0198-0211/87/0704-0197$02.00/0
FOOT & ANKLE

The Biomechanics of Lower Extremity Action in Distance Running

Peter R. Cavanagh, Ph.D.
University Park, Pennsylvania

ABSTRACT

The role of quantitative biomechanical measurements in the evaluation of the running patient is discussed. Many techniques are now available to provide insight into the external mechanics of lower extremity action during running, and results from such measurements are presented for symptom-free subjects at distance running speeds. Details of stride length, stride time, and foot placement are first presented followed by a discussion of kinematic data, including stick figures, angle-time graphs, and angle-angle diagrams for the sagittal plane motion of the hip, knee, and ankle joints. The measurement of rearfoot motion, as an approximation of coronal plane subtalar joint movements, is also discussed. Results from acceleration, force, and pressure measurements are considered, and the assertion is made that bilateral asymmetry in many of these measures is a fairly common finding. There are, at present, few reports in the literature of the application of biomechanical techniques to the evaluation of patients with running injuries. It is anticipated that there will be rapid developments in this area in the future and that this will provide considerable insight into the etiology, diagnosis, and treatment of running injuries.

Only a decade ago, football injuries formed the largest group of sports injuries that occupied the time of most orthopaedic surgeons. The "running boom" of the late 1970's and 1980's has brought a new population of injuries into the office, and this population is different from the classic sports injured group in a number of ways. These patients are often older, less muscular, more questioning, less inclined to follow recommendations that involve the cessation of activity and more resistant to surgical intervention. They frequently have two bags of running shoes, three volumes of training diaries and a list of four or five differential diagnoses of their own. Their injuries are also often considerably more subtle than most sports injuries. X-ray findings are often negative, since soft tissues are involved more frequently than bone. In summary, the patient and the injury are about as different from the classical sports injury as they could be.

The orthopaedist, who has been trained to perform complex surgical procedures from joint replacement to extensive reconstruction, is expected to offer counsel on all aspects of the runner's problem. This often extends from identification of the injury and its cause to possible cure through avenues as broad as orthotic therapy, physical therapy, and changes in footwear and running mechanics. It is not surprising that many clinicians feel ill prepared by a typical medical education and training to deal with the runner and his or her problems. Fortunately, there are a number of places to look for help. An earlier running symposium by Mann and colleagues in this journal[36] should be required reading for all those about to face running patients for the first time. Various texts and many articles on running injuries[4,12,17,34,45,46] are also available, and there has been a general acceptance that a knowledge of the biomechanics of running is important to an understanding of many running injuries.[29,33,46]

Much of the information on running mechanics is scattered through the scientific rather than clinical literature, and much of it does not consider the issue of clinical application. An extensive review of the available literature in the area is given by Williams.[48] It is the purpose of the present article to present a coherent review of the biomechanics of distance running as they are related to the foot. The reader is not required nor expected to be versed in mechanics or mathematics. The techniques that are now available for measurement of the interaction between the foot and the ground allow tremendous insight into lower extremity action in running. In the same way that running shoes have led the way in the application of technology to footwear, it is likely that sports medicine will play a similar role in the routine introduction of biomechanical measurement techniques to orthopaedic practice. This article is intended to show what can be measured and how it gives insight into lower extremity function during distance running.

Since the use of quantitative techniques in clinical situations is in its infancy, this review is, with few

Professor of Biomechanics, Pennsylvania State University, University Park, Pennsylvania. Address requests for reprints to Dr. Cavanagh at The Center for Locomotion Studies, Pennsylvania State University, University Park, Pennsylvania 16802.

exceptions, concerned with patterns measured from the symptom-free individual. There are, at the present time, few studies in the clinical literature where the techniques of biomechanics have been applied in the diagnosis, treatment, and follow-up of running patients. Perhaps this fact alone will stimulate some readers to apply the methods presented here. In a recent paper, Saleh and Murdoch[43] have made a compelling argument for the use of quantitative techniques in gait analysis of prosthetic walking. In running, the problem of visual analysis is compounded by the increased angular velocities at the joints and the very brief ground contact times. Only by the use of quantitative dynamic measurement techniques can we gain true insight into the mechanics of running.

It must be stated at the outset that, even in the symptom-free individual, there is still much that remains unstudied. Our knowledge is almost exclusively what might be called "external mechanics." That is to say that we know what forces are applied to the foot and how it moves, yet we know almost nothing about the effects of these forces and movements on the anatomy of the foot. There are no published reports of cineradiography that could take us beyond a static view of the anatomy. There are only a few electromyographic studies of the muscles of the leg and foot during running.[19,21,35,36,42] Attempts to establish cause and effect relationships between running injuries and running mechanics are mostly based on anecdotal reports. Epidemiological studies of running injuries are still in their infancy. Thus, at the present time, the principal reason to study the biomechanics of running is more to enable us to ask the right questions than to find the right answers. But as more people begin to use the appropriate tools to ask the right questions, more and more right answers will be forthcoming.

To put the action of the foot in proper context it will be necessary to broaden our horizons above the ankle joint. The patterns of motion of the hip and knee joints are intimately related to the action of the foot. It is therefore important to view foot motion in relation to the position of other parts of the body and the timing of their movements. It is also important to realize that almost everything that we care to study about the running gait will change as the speed of running changes.[31] This review concentrates on the speed range used by most noncompetitive runners that the clinician will see—between 3 and 4 meters/sec (or about 9 min/mile—6:40 min/mile in the units that runners prefer). The statements made here must therefore be interpreted with caution outside this speed range. In particular, they will not be appropriate to describe running at the speeds used by elite distance runners who typically compete at speeds between 5 and 6 meters/sec. A large number of studies in the literature have been conducted at 3.83 meters/sec, which is 7 min/mile; thus, many results at this speed will be quoted in the present paper.

STRIDE VARIABLES

Stride Length at Different Speeds

The distance and time between successive contacts of the same foot with the ground are known as the stride length and stride time. Step length and step time refer to similar quantities for successive contacts of opposite feet. As Figure 1 shows, stride length increases and stride frequency decreases fairly linearly over the speed range of interest to us. It should be realized that a runner does not have a single stride length which he or she uses at every speed. Stride length increases as the speed of running increases, and there is always some variation during a run, even on level ground. On uphill terrain, stride length shortens while in downhill running, a longer stride is used.

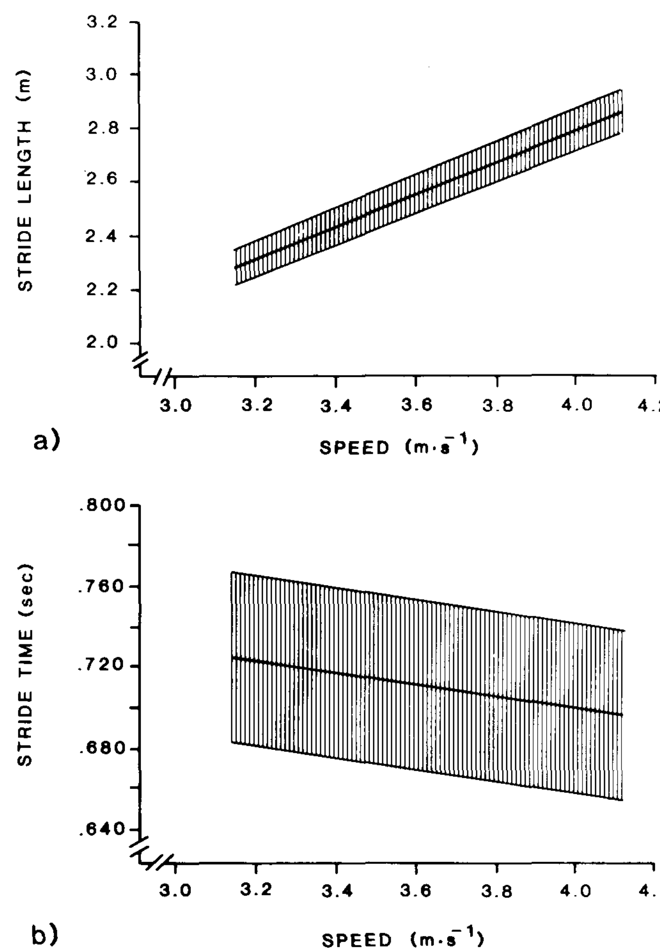


Fig. 1. Stride length (*a*) and stride time (*b*) for a group of asymptomatic runners at speeds between 3 and 4 meters per second. The shaded area represents one standard deviation.

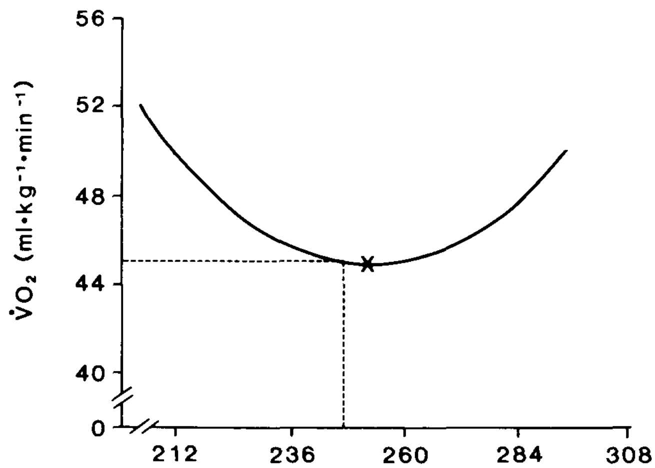


Fig. 2. Stride length (in cm) plotted against oxygen uptake (in ml/kg/min) at a given speed of running. Notice that at both long and short strides the energy cost is higher than at the optimal speed (marked by the × on the curve). The subject's freely chosen stride length (indicated by the dashed line) resulted in an oxygen uptake that was extremely close to optimal.

Optimal Stride Length

It has been shown[10, 26] that, at a given speed of running, each individual has a stride length that is "optimal" in terms of minimizing the metabolic energy needed for running. This finding, illustrated in Figure 2, implies that stride lengths longer or shorter than this optimal will involve a higher energy cost. It is intriguing that most of the runners tested have used a freely chosen stride length that was very close to optimal, suggesting that the body is able to perceive those conditions that involve minimum energy cost. Forcing an individual to lengthen or shorten his or her stride will involve an increased energy cost of perhaps 1% or 2%, but there may be occasions when, in spite of the metabolic penalty, it may be advisable to suggest a change. In the case of heel pain, for example, a shorter stride may be advisable to enforce a more anterior strike on the foot.

Contrary to popular belief, tall subjects do not always have longer optimal strides than shorter individuals. Freely chosen stride length correlates only moderately (r, approximately 0.6) with stature.[20] In our optimal stride length study,[10] the tallest subject in the study had the shortest stride of any of the subjects, and that particular stride was optimal for him. Since there is considerable variation, stride lengths in the speed range of interest may be anywhere from approximately 1.2 to 1.8 times the stature. This may increase to 2.2 times the stature at racing speeds.[7] Because of these variations, the running speed should always be monitored during tests on runners so that variation due to speed effects can be minimized on repeated trials. Also, because of individual differences between subjects and the lack of correlation with anthropometric dimensions, it is unlikely that a judgment about the appropriateness of a particular runner's stride length can be easily made.

Timing of the Phases of the Running Gait

The typical timing of stride events at 3.83 meters/sec is shown in Table 1. Notice that in Table 1, and throughout this article, we use the term "foot strike" rather than "heel strike" to describe initial foot contact with the ground. This is because, as will be apparent later, many runners never strike with their heels.

It is suprising to note from Table 1 that the foot is on the ground for only about two-tenths of a second.[1] In fact, the phase most often implicated in injury—the initial pronatory phase—is usually over in one-twentieth of a second (50 msec). Notice that, at 3.83 meters/sec, a complete cycle of running, defined as the time between two ipsilateral foot strikes, last only seven-tenths of a second. Under these conditions the foot strikes the ground about 5,100 times during an hour of running, a value that, when considered in relation to the magnitude of the forces involved, gives a background for the occurrence of overuse injuries that are so common in running.

