The ultimate goal of gait analysis is to understand the relationship between an individual’s functional capabilities and limitations and gait pattern, with the purpose of enhancing performance while preventing injury. The physiologic aspects of training the running athlete are well-known. The biomechanical variables associated with training are less understood by clinicians. This article reviews the components of gait and helps clinicians apply these concepts to clinical analysis.

Human locomotion involves moving a center of mass (COM) across a given distance. Gait can be monitored subjectively in the clinic, or can be quantified objectively in a modern three-dimensional (3D) gait laboratory setting. The selected method must be able to assess the desired parameters. The relatively young science of biomechanical gait analysis, combined with the inherent variability of individual runners, makes the comprehensive study of running gait challenging. A clinical gait evaluation, however, used in conjunction with a thorough history, physical examination, and functional screen is a powerful tool for shedding light on the dysfunction causing an individual’s symptoms.

THE GAIT CYCLE

The gait cycle describes the time and space parameters that occur in the distinctly different activities of walking and running. The lower body limbs experience both stance and swing periods, but the timing and contact patterns of these stance and swing phases differentiate the two tasks of walking and running. Stance phase begins at contact with the ground. Swing phase initiates as that limb moves into toe-off. The walking gait cycle has a period of double support (both feet in contact with the ground at the same time) and a period of single-leg support. An individual is able to walk at varied speeds while still maintaining these essential characteristics, and as evidenced by the rapid pace of elite race-walkers. Gait is described as running when the gait cycle exhibits single-leg support and double-leg float (flight) periods. Therefore,
walkers always have at least one limb in contact with the ground, and runners either have one limb or no limbs in contact with the ground at respective periods of gait.

Gait cycles can be categorized. The walking gait is subdivided into eight distinct phases: initial contact, loading response, midstance, terminal stance, preswing, initial swing, midswing, and terminal swing (Fig. 1). Although these distinct periods offer a method to categorize gait, they can be combined to reflect the three functional components for each limb: (1) weight acceptance (initial contact, loading response), (2) single limb support (midstance, terminal swing, preswing), and (3) limb advancement (initial swing, midswing, terminal swing). The first two functional components occur during the stance phases with the third occurring during swing. Running gait cycles are broken down into the following phases: stance phase absorption, stance phase generation, swing phase generation, swing phase reversal, and swing phase absorption. The stance phase is typically emphasized in most clinical running observation and can be broken down into (1) initial contact to foot flat, (2) foot flat to heel-off, and (3) heel-off to toe-off.

Definition of temporal–spatial gait parameters allows objective reports of both walking and running; they define where, when, how long, and how rapidly the individual is in contact with the ground. These parameters include stride time, step time, stride length, step length, gait velocity, and cadence. Stride time is the time from initial contact of one foot to initial contact of the same foot. Step time refers to the period of initial contact of one foot to initial contact of the opposite foot. Stride length is the distance covered between initial contact of one foot to initial contact of the same foot, whereas step length reflects the distance from initial contact of one limb to the initial contact of the opposite limb. Gait velocity is stride length divided by stride time, usually expressed as meters per second or miles per hour. Cadence refers to the number of steps taken in a unit of time, usually expressed as steps per

![Fig. 1. The gait cycle. (A) Walking figure. IC, initial contact; ISW, initial swing; LR, loading response; MST, midstance; MSW, midswing; PS, preswing; TST, terminal stance; TSW, terminal swing. (B) Walking gait cycle. IC, initial contact; LR, loading response; IS, initial swing; MS (first instance), midstance; MS (second instance), midswing; PS, preswing; TO, toe off; TS (first instance), terminal stance; TS (second instance), terminal swing. (C) Running figure. 1. Stance phase absorption. 2. Stance phase generation. 3. Swing phase generation. 4. Swing phase reversal. 5. Swing phase absorption. (D) Running gait cycle for running and sprinting. Absorption, from SwR through IC to StR; generation, from StR through TO to SwR. IC, initial contact; StR, stance phase reversal; SwR, swing phase reversal; TO, toe off. (From Novacheck TF. The biomechanics of running. Gait Posture 1998;7:77–95; with permission.)](image-url)
minute. These characteristics can be monitored with pressure mats, force platforms, and 3D motion analysis camera systems. Foot switches (on/off devices to collect foot contact timing) allow temporal data to be collected but ignore spatial data.

