HUMAN LOCOMOTION

The Conservative Management of Gait-Related Disorders

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TABLE of CONTENTS

Chapter One

The Evolution of Bipedality

The process of walking around on two legs is an inherently unstable form of transportation. Watch any toddler attempt a few steps and it is easy to appreciate the complexity of the task. It is also an unusual way to get around: of the more than 4,000 species of mammals on earth today, only one is upright when walking (1). Even Plato commented on the curious nature of our preferred form of transportation by referring to humans as the only "featherless bipeds."

In 1871, Charles Darwin claimed that bipedality was the defining feature separating humans apart from our ape ancestors (2). Darwin theorized that the conversion to bipedal gait freed the hands to allow for tool use, which in turn created an environment that favored rapid brain expansion. This line of reasoning is consistent with that of modern anthropologists such as Mary Leakey (3), who states that bipedality "freed the hands for myriad possibilities: carrying, toolmaking, intricate manipulation... this new freedom of the forelimbs posed a challenge. The brain expanded to meet it. And mankind was formed."

According to the classic theory of bipedal evolution, approximately 2.5 million years ago a seismic shifting of tectonic plates caused a rapid global cooling that quickly converted the once dense forests of eastern Africa into the open grasslands of the savanna. Because food sources became more spread out, our early quadruped ancestors were forced to stand up and walk. This new form of transportation theoretically allowed the early hominids to see over the tall savanna grasses and cover larger distances in search of food.

The problem with the savanna hypothesis is that recent discoveries show that the timing is all wrong. In 2001, a team of French and Kenyan paleontologists announced the discovery of multiple specimens of a 6-million-year-old hominid they named *Orrorin tugenensis* (4). Discovered in the Tugin hills of Kenya, the femur of this early hominid was remarkably humanlike, as it even possessed a groove on the back of the femoral neck for the obturator externus muscle. This groove is only present in bipeds and confirmed that *Orrorin* most definitely walked upright (Fig. 1.1). In 2002, a team of paleontologists led by Michael Brunet (5) unveiled a newly discovered skull from a 7-million-year-old hominid they called *Sahelanthropus tchadensis* (named after the region in Africa where the fossil remnants were discovered). Although no other remains have been found, the skull of this hominid possessed a centrally located

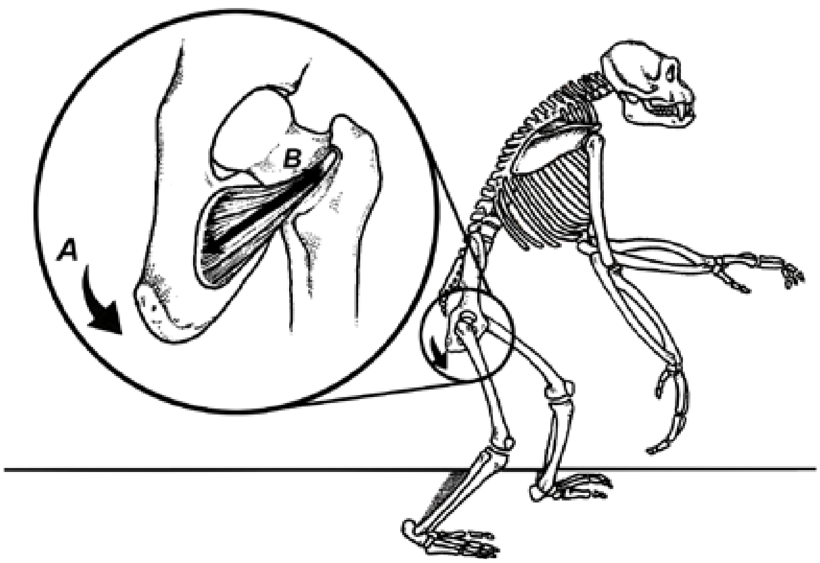

Figure 1.1. In contrast to quadrupedal locomotion, bipedal gait forces the ischium to move down and forward (arrow A), which significantly increases tensile strain placed on the obturator externus tendon where it passes along the posterior aspect of the femoral neck (B).

Chapter Two

Structural and Functional Anatomy

Leonardo da Vinci once said that in addition to being a work of art, the human body is also a marvel of engineering. Da Vinci's statement is particularly true when it comes to the anatomical structures necessary to allow for bipedality, since walking on two legs presents an engineering conundrum: during early stance phase the lower extremity must be supple in order to absorb shock and accommodate discrepancies in terrain, while the latter portion of stance phase requires that these same structures become rigid so they can tolerate the accelerational forces associated with the propulsive period. This is in contrast to quadrupeds, who have the luxury of being able to absorb shock with their forelimbs while their hindlimbs serve to support and accelerate (picture a cat jumping on and off a ledge).

The human body is able to accomplish these contradictory functions through a series of intricate articular interactions that allow the same anatomical structures to behave differently during the early and latter phases of gait. For example, the bones of the foot and ankle play a vital role during early stance as lowering of the medial arch creates a parallelism of the midtarsal joint axes that effectively unlocks the articulations of the midfoot allowing the bony structures to shift in order to accommodate surface irregularities (Fig. 2.1). Lowering of the arch is also indirectly responsible for shock absorption at the knee because it causes the talus to slide medially down the calcaneus (arrow A in Fig. 2.1), causing the ankle to twist inwardly as it follows the talus. The resultant internal rotation of the ankle forces the tibia to rotate beneath the femur, allowing the knee to flex (i.e., the knee is not a pure hinge joint since the tibia must internally rotate in order for the knee to flex properly). Knee flexion, in turn, allows eccentric contraction of the quadriceps muscle to dampen large amounts of vertical forces. To assist the knee in absorbing shock, the gluteus medius muscle is lowering the

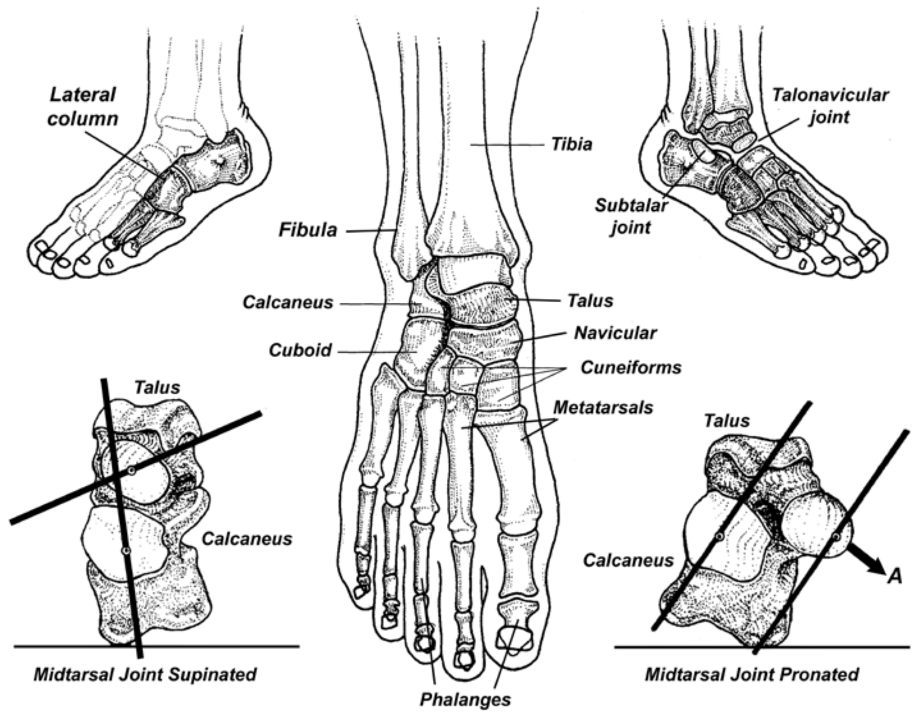

Figure 2.1. Osseous anatomy of the foot and ankle with an anterior view of the right talus and calcaneus pictured in the lower corners. Notice how pronation of the midtarsal joint (lowering of the medial arch) creates a parallelism of their shared axes while supination (arch elevation) creates a malalignment of these same axes. The improved axis alignment associated with lowering of the arch increases the range of motion available to the joint by more than 10°. Conversely, supination effectively locks the joint, as the axes no longer converge. This action is comparable to shifting the alignment of hinges on a door: when properly aligned, the parallelism of their shared axes allows the door to open without resistance while even a slight alteration in the position of one of the hinges will make it difficult to open the door.

Chapter Three

Ideal Motions during the Gait Cycle

The fundamental component of human locomotion is the gait cycle. One complete gait cycle consists of the anatomical interactions occurring from the moment the foot first contacts the ground, until that same foot again makes ground contact with the next step. The human gait cycle consists of two phases: stance phase, in which the lower extremity is contacting the ground; and swing phase, in which the lower extremity is swinging through the air preparing for the next impact (Fig. 3.1). When a person is walking, the gait cycle lasts approximately one second (1). As a result, stance phase occurs in 0.6 seconds and swing phase in 0.4 seconds. Because the distal end of the kinetic chain is fixed by ground-reactive forces during stance phase, motions during this portion of the gait cycle are referred to as closed-chain motions. In contrast, swing phase motions are referred to as open-chain motions since the distal end of the kinetic chain is freely mobile. Because of the complexity of stance phase motions, this portion of the gait cycle has been subdivided into contact, midstance, and propulsive periods (Fig. 3.2).

When walking, the contact period represents the first 27% of stance phase, beginning at touchdown and ending when the entire forefoot makes ground contact. Midstance occupies 27-67% of stance, representing the period in which the body's center of mass is "vaulting" over the stance phase foot. The propulsive period occupies the final 33% of stance phase, beginning the moment the heel leaves the ground and ending when the tips of the phalanges no longer make ground contact. Although running is also divided into the same 3 periods, the increased speed and the need for a more forceful propulsive period changes the timing of the events, as the contact and midstance periods are slightly shorter (occurring in the initial 0-20% and 20-45% of stance phase, respectively), while the propulsive period is extended, occupying the final 55% of stance phase (2). With more than 5,000 cycles performed daily, the gait cycle is one of the most repetitive events in our lives.

