The Influence of Medial and Lateral Placement of Orthotic Wedges on Loading of the Plantar Aponeurosis

AN *IN VITRO* STUDY*

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Abstract

Background: **Repetitive trauma and overuse of the plantar aponeurosis are believed to be causal factors of plantar fasciitis. Therefore, it is important to know how an orthosis influences loading of the plantar aponeurosis. The aim of this study was to quantify strain in the plantar aponeurosis in cadaveric feet with the use of various combinations of orthotic wedges.**

Methods: **An *in vitro* test that simulated static stance was used to determine the loading characteristics of the plantar aponeurosis. A differential variable reluctance transducer was operatively implanted into the plantar aponeurosis of nine fresh-frozen cadaveric lower limbs. Each specimen was mounted in an electromechanical testing machine that applied an axial load of as much as 900 newtons to the tibia. Eight different combinations of test conditions, in which wedges (each with a 6-degree incline) were or were not positioned under the medial and lateral aspects of the forefoot and hindfoot, were evaluated, with the plantigrade foot used as a neutral control.**

Results: **Each of the test conditions that involved a wedge under the forefoot resulted in strain that was significantly different from that in the neutral control. A wedge under the lateral aspect of the forefoot decreased strain in the plantar aponeurosis, and a wedge under the medial aspect increased strain ($p < 0.05$). The test conditions that involved a wedge under the hindfoot but not under the forefoot resulted in strains that were not significantly different from those in the neutral control ($p > 0.05$).**

Conclusions: **A wedge under the lateral aspect of the forefoot transmits loads through the lateral support structures of the foot, locking the calcaneocuboid joint and decreasing strain in the plantar aponeurosis. A wedge under the medial aspect of the forefoot transmits loads through the medial support structures of the foot, which produces a truss-like action that increases strain in the plantar aponeurosis.**

Clinical Relevance: **Orthotic wedges seem to be effective in controlling the load-path pattern in the foot. The results of the tests involving a wedge under the lateral aspect of the forefoot were noteworthy, as the potential of such a wedge for reducing strain in the plantar aponeurosis was not previously known. The data suggest that an orthotic wedge under the lateral aspect of the forefoot thus may be effective for the treatment of plantar fasciitis.**

Biomechanical control of the foot with orthotic wedges has broad applications in the treatment of a variety of disorders. Different combinations of wedges or posts have been used to treat a multitude of anomalies, such as varus deformity of the hindfoot alone[3,7,10,27,32] or with varus[26] or valgus[35] deformity of the forefoot, varus deformity of the forefoot[26], valgus deformity of the forefoot[26,31], and abnormal pronation[3,14]. When used to treat these deformities, the function of the wedges is primarily to improve skeletal alignment in order to normalize standing posture and walking. Orthotic wedges have also been advocated for the treatment of pain in the heel[6,24,25,28].

Plantar fasciitis is among the most common disorders of the foot and ankle. The most frequent site of pain and inflammation at clinical presentation is the attachment of the plantar aponeurosis to the medial prominence of the calcaneal tuberosity; however, pain can occur anywhere along the structure[5]. Repetitive trauma and stress have been implicated as causal factors of plantar fasciitis[18,21-23,34]. Nonoperative modalities are the preferred form of management for most patients[11]. The primary goal of orthotic treatment of plantar fasciitis is to relieve strain in the plantar aponeurosis during weight-bearing[5]. Marr and Pod[24] stated that this objective can be achieved by placing a wedge under the medial aspect of the foot, and several authors have recommended a medial wedge or post with such an objective in mind[6,24,25,28]; however, as far as we know, no scientific studies have been done to validate the premise

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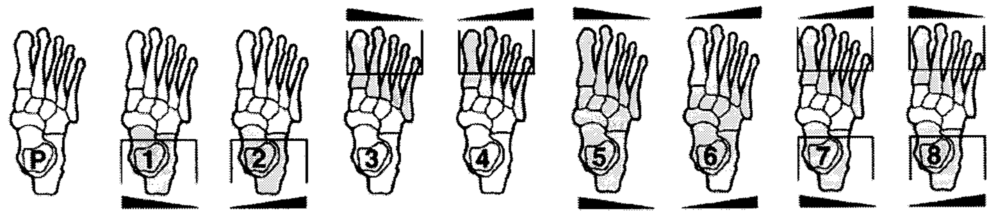

FIG. 1

Illustration of the nine test conditions for a right foot. P = plantigrade (neutral control), 1 = wedge under the medial aspect of the hindfoot, 2 = wedge under the lateral aspect of the hindfoot, 3 = wedge under the medial aspect of the forefoot, 4 = wedge under the lateral aspect of the forefoot, 5 = wedges under the medial aspects of the hindfoot and forefoot, 6 = wedges under the lateral aspects of the hindfoot and forefoot, 7 = wedges under the medial aspect of the hindfoot and the lateral aspect of the forefoot, and 8 = wedges under the lateral aspect of the hindfoot and the medial aspect of the forefoot. All wedges had a 6-degree incline. The triangles indicate the location of the inclination.

that such devices decrease strain in the plantar aponeurosis. Without an understanding of the change in tensile load that may occur with the use of an orthotic wedge, it is difficult to predict the outcome of such treatment or to determine its efficacy.

Because of this void in the literature, many types of orthoses were proposed for the treatment of plantar fasciitis without knowledge of their strain-shielding effect on the plantar aponeurosis. Furthermore, the multitude of antithetical design principles makes it difficult to judge which orthosis will most effectively relieve the pain associated with plantar fasciitis. We[16,17] introduced an *in vitro* measurement technique to quantify the strain-shielding capabilities of the medial longitudinal arch-support mechanisms in foot orthoses. We showed that an orthosis that supports the apical osseous structure of the arch can significantly decrease strain in the plantar aponeurosis ($p < 0.05$)[17]. The University of California Biomechanics Laboratory shoe insert[5] produced a similar reduction in strain by restricting abduction of the forefoot relative to the hindfoot[17]. While there are now two known strain-shielding mechanisms for the plantar aponeurosis, different foot shapes, individual variation, and deformity may predetermine their protective capabilities. Therefore, as some devices may provide a greater degree of control for a particular type of foot, it is important that other orthotic mechanisms be evaluated to determine their potential strain-shielding effects on the plantar aponeurosis.

The plantar aponeurosis is a major component of the static and dynamic structural support mechanisms of the foot[12,13,25]. In neutral stance, both compressive and tensile loads, which provide intrinsic stability to the pedal unit, are produced during load transmission. A truss-like action occurs[19,20], with skeletal segments serving as the compressive members and the plantar ligaments and soft-tissue structures serving as the tensile elements. The plantar aponeurosis, which spans the length of the foot, receives a large portion of the tensile load as the arch of the foot tends to depress and elongate[12,13,29,36]. On the basis of this fundamental principle of foot biomechanics, we developed the hypothesis that, if the support surface of the foot is inclined with wedges, the alteration in the method of load transmission will be exhibited as strain in the plantar aponeurosis.