During the walking gait cycle, approximately 60% of time is spent in stance phase and 40% in swing. Under average walking speeds, each double-limb support comprises 10% of the gait cycle (total, 20%), whereas single-limb stance accounts for the remaining 80% of stance phase. During slower walking speeds the double-limb support phase increases, whereas faster walking speeds reflect shorter double-limb support periods (Fig. 2).<sup>3,5</sup> Running reverses the support characteristics found in walking: less than 40% of the gait cycle is spent in stance and greater than 60% is spent in swing. As running speed increases, the time spent in stance decreases and that spent in swing increases. To increase speed, an initial increase occurs in step length, followed by increased cadence. Increased stride length is associated with an increase in velocity and is limited by leg length and the athlete’s ability to generate sufficient force to move the COM forward. Although cadence is trainable, its direct impact on ground contact time, and thus ground reaction forces (GRFs) acting on the runner, reflect that a preferred cadence might be a protective effect to stem both impact forces and loading rates.<sup>6</sup> Typical values for cadence vary from the low 70s to the mid-90s for all runners, including distance runners and sprinters. Fig. 2 illustrates the relationship of stance time to swing time throughout various speeds.<sup>3</sup>

**KINEMATICS OF GAIT**

The study of kinematics involves the use of 3D motion analysis systems that digitally reconstruct the individual’s body as a multisegment system. After infrared markers are placed at specific anatomic landmarks, their position is triangulated by cameras to calibrate the individual into the system. Construction of the coordinates and orientation of the rigid body segments allow calculation of joint angles of the proximal and distal segment, joint angular velocity, and joint acceleration. Measurements are collected for each joint in all three cardinal planes of motion. Movement of the COM of each body segment impacts the overall COM of the individual, which is critical for balance and energy expenditure.<sup>1</sup> During walking, the COM trajectory is at the highest point in stance, when speed is minimum. During running, the COM reaches

![Fig. 2](image_url). Variation in gait cycle parameters with speed. For each condition, the bar graph begins at initial contact on the left and represents two complete gait cycles or strides. Note that as speed increases, time spent in swing (clear) increases, stance time (shaded) decreases, double float increases, and cycle time shortens. *(From Novacheck TF. The biomechanics of running. Gait Posture 1998;7:77–95; with permission.)*
the highest point in the flight phase, when velocity is maximum.\textsuperscript{5} Although skin artifact is inherent in this data collection process, modern 3D motion analysis systems currently represent the most noninvasive tool for reliable kinematic data collection.\textsuperscript{2,7–11}

Running-joint kinematic patterns differ somewhat from those of walking. Differences are more pronounced in the sagittal plane and somewhat muted in the coronal and transverse planes. However, absolute peak values are not always the primary focus of an analysis, because they are also greatly influenced by the athlete’s training level and speed. The timing of the extremes of mobility (and force data presented in the following section) is a more important descriptor because it better exemplifies individual characteristics.\textsuperscript{3} Although 3D gait analysis produces graphs for each plane of motion of each specific joint (Fig. 3), all body segments are coupled in a closed kinetic chain during stance and an open kinetic chain in swing.

**Sagittal Plane Kinematics**

As the body transitions from walking to running to sprinting, the COM is lowered and the body tilt in space shifted forward. The combined effect is to maximize the propulsion phase. Although Novacheck reported that pelvic orientation stays remarkably similar to maintain economy,\textsuperscript{3} recent studies have shown changes in pelvic orientation with running.

