Characteristics of Gait in Hemiplegia

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The following review examines the walking patterns of patients who have hemiplegia, primarily as a result of a stroke. Attention is given to the changes in the distance and temporal factors of walking, phasic patterns and joint ranges of motion throughout the walking cycle, the ground reaction forces, joint moments of force, joint powers, energy expenditure, and muscle activation patterns. The effect of orthotic intervention on these walking parameters is also addressed. A frequently cited issue regarding the gait patterns of these patients was that their walking patterns exhibit significant deviations from normal healthy individuals. Although hemiplegia is primarily associated with unilateral motor involvement, changes in almost all of the parameters used to assess walking were evident on both the involved and uninvolved sides of the body. Last, although hemiplegia appears to reflect a single diagnostic category, there is large interindividual variability in the patterns of gait deviations, which suggests that the management and treatment of these patients need to address the unique deficits of the individual. Key words: *gait, hemiplegia, stroke, orthotics*

Normal human locomotion or walking is a complex motor task involving the interaction of the neuromuscular system acting on the musculoskeletal system. Walking patterns develop early in life and exhibit developmental changes from childhood through the aging process. It is a task that individuals often take for granted unless it is disrupted by illness, injury, or disease processes. The normal patterns of human locomotion have been examined extensively in the past, and these examinations have covered aspects of gait such as the distance and temporal factors of locomotion, the joint kinematics or the phasic patterns and joint ranges of motion over the cycle, the kinetic measures of ground reaction forces, joint moments and powers and mechanical energy expenditure, the metabolic energy expenditures, and the patterns of muscle activation over the walking cycle. Similar attention has been given to the characterization of pathological gait.

One such pathological gait pattern results from hemiplegia due to cerebrovascular accidents or stroke, head injury, or cerebral palsy. The following review will summarize the basic literature that has characterized the pathological gait patterns of patients with hemiplegia, primarily as a result of a cerebrovascular accident. This review will examine the characteristic changes in the distance and temporal factors of walking, the joint kinematics, kinetic measures, and the electromyographic (EMG) activity. Some attention will also be given to the effects of orthotic intervention on these gait patterns.

In reviewing the literature on gait changes associated with hemiplegia, we consistently found several issues. First, as would be expected, these patients exhibit gait deviations that differ significantly from those of healthy normal individuals. Second, although a diagnosis of hemiparesis may represent a so-called homogenous disease process, there is large interindividual variability in the gait disturbances,

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resulting in a very heterogenous pattern of gait disturbances. In addition, although hemiplegia is typically associated with unilateral involvement of the neuromuscular system, gait deviations have been observed in the mechanics and muscle action of both the affected and unaffected lower limbs. Because of these issues, caregivers need an understanding of the range of gait deviations that are seen in hemiplegic patients to provide more specific treatment and rehabilitation for these individuals by addressing the unique deficiencies of the individuals' neurological deficits.

Distance and Temporal Factors of Walking

The distance and temporal factors of walking of the hemiplegic patient relative to normal participants are best summarized in Tables 1 and 2, respectively. Patients with hemiplegia have been reported to exhibit reduced stride/step lengths relative to normal,[1–8] increased step lengths on the affected lower extremity,[8,9] a moderately wider base of support,[2] and slightly greater toe-out angles.[2] The temporal aspects of hemiplegic gait are characterized by increased stride times,[1–3,6,8,10] reduced walking velocity,[1,2,4,6,8–12] and reduced cadence.[1,6,9–11] Altered stance phase and swing phase periods have been observed in hemiplegic patients, with increased stance periods being observed on both lower limbs relative to normal participants.[1,10,13] On the affected lower extremity, less time is spent in stance and more time is spent in swing.[1,2,8–10,13–19] As a result, the unaffected lower extremity exhibits an increased period of stance and a reduced period of swing.[1,2,8,9,13,16,18–20] Increased periods in double support have also been reported,[1,6,10,13] with a tendency for increased double support time on the unaffected lower extremity.[18] Contradicting reports have been made relative to the amount of time hemiplegic patients spend in single limb support on each leg. Trueblood et al.[6] observed a reduction in single stance time on the involved lower extremity; whereas von Schroeder et al.[10] found that on the affected lower extremity there was an equivalent amount of time spent in single limb support relative to normal participants, but when single limb support was calculated as a percentage of the walking cycle, there was a reduction in the percentage of the cycle spent in single limb support. On the unaffected limb, these authors reported increased time in single limb support that, as a percentage of the gait cycle, was equivalent to the normal participants.[10] As a result of these bilateral differences, these patients have been shown to exhibit marked asymmetries in many of the distance and temporal factors of walking.[1,8,9,16,18]

