Dynamic Foot Biomechanics

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Since clinicians often encounter patients with symptomatic feet, a basic understanding of "normal" foot function during movement is important for comparison and to assist with treatment strategies. This article provides a selected summary of quantitative information and current research findings relevant to the dynamic function of the foot. Functional descriptions of walking and running biomechanics are provided along with quantitative findings from current biomechanical studies. A classical description of the biomechanics of gait as found in clinical literature is followed by an overview of quantitative findings which document kinematic and kinetic characteristics during walking. The foot kinematics and kinetics which occur during running are presented in the final section. Extensive databases are still not available for most of the biomechanical parameters which affect foot motion. However, as advances in biomechanical methods continue and more clinicians include quantitative techniques in their routine evaluations, more insight into dynamic foot function will be provided.

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The foot is in a location that requires it to form a vital dynamic connection between the human body and the ground. All upright locomotion accomplished by the human being requires the foot to continually adapt to and couple with the body's surroundings. Traditionally, the dynamic attributes of the foot have been inferred from cadaveric study and qualitative clinical evaluation. Biomechanical methods for dynamic analysis have advanced, providing a more quantitative and precise description of foot function during movement, especially during walking.

In this article, static biomechanics of the foot introduces a framework for a selected summary of quantitative information relevant to the dynamic function of the foot. Of all the activities which require foot movement, the most frequently performed activity for healthy people is walking. Much research has been conducted in the analysis of walking, and the majority of this article will concentrate on dynamic foot biomechanics during this activity. A classical description of lower extremity biomechanics during walking as found in clinical literature is followed by an overview of quantitative findings which document kinematic and kinetic characteristics of the foot during walking. The final section presents foot kinematics and kinetics which occur during another common activity—running.

STATIC BIOMECHANICS OF THE FOOT

In order to put the dynamic characteristics of the foot in context, the basic biomechanical characteristics of the foot under static, or nonambulatory, conditions will be summarized. When the foot is free to move and the leg is fixed (open kinetic chain), the subtalar joint is a gliding joint in supination. The calcaneus moves anteriorly, inferiorly, and medially under the talus. During pronation, it moves posteriorly, superiorly, and laterally (45). The talus can be palpated laterally during supination as the calcaneus moves medially, and palpated medially during pronation as the calcaneus moves laterally. The subtalar joint is a hinge or uniaxial joint with an axis running downward, posteriorly, and laterally (24). This joint orientation results in triplanar motion, providing pronation and supination. Inman has reported a wide variation in the sagittal orientation of the subtalar joint with respect to the horizontal plane (45 ± 22°) and its rotation from the long axis of the foot (ranging from 4 to 47°) (24). The subtalar joint contributes more to inversion/eversion the more closely its axis is aligned with the longitudinal axis of the foot, and contributes more to abduction/adduction the more closely its axis is aligned with the longitudinal axis of the leg. The subtalar joint orientation also affects motion in the transverse view, providing more eversion/inversion and less dorsiflexion/plantar flexion as the subtalar joint axis approaches the longitudinal axis of the foot.

The transverse tarsal joint connects the hindfoot (talus and calcaneus) and midfoot (navicular and cuboid) in plane or gliding joints. It

amplifies and is interdependent with motions of the ankle joint and subtalar joint. As the subtalar joint moves into pronation, the transverse tarsal joint is pulled toward pronation resulting in flattening of the medial longitudinal arch and increasing flexibility of the foot. As the subtalar joint moves into supination, the transverse tarsal joint is pulled toward supination, resulting in elevation of the arch and increasing rigidity of the foot. Tarsometarsal joints are gliding joints, providing motion which is translatory or planar. The first ray connects the first metatarsal and medial cuneiform bones. The axis of motion is uniaxial and triplanar, combining dorsiflexion and inversion or plantar flexion and eversion with little contribution to abduction or adduction. Rays two through four provide flexion extension motion while the fifth ray allows pronation and supination to occur between the metatarsal and the cuboid. The metatarsophalangeal joints are biaxial, providing pure dorsiflexion/plantar flexion and abduction/adduction. The interphalangeal joints are hinge joints, permitting pure flexion/extension.

When the foot is weight bearing, the foot functions in a closed kinetic chain. Movement of the foot will cause the tibia and fibula to move, forcing the femur into rotation unless the twisting force or torque is absorbed at the knee joint. When the triplanar motion of pronation and supination occurs with the foot fixed, the leg inwardly rotates, medially deviates, and slightly inclines the leg forward. Supination produces the opposite results.

Along with variation in the location of different joint axes is variation in the static varus/valgus position of the forefoot with respect to the rearfoot. In an investigation of 234 asymptomatic feet, 86.7% were in varus, 8.8% were in valgus, and 4.6% demonstrated a neutral forefoot-rearfoot relationship (20). No significant differences were found in the varus/valgus position between genders or legs. Passive motion ranges for the different joints of the foot are summarized in Table 1. The static characteristics of each individual foot will influence its dynamic response to walking and running.

Joint	Motion	Range (degrees)
Subtalar	Inversion	5–50
	Eversion	5–26
	Total	10–65 ($\bar{X}$ = 40°)
Transverse tarsal	Inversion	0–20
	Eversion	0–10
Metatarsophalangeal	Maximum extension	40–90
	Maximum flexion	40–50
Interphalangeal	Maximum flexion	30–90

TABLE 1. *Passive motion ranges for joints of the foot (24,44).*

WALKING—FOOT KINEMATICS

The foot has been traditionally viewed as a static tripod or a semirigid support for body weight. But, since it has evolved primarily for walking, it is very definitely a dynamic mechanism. Three basic dynamic functions are required of the foot during walking. The body requires a flexible foot to accommodate the variations in the external environment—a semirigid foot which can act as a spring and lever arm for the push-off during gait and a rigid foot to enable body weight to be carried with adequate stability. The role of the foot complex in successful performance of all these requirements can only be understood when studied in relation to the biomechanics of the lower limb during walking.