The specific objective of the present study was to quantify strain in the plantar aponeurosis with different combinations of orthotic wedges in simulated static stance, in order to determine which wedge positions may contribute to the reduction of strain for the treatment of plantar fasciitis. Eight different wedge conditions were tested, with the plantigrade foot serving as the neutral control.

Materials and Methods

Nine fresh-frozen cadaveric lower limbs from five men and four women were disarticulated at the knee. The individuals had ranged in age from forty-five to seventy-three years (mean, fifty-eight years) at the time of death. The mean mass of the six right and three left legs was 3.1 kilograms (range, 2.0 to 4.0 kilograms). All tissue was preserved by freezing at –20 degrees Celsius.

Test Conditions

Eight combinations of wedges with a 6-degree incline and the plantigrade foot, which served as the neutral control, were evaluated, for a total of nine test conditions (Fig. 1). A wedge was placed under the medial aspect of the hindfoot, under the lateral aspect of the hindfoot, under the medial aspect of the forefoot, or under the lateral aspect of the forefoot, or wedges were placed under the medial aspects of the hindfoot and forefoot, under the lateral aspects of the hindfoot and forefoot, under the medial aspect of the hindfoot and the lateral aspect of the forefoot, or under the lateral aspect of the hindfoot and the medial aspect of the forefoot. For the ninth test condition, no wedges were placed at all (the plantigrade foot). Together, the number of specimens and the number of test conditions formed a balanced (nine-by-nine) Latin square testing sequence. This served as the schedule by which to vary the order of the tests.

Orthotic Test Wedges

A 6-degree inclined plane was used to simulate the orthotic wedge condition. This closely approximates the most common wedge used clinically, which is five, six, or seven millimeters high at its thickest region. Four wedges that had been fabricated from Plexiglas were milled to the designated 6-degree angle. A slip-resistant material (Cat's Paw SoleGuard; Biltrite, Ripley, Mississippi) was applied to the top surface to resist slippage of the foot during testing. In addition, the load platform of the testing machine also had a layer of slip-resistant material to ensure that the wedges did not shift during loading. One wedge sufficed as either the forefoot or hindfoot wedge, while all four positioned next to each other made a complete wedged surface for the entire foot. The size and versatility of this system made it simple to use and convenient for the multiple tests that were conducted.

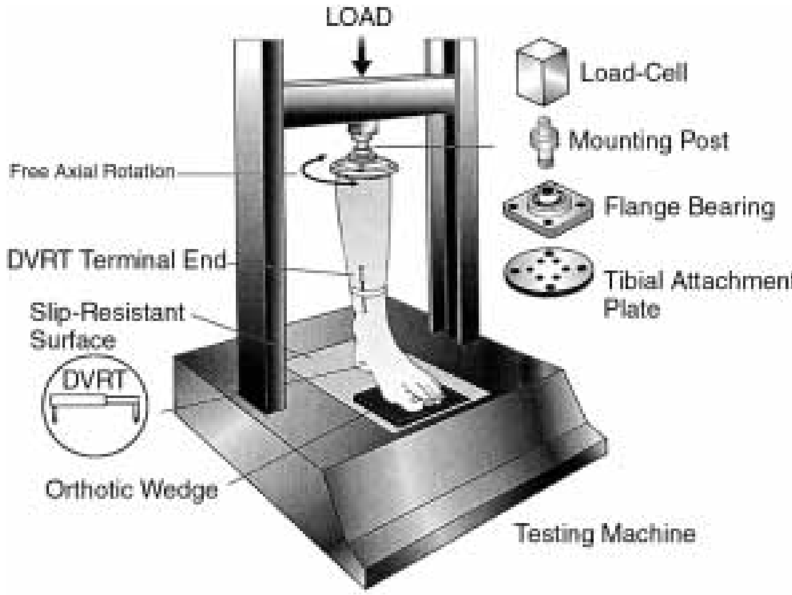

FIG. 2

Illustration of the *in vitro* test setup. DVRT = differential variable reluctance transducer.

Apparatus

The loading system consisted of a Scott electromechanical testing machine (model CRE/500; GCA Precision Scientific, Chicago, Illinois) that was equipped with a strain-gauge-type compression-strain load-cell (Fig. 2). Data were collected on a microcomputer through an amplifier-signal conditioner and an analog-to-digital conversion card (Real Time Devices, State College, Pennsylvania). A data-acquisition software package (DataGauge; MicroStrain, Burlington, Vermont) was used to collect load data from the load-cell and strain readings from the differential variable reluctance transducer (MicroStrain). Subchondral bone was removed from the proximal surface of the tibia with an orthopaedic handsaw to produce a flat surface for attachment of the fixture. A specially designed circular tibial compression plate was fixed to the surface with 7.2-centimeter cancellous-bone screws. A flange bearing attached to the plate permitted free rotation of the tibia while an axial load was applied. The collar of the bearing mated with a post that was connected to the load-cell. A self-centering feature of the flange bearing allowed subtle adjustment of malalignment of the cut surface of the tibia in order to orient the leg perpendicular to the base of the machine.

Specimen-Loading Parameters

For each test, an axial load was applied in a vertical direction through the crosshead of the testing machine. The limb was constrained proximally with the fixture that attached to the testing machine and distally with the slip-resistant material beneath the plantar surface of the foot. The fixture attached to the limb and the testing machine permitted free rotation of the tibia and did not restrain subtalar motion and motion of the foot, which were actively observed during testing. The slip-resistant material increased the friction between the skin on the plantar surface of the foot and the surface of the testing machine and wedges, but it did not interfere with movements of the foot. The self-centering component of the flange bearing of the proximal attachment permitted the transmission of load in three directions: perpendicular to the base of the machine, anteroposterior, and mediolateral. The excursion of the flange bearing, which did not reach the maximum of 10 degrees during any test, ranged from 0 to 3 degrees. Translation of the limb was not observed during any test after the cyclical loading or preconditioning of the specimen that was conducted just before each set of test runs. The slip-resistant surface resisted translation of the foot in two directions and rotation of the foot about an axis perpendicular to the loading platform. The foot was placed so that the center of the ankle joint was on the load line of the testing machine. Vertical alignment of the specimen was determined with a plumb line from the center of the point of load application on the flange bearing to the axis of the talocrural joint in the sagittal plane and through the bisection of the calcaneus in the coronal plane. Delineation lines for placement of the foot were marked on the loading platform to maintain a consistent position for the foot between the changes in wedges. Four lines were drawn: (1) the longitudinal axis of the foot, which was a line bisecting the calcaneus and extending through the second toe; (2) a line from the medial border of the calcaneus to the medial border of the head of the first metatarsal; (3) a line from the lateral border of the calcaneus to the lateral border of the head of the fifth metatarsal; and (4) a line indicating the posterior border of the calcaneus. As the wedges were introduced, the lines were transposed to the new unmarked surface to maintain the same perimeter of foot position.

