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COUGHLIN AND MANN'S
Surgery of the Foot and Ankle

10TH EDITION

VOLUME ONE

2-Volume Set

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PREFACE

As we bring you the tenth edition of this enduring text, we should reflect on its journey and past editors, since they are the foundation upon which our current understanding and practice of foot and ankle surgery rests. It has been over 60 years since the first edition of *Surgery of the Foot* was published in 1959. Written by Henri L. DuVries of Chicago, the book captured his unique perspective as a physician who initially trained as a Podiatrist and later obtained his Doctor of Medicine, as well as his 30 years of experience treating disorders, deformities, and injuries of the foot. It became a classic reference text in the treatment of common foot disorders.

The initial edition was revised in 1965 to include several contributors. Most notable were Verne T. Inman, MD, who was Chair of the Department of Orthopaedic Surgery at the University of California, San Francisco (UCSF), and Roger A. Mann, MD, who was senior resident at the time. The third edition, edited by Dr. Inman and published in 1973, expanded the content to include the ankle joint and incorporated Dr. Inman's interest in biomechanics of the foot and ankle.

Not long after, in 1978, Dr. Mann edited the fourth edition, *DuVries' Surgery of the Foot*. His unique exposure to Dr. Inman as a resident at UCSF and as a fellow under Dr. DuVries allowed him to blend the strengths of each: Dr. Inman's biomechanical basic science background and Dr. DuVries's extensive clinical experience. This was followed in 1986 by the fifth edition, also edited by Dr. Mann.

Dr. Michael J. Coughlin, Dr. Mann's first Foot and Ankle Fellow in 1978, joined Dr. Mann as editor of the sixth edition in 1993, with the name expanded to *Surgery of the Foot and Ankle*. Their combined commitment to excellence in patient care through meticulous surgical technique and exacting postoperative care, to continued improvement based on scientific study of pathoanatomy, indications, and outcomes, and to education through training multiple generations of fellows, shines through and lives on in the text. By this time, the knowledge base of foot and ankle surgery had grown, and the text was divided into two volumes. Following the model of prior editions, many contemporary experts in the field contributed to the work. Drs. Coughlin and Mann co-edited the seventh edition in 1999.

I was honored to be Dr. Mann's Foot and Ankle Fellow in 2002, one evening turning in my chief-resident pager at UCSF and the next morning picking up my fellow pager at Dr. Mann's office in Oakland, California, beginning the most educational and productive 6 months of my training. From Dr. Mann I learned to simplify clinical problems and apply reliable treatments. He approached all aspects of foot and ankle care with curiosity and honesty, all the while studying his own outcomes with the goal of discovering ways to improve patient care. It was an exciting time, and I was trusted to research and publish some of the first studies on three-component total ankle replacement to come out of the United States. It was also during this time I met Dr. Coughlin, while he, Dr. Mann, and other luminaries worked out the intricacies of introducing the STAR ankle replacement in the United States. Dr. Coughlin approaches problems in orthopedics with a rigor and clarity that both inspires creativity and encourages confidence to take on all clinical challenges.

In 2005, the eighth edition saw the addition of Dr. Charles L. Saltzman (prior fellow of Kenneth A. Johnson, MD) as co-editor, along with Drs. Coughlin and Mann. This edition included the colorization of figures and graphs. The ninth edition in 2014 honored Dr. Mann as editor emeritus and renamed the text *Mann's Surgery of the Foot and Ankle*. Dr. Robert B. Anderson (prior fellow of John Gould, MD) joined Drs. Coughlin and Saltzman as co-editors, and the text was complemented by numerous surgical videos, many adapted from the classic *Video Textbook of Foot and Ankle Surgery*.

The tenth edition stands on the shoulders of the prior nine iterations. Dr. Coughlin has continued his contribution to the work through planning, guidance, and recruitment, maintaining the direct lineage to Drs. Mann, Inman, and DuVries. In respect to his career-long contribution to the work, the tenth edition title is now *Coughlin and Mann's Surgery of the Foot and Ankle*. A new generation of authors complements returning experts in the field, with over 65 authors in all, covering all aspects of foot and ankle surgery. Topics are divided into 11 parts over two volumes: General Considerations, Forefoot and Heel, Integument, Tumors, Nerve Disorders, Arthritis, Tendon Disorders and Postural Malalignment, Diabetes and Infection, Sports Medicine, Pediatrics, and Trauma. New chapters include Scientific Evidence-Based Foot and Ankle Care, Hallux Varus and Complication of Bunion Repair, Complex Regional Pain Syndrome, Avascular Necrosis and Total Talus Replacement, and Osteochondral Lesions, rounding out a total of 47 chapters. Over one-hundred sixty surgical videos, with 25% new video content, provide a real-time complement to the text.

The text strives to be an up-to-date reference for foot and ankle surgeons in all levels of their career. New techniques and updated references are presented along with historical context, established reliable methods, and recommendations from experts in the field. We believe it will be both a quick reference for seasoned surgeons before a challenging case as well as a sound basis for resident and fellow foot and ankle education.

Andrew Haskell, MD

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PART I General Considerations

1

Biomechanics of the Foot and Ankle

Debbie Y. Dang

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The human foot is composed of 28 bones and 33 joint articulations. Together, the feet account for more than 25% of the total number of bones in the human body! These numbers only hint at the intricate anatomic and physiologic relationships that make bipedal locomotion possible. The foot and ankle surgeon needs a firm understanding of these relationships in order to diagnose, counsel, and treat patients.

This chapter, focused on foot and ankle biomechanics, is intended to provide the reader a foundation upon which to build an understanding of subsequent topics. First, a review of the gait cycle will be presented. Next, the biomechanics of the foot and ankle as it relates to gait will be outlined. And from these, surgical implications regarding the foot and ankle will be drawn. For greater details, the final sections in the chapter touch upon the kinematics and kinetics of human locomotion.

This chapter is based on the assumption that the reader has accurate knowledge of the anatomy of the foot and ankle. For review, the reader is encouraged to refer to anatomy textbooks that depict in detail the precise anatomic structures constituting the foot and ankle.[1,2]

GAIT CYCLE

At the most fundamental level, *gait* is defined as the manner in which a person walks. Incredible advancements in computing, sensor, and imaging technologies have made it possible for nuances in human motion to be captured and mimicked in two- and three-dimensional (3D) spaces. One need to only watch some of the latest Hollywood movies to see how seamlessly computer-generated movements integrate with natural human locomotion. At the root of these advancements are the basic principles of gait that were initially studied and outlined by surgeon scientists in the last century. Outlined below are the fundamental aspects of the gait cycle that a surgeon should understand. More detailed descriptions of gait analysis can be found in the literature.[3,4]

The walking cycle, being one of continuous motion, is difficult to appreciate in its entirety because so many events occur simultaneously. For simplicity, the cycle can first be considered from the standpoint of one limb. The gait cycle begins when the heel of that limb contacts the ground and ends when it again contacts the ground on the subsequent step. Within this cycle, are two phases: **stance phase** occurs while the foot is in contact with the ground and **swing phase** occurs while the foot is not in contact with the ground. For a given limb, more time is spent in stance phase (about 62%) than in swing phase (about 38%) (Fig. 1-1).

Each phase can further be divided into intervals. During stance phase, the first interval begins when the heel strikes the ground (initial contact) and continues to when the foot lies flat (loading response). The second interval occurs as the foot lies flat and the body weight passes over and is briefly aligned with the foot (mid stance). The third interval happens as the body "falls" over the foot, the heel rises from the floor (terminal stance), and the toes begin to leave the floor (pre-swing).

Continuing to follow the same limb, swing phase then occurs. It begins with initial swing, or toe-off, when the foot leaves the ground to when the knee is at maximum flexion during gait. Then mid swing happens between when the knee is at maximum flexion to when the tibia is vertical. This continues to terminal swing, from when the tibia is vertical to just before the heel hits the ground again to begin another cycle.

Of course, as one limb is experiencing one phase of the gait cycle, the other limb is in some part of the opposite phase. It is helpful now to think of what is occurring with the contralateral limb during the phases of gait. Let's consider the left and right limbs. During the first interval of stance phase, as the left foot experiences a loading response, the right foot is experiencing initial swing, or toe-off, both feet are momentarily touching the ground, providing double limb support. Then, as the body weight passes over the left foot, and the left foot experiences mid stance during the second interval, the right leg is in mid swing. During this time, there is single-limb support on the left leg. In the third interval, as the left foot experiences heel rise and pre-swing, the right foot is in terminal swing. Again, at this point, there is double limb support, where both feet are touching the ground.

FOOT AND ANKLE BIOMECHANICS

Now, with a basic understanding of the gait cycle, let's consider the mechanical principles that govern gait. The foot is often viewed as a rigid base that supports the rest of the body, when in fact, it is a dynamic structure that changes to accommodate different needs during movement.

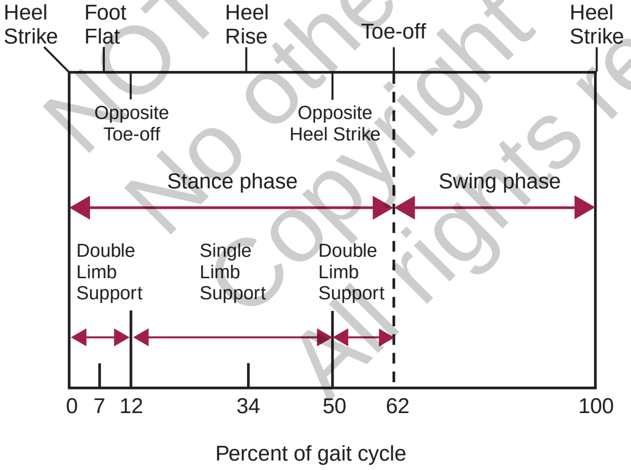

Fig. 1-1 Phases of the walking cycle. Stance phase constitutes approximately 62% and swing phase 38% of cycle. During stance phase of walking, there are two periods of double limb support and one period of single limb support. Stance phase is further divided into three intervals: from heel strike to foot flat at approximately 7% of the gait cycle, foot flat to heel rise at approximately 34% of the gait cycle, and heel rise to toe-off at approximately 62% of the gait cycle.

Consider the gait cycle described in the previous section. During stance phase, at heel strike, the foot needs to be supple in order to absorb the impact energy that comes with the weight of the body contacting the floor. And within moments, at the end of stance phase, that same foot transforms into a rigid structure over which the weight of the body can pass and fall into heel strike of the other foot. How does this transition happen?

Just as the gait cycle itself is the result of multiple concurrent events, the structural changes that occur during locomotion are the result of a confluence of anatomic relationships. There are passive relationships based on bone and joint anatomy, and there are also active relationships between muscle groups and their actions over each joint. The following sections explore these relationships separately before putting the information together so that it is more easily appreciated how the foot transitions from being supple to being rigid over the course of a gait cycle.

Joint Mechanics

To start, it is simplest to focus attention on one joint at a time, then to put together the knowledge about each joint to see how one affects the others during the foot's transition from suppleness to rigidity.

The Ankle Joint

The ankle joint is oriented obliquely in the transverse (or axial) plane as well as in the coronal plane. In the transverse, or axial plane, the ankle joint is externally rotated in relation to the sagittal plane. In the clinical literature, this rotation is described as tibial torsion and affects the degree to which the foot is internally or externally rotated, or the degree to which there is in-toeing or out-toeing.

In the coronal plane, the ankle axis is best approximated by a line connecting the tips of the medial and lateral malleoli (Fig. 1-2). The actual axis passes just distal to the tip of each malleolus. In the coronal plane, Inman found that the axis of the ankle may deviate 88 to 100 degrees from the vertical axis of the leg, with an average of 93 degrees.[5] It slants from proximal medial to distal lateral (Fig. 1-3).

The obliquity of the ankle joint contributes to the relative rotation of the leg and the foot depending on the position of the joint.[6]

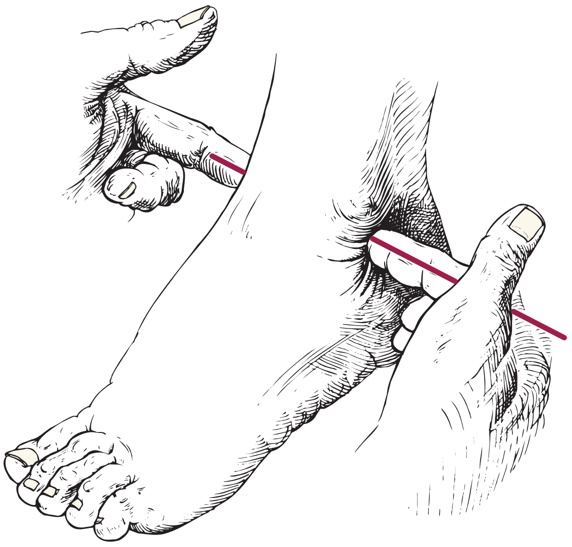

Fig. 1-2 Estimation of obliquity of empirical ankle axis by palpating tips of malleoli.

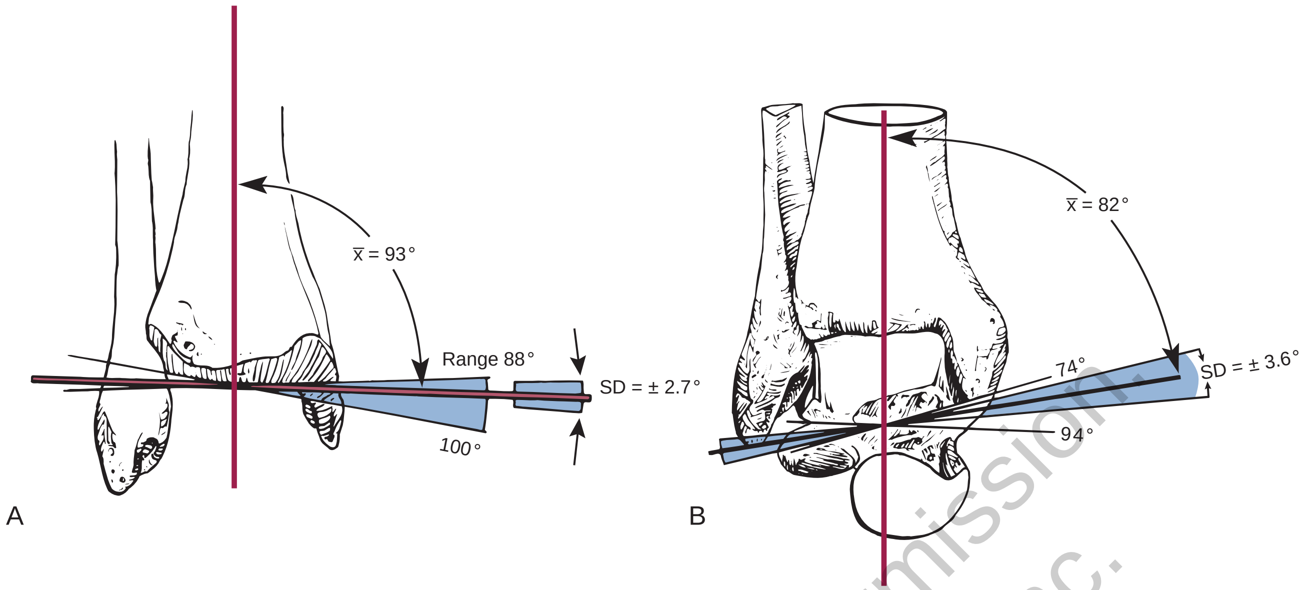

Fig. 1-3 **A,** Variations in angle between midline of tibia and plafond of mortise. **B,** Variations in angle between midline of tibia and empiric axis of ankle. *SD*, Standard deviation; $\bar{x}$, arithmetic mean. (From Inman VT: *The joints of the ankle*, Baltimore, 1976, Williams & Wilkins.)

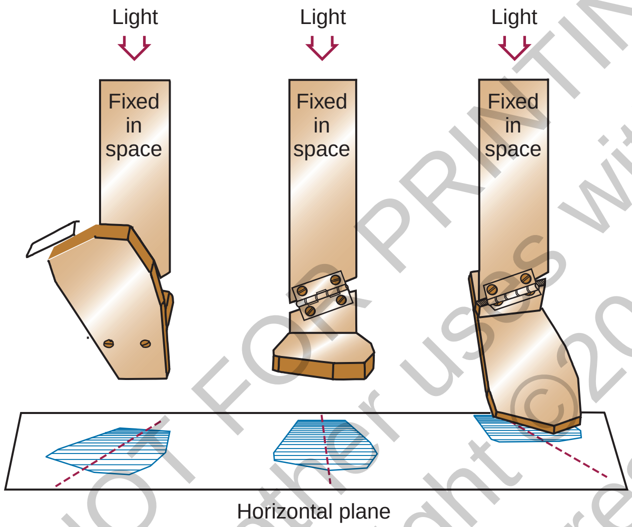

Fig. 1-4 Effect of obliquely placed ankle axis on rotation of foot in horizontal plane during plantar flexion and dorsiflexion, with foot free. Displacement is reflected in shadows of the foot.

When the leg is held still and the foot is allowed to move, dorsiflexion of the ankle causes the foot to deviate outward. Plantarflexion causes the foot to deviate inward (Fig. 1-4). The amount of rotation varies with the obliquity of the joint and the degree of dorsiflexion and plantar flexion. Conversely, when the feet are fixed to the floor, ankle dorsiflexion causes the tibia to rotate internally, and plantar flexion causes the tibia to rotate externally (Fig. 1-5).

It is important to note that an oblique ankle axis does not fully account for all of the leg rotation and foot positions during movement. For example, when the magnitudes of the various displacements are studied, it becomes clear that the rotation of the leg attributable to ankle axis obliquity is much smaller than the degree of horizontal rotation of the leg that actually occurs. This additional rotation occurs due to the interplay of joints both proximal and distal to the ankle joint as discussed below.

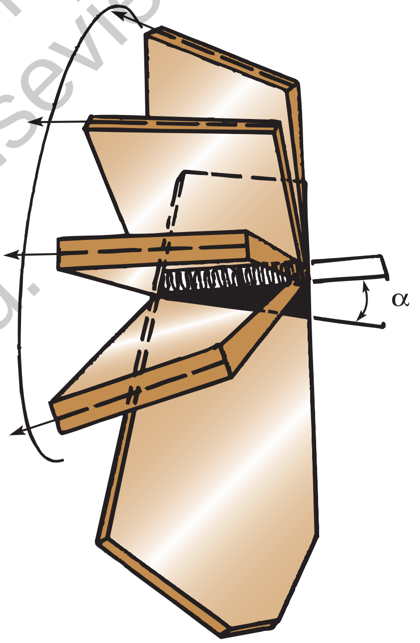

Fig. 1-5 Foot fixed to floor. Plantar flexion and dorsiflexion of ankle produce horizontal rotation of leg because of obliquity of the ankle axis.

