Running Movement and Biomechanical Risks of Common Lower Limb Injury

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BIOMECHANICS OF RUNNING

INTRODUCTION

Running is a popular sport worldwide, but while it is a positive way of enhancing health and fitness, it is

Running injuries occur as a result of cumulative stresses from forces placed on the body via intrinsic means (biomechanics, age etc.) as well as extrinsic factors (training volume, nutrition, type of training etc.)

Running technique, as well as running biomechanics, have been shown to affect a runner's performance. Evaluating Running Biomechanics and kinematic patterns can help to identify the forces placed on a runner as well as identify if these forces are potentially causing or aggravating their symptoms. A thorough biomechanical evaluation will enable the identify their problem areas as well as develop a comprehensive management plan to address any impairments observed. It is important to note that most of the research to date has been conducted on populations of non-symptomatic runners and as such one should keep in mind that these findings might not correlate directly with a symptomatic population.

There is a need for greater balance because the double support period presentin walking is not present when running. There is also the addition of a double float period during which both feet are off the ground, not making contact with the support surface.

The amount of time that the runner spends in float, increases as the runner increases in speed. The muscles must produce greater energy to elevate the head, arms and trunk (HAT) higher than in normal walking, and to support HAT during gait cycle. The muscles and joints, must also be able to absorb increased amounts of energy to control the weight of HAT.

During the running gait cycle, the Ground reaction force (GRF) at the center of pressure (COP) have been shown to increase to 250% of the body weight.

The Running Gait Cycle

Running is similar to walking in terms of locomotor activity. However, there are key differences. Having the ability to walk does not mean that the individual has the ability to run.

Running requires:

1. Greater balance
2. Greater muscle strength
3. Greater joint range of movement

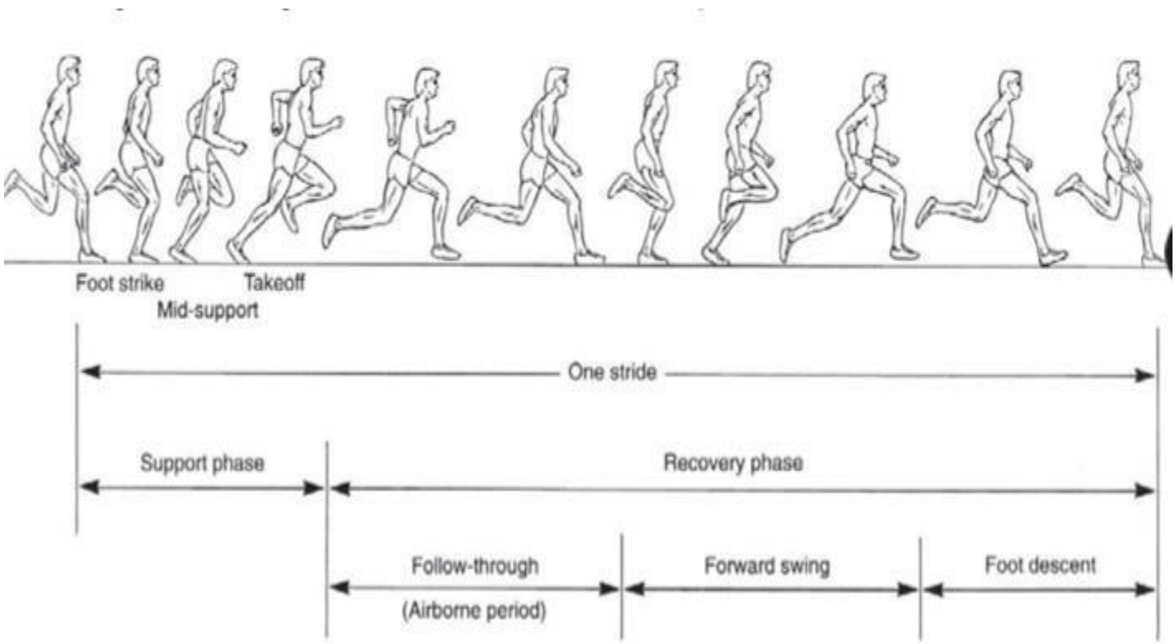


Figure 13-1
The running cycle and its various components.

Biomechanical Observation

The following kinematic patterns should be observed when analyzing the video.At present, there are no set norms or ideal values and as such, every aspect should be analyzed and then compiled into a “bigger picture” analysis.

Side View

1. Foot strike
2. Foot inclination angle at initial contact
3. Tibial angle at loading response
4. Knee flexion during stance
5. Hip extension during late stance
6. Trunk Lean
7. Overstriding
8. Vertical Displacement of Centre of Mass

Posterior View

1. Base of Support
2. Heel Eversion
3. Foot Progression Angle
4. Heel Whips
5. Knee Window
6. Pelvic Drop

Joint Motion

1. Beginning of stance phase – hip is in about 50° flexion at heel strike, continuing to extend during the rest of the stance phase. It reaches 10° of hyperextension after toe off.
2. The hip flexes to 55° flexion in the late swing phase.
3. Before the end of the swing phase, the hip extends to 50° to prepare for theheel strike.
4. The knee flexes to about 40° as the heel strikes, then flexes to 60° during theloading response.
5. The knee begins to extend after this, and reaches 40° flexion just before toe-off.
6. During swing phase and the initial part of the float period, the knee flexes to reach maximum flexion of 125° during the mid swing.
7. The knee then prepares for heel strike by extending to 40°The ankle is in about 10° of dorsiflexion when the heel strikes, and then dorsiflexes rapidly to 25° DF.
8. Plantarflexion happens almost immediately, continuing throughout the rest of the stance phase of running, and as it enters swing phase also.
9. Plantarflexion reaches a maximum of 25° in the first few seconds of of swing phase.
10. The ankle then dorsiflexes throughout the swing phase to 10° in the late stage of swing phase, preparing for heel strike.
11. The lower limb medially rotates during the swing phase, continuing to medially rotate at heel strike.
12. The foot pronates at heel strike.
13. Lateral rotation of the lower limb stance leg begins as the swing leg passes by the stance leg in mid stance position.

Table 13-1
Typical Range-of-Motion Values for the Lower Extremities During Running

Running Phase	Joint	Motion
Foot strike* to midsupport†	Hip Knee Ankle	45° to 20° flexion at midsupport 20° to 40° flexion by midsupport 5° plantar flexion to 10° dorsiflexion
Midsupport to takeoff‡	Hip Knee Ankle	20° flexion to 5° extension 40° to 15° flexion 10° to 20° dorsiflexion
Follow-through§	Hip Knee Ankle	5° to 20° hyperextension 15° to 5° flexion 20° to 30° plantar flexion
Forward swing¶	Hip Knee Ankle	5° to 20° hyperextension 15° to 5° flexion 20° to 30° plantar flexion
Foot descent‖	Hip Knee Ankle	65° to 40° flexion 130° to 20° flexion 0° to 5° dorsiflexion to 5° plantar flexion

Muscle Activity

1. Gluteus maximus and gluteus medius are both active at the beginning of stance phase, and also at the end of swing phase.
2. TFL is active from the beginning of stance, and also the end of swing phase. It is also active between early and mid-swing.
3. Adductor Magnus is active for about 25% of cycle, from late stance to early part of swing phase.
4. Iliopsoas activity occurs during swing phase for 35-60% of cycle.
5. Quadriceps works in an eccentric manner for the initial 10% of the stance phase. Its role is to control knee flexion as the knee goes through rapid flexion.
6. It stops being active after the first part of the stance phase, there is then no activity until the last 20% of swing phase. At this point it becomes concentric in behavior so it can extend the knee to prepare for heel strike.
7. Medial Hamstrings become active at the beginning of the stance phase (18- 28% of stance), they are also active throughout much of the swing phase (40- 58% of initial swing then the last 20% of swing).

8. They act to extend the hip and control the knee through concentric contraction. In late swing, the hamstrings act eccentrically to control knee extension and take the hip into extension again.
9. Gastrocnemius muscle activity starts just after loading at heel strike, remaining active up until 15% of the gait cycle (this is where its activity begins in walking). It then re-starts its activity in the last 15% of the swing phase.
10. Tibialis anterior muscle is active through both stance and swing phases in running. It is active for about 73% of the cycle (compared to 54% when walking). The swing phase when running, is 62% of the total gait cycle, compared to 40% when walking, so TA is considerably more active when running. Its activity is mainly concentric or isometric, enabling the foot to clear the support surface during the swing phase of the running gait.

Runner Classification by Foot Strike

1. Rearfoot strikers—posterior 1⁄3 of shoe strikes ground first.
2. Midfoot strikers—middle 1⁄3 of shoe strikes ground first.
3. Forefoot strikers—anterior 1⁄3 of shoe strikes ground first.

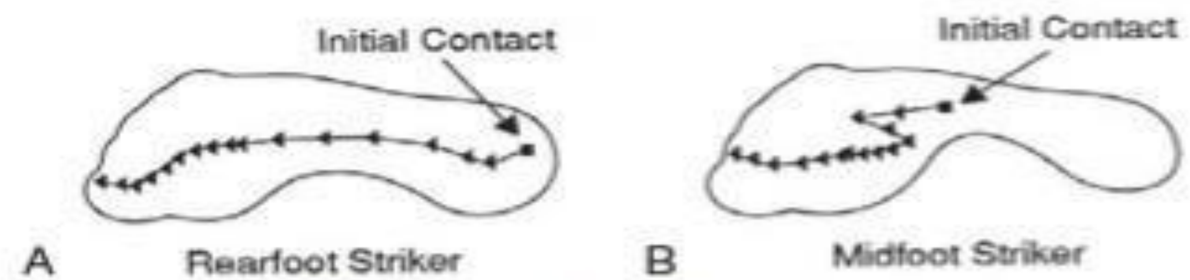


Figure 13-7
Center of pressure paths for a rearfoot and midfoot striker.

Elastic Support Strategy

Joanne Elphinstone (2013)[4] describes this as a mechanism for transferring force from the lower control zone to the upper control zone and back again.

In runners the diagonal elastic support mechanism is utilized. This is produced by aconstant diagonal stretch and release that is enabled by the body’s counter rotation.The force continually flows up and down these force pathways alternately. The pattern of force distribution prevents force being concentrated in one area, but allows wide distribution of force throughout the body.

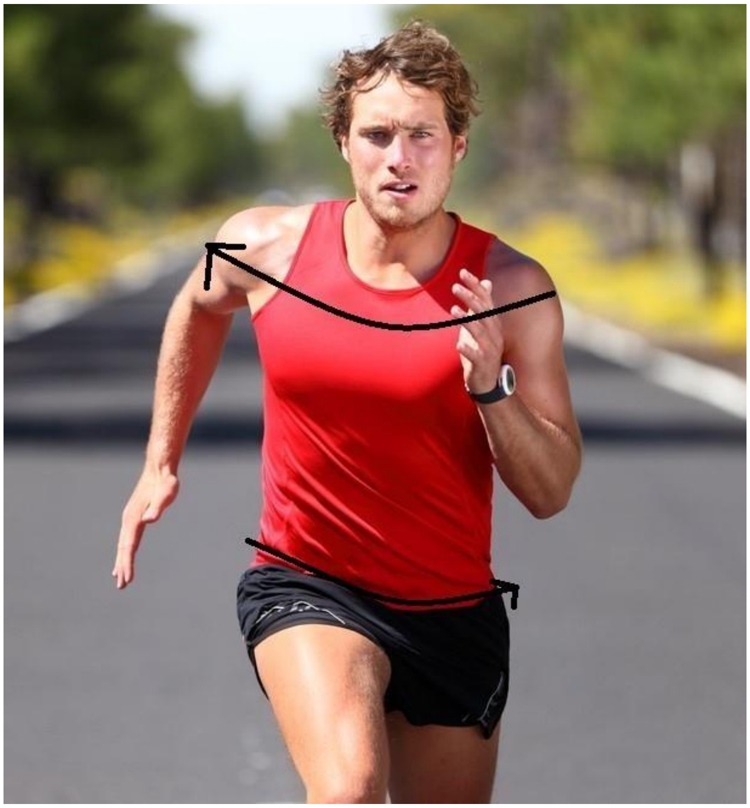
Elphinsone maintains that it is crucial to have a well-functioning central core area to allow this pattern of force distribution to take place efficiently.

Rotation through the Kinetic Chain

The kinetic chain can be described as a series of joint movements, that make up a larger movement. Running mainly uses sagittal movements as the arms and legs move forwards. However, there is also a rotational component as the joints of the leg lock to support the body weight on each side.

There is also an element of counter pelvic rotation as the chest moves forward on the opposite side. This rotation is produced at the spine, and is often referred to as the spinal engine. This is also linked to running economy. This counter rotation enables the spinal forces to be dissipated as the foot hits the ground.

Runners may complain of a feeling of restriction in hamstrings or shoulders, however, when examined it may be found that there is actually limitation in rotation of the.



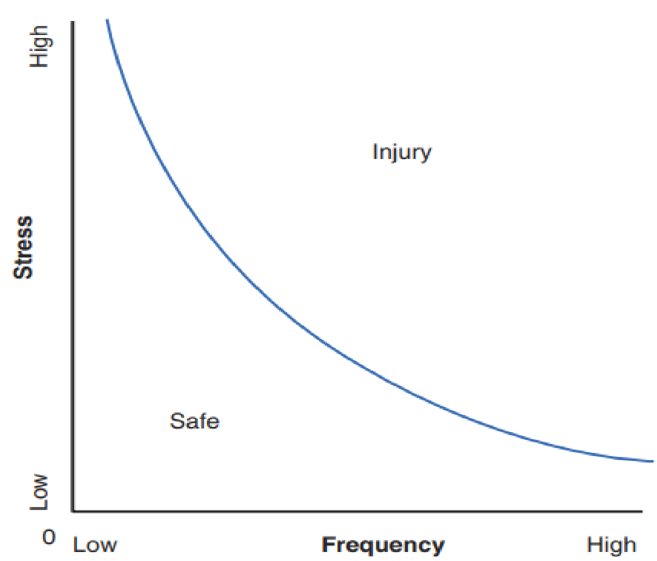
Incidence of Running-Relate Injuries

Although runners occasionally sustain acute injuries such as ankle sprains and muscle strains, the majority of running injuries can be classified as cumulative microtrauma (overuse) injuries. Running is one of the most popular activities, and overuse injuries of the lower extremity occur regularly. There is no agreed on standardized definition of an overuse running injury, but several authors have defined it as a musculoskeletal ailment attributed to running that causes a restriction of running speed, distance, duration, or frequency for a least 1 week. Using slight variations of this definition, various epidemiological studies have estimated that 27% to 70% of recreational and competitive distance runners sustain an overuse running injury during any 1-year period. The runners in these studies vary considerably in their running experience and training habits, but generally they run a minimum distance of 20 km per week (12 mi per week) on a regular basis and have been running consistently for at least one to three years.

