Chapter 19

GAIT ANALYSIS AND TRAINING OF PEOPLE WITH LIMB LOSS

JASON M. WILKEN, PhD, PT*; and RAUL MARIN, MD†

*Director, Military Performance Laboratory, Department of Orthopaedics and Rehabilitation, Center for the Intrepid, Brooke Army Medical Center, 3851 Roger Brooke Drive, Fort Sam Houston, Texas 78234

†Colonel (Retired), Medical Corps, US Army; Physician, Center for the Intrepid, Brooke Army Medical Center, 3851 Roger Brooke Drive, Fort Sam Houston, Texas 78234; formerly, Medical Director, GaitLab, Physical Medicine and Rehabilitation Teaching Staff, and Chair, Internal Review Board, Department of Orthopaedics and Rehabilitation, Walter Reed Army Medical Center, 6900 Georgia Avenue, NW, Washington, DC

INTRODUCTION

The systematic study of human ambulation began in the mid-1800s with the appearance of photography and observation-based studies.[1] Subsequently, kinetic (forces)[2] and kinematic (joint movement)[3] studies began near the onset of World War II. The large number of amputees resulting from this conflict became a catalyst for the use of technology for the assessment of amputee gait. Inman and his colleagues introduced the use of movies in the coronal, sagittal, and transverse planes, as well as the use of force plates and electromyography (EMG) into the evaluation of able-bodied and amputee gait.[4]

The intent of amputee gait assessment is the identification of gait parameters that deviate or differ from able-bodied gait (ie, from "normal"). This identification, in turn, provides an opportunity to develop corrective strategies designed to enhance the efficiency, comfort, and cosmesis of amputee gait. Attainment of "normality" is not an end in itself, but rather a guide toward the optimization of function.

This chapter reviews the terminology of gait assessment and discusses both observational and computerized gait assessment. The authors' current understanding of able-bodied gait and documentation of gait performance observed in current military amputees is presented. Also included is a discussion of several special considerations about amputee gait and how gait laboratory assessment translates into effective therapeutic interventions.

TERMINOLOGY OF GAIT ASSESSMENT

A clear understanding of the terminology used in the description and evaluation of gait is essential prior to the assessment of patient performance. A clear or well-defined "common language" among the various disciplines forms the foundation on which a systematic and reliable assessment is built. Such assessment, in turn, leads to the effective planning and execution of interventions. Furthermore, common language facilitates a systematic tracking of progress because it allows for consistent pre- and postintervention comparison and monitoring.

The gait cycle consists of one stride that is divided into two periods known as stance and swing periods. During stance, the primary responsibility of the limb is to support the superincumbent body weight, while during swing, the task of limb advancement is accomplished. Figure 19-1 provides a visual representation of the gait cycle.[5] Note that double support forms 20% to 30% of the gait cycle and that the terminology for describing the phases of gait has changed in recent years. To account for the frequent absence of heel strike in pathologic gait and to allow for a more consistent framework for discussing gait deviations, the classic terminology of heel strike, foot flat, midstance, heel-off, toe-off; and early, mid, and terminal swing is used less frequently. Rather, initial contact, loading response, midstance, terminal stance, and pre-swing are commonly used to describe the stance phase. Initial, middle, and late swing are used to characterize the swing phase.

One of the easiest temporal and spatial parameters to measure quantitatively is walking speed (meters/second). This parameter is critical because velocity strongly influences the loading, motion, and alignment of the different joints throughout the gait cycle. Furthermore, each individual selects his or her natural rate of walking so that the speed selected minimizes energy expenditure.[4] When the gait efficiency of an individual is plotted against various walking speeds ranging from very slow to very fast, a parabolic curve is displayed demonstrating that speeds slower or faster than the self-selected walking velocity (SSWV) will increase energy expenditure.[6]

There are several other temporal and spatial measurements that are used in the assessment of gait. Step length is the distance between a person's heels during the double support phase of gait. Stride length is the distance traveled by the heel from initial contact of one foot to the subsequent initial contact of the same foot. One stride length contains two step lengths. The number of steps that occur within a given period of time (typically per minute) is known as cadence. Cadence can be multiplied by the individual's stride length to determine walking velocity.[5]

Bipedal human locomotion is the result of evolutionary changes that allowed—among other things—the freeing of the upper limbs. This evolutionary change from quadruped to biped locomotion involved changes in the action of forces across levers responsible for locomotion. Torque from application of force distant to the axis of rotation that results in motion about the axis of rotation (if not opposed) is referred to in this chapter as a joint moment or simply moment. The convention of internal joint moments will be used throughout this chapter. Internal joint moments are commonly used because they are associated with the internal (typically muscular) forces that are necessary to produce or resist rotational motion at a given joint. These forces are in direct response to external moments, which are generated by the forces

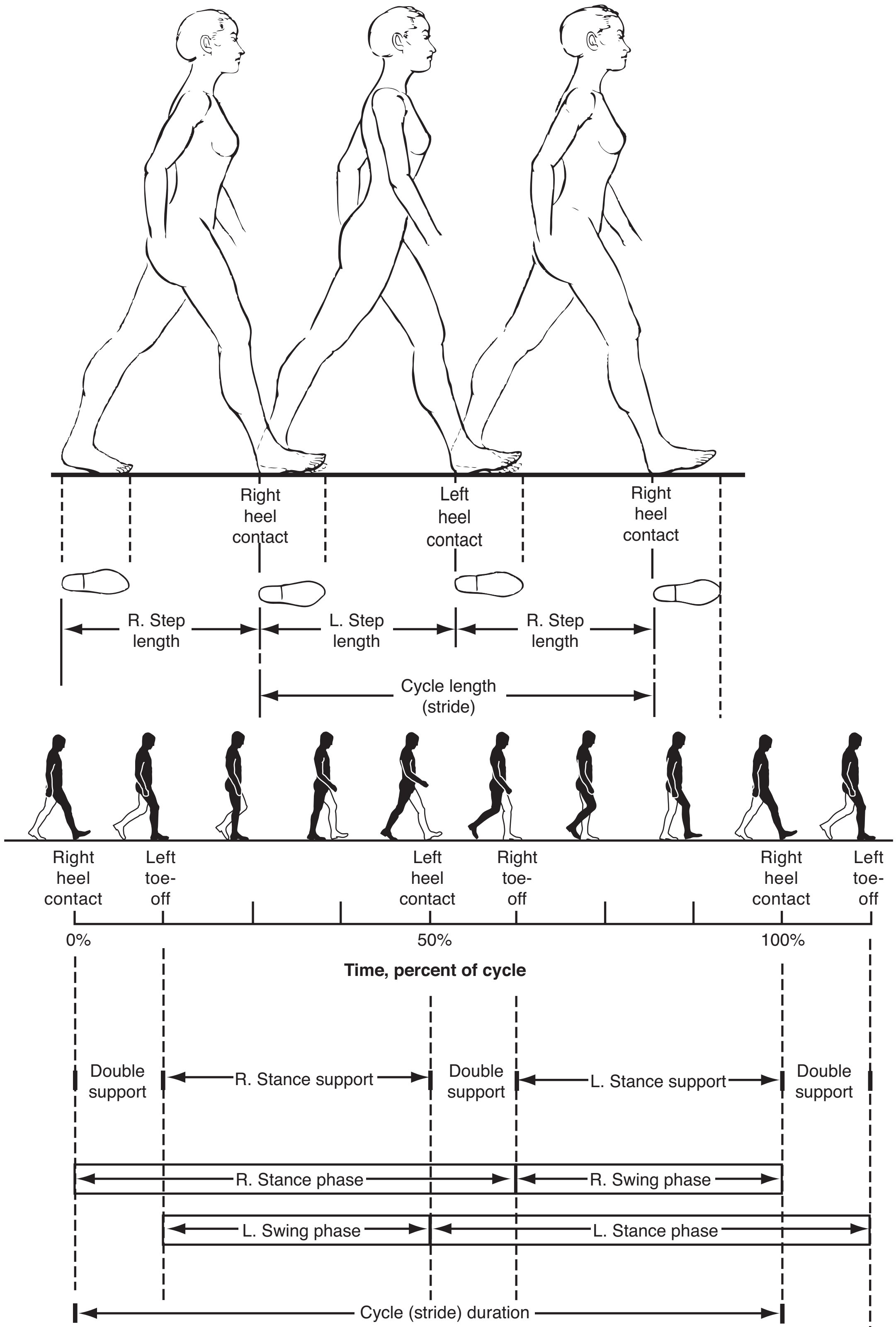

Figure 19-1. Time dimensions of the gait cycle.
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that push against the body (eg, ground reaction force). In standard reporting of gait laboratory data the term moment refers to net or resultant joint internal moment or torque produced by muscular, ligamentous, or bony force.

