Chapter E-1

Gait: Development and Analysis

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**SUMMARY**

One of the first questions asked by many parents whose child has recently been diagnosed as having a motor impairment or whose child has experienced an injury that affects movement is: "Will my child walk?" or "Will my child walk again?" The reason for emphasis on the task of walking apart from many other abilities (which are sometimes more functional), I believe, is the measure of independence and social acceptance with which the task of walking is associated. The ultimate goal for families, and sometimes pediatric therapists as well, becomes independent walking for the child. Although the variables that contribute to this important ability are numerous and complex, *a child is usually able to stand and walk by 8 to 17 months of age* (average 12 months; see Chapter 2). Normative data from Canadian infants on the Alberta Infant Motor Scale (AIMS) indicate that 50% of infants will achieve independent standing by the age of...
10.5 months, take first steps by 11 months, and walk independently by the age of 11.5 months. Furthermore, the pattern is mature in typically developing children by the age of 3.5 years, all without instruction. The complexity of the task has been studied extensively since the days of Saunders and Murray and study continues today. Improved computerized techniques and new theories of motor control and balance contribute to better understanding of the components of walking. Understanding how we walk was once required only for those in the fields of child development and rehabilitation. Today, walking is studied by many disciplines, including engineering. The extent and duration of the research in the area is a statement of its continued importance.

The purpose of this chapter is to provide an overview of the aspects of gait important to physical therapists providing intervention for children. These include the development of gait and its refinement in childhood, the typical components of walking as identified by gait analysis, and a description of some of the common gait deviations found in children with physical disabilities. Guidelines for when a computerized gait analysis is desirable will also be addressed. Those interested in more extensive information are referred to a number of excellent resources on the topic.

DEVELOPMENT OF GAIT
Gage suggests that typical walking has five major attributes: (1) stability in stance, (2) sufficient foot clearance in swing, (3) appropriate prepositioning of the foot for initial contact, (4) adequate step length, and (5) energy conservation. When independent ambulation begins in the toddler, only the prerequisites for these attributes are present. The prerequisites, however, are likely to be more important to the attainment of walking than the attributes themselves. The attributes develop over time with typical growth, maturation, and refinement of the skill. Scales of motor development illustrate that even during the time period between first steps and independent walking important balance abilities are achieved. This includes rudimentary anticipatory postural adjustments for gait initiation. Loss of the five attributes occurs in atypical gait because of a loss or failure to achieve the prerequisites. The prerequisites related to the development and skill of locomotion have already been suggested in Chapter 2 during the review of the eight subsystems proposed on the basis of the dynamic systems theory. These include adequate motor control and central nervous system (CNS) maturation (implying an intact neurologic system), adequate range of motion (ROM), strength, appropriate bone structure and composition, and intact sensation (proprioception). More recent research suggests that muscle activation patterns for pelvis stabilization may also be a prerequisite. As suggested by Campbell (Chapter 2), the development of a motor pattern (in this case walking) "depends on a combination of mechanical [including structural], neurologic, cognitive, and perceptual factors." When other factors are controlled, chronologic age is less significant than previously thought. Changes in body dimensions and speed, however, are important. A brief description of the neurologic and biomechanical factors that contribute to the development and refinement of walking follows.

Neurologic Factors
Campbell (Chapter 2) and Westcott (Chapter 3) have discussed the primary neurologic factors related to walking and other motor tasks. To briefly review, the basic neural organization and function used to execute locomotion is hypothesized to be controlled by a central pattern generator (CPG) located in either the spinal cord or in the brainstem. Modulation of the
CPG depends on both descending neural input and peripheral input, which modify the output to adapt the execution to stability requirements and the demands of the specific task and environment. Most CPG research involves animals (including mammals and other vertebrates). CPGs are thought to exist for many movements, and the CPG for walking is believed to organize the activation and firing sequence of muscles during gait. The presence of stepping movements in the human fetus, the early onset of infant stepping in premature infants, the parallels between developing animals and humans, and the parallels between the sequencing and timing of infant stepping, kicking, and locomotion in infants with and without disabilities have led individuals to suggest that the neural foundations for locomotion in humans are established during embryogenesis as in other animals. Postnatally, the major periods of accelerated brain growth in relation to body growth occur between 3 and 10 months of age, and the myelination that develops during this time probably contributes to the neural organization required for independent locomotion. Sufficient maturation of information processing capabilities of the CNS for both internal and external stimuli is also believed to play a role.

Biomechanical Factors

Adequate ROM, strength, appropriate bone structure and composition, and body composition also affect the emergence of locomotion and its refinement. These variables have significant ramifications as mechanical factors in the development of walking. ROM, strength, bony structure, and the ability to manage gravitational and inertial forces of the lower extremities affect the early patterns of walking. In the presence of typical motor control and maturation, constraints in any of these mechanical variables change patterns; as the constraints change, so do the movement and the muscle activity involved in motor control. Typical growth is periodic and often occurs in spurts. Changes seen in the development of many skills may be the result of critical dimension changes in the body of the growing child, and adaptation to those changes. In 1984, Thelen and colleagues described how simple physical growth could be an explanation for the disappearance of the stepping reflex. This study was republished in 2002.

More recent work also corroborates the importance of mechanical factors, suggesting that infants often have to physically grow into the body dimensions needed for optimal functioning and develop adequate nerve conduction time for activation of CPGs. An extensive review of musculoskeletal development and adaptation can be found in Chapter 5.

Determinants of Walking

The seminal work of Sutherland and colleagues identified five important determinants of mature walking. The determinants distinguish walking that is considered "mature" (age 3 years and older) from walking that is "immature" (age 2.5 years and younger). The most important of the 13 variables analyzed were the duration of single-limb stance, walking velocity, cadence, step length, and the ratio of pelvic span to ankle spread. The variable with the least discriminating power was the presence or absence of a heel strike at initial contact.

The duration of single-limb stance is, as the name implies, the length of time during which only one foot is on the ground during stance phase. As a child's walking pattern matures the duration of single-limb stance increases, implying a measure of increasing stability and increasing control. The most rapid change occurs between ages 1.5 and 3.5 years.

The remaining variables (with the exception of pelvic span to ankle spread) are temporal distance parameters that are influenced by growth. A linear relationship between step length and leg length is present from age 1 to age 7. A linear relationship between step length and age exists.
from age 1 to age 4. Walking velocity also increases with age up to age 7, although the rate of change decreases from age 4 to age 7. Cadence gradually decreases with age throughout childhood. The most rapid reduction in cadence occurs between the ages of 1 and 2 years. Because of this strong correlation among age, leg length, and stature, methods of normalization have been explored and endorsed to better understand their influence on these and other temporal distance parameters. Using a geometric normalization technique on Sutherland's data, it was demonstrated that preferred stride length and cadence increase and decrease, respectively, up to age 3.5 and then remain essentially constant into adulthood. The variability of individual step length from step-to-step in infants has brought into question the practice of averaging step lengths in this age group as well, implying that averaging step lengths across strides minimizes balance control and strength present in early walkers.

Refinement of Gait by Age

Birth to Age 9 Months

The disproportionate contribution of fat content to overall increases in body mass over the first 8 months of postnatal life causes infants to be relatively weak during a time when they are developing the motor control and coordination skills to progress toward independent ambulation. Studies suggest that bigger, fatter infants achieve locomotor milestones later than their smaller peers for this reason. From birth to 6 months of age, the body fat of the infant rises from 12% to 25% of body mass. Adipose tissue is a major component of weight gain during the first 4 months of postnatal life, with increases in lipids accounting for more than 40% of total weight gain during this time. Between 4 and 12 months of age increased lipid content accounts for approximately 20% of increased weight. With increasing age and mobility, fat content drops and muscle mass increases.

Not only mass but also body proportions change as infants grow. During the first few months of life, the fastest rate of growth occurs in the extremities as opposed to the head and trunk. Leg length increases 130% from birth to 2 years, but crown-to-rump length increases only 69% in the same period. Inertial characteristics of body segments change with growth and the effects of gravity. It has been demonstrated that the resistance to motion in limb segments (segment moments of inertia) more than doubles and triples during the first 6 months of life. The rate of change in gravitational moments appears to decrease relative to the rate of change in the developing extensor muscle moment in a cephalocaudal progression, which also influences the ability to move against gravity. The body structure of infants as they develop the ability to stand upright with support and to cruise independently affects their posture and movement patterns. Flexion "contractures" are present at the hips, and range of hip external rotation is slightly greater than range of internal rotation. Range of hip abduction is slightly increased at 8 to 9 months of age but has been slowly decreasing since birth. Femoral anteversion* (a forward or anterior orientation of the head and neck with respect to the frontal plane) and femoral antetorsion (a twist of the bone in its longitudinal axis) of the hips are both present. The magnitude of the femoral torsion is 40 to 50 degrees.

*Femoral anteversion and femoral antetorsion are often used as synonymous terms in orthopedic literature to refer to the torsion or the medial twist between the axis of the head and neck of the femur and the axis of the femoral condyles.
Structurally, the knees in the frontal plane exhibit genu varum, or bowing in the tibiofemoral angle (Figure E-1-1). The tibia and fibula exhibit neutral alignment about the longitudinal axis, which represents slight internal torsion relative to adult values. A medial inclination of the talotibial articulation is present in the infant, producing an everted talocrural mortise. The medial inclination of the joint is manifested in an everted heel position in weight bearing. Supported walking at this age is characterized by wide abduction, external rotation and flexion at the hips, bowed legs, and an everted heel position.