It has been suggested that many of the gait deviations observed in the hemiplegic patient may be related to the reduced walking velocity.[7,9,20,21] Significant velocity-dependent relations have been found to exist between the distance and temporal measures of gait, including stride length and stride frequency.[7,15] Parameters that have not been found to be velocity dependent are the asymmetries in the step length, the longer duration of stance, and the shorter swing on the unaffected leg.[9] Olney et al.[15] found that walking velocity on the affected side was inversely correlated with the proportion of stance and double sup-

The joint kinematics of hemiplegic patients exhibit differences from normal participants in both the stance and swing phases of gait, and they also exhibit large interindividual variability.

port, with faster walking velocities being associated with shorter periods of double support when body weight was being transferred to the uninvolved limb.[15] Stance on the unaffected lower leg and contact time of the affected foot were also found to be correlated with walking velocity.[15] Carlsoo et al.[20] found that differences in stance between the affected and unaffected lower limbs were smaller in participants that exhibited more normal walking velocities.

Improvements in the absolute values of the distance and temporal measures of walking have been observed with increasing time from the onset of the stroke,[1,8,21] the stage of motor recovery, and clinical measures of functional ambulation.[1,22] With respect to the time from the onset of the infarct, the most rapid improvements have been observed over the first 6 weeks to 3 months, with slower changes being evident up to 1 year.[8,10,23,24] The time course of recovery in walking abilities has been found to be influenced by the degree of walking dysfunction and the degree of leg paresis at the time of admission.[23] Mizrahi et al.[21] observed a reduction in contact time of the plegic limb in 80% of the participants and a similar reduction in the double contact time in a 2-month poststroke gait follow-up. There was an increase in walking velocity in 90% of the participants, and there was an increase in the average stride length in 75% of the participants. Time and distance asymmetries improved in 40% and 70% of the participants, respectively.[21]

Relative to the level of motor recovery, Brandstater et al.[1] noted that the distance and temporal variables of the hemiplegic patients tended to approach those of normal participants with increasing stages of motor recovery, which were measured using the Brunnstrom method with the Fugl-Meyer scoring. These authors found that it was difficult to discriminate between stages 3 and 4 with most of the variables, with the exception of the symmetry of the swing period and the stance:swing ratio symmetry. However, increasing symmetry was observed with increased motor recovery.[1] Gaviria et al.[14] also indicated that the major factor affecting stride length and walking velocity was the severity of motor involvement. These authors suggested that total support, single support, and step duration were indicators of the severity of motor involvement.[14] Holden et al.[22] reported that the amount of assistance (physical assistance to independent ambulators) needed for functional ambulation was significantly related to measures of walking velocity, cadence, stride/step length, and the ratio of the stride length to lower extremity length. It was suggested that these five measures were meaningful in measuring treatment outcomes in these patients.

Joint Kinematics

Kinematics analyses are concerned with providing a detailed description of human movement without regard for the actual forces that are causing the movement.[25]

This type of analysis may include the linear and angular displacements of the body and its segments, velocities, and accelerations.[25] This section on joint kinematics will examine the phasic patterns of joint movements or the shape of the joint angle curves over the walking cycle and the ranges of motion of specific joints. As was evident in the analysis of the distance and temporal factors of walking, the joint kinematics of hemiplegic patients exhibit differences from normal participants in both the stance and swing phases of gait, and they also exhibit large interindividual variability. Olney et al.[15] examined the joint angle profiles or the shape of the joint angle curves on both the affected and unaffected lower limbs of the hemiplegic patients walking at slow, medium, and fast speeds and found that the joint profiles exhibited most of the phases found in normal gait. Differences were, however, found in the amplitudes of the curves or actual joint ranges of motion of the curves. Reduced ranges of motion were observed even in the fastest speed group, with the lowest amplitudes being evident on the affected lower extremity.[15] The following sections will address the joint ranges of motion of the hips, knees, ankles, upper limbs, trunk, and pelvis during the cycle. Attention will be given to the joint movements during stance and subsequently during swing.

During the stance phase, the sagittal plane range of motion at the hip appears to exhibit a greater variety of atypical joint motions across groups of hemiparetic patients than does swing phase hip movement. During the stance phase, it has been reported that these patients may exhibit hip motions that range from not being significantly different from normal[12] to having reduced hip flexion at initial contact[12] or to exhibiting more hip flexion than normal at initial contact.[2,9,13] It has also been reported that the magnitude of peak hip extension during stance, particularly late stance and push-off, is reduced relative to normal participants[9,12,15] and that the hip may actually be flexed at toe-off.[2]

