Review Paper

The biomechanics of running

Tom F. Novacheck

Motion Analysis Laboratory, Gillette Children's Specialty Healthcare, University of Minnesota, 200 E. University Ave., St. Paul, MN 55101, USA

Abstract

This review article summarizes the current literature regarding the analysis of running gait. It is compared to walking and sprinting. The current state of knowledge is presented as it fits in the context of the history of analysis of movement. The characteristics of the gait cycle and its relationship to potential and kinetic energy interactions are reviewed. The timing of electromyographic activity is provided. Kinematic and kinetic data (including center of pressure measurements, raw force plate data, joint moments, and joint powers) and the impact of changes in velocity on these findings is presented. The status of shoewear literature, alterations in movement strategies, the role of biarticular muscles, and the springlike function of tendons are addressed. This type of information can provide insight into injury mechanisms and training strategies. © 1998 Elsevier Science B.V.

Keywords: Running; Biomechanics; Kinematics; Kinetics; Electromyography; Energy; Injury

1. Introduction/history

To avoid the misconception that the analysis of running is a new area of interest, one need only examine the art of Grecian vases and consider the writings of Aristotle, 'Further, the forces of that which causes movement and of that which remains still must be made equal... For just as the pusher pushes, so the pusher is pushed—i.e. with similar force' [1]. Leonardo da Vinci's interest in accuracy in painting in the 15th and 16th centuries increased focus on human movement and was followed by Newton's proclamation of his three laws in the 17th century. In 1836, the Weber brothers (Wilhelm and Eduard) set the agenda for future research with the most detailed treatise on walking and running gait to date. They listed 150 hypotheses including that the limb can act as a pendulum. More sophisticated tools were needed than were currently available to test them. Etienne Jules Marey (1830–1904) was a prolific pioneer of instrumentation. He was among the first to employ photog-

raphy and use it as a true photogrammetric tool. He also designed and built the first serious force platform. The reader is referred to Cavanagh's historical review [2] for further insight into the contributions and historical significance of the works of Braune, Fischer, Muybridge, Hill, Fenn, Elftman, and Hubbard.

The explosion of interest in running has prompted a comparable explosion of research and assessment. This has been potentiated by technical advances including faster cameras and marker systems which eliminate the need to hand digitize frame after frame of video. The growth of this field has been spurred by the vast growth in participation in distance running in the late 1960's and early 1970's. Approximately 30 million Americans run for recreation or competition. The rate of injury is significant. Each year between 1/4 and 1/2 of runners will sustain an injury that is severe enough to cause a change in practice or performance [3,4]. This may lead the runner to seek consultation, alter training, or use medication.

Because running shoe companies now had a large new market, they spent part of their profits to support research. The increased incidence of injury highlighted the lack of understanding of the pathophysiology and biomechanics of chronic running injuries. These injuries are due to repetitive application of relatively small loads over many repetitive cycles (in sharp distinction to acute traumatic events such as ACL ruptures in football—a single large load). The tissues respond differently as well [5–7].

It is often the number of repetitions that is problematic. A variety of intrinsic and extrinsic factors have been blamed for the development of these types of injuries [3,4,8]. In addition, particular patterns of injury have been noted. James and Jones [8] noted that almost 75% of complaints fell into six categories (Fig. 1). Interestingly, one might intuitively think that particular anatomic abnormalities lead to specific injury patterns (e.g. hyperpronation predisposing to posterior tibial syndrome or genu varum leading to iliotibial band syndrome), but few such relationships have been found. Given the assumption that greater understanding will improve diagnosis and counseling, the quandary for the last two to three decades has been how to make more sense out of why and how injuries occur.

The volume of literature is extensive; therefore, not all material can be reviewed or referenced. For the most part this treatment of the topic will be restricted to biomechanics and its application to the study of running gait. Clinical information will be reviewed to the extent that it focuses one's attention on the issues at hand. The reader is referred to articles and chapters dedicated to the pathophysiology and management of chronic running injuries [3–7,9–14]. Running Injuries [15] edited by Gary N. Guten, MD provides a relevant, recent review of clinical material. These clinical and pathophysiological issues lie outside the scope of this article. Several prior review articles [16–21] dedicated to the biomechanics of running gait are recommended. These have been invaluable to this author over the years and are highly recommended. The Biomechanics of Distance Running edited by Cavanagh [22] is an essential reference.

Unfortunately, a significant void exists between the world of the biomechanist and the realm of the clinician. A look at the available literature reveals that the link between the field of biomechanics and the clinical realm is difficult to identify. It seems that Dr Stan James (Eugene, OR, USA) has been the clinician who has exhibited the greatest understanding of the biomechanics of running gait [23]. He has also used biomechanical insight to shed light on running injury patterns [8,24] as have several biomechanists [25,26]. Even though shoe manufacturers have lead the way in some areas of biomechanics research, one must wonder whether a broad spectrum of focus is maintained by that approach.

2. Gait cycle

How does one go from a standstill to maximum forward velocity during sprinting? How does the movement strategy change between walking and running locomotion? The demarcation between walking and running (Fig. 1, point A) occurs when periods of double support during the stance phase of the gait cycle (both feet are simultaneously in contact with the ground) give way to two periods of double float at the beginning and the end of the swing phase of gait (neither foot is touching the ground). Generally as speed increases further, initial contact changes from being on the hindfoot to the forefoot (Fig. 1, point B). This typically marks the distinction between running and sprinting. In practicality, the difference between running and sprinting is in the goal to be achieved. Running is performed over longer distances, for endurance, and with primarily aerobic metabolism. Jogging, road racing, and marathons are examples. Approximately 80% of distance runners are rearfoot strikers. Most of the remainder are characterized as midfoot strikers [27]. Sprinting activities are done over shorter distances and at faster speeds, with the goal of covering a relatively short distance in the shortest period of time possible without regard for maintaining aerobic metabolism. Elite sprinters perform with a forefoot initial contact, and in fact, the hindfoot may never contact the ground. For sprinting, the body and its segments are moved as rapidly as possible throughout the entire race. For distance running on the other hand, the body is moved at a more controlled rate in relation to the energy demand of the race.

The gait cycle is the basic unit of measurement in gait analysis [28]. The gait cycle begins when one foot comes in contact with the ground and ends when the same foot contacts the ground again. These moments in time are referred to as initial contact. Stance ends when the foot is no longer in contact with the ground. Toe off marks the beginning of the swing phase of the gait cycle. Each of these phases for both walking and running is subdivided further as seen in Fig. 2. Because the stance phase in walking is longer than 50% of the gait cycle, there are two periods of double support when
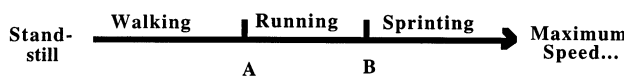

Fig. 1. Forward human locomotion. At point A, stance phase equals 50% of gait cycle. Periods of double support in walking give way to periods of double float in running. Point B for the purposes of the kinematic and kinetic sections of this article represents a change from hindfoot to forefoot initial contact.
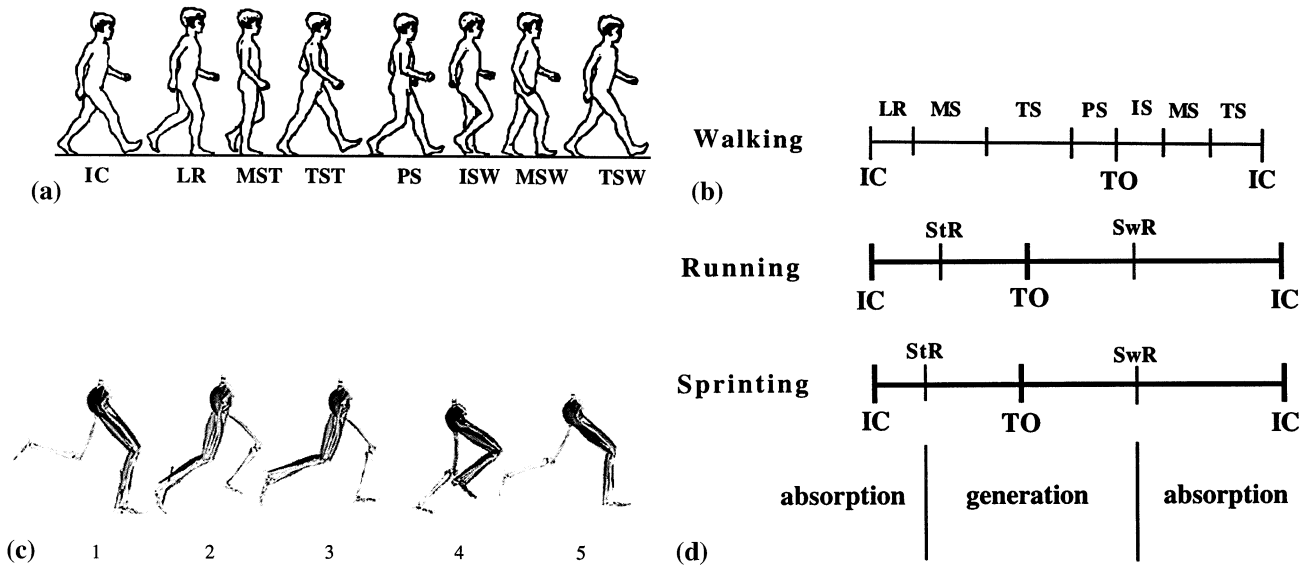
Fig. 2. The gait cycle. 2a. Walking figure. 2b. Walking gait cycle: *IC, initial contact; LR, loading response; *TO, toe off; MS, midstance; TS, terminal stance; PS, preswing; IS, initial swing; MS, midswing; TS, terminal swing. 2c. Running figure: 1. Stance phase absorption. 2. Stance phase generation. 3. Swing phase generation. 4. Swing phase reversal. 5. Swing phase absorption. *Musculoskeletal animation produced using SIMM (Software for Musculoskeletal Modelling, Musculographics, Chicago, Illinois). 2d. Running gait cycle: *for running and sprinting; IC, initial contact; TO, toe off; StR, stance phase reversal; SwR, swing phase reversal; absorption, from SwR through IC to StR; generation, from StR through TO to SwR.

both feet are on the ground (Fig. 3), one at the beginning and one at the end of stance phase.

In running, toe off occurs before 50% of the gait cycle is completed. There are no periods when both feet are in contact with the ground. Instead, both feet are airborne twice during the gait cycle, one at the beginning and one at the end of swing [30,31], referred to as double float. The timing of toe off depends on speed. Less time is spent in stance as the athlete moves faster. In our study, toe off occurred at 39 and 36% of the gait
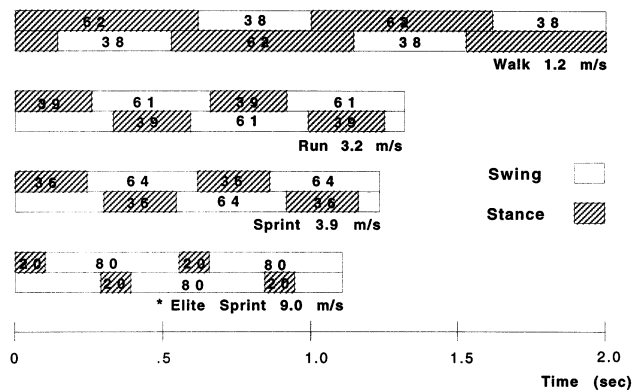

Fig. 3. Variation in gait cycle parameters with speed of movement. For each condition, the bar graph begins at initial contact on the left and represents two complete gait cycles or strides. Note that as speed increases, time spent in swing (clear) increases, stance time (shaded) decreases, double float increases, and cycle time shortens. Information for this graph comes from data collected at the Motion Analysis Lab at Gillette Children's Specialty Healthcare. *Data for elite sprinting is from Vaughan [16].

cycle for running and sprinting, respectively. Faster runners and elite sprinters spend much less time in stance than that (Fig. 3). World class sprinters toe off as early as 22% of the gait cycle [32].