Figure E-1-1. An 8-month-old infant with physiologic varus in the tibiofemoral angle.

The postural control, development of antigravity muscle strength, and control of gravitational moments that are gained over the first 9 months of life are important precursors to the development of independent ambulation. Standing balance is also a necessary precursor to upright motor skill. The frequency and the amount of practice of activities, such as kicking, have been shown to affect the age of onset of ambulation, in infants with and without disabilities. Antigravity strength of the hip flexors in the lower extremities is built early on in the developmental process by kicking from the supine position.

Hip extensor strength in both concentric and eccentric types of muscle contractions similar to those used in ambulation begins from activities in the prone position but gradually builds as the infant begins creeping and kneeling activities. Bly describes infants of 8 months of age exhibiting the ability to rise from a kneeling to standing position, which requires closed chain hip and knee extension. Control of gravitational moments at the hip by the hip flexor and extensor muscles has typically occurred by 8 to 9 months of age. Cruising along furniture builds strength of the hip abductors.

By 8 months of age, the visual, proprioceptive, and vestibular systems work together to consistently bring the center of mass back to a stable position after perturbation in a seated position. Needed postural corrections in response to visual flow for a particular skill often predate the development of that skill. For example, Butterworth and Hicks demonstrated that infants who could sit alone but could not stand independently exhibited the same appropriate postural adjustments to visual information that simulated movement as did infants who were able to walk independently.

Age 9 to 15 Months

Lower extremity alignment and body structure at the onset of ambulation are characterized by a standing posture with a wide base of support and the hips positioned in abduction, flexion, and slight external rotation. The tibias display mild internal torsion, and varus is still present in the
tibiofemoral angle; the heel position in weight bearing remains everted because of the inclination of the mortise joint. The child's center of mass is proportionately closer to the head and upper trunk (at the lower thoracic level) than in an older child, whose center of mass is located at midlumbar level, or in an adult, whose center of mass is located at the sacral level. Although differential growth rates of the extremities are allowing the head to become relatively smaller than the rest of the body, the head is still proportionately large. The ratio of body fat to muscle mass is still high, contributing to weakness relative to the demands of upright posture. Coming upright against the force of gravity puts new demands on muscle strength. Muscles (particularly the abdominals, hip flexors, knee extensors, and ankle dorsiflexors) must work in new antigravity positions, which further increases functional weakness. Despite the structural limitations and the typical inability to walk at constant average forward or lateral speed, infants exhibit the necessary postural responses to compensate for visual and support-surface perturbations that are inherent in the task of upright locomotion.

The rate-limiting factors associated with the ability to demonstrate upright locomotion are (1) sufficient extensor muscle strength to support the body's weight on a single-limb base of support, (2) dynamic balance, and (3) postural control in the form of anticipatory and integrative postural adjustments. The critical dimension of the body to be controlled is the head within the limits of stability of the base of support. The base of support in an infant at the onset of ambulation is wide for both structural and stability reasons. Mediolateral (side-to-side) stability is achieved, but anteroposterior stability is limited. Progression takes place in the sagittal plane; if the head moves outside the limits of stability at the base of support, balance is lost. Recent work has demonstrated that even though forward progression takes place, infants do not walk with either constant forward or lateral speeds, step lengths, or other joint parameters.

Initial anticipatory postural adjustments used for gait initiation during the postural phase (mainly the lateral shift of body weight toward the stance limb) are present at the onset of walking, but those required during the locomotor phase (control of the swing side pelvic drop) are not. Because the hip is the interface between the locomotor movements of the legs and head-arms-trunk postural control, strength and control at the hip is very important. Studies have shown that at the onset of ambulation, hip strength is inadequate to control the gravitational forces during gait and maintain balance. Infants actually "walk by falling," but they also learn adaptive strategies from the falling experience.

If balance is perturbed, control is regained by rudimentary mechanisms requiring torque adjustments at multiple joints.

The pattern displayed by a beginning walker is similar to that of an experienced walker on a slippery surface (i.e., small steps, a widened base of support, and maintenance of the body and limbs very upright in an extended, stiff position). Postural adjustments are made using movements of the entire body. The work of Sutherland and colleagues characterizes infant ambulation as consisting of a wide base of support, increased hip and knee flexion, full foot initial contact in plantar flexion, a short stride, increased cadence, and a relative footdrop in swing phase.

Motorically, the initial pattern and execution of the first steps are thought to be related to the patterns used in stepping and kicking during early infancy and may be constrained by this pattern. That is, just as the structural makeup of the body drives the initial posture, so may the primitive generator for kicking and stepping drive the muscle activity for early walking. The reduced frequency of kicking in children with Down syndrome has been shown to be correlated with a later age of walking than in typically developing infants. Both kicking and stepping are alternating reciprocal patterns of movement between the limbs. Each has a flexion phase (in gait,
analogous to swing) and an extension phase (in gait, analogous to stance). Thelen and colleagues\textsuperscript{121} suggest that the ability to generate steplike patterns is continuous from the newborn period through independent locomotion and that the pattern demonstrated in beginning upright ambulation is a modified, more flexible version of the earlier pattern that has been modified by changes in strength, neurologic maturation, and the mechanics of upright posture.

The electromyographic (EMG) patterns of activity at the onset of independent ambulation demonstrate a combination of significant co-contraction across antagonistic muscle groups—the anterior tibialis and gastrocnemius during swing phase and the quadriceps and hamstrings muscles during stance phase. Reciprocal patterns in single stance also appear in conjunction with the co-contraction of the antagonistic muscle groups as a feature of the redundancy and variability within the motor system.\textsuperscript{27,71,72,112,113,117} Coactivation patterns result from the need for stability and decreases are noted with as little as one month of walking experience. After 3 months of independent walking, further refinement of excessive co-contraction is noted. Stable patterns emerge via exploration, selection, and practice. With practice and time, the high levels of co-contraction decrease and reciprocal patterns increase.

As stated previously, Thelen and colleagues suggest that it is not the pattern-generating capabilities or motor control postural abilities that constrain the onset of independent locomotion. Development of sufficient extensor strength is believed to be the critical variable.\textsuperscript{121} This belief is consistent with other views for the requirements of walking, including stance phase stability,\textsuperscript{41} the need to maintain a net extensor muscle support moment as described by Winter,\textsuperscript{131,132} and the inability of the hip to completely control the gravitational forces for balance.\textsuperscript{21}

**Age 18 to 24 Months**

As the child grows, body structure changes, increases in strength, neurologic maturation, and walking experience all play a part in altering the walking pattern. By 18 months of age, the varus angulation of the tibiofemoral angle in the frontal plane has resolved and the limb is straight (Figure E-1-2)\textsuperscript{115} (see also Chapter 5). No change is noted in excessive femoral antetorsion, although limitation in hip extension ROM is reduced to an average of 4 degrees, indicating that remodeling is under way.\textsuperscript{27,80} Range of hip abduction is no longer excessive. Because of decreased abduction and improved stability, the base of support has decreased. Dynamic balance and strength have also improved. A heel strike has not consistently emerged in the 18-month-old child,\textsuperscript{112,113} but it has been reported with kinematics.\textsuperscript{50} The lessened base of support allows for more anterior-posterior movement over the planted foot. When this occurs a maturation of the ankle plantar flexion moment also occurs.\textsuperscript{50} Heel position remains everted.\textsuperscript{86,124} The viscoelastic and inertial properties of the stretched stance limb begin to be exploited to propel the leg in swing.\textsuperscript{117} A knee flexion wave begins to emerge during initial stance phase as a heel strike develops and knee extensor contraction absorbs some of the impact of floor contact.\textsuperscript{50,112,113} Knee extension in the middle of stance phase also emerges. The duration of stance phase remains prolonged and cadence is increased relative to mature gait but is significantly faster than that observed in 12-month-old infants.\textsuperscript{60}
The efficiency of locomotion slowly improves during the period from 18 to 24 months of age as the center of mass descends from a position high above the lower extremities to one in close proximity to the chief motor power in the legs. Stability of any body is inversely related to the distance of its center of mass from its base of support. Between the first and second years of life, the legs are growing proportionately longer, becoming the most rapidly increasing dimension of the body. These events bring the center of mass closer to the proximal end of the lower limbs. This, in conjunction with the repetition and practice of walking experience, results in a reduction of early high levels of muscle co-contraction and energy recovery as quickly as 3 to 5 months after walking begins.\textsuperscript{27,60}

Controversy exists over whether heel strike develops as a result of neurologic maturation or gradual changes in body structure, base of support, improved strength, and improved stability.\textsuperscript{39,40,117,121} It is likely that each variable is important. \textbf{A consistent heel strike develops by 24 months of age.}\textsuperscript{112,113} Requirements include refined motor control, strength, and dynamic balance to sustain stability on a small area of contact (the heel). Children with impaired walking ability lack a heel strike at initial contact. This may be caused by either lack of motor control or the inherent choice of maintaining stability by use of a larger area of initial contact.\textsuperscript{41}

With 6 to 12 months of walking experience new strategies are employed for balance and postural control. Walking experience also plays a role. Decreases in peak head and trunk oscillations are noted during the first weeks and months of independent walking.\textsuperscript{67} A backward inclination of the trunk is noted at this age rather than the anticipatory forward inclination noted in older children and adults.\textsuperscript{6} Velocity normalized for height is increasing during this time. The deficit between muscular strength at the hip and vertical acceleration of the center of mass diminishes, which improves the control of the gravitational forces and the postural capacity of the musculoskeletal system as a whole.\textsuperscript{21} As a result, single limb stance is more stable.