To the clinician performing an observational gait assessment, the temporal and spatial components of gait are—at most—approximations of observed deviations from what an expected symmetric gait should be. Computerized gait assessment, however, provides for precise quantifiable measurements of these variables.

OBSERVATIONAL GAIT ASSESSMENT

In the military setting a wide range of tools is available to assess amputee gait performance. These tools range from the brief informal visual assessment commonly used in the clinic to the application of advanced motion capture technologies. Selection of the most appropriate assessment approach is based primarily on the desired outcome of that particular evaluation session, which is determined using the patient's and therapist's rehabilitation goals, as well as the standardized assessment protocols at each facility. Observational assessment is—by its very nature—prone to inaccuracies. When a patient is walking, the speed at which motion occurs is commonly faster than what can be processed by the observing clinician. Intuitively, one would think that this limitation can be addressed by the presence of multiple observers under the belief that what one observer may miss may be caught by another observer. However, Krebs et al noted that single observer observational gait analysis is more consistent than multiple observer gait analysis and that this consistency could be increased for both methods by video recording, particularly if the replayed video was in slow motion.[7]

If computerized gait analysis is used as the gold standard by which observational gait analysis is to be compared, then it is apparent that—although convenient—observational gait analysis lacks adequate sensitivity. Saleh and Murdoch demonstrated that visual observers recorded only 22% of the gait laboratory-determined gait deviations with the measurement system detecting 3.4 times more deviations than visual observation alone.[8]

Although the above discussion makes a strong case for the application of computerized gait analysis, the complexity, cost, and limited availability of this methodology significantly hampers its clinical utility. Thus, the average clinician is left with his or her observations to assess the degree of pathology and the intervention required to address that pathology. A report by Gage in 2004 recommends that a systematic and well-organized observational gait analysis recording form be used for all such evaluations.[5] The Rancho Los Amigos full body observational gait analysis form is perhaps the most well-known tool to accomplish this evaluation.[9]

COMPUTERIZED GAIT ASSESSMENT

Gait laboratory assessment in the military setting typically consists of video-based observational analysis in combination with quantitative assessment of temporal-spatial, kinematic, and kinetic parameters. This assessment differs significantly from the typical evaluation performed in the clinic in that it provides quantitative information that can be used to identify gait deviations, provide insight into joint loading and torques, and objectively track progress through pre- and postintervention comparisons. Computerized gait analysis is the standard of care for military amputees and is part of a regular assessment plan. Clinical experience suggests that the best results are obtained in cases where the gait pattern is complex or deviations are subtle, the altered gait pattern is believed to be the source of decreased function or pain, and the patient is highly motivated and follows the resulting treatment plan.

Data Collection

To allow data sharing, facilitate troubleshooting, and support collaborative research projects, the Military Advanced Training Center and the Center for the Intrepid military amputee patient care sites use similar assessment tools for quantifying gait performance.

To provide an accurate assessment of amputee gait kinematics (motion), both sites use passive marker-based optoelectronic motion tracking systems. Passive tracking systems rely on the placement of small (9 mm) reflective markers on the patient's body to accurately track joint and segment positions in three-dimensional space. To independently monitor the three-dimensional motion of each segment, a minimum of three tracking markers is placed on each segment. These tracking markers are supplemented with additional markers that are used to identify bony landmarks and define the orientation of each segment (eg, thigh). The total number of markers used during a session can range from 6 to 70, depending on the type of assessment being performed.

Between 12 and 26 motion capture cameras emit infrared light that is reflected off the surface of the tracking markers and then recorded on the camera's image sensor. Image processing is performed within the camera to identify the centroid—or center—of the marker as a two-dimensional coordinate (vertical and medial-lateral position). The information from each camera is then sent to a central data collection computer. By combining two-dimensional images from the cameras, the system can triangulate the three-dimensional position of the marker with an accuracy of 0.7 millimeters. These marker position data are then used to quantify the magnitude and direction of motion that occur at individual joints and are presented in graphical format for interpretation by the clinical staff.

Force plates are used in all amputee center gait laboratories to assess gait kinetics (or forces). Typically eight force plates are embedded in the main walkway allowing the ability to measure location, direction, and magnitude of force applied at the foot floor interface. Although ground reaction force data are useful as an independent assessment tool, the force data are typically combined with marker data to calculate joint moments as well as power generation and absorption. Insight into the compensatory responses exhibited by amputees is provided using an approach called inverse dynamics. When using this method, joint position, segment motion, mass properties, orientation, and external forces acting on the segment are used to estimate lower extremity joint reaction forces, moments, and powers. Although not providing direct information regarding activation of individual muscles, the approach provides insight into the net effect of muscle activation at each joint of interest.

Because of the large effect of walking velocity on gait performance, gait kinetics in particular, patients are asked to walk along the laboratory walkway at both controlled and SSWV during standard clinical assessment. Collecting gait data at controlled speeds allows the opportunity to determine whether deviations from normal self-selected gait are a result of the amputation or simply the result of changes in walking velocity as compared to uninjured individuals.

Energy Expenditure

Metabolic cost assessment, another assessment technique that can provide insight into gait performance, is commonly used to assess the effectiveness of prosthetic components (as well as training interventions). Metabolic cost assessment is typically performed as patients ambulate on a treadmill while walking velocity is incrementally increased. Patients are typically tested at three to six controlled walking velocities approximately centered about the individual's SSWV. Such assessment allows the ability to determine the patient's gait efficiency and his or her ability to accommodate ambulation at multiple speeds.

Electromyography

EMG can be used to provide insight into the timing, magnitude, and patterns of muscular activation used by amputees. Use of surface electrodes allows the ability to quantify muscle activation while minimizing the risk of discomfort and infection that can be associated with indwelling electrodes. Although challenges such as appropriate normalization of EMG data and reduced access to residual limb musculature must be overcome, the assessment of muscle activation provides insight that other approaches cannot. Unlike the inverse dynamic approach that is unable to account for co-contraction and force production by passive structures (bony or ligamentous restraint), EMG allows the ability to directly record the timing and magnitude of muscle activation.

ABLE-BODIED ("NORMAL") GAIT

In 1985 Perry described four essential requirements for normal gait that—when absent in part or in whole—produce deviations from the norm.[10] These requirements are (1) symmetric or proportionate step length, (2) stance stability, (3) swing clearance, and (4) adequate foot position before initial contact. Gage added energy conservation as a fifth essential requirement in 1991.[11]

Energy conservation during ambulation is multifactorial in nature. The body is designed to be able to take advantage of the ground reaction force for joint stability and efficiency. Efficiency is also increased by the predominant use of eccentric muscle contractions during movement. Furthermore, when the less efficient concentric contractions occur, the muscle begins from a stretched or elongated position to allow the recovery of kinetic energy stored in elastic connective tissues. Energy conservation also occurs through muscles crossing two joints that serve the function of linking and transferring forces across body segments.

