Orthotics and Prosthetics, **Volume 38, Number 3, Autumn, 1984, pp. 12–40.**

Orthotic Control of Ground Reaction Forces During Running (A Preliminary Report)

by John Glancy, CO

This discussion of the control of ground reaction forces is presented on the assumption that the best approach to improving the runner's environment is to prevent injuries by the prophylactic use of orthotic devices. It is now readily acknowledged that ground reaction forces can adversely affect the foot/ankle complex when one or more of a variety of abnormal conditions are present. Also, there is equal acceptance that ground reaction forces when malfunction of a part of the foot/ankle complex is present, can have debilitating effects higher up in the kinetic chain. This discussion will be limited to the effects of ground reaction forces to the foot and ankle during the heel-to-toe running cycle.

The prevention of running injuries by orthotic means may seem academic with respect to runners attending sports medicine clinics who come for immediate treatment of injuries they have already sustained. Under these circumstances, the use of orthotic devices may still be thought of as prophylactic if the cause of a particular injury is determined to be biomechanical in nature and if the device successfully arrests or reverses the causative condition and/or prevents reoccurrence of the injury.

There are estimated to be over 25 million adult Americans now running or jogging regularly.[2 (Part 1)] If the patients referred to the University of Indiana Division of Orthotics are representative of runners in general, many of their injuries are related to foot and/or ankle conditions that were present prior to their running activities. These conditions are often reported to have been asymptomatic before the patient took up running. In fact, the majority report that they do not experience symptoms except when they run or soon after a run. The increase in impact with the ground when running, by a factor of two to three times body weight at heel strike, is known to be related to the development of symptoms. Burdett's study predicts peak Achilles tendon forces ranging from 5.3 to 10 times body weight.[3] Also related are pre-existing biomechanical "imperfection(s)" of which the runner has been unaware and/or a long-forgotten prior injury has led to malalignment of the foot or ankle and which predisposes the runner to injury.[2 (Part II, p. 26)]

Sports medicine patients are unique in terms of individual motivation. They dislike hearing that they must give up or slow down their weekly running schedule, even for a few days. Often, many attempt to "run through" an injury, causing the condition to become chronic or resulting in a more severe level of injury. This mind-over-matter attitude, when pathomechanical factors are involved, coupled with attrition, can lead to a no-win situation, both immediate and long-term.

Since the general fitness of most runners is well above average, their very fitness often masks the cause of their complaints which, as a consequence, can be very subtle and difficult to define. Dealing with these subtleties sharpens one's powers of clinical observation. The insights and developments reported in this paper are to a substantial extent the product of the patience, persistence, conscientiousness, and accurate feedback of the runners themselves. I am indebted to them because the benefits from these new insights and developments are already being shared by orthotically handicapped patients. For example, the promising results to date with the flexible polypropylene insert and cushion heel wedge (Figures 6 and 7) for juvenile rheumatoid arthritis patients are especially gratifying.

THE RUNNING CYCLE VERSUS THE WALKING CYCLE

The purpose of comparing the walking and running cycles is twofold: first, to identify differences in kind from differences in degree between the two cycles; second, once identified, to try to understand the biomechanical relevance of either type of difference as regards injuries to the lower limb while running, particularly to the foot/ankle complex.

The stance phase is reduced from approximately two-thirds of the walking cycle to one-third in the running cycle. As one might anticipate, whether during walking or running, the velocity, cadence and stride length increase as the speed of gait increases. Also, as the gait speed increases, the period of stance phase decreases, and the period of swing phase increases in the walking cycle. As a result, the greater the increase in gait speed, the less time the foot spends on the ground.[20, 21] An increase in the speed of gait, then, in spite of its effect upon velocity, cadence and stride, is not the feature that differentiates the running cycle from the walking cycle.

It is the absence of a double-support phase that distinguishes running from walking. There is a period during the running cycle when both feet are off the ground. This period has been named the non-support or "float" phase.[20] What motion, exclusive to running, eliminates the double-support phase? None, per se. An increase in the magnitude of the thrust at push-off by the contralateral limb, generated principally by the gastrocnemius and soleus muscles, "lifts" the body and makes a float phase possible. The forward velocity attained is also an important contributor to the float phase.

From an orthotic standpoint, the replacement of the double-support phase of walking with a float phase when running is the most significant biomechanical feature of the running cycle. It is this biomechanical feature, above all, that makes the control of ground reaction forces important for the following reasons:

- The orientation of the foot/ankle complex to the ground at lift-off determines the balance and direction of the body as it ascends, particularly with respect to the line of progression.
- Without the double-support phase, a person's ability to shift his body's weight efficiently and economically during stance phase is greatly reduced. Adaptation to surface conditions, topography, and fatigue, for example, must be made rapidly upon one foot at a time. This restriction affects both lift-off and heel-strike.
- The orientation of the foot/ankle complex to the ground, when receiving the body as it descends from float phase to stance phase, has a direct bearing on running injuries.
- During its descent, the body is in "free fall." The resulting impact to the lower limb is reported to be 2.5 to 3.0 times the body's weight.[4, 20, 21] Obviously, there is no way of altering the

body's rate of descent for a given running speed. However, the floor reaction forces are always equal and opposite to the resulting vertical, AP shear, ML shear, and torque forces generated at any particular speed. Control of one or more of these floor reaction forces is a viable means of controlling the alignment and/or phasic motions of the foot/ankle complex throughout the stance phase, thereby preventing injury.

Several studies report that the vertical force passing through the foot reaches its highest peak just before lift-off.[4, 20, 21] As a consequence, from an orthotic point of view, the mid-foot is especially vulnerable between heel-rise and lift-off. It is during this period of the running cycle that protection from injury is most difficult to provide without interfering with the dynamics of running.

With biomechanics foremost in mind, then, the only difference in kind between walking and running is the transition from a double-support phase to a float phase. All the motions that are inclusive to normal stance and swing phases of the walking cycle are the same that occur during the stance and swing phases of the heel-to-toe running cycle.

However, there are differences in degree of these motions as they occur within the running cycle. The principal difference is that the range of these motions increase during running, reflecting additional differences in degree in gait, velocity, cadence, and stride length. There is also a difference in degree with respect to the impact and stress to which the lower limbs are subjected between walking and running.

Because the same number and kinds of motion occur in the same sequence in both cycles, the absence of a double-support phase forces the runner to perform the phasic motions of the stance phase without assistive substitution from the contralateral lower limb. Any attempts to substitute one motion for another are not feasible, since each motion—in proper sequence—is essential to a running gait free of injury. As a consequence, when one or more of three conditions are present within the foot/ankle complex, the probability of injury becomes a distinct biomechanical possibility. The three conditions are:

- *Hypermobility,* i.e., excessive range of one or more phasic motions,
- *Hypomobility,* i.e., less than normal range of one or more phasic motions, and
- *The loss* of one or more phasic motions.

Whether resultant injury from these conditions is immediate or long term will be discussed later, along with suggested orthotic management for each.

Numerous questions arise as a result of comparing the walking and running cycles. For example, 70 percent of the runners seen in our Sports Medicine Clinic cannot evert their heels, either passively or actively. There is nothing in the literature to indicate that the absence of this motion is as prevalent in "normal" feet among the walking population. Is it not reasonable, then, to presume that the loss of heel eversion is acquired and that this loss is a product of long term distance running?

Is "cavus" necessarily the best term to use to describe the high longitudinal arch of a runner? Dorland's *Medical Dictionary* defines "pes cavus" as "exaggerated height of the longitudinal arch of the foot, present from birth or appearing later because of contractures or disturbed balance of the muscles."

In an otherwise normal foot, is the presence of a congenital high longitudinal arch (Dorland's "exaggerated height") considered an abnormality? When the subjects are runners, the literature does not associate abnormality with this condition. In fact, substantial evidence has been reported, regarding the demands that running places upon the foot/ankle complex, that has led to a consensus that a high longitudinal arch is favorable to runners. This consensus is based on evidence related to studies of the subtalar joint.[11, 14, 15, 16, 18, 22]

In essence, the more vertical the angle of inclination of the A-P axis of the subtalar joint (*i.e.,* the higher the longitudinal arch), the less pronation or supination of the foot. The correlation between excessive

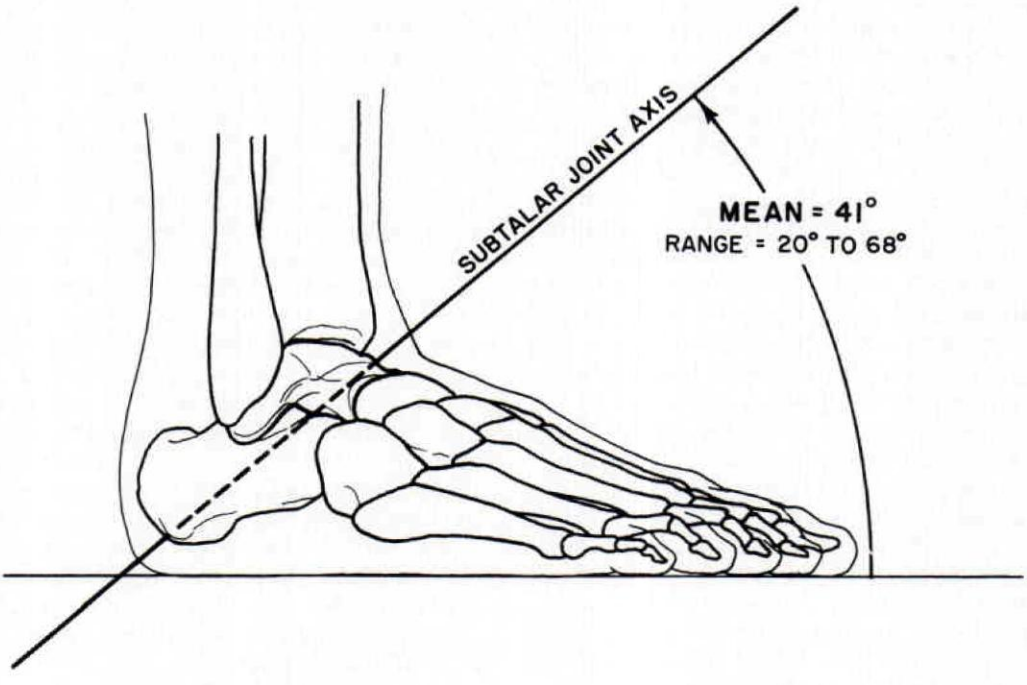

Figure 1. Angle between the axis of the subtalar joint and the horizontal (From Inman [14] in: *Bull. Prosth. Res.*, BPR-10-11, 1969, p. 137). Reprinted with permission of the Veterans Administration.

pronation and injury to runners is well established, hence the broad consensus (Witness the attention to "heel control" by running shoe manufacturers). The answer thus appears straightforward—a high arch is an asset to runners.

However, I have experienced difficulty accepting this answer, particularly with respect to long distance running as a long term activity. How high is high? What precisely is an "exaggerated height?"

First, how high is high is another way of asking "What is the norm?" Isman and Inman[16] reported a mean of 41 degrees for the ankle of the axis of the talocalcaneal joint with respect to a horizontal plane (Figure 1). They found the range of inclination of this axis to be 20 to 68 degrees.[16] It would appear safe to assume, then, that one may be born with quite a "high" longitudinal arch and still be within the bounds of normality. It would also appear that Dorland's term, "exaggerated height," to describe a cavus foot is misleading. Two of Webster's definitions of "exaggerate" are "overstate," and "to enlarge or increase, especially beyond the normal." When the term "cavus" is used with reference to runners' feet, it overstates the case, in light of Webster's second definition.

