CHAPTER 8

Gait Analysis

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Gait, referring in humans to walking and running, is one of the most fundamental actions in life. Rehabilitation clinicians can especially appreciate the complexity of gait in the face of impairment or functional limitation. Often, an individual has difficulty walking, and for some, gait may be functionally impossible. It is the physiatrist's task to determine the specific causes of why a person cannot walk well, not only at the pathophysiology level, but also at the impairment and functional limitation levels as well. The effectiveness of any physiatric treatment relies heavily on the ability to accurately determine these causes.

NOMENCLATURE

An understanding of gait analysis first requires familiarization with the currently accepted terminology. Because gait is habitual in nature, we often focus our analysis on the functional unit of gait, called the gait cycle, or stride. Various temporal and functional parameters within the gait cycle, presented by Perry and colleagues (1) (Fig. 8-1), form a frame of reference to discuss both nondisabled and disabled gait. This standard classification divides the gait cycle into the stance and swing periods. Similarly, the gait cycle is divided into three basic functional tasks. Weight acceptance and single limb support are the functional tasks occurring during stance, whereas limb advancement is the functional task primarily occurring during swing. These functional tasks are further broken down into eight phases during the gait cycle. The phases of initial contact and loading response comprise the functional task of weight acceptance. The phases of mid-stance and terminal stance comprise single limb support. Limb advancement begins in the final phase of stance (preswing) and then continues through the three phases of swing (initial swing, midswing, and terminal swing). The terms "heel strike" and "toe-off," corresponding to initial contact and preswing, respectively, may be inappropriate and inaccurate in many atypical gait patterns.

Gait velocity refers simply to the speed of gait. Stride time is defined from the time of initial contact of one limb to the next initial contact of the same limb. Step time is defined from the time of initial contact of one limb to the time of initial contact of the contralateral limb. Stride length and step length are the distances covered during their respective time frames. The cadence of gait can be expressed in either strides per minute or steps per minute. At an average walking velocity, the stance period comprises about 60% of the gait cycle, whereas the swing period comprises 40%. In walking, at least one foot is on the ground at all times. During the stance period, there are two time intervals when both feet are on the ground, termed double limb support. One of these time intervals occurs from initial contact into loading response and the other during preswing. Single limb support refers to the time interval in stance when the opposite limb is in swing. At an average walking speed, each double limb support time comprises approximately 10% of the gait cycle, whereas single limb support comprises about 40%. Typical values (1) for temporal gait parameters in adult nondisabled subjects, walking comfortably on a level surface, are summarized in Table 8-1. At slower walking velocities, the double limb support times are greater. Conversely, with increasing walking speeds, the double limb support time intervals decrease. Walking becomes running when there is no longer an interval of time in which both limbs are in contact on the ground.

ENERGY CONSERVATION AND THE DETERMINANTS OF GAIT

To the casual observer, nondisabled walking is a smooth and almost effortless task of locomotion. This efficiency is made possible by minimizing the displacement of the body's center of mass (COM) during walking (2,3). The COM, de-
fined as the hypothetical point at which all mass can be considered to be concentrated, is located just anterior to the second sacral vertebrae in the average human lying in the anatomic position (4). During walking, the COM normally travels along a sinusoidal up-and-down and side-to-side path with each step. It reaches its highest point during single limb support and its lowest point during double limb support. With regard to efficiency of walking, the vertical displacement of the COM is far more relevant than the lateral displacement (5). The major mechanisms by which the body minimizes the displacement of the COM during walking are via a series of maneuvers described as the determinants of gait by Saunders and colleagues (6):

• Pelvic rotation in the transverse plane
• Pelvic obliquity in the coronal plane
• Lateral displacement in the coronal plane
• Interchange between knee, ankle, and foot motion

The subsequent figures serve as simple models to illustrate how each determinant contributes to reducing the COM displacement. Figure 8-2 demonstrates a hypothetical "compass" that assumes what walking would be like without any of these determinants (6). The legs are represented as rigid levers without foot, ankle, or knee components, and articulation only at the hip joints. Normally, pelvic rotation in the transverse plane reduces the drop in the COM during double limb support (Fig. 8-3). A slight amount of pelvic obliquity (i.e., Trendelenburg) reduces the peak of the COM during single limb support (Fig. 8-4). Diminution of the lateral displacement of the pelvis are influenced by two factors. One, the body is shifted toward the side of the stance limb during loading. Two, the natural valgus between the femur and tibia allow the feet to be closer together during forward progression (Fig. 8-5).

The interchange between knee, ankle, and foot motion is a mechanism that further reduces the vertical displacement of the COM during walking and helps alter the pattern of COM motion from a series of arcs as in the hypothetical compass gait situation to the actual characteristic smooth sinusoidal appearance (Fig. 8-6). These joint motions are described in detail in a later section. Relevant to this discussion, the ankle moves into controlled plantarflexion from initial contact into loading, and the knee flexes slightly to reduce the peak of COM displacement in single limb support. Also during single limb support, there is progressive ankle dorsiflexion that similarly effectively reduces the peak of COM displacement. The ankle plantarflexes again in double limb support, which effectively raises the COM’s lowest point. All of these actions occur gradually and in rhythm so as to also smooth the curve of COM motion during gait.

If it were not for the combined action of the determinants of gait, the average total vertical displacement of COM would be about twice the value that it actually is (7,8). Many impairments and functional limitations can interfere with one or more of these determinants and thus increase the COM displacement and energy cost of walking.

**ENERGY COST OF GAIT**

At an average comfortable walking velocity of 80 m/min in nondisabled subjects, the energy expenditure is about four times the basal metabolic rate (5). Interestingly, the velocity
FIG. 8-2. Hypothetical "compass" gait. The pelvis is represented by a single bar with a small cuboid representing the body's COM. The legs are rigid bars articulating only at the hip. No foot, ankle, or knee joints are present. The pathway of the COM is a series of interconnected arcs. (Reprinted from Rose J, Gamble JG. *Human walking*, 2nd ed. Baltimore: Williams & Wilkins, 1994, with permission.)

FIG. 8-3. Effect of pelvic rotation in the transverse plane. The slight rotation of the pelvis in the transverse plane during double-limb support reduces the elevation needed by the COM when passing over the weight-bearing leg during midstance. (Reprinted from Rose J, Gamble JG. *Human walking*, 2nd ed. Baltimore: Williams & Wilkins, 1994, with permission.)
that subjects choose as their comfortable speed is also the velocity that requires the least energy per unit distance (9). Walking faster or running usually requires anaerobic metabolism. On the other hand, walking slower requires extra energy, probably for balance support, rather than for propelling the body forward (10). Importantly, the rate of energy expended during comfortable walking is consistent across the nondisabled and disabled gait populations (5). A person with a gait disability tends to walk slower than a person without a gait disability (11). Thus, although the energy expenditure per unit time is consistent in subjects with gait disability, increases in energy expenditure per unit distance are common. For instance, patients with hemiplegia affecting their gait spend the same amount of energy per time during comfortable walking as subjects without gait disability, but they walk slower and spend 37% (12) to 62% (13) more energy per unit distance.