Foot Placement

It is a fairly simple matter to record the placement of the feet during running. Inked markers on the sole of the shoe leave a trace that can be measured quite accurately. The interpretation of such data is complicated by the fact that the instantaneous direction of running is not always in the straight line joining the starting and ending points. This is illustrated in Figure 3 where the direction of running during the course of a 25-meter run for three individuals has been calculated according to the method of Chodera and Levell.[11] Notice that none of the runners travel in a single straight

TABLE 1
Typical Timing of the Phases of Gait at a Speed of 3.83 meters/sec

Phase of gait	Duration of phase (msec)	Time from foot strike (msec)
Ipsilateral foot strike		0
Initial pronatory phase	50	50
Mid-support phase	130	180
Toe off		230 (approx.)
Contralateral foot strike		350
Ipsilateral swing phase	450	230–700
Second ipsilateral foot strike		700

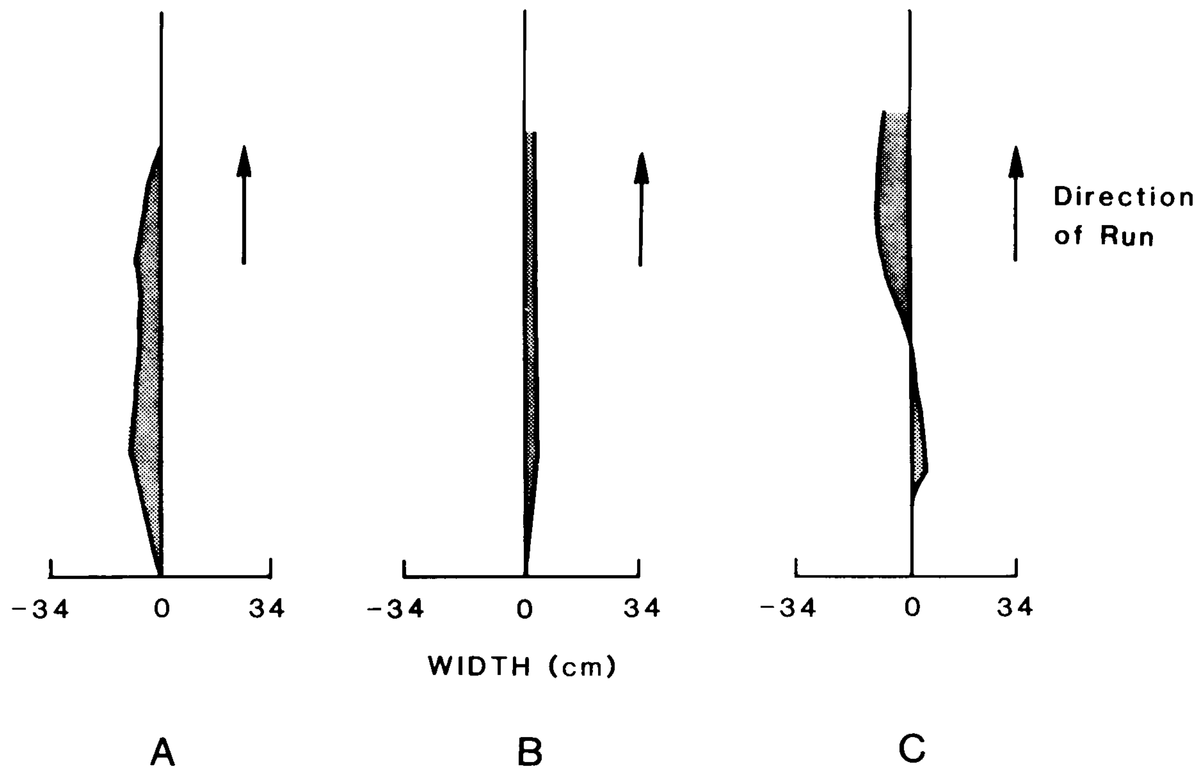


Fig. 3. Instantaneous directions of running for three different symptom-free subjects asked to run between two points. Subject A makes a large initial direction error which is corrected after a period of straight line running, subject B makes a single correction, while subject C oscillated around the desired direction. If foot placement angles were measured during these changes of instantaneous direction, extreme values would be found.

line; they each have a different strategy for getting from the starting to ending points. The "homing process" of one runner is to make a large initial mistake followed by a single correction, while for another it is a series of overcorrections resulting in oscillation about the mean path. Suppose that the angle of the axis of the shoe (that we shall call the foot placement angle) were to be measured with respect to the mean direction at one of the points where direction correction was being made. It is clear that an overestimation of the amount of abduction would be made for the right foot, while the estimate for the left foot would show an overestimation of the amount of adduction. The methods described by Chodera and Levell[11] can be used to overcome this problem.

The variables measured from such foot placement data have traditionally been called the "angle and base of gait."[39] While the term "base of gait" to describe the mediolateral distance between the placement of the feet seems quite appropriate, the present author has always rejected the term "angle of gait" since there are as many angles that can be obtained from various parts of the body during gait as one cares to measure. "Foot placement angle" seems to be a more appropriate term.

Typical values for the base of gait during running approach zero. This pattern, shown in Figure 4, means that both feet basically fall on the same straight line during running. In contrast, during walking the footprints usually fall in two distinct lines. This difference is accounted for by the fact that there is more "functional varus" of the entire support limb in running because of the period of single support. In running, the foot must be placed more nearly under the center of mass of the body to prevent the load on the hip adductors of the support limb from becoming too great. There are individuals who display both positive and negative bases of gait during running. The negative base of gait, which has been called "crossover,"[7] occurs when the right foot is placed to the left of a line defining the midline of the body, and vice versa for the left foot. The runner shown in Figure 4 exhibits crossover on the right side. Dynamic asymmetry, such as this, between the left and right sides of the body is a fairly frequent finding in asymptomatic subjects—an issue which is discussed further in the section on force measurement. One would intuitively expect the placement of the foot in crossover to be a very stressful situation for the structures on the lateral aspect of the knee joint, and one which would involve greater subtalar joint pronation to achieve foot flat. There is, at present, no direct evidence to support either of these suppositions.

Morton[38,39] considered the angle of gait to be a fairly

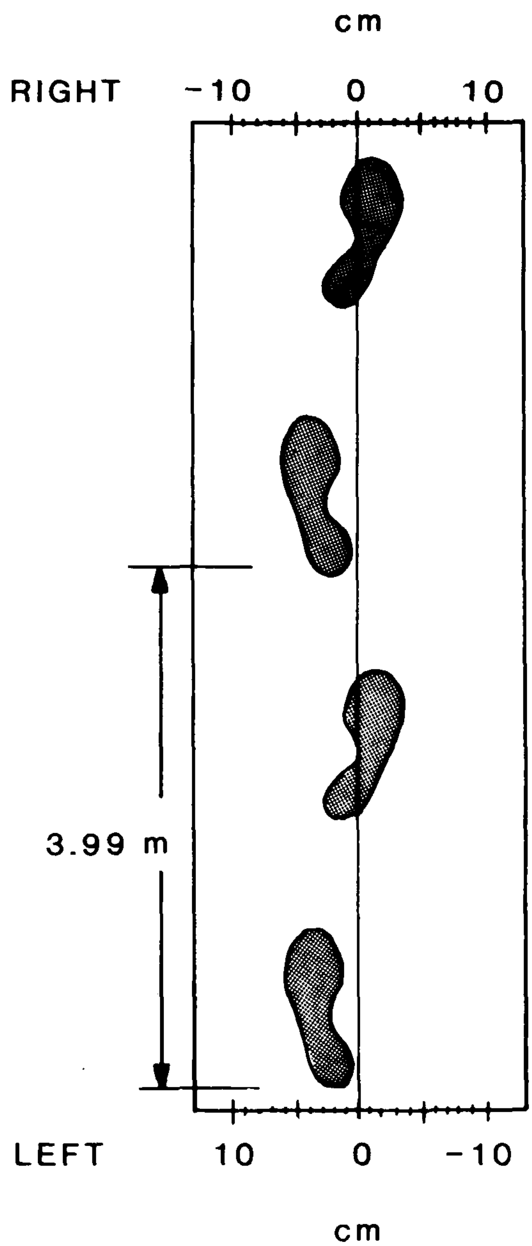


Fig. 4. Foot placement data for a champion cross-country runner at 6 meters/sec. Notice that the feet fall basically along the same straight line. The line drawn through the footprints represents the average mediolateral location of a marker at the L5 level. The right foot is said to exhibit "crossover" because it is placed to the left side of this line. The foot placement angle on the right side is also greater (13.8 and 9 degrees on the L and R, respectively). Note that the mediolateral scale is magnified approximately 20 times with respect to the a-p scale.

simple matter. He believed that it was greatest during standing, intermediate during walking, and least during running. Subsequent research has shown that such a clear relationship does not appear to exist in many individuals.[27] While many features of the running gait, such as ground reaction forces and most joint angles, do change systematically and monotonically as the speed of running increases, foot placement angle does not seem to be one of them. There is a large variation of placement angles at the same running speed in a symptom-free population, and the trends with speed are as varied. Our own studies[27] indicate that the most common trend is toward more abduction of the foot as speed increases (4.7, 5.1, and 9.0 degrees for slow, medium, and fast running, respectively). This pattern was, however, displayed by only half of the population, and almost every other possible permutation was observed. For example, some individuals showed most adduction of the foot at high speeds, while others reached a maximum value of abduction at some intermediate speed, and then exhibited more adduction at higher speeds.

It is certainly a mistake to think of the foot as fixed throughout the entire contact phase in running. Although this issue has received only scant attention in the literature, the fact that the foot tends to adduct during mid-support implies that there are rotational moments in the leg. Measurement of this tendency to adduct the foot in support may, in the future, be found to have some diagnostic value in knee injuries.

While much of the information in this section should be considered informational, the measurement of stride variables can be useful in detecting left/right asymmetries which can lead to insight into possible pathology.

KINEMATICS

Raw Coordinate Data and Stick Diagrams

The field of biomechanics has developed partly because the human eye is an inadequate instrument for studying the fine details of fast human movement. The motion of the parts of the body during the running cycle can be studied using a number of techniques. Historically, high speed cinematography has been employed to record the movement of the body in running. Subsequent frame-by-frame analysis or "digitizing" of the film yields coordinates from which joint angles and landmark motion can be calculated. Although high speed film is still our most accurate form of data collection, the analysis of data from cine-film is laborious—a fact that makes it unsuitable for clinical use. In recent years optoelectronic techniques such as SELSPOT and high speed video methods such as VICON and EXPERTVISION have become available. These forms of instrumentation, in conjunction with a typical laboratory minicomputer or microcomputer, allow the display of quantified movement patterns either simultaneously with the movement or immediately afterward.