The phases of the gait cycle depicted in Figure 19-1 are helpful in understanding the sequence of events that is characteristic of a normal gait cycle. Human walking can be described as a process of alternating acceleration and deceleration of the center of mass, with the bulk of energy utilization occurring at the beginning and at the end of the stance periods. As such, during loading response the quadriceps and tibialis anterior muscles contract eccentrically to allow controlled knee flexion and ankle plantarflexion resulting in a net absorption of force and subsequent deceleration of the body. During midstance the momentum generated by the above described muscle actions takes the body's center of gravity over the weight-bearing leg as the contralateral leg swings like a pendulum with minimal energy utilization. During terminal stance and pre-swing, there is ipsilateral force generation and acceleration via the concentric contraction of the gastroc-soleus complex and similar response of the gluteus maximus in the contralateral side.

Evolutionary changes in human anatomy and biomechanics have ensured that human ambulation occurs with the greatest efficiency at self-selected walking speeds. In 1953 Saunders et al proposed the six determinants of gait as a way of explaining how the body's displacement of the center of mass within a 5-cm sinusoidal pattern of horizontal and vertical displacement conserves energy expenditures during walking.[12] Saunders et al's six determinants of gait follow:

1. Pelvic tilt: in the frontal plane, during stance the weight-bearing gluteus medius allows a drop of the contralateral swing side iliac crest, thus decreasing the rise in the center of mass during single limb support.
2. Pelvic rotation: in the horizontal plane during double limb support, pelvic rotaion allows increased step length while minimizing the drop of the center of mass.
3. Knee flexion at loading response and during stance: reduces the rise of the center of mass as the body transitions over the stance limb.
4. Ankle dorsiflexion at initial contact: effectively elongates the limb allowing a smooth transition onto the stance limb preventing a drop of the center of mass.
5. Ankle plantarflexion at terminal stance/pre-swing: effectively elongates the limb allowing a smooth transition off of the stance limb by preventing a drop of the center of mass.
6. Side-to-side pelvic displacement from stance to stance: shifts the center of mass from one weight-bearing leg to the other.

For decades, these six determinants of gait were believed to effectively characterize the major factors that serve to minimize energy expenditure in human ambulation under the premise that they minimized the excursion or displacement of the center of gravity. Although evidence supports the association between vertical displacement of the center of mass and energy consumption, other studies have questioned the premise that the six determinants of gait (in combination or independently) are the sole factors responsible for controlling the center of mass displacement and thus energy expenditure.[13]

As early as 1983, in a study measuring mechanical cost at three different walking speeds (slow, self-selected, fast), Winter found that energy cost decreases as knee extension increases during stance with a significant positive correlation between energy cost and the degree of knee flexion.[14] In a series of articles, Kerrigan and her co-investigators found that heel rise from foot flat to pre-swing significantly raises the center of mass when it is at its lowest point during the gait cycle with the end result of reducing its overall displacement. They also found that pelvic rotation contributes no more than 12% of the total reduction of the vertical displacement of the center of mass.[15,16] Likewise, other investigators have documented that although knee flexion reduces the center of mass vertical excursion, it significantly increases the energy costs, and that heel rise from foot-flat to pre-swing is responsible for two-thirds of the reduction in the center of mass vertical excursion.[17-19]

It appears then that the six determinants do not have the same influence in reducing the center of mass excursion and that perhaps other as-of-yet to be identified determinants must be contributing to the energy cost savings resulting from the minimization of the center of mass displacement.

GAIT DEVIATIONS IN THE LOWER LIMB AMPUTEE

As in the civilian setting, lower limb military amputees typically demonstrate characteristic gait deviations as a result of the partial loss of lower limb function. Gait deviations observed in the amputee patient are commonly attributed to the following:

- loss of active torque generation;
- loss of somatosensory feedback and limb position awareness;
- additional degrees of freedom added by the mobile interface between the residual limb

and socket;
- pain;
- limitations of current prosthetic devices (foot, socket, and knee if applicable); and
- functional impairments in the contralateral limb.

Fortunately, with adequate training, prosthetic care, and patient motivation, these factors can be minimized or compensated for, yielding a very functional gait pattern that is nearly indistinguishable from that of uninjured individuals.

Many factors influence the maximal gait performance patients attain before discharge from treatment. In military amputees, the frequent presence of significant injury to multiple areas of the body plays an important role in determining the maximal gait performance attained after completion of therapy. Patients often experience significant bony and soft tissue injury to the nonamputated limb at the time of injury, which can result in a nonamputated side that is less functional than the amputated side. In some instances the amputated side is referred to as the "good" or more functional side. Because other injuries vary widely in their impact on the patient's overall gait performance, the current discussion will focus on what is currently deemed the maximal gait performance exhibited by the highest functioning patients and then on commonly observed deviations from that gait pattern.

When assessing gait performance it is important to consider the potential long-term impact of individual gait deviations. Although the patients and clinicians may wish to minimize all gait deviations until they are not readily noticeable, efforts should first be focused on facilitating the most functional and efficient gait while identifying and addressing the source of gait deviations that are most likely to produce long-term morbidity such as osteoarthritis, low back pain, and other commonly observed problems.

STANDARDIZED GAIT ASSESSMENT

To provide a consistent and thorough assessment of amputee gait, a standardized approach is used for the interpretation of gait performance. A comprehensive assessment typically begins with the assessment of temporal-spatial parameters such as walking velocity, step length, step width, stride length, and other gait parameters that allow the rapid detection of gross asymmetries and other deviations from normal. Information such as walking velocity is particularly valuable when interpreting kinematic and kinetic data that are influenced by walking velocity. Kinematic data are then assessed using what is commonly known as a "bottom-up" approach, beginning with the examination of ankle and foot motion and progressing up the kinematic chain. These measures are compared with that of uninjured individuals and high functioning patients that are thought to demonstrate maximal or "optimal" gait performance. After completing the assessment of kinematic data, torque production at the ankle, knee, and hip is assessed in a similar manner, beginning at the ankle and ending at the hips. This is followed by assessment of joint power data before providing a comprehensive assessment of the patient's overall gait pattern and final report generation.

TRANSTIBIAL AMPUTEE GAIT

Due to their relatively young age and high fitness level, service members who have experienced transtibial amputation with adequate residual limb length and minimal contralateral involvement are typically able to achieve a gait pattern that is indistinguishable from uninjured individuals. The highest functioning individuals demonstrate gait patterns that are fluid, functional, and approximate the gait of uninjured individuals in terms of gait kinematics and kinetics: their gait patterns are used as the benchmark against which other amputees are compared. For this reason, deviations from normal gait that are demonstrated by patients with maximal or "optimal" gait performance will be presented before discussing additional deviations that are addressed with training or prosthetic management. Despite the extensive study of amputee gait, the "optimal" gait pattern has yet to be defined.

Initial Contact through Midstance

For most transtibial amputees, the stance phase begins with the lower limb positioned in a manner nearly identical to that of uninjured individuals. The ankle is in a neutral position, the knee is nearly fully extended, and the hip is flexed to between 25 degrees and 35 degrees. Differences from normal gait are, however, observed as soon as the patient begins to bear weight on and control motion of the involved limb and prosthesis.

Ankle Kinematics and Kinetics

State-of-the-art prosthetic feet currently provided to military amputees function—mechanically speaking—in a manner that is significantly different from the physiologic ankle. Shock absorption and "ankle" motion after initial contact are achieved by compression of the heel component rather than through true plantarflexion rotation about a physiologic ankle joint. Observed motion is, therefore, primarily an artifact of proximal displacement of the heel relative to the forefoot resulting in rotation of the shoe relative to the shank. However, meaningful information can be gained from the interpretation of "ankle" kinematic and kinetic data as will be established during presentation of specific gait deviations associated with prosthetic alignment, foot category, and training.