An important aim of improving gait disability may be to reduce the energy required to walk. To this end, the effectiveness of a particular type of rehabilitation treatment can be assessed by evaluating the energy expended during walking. The most direct method to evaluate energy expended is via measuring the oxygen that is consumed during walking. This involves having the subject breathe into a mask that is linked to a gas analyzer. The analyzer determines how much oxygen is being used, and from this, a calculation of energy expenditure is given, based on the knowledge that about 4.83 kcal of energy is expended for every liter of oxygen consumed (5). Alternatively, an estimate of energy expended can be obtained by measuring heart rate before and during walking because the change in heart rate that occurs with walking is linearly correlated with oxygen consumption measurements (14). An easier, although more indirect, method to evaluate the energy required to walk is to measure the comfortable walking speed. This can be performed using a stopwatch and a designated walking distance. This simple measure rests on the fact noted above that subjects with gait disability tend to walk at a consistent energy rate, just slower. Thus, comfortable walking speed relates indirectly to the energy required to walk. Unfortunately, all of these measures, including oxygen consumption, heart rate, and comfortable walking speed, relate not only to biomechanical aspects of walking, but to cardiopulmonary conditioning and psychological factors including mood as well. Quantitative gait measures described in a later section, although useful in determining the mechanisms of the gait impairment, are insufficient in evaluating the efficiency of walking. A so-called biomechanical efficiency quotient was proposed (8,15) based on the concept of minimizing the COM displacement through the determinants of gait. This measure was introduced as a means to specifically evaluate biomechanical walking efficiency in subjects with gait disability, independent of cardiopulmonary conditioning and psychological factors. The quotient is the measured vertical displacement of the COM divided by the predicted vertical displacement, the latter being a function of the subject’s average stride length and height of the pelvis from the ground. Patients with gait disability tend to have higher biomechanical
efficiency quotients than subjects without gait disability and treatments such as an ankle-foot-orthosis tend to reduce the biomechanical efficiency quotient (15).

CONCEPTS FOR UNDERSTANDING GAIT EVENTS

Inasmuch as the determinants of gait result in a more efficient method of human locomotion, they make human walking a rather complex concept to understand. In order to evaluate the mechanisms of gait disability and therefore to identify individualized therapeutic interventions, a basic knowledge of the events during a normal gait cycle is necessary. Kinematics describe the spatial motions of joints and limb segments. Quantitative gait analysis, described in a later section, can be used to quantitate normal kinematics during the gait cycle (16,17). However, observational gait analysis also can provide important qualitative kinematic information. Kinetics describe the moments or torques and forces that cause joint and limb motion, and these are not intuitive from observational gait analysis. Only quantitative gait analysis can provide kinetic information. Similarly, the firing patterns of muscles can be determined only with the aid of dynamic electromyographic (EMG) measurement used in quantitative gait analysis.

Broadly speaking, the study of kinetics includes the study of muscular activity as well as the study of forces, calculated using physics, and provides insight about the causes of the observed kinematics. In quantitative gait analysis, we are often interested in computing the net moments acting on muscles, tendons, and ligaments. A moment about a joint occurs when a force is acting at a distance from the joint through a lever, causing acceleration of the joint angle. For instance, an externally applied extensor moment about the elbow is produced when a weight is placed in the hand. In this case, the lever is the forearm and the elbow will tend to accelerate uncontrollably into extension. The external moment can be mathematically calculated as the product of the weight of the object and the length of the forearm. In order that a joint angle remains stable, all the moments acting about the joint must sum to zero. An internal force from the biceps humerus acting through its forearm lever can provide a resisting internal flexor moment such that the elbow joint is stabilized. Depending on the magnitude of the force through the biceps, the elbow joint angle will extend in a controlled fashion (eccentric contraction), stay the same (isometric contraction), or flex (concentric contraction). These concepts are applied repeatedly in gait analysis. At each point in the gait cycle, the hip, knee, and ankle joints are stabilized, such that all the moments about a particular joint are in a state of equilibrium. The externally applied moments from gravity, inertia, and the ground are countered by internal joint moments generated by muscle activity and/or soft tissue. During the swing period of the gait cycle, most of the external moments occurring about the lower limb joints are a result of gravitational and inertial forces from the individual limb segments. For instance, during swing, both the weight of the foot and the inertial force from the swinging lower leg will generate an external plantarflexor moment that needs to be restrained by an internal dorsiflexion moment provided by the ankle dorsiflexors in order to prevent foot drop.

During the stance period of the gait cycle, most of the external moments occurring about the hip, knee, and ankle joints are produced from the ground reaction force (GRF). In quiet standing, the body weight pushes against the ground. The ground reacts with an equal and opposite GRF, the vector of which passes through the base of support (the feet) up toward the COM of the body. When we walk, the GRF is essentially a result of both the weight of the body and the body's accelerations and decelerations as our COM moves up and down. Knowing where the line of the GRF lies with
respect to the hip, knee, and ankle joints gives us a reasonable approximation of the external moments occurring about each of these joints. The GRF can be directly measured with a force plate, described later in the quantitative gait analysis section. A more exact estimation of the external moments during the stance period, which includes the additional effects of gravitational and inertial forces, is ordinarily performed with quantitative gait analysis. These additional gravitational and inertial forces are small during stance at slow and normal walking speeds and thus can be ignored for now in understanding normal gait function (18). Visualizing where the GRF lies with respect to a joint provides a means to understand what internal moments must be generated in order to stabilize that joint. For instance, if the GRF line lies posterior to the knee, an external knee flexor moment is produced that is the product of the GRF multiplied by the distance of the GRF line from the knee joint. In order to maintain stability so that the knee does not collapse uncontrollably into flexion, an internal knee extensor moment must occur. This moment, provided by the knee extensors, is equal in magnitude to the external flexor moment.

The concept of joint stabilization and the importance of knowing where the GRF lies in relation to the joints are best exemplified during quiet standing. In quiet standing, the GRF extends from the ground through the mid-foot, passing anterior to the ankle and knee joints and posterior to the hip joints (Fig. 8-7). At the hip, the external extensor moment is countered passively by the iliofemoral ligaments. Similarly, at the knee, the external knee extensor moment is countered passively by the posterior capsule and ligaments at the knee. At the ankle, the external dorsiflexion moment can be countered with an internal ankle plantarflexor moment provided by the ankle plantarflexors (or alternatively with an ankle-foot-orthosis with a dorsiflexion stop equivalent). Thus, the only lower extremity muscles that need to be con-
FIG. 8-8. The eight phases of the gait cycle include initial contact, loading response, midstance, terminal stance, preswing, initial swing, midswing, and terminal swing. The GRF vector is represented by a solid line with an arrow. The active muscles are shown during each phase of the gait cycle. The uninvolved limb is shown as a dotted line.

Consistently active during quiet standing are the ankle plantarflexors.

During walking, the GRF line moves in a posterior-anterior direction as the body progresses forward (Fig. 8-8). During loading response, the vector is anterior to the hip and posterior to the knee and ankle. In mid-stance, the vector passes through the hip and knee joints and is anterior to the ankle. During terminal stance, the vector moves posterior to the hip, anterior to the knee joint, and maximally posterior to the ankle. With these dynamics in mind, normal gait function is easier to interpret. The muscles fire in response to the need for joint stability. Furthermore, whether the muscle is firing concentrically or eccentrically depends on the corresponding joint motion at that time. In quantitative gait analysis, whether a muscle group is firing concentrically or eccentrically can be determined by measuring the joint power that is mathematically the product of the joint moment and the joint angular velocity. A positive joint power implies that the muscle group is firing concentrically, whereas a negative joint power implies that the muscle group is firing eccentrically. Interestingly, most of the muscle activity that occurs in walking is eccentric. Also, it is interesting to note that each muscle group undergoes a phase of stretching and/or eccentric contraction before each concentric contraction.

NORMAL KINEMATICS, KINETICS, AND MUSCLE FUNCTION

The following descriptions of normal kinematics, kinetics, and muscle activity are based on data collected from the Spaulding Rehabilitation Hospital Gait Laboratory and are similar to those reported elsewhere. The following general patterns of movement are fairly representative in nondisabled subjects across most ages after the age of 3 years (19,20).