Etiology of Over Injuries in Runners

The training variables most often identified as risk factors for overuse running injuries include running distance, training intensity, rapid increases in weekly running distance or intensity, and stretching habits. Examining how these variables affect the stress–frequency relationship reveals how some of these training variables may lead to overuse injuries. Increasing running distance increases the number of repetitions of the applied stress since the number of steps taken increases. Provided that running speed remained unchanged, the magnitude of the forces and moments produced at various musculoskeletal structures during each step remain unchanged also (neglecting fatigue effects). Thus, running a greater distance places the involved musculoskeletal structures further to the right on the graph by increasing the number of stress applications. Since this portion of the curve has a slight negative slope, locations further to the right on the curve require slightly lower stresses for a structure to enter the injury zone of the curve. Thus, the possibility that one or more structures will enter the injury zone of the graph increases with increasing running distance.

In running, training intensity relates to running speed. Faster running speeds generally produce greater forces and torsional stress to the involved musculoskeletal. When training intensity increases, the stress level applied to all of these structures occurs higher on the stress-frequency graph. Locations higher on this graph require fewer repetitions for a structure to enter the injury zone. In this way, when training intensity increases without a decrease in running distance or frequency, the likelihood of injury also increases.



Overuse injury occurrence due to the theoretical relationship between stress application and frequency of force application.

The stress–frequency relationship can also explain how rapid changes in distance or intensity increase the risk of injury. When a musculoskeletal structure is subjected to a stress-frequency combination that is close to the stress–frequency curve yet below or to the left of the curve, positive remodeling of the structure may occur, shifting the curve upward and to the right as long as detraining does not occur. When these increases in running distance and intensity are gradual, it is possible to shift the stress–frequency curve to outpace the shifting of the structure's location on the graph. However, rapid increases in running distance or intensity may cause the structure to cross the curve from the non-injury region to the injury region even when some positive remodeling and shifting of the curve has occurred.

Performing stretching exercises before running is a training-related variable that has been examined as a possible risk factor for running injuries. Unfortunately, there have been conflicting conclusions drawn regarding the association of this factor with overuse running injuries. A systematic review and meta-analysis reported that research investigating protocols of stretching before exercise and stretching outside the training sessions did not produce a clinically useful or statistically significant reduction in the risk of soft tissue running-related injuries. Without conclusive evidence, other factors, such as training errors, should be considered first as potential contributors to injury.

One of the most appealing aspects of assigning the causes of all overuse running injuries to training variables is that all of these injuries could then be considered preventable, since runners have control over training variables. However, rarely do runners know that they are about to commit a training error that will place them outside of their injury threshold. Therefore, to prevent overuse running injuries and assess and understand the etiology of a current injury, knowledge of the current limits of all of the involved musculoskeletal structures is required. These limits primarily are determined by anatomical and biomechanical variables in addition to the current state of training, strength and flexibility of specific tissues and muscles, and the integrity and injury status of various structures. Of course, it is not possible to know these limits exactly, but it is possible to minimize the risk of injury by thoroughly understanding the key clinical and biomechanical risk factors.

Common Running-Related Injuries

The knee is the most common site of overuse running injuries, accounting for close to half of all running injuries. According to a clinical study of more than 2,000 injured runners (Taunton et al. 2002), 42% had knee injuries, 92% had lower leg injuries, and 22% had injuries superior to the knee. The most common knee injury was patellofemoral pain syndrome and was seen in 331 individuals, followed by iliotibial band friction syndrome (168 cases), plantar fasciitis (158 cases), meniscal injuries (100 cases), and patellar tendinitis (96 cases). Other researchers have reported a fairly similar breakdown for the location of overuse running injuries (Clement et al. 1981; Marti et al. 1988; Rolf 1995). Although very few overuse running injuries have an established etiology (Rolf 1995), the fact that over 80% of these injuries occur at or below the knee suggests that there may be some common mechanisms.

Patellofemoral Pain Syndrome

Patellofemoral pain syndrome (PFPS) is one of the most common injuries in running and jumping sports regardless of age or sex. In a survey of patient chart data from cases of running and jumping-based sports that were referred to an outpatient sports medicine clinic over a 5-year period, PFPS was one of the most common injuries for adults aged 22 to 38 years (Matheson et al. 1989). Activities for these individuals included recreational running, fitness classes, field sports, and racket sports. With respect to running, the knee has been shown to be the most common site of injury, representing approximately 40% of all running related injuries, and PFPS accounts for 46% to 62% of these injuries (Clement and Taunton 1981; Pinshaw et al. 1984; Taunton et al. 2002).

While PFPS is a common problem experienced by active adults, the etiology of PFPS has remained vague and controversial. Unlike other knee dysfunctions such as an anterior cruciate ligament injury, which often have a specific onset and mechanism of injury, patients with PFPS generally report diffuse peripatellar and retropatellar pain of an insidious onset. In addition, the majority of patients often report pain with no discernable cause other than overuse (Dye et al. 1999; Dye 2001; Fulkerson 2002). Dye (2001) has described PFPS as an orthopedic enigma because of the continued misunderstanding of its etiology. Thus, a thorough understanding of all etiological factors is essential for properly treating and preventing this common injury.

For example, female runners are twice as likely to sustain PFPS compared to their male counterparts. One study reported that female runners exhibit increased hip internal rotation angle, which likely led to a reduced peak external knee rotation (rotation of the distal femur on the tibia) angle compared to men. In addition, female runners remained in greater amounts of tibial external rotation compared to men throughout the entire stance phase of gait. These results are in support of Yoshioka et al. (1989) who reported that women exhibit greater static external knee rotation alignment compared to men. Moreover, Tiberio (1987) suggested that excessive internal rotation of the femur may result in malalignment of the patellofemoral joint and lead to anterior knee pain. The increased hip internal rotation demonstrated by the female runners in the aforementioned study by Ferber et al. 2003, coupled with the greater knee abduction (genu valgum) may result in a greater dynamic Q-angle. An increase in the Q-angle is thought to result in higher patellofemoral joint contact forces and place a runner at greater risk for injury (Cowan et al. 1996; Mizuno et al. 2001). These results, amongst others, may also partially explain why female runners are twice as likely to develop PFPS.

Table 1.1 Frequency of the 10 Most Common Injuries in a Study of Injured Runners

Injury	Frequency of injury
Patellofemoral syndrome	16.5%
Iliotibial band syndrome	8.40%
Plantar fasciitis	7.80%
Tibial stress syndrome	4.94%
Patellar tendinitis	4.80%
Achilles tendinitis	4.80%
Gluteus medius injuries	3.50%
Tibial stress fracture	3.30%
Hamstring injuries	2.29%

Data from Taunton and Ryan 2002.

Iliotibial Band Syndrome

Iliotibial band syndrome (ITBS) is the second leading cause of knee pain in runners and the most common cause of lateral knee pain (Noble 1980; Taunton et al. 2002). Anecdotally, this syndrome has been associated with repetitive flexion and extension on a loaded knee in combination with a tight iliotibial band (Noble 1980; Orchard et al. 1996; Birnbaum et al. 2004; Miller et al. 2007; Noehren et al. 2007). Orchard et al. (1996) suggested that frictional forces between the iliotibial band and the lateral femoral condyle are greatest at 20° to 30° of knee flexion, which occurs during the first half of the stance phase of running. However, despite this well-accepted sagittal plane theory (Noble 1980; Orchard et al. 1996; Birnbaum et al. 2004), no differences have been found in the few biomechanical investigations involving knee flexion and extension patterns in runners who had ITBS compared to healthy controls (Miller et al. 2007; Noehren et al. 2007; Ferber et al. 2010).

It is possible that motions in other planes or at other joints may contribute to ITBS. The primary functions of the iliotibial band are to serve as a lateral hip and knee stabilizer and to resist hip adduction and knee internal rotation (Fredericson et al. 2000; Moore and Dalley 2005). As a result of the femoral and tibial attachments, it is possible that abnormal hip as well as foot mechanics, which both influence the knee, could play a role in the development of ITBS. For example, a study reported that female recreational runners who had previously sustained ITBS exhibited significantly greater stance phase peak hip adduction and peak knee internal rotation angles compared with a control group during running (Ferber et al. 2010). These results were similar to those reported for a prospective study conducted in the same laboratory environment with a separate group of subjects (Noehren et al. 2007). The common results between the prospective study and the retrospective study provide strong evidence related to atypical running mechanics and the etiology of ITBS. However, other factors, such as reduced hip abductor muscle strength (Fredericson et al. 2000; Ireland et al. 2003; Niemuth et al. 2005) and atypical anatomical alignment (Horton and Hall 1989),may also play a role in the development of this particular injury.

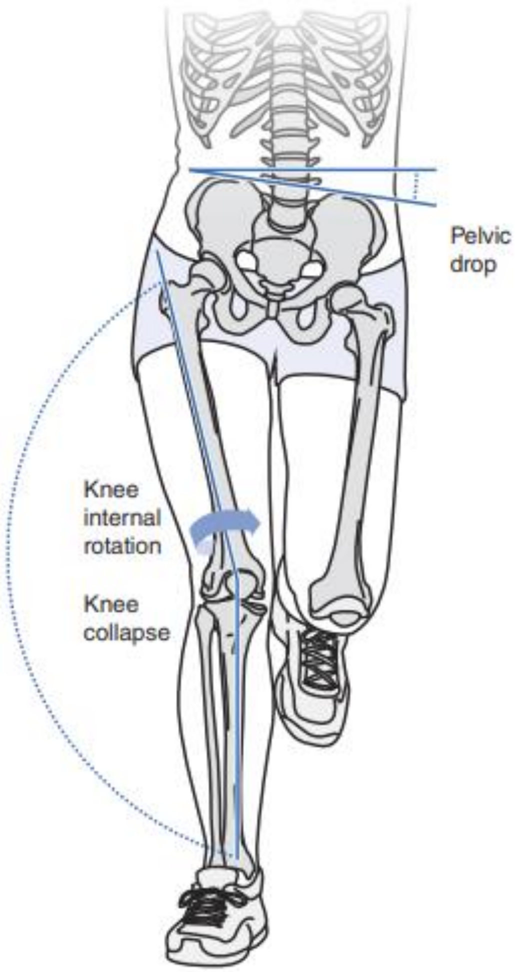


Figure 1.2 A runner exhibiting a combination of excessive pelvic drop, knee collapse, and knee rotation.

Understanding Clinical and Biomechanical Risk Factors

Although there has been a large amount of speculation regarding the mechanisms of running injuries, the exact causes of overuse running injuries have yet to be determined. It could only be stated with certainty that the etiology of these injuries is multifactorial and diverse (Marti et al. 1988; Rolf 1995; van Mechelen 1995). It has occasionally been suggested that particular running injuries, or sites of injuries, are associated with specific risk factors, but some researchers have concluded that there are no specific risk factors that correlate with specific types of injury in a reliable fashion (James and Jones 1990; James 1998). However, different research studies have investigated various risk factors in isolation and have provided reasonable rationale as to why they may be associated with a variety of running injuries.

To provide a comprehensive injury assessment for a running-related injury, four factors must be considered:

1. Biomechanical gait patterns
2. Muscular strength
3. Anatomical alignment
4. Tissue flexibility

Assessing Foot Mechanics

Biomechanics

The stance phase of walking or running gait can be divided into two functional phases. The first half of stance is commonly referred to as the cushioning, or eccentric, phase of gait. The last half of stance is referred to as the propulsion, or concentric phase. When the foot strikes the ground, it is supinated, or locked, to better attenuate the initial shock wave traveling up into the foot. Just before midstance the foot pronates, or unlocks. Then, as the heel lifts off the ground in preparation for toe-off, the foot once again supinates to allow the first ray to become a rigid lever to propel the runner forward

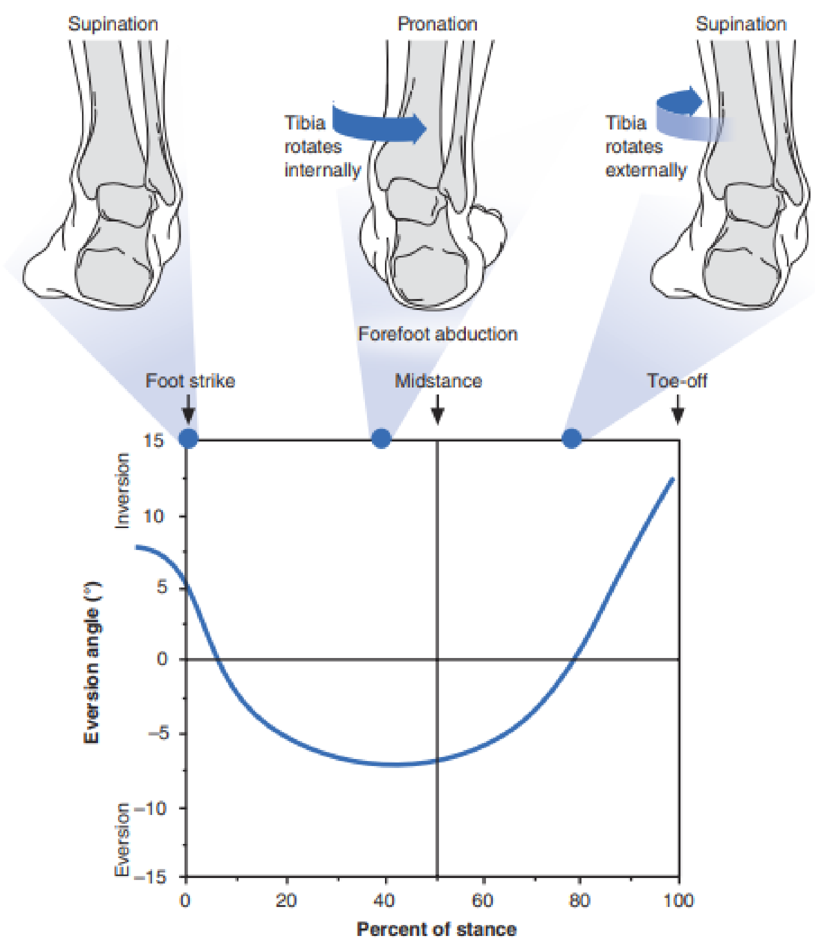


Figure 2.1 Stance phase of walking or running.

Pronation is a combination of ankle dorsiflexion, rearfoot eversion (figure 2.3a), and forefoot abduction, and it occurs during the first half of stance (the cushioning, or eccentric, phase). The reverse movement is supination (figure 2.2b) which the foot moves back toward after heel-lift. As seen in biomechanical graphs, the foot lands in a dorsiflexed position and then rapidly plantar flexes along a sagittal plane to bring the foot completely in contact with the ground (figure 2.4). Once the foot is completely on the ground, the ankle once again dorsiflexes as the shank moves anteriorly over the fixated foot; this dorsiflexion is a component of overall foot pronation. Coupled with this dorsiflexion motion along the frontal plane, as seen in figure 2.5, the foot everts from the instance of heel strike until approximately midstance, and the shank internally rotates during this same period (figure 2.6). So all three components of overall foot pronation simultaneously occur from about 20% of stance until 50% of stance. Any frontal or transverse plane motion before 20% of stance occurs while the foot is not completely in contact with the ground and cannot be considered pronation based on the aforementioned definition. Foot pronation is a necessary and protective mechanism since it allows for

- impact forces to be attenuated over a longer period,
- the foot to accommodate uneven surfaces, and
- the foot to roll inward so that the first ray makes contact with the ground in preparation for resupination after heel lift.