Technique for Implantation of the Transducer

Each limb was removed from the freezer approximately eight hours before testing and was thawed to room temperature in a cold-water bath. A differential variable reluctance transducer was implanted into the plantar aponeurosis of all specimens (Fig. 3). A three-centimeter-long longitudinal incision was made medially; it extended distally from a point 2.5 centimeters from the posterior aspect of the calcaneus and one centimeter dorsally from the plantar surface of the heel. Blunt dissection was used to expose the plantar aponeurosis. The transducer was implanted into the central band of the aponeurosis with a specially designed tool[16], and two barbed prongs (three millimeters long) affixed to each end of the transducer secured the transducer to the tissue. There was no extraneous off-axis loading of the transducer when an axial load was applied to the limb. Careful attention was given to the placement of the transducer to ensure that it was situated in the tautest segment (the focal load path) and was parallel to the major orientation of the fibers. The focal load path was located by close examination of the plantar aponeurosis through the incision during dorsiflexion of the foot and toes. The connector lead exited from the dorsal surface of the foot just proximal to the medial malleolus through a subcutaneous tunnel.

TABLE I

PERCENT STRAIN*

Load (N)	Test Condition†								
	HL/FL	HM/FL	H0/FL	HL/F0	HM/F0	H0/F0	HL/FM	H0/FM	HM/FM
225	0.89 ± 1.66	1.44 ± 1.47	1.61 ± 1.52	1.97 ± 1.19	2.06 ± 1.94	2.14 ± 1.69	2.72 ± 1.63	2.88 ± 1.38	3.05 ± 1.90
450	2.38 ± 1.86	2.91 ± 1.78	3.04 ± 1.98	3.43 ± 1.42	3.49 ± 2.13	3.57 ± 1.99	4.14 ± 1.98	4.31 ± 1.69	4.54 ± 2.18
675	3.44 ± 2.03	3.98 ± 1.99	4.13 ± 2.22	4.47 ± 1.60	4.54 ± 2.29	4.61 ± 2.17	5.22 ± 2.19	5.40 ± 1.89	5.66 ± 2.36
900	4.31 ± 2.18	4.88 ± 2.14	5.02 ± 2.41	5.34 ± 1.76	5.43 ± 2.42	5.49 ± 2.31	6.13 ± 2.33	6.31 ± 2.03	6.58 ± 2.51

*The values are given as the mean percentage and the standard deviation. Any two means that are not underscored by the same line are significantly different ($p < 0.05$) and any two means that are underscored by the same line are not significantly different ($p > 0.05$), according to the results of Duncan's multiple-range test[8,9].

†H = hindfoot, F = forefoot, L = lateral wedge, M = medial wedge, and 0 = no wedge.

Test Procedures

The specimens were preconditioned with cyclical loading just before testing. Fifty cycles were performed at the selected test parameters, which consisted of movement of the crosshead at a speed of 508 millimeters per minute to apply a load of zero to 900 newtons. After the specimens had been preconditioned, the reference length and the zero position of the strain transducer were determined. The reference length corresponded to a condition of simulated partial weight-bearing, with the plantar aponeurosis undergoing slight strain as it initially became load-bearing. The specimen was placed on the surface of the machine in the plantigrade position without constraint at the top, and the zero position of the transducer and its reference length were initialized. This permitted a proportional calibration of each limb. Percent strain was calculated with the formula $([L - L_0]/L_0) \times 100$, where L was the length at any instant and L_0 was the initial length.

The protocol for all tests was the same for each specimen. The cadaveric limb was mounted in the testing machine. Each limb was subjected to eight different wedge conditions and the neutral condition. The foot was raised slightly to allow the appropriate wedge or wedges to be introduced and, after the specimen had been placed on the wedge or wedges, it was subjected to an additional ten cycles of loading to eliminate so-called first-time behavior represented by an initial settling of the limb on the wedge or wedges. The load and the strain were recorded continuously until maximum load was reached. The data recorded at zero load and at 25 percent (225 newtons), 50 percent (450 newtons), 75 percent (675 newtons), and 100 percent (900 newtons) of the maximum load are presented. The mounting fixture ensured that the placement of the limb was the same for each treatment condition, with only the position of the foot changing. The leg was covered in plastic throughout the test procedures to maintain a moist environment. The voltage outputs from the strain transducer and load-cell were transferred through an amplifier-signal conditioner to a data-acquisition board for analog-to-digital conversion sampled at 750 hertz for a 1.5-second test. As data were sampled, they were simultaneously stored in the computer.

Statistical Analysis

Statistical analysis consisted of separate examination of the dependent variables of percent strain and time to load for each of the four load conditions. As described earlier, this analysis was conducted as a Latin square analysis of variance that consisted of subject, treatment, and order for each of the four loads separately. Duncan's multiple-range test was used for follow-up comparisons[8,9]. Descriptive information was also collected, and significance was set at the 5 percent level. Statistical tests were conducted with the SAS software program (SAS/STAT, version 6; SAS Institute, Cary, North Carolina).

Results

Overview of the Test Model

The results of the Latin square, performed at each load separately, indicated a nonsignificant effect of order for both percent strain and time to load at each of the four load levels ($p > 0.05$). The effect of order is presented as a decrease in percent strain over order, even though it accounted for only approximately 2 percent of the variability in the full model. Further examination of the means suggests that the use of a testing-sequence schedule that is designed to control for order would probably be beneficial, with regard to percent strain, in future studies of this type. In addition, the

TABLE II
TIME TO PRESCRIBED LOAD*

Load *(N)*	Test Condition†								
	H0/FM	H0/F0	H0/FL	HM/FM	HM/F0	HL/F0	HL/FM	HM/FL	HL/FL
225	0.38 ± 0.03	0.38 ± 0.02	0.39 ± 0.03	0.42 ± 0.03	0.43 ± 0.02	0.44 ± 0.05	0.45 ± 0.04	0.45 ± 0.04	0.46 ± 0.03
450	0.50 ± 0.04	0.51 ± 0.03	0.52 ± 0.04	0.56 ± 0.05	0.56 ± 0.03	0.58 ± 0.06	0.59 ± 0.05	0.59 ± 0.05	0.61 ± 0.04
675	0.58 ± 0.04	0.59 ± 0.03	0.61 ± 0.05	0.64 ± 0.05	0.65 ± 0.03	0.67 ± 0.06	0.68 ± 0.05	0.68 ± 0.06	0.71 ± 0.04
900	0.65 ± 0.05	0.65 ± 0.04	0.67 ± 0.04	0.71 ± 0.06	0.71 ± 0.04	0.74 ± 0.07	0.75 ± 0.06	0.75 ± 0.06	0.78 ± 0.05

*The values are given as the mean and the standard deviation in seconds. Any two means that are not underscored by the same line are significantly different ($p < 0.05$) and any two means that are underscored by the same line are not significantly different ($p > 0.05$), according to the results of Duncan's multiple-range test[8,9].