Regardless of speed, alternate periods of acceleration and deceleration occur during running referred to as absorption and generation (Fig. 2c,d). As can be seen, these phases do not coincide with the timing of initial contact and toe off. They are out of phase. During the period of absorption, the body's center of mass falls from its peak height during double float. This period is divided by initial contact (IC) into swing phase absorption (Fig. 2c, #5) and stance phase absorption (Fig. 2c #1). The velocity of the center of mass decelerates horizontally during this period as well. After stance phase reversal, the center of mass is propelled upward and forward during stance phase generation (Fig. 2c #2). Kinetic and potential energy increase. The limb is then propelled into swing phase after toe off (swing phase generation—Fig. 2c #3). At swing phase reversal (Fig. 2c #4), the next period of absorption begins. These issues will be discussed further in a subsequent section on the interaction of potential and kinetic energy.

While stance will be plotted before swing for the purposes of this article, not all authors agree with this convention. Many publications depict swing phase first. In fact, DeVita [29] felt strongly that toe off should mark the beginning of the gait cycle and that swing phase be depicted before stance. His reasoning was based on the observation that both net joint torques
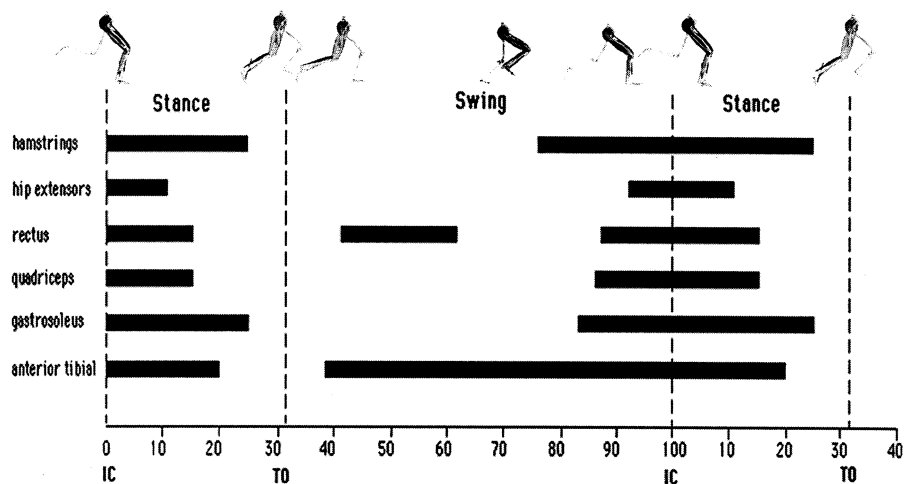
Fig. 4. EMG (adapted from Mann and Hagy [32]). Muscle activity is represented by the solid bars in relation to the gait cycle. Approximately 1.3 gait cycles are depicted in an effort to better visualize the continous nature of running gait (by eliminating the artificial division caused by beginning and ending the cycle at initial contact). Note the greater number of active muscle groups around the time of initial contact (IC) and the lack of muscle activation at the time of toe off (TO).

and EMG activity are greater at the transition from swing to stance than from stance to swing. This suggests that the body's preparation for ground contact is more significant than that needed to leave the ground. This author has also felt that the artificial division of the events surrounding initial contact (by depicting it at the beginning and the end of the cycle) can lead to a loss of insight into the continuous nature of the activity. Both events of course are important, and the issue can be easily resolved by depicting two consecutive cycles adjacent to one another.

3. EMG

Muscle activity during normal walking [33] and running [30,34,35] has been well documented. Typical electromyographic (EMG) activity for running is depicted in Fig. 4.

In general, muscles are most active in anticipation of and just after initial contact. Muscle contraction is apparently more important at that time than it is for the preparation for and the act of leaving the ground. This certainly is consistent with DeVita's contention that the events surrounding IC are more important than those around the time of TO. This is the basis for his recommendation that swing phase be depicted first and stance phase second when graphically representing the running gait cycle [29]. Muscle activity and function will be discussed in more detail in the kinetics section, and the reader can refer back to Fig. 4 to examine the relationship between kinetic findings and EMG activity. Comments here will be brief.

One must remember that there is a delay (approximately 50 ms [36]) between the onset of EMG activity and the development of muscle force. One should also not be confused by what would seem to be early cessation of EMG activity of the gastrosoleus and quadriceps in midstance because muscle force is still present after EMG activity ceases.

3.1. Rectus/quadriceps

The quadriceps and rectus femoris both fire from late swing to midstance to prepare the limb for ground contact and to absorb the shock of that impact during stance phase absorption. The cycle time for the data presented in Fig. 4 is 0.6 s. The onset of quadriceps activity is at 87%, 78 ms before IC, which is consistent with the development of muscle force just before IC. Only the rectus is active in midswing. This is essential to restrain the posterior movement of the tibia as the knee flexes. The biarticular rectus probably plays a role in energy transfer between segments (see later section on biarticular muscles).

3.2. Hamstrings/hip extensors/gastrosoleus

All of these muscle groups have similar activation times as described above. The hamstrings and hip extensors extend the hip in the 2nd half of swing and the first half of stance. The hamstrings also decelerate the momentum of the tibia as the knee extends just prior to IC. Similar to the rectus, the biarticular hamstrings play a role in energy transfer between segments. The hamstrings and gastrosoleus both have important eccentric and concentric functions while the hip extensors probably function only concentrically (see subsequent kinetics section).
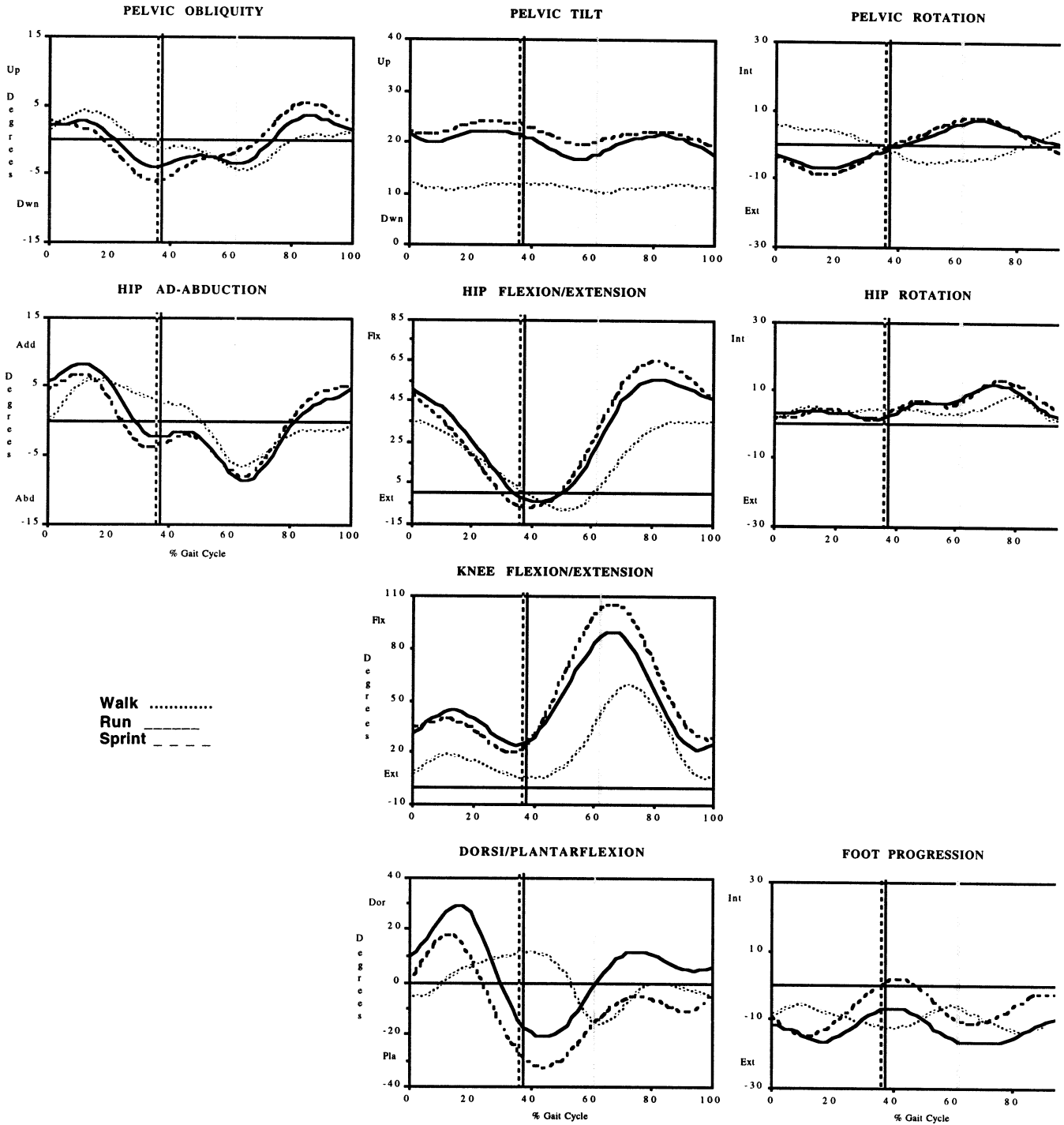
Fig. 5. Kinematics. These graphs show the changing position of the joint listed for one complete gait cycle in all three planes. Each graph begins and ends at initial contact and therefore represents one gait cycle along the *x*-axis. The vertical dashed line represents toe off for each condition. The portion of the graph to the left of the toe off line depicts joint motion during stance phase while swing phase motion is depicted to the right of the dashed line. The position of the joint or body segment in degrees is represented along the *y*-axis. Walking is represented by the lightly-dashed line, running the solid line, and sprinting the heavy-dashed. The corresponding toe-off line is plotted using the same line style. The continuous line connects fifty data points (every 2% of the gait cycle) and represents average data (15 strides) for each of the three conditions. The position of the pelvis is plotted relative to the horizontal and vertical coordinate system of the lab. Hip position represents the position of the femur plotted relative to the position of the pelvis. Knee flexion-extension denotes the angle between the femur and the tibia. 0° indicates full extension (180° between the femoral and the tibial shafts). Dorsiflexion-plantarflexion is the position of the foot relative to the tibia with a 90° angle being plotted as 0°. Foot progression angle depicts the orientation of the foot relative to the lab.