The neurological mechanisms necessary to complete a gait cycle are unusual in that swing phase motions are reflexive and present at birth (e.g., an unbalanced toddler will immediately swing the lower extremity into a protected position), while movements associated with stance phase represent a learned process (3). Scott (4) supports this statement with the clinical observation that children born without sight make no spontaneous attempts to stand up and walk on their own, and will only do so when physically guided. With or without sight, once upright and moving about, children immediately begin experimenting with a wide range of walking and running patterns, subconsciously analyzing the metabolic expense associated with each variation in gait. This is a time-consuming process and perfecting the musculoskeletal interactions necessary to

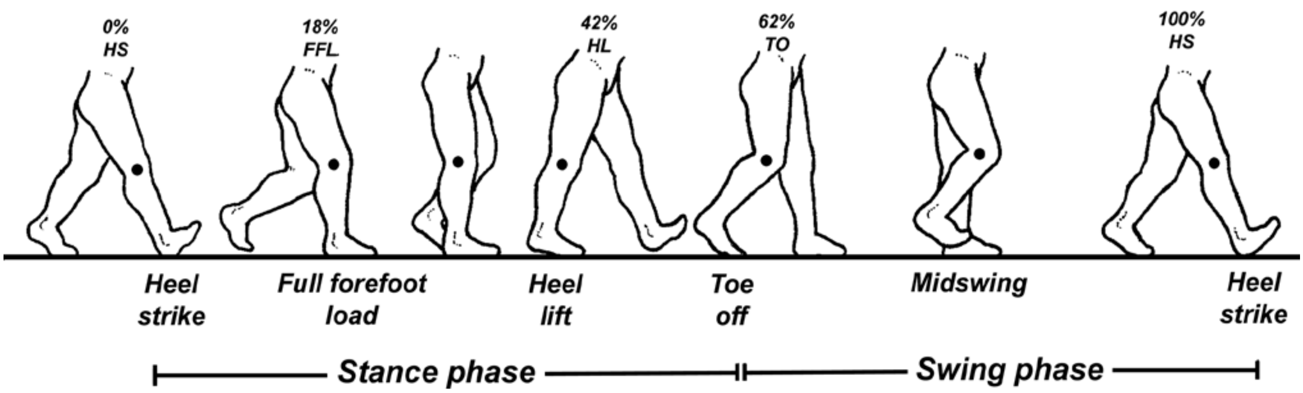

Figure 3.1. Gait cycle of the right leg. Stance phase begins at heel strike (HS) and ends when the great toe leaves the ground. Swing phase continues until the heel again strikes the ground. The length of stride, which refers to the distance between successive ipsilateral heel strikes, is approximately 0.8 times a person's body height and the average cadence is 115 steps/minute. Because size affects stride length and cadence, there is much individual variation in the gait cycle as women typically have slightly shorter stride lengths and a more frequent cadence. Children have particularly high cadences as the average 7 year old takes approximately 143 steps per minute. Because of the prolonged airborne phase, stride lengths while running significantly increase and it is not uncommon for world-class runners to possess stride lengths exceeding 3.5 meters (11 feet 6 inches), which is more than one meter longer than a comparably sized running quadruped (13).

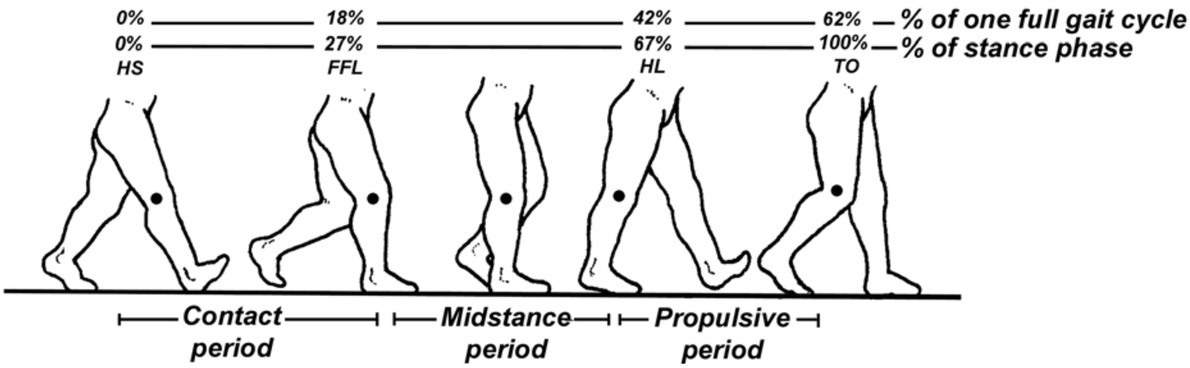

Figure 3.2. The various periods of stance phase. HS, heel strike; FFL, full forefoot load; HL, heel lift; TO, toe off.

become metabolically efficient can take up to a decade to master. Even when considering size differences, the average 3 year old consumes 33% more oxygen when traveling at a fixed speed compared to an adult (5). By the age of 6, children continue to have significantly higher ratios of energy costs versus work performed (6). Fortunately, by age 10, mechanical efficiency has improved and the cost/work ratios of 10 year olds and adults are about equal (6): after almost a decade of practice, children are finally efficient at bipedal locomotion.

In order to create a metabolically efficient gait, Saunders et al. (7) claim that individuals must learn to "translate their center of mass through space along a path requiring the least expenditure of energy." This is accomplished by modifying joint positions in the lower extremity and pelvis in such a way that the pathway of the center of mass through space is flattened. For example, if an individual were to walk with knees locked and the pelvis stiff, the body's center of mass would move through a series of abruptly intersecting arcs (Fig. 3.3A) that would greatly increase the metabolic cost of locomotion because muscles must tense to accommodate the exaggerated angular displacements. Further strain would be placed on the supporting muscles since they would initially absorb, and then accelerate these forces as the curves reverse direction. To lessen the metabolic cost of locomotion, each person incorporates a specific series of articular interactions that effectively decrease angular displacement of the body's center of mass. These actions, or determinants, are listed as follows: pelvic rotation; pelvic tilt; knee flexion/extension during stance phase; hip-knee-ankle interactions; and lateral pelvic displacement. The following illustrations, which were adapted from Saunders et al. (7), demonstrate how each determinant affects translation of the center of mass through space (Figs. 3.3-3.8).

Although the determinants described by Saunders et

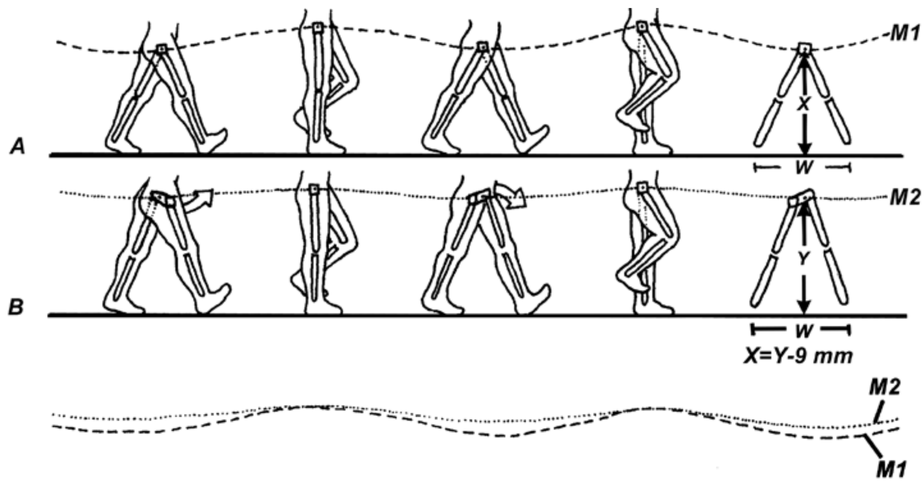

Figure 3.3. Determinants of gait: pelvic rotation. Panel A represents a lateral view of the gait cycle with the knees and hips locked. Notice how the pathway of the center of mass creates an exaggerated sine wave (**M1**), which is metabolically expensive because the hip abductors must raise and lower the center of mass through the exaggerated ranges. By incorporating pelvic rotation (arrows in panel **B**), the pathway of the center of mass is flattened slightly as rotation of the pelvis decreases the amount of hip flexion/extension necessary to achieve the same stride length (**W**). This decreases vertical drop during double-limb support by approximately 9 mm (the difference between the ground and the center of mass in **X** and **Y**), flattening the pathway for the center of mass (compare **M2** and **M1**).

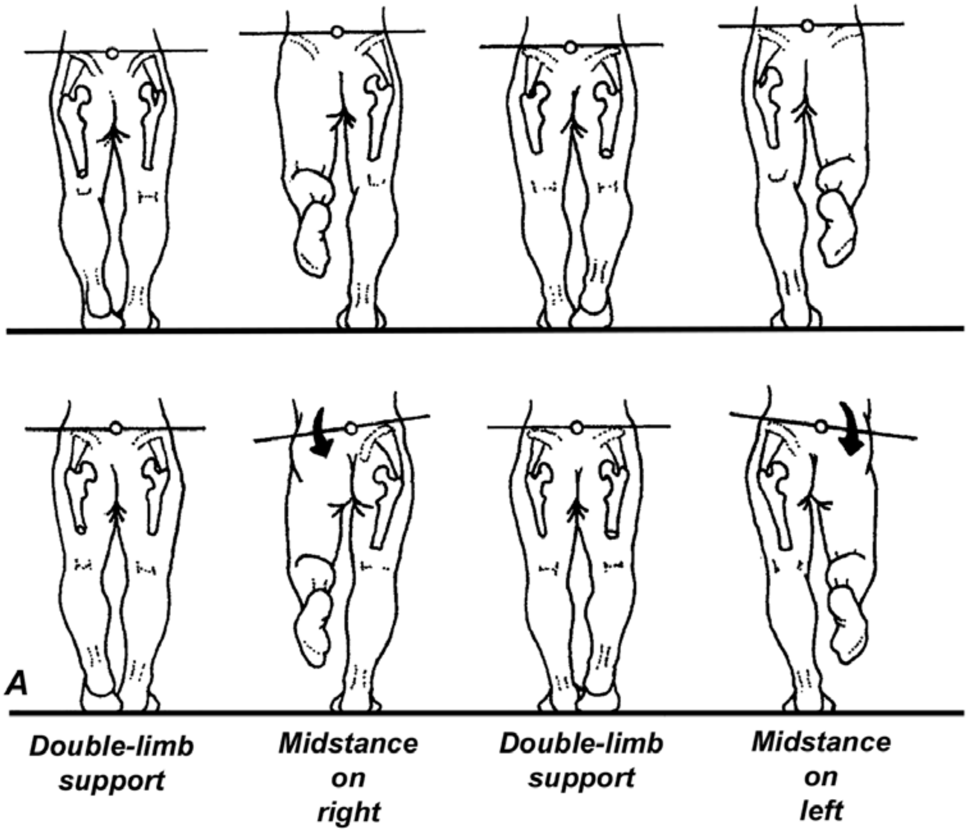

Figure 3.4. Pelvic tilt. Eccentric contraction of the hip abductors during midstance lowers the pelvis on the side of the swing leg (arrows in **B**). This decreases vertical displacement of the center of mass by approximately 3 mm.

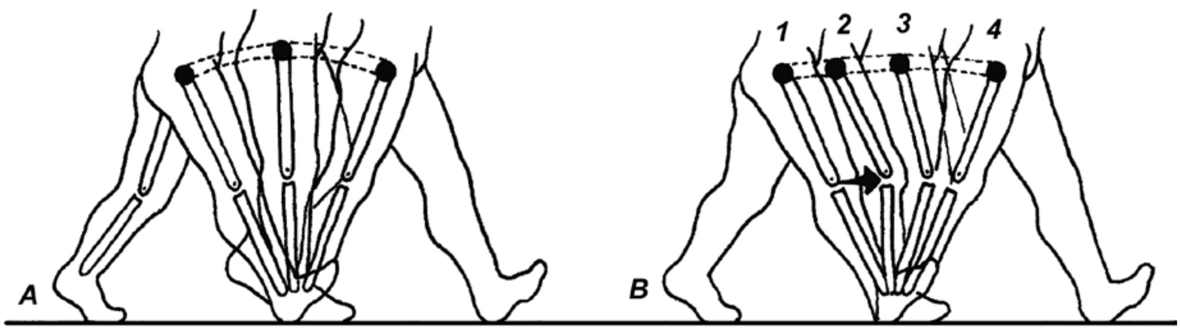

Figure 3.5. Knee flexion/extension during stance phase. Part **A** represents stance phase lower extremity motion without knee flexion while Part **B** represents the same leg with knee flexion/extension. Notice that when the lower extremity is straightened throughout stance phase, the center of mass describes a path along the arc of a circle, with the length of the lower extremity being the radius. This arc is effectively flattened by knee flexion during early stance phase and knee extension during late stance phase.