During loading response most high functioning patients are able to reproduce sagittal plane ankle kinematics that are indistinguishable from uninjured individuals. Although they produce normal kinematics, most patients demonstrate dorsiflexion torque that is increased in both duration and magnitude.[20,21] This is due—in part—to the frequently observed increase in double limb support time, which is believed to impart additional stability, and compression of the prosthetic heel, which results in patients spending a significant amount of time on the heel. Once patients reach a stable foot flat position and transition off the heel component, ankle kinematics and kinetics are nearly identical to that of uninjured individuals until pre-swing. As loading shifts from the heel to the forefoot due to forward rotation of the tibia associated with progression of the trunk over the stance limb, gradual dorsiflexion motion and an increasing plantarflexor moment are observed. Resulting sagittal plane ankle kinematics and kinetics (moment and power) during midstance are commonly normal with respect to both pattern and peak magnitude.

Knee Kinematics and Kinetics

Consistent with the broader amputee population, high functioning patients typically display decreased knee flexion from loading response until pre-swing.[20,22,23] Although knee flexion during the stance phase is present, it is both decreased and delayed relative to nonamputee gait.[22] Although nonamputee gait is commonly seen as ideal, a convincing case has not been made to indicate that decreased knee flexion during stance phase is detrimental and not an effective compensatory strategy. There are several potential explanations why patients choose to decrease knee flexion. First, this altered gait pattern may compensate for the functional loss of leg length because of the distal translation of the residual limb within the socket and compression of the prosthetic foot and shock absorber (if present).[24-29] Second, decreased knee flexion during midstance may assist in the gradual storage of energy in the prosthetic forefoot that can be released for propulsion. With an extended knee the body moves as an inverted pendulum. Forward movement of the body over the fixed foot and rotation of the extended leg result in gradual dorsiflexion of the prosthetic foot and storage of energy that can be returned during pre-swing. In this manner, amputees are also able to effectively utilize the rollover characteristics of the prosthetic foot.[30,31] Third, co-contraction may be used as a strategy to stabilize the mobile bone–socket interface and decrease the degrees of freedom of the residual limb.[24,26-29] Increased activation of the quadriceps muscle, despite an extended knee position, has been previously identified as a strategy to stabilize the knee.[21,22] Although this strategy likely serves to control the knee, it may also be in response to knee flexor torque produced by activation of the biarticular gastrocnemius muscle and an overall stiffening of the limb. Activation has been identified as a strategy to stabilize the bone–socket interface by increasing the stiffness of the soft tissue in the residual limb.[32] Although sagittal plane kinematics of the knee is altered, as previously reported net joint torques are minimally affected.[22]

Hip Kinematics and Kinetics

Sagittal plane kinematics and kinetics of the hip have been identified as the primary source of compensation in civilian transtibial amputees, but are relatively unaffected in high functioning military amputees.[20,21] Hip flexion is occasionally increased at initial contact and toe-off but this does not significantly alter joint torques or powers. The increased hamstring activation associated with reported co-contraction at the knee and other reported changes in muscle activation are either not present or have little effect on the sagittal plane kinematics and net sagittal plane hip moments for military amputees. In the highest functioning transtibial amputees the most frequent deviation in hip kinematics is an occasional increase in hip flexion at initial contact.[21]

Trunk and Pelvic Kinematics

A unique characteristic of very high functioning military transtibial amputees is the ability to produce frontal plane trunk and pelvic kinematics that are similar to that of uninjured individuals. Deviations in frontal plane kinematics are documented in other populations but are typically most evident early in the rehabilitation process for military amputees.[33] This deviation is apparently reduced through the combination of extensive training using manual and verbal feedback in combination with proper prosthetic management. Although normal kinematics can be replicated in high functioning transtibial amputees, as a general rule patients demonstrate increased frontal plane motion at the trunk and pelvis during gait.

Pre-Swing

Gait kinematics and kinetics during pre-swing and swing in the military population are similar to those of civilian amputees with a few key exceptions. For all amputees using dynamic energy storing and return feet, the peak dorsiflexion angle that is achieved is determined by a combination of foot stiffness, alignment, and the ankle joint torque produced by the patient due to loading of the trail limb. As observed in previous studies, with appropriate prosthetic management patients are able to attain normal plantarflexor torque and motion at terminal stance. Patients demonstrate a remarkable ability to control the rate of energy returned from the foot and as a result are able to achieve nearly normal power output at the ankle. This observation is in sharp contrast to literature indicating "marked reduction in energy generating capabilities compared to the normal gastrocnemius-soleus."[20,21,34-36] Although the exact mechanism by which normal ankle powers are achieved is unknown, current explanations focus on training to sustain loading of the trail leg until just before the foot is lifted from the floor. The ability to produce normal ankle power is particularly impressive given that current prosthetic feet only return to the neutral alignment position (typically in slight dorsiflexion) when unloaded.

Swing

With the exception of ankle kinematics, few differences exist between the swing phase kinematics of high functioning transtibial amputees and uninjured individuals. Toe clearance (1–2 cm) is the most important aspect of the swing phase in transtibial amputees. The loss of active control of ankle motion prevents patients from using dorsiflexion during midswing to aid in clearing the toe. This is particularly relevant when considering the effective increase in leg length that occurs due to distal translation of the socket relative to the residual limb during swing.[37] Although microprocessor controlled prosthetic foot and ankle systems are available to aid with toe clearance, they have not been widely accepted by the patients. Instead patients use nearly imperceptible compensatory changes at more proximal joints that allow adequate clearance.

Additional Kinematic and Kinetic Deviations in the Transtibial Amputee

Deviations in ankle kinematics and kinetics are most commonly observed early in the rehabilitation process as patients are beginning to walk or are accommodating to new prosthetic components. In the authors' experience, ankle kinematics and kinetics are primarily influenced by the stiffness (grade) of the prosthetic foot and training to control loading of the prosthetic foot.

Initial Contact Through Midstance

One of the most commonly observed deviations in transtibial military amputees includes ankle plantarflexion motion and ankle dorsiflexion torques that are increased in duration and magnitude relative to other transtibial amputees.[20,21] These deviations are primarily attributed to a prosthetic foot that is too soft or too excessive and poorly controlled loading early in stance phase. In the case of a soft foot, the heel undergoes significant compression resulting in increased time in loading response. Rather than gradually moving from the heel during loading response to foot flat, this transition is delayed as the patient compresses the prosthetic heel. The aforementioned deviations in ankle kinematics and kinetics are typically accompanied by deviations at more proximal joints. Knee flexion is decreased or nearly absent resulting in a gait pattern similar to that of high functioning transfemoral amputees. As the heel compresses, patients lean toward the stance limb resulting in frontal plane rotation of the trunk and pelvis toward the stance limb.

In contrast, some individuals demonstrate an opposite response in which ankle plantarflexion motion and ankle dorsiflexion torques are decreased in duration and magnitude.[21] This is typically observed in patients with a prosthetic foot that is too stiff and/or in cases where patients are experiencing anterior-distal pain or have a poor tolerance for loading early in stance phase. In both instances, patients tend to move quickly to a foot flat position following initial contact. In the case of a foot that is too stiff, it appears as though patients avoid the unstable position of balancing on a firm heel and instead quickly move to a more stable foot flat position. For patients with a painful residual limb, a compensatory response of a foot flat landing and decreased time on the heel is reasonable given the current understanding of limb–socket dynamics. At initial contact a significant anterior and inferiorly directed force is applied at the bone–socket interface resulting in distal translation of the limb within the socket and loading along the anterior aspect of the socket. By landing in a foot flat position, patients are able to decrease the shear force during loading response and direct force along the axis of the residual limb into the soft tissue bulk instead of focusing it on the anterior-distal end. Due to the inability to actively control ankle orientation to bring the forefoot to the floor, patients use increased knee and hip flexion motion to lower the foot to the floor in a controlled manner.[21] Knee extensor torque is not, however, typically increased in military amputees because of efforts to minimize anterior shear forces in the painful patient.

