

and yielded an intraclass correlation coefficient of .99. Within-trial variability for angle calculation was less than 0.4° in all locations, although variability in the accuracy of recorded motion increased with extremes of joint motion and angulation to the field of view of the cameras. The system-calculated distance between markers during dynamic tests of gait yielded a within-trial variability of 2.16 mm, compared with the manufacturer’s claim that a reflective marker could be estimated within 2 mm of its location.

Procedures

An informed consent form with complete explanations of the research procedure was provided for each subject. Reflective markers were placed at appropriate joint positions along the right lower extremity following established protocols. Adhesive-backed markers were positioned at the following sites: two markers were placed on the femur and two were positioned on the tibia, equidistant from the knee joint center and approximately 5 cm apart, along the lateral aspect of the lower extremity on a line that connects the greater trochanter to the lateral malleolus. Two markers were also placed on the distal one-third of the posterior leg. The knee was held in maximum extension during marker placement by active quadriceps contraction. Markers were placed on the shoe to duplicate the position of selected anatomical landmarks. Two markers were placed on the calcaneus approximately 5 cm apart and along the plane of the foot, one directly below the lateral malleolus and the other on the fifth metatarsal head. An illustration of the marker placement system is shown in Figure 1.

The video cameras were optimally positioned along the right side and rear of the treadmill in order to capture motion data from the reflective markers placed along the lower right limb of each subject. Several trial walking sessions were performed by each subject to familiarize the individual with the treadmill and

<table>
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<tr>
<th>AGE</th>
<th>SEX</th>
<th>STN</th>
<th>PROM-I</th>
<th>PROM-E</th>
<th>FF</th>
<th>FS</th>
<th>DF-KE/KF</th>
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<td>31.6</td>
<td>5-M</td>
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<td>11.0°</td>
<td>8.2° V</td>
<td>1.8° V</td>
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<td>(1.7°)</td>
<td>(8.3°)</td>
<td>(3.9°)</td>
<td>(1.7°)</td>
<td>(0.9°)</td>
<td>(1.9/1.4°)</td>
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</table>

STN = subtalar neutral position, V = varus, PROM = passive range of motion, I = inversion, E = eversion, FF = forefoot, TS = neutral tibial stance, DF = dorsiflexion, KE = knee extension, KF = knee flexion.

TABLE 1. Means (SD) for subject demographics.
testing procedures. The treadmill was activated at the designated speeds, 2 and 3 mph, prior to actual data collection, and the subject was instructed to begin walking. Each subject walked continuously on the treadmill while three 5-second trials were recorded at the two selected speeds. Each subject served as his/her own control and was tested with and without orthotic inserts in personal athletic shoes. A 2-minute rest period was allowed between speed changes. A representative average of five gait cycle recordings for each speed condition was used for analysis.

Data Analysis

Specific data points from the angle plots of segmental joint motion and the temporal data from the OrthoTrack™ were statistically analyzed in a two-factor design with independent variables of orthotic control and speed. Dependent variables were 1) time to maximum pronation and heel rise expressed as a percentage of the stance phase; 2) rate of pronation during the first and second 10% of stance; and 3) maximum pronation, ankle dorsiflexion, and knee flexion. A two-factor, repeated measures analysis of variance was used to analyze the angular and temporal data. For significant interactions, appropriate univariate tests were applied to determine simple main effects.

According to previous research, a discrepancy exists between movement of the calcaneus and that of the shoe (8, 27, 36), even though the calcaneus and shoe’s heel follow similar angular excursions (2). We also noted this discrepancy in rearfoot measurement between relaxed barefoot stance and stance while in the athletic shoe. However, minimal discrepancy existed using the same measurements with the insertion of the orthotic. Therefore, an adjustment to the angular rearfoot measures of the nonorthotic condition was made to account for this offset. To make this adjustment, we adopted a normalization procedure previously reported by Novick and Kelly (27). This normalization value represents the difference between the measurements of relaxed calcaneal stance barefoot and in the shoe and orthotic stance barefoot and in the shoe. Group means of 5° eversion for relaxed calcaneal stance and 1.7° for orthotic stance were determined. This provided a mean normalization value of 3.3° compared with 2.39° determined in previous research (27).