The Subtalar Joint

The subtalar joint works in cooperation with the more proximal joints of the lower limb to account for the additional leg rotation not explained by the obliquity of the ankle joint axis. The subtalar joint is a sliding single-axis joint that acts like a mitered hinge connecting the talus and the calcaneus. In the coronal plane, the axis passes from medial to lateral at an angle of approximately 16 degrees. In the sagittal plane, the axis is angled about 42 degrees from horizontal. In the transverse plane, the joint deviates approximately 23 degrees medially from the long axis of the foot[7,8] (Fig. 1-6).

Mechanically, this hinge joint can be modeled by two boards joined by a hinge as in Fig. 1-7A. The vertical board represents the tibia and

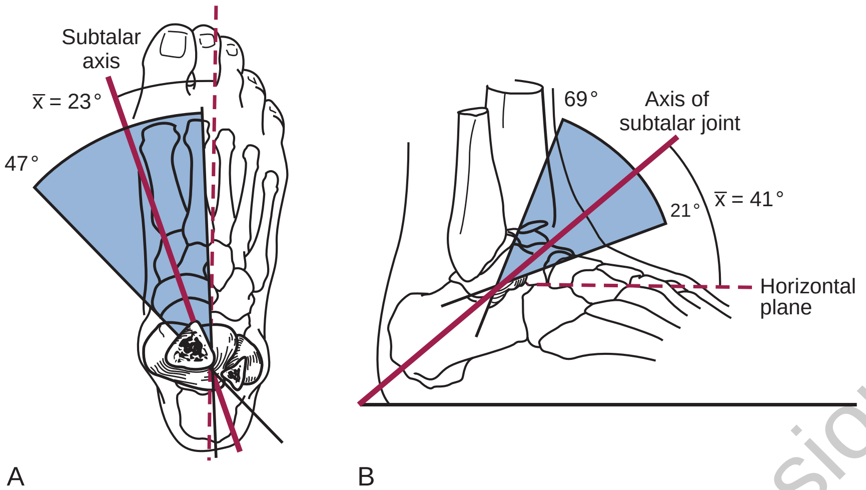

Fig. 1-6 Variations in subtalar joint axes. **A,** In transverse plane, subtalar axis deviates approximately 23 degrees medial to long axis of foot, with range of 4 to 47 degrees. **B,** In horizontal plane, axis approximates 41 degrees, with range of 21 to 69 degrees. $\bar{x}$, Arithmetic mean. (Modified from Isman RE, Inman VT: Anthropometric studies of the human foot and ankle, *Bull Prosthet Res* 10:97, 1969.)

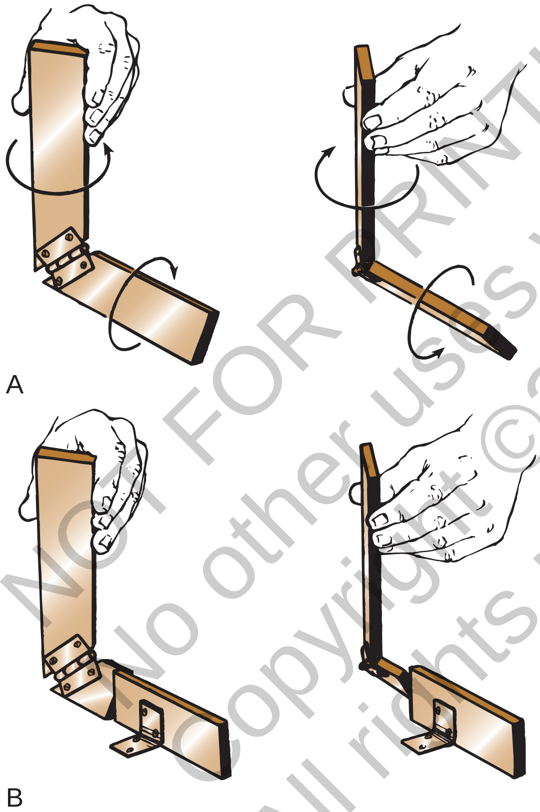

Fig. 1-7 Simple mechanism demonstrating functional relationships. **A,** Action of mitered hinge. **B,** Addition of pivot between two segments of mechanism.

the horizontal board the foot. If the axis of the hinge is 45 degrees, then a simple torque converter has been created. Rotation of the vertical member causes equal rotation of the horizontal member. Changing the angle of the hinge alters this one-to-one relationship such that a more horizontally placed hinge causes a greater rotation of the horizontal member for each degree of rotation of the vertical member. And the reverse is true such that a more vertically oriented hinge causes a smaller rotation of the horizontal member for each degree of rotation of the vertical member. External rotation of the leg is converted to hindfoot inversion through the oblique axis of the subtalar joint.[9]

The importance of this mechanical relationship is apparent when comparing the angle of the subtalar joint in the sagittal plane in individuals with congenital pes planus and those with cavovarus deformity. In individuals with congenital pes planus, the subtalar joint tends to be more horizontally oriented. In these individuals, a more horizontally angled subtalar joint results in greater suppleness of the hindfoot.

Conversely, in individuals with cavovarus feet, the subtalar joint tends to be oriented more vertically. This means that for each degree of rotation of the leg, there is relatively less inversion or eversion of the hindfoot. Individuals with cavovarus feet therefore tend to have stiffer, more rigid joints in the hindfoot.

The Transverse Tarsal Joint Complex

Traveling distally from the subtalar joint, the calcaneocuboid and talonavicular articulations together make up the transverse tarsal joint complex. While each possess some independent motion, from a functional standpoint, they perform together.

The simple model described above for the subtalar joint can be refined further to include the transverse tarsal joint complex (see Fig. 1-7B). The horizontal "foot" segment is divided into a short proximal and a long distal segment, with a pivot between the two segments. This pivot represents the transverse tarsal joint complex. Keeping the longer distal segment fixed to the floor in this model, the rotation at the pivot allows for inward and outward tilt at the shorter proximal segment. Analogously in the foot, the transverse tarsal joint is like a pivot that allows for hindfoot inversion and eversion while the forefoot remains in contact with the ground.

A closer look at the two joints of the transverse tarsal joint complex, the calcaneocuboid and talonavicular joints, reveals that beyond being a pivot between the hind and forefoot, they are responsible for the transition from suppleness to rigidity during gait. Elftman[10] demonstrated that the axes of these two joints are parallel when the calcaneus is in an everted position and are nonparallel when the calcaneus is in an inverted position. The importance of this is that, when the axes are parallel, there is flexibility within the transverse tarsal joint, whereas when

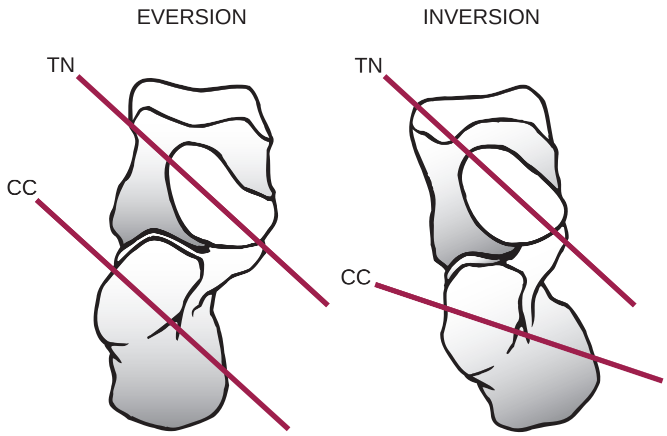

Fig. 1-8 Function of transverse tarsal joint (as described by Elftman H: The transverse tarsal joint and its control, *Clin Orthop Relat Res* 16:41–46, 1960) demonstrates that when the calcaneus is in eversion, the resultant axes of talonavicular *(TN)* and calcaneocuboid *(CC)* joints are parallel. When the subtalar joint is in an inverted position, the axes are nonparallel, giving increased stability to the midfoot.

the axes are nonparallel, there is rigidity at the transverse tarsal joint. Imagine a door where the hinges all line up and will open and close easily, whereas if the hinges of a door diverge, then the door will be stuck in one position. In other words, when the heel everts, the transverse tarsal joint is "unlocked" (supple), and when the heel inverts, the joint is "locked" (rigid) (Figs. 1-8 and 1-9).

Another clinical correlation can be drawn here. The dependency of transverse tarsal joint suppleness on hindfoot position further contributes to the suppleness of the foot in an individual with congenital pes planovalgus compared to someone with cavovarus deformity. Those with a valgus hindfoot will more likely have an "unlocked" and supple midfoot. (A caveat here is that this is not true in patients with advanced adult acquired flatfoot deformity, which is discussed in Chapter 29.) Those with a varus hindfoot will likely have a locked and rigid midfoot.

The talonavicular joint morphology adds additional stability to the longitudinal arch when force is applied across it during the last half of the stance phase. The joint surface has different curvature of radius in the anteroposterior and lateral projections (Fig. 1-10). When force is applied across a joint of this shape, stability is enhanced. This occurs at toe-off, when the plantar aponeurosis (described below) has stabilized the longitudinal arch and most of the body weight is being borne by the forefoot and medial longitudinal arch.

Tarsometatarsal Joints and Columns of the Midfoot

Traveling distally from the transverse tarsal joint complex, the midfoot can be divided further into two columns. The medial column consists of the first through third rays, including the cuneiform-metatarsal joints and their respective metatarsals. The lateral column consists of the fourth and fifth rays, including the cuboid-metatarsal joints, and the fourth and fifth metatarsals.

Anatomically, the tarsal-metatarsal joints in the medial column are relatively stiff compared to those at the lateral column. The medial column is therefore considered rigid, while the lateral column is supple. If the model from the previous section is further elaborated to incorporate the foot columns, then the long distal "foot" segment is divided into a medial and a slightly shorter lateral column (Fig. 1-11). External rotation of the vertical component of the model leads to inversion of the hindfoot portion, which then leads to elevation of the rigid medial forefoot and depression of the lateral forefoot. Internal rotation of the leg produces the opposite effect on the foot such that the hindfoot is everted, the medial forefoot is depressed, and the relatively flexible lateral forefoot remains in contact with the floor.

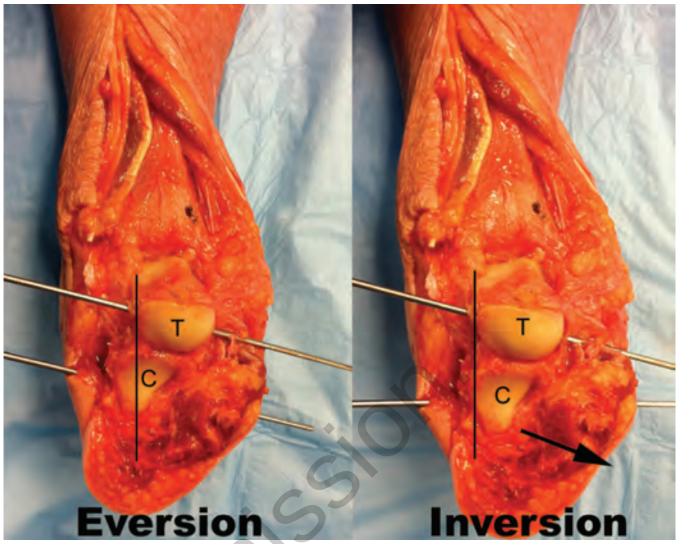

Fig. 1-9 Anatomic specimen with the foot removed at the transverse tarsal joint complex, demonstrating the relationship between talus and calcaneus during hindfoot motion. The talar head *(T)* and calcaneal side of the calcaneocuboid joint *(C)* are shown. The *vertical line* highlights motion of the calcaneus relative to the talus. K-wires mark the axes of the respective joints. When the calcaneus is in the everted position, the talonavicular and calcaneocuboid joint axes are parallel, and the transverse tarsal joint complex is mobile. When the calcaneus is in an inverted position, following the direction of the *arrow*, the talonavicular and calcaneocuboid joint axes diverge, and the transverse tarsal joint complex is locked.

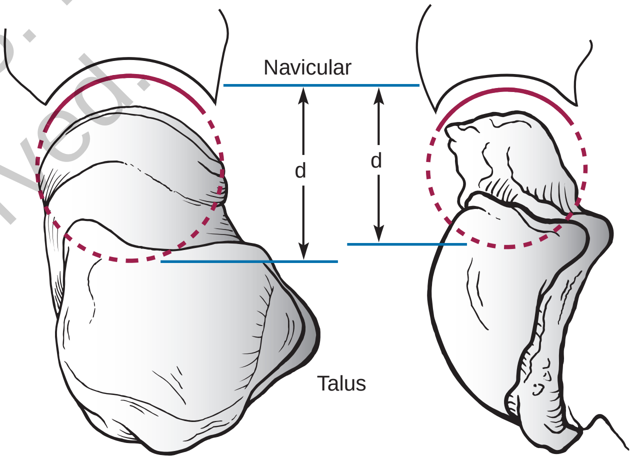

Fig. 1-10 Talonavicular joint. *Left*, Anterior view. *Right*, Lateral view. Relationship of head of talus to navicular bone shows differing diameters of head of talus. (From Mann RA: Intractable plantar keratoses. In Nicholas JA, Hershman EB, editors: *The lower extremity and spine in sports medicine*, ed 2, St Louis, 1995, Mosby.)

Metatarsophalangeal Joints

Finally, at the distal portion the foot are the metatarsaophalangeal joints. The distinguishing feature of the metatarsophalangeal joints is the axis formed by the unequal forward extension of the metatarsals

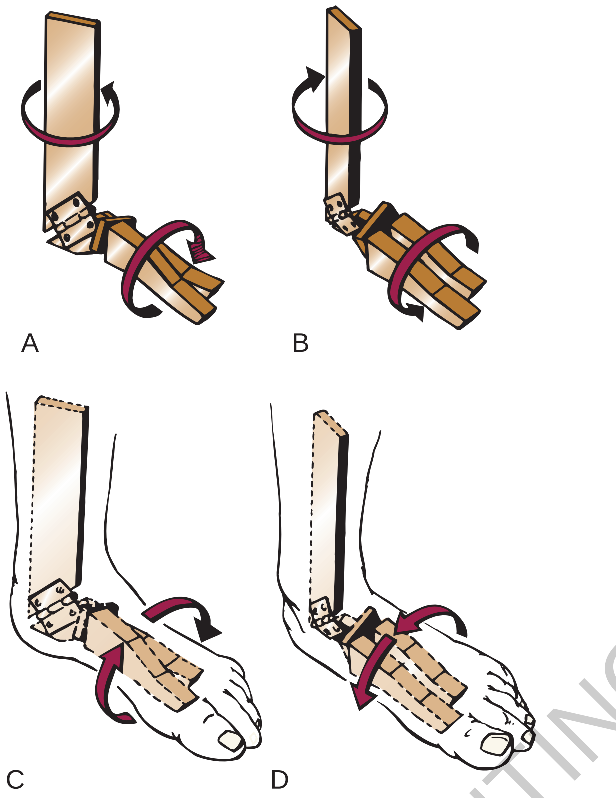

Fig. 1-11 Distal portion of horizontal member replaced by two structures. **A** and **B**, Mechanical analog of principal components of foot. **C** and **D**, Mechanical components inserted into foot and leg.

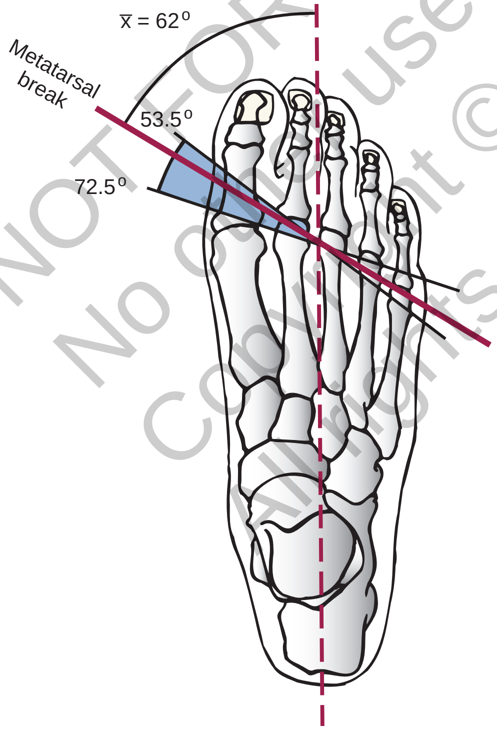

Fig. 1-12 Variations in metatarsal break in relation to longitudinal axis of foot. (From Isman RE, Inman VT: Anthropometric studies of the human foot and ankle, *Bull Prosthet Res* 10:97, 1969.)

(Fig. 1-12). This is referred to as the **metatarsophalangeal break**. The head of the second metatarsal is the most distal head; that of the fifth is the most proximal. Although the first metatarsal usually is shorter than the second (because the first metatarsal head is slightly elevated and is supported by the two sesamoid bones), it often functionally approximates the length of the second.

In the model above, the metatarsophalangeal break is modeled by a gentle taper or cascade in the length of the two columns going from medial to lateral. As the heel inverts and the medial column elevates, the axis of the metatarsophalangeal break allows all of the metatarsal heads to be in contact with the ground, thus evenly distributing body weight across the forefoot. If the metatarsals were all the same length, then as the heel inverts, only the metatarsal head of the fifth ray would be in contact with the ground. This concept is modeled more simply in Fig. 1-13. The angle between the metatarsal break and the longitudinal axis of the foot may vary from 50 to 70 degrees.[11] The more oblique the metatarsal break, the more the foot must supinate and deviate laterally after heel rise.

Progression From a Supple to Rigid Platform

At this point, the reader can take a moment to synthesize the relationships among the joints described in the models above to understand the cascade of events that allow the foot to transition from a supple to a rigid platform during gait.

The first interval in the gait cycle begins with heel strike and continues to when the foot lies flat. Just prior to heel strike, the hindfoot is inverted due to the pull of the tibialis anterior tendon. At heel strike, the anatomic position of the calcaneus being lateral to the tibial mechanical axis leads to eversion at the hindfoot, which leads to internal rotation of the leg proximally.[12] Distally this rapid eversion of the hindfoot unlocks the transverse tarsal joint, leaving a supple midfoot that allows both columns to contact the ground, absorbing the impact energy.

Then during the second interval of the gait cycle, the limb transitions through stance phase and the leg externally rotates, causing the hindfoot to invert, which then locks the transverse tarsal joint, leaving a rigid midfoot wherein the relatively stiff medial column is elevated and the lateral column is depressed.