This reference to abnormality brings us to the non-congenital, or "acquired," cavus foot due to "contractures or disturbed balance of the muscles" (Dorland's words). The musculature of the plantar aspect of the foot must be stronger than that of the dorsal aspect to cause the formation of a cavus foot. Such was frequently the case with growing children stricken with poliomyelitis.

Runners do not "acquire" cavus feet. The runner with high longitudinal arches was born with them. The A-P angle of his subtalar joints is more perpendicular than horizontal with respect to the ground. This is a condition that is generally considered favorable because it provides a built-in restriction to pronation of the mid-foot. A high arch is better suited to resist the superincumbent weight of the body. However, there is also one important disadvantage of a high arch—its relative rigidity makes it a less efficient shock absorber for the greater magnitude of impact forces that are generated by running. There is, then, a trade-off—hypomobility of the longitudinal arch versus a reduction of shock absorbency. Is this trade-off advantageous to the long distance, heel-to-toe runner over the long term?

As runners with high longitudinal arches began to appear with regularity at the Sports Medicine clinic, the clinical staff's unanimity of opinion with regard to the advantages of cavus feet began to fragmentize. From an orthotic overview, this puzzling question arose: Is not the possessor of so-called cavus longitudinal arches in actuality the possessor of what may best be described as feet with functional forefoot drop? The question is directed to the heel-to-toe runners only, i.e., Levels I, II, and III—or 98 percent of all runners.[2, Part III] A quite different set of biomechanical circumstances applies to the long distance (Level IV, "Elite") forefoot runner.

A GENERAL RULE

Assume the proposition that we are now dealing with two distinctly different types of feet, neither of which can be said to be abnormal *per se*. Structurally, the difference is one of degree between the normal non-cavus and the normal cavus foot. However, there are distinct functional differences resulting from these structural variances which have been overlooked.

One functional difference concerns the synchronous relationship of pronation of the longitudinal arch of the foot to transverse internal rotation of the tibia and, conversely, the raising (supination) of the arch to transverse external rotation of the tibia. From an orthotic viewpoint, an important feature of these motions is the fact that, as a general rule, blockage of the one automatically blocks its synchronous mate. That is, inhibition of phasic midfoot pronation also inhibits phasic internal rotation of the tibia, and vice versa. The same applies to phasic, midfoot supination and external rotation of the tibia.[21] There are exceptions and these exceptions are directly related to structural differences.

The initial pronation of the longitudinal arch (as a result of eversion of the heel at heel-strike) is a purely passive mechanism which is initiated by contact with the ground. The limitations of the range of either heel eversion or pronation are not dependent upon muscle control. The range of both motions is controlled, in order of importance, by the congenital placement of the axes of the subtalar and transverse tarsal joints, the geometry of their articulating surfaces, and their connecting ligaments.[21] Both motions are integral parts of the heel-toe running cycle; their occurrence at the beginning of the stance phase sets the alignment of the foot/ankle complex and, in so doing, affects all that follows.

There are great individual variations in the angle of the sagittal axis of the subtalar joint. These variations (Figure 1) alter the relation between the amount of pronation and supination of the foot and the amount of internal and external rotation of the tibia in the transverse plane.[14, 16] These tibial rotations are affected by the variations in the tangent of the angle of inclination in the sagittal plane. When the leg is vertical and the foot is at a right angle to it (flat on the ground), and the axis of the subtalar joint is 45 degrees, the internal-external rotations of the tibia would be equal in magnitude to their respective pronation-supination motions of the longitudinal arch.[14] It is because of this one-to-one relationship of these motions, at this angle of inclination of the axis, that 45 degrees was chosen as the benchmark for our general rule. The A-P angle of the subtalar joint in a living subject cannot currently be determined. Nevertheless, the benchmark gives a factual point of reference in order to discuss the relationship of structure to function.

When the angle of inclination of the A-P axis of the subtalar joint (as viewed in the frontal plane) is closer to the horizontal

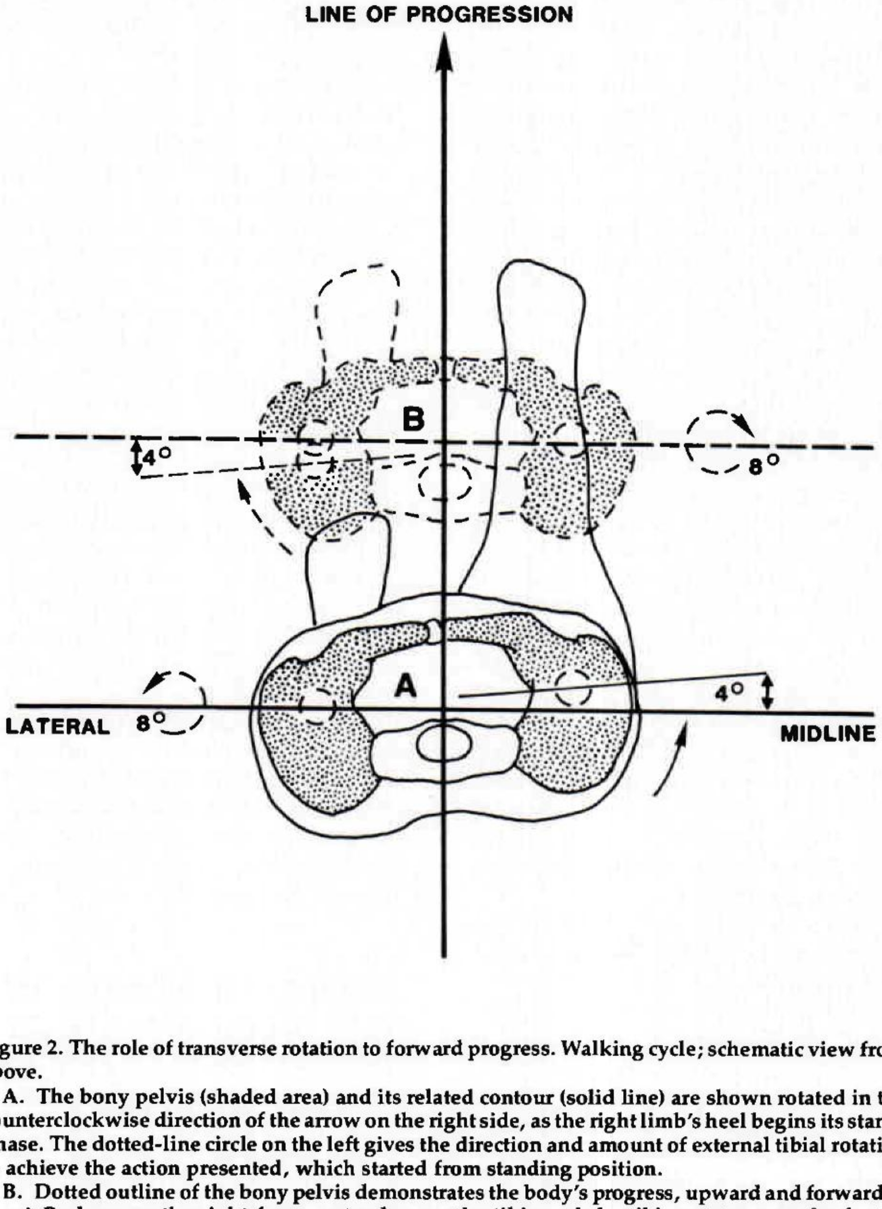

Figure 2. The role of transverse rotation to forward progress. Walking cycle; schematic view from above.

A. The bony pelvis (shaded area) and its related contour (solid line) are shown rotated in the counterclockwise direction of the arrow on the right side, as the right limb's heel begins its stance phase. The dotted-line circle on the left gives the direction and amount of external tibial rotation to achieve the action presented, which started from standing position.

B. Dotted outline of the bony pelvis demonstrates the body's progress, upward and forward in the A-P plane, as the right femur extends over the tibia and the tibia rotates over the foot to mid-stance. The dotted-line arrow on the left shows the clockwise forward rotation of the pelvis to the lateral midline and return to the standing position, as indicated by the placement of the left forefoot's dotted outline. The dotted-line circle on the right shows the eight degrees of external clockwise rotation, half its range, which reverses the tibia's internally rotated position taking the pelvis *et. al.* with it. The preceding eight degrees of internal tibial rotation, having occurred between foot-flat and mid-stance, are not shown.

(less than 45, but not lower than 20 degrees), the greater is the amount of pronation and supination for a given amount of internal or external rotation in the transverse plane[16] (Pes planus being representative of the abnormal non-cavus foot). Also, it can be demonstrated that when the oblique angle of the subtalar joint is closer to the vertical (more than 45 degrees) when viewed from the side, the magnitude of axial rotation to pronation and supination of the midfoot is greater.

It is in the normal cavus foot that we will find exceptions to the general rule. ("Normal" is here defined as an A-P obliquity of the axis of the foot's subtalar joint from 46 to 68 degrees to the horizontal.) The correlation between axial rotations to pronation/supination breaks down when the axis' A-P angle is greater than 45 degrees. For example, a given amount of tibial internal or external rotation effectuates less respective pronation or supination of the midfoot, the closer the axis of the subtalar joint is to the vertical. However, in spite of the fact that the range of pronation/supination is anatomically restricted in such cases, ground reaction forces being transmitted through the cavus foot (especially the vertical component) translate into a disproportionally larger transverse rotary force to the tibia.[14]

The more vertical the A-P angle of the subtalar joint axis, the less the amount of pronation possible. This fact provides insight into the biomechanics unique to the forefoot runner. The forefoot runner's gait does not seem to require depression of the longitudinal arches; they would be subject to injury were pronation to occur with each stride. Suppose we were to assume that the absence of pronation also eliminates the need for internal rotation of the tibia (and femur) in the transverse plane between foot-flat and mid-stance. Such an assumption seems to be valid, since heel-strike—the "trigger" of pronation—is bypassed by forefoot runners. Obviously, external rotation of the tibia (and femur), as well as pelvic rotation in the transverse plane, must be modified under such circumstances to enable the forefoot runner to effect an efficient stride. However, it would appear that a like accommodation cannot be achieved during a heel-to-toe running gait (Figure 5).

Another functional difference presented by the normal, so-called cavus foot is related to the hypomobility of the longitudinal arch of the foot with the condition here referred to as functional forefoot drop. This hypomobility of the midfoot necessitates an adjustment in order to effect heel contact in the standing and/or midstance position. The patient is unconscious of making this adjustment, which he performs by rotating his tibia anteriorly or posteriorly in the A-P plane (Figure 3). The consequences of this adjustment to a runner are discussed in detail under Condition IV.

One might well ask whether a heel-to-toe running gait should be advised for the beginner whose feet have functional forefoot drop. Also, it would appear that current practices of shoe selection and fitting of growing, active youngsters should be reexamined. The impact of this finding upon the design of commercial running shoes would appear to depend on the percentage of feet in the running population with functional forefoot drop. If the percentage is found to be substantial, it would be feasible for manufacturers to provide running shoes similar to the modified shoe shown in Figure 9.