Abnormal Motion during the Gait Cycle

For the previously described ideal movement patterns to occur, several factors related to bony alignment, joint mobility and muscular strength must be present. Specifically:

1. Ontogeny should allow for the formation of an aligned lower extremity, particularly in the transverse and frontal planes.
2. The joints of the feet should form a stable medial longitudinal arch that is neither too high nor too low.
3. When the talonavicular joint is maintained in a neutral position and the calcaneocuboid joint is locked in its close-packed position, the plantar metatarsal heads should all rest on the same transverse plane.
4. The distal extensions of the metatarsal heads should form a smooth parabolic curve.
5. The lower extremities must be of equal length.
6. The articular architecture and ligamentous restraining mechanisms should protect against excessive mobility.
7. The joints of the pelvis and lower extremity should move through certain minimum ranges of motion.
8. Neuromotor coordination must be intact and the supporting muscles must possess adequate strength and endurance.

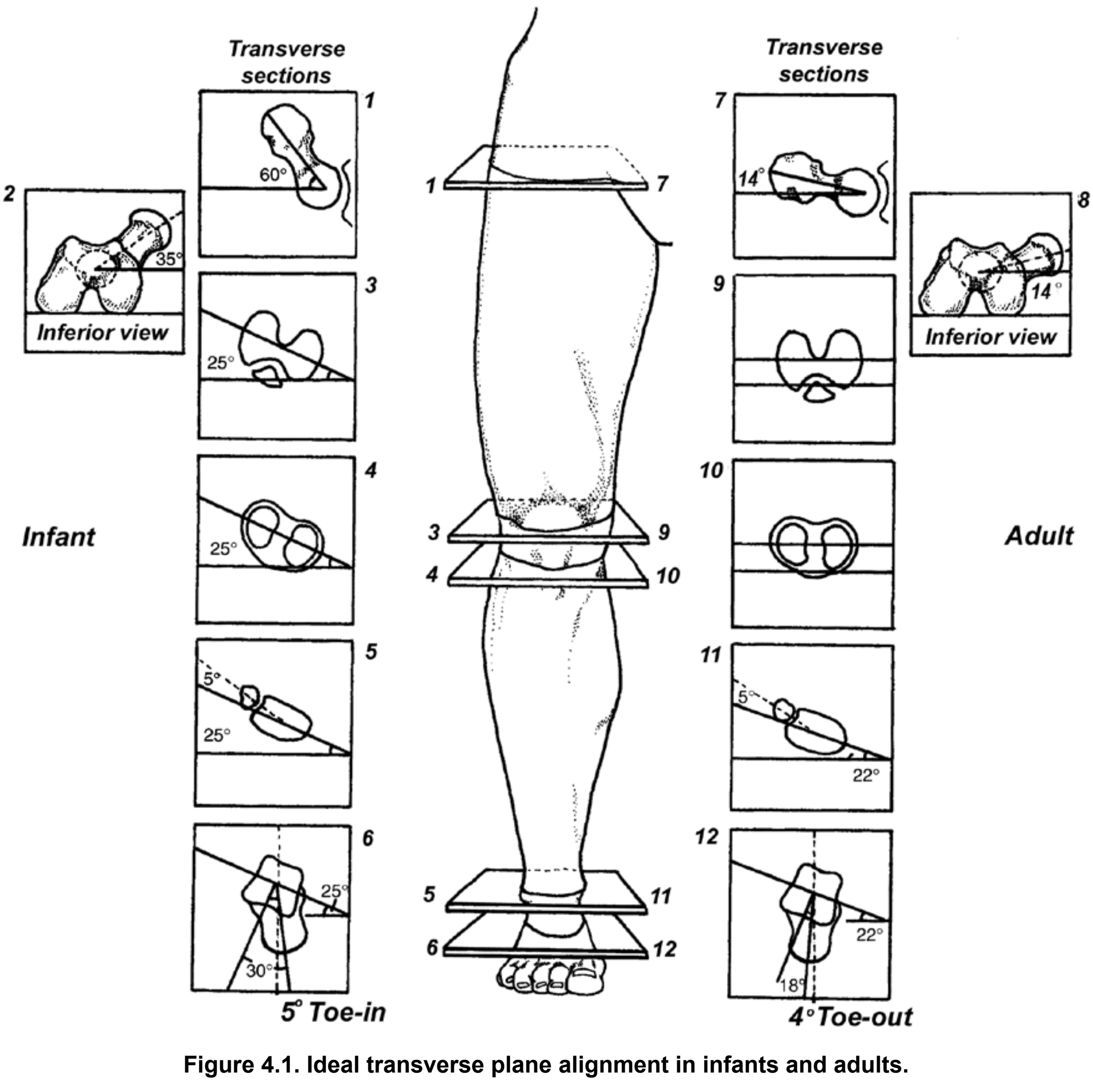

Figure 4.1. Ideal transverse plane alignment in infants and adults.

As expected, whenever there are guidelines defining norm, there are bound to be situations in which individuals deviate from these outlined parameters. The following section reviews abnormal motion associated with each of these categories and, when applicable, discusses conservative treatment interventions.

Developmental Trends in Lower Extremity Alignment

At birth, rotational patterns of the lower extremity differ significantly from those of the adult. During childhood and adolescence, the femur, tibia, and foot undergo specific transformations in the transverse and frontal planes that ideally allow the adult to walk with a relatively straight gait pattern; i.e., the young adult should walk with an approximate 7° toe-out gait pattern, with the tibia being nearly perpendicular to the ground at heel strike. These developmental changes will be discussed separately, beginning with those occurring in the transverse plane.

Transverse Plane Alignment

In the infant, the femoral head is positioned in the acetabulum so the femoral neck is angled approximately 60° posterior to the frontal plane (panel 1 in Fig. 4.1). Notice in panel 2, the femoral neck is anteverted 35° to the transcondylar axis of the distal femur. Because the 35° angle of femoral anteversion partially negates the 60° posterior angling of the femoral neck in the acetabulum (panel 1), the transcondylar axis of the knee joint is externally rotated 25° relative to the frontal plane (panel 3).

Panels 4 and 5 illustrate how the proximal and distal aspects of the tibia are well-aligned in the infant: there is 0° of tibial torsion as the distal femur and proximal tibia are both rotated 25° externally. The degree of tibial torsion is usually 5° less than a line bisecting the medial and lateral malleolus. (Compare the dotted line in panel 5, which represents the transmalleolar axis, to the solid lines in panels 4 and 5, which represent actual tibial torsion.) Panel 6 in Figure 4.1 illustrates the normal talar neck angle relative to the superior articular surface of the talar body. This angle of adduction increases from 20° in the fetus to 30° in the infant (1). Since the entire foot follows the neck of the talus via the articulation with the navicular, the talar neck angle is an important, albeit often overlooked, component of transverse plane alignment.

By combining the various angles for each segment in the lower extremity, the average degree of toe-in present in the infant should be approximately 5° (notice how the talar neck in panel 6 deviates 5° medially from the sagittal plane). This is consistent with a study of 70 infants by Bleck (1), who determined the mean normal internal rotation of an infant's foot with reference to the line of progression was 4.4°. The internally rotated position of the foot in newborns is usually not noticed because most babies lie with their hips externally rotated and because prewalking children who are forced to stand will often turn their feet out in an attempt to maintain balance. As toddlers begin to walk on their own, the normal toe-in pattern becomes more apparent as all segments of the lower extremity rotate into their neutral positions during stance phase. A slight toe-in is beneficial to the early walker as preschool children walking with toe-in gait patterns walk at faster speeds (2) (perhaps because the low gear push-off associated with the toe-in is easier to manage muscularly).

The transverse plane alignment patterns in adults are unlike those present in infants. As illustrated in panels 7 and 8, the angle of femoral neck anteversion has reduced from the 60° angle present in infants, to the 14° angle present in adults. This reduction is significant as it allows the transcondylar axis of the distal femur to align perfectly with the frontal plane (panel 9). First noted in the hominid *Homo rudolfensis*, precise positioning of the tibiofemoral joint in the frontal plane is important for efficient bipedality, as it enables the knee to flex in the sagittal plane, thereby allowing the limb to move in the direction of travel.

Although derotation of femoral neck anteversion from 60° to 14° generally occurs before the age of 8 (3), the major reductions in femoral anteversion occur during the first 3 months of infancy, when extension of the hip (which is necessary to reduce contracture in the hip flexors present at birth) produces an external rotational torque on the proximal end of the femur as the iliopsoas tendon pulls on the lesser trochanter (twisting the proximal femur outwardly). Because the end of the femur is cartilaginous and fixed to the rigid diaphysis, the external torque strain rotates the femoral neck relative to the shaft, producing a decrease in the angle of femoral anteversion. Apparently, this twist occurs in the subtrochanteric region where the plastic cartilage of the proximal femur interfaces with the solid diaphysis (1).

Another important developmental change in alignment occurs in the tibia. By comparing the relative positions of the proximal and distal tibia in panels 10 and 11, notice the distal tibia becomes externally rotated 22° relative to the proximal tibia by adulthood. (The transmalleolar axis, which as mentioned is approximately 5° greater than the degree of tibial torsion, is positioned 27° to the frontal plane.) Jay (4) notes the distal tibia rotates externally at a rate of 1-1.5° per year. This is clinically useful in determining the ideal degree of tibial torsion at a given age, in that a 10-year-old child should present with approximately 10° of external tibial torsion, with a transmalleolar axis of approximately 15°.

The final developmental change to be discussed occurs in the talar neck. From infancy to adulthood, the medial deviation of the talar neck should reduce from 30° to 18°, with most of these changes occurring by age 6 (1). After adding up the various transverse plane alignments, the adult should present with an approximate 4° toe-out pattern when all segments are in neutral alignment. This angle of toe-out is often deliberately increased, as most people externally rotate their hips in order to provide lateral stability during slower walking speeds. A slight toe-out also improves efficiency during locomotion as it allows for a high gear push-off, which improves the force generating capacity of the ankle plantarflexors (5).

While the developmental trends outlined in Figure 4.1 represent ideal ontogenic patterns, various genetic and/or developmental factors may either impair or exaggerate rotational development at any or all segments of the lower extremity. While such torsional deformities may occasionally be inherited, they are more commonly developmental and typically result from faulty intrauterine positioning during the final months of pregnancy (6). Intrauterine constraints are more likely to mold the fetal tissues detrimentally when tight uterine muscles are present (as with a firstborn child) or when a large fetus or multiple fetuses are present (4). Also, excessively tight abdominal muscles, a small pelvis, a prominent lumbar spine, uterine fibroids, or any fetal malposition (such as a breach or transverse lie position) all may impair normal rotation of the limb buds. The resultant torsional deformities are often maintained by various sitting and sleeping postures that act to perpetuate or even produce transverse plane malpositioning (Fig. 4.2).