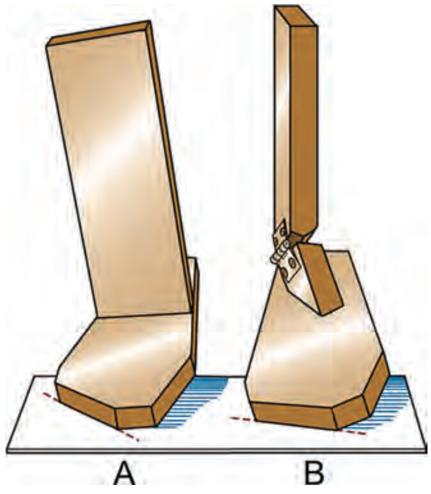

Fig. 1-13 Supination and lateral deviation of foot during raising of heel caused by oblique metatarsophalangeal break. **A**, Wooden mechanism without articulation. If no articulation is present, the leg deviates laterally. **B**, Model including the subtalar joint articulation. In addition to its other complex functions, the subtalar joint functions to permit the leg to remain vertical.

In the third interval, the body weight passes over the rigid midfoot while the heel remains inverted and transverse tarsal joints locked. And since there is an oblique axis to the metatarsophalangeal break, the forefoot remains in contact with the ground, with body weight evenly distributed across the metatarsal heads. In this way, as the heel lifts, the body weight can pass over a rigid and supportive foot.

The next section will introduce the modulators of the joint mechanics described above. Figs. 1–14 to 1–16 provide a visual representation of the multiple events that occur during the intervals of stance phase.

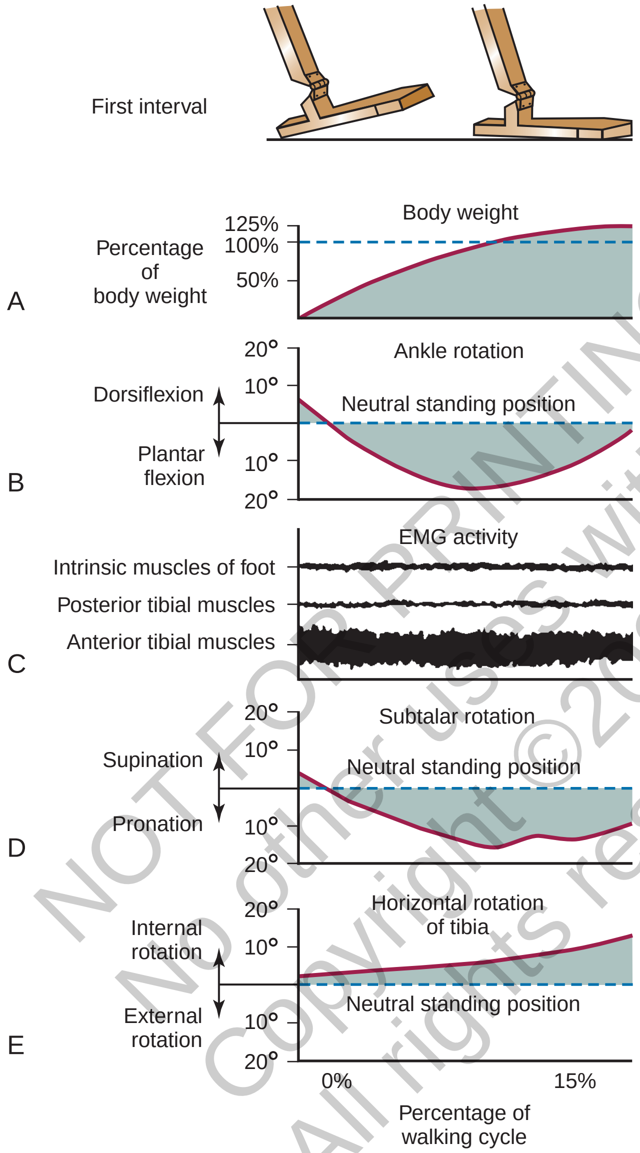

Fig. 1-14 Composite of events of first interval of walking, or period that extends from heel strike to foot flat, and occurs during the first 15% of the walking cycle. The heel's impact and body's center of gravity shift results in vertical floor reaction that transitions from zero to exceeding body weight by 15% to 25% (**A**). The ankle begins in dorsiflexion and progresses through plantarflexion (**B**). The anterior tibial muscles are active as seen in electromyograph *(EMG)* tracings (**C**), as they control the progressive plantar flexion at the ankle. The heel is mostly pronated to allow the foot to absorb the energy as the foot hits the ground (**D**). Accordingly, there is increased external rotation at the tibia as the hindfoot pronates (**E**).

Passive and Active Modulators of Joint Biomechanics

The mechanical relationships at the joints described above are modulated by both passive and active means. The anatomic characteristics at each joint described above are key passive contributors to the typical movements of the foot. The plantar aponeurosis, described below, is another important passive contributor to achieving a rigid platform in the foot during gait. The multiple tendons that cross the ankle into the foot are part of active systems that control foot biomechanics.

The Plantar Aponeurosis

An important passive contributor to foot biomechanics is the plantar aponeurosis, a band of fibrous tissue arising from the tubercle of the calcaneus and passing distally to insert into the base of the proximal phalanx (Fig. 1-17). As the plantar aponeurosis passes the plantar aspect of the metatarsophalangeal joints, it combines with the joint capsule to form the plantar plate.

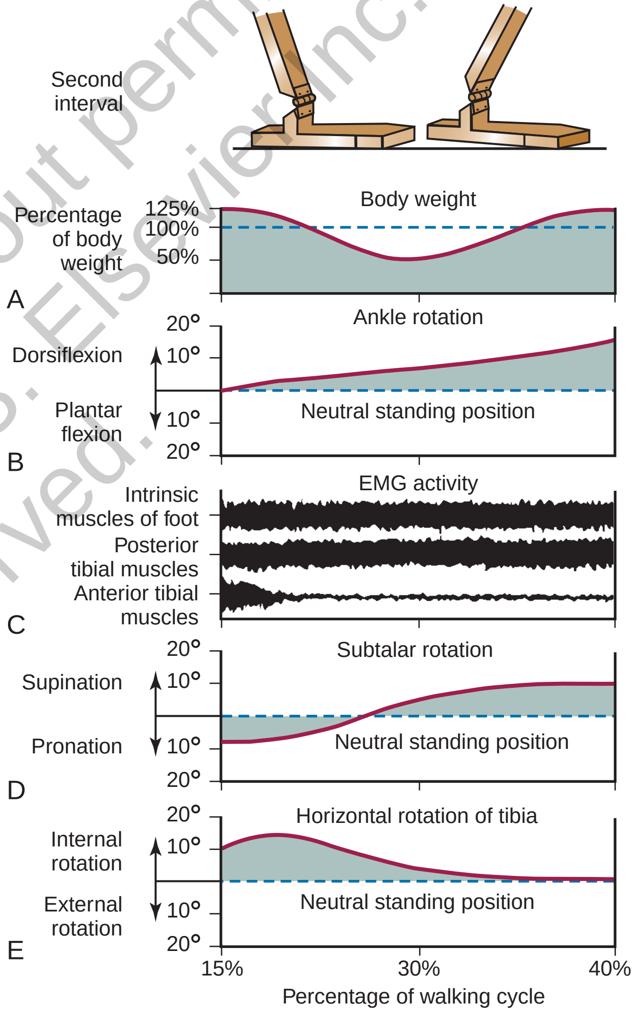

Fig. 1-15 Composite of events of second interval of walking, or period of foot flat, which extends from 15% to 40% of the walking cycle. As the body passes over the foot in stance phase, force plate recordings show that the load on the foot may be as low as 70% to 80% of actual body weight (**A**). The ankle progresses from plantar to dorsiflexion (**B**), and the posterior tibial muscles contract eccentrically, along with the intrinsic foot muscles, to control the forward movement of the tibia (**C**), allowing the contralateral leg to take a longer step. The hindfoot supinates, and the tibia begins to come out of internal rotation (**D** and **E**).

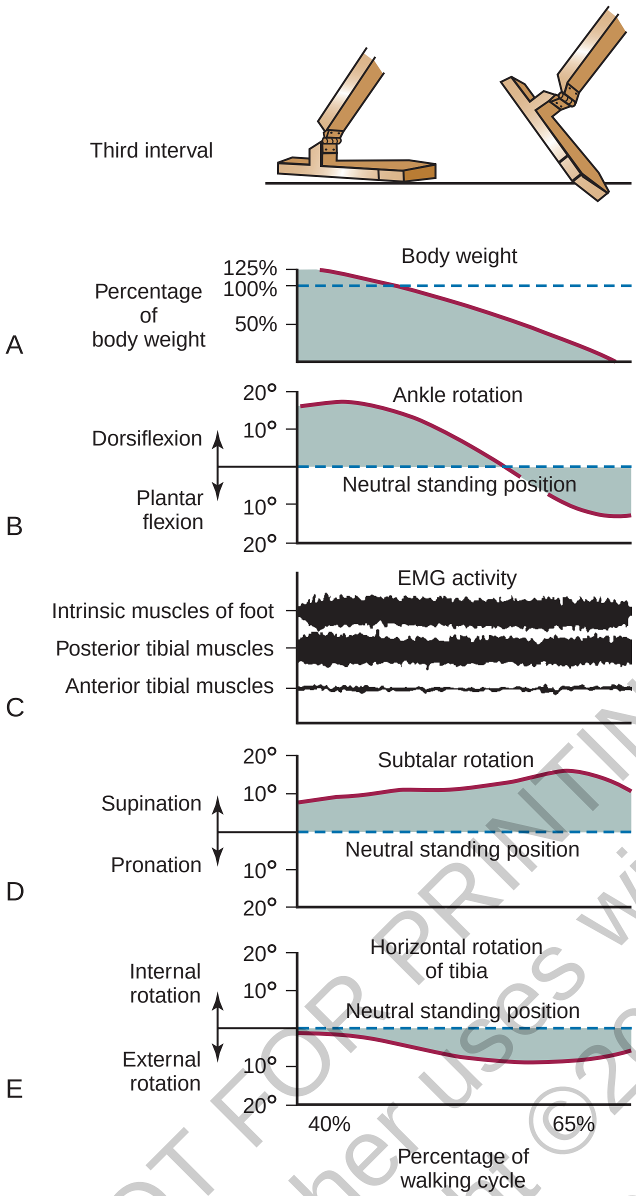

Fig. 1-16 Composite of all events of third interval of walking, or period extending from foot flat to toe-off, and extends from 40% to 62% of the walking cycle. At the end of stance phase, as the other foot begins heel strike, the stance phase foot accordingly does not bear force toward the end of heel rise (**A**). The ankle plantar flexes at toe-off (**B**), driven by concentric contraction of the posterior calf musculature (**C**). The anterior compartment muscles become active in the last 5% of this interval. The subtalar joint remains supinated, and the tibia goes into external rotation (**D** and **E**).

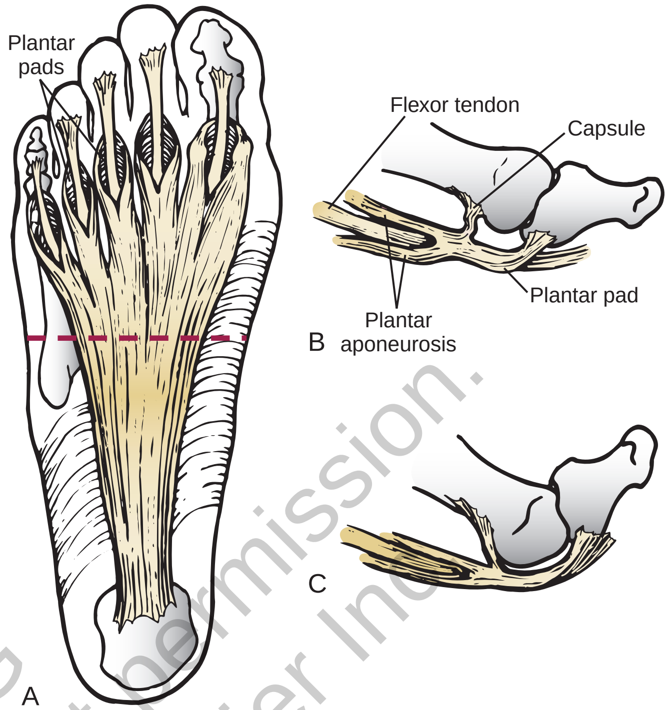

Fig. 1-17 Plantar aponeurosis. **A**, Division of plantar aponeurosis around flexor tendons. **B**, Components of plantar pad and its insertion into base of proximal phalanx. **C**, Extension of toes draws plantar pad over metatarsal head, pushing it into plantar flexion.

The plantar aponeurosis is the most significant stabilizer of the longitudinal arch between heel rise and toe off. As the body moves over the fixed foot and the heel begins to rise, the proximal phalanges dorsiflex, pulling the plantar aponeurosis over the metatarsal heads. This tightens the plantar fascia, resulting in a depression of the metatarsal heads and an elevation of the longitudinal arch (Fig. 1-18). This mechanism is passive in that no muscle function per se brings about this stabilization.

The plantar aponeurosis is most functional on the medial side of the foot and becomes less functional as one moves laterally toward the fifth metatarsophalangeal articulation. Based on its medial attachment to the calcaneus, plantar fascia tightening also contributes to hindfoot inversion, tibial external rotation, and transverse tarsal joint stabilization.[13] These changes stabilize the midfoot and further allow the foot to act as a rigid lever during the toe-off phase of gait.

The function of the plantar aponeurosis is referred to as the **windlass mechanism** of the foot, likening its structure and function to an anchor windlass on a boat.[14] The mechanics of the windlass mechanism can be demonstrated clinically by having an individual stand and forcing the great toe into dorsiflexion. As this occurs, one observes elevation of the longitudinal arch by the depression of the first metatarsal by the proximal phalanx, and, at the same time, inversion of the calcaneus. Careful observation of the tibia demonstrates that it externally rotates in response to this calcaneal inversion.

The Posterior Calf Muscles

The function of the posterior calf group during stance phase is to control the forward movement of the tibia on the fixed foot.[15,16] Control of the forward movement of the stance leg tibia is critical to normal gait because it permits the contralateral leg to take a longer step, increasing stride length and improving walking efficiency. In pathologic states in which the calf muscle is weak, the stride length shortens, and dorsiflexion occurs at the ankle joint after heel strike because it is a position of stability. Paradoxically, the ankle is held more rigidly by secondary stabilizers to make up for the inability to control ankle dorsiflexion.[17]

The posterior calf muscles basically function as a group, although the tibialis posterior and peroneus longus muscles usually begin functioning by about 10% of the stance phase, whereas the other posterior calf muscles tend to become functional at about 20% of the stance phase. As the ankle joint undergoes progressive dorsiflexion from foot flat until heel rise at 40% of the cycle, these muscles contract eccentrically. After heel rise, as ankle plantar flexion begins, they continue to contract, but now via a concentric contraction. It is interesting to note, however, that by 50% of the cycle, the electrical activity in these muscles

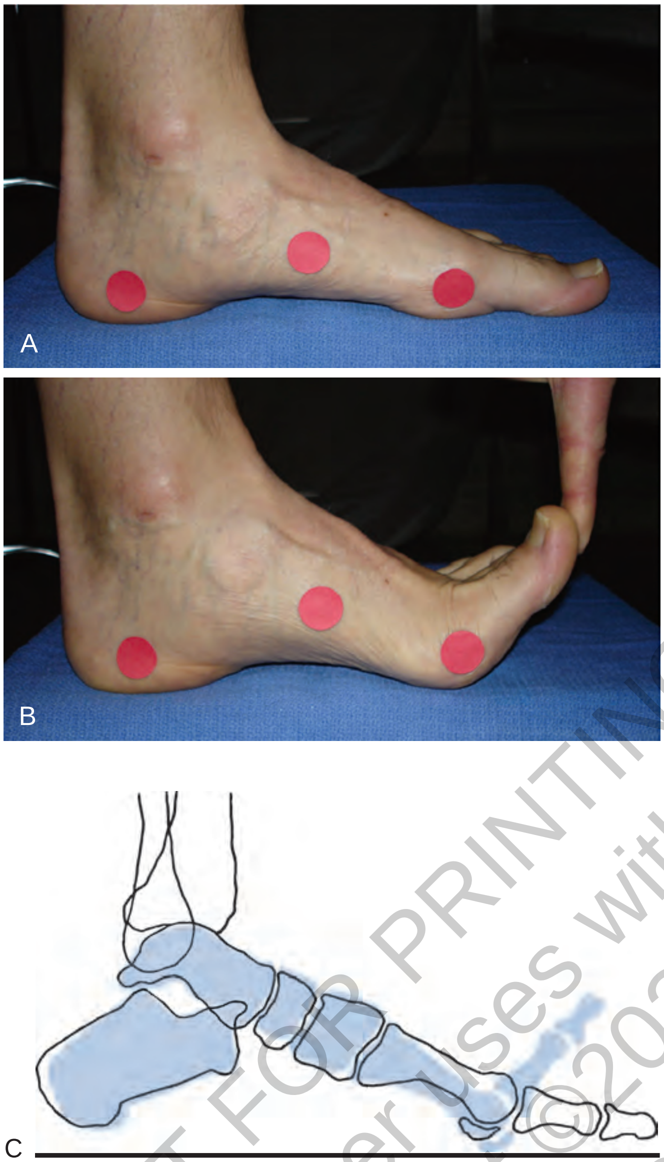

Fig. 1-18 Dynamic function of plantar aponeurosis. **A**, Foot at rest. **B**, Dorsiflexion of metatarsophalangeal joints, which activates windlass mechanisms, brings about elevation of longitudinal arch, plantar flexion of metatarsal heads, and inversion of heel. **C**, Superimposed tracing of lateral radiographs of the foot at rest *(outline)* and with first ray dorsiflexion *(gray figure)*. Notice that dorsiflexion of the first toe tightens the plantar aponeurosis, which results in depression of the metatarsal heads, elevation and shortening of the longitudinal arch, inversion of the calcaneus, and elevation of the calcaneal pitch. (**C**, from Haskell A: Foot and ankle biomechanics. In Miller MD, Thompson SR, editors: *DeLee, Drez, and Miller's orthopaedic sports medicine*, Philadelphia, 2020, Elsevier.)

ceases, and the remainder of the plantar flexion of the ankle joint is a passive event. High-speed motion pictures have demonstrated that during steady-state walking, at the time of toe-off, the foot is lifted from the ground, and the toes do not actively push off.

Muscle activity in the deep posterior compartment contributes to hindfoot inversion (see Fig. 1-15). As the posterior tibial muscle–tendon complex contracts, the hindfoot is pulled into inversion. Activity of the intrinsic muscles of the foot also contributes to midfoot stability and correlates fairly closely with the degree of subtalar joint rotation. In the normal foot, the intrinsic muscles become active at about 30% of the walking cycle, whereas in flatfoot, they become active during the first 15% of the walking cycle[18] (Fig. 1-19).