EXTRAPOLATIONS FROM THE BIOMECHANICS OF THE WALKING CYCLE TO THE RUNNING CYCLE

Why should the orthotist find the synchronous character of certain motions of the tibia, hindfoot, and midfoot of special interest, when his patient is a runner? The three major reasons for his interest have been mentioned previously: first, the conviction that the foot/ankle complex is most vulnerable and most difficult to control between heel-rise and lift-off; second, the absence of a double-support phase in the running cycle leads to the presumption that there are no viable substitutions for hypermobility, hypomobility, or the loss of a particular motion, and third, the fact that the axis of the subtalar joint (the primary control mechanism of the synchronous motion) cannot be identified in living subjects. To determine the importance of each of the above to the running cycle, the orthotist must temporarily divert his attention further up the kinetic chain.

During running, the impact at heel-strike is 2.5 to 3.0 times body weight as compared to 1.2 times body weight during walking. This increase in vertical force can be assumed to generate a proportionately greater eversion moment during running. Because eversion of the heel is the key to midfoot pronation, it is reasonable to assume that its range also increases proportionately to the increased moment. With an increase in the range of pronation, there is a corresponding increase in the range of internal rotation of the tibia. This action would fall within the general rule, as the rule applies to the non-cavus foot.

When analyzed from an orthotic point of view, it becomes apparent that motions in the transverse plane are especially important to runners. The synchronous motions of depression of the longitudinal arch and internal rotation of the tibia, as well as the reverse motions of the raising of the arch with its synchronous external rotation of the tibia, are directly related to the length of stride of a heel-toe gait. What happens when these phasic motions are inhibited?

In order to answer this question, a closer look at the normal walking cycle from a fresh perspective was needed. We begin with a single step.

Imagine looking down from above on the lower half of a body as it begins to walk from a standing position (Figure 2). The left limb is in the mid-stance position as the right heel is about to contact the ground and begin its stance phase of the cycle. In the mind's eye, remove all portions of the skeleton above S1. We now look directly down on the pelvis and lower extremities. Draw an imaginary line in the sagittal plane which parallels the lower limbs. The line should be equidistant between both limbs and pass through the center of the first sacral vertebrae. This line represents the line of progression. Draw a second imaginary line perpendicular to the line of progression, so that it crosses the line of progression and passes through the center of the head of the left femur. This second reference line serves as the body's imaginary lateral midline (as viewed in the frontal plane). If the lower half of the skeleton were in a standing position (that is, if the left limb were to come to a stop and assume the standing position, instead of continuing forward to complete its stride in the cycle), the lateral midline would then pass through the center of the head of both femurs (Figure 2).

The total amount of external transverse rotation of the tibia is known to occur between mid-stance and push-off, whereas half the matching internal rotation occurs during the swing phase and is completed between foot-flat and mid-stance. All of the muscles within the limb in stance phase which are affected by these transverse motions are placed on stretch (eccentric contractions). None of them contributes any of the force necessary to effectuate transverse rotation during the stance phase. The role of the musculature, then, is to control external forces acting upon the limb in stance phase.

What is the significance of this arrangement to running? In seeking possible answers, it is first necessary to define the neutral position of the tibia and femur with respect to the transverse plane, as they relate to each other and to the foot. (Reference to the anatomic position would be confusing, since the tibiae are internally rotated when standing.) For this discussion, then, their neutral position is here defined as that position when all muscles within the limb are at their normal rest lengths, i.e., no muscles are elongated as a result of torque forces, nor are any contracted to generate torque forces. This definition immediately raises another question: At what instance(s), if any, during the walking and/or running cycle, are the tibia and femur in the neutral position? I submit that there can only be two instants when such is the case: just prior to heel strike, and just as the forward progression of the pelvis *et. al.*, moving in the transverse plane, reaches the lateral midline of the body at mid-

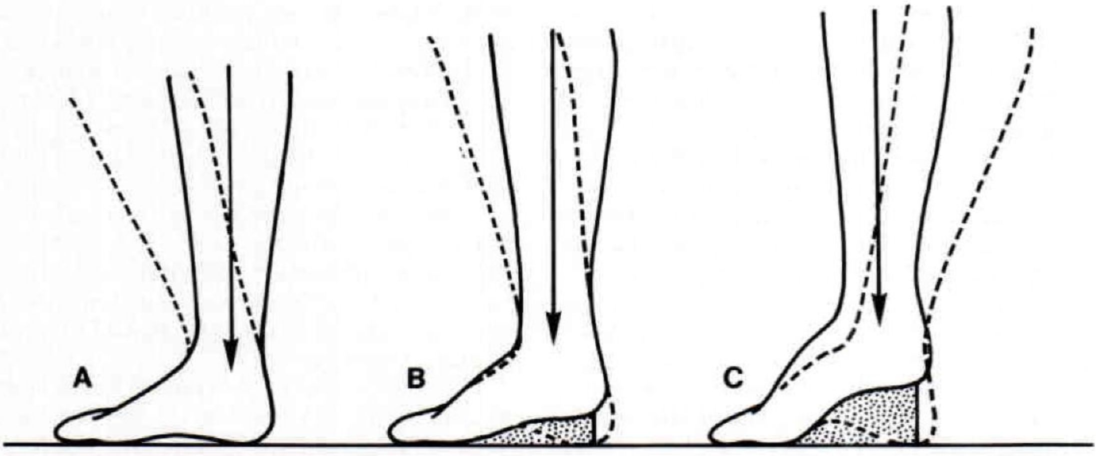

Figure 3. Foot/ankle complex's adjustments to forefoot drop to achieve standing balance in the A-P plane:

A. Tibia is vertical to a non-cavus foot and floor (solid line) for A-P balance. Downward arrow represents CG. Dotted line shows phasic 15 degrees of anterior rotation of the tibia prior to heel-rise at mid-stance.

B. Relationship of moderate forefoot drop to floor when tibia is in vertical position (solid line). Heel cannot reach the floor. Dotted line shows the two adjustments that allow the heel to make contact with the floor: 1. The midfoot and hindfoot are lowered together by rotation at the MP joints (the midfoot is hypomobile); 2. The tibia is rotated anteriorly to the degree necessary to restore standing balance in the A-P plane.

C. Solid-line drawing depicts severe fore-foot drop. Range of anterior tibial rotation is insufficient to achieve heel contact with the floor. The tibia must rotate posteriorly in order to gain the range required for restoring A-P balance. Unfortunately, hyperextension of the knee joints must accompany this adjustment.

A Glaubitz modification under the midfoot and heel (shaded areas, B and C) restores the tibia's normal perpendicular relationship as well as the full functional anterioposterior range to the tibia.

stance (Figure 4-A). It is only during these two instants that all muscles are at their normal rest lengths, due to the position of the tibia and femur whose actions affect them (with respect to motion in the transverse plane). Throughout the rest of the cycle, these same muscles are elongated beyond their normal rest lengths as each, to one degree or another, is "wound" around the tibia and/or femur. Once the foot has left the ground, these muscles are now free to "snap back" to their normal rest lengths. However, it is important to note that they do not go beyond their normal rest lengths ("snap back" being a passive activity), for to do so they would have to contract to generate the necessary force for such action. There is no evidence that concentric contractions are initiated by any of the extrinsic muscles of the foot to effectuate internal tibial rotation during the swing phase.

At heel-strike, then, both the tibia and femur are in the neutral position with respect to the transverse plane. Yet, only one-half of the full range of internal rotation has been completed (*i.e.*, the "snap-back" half of the range needed to match the full range of external rotation which had immediately preceded it). The other half of internal rotation occurs between foot-flat and mid-stance. Why? The answer lies further up the kinetic chain. It is the position of the pelvis at this time in the cycle, with respect to the lower limb which has just begun its stance phase, that points to the answer. In this case, maintaining a straight line of progression and economy of motion are the primary functions of nature's arrangement. The pelvis, with respect to the limb beginning its stance phase, is obliquely behind with references to the line of progression as viewed from above (Figure 2).

It is known that during fast walking, the increase in magnitude of the transverse rotary motions of tibia, femur and pelvis can exceed 50 percent of the range of average walking speed.[17] As our interest is in running, we will use the following ranges throughout the remainder of this discussion: tibia, 16 degrees; femur, 16 degrees; and pelvis, eight degrees. All of these convenient even numbers are within normal ranges for fast walking, and are, therefore, conservative estimates for running. It is not the magnitude of these motions alone that should attract our attention, because their direction and the phasic period(s) within the walking cycle in which they occur are equally important.

As walking speed increases, the magnitude of these motions increases proportionally. Therefore, it would seem to follow that during running, the proportional increase would be even greater, with cadence having the same relevance as it does during walking. A major component of cadence (when speed is the consideration) is an increase in stride length. Walking speed can be increased by stepping up the cadence without an increase in stride length. With the latter gait, however, Point A to Point B may be reached in a shorter time, but the cost in energy expended to distance travelled is disproportionately high. In order to cover more ground in less time (since increasing hip flexion alone results in an inefficient over-stride), a primary mechanism for increasing stride length is to increase transverse rotation of the tibia, femur, and pelvis.

Returning to the walking cycle, we pick it up as the left limb begins to rotate forward in the sagittal plane. The weight being borne by the right foot increases a-pace with the left limb's forward progression to mid-stance (Figure 4). From this point on, the body is being fully supported by the right foot, throughout the swing phase of the left limb. However, with all the body's weight positioned behind and medial to the right foot, forward rotation of the pelvis in the transverse plane is seriously compromised, unless accompanied by the phasic internal rotation of the tibia in the transverse plane, between foot-flat and mid-stance (Figure 5). In turn, the internal rotation of the tibia cannot occur without placing the extrinsic muscles of the foot on stretch as the tibia turns.

Keeping the values assigned previously, the normal course of events, with respect to the line of progression, would be eight degrees of internal rotation of the tibia to accommodate for the oblique, posterior position of the pelvis, which is four degrees counterclockwise to the right limb at foot-flat. Normally, the first eight degrees of external tibial rotation that occurs immediately following heel-rise at mid-stance, reverses the eight degrees of internal rotation the tibia was in, and thereby, along with the femur, reverses the four degrees the pelvis was in, thus bringing the pelvis to the lateral midline. The external rotation of the tibia and femur continues on to the completion of their respective ranges of 16 degrees each. The last eight degrees of external tibial/femoral rotation effectuates a four degree rotation of the pelvis forward of the lateral midline. The completion of the 16 degrees of external tibial/femoral rotation is reached at push-off. The muscles return to their normal rest length during the swing phase, causing the tibia and femur to return to their "neutral" positions (Figure 4).

Applying the same assigned values, what effect would either a partial or total absence of the lowering of the longitudinal arch and the synchronous internal tibial rotation have upon the running cycle? What are the conditions that can inhibit these two phasic motions during the heel-to-toe gait of the runner with non-cavus feet? Any condition that prevents depression of the longitudinal arch at mid-stance can be a causative factor.

For instance, when an insert is well molded to encompass the heel and longitudinal arch in their neutral positions and is sufficiently rigid to receive the full weight of the body at mid-stance without distorting, depression of the arch between foot-flat and mid-stance is blocked, as is the synchronous tibial internal rotation. The immediate result is a shortening of the stride length of the contralateral limb which is in swing phase. This is caused by denying the latter half of the full magnitude of normal transverse internal rotation of the tibia at the talocalcaneal (subtalar) joint. The heel everts at heel-strike, which normally "unlocks" the midfoot with respect to the hindfoot, but the rigid insert, as it firmly encases the hindfoot and midfoot together, maintains a neutral relationship between the two by mechanically "locking" both together. Albeit, the mid-foot still pronates with the eversion of the heel, but the rigidity under the arch prevents its phasic depression. Thus, two important motions, depression of the longitudinal arch and internal rotation of the tibia, are inadvertently eliminated from the normal sequence of events within the walking and/or the heel-to-toe running cycle(s) of non-cavus feet.