In approximately 5-15% of the population, rotational patterns of infancy will persist beyond skeletal maturity (1). If the toe-out gait pattern persists into adulthood (which usually is the result of femoral retroversion), the individual is predisposed to injury because of the increase in pronatory forces placed upon the subtalar joint: normally, when the foot lands in a straight position, shear forces act to create a plantarflectory moment around the ankle axis. However, when the foot lands with a toe-out gait pattern, shear forces are applied more perpendicularly to the subtalar axis and can therefore generate a strong pronatory force at the foot and a valgus torque at the knee. This may explain why individuals with large ranges of external rotation at the hip are predisposed to medial tibial stress reactions (8), and

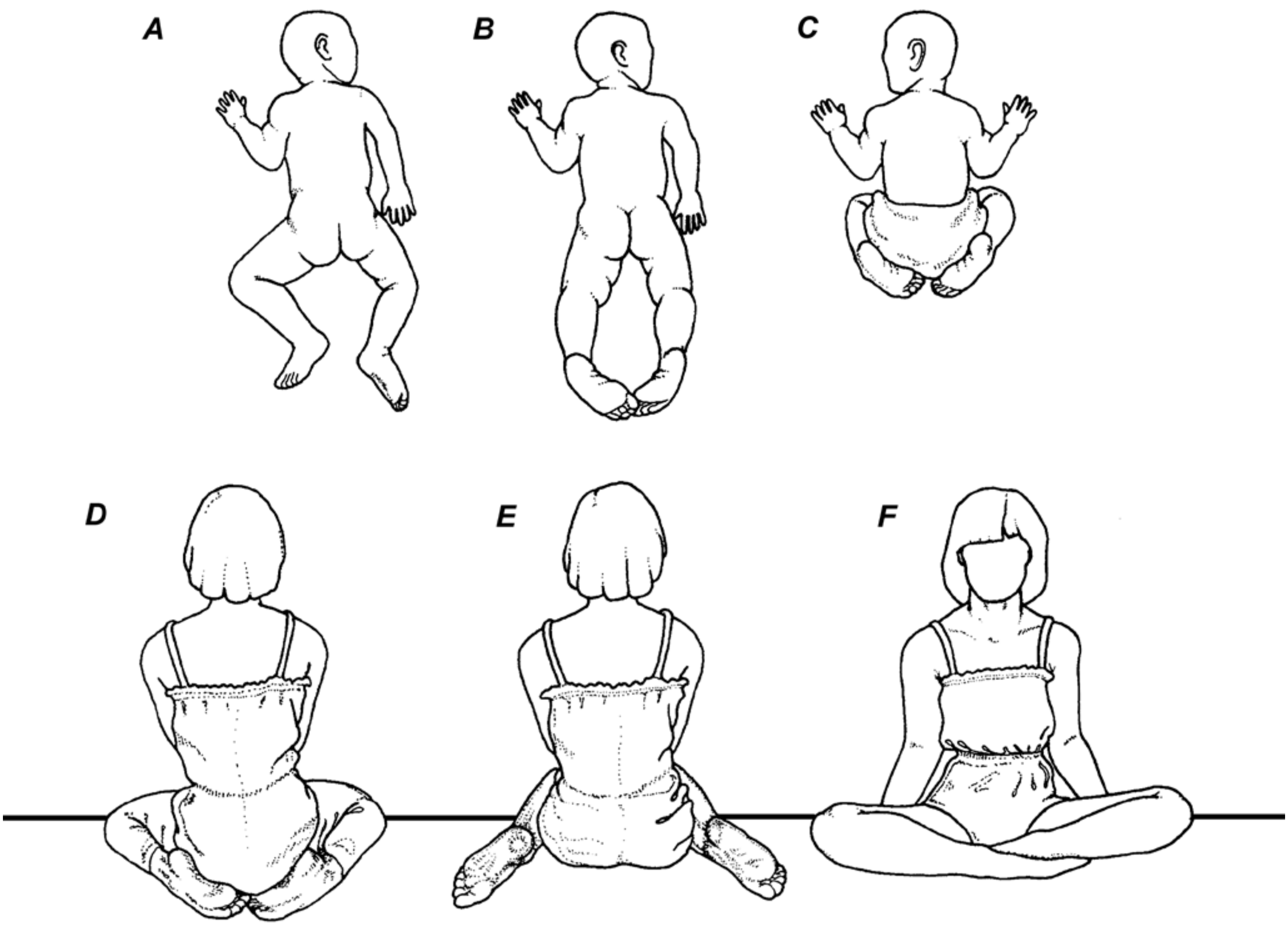

Figure 4.2. Sleeping and sitting postures that may perpetuate or produce various torsional deformities. (**A**) Prone frog leg: produces external rotation deformity at hips, external tibial torsion, and valgus heels. (**B**) Prone, hips extended, feet adducted: produces internal tibial torsion and varus heels. (**C**) Prone, hips flexed, feet adducted: produces external rotation deformity at hips, internal tibial torsion, varus heels. (**D**) Sitting on adducted feet: produces internal tibial torsion and varus heels. (**E**) Sitting with feet abducted (the "television position"): produces excessive external tibial torsion and valgus heels. (**F**) Tailor's position: produces external rotation deformity at hips and varus heels.

Chapter Five

Biomechanical Examination

Because private practitioners are currently unable to access the complex machinery necessary to perform research quality 3-dimensional gait evaluations, they are forced to rely upon specific static and dynamic measurements that theoretically predict triplanar motions present during the gait cycle. For almost a century, researchers have been attempting to refine these measurements. In the early 1900's, Dudley Morton (1) developed a foot classification system based on the respective lengths of the individual metatarsals. His theory that a shortened first metatarsal could identify overpronators has since been disproved (2). In 1949, Hiss (3) published his text *Functional Foot Disorders* in which he incorporated a more complicated classification system where various morphological and dynamic relationships were assessed and related to altered function; e.g., hypermobility of the first ray was identified by evaluating intermetatarsal movement ratios and was believed to be associated with the development of forefoot pathology. Despite problems with reproducibility, many foot specialists continue to use this biomechanical approach today (4). In 1971, Root et al. (5) published what is without doubt the most widely referenced source on biomechanical measurements. In their text *Biomechanical Examination of the Foot*, they describe a range of specific goniometric biomechanical measurements that were theorized to predict 3-dimensional motion patterns present during the gait cycle. In turn, these measurements were used to justify specific aspects of orthotic intervention.

Although Root et al. (5) were pioneers in the field of biomechanics, their described measurements were difficult to reproduce (6) and it was later proven that their off weight-bearing measurements did not reflect the true ranges available during weight-bearing (7). More importantly, the various goniometric measurements suggested by the authors did not predict 3-dimensional lower extremity motions present during the gait cycle (8-10). This was first noted in 1989 by Hamill et al. (9). Using a handheld goniometer, these researchers measured the 16 static and dynamic variables described by Root et al. and then evaluated 3-dimensional motion as 24 subjects walked over a force platform. Unfortunately, not one of the static measurements predicted dynamic motion. In a similar study, McPoil and Cornwall (8) took 17 static and dynamic measurements on 27 young adults and performed 2-dimensional analysis of rearfoot motion during walking. Again, none of the static measurements predicted dynamic motion.

The complete inability of static measurements to predict dynamic function has become a major obstacle in the development of a scientifically justified algorithm for managing gait disorders. To help resolve this issue, several investigators have developed and evaluated a range of biomechanical measurements to determine if any of these measurements correlate with dynamic function. The ideal biomechanical measurement, besides predicting motion, should be repeatable between different examiners (i.e., possess high interrater reliability) and provide information that justifies clinical intervention. This chapter reviews these measurements as they are performed during the standard biomechanical examination.

Supine Examination

This examination should begin by motion palpating the various lower extremity articulations. The presence of joint dysfunction should be noted, and the fixations should be gently mobilized. In addition to relaxing the patient, this helps to reduce any functional deformities that might adversely affect measurements. A quick visual assessment of muscle symmetry should be performed and if asymmetry is noted, particularly in the VMO or calf, circumference measurements should be taken. Next, a simple screening of muscle flexibility can be performed by moving each of the lower extremity joints through specific ranges of motion (refer back to figures 4.113 through 4.129). Excessive tightness/laxity should be noted and side-to-side differences in flexibility should be recorded. Ligamentous integrity of the knee and ankle can be evaluated using the specific orthopedic tests illustrated in Fig. 5.1 through 5.5. It is especially important to check the integrity of the anterior talofibular ligament, since laxity in this ligament increases the transfer of calcaneal eversion into internal tibial rotation (11), potentially producing injury in the proximal structures.

The relative lengths of the metatarsals can be determined by plantarflexing the digits and comparing the locations of the dorsal metatarsal heads (see Fig. 4.76). The presence of shortened/lengthened metatarsals should be noted. A standard goniometer is then used to measure the range of hallux dorsiflexion (Fig. 5.6). Although off weight-bearing measurement of first metatarsophalangeal joint dorsiflexion is not predictive of 3-dimensional motion during the gait cycle, it is useful for monitoring the progression of osteoarthritis and/or for evaluating the efficacy of manual techniques when attempting to

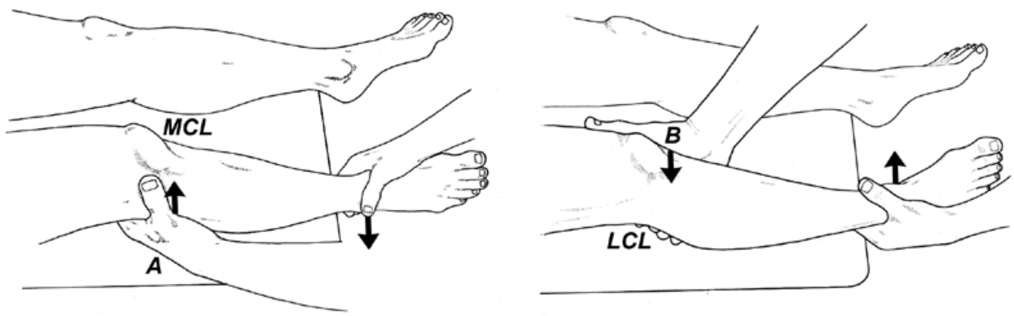

Figure 5.1. Medial collateral ligament (MCL) is tested with the valgus stress test, in which the knee is flexed 30° and a valgus stress is applied at the knee (A). The lateral collateral ligament (**LCL**) is also performed with the knee flexed 30°, only a varus stress is applied at the knee (**B**).