The most basic display of the kinematics of running is shown in Figure 5a, where the location of certain anatomical landmarks on one side of a subject running

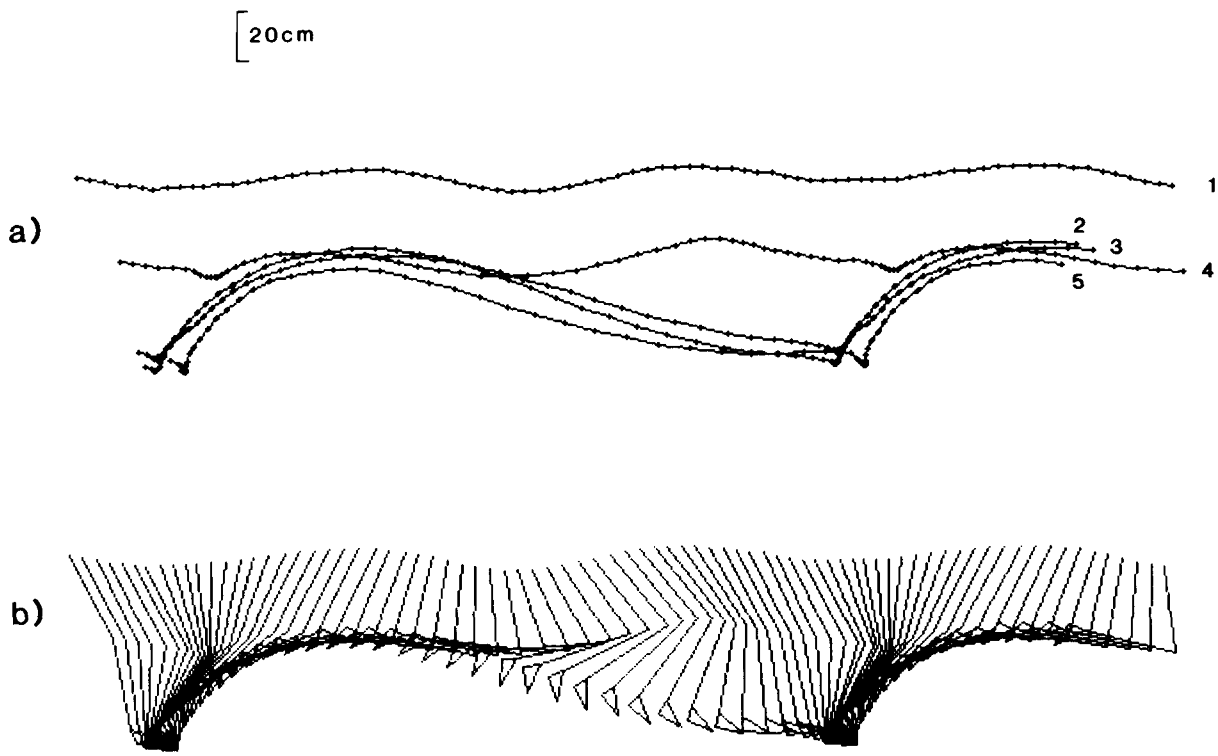


Fig. 5. a, The motion of targets on the hip, knee, ankle, and foot (two targets) during running at 4.2 meters/sec. These data were collected using an EXPERTVISION video-based system. The amplitudes of movement of the hip, knee, and ankle markers are 11, 17, and 43 cm, respectively. *b*, A stick diagram representation where the landmarks, shown in Figure 5a, that were measured at the same instant of time have been joined to show the posture of the limb.

overground are shown in a sagittal plane. Such a display was available to biomechanics researchers almost a century ago.[37] Notice that the amplitude of vertical movement increases distally, with the hip moving up and down by about 11 cm, the knee by 17 cm, and the ankle by 43 cm. The sets of landmarks that were measured at the same instant of time are joined to form a stick diagram in Figure 5b.

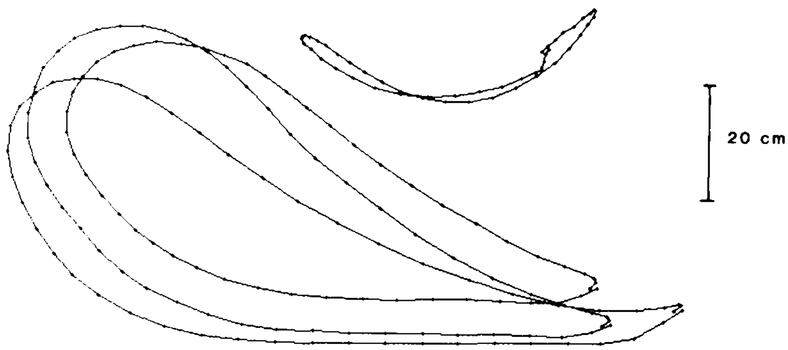
It is frequently convenient to study a subject in the clinic or laboratory running on a treadmill, although there has been considerable debate over the similarity or dissimilarity between treadmill and overground running.[47] The present author is of the belief that treadmill running is an entirely adequate model of overground running for the study of most clinically relevant conditions. The data represented in Figure 6 show the path of the same landmarks seen in Figure 5 during running on a treadmill. This figure is presented here because it resembles the initial output from several of the newer optoelectronic measurement devices that are available for clinical use.

Once the path of points on the foot and leg are known, it is a simple matter to calculate velocities of those points, and this can lead to some interesting results. For example, it is often thought that the foot is moving backward at foot strike during overground running, but the velocity vectors drawn on the foot in Figure 7 show this is to be incorrect. Another example of a useful measure that can be directly obtained from path information is the horizontal distance between the hip and ankle joints at the instant of foot strike. The lateral malleolus is typically about 20 cm in front of the vertical line drawn through the greater trochanter at foot strike during running at 4.9 meters/sec. It is also theoretically possible to calculate accelerations from film, but, because of the mathematical errors involved in the process, the preferred techniques to determine acceleration involve direct measurements. These are discussed below.

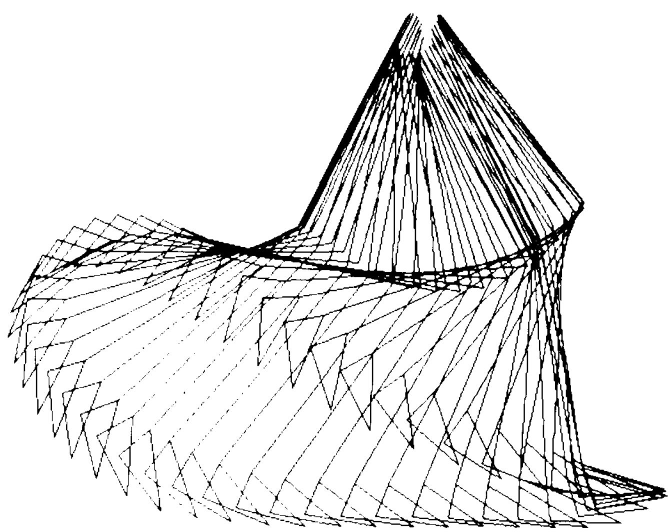
Angle-Time and Angle-Angle Plots

Although the methods of presentation used in Figures 5 and 6 contain all the relevant kinematic information, the diagrams are very difficult to interpret. Frequently they are only used at an early point in the case evaluation to ensure that the analysis is free of obvious errors and to provide clues to major abnormalities in the movement pattern. The more useful plots are usually derivations of these "raw" data in which individual or paired joint or segment angles and velocities are shown.

9



a)



b)

Fig. 6. *a*, The path of the same landmarks shown in Figure 5 during treadmill running at 4.2 meters/sec. *b*, A stick diagram of the angles observed during treadmill running at 4.2 meters/sec.

In Figure 8, the angles of the three major joints of the lower extremity, measured in a sagittal plane, are shown as functions of time according to the conventions indicated in the inset to the diagram. The data in the diagram are calculated from the coordinates shown in Figure 6. From this diagram it is possible to derive ranges of motion used at this speed of running and to follow the time course of individual joint angle changes. One can see, for example, that, in this subject, the knee never quite reaches full extension and flexes to approximately 110 degrees. The "cushioning" action of the knee after foot strike is readily seen as flexion through about 30 degrees occurs in the first 100 msec of the support phase. Ankle motion has a range of about 50 degrees and hip motion approximately 60 degrees.

Since coordination of joint action is such a vital part of all movement, the presentations called angle-angle diagrams proposed by Grieve[22] shown in Figure 9 have much to commend them. Angle-angle diagrams plot the angles at adjacent joints against each other rather than plotting angles versus time. For example, Figure 9a shows the same data plotted in Figure 8 with the hip angle on a horizontal axis plotted against the knee

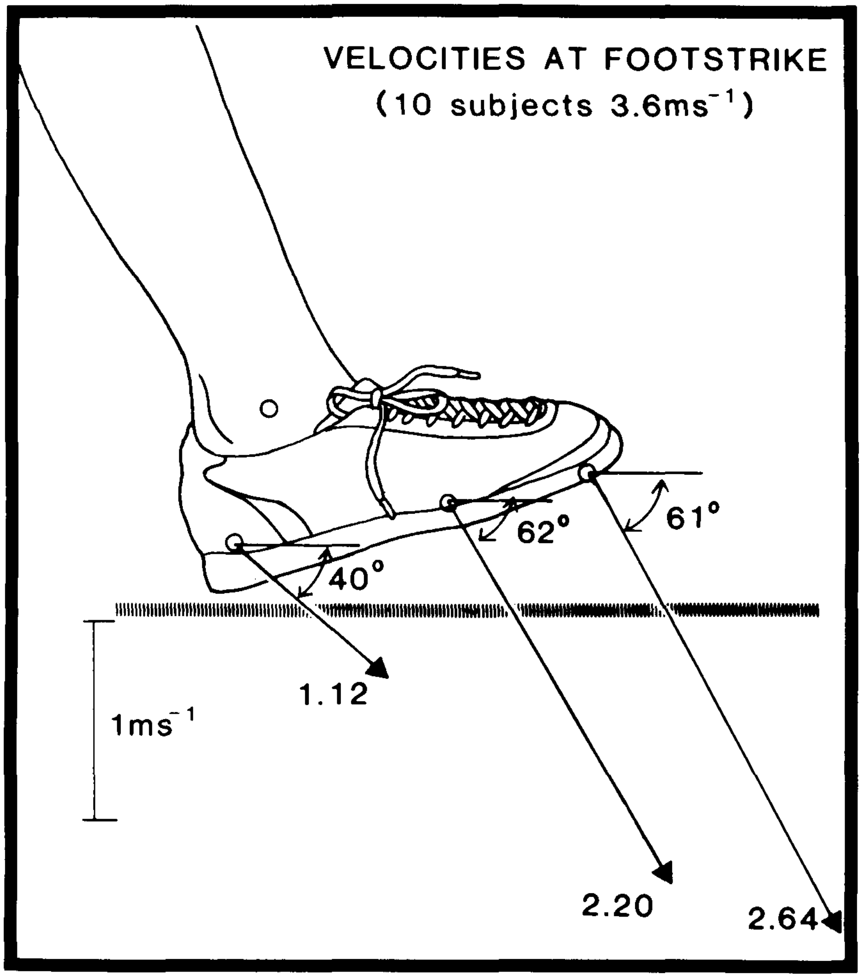


Fig. 7. The velocities of three points on the shoe with respect to the ground at the instant of foot strike during overground running at 3.6 meters/sec (mean results from 10 runners). Notice that the foot is moving forward with respect to the ground at foot strike. (Reproduced from *Sport Shoes and Playing Surfaces* with permission of Human Kinetics Publishers.)

angle on a vertical axis, with the same conventions shown in Figure 8 (hip extension is negative, and knee flexion is an increasing angle). Whenever the diagram is parallel to one of the axes, movement at only one joint is occurring. When the line migrates at 45%, then the two joints are changing at the the same rate. The element of time has, of course, been lost in the cross plot of two angles, and some adjustments have to be made to conventional graph reading procedures since equal lengths on the diagram do not represent equal times.

It is very easy to see from Figure 9a that during the cushioning phase, immediately following foot strike (a on the figure), about 25 degrees of knee flexion occur about an almost fixed hip (between a and b on the figure). This is followed by a "thrust phase" (between b and c in Figure 9a) when hip extension and knee extension occur simultaneously up to the point of toe off (c on the figure). We note that this thrust phase involves about 25 degrees of knee extension (upward migration of the diagram) and almost 60 degrees of hip extension (movement from right to left on the diagram). Hip flexion begins immediately after toe off, and the knee continues to flex until the hip is almost 25 degrees flexed (d in Figure 9a). The maximum knee flexion angle at this speed of running is about 110 degrees. As the knee begins to extend, hip flexion continues for only about another 15 degrees to reach a maximum of 42 degrees (e on the figure) and from that point onward, hip and knee extension are occurring simultaneously (between e and f on Figure 9a). The diagram also clearly shows that the knee is never fully extended and that foot strike (a on the figure) occurs as the knee has already begun to flex about a relatively fixed hip.