Generalized Description

Basic to the discussion of walking kinematics is an understanding of the various lower limb joint axes and their motions. Many authors have provided a clinical description of walking kinematics based primarily on observation. Starting at the pelvis and progressing down the leg, these walking kinematics are described as follows. During walking, pelvic rotation causes the femur, fibula, and tibia to rotate about the long axis of the limb (25). The magnitude of this rotational motion increases progressively from pelvis to tibia. For example, during normal walking on level ground, the pelvis undergoes a maximum rotation in each gait cycle of about 6°, while the tibia undergoes a rotation of about 18° in the same period (26). The lower limb generally rotates internally during the swing phase and early stance phase and then externally until the stance phase is complete and toe-off has occurred (34).

At heelstrike, the tibia is rotated internally about 5° from its neutral position, and the ankle joint is either in its neutral position or in slight plantar flexion (60). Compression of the heel pad occurs at heelstrike, followed by traction on both anterior and posterior calcaneal attachments during terminal stance (47). The foot continues toward the floor immediately following heelstrike, with the dorsiflexors controlling this plantar flexion movement to prevent the foot from slapping down to the foot flat position (a condition called foot drop). From heelstrike to just prior to foot flat, the increasing inward rotation of the tibia and fibula is transmitted through the ankle mortise to the talus (37). Inward rotation of the ankle mortise combined with the plantar flexed position of the ankle tends to shift the forefoot medially from its neutral, toe-out position. Heel contact with the ground is lat-

eral to the center of the ankle joint where body weight is transmitted to the talus, creating a pronatory moment at the subtalar joint which, in turn, stresses the structures of the medial arch. The talus rotates medially on the calcaneus about the subtalar axis, forcing the calcaneus into a pronated position. The foot quickly pronates, moving approximately 10° within the first 8% of stance at an average walking speed (71). In this pronated position, free motion is available at the transverse tarsal joint so that the foot remains flexible, distal to the navicular and cuboid, and can bend into close contact with the supporting surface. At the foot flat position, the lower limb begins to rotate externally. Since the forefoot is now fixed on the ground, the entire external rotation of the ankle mortise is transmitted to the talus. As external rotation continues, the foot supinates, producing increased stability at the transverse tarsal joint and along the longitudinal arch of the foot. Transverse tarsal joint stability is further improved by the increasing body load being carried and by the firm fit of the convex head of the talus into the concave face of the navicular bone (37).

When the leg has passed over the foot, ankle dorsiflexion is initiated. After heel rise, the ankle joint moves back into plantar flexion, forcing the metatarsophalangeal joints to dorsiflex. Since the plantar aponeurosis wraps around the metatarsal heads, a "windlass" effect takes place which increases tension across the longitudinal arch, further elevating the arch and increasing foot stability. Just before toe-off, the combination of weight bearing, windlass effect, and supination ensures that the foot is in a maximally stable position for push-off (14). After toe-off, the leg rotates medially, once again pronating the foot and unlocking the transverse tarsal joint so that the foot returns to its flexible state for the swing phase of gait.

Kinematic Studies

Variables which describe motion, independent of the forces which cause the movement to take place, are referred to as kinematics. Linear and angular displacements, velocities, acceleration, center of rotation for joints, and joint angles are all examples of kinematic variables (53). Kinematic information can be collected using direct measurement techniques (ie., goniometers, accelerometers) and/or with indirect measurement using imaging techniques (ie., cinematography, high speed video, and stroboscopy). Advantages and disadvantages of the different techniques for the study of walking have been described by several authors and will not be detailed in this discussion (66, 72). Instead, results of selected studies relevant to dynamic biomechanics of the foot will be presented.

Walking Cadence and Velocity

Many factors affect foot biomechanics during walking, including walking speed and anthropometric characteristics of the individual (ie., limb length). Natural or free cadence is defined as the number of steps per minute (step/min) when a subject walks as naturally as possible. Average natural cadence ranges from 101 to 122 step/min, with females generally having a cadence 6–9 step/min higher than that of males, probably related to differences in limb length (68). Many studies have documented the changes in foot biomechanics that occur with increasing walking speed (2,67). For this reason, walking velocity must be considered when comparing biomechanical findings among studies. A classification scale for a range of walking velocities for clinical reporting has been suggested by Smidt and is shown in Table 2 (58).

Walking velocity varies with age (23). Kadaba et al report mean walking velocity for young adults of 1.34 ± 0.22 m/sec for men and 1.27 ± 0.16 m/sec for women (30). Normal velocity for a fit person at the age of 80 years is 1.0–1.2 m/sec, which represents a 10–20% decline from the value for younger persons (63). Walking velocity in children has been reported by Ounpuu et al to be 1.19 ± 0.13 m/sec for 31 children (46). Based on the findings of Wheelwright et al, walking velocity also varies from left to right leg (64). In 134 children, left and right swing time and maximum foot velocity differed 8–10%, and double support time following left stride was significantly different from right stride. The authors conclude that some degree of asymmetry exists in normal childhood walking gait, perhaps based on limb dominance.

Rate	Velocity (m/sec)
Very slow	≤0.40
Slow	0.41–0.70
Slow to moderate	0.71–1.00
Moderate	1.01–1.30
Moderate to fast	1.31–1.60
Fast	1.61–1.90
Very fast	>1.90

TABLE 2. *Rating of walking speed (58).*

Displacements

Linear heel movement during walking has been reported by Winter in a study with 14 subjects walking at their natural cadences (68). Vertical displacement of the heel begins well before toe-off and reaches maximum upward velocity just prior to toe-off. The heel reaches its highest displacement shortly after toe-off. Horizontal velocity builds up gradually after heel rise, reaching a maximum level late in the swing phase, and then rapidly decreases just prior to heel contact. Vertical velocity of the heel slows abruptly at about 1 cm above ground level, after which the heel is lowered very gently to the ground.

The forefoot movement during walking differs from that of the heel. An initial rise in the forefoot during

late push-off and early swing phase has been reported by Winter (68). As the leg and foot are swung forward, the forefoot just clears the ground and then rises to a second peak just prior to heel contact. Since the toe is the last to leave the ground, and because of the accompanying leg and foot angles, the toe rises to no more than 2.5 cm above the ground and then drops to only 0.87 cm clearance at midswing. As the knee extends and the foot dorsiflexes, the toe rises to a maximum of 13 cm just prior to heel contact.