The role of walking experience cannot be overestimated. Adolph and colleagues have shown that after controlling for body dimensions and testing age, walking experience explained an additional 19\% to 26\% of variance in improvement in walking skill in a group of 210 infants of differing ages. Previous diary studies have demonstrated that infants may take as many as 9000 steps in a given day and travel the equivalent of 29 football fields.\textsuperscript{24} Both walking experience and the experience of falling assist in the development of prospective control to adapt locomotion to threats to balance such as uneven terrain or obstacles.\textsuperscript{58}

The EMG patterns at this age show decreasing co-contraction in antagonistic muscle groups, implying increased control and stability. The primary changes occur in the duration of stance.
phase activity. The durations of stance phase quadriceps, medial hamstring, and anterior tibialis muscle activity are all decreased in the 18- to 24-month-old as compared with those seen in the 12-month-old. By age 2, the late swing phase–early stance phase EMG activity monitored in the gastrocnemius-soleus complex of the 12- to 18-month-old child has disappeared.

**Age 3 to 3.5 Years**

Between the ages of 3 and 3.5 years, the joint angles associated with walking mature into the adult pattern, but the joint torques and propulsion patterns remain immature. Structurally, the tibiofemoral angle, which was neutral at 18 months of age, now shows maximum valgus alignment (Figure E-1.3). Femoral antetorsion of the hip is decreasing but remains increased in relation to that measured in an adult. The center of mass is closer to the extremities as the rate of lower extremity growth stabilizes. Heel eversion in weight bearing can still be observed but is decreasing. Measurement by motion analysis demonstrates that a heel strike is consistently present in conjunction with a knee flexion wave in early stance. EMG activity has a mature pattern by this age.

**Figure E-1.3.** The tibiofemoral alignment of a 3-year-old showing maximum physiologic valgus.

Balance mechanisms continue to be refined during this period. Children under 4 years of age do not demonstrate integrated postural development and often "collide" with the boundaries of their base of support at gait initiation and during other perturbations. Neither the visual system nor the vestibular systems are mature at this age and are thought to play a role. Torque profiles of perturbation responses and calculation of external work demonstrate that children in this age range continue to exhibit an immature pattern, but the pattern is clearly different from children with less walking experience. The vertical acceleration of the center of mass at foot contact also demonstrates a deficit in the capacity of the stance leg muscles to control balance. However, walking velocity normalized for height is consistent with that of adults.

**Age 6 to 7 Years**

By age 7, the gait patterns by standards of movement or motion are fully mature. Minimal changes are noted when compared with the adult pattern. Dimensionless time and distance parameters are fairly constant. Joint torque and propulsion patterns are also similar to the adult pattern with the exception of the ankle joint where lesser values were noted. Speed-related changes in joint motion and joint torques are noted. Balance and postural control demonstrate renewed stability after a period of disequilibrium often seen between ages 4 and 6 years. Where previous research suggested that postural control reached maturity by age 7, with
the visual, vestibular, and proprioceptive systems becoming more efficiently coordinated.\textsuperscript{21,85,101,134} Recent research suggests maturity of these systems is not present until 10-12 years of age.\textsuperscript{45,79} Study of a cycling task suggests that intermuscular coordination rather than muscular power limits movement speed and that children younger than 10 years of age do not show an adult pattern.\textsuperscript{63} Structurally, the tibiofemoral angle has returned to neutral,\textsuperscript{115} and femoral antetorsion is largely resolved but still slightly higher than that measured in the adult.\textsuperscript{15,37} The inclination of the talotibial joint is no longer present, and heel position is neutral by age 7.\textsuperscript{124} A period of disproportionate growth with respect to body dimensions has also passed. The center of mass is still slightly higher than in the adult, at the level of the third lumbar vertebra.\textsuperscript{76}

COMPONENTS OF TYPICAL GAIT AS MEASURED BY GAIT ANALYSIS

As stated previously, extensive research has been done and continues to be done in the area of gait. Numerous textbooks and basic research articles have been written on the topic.\textsuperscript{41-43,55,75,78,89,90,99,96,107,111,112,113,129-132} This section describes the typical components of mature walking as identified by three-dimensional computerized analysis of movement and forces, electromyography, and energy expenditure in a functional context.

One complete gait cycle refers to a single stride that begins when one foot strikes the ground and ends when the same foot strikes the ground again. The gait cycle is divided into two major phases—stance and swing. Stance phase is associated with the period of time when the foot is on the ground; swing phase is the period of time when the foot is in the air. The stance phase of the gait cycle occupies approximately 60\% of the cycle, and the swing phase occupies approximately 40\%.

Perry\textsuperscript{78} developed a generic terminology for the functional phases of gait that further divided the gait cycle into eight subphases. Each subphase has a functional objective that assists in the accomplishment of one of three basic tasks of the walking cycle: weight acceptance, single-limb support, and limb advancement.\textsuperscript{84} The basic tasks that Perry’s group described are similar to the "attributes" discussed by Gage\textsuperscript{41}: weight acceptance and single-limb support (stability in stance and pre-positioning of the foot for initial contact) and limb advancement (foot clearance in swing, prepositioning of the foot for initial contact, and adequate step length).

Stance phase is divided into five subphases or instantaneous events (Figure E-1-4):

1. Initial contact (0\%–2\% of the cycle)
2. Loading response (0\%–10\% of the cycle)
3. Midstance (10\%–30\% of the cycle)
4. Terminal stance (30\%–50\% of the cycle)
5. Preswing (50\%–60\% of the cycle)
Figure E-1-4. One complete gait cycle depicting stance phase and swing phase. The cycle begins with initial contact of the right foot. Stance phase has four subphases after initial contact; swing phase has three subphases. Initial contact and toe-off are instantaneous events. (Adapted from Gage JR. An overview of normal walking. In Greene WB, ed. Instructional course lectures, volume 39. Park Ridge, IL: American Academy of Orthopaedic Surgeons, 1990.)

Opposite leg toe-off and opposite initial contact occur at 10% and 50% of the gait cycle, respectively. Thus there are two periods of double support during the walking cycle when both feet are on the ground. These occur during loading response (just after initial contact) and preswing (just before toe-off). Each occupies approximately 10% of the gait cycle.

Swing phase begins at toe-off and occurs during the period of single support of the stance limb. Three subphases are identified (see Figure E-1-4):

1. Initial swing (60%–73% of the cycle)
2. Midswing (73%–87% of the cycle)
3. Terminal swing (87%–100% of the cycle)

**Temporal Measurement Definitions and Common Terms**

The following definitions will be helpful in the description of walking as used in gait analysis:

**Cadence:** The frequency of steps taken in a given amount of time, usually measured in steps per minute.

**Concentric muscle contraction:** A shortening contraction that produces acceleration. Positive work results and power generation occurs.

**Eccentric muscle contraction:** A lengthening contraction that produces deceleration. Negative work results and power absorption occurs. The efficiency of negative work or power absorption by a muscle is three to nine times higher than that of positive work.\(^5^5\)
**External load:** Ground reaction forces, inertial forces, and gravitational forces that affect joint motion.

**Isometric muscle contraction:** A stabilizing contraction that produces no net power. Force is produced without length change in the muscle.

**Joint moment:** A force acting at a distance from an axis causing a rotation about that axis is called a *torque* or *moment* (moment equals force times perpendicular distance). In the body, moments are produced external to the body by ground reaction forces, forces related to gravity, and inertial forces. Moments internal to the body are produced by muscle forces, ligamentous forces, or forces produced by joint capsules. Joint moments in this chapter will represent the physiologic response of the body generated in response to an external load or the net internal moments, in accordance with the definition used by Ounpuu and colleagues.\(^{75}\)

**Joint power:** The net rate of energy absorption or generation. Mechanical power is defined as the work performed per unit of time. Joint power is defined as the product of the net joint moment and the joint angular velocity. Muscles are the primary internal power producers in the body. Muscles can also be internal power absorbers. Ligaments usually absorb power.

**Kinematics:** The parameters used to describe motion without regard to forces. These include linear or angular displacements, velocities, and accelerations.

**Kinetics:** The parameters that describe the causes of the movement. These include external and internal forces such as gravitational forces, ground reaction forces, inertial forces, muscle or ligament forces, joint moments, and joint powers.

**Step length:** The longitudinal distance between the two feet. Right step length is measured from the first point of contact of the left foot to the first point of contact of the right foot.

**Stride length:** The longitudinal distance between the initial contact of one foot and the next initial contact of the same foot. It is the sum of the right and left step lengths and represents the distance traversed in one complete gait cycle.

**Walking velocity:** The rate of walking or the distance traversed in a specified length of time. Velocity can be expressed as stride length divided by cycle time or the product of step length and cadence. Because of the correlation with leg length and therefore age, methods of normalization are used to evaluate comparisons of different velocities.