What is important about angle-angle diagrams is that they can be appreciated at two distinct and, if neces-

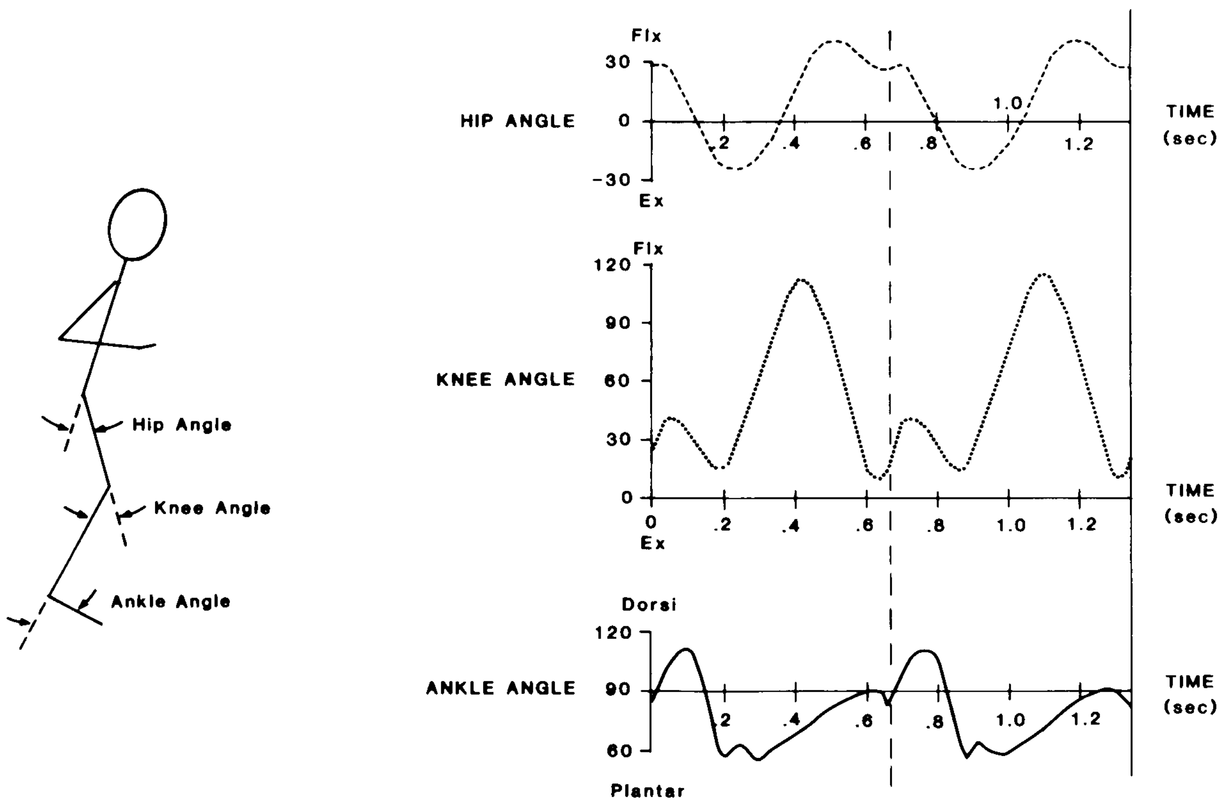


Fig. 8. Hip, knee, and ankle joint angles calculated partly from the data shown in Figure 6 and plotted as functions of time. The conventions shown for angle calculations are inset. Two complete cycles are shown.

sary, separate levels. Quantitative information, such as that discussed in the previous paragraph, can be extracted from them regarding ranges of joint motion and coordinated action. But the angle-angle diagram can be useful on the level of being simply a "shape" that we can easily appreciate, remember, and compare to other shapes. For example, when compared to Figure 9a, the pattern from another runner at exactly the same speed shown in Figure 9b has some important shape differences which are readily apparent. Despite being of similar width and height, Figure 9b has a much deeper indentation in the upper right-hand corner than Figure 9a. Also, rather than moving vertically down from foot strike, Figure 9b moves downward and to the right, back along the exact same trajectory that was followed during late support. These differences are compelling and easily seen on the basis of shape alone. From the earlier discussions, the functional implications are that the runner whose motion is depicted in Figure 9b has more knee flexion immediately following foot strike than the runner in Figure 9a, and his post foot strike cushioning pattern involves knee flexion and hip flexion rather than knee flexion about a fixed hip exhibited by the runner in Figure 9a.

Similar plots can be made for the motion of the ankle and knee joints. Ankle-knee diagrams, such as that shown in Figure 10 in which data from Figure 6 have again been used, plot the ankle angle on a horizontal axis (plantarflexion to the left) against the knee joint on a vertical axis (same conventions as for hip-knee). These diagrams take a little more determination to understand since they tend to cross more frequently, yet they are equally illuminating. The foot strikes the ground at the point marked (a) in Figure 10, and dorsiflexion immediately begins accompanied by knee flexion. The ankle is frequently thought to be plantarflexing after foot strike, but the diagram shows that in this subject, and most other runners, the plantarflexion before foot strike rapidly changes to dorsiflexion at foot strike as the knee flexes during the cushioning phase. Dorsiflexion continues after maximum knee flexion (point b on the curve), but soon the "extensor thrust" of plantarflexion, knee extension, and—as we know from the hip-knee diagram—hip extension takes place (b through c on the diagram). After toe off (c), the ankle angle varies only slightly during the first part of the swing phase but is then gradually dorsiflexed through maximum knee flexion (d), maximum hip flexion (e—

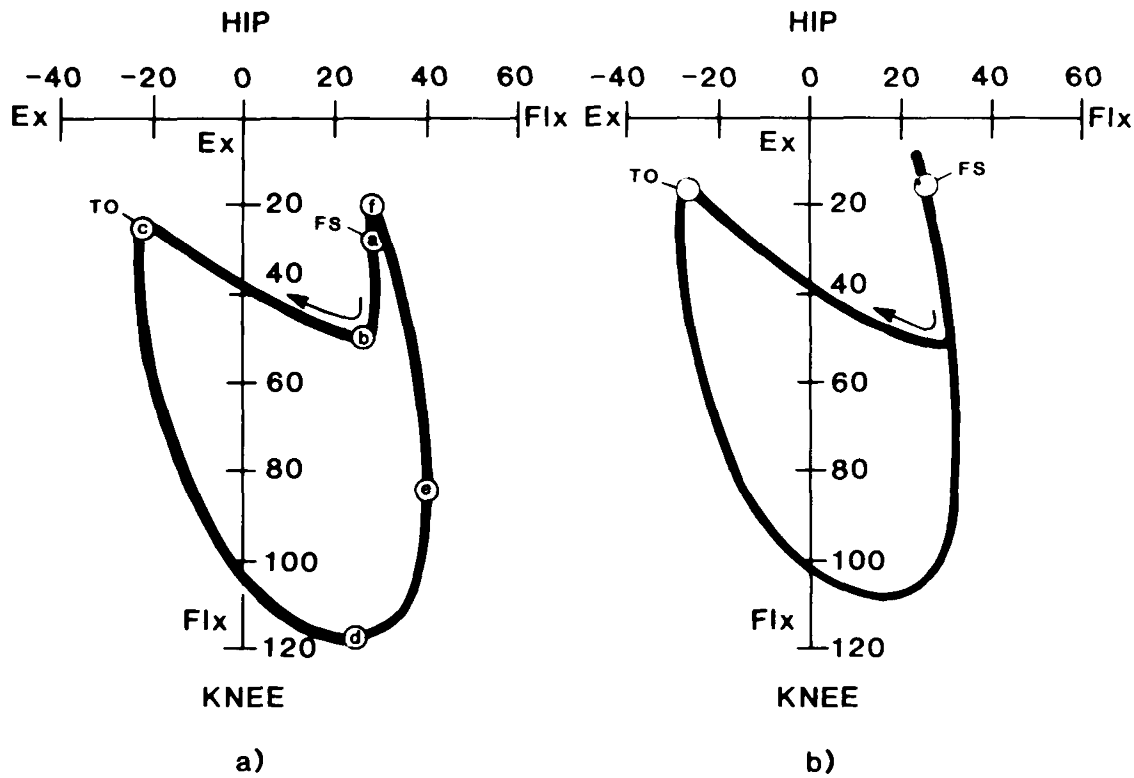


Fig. 9. *a*, A hip-knee diagram for the one cycle of running at 4.2 meters/sec that was shown in Figures 6 and 8. The foot strikes the ground at point a after which, during the "cushioning phase," the knee flexes with no movement at the hip. The "propulsive phase" (between b and c) involves simultaneous knee and hip extension. Toe off occurs at c, the point of maximum hip extension, and the swing phase (c through a) at first involves simultaneous knee flexion and hip flexion (c through d). The points of maximum knee flexion (d) and hip flexion (e) are indicated. Note that the hip joint is extending during the last 65 degrees of knee extension in swing (e through f) and also that the knee is flexing before foot strike (f through a). *b*, Hip-knee diagram for a different subject running at the same speed presented to show how a comparison of shape alone can provide interesting information. The markedly different pattern after foot strike implies the greater use of controlled knee flexion during the cushioning phase, and the use of hip flexion rather than the maintenance of a fixed hip demonstrated by the runner in Figure 9a.

positioned from a knowledge of the hip-knee curve), and only when the knee begins to extend in preparation for foot strike does plantarflexion begin (f through a).

Various types of instrumentation will allow angle-angle diagrams to be generated either immediately or within a few minutes of the run, while the patient is still in the clinic. The use of such a presentation as part of a patient's chart would be valuable in a number of respects. First, it would provide a "hard copy" description of certain aspects of the patient's movement pattern which could be used for comparison on subsequent visits. Second, it could serve as a means of communication between colleagues involved in the treatment of the same patient. Last, it would allow comparison against a database which, while not large at the present time, can be expected to develop rapidly in the future.

Rearfoot Motion

Much diagnostic interest has been centered around the motion of the rearfoot in running.[1,3,15,40] This is because of the widely held belief that excessive pronation of the subtalar joint during the support phase of running may, over the course of many thousands of foot strikes, lead to injuries of the foot, leg, and knee.[29] Unfortunately, this hypothesis is not supported by any quantitative evidence, yet one can hardly argue with the wealth of anecdotal reports that associate reduction in "rearfoot motion" using, for example, in-shoe orthotic devices, with a relief of symptoms.[45, 46]

As Inman's pivotal work has demonstrated,[28] true motion of the subtalar joint would require three-dimensional filming of the foot, and this has not yet been attempted in running. Consequently, most researchers have filmed the leg and foot in a coronal plane and used the designations pronation/supination of the subtalar joint interchangeably with eversion/inversion of the calcaneus, respectively.

The variations seen in the running population are characterized by the tracings from high speed film shown in Figure 11. Looking first at a typical "normal" pattern in Figure 11a, notice that the rearfoot is in about

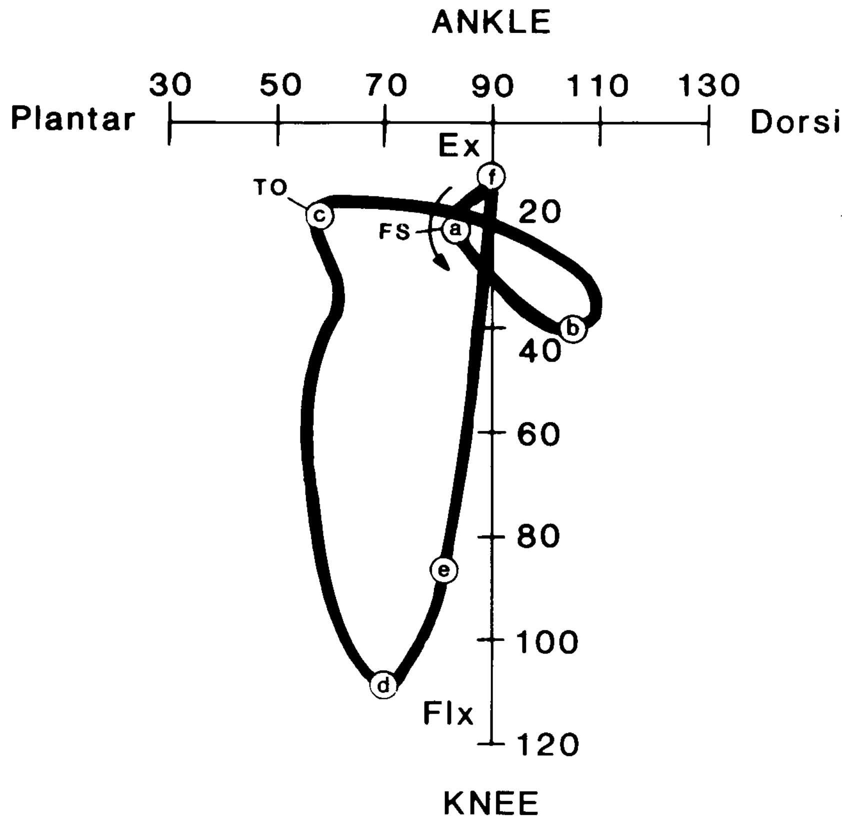


Fig. 10. Ankle-knee diagram drawn from the data in Figure 6. The foot strikes the ground at the point marked (a), and dorsiflexion immediately begins accompanied by knee flexion. Dorsiflexion continues after maximum knee flexion (point b on the curve), but soon the "extensor thrust" of plantarflexion, knee extension, and—as we know from the hip-knee diagram—hip extension, takes place (b through c on the diagram). After toe off (c), the ankle angle varies only slightly during the first part of the swing phase but is then gradually dorsiflexed through maximum knee flexion (d), maximum hip flexion (e—positioned from a knowledge of the hip-knee curve), and only when the knee begins to extend in preparation for foot strike does plantarflexion begin (f through a).