Foot Placement and Arch Movement

Foot placement angles and arch movement are other kinematic characteristics that have been investigated. Foot placement varies substantially on successive steps of the same foot (41). A mean value of 6.8° of foot abduction has been reported, with the average difference between successive foot angles being 2.4°. Redfern and Schumann modelled foot placement during gait to investigate how balance is affected (51). They found that foot placements are dependent upon location of the stance foot with respect to the pelvis in order to help maintain balance during gait.

The medial longitudinal arch was found to lengthen from early stance to foot flat due to the vertical force of body weight.

In a study of dynamic arch movement using an "electro arch gauge," the medial longitudinal arch was found to lengthen from early stance to foot flat due to the vertical force of body weight (33). It then shortens with the decrease in body weight and activation of the arch supporting muscles. As the posterior calf muscles activate for push-off, the arch lengthens again. It finally shortens rapidly because of the windlass action of the plantar aponeurosis as the toes dorsiflex for toe-off. These findings were supported by a biomechanical model of the foot during the stance phase of walking (55). Using this model, Scott and Winter found that the joints that constitute the longitudinal arch extend slightly when the forefoot is loaded. These joints flex as the metatarsophalangeal joints extend during push-off.

FOOT KINETICS DURING WALKING

The study of those forces that cause movement, both internally (muscle activity, ligaments, friction in muscles and joints) and externally (from the ground, active bodies or passive bodies), is referred to as kinetics. Although a large number of researchers have analyzed muscular activity and ground reaction forces during gait, other kinetic parameters such as joint moments, segmental energy, joint reaction, and pressure distribution beneath the foot during walking have received less attention. The findings from electromyography (EMG) studies of the foot muscles during walking will be presented in the first subsection, followed by findings from force plate and pressure distribution studies. Calculated kinetic parameters, such as joint reaction forces, will be included in the final subsection.

Foot Muscle Activity During Walking

Many researchers have investigated the EMG activity of lower extremity muscles during walking. Studies have shown that many of the changes in levels of EMG activity occur at 15–20% of the cycle, when the foot adapts to the supporting surface (3). The tibialis anterior has its major activity at the end of swing to keep the foot in a dorsiflexed position (70). Immediately after heel contact, the tibialis anterior peaks and generates forces to lower the foot to the ground in opposition to the plantar flexing ground reaction forces. The tibialis anterior is the only inverting muscle active during the period of maximum everting stress, when body weight is completely on the heel. In some individuals, the tibialis anterior plays a minor role in pulling the leg forward over the foot shortly after foot flat. A second burst of tibialis anterior activity commences the toe-off and results in dorsiflexion for foot clearance during midswing. The extensor digitorum has almost identical activity to the tibialis anterior. It functions to lower the foot after heel contact and to dorsiflex the foot and toes for clearance during swing. A minor third phase is seen during push-off and appears to be a cocontraction to stabilize the ankle joint.

The gastrocnemius and soleus show a significant phase of activity, evident throughout the single limb support period. It begins just prior to heel contact and rises during stance, reaching peak just before midpush-off (50% stride). From foot flat to 40% stride, the muscles lengthen as the leg rotates forward about the ankle under its control. During push-off, the posterior calf muscles shorten to actively plantar flex the foot and to generate an explosive push-off (estimated at 250% of body weight in tension). Activity rapidly drops until toe-off, where low-level gastrocnemius activity continues into swing, probably showing the gastrocnemius acting as a knee flexor to cause adequate knee flexion prior to swing-through.

The peroneus longus has a small burst of activity during weight acceptance (10% of stride) which appears to stabilize the ankle (possibly as a cocontraction to the tibialis anterior). A larger burst during push-off (50%) shows the peroneus longus acting as a plantar flexor. Low level peroneus

longus activity during early swing is likely a cocontraction to the tibialis anterior to control the amount of foot dorsiflexion and supination.

Other investigators have reported their findings of intrinsic muscle activity in the foot during walking (4, 37). The group of intrinsic muscles covered by the plantar fascia (flexor digitorum brevis, abductor hallucis, and abductor digiti minimi) were shown to be active at 35% of the gait cycle. This part of the gait cycle includes the onset of heel rise, the concentration of body weight on the forefoot, and the beginning of foot resupination. The intrinsic muscle activity assists the foot in becoming more rigid as it resupinates.

Ground Reaction Forces

Gait laboratories usually have force plates, and ground reaction forces are one of the most commonly measured biomechanical parameters in gait. Ground reaction forces show the magnitude and direction of loading applied to the foot structures during locomotion. Since the feet are the first part of the body to contact the ground during walking, they must be able to withstand and transmit these ground reaction forces. Ground reaction force data also provide information necessary for the calculation of ankle joint reaction forces, which will be discussed later.

The magnitude of vertical ground reaction force has been reported to range from 1.1 to 1.3 times body weight depending upon walking speed (13). At termination of gait, push-off force diminishes and braking force increases (28). Footwear has been shown to attenuate the peak vertical ground reaction force values. A rapid loading rate, often seen in vertical ground reaction force during the first 25 msec after contact, has been described as a possible contributing factor in knee joint degeneration because of transmission of impulse through the tibia to the knee joint (50).

The force plate provides only one instantaneous measure of force distribution. This measure, called the center of pressure, identifies the geometric centroid of the applied force distribution. The center of pressure path is created by plotting the instantaneous center of pressure at regular time intervals during the entire stance phase of gait. Studies of the center of pressure show a normal progression of the path from just slightly lateral to the midline of the heel, along the midline of the foot up to the metatarsal heads (13,32). At this point, medial migration occurs so that by toe-off, the center of pressure lies under the first or second toe. This medial migration aspect of the center of pressure path has been described as the most variable among subjects. The center of pressure path is altered by different footwear and foot positions (ie., primarily supinated or pronated during stance phase).

Pressure Distribution Studies

Since the force information provided by force plate systems is not specific to foot anatomical location, the information is limited in the analysis of foot movement. For example, when forces recorded occur underneath both the fore and rear part of the foot simultaneously, the center of pressure may fall at some intermediate point which is not actually loaded. Information about the specific location of pressures as they occur beneath the moving foot is provided by pressure distribution systems. An increasing number of studies in pressure distribution have revealed new information regarding dynamic foot function during walking.