How does the elimination of these two motions effectuate a shortening of the stride of the limb in swing phase? Visualize a limb in stance phase that has been denied the two motions discussed as mid-stance is reached (Figure 5). At this time, the triceps surae of the contralateral limb pushes the contralateral limb off into its swing phase. This thrust provides the power to rotate the pelvis forward in the transverse plane about the vertical axis of the limb in stance phase. However, the tibia is in a neutral position, *i.e.*, the phasic internal tibial rotation that would normally occur between foot-flat and mid-stance has been effectively blocked by the rigid insert. The pelvis, taking with it the limb in swing phase, begins moving forward in the transverse plane about the vertical axis of the limb in stance phase. At this period of the cycle, the first half of the tibia's full range of external rotation is now superfluous, *i.e.*, it is not needed to rotate the tibia to the neutral position as mid-stance is reached because the tibia is already in the neutral position.

The force that the left triceps surae is able to generate, in order to pivot the pelvis *et. al.* forward under the circumstances just described, is more than equal to the task. The point of application of this force, at the left outer rim of the pelvis, gives it an enormous mechanical advantage. The weight of the pelvis *et. al.*, plus the required additional momentum, would apply a greater amount of torque than usual upon the muscles of the right limb that would increase the amount of their elongation to allow the tibia and femur to rotate beyond their normal ranges.

However, in this case, the latter eight

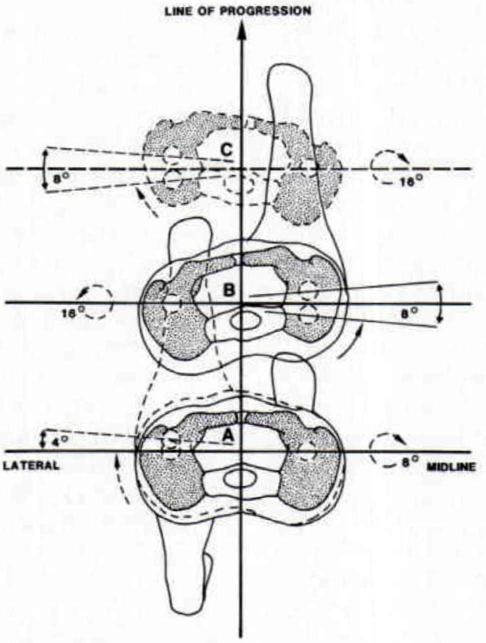

Figure 4. The contribution of rotations in the transverse plane to symmetry of stride. Walking cycle; schematic view from above.

A. The bony pelvis (shaded area) and its related contour (solid line) are shown in the mid-stance position, as indicated by the solid-line right foot and left limb, with its flexed knee, at the midpoint of its swing phase. The dotted outline of the pelvis and left limb have continued forward to heel-strike, completing the swing phase as shown by the dotted arrow on the left side. The dotted-line circle on the right gives the direction and amount of rotation of the right tibia to achieve the action presented. The preceding eight degrees of internal tibial rotation, having occurred between foot-flat and mid-stance, are not shown.

B. The bony pelvis and its related contour depicts the body's progress, upward and forward in the A-P plane, as the left femur extends over the tibia and the tibia rotates over the foot to mid-stance. The arrow on the right side shows the now counterclockwise forward rotation of the pelvis, from its previous position four degrees posterior to the lateral midline at foot-flat, on through to four degrees forward of the lateral midline, as the right limb completes its swing phase. The dotted-line circle on the left gives the direction and amount of rotation of the left tibia (the total range) to achieve the action presented. Again, the preceding eight degrees of internal tibial rotation, so essential to the symmetry of stride, could not be shown.

C. The dotted outline of the bony pelvis demonstrates that the upcoming full stride should be a mirror image of the preceding full stride (shown in B)—the same amount of upward and forward progression of the body through space in the A-P plane, the same amount of pelvic and tibial rotation, but both now in a clockwise direction. Though not shown, the preceding internal tibial rotation was in a counterclockwise direction, the opposite of B's stride.

Note: Since our primary interest is the foot/ankle complex, all reference to transverse rotations of the femur were dropped. However, it is understood that internal/external rotations of the femur are phasic with and closely mimic the range and direction of the tibia throughout the walking and heel-to-toe running cycles.

degrees of external rotation would force the affected muscles to stretch twice their usual required elongation. The stretching of these muscles is a passive action. It is known that muscle fiber cannot be passively stretched beyond 60 percent of its normal rest length without rupturing.[27] The mechanical advantage of the triceps surae's action upon the pelvis is such that the muscles within the right limb can be stretched quite easily beyond the danger point. All else being equal, however, tearing at a muscle tendon junction, especially in persons over 30, is more likely to occur than rupture of muscle fibers. In younger individuals, evulsions occur more easily than ruptures of tendons.[27] Figure 5 also demonstrates that there is another very practical reason for limiting the range of transverse rotations for the purpose of lengthening one's stride under the circumstances just described.

It is now evident that rotation of the tibia, femur, and pelvis in the transverse plane can be directly related to the length of stride. The amount of increase to stride length that transverse rotation can safely contribute is directly related to the fact that these rotations cannot occur without placing a large number of muscles, within the limb in stance phase, on passive stretch (eccentric contractions). The mechanical advantage the action of the triceps surae of the contralateral limb has upon the pelvis, suggests that tendon tissue within the limb in stance phase could be subjected to excessive stress.

A substantial increase in push-off thrust of the triceps surae and anterior tibialis of the contralateral limb would now be necessary. This increase in thrust at push-off would be necessary to replace the assistive force that would have been supplied in the form of phasic eccentric contractions by the musculature of the limb in stance phase. This would seem to be a contradiction of a previous statement that there is no evidence that the musculature of a limb contributes to rotation in the transverse plane during stance phase. Such is not the case. The primary biomechanical function of all eccentric contractions is control. Nevertheless, when a muscle is placed on stretch beyond its normal rest length, its elastic properties enable it to store energy in a manner not unlike an elastic band when stretched. Like the elastic band, a muscle utilizes the energy thus stored to return itself to its normal rest length. Having gained the desired control, nature also uses the energy resulting from the eccentric contraction to return the tibia and femur to their neutral position during swing phase. It is the absence of this secondary source of energy, due to the loss of internal tibial rotation between foot-flat and midstance (bearing in mind that the foot is receiving a rapid increase in load during this period), that necessitates an increased thrust by the contralateral limb. There is an immediate cost, an increase in energy expenditure.

The absence of internal tibial rotation also eliminates placing the musculature of the limb in stance phase into the necessary amount of eccentric contraction to control the tibial/femoral counterclockwise rotations (as viewed from above). As seen from above, the pelvis, as in normal circumstances, would still be positioned obliquely four degrees posterior to the lateral midline of the right limb at mid-stance. Once the pelvis is posterior to a limb in stance phase, the magnitude of external tibial rotation must be limited to one-half of the representative 16 degrees to avoid injury.

From the foregoing analysis, we arrive at these conclusions: Actual forward progression of the pelvis, through space via the transverse plane, is a primary function of external tibial/femoral rotation. The functional range of external rotation for a step forward (using the previously assigned values) is eight degrees (Figure 2). The functional range for a full stride forward (using the same assigned values) is 16 degrees (Figure 4). The occurrence of internal tibial/femoral rotation between foot-flat and mid-stance is a prerequisite to extending a step into a stride.

To inhibit internal rotation of the tibia—that normally occurs between foot-flat and mid-stance—is decidedly not in the best interest of a runner. To do so causes a dilemma because his desire for speed

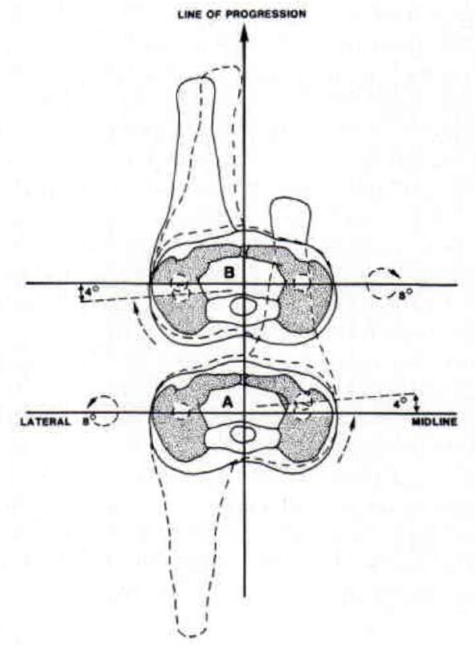

Figure 5. Effect upon stride length when phasic depression of the longitudinal arch is inhibited. Walking cycle; schematic view from above.

A. The bony pelvis (shaded area) and its outer contour (solid line) are shown in the standing position. Dotted-line contours show right limb at heel-strike and left limb at heel-rise, *i.e.*, the beginning of the double-support phase. The pelvis has rotated forward (counterclockwise) in the transverse plane about the left limb, as shown by the dotted arrow on the right side. The range and direction (counterclockwise) of the tibia's rotation are represented by the dotted-line circle on the left side.

B. The left limb has pushed off. The body has progressed through space, upward and forward in the A-P plane, as the right femur extends over the tibia and the tibia rotates over the foot to mid-stance. The solid-line contour of the pelvic region, the heel-strike position of the left limb, and the forefoot portion of the right foot (indicating mid-stance) demonstrate an arrested forward advance of the pelvis at the lateral mid-line following the phasic external rotation of the tibia and femur in the transverse plane. Such an arrest of forward advance of the pelvis can result from wearing a molded rigid insert which inhibits phasic depression of the longitudinal arch and its synchronous medial tibial rotation in the transverse plane, between foot-flat and mid-stance. As a result of the blockage of phasic medial rotation (counterclockwise) of the right tibia (not shown), its external range is shortened by half to eight degrees (dotted-line circle on right side) because the tibia has been out-of-phase in a neutral position from foot-flat to mid-stance. The dotted-line contour of the pelvic region and left limb demonstrate that any attempt at the full range of external rotation (clockwise) of the right tibia cannot effectuate forward progress of the pelvis beyond the lateral midline. Note how straightforward progression would be jeopardized.

urges him to a full-length stride, yet inhibition of internal tibial rotation dictates a shortened stride in order to avoid injury.

We have found why the latter half of internal tibial rotation in the transverse plane occurs between foot-flat and mid-stance. It is the key to maintaining control of the forward advancement of the pelvis during a heel-to-toe gait. The tibia's completion of the latter eight degrees of internal tibial rotation, between foot-flat and mid-stance, ensures a straight line of progression and symmetry of stride. Eight degrees of external tibial rotation reverses the internal tibial rotation, thus bringing the trunk to the lateral midline. Eight degrees of external tibial rotation brings the trunk and contralateral limb to heel-strike position. Symmetry is achieved during each swing phase by a form of "catch-up" previously referred to as "snap-back."

CONDITIONS I–VIII

The following is a list of eight conditions of the foot/ankle complex which appear with regularity in our Sports Medicine Clinic. A discussion of the variety of complaints made by patients who presented one or more of these eight conditions would be beyond the scope of this report. Orthotic management of their symptoms was clinically determined by the process of elimination. That is, all known possible causes (other than biomechanical ones) for each of the patients' complaints were judged to be unrelated. Each condition listed will be discussed individually from an orthotic point of view, followed by the orthotic solution and its rationale.