10 degrees of inversion at foot strike and then everts to a position of about 10 degrees at 30 msec. This finding confirms that some pronation is both normal and essential to the running gait. Another feature of Figure 11a—and indeed of all three patterns shown in the diagram—is that the leg remains at an almost fixed angle in space throughout the foot strike and initial pronation period that is depicted.

A classical case of "excessive" pronation is shown in Figure 11b. Here, the rearfoot continues its initial eversion and reaches an endpoint some 18 degrees beyond the neutral position (defined for this purpose as a zero degree rearfoot angle). Because of the orientation of the subtalar joint axis, this motion will result in considerable internal rotation of the entire lower extremity[28] and will presumably increase the local stresses at the knee joint. The other end of the spectrum is shown by the pattern in Figure 11c. This foot is so functionally rigid that there is hardly any rearfoot motion at all during the part of the support phase shown. In fact, this foot moves into slightly more inversion after foot strike and remains inverted throughout the contact.

In the same way that angle-angle diagrams can be generated in almost "real time," the angles shown in Figure 11 can be monitored continuously while the patient is running on the treadmill in a laboratory or clinic. Rearfoot angle-time plots for the same three runners whose tracings appear in Figure 11 are shown in Figure 12. The shaded area of the graph indicates the changes in rearfoot motion after the last tracing (30 msec) in Figure 12. The curve for the typical subject confirms that after a foot strike, with the rearfoot inverted by about 9 degrees, the foot moved to a position of about 14 degrees of eversion by 40 msec. The inversion after foot strike of the runner with a rigid foot is clearly seen in the graph, and he demonstrates a return to the initial footstrike position only after about 60 msec. The runner with excessive pronation continues to show rearfoot eversion well after the last drawing in Figure 11. His rate of rearfoot motion, shown by the gradient of the curve, is less than the normal pattern, and this is somewhat unusual for a runner who pronates this much. Some authors have suggested[15] that the maximum rate at which rearfoot motion occurs (some-

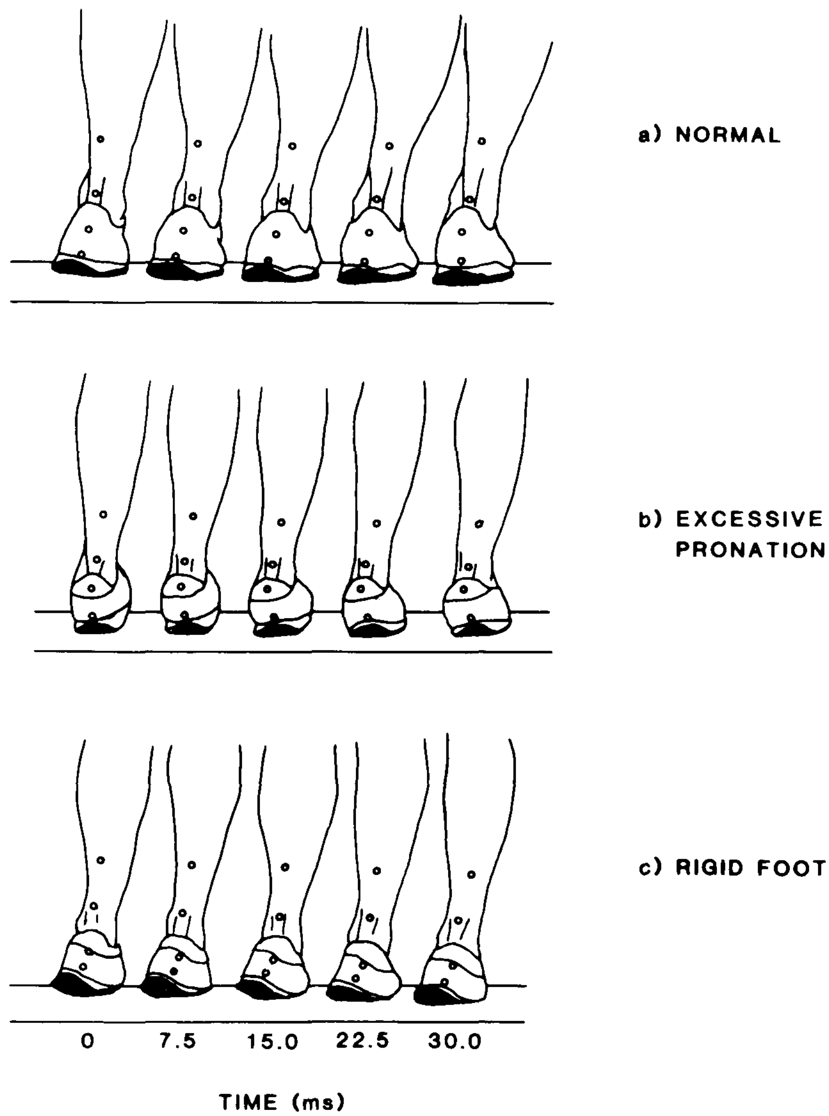


Fig. 11. Tracings of rearfoot motion from high speed film of the first 30 msec after foot strike of three runners who display: a) a typical "normal" pattern, b) excessive pronation, and c) a rigid foot in which the rearfoot remains inverted for most of the support phase. The times marked are milliseconds (one thousandths of a second) after foot strike. Notice that the leg remains at about the same orientation in space throughout this phase.

times called the "maximum velocity of pronation") may be as important in the injury process as the maximum rearfoot angle itself. It is clear that rearfoot angle graphs, such as those shown in Figure 12, could also form a valuable part of a patient's record, as the differences that were seen from the tracings in Figure 11 are accentuated by the quantitative data.

Although the pronatory movement occurs very rapidly, Figure 11 demonstrates that, after about 60 msec, the endpoint of the motion is held for an appreciable period of time during mid-support. This means that it is possible to obtain a first order estimate of the amount of maximum pronation without instrumentation or at least with a simple video system that has a good stop frame capability. Many sports medicine offices have such a facility for this purpose.

The hope for quantitative evaluation of rearfoot motion is that it will provide a more rational basis for the prescription and evaluation of in-shoe orthotic devices. Initial results of experiments using these techniques have shown that the response to varus or valgus correction in the shoe may be more complex than might, at first, be imagined.[3,44] Patients modify their gaits in different ways in response to a change in their dynamic balance. It is fair to state that the authoritative study on changes in foot action as a result of in-shoe intervention remains to be conducted.

Motion at Other Joints of the Foot

In closing this section on motion analysis, it must be pointed out that there are large gaps in our knowledge of the motion of the foot during running. For example,

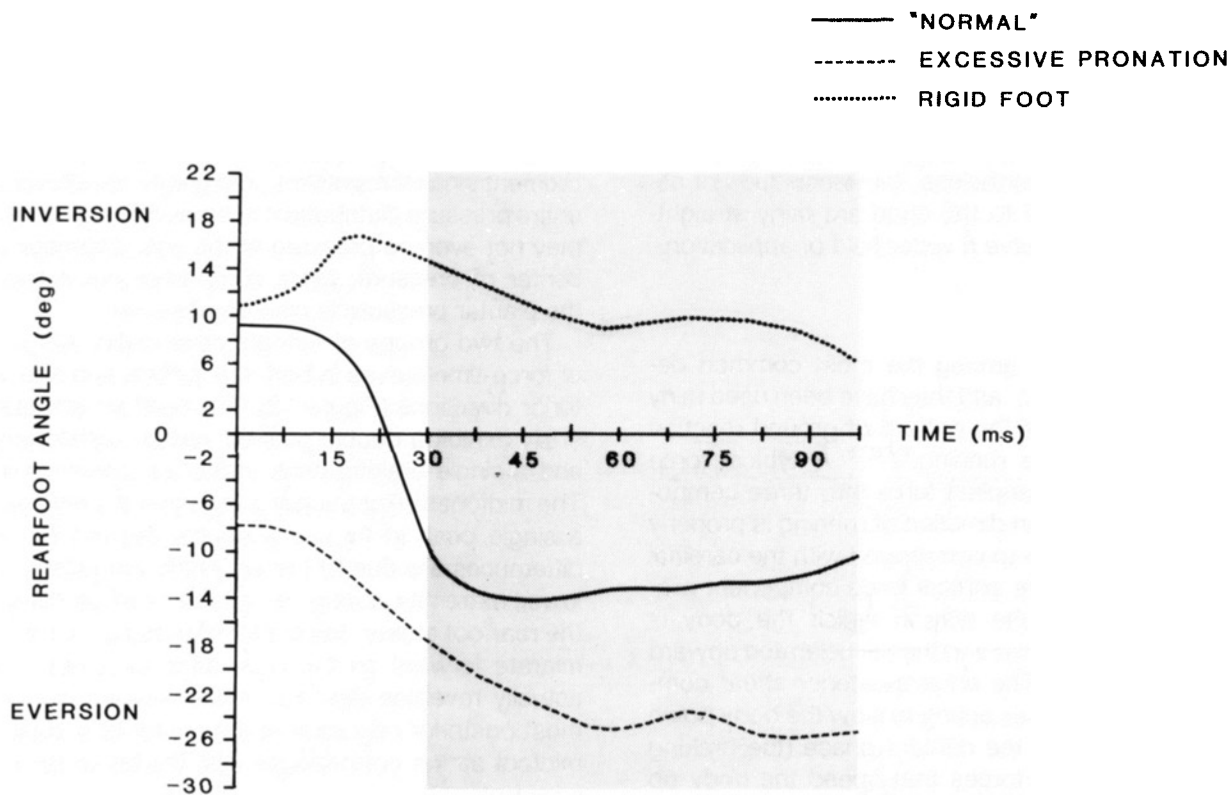


Fig. 12. Rearfoot angle-time graphs for the three runners depicted in Figure 11. The graphs extend to about 100 msec after foot strike. Notice that, although the initial movements following foot strike occur rapidly, the endpoint of the motion is held for a longer period of time. This allows maximum rearfoot angle to be measured with normal video equipment that has single frame mode. However, the curves themselves could form a valuable addition to the patient's record.

there are no published reports of changes in metatarsophalangeal joint angles, and little is known concerning the action of the midtarsal joint in running. The relative motion between foot and shoe is also a topic that deserves further study.

ACCELERATION, FORCE, AND PRESSURE

Thus far, we have considered only those phenomena that are visible—either to the eye, or to instruments designed to improve the temporal resolution of the eye. The physical quantities that are the topics of this section cannot be seen by any means. They can, however, be measured and displayed by a variety of sophisticated instruments.

Acceleration

An accelerometer can be thought of as a device that measures the shock transmitted to an object.[16] Such a measurement is simple enough when the device, which can weigh as little as 2 grams, can be rigidly attached to a tool or to an automobile. But attachment to the human body is problematical, since direct fixation to bone is usually possible only in limited experimental situations.[32] The clinician usually has to be satisfied with an attachment on the surface of the skin, where the soft tissue acts as a mechanical filter of unknown, and possibly variable, properties.[50] The validity of such measurements is still under debate in the scientific community, but accelerometric data do appear to have merit for making comparative statements between two conditions in the same individual. For example, in an experiment relevant to earlier discussions on altered stride length, Clarke et al.[13] have shown that running with shorter than freely chosen stride lengths reduced leg shock, whereas longer strides increased leg shock.