†H = hindfoot, F = forefoot, L = lateral wedge, M = medial wedge, and 0 = no wedge.

effect of subject for percent strain was significant for all loads ($p < 0.001$) and accounted for approximately 74 percent of the variability in the model. Most importantly, the effect of treatment (orthosis) for percent strain was significant for each load separately ($p < 0.001$) and accounted for about 12 percent of the variability. In general, the results indicate that placement of a wedge under the lateral aspect of the forefoot resulted in significantly less strain in the plantar aponeurosis than did placement of a wedge under the medial aspect of the forefoot (Table I). We did not account for the remaining 11 percent of the variability.

The results for time to load were, in general, very similar to those for percent strain, with the effects of both subject and treatment contributing significantly to the model ($p < 0.001$). The effect of subject accounted for approximately 42 percent of the variability, while the effect of treatment accounted for almost 45 percent of the variability. Order accounted for 2 percent of the variability, and the remaining 10 percent was not accounted for.

Measurements of Strain in the Plantar Aponeurosis

It was remarkable that the relative ordering of the treatments remained the same across all four load levels. In general, the control (neutral) condition remained in the sixth position — that is, three test conditions showed more strain and five showed less strain in the plantar aponeurosis (Table I). A complex interrelationship of Duncan groupings was produced from the respective multiple-range follow-up tests. The groupings tended to tighten as the load increased, with five differentiated sets noted at loads of 225 and 450 newtons; only four differentiated sets were found at the higher loads of 675 and 900 newtons (Table I).

A nine-by-nine Latin square design and analysis was employed to evaluate the effect, if any, of test condition on individual differences and experimental order. In general, the results indicated a significant ($p < 0.001$) effect of subject and treatment, but not of order, for each of the four load levels performed separately.

Time to Load

Test conditions involving a wedge generally increased the time to load compared with that for the neutral control (Table II). The times to load were shortest under the test conditions in which the hindfoot was in neutral, including the control condition. Use of wedges under the lateral aspects of the hindfoot and forefoot resulted in a time to load that, at higher loads (450, 675, and 900 newtons), was at least one-tenth of a second longer than that for the neutral control and was significantly different from that under all other test con-

FIG. 3

Radiograph of a right foot, showing the differential variable reluctance transducer (DVRT) implanted in the plantar aponeurosis (arrow). The terminal connector end of the DVRT lead is pictured above the foot.

ditions ($p < 0.05$). Two test conditions, a wedge under the medial aspect of the hindfoot with a neutral forefoot and wedges under the medial aspects of the hindfoot and forefoot, demonstrated, at higher loads, times to load that fell mid-range between the time under the control condition and that with the wedges under the lateral aspects of the hindfoot and forefoot. Compared with the mid-range times to load, the times to load with a wedge under the lateral aspect of the hindfoot with a neutral forefoot, wedges under the lateral aspect of the hindfoot and the medial aspect of the forefoot, and wedges under the medial aspect of the hindfoot and the lateral aspect of the forefoot were increased slightly so that those conditions formed another statistical grouping.

Discussion

Since wedges are now used as an orthotic treatment for plantar fasciitis, it is important to know how they contribute to changes in the strain in the plantar aponeurosis. The aim of the present study was to quantify the strain in the plantar aponeurosis under test conditions involving different combinations of wedges. The wedges produced noteworthy shifts in the pattern of load transmission of the foot compared with that for the neutral control.

Wedge Under the Medial Aspect of the Forefoot

The test conditions that included a wedge under the medial aspect of the forefoot significantly increased the strain in the plantar aponeurosis compared with that for the neutral control or that recorded with a wedge under the lateral aspect of the forefoot ($p < 0.05$). These results indicate that the medial support structures of the foot sustain a greater portion of the load with elevation of the medial (first, second, and third) metatarsals relative to the lateral (fourth and fifth) metatarsals. It can be concluded from these data that the use of a wedge under the medial aspect of the forefoot for the treatment of plantar fasciitis would not provide the desired biomechanical effect of reducing strain in the plantar aponeurosis. This finding is not in agreement with that of Riddle and Freeman[28], who observed a decrease in symptoms with the use of a wedge under the medial aspect of the foot. In their case report of one patient, they indicated that use of a medial-post orthosis (with an inclination of 4 degrees in the hindfoot and 3 degrees in the forefoot) was responsible for the reduction of pain in the heel due to plantar fasciitis. It is difficult to determine the contribution of the orthosis, however, because several treatments were implemented during the orthotic therapy.

The resultant conformations of the foot with a wedge under the medial aspect of the forefoot suggest that a greater portion of the load was directed through the plantar aponeurosis. The use of such a wedge, which was a controlled variable in our experiments, appears to be the primary cause of the increased strain in the plantar aponeurosis, as the addition of a wedge under the medial or lateral aspect of the hindfoot did not significantly affect the resultant strain.

Wedge Under the Lateral Aspect of the Forefoot

A wedge under the lateral aspect of the forefoot decreased strain in the plantar aponeurosis compared with that for the neutral control and that under the test condition involving a wedge under the medial aspect of the forefoot. The data suggest that loads are transferred to the lateral support structures of the foot when the lateral metatarsals are raised higher than the medial metatarsals, thus decreasing strain in the plantar aponeurosis. The support structures engaged by this orthotic position may simulate the calcaneocuboid locking action[7]. While a close examination of the mechanics of the foot would reveal this mechanism, it is surprising that, as far as we know, the association between placement of a wedge under the lateral aspect of the forefoot and the midfoot locking mechanism has not been described in the literature.

Our findings revealed a new type of strain-relief effect in the plantar aponeurosis. The placement of a wedge under the lateral aspect of the forefoot appears to be a promising orthotic control mechanism for shielding the plantar aponeurosis from strain during standing and, most likely, during midstance in walking. While a wedge under the lateral aspect of the forefoot did not reduce strain as much as the medial longitudinal arch-support mechanism or the University of California Biomechanics Laboratory shoe insert did[17], feet that do not respond to treatment with the latter two devices may respond more favorably to placement of a wedge under the lateral aspect of the forefoot. Since individual variation is a determinant of load transmission characteristics, multiple strain-shielding mechanisms offer the clinician alternatives for orthotic treatment when a condition does not respond satisfactorily to a particular orthosis.