DORSI/PLANTARFLEXION
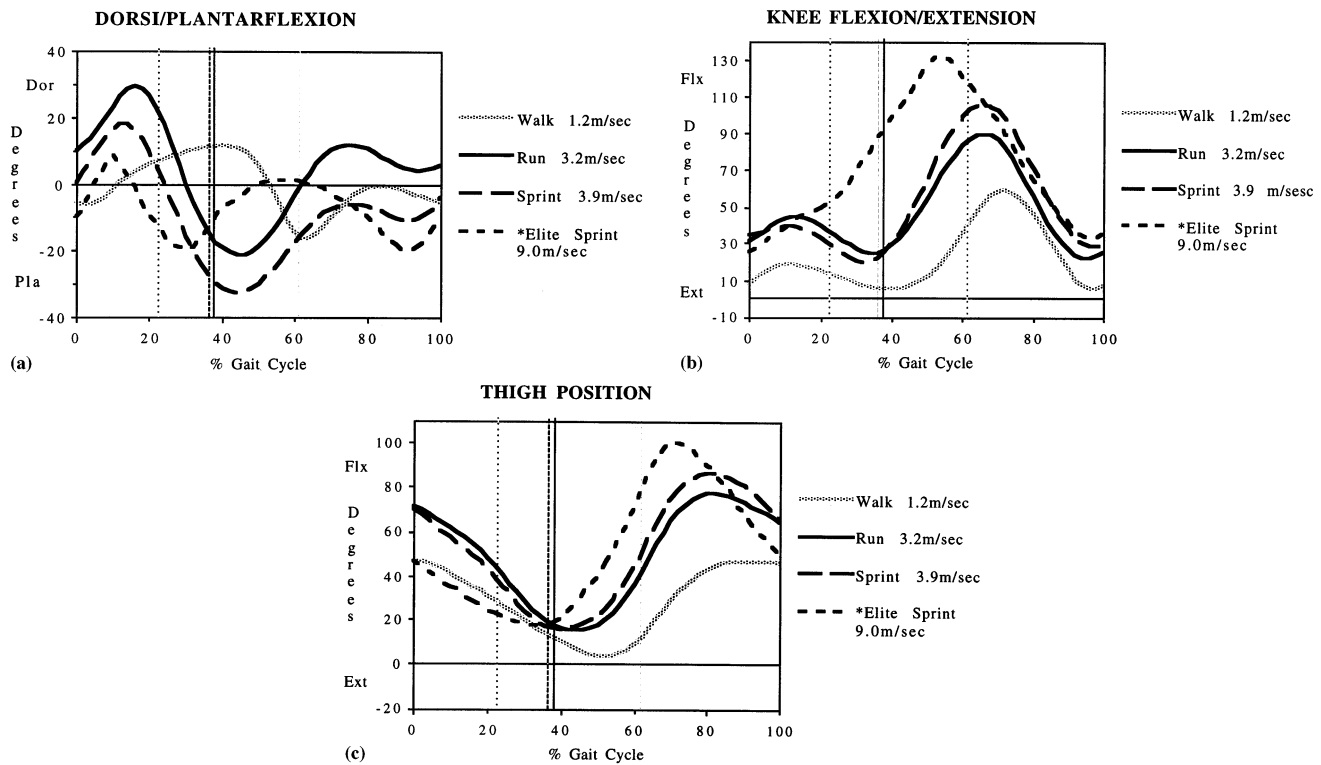
KNEE FLEXION/EXTENSION

THIGH POSITION

Fig. 6. Sagittal plane ankle, knee, and hip motion. 6a. Ankle motion: TO timing, Elite 22%, Sprint 37%, Run 39%, Walk 62%. The three sets of slower data were collected at the Motion Analysis Laboratory at Gillette Children's Specialty Healthcare and represent average data (15 strides) for each of the three conditions. *Elite sprinting data was adapted from Mann and Hagy [32] and represents the average of two elite sprinters with similar velocities. Dorsiflexion-plantarflexion is the position of the foot relative to the tibia with a 90° angle being plotted at 0°. 6b. Knee motion: Knee flexion-extension denotes the angle between the femur and the tibia. 0° indicates full extension (180° between the femoral and the tibial shafts). 6c. Thigh position: Thigh position denotes the position of the thigh relative to the vertical. A 0° angle indicates that the thigh is in a vertical position. This is comparable to the hip flexion/extension angle depicted in Fig. 5 except that in that case the thigh position is plotted in relation to the position of the pelvis and is, therefore, reported as a hip flexion/extension angle.

3.3. Anterior tibialis

The anterior tibialis dorsiflexes the ankle to provide clearance in swing (concentric), to allow ground contact with the hindfoot initially, and to control the lowering of the forefoot to the ground during the first part of stance (eccentric).

4. Kinematics

Kinematics are a description of movement and do not consider the forces that cause that movement. We can graph kinematic variables as a function of the percentage of the total gait cycle or time. For all of the kinematic graphic data presented in the next section, the patterns of movement are important (when in the gait cycle the joint in question is flexing or extending). The peak values in degrees of movement are not important as they depend on the athlete's level of training and speed. The timing of extremes of motion is important to note. This will be true in the subsequent kinetics section as well. Kinematic data can be expressed in

other ways, e.g. angle-angle diagrams [37,38], but these representations begin to have less meaning for the clinician. Motion in all three planes will be considered. One must be cognizant of what the angular measurements represent when reading such graphs. For example, is the hip angle the absolute position of the thigh segment relative to vertical or the angle formed between the orientation of the thigh and that of the pelvis? See the captions for Fig. 5 and Fig. 6 for the conventions employed in this article. Most of the information in the next two sections has been presented elsewhere [31,39,40].

4.1. Sagittal plane kinematics

When observing sagittal plane motion there is a shift into flexion and the center of mass is lowered as the motion changes from walking to running to sprinting. The pattern of movement in the tilt of the pelvis is similar at all speeds (Fig. 5). One might expect a greater amount of pelvic motion with faster velocities, but there is in fact very little increased motion. Pelvic motion is minimized to conserve energy and maintain efficiency in

running and sprinting. However, as speed increases, the pelvis and trunk tilt further forward. The center of mass is lowered, and the horizontal force produced in the propulsion phase is maximized.

The foot and ground exert an equal and opposite force on one another (the ground reaction force). The position and acceleration of the runner's center of mass determines the direction and magnitude of the ground reaction force. Consider for example a sprinter accelerating from a standstill. During the initial phase of acceleration, the body is tilted forward and the center of gravity falls far ahead of the contact point. After several gait cycles, the sprinter reaches maximum velocity and her center of mass then moves backward. An athlete who tried to accelerate with her body upright would fall over backwards because of the direction of the GRF. The forward trunk lean and pelvic tilt keep the ground reaction force in a position to allow forward acceleration.

Sagittal plane hip motion is essentially sinusoidal in walking. Maximum hip extension occurs just before toe-off and maximum flexion occurs in mid to terminal swing. In running and sprinting maximum hip extension is similar to walking, but occurs slightly later in the gait cycle (at the time of toe off). As velocity increases, so does maximum hip flexion leading to a longer step length. Unlike walking, the hip extends during the second half of swing phase during running and sprinting in preparation for initial contact. This difference is to avoid the excessive deceleration that would occur at the time of initial contact if the foot were too far ahead of the center of mass of the body. The ground reaction force vector would be directed excessively posteriorly.

Although the pattern of knee motion in walking, running, and sprinting is very similar, the extremes of motion are very different. In running, during the absorption period of stance phase, the knee flexes to approximately 45°. This is followed by knee extension to an average of 25° during the propulsion phase. In sprinting, the absorption period is shorter and the knee flexes less. Greater knee extension occurs during the propulsion period peaking at 20°. Swing phase also exhibits differences between walking, running, and sprinting. Maximum knee flexion during swing is about 60° in normal walking. This is much less than the average of 90° in running or the 105° in sprinting. The highly trained athlete in a full sprint may exhibit up to 130° of maximum knee flexion (Fig. 6b).

Initial contact during walking and running occurs with the heel. For walking, this occurs despite ankle plantar flexion because of the position of the tibia. In running, greater ankle dorsiflexion is required to achieve initial heel contact. In sprinting, initial contact occurs on the forefoot. Tibial position allows the ankle to be in a more neutral or slightly dorsiflexed position. In walking, the ankle initially plantarflexes as the fore-

foot is lowered to the ground. In contrast, during the absorption phase of running and sprinting, the ankle dorsiflexes as body weight is transferred to the stance leg. Maximum dorsiflexion during stance phase in sprinting is less than in running because of the relatively plantarflexed position at initial contact and the shorter duration of the absorption period. During the generation phase of stance, maximum ankle plantarflexion is greater in sprinting than in running. During swing phase ankle dorsiflexion is less in sprinting than in both walking and running. Dorsiflexion to a neutral position is not necessary for toe clearance during sprinting given the increased amount of hip and knee flexion.

Fig. 6a,b,c show the sagittal plane kinematics for ankle motion, knee motion, and thigh position, respectively, for the slower speeds of movement (walking—1.2 m/s, running—3.2 m/s, and sprinting—3.9 m/s) for the subjects studied at Gillette Children's compared to elite sprinting data (9.0 m/s) adapted from Mann and Hagy [35]. This comparison is included to provide insight into alterations in movement patterns with increasing speed. At the ankle (Fig. 6a), note that the timing of maximum dorsi- and plantarflexion occurs earlier with increasing speed. The range of movement is more restricted in the elite sprinters. Stance phase knee motion (Fig. 6b) is significantly different in the elite sprinters. There is apparently minimal shock absorption or loading response in the elite athletes in that the knee does not extend during the 2nd half of stance, as it does at slower speeds. Apparently that function is completely provided for by the ankle plantarflexors and hip abductors (see subsequent kinetics section). As speed increases the amount of swing phase knee flexion increases. The movement of the thigh (Fig. 6c) is similar at all speeds of movement. Once again, the maxima and minima change with speed (especially in swing) and timing of those peaks occurs earlier with increasing speed of movement. The elite sprinters extend their thighs significantly more in preparation for initial contact (to minimize the loss of speed associated with ground contact).

4.2. Coronal plane kinematics

Overall, coronal plane motion is more subtle than sagittal plane motion. It is however, important in minimizing upper body movement. In this plane, motion of the knee and ankle is restricted by the collateral ligaments. In contrast, significant motion occurs at the hip. As the limb is loaded, the pelvis remains relatively stationary (Fig. 5). The hip adducts relative to the pelvis. This is a shock absorbing mechanism similar to that seen in the sagittal plane at the knee in running and the ankle in sprinting. Throughout the rest of stance phase, the pelvis drops until the start of double

float where it is the most oblique. As the limb begins swing phase, this motion reverses. The pelvis now elevates to obtain foot clearance. Generally, in walking, running and sprinting, the hip is adducted while the limb is loaded in stance phase and abducted during swing. Hip motion in this plane mirrors the movement of the pelvis. This nearly reciprocal motion combined with slight lumbopelvic motion minimizes shoulder and head movement. This is one of the most important mechanisms for decoupling the intense lower extremity motion from the trunk and head. The result is relatively minimal head and trunk motion allowing balance and equilibrium to be maintained.

4.3. Transverse plane kinematics

Motion in the transverse plane, as in the coronal plane, is small in magnitude compared to the sagittal plane. Joint rotations in this plane may be the most difficult to comprehend because they are difficult to see. The movement patterns in the transverse plane are important for energy efficiency (to be discussed in greater detail later). The function and motion of the pelvis in the transverse plane is very different in walking than in running and sprinting. In walking, pelvic rotation is an important method of lengthening the stride. The pelvis is maximally rotated forward at initial contact to achieve a longer step length (Fig. 5). The result is decreased horizontal velocity. During running and sprinting maximum internal pelvic rotation occurs in midswing to lengthen the stride, but by the time of initial contact, the pelvis has rotated exteriorly. This maximizes horizontal propulsion force and avoids the potential loss of speed.

The pelvis in running and sprinting also functions as a pivot between the counter-rotating shoulders and legs. For example, when the right leg is maximally forward in midswing, the left shoulder is rotated forward and the pelvis is neutral.