Sagittal Plane Motion

For each phase, the kinematics, kinetics, and muscle activities are described. Figure 8-8 illustrates the chief actions occurring in each phase with a visual representation of the limb and joint positions, the GRF line, and the muscles that are active during that phase. It also may be useful to refer
to Figure 8-12, later in this chapter, which graphically demonstrates the joint motion, moments, and powers throughout the gait cycle.

Initial Contact

Initial contact with the ground typically occurs with the heel in nondisabled gait. The hip is maximally flexed at 30°, the knee is fully extended, and the ankle is in a neutral position. Because the GRF is anterior to the hip, the hip extensors (gluteus maximus and hamstrings) are firing to maintain hip stability. At the knee, the GRF creates an extensor moment, which is countered by hamstring activity. The foot is supported in a neutral position by the ankle dorsiflexors.

Loading Response

The primary purpose of loading response is to provide weight acceptance and shock absorption while maintaining forward progression. The hip extends and will continue to extend into the terminal stance phase. Because the GRF is anterior to the hip, the hip extensors must be active to resist uncontrolled hip flexion. This hip extension implies that the hip extensors are concentrically active. With the location of the GRF now posterior to the knee joint, an external flexor moment is created. This external moment is resisted by an eccentric contraction of the quadriceps allowing knee flexion to approximately 20°. Because the GRF is posterior to the ankle, an external plantarflexion moment occurs that rapidly lowers the foot into 10° of plantarflexion. This action is controlled by an eccentric contraction of the ankle dorsiflexors. At the end of loading response, the foot is in full contact with the ground.

Midstance

During midstance, the limb supports the full body weight as the contralateral limb swings forward. The GRF vector passes through the hip joint, eliminating the need for hip extensor activity. At the knee, the GRF moves from a posterior to an anterior position, similarly eliminating the need for quadriceps activity. Knee extension occurs and is restrained passively by the knee’s posterior capsule and ligaments and is possibly actively restrained as well by eccentric popliteus and gastrocnemius action. At the ankle, the GRF is anterior to the ankle, thus producing an external ankle dorsiflexion moment. This moment is countered by the ankle plantarflexors, which eccentrically limit the dorsiflexion occurring during this phase.

Terminal Stance

In terminal stance the body’s mass continues to progress over the limb as the trunk falls forward. The GRF at the hip is now posterior, creating an extensor moment that is countered passively by the iliofemoral ligaments. The hip is now maximally extended at 10°. At the knee, the GRF moves from an anterior to a slightly posterior position. As the heel rises from the ground, the GRF becomes increasingly anterior to the ankle joint, and this dorsiflexion moment continues to be stabilized by ankle plantarflexor activity. During this phase, the ankle is plantarflexing; thus, the action of the ankle plantarflexors has switched from eccentric to concentric.

Preswing

The purpose of preswing is to begin propelling the limb forward into swing. This second interval of double limb support is occurring as the contralateral limb now advances through initial contact and loading response. From maximal hip extension, the hip now begins flexing and will continue flexing throughout the swing period. The hip flexors (combined activation of the ilioptoaos, hip adductors, and rectus femoris) are concentrically active. The knee swiftly flexes into 40° of flexion as the GRF progresses rapidly posterior to the knee. Knee flexion may be controlled by rectus femoris activity. Thus, the rectus femoris is simultaneously acting concentrically at the hip and eccentrically at the knee. The ankle continues plantarflexing to approximately 20° with continued concentric activity of the ankle plantarflexors.

Initial Swing

The purpose of initial swing is to continue propelling the limb forward. Hip flexion occurs because of the hip flexion momentum initiated in preswing and the continued concentric activity of the hip flexors. During initial swing, the limb accelerates mainly as a result of concentric hip flexor activity. The knee continues to flex to approximately 65°. This knee flexion occurs passively as a combined result of hip flexion and the momentum generated from preswing. The ankle dorsiflexors are concentrically active as the ankle dorsiflexes.

Midswing

In midswing the limb continues to advance forward, primarily as a pendulum from inertial forces generated in preswing and initial swing. The hip continues to flex, now passively, as a result of the momentum generated in initial swing. The knee begins to extend passively as a result of gravity. The ankle remains in a neutral position with the continued activity of the ankle dorsiflexors.

Terminal Swing

At terminal swing the previously generated momentum has to be controlled to maintain sufficient stability before the upcoming weight acceptance phase. At the hip and knee joint, strong eccentric contraction of the hamstrings deceler-
ate hip flexion and control knee extension. The ankle dorsiflexors remain active to ensure a neutral ankle position at initial contact.

**Coronal and Transverse Plane Motion**

Most lower extremity motion during gait occurs in the sagittal planes. The joint motions and kinetics about the hip in the transverse plane and about the knee and ankle in both the coronal and transverse planes are normally quite small. Although significant motion and associated moments occur in these planes in various gait disabilities, it is difficult to reliably measure these parameters with current quantitative gait analysis techniques. However, significant coronal plane motion and kinetics do occur about the hip (and pelvis) normally and can be accurately evaluated with quantitative analysis. At initial contact both the pelvis and hip are in neutral positions in the coronal plane. During loading response, GRF passes medially to the hip joint center as the opposite limb is unloading. This medial GRF causes an external adductor moment, which tends to allow the contralateral side of the pelvis to drop slightly (the slight Trendelenburg noted previously as one of the determinants of gait). This motion is controlled by eccentric contraction of the hip abductors. During mid-stance and terminal stance, the GRF is still medial to the hip; however, now the contralateral side of the pelvis is lifted concentrically by the hip abductors. During preswing, unloading of the limb causes the ipsilateral side of the pelvis to drop again.

**GENERAL APPROACH TO EVALUATING A PATIENT WITH AN ATYPICAL GAIT PATTERN**

Although a number of atypical gait patterns have been described, each patient has a unique set of impairments, functional limitations, and associated compensations causing these patterns. Examples of atypical gait patterns associated with distinct diagnoses are described in the following sections. Especially in the case of upper motor neuron (UMN) pathology, a stereotypical description of the gait pattern may be sufficient for an initial classification but is too imprecise for determining the mechanisms in an individual patient. It is important to determine these mechanisms in individual patients because they are the basis for directing optimal rehabilitation treatment.

It should be noted that an atypical gait pattern may or may not be functionally significant and thus may or may not be considered a true gait disability. Thus, the atypical gait pattern first should be evaluated with respect to each of the following:

- Energy requirement
- Risk of falling
- Biomechanical injury
- Cosmesis

Treatment to change the gait pattern should be prescribed if the pattern is functionally significant with respect to these four criteria. For instance, the pattern of knee recurvatum (or hyperextension) can be functionally significant if it increases the energy required to walk by not allowing the peak of the COM to be minimized during single limb support. Alternatively, knee recurvatum may or may not be associated with increased forces across the posterior capsule and ligaments of the knee (21), which would predispose to biomechanical injury. Another example is equinus during the swing period, which may or may not predispose to falling depending on the associated compensations. To this end, the associated compensatory gait patterns also need to be evaluated with respect to these four criteria. In the case of equinus in swing, a compensation at the pelvis such as hip hiking would interfere with the pelvic obliquity determinant and thus increase the energy required to walk. Finally, an atypical gait pattern should be evaluated with respect to cosmesis. For this assessment, the patient’s own perceptions are far more important than the clinician’s perceptions.