Wedge Under the Hindfoot

In the tests in which there was a wedge under only the hindfoot, there was no significant difference between the strain associated with the wedge under the medial aspect and that associated with the wedge under the lateral aspect ($p > 0.05$). There was a slight decrease in the strain for both of those conditions compared with that for the neutral control; however, the trend was not significant ($p < 0.05$). It is of interest that, although we did not find that placement of a wedge under the hindfoot reduced strain significantly, clinical resolution of symptoms has been reported with the use of a wedge under the medial aspect of the hindfoot[6,24,25,28]. The hindfoot medial-post technique developed by Marr and Pod[24] was designed to "reduce the pull on the medial band of the plantar aponeurosis." Those authors reported relief of symptoms with that orthosis. Clancy[6] stated that a three-millimeter-high wedge under the medial aspect of the heel decreased loading of the plantar fascia in runners who had plantar fasciitis. He implied that the symptoms were reduced with that treatment method.

While our tests did not show a significant decrease in strain in the plantar aponeurosis, it is not known how much of a reduction may yield clinical relief of pain. A wedge under the lateral aspect of the forefoot may provide more reliable clinical results; in the present study, it significantly decreased strain in the plantar aponeurosis compared with that associated with a wedge under the hindfoot.

Time to Load

Comparison of the means for percent strain and time to load suggest that there are some parallels between these values and the loading behavior of the specimen (Tables I and II and Fig. 4). Two of the test conditions in which a wedge was placed under the lateral aspect of the forefoot (with another wedge under

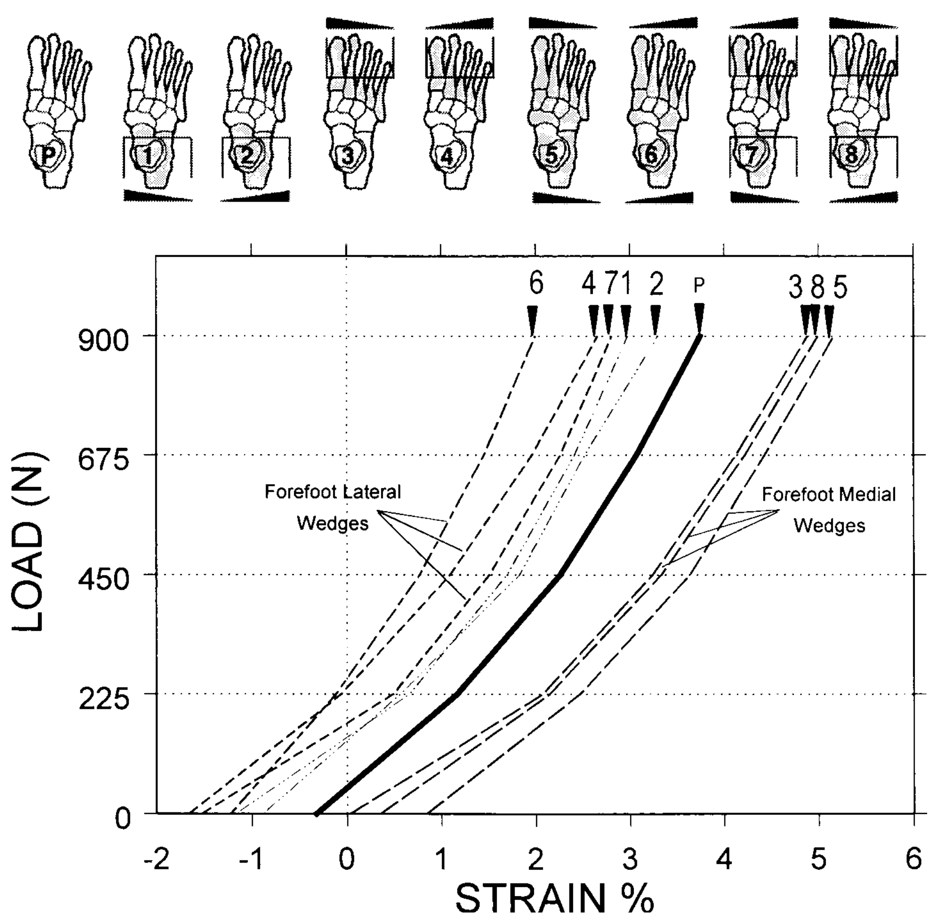

FIG. 4

Typical load-versus-strain curves for all nine test conditions in one specimen. A wedge under the medial aspect of the forefoot (conditions 3, 5, and 8) significantly increased strain in the plantar aponeurosis ($p < 0.05$). A wedge under the lateral aspect of the forefoot (conditions 4, 6, and 7) significantly decreased strain in the plantar aponeurosis compared with that in the neutral control (P) ($p < 0.05$). P = plantigrade (neutral control), 1 = wedge under the medial aspect of the hindfoot, 2 = wedge under the lateral aspect of the hindfoot, 3 = wedge under the medial aspect of the forefoot, 4 = wedge under the lateral aspect of the forefoot, 5 = wedges under the medial aspects of the hindfoot and forefoot, 6 = wedges under the lateral aspects of the hindfoot and forefoot, 7 = wedges under the medial aspect of the hindfoot and the lateral aspect of the forefoot, and 8 = wedges under the lateral aspect of the hindfoot and the medial aspect of the forefoot.

either the medial or the lateral aspect of the hindfoot) produced the longest time to load and the lowest strains compared with those for the neutral control. Of the three test conditions in which a wedge was placed under the medial aspect of the forefoot, only one — a neutral hindfoot with a wedge under the medial aspect of the forefoot — had a notable countertype performance: it produced one of the highest values for percent strain and, simultaneously, the fastest time to load. The two test conditions in which the forefoot was in neutral and a wedge was placed under either the medial or the lateral aspect of the hindfoot had counterpart performance in that they both were generally in the mid-range of order for both percent strain and time to load. While it appears that there may be a relationship between time to load and tensile loading properties of the plantar aponeurosis, the nonparallel performances of some of the other test conditions limit any definitive conclusions from being drawn in this regard.

Although the quantification of strain was the focus of this study, various aspects of the time-to-load data offer new information on the structural stiffness properties of the foot in quasi-static stance. If the foot is considered a structural spring, as has been suggested[1,2,15], a stiff structural system takes less time to load than a flexible one. The Duncan groupings (Table II) can be used to characterize the stiffness properties of the redundancies in the load path created by the wedge test conditions. The present study showed that the foot that had wedges under the lateral aspects of the hindfoot and forefoot was more flexible than the neutral control, which had one of the fastest times to load. Five other test conditions — wedges under the medial aspects of the hindfoot and forefoot, a wedge under the medial aspect of the hindfoot alone, a wedge under the lateral aspect of the hindfoot alone, wedges under the lateral aspect of the hindfoot and the medial aspect of the forefoot, and wedges under the medial aspect of the hindfoot and the lateral aspect of the forefoot — also produced a more flexible foot compared with the neutral control. In contrast, under the three test conditions that did not involve a wedge under the hindfoot, the foot was relatively stiff structurally.