Another important motion, pronation/supination, occurs in an oblique plane in the foot. In the graphs of Fig. 5, the portion of this motion that occurs in the transverse plane is reflected as changes in the foot progression angle during stance phase. The sagittal plane component of pronation/supination is incorporated in the ankle dorsi/plantarflexion plot (since the foot is modeled as a rigid body in this model). Pronation occurs during the absorption phase while the limb is loaded. The foot then supinates during the generation phase providing a stable lever for push-off. In addition to the boney and ligamentous structures of the foot, the posterior tibialis helps to control this motion. This complex motion is difficult to quantify biomechanically because the motions are small in magnitude, and the body segments about which they occur are small and defy accurate localization (see later section).

5. Kinetics

Winter and Bishop outlined the major goals associated with athletic events [26] providing an overall outline in which to organize one's thinking about the tasks that muscles must perform. They are

1. shock absorption and control of vertical collapse during any weight acceptance phase;
2. balance and posture control of the upper part of the body;
3. energy generation associated with forward and upward propulsion;
4. control of direction changes of the center of mass of the body.

The study of kinetics begins to answer the 'how and why' of the movement we observe.

5.1. Center of pressure (COP)

One method of evaluating force application to the foot is the assessment of center of pressure and the mapping of pressure distribution. Significant variability is seen, especially between mid- and rearfoot strikers [17]. Pressure distribution mapping can be represented graphically in many ways. One representative method is seen in Fig. 7.

Pressure is generally initially focused on the lateral border of the heel. It moves fairly rapidly to the medial aspect of the heel and to the forefoot where two peaks of pressure of nearly equal magnitude under the first and second metatarsal heads are seen. Of course, this type of mapping analysis is altered significantly by the
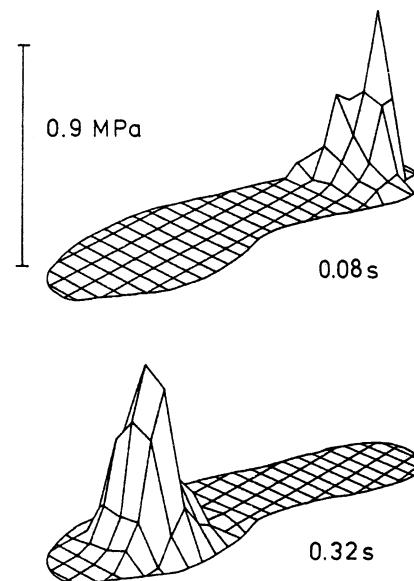

Fig. 7. One representative method of mapping plantar pressure distribution. Indicated times represent elapsed time from the moment of initial contact for this rearfoot striker. (from Cavanagh [17], with permission).
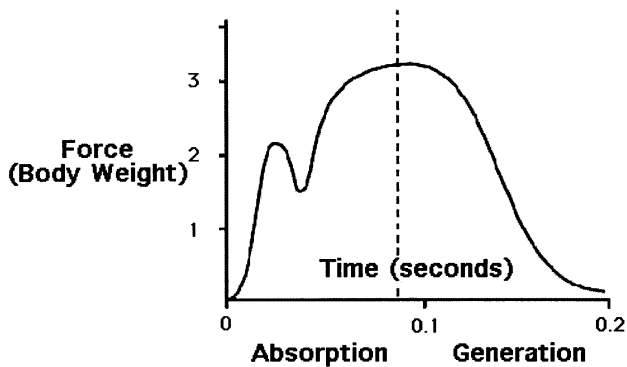
Fig. 8. Schematic of a typical ground reaction force (vertical component) in a runner (rearfoot striker). The first peak is a passive force peak associated with the shock of contact with the ground. This is attenuated by the heel pad and shoewear and modified by passive characteristics of the running surface. It is generally smaller and of shorter duration than the second peak due to active muscle forces (as depicted in this example). This second peak is centered about stance phase absorption (vertical dashed line) marking the end of deceleration (absorption) and the beginning of acceleration (generation).

wearing of shoes. The reader must recognize that as long as the motion of the body's center of mass is unchanged, shoewear does not alter the measured ground reaction force. Shoewear clearly can change the pressure application to different anatomic structures of the foot.

5.2. Raw force plate data

Raw force plate data can be analyzed and depicted relative to the three dimensional laboratory coordinates. This type of analysis has been extensively reported. Several articles are particularly recommended [16,21,41,42]. Fig. 8 is a schematic representation of a typical vertical component of the ground reaction force in a runner who is a rearfoot striker. No one foot strike pattern can be taken as representative of runners in general. By comparison, midfoot strikers generally have no initial vertical force peak. Maxima and minima are velocity dependent [43].

5.3. Sagittal plane joint moments and powers

By combining kinematics with the measured ground reaction force, net joint moments and powers can be calculated. The mathematical method used for this calculation is 'inverse dynamics'.

Sagittal plane kinetic findings are of the most interest. During running, the ankle moment pattern is similar to that in walking (Fig. 9). Initial contact is with the heel. The forefoot is lowered to the ground under the control of eccentric contraction of the anterior tibial muscles. The onset of the ankle plantarflexion moment occurs at 5–10% of the running gait cycle. In contrast, during sprinting there is no initial dorsiflexor moment

because initial contact is on the forefoot followed by immediate dorsiflexion. The total energy absorbed at the ankle is greater in sprinting than in running (see ankle joint power plot). The period of absorption is followed by a period of power generation whether one is walking, running, or sprinting,. The power generated provides energy for forward propulsion. The magnitude of the ankle power generation is directly related to the athlete's speed.

The knee moment pattern is very similar in sprinting and running. To prepare for initial contact, the hamstrings become dominant in the second half of swing producing a knee flexor moment. This moment controls rapid knee extension. Shortly after initial contact, the quadriceps become dominant producing a knee extensor moment. The magnitude of the peak knee extensor moment tends to be greater in running than in sprinting. This is related to the runner's greater degree of knee flexion as the limb is loaded.

In running, as the knee flexes following initial contact, the quadriceps contract eccentrically. This is seen as power absorption and reflects their essential role as shock absorbers. In sprinting, however, the ankle plantarflexors absorb much of the shock of contact with the ground. Therefore, little power is absorbed at the knee. In both running and sprinting, the knee extends in the second half of stance phase. The quadriceps contract concentrically and power is generated. In swing phase very little power is generated by the muscles crossing the knee. Instead the muscles absorb power to control the movement of the swinging leg. The rectus femoris contracts eccentrically in early swing to prevent excessive knee flexion. During late swing phase the hamstrings contract eccentrically to control the momentum of the tibia and prevent knee hyperextension as the knee is rapidly extending.

The hip moment pattern is similar in all conditions of forward locomotion. Just prior to and just after initial contact, the hip extensors are dominant. In contrast, the hip flexors are dominant in the second half of stance through the first half of swing. Both the hip flexors and extensors are responsible for increased power generation in running and sprinting. Peak hip flexion occurs in the second half of swing in both running and sprinting. After peak flexion occurs the hip extensors contract concentrically to extend the hip in preparation for initial contact. The power graph depicts power generation for running and sprinting prior to initial contact. The hip extensors continue to generate power through the first half of stance phase. The hip continues to rapidly extend. Following this, the hip flexors become dominant and decelerate the backward rotating thigh in preparation for swing. During this time, the psoas tendon is stretched. The energy absorbed in stretching the tendon is returned at toe off.
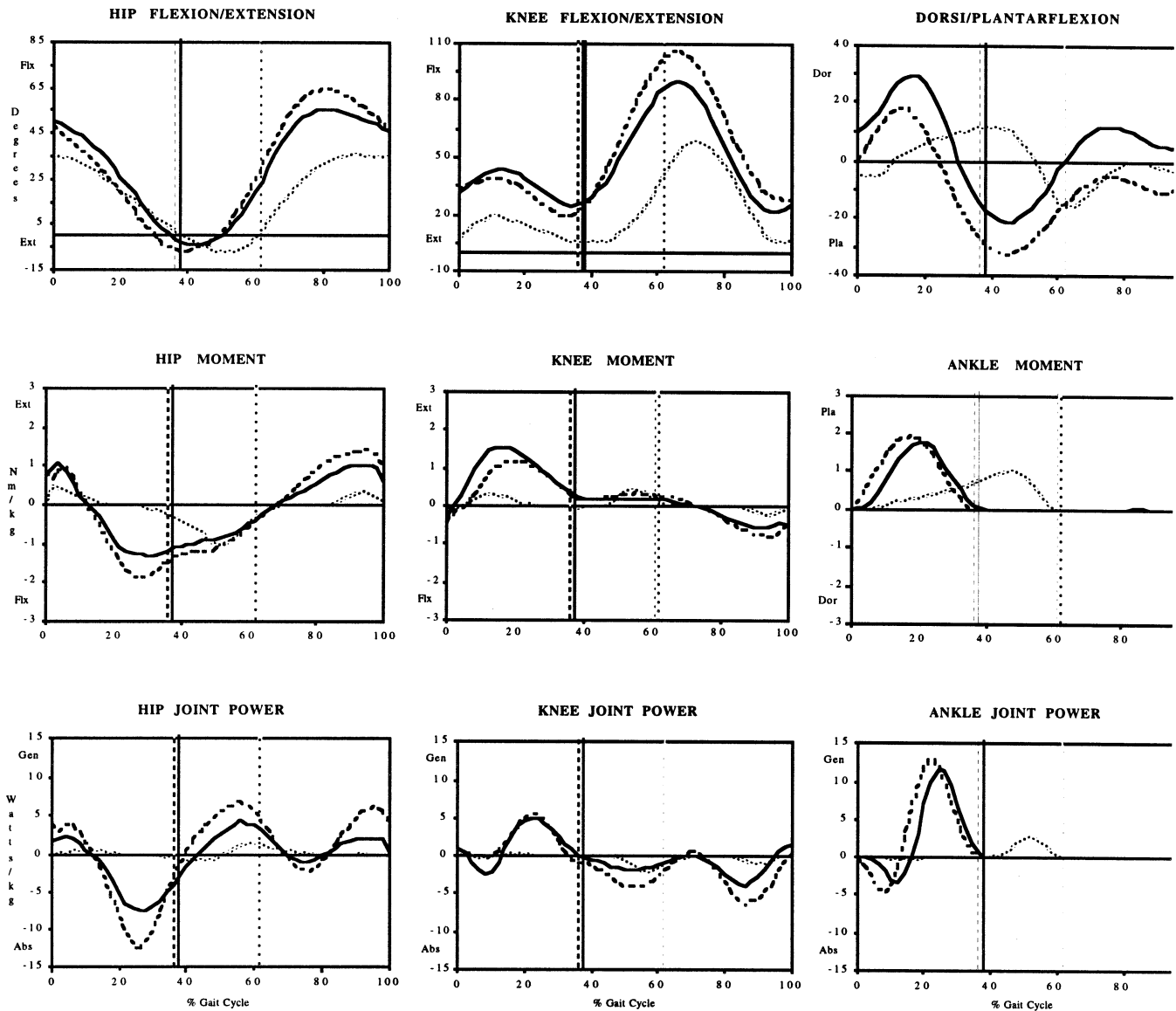
Walk ....
Run _____
Sprint _ _ _ _

Fig. 9. Sagittal plane joint moments and powers. Joints are organized by column. For convenience, the corresponding joint motion is shown in the first row. The net joint moment is shown in the second row. Joint moments are labelled as the internal moment (some authors use the convention of labelling as the external moment). In the net joint power plots (third row) periods of power absorption are negative (deflections below the zero line) while periods of power generation are positive. Results are normalized by dividing by body weight in kg.