Pre-Swing

Improper stiffness (grade), alignment of the prosthetic foot, or poor muscular control also results in gait deviations during pre-swing. As previously mentioned, the alignment and stiffness of the foot can significantly influence the peak dorsiflexion angle and response of the foot during pre-swing. A forefoot that is too soft or is aligned in excessive dorsiflexion can allow excessive peak dorsiflexion motion and is commonly accompanied by patient reports of insufficient foot length or energy return. Rather than exhibiting increased peak dorsiflexion, patients commonly accommodate by decreasing loading of the trail limb, which has the negative consequence of decreased plantarflexor power. Although compensations such as increased knee and hip extension are additional strategies that may be used to accommodate a prosthetic foot that is too soft, such responses are not consistently observed.

A forefoot that is too stiff or is aligned in excessive plantarflexion does not allow sufficient dorsiflexion motion and is accompanied by increased knee extension during mid and late stance.[21] This is due to increased resistance to forward rotation of the tibia over the foot. Rather than increasing the loading on the forefoot to attempt to deflect the foot, patients typically compensate by extending the knee. As a result, dorsiflexion motion is decreased, which leads to decreased peak plantarflexor power as the foot is unloaded at terminal stance.[21]

TRANSFEMORAL AMPUTEE GAIT

The combined loss of ankle and knee function and a dynamic bone–socket interface results in more pronounced gait deviations in patients that have undergone transfemoral amputation or knee disarticulation procedures. Although military transfemoral amputees have access to the latest prosthetic technologies and training, they typically have gait deviations that are readily apparent to an untrained observer. Appropriate prosthetic and physical therapy management is essential to allow restoration of a functional, efficient gait pattern. Contracture prevention, strength training, and appropriate socket fit are critical to provide patients the range of motion, strength, and control of the prosthesis necessary to produce a good gait pattern.

Initial Contact Through Midstance

Ankle Kinematic and Kinetic Data

Sagittal plane ankle kinematics and kinetics of the high functioning transfemoral amputee closely mimic those of the transtibial amputee. They typically demonstrate near normal sagittal plane ankle kinematics but exhibit the frequently observed pronounced dorsiflexion moment at loading response.[38-40]

Knee Flexion Kinematic and Kinetic Data

Sagittal plane knee kinematics differs significantly from that of uninjured individuals. Like most civilian amputees, military transfemoral amputees have little if any knee flexion and exhibit knee flexor torques throughout the stance phase.[38-41] This is primarily attributed to a training philosophy that encourages patients to utilize hip extensor activation to stabilize the prosthetic knee by pushing it into full extension during stance. There are three perceived benefits of this training. First, by not relying on the 10 degrees to 15 degrees of stance flexion allowed by most prosthetic knees, there is a greater margin for error when performing challenging tasks such as ambulating on uneven terrain. If patients land on an object that serves to push their knee into flexion, they are already actively creating stability rather than relying on the response of the prosthesis. If the knee begins to buckle, the muscles are already tensioned leading to a faster response time and decreased likelihood of falling. Second, it encourages activation of hip extensor musculature for propulsion decreasing demand on the contralateral limb and easing the transition to high-level activities, such as running, during which hip extensor torque is a primary means of propulsion. Third, the repeated activation of residual limb musculature that is encouraged with this approach is thought to reduce muscle atrophy (maintain muscle volume) and provide a more rigid limb–socket interface.

Although stance flexion was thought to play an important role in minimizing center of mass excursion and therefore metabolic cost, its value has recently been questioned. More recently the focus on stance flexion has revolved around its role as a shock absorbing mechanism. Although likely useful for some patients, in the authors' experience, stance flexion is not necessary because of adequate shock absorption provided by the inclusion of rotation and shock absorbers, deflection of the prosthetic heel, and the bone–socket displacement that occurs in current prosthetic socket designs.[30] This ability to adequately absorb impact forces is clearly evident in ground reaction force curves that typically lack the pronounced vertical impact peaks observed in normal gait.[9]

Hip Kinematic and Kinetic Data

As a result of the closed kinematic chain present during stance, altered kinematics of the knee results in deviations in hip kinematics. Unlike normal gait in which the hip flexion angle changes little during the first 5% to 10% of the gait cycle, transfemoral amputees commonly demonstrate a more constant hip extension velocity (straight line on kinematic curve) from initial contact until peak hip extension. In uninjured individuals the minimal changes in hip flexion angles during early stance are the result of stance flexion moving the entire limb into a more flexed position resulting in a delayed transition toward extension at the hip. In transfemoral amputees, the absence of stance flexion results in a consistent hip extension velocity as the pelvis progresses over the fully extended leg.

Trunk and Pelvic Kinematics

Typically only high functioning amputees with well-fitting sockets are able to minimize deviations in frontal, sagittal, and transverse plane pelvic and trunk kinematics that are commonly seen in civilians.[33,42-44] It is unclear whether deviations in pelvic motion are the result of limitations in prosthetic technology, training, or both. Although the use of ischial containment socket designs allows freedom of hip and pelvic motion not otherwise attainable with an ischial weight-bearing quadrilateral socket, most patients demonstrate a compensated gluteus medius gait pattern. Rather than the contralateral side of the pelvis decreasing in height during loading response, the contralateral shoulder and pelvis elevate resulting in trunk and pelvis rotation toward the stance limb.[42,44] This altered trunk and pelvic motion, which is commonly the most visually apparent gait deviation, is readily recognized by the untrained observer. Two primary causes of this increase in motion are believed to be (1) hip abductor weakness and the (2) mobility of the femur within the socket.[30,45] By bringing the center of mass over the hip joint center the hip abductor moment is decreased, thus reducing the torque requirements of the gluteus medius. Similarly, frontal plane trunk and pelvic rotation toward the stance limb allows the patients to axially load through the socket rather than relying on compression of the lateral soft tissues or co-contraction of thigh muscles to stabilize the femur in the socket.

Although some patients are able to achieve normal pelvic orientation (15 degrees anterior pelvic tilt) and excursions (approximately 5 degrees) throughout the gait cycle, a majority of patients demonstrate increased anterior pelvic tilt and sagittal plane motion.[33] Commonly, patients move into a position of increased anterior pelvic tilt during late stance as they approximate hip end range of motion and then rapidly posteriorly tilt the pelvis to help initiate swing. Because of the perceived increased risk of low back pain associated with increased sagittal plane motion of the pelvis and lumbar spine, efforts are under way to more effectively train patients to decrease this motion.

Terminal Stance Heel Strike

Although transfemoral amputees approximate the nearly normal sagittal plane ankle motion and torque exhibited by transtibial amputees, subtle differences result in ankle joint power peaks that are consistently less than that exhibited by high functioning transtibial amputees.[38,40] It is, however, unclear whether this difference is due to the loss of active control of the knee that alters loading of the foot, or the use of low profile feet that tend to be less dynamic.

During terminal stance, the knee joint angle rapidly moves from the extended position during midstance to approximate normal peak flexion angles during swing.[38,40] The rapid transition from extension into flexion is controlled by extensor torque produced by resistance within the prosthetic knee resulting in a burst of negative power that peaks around toe off.

Because it is the only source of active power generation in the involved limb, deviations are commonly seen in hip kinematics and kinetics. It is common for patients to demonstrate increased hip extension motion and increased hip flexor torque in late stance as compared to uninjured individuals. The most commonly observed deviation in sagittal plane pelvic motion is rapid motion into anterior pelvic tilt during terminal stance. Rather than isolating the motion to the hip joint they extend through the lumbar spine. Although functional, this motion pattern is discouraged because it is believed to contribute to the high prevalence of low back pain in amputees.

SPECIAL CONSIDERATIONS

Energy Expenditure

The energy expenditure of an amputee is influenced by extrinsic factors such as prosthetic malalignment, leg length discrepancies, and the limb–socket interface mismatches previously discussed.