From the examples above, it is clear that a detailed evaluation of the patient is required. The summary of the patient’s history and musculoskeletal examination, observational gait analysis, and information from a quantitative gait analysis assist in determining the functional significance of the gait patterns and help identify specific causes for each pattern. Based on these results, a detailed treatment plan can be prescribed.

**STATIC EVALUATION**

**History**

The initial part of a comprehensive gait evaluation should include a focused history and physical examination. Based on this evaluation, the underlying diagnosis (or diagnoses) can be classified as a UMN pathology, lower motor neuron (LMN) pathology, orthopedic disorder, amputation (the evaluation of which is described in another chapter), cerebellar or basal ganglia related disorder, or psychogenic cause, to name a few. It is helpful to anticipate certain gait patterns associated with these diagnoses as well as to anticipate the need for various components of a quantitative gait analysis. For example, the use of dynamic EMG is particularly useful in detecting inappropriate firing patterns in patients with UMN pathology but may not be necessary for every patient with one of the other mentioned diagnostic categories. The reason for referral should be identified, and the patient’s chief complaint with regard to walking should be considered. Any previous medications, neurolytic procedures, or surgeries affecting the lower extremities should be noted. Also, a detailed history of strengthening and stretching exercises previously and currently being performed should be ascertained. Finally, the use of assistive devices and/or orthotics should be recorded.

**Physical Examination**

The physical examination should focus on the neurologic and musculoskeletal system and include a static evaluation
of the patient’s strength, joint range of motion, tone, and proprioception. Although static evaluation is a routine part of a gait consultation and should be included in the assessment of every patient with an atypical gait pattern, it is generally agreed upon that, especially in the case of an UMN pathology, the static evaluation has limited usefulness in determining the underlying mechanisms responsible for the atypical gait pattern (22–24). Thus, often the results from the static evaluation need to be combined with those obtained from quantitative gait analysis to provide dynamically relevant information on which to base treatment.

Strength

Classic evaluation of strength involves quantitative manual muscle testing about each joint. It requires the ability of the patient to cooperate with resistive movement of the examiner (25), which are often difficult in patients presenting with a UMN pathology. The patient with a UMN pathology has impaired voluntary muscle control in the affected limbs so that selectively activating an agonist while simultaneously relaxing the antagonist may be impossible. Thus, the result is a limited relationship between static strength performance and dynamic strength associated with gait. For example, a patient with hemiplegia affecting his or her gait may not be able to dorsiflex the foot during static examination, yet when walking may be able to actively dorsiflex during the swing period of the gait cycle (26,27), presumably under the control of primitive reflexes. Conversely, a patient with normal dorsiflexion strength of the ankle during static evaluation may demonstrate an equinus gait during the swing period of gait.

Range of Motion

Determining the passive range of motion at each joint is the traditional method to assess soft tissue contracture and should be performed in at least the lower extremities in all patients presenting with a gait disability. However, it is important to note that this static testing is somewhat limited, particularly in the case of UMN pathology. Differentiating between contracture of a one-joint and a two-joint muscle is difficult in patients with a UMN pathology, undoubtedly because of impaired selective control of these muscles. Furthermore, there seems to be a limited relationship between static range of motion observed with passive ranging and the dynamic range of motion that occurs during gait.

Three clinical tests are commonly performed in patients with UMN pathology to screen for contractures of a two-joint muscle. The Duncan-Ely test differentiates between a rectus femoris and a iliopsoas contracture given that the rectus femoris is both a hip flexor and knee extensor. In this test, the patient is placed in a prone position and the knee is rapidly flexed. With a contracture of the rectus femoris, the hips will flex and the buttocks will rise off the table. Although this test is somewhat useful, EMG studies have demonstrated that this test induces activity not only in the rectus femoris, but also in the iliopsoas in some patients with cerebral palsy affecting their gait (28). The Silverskiold test is used to differentiate between a contracture of the soleus and the gastrocnemius muscle. Whereas the soleus is a one-joint muscle, the gastrocnemius is both a knee flexor and ankle plantarflexor. With the patient in the sitting position, the knee is flexed at 90° and the foot is brought to maximal dorsiflexion. With a gastrocnemius contracture, some of the ankle dorsiflexion will be lost when the knee is extended. It has been shown that this test is not always clinically reliable (29). The Phelps test differentiates a contracture of the gracilis from the other hip abductors, given that the gracilis is the only hip adductor that crosses the hip and the knee. With the patient in a prone position, the knees are flexed and the hips are brought into an abducted position. A gracilis contracture is present when the hip adducts when one knee is extended.

Structural deformities also may contribute to reduced range of motion and gait. If indicated, further clinical tests and x-rays are helpful to document common problems such as femoral anteversion, knee valgus and varus, tibial torsion, and foot abnormalities. These structural problems are sometimes associated with other underlying diagnoses and impairments and can have significant impact on the patient’s walking.

Tone

Tone in all muscle groups should be assessed in each patient presenting with a gait disability. Clinical examination of tone involves testing for resistance by passively moving a joint through its range of motion. This assessment is fairly subjective and is dependent on time of day, temperature, and limb position. Thus, as in the other static tests, there is often a limited association between what is observed statically and what actually occurs during gait.

Proprioception

Joint sense position should be evaluated in all patients in whom a neurologic diagnosis is suspected. If this is impaired, it is important to also evaluate the degree of impairment by evaluating joint position sense not only at the great toe, but at the ankle, knee, and hip as well.

OBSERVATIONAL GAIT ANALYSIS

Observational gait analysis is common practice for physiatrists. The observer describes the gait after watching the patient walk without the aid of any electronic devices. However, it is often difficult to appreciate all limb segment and joint motions throughout the different phases of gait because of the difficulty in concurrently observing the multiple body segments and joint motion (30). Videotaping can be an important part of observational gait analysis because it allows repeated viewing of the patient’s gait pattern without causing
undue patient fatigue. The patient should be observed from the side and from behind. Stride and step length, width, and symmetry should be noted. By concentrating on one joint at a time, including hip, knee, and ankle, atypical motions may be easier to identify. Having the patient walk at faster speed sometimes exaggerates an atypical motion. Observational gait analysis can identify obvious atypical gait patterns, such as excessive ankle plantarflexion or reduced knee flexion in swing. However, in certain cases this approach may not show all atypical patterns. For example, an increased lumbarlordosis or anterior pelvic tilt due to a hip flexion contracture may be apparent only via quantitative analysis. Moreover, quantitative gait analysis can be quite helpful in delineating the specific causes for each atypical pattern and thus help direct the appropriate treatment.

A number of terms are commonly used to characterize various atypical gait patterns that are obvious from observational assessment alone. For instance, antalgic gait has been described as a pattern common to patients with pain in one lower extremity. In this pattern, gait is modified to reduce weight bearing on the involved side. The uninvolved limb is rapidly advanced to shorten stance on the affected side. Gait is often slow and steps are short in order to limit the weight-bearing period. Steppage gait is a compensatory gait pattern used to describe excessive hip and knee flexion to assist a “functionally long” lower leg to clear the ground in swing. Festinating gait have been described as a characteristic pattern of Parkinson’s disease, in which there is a tendency to take short accelerating steps. Shuffling gait is also common in Parkinson’s disease and refers to the feet shuffling during swing. Ataxic gait, associated with cerebellar pathologies, peripheral neuropathies, and dorsal column pathologies, is a broad term used to describe a pattern of apparent poor balance, a wide base of support, and variable motions from stride to stride.