Although a great deal of individual variability exists in foot pressures during walking, the usual location of peak pressure is beneath the heel. A comparison of mean regional peak pressures found by several different investigators is summarized in the Figure. Differences in values reported result from the variety of techniques and subject samples utilized by investigators (52). These pressure studies have shown that all metatarsal heads are loaded during the stance phase of gait. This finding negates the classical concept of tripod stance, which would not allow pressure beneath the middle metatarsal heads. Pressures during the first step have been shown to be similar to midgait pressures. The pressures recorded for 60 male subjects, 40–81 years of age, during first step and during midgait walking showed significant differences under the toes, under the first metatarsal head, and in the heel regions between the two conditions. This means that for most of the foot, the pressures which occur during walking are similar to those present at the first step. The implications for the work are that patients who are unable to walk the distances required for normal pressure measurement, such as those with balance problems, can still be measured using their first step if regional conversions are utilized.

Pressure distribution beneath the foot is affected by a number of different variables. With increasing speed, pressures increase and shift medially, and the toes contribute more by assuming more loading (15). Walking barefoot alters both kinetic and kinematic parameters when compared with walking in shoes. Structural characteristics of the foot, such as arch type, also affect pressure distribution. The more rigid, high-arched foot tends to concentrate pressure beneath the heel and forefoot, with minimal pressure beneath the midfoot (12). This absence of midfoot pressure is present even in the higher loading conditions that occur with increasing speed of locomotion. The flexible flat-arched foot shows more spreading of pressure, including the area beneath the midfoot.

The classic Morton foot structure, characterized by a second metatarsal head which is more distally placed than the first, has also been shown to influence pressure distribution (54). Rodgers and Cavanagh re-

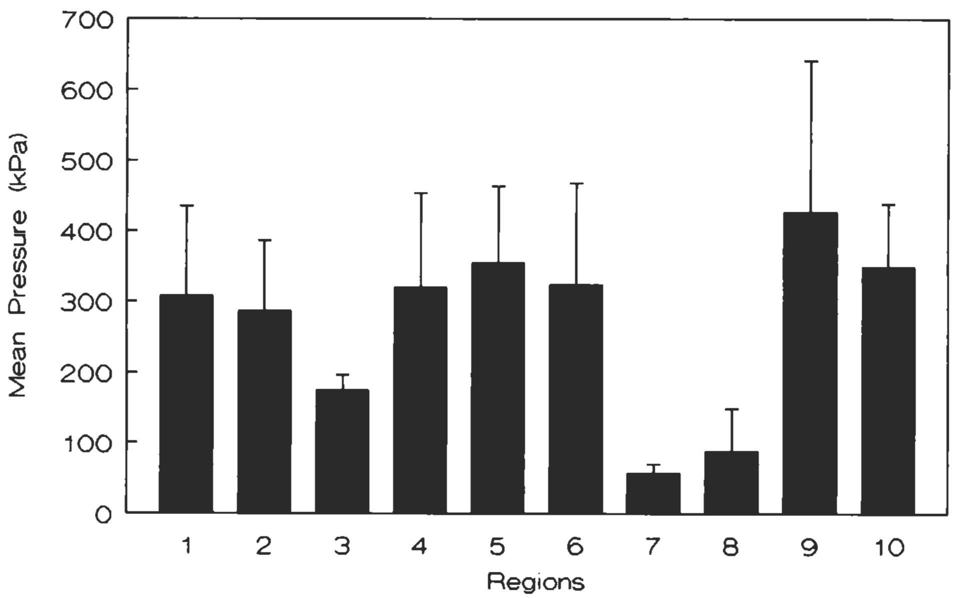

FIGURE. *Means (and standard deviations) for pressures under the foot during walking based on pressure studies (6,15,21,52,59). The regions are hallux (1), medial toes (2), lateral toes (3), first metatarsal (4), second metatarsal (5), lateral metatarsal (6), medial midfoot (7), lateral midfoot (8), medial heel (9), and lateral heel (10).*

ported second metatarsal head pressures which were significantly higher in Morton foot subjects when compared with nonMorton control subjects. This finding suggests that individuals with a Morton foot structure may be more prone to second metatarsal pressure problems (ie., stress fractures or inflammation) than other foot structures. Pressure studies have also been useful in identifying areas of concentrated pressure which may lead to pressure ulcers for individuals with insensitive feet (10). The utilization of pressure studies to screen patients at risk because of foot structure or disease presents a direct clinical application of the biomechanical technique. The relationship of pressure distribution to static radiographic foot measures has been explored by Morag et al (40). In diabetic patients with a history of ulceration, the researchers found that a low Morton's index (head of metatarsal head 2 less distal to metatarsal head 1), a lower sesamoid height (less plantar soft tissue), a small inclination angle of the first phalanx, and a large deviation of the fibular sesamoid from the midline were features which tended to be associated with higher metatarsal head pressures.

Joint Moments/Joint Reaction Forces

Indirect methods, such as modeling, have been used to calculate gait kinetics when direct methods are not feasible. These methods are necessary to calculate forces within the joint since force transducers cannot presently be safely used in subjects. Scott and Winter (55), Winter (68), and Winter and Robertson (69) have made significant contributions in the calculation of joint moments of force and energy patterns during walking. The mean maximum ankle joint moment (normalized to body mass) generated during walking was found to be a plantar moment of 1.6 Nm/kg, occurring between 40 and 60% of the gait cycle. Plantar flexors were found to absorb energy during early and midstance phase as the leg rotates over the foot. Late in stance, these same muscles plantar flex rapidly (producing the plantar moment) and generate an explosive burst of energy (push-off). The joints that constitute the longitudinal arch extend slightly when the forefoot is loaded. During push-off, these joints flex as the metatarsophalangeal joints extend. A model was used to calculate the magnitude of these joint moments of force for a small sample ($N = 3$) of subjects (55). The magnitude of the moments depended largely on the distribution of the load under the foot which varied considerably for the small sample of subjects tested.

Nashman and Vaughan reported on the ankle joint moments during walking in 10 subjects (42). They found that the plantar flexion-dorsiflexion moment was characterized by a small dorsiflexor moment at heel contact followed by a plantar flexor moment increasing from foot flat to a peak during push-off. This moment was consistent both within individuals and also across the subject population. They also found that the inversion-eversion moment was consistent within individuals, but across subjects was variable.