With respect to the midfoot and associated deformation and forefoot abduction during the first half of stance, Dugan and Bhat (2005) investigated the 3-dimensional (3D) motion between rearfoot, midfoot, and forefoot segments in 18 healthy male subjects while walking barefoot. They reported that the contribution of midfoot motion was equivalent to 25% to 45% of the combined motions for both the forefoot and rearfoot segments and that total transverse plane (rotational) motion for the forefoot exceeded that of the rearfoot segment. Cornwall and McPoil (2004) also investigated the relative movement of the midfoot, measuring arch deformation and the possible relationship to dynamic rearfoot eversion. They concluded that the midfoot undergoes significant vertical and medial displacement during barefoot walking and that this motion is highly correlated with rearfoot eversion. Thus, lack of motion at the arch requires greater relative rearfoot eversion to compensate, while midfoot hypermobility requires less relative rearfoot eversion. These results were supported by Hunt et al. (2001) and Leardini et al. (2007), who reported that the arch undergoes significant deformation during the first 74% of the stance phase of walking.

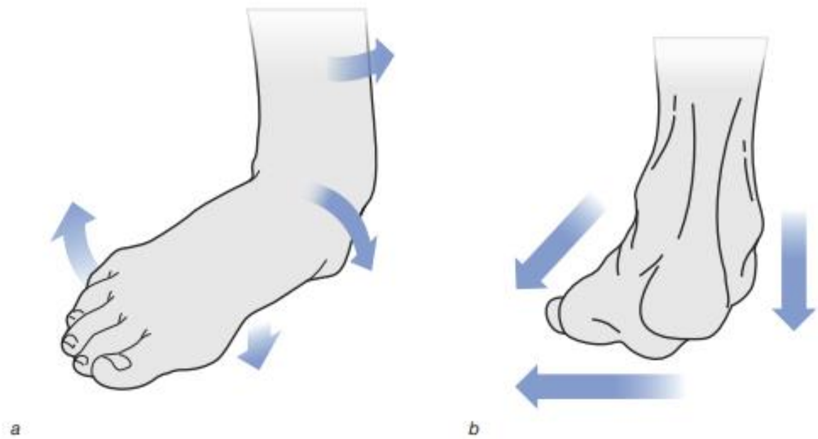


Figure 2.2 *(a)* Foot pronation and *(b)* supination.

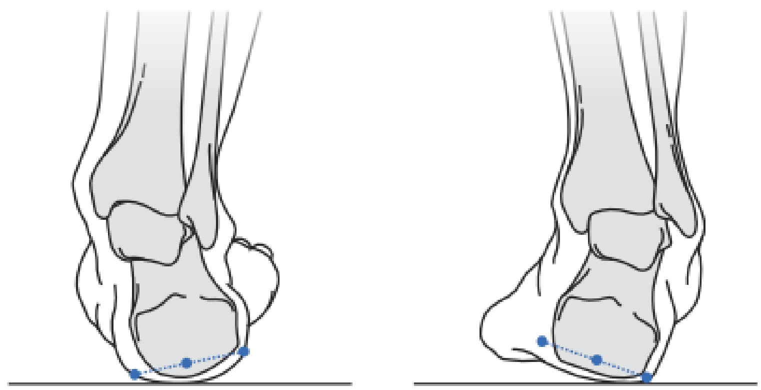


Figure 2.3 Rearfoot *(a)* eversion and *(b)* inversion.

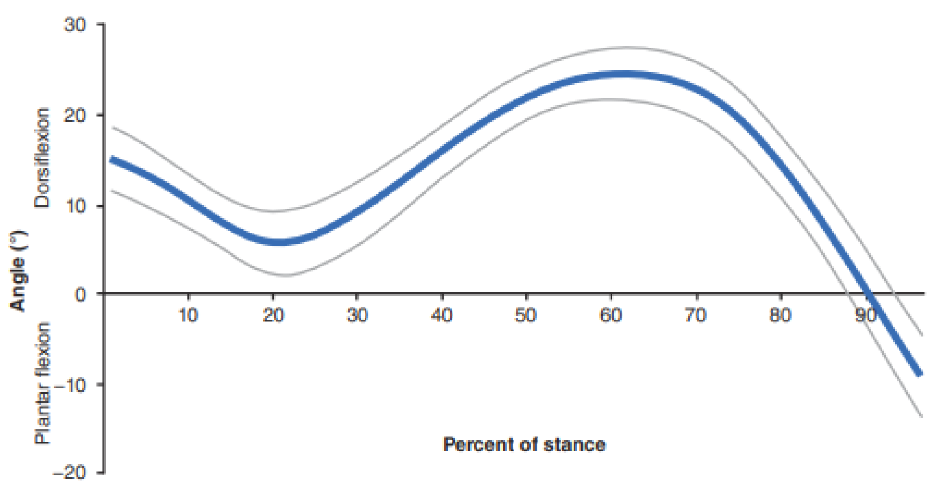


Figure 2.4 Biomechanical motion of ankle sagittal plane motion during the stance phase of running gait. A stance of 0% indicates heel strike, and a stance of 100% indicates toe-off.

Note: Blue line = mean; gray lines = ± 1 SD.

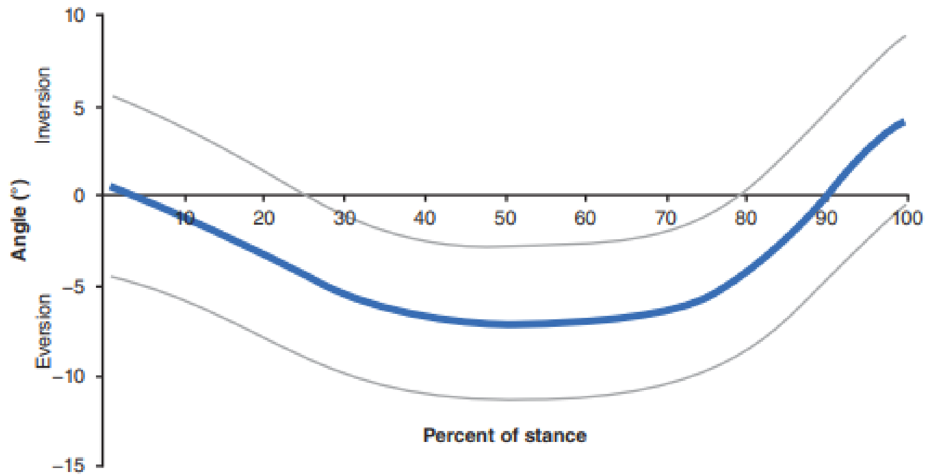


Figure 2.5 Biomechanical motion of ankle frontal plane motion during the stance phase of running gait. A stance of 0% indicates heel strike, and a stance of 100% indicates toe-off.
Note: Blue line = mean; gray lines = ± 1 SD.

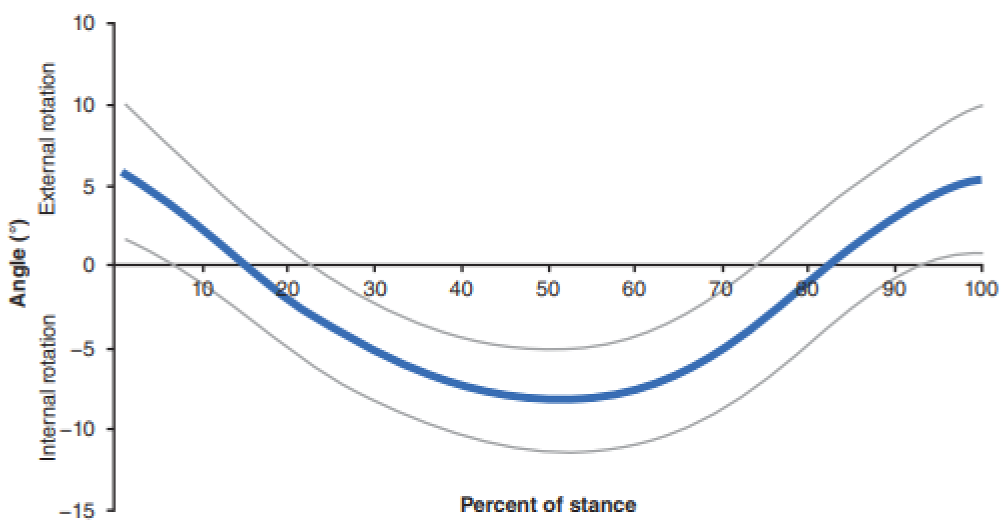


Figure 2.6 Biomechanical motion of tibial transverse plane motion during the stance phase of running gait. A stance of 0% indicates heel strike, and a stance of 100% indicates toe-off.
Note: Blue line = mean; gray lines = ± 1 SD.

Rearfoot eversion influences lower extremity mechanics via tibial rotation. During closed-chain pronation, when the calcaneus is fixed to the ground, it cannot abduct relative to the talus. Therefore, to obtain the transverse plane component of subtalar joint pronation, the talus adducts and medially rotates. Due to the tight articulation of the ankle mortise, the tibia internally rotates as the talus adducts (see figure 2.7). During this cushioning phase of stance, the knee joint flexes, which is also associated with tibial internal rotation. During gait, there is a direct relationship between the degree of pronation and internal tibial rotation for runners who exhibit a heel–toe footfall pattern. Because there is normally more rearfoot eversion than tibial internal rotation, this ratio has been reported to vary between 1.2 and 1.8. In other words, every 1° of tibial internal rotation is associated with approximately 1.2° to 1.8° of rearfoot eversion.

Researchers (Subotnick 1995; Hreljac 2004) suggested that a higher level of pronation is favorable during running if it falls within so-called normal physiological limits and does not continue beyond midstance. After midstance, it is necessary for the foot to become more rigid and supinate (tibia and talus externally rotate and rearfoot inverts) in preparation for toe-off (figure 2.7). Severe over-pronators, or runners who exhibit prolonged pronation, may be at an increased risk of injury due to the potentially large torques generated in the lower extremity and the subsequent increase in internal tibial rotation. This situation results in a mechanical dilemma at the knee because knee extension begins around midstance and must be accompanied by tibial external rotation to maintain joint congruity. However, since the tibia is internally rotated with the rearfoot, the hip must excessively internally rotate to maintain proper knee and patellofemoral joint position. The compensatory hip internal rotation may alter normal patellofemoral alignment and cause excessive contact pressures on the patella. This excessive pressure eventually leads to cartilage breakdown and anterior knee pain.

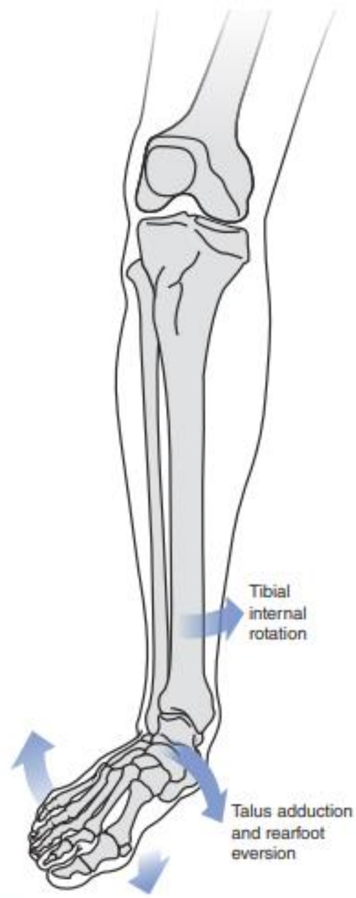


Figure 2.7 Tibial internal rotation, talus adduction, and knee flexion during the stance phase of gait.

Strength

When in a pronated position, the subtalar and talocrural joints are more mobile and require more muscle work to maintain stability compared to a supinated position. Several muscles are involved in this dynamic support, including but not limited to the tibialis posterior, peroneus longus, and tibialis anterior. The tibialis posterior is believed to play a key role in controlling rearfoot eversion and providing dynamic support across the midfoot and forefoot during the stance phase of gait. Furthermore, it has been postulated that when the tibialis posterior muscle is weak, greater rear foot eversion is measured. However, the direct connection between reduced strength and altered biomechanics is difficult to discern based on a review of the current literature.

The proximal origin of tibialis posterior lies on the interosseous membrane and posterior surfaces of the tibia and fibula. The muscle has multiple distal insertions including the navicular tubercle, the plantar surface of the cuneiforms and cuboid, and bases of the second, third, and fourth metatarsals (figure 2.8).

Biomechanical research conducted on patients with posterior tibialis tendon dysfunction (PTTD) while involving a fatigue protocol highlights the importance of this muscle in controlling rearfoot, midfoot, and forefoot mechanics during gait (Rattanaprasert et al. 1999; Tome et al. 2006; Ness et al. 2008). A tibialis posterior fatigue protocol and repeated bouts of exercise, show that the force output from this muscle is reduced by over 30%. For the fatigue protocol, subjects were seated in a chair while their right foot was placed in a custom-built device that allowed them to perform concentric and eccentric foot adduction contractions with adjustable resistance. Then with the ankle positioned in 20° plantar flexion, subjects performed sets of 50 concentric and eccentric contractions at 50% of maximal voluntary contraction (MVC) force through a 30° range of motion. The subjects were allowed 10 seconds of rest between each set, and after every four sets, maximum voluntary isometric force output was measured again. The sets were continued until subjects' isometric force output had dropped below 70% of the pre-fatigue values or they were unable to complete two consecutive sets.

Little to no change in rearfoot or forefoot kinematics was observed as a result of the reduction in force output. Specifically, only a 0.7° increase in peak rearfoot eversion was reported as statistically significant, but this change was smaller than the precision error of a within-day gait analysis of 0.9°. Therefore, the results were not clinically relevant, and it is possible that other muscles, such as the tibialis anterior, may have compensated for the lack of tibialis posterior force production, thereby resulting in no change in discrete kinematic variables. However, inspection of the data also revealed that 24 out of 29 participants demonstrated an increase in peak rearfoot angle following fatigue (ranging from 0.5° to 2.0°).

Despite these results not being statistically significant, there was a pattern consistent enough to be of interest, and we decided to explore the data further. Since such a consistent change was observed, it raises the question of what other mechanisms and potential explanations can account for these systematic changes. Thus, we reanalyzed the data based on joint coupling and coupling variability. This analysis revealed increases in coupling motion of the shank in the transverse plane and forefoot in the sagittal plane and transverse plane relative to frontal plane motion of the rearfoot. In addition, an increase in joint coupling variability was measured between the shank and rearfoot and between the rearfoot and forefoot during the fatigue condition. It was concluded that once the tibialis posterior muscle was fatigued, fewer muscles were functioning to achieve a desired movement pattern, and alterations in joint coupling and coupling variability resulted.