RESULTS

Temporal Changes

There was a main effect for orthotic for the relative period of stance from maximum pronation to heel rise ($F = 7.50; df = 1.9; p < .05$). The duration of this period of stance increased with insertion of the orthotic device. There was a main effect for speed for the rate of rearfoot pronation during the first 10% of stance ($F = 14.69; df = 1.9; p < .01$) and second 10% of stance ($F = 24.98; df = 1.9; p < .01$). Rearfoot velocity increased during both the first and second portions of initial pronation as walking speed increased, although no orthotic effect was noted. A summary of the above data is shown in Table 2.

Rearfoot Angular Changes

There was a main effect for orthotic at heel strike ($F = 17.45; df = 1.9; p < .01$). The rearfoot angle at heel strike was 1.08° of inversion without the orthotic compared with 4.39° with the orthotic. There was also a main effect for orthotic for rearfoot angle at maximum pronation ($F = 18.47; df = 1.9; p < .01$), heel rise ($F = 22.14; df = 1.9; p < .01$), and toe off ($F = 13.57; df = 1.9; p < .01$). Angular position of the rearfoot at maximum pronation changed from an everted position of 10.30 to 6.95° with orthotic usage. There was a reduction in the position of the rearfoot at heel rise from 6.17 to 2.12° eversion in the orthotic condition, and, at toe off, angular position changed from 2.15 to 5.60° of inversion with use of the orthotic.

Ankle Angular Changes

There was a significant interaction effect ($F = 17.98; df = 1.9; p < .01$) for maximum dorsiflexion. Univariate analysis indicated significant difference at the fast speed between the cell means for the orthotic and nonorthotic conditions ($F = 63.43; df = 1.9; p < .01$) (Figure 2). A difference in angular movement also existed between the two speeds in the nonorthotic condition ($F = 14.37; df = 1.9; p < .01$). However, there was no significant difference between the fast and slow speed, in the orthotic condition or between the orthotic conditions at the slow speed. At the fast speed, maximum dorsiflexion increased from 10.56° without the orthotic to 11.53° with the device. Peak dorsiflexion decreased from 12.78 to 10.56° in the nonorthotic condition as speed increased.

<table>
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<tr>
<th></th>
<th>No-orthotic</th>
<th>Orthotic</th>
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<tr>
<td></td>
<td>Slow</td>
<td>Fast</td>
</tr>
<tr>
<td>Maximum pronation</td>
<td>54  (15)</td>
<td>50  (13)</td>
</tr>
<tr>
<td>Heel rise</td>
<td>80  (4)</td>
<td>78  (5)</td>
</tr>
<tr>
<td>1st 10%</td>
<td>46.20 (17.65)</td>
<td>61.50 (22.27)</td>
</tr>
<tr>
<td>2nd 10%</td>
<td>54.40 (23.70)</td>
<td>64.50 (26.32)</td>
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TABLE 2. Means (SD) for time to maximum pronation and heel rise expressed as a percentage of stance time and velocity of pronation during the first and second 10% of stance in °/sec.
FIGURE 2. Significant orthotic by speed interaction of the ankle at maximum dorsiflexion. Standard deviations are indicated in parentheses. Levels of significance are *# = p < .01.

Knee Angular Changes

There was a significant main effect for speed for maximum knee flexion ($F = 13.33$; $df = 1.9$; $p < .01$). Stance knee position changed from the initial contact angle to $5.73^\circ$ of flexion at the slow speed to $10.37^\circ$ at the fast speed when averaged over conditions. Figure 3 shows the angular movement of the rearfoot, ankle, and knee in one subject for the no orthotic condition at the faster speed of gait. The angle plots illustrate the various contact positions during the gait cycle. Each angle plot represents two out of five gait cycles.