Intrinsic factors are those inherent to the amputee such as age and the presence of vascular disease. In addition, amputee specific gait deviations that result from the loss of sensory perception in the amputated limb (particularly proprioception), loss of the shock absorbing and propulsion properties of the foot and ankle (for transfemoral and transtibial amputees), and loss of the loading response at the knee (transfemoral amputee) lead to unavoidable deviations from the energy-efficient gait pattern of nonamputees. These losses can result in deviations that result in an increase in energy expenditure. As stated previously, although changes in the six determinants of gait have been used to explain these changes, some of these may actually have no influence and other factors yet to be identified may be more influential. Table 19-1 shows the percent increase in energy cost for the different levels of lower limb amputation.[46] Note that these average values will increase when pathological deviations exist. The end result shows that to minimize these average energy cost increases, the amputee slows down the ambulation speed. Thus, just as in able-bodied individuals, the U-shaped curve seen when gait efficiency is plotted against various walking speeds is also seen in the amputee, although at a higher baseline. That is, the amputee consumes more energy when walking at slower or faster speeds than the SSWV with the energy expenditure of that SSWV being higher than in able-bodied individuals.[47] As a general rule, appropriate prosthetic management of the amputee aims to keep the energy costs of ambulation at or below the values listed in Table 19-1 by minimizing previously discussed gait deviations.

In a review of the influence of prosthetic alignment and prosthetic components on energy expenditure, Schmalz et al found that malalignments affect transfemoral more than transtibial amputee gait.[48] Furthermore, it was also found that there were no differences in energy expenditure when different prosthetic feet were worn by transtibial amputees and that microprocessor controlled knees reduced energy costs when compared to conventional hydraulic knees. Buckley et al also found that microprocessor knees tend to lower energy costs when compared to conventional pneumatic knees.[49] Taylor et al found that although energy cost differences exist between these two prosthetic types at slower or SSWVs, there were energy cost savings when walking at higher speeds.[50] Barth et al found no differences in metabolic costs among different feet in transtibial amputees although significant differences in temporal-spatial, kinetic, and kinematic parameters were found among the various feet tested.[51] Furthermore, vascular amputees had higher metabolic costs than traumatic amputees. However, these studies had small sample sizes and lacked randomization or stringent controls. Large sample comprehensive randomized studies with stringent controls are needed to fully and conclusively assess the purported advantages of currently available advanced prosthetic components.

Effect of Prosthetic Mass

As shown in Table 19-1, the average energy costs of prosthetic ambulation increase as the level of limb loss increases. This reality, with the evidence showing that adding weight to the lower extremities of nonamputees increases metabolic costs of walking, has triggered the development of lightweight prosthetic components over the past decade.[52]

The concerted effort to decrease prosthetic weight so as to decrease the metabolic costs of amputee gait, however, appears to have little scientific basis. Investigators have consistently demonstrated that no relationship exists between metabolic costs and prosthetic limb mass,[53-56] while others have also shown that no relationships exist between prosthetic mass and temporal-spatial gait parameters such as stride length, stride frequency, or self-selected walking speed.[55,57,58]

Selles et al found that although variations in prosthetic mass do not seem to affect joint angles (ie, kinematics), changing limb mass does influence the forces being applied across the joints (ie, kinetics).[59] Gitter et

TABLE 19-1

AVERAGE ENERGY COSTS OF PROSTHETIC AMBULATION

	Percentage Increase
Unilateral transtibial	9%–28%
Unilateral transfemoral	40%–655%
Bilateral transtibial	41%–100%
Bilateral transfemoral	280%
Transtibial plus transfemoral	75%
Unilateral hip disarticulation	82%
Hemipelvectomy	125%

al explained these changes in the kinetics of gait, however, in 1997 when they demonstrated that although an increase in the mechanical energy occurs across the prosthetic side hip joint during the pre-swing and swing phases of gait to accelerate the heavier prosthesis into swing, there is also a net absorption of energy during the terminal swing phase.[57] Thus, the force needed to accelerate the heavier prosthesis during swing is counteracted by an equal degree of energy recovery as the limb decelerates during terminal swing with a net effect of zero loss of energy.

To best serve the patient, the clinician should consider the location of the mass of the prosthesis rather than concentrate on the actual total weight of any given prosthesis. Lehmann et al demonstrated that the economy of gait decreases when the center of mass of the prosthesis is located distally.[55] Furthermore, investigators have also determined that intermediate center of mass locations were the most energy-efficient locations for the prosthetic center of mass.[52,56]

Effect of Prosthetic Components

In an extensive review of the literature, van der Linde et al classified all the studies evaluating the effect of prosthetic components on amputee gait using established quality assessment criteria.[60] The literature was divided into three categories:

1. A-level studies: studies meeting the most stringent criteria, thus defining these as studies of the highest quality.
2. B-level studies: studies partially meeting the selection criteria, thus defining these as studies of moderate quality.
3. C-level studies: studies not meeting most of the criteria, thus defining them as studies of low-level quality.

A total of 40 of the 356 articles identified met the criteria warranting full assessment. These studies were then divided into prosthetic foot, prosthetic knee, prosthetic socket, and prosthetic mass focus studies. Within the prosthetic foot studies, one met the A-level criteria, 15 met the B-level criteria, and 5 met the C-level criteria. As a whole, foot type did not influence the SSWV among individuals with traumatic transtibial and transfemoral amputation. There were few foot-specific effects on gait performance and the few foot-specific effects that were observed occurred when energy-storing feet were compared with the Solid Ankle Cushion Heel foot. In general, although variability existed resulting from study group selection bias, the energy-storing feet appeared to provide for a higher SSWV. However, the evidence for a slight decrease in oxygen consumption is not convincing. Likewise, the evidence for a higher degree of patient satisfaction with the energy-storing feet is not fully convincing.

In this review, the authors also reported that as a whole, prosthetic knees offering advanced swing control (pneumatic, microprocessor-controlled, hydraulic) provide improved symmetry and gait velocity when compared to constant friction mechanical knees, particularly in active amputees. For the geriatric and more sedentary amputee, however, a locked knee with circumduction or hip hiking during swing may be more efficient because it provides improved knee stability required for this type of patient who is at risk of falls.[60]

Although van der Linde et al's literature review used stringent criteria to select methodologically sound articles in their final analysis, only five manuscripts were included addressing the prosthetic knee, two of which were from the early 1980s, two from the mid 1990s, and one from 2000. Furthermore, their conclusions compared the pneumatic, hydraulic, and microprocessor knees versus the constant friction knee. Studies comparing microprocessor-controlled knees versus mechanical pneumatic or hydraulic knees and studies comparing various types of microprocessor-controlled knees were thus not included in their review.[60]

In an assessment of the C-Leg (Otto Bock Healthcare, Minneapolis, Minn) versus two hydraulic-based mechanical knees utilizing 10 amputees, Kastner et al reported that when using the C-Leg, amputees had the most normal kinematic parameters at various speeds and their walk was more efficient as demonstrated by having the fastest time in a 1,000-meter walk test.[61] A more recent eight-subject study compared two microprocessor-controlled knees (C-Leg; RHEO KNEE, Ossur, Aliso Viejo, Calif) and a commonly used hydraulic mechanical knee (Mauch SNS, Ossur, Aliso Viejo, Calif).[38] The outcome measures included (*a*) metabolic cost, (*b*) kinetic and kinematic data, and (*c*) EMG and accelerometer data. The investigators found that metabolic costs decreased the most with the RHEO KNEE followed by the C-Leg. Furthermore, the microprocessor-controlled knees demonstrated biomechanical advantages in several kinetic and kinematic parameters.[38] EMG analysis revealed that there was decreased muscular activity of the gluteus medius on the affected side with the RHEO KNEE compared to the Mauch SNS and C-Leg. Finally, accelerometer data demonstrated smoother transitions from swing to stance and vice versa with the microprocessor knees.[38] Although these studies seem to demonstrate differences among the knees studied, these results need further corroboration. This is highlighted by the current self-selection of conventional hydraulic knees by military amputees.