5.4. Coronal plane joint moments and powers

Although the magnitudes of coronal plane moments are substantial, the muscles and ligaments that create them function primarily as stabilizers. There is minimal motion; therefore, power generated and absorbed are much less than in the sagittal plane. Coronal plane kinetic data is not graphically depicted in this review. The reader is referred to prior publications [31,39].

During stance phase, a continuous hip abductor moment is produced primarily by the gluteus medius. The hip adducts in the absorption phase because the ground reaction force falls medial to the hip and the hip abductor moment is less than the external adduction moment due to gravitational and acceleration loads. The gluteus medius contracts eccentrically to control this motion. During the propulsion phase, the gluteus medius contracts concentrically abducting the hip and

Walk
(1.2 m/sec)
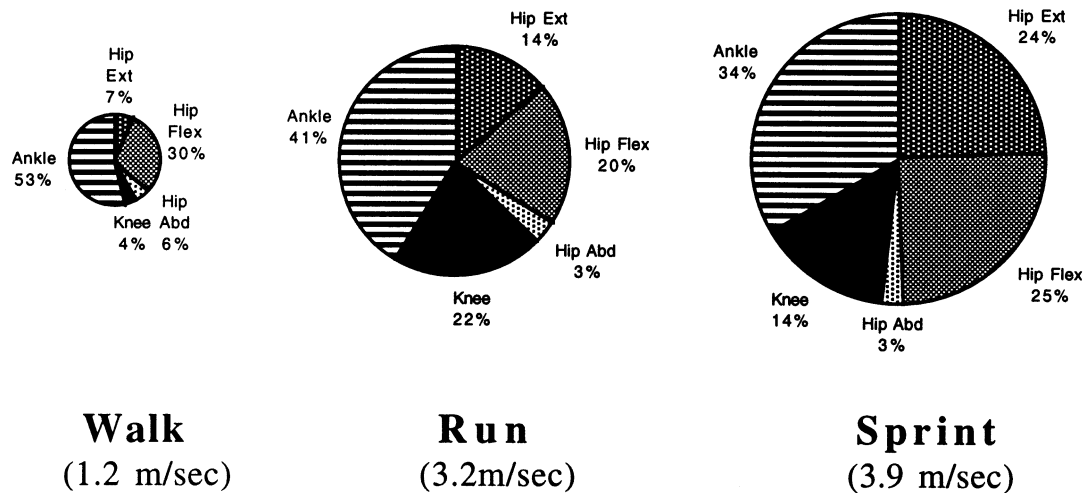
Run
(3.2m/sec)

Sprint
(3.9 m/sec)

Fig. 10. Energy sources. The area of each pie chart corresponds to the total amount of positive work (area underneath the power curve positive peaks—see Fig. 9) measured for the lower extremity joints for each condition. The size of each portion of the chart corresponds to the percentage contribution from each peak included in the calculation.

generating power. At the knee and ankle, moments are generated but little motion occurs. Therefore, ligaments, bone on bone contact forces, and tendons neither generate nor absorb significant power.

5.5. Conclusions based on kinetics

The movement strategy changes as one increases speed. This is apparent by considering the sources of power generation for forward propulsion (Fig. 10). By examining the power curves, one can see that the main sources of power generation are from
1. the hip extensors during the second half of swing and the first half of stance;
2. the hip flexors after toe off;
3. and the knee extensors, hip abductors, and ankle plantarflexors during stance phase generation.

Essentially, the hamstrings and gluteus maximus pull the body forward by actively extending the hip after swing phase reversal when the foot is ahead of the body. Then, during the second half of stance phase, the quadriceps and gastrosoleus contract to push us forward by extending the knee and plantarflexing the foot. The hip abductors are contracting to stabilize the hip and perhaps to provide lift (unproven). Finally, the psoas propels the limb into swing by pulling the thigh forward. The total amount of power generated increases as speed increases, and the relative contribution from each of these muscle groups changes such that relatively more power is generated proximally as speed increases. The muscle mass of a horse (an animal that is able to run much faster than a human) is concentrated proximally. The muscles have long tendons connecting them to their distal lever arms. This anatomic feature is not only consistent with the finding mentioned above,

but it also concentrates weight proximally in order to minimize the inertia of the rapidly moving distal segments.

One should also note that each of these essential power generators stretches eccentrically just prior to generating their burst of power (Fig. 9). It has been shown that tendons stretch and then efficiently return most of that energy when they recoil. In addition, it is well known that muscles that are pretensioned and then contract generate more power per unit of activation than those that are not. In essence one can consider the tendons as the springs and the muscles as the tensioners of the springs (see later section).

The role of the arms in running has been an area of debate. Hinrichs [44] concluded that the arms provide lift and that in distance running, they do not contribute to forward propulsion. They help the runner maintain a more constant horizontal velocity by acting as a counterbalance for the rotating lower extremities [45].

6. Potential and kinetic energy

The relationship between potential and kinetic energy is critically different between walking and running activities (Fig. 11). In walking, the two are out of phase. When potential energy is high, kinetic energy is low, and vice versa. Walking has been referred to as controlled falling (from the zenith of the center of mass in midstance to its nadir during double support) and is similar to a swinging pendulum. In running on the other hand, the two are in phase. Running has been likened to an individual on a pogo stick [21], propelling oneself from a low point during the middle portion of stance (stance phase reversal) to a peak during double float.

Because of this difference, the body completely alters the methods it uses to maintain energy efficiency. Large fluctuations in total energy going into and out of the system would be disadvantageous regardless of the pace of movement. Efficiency in walking is maintained by the effective interchange between potential and kinetic energy. They are out of phase. In running, because the two are in phase, this is not possible. Instead, efficiency is primarily maintained in two ways [16,31,39].

1. The storage and later return of elastic potential energy by the stretch of elastic structures (especially tendons);
2. The transfer of energy from one body segment to another by two joint muscles such as the rectus femoris and the hamstrings.

These two concepts will be addressed separately in the next two sections. These mechanisms do not occur without some cost of their own. It is the repetitive cycling of tendon stretch and recoil that is responsible for many of the chronic overuse syndromes in runners [7] (see subsequent 'injuries' section).
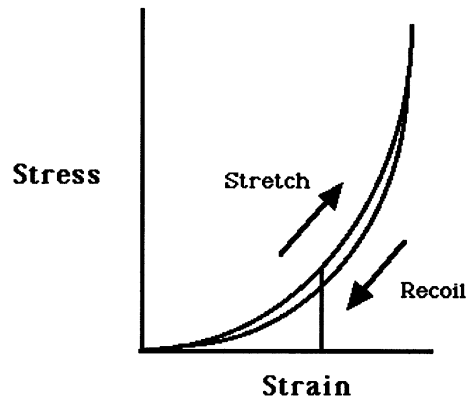
Fig. 12. Hysteresis curve for tendon. Tendons efficiently recoil in a springlike fashion returning approximately 95% of the energy stored when stretched. For any given amount of stretch (strain), the difference in stress is dissipated as heat.

During running, potential and kinetic energy peak in midswing. As the center of mass falls toward the ground, potential energy is lost. As the foot contacts the ground, kinetic energy is lost. Much of the lost potential and kinetic energy is converted into elastic potential energy and stored in the muscles, tendons, and ligaments (see later section on the tendon as the musculotendinous spring). During the generation phase, the center of mass accelerates upward and both potential and kinetic energy increase. Energy for this movement is supplied by the active contraction of the muscles and the release of the elastic potential energy stored in the ligaments and tendons. The storage of energy in the elastic structures of the lower extremities thus plays a more important role in running and sprinting than in walking.

7. Tendons as springs

As mentioned above, each of these musculotendinous units absorbs power by stretching (eccentric) just before they shorten (concentric) to generate power. Recent animal studies have indicated that the changes in the length of the muscle belly itself are relatively minimal [46]. Instead, they function as tensioners of the musculotendinous springs, their tendons. Most of the change in length comes from the stretch and recoil of their respective tendons. Therefore, most of the work is done by the tendons. An excellent source for information on this topic is provided by McMahon [47].

Tendons are, in fact, excellent biological springs (Fig. 12). In this way, we should begin to think of tendons as springs and muscles as the tensioners of the springs. The analogy of a runner to a person on a pogo stick [21] starts to make even more sense! If we consider the Achilles' tendon, for example, we can begin to understand the way that it stretches during the first portion
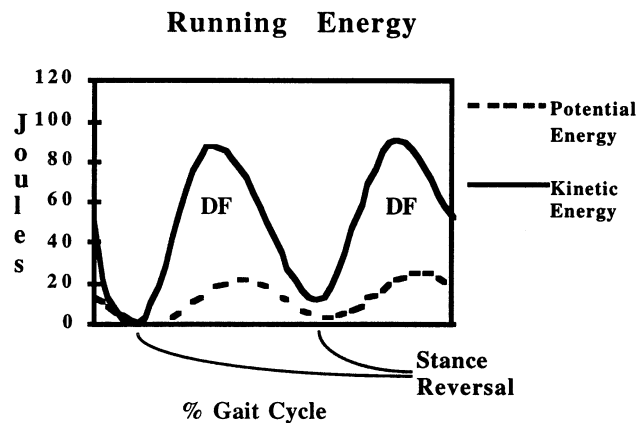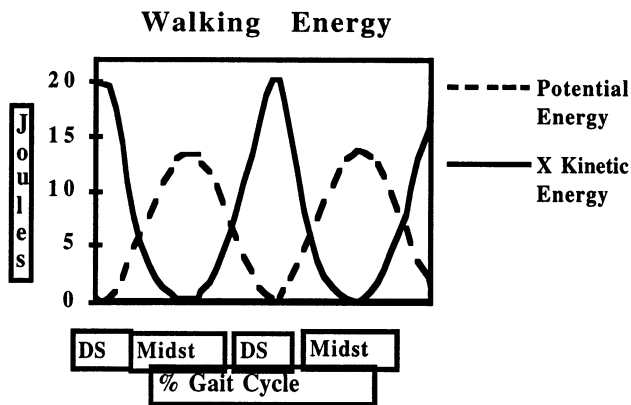

Fig. 11. The relationship between potential and kinetic energy in walking and running. The relationship between potential and kinetic energy is one of the crucial differences between walking and running. In walking, they are out of phase. In running, they are in phase.

of the stance phase of the gait cycle and recoils to return that energy back to the individual at the time of push-off. It functions in a similar fashion to the spring of the pogo stick (except that the tendon stretches under tension while the pogo stick spring compresses under pressure). Each returns the energy that it absorbs back to the system.

R. McNeil Alexander found that the total energy turnover in each stance phase of a 70 kg man is 100 J when running at 4.5 m/s [48]. He estimated that 35 J are stored as strain energy in the heelcord and 17 J in the arch of the foot. More is stored in the stretch of the quadriceps and patellar tendons. Therefore, less than half of the energy has to be removed by the muscles acting as brakes and returned by them doing work. The muscles must still exert the tension, but they shorten/lengthen less. The idea that the body's system of muscle, tendon, and ligament springs behaves like a single linear spring ('leg spring') is supported by the work of Farley and Gonzalez [49]. They concluded that the most important adjustment to the body's spring system to accommodate higher stride frequencies is that the leg spring becomes stiffer (for stride frequencies from 26% below to 36% above preferred frequency, the stiffness increased by 2.3-fold from 7.0 to 16.3 kN/m). They also noted that vertical displacement decreased with increased stride frequency.