DISCUSSION

Temporal Changes

Results of this study indicated a significant increase in the relative duration of maximum pronation to heel rise for the orthotic condition. A foot with structural abnormalities compensates during stance with initial excessive pronation when compared with movement that occurs in the normal foot. Normal progression from pronation to supination then occurs gradually at $50\%$ of stance (13). This increase in time to heel rise is the result of less pronation with the use of the orthotic, where peak pronation occurred proportionally earlier in stance, as well as the tendency for increased dorsiflexion with orthotic use. In essence, this delay through midstance lengthens the time interval for resupination of the foot from the maximally everted position to the inverted position at toe off. The device also positions the foot in a relatively plantarflexed orientation during stance (3). With the foot stable, fixed, and reoriented on the orthotic, the tibia is now able to move forward with the body for a longer period of time before maximum dorsiflexion is reached, and the heel is forced to leave the support surface (29).

As expected, speed of walking demonstrated a significant effect on the rate of pronation during the first and second $10\%$ of the stance phase. Pronation is rapid once the heel strikes the floor. The velocity of movement gradually lessens and then stops when the full foot is in contact with the walking surface. The calcaneus, which everts during pronation, has been shown to move significantly faster at fast speeds (36) and at a slower velocity of movement when orthotics are worn (27). During the first and second portions of initial pronation at the fast speed, velocity of movement was $62.25$ and $65.05^\circ$/sec, respectively (Table 2). This represented an increase in rate of pronation compared with the slow speed. These values are, however, substantially less than the values reported previously for running and normal walking with orthotics, which were $464.00$ and $91.00^\circ$/sec, respectively (27, 36). The differences noted in velocity may be due to measurement technique. In our study, velocity was measured from the time of heel contact through the first $20\%$ of the stance phase. The calcaneus moved from the inverted angular position at heel strike to vertical and then continued to maximum eversion. The average contact angle in the orthotic condition in this study was $4.30^\circ$ of inversion. In previous research, however, velocity was measured from the vertical position of the calcaneus to maximal eversion (27), hence omitting the zero velocity of the calcaneus at heel strike from any averaged calculation. Another factor that affects the rate of pronation is the athletic shoe sole
composition and heel flare. It was shown in a study of three shoe designs that a rounded or less flared heel and a soft composition sole reduce the initial pronation velocity during running (24). Other studies that found reduced pronation velocities used either one shoe model for testing that consisted of a combination, curved last, medium firmness midsole, and a flat outersole (36) or used shoes with a firm heel counter and provided extra material for shoes with uneven sole wear to restore a flat surface (27). The shoes of each participant in the present study were inspected for signs of excessive wear, but no attempt was made to limit shoe style or make adjustments for individual wear patterns. Each subject participated with shoes that were in good condition and were constructed of various sole materials with heels of differing flare and dimensions that might have affected the velocity measures.

Rearfoot Angular Changes

The use of the orthotic device in walking effectively altered rearfoot movement during stance. The results of this study are in agreement with other studies that documented reduced maximum pronation with the use of a functional orthotic (8, 27, 33, 36). Maximum eversion angle, representing subtalar pronation via the position of the calcaneus, was reduced from 10.30 to 6.95° with orthotic control. Similar reductions of 3 to 4° in the calcaneal angle were found in studies that used rigid orthotic devices (3, 27), while studies that used soft or semirigid devices found less change in the eversion angle between conditions (8, 33, 36). Rigid devices are designed to provide motion control and functional support of the foot, semi-rigid orthotics provide motion control and some cushioning, and soft orthotics provide mostly cushioning and some functional control (36). Seventy-five percent of the subjects in this study wore rigid orthotics, and the remainder wore semirigid devices. The foot is held in a more favorable and functional position by the medial post of the orthotic, and excessive pronation is prevented. Total rearfoot movement, however, demonstrated less change with orthotic use in this study than was shown previously (27, 36). Results of this study showed that orthotic intervention reduced maximum pronation by 4.0° at the slow speed and 2.7° at the fast speed. This was accomplished by providing rearfoot stability, support, and a physical block to movement, but the repositioning of the foot did not appear to alter the total functional range of motion. Total rearfoot movement from the maximally inverted position to the maximally everted position of the heel demonstrated a very small change of 0.57° with the orthotic device. A similar change of 0.40° was found in total rearfoot movement with the orthotic during running (36). These findings differ from the change of 2.55° found for total rearfoot movement with the orthotic during walking reported in a previous study (27). The amount of total rearfoot movement was not affected by changes in walking speed, either with or without the orthosis, as has been reported previously during running (36).