Based on the redundancy of the various muscles that serve to control frontal plane rearfoot and transverse plane tibial motion, a potential strategy for the foot may be to increase coupling variability to avoid injury when the function of some muscles is compromised. Therefore, it can be hypothesized that with a diminished ability of the tibialis posterior muscle to produce a vigorous contraction, a concomitant reduction in joint contact force and a resulting increase in joint coupling variability could result. In other words, the reduced function of the tibialis posterior muscle after fatigue could result in less control of the ankle joint movement since fewer muscles are functioning to achieve a movement pattern that minimizes injury potential or pain while running.

Reduced force output from the tibialis posterior does not necessarily or automatically result in a greater peak rearfoot eversion angle. However, lack of strength from the tibialis posterior could be the root cause of several different musculoskeletal injuries based on its interrelationship with other muscles and the aforementioned changes in coupling among the rearfoot, tibia, and forefoot. For example, the tibialis posterior and soleus are the two primary stabilizing muscles at the ankle joint. Collectively, these two muscles have two main functions: to minimize torsional forces at the ankle and lower leg and to control rearfoot eversion during the stance phase of gait. In addition, the tibialis posterior muscle attaches to multiple sites on the plantar surface of the foot (figure 2.8) and thus serves to dynamically support the medial longitudinal arch. If the tibialis posterior muscle cannot produce adequate force, greater stress is placed on the soleus muscle to accomplish the aforementioned tasks, which partially explains the development of Achilles tendinopathy and overall calf pain and tightness. In addition, weakness of the tibialis posterior directly increases stress to the plantar fascia, which serves to statically support the arch of the foot and can help explain the development of plantar fasciitis.

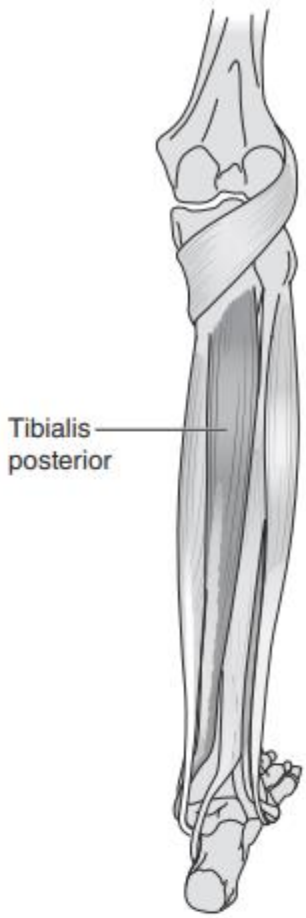


Figure 2.8 Tibialis posterior.

Anatomical Alignment

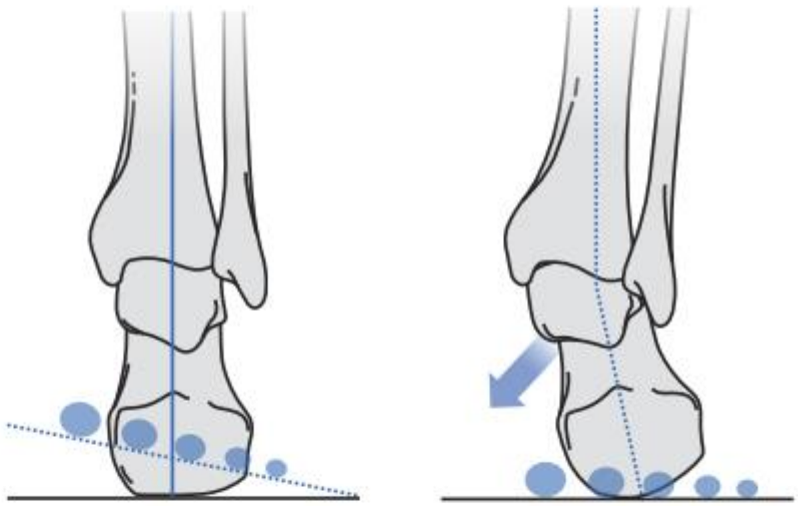
While excessive rearfoot motion during gait has received much attention in the literature, its relationship with anatomical structure remains unclear. Overall, one cannot assume that there is a direct relationship between anatomical foot structure and dynamic foot biomechanics. For instance, one study reported that a greater standing rearfoot angle was associated with greater measures of rearfoot eversion during walking. In contrast, Cornwall and McPoil (2004) showed no relationship between static measures and dynamic rearfoot motion.

The conflicting findings may be due to neglecting the role of muscular support when studying the relationship between the static and dynamic behavior of the rearfoot. For example, subjects with pronated foot posture have been shown to exhibit increased tibialis posterior activity as compared with those with a normal foot structure. Individuals with structural deficiencies, such as excessive rearfoot valgus, may rely more heavily on muscular contributions to control rearfoot kinematics during gait. Thus, it might be expected that these subjects would undergo greater changes in rearfoot kinematics after fatiguing exercise of a major invertor muscle. However, based on the literature, this assumption is wrong.

In the aforementioned tibialis posterior fatigue studies, there was a poor relationship between the standing rearfoot angle and changes in rearfoot walking kinematics following fatigue. Therefore, subjects who had greater standing rearfoot valgus angles did not rely more on tibialis posterior to control rearfoot motion during walking. These results suggest that the anatomical structure of the foot is not associated with an increase in muscular activity required to maintain normal foot kinematics during gait. However, other muscles may have compensated for reduced force output from the tibialis posterior. Therefore, it is possible that compensation strategies masked the true relationship between anatomical structure and tibialis posterior contribution, which explains the increase in rearfoot, tibia, and forefoot coupling variability. Moreover, several structural aspects of the foot were not included in the previously mentioned studies, such as forefoot alignment, which are important to consider.

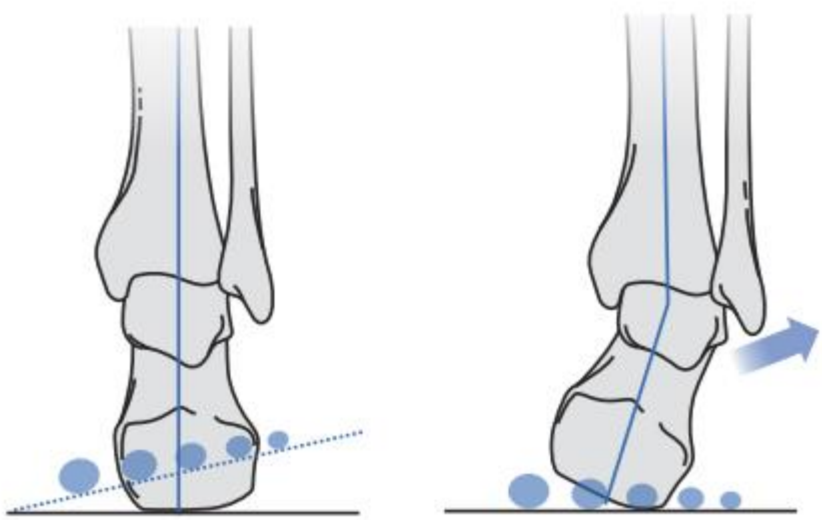
Another structural measure to consider is forefoot orientation relative to the rearfoot. Clinically, forefoot varus contributes to decreasing the medial longitudinal arch and therefore resembles pes planus (flat feet). During the stance phase of running the midfoot and forefoot are completely pronated in an attempt to bring the first metatarsal head in contact with the ground (figure 2.2a). The forefoot varus position necessitates greater rearfoot eversion to bring the first digit toward the ground to act as a rigid lever in preparation for toe-off (figure 2.9).

In contrast, forefoot valgus contributes to increasing the medial longitudinal arch and therefore resembles pes cavus (high arches). During the stance phase of running this places the rearfoot in a more supinated (reduced eversion or an inverted position) position so that the lateral aspect of the foot is in contact with the ground (figure 2.10). A study by Buchanan and Davis (2005) measured forefoot varus and valgus and standing rearfoot angle on 51 individuals. Forefoot angles were obtained with the subject prone on the table and their foot in a neutral position. Varus (positive degrees), neutral (0°), or valgus (negative degrees) was measured as the angle between the bisection of the calcaneus and an imaginary perpendicular line drawn through the metatarsal heads (figure 2.11). Rearfoot angles were obtained with the subject standing and measured as the angle between the bisection of the lower one third of the leg and the bisection of the calcaneus. The authors reported that a forefoot varus angle was present in 92% and a forefoot valgus angle was present in 8% of the cases. These results were also in agreement with data reported by Donatelli et al. (1999) and Garbalosa et al. (1994). Buchanan and Davis (2005) also examined the relationship between these two anatomical measures and reported a strong relationship. Specifically, a rearfoot valgus standing posture is most likely associated with a forefoot varus alignment, whereas a rearfoot varus standing posture is associated with a forefoot valgus alignment.



a *b*

Figure 2.9 Example of the relationship between forefoot varus, midfoot collapse, and rearfoot eversion during pronation. With (*a*) forefoot varus, the first metatarsophalangeal (MTP) joint is rotated upwards, relative to a neutral rearfoot, and a (*b*) rearfoot valgus position is necessary to bring all metatarsals in contact with the ground.



a *b*

Figure 2.10 Example of the midfoot valgus, rearfoot inversion, and lack of midfoot collapse during pronation. With (*a*) forefoot valgus, the fifth MTP is rotated upwards, relative to a neutral rearfoot, and a (*b*) **rearfoot varus** position is necessary to bring all metatarsals in contact with the ground.

Flexibility

While there is limited research concerning the interrelationship between tissue flexibility and overall foot biomechanics, first ray mobility has been examined with respect to rearfoot motion. Cornwall and McPoil (2004) conducted a study to determine whether hypo- or hypermobility of the first ray influences rearfoot eversion during walking. Static measure of first ray mobility in 82 individuals (N = 164 feet) was measured and then classified as being hypomobile (n = 31), normal (n = 111), or hypermobile (n = 22). It was reported that a hypomobile first ray resulted in significantly more rearfoot eversion than those with either normal or hypermobile first rays. Thus, with a hypomobile first ray, greater rearfoot eversion is necessary to achieve overall foot pronation and should be considered when quantifying typical and atypical foot pronation mechanics. It is also interesting to note that the 82 subjects involved in this study were the same 82 subjects involved in a previous study that measured rearfoot standing posture, which found no correlation to rearfoot mechanics (Cornwall and McPoil 2004). A combined analysis of rearfoot standing posture and first ray mobility could have provided a more comprehensive understanding of rearfoot eversion biomechanics than the isolated analysis conducted.

Finally, adequate gastrocnemius and soleus flexibility is a critical component for proper foot biomechanics. Specifically, since talocrural joint dorsiflexion is a component of overall foot pronation, reduced relative motion between the gastrocnemius and soleus muscles can result in reduced ankle dorsiflexion and knee extension. Since the gastrocnemius muscle crosses the knee and ankle joints, reduced gastrocnemius muscle flexibility also influences the kinematic patterns of the lower extremities during gait. This study (You et al. 2009) reported that gastrocnemius tightness, defined as less than 12° for passive dorsiflexion with the knee extended, resulted in several different compensatory patterns including greater hip and knee flexion angles at the time of maximal ankle dorsiflexion and reduced knee energy absorption but increased ankle energy absorption during the first half of stance. Thus, reduced calf muscle flexibility can result in a redistribution of forces throughout the lower extremity and must be considered when trying to understand injury etiology.

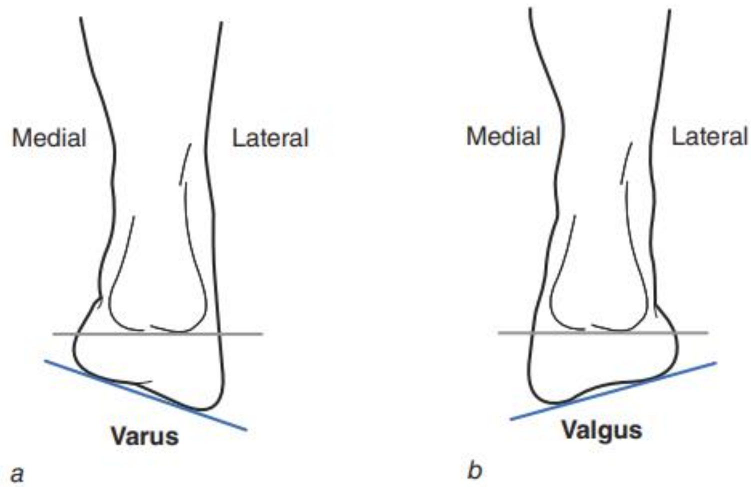


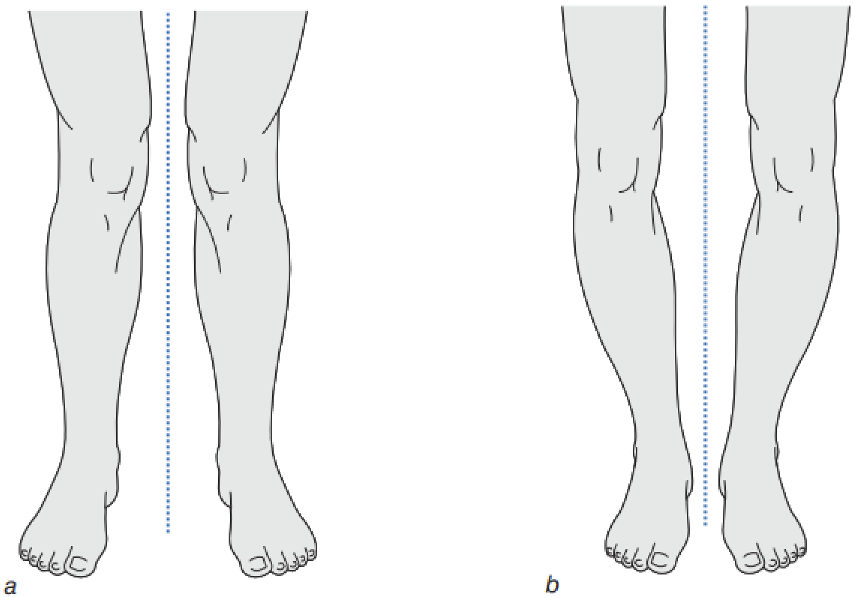
Figure 2.11 Forefoot angle measurement in prone relative to rearfoot angle in standing for *(a)* forefoot varus and *(b)* forefoot valgus.