The Anterior Calf Muscles

The anterior calf muscles contract eccentrically and function to slow the rapid ankle plantarflexion as the foot goes from heel strike to flatfoot during the first interval of the gait cycle. Anterior compartment musculature weakness results in a footdrop gait, characterized by accentuated hip flexion or circumduction of the hip during swing phase to avoid the toes of the dropped foot hitting the floor during swing-through.

During swing phase, dorsiflexion occurs at the ankle joint. Beginning at about 55% of the cycle and throughout swing phase, the anterior compartment muscles contract concentrically to dorsiflex the ankle. The medial insertion of the tibialis anterior tendon pulls the hindfoot into slight inversion during swing phase such that the calcaneus is slightly inverted at initial heel strike, before everting as described above (see Figs. 1-14–1-16). This is why most people will wear down the outer edge of the heel in their shoes asymmetrically.

Ligaments of the Ankle

The ankle joint is stabilized by ligaments whose configuration and alignment permit free movement of the ankle and subtalar joints to occur simultaneously. Because the configuration of the trochlear surface of the talus is curved to produce a cone-shaped articulation whose apex is directed medially, the single fan-shaped deltoid ligament is adequate to provide stability to the medial side of the ankle joint (Fig. 1-20). However, on the lateral aspect of the ankle joint, there is a larger area to be covered by a ligamentous structure. The lateral ligaments are divided into three bands: the anterior and posterior talofibular ligaments, and the calcaneofibular ligament.

Fig. 1-21 demonstrates the anterior talofibular and calcaneofibular ligaments in relation to the subtalar joint axis. The calcaneofibular ligament is parallel to the subtalar joint axis in the sagittal plane. As the ankle joint is dorsiflexed and plantar flexed, this relationship between the calcaneofibular ligament and the subtalar joint axis does not change. It is important to appreciate that, when the ankle joint is in neutral position, the calcaneofibular ligament is angulated posteriorly, but as the ankle joint is brought into more dorsiflexion, the calcaneofibular ligament is brought into line with the fibula, thereby becoming a true collateral ligament. Conversely, as the ankle joint is brought into plantar flexion, the calcaneofibular ligament becomes horizontal to the ground. In this position, it provides little or no stability for resisting inversion stress.

The anterior talofibular ligament, on the other hand, is brought into line with the fibula when the ankle joint is plantar flexed, thereby acting as a collateral ligament. When the ankle joint is brought up into dorsiflexion, the anterior talofibular ligament becomes sufficiently horizontal so that it does not function as a collateral ligament. It can thus be appreciated that, depending on the position of the ankle joint, either the calcaneofibular or the anterior talofibular ligament will be a true collateral ligament with regard to providing stability to the lateral side of the ankle joint.

The relationship between these two ligaments has been quantified and is presented in Fig. 1-22. This demonstrates the relationship of the angle produced by the calcaneofibular and the anterior talofibular ligaments to one another. The average angle in the sagittal plane is approximately 105 degrees, although there is considerable variation, from 70 to 140 degrees. This is important because, from a clinical standpoint, it partially explains why some persons have lax collateral ligaments. If we assume that when the ankle is in full dorsiflexion the calcaneofibular

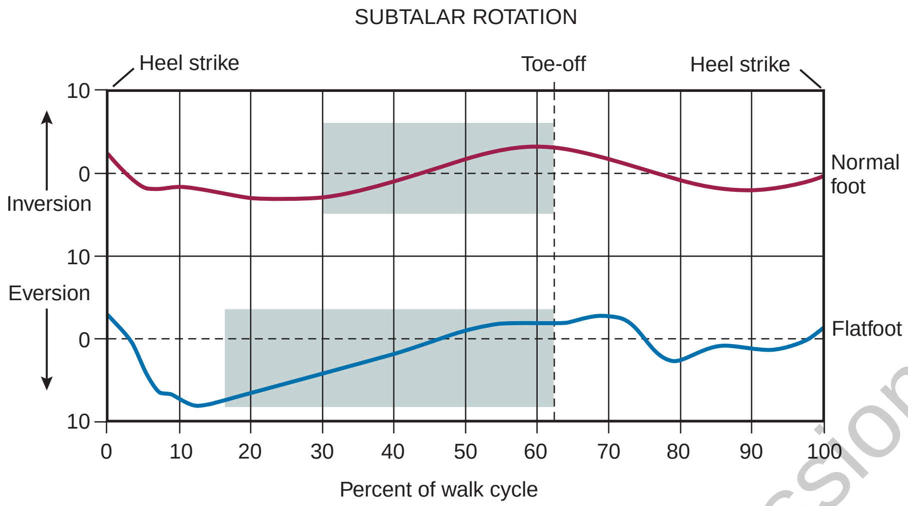

Fig. 1-19 Subtalar joint motion in normal foot and flatfoot. Shaded areas indicate period of activity of intrinsic muscles in normal foot and flatfoot.

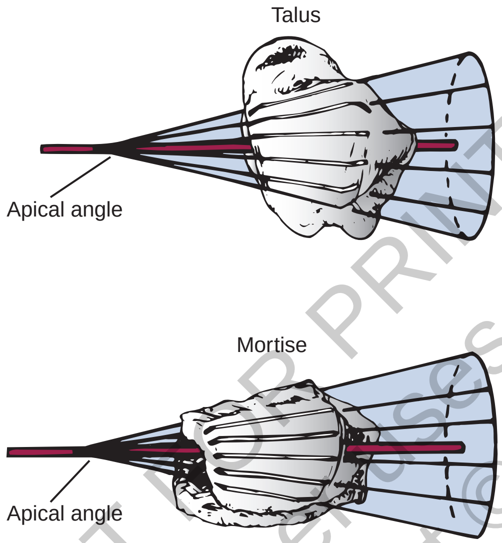

Fig. 1-20 Curvature of trochlear surface of talus creates cone whose apex is based medially. From this configuration, one can observe that the deltoid ligament is well suited to function along the medial side of ankle joint, whereas laterally, where more rotation occurs, three separate ligaments are necessary. (From Inman VT: *The joints of the ankle*, Baltimore, 1976, Williams & Wilkins)

ligament provides most of the stability and that in full plantar flexion the anterior talofibular ligament provides stability, then as we pass from dorsiflexion to plantar flexion and back there will be a certain period in which neither ligament is functioning as a true collateral ligament. If we assume there is an average angle of approximately 105 degrees between these ligaments, then generally speaking, an area in which an insufficient lateral collateral ligament is present is unusual; however, if we have angulation of 130 to 140 degrees between these two ligaments, there is a significant interval while the ankle is passing from dorsiflexion to plantar flexion and back in which neither ligament is functioning as a collateral ligament. This may explain why some persons are susceptible to chronic ankle sprains. Some patients who are thought to have ligamentous laxity may, in reality, possess this anatomic configuration of lateral collateral ligaments.

Putting It Together: Joint Mechanics and Modulators During Gait

At this point, the mechanics of individual joints at the ankle and foot have been described. Additionally, the previous section outlined important modulators of these joints. The following sections put this information together to describe what occurs during the intervals of gait.

First Interval

The first interval occurs during approximately the first 15% of the walking cycle and is defined from the moment of initial heel strike to when the foot becomes flat on the floor. Typically, the opposite heel has lifted from the floor but weight remains on the forefoot. During the first interval, the foot helps to absorb and dissipate the forces generated by the foot striking the ground.

The ankle joint undergoes rapid plantar flexion from heel strike until foot flat is achieved. At approximately 7% of the walking cycle, dorsiflexion begins (see Fig. 1-14B).

As the foot is loaded with the weight of the body during the first interval, the calcaneus rapidly everts and the longitudinal arch flattens. This flattening of the arch originates in the subtalar joint and reaches a maximum during this interval (see Fig. 1-14D). The hindfoot is often mildly supinated at initial ground contact associated with ankle dorsiflexion during swing-through. The hindfoot moving from supination to pronation during the first interval is a passive mechanism, and the amount of motion appears to depend entirely on the configuration of the articulating surfaces, their capsular attachments, and ligamentous support. No significant muscle function appears to play a role in restricting this motion at initial ground contact.

The subtalar joint links rotation of the hindfoot to rotation of the leg. During the first interval, eversion of the calcaneus is translated by the subtalar joint into inward rotation that is transmitted proximally across the ankle joint into the lower extremity (see Fig. 1-14E). Distally, this hindfoot eversion unlocks the transverse tarsal joint, allowing the midfoot joints to become supple (see Fig. 1-14D). This allows the flattening of the longitudinal arch that contributes to energy dissipation during this phase.

At heel strike, the center of gravity of the body is decelerated by ground contact, then immediately accelerated upward to carry it over the extending lower extremity. The heel's impact and body's center of

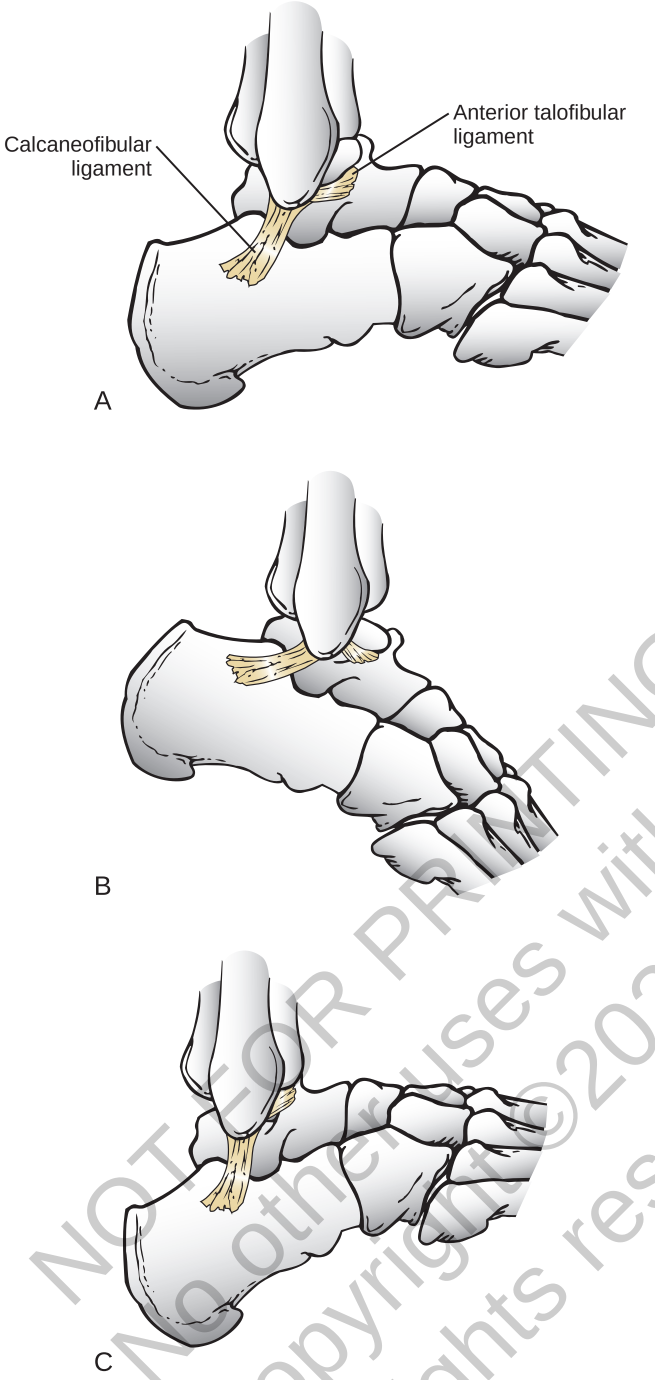

Fig. 1-21 Calcaneal fibular ligament and anterior talofibular ligament. **A**, In neutral position of ankle joint, both anterior talofibular and calcaneofibular ligaments provide support to joint. **B**, In plantar flexion, anterior talofibular ligament is in line with fibula and provides most of support to lateral aspect of ankle joint. **C**, In dorsiflexion, calcaneofibular ligament is in line with the fibula and provides support to the lateral aspect of ankle joint. (From Inman VT: *The joints of the ankle*, Baltimore, 1976, Williams & Wilkins)

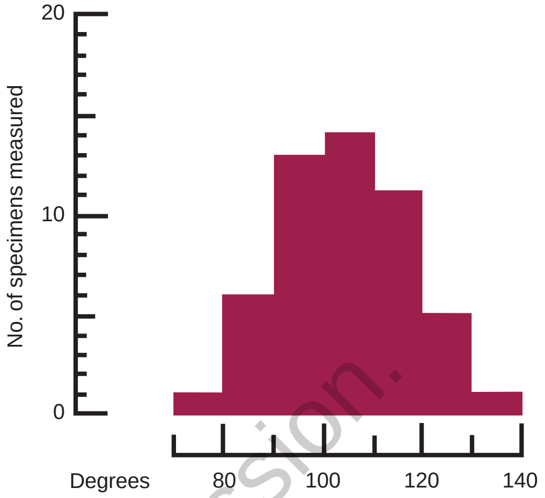

Fig. 1-22 Average angle between calcaneofibular and talofibular ligaments in sagittal plane. Although the average angle is 105 degrees, there is considerable variation, from 70 to 140 degrees. (From Inman VT: *The joints of the ankle*, Baltimore, 1976, Williams & Wilkins)

gravity shift accounts for a vertical floor reaction that exceeds body weight by 15% to 25% (see Fig. 1-14A).

Eccentric contraction of the anterior compartment leg muscles slows the rapid ankle plantar flexion during this phase from heel strike until a foot-flat position is reached. The posterior calf muscles all are electrically quiet, as are the intrinsic muscles in the sole of the foot (see Fig. 1-14C). There is no muscular response in those muscles usually considered important in supporting the longitudinal arch of the foot. Weakness of the anterior compartment muscles leads to a loss of this deceleration and a characteristic slap foot gait.

Second Interval

The second interval extends from 15% to 40% of the walking cycle. During this interval, the body's center of gravity passes from behind to in front of the weight-bearing leg. It reaches a maximum height as it passes over the leg at about 35% of the cycle, after which it commences to fall. During this interval, the foot transitions from a flexible, energy-absorbing structure to a more rigid one, capable of bearing the body's weight.

The ankle joint undergoes progressive dorsiflexion during the second interval, reaching its peak at 40% of the walking cycle. This is when the force across the ankle joint has reached a maximum of 4.5 times body weight. Heel rise begins at 34% of the cycle as the contralateral leg passes by the stance foot and precedes the onset of plantar flexion, which begins at 40% (see Fig. 1-15B).

During the second interval, the subtalar joint progressively inverts. This starts at about 30% of the cycle in a normal foot and at about 15% of the cycle in a flatfoot (see Fig. 1-15D). Multiple factors contribute to this inversion, but precisely which plays the greatest role is unclear. Above the subtalar joint, the swinging contralateral limb externally rotates the stance limb. This external rotation torque is translated by the subtalar joint into hindfoot inversion. The oblique nature of the ankle joint axis, the oblique setting of the metatarsal break, and the function of the plantar aponeurosis also contribute to hindfoot inversion. Inversion of the subtalar joint is passed distally into the midfoot, increasing the stability of the transverse tarsal articulation and transforming the flexible midfoot into a rigid structure.

During this interval, full body weight is not borne on the foot, smoothing the transition to single limb support. Force plate recordings show that the load on the foot may be as low as 70% to 80% of actual body weight (Fig. 1-15A).

During the second interval, important functional changes occur in both the foot and leg, which are the result of muscular action. The posterior and lateral compartment leg muscles (triceps surae, peroneals, tibialis posterior, long toe flexors) and intrinsic muscles in the sole of the foot demonstrate electrical activity (see Fig. 1-15C). Intrinsic muscle activity of the normal foot begins at 30% of the cycle, whereas in flatfoot, activity begins at 15% of the cycle. The posterior calf musculature slows the forward movement of the tibia over the fixed foot, which permits the contralateral limb to increase its step length. Weakness of the posterior compartment muscles may lead to premature contralateral heel strike and shortened stride length.

Third Interval

The third interval constitutes the last of the stance phase and extends from 40% to 62% of the walking cycle.

The ankle joint demonstrates rapid plantar flexion during this interval as the foot essentially extends the effective limb length. The subtalar joint continues to invert during this interval, reaching its maximum at toe-off (see Fig. 1-16D). This completes the conversion of the forefoot from the flexible structure observed in the first interval at the time of weight acceptance to a rigid structure at the end of the third interval in preparation for toe-off. The inversion is a continuation of the processes that began in the second interval. These include external rotation of limb above the foot passing across the ankle and subtalar joints as well as mechanisms in the foot such as the obliquity of the ankle joint, the function of the plantar aponeurosis, and obliquity of the metatarsal break. Distally, the transverse tarsal joint is converted from a flexible structure into a rigid one by the progressive inversion of the calcaneus. The talonavicular joint also is stabilized during this period by the pressure placed across the joint by both body weight and the intrinsic force created by the plantar aponeurosis.

At the beginning of the third interval, force plate recordings demonstrate an increase in the percentage of body weight borne by the foot resulting from the center of gravity falling. The load on the foot exceeds body weight by approximately 20%. Later in the interval, the vertical floor reaction force falls to zero as the body's weight is transferred to the opposite foot (see Fig. 1-16A).

Ankle plantar flexion during the third interval is caused primarily by the concentric contraction of the posterior calf musculature, in particular the triceps surae (see Fig. 1-16B). The plantar flexion leads to relative elongation of the extremity. Although full plantar flexion at the ankle joint occurs during this interval, electrical activity is observed only until 50% of the cycle, after which there is no longer electrical activity in the extrinsic muscles (see Fig. 1-16C). The remainder of ankle joint plantar flexion occurs because of the transfer of weight from the stance leg to the contralateral limb. The intrinsic muscles of the foot are active until toe-off. Although the intrinsic muscles help to stabilize the longitudinal arch, the main stabilizer is the plantar aponeurosis, which is functioning maximally during this period as the toes are brought into dorsiflexion and the plantar aponeurosis is wrapped around the metatarsal heads, forcing them into plantar flexion and elevating the longitudinal arch. The anterior compartment muscles become active in the last 5% of this interval, probably to initiate dorsiflexion of the ankle joint immediately after toe-off.

Component Mechanics of Running

While the mechanical relationships described for each joint remain consistent, there are some differences between walking and running that are described below.

During running, the stance phase is diminished from approximately 0.6 second while walking to 0.2 second while sprinting (Fig. 1-23). During this brief period of stance phase, the forces involved in the vertical plane are increased to 2.5 to 3 times body weight. The range of motion of the joints is increased approximately 50%, and the muscles in the lower extremity must control these motions over a short time when measured in real time but over a considerable period when expressed as percentage of the gait cycle. It is probably because of the increased forces and muscle action required over a shorter period of time, and the repetitive nature of sport, that overuse injuries occur during running.