No definitive statements could be made, with use of the time-to-load data, about the differences between the structural properties produced by the truss-like action and those produced by the lateral midtarsal locking mechanism. Close examination of the means for time to load show that the truss-like action of the medial arch appears to produce slightly greater structural stiffness than the lateral midtarsal locking mechanism. The truss-like action, which is activated by dorsiflexion of the tarsometatarsal joints, is simulated by a wedge placed under the medial aspect of the forefoot. This test condition produced the fastest times to load, whereas two of the loading conditions involving a wedge under the lateral aspect of the forefoot (in addition to a wedge under the medial or lateral aspect of the hindfoot) had the longest times to load. One explanation for the increased time to load with a wedge under the lateral aspect of the forefoot may be that the wedge increased the range of subtalar pronation. If the subtalar joint allows additional pronation to occur during loading, it is possible that time to load could also be increased[26].

TABLE III

Pearson Correlations for Percent Strain Versus Time to Load

Test Condition*	Load Level† 225 N	450 N	675 N	900 N
HL/FL	0.64	**0.79**	**0.84**	**0.86**
H0/F0	**0.75**	**0.81**	**0.82**	**0.83**
H0/FL	0.61	**0.79**	**0.74**	**0.75**
HM/F0	0.51	0.65	**0.71**	**0.73**
HM/FM	**0.66**	**0.66**	0.65	**0.66**
HM/FL	0.36	0.52	0.60	0.65
HL/F0	0.16	0.45	0.54	0.60
H0/FM	0.26	0.44	0.51	0.57
HL/FM	0.46	0.50	0.48	0.48

*H = hindfoot, F = forefoot, L = lateral wedge, M = medial wedge, and 0 = no wedge.

†The values in boldface are clinically relevant and significant ($p < 0.05$).

Pearson's correlations between strain and time to load show that certain load-path patterns created by an inclined surface produce a more predictable performance than others (Table III). This is of clinical relevance because it is important to know how the foot will respond to a particular orthotic interface, such as a wedge, and the control potential of the orthosis.

As expected, the degree of correlation increased with an increase in load. At 225 newtons, there was a significant correlation between strain and time to load under only two test conditions, the neutral control and wedges under the medial aspects of the hindfoot and forefoot ($r = 0.75$ and 0.66, respectively). At the higher loads (450, 675, and 900 newtons), significant correlations were found under four and five test conditions. The two test conditions with the highest correlations were wedges under the lateral aspects of the hindfoot and forefoot and the neutral control, followed by a wedge under the lateral aspect of the forefoot alone, a wedge under the medial aspect of the hindfoot alone, and wedges under the medial aspects of the hindfoot and forefoot. There were significant correlations under all of these test conditions ($p < 0.05$). The correlations between strain and time to load under the test conditions that involved wedges placed in opposite positions at the hindfoot and forefoot (the medial aspect of the hindfoot and the lateral aspect of the forefoot or the lateral aspect of the hindfoot and the medial aspect of the forefoot) were not significant ($p > 0.05$). The test conditions that involved a wedge under the medial aspect of the forefoot alone or the lateral aspect of the hindfoot alone also did not demonstrate a significant correlation between strain and time to load. The Pearson correlations are important because they represent another aspect of the control capabilities of orthotic wedges. The countertype response of strain versus time to load may denote a so-called stability factor of a load pathway since a higher correlation tends to be more predictable.

Biomechanical Control Mechanisms of Orthotic Wedges

The rationale for the loading behavior of the foot with various orthotic wedges can be interpreted from fundamental biomechanical concepts of foot function. The work of Bojsen-Møller[4] provides a plausible explanation for the distinct performances of the forefoot wedge tests. Through the use of a high-speed camera and an illuminated glass plate on a walkway, that author was able to visualize contact of the foot and the strain in the plantar aponeurosis. During high-gear push-off, when the area of contact shifts from the heel to the medial aspect of the ball of the foot, tension was observed in the plantar aponeurosis through the skin. During low-gear push-off, when the area of contact shifts from the heel to the lateral part of the ball of the foot, tension in the plantar aponeurosis dissipated. Comparable observations were noted during our biomechanical tests that involved wedges under the forefoot. Conformations of the foot created by a wedge under the medial aspect of the forefoot emulated the initial loading parameters of high-gear push-off, while those created by a wedge under the lateral aspect of the forefoot emulated low-gear push-off. The consequential reorientation of the load produced by a wedge under the lateral aspect of the forefoot simulated pronation of the calcaneocuboid joint, which moved into the close-packed position, thereby decreasing strain in the plantar aponeurosis. While this osseous locking mechanism requires restraint from plantar ligaments[27], it appears that either the lateral component of the plantar fascia or, more likely, other support structures must be load-bearing, since the transducer showed a reduction in strain. Placement of a wedge under the medial aspect of the forefoot closely resembles inversion of the foot, which moves the calcaneocuboid joint into the loose-packed position, thereby increasing strain in the plantar aponeurosis.

Because an orthotic wedge produces an alternative load path in the foot, the present study inherently dealt with two accepted structural components, the medial truss-like action and the lateral midtarsal locking mechanism. The medial column of the foot relies on a truss mechanism to transmit loads[19,20]. According to Sarrafian[30], internal rotation of the leg, dorsiflexion of the foot, and hyperextension of the toes at the metatarsophalangeal joints activate this loading mechanism. When