Assessing Knee Mechanics

Biomechanics

During foot pronation, when the rearfoot is fixed to the ground, the calcaneus cannot abduct relative to the talus. Therefore, to obtain the transverse plane component of subtalar joint pronation, the talus adducts or medially rotates. Due to the tight articulation of the ankle mortise, the tibia internally rotates as the talus adducts. These mechanically linked motions occur during the first half of the stance phase of gait along with knee joint flexion, internal rotation, and adduction (figures 4.1-4.3). However, it is important to note that for most of the stance phase of gait, the knee undergoes only 4° to 6° of frontal plane motion and remains primarily in an inwardly collapsed position commonly called a genu valgum position (figure 4.4a). Rearfoot eversion, tibial internal rotation, knee flexion, and knee internal rotation (distal femur rotation on the tibia) occur relatively synchronously. During the second phase of gait, the propulsive phase of stance, these motions reverse, and the rearfoot inverts as the tibia and knee externally rotate when the knee extends

Although the relative timing between the foot and the knee is nearly synchronous, minor variations in relative timing have been reported, and several investigations have reported relative asynchrony between these motions. For example, several authors have reported that peak rearfoot eversion occurs between 39% and 54% of stance while peak knee flexion occurs between 36% and 45% of stance (James et al. 1978; Van Woensel and Cavanaugh 1992; McClay and Manal 1997; Stergiou et al. 1999; De Wit and De Clercq 2000). Moreover, inspection of figure 4.3 (knee rotation) in comparison with figure 2.6 (tibial rotation) shows that knee external rotation occurs in an asynchronous manner as compared to the tibia. Specifically, tibial external rotation begins near 50% of stance, coincident with rearfoot inversion, whereas knee external rotation occurs closer to 70% of stance. Previous studies have also measured the relative motion between rearfoot and knee in much the same way as was previously discussed in chapter 2 for the rearfoot and tibia.



a b

Figure 4.4 *(a)* Genu valgum and *(b)* genu varum.

The degree of coupling between the rearfoot and knee is influenced by the orientation of the subtalar joint axis in the sagittal plane. Using intracortical bone pins placed into the calcaneus and talus of cadaveric feet, Manter (1941) and Root et al. (1966) have reported that the orientation of the subtalar joint axis in the sagittal plane is approximately 41° to 42°, relative to the horizontal, with a range from 25° to 51°. If the subtalar joint axis were oriented at 45° in the sagittal plane, close to the average seen in these cadaver studies, one would expect equal amounts of rearfoot eversion and transverse plane tibial and knee internal rotation. However, as discussed in chapter 2, there is approximately 1.2° to 1.8° of rearfoot eversion for every degree of tibial internal rotation; the knee and tibia both exhibit less relative rotation than the frontal motion of the rearfoot. To explain this discrepancy, consider that the early studies of rearfoot and knee coupling involved measurements of cadaveric feet in a non-weightbearing position, likely resulting in a higher orientation of the axis than is present during stance.

Lundberg et al. (1989) measured the orientation of the subtalar joint axis during stance in healthy subjects. Using an interesting approach, tantalum balls were injected into various bones of the foot, and relative 3D joint positions were measured using 2D biplanar radiographs. The authors reported an average orientation of subtalar joint axis of only 32° in the sagittal plane, ranging from 14° to 40°. Contrary to the earlier studies, but consistent with the biomechanical studies already discussed, these data suggest that the knee exhibits less relative rotation than the frontal motion of the rearfoot and less relative overall internal and external rotation than the tibia. Again, comparison of figure 4.3 and 2.6 support these findings and demonstrate these interrelationships. The clinical interpretation of the range of variation among individuals is that there is no norm and an individualized assessment must be performed for each patient to gain insight into the individual's specific biomechanical coupling pattern.

The rationale for understanding the timing of joint movements is based on the notion that asynchrony in these motions may result in injury. For example, Tiberio (1987) first proposed an interesting hypothetical mechanism for PFPS related to atypical or asynchronous joint coupling. He theorized that if the time to peak rearfoot eversion is prolonged and continues beyond midstance, tibial internal rotation would also be prolonged. These atypical mechanics result in a mechanical dilemma at the knee because knee extension begins around midstance and must be accompanied by tibial external rotation to maintain joint symmetry and congruity. However, since the tibia is continuing to internally rotate with the rearfoot, the femur must excessively internally rotate to obtain the relative knee external rotation needed for the second half of the stance phase. The result would be an increased amount of dynamic knee genu valgum (abduction), since knee internal rotation and abduction are mechanically linked, and an increased potential for other tissues to undergo atypical stress. Tiberio also went on to hypothesize that the compensatory femoral internal rotation may alter normal patellofemoral alignment and cause excessive contact pressures on the patella. This excessive pressure is thought to eventually lead to cartilage breakdown and anterior knee pain (Buchbinder et al. 1979; Cowan et al. 1996; Mizuno et al. 2001). However, one mechanism to resist or minimize the international rotation and concomitant patellar contact pressures would be adequate hip rotator strength (discussed in chapter 5) as well as those muscles responsible for controlling the knee.

Strength

Many muscles cross the knee joint and serve to provide dynamic control and stabilization. Most pertinent to clinical gait analysis are the hamstring and quadriceps muscles. The hamstrings primarily function during the swing phase of gait to eccentrically control knee extension via control of the lower leg, or shank, in preparation for heel strike. Thus, weakness of the hamstring musculature results in reduced hip extension at toe-off (figure 4.5a) to minimize the inertial motion of the shank and concomitant demand on the hamstring muscles during the swing phase. Moreover, since hip extension at toe-off is reduced, a reduced stride length and increased stride frequency will be measured on a step-by-step basis.

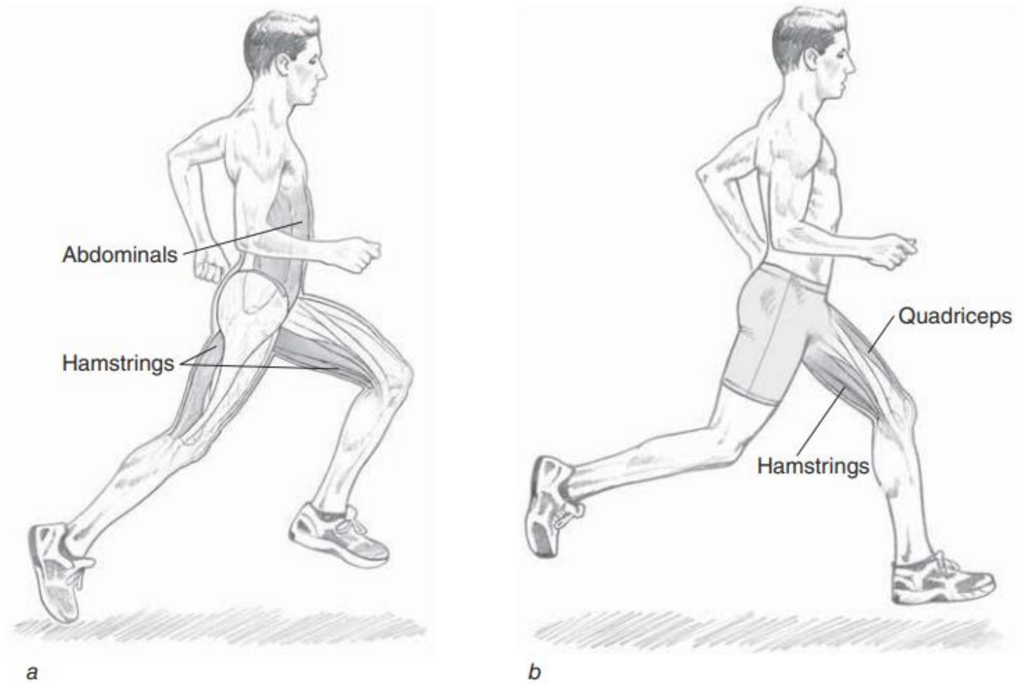


Figure 4.5 Muscle activity during the *(a)* toe-off and *(b)* forward swing phase of running gait.

During the stance phase of gait, the hamstring muscles serve to stabilize the knee and maintain relative position of the tibia and femur with respect to translational motion. Pandy and Shelburne (1997) and Li et al. (1999) reported that during stance anterior tibial shear increases in magnitude from full extension to 15° flexion, then decreases as the knee flexes. The knee is near full extension at two points during stance: after heel strike (figure 4.6a) and near the end of stance (figure 4.6b). Thus, the hamstring muscles must serve as dynamic synergists to the anterior cruciate ligament and assist in reducing anterior tibial shear at these two points (Pandy and Shelburne 1997; Osternig et al. 2000). Weakness of the hamstring muscles allows for increased anterior tibial translation and increased shear forces to the anterior cruciate ligament and menisci of the knee joint during the first 20% of stance. Interestingly, the increase in shear forces is coupled with reciprocal action and down-regulation in force output from the quadriceps muscles (Osternig et al. 2000; Ferber et al. 2002b). The hamstring muscles, however, do less overall work during the stance phase of gait than the quadriceps (Anderson and Pandy 2003).

From foot flat to 20% of stance, the vasti muscles (medialis, lateralis, intermedius), along with the gluteus maximus and medius (discussed further in chapter 5) produce the majority of support and prevent the knee from collapsing against the downward pull of gravity and during knee flexion. Interestingly, rectus femoris, a biarticular muscle, contributes very little to overall support. In contrast, the uni-articular vasti muscles remain active throughout most of midstance until heel lift and, as previously discussed in chapter 2, generate nearly all support in late stance along with the soleus and gastrocnemius. All other muscles crossing the knee joint, including sartorius, gracilis, popliteus, and iliotibial (IT) band (via action of the tensor fasciae latae), serve to minimize transverse plane motions of knee internal and external rotation.

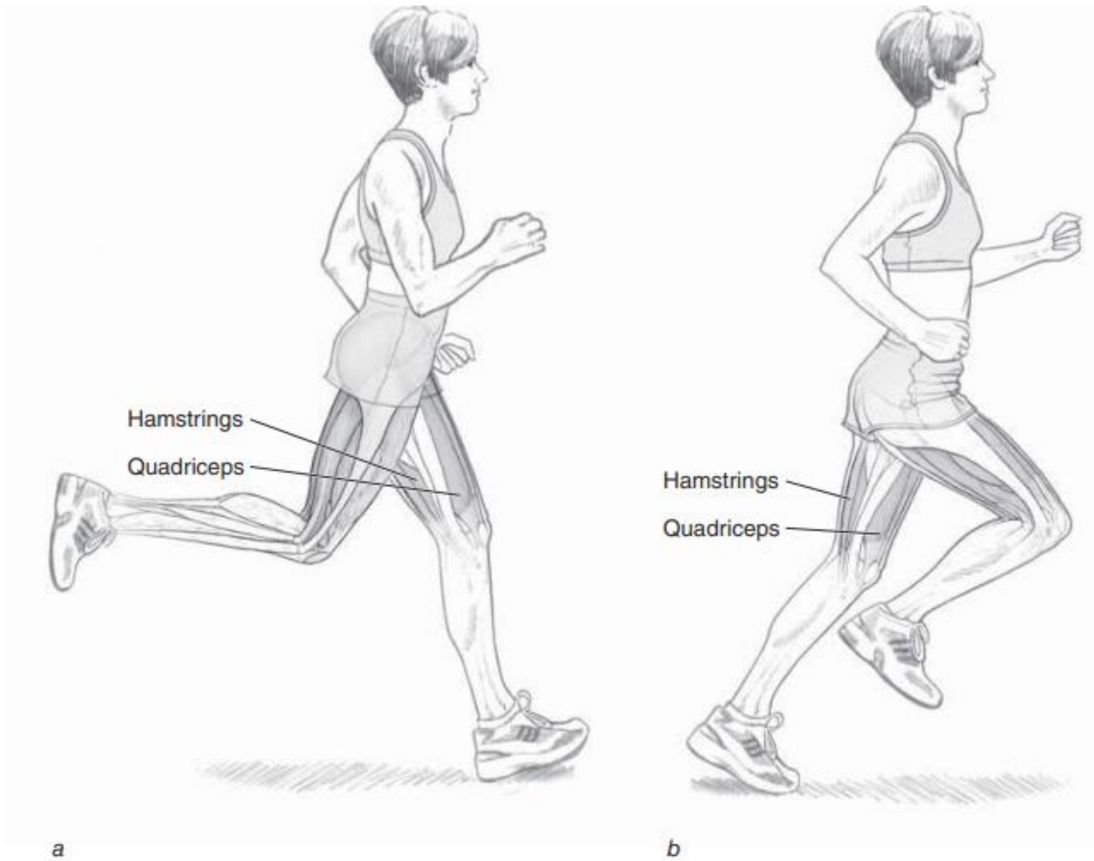


Figure 4.6 Muscle activity *(a)* immediately after heel strike and *(b)* near the end of the stance phase. At these two points the knee is near full extension, and the hamstring muscles must act as stabilizers.

Thus, considering the knee undergoes approximately 40° of sagittal plane flexion, as opposed to 3° of frontal plane and 9° of transverse plane motion, and based on the dominance of the vasti muscles in supporting and controlling the knee joint, adequate strength and function of these muscles is critical for typical knee mechanics. Reduced force output from the vasti muscles because of muscle weakness results in increased knee flexion during the stance phase of gait since these muscles are unable to eccentrically control knee flexion. Arnold et al. (2005) conducted a study utilizing a 3D, muscle-actuated dynamic simulation of gait to understand the relative contributions of hip and knee muscles during normal gait. Overall, the vasti muscles were reported to play a large role in eccentrically controlling knee flexion during stance. Since anterior tibial shear is reduced as the knee flexes, the increase in knee flexion causes a decrease in shear forces and a reduced demand on the hamstring muscles (Osternig et al. 2000; Ferber et al. 2002a; Ferber et al. 2002b; Ferber et al. 2003). On the other hand, reduced hamstring strength, and a concomitant inability to control knee extension at or near heel strike, results in a more flexed knee at heel strike and throughout stance, concomitant reductions in anterior tibial shear, and thus greater demand on the quadriceps muscles to eccentrically control the knee.

Anatomical Alignment

It has been postulated that differences in knee anatomical structure may predispose runners to differences in running mechanics, which over many repetitions may lead to certain injuries. Specifically, a larger Q-angle (figure 4.7) has been reported to be associated with an increase in lateral patellar contact forces. Therefore, an increased Q-angle is thought to lead to a more pronounced genu valgum position during the stance phase of gait and plays a partial role in the etiology of knee-related injuries (DeHaven and Lintner 1986; Messier et al. 1991; Almeida et al. 1999). However, there is very little evidence within the scientific literature regarding the interrelationship between Q-angle and lower-extremity biomechanics.

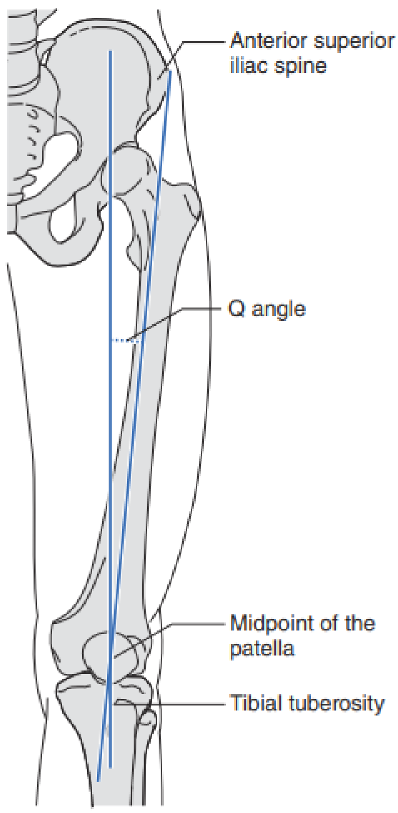


Figure 4.7 Measurement of the Q-angle.