Three different types of knee patterns during stance have been reported in hemiplegic gait analyses. Some participants have been found to exhibit increased knee flexion during the stance phase relative to normal participants,[4,6,12] particularly at initial contact.[2] Other participants have been observed to exhibit reduced knee flexion during early stance followed by knee hyperextension in late stance and delayed movement into knee flexion in preparation for swing.[12,13] The magnitude of knee flexion at toe-off was also found to be less than normal.[2] The third group of participants exhibited excessive knee hyperextension throughout most of the stance phase.[6,12,26,27] It has been suggested that the knee hyperextension that is observed during the stance phase in different patients may be caused by different mechanisms.[12] In some cases, the knee hyperextension may be caused by early calf muscle activity, resulting in the lower leg being pulled posteriorly and forcing the knee into hyperextension.[12] In other patients, it has been suggested that the knee hyperextension is a compensatory mechanism to provide a stable limb for weight bearing, in which case the hamstrings are acting to stabilize the knee.[12]

The literature on the stance phase ankle movements have found several patterns of initial floor contact. Initial contact on the hemiparetic side is typically made with the

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foot flat,[10,17,26] with the ankle in a plantarflexed position,[2,12,13] which results in initial toe contact[26] or a moderate decrease in toe elevation.[26] After initial contact, the ankle has been reported to exhibit irregular movements into dorsiflexion in stance,[12] reduced dorsiflexion in midstance and push-off,[9] and/or increased plantarflexion in stance.[12] Reduced ankle plantarflexion at toe-off on the paretic lower extremity has also been observed.[2,9]

The swing phase patterns of hip, knee, and ankle motions on the hemiplegic side have been characterized by limited or reduced hip flexion[2,12,17] and an upward tilt of the hip,[26] a lack of or reduced knee flexion,[2,6,9,12,17,26] and reduced dorsiflexion or continuous ankle plantarflexion.[2,9,12,17] The increased leg length produced by the limited hip and knee flexion and reduced ankle dorsiflexion results in reduced floor clearance by the foot during swing, which produces dragging of the toes or circumduction of the leg.[2,9,12,17,26] As a result, hemiplegic gait has been characterized as exhibiting a stiff knee during swing.[9,17] It has been suggested that the upward tilt of the hip is also a compensation to ensure toe clearance as the involved lower extremity is swinging forward.[17,26] The limited range of knee flexion during swing is not limited to the involved lower extremity; it is also seen in the uninvolved lower extremity in cases of severe impairment.[15,17]

Although little has been reported on the movements of the upper limbs of the hemiplegic patient over the walking cycle, it has been reported that there are alterations in the upper extremity kinematics. It has been reported that movements of the arms are slight,[26] with the shoulders remaining relatively fixed in extension and the elbow remaining flexed.[12]

Little attention has also been given to the trunk movements during the walking cycle. It has been reported that the trunk is flexed forward during stance, which moves the center of gravity forward when there is recurvatum of the knee.[28] Carlsoo et al.[20] reported that forward trunk lean at push-off of the involved leg was a compensation for weak musculature on the paretic leg and that this forward trunk lean was not evident during push-off of the uninvolved leg. It has also been reported that there is lateral shift of the trunk over the uninvolved stance to assist in weight shifting during swing of the affected leg.[28]

The ranges of joint motion over the walking cycle in hemiparetic patients have also been found to be influenced by walking velocity. Olney et al.[15] found that the magnitude of knee flexion during swing and maximum hip extension throughout the gait cycle decreased with reduced walking velocity, particularly on the affected side. The maximum ankle plantarflexion on the unaffected side was also found to be related to walking velocity.[15] On the unaffected leg, increasing velocity was associated with increasing angles of ankle plantarflexion.[15] Wagenaar and Beek[7] also observed significant speed-dependent relationships between the total range of and phase differences in trunk rotation with changes in walking speed in both hemiplegic and normal participants, whereas the maximum amplitude of pelvic and thoracic rotations was not speed dependent. The magnitude of thoracic rotations was significantly larger than in the healthy participants at velocities of 0.75 and 1.0 m/s, whereas no significant differences were observed in the total range of pelvic rotation, the total range of trunk rotations, nor the phase differences in the trunk rotations.[7] These authors observed three different patterns of gait deviations in the hemiplegic patients based on the speed of walking. One group of patients exhibited excessive pelvic rotations, a second group exhibited a lack of timing in the trunk rotations, and the third group exhibited difficulties with relative pelvic as well as trunk rotations.[7] As a result, it was suggested that walking speed should be a basic variable in the evaluation of the gait in stroke patients.[7]

Kinetic Analysis

Kinetic analysis is the study and analysis of the forces, powers, and resultant energetics used in human locomotion.[25] Knowledge of the patterns of forces is considered necessary for understanding the cause of any movement.[25] Included in kinetic analyses are the ground reaction forces, the joint moments, joint powers, and mechanical energies.