Using healthy adult subjects, Franz and colleagues\textsuperscript{12} recently showed that despite an increased stride length in running, hip range of motion is unchanged from walking to running. Instead, the compensations in sagittal plane involved an increase in anterior pelvic tilt and thigh angle (thigh-in-space angle). Furthermore, they observed a greater anterior pelvic tilt in subjects who displayed reduced utilized peak hip extension. Thus, this limitation of hip flexor mobility shifts pelvic orientation and may place the lumbar spine in a compromised position for core muscle activation and have implications for low-back pain.\textsuperscript{13} Both Chumanov and colleagues\textsuperscript{14} and Thelen and colleagues\textsuperscript{15} report that this altered pelvic orientation and relationship between the hip flexors and extensors may also place additional strain on the hamstrings. Preservation of pelvic alignment during running is a key variable to consider in these injury populations.

In walking, the hip shows peak extension just before toe-off and peak flexion in mid-to terminal swing. In running, the peak range of extension is similar to that of walking; however, peak extension occurs at toe-off. Increased peak hip flexion is seen during running to advance the limb in swing. Overall hip flexion/extension and abduction/adduction mobility are increased in running (approximately 60° and 15°, respectively) compared with walking (approximately 40° and 10°, respectively).

Unlike in walking, in running the hip must extend in the later part of swing so as to place the foot in the correct orientation under the body. If this did not occur, foot contact would be too far ahead of the COM and shift the GRFs posterior, thus causing deceleration. Correct hip position at contact can be best visualized as riding on a skateboard or scooter. Planting the stance leg close to the COM at contact tends to preserve the GRF close to the COM and minimize the deceleration component. Placing the foot excessively anterior will cause an abrupt braking force. This characteristic is accentuated in sprinters in that the hip in a more extended position at contact will continue to produce net acceleration with the COM ahead of the contact limb.

The knee exhibits similar patterns in walking and running, flexing once to absorb in stance and once to clear the limb in swing. Although the patterns are the same, the amount of knee flexion in swing increases greatly from approximately 60° in walking to upwards of 90° in running. Some sprinters achieve 105° to 130° of knee flexion in
Fig. 3. Sample plots of kinematics are shown for hip, knee, and pelvis in all three cardinal planes of movement. All data are plotted in percent (%) gait cycle. Data are from a sample of healthy runners collected at the University of Virginia Center for Endurance Sport Gait Laboratory.
In stance, the running knee is flexed approximately 25° at contact and continues flexing to 45° peak in midstance. Less peak flexion in stance occurs in sprinters because of the lower ground contact time.

In walking, the tibia is positioned so that the contact is made at the heel with the ankle in plantar flexion. The position of the tibia is more vertical in running, requiring more dorsiflexion of the ankle to achieve contact. Moving past initial contact, the walking ankle initially plantar flexes to achieve foot flat, whereas the running ankle moves into dorsiflexion as the limb is loaded. Because of shorter contact times, sprinters tend to run more on the forefoot, and thus have less peak dorsiflexion at the ankle in midstance. Moving into propulsion, sprinters have increased plantar flexion at toe-off and decreased need for dorsiflexion to clear the limb in swing because of increased mobility of the knee.

To summarize, net ankle mobility is higher in running (50°) than walking (30°). The timing of peak ankle values occur earlier in the stance phase as running speed increases.

Full dorsiflexion of the metatarsal joints (normally 85°) is not needed for walking or running; however, a limitation (<30° metatarsophalangeal [MTP] at the first ray) can cause significant changes. A significant amount of leverage, and thus stability, is provided from the first ray and its musculotendinous structures from late stance through toe-off. An individual with restricted MTP extension cannot rollover through the metatarsal heads at toe-off, causing the foot to supinate early and thus shifting the base of support laterally on the foot away from the stable first ray. This process introduces a “heel whip” into the gait cycle that forces rotational motion into the entire kinetic chain.

A clinical examination performed with the ankle in a relaxed plantar flexed position can reveal arthrokinematic restriction in mobility of the MTP joints. However, all of the lower leg tissues are continuous in nature. Assessing MTP mobility with the ankle in slight dorsiflexion better replicates the position of the ankle at toe-off and offers the unique ability to examine the combined effects of the calf musculature, Achilles, and plantar fascia together. A limitation in one or all of these structures can limit mobility of the MTP even when the joint itself might move freely. Identification and treatment of the soft tissue structure at fault allows unimpeded progression of the runner through the forefoot.