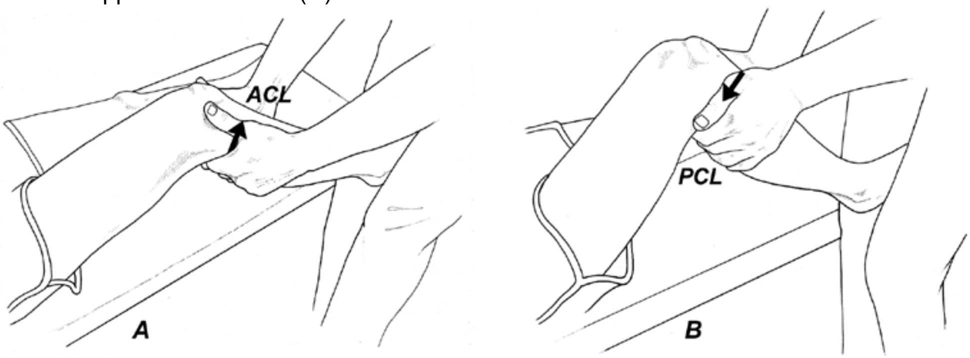

Figure 5.2. The anterior cruciate ligament (ACL) is tested with the modified anterior drawer test (A). This test is performed with the patient's foot grasped between the examiners knees. The ACL is tested with the knee flexed 30° and the examiner pulling the proximal tibia forward while simultaneously palpating the femoral condyles with the thumbs. The posterior cruciate ligament (**PCL**) is tested by flexing the patient's knee to 90° and applying a posteriorly directed force. Laxity should be compared bilaterally.

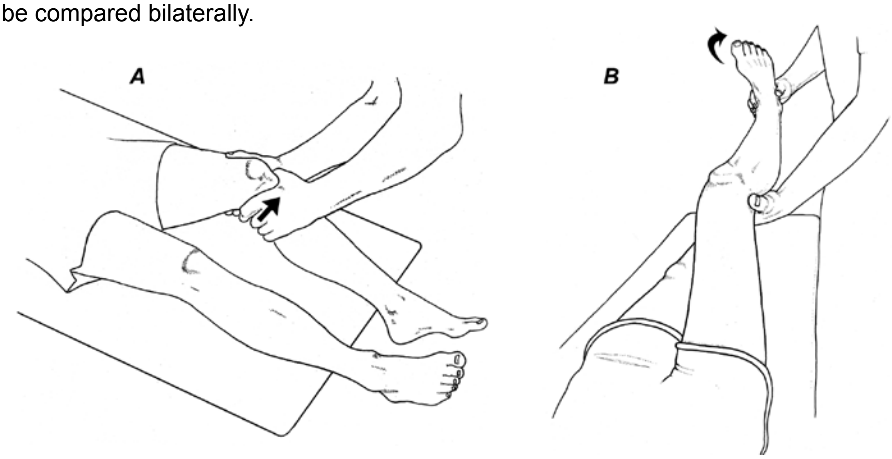

Figure 5.3. Specialty tests. Because the diagnostic accuracy of the anterior drawer test is poor, the ACL may also be evaluated with **Lachman's test (A).** To perform this test, the patient is positioned with the knee slightly flexed and the leg externally rotated 20° while the examiner stabilizes the distal femur with one hand while attempting to displace the proximal tibia anteriorly with the other hand (arrow). The hand on the proximal tibia is positioned with the thumb and index finger placed on the joint line. The integrity of the posterolateral ligament complex can be evaluated with the **Dial Test (B).** This test is performed by flexing the knee of the supine patient 30° with one hand palpating the joint line while the opposite hand attempts to externally rotate the leg (arrow). If laxity of the posterolateral ligament complex is present, the forefoot will abduct excessively. This test is also helpful for evaluating the anterior and posterior cruciate ligaments. To test the anterior cruciate ligament, the knee is flexed 90° and the examiner attempts to internally rotate the knee by adducting the forefoot. The posterior cruciate ligament is also tested with the knee flexed 90°, only now the dial test is performed by abducting the forefoot. By visually aligning the distal pole of the patella with the tibial tuberosity, the examiner can evaluate even subtle differences in laxity.

Figure 5.4. Possible tears in the menisci are evaluated with McMurray's test. The medial meniscus is evaluated by extending the fully flexed knee while externally rotating the leg (**A**) as a valgus stress is applied at the knee (**B**). An audible clunk is indicative of a tear. The lateral meniscus is tested by internally rotating the leg while applying a varus stress at the knee.

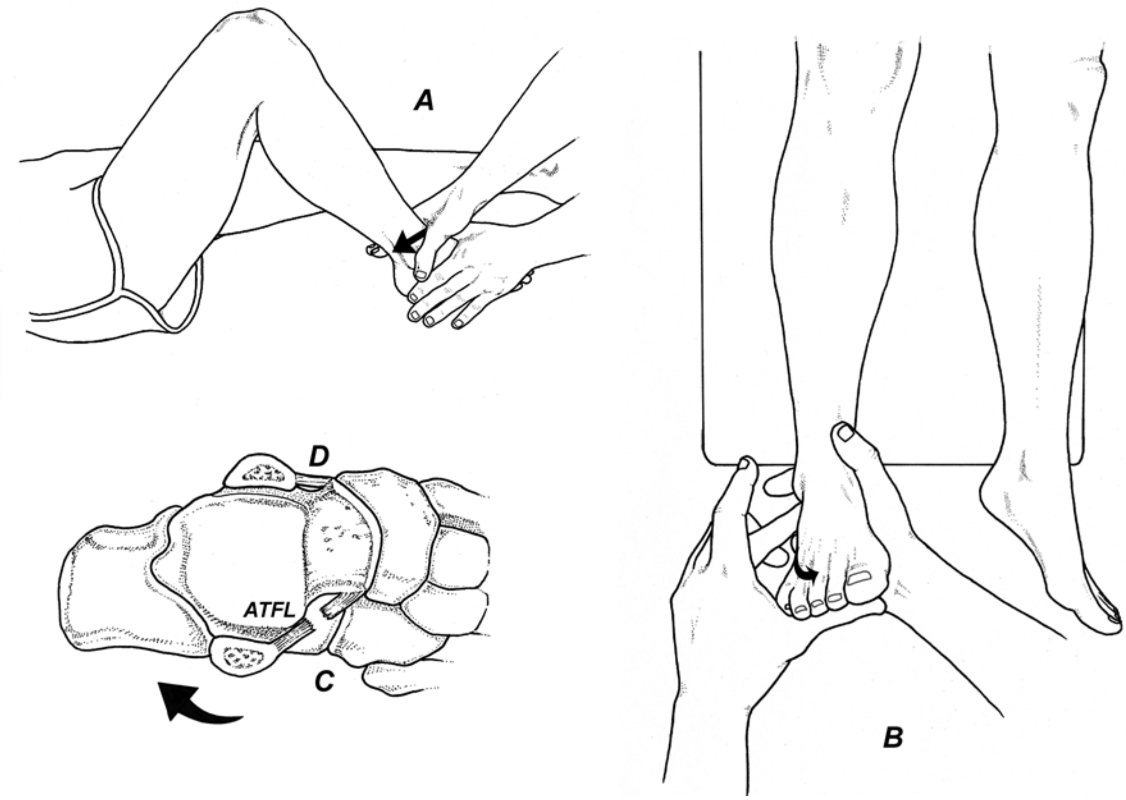

Figure 5.5. The modified anterior drawer (A) and a varus stress test (B) evaluate the integrity of the anterior talofibular ligament (ATFL) and the calcaneofibular ligament, respectively. The anterior drawer test is performed by stabilizing the patient's dorsal foot with one hand, while the opposite hand displaces the tibia straight posteriorly (arrow in **A**). When an isolated tear of the ATFL is present (**C**), the leg externally rotates with the applied stress (curved arrow). If both the deltoid (**D**) and anterior talofibular ligaments are torn, the leg will displace straight posteriorly with no rotational component to the motion. To test the calcaneofibular ligament with the varus stress test, the examiner applies a varus stress at the talocrural joint while the opposite hand palpates the joint line for gapping. As with all stress tests, laxity should be compared bilaterally.

Chapter Six

Foot Orthotics

Despite numerous methods of modifying motion during the gait cycle (e.g., gait training, agility drills, isokinetic exercises), the most popular technique to alter lower extremity movement is with the use of custom and prefabricated foot orthotics. Both custom and prefabricated orthotics are available in a variety of materials, ranging from the relatively rigid graphites to the more flexible leather laminates. Foot orthotics are particularly commonplace in the sporting community, where they are used to treat a wide variety of injuries, including patellofemoral pain syndrome, plantar fasciitis, and Achilles tendinitis. In spite of their clinical popularity, there continues to be significant controversy regarding their exact mechanism of action. Researchers argue as to whether orthotics are effective because they improve skeletal alignment, lessen lower extremity rotation, decrease velocity of motion and/or alter the mechanical efficiency of the stabilizing muscles. Furthermore, practitioners have strong opinions when it comes to deciding which casting technique should be used when fabricating an orthotic: some claim orthotics should only be made from off weight-bearing neutral position casts, while others claim only full weight-bearing foam step-in techniques are appropriate. Even the most fundamental questions concerning orthotic material selection, fabrication techniques, and the clinical indications for specific additions remain unanswered. Obviously, this controversy greatly affects the consistency in which orthotic treatment protocols are applied, which in turn has a negative effect on clinical outcomes.

Fortunately, in the past few years, advances in 3-dimensional imaging have allowed for an improved understanding of the various mechanisms responsible for orthotic efficacy. In many situations, the results have been surprising. The purpose of this chapter is to review the different ways foot orthotics affect lower extremity kinematics and relate this information to the design and fabrication of the orthotics themselves. Specific orthotic casting methods, materials, orthotic additions, sport specific variations, and in-office manufacturing techniques will be reviewed in detail.

Mechanism of Action

It has been a long-held belief that the main reason orthotics work is because they improve skeletal alignment. In fact, the origin of the word orthotic stems from the Greek word *ortho*, meaning straight. It seems intuitive that if you support a pronated foot with an orthotic, the calcaneus will become more vertical (i.e., straighter), causing the lower extremity to rotate externally, thereby improving alignment at the knee. These coupled movements have been theorized to continue along the entire kinetic chain, with some authors claiming that foot orthoses may help to lessen low back pain (1).

Surprisingly, a significant body of information suggests that this may not be the case, as most 3-dimensional research suggests that orthotic intervention produces very little change in frontal plane rearfoot motion during the gait cycle (2-7). In an exceptionally detailed study, Stacoff et al. (6) surgically implanted intracortical pins into various bones of the lower extremity (which is the gold standard for studying 3-dimensional motion) and concluded that orthotics produce little to no change in rearfoot eversion during the gait cycle. More recently, Nawoczenski et al. (7) performed an interesting 3-dimensional analysis and determined that individuals with low arches actually have increased ranges of rearfoot eversion while wearing orthotics.

The inability of orthotics to modify frontal plane movement of the rearfoot caused investigators to come up with alternative theories to explain the mechanisms responsible for the beneficial clinical outcomes associated with their use. Messier and Pittala (8) suggest that orthotics work because they alter the velocity of pronation, while others claim orthotics work because they decrease tibial rotation and/or decrease impact and loading rates of vertical ground-reactive forces (9,10). It is also suggested that orthotics work because they improve proprioception, decrease ankle inversion moments, decrease genu valgum and/or decrease external rotation moments at the knee (11, 12). To add to the controversy, some experts question whether the effectiveness of orthotic intervention is related to the actual shell of the orthotic or the applied post material (13).