Ground reaction forces

Ground reaction force is the force exerted by the body onto the floor surface and is commonly measured by a force transducer embedded in the walking surface.[25] This type of analysis provides information on the vertical forces imparted by the foot and the anterior–posterior and medial–lateral shear forces. A number of researchers have examined the force patterns of the hemiplegic patient during walking and found that these patients exhibit patterns that differ from normal healthy participants on both the involved and uninvolved legs and they exhibit large interindividual variability.[19,20,27] As a result, there is a lack of bilateral symmetry in the force curve patterns on both the involved and uninvolved lower limbs.[20,29] However, it was found that hemiplegic patients exhibit their own characteristic pattern.[20]

Vertical force patterns

The results of the vertical force curves obtained from stroke patients suggest that there is reduced vertical loading and more variable loading of the paretic foot at initial contact.[19,27] The results of the vertical forces obtained from the uninvolved leg have been found to exhibit similar characteristics.[19,27] Carlsoo et al.[20] suggested that three characteristic vertical force patterns are seen in hemiplegic gait. The first classification of participants exhibited vertical curves similar to normal participants with two vertical peaks occurring at weight acceptance and push-off and an intermediate trough occurring during midstance. The second group of patients exhibited a vertical force component that was relatively constant throughout the stance phase and exhibited several irregularly occurring peaks and crests. The third group exhibited a single vertical peak in the first part of the stance phase that gradually decreased to zero during late stance.[20] Lehmann et al.[9] indicated that the maximum vertical forces of the hemiplegic patients were not significantly different from normal participants walking at similar walking speeds, although the hemiplegic patients exhibited either three peaks or a plateau of vertical loading versus the normal bimodal shaped curve.[9] Unlike the preceding reports, Hesse et al.[29] reported that there was increased vertical loading after initial contact on the involved leg relative to the uninvolved leg. These authors also observed reduced vertical push-off forces at terminal stance, delayed loading at initial contact, and premature unloading at terminal stance while in stance on the involved limb.[29]

Anterior–posterior shear forces

The anterior–posterior shear forces have also been found to differ from those of normal participants.[20] The retrograde force after initial contact was found to be of short duration on the involved leg, and the distance between the horizontal peaks was increased.[20] The magnitudes of the anterior–posterior forces were greater on the affected lower extremity than on the unaffected limb.

Medial–lateral shear forces

The medially directed shear force at initial contact was typically nonexistent.[20] During most of stance, the forces were directed laterally on both the affected and unaffected legs, which indicates that the body's center of gravity was located medial to the support foot as would be seen in healthy normal participants.[20] The participants were equally divided in terms of the location of the center of gravity relative to the support leg.[20] One half of the hemiplegic patients positioned the body's center of gravity closer to the healthy leg when it was in stance than when the affected limb was in stance, whereas the other hemiplegic participants exhibited just the opposite.[20] However, Iida and Yamamuro[30] indicated that the medial–lateral displacement width of the center of gravity of the body was significantly greater in hemiplegic patients than in healthy normal participants and that the magnitude of the center of gravity displacement was greater in patients who were categorized in the lower Brunnstrom stages.

Plantar dynamics

Gaviria et al.[14] examined the plantar dynamics of hemiplegic patients and observed most of the differences on the uninvolved leg. The plantar dynamics of the involved leg exhibited a transfer of initial contact from the hind to the forefoot, increased lateral plantar support, limited roll-over, and reduced or absent push-off at terminal stance.[14] There was a tendency for heel support to disappear on the paretic side with reduced functional abilities. The foot flat interval was found to increase to cover the entire stance phase, particularly on the uninvolved side. It was suggested that hindfoot dynamics were not discriminatory of motor status. Midfoot dynamics, however, were found to be indicative of motor involvement, with the transfer area increasing and forces increasing on the uninvolved foot with increasing severity of motor involvement. The forefoot dynamics indicated that on the uninvolved side there was a delay in the maximal anteromedial force and in the anteromedial and anterolateral force occurrence. On the involved side, the maximal anteromedial force was lower and the anterior force difference increased, which resulted in altered propulsion dynamics.[14]

Joint moments

The joint moments of force or torque are the net result of the muscular, ligamentous, and friction forces that act to affect joint rotation.[31] The convention for describing the joint moments is to refer to positive moments as being extensor or plantarflexor because they attempt to push the body upwards away from the ground and flexor moments as being negative because they tend to collapse the limb.[31] To date, little attention has been given to the joint moment profiles in the hemiplegic patient. Several studies have reported that the knee moments of hemiplegic patients differed significantly from normal participants.[19,27] Unlike normal participants who exhibited a negative flexor moment during early stance followed by a positive extensor moment throughout the remainder of stance, the hemiplegic patients exhibited a positive extensor moment throughout the cycle on the affected lower extremity.[19,27] Similar knee moment patterns have been observed in the uninvolved leg.[19,27] The knee moment curves observed by Lehmann et al.[9] in hemiplegics and normal participants walking with a similar velocity also indicated that the hemiplegic patients exhibited an extensor moment throughout most of stance. In most of the hemiplegic patients, the mean total knee extension moment at midstance was higher than in the normal group, although this difference was not statistically significant. The mean total knee flexion moment during midstance was similar in the two groups. This pattern of moments occurred despite the fact that the knee was flexed during midstance, and it was suggested that this was a result of increased hip flexion to move the center of gravity of the trunk forward.[9]

The use of AFOs has been associated with increased walking velocities, increased step length on the involved lower extremity, reduced cadence, increased stance on the involved lower extremity, and changes in the duration of the phases of stance relative to not wearing an AFO.