**Coronal and Transverse Plane Kinematics**

Motion in the mediolateral direction during walking and running is more subtle than in the sagittal plane. Nonsagittal plane mobility in gait acts to moderate stability, force attenuation, and economy in gait. Motions of pronation and supination are visualized not as isolated events occurring only in the foot, but as an interconnected multisegment system that functions together. An alteration in one specific location can impact the joints both above or below that segment. Table 1 summarizes the coupled motions occurring in the lower quarter during the pronation and supination phases.

The hip adducts relative to the pelvis during the stance phase and is abducted relative to the pelvis during the swing phase. Thus, if the runner is on the right foot during midstance, the right hip will be adducted, the contralateral side of the pelvis will be lower, and the lumbar spine will be slightly side bent to the right side. This coupled motion between the hip, pelvis and lumbar spine acts to minimize motion to the trunk and head for balance and equilibrium.

Although some mobility here is beneficial, too much or too little may cause additional problems. Step width may influence this motion, because a crossover or adducted gait may induce excessive pelvic drop on the contralateral side. Weak hip
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Abbreviations: DF, dorsiflexion; Lat, lateral; MTJ, midtalar joint; Opp, opposite; PF, plantarflexion; rot, rotation; STJ, subtalar joint.

abductors can cause increased hip adduction and pelvic drop (Trendelenberg). Weak hip stability can also lead to a compensatory lateral trunk shift to move COM laterally and limit pelvic drop. Altered hip stability recruitment patterns decrease running economy and have been shown to be associated with injury, including patellofemoral dysfunction.\textsuperscript{16,17} Too little motion, with little to no adduction between the hip and pelvis, can also be problematic. Although commonly seen when sprinting all out, this should not be a goal of typical gait. Additionally, too little motion may require additional stabilization at the knee (high metabolic cost) or contribute to inadequate shock transmission. For example, using a wide stance width with little coronal plane mobility at the hip and pelvis can lead to decreased shock absorption. This positioning could create a situation where additional forces could be transferred to the pelvis. Problem-solving through this unique combination of gait attributes can show the biomechanical flaws placing additional stress at the pelvis and thus contributing to an injury such as osteitis pubis or sacral stress fractures. If the cause of this gait abnormality is not corrected in the run technique, the abnormal loading pattern will persist and conditions will become repeat or chronic.

Foot and ankle mechanics are a critical aspect of the evaluation. The foot and ankle function as a mobile lever. Their ability to move into an open packed position (full pronation) to absorb and dissipate shock is critical to achieve contact and avoid excessive shock transmission up the kinematic chain. Likewise, the foot must be able to resupinate at the correct time in push-off to maximize force transfer.

In a clinical setting, barefoot gait evaluation can yield a plethora of information about the foot, but clinicians must be aware of the complex foot mechanics.\textsuperscript{18} In running, it is typically taught that the foot moves from a supinated position at contact (in 6°–8° calcaneal inversion) into 6° to 8° eversion at midstance, and then immediately begins to supinate into a more rigid state.\textsuperscript{1} This premise was based largely on using rearfoot mobility to define the pronatory status of the foot. McPoil and Cornwall\textsuperscript{19} accordingly note that rearfoot pronation peaks at 37.9% of stance phase of walking. Recent work has shown that using the rearfoot to define the pronatory status of the foot may be less specific than using a dynamic foot measurement with motion analysis. Peak composite foot deformation does not occur until approximately 78% of stance in walking and until 52% to 54% in running.\textsuperscript{20} Maximal deformation of the foot in the gait cycle occurs at maximal GRF application.