Another possible location for an accelerometer is on the shoe; this has provided some interesting information relative to foot strike.[23,24] During running at 3.8 meters/sec, the acceleration of the shoe can be in excess of 50 g, although typical values are about 25 g. There is a very large variation between individuals, as anyone who has run with different running partners can appreciate. Some runners seem subjectively to come crashing in to foot strike, while others strike the ground very gently. Measured values as low as 10 g have been recorded from the shoe at foot strike. Accelerometer measurements may in the future turn out to have some diagnostic value. If the feet of two individuals, or the right and left feet of the same runner, are experiencing markedly different accelerations, it is tempting to believe that the extremity with the higher accelerations would be more at risk for injury. Such speculation remains to be proven because part of the problem with accelerometry is that we have no clear idea of injury thresholds, and these are certain to be different between individuals. Nevertheless, the techniques of accelerometry as applied to the shoe are fairly straightforward, and they deserve a wider field of application.

Force Measurements

Force platforms are among the most common devices in gait laboratories, and they have been used fairly extensively to describe the pattern of ground reaction forces during distance running.[2,7,9,18] A typical force platform resolves the applied force into three components, which if the mean direction of running is properly arranged, can be made to correspond with the cardinal planes of the body. The vertical force component provides information on the way in which the body is cushioned after foot strike and then accelerated upward into the flight phase. The anteroposterior shear component reflects the forces acting to slow the body down during the first half of the contact phase (the braking force) and then those forces that speed the body up again before takeoff. The mediolateral shear forces indicate how the center of mass of the body is transferred from side to side during ground contact.

Unlike an accelerometer attached to the shoe, which measures only what is experienced by the foot, the force platform measures the net force applied to the total body center of mass, and also allows us to calculate the acceleration of the center of mass. This fact is easy to forget since the foot is the part of the body that contacts the platform. The contrast between accelerometry and force platform measurements can best be seen by the fact that the force platform tells us that the peak acceleration experienced by the total body center of mass during running at 3.8 meters/sec is approximately 3 g, while, as we saw earlier, the acceleration of the foot was typically 25 g.

Mean force-time curves for all three force components from two groups of subjects running at 4.5 meters/sec are shown in Figure 13. The groups were distinguished by their first point of contact between the foot and ground and designated "rearfoot strikers" and "midfoot strikers."[9] This designation will be clearer by viewing the center of pressure patterns for the same runners, also collected from a force platform, shown in Figure 14. The rearfoot strikers were defined as those runners who made initial contact with the ground in the posterior third of the shoe. The initial contact for the midfoot strikers was in the middle third of the shoe. It is interesting to note from the center of pressure plots that initial contact between shoe and ground in all runners was on the lateral border of the shoe. We have rarely seen a subject who makes initial contact with the posterior border of the shoe.

The center of pressure is a somewhat global measurement since it represents the geometric center of the entire pressure distribution. In some limiting cases there may not even be pressure at the actual location of the center of pressure. More comprehensive mapping of the plantar pressure is discussed below.

The two groups of runners have distinctive patterns of force-time curves in both the vertical and anteroposterior directions (Figure 13). The rearfoot strikers generally exhibit a double peaked vertical component (*Fz*) and a single braking peak in the a-p component (*Fy*). The midfoot striker usually shows just the reverse, with a single peak in *Fz* and a double peak in *Fy*. These differences are due to the very different actions of the lower extremity during the phase of initial contact. In the rearfoot striker, the center of pressure continues to migrate forward on the shoe after footstrike, while it actually reverses direction in the midfoot striker. The most posterior migration of the center of pressure in a midfoot striker corresponds with the fall to zero of the a-p force.

These different strike patterns have considerable implications for the use of the musculature of the leg in the process of cushioning. In a rearfoot striker, the large initial spike in the vertical force occurs as the heel is hitting the ground, a peak that Nigg[41] has referred to as the "impact peak." This period is so brief (the force peaks in about 20 msec) that any muscular activity associated with it must have been initiated before foot strike. In the midfoot striker, this impact peak is absent or considerably reduced, suggesting that much of the impact is indeed being cushioned by the posterior musculature of the leg, a fact which electromyographic studies should confirm.

The peak later in support, called the active peak by Nigg,[41] is similar in both groups of runners, and it is also generally the larger peak at slow to medium speeds of running. This latter fact is somewhat of a surprise, because by this time the center of pressure is underneath the ball of the foot. Intuition would suggest that the impact peak would be greater, but this is not the case.[9] It has been suggested[30] that approximately 80% of the population are rearfoot strikers at distance running speeds, but there is no clear idea at present of which pattern is more economical from an energy standpoint or more beneficial from an injury prevention point of view.

It is not uncommon to see differences between the left and right foot of the same individual, both in the

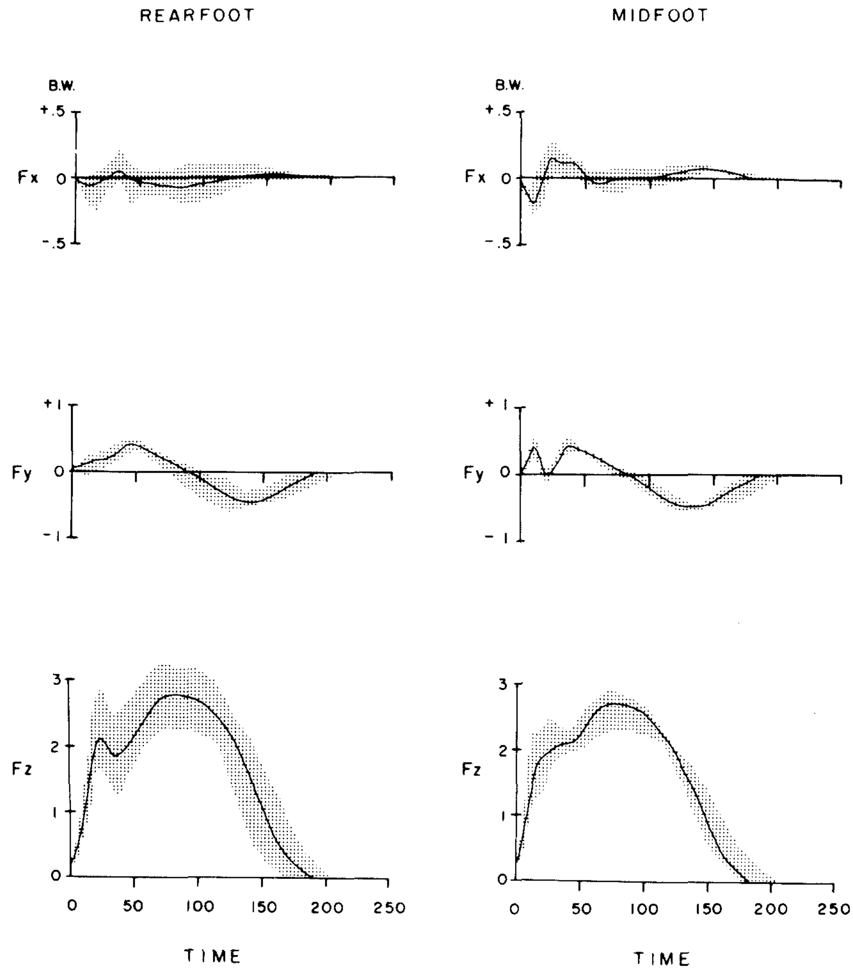
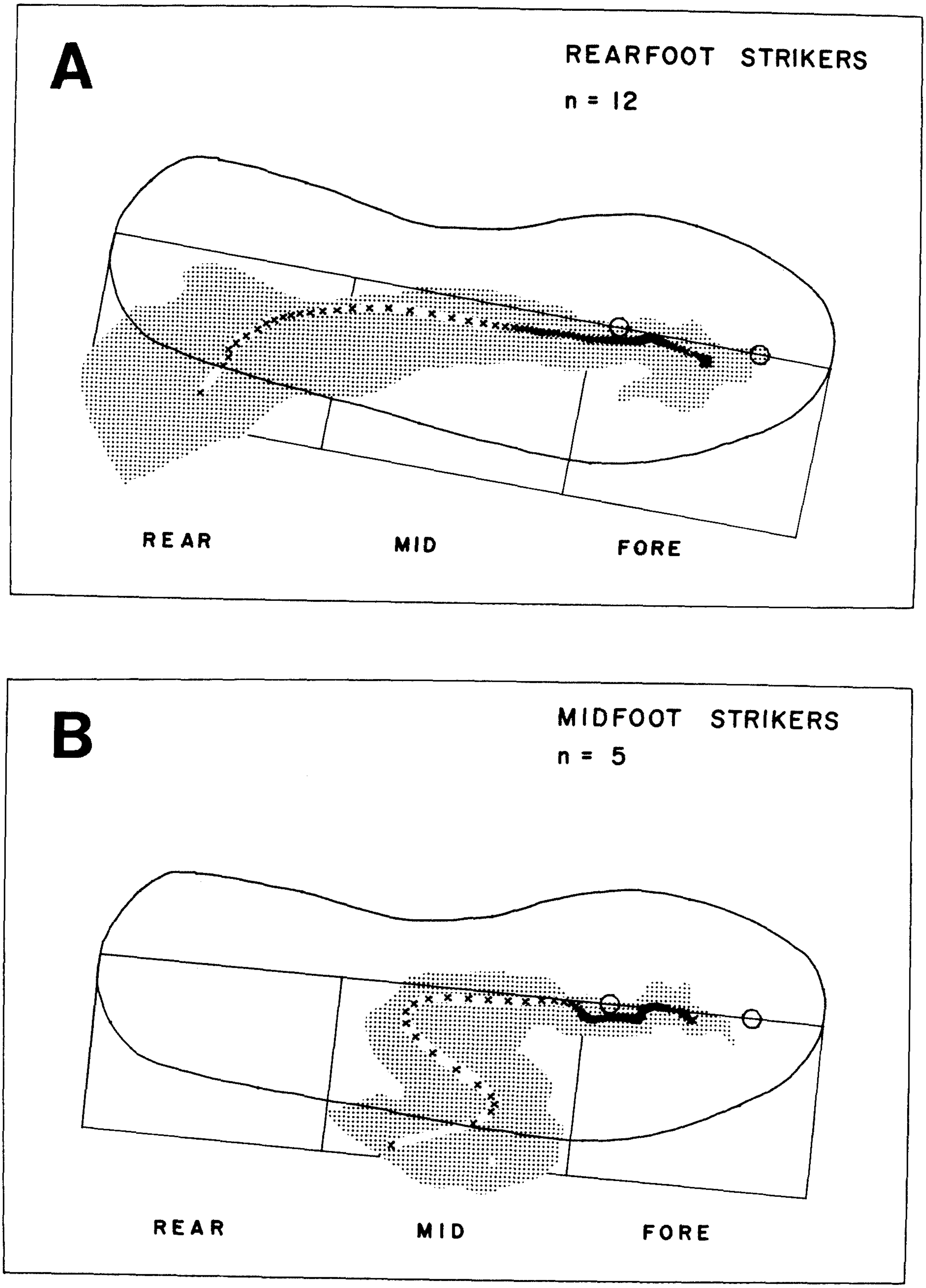


Fig. 13. Mean force-time curves from rearfoot strikers (N = 12) and midfoot strikers (N = 5) running at 4.5 meters/sec. All force values are expressed in terms of body weight and time is in milliseconds after foot strike. The rearfoot striker generally exhibits a double peaked vertical component (*Fz*) and a single braking peak in the a-p component (*Fy*). The midfoot striker usually shows just the reverse, with a single peak *Fz* and a double peaked *Fy*. There is a great deal of variability in the mediolateral curves in both groups. The shaded area shows the range encountered in each group. (Reproduced from *Journal of Biomechanics* with permission of Pergamon Press.[9])

center of pressure patterns and in the force time patterns. The most extreme differences that we have encountered are the Fz versus time curves for an elite 10,000 meter runner[7] shown in Figure 15. These data collected at a race pace (about 6 meters/sec) revealed that the athlete sustained a force of over 1.5 body weight greater in the right leg during the impact phase. This startling difference appeared to be due to the different action of the limbs, with the right leg acting as a stiff "strut" at foot strike, while the left leg was more of a "shock absorber." Although there was apparently no greater incidence of injury on the right side, such a finding must be taken into account when assessing the injury status of a runner.