Mechanical powers

Mechanical power examines the concentric and eccentric phases of muscle contractions as the muscles generate and absorb mechanical energy that is needed for movement.[31] Power analyses obtained from normal healthy participants have found that there are two important power phases at the ankle.[31] Between 5% and 40% of the walking cycle, there is a small negative A1 burst of power that is the result of energy absorption at the ankle, followed by large positive A2 burst at push-off. This A2 burst, due to the ankle plantarflexors, is the most important energy generation resulting in 80%–85% of the energy generated during the cycle.[32] The knee has been found to exhibit four phases of power absorption and generation. The first phase of power absorption, K1, occurs over the first 15% of the cycle and is associated with eccentric work at weight acceptance. From 15%–40% of the cycle as the knee is extending, there is a positive K2 burst resulting from concentric contraction of the knee extensors. At push-off and early swing, there is energy absorption caused by eccentric contraction of the quadriceps to provide some control for the collapsing knee joint. The fourth power burst, K4, which is a power absorption, occurs during terminal swing and is associated with eccentric contraction of the hamstrings. The hip exhibits two bursts of generation (H1, H3) and one of absorption (H2). H1 occurs following initial contact and is associated with concentric contractions of the hip extensors. H2 occurs throughout the remainder of stance and is associated with eccentric work of the hip flexors. The highest level hip power generation occurs at H3, which is the result of concentric work of the hip flexors during late stance and early swing, acting to pull the leg forward.[31,32]

The joint powers obtained from 30 hemiplegic patients showed that, with one exception, the patients exhibited a positive A2 burst at push-off on both the involved and uninvolved ankles due to the ankle plantarflexors.[15] At the knee, a small positive K2 burst by the knee extensors was evident on the affected side for those individuals who were walking at fastest walking speeds. The K3 power burst was also evident on the affected side and in the patients with the fastest walking speeds; the negative power burst was larger than that of normal participants. It was suggested that the correlation between the K3 burst and speed indicates that patients with better walking abilities flex their knees at terminal stance while there is still weight on the foot. The H1 energy generation by the hip extensors after initial contact was positive only on the affected side, whereas the H3 pull-off power burst was found in all patients on the involved limb but was lower in the patients with the slowest walking velocity. It was found that the proportion of work performed by the unaffected and affected sides was a ratio of 60:40, primarily due to the work differences between the affected and unaffected ankles and differences at the hip. It was also suggested that the longer duration of swing on the affected side was due to the fact that less power was being put into the affected leg during late stance and early swing.[15] Similar results were seen in an earlier single case study in which low levels of power generation were observed on both legs.[4] Small amounts of positive work were observed in the hip flexor muscles (H3) and ankle plantarflexors (A2) on both limbs and in the knee extensor muscles (K2) on the unaffected leg.[4]

Energy expenditures

Hemiplegic walking has been found to require between 50% to 67% more metabolic energy expenditure than that of normal participants at the same walking velocity.[33] As opposed to metabolic energy expenditures, mechanical energy expenditure is a measure of the body's ability to do work.[25] The mechanical energy of a body segment is made of potential and kinetic translational and kinetic rotational energy components.[25] In the human body, there may be exchanges of energy within a segment (exchanges between potential and kinetic energy) or transfers of energy between adjacent segments. The total energy of the body is calculated by algebraically summing up the energies of the individual body segments at each point in time.[25] Several studies have examined the mechanical energy costs associated with hemiplegic gait and found the costs to be higher than that of normal gait, ranging between 0.76 to 3.9 J/kg/m, whereas normal participants have an energy cost of 1.1 J/kg/m with 70% energy conservation.[4,30,34,35]