Considerable alterations occur around the ankle joint when comparing jogging or running with walking. The gait cycle time progressively decreases from 1 second to 0.6 second. The ankle's total arc of motion increases from 30 degrees during walking to 45 degrees during running. This motion occurs during 0.6 second for walking and 0.2 second for running. The direction of motion also changes; during walking, plantar flexion occurs at heel strike, whereas during jogging and running, there is progressive dorsiflexion. Rapid plantar flexion occurs at toe-off during all speeds of gait.

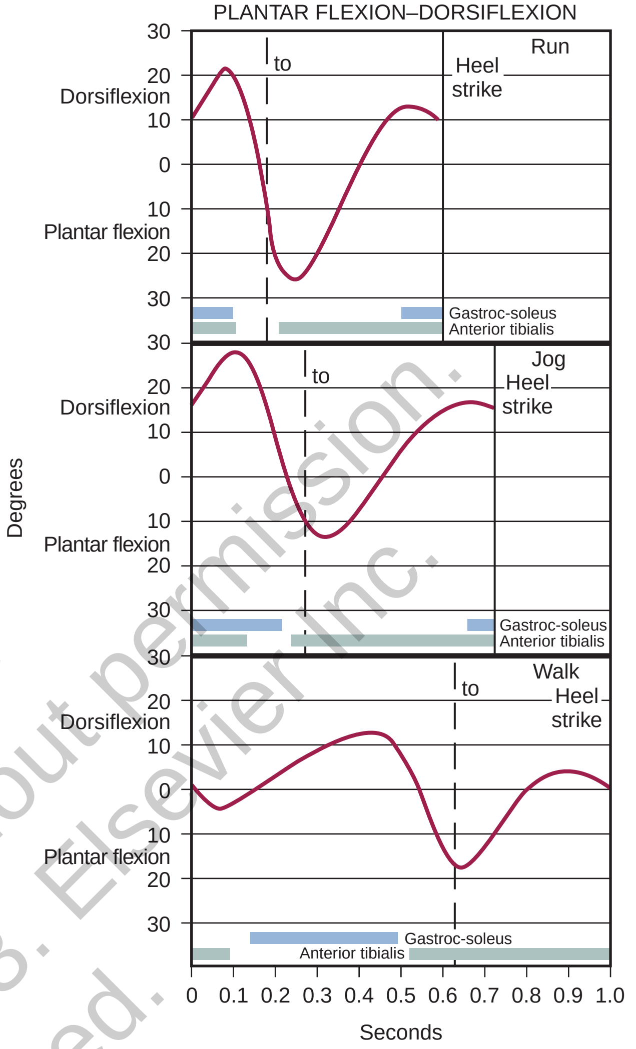

Fig. 1-23 Ankle joint dorsiflexion–plantar flexion during running, jogging, and walking. Note that time of walking cycle decreases from 1 second for walking to approximately 0.6 second for running. Stance-phase time decreases significantly, as well. Muscle function is characterized by gastrocnemius–soleus muscle group and anterior tibial muscle. Note that gastrocnemius–soleus muscle group becomes active in late swing phase for jogging and running, compared with stance-phase muscle for walking. (From Mann RA: Intractable Plantar Keratoses. In Nicholas JA, Hershmann EB, editors: *The lower extremity and spine in sports medicine*, ed 2, St Louis, 1995, Mosby)

Along with this increase in the range of motion and in the forces generated during running, the muscle function in the lower extremity also is altered. In real time, the phasic activity of most muscles decreases; however, when considered as a percentage of the gait cycle, the period of activity of these muscles increases considerably. Generally speaking, at initial ground contact, the majority of the muscles about the hip, knee, and ankle joints are active, and their period of activity, which begins during the late float phase, increases as the speed of gait increases. This is probably related to the rapid motion required by these joints in preparation for the impact of ground contact. During walking, there is adequate time for most of the preparation for ground contact to be carried out rather passively, but with the markedly increased range and speed of motion of these joints during running, muscle function plays a more active role.

As the speed of gait increases, the muscle function in the posterior calf group changes significantly. During walking, the posterior calf group functions in stance phase, and during jogging and running, it performs in late swing phase; its activity is ongoing from the time of initial ground contact through most of the stance phase. The muscle group controls the ankle dorsiflexion that occurs after initial ground contact, the forward movement of the tibia, and brings about plantar flexion of the ankle joint. Similar changes in both the magnitude of motion and muscle function occur about the hip and knee joints as well. During running and changing direction, as well as acceleration and deceleration, the toes play an active role in push-off, whereas push-off is minimal during steady-state walking.

CLINICAL IMPLICATIONS OF FOOT AND ANKLE BIOMECHANICS

Now with an understanding of the anatomic relationships that guide the mechanics of gait, the reader has a foundation for understanding the surgical implications of foot and ankle biomechanics. The principles outlined above govern how a surgeon maintains or alters the anatomic relationships among joints to permit optimal movement. The concepts discussed below serve as an introduction to subsequent chapters that provide greater detail on each specific topic.

Ankle Arthrodesis

Because all of the joints of the lower limb work together during gait, it is important that the above anatomic facts be considered when carrying out an ankle arthrodesis. The following paragraphs outline specific considerations with respect to transverse rotation, varus or valgus tilt, and dorsiflexion. Each of these variables impacts the amount of stress experienced at joints proximal and distal to the fused joint.

Transverse Rotation

If the ankle is placed into excessive internal rotation, the patient experiences difficulty when the center of gravity passes over the foot. The position of internal rotation places increased stress on the subtalar and mid tarsal joint region, which may become painful as a result of increased stress. Knee pain, and possibly hip pain, may also develop secondarily as a result of attempts to externally rotate the lower limb to help compensate for the abnormal position of the foot. Conversely, if the ankle is placed in too much external rotation, the patient tends to roll over the medial border of the foot. This position permits the patient to easily roll over the foot, but in turn, it places increased stress on the medial side of the first metatarsophalangeal joint, which can lead to hallux valgus deformity. It may also cause increased stress along the medial side of the knee joint, though no studies have demonstrated this specific relationship. Arthrodesis of the ankle joint so that the forefoot is in 5 to 10 degrees of external rotation is recommended.

Varus or Valgus Tilt

The degree of varus or valgus tilt of the ankle joint affects the degree of subtalar joint motion and the overall alignment of the tibia and knee. If the subtalar joint is stiff and unable to compensate for any malalignment, then it is imperative to place the ankle joint into sufficient valgus to obtain a plantigrade foot. If the ankle joint is placed into a varus potion, the patient will walk on the lateral border of the foot. This not only causes the patient discomfort because of localized weight bearing in a relatively small area, but the persistent varus position of the talar joint keeps the transverse tarsal joint in a semirigid state, resulting in a rather immobile forefoot that is difficult for the body to pass over during the stance phase.

Dorsiflexion

The degree of dorsiflexion should account for the patient's ability to clear the ground during swing phase as well as comfortably contact the ground during stance phase. In the absence of pathology at the adjacent joints, a neutral position is considered the position of choice. If the ankle is placed into too much dorsiflexion, then the impact of ground contact is concentrated in one small area of the heel, which may result in chronic pain. If the ankle is placed into excessive plantar flexion, then the involved limb is lengthened, which, in turn, causes a knee joint hyperextension, uneven gait pattern, and stress across the midfoot. However, sometimes plantar flexion is desirable. For example, if there is a short lower extremity or an unstable knee joint as a result of weakness or loss of quadriceps function, then the ankle joint should be placed into plantar flexion (10–15 degrees) to help give stability to the knee joint. After an ankle arthrodesis, patients usually develop increased motion in the sagittal plane, which helps to compensate for loss of ankle motion. In a study of 81 ankle fusions, the sagittal arc of motion of the talar-first metatarsal complex averaged 24 degrees (9–43 degrees), at the talonavicular joint 14 ± 5 degrees, and at the talocalcaneal joint 8 ± 6 degrees (Fig. 1-24).[19]

Hindfoot Alignment

Recall that the flexibility of the subtalar joint to go from inversion to eversion directly influences the stiffness of the transverse tarsal joint and the foot's ability to transition from being supple to being rigid during gait. Loss of subtalar joint motion may result from trauma, arthritis, congenital abnormality, and surgery. This loss of motion causes increased stress placed at the joints above (ankle) and below (transverse tarsal) the immobile joint. In turn, these changes brought on by subtalar joint stiffness may lead to chronic pain. In chronic states, increased stress caused by a stiff subtalar joint may result in secondary changes. For example, in some individuals, the ankle may take the form of a ball-and-socket joint (Fig. 1-25). And in patients with a subtalar coalition, beaking may occur in the talonavicular joint (Fig. 1-26).

Subtalar joint stiffness can also be due to iatrogenic causes, whether intentional or not. When a subtalar joint is fused, the transverse rotation that occurs in the lower extremity is partially absorbed in the ankle joint because it no longer can pass through the subtalar joint into the foot. The fixed varus or valgus alignment of the subtalar joint will affect the position of the forefoot. If the subtalar joint is placed into too much varus, then the forefoot is rotated into supination, and the weight-bearing line of the extremity then passes laterally to the calcaneus and fifth metatarsal. This causes increased stress on the lateral collateral ligament structure at the ankle joint and abnormal weight bearing along the lateral aspect of the foot. This position also holds the forefoot in a semirigid position, the patient must either vault over the foot or place it in external rotation to roll over the medial aspect.

For a subtalar joint arthrodesis, the position choice is valgus tilt of about 5 degrees because this permits satisfactory stability of the ankle

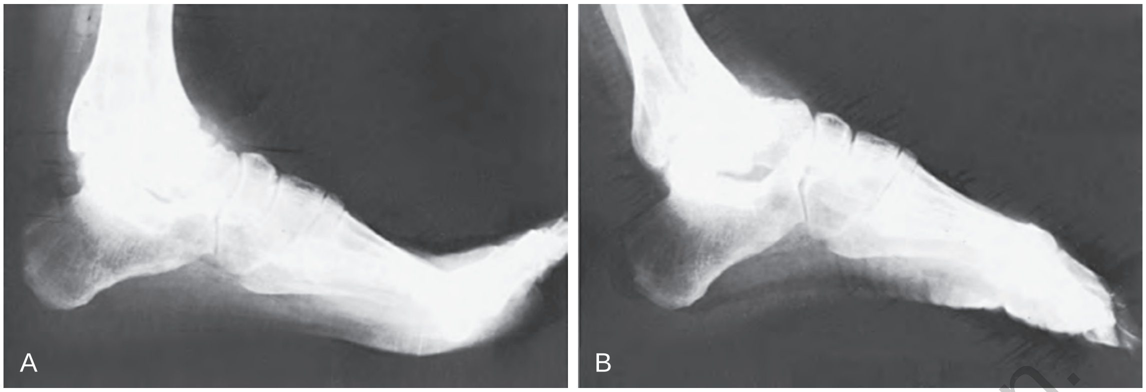

Fig. 1-24 Increased motion in transverse tarsal and subtalar joints to compensate for ankle arthrodesis. **A**, Dorsiflexion. **B**, Plantar flexion.

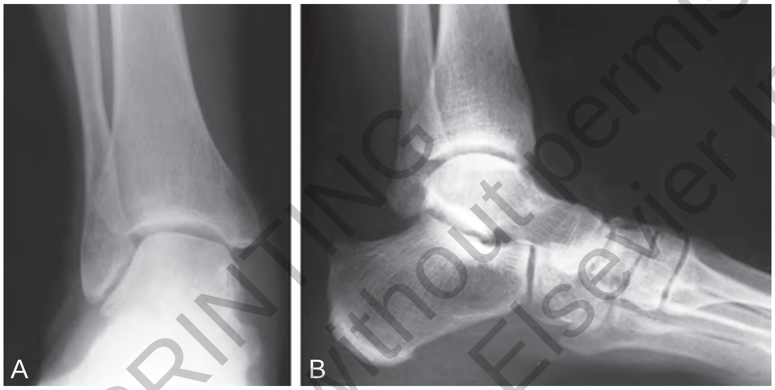

Fig. 1-25 Etiology of a ball-and-socket ankle joint in adults. **A,** As a result of a congenital abnormality of the subtalar joint that eliminated subtalar motion, the ankle joint absorbed transverse rotation that normally occurs in subtalar joint. **B,** A congenital talonavicular fusion, which results in loss of subtalar joint motion, causes the ankle to absorb transverse rotation, resulting in a ball-and-socket ankle joint.

joint, and the weight-bearing line of the body will pass medial to the calcaneus. Computational studies also suggest that this degree of valgus tilt maximizes ankle dorsiflexion and plantar flexion strength.[20] Stress on the lateral collateral ligament structure is therefore minimized. This position results in slight pronation of the forefoot, which permits even distribution of weight on the plantar aspect of the entire foot. And the slight valgus, or everted, position allows the transverse tarsal joint to remain unlocked, and therefore flexible, so that the body can more easily pass over it during stance phase. A more supple midfoot potentially mitigates the progression of arthritis at the transverse tarsal and more distal joints. On the other hand, an inverted, varus hindfoot would result in a more rigid midfoot that is less able to absorb impact energy during gait and be more prone to arthritis.

Midfoot Alignment

Just as subtalar joint motion influences flexibility at the transverse tarsal joint, so too does movement at the talonavicular or transverse tarsal joint complex affect subtalar joint motion.

When surgical stabilization of the talonavicular or transverse tarsal joint is performed, motion at the subtalar joint is largely eliminated. For motion to occur in the subtalar joint, rotation of the navicular over the head of the talus must occur. If it cannot, then there is essentially no subtalar joint motion. A cadaveric study revealed that with an isolated talonavicular fusion, the subtalar joint loses over 90% of its motion. In the same study, while isolated calcaneocuboid joint fusion is reported to lead to about a 30% loss in talonavicular joint motion, it leads to less than 10% loss of motion at the subtalar joint.[21] These findings underscore the importance of the talonavicular joint for both hindfoot and midfoot movement.

Of course, the transverse tarsal joint also affects the bones and joints distal to it, controlling forefoot position. A fixed, supinated transverse tarsal joint leads to overloading of the lateral column since the rigid medial column is elevated and cannot take equal share in ground forces. In order to place the forefoot in a plantigrade position, a neutral to slightly pronated placement of the transverse tarsal joint is required. In this position, the rigid medial column of the forefoot is allowed to contact the ground, thus sharing the load with the more flexible lateral column. The foot is therefore in a plantigrade position.

When a triple arthrodesis is carried out, the position of choice is 5 degrees of valgus for the subtalar joint and neutral position for the transverse tarsal joint.[20] It should be emphasized that it is better to err on the side of too much valgus and pronation to keep the weight-bearing line medial to the calcaneus, because that produces a more flexible plantigrade foot. These basic principles also apply when carrying out a pantalar arthrodesis.

Since intertarsal and tarsometatarsal joints of the medial column are naturally rigid, arthrodesis or surgical stabilization of these joints can be performed with minimum loss of function or increased stress on the other joints in the foot. When these joints are fixed, care should be taken to ensure that the medial column remains plantigrade on

Fig. 1-26 Talar beaking after increased stress as result of subtalar coalition.

weight bearing. The tarsometatarsal joints of the lateral column (at the fourth and fifth rays), on the other hand, are flexible, and efforts to maintain their flexibility are preferred to fusion in order to maintain suppleness in the forefoot.

Forefoot Principles

When operating on the forefoot, the surgeon needs to keep in mind the metatarsophalangeal break and the plantar aponeurosis, both described earlier in this chapter. Recall that the plantar aponeurosis is important for stabilizing the medial longitudinal arch through the windlass mechanism wherein the metatarsal heads act as a pulley over which the plantar aponeurosis is tightened when the metatarsophalangeal joints are dorsiflexed. This contributes to the rigidity of the foot as the body's weight passes over it during gait. Also recall that the metatarsophalangeal break refers to the cascade in the length of the metatarsal heads that allows for even distribution weight across the forefoot during stance phase.

With these principles in mind, surgeries on the forefoot can be thoughtfully planned. Removal of the proximal phalanx of the great toe or the first metatarsal head severely disrupts the plantar aponeurosis, leading to weight being transferred to the lesser metatarsal heads. Surgical techniques that remove the proximal phalanx base but preserve the plantar plate may lessen (but not eliminate) this effect. Similarly, removal of the proximal phalanx of the lesser toes will cause similar problems of instability, but to a lesser degree as one moves laterally across the foot. Removal of the proximal phalanges or metatarsal heads is generally reserved for extreme cases of deformity such as those that occur in rheumatoid arthritis.

Metatarsal osteotomies for conditions such as hammertoe, fracture nonunion correction, or avascular necrosis of the metatarsal head should be performed in a way that restores or preserves the metatarsophalangeal break. For example, in a patient with transfer metatarsalgia at the second metatarsal head due to the combination of having a long second metatarsal and a first ray condition such as hallux valgus or hallux rigidus, a shortening osteotomy may be performed. The resulting length of the second metatarsal should be just slightly longer, equal to, but not shorter than that of the third metatarsal lest there be resulting transference of pain to the adjacent third metatarsal. In general, when weight is transferred to adjacent metatarsal heads, increased stress and callus formation can ensue. In the neuropathic patient, this can have detrimental consequences including ulceration and infection.

When a transmetatarsal amputation is indicated, the bones should be resected in a way that preserves the metatarsal cascade so that weight is evenly distributed at the residual metatarsal lengths. Neglecting to do this will lead to wound problems, pain, and the need for revision procedures.

Arthrodesis at the first metatarsophalangeal joint for conditions such as hallux rigidus or recurrent hallux valgus can be performed while allowing a patient to maintain normal gait. The joint should be positioned in 10 to 15 degrees of valgus to allow for shoe wear, and 15 to 25 degrees of dorsiflexion in relation to the first metatarsal shaft to allow the foot to progress through terminal stance and into pre-swing. The degree of dorsiflexion may depend, to a certain extent, on the heel height of the shoe that the patient desires to wear. Arthrodesis of the first metatarsophalangeal joint has theoretical risk of resulting adjacent joint arthritis at the first tarsometatarsal and interphalangeal joints. Arthrodesis of the interphalangeal joint of the toes is thought to have negligible effects on gait biomechanics.