In 2003, Mundermann et al. (13) tested these theories by performing a detailed 3-dimensional examination comparing a range of kinetic and kinematic variables in 21 pronators wearing one of four test conditions: a full-length varus post, a custom molded orthotic shell, the same shell with a varus post, and an unposted flat insole. After having the subjects fitted with reflective skin markers, kinematic data was recorded with 7 high-speed cameras and ground-reactive forces were measured using a force plate. The subjects then ran 200 meters on an indoor track in each of the four test conditions as researchers measured 15 different variables that included maximum foot eversion, maximum foot eversion velocity, tibial rotation (range and velocity), foot inversion (range and velocity), ankle inversion moments, knee abduction moments and vertical impact forces. The results were interesting as the post, shell and post, and shell alone all altered different parameters of gait compared to the flat insole control. Specifically, the post alone condition significantly decreased rearfoot eversion during the contact period, and slightly decreased tibial internal rotation. In addition, the post alone appreciably lessened the maximum ankle inversion moment, which would decrease strain on the muscles responsible for decelerating subtalar pronation during the contact period; e.g., tibialis posterior.

When used by itself, the custom molded polypropylene shell (molded from a neutral position positive model) had different actions. Although it had no effect on frontal plane motion of the rearfoot (range or velocity), it markedly lessened the maximum ankle inversion moment, which as mentioned, decreases strain on tibialis posterior. The shell also appreciably reduced the range of tibial internal rotation and, more importantly, increased the angle of maximum foot inversion during the propulsive period. The authors state that this finding suggests that molding plays a significant role during late stance phase. Also of note, the custom molded shell significantly improved shock absorption as it decreased the vertical impact peak and vertical loading rates present during early stance. Mundermann et al. (13) speculate that custom molding of the orthotic shell increased shock absorption by providing a larger contact area between the foot and the orthotic. They go on to reference Perry and Lafortune's (14) belief that the unaltered range of rearfoot eversion associated with using the shell only might allow the subtalar joint to act as a more efficient shock absorber.

When analyzed with the post and shell condition, the subjects consistently demonstrated the same kinetic and kinematic changes present with the shell alone; i.e. decreased vertical force, minimal changes in contact period frontal plane motion of the rearfoot, a decreased inversion ankle moment and an increased maximum foot inversion angle during the propulsive period.

In a follow-up study published the same year by the same authors, Mundermann et al. (15) again studied the effect of post vs. shell vs. control in 21 pronators, only this time they also included EMG data of muscle function and evaluated comfort in each of the test situations with a visual analog scale. The findings regarding kinetic and kinematic data were similar to the previous study in that the post alone significantly decreased maximum foot eversion, slightly increased vertical impact peak and improved ankle inversion moments. Conversely, the shell alone improved shock absorption, lessened ankle inversion moments and allowed for greater supination during propulsion. Again, when the shell and post were combined, the effect of the shell was dominant. What is interesting about this study is that comfort was significantly higher for the shell alone compared to all other conditions. The authors state, "in general, custom molding of foot orthoses appears to be a feature that increases comfort." In both studies, foot orthoses produced significant and systematic changes in kinetic, kinematic, comfort and EMG variables. Although these outcomes conflict with prior research suggesting orthotics do not significantly alter motion (6,7,16), Mundermann et al. (15) claim the controlled nature of their studies made for more accurate outcomes.

The conclusion that custom molding improves comfort is clinically supported by the fact that studies comparing efficacy of custom versus prefabricated orthotics often report lower dropout rates for the individuals wearing custom orthotics (17). In an *in vivo* study of foot pressures associated with different orthotics, Redmond et al. (18) evaluated plantar pressures in 22 healthy individuals as they wore either thinly-soled athletic shoes alone, flat stock insoles modified with 6° varus posts or functional foot orthotics made from neutral position off weight-bearing plaster casts that were manufactured with 6° varus posts. The authors note that plantar pressures in subjects wearing the posted stock insoles differed only slightly from the shoe condition but the custom foot orthotics significantly lowered all pressure related variables with reductions ranging from 14 to 21%. The total contact area was increased by 38% in the custom orthotic group and 16% in the posted insoles. Maximum heel forces were reduced 8% in the custom orthotic group as the rearfoot contact area was increased. The authors note that despite the varus posts, neither the custom orthotics nor the posted insoles shifted force laterally and the custom orthotics worked primarily by supporting the arch. By elevating the arch, the custom orthoses increased maximum force displaced beneath the arch but because the closely fitted contours of the custom orthotics spread forces evenly over the entire arch area, there was no significant change in pressure peaks through the midfoot.

In another interesting evaluation of 3-dimensional motion with custom orthotics, Williams et al. (19) studied the effect of standard foot orthoses and inverted orthoses on lower extremity function (details concerning the manufacture of inverted orthotics will be discussed briefly). The standard foot orthotics were made from graphite shells with 4° varus posts, while the inverted orthotics were made from the same materials but possessed extremely high post angles; i.e., the rearfoot of the orthotic shell was inverted either 15° or 25° (individuals with less than 10° resting calcaneal stance eversion were posted at 15° while those with greater than 10° calcaneal eversion were posted at 25°). Despite the extremes of these post angles, the subsequent 3-dimensional gait analysis revealed absolutely

Chapter Seven

Shoe Gear

Unlike the majority of mammals possessing insensitive hooves or thick pads, humans are forced to traverse the terrain on our comparably soft and extremely sensitive feet. Because they are so easily injured, some argue that the human foot was poorly designed to manage the stresses associated with wandering across our often-treacherous landscapes, since feet were originally intended to help us get around the thick branches present in cool dense forests, where arboreal climbing skills were key to our early survival. The difficulties associated with prolonged bipedality presented significant challenges to our feet, as ground-reactive forces were drastically amplified (increasing the risk of puncture wounds) and the higher ground temperatures outside the forest increased the potential for thermal injury and/or surface heat loss. Given the potential for lacerations, abrasions, and/or thermal injury, it seems odd that for almost all of our 7 million year history as bipeds, we walked around the planet barefoot.

Although we perceive our feet as being delicate structures in need of protection, when barefoot from birth, the human foot is remarkably resilient. As demonstrated by Robbins et al. (1), the skin beneath the heel and hallux is designed with a tight trabecular tethering of the epithelial membrane that resists perforation. The authors tested the integrity of this membrane by measuring skin resiliency and perceived pain as barefoot subjects were exposed to a "penetrometer" (a pointed device that applied 9 kg of pressure through a 10 mm spherical ball). Surprisingly, when the heel was compressed with the device, only 6% of the subjects reported pain; when the penetrometer compressed the distal hallux, just 32% reported pain. This contrasts to compression of the plantar first metatarsophalangeal joint, in which 66% of the subjects reported significant discomfort. The authors state that painful plantar stimulation beneath the metatarsal heads is necessary to initiate a protective muscular response in the digital flexors, which has been proven to distribute pressure over a broader area. In contrast, the heel, which is the initial point of contact while walking, is relatively insensitive to small objects applied with light to moderate force. The plantar heel is also relatively impervious to thermal injury, since the fat pads located beneath the heel and forefoot possess 4-times the polyunsaturated fat of regular adipose tissue, which, due to its lower freezing point and viscosity, prevents against heat loss to the environment and dissipates shock even at subzero temperatures (2).

In a study comparing lifelong shod feet with the feet of people who have never worn shoes, D'Aout et al. (3) confirm that the unshod forefoot is 16% wider than the shod forefoot. The increased width allows for improved distribution of ground-reactive forces during the propulsive period of barefoot walking. In their analysis of plantar pressures centered beneath the forefoot in lifelong shod versus unshod individuals, the authors confirm that regular shoe use is associated with significantly higher peak pressures beneath the second and third metatarsal heads. This is consistent with an analysis of skeletal remains dating back 100,000 years, confirming metatarsal pathology is more severe in shod populations (4). To enhance protection against perforation, the skin of an unshod foot becomes extremely tough and is remarkably similar to leather. These features allowed the feet of our earliest ancestors to easily manage the stresses associated with moving around sub-Saharan Africa.

Surprisingly, our unshod feet could even handle the extremely cold temperatures and jagged mountainous terrain associated with traversing Eurasia, as evidence suggests that we did not begin routinely using protective footwear until 30,000 years ago. This means that for 80,000 years following our exodus from Africa, we crossed the Swiss and Italian Alps and quickly spread through the harsh climates of Europe and Asia without protective shoe wear.

Determining the exact date that we began routinely using shoes has been difficult, as the early shoes were made of leather, grass and other biodegradable materials that left no fossil evidence. Although Neanderthals and *Homo Erectus* were suspected of occasionally using insulated foot coverings, the first direct evidence of shoe use dates back only 3,500 years, to a leather shoe found in an Armenian cave (Fig. 7.1). While primitive sandals and moccasins discovered in Oregon and Missouri have been carbon-dated to 10,000 years ago, the actual time period that our ancestors first introduced protective shoe wear remains a mystery.

To get around the fact that ancient shoes rapidly decayed leaving no evidence of use, Trinkaus and Shang (5) decided to date the initiation of shoe wear by searching for changes in the diaphyseal diameter of the second through fourth proximal pedal phalanges in our early ancestors. Because shoe use lessens strain on the digital flexors, the authors theorized that habitual shoe use would

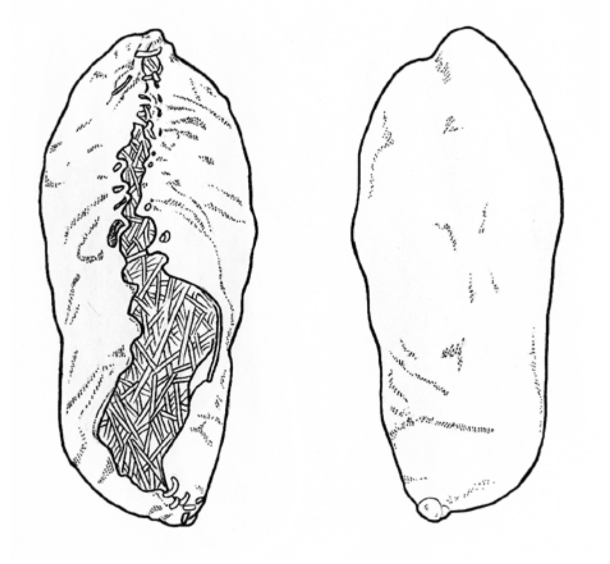

Figure 7.1. The earliest shoes resembled stitched leather bags. Drawn from a photograph in Pinhasi et al. (36).

be associated with the sudden appearance of a thinning of the proximal phalanges. By precisely measuring all aspects of phalangeal shape and composition, the authors discovered a marked decrease in the robusticity of these bones during the late Pleistocene era, approximately 30,000 years ago (Fig. 7.2). Because there was no change in overall limb robusticity, the anatomical inference is that shoe gear resulted in reduced strain on the long and short digital flexors, eventually resulting in the development of narrower proximal phalanges. The authors state that because there is no evidence of a meaningful reduction in biomechanical loads placed on human lower limbs during the late Pleistocene era (e.g., reduced foraging distances), the only logical conclusion is that the slender phalanges could only have resulted from the use of shoes. The authors evaluated numerous skeletal remains from different periods and concluded that based on the sudden reduction in diaphyseal diameter of the second through fourth proximal phalanges, the use of footwear was habitual sometime between 28,000 and 32,000 years ago.