Olney et al.[34] found that total energy conservation was low in these patients, ranging between 22% to 66%. The energy levels of the legs were low in magnitude, unlike those of healthy participants, but the energy pattern was similar to that of healthy participants. The total energy pattern was dominated by the actions of the head, arms, and trunk (HAT). Three types of disturbances were identified in the HAT that produced the reduction in energy conservation. The first disturbance was a lack of within-segment energy exchange, which means that that there was little exchange between the potential and kinetic energy. The second disturbance was a result of low magnitudes of kinetic energy that resulted in limited energy exchange within the HAT, although there was some mirroring of the energy patterns. In the third pattern, instead of the normal double-humped total energy pattern associated with the large energy contributions of the swing leg, there was a single-humped curve that coincided with the swing of the affected limb. This pattern was termed the hip-hiking pattern that was associated with the rise and fall of the potential energy curve caused by the hip hiking of the trunk. It was suggested that management of the first type of disturbance should be directed at regaining the sinusoidal pattern of rise and fall of the body over the walking cycle, thus improving the conservation of energy. In the case of the second disturbance where there is low kinetic energy with some energy exchange, treatment should attempt to increase the walking velocity by increasing the power output of muscle groups. The last type of energy disturbance requires attention to reduce the magnitude of hip hiking, while also attempting to increase walking velocity and muscle function.[34]

Electromyography

A great deal of attention has been given to the patterns of muscle activity or EMG during the walking cycle in hemiplegic patients. As with many of the previously discussed gait parameters, the magnitudes and phasic patterns of the muscles in the lower extremity have been found to differ significantly from those of normal healthy individuals,[12,16,19,20,27,36,37] to exhibit bilateral differences represented by abnormalities on both the involved[12,16,17,19,20,27,36,37] and uninvolved lower limbs,[17,19,20,27] and to exhibit marked interindividual variation over the walking cycle.[16,26,36,37] As a result, there is no way to provide a compilation of the EMG patterns exhibited by hemiplegic patients. However, some common general characteristics have been identified that include a reduction in the magnitude of the EMG obtained from the muscles on the paretic limb, premature onset and prolonged duration of firing over the walking cycle, and peaks of activity that differed from normal.

It has been suggested that the muscles of the lower extremity tend to align into the flexor and extensor locomotor patterns, having a preference for one phase of the walking cycle, which may not be the typical phase associated with that muscle group.[5] Muscles associated with the extensor locomotor patterns are the gluteus medius, gluteus maximus, vastus lateralis, tibialis posterior, long head of the biceps, vastus intermedius, medial hamstrings, and soleus. Muscles with dominant swing phase activity were the iliacus, sartorius, tibialis anterior, tensor fascia latae, rectus femoris, and extensor digitorum longus.[5] These muscles were found to initiate and terminate activity at the same time in their respective phases of the gait cycle.[5] Functionally significant stance phase EMG levels observed in 75% of the patients included the gluteus medius, gluteus maximus, vastus lateralis, and soleus. During swing, 69% of the patients exhibited functionally significant EMG levels in the tensor fascia latae, tibialis anterior, and iliacus.[5] These authors did, however, indicate that all of these muscles with the exception of the gluteus medius were continuous throughout the gait cycle in some patients.[5]

Another study, which suggested that the abnormal EMG patterns are related to the flexor and extensor mechanisms, classified the EMG patterns into six categories on the basis of flexor and extensor muscle activity.[36] The disturbances in the EMG patterns decreased from Category 1 to 6. Category 1 participants exhibited no activity in the tibialis anterior during stance or swing and continuous activity in the triceps surae. Participants classified as Category 2 exhibited activity in the tibialis anterior during swing and low activity in the triceps surae during stance and early swing. Both muscle groups were active during stance and midswing. Participants in Category 3 exhibited activity in both the triceps surae and tibialis anterior during stance and the middle of swing, although the magnitude of the triceps surae was less during swing than in stance. Category 4 was characterized by tibialis anterior activity during swing and the first and third parts of stance. The triceps surae exhibited similar stance phase activity and were also active during terminal swing. In Category 5 patients, both muscles were active during stance with some reciprocal activity. The tibialis anterior was also active during initial swing. Patients classified as being in Category 6 exhibited patterns similar to normal with triceps surae activity in midstance and tibialis anterior activity at initial and terminal swing. It was suggested that the preservation of the extensor mechanism provides some support during stance, and the flexor mechanism is essential for swing.[36]

Peat et al.[16] found that all the muscles on the involved lower extremity tended to exhibit peak EMG activity at the same time during stance after weight acceptance on the involved limb. Low levels of activity were observed at initial contact in the tibialis anterior, triceps surae, and vastus lateralis and increased to a peak at midstance. The medial hamstrings, however, were active for a longer period of time with peak activity being observed between foot flat and midstance. During swing, all of the muscles exhibited low levels of muscle activation.[16]