In regards to the prosthetic socket, only one C-level study was found that suggested increased symmetry of gait with vacuum versus suction sockets. The methodology of this study, however, was poor with no control for other prosthetic components. The literature analysis dealing with prosthetic mass supported the points already discussed in the section above.[60]

In summary, when it comes to prosthetic prescription guidelines, the clinician must rely on a consensus team decision and patient input due to the current paucity of scientific evidence for the prescription of prosthetic devices to injured service members.

Effect of Gait Training

A review of the Ovid database over the past 10 years revealed only two articles on this topic, both by the same investigators.[62,63] Both papers originate from the same research with the first paper describing the effect of a gait reeducation program on self-selected walking speed and pain-free or adaptive equipment free ambulation. The second paper described the effects of the gait reeducation program on sagittal plane kinetic and kinematic variables.

The sample size for both papers was 9 transfemoral trauma or tumor-related amputees with a mean age of 33 years. All had worn their respective prosthesis for 18 months. Measurements were made at baseline (before the start of training), at the conclusion of the training program, and at 6 months follow-up. The training program averaged 10 months and incorporated traditional motor skills based gait training with a psychologically based self-awareness and confidence building outreach component. Training sessions occurred weekly for 1 to 2 hours.

The investigators reported that the gait re-education program led to improvements in SSWV, gait symmetry, and an associated increase in the amount of muscle work on the amputated side. Three subjects were able to walk without walking aids, seven subjects learned to jog, and all had either a decrease or a disappearance of pre-training low back pain. These findings persisted until the 6-month follow-up. Although the findings described in these two studies are promising, more studies are needed in this area.

GAIT LABORATORY DEPENDENT THERAPEUTIC INTERVENTIONS

Although rarely implemented in the past, the use of advanced technologies holds great promise for rehabilitation of the military amputee.[64,65] The combined resources of real-time feedback and virtual reality are being implemented with the goal of improving gait performance. Current efforts are based on promising work that suggests that improved symmetry of loading and decreased metabolic cost can be attained in a relatively short period of time using visual feedback.[64,65] It is anticipated that the use of the virtual reality and a real-time feedback could be effective in accelerating the rehabilitation process, especially with gait training. This method also could prove valuable in prosthetic device modifications, gait strategies used by amputees, and selection of an environment to achieve optimal efficient prosthetic gait.

SUMMARY

This chapter has attempted to provide the reader with an overview of the terminology of gait, as well as observational and computerized gait assessment. The able-bodied "normal" gait has been discussed as a construct from which to understand gait deviations in the amputee. An indepth discussion of transtibial and transfemoral amputee gait has been provided with specific discussions concerning kinematic and kinetic parameters of joints throughout the gait cycle. Further discussion of specific considerations in amputee gait such as energy expenditure and prosthetic mass effect has also been provided. The chapter concludes with a discussion of how gait laboratory evaluation assists in the management of lower extremity amputees.

REFERENCES

1. Weber W, Weber E. *Mechanik du Meschlichen*. Gottingen, Germany: Gehwerkzeugen; 1836.

2. Schwartz RP, Heath AL, Misiek W, Wright JN. Kinetics of human gait: the making and interpretation of electrobasographic records of gait. *J Bone Joint Surg Am*. 1934;16:343–350.

3. Bernstein NA. Biodynamics of locomotion. In: Whiting HTA, ed. *Human Motor Actions: Bernstein Reassessed*. Amsterdam, The Netherlands: North-Holland; 1984.

4. Inman VT, Ralston HJ. Human walking. In: Lieberman JC, ed. *Human Walking*. Baltimore, Md: Williams and Wilkins; 1981.

5. Gage JR. A quantitative description of normal gait. In: Gage JR, ed. *The Treatment of Gait Problems in Cerebral Palsy*. 2nd ed. London, England: Mac Keith Press; 2004.

6. Rose J, Ralston HJ. Energetics of walking. In: Rose J, Gambel JG, eds. *Human Walking*. Baltimore, Md: Williams and Wilkins; 1994.

7. Krebs DE, Edelstein JE, Fishman S. Reliability of observational kinematic gait analysis. *Phys Ther*. 1985;65:1027–1033.

8. Saleh M, Murdoch G. In defence of gait analysis. Observation and measurement in gait assessment. *J Bone Joint Surg Br*. 1985;67:237–241.

9. Perry J. Gait analysis systems. In: Perry J, ed. *Gait Analysis: Normal and Pathological Gait*. Thorofare, NJ: Slack; 1992.

10. Perry J. Normal and pathologic gait. In: Bunch WH, ed. *Atlas of Orthotics*. St. Louis, Mo: Mosby; 1985.

11. Gage JR. *Gait Analysis in Cerebral Palsy*. London, England: Mac Keith Press; 1991.

12. Saunders JB, Inman VT, Eberhart HD. The major determinants in normal and pathological gait. *J Bone Joint Surg Am*. 1953;35:543–558.

13. Kerrigan DC, Viramontes BE, Corcoran PJ, LaRaia PJ. Measured versus predicted vertical displacement of the sacrum during gait as a tool to measure biomechanical gait performance. *Arch Phys Med Rehabil*. 1995;74:3–8.

14. Winter DA. Knee flexion during stance as a determinant of inefficient walking. *Phys Ther*. 1983;63:331–333.

15. Kerrigan DC, Della Croce U, Marciello M, Riley PO. A refined view of the determinants of gait: significance of heel rise. *Arch Phys Med Rehabil*. 2000;81:1077–1080.

16. Kerrigan DC, Riley PO, Lelas JL, Della Croce U. Quantification of pelvic rotation as a determinant of gait. *Arch Phys Med Rehabil*. 2001;82:217–220.

17. Della Croce U, Riley PO, Lelas JL, Kerrigan DC. A refined view of the determinants of gait. *Gait Posture*. 2001;14:79–84.

18. Gard SA, Childress DS. The influence of stance-phase knee flexion on the vertical displacement of the trunk during normal walking. *Arch Phys Med Rehabil*. 1999;80:26–32.

19. Ortega JD, Farley CT. Minimizing center of mass vertical movement increases metabolic cost in walking. *J Appl Physiol*. 2005;99:2099–2107.

20. Seroussi RE, Gitter A, Czerniecki JM, Weaver K. Mechanical work adaptations of above-knee amputee ambulation. *Arch Phys Med Rehabil*. 1996;77:1209–1214.

21. Winter DA, Sienko SE. Biomechanics of below-knee amputee gait. *J Biomech*. 1988;21:361–367.

22. Powers CM, Rao S, Perry J. Knee kinetics in transtibial amputee gait. *Gait Posture*. 1998;8:1–7.

23. Sanderson DJ, Martin PE. Lower extremity kinematic and kinetic adaptations in unilateral below-knee amputees during walking. *Gait Posture*. 1997;6:126–136.

24. Commean PK, Smith KE, Vannier MW. Lower extremity residual limb slippage within the prosthesis. *Arch Phys Med Rehabil*. 1997;78:476–485.

25. Erikson U, Lemperg R. Roentgenological study of movements of the amputation stump within the prosthesis socket in below-knee amputees fitted with a PTB prosthesis. *Acta Orthop Scand*. 1969;40:520–529.

26. Grevsten S, Erikson U. A roentgenological study of the stump-socket contact and skeletal displacement in the PTB-Suction Prosthesis. *Up J Med Sci*. 1975;80:49–57.

27. Grevsten S, Eriksson U. Stump-socket contact and skeletal displacement in a suction patellar-tendon bearing prosthesis. *J Bone Joint Surg Am*. 1974;56:1692–1696.

28. Lilja M, Johansson T, Oberg T. Movement of the tibial end in a PTB prosthesis socket: a sagittal X-ray study of the PTB prosthesis. *Prosthet Orthot Int*. 1993;17:21–26.

29. Narita H, Yokogushi K, Shii S, Kakizawa M, Nosaka T. Suspension effect and dynamic evaluation of the total surface bearing (TSB) transtibial prosthesis: a comparison with the patellar tendon bearing (PTB) transtibial prosthesis. *Prosthet Orthot Int*. 1997;21:175–178.