**Kinematics**

Kinematics in gait analysis can be collected in either two or three dimensions. Common to all types of computerized motion analysis is a reference system such as the use of external markers or targets placed on the body and aligned with respect to specific anatomic landmarks. Two-dimensional motion systems provide joint angles that are a direct measure of the motion of the marker set placed on the skin. Three-dimensional systems reference the marker coordinate system to an internal coordinate system based on an estimation of the locations of the anatomic joint centers. The outputs, regardless of two- or three-dimensional technique, are typically displayed as a series of graphs of a single gait cycle for each joint in a given plane of motion. Figure E-1-5 gives an example of three-dimensional motion data.
**Figure E-1-5.** Representative typical three-dimensional kinematic data from the James R. Gage Center for Gait and Motion Analysis of Gillette Children’s Specialty Healthcare. In this figure, one complete gait cycle is depicted and is normalized to 100% of the stride. Degrees of motion are on the ordinate. The gray band represents the mean and plus or minus one standard deviation of composite data collected from children ages 4 to 17 years. Stance phase is separated from swing phase in each graph by the vertical bar. Each graph represents the same walking cycle. Each row displays a different joint: From top to bottom are trunk, pelvis, hip, knee, and ankle. Each column displays the movements of a different joint in the same plane of motion; from left to right are the coronal plane (front view), sagittal plane (side view), and transverse plane.
(rotation view). Note: The pelvis is measured with respect to laboratory coordinates. The trunk and hip are measured with respect to the pelvis, the knee with respect to the thigh, and the sagittal plane ankle with respect to the shank. Foot rotation graph represents foot progression angle with respect to laboratory coordinates, not rotation of the foot with respect to the tibia.

**Kinetics**

Kinetics represent the parameters that describe the causes of the movement. These include external and internal forces such as gravitational forces, ground reaction forces, inertial forces, muscle or ligament forces, joint moments, and joint powers. Kinetic data as part of two- or three-dimensional gait analysis are obtained from a combination of force plate and kinematic information. They are displayed as joint moments and joint powers. The force plate provides information regarding the ground reaction force, and the kinematics provide information regarding the joint angular velocities. Anthropometric measurements of the body are also required to locate the joint centers. The method commonly used to calculate joint moments and powers is called *inverse dynamics* and is based on a linked segment model approach. A ground reaction force method is another method sometimes used.

A moment or torque is a force acting at a distance that causes an object to rotate. A joint moment represents the physiologic response of the body generated in response to an external load. At Gillette Children’s, what is displayed on the graphs is the net joint moment, which represents the sum of all internal joint moments in a particular plane at a particular joint. The moments refer primarily to the muscle forces that are acting to control segment rotation, but internal joint moments can be generated by ligaments, joint capsules, and fascia as well. The net joint moment depicts which muscle group is dominant but does not denote the relative contributions of muscle groups on either side of the joint. For example, in the sagittal plane, a net extensor moment at the hip during stance phase implies that the hip extensors are dominant. Hip flexors may or may not contribute, but the overall moment is an extensor moment. The hip extensor muscles are active to counteract an external moment created by the ground reaction force that is tending to flex the hip (Figure E-1-6). The ground reaction force tends to flex the hip because it is anterior to the joint center.

![Figure E-1-6](image-url) An example of an internal joint moment produced at the right hip. Because the ground reaction force falls anterior to the hip joint, the joint would flex without internally produced resistance. The joint moment graph demonstrates a net extensor moment at the hip (shaded) as the body's internal resistance to the external force. The internal moment is produced by dominant action of the hip extensor muscles. (Adapted in part from Gage JR. Gait analysis in cerebral palsy. London: Mac Keith Press, 1991.)
Power, in mechanical terms, is defined as the rate of doing work. Joint power is defined as the product of the net joint moment and the joint angular velocity. Muscles are the primary internal power producers in the body. A muscle's ability to produce power is affected by its cross-sectional area. Other factors that affect power include muscle fiber type, the length-tension ratio, and the degree of fatigue (see Chapter 5). Muscles can also be internal power absorbers. Ligaments usually absorb power. The power graphs display whether power is generated (positive work) or absorbed (negative work). Concentric muscle action is associated with power generation, and eccentric muscle action is associated with power absorption. In the previous example, the hip is extending in the presence of a net extensor moment; therefore, power generation occurs (Figure E-1-7). Typical sagittal plane kinematics and kinetics can be found in Figure E-1-8.

**Figure E-1-7.** An example of power generation at the right hip. Because the hip is extending and a net extensor moment is present, power generation (*shaded*) occurs. The units for power are watts per kilogram. By convention, power generation is represented by positive deflection on the graph; power absorption is negative. (Adapted in part from Gage, JR. Gait Analysis in Cerebral Palsy. London: Mac Keith Press, 1991.)
Figure E-1-8. Sagittal plane kinematics and kinetics of the hip (A), knee (B), and ankle (C). Each graph represents the same walking cycle, with the mean (solid line) and one standard deviation (dashed line). The top row shows the kinematic graphs; the middle row, the joint moments; and the bottom row, the joint powers. Units for the kinematics are degrees. The units for joint moments (newton-meters/kilogram) and powers (watts/kilogram) are standardized for body weight.
Electromyography
The electrical signal associated with the neuromuscular activation of a muscle is measured by electromyography.\textsuperscript{11,132} It represents the pattern of motor unit activation. Electromyographic data can provide information about the timing of muscle activity and, in some cases, about the intensity of muscle contraction. Under some conditions, EMG amplitude has been shown to be related to force,\textsuperscript{61,126} but the usefulness of this aspect for assessment of walking is limited because the relationship is valid only under isometric conditions and when no coactivation is occurring.

The amplitude of an EMG signal is affected by the rate of motor unit firing and the number of motor units active at any given time. The type of motor units firing and the proportion of different motor units firing also affect amplitude. Walking speed has been shown to influence both the timing of the muscle firing and amplitude.\textsuperscript{99} In addition, many external factors, including electrode location, type of electrode used, interelectrode distance, skin temperature, and amount of subcutaneous fat, also affect the amplitude of the signal.\textsuperscript{11} Comparing EMG amplitudes across muscles, or within or between subjects, should be done with caution and adequate understanding of the complexity involved in interpretation. Understanding the relationship of muscle activity and force with joint kinematics and kinetics using engineering principles and neural network representations continues to be a focus of research.\textsuperscript{10,26,100,135}

Each muscle has a particular time during the walking cycle when activity is present or absent and a particular pattern of increasing or decreasing motor unit activity. These timings have been documented for both children and adults.\textsuperscript{75,78,99,112,133} An example can be found in Figure E-1-9. EMG data can be collected using surface or fine wire (indwelling) electrodes. Reviews on the advantages and disadvantages of each have been published.\textsuperscript{59}

![Figure E-1-9](Figure E-1-9. The phasic (on/off) EMG activity of the major muscles used during walking. (Based on data from the Center for Gait and Motion Analysis, Gillette Children's Specialty Healthcare.))

KINEMATICS, KINETICS, AND ELECTROMYOGRAPHY IN NORMAL GAIT
A detailed description of activity in the lower extremities during each phase of the gait cycle is presented in this section, including a summary of hip, knee, and ankle kinematics and kinetics with associated muscle activity. Sagittal plane events are emphasized. The reader should refer to the kinematic and kinetic plots associated with each description.
**Sagittal Plane**

At initial contact (0% of the gait cycle) the ankle is in a position of neutral dorsiflexion, the knee is in minimal flexion, and the hip is in approximately 35 degrees of flexion. The objective of this event is appropriate prepositioning of the foot to begin the gait cycle. The ground reaction force at initial contact is passing through the heel and is anterior to both the knee and the hip. The gluteus maximus and hamstrings muscles are active to control the external flexor moment at the hip. The hamstrings also assist in preventing knee hyperextension. Anterior tibialis and quadriceps activity initiates the loading response (Figure E-1-10; video 1).

**Figure E-1-10.** Initial contact of the gait cycle. (From Gage JR. Gait analysis in cerebral palsy. London: Mac Keith Press, 1991.)

Loading response (Figure E-1-11; video 2) is a period of acceptance of body weight while maintaining stability and progression. The purpose of loading response is to cushion or absorb the impact of the body's moment of inertia. It is the first period of double support and occurs between 0% and 10% of the gait cycle, beginning after initial contact and ending when the entire foot is in contact with the floor. The ankle is plantar flexing at this time under controlled eccentric contraction of the anterior tibialis muscle. The internal moment at the ankle is a net dorsiflexor moment because the dorsiflexor muscles are dominant. The power curve depicts absorption because the anterior tibialis muscle is contracting eccentrically. Gage, Gage and Schwartz, and Perry refer to loading response as the first "rocker" of ankle stance phase.
Figure E-1-11. Loading response (0% to 10% of the gait cycle). (Adapted from Gage JR. Gait analysis in cerebral palsy. London: Mac Keith Press, 1991.)

During loading response, the knee undergoes an initial phase of flexion to approximately 15 degrees (average value). Both hamstring activity and quadriceps muscle activity are present. Because the quadriceps muscles are acting eccentrically to decelerate knee flexion, the power graph depicts absorption. Whenever the joint movement and the joint moment are opposite each other, power absorption is occurring.

Concentric action of the gluteus maximus, gluteus minimus, and hamstring muscles extends the hip. The hamstrings are able to work as hip extensors because knee motion is stabilized by the single-joint muscles of the quadriceps. The net internal joint moment is extensor, and the power graph depicts generation. Because the external ground reaction force falls anterior to the hip joint, without the action of the hip extensors the joint would collapse into flexion.