8. Biarticular muscles

The second mechanism, transfer of energy between body segments by two-joint muscles, also contributes to energy efficiency. Elftman is the first to be recognized for proposing this principle. Consider the hamstrings in the second half of swing phase. The hip and knee are both extending (Fig. 5) while the hamstrings are contracting (Fig. 4). An extensor moment is produced by the hamstrings at the hip while they generate a flexor moment at the knee (Fig. 9, row 2). The moment produced at the knee is opposite the knee motion. In effect the hamstrings absorb energy at the knee and generate energy at the hip (Fig. 9, row 3). However since the overall change in length of the hamstrings is minimal, the hamstrings as a whole can be considered to neither absorb nor generate energy. In this instance the hamstrings can be thought to function as an 'energy strap', transferring energy from the moving tibia to the pelvis to aid in hip extension. As the knee extends, energy from the tibia is supplied to the pelvis to augment hip extension. A similar type of analysis can be done for the two joint rectus femoris during the first half of swing. This can be visualized more easily by overlaying the power curves for the hip, knee, and ankle (Fig. 13).

Jacobs et al [50] hypothesize that biarticular leg muscles play an effective role in power transfer from proximal joints to distal joints in order to cause an efficient conversion of successive rotational motions of body segments into translation of the body center of gravity. Prilutsky and Zatsiorsky [51] concluded that at push off the two joint rectus femoris and gastrocnemius transfer mechanical energy from the proximal joints of the leg to the distal ones to help extend the distal joints. At initial contact and the first half of stance, they transfer energy from the distal to proximal joints to help dissipate the mechanical energy of the body.

The passive transfer of energy between adjacent segments at the joint and via single joint muscles may also be an important method of maintaining energy efficiency. Robertson and Winter [52] reported that mechanical energy transfer by this method was significant for walking. It is likely that it is as or more important in running, but the author is not aware of any publications on this topic.

9. Economy of motion

It is generally accepted that one of the most important determining factors of the manner in which the individual moves is to maximize energy efficiency. In general it is held that for aerobic, steady state conditions, one chooses the movement strategies which are most economical in regard to energy usage. Economy of movement has also been felt to be a driving force for the evolution of limb structure in terrestrial animals. Despite these beliefs, interindividual variability in walking and running aerobic demand is significant. The source of this variability has been pursued extensively. This author highly recommends the review of the essen-
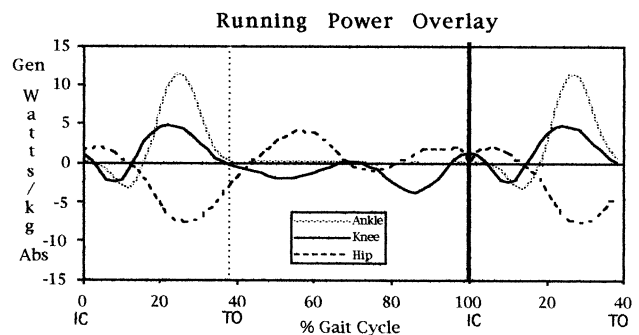

Fig. 13. Power overlay—biarticular muscles. Data represents a single subject running at 3.2 m/s. Notice the flow of energy via the two joint muscles (primarily rectus femoris and hamstrings) depicted by the nearly mirror images of the knee and hip power curves, i.e. when power is absorbed at one joint, it is generated at the other. In this way, biarticular muscles act as 'energy straps' by harnessing the momentum of a moving body segment and transferring that energy to the next adjacent joint.
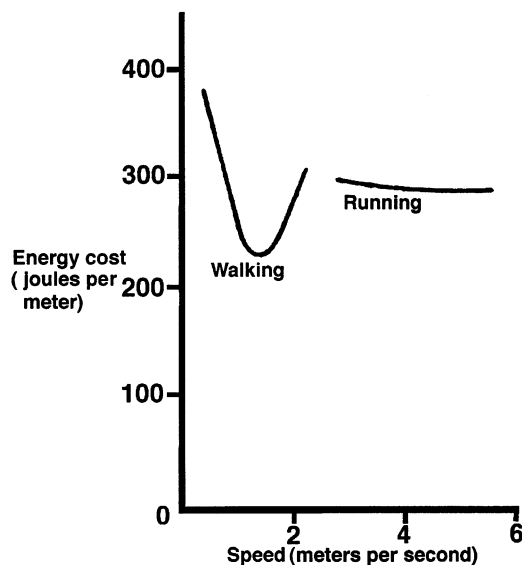
Fig. 14. O_2 curve.

tial aspects of this topic by Martin and Morgan [53]. They examined the economy of movement in regards to the effects of four general categories—body structure, kinematics, kinetics, and biomechanical feedback/training.

It is clear that each individual walks at his/her most economical speed. Fig. 14 (adapted from Alexander [21]) reveals the inverted U-shaped speed/economy relationship for walking. Minimum aerobic demand occurs at approximately 1.3 m/s. Between 1.1 and 1.4 m/s, the speed-energy expenditure curve is nearly flat allowing variability of about ± 15% difference in walking speed without compromising efficiency.

In contrast, no such relationship exists for running speed and energy cost. Oxygen cost changes little over a wide range of chosen running speeds. Economy of movement is obviously maintained by mechanisms that remain elusive, but they undoubtedly include choices of stride length/cadence [54], muscle shortening velocity [55], and sources of mechanical power output [56]. A third conclusion that can be drawn from Fig. 14 is that it would be extremely uneconomical to continue to walk at speeds exceeding 3 m/s. As previously discussed, this difference is at least partially explained by the method which is most important for maintaining energy efficiency. In walking, the transfer between potential and kinetic energy is most important. Since this is not possible in running, energy transfer by the biarticular muscles and the storage and later return of elastic potential energy in tendons are most important.

Kram and Taylor [57] concluded that economy of running involves little relationship to work done against the environment. Instead the two most important factors are a direct relationship to work done by muscles and tendons to lift and accelerate the body and

limbs (i.e. the cost of supporting weight) and an inverse relationship to stride rate (i.e. the shorter the time course for application of force, the higher the energy cost). Finally, as noted by Martin and Morgan [53], 'The question regarding our ability to significantly improve economy through biomechanical training remains unanswered' (p. 472).

10. Foot biomechanics

Lay and sports medicine literature has blamed excessive pronation for nearly all maladies of the lower extremities (and the spine, for that matter). It is felt that abnormal movement of this joint occurring over the course of thousands of repetitious cycles leads to overuse syndromes due to increased internal rotation of the tibia via the mitered hinge effect [58]. There is a large amount of empiric clinical support for this notion in that shoes or orthotics designed to diminish hyperpronation frequently eliminate painful conditions. Unfortunately, there is little quantitative evidence of this type of abnormal biomechanics [17].

Several foot and ankle biomechanics review articles examine normal and pathological mechanics of hind- and midfoot motion [59,60]. Roger Mann [61], of course, has lead the way in educating biomechanists and the medical community in this area. Czerniecki's review [58] correlates foot and ankle biomechanics with Perry's three stance phase ankle rockers combining two separate areas of knowledge. Sagittal plane ankle motion is accompanied by rotation in the transverse plane and rotation of the foot about its long axis due to the oblique orientation of the ankle joint. During gait, when the foot is fixed to the ground, ankle dorsiflexion causes internal rotation of the tibia and pronation of the foot. The subtalar joint also has an oblique axis of rotation and is therefore also responsible for the complex movement of pronation/supination of the foot relative to the tibia. Again via the mitered hinge effect, rotational torques about the longitudinal axis of the foot are transmitted to the tibia resulting in rotational torques about its longitudinal axis. At initial contact, the hindfoot is typically inverted. Pronation then occurs as the limb is loaded during the absorption phase. Pronation 'unlocks' the transverse tarsal joint increasing the flexibility of the foot allowing it to function more effectively as a shock absorber. Peak pronation normally occurs at 40% of stance phase (Fig. 15). The foot then begins to supinate and reaches a neutral position at 70% of stance. The transverse tarsal joint is then 'locked'. The generation phase has been reached, and the foot is now more rigid allowing it to act more effectively as a lever for push-off. The 'hyperpronator' may not begin to supinate or reach a neutral position until later—well after power generation was to have
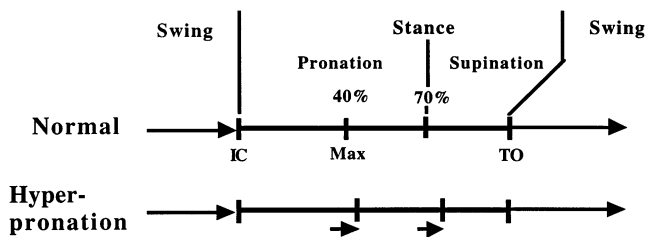
Fig. 15. Abnormal excessive pronation. This schematic illustrates the difference between normal foot mechanics and that of the hypermobile, 'pronated' foot. Time to max pronation is delayed beyond 40% of stance. Likewise, the period of pronation is prolonged delaying the onset of supination. This diagram does not reflect the degree of excessive movement nor does it account for abnormal pronation velocity.

begun. In this case, the foot would not be an effective lever.

The foot in most laboratories is modeled as a single rigid body. Therefore, motion of the ankle, subtalar, and transverse tarsal joints is measured and reported as a single joint. This is clearly too simplistic! Gerald Harris and his co-workers [62,63] and others [64,65] have reported on their efforts to develop an accurate and precise 3-D measurement system to quantify the position of these body segments and of the motion of the joints between them. Hindfoot motion has frequently been measured using two dimensional video measurement. In this type of analysis, eversion and inversion angles of the calcaneus are used to designate pronation and supination movement of the subtalar joint [17] (Fig. 16). These pairs of terms have essentially been used interchangeably.
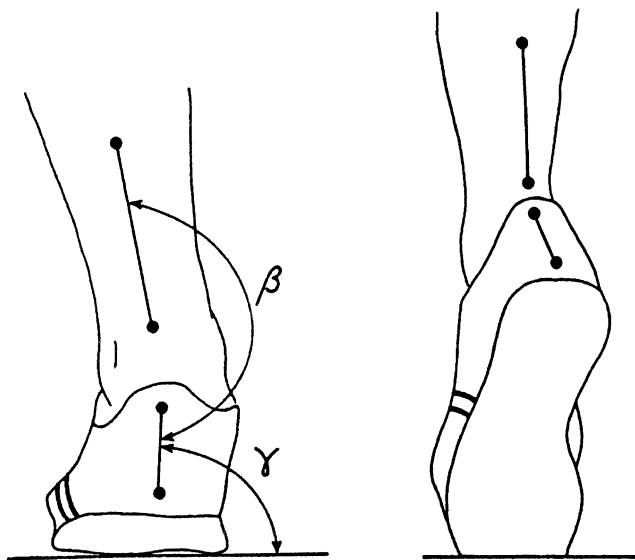

Fig. 16. Typical rearfoot eversion angle. Two dimensional assessment of the rearfoot eversion angle commonly reported in running biomechanics literature. Major sources of error include rotational motion out of the plane of view and the inability to visualize the rearfoot position (obscured by shoewear).

Areblad et al. [64] showed changes in variables of up to 1° for every 2° of change of the alignment angle when comparing angles derived from a 3-D model to the projected angles of the lower extremity gathered with a 2-D model. These errors were found to be most influenced by differences in the alignment of the longitudinal axis of the foot with the camera axis. This points out the concern surrounding the use of 2-D systems.