It has been theorized that Q-angle measures greater than 15° should be considered pathological and be associated with atypical gait mechanics (Caylor et al. 1993; Byl et al. 2000; Heiderscheit et al. 2000). Unfortunately, there is no research data to support this theory. Kernozek and Greer (1993) conducted a study to understand the relationship between Q-angle and rearfoot motion in walking. They conducted a 2D video analysis on 20 women walking on a treadmill and measured static and dynamic Q-angle measures. Overall, the authors reported poor to very poor correlations among the variables and concluded that Q-angle had little to do with predicting or influencing rearfoot motion during gait.

Heiderscheit et al. (1999) assessed the influence of Q-angle on the 3D rearfoot, tibia, and knee joint coupling and coupling variability during gait. Thirty-two subjects had clinical measures of static Q-angle and ran over-ground. Using a unique continuous relative phase (CRP) approach to measure segment couplings to assess between-trial consistency for the stance phase, they reported no differences in CRP variability among subjects with varying Q-angles. In a followup study, Heiderscheit et al. (2000) reanalyzed the same data for differences in specific joint angles. Specifically, they hypothesized that an individual with a Q-angle greater than 15° would exhibit increased rearfoot eversion and tibial internal rotation as a result of a greater knee genu valgum angle as compared with those with a Q-angle less than 15°. However, Q-angle measures did not increase the maximum segment or joint angles during running, but the highQ-angle group demonstrated an increased time to maximum tibial internal rotation. Thus, based on these studies investigating gait biomechanics and the interrelationship to static measures of Q-angle, the conclusion has been that no relationship exists. However, one significant reason for the findings may be the lack of evidence to define a high or low Q-angle.

As previously mentioned, there is no evidence to support the theory that a Q-angle measure greater than 15° is pathological and increases injury risk. In a review article, Livingston (1998) also stated that a high-risk Q-angle measure of 15° to 20° was not based on any scientific literature and was more speculative than evidence based. Subsequently, one study sought to establish a normative range for Q-angle in an asymptomatic population, and adjusted these normative values in accordance with individual-specific mediolateral patellar displacement (Herrington and Nester 2004). Adjusting for patellar displacement is important because an accurate Q-angle measurement depends on a patella centralized within the femoral trochlear groove. If the patella is laterally displaced, the Q-angle measured will be falsely low, and if it is medially displaced, the angle will be falsely high (figure 4.8). These authors reported Q-angle values between 11° and 14° (±5o) for both the left and right knees and for male and female subjects. When correcting for a laterally displaced patella, present in 68 of the 109 subjects tested, a reduction in Q-angle close to 1° was found, but no change was seen when correcting for a medially displaced patella, which was present in 28 subjects.

Moreover, in Livingston's review article (1998), minimum Q-angle values from a variety of peer-reviewed manuscripts for healthy females ranged from 2.5° to 10°, and maximum values ranged from 15° to 26°. As well, while it is commonly assumed that women tend to have larger Q-angles than men, minimum values for healthy males ranged from 0° to 8°, and maximum values ranged from 15° to 27°. Other authors have reported a variety of Q-angle measures ranging from 8° to 14° and 11° to 20° based on measures from 50 men and women, respectively (Horton and Hall 1989), and another study reported ranges of 5° to 16° and 6° to 17° based on measures from 50 men and women, respectively (Livingston and Mandigo 1997). Woodland and Francis (1992) measured the Q-angles of 269 men and 257 women and reported mean values of 13° and 17°, respectively. Thus, when considering Q-angle, values in excess of 15° would fall within normal limits, and the data do not support an across-the-board difference in Q-angle between men and women.

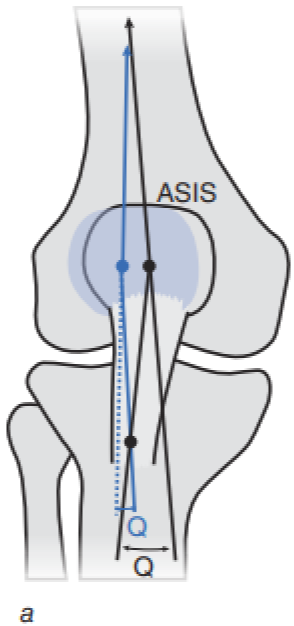


Figure 4.8 *(a)* Medially and *(b)* laterally displaced patella and change in Q-angle.

Based on data from L. Herrington and C. Nester, 2004, "Q-angle undervalued? The relationship between Q-angle and medio-lateral position of the patella," *Clinical Biomechanics* 19(10): 1070-1073.

Limited evidence suggests that a very large Q-angle measure (greater than 20°) may increase the risk of injury. Rauh et al. (2007) measured the Q-angle for 393 high school cross-country runners and followed them during a crosscountry season to track lower extremity injuries from practices or competitions. They concluded that runners with a Q-angle greater than 20° were at 1.7 times greater risk of injury compared with runners whose Q-angle was between 10° and 15°. Moreover, runners with more than 4° side-to-side difference in Q-angle measures were 1.8 times more likely to sustain an injury than runners with a smaller side-to-side difference. In contrast, another prospective study (Raissi et al. 2009) followed 66 athletes for 17 weeks and reported that Q-angle was not a predictive factor, nor did it increase the relative risk for incurring a running-related injury. However, the measures of Q-angle only ranged between 11° and 13° for both injured and noninjured runners.

In summary, a Q-angle less than 20° should be considered typical based on the aforementioned studies. The long-standing theory of a 15° Q-angle being related to injury is not supported by the literature, and practitioners must take care when attributing a running-related injury solely to a nonmodifiable factor such as Q-angle. Other factors, such as muscle flexibility should also be incorporated into the assessment to better understand atypical mechanical loading to the knee joint.

Flexibility

While several other muscles and tissues, including the IT band, adductor magnus, and sartorius, cross the knee joint, the paucity of research involving these tissues does not allow us to discuss their potential contributions to typical or atypical knee joint gait mechanics. Previous research has focused primarily on hamstring and quadriceps muscle flexibility in attempts to understand injury etiology.

Flexibility of the hamstring muscles is a critical component to injury prevention. Chumanov et al. (2011) investigated whether the hamstrings are susceptible to injury during the swing phase, when the hamstrings are eccentrically lengthening, or during stance, when knee stabilization and resistance to anterior tibial shear is needed. Using a modeling technique, they measured hamstring lengthening from mid to late swing and shortening under load throughout stance. The authors concluded that the large inertial loads produced from knee extension during the swing phase of gait make the hamstrings most susceptible to injury. Moreover, lateral hamstring (biceps femoris) loading increased significantly with speed and loading was greater during swing than stance. Theoretically, reduced hamstring muscle flexibility and tissue length result in a shortened stride length, increase stride frequency, and reduced knee flexion position at heel strike similar to reduced hamstring muscle strength. The key factor then is to establish the normative range for hamstring tissue flexibility.

Youdas et al. (2005) examined the factors of gender and age over 10-year increments on hamstring muscle length via measurements of passive straight-leg raise and popliteal angle. Overall, females exhibited greater overall hamstring flexibility for both measures (8° greater and 11° greater, respectively) but no differences were measured across age. The mean hamstring muscle length for passive straight-leg raise was 69° (± 7°) for men and 76° (± 10°) for women. When measuring hamstring flexibility via popliteal angle, the authors reported values of 141° (± 8°) for men and 152° (± 11°) for women. Using these hamstring flexibility values, practitioners and runners can determine if they are at risk for atypical knee joint loads or atypical running biomechanics.

An interesting study by Silder et al. (2010) investigated whether residual scar tissue at the hamstring musculotendon junction after a previous injury would influence strength, neuromotor activation patterns, and joint kinematics. While magnetic resonance (MR) imaging revealed significantly enlarged proximal biceps femoris tendon volume for the injured hamstring muscle, no significant between-limb differences were found for any other variables. The authors concluded that previous hamstring injuries and residual scar tissue did not negatively influence function or gait mechanics. However, it is unknown whether the hamstring muscles exhibited reduced flexibility concomitant with the enlarged tendon volume.

Few investigations have been conducted to evaluate quadriceps muscle flexibility as a risk factor for injury. A review article from van der Worp et al. (2011) suggested only poor to moderate evidence for nine factors related to patellar tendinopathy, including reduced quadriceps flexibility. However, the authors stressed a clear need for high-quality studies. Considering the knee undergoes approximately 40° of knee flexion during the stance phase of gait, and the vasti muscles produce the majority of support to the knee joint, adequate flexibility is critical to allow for normal range of knee motion. If the vasti muscles are inflexible, there would be a reduced amount of knee flexion during stance in contrast to the changes due to reduced vasti strength. Unfortunately, even fewer studies have established normative ranges for quadriceps muscle flexibility, but Harvey (1998) measured quadriceps flexibility via knee flexion angle during the Thomas test and reported a mean angle of 53°.

In summary, while muscle flexibility has been shown to play a minor role in injury etiology and prevention as the primary causative factor, there is still some research needed to help identify when the degrees of hamstring and quadriceps muscle flexibility are within normal limits and not a factor in injury prevention and rehabilitation. Considering that limited hamstring flexibility has been linked to a shortened stride length, optimal flexibility is important from a running performance standpoint.

Assessing Hip Mechanics

Biomechanics

For the first half of stance, while the knee joint undergoes flexion, internal rotation, and abduction, the hip functions in a similar manner (figures 5.1-5.3). Specifically, at heel strike the hip is approximately 30° flexed and remains in this position for about the first 30% to 40% of stance. Thus, it is flexion at the knee joint that primarily brings the center of mass downward during this period of time. For the remaining 60% to 70% of stance, the hip undergoes extension and is nearly fully extended at toe-off (figure 5.1).

Throughout the beginning of stance, the hip also adducts and internally rotates (figure 5.2). However, since the swing leg is bringing the pelvis anteriorly and the transverse plane motion of the pelvis is opposite (external rotation) to the femur, there is little internal rotation (2°-3°) at the hip joint as compared with the 9° rotation at the knee joint. Thus, overall transverse plane motion at the hip is primarily composed of external rotation, which begins at approximately 20% of stance. At toe-off the hip is in an externally rotated position as a result of pelvic transverse plane motion (figure 5.3). For the second half of stance, the hip abducts in a manner similar to the knee but greater in overall magnitude. Thus, just as the coupling and timing relationship between the knee and rearfoot was asynchronous, the timing between the hip and knee joints is asynchronous as well.

Several authors have also reported these asynchronous patterns and out-of phase relationships during the stance phase of gait for hip internal rotation and tibial internal rotation (Hamill et al. 1999; Dierks and Davis 2007). Unlike the relationship between rearfoot eversion and tibial internal rotation, both the hip's and knee's transverse plane relationships remain uncoupled throughout stance, and studies have reported this relationship to be largely nonsystematic across subjects. These results are surprising because these motions are thought to occur synchronously. It has been theorized that the out-of-phase nature of these relationships may be a function of the velocities of these motions, with the tibia and knee rotating more quickly than the hip. With this lack of movement synchrony and timing, it is no surprise that the knee joint is most susceptible to injury. There is a large amount of coupling variability between the knee and ankle joints, and there is relatively no synchronous coupling between the knee and hip joints. Considering that the tibia and knee rotate much more quickly and to a larger degree than the hip joint, the knee is much more susceptible to injury even under typical biomechanical circumstances.

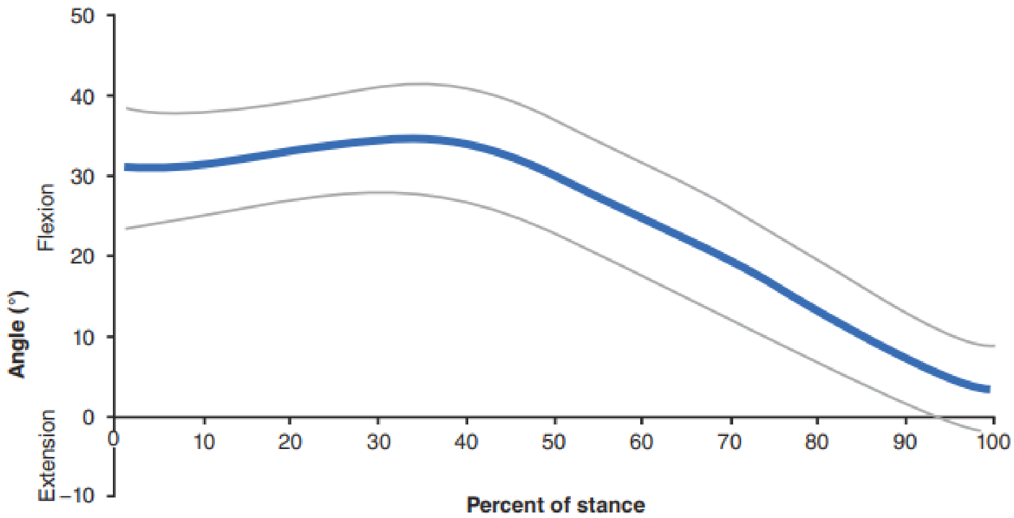


Figure 5.1 Biomechanical motion of hip sagittal plane motion during the stance phase of running gait. A stance of 0% indicates heel strike, and a stance of 100% indicates toe-off.
Note: Blue line = mean; gray lines = ± 1 SD.

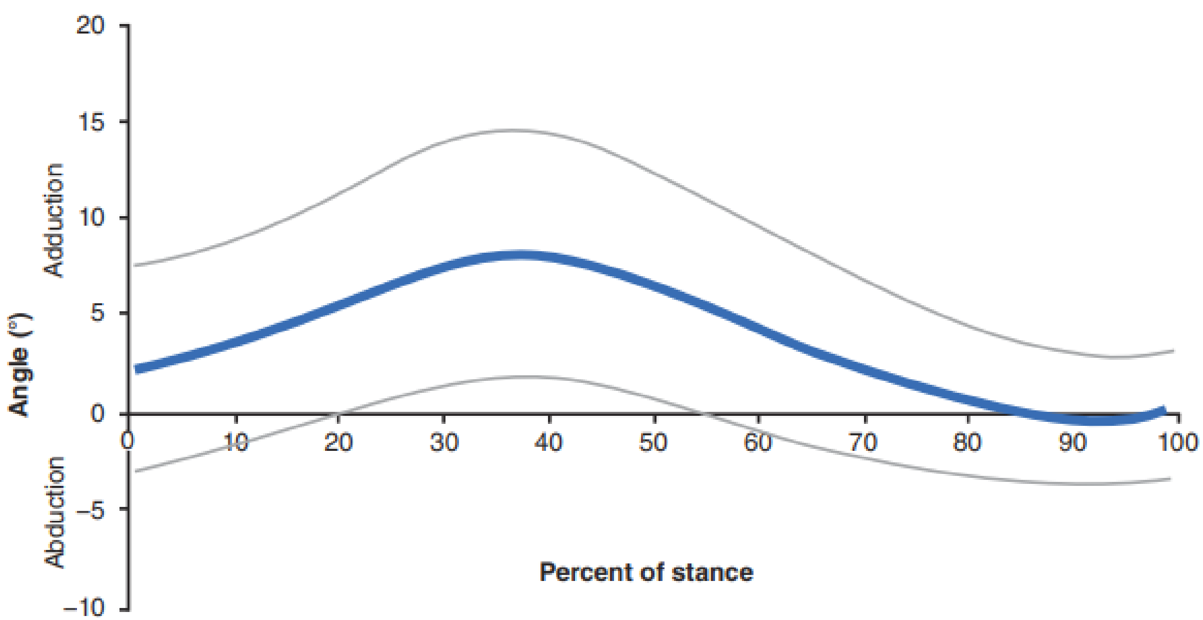


Figure 5.2 Biomechanical motion of hip frontal plane motion during the stance phase of running gait. 0% of stance indicates heel strike, and 100% of stance indicates toe-off.
Note: Blue line = mean; gray lines = ± 1 SD.