Midstance (Figure E-1-12; video 3) is the beginning of the single-support phase of the gait cycle and extends from the period of 10% to 30% of the cycle. The goal during midstance is to maintain trunk and limb stability and allow smooth progression over a stationary foot when the entire plantar surface is in contact with the floor. The ankle is in a period of increasing dorsiflexion that is controlled by eccentric contraction of the soleus muscle. The moment graph demonstrates a dominant plantar flexor moment and the power graph a period of absorption. The second rocker of ankle stance phase occurs during midstance.\textsuperscript{41-43}
The knee extends during midstance. The vastus medialis, vastus intermedius, and vastus lateralis muscles work initially to stabilize the knee until the ground reaction force passes anterior to the knee joint. Once the ground reaction force is anterior to the knee, extension is passive. The joint moment at the knee is an extensor moment. The power graph shows initial decreasing absorption followed by generation. This period of power generation is the only one produced at the knee during the entire gait cycle.

The hip continues to extend during midstance. The joint moment is extensor, and power is generated, implicating concentric action of the hip extensor muscles. The internal extension moment, however, is decreasing during this time. When the ground reaction force becomes posterior to the hip joint, a transition occurs from a concentric (power generation) extensor moment to an eccentric (power absorption) flexor moment.

Terminal stance (Figure E-1-13; video 4) is the second half of the single-support phase and occurs from 30% to 50% of the gait cycle. This period begins when the ground reaction force passes anterior to the knee and posterior to the hip and often occurs with heel rise. During this phase of the gait cycle, forward progression of the tibia is arrested and further increase in dorsiflexion is limited. The ankle begins a period of decreasing dorsiflexion by concentric contraction of the gastrocnemius and soleus muscles, producing power generation. One of the primary power productions that propels an individual through the walking cycle is generated at the ankle during terminal stance (36% of the total power generation produced during the walking cycle). Heel rise marks the period of the third rocker of ankle stance phase.
The knee moves from relative extension to increasing flexion during terminal stance. An internal net flexor moment is dominant because the ground reaction force falls in front of the knee. Power is absorbed. The internal flexor moment is produced by a combination of ligamentous resistance and flexor activity of the gastrocnemius muscle.

The hip continues to extend during terminal stance. Because the ground reaction force is posterior to the joint center, extension is resisted by an internal flexor moment. Power is absorbed, suggesting that the flexor moment is produced by tension force of the iliofemoral ligament.

Preswing (Figure E-1-14, video 5) is the second period of double support during the walking cycle and occurs at 50% to 60% of the cycle. The function of preswing is to advance the limb into swing; preswing ends at toe-off. The stance phase extremity is unweighted as weight is accepted on the opposite limb. The ankle is now in true plantar flexion, and the plantar flexor moment remains dominant. The magnitude of the moment is rapidly decreasing, however, and power generation falls quickly to zero.
Figure E-1-14. Preswing (at 50% to 60% of the gait cycle). (Adapted from Gage JR. Gait analysis in cerebral palsy. London: Mac Keith Press, 1991.)

The knee joint is flexing during preswing and reaches approximately 45 degree (average value) at toe-off. The internal muscle moment is extensor, and power is absorbed. Activity of the rectus femoris muscle is probably responsible for this activity and assists the deceleration of the moment of inertia of the shank.

The hip begins to flex in preswing. The dominant moment is a flexor moment, and power is generated. Concentric action of the hip flexor muscles, primarily the iliopsoas, produces the activity. Occasionally, the rectus femoris muscle is active to augment hip flexion. This usually occurs at faster walking speeds. Peak flexor power of the hip is generated at toe-off. Hip musculature (both extensors and flexors) is responsible for the majority of positive work performed during the walking cycle (56%), with most being produced during stance phase.75

The objective of initial swing (Figure E-1-15; video 6) is foot clearance and limb advancement. Initial swing occurs from 60% to 73% of the gait cycle. Maximum plantar flexion occurs at the ankle in initial swing. At the same time, peak knee flexion occurs, uniquely timed for optimal clearance of the foot. The ankle then begins to dorsiflex by activity of the anterior tibialis muscle. The dominant muscle moment is a dorsiflexor moment. Power output is negligible. The hip flexes during initial swing, which also assists foot clearance. A flexor moment remains dominant, and power generation is occurring by concentric activity of the hip flexors.
Figure E-1-15. Initial swing (at 60% to 73% of the gait cycle). (Adapted from Gage JR. An overview of normal walking. In Greene WB, ed. Instructional course lectures, volume 39. Park Ridge, IL: American Academy of Orthopaedic Surgeons, 1990.)

The goals of midswing (Figure E-1-16; video 6) remain foot clearance and limb advancement, and it occurs between 73% and 87% of the gait cycle. The ankle is dorsiflexing during this time by concentric action of the anterior tibialis muscle. The knee is extending by inertial forces without muscle activity. The hip is flexing.
Figure E-1-16. Midswing (at 73% to 87% of the gait cycle). (Adapted from Gage JR. An overview of normal walking. In Greene WB, ed. Instructional course lectures, volume 39. Park Ridge, IL: American Academy of Orthopaedic Surgeons, 1990.)

The primary purpose of terminal swing (Figure E-1-17; video 6) is prepositioning of the limb for weight acceptance. This phase occupies the period between 87% and 100% of the gait cycle. The ankle begins to plantar flex by eccentric action of the anterior tibialis muscle to position the ankle in neutral. A flexor moment is dominant at the knee with power absorption as knee extension is controlled by eccentric action of the hamstrings to decelerate the forward swing of the thigh. The quadriceps muscles also become active to assist with control of the knee. Minimal movement is noted at the hip at this time.
Figure E-1-17. Terminal swing (at 87% to 100% of the gait cycle). (Adapted from Gage JR. An overview of normal walking. In Greene WB, ed. Instructional course lectures, volume 39. Park Ridge, IL: American Academy of Orthopaedic Surgeons, 1990.)

Coronal Plane
The function of hip and pelvis motion in the coronal plane is to optimize the vertical excursion of the center of mass (Figure E-1-18; see Figure E-1-5). At initial contact the pelvis in the coronal plane is level and the hip is in neutral abduction and adduction. The stance side of the pelvis rises 5° at the beginning of loading response in conjunction with adduction of the stance limb. The dropping of the pelvis on the unsupported limb is caused by the ground reaction force on the supported limb, which produces an external adduction moment at hip, knee, and ankle. The external adduction moment is resisted by eccentric control of the hip abductors. This allows the stance side of the pelvis to rise and the unsupported side to drop.

Figure E-1-18. Coronal plane hip joint kinetics. (From Gage JR. Gait analysis in cerebral palsy. London: Mac Keith Press, 1991.)
The pelvis and hip motion reverses in midstance as concentric control by the hip abductors of the stance limb acts to raise the pelvis. This action assists clearance on the swing side. During preswing the hip on the stance side goes into abduction in preparation for toe-off.

Ankle and foot motion in the coronal plane is complex, but it is also very important to efficient gait. It is not routinely measured during full body gait analysis because of inadequate multisegment foot models appropriate for the pediatric client with neuromuscular issues. The literature is confusing because of variations in the terms used to describe motions occurring in the foot. The following description is consistent with nomenclature used by Perry.\(^78\)

At initial contact, as stated earlier, an external adduction moment is present at the ankle and foot. The position of the anatomic body of the calcaneus, which accepts floor contact, is lateral to the tibia, which transmits body weight onto the talus from above. This causes the calcaneus to evert on the talus and reduces the support the calcaneus provides the talus (Figure E-1-19, A). The eversion of the calcaneus on the talus occurs during the initial part of the gait cycle (loading response and early midstance). It is controlled by ligaments surrounding the subtalar joint, as well as eccentric muscle action of the anterior tibialis and posterior tibialis muscles. Maximum eversion occurs at the onset of single-leg stance (15% of the gait cycle). Motion reverses during the remainder of midstance under muscular action of the posterior tibialis and the soleus and the gradual shift of weight bearing to the forefoot. Subtalar joint neutral is reached by 40% of the gait cycle—the midpoint of terminal stance. Inversion locks the midtarsal joint and provides increased stability of the foot during weight bearing on the forefoot. It also moves the calcaneus back under the talus (see Figure E-1-19, B). Peak inversion occurs at the end of terminal stance when ankle power generation is at its peak. Excessive inversion is avoided by co-contraction of the peroneal muscles (peroneus longus and peroneus brevis) during terminal stance and preswing. The subtalar joint is typically in a neutral position during swing phase until slight inversion begins again during the last 20% of the gait cycle.

**Figure E-1-19.** Subtalar action during stance phase. A, The offset in alignment between the body of the calcaneus, which accepts floor contact, and the tibia, which transmits body weight, causes the calcaneus to evert on the talus. The long axis of the calcaneus and the long axis of the talus diverge from each other as the talus rotates inward. B, Subtalar joint inversion during terminal stance repositions the calcaneus under the talus, and the long axis of the bones converge but are not parallel.

**Transverse Plane**

The end result of transverse plane motion is stride elongation (see Figure E-1-5). This is accomplished by internal pelvis and hip rotation under the control of the adductor magnus muscle. During the first half of stance phase internal rotation occurs at the pelvis and hip that reverses in the last half of stance. The pelvis is at its maximum posterior position at toe-off. The foot is positioned 5 to 10 degrees (average value) external to the line of progression throughout the entire walking cycle. Subtalar joint action produces rotation of the tibia as part of the closed
chain mortise joint. The reader is referred to other texts for explanation of the transverse plane motions at the ankle and knee.\textsuperscript{55,78}

**Energy Expenditure**

The purpose of many of the events in the walking cycle in all three planes of motion is to optimize energy expenditure or reduce the vertical translation of the center of mass. Gage\textsuperscript{41,42} includes energy conservation as one of the five attributes of normal gait and states that variation in this attribute encompasses the deviations of the other four attributes.