30. Convery P, Murray KD. Ultrasound study of the motion of the residual femur within a transfemoral socket during gait. *Prosthet Orthot Int*. 2000;24:226–232.

31. Hansen AH, Sam M, Childress DS. The effective foot length ratio: a potential tool for characterization and evaluation of prosthetic feet. *J Prosthet Orthot*. 2004;16:41–45.

32. Kegel B, Burgess EM, Starr TW, Daly WK. Effects of isometric muscle training on residual limb volume, strength, and gait of below-knee amputees. *Phys Ther*. 1981;61:1419–1426.

33. Michaud SB, Gard SA, Childress DS. A preliminary investigation of pelvic obliquity patterns during gait in persons with transtibial and transfemoral amputation. *J Rehabil Res Dev*. 2000;37:1–10.

34. Czerniecki JM, Gitter AJ, Beck JC. Energy transfer mechanisms as a compensatory strategy in below knee amputee runners. *J Biomech*. 1996;29:717–722.

35. Gitter A, Czerniecki JM, DeGroot DM. Biomechanical analysis of the influence of prosthetic feet on below-knee amputee walking. *Am J Phys Med Rehabil*. 1991;70:142–148.

36. Postema K, Hermens HJ, de Vries J, Koopman HF, Eisma WH. Energy storage and release of prosthetic feet. Part 1: Biomechanical analysis related to user benefits. *Prosthet Orthot Int.* 1997;21:17–27.

37. Board WJ, Street GM, Caspers C. A comparison of transtibial amputee suction and vacuum socket conditions. *Prosthet Orthot Int*. 2001;25:202–209.

38. Johansson JL, Sherrill DM, Riley PO, Bonato P, Herr H. A clinical comparison of variable-damping and mechanically passive prosthetic knee devices. *Am J Phys Med Rehabil*. 2005;84:563–575.

39. Segal AD, Orendurff MS, Klute GK, et al. Kinematic and kinetic comparisons of transfemoral amputee gait using C-Leg and Mauch SNS prosthetic knees. *J Rehabil Res Dev*. 2006;43:857–870.

40. van der Linden ML, Solomonidis SE, Spence WD, Li N, Paul JP. A methodology for studying the effects of various types of prosthetic feet on the biomechanics of transfemoral amputee gait. *J Biomech*. 1999;32:877–889.

41. Boonstra AM, Schrama JM, Eisma WH, Hof AL, Fidler V. Gait analysis of transfemoral amputee patients using prostheses with two different knee joints. *Arch Phys Med Rehabil*. 1996;77:515–520.

42. Cappozzo A, Figura F, Gazzani F, Leo T, Marchetti M. Angular displacements in the upper body of AK amputees during level walking. *Prosthet Orthot Int*. 1982;6:131–138.

43. Macfarlane PA, Nielsen DH, Shurr DG, Kenneth M. Gait comparisons for below-knee amputees using a flex-foot (TM) versus a conventional prosthetic foot. *J Prosthet Orthot*. 1991;3:150–161.

44. Tazawa E. Analysis of torso movement of trans-femoral amputees during level walking. *Prosthet Orthot Int*. 1997;21:129–140.

45. Ryser DK, Erickson RP, Cahalan T. Isometric and isokinetic hip abductor strength in persons with above-knee amputations. *Arch Phys Med Rehabil*. 1988;69:840–845.

46. Shah SK. Cardiac rehabilitation. In: DeLisa J, Gans BM, Wash NE, eds. *Physical Medicine & Rehabilitation: Principles and Practice*. Philadelphia, Pa: Lippincott Williams & Wilkins; 2004.

47. Pagliarulo MA, Waters R, Hislop HJ. Energy cost of walking of below-knee amputees having no vascular disease. *Phys Ther*. 1979;59:538–543.

48. Schmalz T, Blumentritt S, Jarasch R. Energy expenditure and biomechanical characteristics of lower limb amputee gait: the influence of prosthetic alignment and different prosthetic components. *Gait Posture*. 2002;16:255–263.

49. Buckley J, Spence W, Solomonidis S. Energy cost of walking: comparison of "intelligent prosthesis" with conventional mechanism. *Arch Phys Med Rehabil*. 1997;78:330–333.

50. Taylor MB, Clark E, Offord EA, Baxter C. A comparison of energy expenditure by a high level trans-femoral amputee using the Intelligent Prosthesis and conventionally damped prosthetic limbs. *Prosthet Orthot Int*. 1996;20:116–121.

51. Barth DI, Schmacher L, Thomas SS. Gait analysis and energy cost of below-knee amputees wearing six different prosthetic feet. *J Prosthet Orthot*. 1992;4:63–75.

52. Skinner HB, Mote CD. Optimization of amputee prosthetic weight and weight distribution. *Rehabil Res Dev Programs Rep*. 1989;26(suppl):32–33.

53. Czerniecki JM, Gitter A, Weaver K. Effect of alterations in prosthetic shank mass on the metabolic costs of ambulation in above-knee amputees. *Am J Phys Med Rehabil*. 1994;73:348–352.

54. Gailey RS, Nash MS, Atchley TA, et al. The effects of prosthesis mass on metabolic cost of ambulation in non-vascular trans-tibial amputees. *Prosthet Orthot Int*. 1997;21:9–16.

55. Lehmann JF, Price R, Okumura R, Questad K, de Lateur BJ, Négretot A. Mass and mass distribution of below-knee prostheses: effect on gait efficacy and self-selected walking speed. *Arch Phys Med Rehabil*. 1998;79:162–168.

56. Lin-Chan SJ, Nielsen DH, Yack HJ, Hsu MJ, Shurr DG. The effects of added prosthetic mass on physiologic responses and stride frequency during multiple speeds of walking in persons with transtibial amputation. *Arch Phys Med Rehabil*. 2003;84:1865–1871.

57. Gitter AJ, Czerniecki J, Meinders M. Effect of prosthetic mass on swing phase work during above-knee amputee ambulation. *Am J Phys Med Rehabil*. 1997;76:114–121.

58. Hale SA. Analysis of the swing phase dynamics and muscular effort of the above-knee amputee for varying prosthetic shank loads. *Prosthet Orthot Int*. 1990;14:125–135.

59. Selles RW, Bussmann JB, Klip LM, Speet B, Van Soest AJ, Stam HJ. Adaptations to mass perturbations in trans-tibial amputees: kinetic or kinematic invariance? *Arch Phys Med Rehabil*. 2004;85:2046–2052.

60. van der Linde H, Hofstad CJ, Geurts AC, Postema K, Geertzen JH, van Limbeek J. A systematic literature review of the effect of different prosthetic components on human functioning with a lower-limb prosthesis. *J Rehabil Res Dev*. 2004;41:555–570.

61. Kastner J, Nimmervoll R, Kristen H, Wagner P. What are the benefits of the C-Leg? A comparative gait analysis of the C-Leg, the 3R45 and the 3R80 prosthetic knee joints. *Med Orth Tech*. 1999;119:131–137.

62. Sjödahl C, Jarnlo GB, Persson BM. Gait improvement in unilateral transfemoral amputees by a combined psychological and physiotherapeutic treatment. *J Rehabil Med*. 2001;33:114–118.

63. Sjödahl C, Jarnlo GB, Söderberg B, Persson BM. Kinematic and kinetic gait analysis in the sagittal plane of trans-femoral amputees before and after special gait re-education. *Prosthet Orthot Int*. 2002;26:101–112.

64. Davis BL, Ortolano (Cater) M, Richards K, Redhed J, Kuznicki J, Sahgal V. Realtime visual feedback diminishes energy consumption of amputee subjects during treadmill locomotion. *J Prosthet Orthot*. 2004;16:49–54.

65. Dingwell JB, Davis BL, Frazier DM. Use of an instrumented treadmill for real-time gait symmetry evaluation and feedback in normal and trans-tibial amputee subjects. *Prosthet Orthot Int*. 1996;20:101–110.