As mentioned in the section on force plate studies, the ground reaction forces during gait are transmitted proximally to the rest of the body through the foot, compressing each joint along the way. These compressive forces have been shown by Radin et al to contribute to the formation of osteoarthrosis (49,50). Joint reac-

Compressive forces have been shown to contribute to the formation of osteoarthrosis.

tion studies of the ankle have been few, probably because this joint demonstrates osteoarthritic changes less often than the hip and knee joints. Stauffer et al have shown ankle joint compressive forces of approximately

3 times body weight from heelstrike to foot flat (62). A further rise to a peak value of 4.5 to 5.5 times body weight occurs during heel-off when the plantar flexors are undergoing strong contraction. Seireg and Arvikar have derived maximal ankle joint reaction forces of 5.2 times body weight from mathematical models (57). Procter and Paul found a peak of 3.9 times body weight for ankle joint reaction force during walking (48).

Stauffer et al also reported ankle shear forces of 0.6 times body weight in a posterior direction (62). After heel rise, talocrural shear forces were anterior and reduced to less than half of the previous posterior forces. Subtalar joint reaction forces have been calculated by Seireg and Arvikar (57). The peak resultant force in the anterior facet of the talocalcaneonavicular joint was 2.4 times body weight and 2.8 times body weight for the posterior facet. Peaks for both locations occurred in late stance phase of the gait cycle.

FOOT KINEMATICS DURING RUNNING

A considerable amount of research has been conducted in the area of running biomechanics and is presented in a detailed review by Williams (65). The position of other body parts and the timing of their movements are basic to an understanding of the motion of the foot. Although other body parts (primarily the hip and knee) have received most of the attention, several investigators have contributed to a functional description specific to foot motions during running at moderate speeds (5,36).

Generalized Description

For the running gait in which heel strike occurs, initial contact is at the lateral heel with the foot slightly supinated. This position results from swinging the leg toward the line of progression. Slight plantar flexion of the subtalar joint occurs along with a supinated to a pronated position between heelstrike and 20% into support phase. The foot remains pronated between 55 and 85% of the support phase. Maximum pronation occurs between 35 and 40% of support phase, approximately the time when total body center of gravity passes over the base of support. Full pronation marks the end of the absorbing and braking period of support as the foot begins its propulsive period. Maximum ankle dorsiflexion occurs 50–55% into support phase when the center of gravity is forward of the support leg. The foot begins to supinate and returns to a neutral position from 70 to 90% of the support phase. The foot then assumes a supinated position for push-off.

Kinematic Studies

Several stride variables which directly affect running kinematics and kinetics have been described by Cavanagh (8). These include stride length at different speeds, optimal stride length, timing of the phases of running gait, and foot placement. Timing of the biomechanical events in running is variable since it depends on running speed, type of shoe, and individual anatomic variations. For example, Kaelin et al reported interindividual ($N = 70$) and intraindividual variabilities (20 repetitions each for six subjects) for several parameters during running (31). The maximum pronation angle during foot-ground contact showed a range of 20° among the 70 subjects, but only 7 to 12° within the same individual. Vertical touchdown velocity of the foot during running varied between 0.64 and 2.3 m/sec for the 70 subjects. Scranton et al reported an average duration of support phase for jogging of 0.2 sec and for sprinting of 0.1 sec (56).

Clinical evaluations have suggested a relationship between pronation of the foot during running and a variety of lower extremity problems, such as shin splints and knee pain. At present, quantitative data do not support the relationship, although this may result from inadequate analytical techniques. For example, studies of rearfoot motion have been conducted in two-dimensions, although pronation occurs in more than one plane. McClay and Manal found no difference between two-dimensional and three-dimensional peak eversion angles during running (38). However, foot abduction angle did influence the two-dimensional analysis of eversion. Clarke et al have reviewed several different studies of rearfoot movement in running (16). They reported an average maximum pronation angle of 9.4° over all studies. The authors suggest that a maximum pronation angle of 13° and total rearfoot motion greater than 19° during running would be considered excessive. However, at present, no single parameter reliably predicts safe rearfoot movement during running.

Several investigators have studied the effects of footwear and orthotics on pronation. Stacoff et al reported on the effect of running shoes and spikes on the shoe/foot motion in running (61). They found that, at touchdown, the torsional movements with both shoe types are quite different from those of running barefoot. With shoes, the torsion angle is reduced back to zero (more so with running shoes than with spikes), and the pronation angle is increased beyond the barefoot values. The authors suggest improvement of running shoes with respect to torsion and the spikes with respect to pronation. Eng and Pierrynowski found reductions of 1–3° in the talocrural/subtalar joint motion (frontal and transverse planes) with soft foot orthotics (18).

Nigg et al have investigated the relationship of arch height to angular motion of the leg in running (44). They found that arch height does not influence either maximal eversion movement or maximal internal leg

rotation during running stance. Their findings support the assumption that foot eversion is coupled to internal rotation of the tibia during the first half of running stance.

FOOT KINETICS DURING RUNNING

Direct measurement of running kinetics poses more difficult technical problems than during the slower speeds of walking gait. Targeting a force plate is more difficult at the higher speeds without altering the normal running gait patterns. The faster motion requires more distance for running and, therefore, longer cables or telemetry systems must be utilized for EMG data collection. Treadmill running has been used for EMG data collection, although the pattern of running is different from that seen over natural terrain or on a track. Because of these problems, few researchers have directly measured foot muscle activity during running. More research has been conducted in ground reaction forces and pressure distribution during running. Indirect calculations of foot muscle forces, segmental moments, and joint reaction forces during running have been performed by a few researchers.

Foot Muscle Activity During Running

Studies have shown that EMG activity increases with running as compared with walking. Miyashita et al have reported that integrated EMG activity of the tibialis anterior and gastrocnemius increases exponentially with increasing speed (39). Ito et al report that with increasing running speed, the integrated EMG increased during swing, but remained the same during the contact phase (27).

Ground Reaction Force Studies

Several authors have suggested a link between common running injuries and the impact forces at foot strike which can occur thousands of times during running (29,35). Force plate analysis has shown that peak vertical loading force during running is more than twice that of walking and occurs at least twice as fast. Perry extrapolates that the forces imposed on the supporting tissues would reflect a four-fold increase in strain (47). Since microtrauma is cumulative, running creates symptoms that do not arise with ordinary walking.