Force platforms are fairly expensive instruments, but they are very straightforward to use and provide an important quantitative picture of the events during ground contact. It is generally advisable to collect data from many transits of the platform and average them together to provide a mean response. The same comment could well be made of all the measurements discussed thus far because the variability of the running process is something that is just beginning to be explored.[6] Presently available force platforms are fairly small (40 × 60 cm), and care must be taken that the subjects do not "target" for the platform by rapid gait alterations immediately before contact. Footwear will also affect the patterns recorded,[2,14,40] although not as drastically as one might have imagined. It is nevertheless important to ensure that patients use the same footwear on repeated visits when force platform measurements will be made.


A
REARFOOT STRIKERS
n = 12
REAR
MID
FORE
B
MIDFOOT STRIKERS
n = 5
REAR
MID
FORE

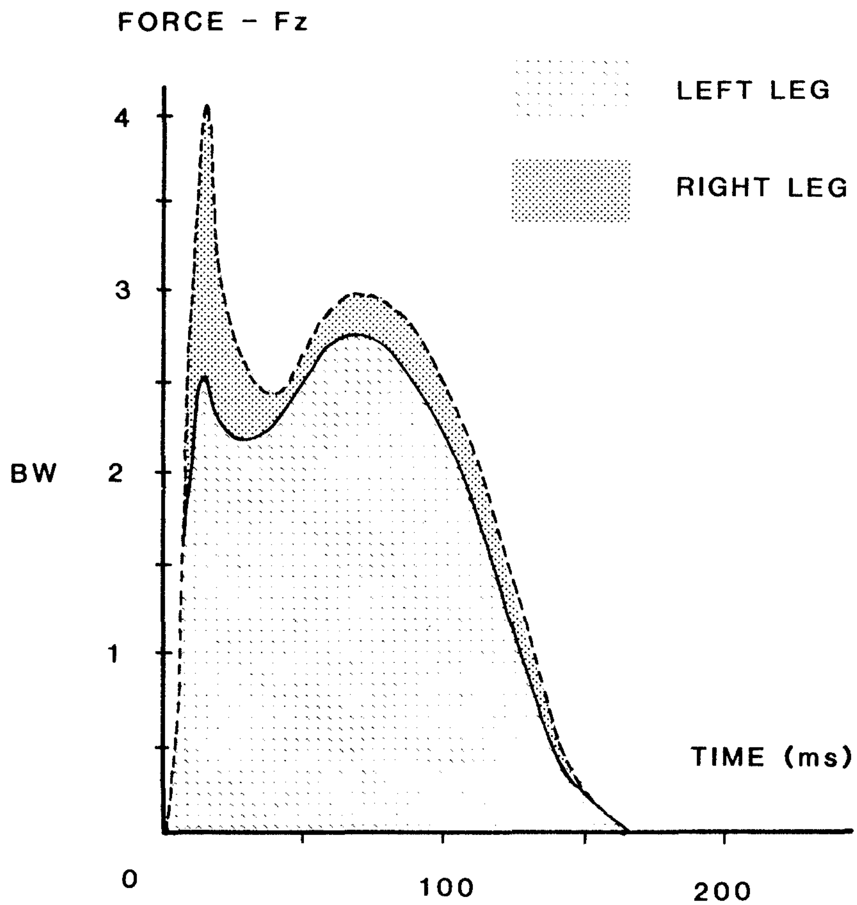


Fig. 15. The vertical component of ground reaction force (*Fz*) under the left and right feet of a champion 10,000-meter run athlete running at 6 meters/sec. The forces during the impact phase on the right side are 1.5 times body weight higher than those on the left side. Although this is an extreme example, asymmetries are not an uncommon finding in both elite and non-elite athletes.

Pressure Distribution

A complete picture of the interaction between the foot and the ground can only be obtained by a study of the forces acting on the various anatomical structures individually, not just a knowledge of the net force such as given by a force platform. The data presented in Figure 16 represent initial measurements made in our own laboratory of a subject running barefoot at 3.8 meters/sec across a 1,000 element pressure distribution platform.[8] Each sensor in the array is 0.5 × 0.5 cm, and associated computer hardware and software enable contour maps, such as that shown in Figure 16, to be generated and animated into films and videotapes. The contour interval in the diagram is 100 kPa, and peak pressures in excess of 1 mPa (145 psi) can be identified under both heel and metatarsal heads.

Such presentations provide entirely new information about the action of the foot. After an initial focus of pressure develops with its center on the lateral border of the heel (Fig. 16a), the pressure moves rapidly to the medial border of the heel, and load bearing under the heads of the lateral metatarsals begins (Fig. 16b). The subsequent decay of pressure under the heel and the development of two separate peaks of approximately equal magnitude under the first and second metatarsal heads are clearly seen (Fig. 10c through f). In this subject, none of the lesser toes are substantially involved in the weight bearing process, and the slight medial transfer of load before toe off is apparent.

The data shown in Figure 16 cannot be generalized to running in shoes for a number of reasons. Runners deprived of their shoes frequently make drastic alterations in running style. The clear heel strike pattern

Fig. 14. Center of pressure patterns collected from the right side using a force platform for the same runners whose force-time curves are shown in Figure 13. The rearfoot strikers (*A*) all made initial contact with the ground in the posterior third of the shoe while the initial contact for the midfoot strikers (*B*) was in the middle third of the shoe. Notice that in all runners, the initial contact was on the outside border of the shoe. The shaded area shows the range encountered in each group. (Reproduced from *Journal of Biomechanics* with permission of Pergamon Press.[9])

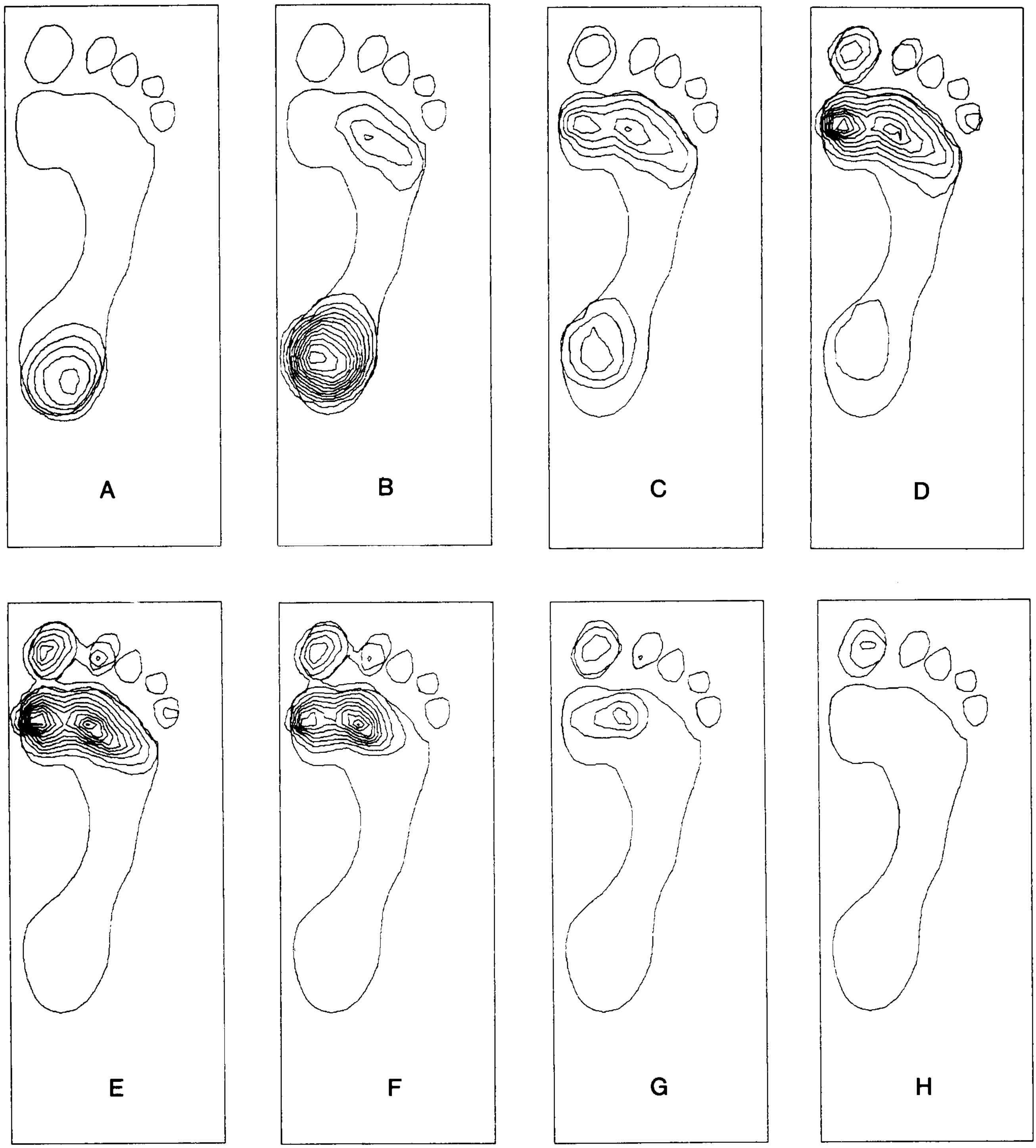


Fig. 16. The plantar pressure distribution at 8 instants of time during the contact phase of a single subject running barefoot at 3.8 meters/sec. The contour interval in the diagram is 100 kPa. Peak pressures in excess of 1 mPa (145 psi) can be identified under both heel and metatarsal heads. After an initial focus of pressure develops exclusively on the lateral border of the heel (A), the pressure moves rapidly to the medial border of the heel, and load bearing under the heads of the lateral metatarsals begins (B). The subsequent decay of pressure under the heel and two separate peaks of approximately equal magnitude under the first and second metatarsal heads develop (C through F). In this subject, none of the lesser toes are substantially involved in the weightbearing process, and the slight lateral transfer of load before toe off is apparent. The pressure contours appear medial to the foot outline because some foot motion occurs during support which cannot be represented on the diagram by a single footprint.

shown in Figure 16, a result of specific instructions from the experimenter, is unusual in a barefoot runner. Also, the pressure distribution inside the running shoe is likely to show lower peaks than a contact between the bare foot and a flat rigid surface. It is important to realize that the shoe cannot reduce the forces involved in the contact with the ground: if the external mechanics of the motion of the body's mass center remain the same, the forces will remain the same. The shoe can, however, reduce the pressure acting on the various bony structures in the foot by transferring force to other regions. The resulting pressure will therefore be less.

There are a number of systems for the measurement of pressure distribution between the foot and the floor, or between the shoe and the floor. Progress has, however, been slower in the development of a suitable in-shoe device that will accurately measure those pressures that are actually experienced by the foot during running.[25] Until such devices are available, the study of the interaction of the bare foot with the ground should provide a valuable addition to the profile of foot-ground interaction. This is particularly true during the latter part of the support phase in the forefoot region in which patterns are likely to be least affected by the style changes which accompany barefoot running discussed earlier.

JOINT MOMENTS AND JOINT FORCES

By combining the movement data, such as that shown in Figure 6, with information from a force platform, such as that in Figures 13 and 14, with anthropometric data on the subject, it is possible to calculate the net moments or torques that act at the joints of the leg and foot.[49] Unfortunately, the calculations cannot indicate the contributions either of individual muscles or even of the flexors or extensor groups. The values calculated are just the net result of all muscle action. Such procedures can, however, lead to the calculation of power flow across joints,[39,49] and these techniques have proved useful to an understanding of the overall mechanics of the running process. Their usefulness in understanding the mechanics of injury in running remains to be demonstrated.