Various gait patterns associated with the use of assistive gait devices are easily noted with observational analysis. The specific indications and use of each of these type of devices are described in detail in another chapter. A cane essentially increases the base of support by providing an additional point of contact with the ground. When pathology, impairment, and functional limitation involve bilateral extremities, two canes or crutches are occasionally used. In this situation, an alternating two-point gait is commonly used in which one cane and opposite lower limb are in contact with the ground alternating with the opposite cane and lower limb in each successive step. In three-point gait, contact with one limb that fully bears weight onto the ground alternates with full weight-bearing through two crutches that make simultaneous contact with the ground. In four-point gait, which provides maximal stability and base of support (at the cost of reduced speed of locomotion), there is always three points of support on the ground at all times. It is initiated by forward movement by an upper extremity crutch, followed by forward movement of the other crutch followed by forward movement of the other lower limb.

QUANTITATIVE GAIT EVALUATION

Modern-day quantitative gait analysis systems typically include measurement of three primary components: kinematics, kinetics, and muscle activity. Quantitative gait analysis also can include other components such as footswitches and oxygen consumption monitoring to measure overall energy expenditure. To measure these various components, a variety of equipment is used, including optoelectronic motion analysis systems to measure kinematics, force plates to help measure kinetics, and a multi-channel dynamic EMG apparatus to measure electrical muscle activity in multiple muscles during gait. Given the previously described limitations of static evaluations and of observational gait analysis, quantitative gait analysis can be a particularly useful clinical tool for developing a treatment plan.

Modern quantitative gait analysis is clearly recognized as useful in outlining an effective orthopedic surgical treatment plan in patients with spastic parietic gait from cerebral palsy (23,31–33). Children with cerebral palsy often undergo tendon lengthening or transfer procedures to improve range of motion in the lower extremities in an effort to improve gait disability. The results from a detailed quantitative gait analysis can help determine the best surgical plan (i.e., which tendons should be lengthened or transferred) to provide the most optimal gait. In the same way that quantitative gait analysis is helpful in orthopedic decision making in patients with UMN pathology, it should be similarly useful in directing these patients’ rehabilitation management. Many physiatric treatments, as described in other chapters, include intramuscular neurolytic techniques, strengthening, bracing, functional electrical stimulation, stretching, modalities, and many other management techniques aimed at (a) strengthening or compensating for weakness, (b) stretching a contracture, and/or (c) reducing tone in a spastic muscle. The outcome of these treatments ultimately rely on the proper determination of the specific underlying impairment or functional limitation causing the gait disability. In some instances, rehabilitation treatments are aimed at improving motor control through, for example, EMG biofeedback or neuromuscular re-education. In these instances, quantitative gait analysis is especially helpful in determining which specific muscle groups are firing at inappropriate times.

Unfortunately, skepticism still persists about the value of quantitative gait analysis in defining a physiatric therapeutic plan because there have been few reports about the value of quantitative gait analysis as a useful evaluation tool in rehabilitation. Human gait is complex. Quantitative analysis offers a clinical tool to better understand these complexities and thus prescribe an optimal rehabilitation treatment program (34). Some of the reluctance in using quantitative gait analysis may be due to the heavy time commitment necessary to understand and interpret the data and the necessity
for teamwork between many disciplines, including medicine and engineering. The cost for gait analysis systems is declining, and the technology required for acquiring and analyzing the data is continually improving. A rapid expansion of computer and optoelectronic technology has brought dramatic changes in image-based motion analysis in the past 10 years. It is anticipated that quantitative gait analysis will soon become a routine clinical evaluation, much like electrodiagnosis has become a routine clinical extension of our physiatric examination. Formal training in quantitative gait analysis, which is already a mandatory part of our physiatric residency curriculum, is likely to become the norm.

Systems to Evaluate Temporal Parameters of Gait

Common temporal parameters such as velocity, cadence, and stride length can be measured to monitor a patient’s progress outside of a sophisticated gait laboratory. As noted previously, velocity can be measured simply with a stopwatch as a patient traverses a designated distance. Similarly, step and stride length can be measured with sophisticated equipment if the walkway is sprinkled with talcum powder. Computerized stride analyzers may provide this same information in a more automated fashion (35,36). They usually consist of instrumented insoles with footswitches (i.e., pressure sensitive transducers), typically attached to the heel, toe, and occasionally the metatarsal region. They are connected to data boxes worn by the patient either around the waist or the ankle. These sensors measure the duration of floor contact via opening and closing switches. After acquisition, data transfer and analysis are typically performed using a personal computer.

Footswitches are also commonly used in gait laboratories to help determine the beginning and end of the stance period, allowing calculation of temporal gait parameters such as the duration of the stance and swing periods, single and double support time, and cadence. These parameters are useful in interpreting the temporal relationships of kinematic, kinetic, and particularly dynamic EMG data. Although this same information can be obtained directly from force plate data, footswitches are particularly helpful in the gait laboratory when force plate data cannot be obtained.

Foot Pressure Systems

Foot pressure systems are electronic instruments to measure pressure distributions in the soles of the feet. The systems work via a large number of capacitive or force sensitive sensors in foot insoles or platforms and are linked to a computer by either cable or radiowave telemetry. Several commercial systems are available and used clinically and for research. These systems may help direct appropriate shoe wear and orthotic prescriptions by providing information about abnormal pressure distributions, particularly in patients with structural foot deformities or in patients at risk for developing skin ulcerations in the feet because of diabetes mellitus or other underlying vascular and peripheral neuropathy disorders.

Kinematics

Electrogoniometers

Electrogoniometers are computerized versions of simple goniometers, which are commonly used in clinical practice to assess joint range of motion. An electrogoniometer consists of one or more potentiometers placed between two bars, with one bar strapped to the proximal limb segment and the other strapped to the distal limb segment (Fig. 8-9). The potentiometer, which is placed over the joint, provides a varying electrical impulse, depending on the instantaneous angle between the two limb segments. This electrical impulse information is then interfaced to an analog-to-digital converter in a personal computer to plot joint angle information over time. A combination of three potentiometers allows for measuring three rotations between limb segments (37).

A major disadvantage of current electrogoniometers is relatively poor accuracy because they are difficult to apply, particularly about the hip and ankle. Unfortunately, even in the case of good accuracy, the results obtained from electrogoniometers provide only relative joint angle information, not absolute positions of the joints of limb segments. Because of these limitations, electrogoniometers cannot be used in conjunction with force plate data to evaluate joint kinetic data.

Cinematography

Historically, gait analysis was performed using sequential photographs or motion pictures. Markers placed over various anatomic landmarks can be used to help identify the location of limb segments and joints. The location of markers can

FIG. 8-9. Electrogoniometer. A potentiometer placed at the joint center records varying electrical impulses depending on the relative position of the proximal and distal segments.
then be manually digitized, frame by frame, so that the marker position in two dimensions can be determined. In both cinematographic and optoelectronic systems, a single camera provides two-dimensional information. By using two cameras, triangulation can be performed to determine the three-dimensional position of each marker. Although the cinematographic system is theoretically as accurate as what can be obtained with modern-day optoelectronic systems, the time necessary to manually digitize and process the data is of such great magnitude that it makes this procedure unfeasible for routine clinical evaluation.