Force plate data for jogging and running are much more variable from step to step when compared with walking. The pattern and magnitude of the vertical ground reaction force during running also differ significantly from those which occur during walking. Variables that affect vertical ground reaction force data include touchdown velocity of the heel, position of the foot and lower leg before contact, and movement of these structures during impact (43). Two distinct peaks are usually present in the vertical ground reaction force curve for heel-toe running (heel strikers): the impact force peak and the active force peak (11,18,19). Typical peak vertical ground reaction force values for distance running speeds are 2.5–3.0 times body weight. DeClercq et al found that although the amplitudes of the vertical ground reaction forces do not differ notably in barefoot running compared with shod running, barefoot running imposes a maximal deformation to the fatty heel tissue, reducing its functional role from shock reduction toward local protection of the heel bone (17). The heel pad deforms to a maximal percentage deformation of 61 ± 6% in barefoot running, while running in shoes produces a deformation of 36 ± 3%. Shoes appear to increase the effective stiffness of the heel pad.

The pattern of force is dependent upon the orientation of the foot at initial contact, which is determined by whether the runner is a forefoot, midfoot, or rearfoot striker (11). Most runners initially contact the ground with the outside border of the shoe, while some make contact with the rear lateral shoe (rearfoot strikers). Harrison et al report that mean foot contact time is reduced in forefoot strikers as compared with rearfoot strikers (0.20 seconds compared with 0.19 seconds) (22). Cavanagh and Lafortune also found slightly shorter contact times for the midfoot strikers compared with the rearfoot strikers (11). In addition, rearfoot strikers demonstrate a sharp initial spike in vertical ground reaction force generally absent from the midfoot striker patterns. Midfoot strikers produced two positive peaks in the anterior/posterior force during the braking phase. The mean peak to peak amplitude for mediolateral ground reaction force was three times greater in the midfoot strikers than that for the rearfoot strikers (0.35 body weight and 0.12 body weight, respectively). These findings indicate that the loading rates within the muscles and joints are affected by the type of initial foot contact during running.

The path of the center of pressure also depends upon the type of initial foot contact during running. The center of pressure path for rearfoot strikers follows from the rear lateral border to the midline within 15 msec of contact (11). The center of pressure path then continues along the midline to the center of the forefoot where it remains for almost 2/3 of the entire 200 msec contact phase. Midfoot strikers running at the same running speed make initial contact at 50% of shoe length. The center of pressure path then migrates posteriorly as the rear part of the shoe makes contact with the ground. This posterior movement coincides with a drop in the anterior/posterior ground reaction force. When the end of posterior migration is reached, the center of pressure rapidly moves to the forefoot where it remains for most of the contact phase.

Pressure Distribution Studies

Very little information is available regarding pressure distribution under the foot during running. Cavanagh and Rodgers have shown that pressure patterns during running vary with foot type (ie., different arch heights or metatarsal lengths) (12). The increased loading which occurs with running remains concentrated under the heel and forefoot in the more rigid high-arched foot. In the more flexible flat-arched foot, the increased load is spread beneath the entire foot, including the midfoot region. Cavanagh and Hennig found that the average peak pressure during the contact phase of running (868 kPa) occurred under the heel for a sample of 10 rearfoot strikers (9). Although pressures were much higher beneath the heel of these rearfoot strikers, more of the contact time was spent on the forefoot.

Muscle Forces, Segmental Impulse, and Joint Reaction Forces

Several investigators have developed mathematical models to predict muscle forces during running. Forces generated by the dorsiflexors and the gastrocnemius have been calculated by Harrison et al (22). They report peak forces in the dorsiflexors of 0.52 times body weight, which are active only during the first 10% of the stance phase. The gastrocnemius generated a substantially greater peak force of 7.53 times body weight. Calculations by Burdett revealed that the gastrocsoleus group had the highest predicted force (5.3–10.0 times body weight) of the ankle muscle groups (7). Predicted forces in the tibialis posterior, flexor digitorum longus, and flexor hallucis longus group ranged from 4.0 to 5.3 times body weight. The peroneus tertius and extensor digitorum longus did not show any predicted force during the stance phase of running.

Impulse is the effect of a force acting over a period of time and is determined mathematically as the integral of the force-time curve. Ae et al calculated the impulse generated by different body segments during running (1). The researchers found that the foot generated the largest mean impulse compared with other body segments. This impulse increased with faster running, suggesting that the foot plays an important role in projecting the body and increasing running velocity.

The foot generated the largest mean impulse compared with other body segments.

Ankle joint reaction forces during running have also been calculated by several investigators. Harrison et al reported maximum ankle joint reaction of 8.97 and 4.15 times body weight for the compressive and shear components, respectively (22). Burdett predicted that compressive forces on the foot along the longitudinal axis of the leg reached peak values from 3.3 to 5.5 times body weight during running (7). In addition, he reported mediolateral shear forces which ranged from a medial force of 0.8 times body weight to a lateral force of 0.5 times body weight. Furthermore, the vertical reaction forces and other calculated forces were determined to be approximately 2.5 larger in running (at a 4.47 m/sec pace) when compared with walking.

SUMMARY

This article has described current findings related to the dynamic biomechanics of the asymptomatic foot during walking and running. Functional descriptions of walking and running biomechanics have been provided along with quantitative findings from current biomechanical studies. Extensive databases are still not available for most of the biomechanical parameters which affect dynamic foot motion. However, as advances in biomechanical methods continue and more clinicians include quantitative techniques in their routine evaluations, more insight into dynamic foot function will be provided. As our understanding improves, we can provide more effective therapeutic approaches to lower extremity injury and better design protective shoewear. We may eventually see computer-aided design of shoes and therapeutic exercise programs based on biomechanical measurements performed in the clinic. JOSPT