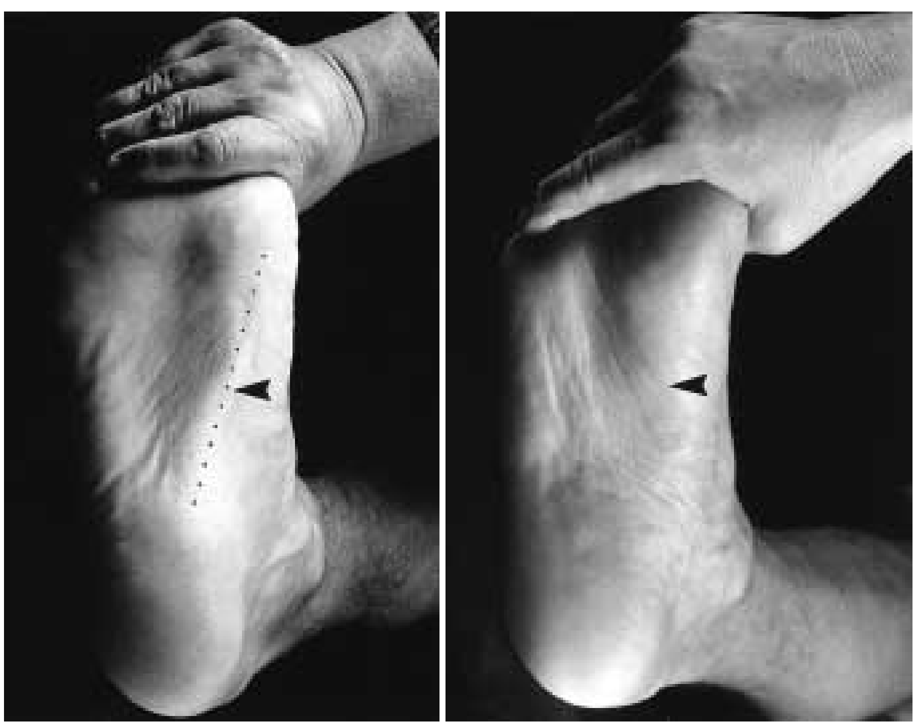

FIG. 5-A FIG. 5-B

Figs. 5-A and 5-B: Photographs of the plantar aspect of the foot. The arrowheads indicate the region where changes in the tension in the plantar aponeurosis can be observed or palpated.

Fig. 5-A: The plantar aponeurosis becomes taut when the forefoot is positioned in varus relative to the hindfoot. The dotted line indicates the region of increased tension.

Fig. 5-B: The plantar aponeurosis becomes slack when the forefoot is positioned in valgus with respect to the hindfoot.

ground-reaction forces to the medial (first, second, and third) metatarsals are increased, this region of the foot is dorsiflexed with respect to the tibia, actuating the truss mechanism to a greater extent than contact with a planar surface. In the present study, the test conditions in which a wedge was placed under the medial aspect of the forefoot produced the truss action by increasing the dorsiflexion moment in the medial tarsometatarsal joints. In line with the theory of this action, the strain in the plantar aponeurosis increased significantly compared with that for the neutral control. This confirms that the analogy of the truss for the medial structure of the foot is reasonable. Additional evidence for the truss action was apparent under the test conditions in which a wedge was placed under the lateral aspect of the forefoot. The decreased strain in the plantar aponeurosis observed under these conditions suggests that the truss action was lessened as the lateral midtarsal locking mechanism was engaged to transmit loads through the foot. The results of the present study clearly support these accepted biomechanical principles of foot function and further define their role.

Clinical Observations

The changes in tensile load measured in the plantar aponeurosis during the various test conditions that involved a wedge under the forefoot can also be observed with clinical examination. With the subject seated on an examination table, the toes are passively extended to make the plantar aponeurosis taut. With the toes maintained in extension, the forefoot is passively positioned in varus to reproduce the condition of a wedge under the medial aspect of the forefoot, and then it is positioned in valgus to reproduce that of a wedge under the lateral aspect of the forefoot. While the forefoot is moved through the range in this manner, the modulation of tension can often be visualized through the skin and can also be palpated readily when the thumb of the other hand is pressed firmly into the medial aspect of the plantar aponeurosis at the level of the midfoot (Figs. 5-A and 5-B). Tension decreases dramatically as the forefoot is passively positioned in valgus (Fig. 5-B), and the aponeurosis becomes taut when the forefoot is positioned in varus (Fig. 5-A). This assessment is of value clinically to determine the responsiveness of the foot to this type of positioning and to approximate the reduction in tension that an orthosis might produce in the plantar aponeurosis.

Patients who have plantar fasciitis often adopt an antalgic gait pattern in an effort to avoid pain. This compensatory response is usually characterized by walking on the lateral border or the ball of the foot. Although the mechanics of these gait deviations and the mechanisms responsible for the partial pain relief associated with them are still uncertain, the data from this study provide some support for a theoretical explanation. The forefoot and hindfoot varus attitude assumed by patients who walk on the lateral aspect of the foot diverts the load-transmission pathway to lateral skeletal and ligamentous support structures. Since the medial load-bearing structures of the foot are not in contact with the ground, the truss-like mechanism normally present during stance is not activated. Thus, strain in the plantar aponeurosis, and the resultant pain, are comparatively less than when the medial part of the ball of the foot is in contact with the ground. This concept is supported by the work of Bojsen-Møller[4], who observed a decrease in strain in the plantar aponeurosis during low-gear push-off in walking, when the area in contact with the ground moves from the heel to the lateral portion of the ball of the foot. Therefore, misconceptions may have developed with regard to the use of a wedge under the medial aspect of the foot for the treatment of plantar fasciitis through clinical observations that pain was temporarily relieved when the medial aspect of the foot was raised. However, the findings of the present study do not support the assumption that a wedge under the medial aspect of the foot decreases strain in the plantar aponeurosis.

Irregular surfaces and slopes of different grades are encountered during the course of daily activities. The foot is capable of adapting to most of these situations through an individualized loading mechanism for each foot-ground interface encountered. The present study revealed some of the possible load-pattern characteristics that the foot may assume during standing on various inclines. The load-redundancy path appears to be regulated in part by the surface with which the foot reacts and the resultant conformation of the foot. Furthermore, an orthotic wedge appears to be a reliable control mechanism for the alteration of the load transmission pattern in the foot, particularly if strain in the plantar aponeurosis is to be modified.

Limitations of the Experimental Model

Consideration should be given to our method of establishing the zero position of the strain transducer when the strain values reported in the present study are compared with those in other studies. Also, the measurements of local strain quantified during the tests may differ from strain values in other regions of the plantar aponeurosis. The test design attempted to simulate static stance *in vitro*, and therefore it is only a limited representation of what may occur during standing or midstance. Orthotic wedges or posts configured to a custom foot orthosis may affect strain in the plantar aponeurosis differently than the conditions tested in the present study. The tests emulated standing on an inclined plane with primary interactions to the forefoot and hindfoot. A custom orthosis with a wedge that interacts with the midfoot may provide additional changes in strain in the plantar aponeurosis.

In summary, this investigation offers new insight into the loading behavior of the plantar aponeurosis and the foot when the latter is in contact with various inclines. It appears that the load-transmission path of the foot adapts to a change in surface through the use of different structural support mechanisms. The medial support structures of the foot seem to act like a truss, as Lapidus[19,20] suggested. A wedge under the medial aspect of the forefoot accentuated this action, while a wedge under the lateral aspect of the forefoot minimized the supportive function of the truss. In the latter case, the calcaneocuboid joint locks to accept a greater portion of the load. For the foot to accommodate to a medially or laterally sloped surface, the two structural mechanisms must work in synergy, alternating their supportive role from a primary to a secondary support system as the situation demands.