The first shoes were most likely similar to the shoes discovered in the Armenian cave, in that they were simple leather bags partially filled with grass to insulate the foot from cold surfaces. Because shoe gear varied depending on the region, the earliest shoes worn in tropical environments were most likely similar to the 3,000-year-old sandals recently found in Israel. Once discovered, use of protective shoe wear quickly spread. The early Egyptians were believed to be the first civilization to create a rigid sandal, which was originally made from woven papyrus leaves molded in wet sand. Affluent citizens even decorated their sandals with expensive jewels.

The Greeks also prized their shoe wear. Although the first Olympic athletes competed barefoot, the average individual routinely wore ornamental sandals. Analysis of Greek art reveals that shoes and sandals were used as status symbols to identify the social status of the wearer (this was also true of the Egyptians). While the Egyptians were credited with adding the first heel to a sandal, it was the Greeks that developed the first high heel shoe, and it is suggested that Greek prostitutes wore the elevated heels because the heavy heel produced a clicking sound that announced their presence to potential clients. The trend for wearing high heels was only temporary and did not come back into fashion until the reign of Queen Elizabeth, when women wore platform heels as high as 24 inches. Because of the frequency of serious falls while wearing the excessively elevated heel, they quickly fell out of favor. Surprisingly, high heels became popular among French men in the 1700s, as Louis XIV was shorter than average and favored wearing 3-inch heels that flared out at the bottom. This type of heel is still around today and is known as the Louis heel. Despite evidence that regular use leads to the development of forefoot arthritis (6), high heel shoes remain popular.

Although the affluent Greeks and Egyptians had separate shoes/sandals made for their right and left feet, the practice of wearing different shoes on each foot was short-lived, and throughout the Dark and Middle Ages, shoes were made to be worn on either foot. Improvements in manufacturing techniques before the American Civil War changed that. By modifying a duplicating lathe used to mass produce wooden gunstocks, a Philadelphia shoemaker was able to manufacture mirror-image lasts for

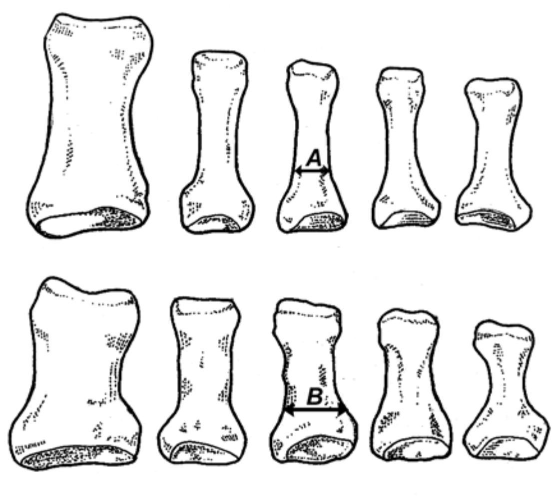

Figure 7.2. Dorsal view of the proximal phalanges from the early (bottom row) and late (top row) Pleistocene era. Trinkaus and Shang (5) claim that the decreased strain on the toes associated with regular shoe use produced bony remodeling with a gradual narrowing of the proximal phalanges (compare **A** and **B**).

the production of separate shoes for each foot. Using this new technology, the Union Army supplied over 500,000 soldiers with matching pairs of right and left leather shoes. The basic components of a leather shoe are illustrated in figure 7.3.

Leather continued to be the most popular material used for making shoe gear until the 1890s, when Charles Goodyear accidentally dropped rubber into heated sulfur creating vulcanized rubber. Prior to his serendipitous discovery, rubber was a relatively useless material. The newfound resiliency of this material would have numerous applications, including the production of the first sneaker. Although alternate names for the new foot wear include tennis shoes, trainers and runners, the term sneaker remains the most popular, and its origin can be traced back to an 1887 quote from *The Boston Journal of Education* (7): “It is only the harassed schoolmaster who can fully appreciate the pertinency of the name boys give to tennis shoes-sneakers.” Apparently, the soft rubber soles allowed schoolchildren to quietly sneak up on unsuspecting teachers.

Spalding manufactured one of the earliest sneakers: the Converse All-Star. Used by athletes at Springfield College to play the newly invented game of basketball, the sneaker was immediately popular. Since their introduction in 1908, more that 70 million pairs of Converse sneakers have been sold worldwide. In 1916, the U.S. Rubber Company (currently named Uniroyal) introduced Keds, a sneaker made with a flexible rubber bottom and canvas upper comparable to the Converse All-Star. The first orthopedic sneaker was developed by New Balance shortly before the Great Depression. This company continues to be the world’s largest manufacturer of sneakers made with different widths. The German shoemaker Adi Dassler formed Adidas in the 1930’s, while his brother Rudi formed Puma in the 1940’s. Adidas was the more popular company and was the dominant manufacturer of sneakers until the 1960’s, when Phil Knight and Bill Bowerman created Blue Ribbon Sports. Renamed Nike Inc. in 1978, after the Greek goddess of victory, this company has remained the world’s largest producer of sneakers and sporting apparel for more than 40 years, with 2009 revenues exceeding $19 billion (8).

The design of the earliest sneaker was simple: a thin rubber sole was covered with a canvas upper, providing nominal cushioning and protection. In contrast, modern sneakers are made with synthetic leather or mesh uppers, foam midsoles and synthetic rubber outsoles to resist abrasion and improve traction (Fig. 7.4). Because runners usually make initial ground contact with the lateral heel, this area is often reinforced with a durable synthetic carbon rubber. The upper, in addition to providing space for the toes, also possesses an elaborate lacing system that has the ability to modify motion (Fig. 7.5). In their detailed analysis of foot motion and pressure distribution in runners wearing the same type of sneaker tightened with different lacing techniques, Hagen and Hennig (9) demonstrate that the high 7-eyelet lacing pattern secured with moderate tension

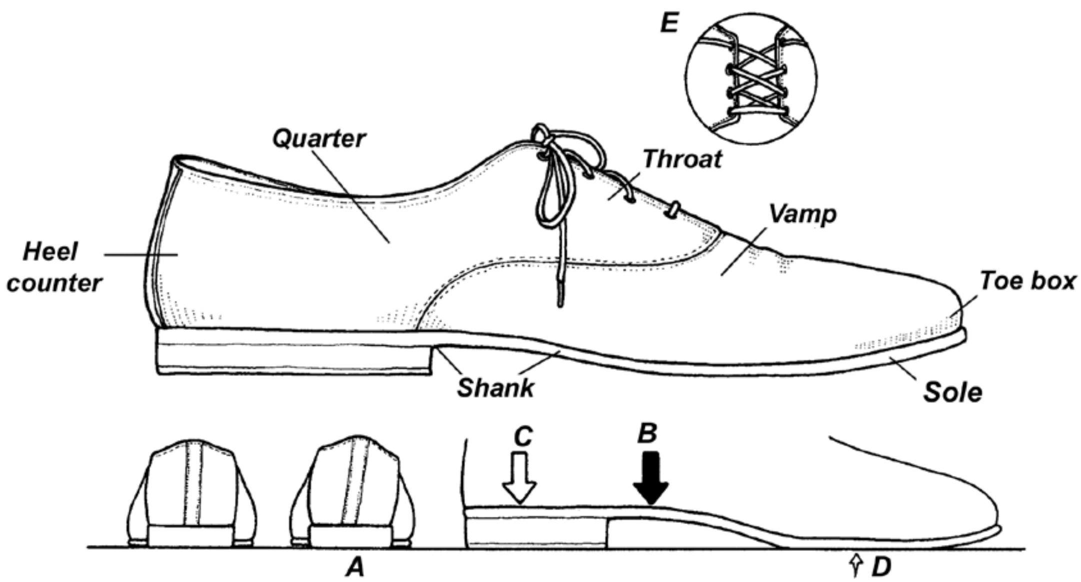

Figure 7.3. Components of a well-made leather shoe. The heel counter should fit securely and the bisection of the shoe should be vertical to the supporting surface. Poor quality control often allows for an asymmetrical heel counter that is either inverted or everted relative to the table top; see **A**. In addition, the shank should be able to resist forceful compression without deforming (**B**), and it should be angled in such a way that when the heel seat is compressed (**C**), the plantar forefoot lifts no more than a few millimeters (**D**). The toe box should provide ample space so as not to compress a dorsomedial or lateral bunion. Because it allows for greater separation of the upper, Blucher lacing (**E**) may be necessary to accommodate the midfoot in individual’s with high arches.

Chapter Eight

Treatment Protocols

As demonstrated throughout this book, the human body is remarkably well-designed to handle the forces associated with walking and running. In fact, contrary to popular belief, long distance running does not cause degenerative changes in our joints, and may even protect us from developing osteoarthritis. The long-held belief that the amplified impact forces associated with running would accelerate the development of osteoarthritis was disproved in a 25-year study by Chakravarty et al. (1). By comparing tibiofemoral joint space on serial x-rays taken between 1984 and 2002 on 45 runners and 53 age-matched controls, the authors confirmed that running altered neither the severity nor prevalence of knee osteoarthritis. In a detailed meta-analysis of the literature evaluating the progression of knee arthritis and exercise, Bosomworth (2) confirms that not only does running not accelerate the development of knee osteoarthritis, it may be protective: his meta-analysis confirmed that aging runners present with reduced rates of lower extremity disability and all-cause disability.

Although a significant body of literature confirms that running does not cause osteoarthritis, epidemiologic studies prove that runners are much more likely to develop a wide range of musculoskeletal injuries, including sprains, strains and fractures. In a thorough review of the literature, van Mechelen et al. (3) report that the annual injury rate for runners is somewhere between 37% and 56%. This is consistent with research by Lysholm and Wiklander (4), who followed 60 runners for 12 months and reported 39 of the 60 athletes developed 55 injuries. The authors reported that sprinters were most likely to be injured (with hamstring injuries being prevalent), followed by middle distance runners (who complained of backache and hip problems) and lastly, long distance runners, who presented with a greater frequency of foot injuries.

As noted by van Mechelen et al. (3), 50 to 75% of running injuries are overuse in nature and the most common cause for injury is exceeding the mileage limits we were designed to tolerate. Several studies show the potential for injury dramatically increases when we run more than 35-40 miles per week (5,6). This number makes perfect sense when you consider the early hominids had foraging distances of only 8-9 miles per day (7). Moreover, because the metabolic cost of running was so high, our ancestors rarely ran as their preferred form of travel was walking and when necessary, a hybrid pendular style shuffle jog (refer back to figure 3.10). The impulsive form of running that we are so fond of was extremely rare among our ancestors (8). The reason for this is simple: if the early hunter-gatherers ran the high mileage common among modern long distance runners, we would not have survived as a species because the metabolic cost of travel would have been so high we would have lacked the energy necessary to care for our children (8). The infrequent use of impulsive running in our hominid ancestors explains why the knee, the body's best shock absorber, is by far the most frequently injured joint while running: it was designed to manage the 2-fold increase in force associated with walking and shuffle jogging, not the 5-fold application of force associated with impulsive running.