**Optoelectronic Motion Analysis**

Modern-day quantitative gait analysis typically involves a sophisticated computerized video camera apparatus, referred to as an optoelectronic motion analysis system. These systems measure the three-dimensional location of an individual marker in a manner similar to that in cinematography, but with far greater ease and speed. The system automatically digitizes the position of each marker from each video camera and then automatically triangulates the information to provide a three-dimensional position of each marker at each frame. A layout of a typical laboratory space that includes an optoelectronic motion analysis system is illustrated in Figure 8-10. Typically, an optoelectronic system can detect the true three-dimensional position of a marker within a few millimeters in each of the three axes. The specific type of camera or lenses that are used, the algorithms used to digitize or identify markers, the size of the markers, and the laboratory environment are all factors that determine the specific accuracy of any given system. Marker position is typically determined at every 1/50, 1/100, or 1/200 of a second, depending on the speed of the cameras used. Multiple

![Optoelectronic motion analysis system](image)

**FIG. 8-10.** Optoelectronic motion analysis system. Patient walks along a walkway with reflective markers attached to specific anatomic reference points. Camera pairs record the three-dimensional locations of the reflective markers. Force plates located in the center of the walkway record GRFs. Computer programs combine three-dimensional coordinates and GRFs to calculate joint kinetics and kinematics.
markers are affixed to the skin of the pelvis and the lower extremities in relationship to bony landmarks. Similar to the cinematographic method, two cameras are necessary to visualize each marker to obtain its three-dimensional position. Often a camera cannot visualize a marker during a particular part of a movement because of limb rotation or because another limb segment gets in the way. For this reason, sometimes a laboratory uses more than two cameras to ensure that at every given frame of movement, at least two cameras can visualize each of the markers. In the case where three or more cameras visualize a marker, an algorithm must be used to determine the true position of the marker because there is invariably some error such that not all cameras converge on the identical three-dimensional position. Other laboratories strategically position the markers so that the same two cameras can visualize a particular marker throughout the movement.

Currently, there are two different types of optoelectronic systems used for quantitative gait evaluation: (a) active marker systems, where the markers are actively illuminated by a computer, and (b) passive marker systems. A built-in advantage of an active marker system is that the computer knows in advance which marker it is illuminating at any given frame so that the markers are automatically identified as the lateral femoral epicondyle marker, the lateral malleolus marker, etc. The main disadvantage of current active marker systems, however, is that the illuminators require power; thus, multiple wires connected to a power source need to be attached to the patient, which tend to encumber the patient's gait. In contrast, passive marker systems require only that a small infrared reflective piece of material be placed over each anatomic landmark. Although passive markers do not encumber the patient, they do require some additional type of system to determine which marker is which. Fortunately, sophisticated computer software programs have been developed that automate this procedure. Thus, passive marker optoelectronic systems have become the preferred systems for routine clinical practice and are readily commercially available with all necessary software programs.

In order to obtain estimates of joint motion, the optoelectronic system is coupled to a biomechanical or mathematical model that defines where on the body the markers are optimally placed (Fig. 8-11). A simple model to measure knee motion might involve placement of one marker over the greater trochanter, one marker over the lateral femoral epicondyle, and one over the lateral malleolus. The angle formed between the line connecting the greater trochanter with the lateral femoral epicondyle and the line connecting the lateral femoral epicondyle and the lateral malleolus would represent knee flexion. However, this model would be too simplistic in that knee varus or valgus could easily be misread as true knee flexion. To accurately define sagittal motions such as knee flexion, geometry dictates that three

![FIG. 8-11. An example of marker arrangement. Markers are placed on a variety of anatomic landmarks allowing for the collection of three markers or marker equivalents per rigid body segment.](image-url)
markers (or marker equivalents) be placed on each limb segment, assumed to be rigid, to define the three-dimensional coordinate system for that segment (Fig. 8-12). A marker equivalent could be some imaginary anatomic point calculated on the basis of the position of real markers. For instance, three markers could be used to define a plane in the pelvis. From this and the known geometry of the pelvis, the location of the hip joint center can be calculated. The hip joint center then becomes an imaginary marker equivalent and can be used in defining the thigh segment coordinate system. Marker locations are often chosen in order to facilitate estimating joint centers as well to ensure that the markers can be visualized by the camera system.

Typically, markers are placed over bony landmarks to ensure consistent applications as well as to reduce skin movement artifact. With three markers or marker equivalents for each body segment, the segment can be represented in the form of a local coordinate system whose orientation is determined with respect to a global coordinate system. The local coordinate system is defined by three mutually perpendicular vectors. Joint angle information then can be ascer-

tained from the proximal and distal limb segment local coordinate systems. Several methods exist for determining joint angle information. Commonly, one axis is chosen to be parallel to the proximal segment local coordinate system axis, and a second axis is chosen to be parallel to the distal segment local coordinate system axis (38). In this way, a medial/lateral axis is selected from the proximal segment local coordinate system and is considered to be the axis about which joint flexion/extension occurs. A longitudinal axis chosen from the distal segment local coordinate system represents the axis about which internal/external rotation occurs. Finally, an axis formed mutually perpendicular to these two axes is considered the axis about which abduction/adduction occurs (Fig. 8-13).

**Kinetics**

Joint moments and power are commonly measured with quantitative gait analysis. The concept of a joint moment has already been described. A joint power, also referred to previously, represents the net rate of generating or absorbing
energy and is the mathematical product of the joint moment and joint angular velocity. A positive joint power implies that the muscle contraction is concentric because the joint angular velocity and moments are in the same direction. A negative power implies that the muscle contraction is eccentric because angular velocity and joint moments are in opposite directions. Joint kinetics are calculated in part using inverse dynamic techniques according to Newton’s second law of motion, which essentially calculate the joint moments based on the motion and mass characteristics of the limb segments. Although theoretically the kinetics could be calculated from the kinematic data alone, these calculations would be extremely complicated and prone to error. Kinetics are therefore typically calculated using a combination of GRF data along with inverse dynamic techniques. Thus, kinetic calculations are usually based on (a) knowledge of the position of the joint in relationship to the GRF, (b) estimates of body segment masses and moments of inertia, and (c) knowledge of the body segment positions, velocities, and acceleration.

GRFs are measured using force plates that are comprised of piezoelectric or strain-gauge transducers. One or more force plates are imbedded in the ground of the walkway (see Fig. 8-10). As the patient walks, he or she steps on the force plate. To obtain useful GRF data, only one foot must strike the plate without interference from the other foot or an assistive device. Also, to feasibly assess joint kinetics, kinematic measurements must be collected synchronously with force plate data. The locations of the force plates are predetermined within a calibrated volume where the kinematic data are measured. A combination of various measurements taken on the patient are used in conjunction with look-up tables, based on cadaver data, to estimate body segment masses and moments of inertia (4.39). Clinical gait laboratories report joint moments as either external or internal. An external moment refers to the net external load applied to the joint measured via inverse dynamic techniques. The internal moment, which is equal and opposite in sign to the external moment is the presumed moment due to the muscle activity and/or soft tissues to fulfill the requirement that the joint is in equilibrium. For example, an external dorsiflexion moment about the ankle during the stance period of a gait cycle implies that an equal and opposite internal moment provided by the ankle plantarflexors or heel cord is present to maintain joint stability. Similarly, an external flexor moment about the hip during the stance implies that the hip extensors must be active in order to maintain stability. The typical kinetics and kinematics at the hip, knee, and ankle are shown in Figure 8-12. This type of graphic format is typically used for reporting quantitative kinetic and kinematic gait information in the clinical setting.

**Dynamic Electromyography**

Quantitative gait analysis also includes measurements of muscle activity during walking obtained using dynamic EMG measurement. When combined with kinematic and kinetic data, dynamic EMG provides useful information about whether a muscle is firing appropriately and if not, how this nonphasic activity impacts on gait, particularly in patients with spastic parietic gait. Because muscle activity does not linearly relate to the magnitude of force generated, quantifying the amplitude of activity is not practical in patients. However, relative normalization to the peak level activity over the gait cycle or the peak level activity, whether it occurs during strength testing or during walking, improves the clinical usefulness of the EMG data (1).