I. *Slight to mild excessive pronation in the standing position*

The patient has sufficient flexibility to raise his longitudinal arch voluntarily, while standing, without raising his heel or forefoot from the floor. When viewed from the back, his Achilles tendons indicate that the usual related heel eversion occurs. The patient is free of symptoms in all activities except running.

II. *Severe pronation in the standing position*

Hypermobility: Patient can raise the longitudinal arch voluntarily with ease. Transverse external/internal rotation of the tibia and femur are very apparent with the raising and lowering of the arch. The knee also shifts in and out of a valgum position with voluntary lowering and raising of the arch. Severe eversion of the heels is evident. The patient reports that his feet are often bothersome during the business day.

III. *Moderately rigid flat feet, usually in conjunction with bowed tibia*

This condition is usually reported to have been asymptomatic until the patient began running.

IV. *Functional forefoot drop ("cavus" foot), unilateral or bilateral*

Previously unknown to patient. The condition is hidden by involuntary, excessive dorsiflexion or plantar flexion at the ankle joint in order to effect heel contact in the standing and mid-stance positions. Patient was without symptoms prior to taking up running.

V. *Limited dorsiflexion*

Hypomobility: inability to rotate tibia forward beyond the neutral or mid-stance position. Not responsive to Achilles tendon stretching exercises.

VI. *Loss of eversion of heel*

Hypomobility: os calsis cannot evert beyond neutral position.

This condition is most common among the runners seen in our clinic.

VII. *Heel in fixed inversion*

Hypomobility: range of fixed inversion of os calsis seen as much as 20 degrees from neutral position. Amount of inversion may or may not be the same bilaterally. The patient is symptom free during activities other than running.

VIII. *Hypermobile transverse tarsal joint*

Abnormal pronation of midfoot occurring in this joint. This condition has been seen in isolation. When not a part of a gen-

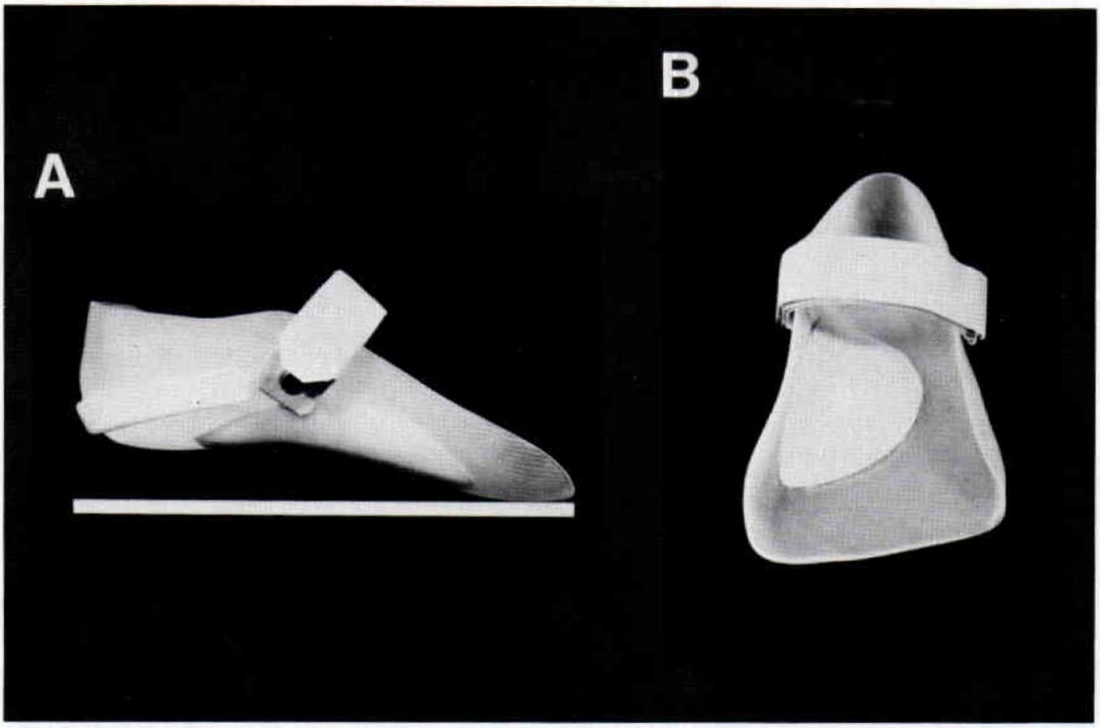

Figure 6. The Flexible Polypropylene Insert: A. Viewed from the medial side, showing the molded reinforced heel portion, the Velcro® instep control strap and the PPT foam longitudinal pad seen through the thin polypropylene. Note the amount of forefoot drop. B. Three-quarter anterior view.

eral condition of hypermobility, it seems to be the result of long-term running, following the loss of phasic eversion of the heel and/or limited anterior tibial rotation due to functional forefoot drop.

CONDITION I

Comments from an orthotic point of view

If a patient has normal range of eversion/inversion of the os calsis and is asymptomatic throughout his daily activities, other than running, his range of pronation when standing is directly related to the angle of his subtalar joint axis and therefore natural to him. However, introduce to such a pair of feet a regimen of running, which automatically increases the impact to the lateroposterior border of the heel by a factor of 2.5 to 3.0 times body weight, and injury would seem to be only a matter of time.

This occurs for the following biomechanical reasons: The increased impact to the lateroposterior border of the os calsis increases the moment of heel eversion by the same factor of 2.5 to 3.0. It is reasonable to assume that such a force causes a greater degree of eversion of the heel to occur, particularly as supportive musculature tires during a long run. The increase in the range of heel eversion automatically increases the amount of pronation of the midfoot.

The angle of the axis of the subtalar joint normally determines the "neutral" position of the longitudinal arch. However, given repeated applications of force to the os calsis (of the magnitudes known to occur when running) for several thousand cycles in rapid succession and as muscles fatigue, such forces are more than likely to weaken the ligaments supporting the multiple jointings within the midfoot. Since hindfoot eversion and midfoot pronation are passive motions, i.e., they are not initiated by muscular activity but by external floor reaction forces, control of the range of either motion is determined by the A-P angle of the subtalar joint axis and by the geometry of the articulations involved. An increase of the interspaces between these articulations, due to weakened ligaments, can lead to serious breakdown of the biomechanical checks that keep these motions within normal ranges. That attrition can be an important contributing factor to the cause of injuries, when biomechanical malalignment and/or weakened ligaments are present, is a conclusion that is hard to dismiss.

Orthotic solution to Condition I

1. A longitudinal arch pad is cemented to the upper surface of the removable innersole of a new running shoe(s). It is important to this conservative treatment that

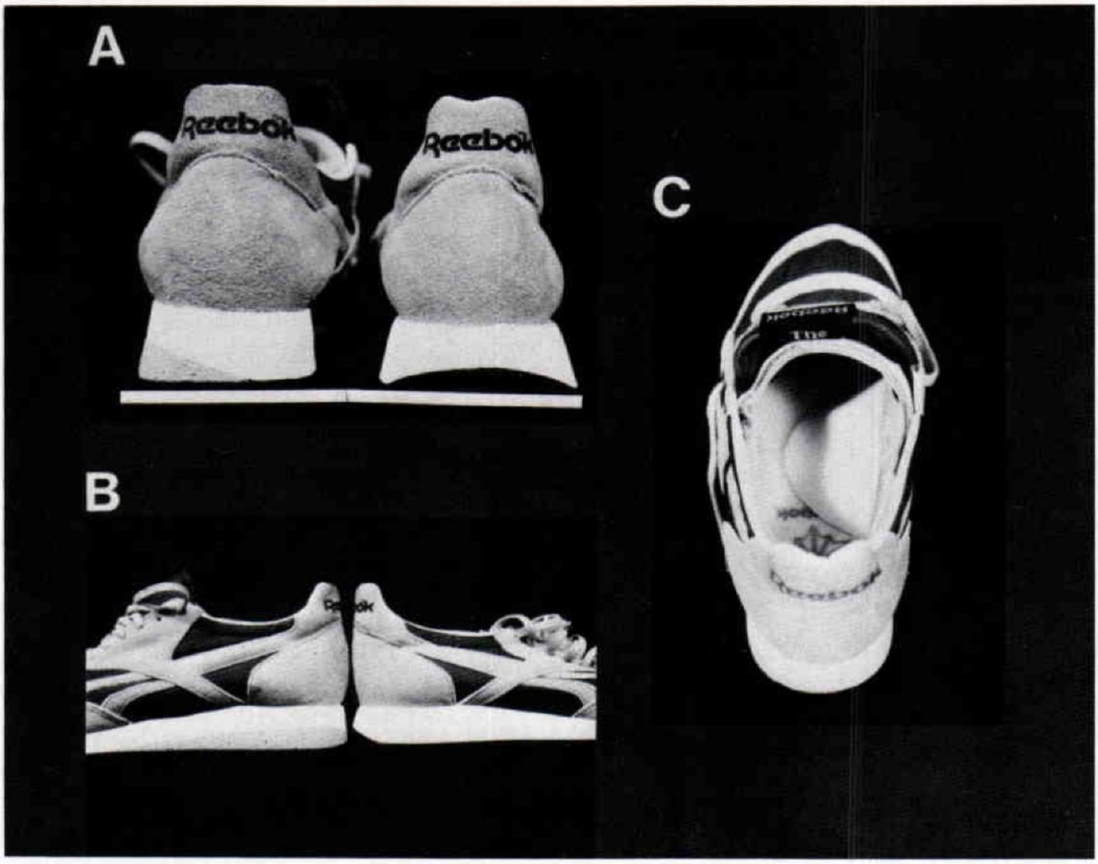

Figure 7. The Lateral Cushion Heel Wedge. A. Posterior view. The cushion wedge is shown on the posteriolateral portion of the left shoe. Note the downward inclincation to the lateral side. B. Lateral view showing the tapering of the cushion wedge to a zero point, distal to the fifth metatarsal head. C. A flexible polypropylene insert shown in a running shoe. The thinness of the polypropylene is evidenced by the design on the shoe's innersole showing through the bottom of the insert.

the running shoes, if previously worn, have not been distorted by the patient's pronated stride. If the running shoe does not have a removable innersole, the arch pad is cemented onto the inner surface at the proper location. The arch pad is made of 0.5-inch thick PPT™ foam.

2. A lateral cushion heel wedge is applied to the lateral border of the running shoe(s) (Figure 7). It is also important not to attempt to apply the cushion wedge to a distorted shoe(s). (See Orthotic solution to Condition II for a description of the lateral cushion heel wedge.)

Rationale for use of lateral cushion heel wedge for Condition I

The PPT™ arch pad offers dynamic control of the compressive effect of the body's weight upon the foot during mid-stance, but it cannot control excessive pronation at any point between foot-flat and mid-stance, because pronation has already occurred by then, i.e., at heel-strike, along with the eversion of the heel. The lateral cushion heel wedge controls heel eversion at heel-strike by limiting its range to normal dynamic requirements of the runner's cycle. (See Condition II for a description of how the lateral cushion heel wedge functions.)

CONDITION II

Comments from an orthotic point of view

We have just concluded that the purpose of the latter half of internal tibial rotation occurring between foot-flat and mid-stance is related to bringing the pelvis (when obliquely posterior to the limb in stance phase) forward to the lateral midline. In the lateral midline position, the pelvis is perpendicular to the line of progression, and the tibia and femur are in their neutral position in relation to the line of progression as well as to the right foot in stance phase (Figure 4).