REFERENCES

1. *Ae M, Miyashita K, Yokoi T, Hashihara Y: Mechanical power and work done by the muscles of the lower limb during running at different speeds. In: Jonsson B (ed), Biomechanics X-B, pp 895–899. Champaign, IL: Human Kinetics Publishers, 1987*
2. *Andriacchi TP, Ogle JA, Galant JO: Walking speed as a basis for normal and abnormal gait measurements. J Biomech 10:261–268, 1977*
3. *Basmajian JV: Muscles Alive (5th Ed), Baltimore: Williams & Wilkins, 1985*
4. *Basmajian JV, Stecko G: The role of muscles in arch support of the foot. J Bone Joint Surg 45A:1184–1190, 1963*
5. *Bates BT, Osternig LR, Mason B: Lower extremity function during the support phase of running. In: Asmussen E, Jorgensen K (eds), Biomechanics VI, pp 31–39. Baltimore: University Park Press, 1978*
6. *Betts RP, Franks CI, Duckworth T: Analysis of pressures and loads under the foot. Part II: Quantification of the dynamic distribution. J Clin Phys Physiol Meas 1(2):113–124, 1980*
7. *Burdett RG: Forces predicted at the ankle during running. Med Sci Sports Exerc 14:308–316, 1982*
8. *Cavanagh PR: The biomechanics of lower extremity action in distance running. Foot Ankle 7:197–217, 1987*
9. *Cavanagh PR, Hennig EM: Pressure distribution measurement. A review and some new observations on the effect of shoe foam materials during running. In: Nigg BM, Kerr BA (eds), Biomechanical Aspects of Sports Shoes and Playing*

Surfaces, pp 187–190. Calgary: The University of Calgary Press, 1983
10. Cavanagh PR, Hennig EM, Rodgers MM, Sanderson DJ: The measurement of pressure distribution on the plantar surface of diabetic feet. In: Whittle M, Harris D (eds), Biomechanical Measurement in Orthopaedic Practice, pp 159–166. Oxford: Clarendon Press, 1985
11. Cavanagh PR, Lafortune MA: Ground reaction forces in distance running. J Biomech 13:397–406, 1980
12. Cavanagh PR, Rodgers MM: Pressure distribution underneath the human foot. In: Perren SM, Schneider E (eds), Biomechanics: Current Interdisciplinary Research, pp 85–95. Dordrecht, The Netherlands: Martinus Nijhoff Publishers, 1985
13. Cavanagh PR, Williams KR, Clarke TE: A comparison of ground reaction forces during walking barefoot and in shoes. In: Morecki A (ed), Biomechanics VII, Baltimore: University Park Press, 1981
14. Chan CW, Rudins A: Foot biomechanics during walking and running. Mayo Clin Proc 69:448–461, 1994
15. Clarke TE: The pressure distribution under the foot during barefoot walking. Unpublished doctoral dissertation, The Pennsylvania State University, University Park, PA, 1980
16. Clarke TE, Frederick EC, Hamill CL: The study of rearfoot movement in running. In: Frederick EC (ed), Sport Shoes and Playing Surfaces, Champaign, IL: Human Kinetics Publishers, 1984
17. DeClercq D, Aerts P, Kunnen M: The mechanical characteristics of the human heel pad during foot strike in running: An in vivo cineradiographic study. J Biomech 27:1213–1222, 1994
18. Eng JJ, Pierrynowski MR: The effect of soft foot orthotics on three-dimensional lower-limb kinematics during walking and running. Phys Ther 74:836–844, 1994
19. Frederick EC, Hagy JL, Mann RA: Prediction of vertical impact force during running. J Biomech 14:498, 1981 (abstract)
20. Garbalosa JC, McClure MH, Catlin PA, Wooden M: The frontal plane relationship of the forefoot to the rearfoot in an asymptomatic population. J Orthop Sports Phys Ther 20:200–206, 1994
21. Grieve DW, Rashdi T: Pressures under normal feet in standing and walking as measured by foil pedobarography. Ann Rheum Dis 43:816–818, 1984
22. Harrison RN, Lees A, McCullagh PJJ, Rowe WB: Bioengineering analysis of muscle and joint forces acting in the human leg during running. In: Jonsson B (ed), Biomechanics X-B, pp 855–861. Champaign, IL: Human Kinetics Publishers, 1987
23. Imms FJ, Edholm OG: Studies of gait and mobility in the elderly. Age Ageing 10:147–156, 1981
24. Inman VT: The Joints of the Ankle, Baltimore: Williams & Wilkins, 1976
25. Inman VT, Mann RA: Biomechanics of the foot and ankle. In: Inman VT, DuVries HL (eds), Surgery of the Foot, pp 3–22. St. Louis, MO: C.V. Mosby Company, 1973
26. Inman VT, Ralston HJ, Todd F: Human Walking, Baltimore: Williams & Wilkins, 1981
27. Ito A, Fuchimoto T, Kaneko M: Quantitative analysis of EMG during various speeds of running. In: Winter DA, Norman RW, Wells RP, Hayes KC, Patla AE (eds), Biomechanics IX-B, pp 301–306. Champaign, IL: Human Kinetics Publishers, 1985
28. Jaeger RJ, Vanitchatchavan P: Ground reaction forces during termination of human gait. J Biomech 25:1233–1236, 1992
29. James SL, Bates BT, Osternig LR: Injuries to runners. Am J Sports Med 6:40–50, 1978
30. Kadaba MP, Ramakrishnan HK, Wootten ME: Measurement of lower extremity kinematics during level walking. J Orthop Res 8:383–392, 1990
31. Kaelin X, Unold E, Stussi E, Stacoff A: Interindividual and intraindividual variabilities in running. In: Winter DA, Norman RW, Wells RP, Hayes KC, Patla AE (eds), Biomechanics IX-B, pp 356–360. Champaign, IL: Human Kinetics Publishers, 1985
32. Katoh Y, Chao EYS, Laughman RK, Schneider E, Morrey BF: Biomechanical analysis of foot function during gait and clinical applications. Clin Orthop 177:23–33, 1983
33. Kayano J: Dynamic function of medial foot arch. J Japan Orthop Assoc (Nippon Seikeigeka Gakkai Zasshi) 60: 1147–1156, 1986
34. Manley MT: Biomechanics of the foot. In: Helfet AJ, Lee DMG (eds), Disorders of the Foot, pp 21–30. Philadelphia: J.B. Lippincott Company, 1980
35. Mann RA: Biomechanics of the foot. In: American Academy of Orthopedic Surgeons (eds), Atlas of Orthotics: Biomechanical Principles and Application, pp 112–125. St. Louis: C.V. Mosby Company, 1985
36. Mann RA, Baxter DE, Lutter LD: Running symposium. Foot Ankle 1:190–224, 1981
37. Mann R, Inman V: Phasic activity of intrinsic muscles of the foot. J Bone Joint Surg 46A:469–481, 1964
38. McClay I, Manal K: A comparison of three-dimensional lower extremity kinematics during running between pronators and normals. Proc Am Soc Biomech 18:127–128, 1994
39. Miyashita M, Matsui H, Miura M: The relation between electrical activity in muscle and speed of walking and running. In: Vredenbregt J, Wartenweiler JW (eds), Biomechanics II, pp 192–196. Baltimore: University Park Press, 1971
40. Morag E, Boulton AJM, Young MJ, Galvin KT, Pammer SE, Cavanagh PR: Predicting function from structure: The relationship between plantar pressure and radiographic measures of the foot. Proc Am Soc Biomech 18:127–128, 1994
41. Murray MP, Kory RC, Sepic S: Walking patterns of normal women. Arch Phys Med Rehabil 51:637–650, 1970
42. Nashman JL, Vaughan CL: Factor analysis of inversion-eversion moments at the ankle during walking. J Biomech 27:811, 1993 (abstract)
43. Nigg BM: Biomechanical analysis of ankle and foot movement. Med Sci Sports Exerc 23:22–29, 1987
44. Nigg BM, Cole GK, Nachbauer W: Effects of arch height of the foot on angular motion of the lower extremities in running. J Biomech 26:909–916, 1993
45. Otis CA: Biomechanics of the foot and ankle under static conditions. Phys Ther 68:1815–1821, 1988
46. Ounpuu S, Gage JR, Davie RB: Three-dimensional lower extremity joint kinetics in normal pediatric gait. J Pediatr Orthop 11:341–349, 1991
47. Perry J: Anatomy and biomechanics of the hindfoot. Clin Orthop 177:9–15, 1983
48. Procter P, Paul JPL: Ankle joint biomechanics. J Biomech 15:627–634, 1982
49. Radin E, Boyd R, Martin B, Burr DB, Caterson B, Goodwin C: Mechanical factors influencing articular cartilage damage. In: Peryon JG (ed), Etiology of Osteroarthrosis, pp 92–101. Paris: CIBA-Geigy, 1985
50. Radin E, Whittle M, Yang KH, Jefferson RJ, Rodgers MM, Kish VL, O'Conner JJ: The heel strike transient, its relationship with the angular velocity of the shank, and the effects of quadriceps paralysis. In: Proceedings of the American Society of Mechanical Engineers (ASME) (Vol 2), pp 121–123. New York, NY: ASME, 1986
51. Redfern MS, Schumann T: A model of foot placement during gait. J Biomech 27:1339–1346, 1994
52. Rodgers MM: Plantar pressure distribu-