Ankle Angular Changes

It has been suggested that during running, the orthotic device reorients and elevates the heel within the shoe, and, as a result, a strong relationship exists between pronation and maximum ankle dorsiflexion (3). The medial longitudinal arch is elevated as the foot rests upon the orthotic, and the medially deviated subtalar joint axis that contributes to pronation is realigned to a more lateral position (16). Dorsiflexion and subsequent ankle motion now occur from a relatively supinated position throughout stance. The movement of the tibia must also overcome the elevation of the heel upon the orthotic, which sets the foot in a slightly plantarflexed position and serves as a functional heel lift. The increased time to maximum dorsiflexion period found in running enables a more gradual initial pronation and a longer period for the leg to move over the fixed foot (3). A slight increase in the maximum dorsiflexion period was noted for the orthotic condition at the fast speed. The prolonged period until maximum dorsiflexion is supported by the increased time to heel rise documented in this study. Other studies have documented ankle dorsiflexion during slow, free, and fast walking (22, 23). Generally, at slow speeds, a greater portion of the trailing stance foot is in contact with the walking surface than at free or fast speeds (22). The leg moves over the fixed foot for a longer period of time during slow walking, and greater dorsiflexion occurs. The elevated and inverted position of the heel on the orthotic contributes to increased dorsiflexion due to a delay in heel rise. Increased dorsiflexion and time to heel rise were noted for the orthotic condition at both speeds. The orthotic device effectively controlled ankle and leg movement during walking, supported the foot during maximum pronation, and limited changes in dorsiflexion between speeds. This suggests a neutralizing effect for speed as the orthotic duplicated the position and kinematic responses of the ankle during slow walking. This may be an attempt to reduce the speed of ankle dorsiflexion in order to prolong movement of the tibia over the foot so that the body’s center of mass progresses smoothly over the base of support.

Knee Angular Changes

Speed influenced the angular changes that occurred at stance phase knee flexion. Increased knee flexion with increased speed has been noted in prior studies (3, 17,
22) and corresponds to changes in the velocity of lower limb movement during fast walking or running. Increased knee flexion during stance is related to the need for additional shock absorption at faster speeds (32). The slight amount of change in stance phase knee flexion documented in this study at slow speeds, 5.73° compared with 10.37° at fast speeds, agrees with others that showed little or no stance phase knee flexion with slow walking (17, 22).

Other studies have documented 9–11° of stance phase knee flexion during normal walking (30, 37, 38). The total amount of stance phase knee flexion at fast speeds documented in this study was less than values of 15 to 24° previously reported (17, 22, 30).

**Interaction of Rearfoot, Ankle, and Knee Angular Changes**

Although the orthotic did not reduce stance phase knee flexion at either treadmill speed, reductions in calcaneal eversion and, therefore, subtalar pronation did occur. Through our knowledge of biomechanics, we expect equivalent amounts of calcaneal eversion, talar adduction, and tibial internal rotation to occur during gait. A 1:2 relationship exists between tibial internal rotation and stance knee flexion in normal lower extremity alignment (5, 26). Accordingly, stance phase knee flexion of 20° would occur with 10° of corresponding distal leg and foot movement. Results of this study indicated that the orthotic provided a more normal biomechanical relationship between the foot and knee by reducing maximum calcaneal eversion. At the slow speed with the orthotic, calcaneal eversion was proportional to stance knee flexion in a 1.02:1 relationship, while at the slow speed without the orthotic, a 1.88:1 proportion was determined. The maximum eversion angle for the orthotic at the fast speed was 3.34° less than the knee angle, or a ratio 1:1.45, but at the fast speed in the nonorthotic condition, maximum eversion angle remained slightly more than stance knee flexion, providing a ratio of 1.03:1 (Table 3).