The mechanisms that the body uses to conserve energy are optimizing the excursion of the center of mass, control of momentum, and active or passive transfer of energy between body segments. The vertical and horizontal displacements of the center of mass are almost sinusoidal and are equal and opposite during typical walking.\textsuperscript{131} The body accomplishes this through the three pelvic rotations (rotation, tilt, and obliquity) and coordinated knee and ankle motion. Inman and colleagues\textsuperscript{55} demonstrated that without pelvic rotation and with stiff limbs, the center of mass of the body would be lifted approximately 9.5 cm with each step. Normal vertical excursion of the body averages approximately 4.5 cm.

Determination of energy expenditure in gait has been a topic of research since the 1950s.\textsuperscript{29,77,83} Various estimates are used to measure energy. Ralston\textsuperscript{83} hypothesized that individuals naturally select a speed of walking that allows a minimum of energy expenditure. More recent studies also support this hypothesis.\textsuperscript{25,33} This has direct implications for the child with a motor impairment.

Research in energy expenditure specifically in children has revealed that younger children consume more energy than teenagers and adults (mass-specific gross rate of oxygen consumption, mL/min-kg).\textsuperscript{33,62,128,125} Despite the fact that children and adults walk in geometrically similar ways, size, changes in morphology, muscular efficiency, and motor skill during growth potentially have an effect on the expense of locomotion. Smaller children perform a greater amount of work per unit mass and per unit time to maintain a given walking speed than larger children, and have a more variable basal metabolic rate. Children under the age of 5 years have more lean body mass and a greater surface-area-to-body-mass ratio.\textsuperscript{33,82} All result in a higher resting energy expenditure. Because of this, a given walking speed is not functionally equivalent at different ages. However, when alternative normalization methods are used to eliminate effects of body size, differences between children of different ages and adults disappear.\textsuperscript{5,33,97} This suggests that changes in the neuromuscular system play a relatively minor role in energy expenditure differences after 3 years of age. Controversy remains over which normalization scheme is most appropriate for use in clinical populations—gross expenditure (resting expenditure + movement expenditure) or net expenditure (gross expenditure – resting expenditure).\textsuperscript{19,20,97,122} A comparison between mass normalized gross rate of oxygen consumption and net rate of oxygen consumption with a nondimensional normalization scheme is found in Figure E-1-20.
Figure E-1-20. A, Typical values of the rate of metabolic oxygen consumption versus velocity for 150 children 2 to 19 years of age. The units for oxygen consumption are normalized by body weight. No normalization of velocity is used. The mean and 2 standard deviations are displayed. B, The same data presented normalized using nondimensional variables for both rate of oxygen consumption and velocity.

An alternative to the use of measuring oxygen consumption to estimate energy expenditure involves calculation of the mechanical work required for walking. Kinematic measurements, anthropometric measurements, and kinetic analyses of internal and external loads are required for use of this method. The advantage of mechanical energy estimation is that the energy requirements of individual joints can be calculated. One of the disadvantages, particularly in the assessment of atypical gait, is that the use of external loads to judge the work associated with walking does not measure the body's ability to efficiently respond to the external loads. Co-contraction associated with spasticity is unaccounted for. A review of mechanical and metabolic estimations of energy expenditure can be found elsewhere.

Heart rate is often used as a substitute clinical measure for oxygen uptake and therefore metabolic energy expenditure because of the linear relationship of heart rate to oxygen uptake. Low mechanical efficiency, however, creates disproportionately high submaximal heart rates in individuals with cerebral palsy (CP), making submaximal heart rate a poor predictor of aerobic capacity in this group. Recent work has demonstrated that day-to-day variability in heart rate in typically developing children is approximately 6%. A lack of correlation between measures using heart rate and direct measures of oxygen cost was also noted. I have observed high and inconsistent heart rate in relation to oxygen uptake in various clinical populations including children with CP. Caution should be taken if consideration is being given to this method of estimation of energy expenditure. See Chapter 6 for more information on aerobic capacity.

USE OF GAIT ANALYSIS IN ASSESSING IMPAIRMENTS

Gait analysis is a useful tool in examining walking impairments in children with physical disabilities because it provides objective measurement of the magnitude of deviations. It also allows the interpreter to analyze data from all planes of motion for a single gait cycle. Just as computed tomography has improved on radiography, gait analysis improves on the evaluator's
clinical examination and visual observation. Gait analysis data in the hands of a critical evaluator will never be used in isolation. Clinical examination by a physician, physical therapist, or trained kinesiologist, and x-rays are vital aspects of the evaluation. The family’s goals for treatment are important as well.

Guidelines for Referral
Guidelines for determining when formal gait analysis should be recommended in the plan of care for a child with a walking disability depend on many factors. The child’s age, diagnosis, progress in physical therapy, and goals of intervention are all taken into consideration. Analysis is best when the child’s gait pattern is mature, but waiting for a mature walking pattern is not always possible. One key to remember is that current gait analysis techniques provide a single "snapshot" in time. Even if a gait pattern is consistent from cycle to cycle on a given day, if it is not consistent from one week or one month to the next, the value of that single snapshot for decision making is more limited. The utility of a gait analysis for children, adolescents, or adults who are making rapid progress in rehabilitation is more limited in scope for this reason.

Typically, the impetus for referral for gait analysis is a change in treatment. That change in treatment may or may not be surgical in nature. Bracing or prosthetic changes, medication changes that may affect balance or walking, can be indicators as well. A three-dimensional gait analysis may not be the best measurement tool for assessing results of changes in a physical therapy program unless the intervention specifically affects the child's walking.

Children under age 3 are not candidates for three-dimensional gait analysis because of their small physical size, their reluctance to cooperate for the duration of the required testing, and their immature gait patterns. Children between 3 and 5 years of age can be tested in the laboratory if their size, behavior, and walking are adequate for the testing process. Because their age and maturity of gait pattern are not ideal, analysis is usually reserved for use as a baseline before surgical intervention. In this age range, interventions, such as selective dorsal rhizotomy, have increased the number of children who have a gait analysis before the age of 5 years.

The optimal age for a formal three-dimensional gait analysis is typically from age 6 onward. Changes in treatment, whether they are bracing or prosthetic changes, medication changes, or surgical interventions, can all be evaluated. Analysis before a treatment change provides a baseline to assist in the decision-making process. Sometimes, no specific treatment changes are recommended. Analysis after treatment provides evaluation of the effects of that treatment. Many people do not appreciate the importance of posttreatment assessment. What they fail to realize is that effective change and improvements in treatment techniques cannot occur without posttreatment assessment. The combined knowledge of previous posttreatment examinations plays an important part in the ability to accurately identify problems and solutions to pretreatment problems.

Assessing Gait Impairments
A gait analysis laboratory should be considered a measurement tool. Information from a gait laboratory can assist in differentiation of primary impairments from secondary compensations. Primary impairments are those abnormalities that are a direct result of CNS injury such as spasticity; secondary impairments are those impairments like soft tissue tightness, which develop over time as a result of the primary impairments, and compensations, which refer to the mechanisms used by the individual to circumvent the primary abnormalities. In addition to use in treatment of conditions such as CP and myelomeningocele, gait data can also be used to assess
the effects of orthotic devices in various populations or of different prosthetic devices in individuals with amputations. Gait analysis at repetitive intervals can assist in the examination of the progression of a particular deformity or condition or the effects of a controlled treatment regimen (including surgery, medication, and physical therapy).

Specific physical therapy recommendations are often difficult to determine with gait laboratory data alone. Assessment during a gait analysis typically includes only a limited documentation of physical findings and is not a complete physical therapy examination. The EMG information from most gait laboratories does not provide information about strength of muscle contraction and cannot be used to determine which muscles are to be strengthened and how. A thorough knowledge of muscle mechanics and typical gait combined with manual muscle testing allows the therapist to answer these questions. EMG activity can, however, determine whether muscle activity is present or absent and the timing of that muscle activity, which can be useful to determine whether a particular muscle is available for strengthening, and to conjecture about what the consequences of that strengthening might be. Kinetic analysis provides joint moment and power information that is sometimes useful for this purpose. Coordination of muscle timing and co-contraction information can be incorporated into treatment as well. In summary, the gait laboratory can be an objective measuring tool for the physical therapist.

Returning to the prerequisites for development of any skill, gait deviations can be created by abnormal motor control, spasticity, loss of ROM, decreased strength, loss of sensation, and bony deformity. Each can result in primary or secondary impairments and compensatory mechanisms used to produce or to maintain useful function. Gait analysis can be used to distinguish among them. It can be used to identify areas of bony abnormality and loss of functional ROM and to document areas of relative weakness or spasticity. The information is always used in conjunction with a clinical examination. The use of gait analysis in assessment of impairments in CP is discussed here as an example of how information is used and interpreted. Limitations of gait analysis are also discussed. Gait analysis is in no way limited to examination of CP or any one gait pathology.

Common Gait Deviations in Cerebral Palsy

Bony Deformity

Bony deformities are assessed best by clinical examination and radiography. Gait analysis data measure the functional effect of the bony deformity on the individual's walking. Bony deformities are examples of secondary impairments because they are not caused directly by the CNS lesion. They can result from failure of physiologic bone remodeling, the effects of spasticity and muscle imbalance, disuse, attempts to function, or any combination of these factors. These deformities (except for leg length discrepancies) are best assessed in the transverse plane data. Once they have occurred they become a primary focus for orthopedic treatment because they cannot be corrected by conservative management. True bony deformity cannot be corrected by physical therapy. Three common bony abnormalities seen in children with CP are internal femoral torsion, tibial torsion, and subtalar joint subluxation.