The calculation of forces at the articular surfaces of the various joints of the leg and foot, usually called joint forces or bone on bone forces, is a complex and much debated topic in biomechanics. Such calculations always involve a mathematical model developed with the aid of many assumptions and simplifications which cannot always be justified given the complexity of the anatomy and its pattern of activation. At the present time, estimates of joint forces are therefore subject to wide error bounds. Burdett[5] has conservatively estimated the force in the talocrural joint during running at 4.5 meters/sec to be between 9.0 and 13.3 times body weight. The difference between the 3 times body weight measured at the feet and these surprisingly large articular forces is due to the action of muscles crossing the joint. If joint force estimates are to be useful in the future they will have to be combined with quantitative individual anatomy from a scanning procedure, and they must rely heavily on an improved understanding of the quantitative use of electromyographic data. Such advances are not likely to be quickly made.

CONCLUDING REMARKS

As mentioned at the outset, this article has attempted to describe a range of biomechanical measures as they are found in the foot and leg of the typical asymptomatic distance runner. As the clinician anxious to apply these methods will be painfully aware, there are at present neither extensive databases giving confidence limits for large populations of asymptomatic individuals, or clear indications of the patterns that may be expected from specific pathologies. It is also clear that many of the mechanical changes that represent the difference between running "in pain" and running "pain free" may be at a level of subtlety that we have not yet explored.

It would be wrong to advocate measurement for measurement's sake, and that is not what is being suggested here. The reason for the lack of databases is that no one has taken the time to gather them. The scientists have principally worked on problems of academic interest with little clinical relevance, while the clinicians have been slow to embrace quantitative techniques in their routine evaluation of the running patient. There is, however, a change in the air. Both groups are realizing that the most successful approach to the injured runner is a team approach in which the quantitative skills of the biomechanist can complement clinical judgments. It is this author's firm belief that, in some few years time, it will be possible to write a fairly extensive review of the literature pertaining to the quantitative biomechanical analysis of running injuries.

ACKNOWLEDGMENTS

Bob Snyder, Frank Buczek, Randall Dick, Dave Sanderson, and Monica Milliron assisted with the preparation of material for figures contained in this article. Dave Sanderson also made valuable comments on a draft of this manuscript. The excellent artistic and photographic assistance of Brenda Palmgren is gratefully acknowledged.

Most of the author's work in the area of the biomechanics of running is supported by PUMA A.G. and the United States Olympic Committee. The generosity of both of these organizations is greatly appreciated.

REFERENCES

1. **Bates, B.T., Osternig, L.R., and Mason, B.R.:** Lower extremity function during the support phase of running. *In* Biomechanics VI-B. Asmussen, E., and Jorgensen, K. (eds.), Baltimore, University Park Press, 1978, pp. 30–39.
2. **Bates, B.T., James, S.L., Osternig, L.R., and Sawhill, J.A.:** Effects of running shoes on ground reaction forces. *In* Biomechanics VII-B. Moreki, A., Fidelus, K., Kedzior, K., and Wit, A. (eds.), Baltimore, University Park Press, 1981, pp. 226–233.
3. **Bates, B.T., Osternig, L.R., Mason, B., and James, S.L.:** Foot orthotic devices to modify selected aspects of lower extremity mechanics. Am. J. of Sports Med., **7**:388–342, 1979.
4. **Brody, D.M.:** Running injuries. Clin. Symp., **32**:2–36, 1980.
5. **Burdett, R.G.:** Forces predicted at the ankle during running. Med. Sci. Sports Exercise, **14**:308–316, 1982.
6. **Cavanagh, P.R.:** Studies of the locomotor mechanics of elite distance runners: present status and future directions. Proceedings of The Gatorade Symposium on Future Directions in Exercise/Sport Research, Champaign, IL, Human Kinetics Publishers, in press.
7. **Cavanagh, P.R., Andrew, G.C., Kram, R.K., Rodgers, M.M., Sanderson, D.J., and Hennig, E.M.:** An approach to biomechanical profiling of elite distance runners. Int. J. Sport Biomech. **1**:36–62, 1985.
8. **Cavanagh, P.R., Hennig, E.M., Rodgers, M.M., and Sanderson, D.J.:** The measurement of pressure distribution on the plantar surface of diabetic feet. *In* Biomechanical Measurement in Orthopaedic Practice. Whittle, M., and Harris, D. (eds.), Oxford, Clarendon Press, 1985, pp. 159–168.
9. **Cavanagh, P.R., and Lafortune, M.A.:** Ground reaction forces in distance running. J. Biomech., **15**:397–406, 1980.
10. **Cavanagh, P.R., and Williams, K.W.:** The effect of stride length variation on oxygen uptake during distance running. Med. Sci. Sports Exercise, **14**:30–35, 1982.
11. **Chodera, J.D., and Levell, R.W.:** Footprint patterns during walking. *In* Perspectives in Biomedical Engineering. Kenedi, R.M. (ed)., Baltimore, University Park Press, 1973, pp. 81–90.
12. **Clancy, W.G.:** Runners injuries. Part II. Evaluation and treatment of specific injuries. Am. J. Sports Med., **8**:287–289, 1980.
13. **Clarke, T.E., Cooper, L.B., Hamill, C.L., and Clarke, D.E.:** The effect of varied stride rate upon shank deceleration in running. J. Sport Sci., **3**:41–49, 1985.
14. **Clarke, T.E., Frederick, E.C., and Cooper, L.B.:** The effects of shoe cushioning upon ground reaction forces in running. Int. J. Sports Med., **4**:376–381, 1983.
15. **Clarke, T.E., Frederick, E.C., and Hammill, C.L.:** The study of rearfoot movement in running. *In* Sport Shoes and Playing Surfaces. Frederick, E.C. (ed.), Champaign, IL, Human Kinetics Publishers, 1984, pp. 166–189.
16. **Cobbold, R.S.C.:** Transducers for Biomedical Measurements: Principles and Applications. New York, John Wiley & Sons, 1974.
17. **D'Ambrosia, R., and Drez, D. (eds.):** Prevention and treatment of running injuries. Thorofare, NJ, Charles B. Slack, 1982.
18. **Dickinson, J.A., Cook, S.D., and Leinhardt, T.M.:** The measurement of shock waves following heel strike while running. J. Biomech., **18**:415–422, 1985.
19. **Dietz, V., Schmidtbleicher, D., and Noth, J.:** Neuronal mechanisms of human locomotion. J. Neurophysiol., **42**:1212–1222, 1979.
20. **Elliot, B.C., and Blanksby, B.A.:** A cinematographic analysis of overground and treadmill running by males and females. Med. Sci. Sports, **8**:84–87, 1976.
21. **Elliot, B.C., and Blanksby, B.A.:** The synchronization of muscle activity and body segment during a running cycle. Med. Sci. Sports Exercise, **11**:322–327, 1979.
22. **Grieve, D.W.:** The assessment of gait. Physiotherapy (UK), 452–460, 1969.
23. **Gross, T.S.:** The acceleration of the shoe at foot strike during running. Proceedings of the ACSM, Mid-Atlantic Chapter, 1985.
24. **Gunter, R.:** Uber stosserschuterungen beim Gang des Menschen. Int. Z. Angew. Phsyiol., **25**:130–141, 1968.
25. **Hennig, E.M., Cavanagh, P.R., Macmillan, N.H., and Albert, H.T.:** A piezoelectric method of measuring the vertical contact stress beneath the human foot. J. Biomed. Eng., **4**:213–222, 1982.
26. **Hogberg, P.:** How do stride length and stride frequency influence the energy-output during running? Arbeitsphysiol., **14**:437–441, 1952.
27. **Holden, J.P., Cavanagh, P.R., Williams, K.R., and Bednarski, K.N.:** Foot angles during walking and running. *In* Biomechanics IX-B. Winter, D., Norman, R., Wells, R., Hayes, K., and Patla, A. (eds.), Baltimore, University Park Press, 1985, pp. 451–456.
28. **Inman, V.T.:** The Joints of the Ankle. Baltimore, Williams & Wilkins, 1976.
29. **James, S.L., Bates, B.T., and Osternig, L.R.:** Injuries to runners. Am. J. Sports Med., **6**:40–50, 1978.
30. **Kerr, B.A., Beauchamp, L., Fisher, V., and Neil, R.:** Footstrike patterns in distance running. *In* Biomechanical Aspects of Sport Shoes and Playing Surfaces. Nigg, B.M., and Kerr, B.A. (eds.), Calgary, Canada, University Printing, 1983, pp. 135–142.
31. **Luhtanen, P., and Komi, P.:** Force-, power-, and elasticity-velocity relationships in walking, running and jumping. Eur. J. Appl. Physiol., **44**:279–289, 1980.
32. **Maclellan, G.E.:** Skeletal heel strike transients, measurement, implications and modifications by footwear. *In* Sport Shoes and Playing Surfaces. Frederick, E.C. (ed.), Champaign, IL, Human Kinetics Publishers, Inc., 1984, pp. 76–86.
33. **Mann, R.A.:** Biomechanics of walking, running and sprinting. Am. J. Sports Med., **8**:345–350, 1980.
34. **Mann, R.A.: Biomechanics of running.** *In* The Foot and Leg in Running Sports. Mack, R.P. (ed.), St. Louis, C.V. Mosby, 1982, pp. 1–29.
35. **Mann, R.A., and Hagy, J.L.:** The function of the toes in walking, jogging, and running. Clin. Orthop., **142**:24–29, 1979.
36. **Mann, R.A., Baxter, D.E., and Lutter, L.D.:** Running symposium. Foot Ankle, **1**:190–224, 1981.
37. **Marey, E.J.:** Movement. New York, D. Appleton and Co. 1895.
38. **Morton, D.J.:** The angle of gait: a study based upon examination of feet of Central African natives. J. Bone Joint Surg., **14**:741–745, 1932.
39. **Morton, D.J.:** The Human Foot: Its Evolution, Physiology, and Functional Disorders. New York, Columbia University Press, 1935.
40. **Nigg, B.M., Eberle, G., Frey, D., Luethi, S., Segesser, B., and Weber, B.:** Gait analysis and sport-shoe construction. *In* Biomechanics VI-A. Asmussen, E., and Jorgensen, K. (eds.), Baltimore, University Park Press, 1978, pp. 303–309.
41. **Nigg, B.M.:** Biomechanics of Running Shoes. Champaign, IL, Human Kinetics Publishers, Inc., 1986.
42. **Nilsson, J., Thorstensson, A., and Halbertsma, J.:** Changes in leg movements and muscle activity with speed of locomotion and mode of progression in humans. Acta Physiol. Scand., **123**:457–475, 1985.
43. **Saleh, M., and Murdoch, G.:** In defense of gait analysis: observation and measurement in gait assessment. J. Bone Joint Surg., **67B**:237–241, 1985.

44. **Smith, L., Clarke, T., Hammill, C., and Santopietro, F.:** The effects of soft and semi-rigid orthoses upon rearfoot movement in running. Med. Sci. Sports Exercise, **15:**171, 1983.
45. **Subotnick, S.I.:** A biomechanical approach to running injuries. *In* The Marathon: Physiological, Medical, Epidemiological, and Psychological Studies. Milvy, P. (ed.), New York, New York Academy of Sciences, 1977, pp. 328–345.
46. **Subotnick, S.I.:** Cures for common running injuries. Mountain View, CA, World Publications, Inc., 1979.
47. **Van Ingen Schenau, G.J.:** Some fundamental aspects of the biomechanics of overground versus treadmill locomotion. Med. Sci. Sports Exercise, **12:**257–261, 1980.
48. **Williams, K.R.:** Biomechanics of running. *In* Exercise and Sport Science Reviews. Terjung, R.L. (ed.), New York, Macmillan Publishing, 1985, pp. 389–442.
49. **Winter, D.A.:** Moments of force and mechanical power jogging. J. Biomech., **16:**91–97, 1983.
50. **Ziegart, J.C., and Lewis, J.L.:** The effect of soft tissue on measurements of vibrational bone motion by skin-mounted accelerometers. J. Biomech. Engineering, **218:**1979.

Announcement

The American Orthopaedic Foot and Ankle Society Summer Meeting, July 17–19, 1987, Santa Fe, New Mexico. Inquiries: 222 S. Prospect Ave., Suite 127, Park Ridge, IL 60068.