KINETICS OF GAIT

Although kinematics can be visually evaluated in the laboratory and clinic, they do not show why individuals move the way they do. Kinetics reflect the cause of movement, and therefore the forces, power, and energy that affect the manner in which an individual moves.\textsuperscript{21} GRFs measured with force plates imbedded in the ground or treadmill refer to the forces that act on the body throughout the stance phase.\textsuperscript{22} Analysis of the GRFs acting on the COM is typically broken down into vertical, mediolateral, and anteroposterior force plots (Fig. 4). The origin of force on the foot is termed center of pressure (COP). Processing the COP, GRFs, and joint kinematics together allows calculation of joint kinetics (joint moments). More specifically, joint kinetics show how the external GRFs, inertia, and gravity interact with the internal recruitment of muscles, tendons, ligaments, and bony structures that stabilize the joint. Joint power indicates the velocity of the joint moment, or the rate of the work exhibited by the muscles. Although monitoring kinetics is not possible outside a laboratory setting, understanding these attributes helps clinicians appreciate why runners move the way they do at various points in the gait cycle.
Fig. 4. Sample plot shows raw force data for individuals in the vertical, anteroposterior, and mediolateral components, respectively. All data are plotted in percent stance, with percent body weight (BW) as the unit of measure. For the vertical ground reaction force (GRF), the first peak represents the collision of shoe and lower leg with the ground. The magnitude of the impact varies greatly with contact style, cadence, and slope of the running surface. The more prominent second represents active peak; this is the point in the gait cycle where external forces acting on the body are at their peak, thus triggering maximum internal force generation of the musculotendinous structures to balance this response. Note that the active peak on the vertical GRF always corresponds to the “0” point on the anteroposterior GRF as long as speed is at steady state. Data are from a sample of healthy runners collected at the University of Virginia Center for Endurance Sport Gait Laboratory.
The vertical GRF has two distinct peaks for walking gait and one distinct peak for running. With an average comfortable walking speed, the GRF spikes at initial contact to form a small impact peak, and then increases through the absorption phase peak to approximately 100% to 110% body weight, dips to approximately 80% COM in double support, and finally increases back to approximately 100% to 110% in the force generation phase. Thus, in walking, the typical peak vertical forces are similar to those experienced in a single-leg stance. During running, vertical GRF spikes sharply at contact to produce an impact peak, may slightly decrease (depending on contact style), and then continues to an active peak of 2.2 to 2.6 times the body weight in a typical distance runner.

The impact peak is formed through collision of the shoe, foot, and lower leg mass with the ground. The active peak is influenced by the mass of the runner, landing velocity, and leg stiffness. The presence of an impact peak, or its prominence, can be moderated by alteration in contact style. Runners with a more pronounced heel contact exhibit larger impact peaks, whereas sprinters or toe runners may have no discernable impact peak. Cadence also has an effect, with increased turnover exhibiting muted or even absent impact peaks, and decreased turnover accentuating impact peaks. Cadence change has been shown to be a clinically significant component of a plan to decrease the incidence of bone-strain–related injury. Gottschall and Kram reported that impact force peaks were dramatically larger for downhill running and smaller for uphill running.

Do increased impact forces lead to injury or impact performance? Hreljac is widely cited as identifying impact peak force as the variable that distinguished injured from uninjured runners. However, this concept has been sharply contested by Nigg, who states that impact forces help in pre-tension, or tuning the muscle contraction of the leg before impact. Furthermore, some level of impact force is necessary for the integrity of bone and cartilage. Nigg does speculate, however, that excessive running without sufficient recovery may adversely affect the remodeling rate of bone. Impact peak forces are comparatively small and experienced for a short time during each gait cycle (<8% of each stance phase). Joint contact forces during the impact peak are three to five times smaller than they are during the active force peak, and are therefore within the normal operating ranges for joint function and unlikely to contribute to the development of running injury.

As depicted by the vertical GRF graphs, peak forces on the body are sustained during midstance, not initial contact. The active force peak of the vertical GRF reflects the number of motor units required to hold the body at a given point and occurs at the lowest point of the COM in the stance phase. Decreased active peak for the same speed means a decrease in motor unit activation and metabolic demand to hold speed. The active peak of the GRF can be modified through mass, velocity of the runner, and contact time. Unlike the impact peak, the active peak is not altered with incline or decline as long as steady state velocity is maintained.