**Internal femoral torsion (femoral antetorsion)**

Visually, internal femoral torsion appears as internal rotation of the femurs during walking and is measured as such on kinematic analysis. Antetorsion (forward torsion) or internal femoral torsion is a true structural twist deformity of the long axis of the femur. The cause is a combination of (1) persistent physiologic alignment in the infant because of delayed weight
bearing and (2) abnormal muscle forces created by spasticity.\textsuperscript{15,41,42} Internal femoral torsion is not synonymous with valgus angulation of the femoral neck-shaft angle (coxa valga) or the anterior or forward position of the head and neck of the femur relative to the frontal plane (anteversion). Clinically, however, femoral anteversion is a surrogate measurement for the amount of internal torsion present. This is measured as the degree of internal femoral rotation in the prone position required to position the greater trochanter most "lateral" or parallel to the supporting surface.\textsuperscript{91} Usually, in the presence of internal femoral torsion there is loss of passive external rotation range and the range of internal rotation is excessive. Often the ratio of internal to external rotation is equal to or greater than 3:1. Internal femoral torsion and the degree of internal rotation measured on kinematic data are correlated, but they are not synonymous. Not all individuals with an internal twist in the femur walk with excessive internal rotation through the hip joint, but it is common (Figure E-1-21; video 7).

Figure E-1-21. An example of kinematic graphs from a 14-year-old child with spastic hemiplegia cerebral palsy. Gait cycle percentage is on the abscissa and the degrees of motion in the ordinate. The vertical bar at about 60% of the gait cycle delineates stance from swing. The gray band indicates the typical mean \pm 1 standard deviation of typical motion. The red line denotes the left side, and the green line denotes the right side. The pathology is coming primarily from the right side. The transverse plane (right column) demonstrates internal hip rotation on the right. This child has evidence of internal femoral
torsion on physical examination. Internal foot progression angles are noted on both sides. Because hip rotation is external on the left, the internal foot progression angle on the left is the result of internal tibial torsion or foot deformity. Physical exam identifies an internal tibial torsion. The sagittal plane kinematics (center column) demonstrate the typical findings of a child with hemiplegia (progressive anterior tilt from initial contact until preswing, incomplete hip and knee extension in stance, restricted knee flexion in swing and ankle plantarflexion).

**Tibial torsion**
External tibial torsion is more common than internal tibial torsion, but both can be seen in clinical practice. External tibial torsion is an external rotation or torsion of the long axis of the tibia. Internal tibial torsion is an internal rotation or torsion of the long axis of the tibia. Both are measured by physical examination of the transmalleolar axis or thigh-foot axis, but variability in these measurements between and within examiners is known to be high. Functional methods of measurement using the motion capture system for determination of the knee axis have improved the precision of this measurement. Both external and internal tibial torsion are true bony deformities. External tibial torsion often develops as a secondary impairment to internal femoral torsion. Limited knee motion that results in repetitive dragging of the foot in an externally rotated posture for clearance can also result in the deformity. In the presence of internal femoral torsion, external tibial torsion is difficult to observe visually because the foot progression angle may not appear abnormal. The knee sometimes has a valgus appearance, but it is not a coronal plane abnormality. The internal torsion of the femur is compensated by external torsion of the tibia so that the foot remains in the direction of progression (Figure E-1-22). Internal tibial torsion may remain from a lack of proper remodelling from infancy, but can develop as a secondary deformity as well (see Figure E-1-21).

![Figure E-1-22](image.png)

**Figure E-1-22.** A 15-year-old with spastic diplegic cerebral palsy. Note the internal hip position on the right side (knee in) and the foot progression angle that is neutral to the line of progression. His foot progression angle is external on the left with a thigh alignment that appears neutral. This individual has an external tibial torsion on both the right and left sides.

**Pes valgus (subtalar joint subluxation)**
Pes valgus is most common in children with spastic diplegic or spastic quadriplegic types of CP and less often occurs in children with spastic hemiplegia. Caused by a relative subluxation of the talus on the os calcis, it usually develops because of plantar flexor muscle tightness and a relative muscle imbalance between the posterior tibialis and peroneals (with a dominance of peroneal overactivity). Visually, the calcaneus is everted. Because of inadequate modelling of the foot,
kinematic data output from most commercial motion capture systems do not identify subtalar joint subluxation. The effect of the deformity is measured in the plantar flexion-dorsiflexion and foot rotation graphs. Physical and radiographic examination is required.

**Inadequate Range of Motion and Spasticity**

Gage\(^4\) often refers to CP as a condition that preferentially affects two-joint muscles because it is primarily in the two-joint muscles that spasticity contributes to the abnormalities associated with walking. Most two-joint muscles are predominantly fast-twitch muscles used for rapid force production. Loss of ROM creates static contracture; spasticity imposes loss of ROM in dynamic situations because of resistance to stretch.\(^64,89\) The effects of inadequate ROM and spasticity can be measured in all three planes of motion but are most prevalently seen in the sagittal plane. Examples measured by gait analysis are an abnormal plantar flexion–knee extension couple, crouch gait, and limited swing-phase knee motion.

**Abnormal plantar flexion–knee extension couple**

During normal walking, the plantar flexion-knee extension couple is a force couple whereby the soleus muscle controls forward momentum and the forward progression of the ground reaction force by eccentric contraction. Children with CP often enter the walking cycle with a footflat initial contact, which rapidly places the gastrocnemius under premature tension at both ends of the muscle. Gage\(^4\) postulated that, in response to the stretch, spasticity is elicited that can restrict tibial advancement, produce knee extension (hyperextension), and reduce the extent of dorsiflexion (plantar flexion). The spastic response is revealed in a biphasic pattern during stance phase of increasing dorsiflexion and then decreasing dorsiflexion in the kinematics and by an abnormal plantar flexor power generation coincident in time with the first decrease of dorsiflexion in the cycle in the kinetics. Occasionally, the EMG activity is biphasic as well (Figure E-1-23). The abnormal power generation elevates the body's center of mass and functionally increases energy expenditure.
Figure E-1-23. Kinematics and kinetics at the ankle joint demonstrating an abnormal plantar flexion–knee extension couple in a child with cerebral palsy (A) compared with normal (B). The ankle kinematics of the patient display a biphasic pattern in stance phase—increasing dorsiflexion–decreasing dorsiflexion—that repeats itself. This results in a biphasic plantar flexor moment as well, and an inappropriate midstance power generation. C, Surface EMG record from the gastrocnemius-soleus complex also exhibits two bursts of activity coinciding with the power generation.

Crouched posture
Caused by hip flexion contractures or tightness, knee flexion contractures or tightness, excessive plantar flexor muscle weakness, or any combination of these conditions, increased flexion is seen at all joints in the sagittal plane in crouched gait (Figure E-1-24; video 8).
Figure E-1-24. Example of a crouched posture of increased hip and knee flexion in the sagittal plane of a 13-year-old child with cerebral palsy. Patient data compare the child's right side (green) with the left (red). The posture at the ankle is asymmetric, with slightly more dorsiflexion on the left side. Total knee range of motion is limited bilaterally.

Limited swing-phase knee motion
Limitation of motion at the knee in swing phase usually begins in preswing, when motion is also inadequate. Momentum for swing phase in normal gait is generated by the acceleration force of the gastrocnemius muscle that drives the ground reaction force behind the knee. When the gastrocnemius force generation is reduced, as is often the case in CP, the hip flexors supply the
momentum to clear the extremity. The rectus femoris muscle is often recruited as a hip flexor. If this muscle is spastic, it maintains its action as a knee extensor. The already diminished knee flexion in preswing is further inhibited by spastic restraint that does not allow further knee flexion (see Figure E-1-24).

**Weakness**

Weakness cannot be measured directly by gait analysis. The type of electromyography used does not provide information regarding strength. Kinematic data measure only the effects; it is up to the clinician to interpret whether the pattern is present as the result of weakness or tightness. Kinetic data do provide a limited measure of weakness if full ROM is available at the joint. Power can be produced only if joint movement is present. The best measure of strength is by physical examination, although this is complicated by the presence of spasticity.

*Hip abductor weakness*

Weakness of the hip abductor muscles in CP, as in any other type of gait disorder, produces an uncontrolled pelvic drop on the swing side and lateral trunk shift over the stance limb. The ground reaction force in the coronal plane produces an external adduction moment at all joints of the stance limb. If abductor strength is insufficient, the lateral trunk shift positions the ground reaction force through the hip joint center so no abductor moment is needed. Hip abductor muscle weakness is frequently (but not necessarily) seen in children with internal femoral torsion because the torsion creates an inadequate lever arm over which the gluteus medius muscle is able to act, imposing functional weakness. On kinematic graphs hip abductor weakness is displayed as increased adduction. Differentiation between hip abductor muscle weakness and hip adductor tightness cannot be done solely using kinematic data.

*Gastrocnemius-soleus weakness*

Plantar flexor muscle weakness primarily affects terminal stance and preswing phases of the gait cycle, but midstance is also affected. During midstance, the soleus is not able to control the progression of the ground reaction force that is anterior to the ankle, and excessive dorsiflexion results. Heel rise is delayed in terminal stance because gastrocnemius power may also be insufficient. Excessive dorsiflexion can drive the knees into flexion as well. Quadriceps muscle activity may be required to maintain an upright posture if the trunk is vertical, resulting in increased energy expense. Clearance is achieved by the proximal hip flexors because the plantar flexor muscles alone are insufficient. This results in an insufficient acceleration force in terminal stance and imposes a slower gait velocity.