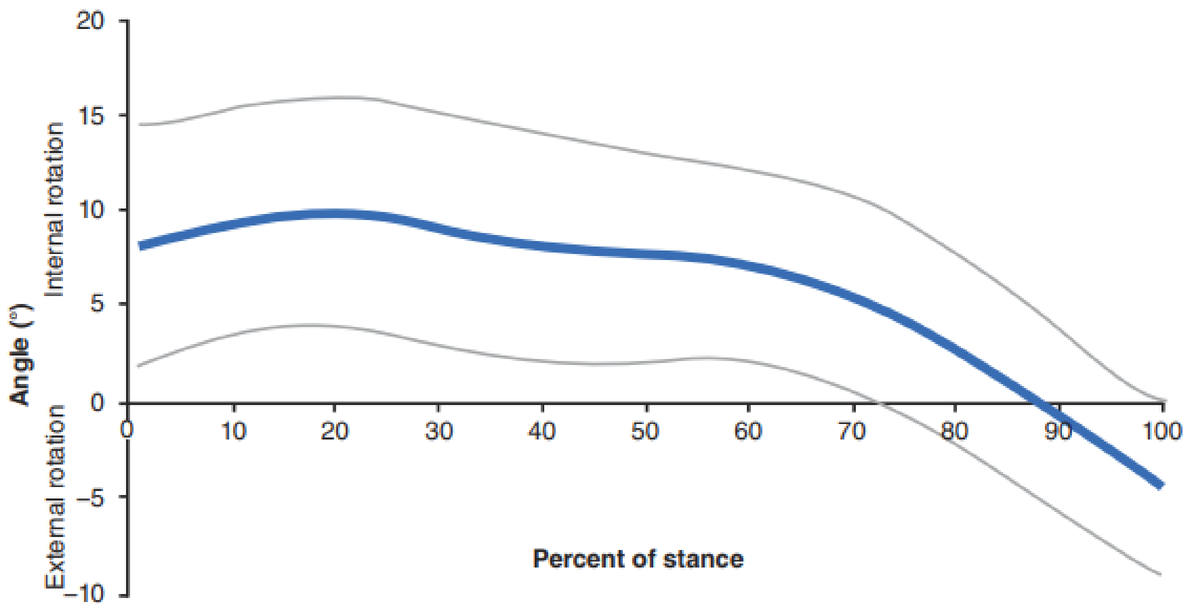


Figure 5.3 Biomechanical motion of hip transverse plane motion during the stance phase of running gait. 0% of stance indicates heel strike and 100% of stance indicates toe-off.
Note: Blue line = mean; gray lines = ± 1 SD.

Strength

As previously mentioned, from foot flat to 20% of stance, the vasti muscles along with the gluteus maximus and medius produce the majority of support and prevent the knee and hip from collapsing against the downward pull of gravity and during knee flexion. Since the hip remains in approximately 30° of flexion for this period of time, the gluteal muscles are contracting isometrically to maintain this static hip flexion position (figure 5.4a). During midstance and with significant passive resistance from noncontractile tissues such as the hip joint capsule and ligaments, the gluteus medius and minimus provide nearly all of the support to accelerate the center of mass upward as the hip joint begins to extend. Thus, gluteus maximus begins to contract concentrically, along with assistance from the hamstring muscles, to produce the overall hip extension motion (figure5.4b) (Anderson and Pandy 2003; Pandy and Andriacchi 2010). In fact, based on per unit of force, the gluteus maximus has greater potential compared with the vasti muscles to control knee flexion, which is discussed later in this chapter.

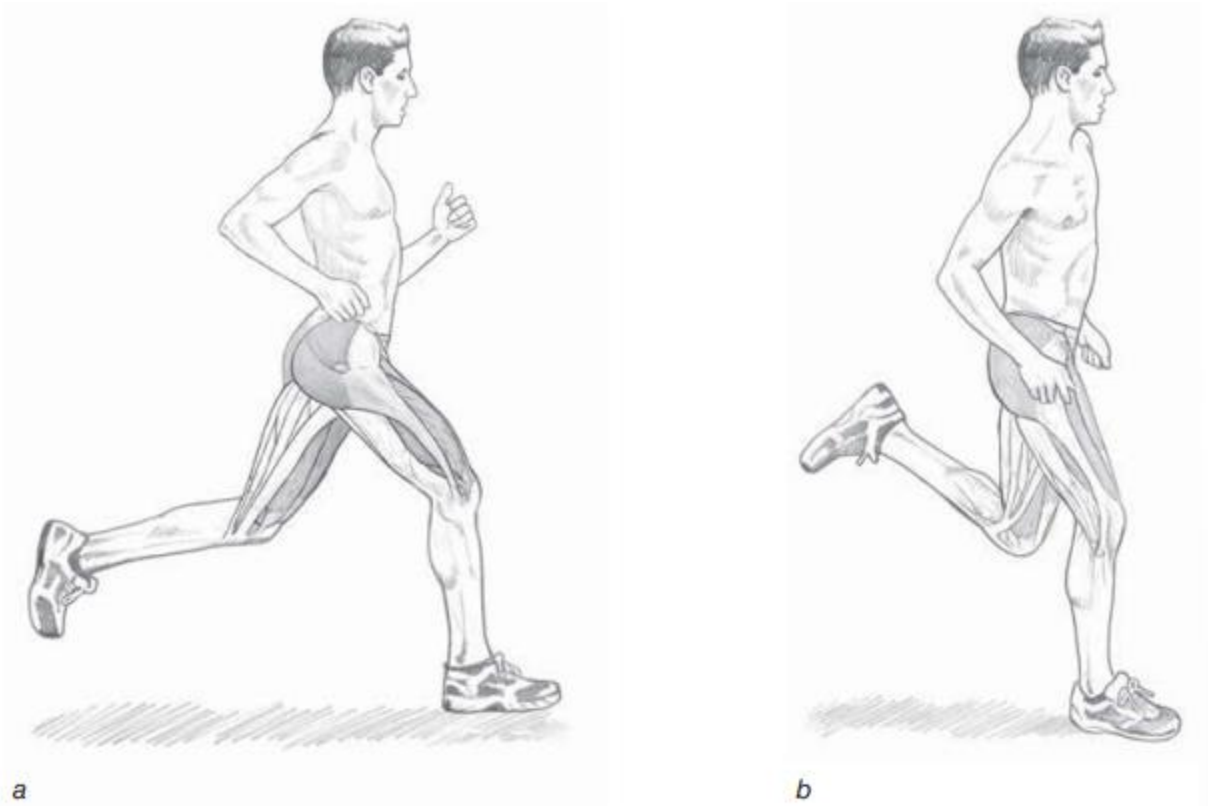


Figure 5.4 Gluteal and hamstring activity during *(a)* the early stance phase and *(b)* midstance.

With respect to frontal plane motion, Pandy et al. (2010) reported that the body's center of mass accelerates outward and downward during the first half of stance. The vasti muscles and the gluteus maximus serve to support the downward acceleration and prevent collapse. Interestingly, only the gluteus medius muscle functions to accelerate the body medially (inward) and thus maintains overall mediolateral balance (figure 5.5). Based on these data, several authors have hypothesized that a primary contributing factor to running-related injuries is weakness of the gluteus medius musculature and the subsequent alterations in frontal plane hip and knee motion (Ireland et al. 2003; Mascal et al. 2003; Cichanowski et al. 2007; Robinson and Nee 2007; Bolgla et al. 2008; Dierks et al. 2008; Willson and Davis 2008).

The gluteus medius has been theorized to eccentrically control hip adduction, and thus knee genu valgum angle, during the stance phase of gait (Ferber et al. 2003; Ireland et al. 2003; Mascal et al. 2003; Powers 2003; Cichanowski et al. 2007; Robinson and Nee 2007; Bolgla et al. 2008; Dierks et al. 2008; Willson and Davis 2008). Unfortunately, few studies have examined the relationship between gluteus medius muscle strength and hip and knee mechanics. Bolgla et al. (2008) measured gluteus medius strength and knee and hip kinematics and reported that subjects with patellofemoral pain syndrome (PFPS) exhibited reduced muscle force output but no differences in knee genu valgum angle during a stair descent compared with controls.

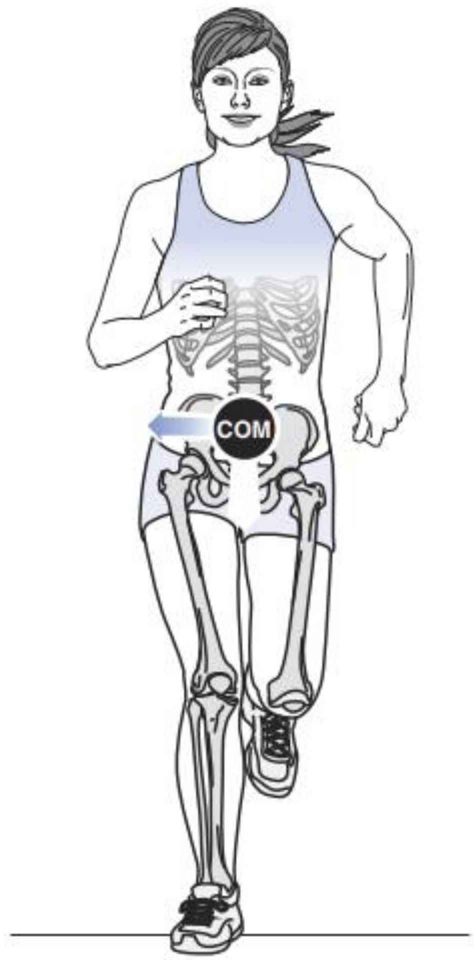


Figure 5.5 During the first half of stance, the body's center of mass is accelerating outward and downward, which requires the hip muscles to act as stabilizers to maintain balance.

Similarly, Dierks et al. (2008) measured reduced gluteus medius isometric force output for subjects with PFPS and control subjects after an exhaustive run. However, in contrast to Bolgla et al. (2008), the PFPS patients in this fatigue study exhibited an increase in peak hip adduction over the course of the run compared to the healthy runners. These data suggest that reduced force output from the gluteus medius will result in an increased hip adduction angle and thus a greater knee genu valgum (abduction) angle during gait. However, Snyder et al. (2009) reported that after a 6-week hip strengthening protocol, healthy female runners exhibited a 13% gain in abductor strength, but the hip adduction angle during running increased by 1.4°, contrary to their hypotheses and the aforementioned studies. So it is unclear exactly how fatigue or reduced force output from this muscle group can influence hip and knee biomechanics during gait.

We sought to test the hypothesis of whether improvements in muscle strength would lead to a reduced peak knee genu valgum angle for runners with PFPS. We conducted an experiment involving

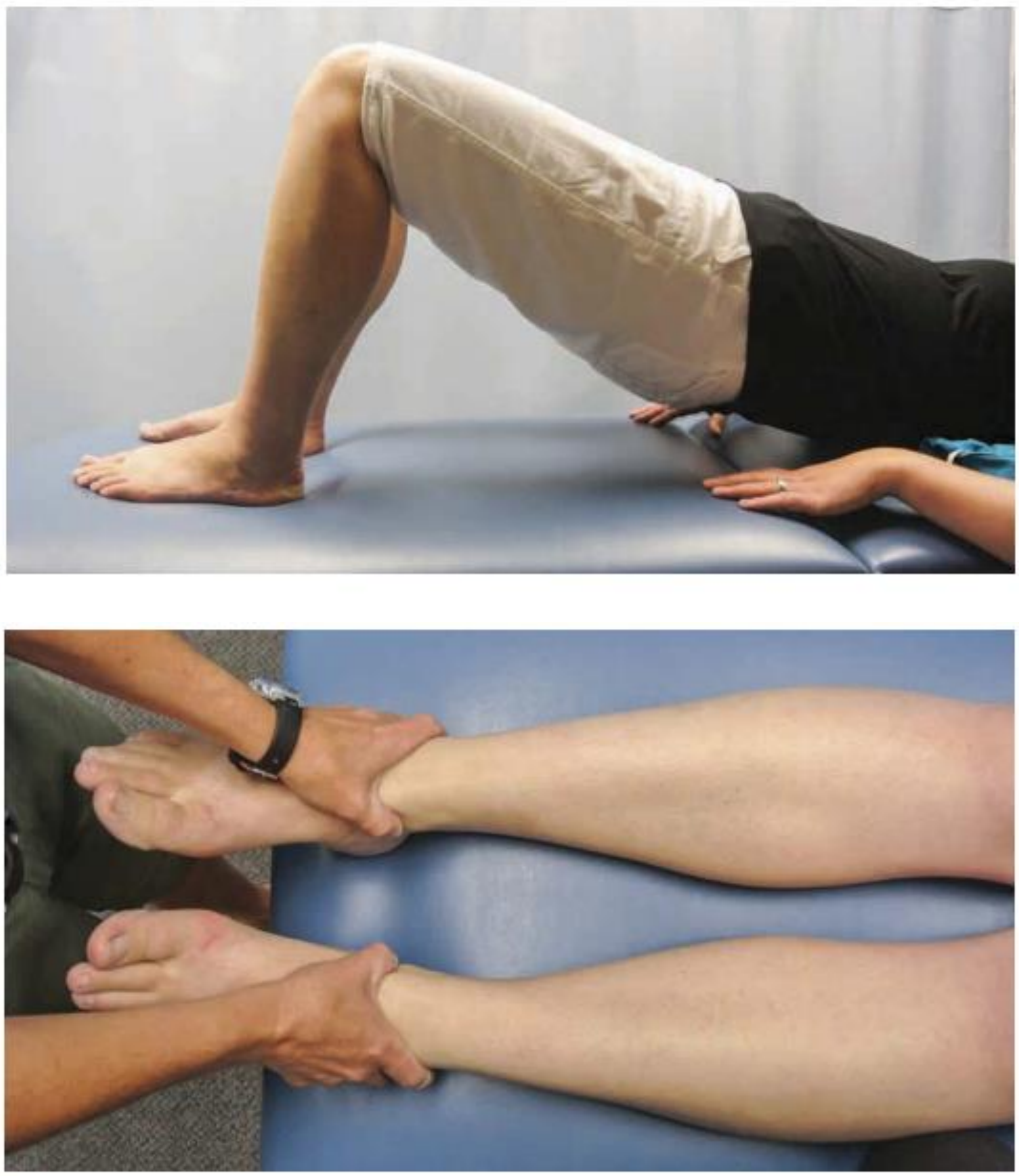
15 runners with PFPS and 10 healthy runners (Ferber, Kendall, et al. 2011). All subjects were involved in a 3-week hip abductor muscle-strengthening protocol composed of two exercises. We found that over a 3-week protocol, the level of pain experienced by the PFPS patients decreased on average 40% and there was a 33% improvement in strength, but no changes in the peak knee abduction angle occurred. However, we took a novel approach to understanding potential changes in gait mechanics and measured the stride-to-stride kinematic pattern. We found a more consistent frontal plane knee movement pattern concomitant with the gains in strength. We theorized that by providing the knee with a more consistent stride-to-stride movement pattern, a more optimal environment is established to allow for tissue healing and pain resolution. We also repeated this study with another group of 20 PFPS runners (Ferber, Bolga, et al. 2011), using the same rehabilitation exercises used by Snyder et al. (2009). Similar to our first study, we found that baseline movement variability was higher for the PFPS involved knee and hip joints compared to controls. Also contrary to the hypotheses, the PFPS affected leg demonstrated reduced knee and hip movement variability after a 6-week strengthening intervention. We concluded that stride-to-stride knee joint variability may be a better indicator of atypical gait patterns compared with peak angles, and it highlighted the importance of understanding the interrelationship between hip abductor muscle strength and biomechanical assessment.