Midstance is the time in the gait cycle when GRFs are highest on the body, with peak internal joint moments generating peak mechanical strain on tissues. The ability of the runner to stabilize these forces is essential to tissue health. Dynamic screening tests performed in single-leg stance offer a functional and running specific midstance posture from which to evaluate stability. From a rehabilitation standpoint, challenging a runner’s multiplane stability in single-leg stance replicates the position and phase in gait at which it is most critical.

In addition to vertical GRF, the anteroposterior (or parallel forces) and mediolateral GRFs are analyzed to describe the sagittal and frontal plane forces, respectively.
Assuming the runner is at steady state, the active peak corresponds to the “0” point on the x-axis. Any negative value will then decelerate the COM, with positive values accelerating the COM. At steady state velocity, the time-integrated brake force equals the propulsive force so that minimum mechanical efficiency is lost. Anteroposterior GRF can be altered through contact style, cadence, or incline so that braking peak increases for running downhill and decreases for running uphill. Likewise, propulsion force increases in uphill running and decreases in downhill running. Excessively elevated parallel braking forces can indicate that the individual’s COM is well behind the foot at contact. This tendency can be remedied by encouraging a more neutral postural alignment and foot strike closer to the COM.

Mediolateral GRF allows quantification of the path of the COM in the frontal plane. Excessive deviation from medial to lateral (or lateral to medial) imparts excess coronal plane forces that the runner is required to stabilize. Increased deviation in this plane is often seen when the runner lacks adequate hip and core stability. This increased instability of COM combined with poor neuromuscular control can lead to a “downward spiral” in the stance. Efforts should target identification of the biomechanical fault and be combined with gait drills to minimize the peak deviation of COM in relation to the stance foot.

**Joint Kinetics**

Quantitatively, motion laboratories use inverse dynamics to combine kinematics and external GRF to produce joint moments and powers. The ability to conceptualize the joint moment at each specific gait phase helps in applying a clinical and dynamic evaluation to a gait analysis. The GRF has a given magnitude for a given gait phase. Additionally, this vector has a point of origin or COP and direction in relation to each joint. The location of this vector dictates what the external GRF is doing to the joint, and what the runner is doing to counter this action. For example, during midstance, the ankle joint is in dorsiflexion and the GRF is anterior to the ankle joint line (Fig. 5). The GRF is imposing an external torque trying to dorsiflex the ankle, while the runner activates the plantar flexors to generate an equal and opposite internal joint torque to maintain position of the ankle at that particular point in time. At any instance in stance, three factors influence the joint kinetics: (1) the magnitude of the GRF, (2) the 3D GRF location in relation to the joint, and (3) the origin of the COP (influenced by foot contact style or compensatory foot function). Fig. 6 shows sample graphs of running joint kinetics in healthy young adults.

The origin of COP is affected by contact style. Because rear-foot runners land on the posterolateral border of the foot, the origin of COP initiates on the lateral aspect of the foot and travels distal for two thirds of the stance phase before traveling medial across the metatarsal heads and finally through the first and second rays. Midfoot strikers’ COP originates on the lateral midfoot, with an initial path posterior toward the base of the foot as the rearfoot approximates the ground. After maximum foot contact, the COP moves rapidly toward the medial forefoot. Forefoot strikers share this same general pattern, except that the initial COP origin is even more distal on the foot.

These described paths are typical values, but compensations can occur because of foot mechanics. An individual who has poor forefoot stability may have an excessive lateral to medial COP shift as he moves into propulsion phase, whereas the COP of an individual with hallux rigidus will track more lateral and exit through the lateral forefoot.