The sesamoid bones in the foot are embedded in the plantar plate complex of the first metatarsophalangeal joint and act to help distribute the weight seen at the first metatarsal head during gait and also act as fulcra to increase the lever arm for the flexor hallucis longus and brevis tendons. Generally, patients can return to athletic activity after isolated removal of one of these bones.[22] In cadaveric studies, it does not appear that hallux valgus or varus results when there is meticulous dissection of the bone.[23] However, in published case series, these deformities do occur.[22,24,25]

Tendon Transfers

When evaluating muscle weakness or loss about the foot and ankle, the diagram in Fig. 1-27 can be useful. It demonstrates the motion that occurs around each joint axis and the location of the muscles in relation to the axes. By considering the muscles in relation to the axes, it is possible to carefully note which muscles are functioning and thereby determine which muscles might be transferred to rebalance the foot and ankle. Generally speaking, if inadequate strength is present to balance the foot adequately, it is important to establish adequate plantar flexion function over that of dorsiflexion; an equinus gait is not as disabling as a calcaneal-type gait. Also keep in mind that it is much more difficult to retrain a muscle that has been a stance-phase muscle to become a swing-phase muscle than to retrain a swing-phase muscle to become a stance-phase muscle. Therefore if possible, an in-phase muscle transfer will produce a more satisfactory result because no phase conversion is necessary.

Ankle Ligaments

Recall the positions of the calcaneofibular and anterior talofibular ligaments. When the ankle is in neutral calcaneofibular ligament, insertion on the calcaneus is posterior and distal to its origin at the fibula. The anterior talofibular ligament, on the other hand, inserts distal and anterior to its origin. When one is evaluating the stability of the lateral collateral ligament structure, the ankle joint should be tested in dorsiflexion to demonstrate the competency of the calcaneofibular ligament and in plantar flexion to test the competency of the anterior talofibular ligament. If both ligaments are completely disrupted, then there will be no stability in either position.[26] Furthermore, to test for stability of the anterior talofibular ligament, the anterior drawer sign should be elicited with the ankle joint in a slight plantarflexed position, when the anterior talofibular ligament is in a position to resist anterior displacement of the talus from the ankle mortise (see Fig. 1-21).

As described earlier, the calcaneofibular ligament lies approximately parallel to the subtalar joint in the sagittal plane. Because motion in the subtalar joint occurs about an axis that deviates from dorsal medial to plantar lateral (see Fig. 1-6), and the calcaneal attachment of the calcaneofibular ligament lies on the subtalar joint axis, motion of the

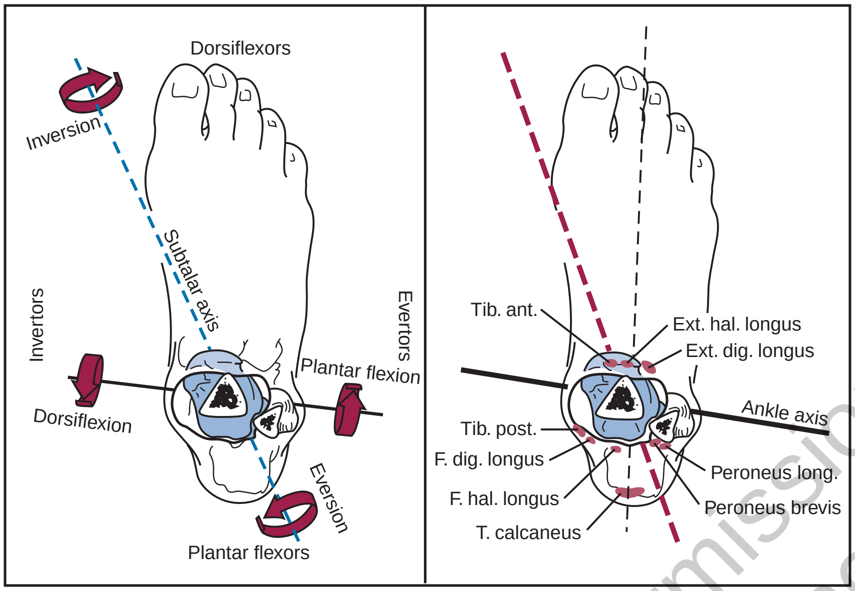

Fig. 1-27 *Left*, Diagram demonstrates rotation that occurs about subtalar and ankle axes. *Right*, Diagram demonstrates relationship of various muscles about subtalar and ankle axes. (From Haskell A, Mann RA. Chapter 23: Biomechanics of the Foot. In American Academy of Orthopaedic Surgeons: *Atlas of Orthoses and Assistive Devices*, ed 4, Philadelphia, 2008, Mosby)

subtalar joint around this axis occurs with minimal change in ligament length. Although it plays a secondary role to the interosseous talocalcaneal ligaments in stabilizing the subtalar joint, its position must be considered during a ligamentous reconstruction to ensure that motion is maintained across both the ankle and subtalar joints.

Rather than two ligaments, on the medial side of the ankle, there is a single fan-shaped deltoid ligament. This is sufficient to provide stability to the medial ankle because the configuration of the trochlear surface of the talus is curved, producing a cone-shaped articulation whose apex is medial. The medial surface area at this joint, is therefore smaller than the lateral (see Fig. 1-20). Reconstructing the deltoid ligament can have a role in preventing valgus deformity at the ankle and hindfoot.[27]

KINEMATICS AND KINETICS OF HUMAN LOCOMOTION

In this chapter, the reader began with an introduction to the basic phases of gait. Next, mechanical principles were illustrated at successive joints, starting from the ankle joint proximally and ending at the metatarsophalangeal joints distally. The passive and active modulators of these events were also introduced. From there, these principles were put together to synthesize biomechanical events as they relate to the phases of gait. Key features important for running gait were identified. And finally, basic clinical correlations were made to give the reader a foundation for connecting biomechanics of the foot and ankle to clinical practice. As a supplement to this knowledge, the chapter will conclude by expanding on the kinematics and kinetics of locomotion.

Kinematics

The main chapter provided details on foot and ankle biomechanics as they relate to gait. However, walking is more than merely placing one foot in front of the other. During walking, all major segments of the body are in motion. Displacements of the body segments occur in a well-preserved fashion and can be accurately described. Kinematics is the study of the motion of these body segments.

It seems almost too obvious to point out that each body segment affects and is affected by the others. Alterations in the foot and ankle will be reflected in patterns of movement in the other segments of the body. Similarly, changes in movements above the foot, such as a stiff arthritic knee, or quadriceps weakness, may be reflected by changes in the behavior of the foot.

It should be noted that while bipedal locomotion requires certain commonalities, each person exhibits individual differences that make for as many unique gaits as there are people. These differences allow us to be recognized by a friend or acquaintance, even from a distance. Accounting for this variation are the myriad combinations of body segment lengths, mass distributions, muscle composition, strength, joint axes, variations in effective lever arms, and the list goes on. At the same time, just as no two people walk exactly alike, gait kinematics will not always be identical even within the same individual.

Because of this diversity, average values of single anthropometric observation of gait kinematic parameters are alone of little value. Instead, it is more important to understand the functional relationship among the various components of kinematics. This is particularly true in the case of the foot, where anatomic variations are extensive. If average values are the only bases of comparison, then it becomes difficult to explain why some feet function adequately and asymptomatically, although their measurements deviate from the average, whereas others function symptomatically, even though their measurements approximate the average. Therefore this section emphasizes functional interrelationships and not on lists of kinematic measurements.

Vertical Body Displacements

The rhythmic upward and downward displacement of the body during walking is familiar to everyone and is particularly noticeable when someone is out of step in a parade. These displacements in the vertical plane are a necessary component of bipedal locomotion. When the legs are separated, as during transmission of the body weight from one leg to the other (double weight bearing), the distance between the trunk and the floor must be less than when it passes over a relatively extended leg, as during midstance.

Smoothing and minimizing vertical oscillations of the body's center of gravity minimizes energy expenditure. Physics principles tell us that much more energy is needed to lift the body against gravity and slow its descent (vertical displacement) than to move perpendicular to gravity's pull (fore–aft or lateral displacement). Because the nature of bipedal locomotion demands such vertical oscillations of the body, they should occur in a smooth manner. The center of gravity of the body does displace in a smooth sinusoidal path; the amplitude of displacement is approximately 4 to 5 cm[28,29] (Fig. 1-28). The body's center of gravity reaches its maximum elevation immediately after passage over the weight-bearing leg and then begins to fall. This fall is stopped at the termination of the swing phase of the opposite leg as the heel strikes the ground.

Much of the coordination of motion between the different segments of the lower limbs results in minimizing the vertical displacement of the body's center of gravity. Although movements of the pelvis and hip modify the amplitude of the sinusoidal pathway, the knee, ankle, and foot are particularly involved in converting what would be a series of intersecting arcs into a smooth, sinusoidal curve.[28] This conversion requires both simultaneous and precise sequential motions in the knee, ankle, and foot.

In a well-functioning system, the body's falling center of gravity is smoothly decelerated, because relative shortening of the leg occurs at the time of impact against a gradually increasing resistance. The knee flexes against a graded contraction of the quadriceps muscle; the ankle plantar flexes against the resisting anterior tibial muscle. After the foot-flat position is reached, further shortening is achieved by pronation of the foot to a degree permitted by the ligamentous structures within.

So, to reemphasize, hindfoot pronation constitutes an important additional factor to that of knee flexion and ankle plantar flexion needed to smoothly decelerate and finally to stop the downward path of the body. If one were forced to walk stiff-kneed or without a mobile foot and ankle, the downward deceleration of the center of gravity at heel strike would be instantaneous. The body would be subjected to a severe jarring force, and the locomotor system would lose kinetic energy.

After reaching its nadir, the center of gravity moves upward to propel it over the stance leg. The leg functionally elongates by transitory extension of the knee, further plantar flexion of the ankle as the heel elevates, and supination of the foot. Elevation of the heel is the major component contributing to upward acceleration of the center of gravity at this time.

Lateral Body Displacements

When a person is walking, the body does not remain precisely in the plane of progression but oscillates slightly from side to side to keep the center of gravity approximately over the weight-bearing foot. Watching someone walk from behind highlights this subtle side-to-side shift of their center of gravity toward the stance limb. When walking side by side with a companion, if one gets out of step with the other, their bodies may bump from this side-to-side sway.

The body is shifted slightly over the weight-bearing leg, with each step creating a sinusoidal lateral displacement of the center of gravity of approximately 4 to 5 cm with each complete stride. This lateral displacement can be increased by walking with the feet more widely separated and decreased by keeping the feet close to the plan of progression (Fig. 1-29). Normally, the slight valgus of the tibiofemoral angle (physiologic genu valgum) permits the tibia to remain essentially vertical and the feet close together while the femurs diverge to articulate with the pelvis, minimizing the lateral displacement.

Horizontal Limb Rotation

In addition to vertical and lateral displacements of the body, a series of axial rotatory movements occur that can be measured in the horizontal (transverse) plane. Rotations of the pelvis and the shoulder girdle are easy to see when watching someone walk. Similar horizontal rotations occur in the femoral and tibial segments of the extremities. The tibias rotate about their long axes, internally during swing phase and into the first interval of stance phase and externally during the latter phases of stance. The degree of these rotations is subject to marked individual variations. In a series of 12 male subjects, the recorded average horizontal rotation of the tibia was 19 degrees during a gait cycle but varied between 13 and 25 degrees.[6]

At heel strike, progressive inward rotation occurs in the lower extremity, which consists of the pelvis, femur, and tibia, and this inward rotation reaches a maximum at the time of foot flat. The internal rotation at heel strike is initiated by the collapse of the subtalar joint into valgus, and its magnitude is determined by the flexibility of the foot and its ligamentous support. After contralateral toe-off, at about 12% of the cycle, progressive outward rotation occurs, which reaches a maximum at the time of toe-off, when inward rotation resumes (Fig. 1-30). Once the foot is on the ground, progressive external rotation is probably initiated by the contralateral swinging limb, which rotates the pelvis forward, imparting a certain degree of external rotation to the stance limb. This external rotation subsequently is passed from the pelvis distally

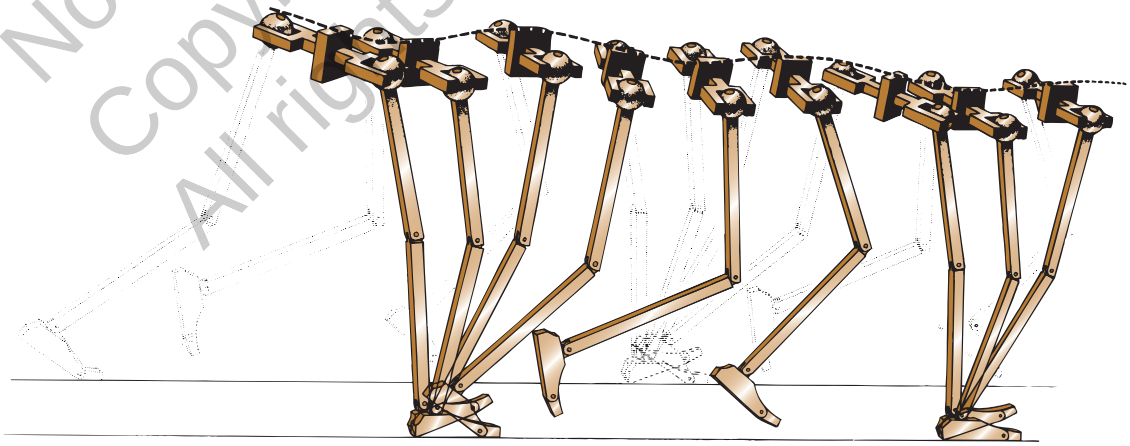

Fig. 1-28 Displacement of center of gravity of body in smooth sinusoidal path. (From Saunders JB, Inman VT, Eberhart HD: The major determinants in normal and pathological gait, *J Bone Joint Surg Am* 35A:543–558, 1953.)

Fig. 1-29 **A,** Slight lateral displacement of body occurring during walking with feet close together. **B,** Increased lateral displacement of body occurring during walking with feet wide apart. (From Saunders JB, Inman VT, Eberhart HD: The major determinants in normal and pathological gait, *J Bone Joint Surg Am* 35A:543–558, 1953.)

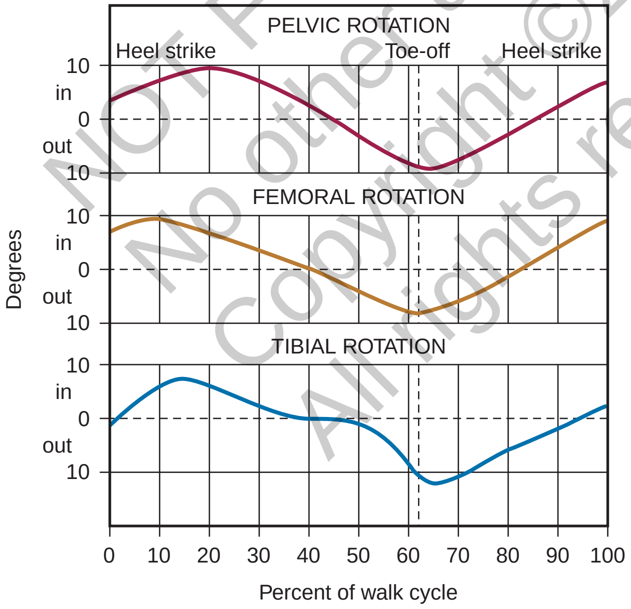

Fig. 1-30 Transverse rotation occurring in the lower extremity during walking. Internal rotation occurs until approximately 15% of cycle, at which time progressive external rotation occurs until toe-off, when internal rotation begins again.

to the femur and tibia, across the ankle joint, and is translated by the subtalar joint into inversion, which reaches its maximum at toe-off. The external rotation is enhanced by the external rotation of the ankle joint axis, the oblique metatarsal break, and the plantar aponeurosis after heel rise begins.

Kinetics

As with all movement, human locomotion occurs at the cost of energy expenditures. Gait kinematics and lower-extremity anatomic interrelationships strive to achieve a system that takes us from one spot to another with the least expenditure of energy.[30,31] Human locomotion is a blending of physical and biologic forces that combine to achieve maximum efficiency at minimum cost. For example, energy conservation during the gait cycle involves having muscles work near their peak efficiency, which tends to be at or longer than their resting length.[7,28,30] Activated muscles are approximately six times as efficient when resisting elongation (eccentric contraction) as when shortening to perform external work.[32–34] Kinetics is the study of these energy expenditures. A number of tools have evolved to study gait kinetics. These are described in detail in the next section, followed by an analysis of kinetics during gait.

Measuring Whole Body Kinetics and Plantar Pressure

Studying the foot's interaction with the ground has a long history, ranging from examining footprints in soil to real-time mapping of plantar pressure under natural conditions. Plantar pressure and ground reaction force measurements are well established in the research realm and have been instrumental in refining our understanding of foot and ankle biomechanics. In conjunction with other technology, including high-speed cameras, video motion-sensing equipment, electrogoniometers, and electromyograph (EMG) devices, the study of the ground-foot interaction has aided the understanding of gait kinetics and kinematics.

Despite improvements in available measurement methods, however, practical collection of clinically novel information remains difficult. The wide variability of normal measures makes clinical comparisons difficult. The large number of measurement systems and equally large number of data analysis techniques make it difficult to generalize results. Although confirmation of areas of excess pressure and monitoring the effects of treatment may prove useful, there is little specificity between plantar pressure patterns and clinical syndromes.

Types of studies. A variety of measurement techniques have been used to study the interaction of the foot with the ground. Indirect techniques rely on correlating other measurable gait parameters to plantar characteristics and offer the advantage of not relying on expensive and often bulky equipment. For example, an estimation of ground reaction force can be made based on a simple-to-measure temporal variable, foot–ground contact time.[35]

Direct measurement techniques rely on physical properties or electronic transducers to translate the interaction between the foot and the ground into a measurable quantity. Multiple direct measurement systems are available that use a variety of strategies to record plantar pressure or ground reaction force. Unfortunately, results obtained with different systems under similar conditions are not always similar, and even qualitative comparisons may not be appropriate.[36] Spatial resolution and sample rate affect the ability of a system to record true peak plantar pressures and to isolate particular areas under the foot.

The earliest direct measurement methods relied on physical properties of a material to capture the interaction of the foot with the ground. Casts of the foot in clay, plaster, or soil were used with the assumption that areas of deeper penetration represented areas of highest pressure.[37,38] Rubber mats incorporating longitudinal ridges,[39] pyramidal projections,[38] or a multilevel grid (such as the Harris-Beath mat)[40,41] use the elastic property of rubber that, when stood or walked on, distorts in proportion to the pressure applied (Fig. 1-31). Although fast, inexpensive, and portable, these methods have low measurement resolution and lack temporal discrimination.[40]

Optically based systems rely on visualizing the plantar aspect of the foot during stance or gait. The simplest allows observation or photographic recording of the plantar foot through a clear platform (Fig. 1-32). This provides an accurate, dynamic, qualitative representation of foot morphology. Addition of a physical transduction device between the foot and glass plate allows quantification of regionalized plantar pressures and adds the temporal component missed using a physical transduction system alone.[38] The pedobarograph places a thin plastic sheet over the clear plate.[42] The sheet is illuminated at the edges, and pressure on the plastic distorts the light in proportion to the pressure applied. The images can be recorded and calibrated to provide a spatial resolution and temporal responsiveness not found with the Harris-Beath mat. However, slow responsiveness at high forces may bias results.[43]

A force plate measures the ground reaction force, that is, the force exerted by the ground on the foot, in three degrees of freedom (vertical force, forward shear, side shear) and allows calculation of the torques around the foot and ankle (axial torque, sagittal torque, coronal torque). Force transducers are configured in orthogonal planes at the corners of a section of floor. The resulting data provide a representation of the

Fig. 1-31 Pressure distribution on plantar aspect of foot as demonstrated by use of barograph. As dots get larger and denser, pressure distribution is greater. (From Elftman H: A cinematic study of the distribution of pressure in the human foot, *Anat Rec* 59:481–491, 1934.)