Knutsson and Richards[26] suggested that the EMG in the hemiparetic patients could be classified into three types of motor disturbances. Patients classified as Type I exhibited mild gait disturbances, with EMG disturbances primarily occurring in the levels and phasic patterns of the tibialis anterior and calf muscles. The EMG levels in both muscle groups were lower than normal during the stance phase, and the calf muscles exhibited premature activation. It was suggested that this pattern was related to a lowered threshold of stretch reflex activation. Type II patients were characterized by an absence or significant reduction in the EMG patterns of two or more muscle groups on the involved lower extremity. The distal muscles of the involved leg demonstrated more disturbed activation patterns than the proximal muscle groups. Of the nine participants in this category, there was a range of EMG disturbances. In addition, these patients exhibited a range of walking capabilities but tended to exhibit knee hyperextension during stance and a lack of knee flexion during swing. Those classified with a Type III EMG pattern exhibited coactivation of several muscles with disorganized shifts in the activity in the different muscle groups, primarily occurring from late swing through the stance phase. However, there was no significant decrease in the level of EMG activity. It was suggested that the moderate gait disturbances seen in the patients in this group were not related to stretch reflexes or to a lack of muscle activity.[26]

In a study of the EMG activity of 27 hemiplegic patients, Waters et al.[37] observed abnormal EMG almost always in the gastrocnemius, soleus, tibialis anterior, tibialis posterior, flexor hallucis longus, flexor digitorum longus, and peroneus brevis, with activity occurring at the same time during the walking cycle. Muscle activity was categorized as no activity; continuous activity in stance and swing; and phasic activity that may be premature, premature and prolonged, or within normal limits. Phasic activity was observed in the gastrocnemius and soleus of 78% and 89% of the patients, respectively, although both muscles were activated prematurely during late stance and early swing. The tibialis anterior exhibited continuous activity in 59.3% of the participants. The EMG activity in the tibialis posterior was distributed across the three major categories, with 44% of the participants exhibiting premature phasic patterns. Similar patterns were observed in the flexor hallucis longus and flexor digitorum longus.[37]

Several studies have found abnormal EMG patterns in the uninvolved leg.[20,27] Marks and Hirschberg[27] indicated that the EMG activity in gluteus maximus, gluteus medius, adductor longus, semitendinosus, vastus lateralis, medial gastrocnemius, and tibialis anterior were increased in all muscle groups and that the phasic patterns of these muscles resembled those on the involved limb more than in normal participants. Carlsoo et al.[20] found prolonged periods of muscle activation in the pretibial, calf, quadriceps, and hamstrings muscles of the uninvolved leg. It was found that often all of these muscle groups were continuously active during stance, particularly when the walking velocity was slow.[20]

Orthotic Intervention

Orthotics are frequently prescribed to improve the walking patterns of patients with residual muscle weakness and/or spasticity after a stroke. Although the primary focus of this review was to examine the previous

Table 1. Summary of the distance factors of walking

Author	**Time from onset/recovery time/functional ambulation category**	**Stride length (m)**		**Step length (m)**		**Step length affected (% stride length) (±*SD*)**	**Step width (cm)**	**Toe-out angle (°) Affected/Unaffected**
		Affected (±*SD*)	**Unaffected**	**Affected**	**Unaffected**			
Data from normal healthly elderly Murray et al., 1969[45]		1.46 ± .16					9 ± 4	8 ± 7
Wall & Ashburn 1979[8]	3 months post 6 months post 9 months post	.676 ± .07 .736 ± .16 .724 ± .19				67.2 ± 12.24 60.6 ± 16.18 59.4 ± 10.02		
Brandstater et al., 1983[1]	Stage 3 Stage 4 Stage 5 Stage 6	.41 ± .12 .44 ± .16 .73 ± .20 .91 ± .09						
Lehmann et al., 1987[9]				.37	.34			
Von Schroeder et al., 1995[10]		1.1 ± .6						
Trueblood et al., 1989[6]		.58 ± .20						
Burdett et al., 1988[2]		.58 ± .19		.31 ± .09	.27 ± .10		12.9 ±4.3	10.7 ± 8.0/7.5 ± 6.5
Iida & Yamamuro 1987[30]	Stage 3 Stage 4 Stage 5 Stage 6	.90 ± .072 .95 ± .099 .90 ± .05 .997 ± .12						