Rise and fall of the COM occurs during walking and running; however, this action yields different results in each task. In walking, kinetic and potential energy are out of phase so that COM moves from its highest point in single-leg stance to its lowest point in double-leg stance. This pendulum-like oscillation of the COM is maintained
by the interchange of potential and kinetic energy and, combined with a given amount of muscle work, propels the body. During running, the body acts like a pogo stick in that potential and kinetic energy are in phase. Potential and kinetic energy are highest in the flight phase. Potential energy is lost as the body falls down to the ground. Kinetic energy is lost as the foot achieves ground contact, but is stored in the musculotendinous structures as elastic energy. After midstance, the tendon springs release approximately 95% of their stored energy in combination with tensioning from the muscles to generate the COM upward and thus raise potential and kinetic energy back to their high point. This storage and rebound of the tendon units is an important component of running gait, because they supply most of the work performed.3,36

Although the muscles act as tensioners to the tendons, they do not exhibit much of a change in length. Rather, uniarticular muscles stabilize joints, whereas biarticular muscles transfer energy to adjacent joints.37,38 The gluteus medius is an example of a uniarticular muscle. The gluteus medius exhibits tension during the stance phase to hold the pelvis stable, although it does not undergo a significant change in muscle length. When looking at a biarticular example, the energy from one segment is transferred through the “energy strap” so that the energy of the femur extending over the tibia can be transferred through the hamstrings to extend the hip in relation to the pelvis.3

During walking, a great deal of mechanical work is dissipated at each contact point because of collision of the foot with the ground.39,40 The contralateral trailing limb restores this energy and the velocity of COM, but the difference must be made up from active muscle work (significant muscle change in length) that is not seen in running. In running, a major energy expense seems to be due to body weight support. The overly flexed position of the knee (compared with walking) increases the joint moment and the knee extensor demand.41,42 During running, the stiffness of the stance leg spring may be altered as much as twofold to adapt either to cadence changes outside the individual’s preferred frequency, or to the surface so that stability
Fig. 6. Sample plots of joint kinetics and powers for the hip, knee, and ankle in all three cardinal planes of motion. All data are plotted in stance and represent a sample of healthy runners collected at the University of Virginia Center for Endurance Sport Gait Laboratory.
and efficiency may be maintained across a wide variety of terrain. This compensation maintains COM in a somewhat consistent oscillatory path throughout a wide range of real-world conditions, because excessive rise and fall or the excessive minimizing of the COM has been associated with increased energy expenditure.

Ankle, knee, and hip power patterns are similar to those in walking, except that the amplitudes of power absorption and generation are related to the speed (greater power for greater velocity). Ankle patterns are fairly similar to those in walking, with the joint moment speed faster because of shorter contact time and a higher GRF. Knee sagittal moments in running are higher than in walking. In running the knee is more flexed throughout the stance phase, thus requiring more internal muscular support to maintain stability through midstance. Hip sagittal plane moments are similar to those in walking, except the amplitude is greater in running.

Arm movement has been discounted as a source of propulsion in distance running, and said to produce only lift. Conversely, Novacheck and colleagues identifies the arms as aiding a constant horizontal velocity through counterbalancing the rotation of the lower extremity. The upper body is unique in that it shows gross stabilization deficits in the lower body. An excessively wide arm swing can be a strategy to provide additional lateral support when core/hip control is diminished. Excessive crossover of the arms may indicate a lack of stability in the transverse plane. Although these cues do not identify the exact cause of the imbalance, they reveal movement patterns that may be the cause or, conversely, the compensation for the default.

Restricted hip flexors, lumbar spine extensors, thoracodorsal fascia, inadequate core muscle support, chronic low back pain, posture dysfunction, or postural fatigue during a run can move the runner into excessive lumbar spine extension. This extended lumbar spine position can alter forces at the lower extremity, because the COM tends to migrate forward with changes in trunk angle. Observation of the runner “fresh” and also several miles into a run can reveal unique gait traits that affect gross movement pattern dysfunction.

Additional tools can be used to obtain gait parameters: accelerometers allow measure of loading rate of structure, dynamic electromyography allows examination of the respective muscles responsible for internal joint moment, and monitoring oxygen consumption allows data on the metabolic economy of gait to be collected. All gait technologies have their role; the key is to use the technology that can obtain the parameters of interest.