**SUMMARY**

The prerequisites for any motor function, including gait, develop as the CNS matures and the body grows, producing physiologic changes in the mechanics and the neurophysiology of the system. The attributes of typical gait are lost in pathologic conditions such as CP because of the primary impairments of loss of selective motor control and balance, abnormal tone and sensation, muscle weakness, and secondary impairments such as bony deformity and loss of ROM.

This chapter was designed to provide an overview of the refinement of gait in childhood, the components of typical gait as measured by gait analysis, and a brief introduction to the use of
gait analysis in examination of impaired gait. A gait laboratory is only a measuring tool and is a supplement, not a substitute, for other tools used to assess gait deviations.

Research in the area of gait continues to be a topic of great interest. As measurement techniques improve, our knowledge of walking and its development becomes more refined. With a clear understanding of typical gait, we can better understand the mechanisms of pathologic gait. Here the science of physical therapy melds into the art of physical therapy practice for the children we treat.

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I would like to thank Dr. Jim Gage and Dr. Steven Koop, whose philosophy and understanding of normal gait and treatment of children with gait pathologies are significant contributions to this chapter. I am privileged to work with both of them.
REFERENCES


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Figure E-1. An 8-month-old infant with physiologic varus in the tibiofemoral angle.

Figure E-1-2. Standing posture of an 18-month-old toddler. The varus in the tibiofemoral angle has resolved, and the limb is straight.

Figure E-1-3. The tibiofemoral alignment of a 3-year-old showing maximum physiologic valgus.

Figure E-1-4. One complete gait cycle depicting stance phase and swing phase. The cycle begins with initial contact of the right foot. Stance phase has four subphases after initial contact; swing phase has three subphases. Initial contact and toe-off are instantaneous events. (Adapted from Gage JR. An overview of normal walking. In Greene WB, ed. Instructional course lectures, volume 39. Park Ridge, IL: American Academy of Orthopaedic Surgeons, 1990.)

Figure E-1-5. Representative typical three-dimensional kinematic data from the James R. Gage Center for Gait and Motion Analysis of Gillette Children's Specialty Healthcare. In this figure, one complete gait cycle is depicted and is normalized to 100% of the stride. Degrees of motion are on the ordinate. The gray band represents the mean and plus or minus one standard deviation of composite data collected from children ages 4 to 17 years. Stance phase is separated from swing phase in each graph by the vertical bar. Each graph represents the same walking cycle. Each row displays a different joint: from top to bottom are trunk, pelvis, hip, knee, and ankle. Each column displays the movements of a different joint in the same plane of motion; from left to right are the coronal plane (front view), sagittal plane (side view), and transverse plane (rotation view). Note: The pelvis is measured with respect to laboratory coordinates. The trunk and hip are measured with respect to the pelvis, the knee with respect to the thigh, and the sagittal plane ankle with respect to the shank. Foot rotation graph represents foot progression angle with respect to laboratory coordinates, not rotation of the foot with respect to the tibia. (Adapted in part from Gage JR. Gait analysis in cerebral palsy. London: Mac Keith Press, 1991.)

Figure E-1-6. An example of an internal joint moment produced at the right hip. Because the ground reaction force falls anterior to the hip joint, the joint would flex without internally produced resistance. The joint moment graph demonstrates a net extensor moment at the hip (shaded) as the body's internal resistance to the external force. The internal moment is produced by dominant action of the hip extensor muscles. (Adapted in part from Gage JR. Gait analysis in cerebral palsy. London: Mac Keith Press, 1991.)

Figure E-1-7. An example of power generation at the right hip. Because the hip is extending and a net extensor moment is present, power generation (shaded) occurs. The units for power are watts per kilogram. By convention, power generation is represented by positive deflection on the graph; power absorption is negative. (Adapted in part from Gage JR. Gait analysis in cerebral palsy. London: Mac Keith Press, 1991.)

Figure E-1-8. Sagittal plane kinematics and kinetics of the hip (A), knee (B), and ankle (C). Each graph represents the same walking cycle, with the mean (solid line) and one standard deviation (dashed line). The top row shows the kinematic graphs; the middle row, the joint moments; and the bottom row, the joint powers. Units for the kinematics are degrees. The units for joint moments (newton-meters/kilogram) and powers (watts/kilogram) are standardized for body weight.

Figure E-1-9. The phasic (on/off) EMG activity of the major muscles used during walking. (Based on data from the Center for Gait and Motion Analysis, Gillette Children's Specialty Healthcare.)

Figure E-1-10. Initial contact of the gait cycle. (From Gage JR. Gait analysis in cerebral palsy. London: Mac Keith Press, 1991.)
**Figure E-1-11.** Loading response (0% to 10% of the gait cycle). (Adapted from Gage, JR. Gait Analysis in Cerebral Palsy. London: Mac Keith Press, 1991.)

**Figure E-1-12.** Midstance (at 10% to 30% of the gait cycle). (Adapted from Gage JR. An overview of normal walking. In Greene WB, ed. Instructional course lectures, volume 39. Park Ridge, IL: American Academy of Orthopaedic Surgeons, 1990.)

**Figure E-1-13.** Terminal stance (at 30% to 50% of the gait cycle). (Adapted from Gage JR. Gait analysis in cerebral palsy. London: Mac Keith Press, 1991.)

**Figure E-1-14.** Preswing (at 50% to 60% of the gait cycle). (Adapted from Gage JR. Gait analysis in cerebral palsy. London: Mac Keith Press, 1991.)

**Figure E-1-15.** Initial swing (at 60% to 73% of the gait cycle). (Adapted from Gage JR. An overview of normal walking. In Greene WB, ed. Instructional course lectures, volume 39. Park Ridge, IL: American Academy of Orthopaedic Surgeons, 1990.)

**Figure E-1-16.** Midswing (at 73% to 87% of the gait cycle). (Adapted from Gage JR. An overview of normal walking. In Greene WB, ed. Instructional course lectures, volume 39. Park Ridge, IL: American Academy of Orthopaedic Surgeons, 1990.)

**Figure E-1-17.** Terminal swing (at 87%–100% of the gait cycle). (Adapted from Gage JR. An overview of normal walking. In Greene WB, ed. Instructional course lectures, volume 39. Park Ridge, IL: American Academy of Orthopaedic Surgeons, 1990.)

**Figure E-1-18.** Coronal plane hip joint kinetics. (From Gage JR. Gait analysis in cerebral palsy. London: Mac Keith Press, 1991.)

**Figure E-1-19.** Subtalar action during stance phase. **A,** The offset in alignment between the body of the calcaneus, which accepts floor contact, and the tibia, which transmits body weight, causes the calcaneus to evert on the talus. The long axis of the calcaneus and the long axis of the talus diverge from each other as the talus rotates inward. **B,** Subtalar joint inversion during terminal stance repositions the calcaneus under the talus, and the long axis of the bones converge but are not parallel.

**Figure E-1-20.** **A,** Typical values of the rate of metabolic oxygen consumption versus velocity for 150 children 2 to 19 years of age. The units for oxygen consumption are normalized by body weight. No normalization of velocity is used. The mean and 2 standard deviations are displayed. **B,** The same data presented normalized using nondimensional variables for both rate of oxygen consumption and velocity.

**Figure E-1-21.** An example of kinematic graphs from a 14-year-old child with spastic hemiplegia cerebral palsy. Gait cycle percentage is on the abscissa and the degrees of motion in the ordinate. The vertical bar at about 60% of the gait cycle delineates stance from swing. The gray band indicates the typical mean ± 1 standard deviation of typical motion. The red line denotes the left side and the green denotes the right. The pathology is coming primarily from the right side. The transverse plane (right column) demonstrates internal hip rotation on the right. This child has evidence of internal femoral torsion on physical examination. Internal foot progression angles are noted on both sides. Because hip rotation is external on the left, the internal foot progression angle on the left is the result of internal tibial torsion or foot deformity. Physical exam identifies an internal tibial torsion. The sagittal plane kinematics (center column) demonstrate the typical findings of a child with hemiplegia (progressive anterior tilt from initial contact until pre-swing, incomplete hip and knee extension in stance, restricted knee flexion in swing and ankle plantarflexion).

**Figure E-1-22.** A 15-year-old with spastic diplegic cerebral palsy. Note the internal hip position on the right side (knee in) and the foot progression angle that is neutral to the line of progression.
His foot progression angle is external on the left with a thigh alignment that appears neutral. This individual has an external tibial torsion on both the right and left sides.

**Figure E-1-23.** Kinematics and kinetics at the ankle joint demonstrating an abnormal plantar flexion–knee extension couple in a child with cerebral palsy (A) compared with normal (B). The ankle kinematics of the patient display a biphasic pattern in stance phase—increasing dorsiflexion–decreasing dorsiflexion—that repeats itself. This results in a biphasic plantar flexor moment as well, and an inappropriate midstance power generation. C, Surface EMG record from the gastrocnemius-soleus complex also exhibits two bursts of activity coinciding with the power generation.

**Figure E-1-24.** Example of a crouched posture of increased hip and knee flexion in the sagittal plane of a 13-year-old child with cerebral palsy. Patient data compare the child's right side (green) with the left (red). The posture at the ankle is asymmetric, with slightly more dorsiflexion on the left side. Total knee range of motion is limited bilaterally.