We can only assume that an equivalent change in the amount of tibial internal rotation and calcaneal eversion occurred, as we were unable to measure limb rotation due to software limitation.

According to Tiberio (40), a reduction in the amount of internal rotation of the tibia would reduce the amount of compensatory movement by the femur. Femoral movement follows the tibia as it remains internally rotated during early contact through midstance due to the biomechanical demands of pronation. Prolonged, excessive pronation will delay obligatory external rotation of the tibia that enables tibiofemoral joint extension during midstance as the body progresses over the fixed foot. Femoral internal rotation maintains the essential joint congruity to achieve full extension in stance. This mechanism enables the lower limb to accommodate prolonged pronation and reduce torsional stress at the knee. Thus, the relative rotational dynamics of the tibia and femur are maintained through stance. The proportional changes noted between stance phase knee flexion and tibial internal rotation in the orthotic condition are in agreement with this theoretical argument. The lack of an orthotic effect at the knee during stance flexion supports the argument that foot orthotics could reduce compensatory rotation of the femur by promoting normal functional movement of the foot, ankle, and leg.

**TABLE 3. Subject means (SD) for joint angles during stance in degrees.**

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<td>Maximum eversion angle</td>
<td>10.40 (5.04)</td>
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<tr>
<td>Total rearfoot movement</td>
<td>10.63 (2.61)</td>
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<tr>
<td>Maximum dorsiflexion</td>
<td>12.78 (2.77)</td>
<td>13.03 (2.53)</td>
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<tr>
<td>Stance knee flexion</td>
<td>5.54 (3.49)</td>
<td>5.93 (3.94)</td>
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CONCLUSION

Results of this study demonstrate that the use of orthotics during walking has a positive effect on rear foot, ankle, and knee motion. The orthotic device supported and stabilized the foot and provided a physical barrier to subtalar movement. Reduced calcaneal eversion was documented with orthotic use throughout the stance phase of gait. The relative duration of stance, particularly heel strike to heel rise, increased with use of the orthotic. The period from maximum pronation to heel rise also increased with orthotic use. This increase in relative durations of stance corresponded with increased dorsiflexion with orthotic use, which allowed a longer period for resupination following maximum pronation. The data from the study allowed us to extrapolate the amount of tibial internal rotation based on its relationship with calcaneal eversion and, hence, determine the ratio of knee flexion to tibial rotation. It was clear that the use of the orthotics normalized these biomechanical relationships by decreasing calcaneal eversion and, thus, approaching the normal 2:1 relationship between knee flexion and tibial internal rotation. Logically, the use of orthotics could, therefore, reduce the need for femoral compensation that maintains congruity at the tibiofemoral joint in response to increases in internal tibial rotation without orthotic use. These combined effects suggest that individuals who wear orthotics to control abnormal foot biomechanics may benefit from routine use during walking as well as running.
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REFERENCES

2. Bates B: Human Movement Studies, Department of Physical Education, University of Oregon, Eugene, OR (personal communication), 1992
13. Inman VT: The joints of the Ankle, Baltimore, MD: Williams & Wilkins Co, 1976
29. Phillips D: College of Podiatric Medicine, University of Osteopathic Medicine, Des Moines, IA (personal communication) 1992
34. Root ML, Oren WP, Weed HJ: Normal and Abnormal Function of the Foot, Los Angeles, CA: Clinical Biomechanics Corp, 1977