Other efforts to evaluate and quantify 3-D motion of the foot illustrate new difficulties. Engsberg and Andrews [66] obtained data in 2-D. It was then reduced to 3-D. The authors acknowledged that this may introduce error. A review of their paper also emphasizes the difficulty of graphical presentation of 3-D data in a 2-D reporting format. It can be extremely confusing! They reported the projection of the direction cosine vector of the equivalent screw displacement onto the xyz coordinate system. The complexity of this type of data presentation essentially limits its utility to a small select group of researchers. Certainly the clinician will not gather any meaning from this information. As a clinician, it does not speak to me! The importance of reporting data about rotations between two body segments in a manner that has anatomical functional meaning was emphasized by Soutas-Little et al. [67], yet in their study the foot and shank were modeled as two rigid bodies. The authors pointed out the limitations of the common approach of defining rearfoot motion as the projection of the angle between a line on the posterior aspect of the shank and a line on the heel (Fig. 16).

It remains to be seen whether an absolute degree of pronation, the timing of pronation, or the maximum velocity of pronation [68] is most important in the development of injury.

11. Shoes

Numerous publications have been written on the topic of running shoe analysis [69–75]. Pink and Jobe [76] recently summarized the status of current thought about the interface between the foot and the shoe. It is essential that shoes not only be tested in the laboratories of shoe companies but also in vivo because individuals modify their movement pattern in complex ways in response to changes in their dynamic balance. This is undoubtedly under neurologic control. It certainly makes the documentation of improvement with alterations in shoewear or in shoe orthotics difficult. Perhaps the vast clinical empiric experience relating improvement with orthotic intervention is due to this type of adaptation!

Bates et al. [73,74] concluded that there was no 'best' shoe for all runners. They correlated injuries and management to shock absorption and control/stabilization.

They found that intra- and intersubject variability was large. The differences between subjects for a given condition were greater than the differences between conditions (different shoe designs) for a given subject. They concluded that the shoe must be on the foot to test its function and that dynamic function must be the basis for evaluation and design.

Winter and Bishop [26] stated that for runners 'footwear is predicted to protect or attenuate the potential damaging forces in three ways'

1. shock absorption at heel contact reducing the initial spike of reaction force (protects against joint cartilage damage);
2. stance phase—protects against the rough ground surface;
3. aligning the forefoot to achieve a uniform force distribution at the major chronic injury sites.

For these reasons, the three main areas of focus for shoe design have been on the attenuation of the shock of heel strike, the control of hindfoot motion during loading response, and forefoot stability in stance phase. An ideally constructed shoe provides both shock absorption and stabilization of the foot. Intrinsic factors of each individual runner such as degree of pronation, flexibility of the foot, and body weight are all important factors which must be considered when selecting a running shoe.

Stability and motion control are addressed in last design, stiffer heel counters, lacing systems, fiberglass midsole plates, and material combinations of varying density in the shoe's midsole [77]. Design features that control the tendency for hyperpronation and maintain neutral forefoot position in midstance can minimize excessive stresses on the medial side of the Achilles' tendon or plantar fascia. Forces may be more uniformly distributed and therefore the potential for injury minimized.

Cushioning and rearfoot control require opposite design features. Therefore, a single shoe design cannot maximize both. For more cushioning, thicker-soled shoes are better than softer ones, but softer materials control pronation poorly.

Interestingly, quantifiable differences between racing and training shoes are negligible despite the commonly held belief that racing shoes provide less shock absorption and control of movement [43]. Perhaps runners adapt their running style to maintain acceptable force levels. On the other hand, actual differences may exist but are undetectable.

Mechanical testing of shoes for shock attenuation has shown a 33% difference between different shoe models [77]. In this report, 75% of shock attenuating capability was found to be retained after 50 miles and only 67% after 100—150 miles. In vivo testing in volunteers showed similar but less severe degradation with 70% retention of cushioning after 500 miles.

Running shoe manufacturers have attempted to develop a cushioning system that not only dissipates energy but also stores it to allow for passive energy exchange (to enhance performance). To this point, to the author's knowledge, the amount of energy return remains quite small compared to the storage characteristics of the stretch/recoil properties of the soft tissue structures of the athlete's lower extremities [78].

While alterations in shoewear or the use of orthotics may decrease foot and lower leg problems, the more proximal problems persist.

Despite advances in shoewear technology, the overall rate of injury in distance runners has not changed significantly. This lends credence to the concept that chronic injury patterns are due to factors other than the forces generated by the shock of initial contact and foot alignment in stance.

12. Injuries

As has been shown, forces are not only higher but they must be attenuated in roughly one-third the time (as compared to walking). Even a slight biomechanical abnormality can induce injury [77]. It should be apparent that injured runners can not be tested to provide insight into the mechanisms by which they became injured. Dynamic analysis in that case would document the compensatory gait mechanisms employed by the runner to avoid pain rather than the gait pattern that lead to injury. Instead, runners would need to be tested prior to injury and then followed clinically for the development of an injury.

Until biomechanical analysis of the forces created in running was available, the forces that create the tissue trauma responsible for chronic injuries was unknown. This led to inaccurate assumptions. The greatest of these was that most injuries occur as a result of the high impact forces at the time of heelstrike. As a result, a tremendous amount of research has focused on footwear and the running surface and how those two factors alter the impact of heelstrike [69–75]. By reviewing Fig. 8, it is easy to see that the passive forces associated with heelstrike are smaller and shorter in duration than the larger, active force phase during the latter 3/4 of the stance phase [40]. This is not to say that attenuating the shock of ground contact is not important. It is essential, however, to understand that absorption does not occur instantaneously like a bowling ball landing on a cement sidewalk! Several different tissues dissipate this force over time during the first half of stance phase (as previously discussed) thereby minimizing the shock to the body [40]. These tissues include

1. Achilles' tendon
2. Plantar fascia
3. Quadriceps mechanism

4. Hip abductors

It is notable that this list is comprised of many of the most common injury sites in distance runners. Inverse dynamics has indeed advanced the state of the art of biomechanical evaluation! It now seems more likely that most of the chronic injuries from jogging are related to the high forces that occur in mid- and late stance [79]. Based on these calculations, training with eccentric knee exercise and concentric plantarflexion can be recommended to help avoid injury. We now have an appreciation for the biomechanical stresses that can give us insight into the etiology of some of the most common injury patterns.

Even though inverse dynamics allows the evaluation of net joint moments about the hip, knee, and ankle, and provides insight into the location and timing of these soft tissues stresses [26,31,39], actual stress levels within specific musculotendinous structures cannot be measured unless strain gauges are implanted. Biomechanical link segment models can be used to estimate force and stress levels. For instance Achilles' tendon forces have been estimated to be approximately 6–8 times body weight [21,80] and patellofemoral contact forces between 7 and 11 times body weight [80]. The development of improved models will allow even more accurate calculation of these and other individual tissue stresses.

In the concluding remarks of his 1987 review article, Cavanaugh wrote, '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'. A number of publications are now available regarding the biomechanical analysis of common injury patterns in runners [25,26,39,40,81,82]. Because of space constraints only one injury pattern, Achilles' tendinopathy, will be presented here.

The Achilles' tendon and its insertion are frequent sites of chronic injury in athletes. Pain along the course of the tendon is the most frequent presenting complaint. Tenderness with or without swelling along its course is common. Acute ruptures are almost always preceded by a prodromal period of low grade pain [10,11].

The Achilles' tendon is one of the anatomic structures that stretches during the first half of stance phase and recoils later in a spring-like fashion. It stores energy as it is stretched and efficiently returns 90% at the time of push off [21]. If initial contact is on the forefoot, the eccentric function of the gastrosoleus/Achilles' tendon complex is exaggerated as the heel is lowered to the ground. The gastrosoleus generates large ankle plantar flexor moments during running compared to those generated during walking (Fig. 9). As mentioned, because there are few other structures involved, peak Achilles' tendon forces have been estimated to be in the range of 6–8 times body weight [21,80]. Peak forces do not occur at initial contact, but in midstance. They are generated by the powerful contraction of the gastrosoleus—not by the shock of initial contact with the ground. These injuries are due to the active muscle forces of midstance not to the passive impact forces at the time of initial contact.

Shoe wear and the type of running surface are much less important factors in the genesis of this type of injury than is commonly believed. Shoewear may play a role in decreasing locally increased stress if you are running on an uneven surface or if you are a hyperpronator. Again, if the shoe can control the position of the hindfoot, the localized stresses both along the medial aspect of the Achilles' tendon and further up the kinetic chain may be decreased.

13. Future directions

Not to be a pessimist, but if one looks at concluding remarks by authors over the years, promises of future knowledge are routinely made. Authors commonly state that greater knowledge will lead to a decreased frequency of injury. These promises are oftentimes overstated and incompletely filled. As Nigg [83] has pointed out, there is, as yet, no evidence that biomechanical research in load analysis has contributed to a decreased frequency of running injuries. This author is hopeful for greater insight and further questioning.

Technical advances in portability will broaden the scope of application. Facilities with the combination of adequate testing space, three dimensional computerized data gathering and reduction, data acquisition speeds in the range of 150–240 Hz testing speeds, and the breadth of technical, engineering, and clinical knowledge to utilize the equipment will contribute the most to the field. If a large cohort of runners underwent dynamic analysis and then were followed for the development of subsequent injury, perhaps some insight into predisposing biomechanical factors for injury would be gained.

Biomechanical models must be improved. The need seems greatest in three overall areas. Firstly, we must improve the ability to calculate individual bone and muscle forces. Secondly, improvements in the measurement of subtle transverse plane motions will finally allow quantification and understanding of the role of this type of motion in the genesis of injury. Finally (and probably most importantly), improvements in the analysis of the complex three dimensional movements of the foot and ankle will unlock the secrets that are hiding inside the runner's shoe. We need accurate 3-D foot models and a testing methodology that is readily available to a large number of laboratories.

Miller also points out the continuing need to evaluate individual body segment contributions to running [38]. Perhaps the techniques reported by Kepple [84] hold promise in determining contributions of joint moments to vertical and forward progression of the body's center of mass.

Finally, we must close the gap between practitioners and biomechanists. It will be important to standardize terminology and to agree on reporting conventions, e.g. is full knee extension 0° or 180°? If information is presented in a format familiar to clinicians, more practising physicians will use it. Data presented as tables of numbers is essentially meaningless for the clinician. Plots on graphs are not much better. Electronic communication will augment the use of animation, video, and live action to display data. Once new biomechanical knowledge is gained, it is the responsibility of the research community to present it to clinicians in an understandable manner. Much can be gained if the biomechanist and pathophysiologist come out of the laboratory, the clinician pulls him/herself out of the clinic, and they all meet on the track.