Muscle activity is measured using either surface electrodes affixed to the skin or fine-wire electrodes inserted in the muscles. Surface electrodes are adequate in studying activity in large superficial muscle groups. In addition to the fact that surface electrodes are less invasive than fine-wire electrodes, a major advantage of surface electrodes is that the data obtained are more easily replicated. This latter advantage is undoubtedly due to the fact that surface electrodes, as compared with fine-wire electrodes, sample data from an inordinately greater number of muscle fibers, representing a far greater number of motor units. Because of this same fact, fine-wire electrodes are not as prone as superficial electrodes to interference or “cross-talk” from nearby muscles. Surface electrodes are commonly used for many large
superficial muscles in the lower extremity. Fine-wire electrodes are necessary for analyzing activity from smaller, deeper muscles, such as the iliopsoas and posterior tibialis. In addition, fine-wire electrodes are useful for differentiating activity from overlapping muscles such as the rectus femoris and vastus intermedius.

Surface EMG is typically recorded using disposable, gelled electrodes attached to the patient’s skin overlying the muscle to be sampled. Usually, bipolar electrodes are used and the signal recorded is the potential difference between the two electrodes. Fine-wire EMG is often recorded using a wire bipolar electrode consisting of two thin, insulated wires with bared tips. The wires are placed through the shaft of a 25-gauge needle with the two ends bent over the needle and the bared tips staggered so as to avoid contact between them. The needle is inserted through the skin into the muscle, and then quickly removed, leaving the fine-wire in place. When in place, the bend in the wires provides a means for the electrodes to “catch” on the muscle fascicles. Again, the signal recorded is the potential difference between the two electrode ends. At the end of the study, the wires are removed with a gentle pull.

Preamplified EMG signals, either from fine-wire electrodes or surface electrodes, can be transmitted by cable or radiowave telemetry to a receiver that is connected to a computer system. The EMG signals are usually filtered to remove artifacts created by the mechanical movement. The signals are displayed and the gait cycle events identified. Some laboratories report raw EMG signals, whereas others report rectified and smooth EMG activity as well. The timing of the activity is typically what is important in the assessment. The normal timings of activity of major muscle groups are summarized in Figure 8-14. Muscle timing errors in patients with UMN pathology traditionally are classified into seven categories: premature onset, delayed onset, curtailed period, prolonged, absent, out of phase, or continuous (40). Although these categorizations are useful in describing activity in each muscle, it is important to note that they do not necessarily imply pathology about that particular muscle. In some instances, muscle activity differs from that of a nondisabled subject because of compensatory actions. As an example, prolongation of quadriceps activity into the mid and terminal stance phases would be compensatory in a patient with an excessive external knee flexor moment. Thus, muscle firing patterns are optimally assessed in conjunction with the kinetics to help dissociate impairment from compensatory action.

**Overall Gait Analysis**

The overall gait laboratory analysis procedures takes approximately 2 hours for data acquisition and an additional 2 hours for analysis and interpretation. The majority of the acquisition time is spent applying and confirming placement of the multiple markers and EMG electrodes. The patient is typically evaluated under several conditions, i.e., barefoot, with shoes, and with and without an orthosis or assistive device.

**EXAMPLES OF EVALUATION APPROACH TO SPECIFIC ATYPICAL GAIT PATTERNS**

**Gait Patterns Associated with UMN Pathology**

A number of atypical gait patterns can be observed in patients with hemiparetic, paraparetic or diplegic impairments affecting their gait, regardless of the underlying UMN pathology (41–45). These atypical patterns include but are

<table>
<thead>
<tr>
<th>Muscle Groups</th>
<th>Percent of Gait Cycle</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>0</td>
</tr>
<tr>
<td>Tibialis Anterior</td>
<td></td>
</tr>
<tr>
<td>Vasti, Long</td>
<td></td>
</tr>
<tr>
<td>Hamstrings, and</td>
<td></td>
</tr>
<tr>
<td>Hip Extensors</td>
<td></td>
</tr>
<tr>
<td>Rectus Femoris</td>
<td></td>
</tr>
<tr>
<td>Ankle</td>
<td></td>
</tr>
<tr>
<td>Plantarflexors</td>
<td></td>
</tr>
<tr>
<td>Hip Flexors</td>
<td></td>
</tr>
</tbody>
</table>

**FIG. 8-14.** General muscle group activity as a percentage of the gait cycle.

- Indicates that a muscle or muscle groups are active
not limited to reduced knee flexion in swing, also referred to as stiff-legged gait, excessive knee flexion in stance referred to as crouched gait, equinus or excessive ankle plantarflexion occurring during one or more phases in either stance and/or swing, and knee hyperextension or recurvatum occurring in one or more phases of stance. Also common are presumably compensatory atypical gait patterns, including hip hiking, circumduction, and steppage gait. The important point to remember is that the causes of each of these atypical gait patterns are not necessarily the same from individual to individual and thus often necessitate a detailed evaluation including quantitative gait analysis.

**Spastic Paretic Stiff-Legged Gait**

Spastic paretic stiff-legged is a classic atypical gait pattern observed in patients with UMN pathology. Stiff-legged gait can be functionally significant from several views. From an energy standpoint, a lack of knee flexion in swing creates a large moment of inertia that significantly increases the energy required to initiate the swing period of the gait cycle. Additionally, associated compensatory actions to clear the stiff limb such as vaulting on the unaffected side and excessive pelvic motion can increase the vertical COM displacement, thereby increasing energy expenditure. From a biomechanical standpoint, these same compensatory actions could place the unaffected knee at risk for posterior capsule damage or the lower back to injury. Finally, lack of knee flexion may cause toe drag during swing, which could increase the risk of falling.

One cause of stiff-legged gait is inappropriate activity in one or more heads of the quadriceps during the pre- and/or initial swing phases of gait (46–49). Reduced knee flexion also may be caused by weak hip flexors, inappropriate hamstring activity, and/or insufficient ankle plantarflexor muscle action (49). For many patients with spastic paretic stiff-legged gait who undergo a quantitative gait analysis, the cause of the stiff-legged gait is not at all obvious from the static or observational gait evaluations. For instance, patients with increased knee extensor tone often can be found to have quiescent quadriceps EMG activity during preswing and initial swing. Conversely, a patient with normal knee extensor tone can have inappropriate activity during these phases in one or more heads of the quadriceps. In the latter case, if the inappropriate activity is limited to just one head, an intra-muscular neurolytic procedure would be a reasonable treatment to improve the gait pattern. On the other hand, quantitative gait analysis may point to dynamically significant weak hip flexors, indicated by slow progression into hip flexion and poor hip power generation in preswing. These findings commonly are not correlated with hip flexion strength evaluated by static testing. In this case, hip flexion strengthening would be the optimal prescription. In another scenario, a reduced external ankle dorsiflexion moment during stance would imply insufficient ankle plantarflexor muscle action, in which case an ankle-foot-orthosis with a dorsiflexion stop might be the most appropriate treatment. A quantitative gait analysis also can help provide information about the functional significance of the atypical gait pattern. For instance, the risk for injury to the posterior capsule and ligaments of the unaffected knee can be assessed by measuring the extensor moment during that limb’s stance period. Finally, a follow-up quantitative gait assessment may be useful in quantifying the improvement in knee flexion as well as ascertaining that the treatment itself did not cause any new problems.