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References

1. **Alexander, R. M.,** and **Bennet-Clark, H. C.:** Storage of elastic strain energy in muscle and other tissues. *Nature,* 265: 114-117, 1977.
2. **Alexander, R. McN.:** Elastic energy stores in running vertebrates. *Am. Zool.,* 24: 85-94, 1984.
3. **Bates, B. T.; Osternig, L. R.; Mason, B.;** and **James, L. S.:** Foot orthotic devices to modify selected aspects of lower extremity mechanics. *Am. J. Sports Med.,* 7: 338-342, 1979.
4. **Bojsen-Møller, F.:** Calcaneocuboid joint and stability of the longitudinal arch of the foot at high and low gear push off. *J. Anat.,* 129: 165-176, 1979.
5. **Campbell, J. W.,** and **Inman, V. T.:** Treatment of plantar fasciitis and calcaneal spurs with the UC-BL shoe insert. *Clin. Orthop.,* 103: 57-62, 1974.
6. **Clancy, W. G.:** Runners' injuries. Part two. Evaluation and treatment of specific injuries. *Am. J. Sports Med.,* 8: 287-289, 1980.
7. **D'Ambrosia, R. D.:** Conservative management of metatarsal and heel pain in the adult foot. *Orthopedics,* 10: 137-142, 1987.
8. **Duncan, D. B.:** A significance test for differences between ranked treatments in an analysis of variance. *Virginia J. Sci.,* 2: 171-189, 1951.
9. **Duncan, D. B.:** Multiple range and multiple F tests. *Biometrics,* 11: 1-42, 1955.
10. **Eggold, J. F.:** Orthotics in the prevention of runner's overuse injuries. *Phys. and Sports Med.,* 9: 124-128, 1981.
11. **Gill, L. H.,** and **Kiebzak, G. M.:** Outcome of nonsurgical treatment for plantar fasciitis. *Foot and Ankle Internat.,* 17: 527-532, 1996.
12. **Hicks, J. H.:** The foot as a support. *Acta Anat.,* 25: 34-45, 1955.
13. **Hicks, J. H.:** The three weight-bearing mechanisms of the foot. In *Biomechanical Studies of the Musculo-Skeletal System,* pp. 161-191. Edited by F. G. Evans. Springfield, Illinois, Charles C Thomas, 1961.
14. **James, S. L.; Bates, B. T.;** and **Osternig, L. R.:** Injuries to runners. *Am. J. Sports Med.,* 6: 40-50, 1978.
15. **Ker, R. F.; Bennett, M. B.; Bibby, S. R.; Kester, R. C.;** and **Alexander, R. M.:** The spring in the arch of the human foot. *Nature,* 325: 147-149, 1987.
16. **Kogler, G. F.; Solomonidis, S. E.;** and **Paul, J. P.:** In vitro method for quantifying the effectiveness of the longitudinal arch support mechanism of a foot orthosis. *Clin. Biomech.,* 10: 245-252, 1995.
17. **Kogler, G. F.; Solomonidis, S. E.;** and **Paul, J. P.:** Biomechanics of longitudinal arch support mechanisms in foot orthoses and their effect on plantar aponeurosis strain. *Clin. Biomech.,* 11: 243-252, 1996.
18. **Kwong, P. K.; Kay, D.; Voner, R. T.;** and **White, M. W.:** Plantar fasciitis. Mechanics and pathomechanics of treatment. *Clin. Sports Med.,* 7: 119-126, 1988.
19. **Lapidus, P. W.:** Misconception about the "springiness" of the longitudinal arch of the foot. Mechanics of the arch of the foot. *Arch. Surg.,* 46: 410-421, 1943.
20. **Lapidus, P. W.:** Kinesiology and mechanical anatomy of the tarsal joints. *Clin. Orthop.,* 30: 20-36, 1963.
21. **Leach, R. E.; Seavey, M. S.;** and **Salter, D. K.:** Results of surgery in athletes with plantar fasciitis. *Foot and Ankle,* 7: 156-161, 1986.
22. **Lester, D. K.,** and **Buchanan, J. R.:** Surgical treatment of plantar fasciitis. *Clin. Orthop.,* 186: 202-204, 1984.
23. **Lutter, L. D.:** Surgical decisions in athletes' subcalcaneal pain. *Am. J. Sports Med.,* 14: 481-485, 1986.
24. **Marr, S. J.,** and **Pod, F. A.:** The use of heel posting orthotic techniques for relief of heel pain. *Arch. Orthop. and Traumatic Surg.,* 96: 73-74, 1980.
25. **Marshall, P.:** The rehabilitation of overuse foot injuries in athletes and dancers. *Clin. Sports Med.,* 7: 175-191, 1988.
26. **Michaud, T. C.:** *Foot Orthoses and Other Forms of Conservative Foot Care.* Baltimore, Williams and Wilkins, 1993.
27. **Novick, A.,** and **Kelley, D. L.:** Position and movement changes of the foot with orthotic intervention during the loading response of gait. *J. Orthop. and Sports Phys. Ther.,* 11: 301-312, 1990.
28. **Riddle, D. L.,** and **Freeman, D. B.:** Management of a patient with a diagnosis of bilateral plantar fasciitis and Achilles tendonitis. A case report. *Phys. Ther.,* 68: 1913-1916, 1988.
29. **Sarrafian, S. K.:** Functional characteristics of the foot and plantar aponeurosis under tibiotalar loading. *Foot and Ankle,* 8: 4-18, 1987.
30. **Sarrafian, S. K.:** *Anatomy of the Foot and Ankle: Descriptive, Topographic, Functional.* Philadelphia, J. B. Lippincott, 1993.
31. **Schoenhaus, H. D.,** and **Jay, R. M.:** Cavus deformities: conservative management. *J. Am. Podiat. Assn.,* 70: 235-238, 1980.
32. **Smith, L. S.; Clarke, T. E.; Hamill, C. L.;** and **Santopietro, F.:** The effects of soft and semi-rigid orthoses upon hindfoot movement in running. *J. Am. Podiat. Med. Assn.,* 76: 227-233, 1986.
33. **Snider, M. P.; Clancy, W. G.;** and **McBeath, A. A.:** Plantar fascia release for chronic plantar fasciitis in runners. *Am. J. Sports Med.,* 11: 215-219, 1983.
34. **Torg, J. S.; Pavlov, H.;** and **Torg, E.:** Overuse injuries in sport: the foot. *Clin. Sports Med.,* 6: 291-320, 1987.
35. **Valmassy, R.:** Orthoses. In *Sports Medicine of the Lower Extremity.* Edited by S. Subotnick. New York, Churchill Livingstone, 1989.
36. **Wright, D. G.,** and **Rennels, D. C.:** A study of the elastic properties of plantar fascia. *J. Bone and Joint Surg.,* 46-A: 482-492, April 1964.