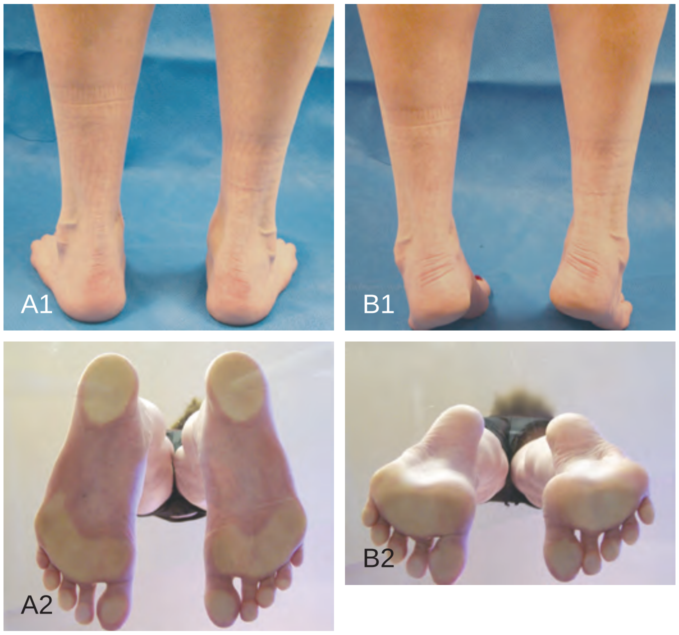

Fig. 1-32 Feet and legs of person standing on barograph. **A**, Weight bearing with muscles relaxed. **B**, Rising on toes.

average forces experienced by the foot over the gait cycle (Fig. 1-33). One advantage of this type of system is that shear forces and torques can be measured in addition to vertical force. The limitations include the lack of ability to map specific regions of plantar pressure. This limitation can be circumvented with the addition of an optical diffraction system, as described above, or with a series of smaller force plates placed in tandem.[44]

The ability to place pressure transducers on discrete parts of the foot has become possible, as their size has shrunk. They can be placed on strategic points of the foot, or an array can be created to map the pressures exerted by the foot during stance or gait. These data provide a spatial and temporal map of plantar pressure over the gait cycle[39,45] (Fig. 1-34). Many of these systems use a floor mat or platform built into the floor with a grid of pressure-sensitive transducers. An alternative is to place a thin film containing a pressure transduction array into a shoe (Fig. 1-35). In this way, the plantar pressures experienced by the foot can be measured in a wider variety of settings and under multiple impacts as well as account for the effect of shoe wear.[46,47] For example, feet experience 10% to 50% higher plantar pressures in a flat, flexible shoe compared with a soft shoe with a firm rubber sole.[48] The floor mat and in-shoe methods correlate well when the shoe used has a firm sole or when barefoot.[49]

A number of system-specific and analysis-dependent factors affect the results of pressure transducer array measurements, including pressure transducer density, responsiveness, linearity, resolution, and range of the transducers. Methods of analyzing the data also differ, including reporting results as force versus pressure, peak values versus sum of values over time, and strategies of regionalizing the foot's plantar surface. Increasing pressure transducer density provides better spatial representation of plantar pressure, whereas systems with relatively lower transducer density may underestimate measurements, such as peak pressure, because the true peak may be missed. Some transducers may have a nonlinear response at the extremes of their measurable range or have a low-level cutoff. The maximum sample rate affects contact time measurements, and low sample rates may underestimate peak pressure measurements because the true peak pressure may be missed.

Recently, more studies have been published using finite element modeling as a method to predict parameters associated with gait kinetics. While a detailed discussion of this method is beyond the scope of this chapter, it can be simplified as follows. In the case of studying foot and ankle biomechanics, a geometric model of the foot and ankle is generated from 3D imaging (usually computed tomography or magnetic resonance images). Theoretically, at every spatial location (or element) within this geometric model, an output parameter, such as the amount of pressure experienced at that location, can be modeled as a mathematical function of multiple variables. There are infinite elements that can be modeled with such an equation. In finite element modeling, only a set number of elements is studied. The geometric model is thus fitted with a "mesh" connecting each of the studied elements. A coarser mesh has fewer data points but is more easily studied; a finer mesh provides higher resolution at the cost of greater computational time and energy. In this method, "boundary" and "loading" conditions—the forces applied in the model—are usually derived from gait analysis methods described above. Model validation is typically achieved by comparing the model output with in vivo measurements.[50]

Data representations. Output from the different measurement systems reflects the nature of their measurement mechanisms. The Harris mat reports pedal pressure but does not vary with time. The force plate reports a true ground reaction force but is not spatially discriminative. The optical systems and the transduction arrays each report pedal pressure that varies with time. The data measured by these systems are subject to sensor density, resolution, and sample rate limitations discussed above. To simplify the information and allow comparisons between subjects or after treatments, a variety of derivative parameters have been defined based on these raw data. Not all systems or measurement methods are able to derive all of these measurements.

The ground reaction force is a vector quantity varying temporally and spatially over the gait cycle that represents the average reciprocal force exerted by the floor in response to the foot. It has a magnitude and direction, and the starting point may be projected onto a representation of the plantar foot at the point of average maximum vertical force (Fig. 1-36). The ground reaction force can be deconstructed into vertical force, anterior–posterior shear, and medial–lateral shear. The vertical ground reaction force represents the force of the ground pushing upward on the foot and can be calculated from systems that measure plantar pressure for the whole foot or for defined regions of the foot.[51] Typically, it has two peaks; the first peak occurs as the body weight is transferred from dual- to single-leg stance and the second as the body weight moves forward over the metatarsal heads. Studies of ground reaction forces may focus on the magnitude of one or the other vertical peak or the timing of the peaks and valleys. Torque (moment) and power around a joint also can be calculated from the ground reaction force, joint geometry, timing parameters, and kinematics.

Another frequently reported measurement is the maximum pressure recorded, or peak pressure. It is usually reported over a spatially subdivided map of the plantar foot. Peak pressure for areas such as heel, individual or grouped metatarsal heads, and toes are common. Alternatively, peak pressure can be reported as a temporally varying measure by displaying its location and magnitude on a diagram of the foot. Peak force can be calculated from peak pressure because the size of the pressure transducers is known. Calculated joint moments represent the torque applied by muscles to counteract the measured ground reaction force, and joint power is calculated from the joint moment and angular velocity (Fig. 1-37).

Timing measurements can also be made. The time intervals from heel strike to metatarsal strike, toe strike, heel-off, metatarsal-off, and toe-off can be calculated. The pressure·time integral, or impulse, for the

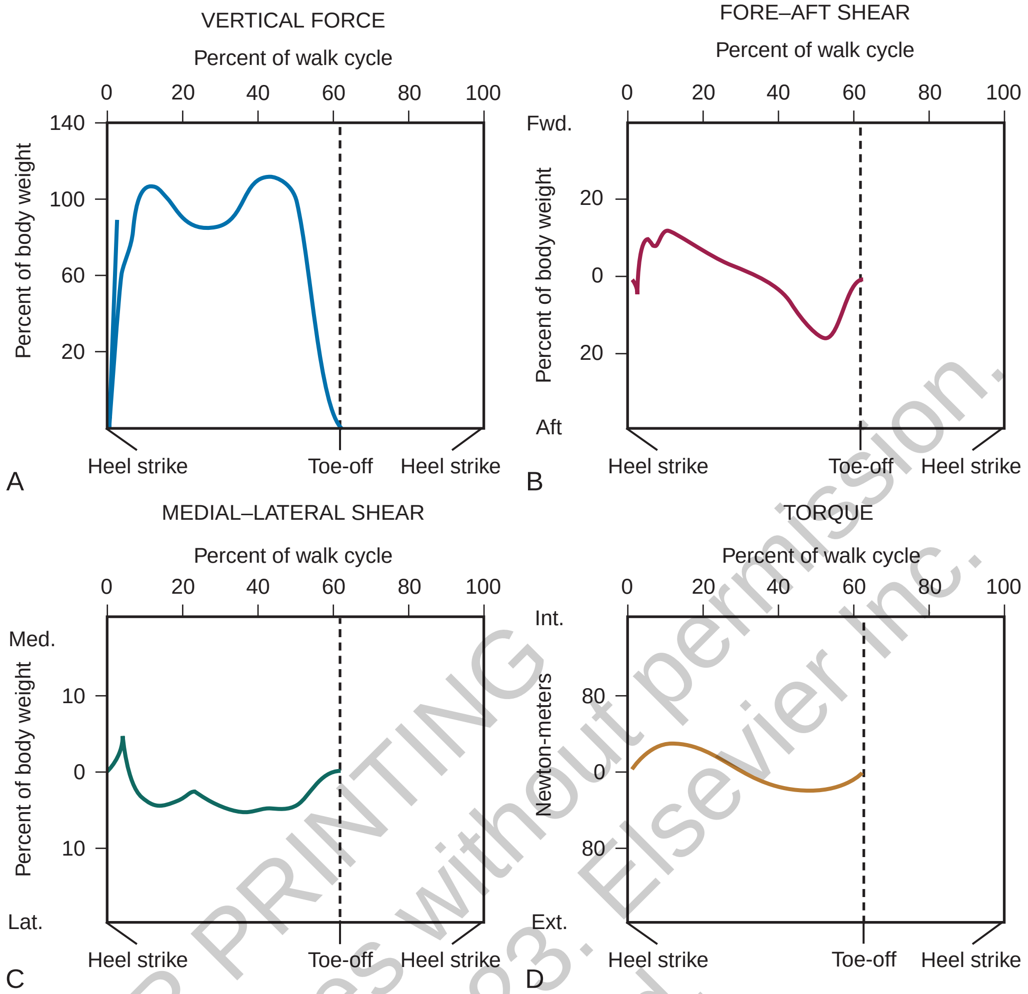

Fig. 1-33 Ground reaction to walking. **A**, Vertical force. **B**, Fore–aft shear. **C**, Medial–lateral shear. **D**, Torque. *Ext.*, External; *Fwd.*, forward; *Int.*, internal; *Lat.*, lateral; *Med.*, medial.

whole foot or defined regions can be calculated. This may be standardized for each region as a percentage of the total impulse for a given foot. The impulse may characterize plantar loading better than peak pressure by taking both pressure and time into consideration.

Finally, the pattern of plantar loading can be categorized based on the pressure measurements. Patients may tend to load the medial ray, the medial and central rays, the central rays, or the central and lateral rays.[52] Put another way, there is an inverse relationship between peak pressure under the first metatarsal head and toe relative to the lesser metatarsal heads.[53] As walking speed increases, a medialization of forefoot pressure occurs such that peak pressure under the first metatarsal head increases and that under the lesser metatarsal heads decreases.[54]

Measurement variability. Many sources of variability affect the results of these measurements. Separating important clinical or research findings from differences based on testing apparatus, measurement methodology, patient demographic factors, or analysis methodology requires an understanding of how these factors affect the measured results. Differences between the different testing apparatus have been described above. Other sources of variability can be divided into methodology, analysis, and patient-specific factors.

Walking speed affects the magnitude of plantar pressures during gait. Velocity is linearly related to peak vertical and fore–aft ground reaction forces[55,56] and inversely related to the pressure·time integral.[57] As velocity increases, peak pressures on the heel, medial metatarsal heads, and the first toe increase while peak pressure in the fifth metatarsal head decreases.[52,54] This medialization may be related to increased magnitude and velocity of hindfoot eversion and medial shear force at heel strike. Timing measurements also change with increasing speed. The normalized time to peak pressure is decreased on the heel but unchanged in the midfoot and forefoot, suggesting the rollover process is mainly accelerated by reducing the time from heel strike to foot flat.[54] To minimize variability introduced because of walking speed, subjects may walk at a fixed rate or at their natural pace.[57]

Deviations from a normal gait pattern can occur if the subject has to take a long or short stride in an effort to place the foot on the appropriate measurement area of floor-based systems. To minimize this effect, the measurement platform is placed flush with the floor and hidden from the subject with a thin, uniform floor covering. The traditional midgait method uses a short lead-up walk before the foot strikes the measurement platform. A three-step or two-step lead-up is as reproducible, but a one-step lead-up is not adequate.[58,59]

Variability of the measurements is also dependent on the type of gait. For example, plantar pressures measured when standing differ from pressures measured during gait.[37] Variations in walking patterns, such as a shuffling-type gait, alter the peak forces on the foot.[60] Gait pattern alteration can be seen in certain conditions, such as after ankle fracture fixation or with concurrent knee pathology.[49,55]

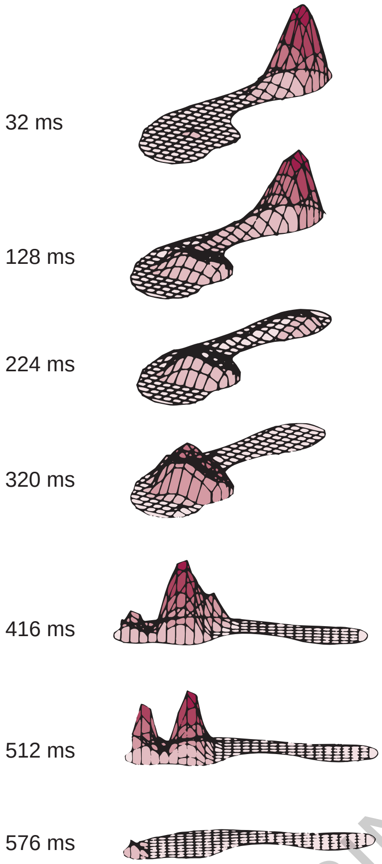

Fig. 1-34 Pressure distribution under bare foot during walking. Height of display above ground is proportional to pressure. (From Clarke TE: *The pressure distribution under the foot during barefoot walking* [doctoral dissertation], University Park, PA, 1980, Pennsylvania State University.)

Drift and calibration of the measurement systems affect the variability of measurements. Plantar pressure measurement systems need to be calibrated to allow comparisons between systems. Transducer output varies between different transducers, with temperature, when an in-shoe system is removed and reinserted, and with the number of trials performed. Pressure can vary by as much as 20% with repeated measurements on the same insert.[48] There may be an offset that drifts with time.[59] The measurements may be adequate for relative ranking purposes but need repeated calibration with a fixed system if accurate values are needed.

Variability is also introduced in the methods by which the acquired data are analyzed. For example, peak pressure can be reported for the whole surface of the foot during a gait cycle, but the clinical utility of this is limited because different regions of the foot experience different plantar pressures during the gait cycle. Subdividing the regions of the plantar foot and recording peak pressures in each of these areas over the gait cycle provides more meaningful data. The heel is often represented as a single region but may be subdivided into medial, central, and lateral.[54] Midfoot peak pressures may be useful in pathologic conditions, such as rocker-bottom deformity, and can classify foot morphology into planus, normal, and cavus categories.[54] The base of the fifth metatarsal can be included as part of the midfoot or can be identified as a separate pressure zone. Improvements in sensor technology have allowed measurement of individual metatarsal heads and toe forefoot pressures.[53,61] Definition of these regions (masks) is still a manual process and is repeated for each trial. Having a single person define the regions may decrease variability.[61]

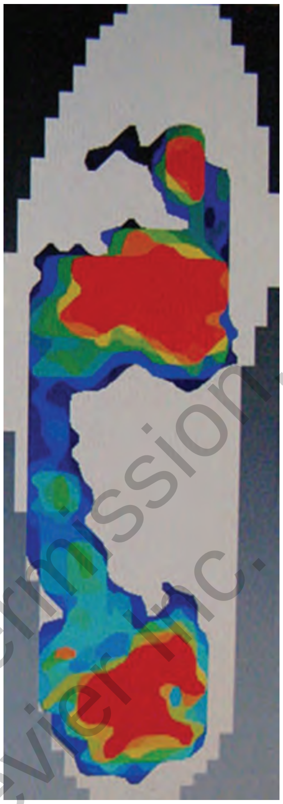

Fig. 1-35 Peak plantar pressure map using an in-shoe thin-film pressure transducer. *Red* represents areas of relatively high pressure, and *violet*, areas of low pressure. (Courtesy Ken Hunt, MD.)

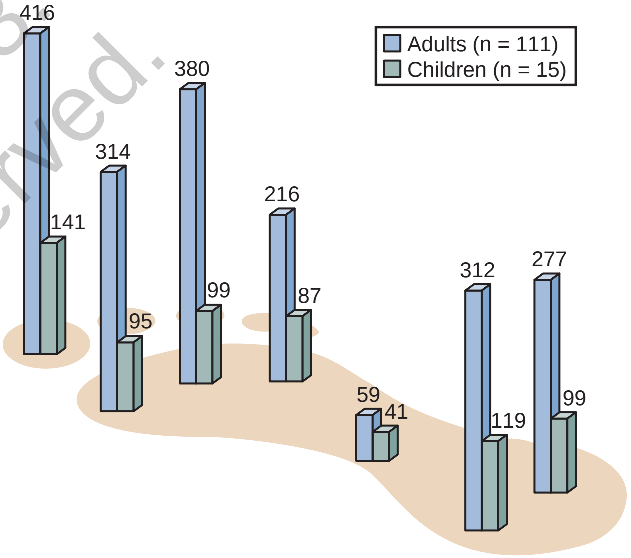

Fig. 1-36 Peak pressure values under selected foot regions demonstrate impact in heel region, minimal weight bearing in midfoot, buildup of pressure beneath metatarsal heads, and transfer of weight to great toe region. (From Hennig EM, Rosenbaum D: Pressure distribution patterns under the feet of children in comparison with adults, *Foot Ankle* 11:306–311, 1991.)