From its lateral midline position, the pelvis continues its transverse rotation about the right limb, forward of the lateral midline. The left foot now completes its stride as its heel contacts the ground. The force generated by the weight of the pelvis and its forward momentum has externally rotated tibia and femur in the right limb simultaneously. As the double support period is reached (as viewed from above), the pelvis' position, with respect to the right limb, is forward, *i.e.*, the reverse of its relationship in the preceding double support period. Thus, the symmetry of internal/external rotation of the tibia and femur, which is essential to a straight line of progression, is maintained throughout the cycle. If one is in doubt as to the contribution of these transverse rotations to economy and efficiency of gait, observe a toddler walking or running. His side-to-side wobbling gait is largely due to the lack of (as yet unlearned) rotary motions in the transverse plane.

Excessive depression of the longitudinal arch is accompanied automatically by excessive internal rotation of the tibia. This condition places the foot's related extrinsic muscles continuously on stretch throughout the stance phase. This constant stretching during weightbearing causes permanent elongation of the muscle tissue. Unlike a normal, in-phase, eccentric contraction, the rest length of the muscle tissue is overextended, thus reducing the control associated with eccentric contractions. The motive power of concentric contractions is also seriously compromised. The most difficult dysfunction to treat, that results from such hypermobility, is the hindfoot's loss of control of the midfoot between heel-rise and lift-off.

Orthotic solution to Condition II

A flexible polypropylene shoe insert is vacuum-formed of $^{1}/_{16}$ inch polypropylene over a plaster of Paris model of the patient's foot. The form is trimmed to cup the entire heel. The medial and lateral trim lines are extended to encompass approximately one-third of the dorsal surface of the foot. The distal trimline is at the distal edge of the metatarsal heads, in order to permit freedom for full flexion of the MP joints and to avoid any impingement of the metatarsal heads by the edge of the insert. Before forming the polypropylene, a stock, preformed, firm rubber longitudinal pad is cemented to the model under the arch position. Once the polypropylene is formed, a longitudinal pad of the same shape (3/8 to 1/2 inch thick at its center) is made of PPT™ foam. The trimmed insert is placed over the model and the outline of the firm rubber pad is traced on its outer surface. The PPT™ foam longitudinal pad is cemented to the inside of the insert within the outline. A Velcro® strap is attached to cross over the proximal portion of the instep when additional control is needed (Figure 6).

Out-of-phase flexibility of the 1/16 inch polypropylene insert is controlled by a snugly laced running shoe. Since the uppers of running shoes are made of woven materials, when drawn in by the laces the upper hugs the insert firmly from the sole level up about the sides and over the top of the instep. The flexibility of the materials used in running shoes is such that the uppers quickly assume the shape of malaligned feet, whereas the polypropylene insert, although flexible, is not stretchable and will not assume unwanted shapes. There is a "marriage" of the design characteristics of both the insert and the running shoe, each complementing the other to ensure the integrity of hindfoot to midfoot throughout the running cycle. Thus, the normal ranges of phasic motions are not inhibited, but abnormal magnitudes and/or out-of-phase motions are not permitted.

To achieve the control just described, it is essential that the dynamics of both heel and midfoot be free to perform their motions in proper sequence and within their normal ranges throughout the running cycle. This is accomplished by encompassing the heel and midfoot in a single, intimately fitting polypropylene form of the foot in the neutral weightbearing position. The mechanical encasement of heel and midfoot of a severely pronating foot mimics the normal funciton of the heel at heel-strike, *i.e.*, the midfoot and forefoot follow the direction of the heel at heel-strike, but the normal amount of depression of the longitudinal arch will not occur prior to the rapid buildup of weight between foot-flat and mid-stance. The intimately formed insert holds the hypermobile midfoot to its normal relationship to the heel without inhibiting the dynamics of a normal amount of depression of the arch, which is allowed as the PPT™ pad under the arch compresses. With controlled depression of the midfoot, synchronous internal tibial rotation also occurs in the proper phase of the cycle. During this same period, the insert offers sufficient resistance to prevent destructive magnitudes of either of these two motions.

As the heel begins to rise and the center of gravity moves over the forefoot, the normal action of "locking" midfoot to hindfoot by the inversion of the heel is assimilated by the intimate encasement of both these regions of the foot. The design again mimics nature by ensuring that, as the heel inverts, the midfoot must go with it—*i.e.*, the longitudinal arch rises and the tibia synchronously rotates lateralward, in the transverse plane, in their normal sequence in the cycle. Thus, excessive pronation is prevented from occurring, without inhibiting normal dynamics, at a time in the cycle that is particularly destructive. The control just described is particularly important in cases where the transverse tarsal joint is hypermobile because, as the heel inverts and the tibia rotates externally, the midfoot will continue to pronate out-of-phase. In such situations, the forces acting upon the extrinsic muscles that insert onto the midfoot are of a high magnitude, and these muscles are in danger of being stretched beyond their passive limits. Also, the flexibility of the insert's distal portion allows normal forefoot abduction to occur, in phase, without interference.

Although the design of the flexible polypropylene insert returns control of the midfoot to the heel without interfering with normal dynamics of the foot, there is one other essential control that it cannot provide. The insert cannot control the magnitude of the impact to the posteriolateral border of the heel, which is generated by the ground reaction force at heel-strike. In short, neither this nor any other insert can control eversion of the heel. This is because the motion is passive, that is, the motion is caused by an external force prior to the

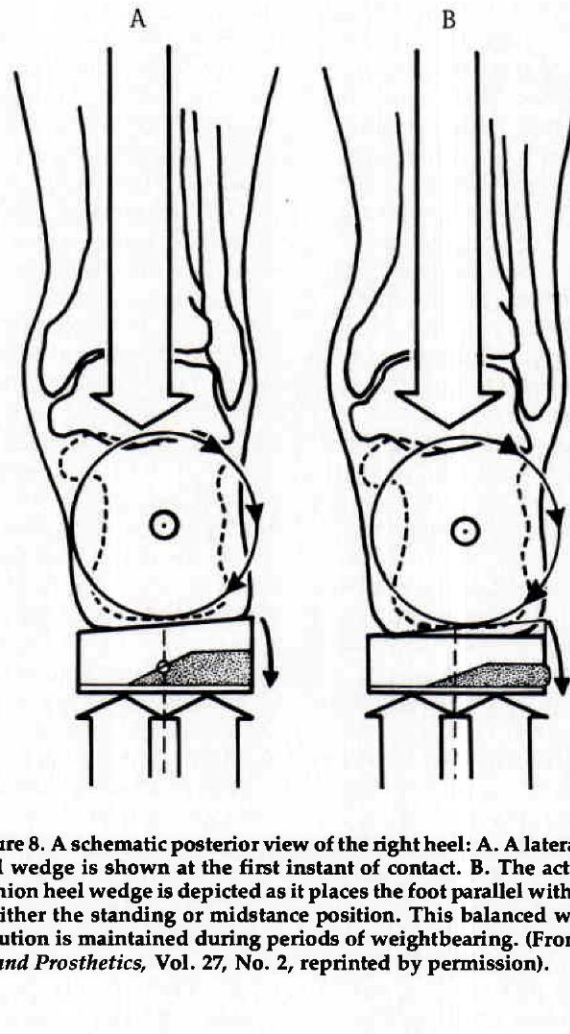

Figure 8. A schematic posterior view of the right heel: A. A lateral cushion heel wedge is shown at the first instant of contact. B. The action of the cushion heel wedge is depicted as it places the foot parallel with the floor, in either the standing or midstance position. This balanced weight distribution is maintained during periods of weightbearing. (From *Orthotics and Prosthetics,* Vol. 27, No. 2, reprinted by permission).

weightbearing phase. The floor reaction force will evert the heel to whatever degree allowed by the axis of the subtalar joint, regardless of the surface contouring within the shoe on which the heel rests.

Orthotic solution to Condition II: The lateral cushion heel wedge (U.S. Patent No. 3738373)

A lateral cushion heel wedge[9] (Figure 7) is incorporated into the shoe heel, in conjunction with the flexible insert, to control unwanted effects from impact at heel-strike. The cushion heel wedge is an integral part of the dynamic control system for the hypermobile foot. The control provided by the wedge is twofold: it reduces the magnitude of impact, and it limits heel eversion at heel-strike to normal dynamic requirements (Figure 8).

Design Rationale

Whereas, conventionally, a solid medial heel wedge is used to block valgus motion of the heel statically by placing the heel in a position of inversion upon weightbearing, the action of the lateral cushion heel wedge is diametrically opposite in function. When two shoes, one with a solid medial wedge and the other with a lateral cushion heel wedge of equal height, are placed side by side on a flat surface and viewed from behind, the medial side is higher on the shoe with the medial wedge, but the lateral side is higher on the shoe with the lateral cushion heel wedge. At heel-strike, the foot of the runner wearing the solid medial heel wedge is positioned upon an unyielding, inclined plane which prevents the heel from achieving a balanced or neutral position in the M-L plane. The foot of the same runner wearing a lateral cushion heel wedge would experience the following:

The instant the posteriolateral border of the shoe heel contacts the ground, the lateral cushion heel wedge begins to compress as the superincumbent weight from above rapidly increases. As the cushion wedge compresses, it absorbs and thereby delays for an instant the progress of the initial weight passing through it down to the ground.

Within that first instant, the lateral half of the heel due to its greater thickness, contacts the ground slightly before the solid medial half has made contact. By the time the cushion wedge has compressed to a level parallel with the ground, the solid medial half of the heel has also made contact.

The lateral cushion heel wedge cannot change the rate of descent of the body's weight to either side of the heel. It can, however, and does, effect a change in the amount of weight that reaches the ground at a given instant in time, *e.g.*, at the instant of heel-strike. The change is due to the "storing" of an unknown amount of weight by the lateral cushion heel wedge. The result is a slight difference in the amount of force making contact with the ground at any given instant in time, between heel-strike and heel-rise, through the medial and lateral halves of the shoe heel. In turn, the floor reaction force is proportionately imbalanced either side of the shoe heel. With the medial half of the heel receiving a greater force than the lateral half, a force-couple is produced which acts dynamically to maintain the os calsis in a position parallel to the ground (Figure 8).

As the tibia rotates anteriorly and internally following foot-flat, bringing the body's center of gravity forward and medialward over the midfoot, the load upon the os calsis increases rapidly. However, this rapid buildup of weight upon the shoe heel cannot affect the imbalance of ground reaction forces caused by the lateral cushion heel wedge, once the force-couple, acting upon the subtalar joint, is activated at the instant of heel-strike. With the preponderance of ground reaction force now passing medial to the axis of the subtalar joint, a moment of inversion is now acting upon the subtalar joint, instead of the phasic eversion moment that would otherwise be generated at heel-strike. The heel does not go into inversion, however, because the superincumbent weight and the floor reaction force on the medial side of the heel have equalized.

In order to produce a force-couple to control dynamically the eversion moment about the A-P axis of the subtalar joint at heel-strike, the action of the cushion heel wedge must be very quick. For example, Cavanaugh and Lafortune[4] report that a runner, running at a speed of six minutes a mile, travels from heel-strike to midstance in 42 milliseconds (ms). The mean ground contact time for 12 heel-toe runners tested was 188 ms per each foot at the same speed.[4]

A most pertinent finding from the same study was that the Z force was not recorded before a magnitude of 50 Newtons (110.23 lbs) was reached. This magnitude of vertical force was reached approximately two ms after heel-strike. These figures indicate that the lateral cushion heel wedge must compress much faster than two ms in order to shift successfully the preponderance of oncoming superincumbent weight to the medial side of the heel, before rapid buildup can overpower the lateral cushion wedge and result in a failure to control eversion of the heel. The speed with which the lateral cushion heel wedge must react at heel-strike is indicative of the need for the wedge to be incorporated into the shoe heel. For instance, were the cushion wedge placed inside the heel of the shoe, or under the heel portion of an insert, the delay in reaction time, in either case, would be sufficient to render it useless to runners.