tion measurement during barefoot walking: Normal values and predictive equations. Unpublished doctoral dissertation, The Pennsylvania State University, University Park, PA, 1985

53. *Rodgers MM, Cavanagh PR: Glossary of biomechanical terms, concepts, and units. Phys Ther 64:1886–1902, 1984*
54. *Rodgers MM, Cavanagh PR: Pressure distribution in Morton's foot structure. Med Sci Sport Exerc 21(1):23–28, 1989*
55. *Scott SH, Winter DA: Biomechanical model of the human foot: Kinematics and kinetics during the stance phase of walking. J Biomech 26:1091–1104, 1993*
56. *Scranton PE, Rutkoski R, Brown TD: Support phase kinematics of the foot. In: Bateman JE, Trott A (eds), The Foot and Ankle, pp 195–205. New York: Thieme-Stratton, 1980*
57. *Seireg A, Arvikar RJ: The prediction of muscular load sharing and joint forces in the lower extremities during walking. J Biomech 8:89–102, 1975*
58. *Smidt GL: Rudiments of gait. In: Smidt GL (ed), Gait in Rehabilitation, pp 1–19. New York: Churchill Livingstone Inc., 1990*
59. *Soames R: Foot pressure patterns during gait. J Biomed Eng 7:120–126, 1985*
60. *Soderberg GL: Kinesiology: Application to Pathological Motion, Baltimore: Williams & Wilkins, 1986*
61. *Stacoff A, Kalin X, Stussi E: The effects of shoes on the torsion and rearfoot motion in running. Med Sci Sport Exerc 23:482–490, 1991*
62. *Stauffer RN, Chao EYS, Brester RC: Force and motion analysis of the normal, diseased, and prosthetic ankle joint. Clin Orthop 127:189–196, 1977*
63. *Sudarsky L: Geriatrics: Gait disorders in the elderly. N Engl J Med 322(20): 1441–1446, 1990*
64. *Wheelwright EF, Minns RA, Law HT, Elton RA: Temporal and spatial parameters of gait in children. I: Normal control data. Dev Med Child Neurol 35: 102–113, 1993*
65. *Williams KR: Biomechanics of running. In: Terjung RL (ed), Exercise and Sport Science Reviews, pp 389–442. New York: Macmillian Publishing, 1985*
66. *Winter DA: Biomechanics of Human Movement, New York: John Wiley & Sons Inc., 1979*
67. *Winter DA: Kinematic and kinetic patterns in human gait: Variability and compensation effects. Hum Mov Sci 3:51–56, 1984*
68. *Winter DA: The Biomechanics and Motor Control of Human Gait (2nd Ed), Waterloo, Canada: University of Waterloo Press, 1991*
69. *Winter DA, Robertson DGE: Joint torque and energy patterns in normal gait. Biol Cybern 29:137–142, 1978*
70. *Winter DA, Yack HJ: EMG profiles during normal human walking: Stride-to-stride and inter-subject variability. Electroencephalogr Clin Neurophysiol 67:402–411, 1987*
71. *Wright DG, Desai ME, Henderson BS: Action of the subtalar and ankle-joint complex during the stance phase of walking. J Bone Joint Surg 46A:361–382, 1964*
72. *Yack HJ: Techniques for clinical assessment of human movement. Phys Ther 64:1821–1830, 1984*