**Dynamic Knee Recurvatum**

Hyperextension of the knee during the stance period, referred to as dynamic knee recurvatum, is a common observation in patients with UMN pathology. This atypical gait pattern may be caused by one or more of the following impairments: quadriceps weakness or spasticity, ankle plantarflexor weakness or spasticity, dorsiflexor weakness, and heel cord contracture (1,50). A primary functional concern for patients with dynamic knee recurvatum is that the hyperextension may produce an abnormal external extensor moment across the knee, placing the capsular and ligamentous structures of the posterior aspect of the knee at risk for injury. Injury to these issues may cause pain, ligamentous laxity, or bony deformity. Not all patients have an abnormal knee external moment, however, in which case the risk for injury is probably less (21). Knee recurvatum is also important from the standpoint of energy expenditure. The lack of knee flexion can cause a greater displacement of the COM because of the lack of knee flexion during the stance period.

Although multiple factors may contribute to knee recurvatum, it is useful to determine the primary cause in each patient so as to prescribe an optimal treatment plan. In some cases, dynamic recurvatum may be advantageous by providing a control mechanism for an otherwise unstable limb during the stance period of the gait cycle. If the associated knee extensor moment is small, then attempts to improve this atypical pattern may not be the appropriate treatment plan. Thus, quantitative gait analysis provides information that can help assess the functional significance of the atypical gait pattern as well as information that can help delineate the pattern’s underlying impairment(s).

**Diplegic-Crouched Gait**

Crouched gait is defined as excessive knee flexion during the stance period of the gait cycle and is most commonly described in diplegic gait specific to cerebral palsy (51–53). Associated gait patterns are adduction and internal rotation at the hips, as well as equinus and forefoot abduction during stance. Reduced knee flexion in swing is also common. Dynamically, hamstring spasticity has been implicated as the principal cause of excessive knee flexion in stance (51–53). However, clinical experience suggests that dynamically tight hip flexors, plantarflexor weakness, and heel cord contracture also may be causative. These potential causes are
best evaluated using the combined information obtained from static evaluation and observational and quantitative gait analysis.

**Equinus Gait**

Excessive ankle plantarflexion or equinus occurring in either stance or swing is common in patients with neurologic lesions. The differential cause for this pattern is inappropriate soleus, gastrocnemius or posterior tibialis activity, heel cord contracture, or weakness of the ankle dorsiflexors. As in the other atypical gait patterns described, the functional significance of the pattern needs to be determined. For instance, excessive plantarflexion during stance may inhibit tibial advancement, thereby interfering with forward progression necessary for efficient ambulation. During swing, excessive plantarflexion may place the patient at increased risk for tripping and falls. Dynamic EMG is useful in identifying the presence of inappropriate soleus, gastrocnemius, or posterior tibialis activity as a cause of the excessive plantarflexion. For equinus in swing, the lack of ankle plantarflexor activity suggests either a heel cord contracture or weak ankle dorsiflexors as a cause. Each patient also should be evaluated for functionally significant compensatory mechanisms as well as increased hip flexion and hip hiking in swing. Finally, it is important to consider the possibility that the excessive ankle plantarflexion itself is a compensatory response for some other impairment or functional limitation such as weakness. This scenario has been reported to occur in muscular dystrophy (54) and is likely to also occur in patients with weakness from UMN. Thus, a reduction in the peak knee flexor moment in a particular patient with excessive ankle plantarflexion during stance may indicate that the ankle plantarflexion is occurring as a compensation for weak knee extensors. Again, because of the complexities of gait, these possibilities are best assessed using quantitative gait analysis including kinetics.

**Gait Patterns Associated with LMN and Orthopedic Disorders**

Unlike in most patients with UMN pathology, the atypical gait patterns associated with specific peripheral nerve injuries cause discrete patterns of muscle weakness and associated characteristic atypical gait patterns. The following examples illustrate atypical gait patterns that arise from weakness of one specific functional muscle group. Unlike in patients with UMN pathology, in order to determine the underlying impairment and functional limitation responsible for the atypical gait pattern, static evaluation and observational analysis are usually adequate. Kinetic assessment is often useful, however, in helping to determine the functional significance of an atypical gait pattern.

**Gait Associated with Femoral Neuropathy**

Selected quadriceps weakness, which can occur in femoral neuropathy in diabetes, femoral nerve entrapment, or polio-

myelitis, impairs weight-bearing stability during stance. The quadriceps eccentrically contract to control the rate of knee flexion during the loading response of the limb. With weakness, the knee would tend to “buckle.” The effective compensatory action is to position the lower extremity such that the GRF lies anterior to the knee joint, imparting an extension moment during stance phases. This is first achieved during initial contact by plantarflexing the ankle. Contraction of the hip extensors also can help to hold the knee in hyperextension. As noted previously, quantitative gait analysis may be useful in evaluating the associated knee extensor moment, which, if excessive, could place the posterior capsule and ligamentous structures at risk for injury.

**Atypical Gait Patterns Associated with Weak Ankle Dorsiflexion**

Dorsiflexion weakness also has a characteristic gait pattern. Clinical conditions in which this is seen is peroneal nerve palsy occurring as a result of entrapment at the fibular head or more proximally as an injury to a branch of the sciatic nerve, or in an L5 radiculopathy. If the ankle dorsiflexors have a grade of 3 or 4/5, the characteristic clinical sign is “foot slap” occurring soon after initial contact, due to the inability of the ankle dorsiflexors to eccentrically control the rate of plantarflexion after normal heel contact. If the ankle dorsiflexors have less than 3/5 strength, toe drag and/or a steppage gait pattern with excessive hip flexion in swing is likely. The cause of these patterns can usually be determined with a careful history, physical examination, and standard electrodiagnostic procedures (as opposed to a dynamic EMG assessment).

**Atypical Gait Patterns Associated with Generalized LMN Lesions**

More generalized LMN lesions commonly involve variable weakness patterns and thus often have unpredictable and often complex associated gait patterns. Poliomyelitis and Guillain-Barré syndrome are examples. For these diagnoses, kinetic assessment can be particularly useful in determining excessive joint moments, implying excessive soft-tissue strain or the need for increased compensatory muscle action in another muscle group.

**Trendelenburg Gait**

Trendelenburg gait (gluteus medius gait), describes a pattern of either excessive pelvic obliquity during the stance period of the affected side (so-called uncompensated Trendelenburg gait) and/or excessive lateral truncal lean during the stance period of the affected side (so-called compensated Trendelenburg gait). Weakness or reluctance to use the gluteus medius can cause this atypical gait pattern. The most common cause of Trendelenburg gait is osteoarthritis of the hip. In this case, the gait pattern (regardless of whether it is
compensated or uncompensated) occurs as a compensatory response to reduce the overall forces across the hip during stance. This can be seen as a reduction in the external hip adductor moment, which ordinarily occurs in the stance period.

**Atypical Gait Patterns Associated with Orthopedic Conditions**

In cases of specific orthopedic conditions, the atypical gait pattern is fairly predictable and the cause directly relates to the structural abnormality. For instance, the cause of absent knee flexion during gait may simply be the result of a knee fusion. Studies about the diagnostic use of quantitative gait analysis in structural abnormalities are scant. Nevertheless, quantitative analysis may be useful in evaluating complex orthopedic conditions involving multiple joints and in evaluating the functional relevance of associated gait patterns. For instance, one study demonstrated that kinematic and kinetic measurements were helpful in documenting the effects of gait training in patients with symptomatic knee hyperextension due to postero-lateral ligament complex injury (55). Other studies have reported the usefulness of quantitative gait assessments to identify abnormal joint forces in patients with anterior cruciate ligament–deficient knees and osteoarthritis of the knee, which may help to identify patients with risk of further deterioration (56,57).

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