CONDITION III

Comments from an orthotic point of view

The oblique angle at which the body's weight is received by the feet is due to the bowed tibiae. Were nature not to make this accommodation, the feet would be forced to receive the superincumbent weight in an untenable varus position. The oblique angle of the tibiae in a medial direction dictates that the feet accommodate by pronating. The lateral to medial direction, from which such feet receive the body's weight, due to the tibiae's lateral bowing, generates a moment that everts the os calsis about the axis of the subtalar joint, a moment of greater mechanical advantage than under conventional circumstances. The pronation accompanying bowed tibiae is both necessary and "natural." Consequently, no attempt should be made to alter the alignment of the foot/ankle complex which the patient presents when standing.

With few exceptions, these patients were asymptomatic prior to taking up running, and they continue to be without discomfort except when running. What phase(s) of the running cycle are most abusive to the runner with bowed tibiae and pronated feet? Decidedly, at heel-strike and, if the transverse tarsal joint is weakening, from heel-rise to push-off. At heel-strike, the moment to evert the heel is 2.5 to 3.0 times the runner's body weight versus 1.2 times when walking. Since these individual's heels are already aligned well beyond the normal eversion range at impact, with respect to their tibiae, it is reasonable to assume that heel-strike is the most damaging phase of the cycle to the foot/ankle complex. Once the heel leaves the ground and begins to rotate into inversion to lock the midfoot in preparation for push-off, the longitudinal arch, already depressed beyond the normal range of conventional alignment, may remain depressed if the os calsis' inversion action cannot control the midfoot due to a loose transverse tarsal joint.

Orthotic solution to Condition III

- Bilateral, lateral cushion heel wedges to running shoes, which will check further eversion of the os calsis at heel-

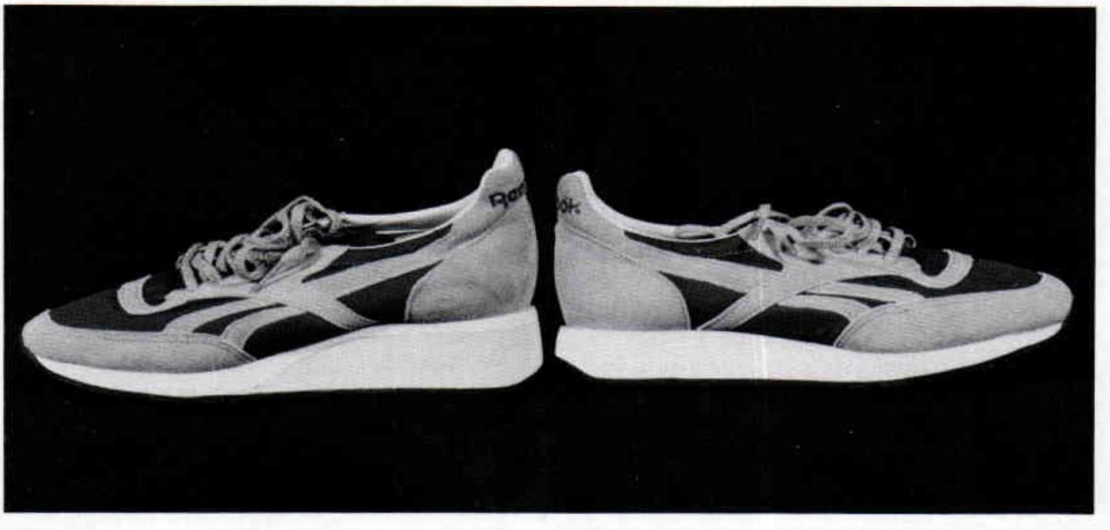

Figure 9. Lateral view of running shoe with Glaubitz modification for functional drop foot on left. Standard shoe on right.

strike in the manner previously described.

- Commercial Spenco® compressible arch supports to resist further unwanted depression of the longitudinal arch, if required, for the reason previously stated. A firmer arch support would be too restrictive, uncomfortable and, not incidentally, would inhibit the dynamics natural to such feet.
- If the above fails to give relief, the problem is a lack of anatomic control at lift-off. A flexible polypropylene insert with reinforced heel portion and Velcro® instep strap is required to restore contol of the midfoot (Figure 6).

CONDITION IV

Comments from an orthotic point of view

This condition is not easy to detect because it is natural to the patient, and he/she has throughout his/her life unconsciously masked the condition in order to effect heel contact in the standing and/or midstance positions. This the patient does in either of two ways: 1) anterior tibial rotation, the more common adjustment, is accompanied by posteriorward rotation of the pelvis to bring the center of gravity back over the feet to maintain postural balance in the A-P plane (When viewed from the side, the patient's posture is mildly reminiscent of the balancing stance of the paraplegic); 2) posterior tibial rotation is accompanied by hyperextension of the knees to bring the center of gravity forward over the feet to maintain balance in the A-P plane (Figure 3).

To the heel-to-toe runner, the unconscious postural accommodation for functional forefoot drop, *i.e.*, anteriorward rotation of the tibia at midstance, is significant for both their immediate and longterm effects. An immediate result of this accommodation is a shortening of the stride length of the contralateral limb, because at midstance the tibia of the limb in stance phase, which is normally identical to standing alignment, may have no tibial anterior rotation range left. The amount of anterior tibial rotation left at midstance is proportional to the amount of forefoot drop. For example, for each 1/16 inch the plantar surface of the forefoot is lower than the heel's plantar surface, the tibia will rotate forward approximately two degrees to achieve a plantigrade position. Thus, a 1/4 inch difference between the forefoot and the heel requires approximately five degrees of anterior tibial rotation, a 1/2 inch difference requires approximately ten degrees, etc. The greater the amount of forefoot drop, the more of the normal maximum range of 20 degrees of anterior tibial rotation (dorsiflexion) about the talocrural joint[13, p .28] is used out-of-phase to achieve balance in the A-P plane during midstance.

During normal walking or heel-to-toe running, between midstance and toe-off, the tibia rotates anteriorly 15 to 20 degrees before the heel leaves the ground. The tibia of a runner with a 1/2 inch forefoot drop, for example, is already in a position of ten degrees of anterior rotation at midstance. If he is running fast, it is reasonable to assume he will utilize the full 20 degree range of tibial anterior rotation as he extends the stride length of the limb in swing phase. Consequently, since 50 percent of the total range has previously been used in an out-of-phase manner, the stride of the limb in swing phase will be shortened proportionally. The runner has but one option to make up for the deficiency in stride length—he must extend the weight of his body vertically by "vaulting" over the MP joints of his forefoot on the stance phase side. Preceding the vaulting, the heel must rise prematurely, which action prolongs the time (by 50 percent) that the midfoot must bear the brunt of the body's weight during the running cycle. The increased stress placed upon the transverse tarsal joint under such circumstances is not hard to imagine, especially as the extrinsic muscles of the foot tire during a run. Also, it is not unreasonable to assume that energy expenditure increases substantially due to having to raise the body higher with each cycle of the run—a use of energy that is counterproductive to efficient forward progression. Functional forefoot drop ("cavus" foot) also subjects the Achilles tendon to severe stretching which can cause tendonitis and, if ignored, microtearing of the tendon.

When the condition is unilateral (which in our experience has been infrequent), there is an additional factor to be considered, *i.e.*, once the heel of a limb with forefoot drop is off the ground, the involved limb is functionally longer than its opposite member. Since the full range of in-phase anterior tibial rotation is blocked, the normal 15 to 20 degree range of knee flexion that simultaneously occurs between heel-rise and toe-off is limited to a proportional degree; hence the functionally 'longer' limb and further need to vault over the MP joints.

This functional leg length discrepancy and the vaulting that accompanies it causes an asymmetrical running gait which is manifested in three important ways: The involved limb causes the uninvolved limb to drop lower to achieve heel-strike, thereby increasing the impact; the trunk will flex laterally to the low side and place abnormal stress upon the hip abductors on the involved side, particularly the gluteous medius muscle; the stride length of the uninvolved lower limb will be shorter. When this condition involves both lower limbs, stride length is shortened bilaterally, and a highly wasteful increase in energy expenditure due to bilateral vaulting is a consequence.

Orthotic solution to Condition IV

An adaptation to the Glaubitz shoe modification[8, 24] is used to restore the tibia to its normal position perpendicular to the foot and ground when standing and/or at midstance during the cycle. The construction of running shoes is such that they do not lend themselves to the conventional Glaubitz forefoot drop modification (Figure 9).

For mild forefoot drop, a lift contoured to the heel and midfoot is placed inside the shoe. When the patient is standing or the limb is in the midstance position with the Glaubitz modification, the aphasic anterior tibial rotation adjustment is no longer necessary in order for his heel to make contact with the ground. Thus, the normal range of anterior tibial rotation can now occur in phase, *i.e.*, between midstance and toe-off, because the tibiae's range is no longer dissipated by an out-of-phase motion to lower the heel to make contact with the ground.

When the forefoot drop is unilateral, a matching lift is applied in or to the shoe on the non-involved side. This is necessary to level the pelvis. It is also necessary to prevent the development of genu recurvatum in the involved limb. These heel lifts are made of firm, lightweight crepe. A softer material is not used because, when resting upon the average running shoe's .75 to 1-inch polyvinylacetate (PVA) foam midsole, the additional softness causes M-L instability.

CONDITION V

Comments from an orthotic point of view

When the foot is non-weightbearing, it cannot dorsiflex, either passively or actively, beyond the neutral position. The runner is unable to rotate the tibia(e) forward beyond the neutral position at midstance, causing premature heel-rise and the necessity to vault over the M-P joints. The tight Achilles tendon does not respond to stretching exercises. This condition results in a shorter stride length and an increase in the percentage of time within a cycle that the midfoot must support the superincumbent weight. Great stress is placed upon the transverse tarsal joint(s), particularly as the extrinsic and intrinsic musculature of the foot fatigue. Aphasic pronation is likely to occur following heel-rise in the attempt to compensate for the absent phasic anterior tibial rotation.