Economy of Motion

Metabolic efficiency is often discussed in terms of physiologic training; however, biomechanical constraints do have a role. Efficiency equals mechanical work/metabolic work. Martin and Morgan identified four primary areas of study: body structure, kinematics, kinetics, and biomechanical feedback/training. However, they were unable to determine how to improve biomechanical economy. Individuals seem to freely choose their most economic speed when walking, as shown by the inverted U theory in Fig. 7. Between 1.1 and 1.4 m/s, a relatively flat energy curve range, is observed. Slower speed is more metabolically costly for distance given, whereas excessively fast walking speeds show the same trend. This fact provides evidence that a freely chosen range of walking velocities are optimal. During running, no single range of velocities seems as optimal, in that the faster velocity is offset by additional distance covered. Regarding metabolic economy in running, little correlation exists between the energy invested to cover a given distance and distance travelled. Mechanisms that do affect metabolic economy are believed to include stride length, stride frequency, muscle shortening velocity, and mechanical power output. Kram and
Taylor\textsuperscript{32} claim the two most important biomechanical constraints in running economy are the total mechanical work done by the body to support the COM in stance, and the inverse relationship with ground contact time (the lower the time of contact, the higher the expenditure).

Current literature supports three critical aspects with respect to biomechanical economy: (1) minimizing the active muscle mass recruited, (2) aligning the legs with the net force vector, and (3) maximizing the effects of elastic recoil. Increased vertical GRF reflects an increase in motor units to go from point $A$ to point $B$ at a given velocity. Greater force equals greater metabolic cost. Decreased ground contact time minimizes the change in vertical COM height, which in turn increases leg stiffness and causes an increased peak GRF. Optimum ground contact time should be short enough to minimize muscle work but long enough to allow for the release of elastic recoil from the tendons.\textsuperscript{32,48} In this manner, the average force production from each leg (the area under the curve) is maximized without increasing the peak force produced. Furthermore, a change in the vertical component is accompanied through a change in the horizontal component (if peak vertical GRF decreases, then acceleration and braking forces also decrease).

Aligning the force vector closer to the leg decreases the external joint moment and minimizes muscle cost.\textsuperscript{48} This function is achieved through running with straighter limbs during the stance phase. Using straighter limbs to avoid excessive energy loss with increased joint excursion from flight to stance-phase absorption can help maximize the effects of elastic recoil.\textsuperscript{48} Additionally, individualized drills and neuromuscular activities that compliment runners’ structure will train them to use elastic recoil in gait. Although sufficient data from multiple works reinforce these concepts, many questions remain unanswered because of the multifactorial and complimentary nature of gait attributes.\textsuperscript{49}

Observational gait analysis, although not as specific as 3D analysis, can improve a clinician’s ability to bridge the outcome of special tests of dynamic function in
gait. The use of slow-motion video capture can vastly improve visualization of gait form. Requirements include a high-quality video camera, a tripod, and video editing software. Video software makes it possible to edit, slow, or freeze-frame footage for specific analysis. Observational video analysis should comprise 30- to 45-second long data collections from multiple angles: front upper body, front lower body, side upper body, side lower body, rear full body, and posterior view from the knee down for a closer look at the foot function. The most important thing is to be consistent when looking at gait; each evaluation should be approached the same each time, irrespective of diagnosis. Often the site of symptom presentation is not the location of biomechanical fault. Altered upper body and trunk movement patterns may reflect alteration in lateral or rotational plane instability. Special attention to the joint above and below the symptom presentation is essential. Through the course of the video evaluation, the observed asymmetry or gait abnormality should be correlated to the clinical examination findings. The goal is to identify a particular biomechanical pattern that could be affecting the individual’s symptoms.

Examining individuals dynamically provides a perspective on how they use their combination of strength, flexibility, and muscle memory to achieve gait. Ignoring the biomechanical cause of the imbalance will likely result in the dysfunction becoming adopted into their repetitive gait pattern. Modern gait analysis is a tool to produce objective, quantitative parameters that identify the source of dysfunction in an individual or population. Understanding the concepts behind repetitive loading of running will enable a more directed approach to diagnosis and treatment intervention.

REFERENCES