Anatomical Alignment

We do not measure aspects of hip or pelvis anatomical structure within our clinical and biomechanical assessment. The primary reason for this decision is that there are very few scientific investigations related to the reliability or validity of hip and pelvic alignment measures. However, we do measure leg length discrepancy (LLD) based on the scientific literature and its relationship to biomechanics and joint injury.

Leg length discrepancy is assessed with subjects positioned supine after the Weber-Barstow maneuver, (figure 5.6) is used to set the pelvis (McGee 2007). See figure 5.6. Overall, it is accepted that a LLD is present when more than a 2 cm side-to-side difference is radiographically confirmed (Perttunen et al. 2004). However, this discrepancy has not been well defined, and the literature does not support this value through comprehensive research studies. We conducted a study to provide a comprehensive database from a large population of runners with selected anatomical alignment measures commonly associated with running injuries (Kendall et al. 2008). We measured 221consecutive patients that presented to the clinic for various musculoskeletal running injuries. The average LLD was 0.45 cm (± 0.87 cm), and only 3.17% of patients exhibited a LLD greater than 1.5 cm. Thus, we concluded that LLD, unless radiographically confirmed, is rare within an active, injured population.

LLD may influence overall gait mechanics. Unfortunately, very few studies have been conducted to confirm or refute this supposition. Perttunen et al. (2004) measured plantar pressures and 2D ground reaction forces on 25 patients with confirmed limb length discrepancy of more than 2 cm. Overall, these authors reported reduced stance time in the short leg and greater peak plantar pressure under the first ray in the long leg. They concluded that loading of the long limb is greater and the forefoot of the long limb experiences greater loading than the short limb. Finally, White et al. (2004) also reported that the shorter limb sustains a greater proportion of load and loading rates compared with the longer limb. However, some anatomic LLDs investigated in this study were between 1-1.5 cm, which are within normal limits for the vast majority of runners, and therefore may not be clinically relevant.



a

b

Figure 5.6 *(a)* the Weber-Barstow maneuver and *(b)* assessment of leg length discrepancy.

Flexibility

The key flexibility measures for the hip are the internal rotation range of motion and tissue flexibility of the rectus femoris, iliopsoas, and iliotibial (IT) band. Biomechanically, the hip internally rotates during early stance and therefore requires sufficient hip rotation range of motion to allow this motion to occur. If the hip cannot adequately internally rotate, increased torsional stress is created within the hip, knee, and ankle joints. To minimize this torsional stress, the hip could return to an externally rotated position, but the relative external rotation of the pelvis, as the swing leg is moving forward and driving the pelvis into an externally rotated position, prevents the unloading, or external rotation, of the hip joint. Thus, sufficient hip internal range of motion flexibility is critical for the reduction in torsional stress.

The rationale behind increased torsional stress being a predictor of injury was first proposed by Holden and Cavanagh (1991). These authors calculated the free moment, the rotational force the foot applies to the ground during the stance phase of gait. Specifically, they reported that the free moment was greatest during the first half of stance and acted to resist foot abduction, a component of overall foot pronation. Additionally, they reported that with greater amounts of rearfoot eversion, the free moment increased. In chapter 6 we discuss the free moment in greater detail.

Unfortunately, few studies have investigated the role of hip rotator strengthening for the treatment of musculoskeletal injuries, and none have determined whether improvements in rotator range of motion influences torsional forces. However, Cibulka and Threlkeld-Watkins (2005) reported a case study involving a 15-year-old who had been experiencing anterior right knee pain for 8 months. Clinical investigation revealed reduced internal rotation flexibility of the right hip and external rotator strength compared to the left, and a rehabilitation program was designed to improve these factors. After 6 visits, within 14 days, the patient's side-to-side asymmetry was negligible, and the pain was gone. While this study is the only one involving a focused, hip rotator rehabilitation program, and the limitations of being a case study are evident, it does provide evidence of the role of improved rotator flexibility for the treatment of knee pain.

On the other hand, increased or excessive hip internal rotation range of motion must be matched with appropriate rotator muscle strength. If the excessive hip rotation range of motion is measured, increased mechanical internal rotation can occur during the stance phase of gait, resulting in increased torsional stress at the hip and especially knee joint. Souza and Powers (2009) reported that 19 female athletes with PFPS exhibited significantly greater than average hip internal rotation range of motion and reduced isotonic hip muscle strength in 8 of 10 hip strength measurements as compared with controls. The authors suggested that excessive internal rotation range of motion must be matched with improvements in overall hip muscle strength to minimize atypical biomechanical motion.

With respect to rectus femoris and IT band flexibility testing, the special tests used by clinicians for assessing these tissues are highly subjective and involve either a positive or negative assessment, making it difficult to apply within evidence-based medicine. While some studies have used either goniometers or inclinometers to quantify the modified Thomas test or Obers test, very few have established normative values.

For rectus femoris and iliopsoas, Corkery et al. (2007) reported on values for various muscle lengths for 72 college-age students using a goniometer. They reported on the modified Thomas test for iliopsoas and the Thomas test for rectus femoris. The modified Thomas test methodology Corkery used involved having the subject lie supine with the body completely on a table that prevented the thigh from dropping below the table (figure 5.7), and thus an average angle of 2.3° (± 1.8°) above the horizontal was reported for the iliopsoas. The Thomas test involved sitting on the edge of the table and allowing the thigh to drop (figure 5.8). They reported an average angle of 52.8° (± 10.5°) for the rectus femoris. Harvey (1998) assessed the flexibility of the iliopsoas using a goniometer for 117 elite athletes and allowed the thigh to drop below the horizontal. They reported an average angle of 11.91° below the horizontal.

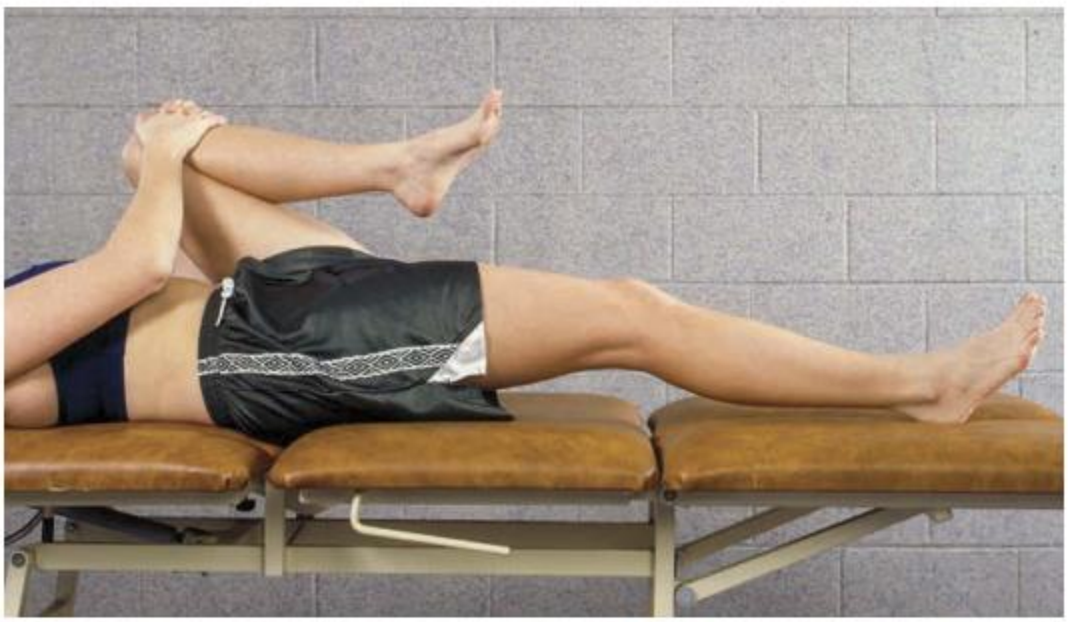


Figure 5.7 Modified Thomas test.

Figure 5.8 Thomas test.

We conducted a study (Ferber et al. 2010) to compare the subjective evaluation of iliopsoas flexibility to the instrumented measurement of a digital inclinometer in the hopes of establishing normative values and providing a critical criterion. Three hundred recreational athletes were classified subjectively as either positive or negative for IT band and iliopsoas tightness using the Ober's and Thomas test, respectively. A digital inclinometer measured thigh position relative to the horizontal to the nearest tenth of a degree. For iliopsoas flexibility, the average inclinometer angle was consistent with Harvey (1998) and was $-10.60 \pm 9.61°$. Interestingly, 208 limbs were subjectively assessed as positive ($0.34 \pm 7.00°$) and 392 limbs were assessed as negative ($-15.51 \pm 5.82°$) and the between-clinician agreement was 95.0%. Thus, a Thomas test measuring less than 10° below the horizontal should be considered atypical and related to reduced iliopsoas tissue flexibility. A modified Thomas test greater than 2.3° above the horizontal should be considered atypical and related to reduced rectus femoris tissue flexibility.

It has been hypothesized that reduced iliopsoas tissue flexibility would result in reduced peak hip extension, stride length, and gait speed during gait (Watt et al. 2011). A study by Kerrigan et al. (2003) performed a double-blinded, randomized, controlled trial wherein 96 healthy elderly individuals were allocated into a treatment and control group. The treatment group received a one-time instruction in hip flexor stretching, and the control group received a one-time instruction in shoulder abductor stretching. After a 1.6° (± 3.0°) improvement in hip flexor flexibility, the treatment group demonstrated a 2°, but statistically nonsignificant, increase in peak hip extension during walking. Interestingly, a significant improvement in peak ankle plantar flexion angle was measured and attributed to improvements in hip flexor contracture rather than changes at the ankle. Watt et al. (2011) conducted another double-blinded, randomized, controlled trial for 74 frail older adults over a 10-week program of twice-daily hip flexor stretching. Similar to Kerrigan et al. (2003) the authors reported significant improvements in passive iliopsoas flexibility and significant increases in gait velocity and stride length during walking, but there were no significant changes in peak hip extension. Based on these intervention studies, improvements in hip flexor flexibility do not seem to coincide with significant improvements in peak hip extension. Future studies that involve a more active population and take into consideration trunk mechanics and flexibility of other muscles are necessary.

Very few studies have been conducted to establish a normative range for IT band tissue flexibility using the Obers test (figure 5.9). In the aforementioned study (Ferber et al. 2010) we also measured the modified Obers test using a digital inclinometer to establish IT band flexibility. We reported an average angle of −24.59° for all 300 participants regardless of whether they were subjectively deemed to exhibit a positive or negative position. Of the 600 limbs of interest, 168 were subjectively assessed as positive while 432 were assessed as negative. The critical criterion for the Obers test was −23.16°. Hudson and Darthuy (2009) also measured IT band tightness in a group of 12 control subjects and 12 subjects with PFPS using a bubble inclinometer. The authors reported a range of −20.3° to −21.4° for the controls and −14.9° to −17.3° for the PFPS group, which are similar to the results of our study. Thus, an Obers test measuring less than 20° below the horizontal should be considered atypical and related to reduced IT band tissue flexibility.

It has been theorized that the primary functions of the IT band are to serve as a lateral hip and knee stabilizer and to resist hip adduction and knee internal rotation (Fredericson et al. 2000). The iliotibial band originates from the fibers of the gluteus maximus, gluteus medius, and tensor fasciae latae muscles and attaches proximal to the knee joint into the lateral femoral condyle and distal to the knee joint into the intercondylar tubercle of the tibia (Birnbaum et al. 2004). As a result of the femoral and tibial attachments, it is possible that abnormal hip as well as foot mechanics, which both influence the knee, could play a role in the development of IT band syndrome or IT band tightness.

Because the iliotibial band attaches to the lateral condyle of the tibia, it is postulated that excessive rearfoot eversion resulting in greater tibial internal rotation could increase the strain in the iliotibial band. Miller et al. (2007) reported that at the end of an exhaustive run, runners with IT band syndrome demonstrated a greater rearfoot inversion angle at heel strike compared with controls, which the researchers hypothesized contributed to a greater peak knee and tibial internal rotation velocity and thus increased torsional strain to the IT band. In contrast, Messier et al. (1995) reported that runners with a history of IT band syndrome exhibited no difference in rearfoot mechanics compared with healthy runners. Moreover, since the gluteus medius muscle is the primary abductor of the hip joint, weakness of this muscle may lead to an increased hip adduction angle, thereby potentially increasing the strain on the IT band. Fredericson et al. (2000) reported that runners with IT band syndrome had significantly reduced hip abductor muscle strength in the affected limb compared with the unaffected limb, as well as compared with healthy controls.

However, very few studies have investigated whether atypical hip mechanics may play a role in the etiology of ITBS.

A prospective study by Noehren et al. (2007) and a retrospective study by Ferber et al. 2010 examined proximal (hip), distal (rearfoot), as well as local (knee) mechanics in the development of IT band syndrome. For both studies, the IT band syndrome group exhibited significantly greater peak knee internal rotation angle and peak hip adduction angle compared to the healthy runners, but no significant differences in peak rearfoot eversion angle or peak knee flexion angle were observed between groups. The common results between the prospective study and the retrospective study provide strong evidence related to atypical running mechanics and the etiology of ITBS.

a

b

Figure 5.9 Obers test *(a)* start position and *(b)* lowering the leg to test IT band flexibility.

Summary

There are still many unanswered questions about running biomechanics and kinetic-chain interrelationships that have not been answered in the scientific literature. One excellent example is the tibial and femoral torsion caused by some runners' anatomical structure and this pattern's relationship to hip and knee rotation. The central hypothesis is that excessive tibial torsion, or a more externally rotated tibia, is related to decreased knee internal rotation and a more toed-out foot progression angle when running, which can lead to increased torsional stress. In accordance, we hypothesize that excessive femoral torsion, or a more internally rotated femur often called anteversion, is related to decreased hip internal rotation.

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