Orthotic solution to Condition V: The rocker bar

The rocker bar or "rollover"[23] is used for the same purpose as it is used generally, *i.e.*, to relieve stress upon the transverse tarsal joint by mechanically raising the heel at the proper time within a walking cycle. This is

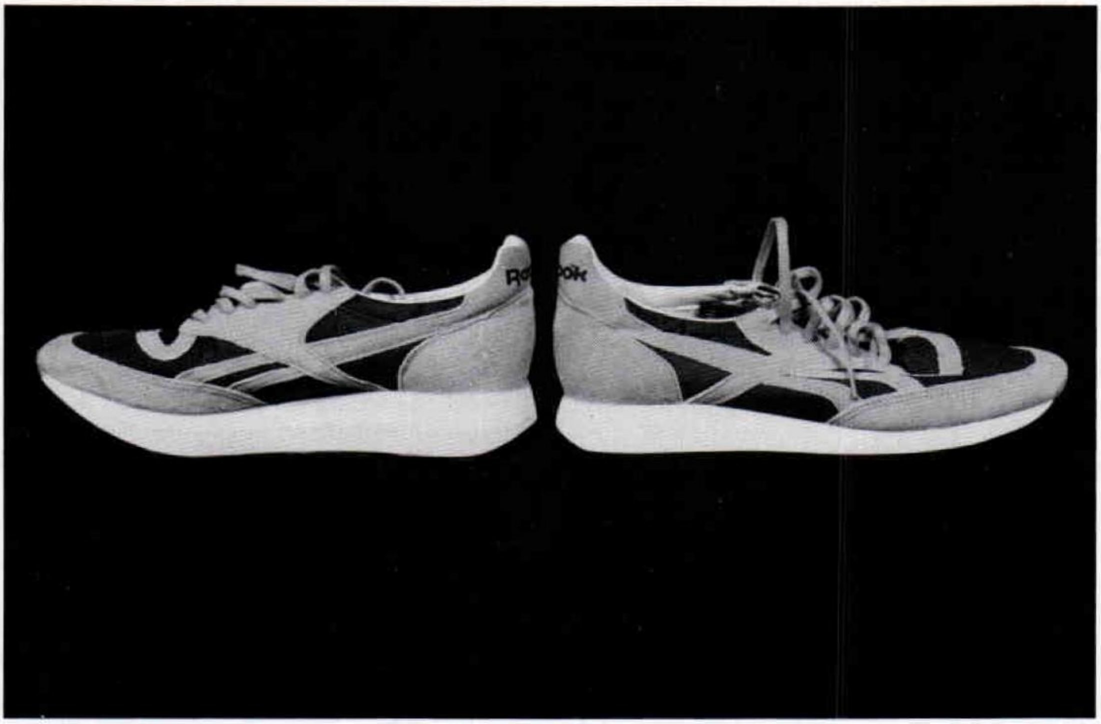

Figure 10. Lateral view of running shoe with rocker bar or "rollover" on left. Standard shoe on right.

done by 'rolling' the body's weight smoothly onto the forefoot (Figure 10). This action of the rocker bar is of primary importance to a runner, as it eliminates the necessity to vault over the M-P joints and restores the normal length of his stride. The timing of heel-rise is regulated by proper positioning of the rocker bar's apex with respect to mid-stance and the M-P joints.

Applying a rocker bar to a running shoe is not as simple a procedure as applying one to a conventional shoe. The process is complicated by two features of running shoes: their gridded contact soles and the need to preserve these soles' high friction surface, and the "wedgie"-like flatness of their contact surface.

The preservation of these two features, friction and flatness, necessitates removal of the gridded sole and the addition of a full-length, ½ to ⅝ inch thick, lightweight, firm crepe to the midsole. The addition is then sanded to form an apex directly behind the metatarsal heads to zero inches at the toe end and angled at the heel end to match the original heel contour. The gridded friction sole is then recemented over the added crepe rocker midsole. The rocker should raise the heel a minimum of 15 degrees (Figure 10).

Care must be taken not to introduce M-L instability by increasing the sole height to a point that magnifies the M-L moments acting upon the foot during stance phase. To prevent unwanted M-L instability, grind away an appropriate amount of the PVA foam midsole before cementing the firm crepe material to it. Unless there is a leg length discrepancy, a matching lift must be added to the opposite shoe in order to maintain a level pelvis.

CONDITION VI

Comments from an orthotic point of view

The os calsis cannot evert beyond the neutral position. This condition is very common among runners seen in the Indiana University Sports Medicine Clinic. All but "elite" runners have a heel-toe running gait. A three-year videotape study of the Boston Marathon showed that 98 percent of the participants "generally strike the ground on the outer heel."[2, Part III]

Since depression of the midfoot (longitudinal arch) is known to have a shock absorption function, and that eversion of the heel automatically places the foot in a pronation mode, prior to the occurrence of the depression, it is reasonable to assume that everting of the os calsis at heel-strike is an integral part of the shock absorption mechanism of the foot. It would appear to follow, then, that inability to evert the os calsis may seriously inhibit the shock absorption function of the foot. Furthermore, it is likely that detrimental consequences may affect the entire kinetic chain, particularly with regard to attritional factors over the long term. When one considers the force with which a runner's heel strikes the ground, it is difficult not to reach the conclusions expressed above.

Orthotic solution for Condition VI: The lateral cushion heel wedge

It has already been described how the lateral cushion heel wedge prevents excessive pronation, when used in conjunction with an insert, by dynamically controlling heel eversion beyond normal range. The rationale for the use of the lateral cushion heel wedge, when eversion of the heel is not anatomically possible, is based upon its ability to reduce the magnitude of impact at heel-strike.

Prior to heel-strike, the descent of the body may be said to be "falling free." The first point of contact is the lateroposterior border of the heel. The ground reaction force, at this point lateral to the axis of the subtalar joint, generates a moment of eversion to the heel. The magnitude of this moment is linked to the velocity of the body's fall. For example, the force of impact to the lateroposterior border of a shoe heel of the currently best shock absorption rated running shoe[6] absorbs all but 1.62 Gs of the impact force, which is normally at least 2½ times the runner's weight. The kinetic chain is subjected to a shock of 1.62 times the body weight when this shoe is worn, instead of 2½ times the body weight.

A runner wearing shoes with the above rating, for example, who weighs 150 pounds, strikes the ground with an impact of 375 pounds. Were he wearing a running shoe incapable of absorbing any portion of a 2.5 G force, the ground reaction force would be equal to the full impact of 375 pounds.

Let us assume that the lateroposterior point of contact of his shoe heel is a half-inch from the axis of his subtalar joint. Multiplying the half-inch radius to the force line by 375 pounds produces a moment of 187.5 inch pounds, or 15.6 foot pounds. The shoes with the above rating are said to absorb all but 1.62 Gs (or 35 percent less) of the 375 pounds, which would reduce the ground reaction force to 244 pounds.[6] When the 244 pounds is multipled by the half-inch radius, the product is now 122 inch pounds, or 10.1 foot pounds of moment to evert the heel. Whether it is 15.6 or 10.1 foot pounds of moment acting to evert a heel that anatomically cannot evert, either one is very jarring to the kinetic chain. A distance runner may run 130 km (80 miles) per week in training, which would subject each lower limb to approximately 40,000 such impacts over each seven-day period.[4]

The lateral border of a running shoe with a lateral cushion heel wedge is 3/16 inch thicker at its heel portion than at the shoe heel's medial border. This lateral wedge tapers to zero inches to a point just behind the head of the fifth metatarsal (Figure 7). From the instant of contact at heel strike, the posteriolateral border of the cushion wedge compresses rapidly, well before the first peak load is reached. The extra thickness of cushion provides an instant delay in transfer of the initial superincumbent load through the lateral portion of the shoe heel to the ground. As previously described, within the minute time frame that immediately follows heel-strike, an imbalance of floor reaction forces is effectuated by the cushion's action, *i.e.*, a higher magnitude under the medial side, there being no delay of the descending superincumbent load on its side.

However, it is the lateral border of the heel that first makes contact with the ground and therefore, the first imbalance of floor reaction forces is lateral to the subtalar axis and generates a moment of eversion. Also, the time frame in which this initial imbalance operates is extremely short, since the superincumbent weight is still falling at a rapid rate.

Whether or not the os calsis is anatomically able to evert, a moment to evert it is generated at each heel-strike. However, the lateral cushion heel wedge has served its purpose, for as it continues to compress, the medial portion of the heel is receiving a rapid and equal increase in vertical load. Since the free-fall weight has been uninterrupted on the medial side from the beginning, by the time the cushion has compressed to the point that the shoe heel is parallel to the ground, the floor reaction force is of greater magnitude on the medial side of the subtalar joint axis. The result is a moment of inversion to the heel which is in effect throughout the short time the heel remains in contact with the ground.

Thus, the lateral cushion heel wedge efficiently reduces the initial shock at heel-strike and simultaneously reduces the eversion moment to the heel, while maintaining the heel parallel to the ground during weightbearing.

The sequence of events just described occurs so rapidly during running that the role of the lateral cushion heel wedge for Condition VI is exclusively that of a very efficient shock absorber. This shock absorber effect is due to the anatomical inability of the os calsis to respond to a relatively light eversion moment which the wedge effectuates in the first instant of heel contact. The shock absorption is twofold: The subtalar joint is relieved of a high magnitude jolt in the form of an eversion moment, and the rest of the kinetic chain is relieved of a severe vertical jolt normally generated by the ground reaction force at heel-strike.

CONDITION VII

Comments from an orthotic point of view

The amount of fixed inversion of the os calsis seen in clinic has been as much as 20 degrees. The amount of inversion may or may not be the same bilaterally.

Biomechanically, this condition presents the same problems as Condition VI, limited heel eversion. However, it is quite possible that the eversion moment normally generated at the instant of heel-strike may cause additional premature inversion of the heel due to its inverted position. If this is the case, the lateral cushion heel wedge can inhibit this out-of-phase heel motion by reducing the impact at heel-strike.

Orthotic solution for Condition VII: The lateral cushion heel wedge

The rationale is the same as that for Condition VI.

CONDITION VIII

Comments from an orthotic point of view

When standing, the patient presents an abnormally pronated midfoot, but when viewed from behind the os calsis is not everted (the Achilles tendon is perpendicular to the ground). The integrity of hindfoot to midfoot has been lost. The most serious consequence of this condition is loss of the heel's ability to "lock" the transverse tarsal joint as it rotates into inversion following mid-stance. The result is that as the hindfoot inverts and the limb simultaneously rotates externally in the transverse plane, the midfoot will rotate medially in the frontal plane as the center of gravity advances over the midfoot.

With the flexing of the metacarpophangeal joints to the toe-off position, the center of gravity moves forward over the forefoot, the region where maximum vertical forces occur at push-off.[4] It is at this end of the stance phase that the runner's pronated midfoot is most vulnerable. The "Spanish windlass" action of the plantar aponeurosis, which passively raises the longitudinal arch as flexion occurs at the MP joints, is also inoperative when hindfoot/midfoot integrity is lost.

Orthotic solution to Condition VIII: The flexible polypropylene insert

The rationale and use of the flexible polypropylene insert are essentially the same as described for Condition II, severe pronation of the midfoot. However, due to the hypermobility of the transverse tarsal joint, restoring control of the midfoot to the hindfoot requires two additional components (previously mentioned) to the flexible insert design: a corrugation-like reinforcement is incorporated into the heel portion when the 1/16 inch thick polypropylene insert is vacuum-formed; a Velcro® strap which passes over the proximal area of the instep is attached to the insert (Figure 6). The purpose of adding one or both components is to prevent mediolateral spreading of the heel portion of the insert, which reduces the intimacy of fit, resulting in loss of control. The amount of laxity present in the transverse tarsal joint is usually a reliable indicator of whether the Velcro® instep strap will be necessary. Also, the Velcro® strap is mandatory for heavy individuals. Molding the insert out of thicker polypropylene for a heavier person is neither feasible nor tolerable, because its rigidity interferes with the dynamics of the running cycle.

During the past two years, 30 runners seen in our Sports Medicine Clinic have been using the flexible polypropylene insert with gratifying success. There are 65 runners now using the lateral cushion heel wedge; many of them have worn out several pairs of running shoes with cushion heel wedges during this period.

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ACKNOWLEDGMENTS

I wish to express my appreciation for the cooperation of all the runners who have attended our Sports Medicine Clinic. I am particularly indebted to Craig Pinkus and Robert Weddle, whose patience, grit, and consistently clear feedback were so helpful. Also, my thanks to Professor Damian Howell, Warren Springer, and Norman Berger for their thoughtful commentary on earlier drafts. I owe special thanks to my wife, Helen, for her typing and editorial suggestions throughout the preparation of this report.

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