Acknowledgements

References

[1] Encyclopedia Britannica, 1952:234.
[2] Cavanagh PR. The mechanics of distance running: a historical perspective. In: Cavanagh PR, editor. Biomechanics of Distance Running. Champaign, IL, Human Kinetics Publishers, 1990; Chapter 1:1–34.
[3] Hanks GA, Kalenak A. Running injuries. In: Grana WA, Kalenak A, editors. Clinical Sports Medicine. Philadelphia: W.B. Saunders Company, 1982:458–65.
[4] Renstrom PA. Mechanism, diagnosis, and treatment of running injuries. AAOS Instructional Course Lectures 1993;42:225–34.
[5] O'Brien M. Functional anatomy and physiology of tendons. Clin Sports Med 1992;11(3):505–20.
[6] Booth FW, Gould EW. Effects of training and disuse on connective tissue. Exer Sport Sci Rev 1975;3:83–112.
[7] Leadbetter WB. Cell-matrix response in tendon injury. Clin Sports Med 1992;11(3):533–78.
[8] James SL, Jones DC. Biomechanical aspects of distance running injuries. In: Cavanagh PR, editor. Biomechanics of distance running. Human Kinetics Books, 1990:249–269.
[9] Brewer B. Mechanism of injury to the musculotendinous unit. AAOS Instructional Course Lectures, 1960;14.
[10] Clement DB, Taunton JE, Smart GW. Achilles tendinitis and peritendinitis; etiology and treatment. Am J Sports Med 1984;12(3):179–84.
[11] Galloway MT, Jokl P, Dayton DW. Achilles tendon overuse injuries. Clin Sports Med 1992;11(4):771–82.
[12] Hoeberigs JH. Factors related to the incidence of running injuries. Sports Med 1992;13(6):408–22.
[13] Järvinen M. Epidemiology of tendon injuries in sports. Clin Sports Med 1992;1111(3):493–504.
[14] Kibler WB. Clinical implications of exercise: injury and performance. AAOS Instructional Course Lectures 1994;43(4):17–24.
[15] Guten GN. Running Injuries, Guten, GN, editor. Philadelphia, PA: W.B. Saunders, 1997.
[16] Vaughan CL. Biomechanics of running gait. Crit Rev Eng 1984;12(1):1–48.
[17] Cavanagh PR. The biomechanics of lower extremity action in distance running. Foot Ankle 1987;7(4):197–217.
[18] Williams KR. Biomechanics of running. Exer Sport Sci Rev 1985;13:389–421.
[19] Anderson T. Biomechanics and running economy. Sports Med 1996;22(2):76–89.
[20] Adelaar RS. The practical biomechanics of running. Am J Sports Med 1986;14(6):497–500.
[21] Alexander RM. Running. The human machine. Natural History Museum Publications, London 1992:74–87.
[22] Cavanagh PR. Biomechanics of distance running, Cavanagh PR, editor. Champaign, IL: Human Kinetics, 1990.
[23] James SL, Brubaker CE. Biomechanics of running. Orthop Clin North Am 1973;4(3):605–15.
[24] James SL. Running injuries to the knee. AAOS J 1995;3(6):309–18.
[25] Andriacchi TP, Kramer GM, Landon GC. The biomechanics of running and knee injuries. Finerman G, editor. AAOS Symposium on Sports Medicine: The Knee. St. Louis, CV Mosby, 1985:23–32.
[26] Winter DA, Bishop PJ. Lower extremity injury-biomechanical factors associated with chronic injury to the lower extremity. Sports Med 1992;14(3):149–56.
[27] Kerr BA, Beauchamp L, Fisher V, Neil R. Footstrike patterns in distance running. Nigg BM, Kerr BA, editors. Biomechanical Aspects of Sport Shoes and Playing Surfaces. Calgary, Canada: University Printing. 1983:135–142.
[28] Gage JR. An overview of normal walking. AAOS Instructional Course Lectures 1990;39:291–303.
[29] DeVita P. The selection of a standard convention for analyzing gait data based on the analysis of relevant biomechanical factors. J Biomech 1994;27(4):501–8.
[30] Ounpuu S. The biomechanics of running: a kinematic and kinetic analysis. AAOS Instructional Course Lectures 1990;39:305–18.
[31] Novacheck TF. Walking, running, and sprinting: a three-dimensional analysis of kinematics and kinetics. AAOS Instructional Course Lectures 1995;44:497–506.
[32] Mann RA, Hagy J. Biomechanics of walking, running, and sprinting. Am J Sports Med 1980;8(5):345–50.
[33] Inmann VT, Ralston HJ, Todd F. Human walking. Baltimore: Williams and Wilkins, 1981;103–117.
[34] Ounpuu S. The biomechanics of walking and running. Clin Sports Med 1994;13(4):843–63.
[35] Mann RA. Biomechanics of walking, running, and sprinting. Am J Sports Med 1980;8(5):345–50.
[36] Sherif MH, Gregor RJ, Liu LM, Roy RR, Hager CL. Correlation of myoelectric activity and muscle force during selected cat treadmill locomotion. J Biomech 1883;16:691–701.

[37] Grieve DW. The assessment of gait. Physiotherapy 1969;55:452–60.

[38] Miller DI. Biomechanics of running: what should the future hold?. Can J Sport Sci 1978;3:229–36.

[39] Novacheck TF. The biomechanics of running and sprinting. In: Guten GN, editor. Running Injuries. Philadelphia, PA: W.B. Saunders, 1997;2:4–19.

[40] Novacheck TF, Trost JP. Running: Injury mechanisms and training strategies, 1997, Instructional Videotape, Gillette Children's Specialty Healthcare Foundation.

[41] Munro CF, Miller DI, Fuglevand AJ. Ground reaction forces in running: a reexamination. J Biomech 1987;20(2):147–55.

[42] Cavanagh PR. Ground reaction forces in distance running. J Biomech 1980;13(5):397–406.

[43] Williams KR, Cavanagh PR, Ziff JL. Biomechanical studies of elite female distance runners. Int J Sports Med 1987;8(2):107–18.

[44] Hinrichs RN. Upper extremity function in distance running. In: Cavanagh PR, editor. Biomechanics of Distance Running. Champaign, IL: Human Kinetics, 1990;4:107–133.

[45] Novacheck TF, Trost JP, Schutte L. Running and Sprinting: A Dynamic Analysis (Video/CD-ROM). St. Paul, MN, Gillette Children's Hospital, 1996.

[46] Roberts TJ, Marsh RL, Weyand PG, Taylor CR. Muscular force in running turkeys: the economy of minimizing work. Sci Mag 1997;275:1113–5.

[47] McMahon TA. Spring-like properties of muscles and reflexes in running: multiple muscle systems. Biomech Movement Org 1990;37:578–90.

[48] Ker RF, Bennett MB, Bibby SR, Kester RC, Alexander RM. The spring in the arch of the human foot. Nature 1987;325:147–9.

[49] Farley CT, Gonzalez O. Leg stiffness and stride frequency in human running. J Biomech 1996;29(2):181–6.

[50] Jacobs R, Bobbert MF, van Ingen Schenau GJ. Mechanical output from individual muscles during explosive leg extensions: the role of biarticular muscles. J Biomech 1996;29(4):513–23.

[51] Prilutsky BI, Zatsiorsky VM. Tendon action of two-joint muscles: transfer of mechanical energy between joints during jumping, landing, and running. J Biomech 1994;27(1):25–34.

[52] Robertson GE, Winter DA. Mechanical energy generation, absorption and transfer amongst segments during walking. J Biomech 1980;13:845–54.

[53] Martin PE, Morgan DW. Biomechanical considerations for economical walking and running. Med Sci Sports Exer 1992;24(4):407–74.

[54] Cavanagh PR, Williams KR. The effect of stride length variation on oxygen uptake during distance running. Med Sci Sports Exer 1982;14:30–5.

[55] Hill AV. The maximum work and mechanical efficiency of human muscles, and their most economical speed. J Physiol 1922;56:19–41.

[56] Shorten MR, Wootton SA, Williams C. Mechanical energy changes and the oxygen cost of running. Engin Med 1981;10:213–7.

[57] Kram R, Taylor CR. Energetics of running: a new perspective. Nature 1990;346:265–86.

[58] Czerniecki JM. Foot and ankle biomechanics in walking and running. A review. Am J Phys Med Rehab 1988;67(6):246–52.

[59] Chan CW, Rudine A. Foot biomechanics during walking and running. Mayo Clin Proc 1994;69(5):448–61.

[60] Rodgers MM. Dynamic biomechanics of the normal foot and ankle during walking and running. Phys Ther 1988;68(12):1822–30.

[61] Mann RA. Biomechanics of the foot and ankle. Mann RA, editor. Surgery of the Foot, 5th Ed. CV Mosby, 1986;1:1–30.

[62] Abuzzahab FS Jr, Harris GF, Kidder SM. A kinetic model of the foot and ankle. Abstract presented at North American Society of Gait and Clinic Movement Analysis, 2nd Annual Meeting, Chicago, IL, 9–12 April 1997.

[63] Abu-Faraj ZO, Sampath G, Smith PA. A clinical system for analysis of pediatric foot and ankle motion. Abstract presented at North American Society of Gait and Clinical Movement Analysis 2nd Annual Meeting, Chicago, IL, 9–12 April 1997.

[64] Areblad J, Nigg BM, Ekstrand J, Olsson KO, Ekstrom H. Three-dimensional measurement of rearfoot motion during running. J Biomech 1990;23(9):933–40.

[65] Scott SH, Winter DA. Talocrural and talocalcaneal joint kinematics and kinetics during the stance phase of walking. J Biomech 1991;24(8):743–52.

[66] Engsberg JR, Andrews JG. Kinematic analysis of the talocalcaneal/talocrural joint during running support. Med Sci Sports Exer 1987;19(3):275–84.

[67] Soutas-Little RW, Beavis GC, Verstraete MC, Markus TL. Analysis of foot motion during running using a joint co-ordinate system. Med Sci Sports Exer 1987;19(3):285–93.

[68] Clarke TE, Frederick EC, Hammill CL. The study of rearfoot movement in running. Frederick, EC, editor. Sport Shoes and Playing Surfaces. Champaign, IL: Human Kinetics, 1984:166–189.

[69] Hennig EM, Cavanagh PR, MacMillian NH. A piezoelectric method of measuring the vertical contact stress beneath the human foot. J Biomed Eng 1982;4(3):213–22.

[70] Cavanagh PR. The Running Shoe Book, Anderson World, California, 1980.

[71] Nigg BM, Denoth J, Luethi S, Stacoff A. Methodological aspects of sport shoe and sport floor analysis. Presented at 8th International Congress on Biomechanics, Nagoya, 20–24 July 1981.

[72] Nigg BM. External force measurements with sport shoes and playing surfaces. Nigg DM, Kerr BA, editors. Biomechanical Aspects of Sport Shoes and Playing Surfaces. University of Calgary, Canada, 1983:11.

[73] Bates BT, Osternig LR, Sawhill JA, James SL. An assessment of subject variability, subject-shoe interaction and the evaluation of running shoes using ground reaction force data. J Biomech 1983;16:181.

[74] Bates BT, Osternig LR, Sawhill JA, James SL. Design of running shoes. Proceedings of the International Conference on Medical Devices Sports Equipment. Shoup and Thacker, editors. ASME, New York, 1980.

[75] Clarke TE, Frederick EC, Cooper LB. Biomechanical measurement of running shoe cushioning properties. Nigg DM, Kerr BA, editors. Biomechanical Aspects of Sport Shoes and Playing Surfaces. University of Calgary, Canada, 1983:25.

[76] Pink MM, Jobe FW. The foot/shoe interface. In: Guten GN, editor. Running Injuries. Philadelphia, PA: W.B. Saunders, 1997;3:20–29.

[77] Cook SD, Kester MA, Brunet ME, Haddad RJ Jr. Biomechanics of running shoe performance. Clin Sports Med 1985;4(4):619–26.

[78] Shorten MR. The energetics of running and running shoes. J Biomech 1993;26(1):41–51.

[79] Winter DA. Moments of force and mechanical power in jogging. J Biomech 1983;16(1):91–7.

[80] Scott SH, Winter DA. Internal forces at chronic running injury sites. Med Sci Sports Exer 1989;22(3):357–69.

[81] Orchard JW, Fricker PA, Abud AT, Mason BR. Biomechanics of iliotibial band fraction syndrome in runners. Am J Sports Med 1996;24(3):375–9.

[82] Kibler WB, Goldberg C, Chandler TJ. Functional biomechanical deficits in running athletes with plantar fasciitis. Am J Sports Med 1991;19(1):66–71.

[83] Nigg BM, Bobbert M. On the potential of various approaches in load analysis to reduce the frequency of sports injuries. J Biomech 1990;3(Suppl 1):3–12.