Because we were not designed for impulsive running long distances, the best way to prevent injury is to run less than 35-40 miles per week. Of course, this is not always possible because in order to run faster, even recreational athletes run more than 60 miles per week, and it is common for professional runners to run between 120 and 150 miles per week. Unfortunately, running high mileage can have a detrimental effect on our musculoskeletal system, as an MRI study of runners' knees performed before and after a marathon revealed significant cartilage proteoglycan and collagen breakdown, particularly in the patellofemoral and medial tibiofemoral joints (9). The authors note that because the biochemical changes in the articular cartilage were present even after 3 months of reduced activity, the runners were at higher risk for degeneration. This may explain why the meta-analysis of exercise and arthritis performed by Chakravarty et al. (1) found that although moderate exercise did not predict knee osteoarthritis, elite athletes were shown to be at increased risk for developing degenerative joint changes.

To maximize the beneficial effect of training while minimizing weekly mileage (and hence, our potential for injury), experts at Furman University (10) developed a training approach that allows for improved performance with minimal strain on the musculoskeletal system. Because faster runners do not have the luxury of running reduced mileage, alternate methods of cross-training are recommended.

Should an injury develop, a thorough biomechanical exam should identify the specific problems with strength, flexibility, proprioception and/or bony alignment that might be responsible for producing the injury, and the appropriate treatment protocol initiated. Identifying the cause is essential, as once injured, reinjury rates among runners can be as high as 70% (3). This is consistent with research confirming the best predictor of future injury is prior injury (5,11). To assist the practitioner in prescribing the best possible treatment protocols, the following section reviews treatment options for a few of the more common gait-related injuries. Obviously, this list is not meant to review all possible treatments and/or injuries, it merely provides a review of the more effective biomechanical treatment interventions.

Achilles Tendinitis

Despite its broad width and significant strength, the Achilles tendon is injured with surprising regularity. In a study of 69 military cadets participating in a 6-week basic training program (which included distance running), 10 of the 69 trainees suffered an Achilles tendon overuse injury (12). The prevalence of this injury is easy to understand when you consider the tremendous strain runners place on this tendon. For example, during the push-off phase of the running cycle, the Achilles tendon is exposed to a force of 7 times body weight. This is close to the maximum strain the tendon can tolerate without rupturing (13). In addition, when you couple the high strain forces with the fact that the Achilles tendon significantly weakens as we get older, it is easy to see why this tendon is injured so frequently.

Achilles tendon overuse injuries are divided into several categories: insertional tendinitis, paratenonitis and non-insertional tendinosis. As the name implies, insertional tendinitis refers to inflammation at the attachment point of the Achilles on the heel. This type of Achilles injury typically occurs in high-arched, inflexible individuals, particularly if they possess a Haglund's deformity (see Fig. 4.45). Because a bursa is present near the Achilles attachment, it is common to have an insertional tendinitis with the retrocalcaneal bursitis. Until recently, the perceived mechanism for the development of insertional tendinitis seemed straightforward: the individual possessing a high arch and tight gastrocnemius/soleus creates an increased tensile strain in the posterior aspect of the Achilles tendon when the ankle is maximally dorsiflexed during late midstance/early propulsion. Although logical, recent research has shown that just the opposite is true (14). By placing strain gauges inside different sections of 6 cadaveric Achilles tendons and loading the tendons with the ankle positioned in a variety of angles, researchers from the University of North Carolina discovered the posterior portion of the Achilles tendon is exposed to far greater amounts of strain (particularly as the ankle was dorsiflexed) while the anterior aspect of the tendon, which is the section most frequently damaged with insertional tendinitis, was exposed to very low loads. The authors suggest that the lack of stress on the forward aspect of the Achilles tendon (which they referred to as a tension shielding effect) may cause that section to weaken and eventually fail. As a result, the treatment of an Achilles insertional tendinitis should be to strengthen the anterior aspect of the tendon. This can be accomplished by performing the eccentric load exercise illustrated in figure 8.1. It is particularly important to work

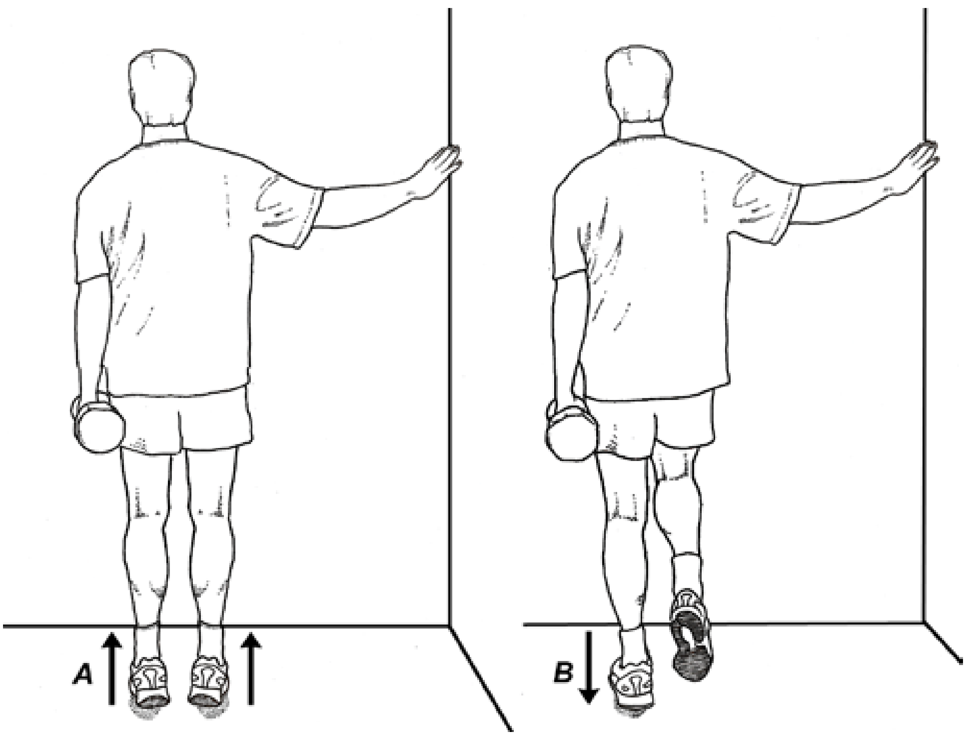

Fig. 8.1. Insertional Achilles tendinitis protocol. While standing on a level surface, the subject maximally plantarflexes both ankles (**A**) and slowly lowers the involved leg (**B**). Three sets of 15 repetitions are performed daily with enough weight to produce fatigue.

the Achilles tendon with the ankle maximally plantarflexed, because this position places greater amounts of strain on the more frequently damaged anterior aspect of the tendon.

When lateral insertion injuries occur in individuals with cavovarus feet, an effective treatment protocol is to incorporate valgus wedges. If a plantarflexed first ray is associated with the cavovarus foot, a sub-1 balance should be incorporated. In contrast, if an Achilles injury occurs in an individual who pronates excessively throughout stance phase, varus posts and/or orthotics should be considered. In an interesting EMG study of orthotics and muscle activation patterns in individuals with and without Achilles injuries, Wyndow et al. (24) confirmed that individuals with Achilles injuries are more likely to present with a premature cessation of soleus muscle activity, while the lateral gastrocnemius stays active for a prolonged period. The authors state the asymmetrical tendon tension created by the altered muscle activation pattern could cause or perpetuate an Achilles tendon injury. When muscle activity was reevaluated when the Achilles injured group wore orthotics, muscle activation patterns became more symmetrical, resulting in a more even transfer of force throughout the Achilles tendon.

Even though heel lifts are frequently prescribed to treat this injury, recent evidence suggests that when used on individuals with reduced ranges of ankle dorsiflexion, heel lifts actually increase EMG activity in the medial gastrocnemius and tibialis anterior muscles (15). Because this is consistent with prior research suggesting that heel lifts may increase strain on the Achilles tendon (16), the routine prescription of heel lifts for the treatment of Achilles tendon disorders needs to be reconsidered.

Rather than accommodating tightness in the gastrocsoleus complex with heel lifts, a better approach is to lengthen these muscles with gentle stretches. This can be accomplished with both straight and bent knee stretches to lengthen the gastrocnemius and soleus muscles, respectively. Because overly aggressive stretching can damage the insertion by pulling on it too vigorously, the stretches should be performed with mild tension and held for 35 seconds. They should be repeated every few hours and best results occur when deep tissue massage is performed prior to stretching. An alternate method to stretch the gastrocsoleus is to have the patient stand on a slant board for extended periods. The slant board should be angled to produce a mild stretch and adjusted as the range of ankle dorsiflexion increases. If calf inflexibility is extreme and does not respond to stretching, a night brace should be prescribed.

An important consideration in the treatment of insertional Achilles tendinitis relates to the forward pitch of the sneaker's heel counter (this is the back upper portion of the sneaker that touches the Achilles tendon). Over the past few years, many sneaker manufacturers have added a forward angulation to the upper portion of the heel counter that causes it to project directly into the Achilles tendon. This addition often pushes into the back the Achilles insertion causing chronic inflammation, particularly if a Haglund's deformity is present. Treatment in this situation is to look for sneakers that do not contact the Achilles insertion. A simple alternative is to cut off the upper back section of the heel counter so it no longer touches the tendon. It is also important to avoid heavy motion control sneakers when treating Achilles tendon injuries because their inherent stiffness increases the length of the lever arm from the ankle to the forefoot, thereby increasing strain on the Achilles tendon.

The second type of Achilles tendon overuse injury is paratenonitis. This injury represents an inflammatory reaction in the outer sheath of cells that surround the tendon. The first sign of this injury is a palpable lump that forms a few inches above the Achilles attachment. Treatment for Achilles paratenonitis is to reduce the swelling with frequent ice packs. Night braces are also effective with paratenonitis because tissues immobilized in a lengthened position heal more rapidly (17). If the paratenonitis worsens, it may eventually turn into a classic Achilles non-insertional tendinosis. This injury involves degeneration of the tendon approximately 2-4 cm above the attachment on the heel. Because this section of the tendon has such a poor blood supply, it is prone to injury and tends to heal very slowly.

Unlike paratenonitis, non-insertional tendinosis represents a degenerative noninflammatory condition. Apparently, repeated trauma from overuse causes fibroblasts to infiltrate the tendon, where, in an attempt to heal the injured regions, they begin to synthesize collagen. In the early stages of tendon healing, the fibroblasts manufacture almost exclusively type 3 collagen, which assists in the repair process but is relatively weak and inflexible compared to the type 1 collagen found in healthy tendons. As healing progresses, greater numbers of fibroblasts appear and collagen production shifts from type 3 to type 1. Unfortunately, the tendon is frequently unable to adequately remodel and a series of small partial ruptures begin to form that can paradoxically act to lengthen the tendon. An asymmetrical increase in the range of ankle dorsiflexion on the side of the injured Achilles tendon is a clinical sign indicative of advancing tendinopathy. Although the classic treatment for Achilles tendinosis is 6 weeks of rest (which theoretically allows the fibroblasts more time to remodel), a randomized controlled trial by Silbernagel et al. (18) reveals that tendinopathy patients who continue to exercise but monitor pain by not allowing tendon discomfort to exceed 5 on a scale of 10 do just as well as a non-exercising tendinopathy control group, even at the 12-month follow-up. The authors emphasize that a training regimen of continuous but pain-monitored tendon-loading physical activity represents a valuable option for patients with Achilles tendinopathy.