Subject-specific characteristics also introduce variability. Children's feet have a dramatically different loading pattern and lower peak pressure because of high relative foot area[62] (see Fig. 1-36). Differences in joint mobility and forefoot pressure based on a subject's ethnicity have

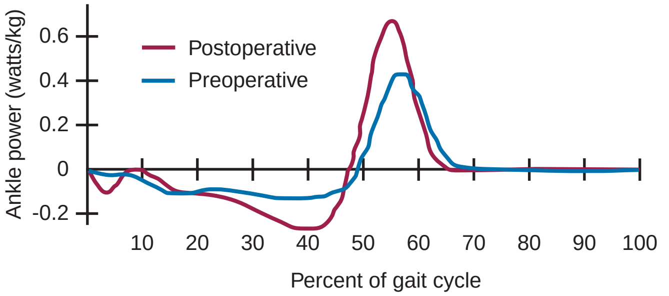

Fig. 1-37 Joint power generated (positive) or absorbed (negative) during the gait cycle before and after total ankle replacement. (From Brodsky JW, Polo FE, Coleman SC, Bruck N: Changes in gait following the Scandinavian total ankle replacement, *J Bone Joint Surg Am* 93:1890–1896, 2011.)

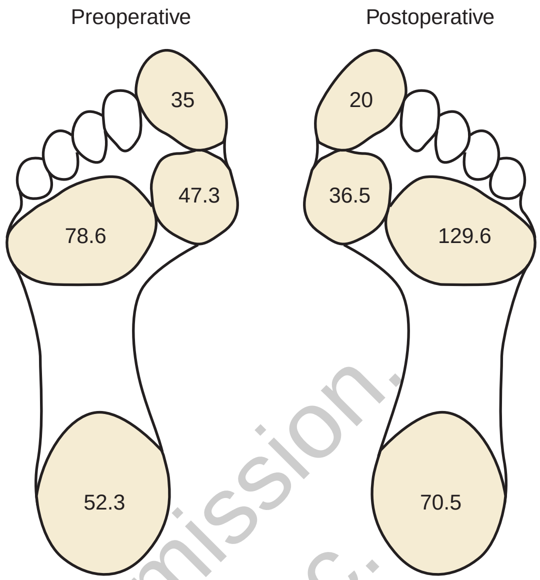

Fig. 1-38 Peak forces (in newtons) measured in four areas of foot before and after silicone arthroplasty of first metatarsophalangeal joint. Preoperatively, there is significant weight bearing by first metatarsal and great toe relative to lateral metatarsals. Postoperatively, there is decreased weight bearing by first metatarsal and great toe and increased weight bearing beneath lesser metatarsal head region. This demonstrates effect of loss of windlass mechanism, by which pressure is transferred to great toe, which, in turn, depresses first metatarsal head. (From Beverly MC, Horan FT, Hutton WC: Load cell analysis following silastic arthroplasty of the hallux, *Int Orthop* 9:101–104, 1985.)

been shown in neuropathic diabetics.[63] The patient's dominant side may experience greater static and dynamic vertical force,[46] although others have found no side dominance.[64] Foot morphology also affects plantar pressure; cavus feet have different midfoot loading characteristics and rate and degree of hindfoot eversion than flatfeet.[54] During running, fatigued subjects tend to have decreased step time, decreased peak and integral force and pressure under heel, and medialization of forces.[47] After a hindfoot fusion, greater contact force at heel strike has been observed.[65,66] This could be due to the inability of the calcaneus to move into a valgus position after heel strike.[67]

The effect of body weight on plantar pressure is less direct than might be expected. Although some have correlated maximum vertical force during gait and body weight,[46] many other studies found little correlation.[36,62,68] In children, the correlation of body weight to peak plantar pressures is clear and plays a greater role in determining peak pressure than in adults.[52,62] The area of peak pressure most highly correlated with body weight in children is the fourth metatarsal head[52] and in adults may be the fourth metatarsal head or the midfoot.[52,62]

Individuals load the foot with different spatial patterns as well. After heel strike, the forefoot may be loaded more medially or laterally across the metatarsal heads and may load the metatarsals and toes simultaneously or in turn. A variety of classification systems have been proposed to group these types of loading, and biomechanical theories have been proposed to explain the different loading patterns.[51,52,61] Finally, there is an inherent variability in an individual's gait from step to step that ranges from less than 1% for vertical ground reaction force to much higher for timing-dependent variables and values calculated as a product of measures.[64] Measured values may vary by more than 10% under identical testing conditions. Averaging data from as few as three trials improves the reliability of the measurement.[69]

Kinetics of Walking

Force plates measure the force felt by the floor produced by displacement of the body's center of gravity. By Newton's law of equal and opposite forces, this is the same force experienced by the foot and represents the effect of gravitational forces on the whole body while walking.[9] The principle of the force plate is seen when one stands on a bathroom scale and flexes and extends the knees to raise and lower the body. The measurement fluctuates abruptly as vertical floor reaction is registered.

Plantar pressure kinetics. Research on plantar pressure during gait has proved useful in numerous clinically relevant areas, including forefoot pressure involving a number of clinical syndromes. Increased forefoot pressures may lead to metatarsalgia or neuropathic ulceration and is mitigated by simple insole modifications. Diabetic and neuropathic foot ulceration correlate with areas of increased vertical and shear forces.[70] The weight-bearing pattern in these patients tends to shift from the medial to the lateral border of the forefoot, and the load taken by the toes is reduced.[71] The rheumatoid foot demonstrates similar findings.[67] A soft pad placed proximal to the metatarsal heads decreases metatarsal head pressure from 12% to 60%.[72] Placement of a half-inch lateral heel wedge decreased pressure under the third through fifth metatarsal heads by 24% and increased pressure under first and second metatarsal heads by 21%.[48] A half-inch medial heel wedge decreased the pressure under the first and second metatarsal heads by 28% and under the first toe by 31%.

Patients with hallux valgus may develop transfer metatarsalgia as plantar pressure increases under the lesser metatarsal heads and decreases under the first toe in relation to the size of the deformity.[73,74] This is associated with impaired maximum walking speed.[75] Those patients with hallux valgus and lesser toe metatarsalgia have greater peak pressure and peak pressure·time integral under the second through fifth metatarsal heads than those without metatarsalgia.[76] Measurement of plantar pressure may be predictive because no patients with less than 20 N/cm[67] peak pressure had metatarsalgia, and all patients with more than 70 N/cm[67] peak pressure had metatarsalgia. Hallux valgus correction with proximal first metatarsal osteotomy and distal soft tissue procedure decreases peak pressure under the second and third metatarsal heads.[77] After a distal chevron osteotomy for mild-to-moderate hallux valgus, the degree of plantar displacement of the distal first metatarsal osteotomy correlates with increased pressure under the first metatarsal head and to a decrease in clinical metatarsalgia.[78] Procedures that destabilize the first metatarsophalangeal joint, such as Keller resection arthroplasty and silicone (Silastic) implant arthroplasty, increase pressure on the lesser metatarsal heads[74,79–81] (Fig. 1-38).

The Achilles tendon and plantar fascia also influence plantar pressure and gait biomechanics.[82] The Achilles tendon contributes to heel rise, leading to a reduction in the vertical displacement of the center of gravity and minimizing energy expenditure.[83] During the stance phase, energy is stored in the gastrocnemius–soleus complex as the ankle dorsiflexes, and the tendon is elastically stretched and is returned after heel rise as the ankle plantar flexes. This elastic recoil facilitates shortening of the gastrocnemius–soleus complex at rates well above those possible by maximal muscle contraction and allows the muscles to act at a rate and length of maximum efficiency over the gait cycle.[84,85] Gastrocnemius–soleus work increases with step length, effectively lengthening the limb by plantar flexing the ankle.[85] A chronically elongated or ruptured tendon leads to a paradoxically rigid ankle by recruiting other ankle stabilizers.[17] The time to initial peak vertical force is shortened, highlighting a loss of shock absorption, but the second peak vertical force, representing metatarsal head pressure, is not diminished.[16] In diabetic patients with plantar ulceration, adding Achilles tendon lengthening to total contact casting leads to increased rate of healing and decreased recurrence of neuropathic ulcers.[86] Ankle dorsiflexion is increased, and both plantar-flexion torque and peak plantar pressure are reduced after Achilles tendon lengthening initially, but plantar-flexor torque and peak pressure return by 7 months even though accentuated dorsiflexion remains.[86] This suggests that the decrease in peak plantar pressure may be related to a weakening of ankle plantar flexors rather than to an increase in ankle dorsiflexion.

Whole body kinetics. The only forces that can produce motion in the human body are those created by gravity, by muscular activity, and, in a few instances, by the elasticity of specific connective tissue structures. A force plate instantaneously records the forces imposed by the body through the foot onto the floor. These measurements include vertical floor reactions, fore and aft shears, medial and lateral shears, and horizontal torques. During the stance phase of walking, the floor reactions in all four categories are continuously changing. Fig. 1-33 demonstrates the force plate data obtained during normal walking. The slower an individual walks, the less the center of gravity moves, and the resultant forces are less. Conversely, the faster the gait, the greater the movement of the center of gravity, and hence a larger force is experienced. When shoes are donned, these forces are transmitted through the interface between the sole of the shoe and the walking surface. This can attenuate rapid spikes, such as the heel striking the ground, and distribute the force over a larger area of the foot, diminishing peak plantar pressures.

The vertical element of ground reaction force is the largest of the component vectors and represents the force required to oppose the pull of gravity. It demonstrates an initial spike and rapid decline as the heel contacts the ground. Shoe material can alter the magnitude of the spike: a softer heel will result in a smaller initial spike and a harder heel in a larger spike. The vertical ground reaction force curve then has two peaks during the stance phase. The first whole body vertical force peak is 10% to 15% greater than body weight and is caused by the upward acceleration of the body's center of gravity. This is followed by a dip to approximately 20% less than body weight as the center of gravity reaches the top of its trajectory and begins to fall. A second peak of 10% to 15% greater than body weight results from resisting the falling of the center of gravity as the body moves over the stance leg. After this, the force rapidly declines to zero at toe-off as weight transfers to the opposite limb (see Fig. 1-33A). We see from the lack of a vertical ground reaction peak at the end of stance phase that the toes do not push off but rather are lifted from the floor as the weight transfers to the other side.

Forward shear occurs at initial heel strike representing the braking of the body as it resists forward momentum. After the center of gravity has passed in front of the weight-bearing foot, an aft shear occurs. The aft shear reaches a maximum as the opposite limb strikes the ground at 50% of the walking cycle. The aft shear approaches zero at the time of toe-off, once again showing the lack of push off during normal walking gait. The magnitude of the fore–aft shear, however, is only about 10% to 15% of body weight (see Fig. 1-33B).

Medial shear is the force exerted toward the midline at the time of heel strike, after which there is a persistent lateral shear until opposite heel strike at 50% of the cycle. A medial shear is not seen in persons with an above-knee amputation in whom a lateral shear mode is always present because of lack of abductor control of the prosthesis. The magnitude of the medial–lateral shear is about 5% of body weight (see Fig. 1-33C).

Lower extremity rotation during the stance phase causes a torque of the foot against the ground. After heel strike, there is an internal torque that reaches maximum at the time of foot flat, after which there is a progressive external torque that reaches a maximum just before toe-off. This torque corresponds to the inward and outward rotation of the lower extremity (see Fig. 1-33D). The majority of this rotation occurs with the foot firmly placed on the floor. The rotations, therefore, generate an internal torque of 7 to 8 newton-meters, which is of considerable magnitude.[9] The ankle and subtalar joints facilitate the transmission of rotatory forces between the foot and lower limb.

The movement of the ground reaction force vector along the bottom of a normal foot follows a consistent pattern[87] (Fig. 1-39). After heel strike, it moves rapidly forward until it reaches the metatarsal area, where it dwells for about half of the stance phase, then passes distally to the great toe. In a patient with a rheumatoid arthritis–related hallux valgus deformity and significant metatarsalgia, the center of pressure remains in the posterior aspect of the foot, avoiding the painful metatarsal area, then rapidly passes over the metatarsal heads along the middle of the foot[88] (Fig. 1-40). In patients with amputation of the great toe, the center of pressure passes in a more lateral direction[89] (Fig. 1-41).

Kinetics of Running

The forces involved during running are considerable, reaching 2.5 to 3 times body weight (Fig. 1-42). The larger forces generated are related to increased displacement of the center of gravity as the speed

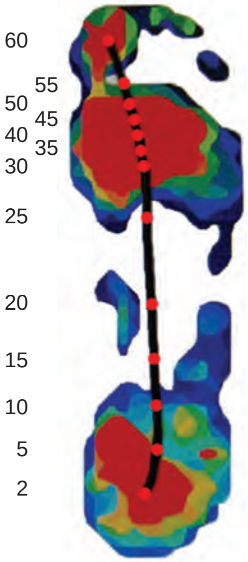

Fig. 1-39 Peak plantar pressure map with superimposed path of instantaneous center of ground reaction force *(black line)*. The *red dots* and corresponding labels represent the location of the ground reaction force at a given percentage of the gait cycle.

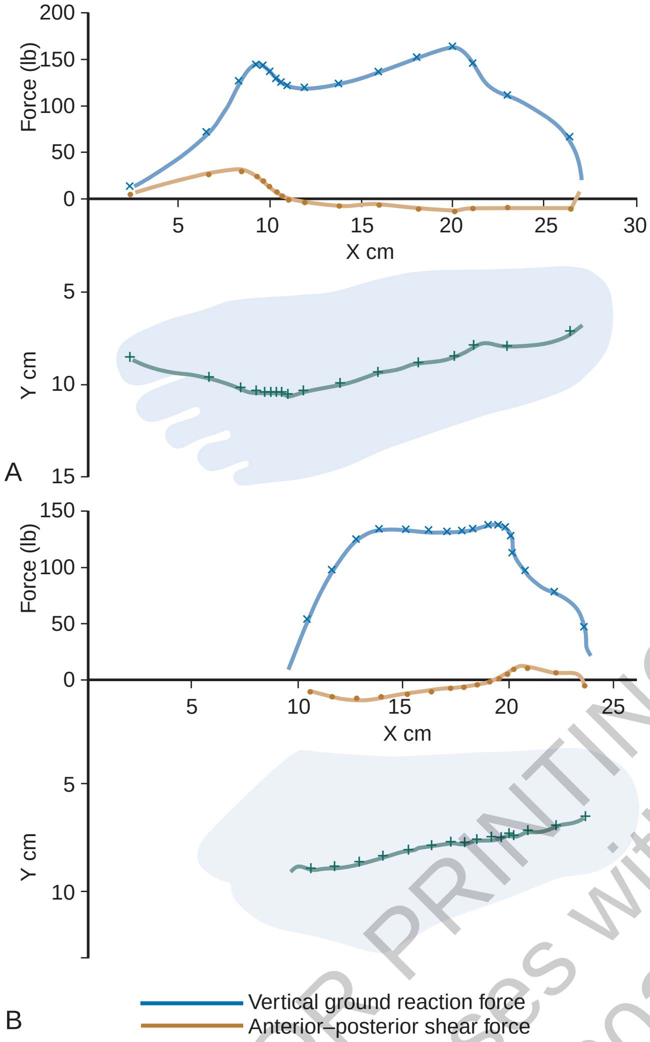

Fig. 1-40 Progression of center of pressure in normal and abnormal foot, beginning at the right and progressing to the left. *Blue line* is the vertical component of ground reaction force, and *tan line* is anterior–posterior shear component of ground reaction force. Marks along the path represent -second intervals. **A**, Note progression of center of pressure from heel toward toes during normal walking cycle. The center of pressure moves rapidly from the heel, dwells in metatarsal head region, then passes rapidly to the great toe at toe-off. **B**, Progression of center of pressure in a patient with rheumatoid arthritis with severe hallux valgus deformity and significant metatarsalgia. Note that the center of pressure remains toward the heel, then rapidly progress across the metatarsal head area with little or no pressure borne by the great toe. Patients with rheumatoid arthritis or significant metatarsalgia keep their weight in the posterior aspect of foot to avoid pressure over the painful portion of foot, which may lead to a shuffling gait. (From Grundy M, Tosh PA, McLeish RD, Smidt L: An investigation of the centres of pressure under the foot while walking, *J Bone Joint Surg Br* 57:98–103, 1975.)

of gait increases. At initial ground contact, increasing the range of motion at the ankle, knee, and hip joints helps absorb these larger forces. As the speed of gait further increases, the degree of motion in these joints also increases to help absorb the added impact. Muscles are active over a greater percentage of the gait cycle during running. The gastrocnemius–soleus contribution to forward propulsion is minimal during normal walking but plays a larger role as walking speed increases.[90]

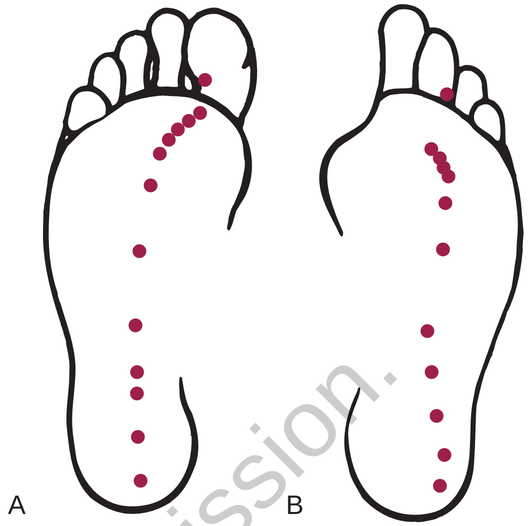

Fig. 1-41 Movement of center of pressure after amputation of great toe. **A**, Normal progression of center of pressure. **B**, Abnormal movement of center of pressure after amputation of great toe. Note that pressure tends to dwell more laterally in the metatarsal area, then passes out toward the third toe rather than the great toe. (From Mann RA, Poppen NK, O'Konski M: Amputation of the great toe. a clinical and biomechanical study, *Clin Orthop* 226:192–205, 1988.)

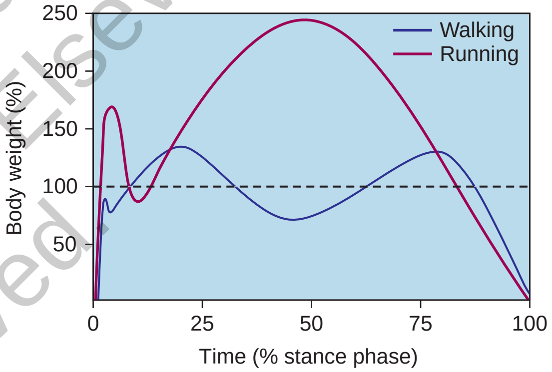

Fig. 1-42 Comparison of vertical ground reaction for walking *(blue line)* and jogging *(red line)*. The horizontal axis is scaled as a percentage of total time in stance phase for walking (0.6 sec) and running (0.24 sec). The vertical axis is shown as a percentage of body weight. (From Haskell A: Foot and ankle biomechanics. In Miller MD, Thompson SR, editors: *DeLee, Drez, and Miller's orthopaedic sports medicine*, Philadelphia, 2020, Elsevier.)

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