Table 2. Summary of the temporal factors of walking

Author	Time from onset/recovery time/functional ambulation category/level of assistance	Stride time (s)	Stance: swing ratio Affected/Unaffected	Velocity (m/s)	Cadence (steps/m)	Stance (%GC) Affected/Unaffected
Data from normal healthy elderly Murray et al., 1969[45]		1.08 ± .11		1.39 ± .23		
Wall & Ashburn, 1979[8]	3 months post 6 months post 9 months post	1.88 ± .32 1.69 ± .21 1.60 ± .24		.378 ± .10 .442 ± .12 .462 ± .16		
Brandstater et al., 1983[1]	Stage 3 Stage 4 Stage 5 Stage 6	2.8. ± .7 2.8 ± .7 2.0 ± .4 1.4 ± .1	2.4 ±1.3/8.1 ± 3.2 2.6 ± 1.7/6.0 ± 3.1 1.7 ± .5/2.9 ± .7 1.7 ± .3/2.1 ± .4	.16 ± .07 .17 ± .08 .40 ± .15 .65 ± .11	45 ± 9 45 ± 9 63 ± 11 85 ± 9	
Cozean, Pease, & Hubbell, 1988[3]		2.54		.52		
Mizrahi et al., 1982[21]				.17 ± .08		
Bohannon, 1987[11]				.39 ± .26	63.8 ± 26.2	
Von Schroeder et al., 1995[10]		1.6 ± .7		.44 ± .23	84.8 ± 22.4	67.0 ± 6.6/70.8 ± 8.6
Trueblood et al., 1989[6]		2.5 ± 1.0		.17 ± .11	54.7 ± 18.9	64.6 ±11.0/not reported
Burdett et al., 1988[2]		2.7 ± .9		.23 ± .11		
Holden et al., 1984[22]	Level of assistance 1 2 3 4 5			 .14 ± .08 .23 ± .14 .24 ± .17 .38 ± .23 .64 ± .28	 34.0 ± 12 40.0 ± 19 39.0 ± 20 48.0 ± 22 69.0 ± 17	
Iida & Yamamuro 1987[30]	Stage 3 Stage 4 Stage 5 Stage 6			.31 ± .01 .38 ± .09 .35 ± .04 .41 ± .09	68.7 ± 3.4 79.0 ± 14.7 78.9 ± 9.3 81.2 ± 9.6	
Peat et al., 1976[16]						67/80

Note: GC = gait cycle.

research on the walking patterns of these patients, it would be an oversight not to provide at least a brief overview of the effects of orthotics on their walking patterns. The following section is by no means a comprehensive analysis of the literature on the effects of orthotic intervention on gait, but it is an examination of some of the salient issues related to orthotic prescription. The use of orthotics is predicated on reducing the amount of ankle varus at initial contact; restraining excessive spasticity, exaggerated motions, or contractures; and providing knee stability.[38] Lehmann[39] suggested that there should be three basic goals in prescribing an ankle-foot orthosis (AFO): (1) to provide adequate medial–lateral stability during stance, (2) to provide adequate toe clearance during swing, and (3) to approximate normal gait while reducing energy expenditure. He also suggested that the effect of the orthosis on knee stability was an important issue.

The use of AFOs has been associated with increased walking velocities,[9,33,40] increased step length on the involved lower extremity,[2,40] reduced cadence,[40] increased stance on the involved lower extremity,[40] and changes in the duration of the phases of stance relative to not wearing an AFO. Lehmann et al.[9] indicated that the improvements in walking velocity and the duration of stance phase characteristics were dependent on the ankle angle of the AFO. It was found if the AFO was set in 5 degrees of dorsiflexion, there was a significant improvement in walking velocity and an increase in the duration of heel-strike phase relative to a no-AFO condition or when the AFO was set in 5 degrees of plantarflexion.[9] Either ankle setting of the AFO was associated with a significant reduction in the midstance phase, whereas 5 degrees of AFO ankle plantarflexion resulted in an increase in the duration of the push-off phase, relative to not wearing an AFO.[9] In contrast to the preceding results, Burdett et al.[2] did not find a significant change in the walking velocity or other distance and time factors of walking between an air splint, an AFO, and no appliance.

Kinematic analyses suggest that sagittal plane hip and knee joint angles at specific points in the gait cycle are not affected by the use of an AFO.[2] As would be expected, changes have been observed in the ankle joint angles. In contrast, Yamamoto, Miyazaki, and Kubota[41] observed changes in the range of knee flexion when the AFO was worn, which indicates that knee flexion during initial stance increased when an AFO was worn. Burdett et al.[2] examined 19 hemiplegia patients who were walking without an AFO, with an AFO, and with an air stirrup splint. These authors found that during the stance phase, there was less plantarflexion at initial contact when the patients wore the AFO relative to the other two conditions. Patients who wore either appliance had less ankle plantarflexion during midswing than when they walked without any brace. When the air splint was worn, there was significantly less plantarflexion at toe-off than without any brace. No significant differences were found in the calcaneal angles at foot strike, after foot strike, or at heel-off among the three walking conditions.[2] These results indicated that the AFO assisted the dorsiflexors during initial contact and midswing but did not have significant effect on the magnitude of inversion at initial contact.[2] The air stirrup splint resulted in significantly reduced ankle plantarflexion at toe-off and during swing and significantly less inversion at initial contact. As a result, it was concluded that the air stirrup splint may be effective in controlling medial–lateral stability. Lehmann et al.[9] supported this notion and suggested that AFOs increase medial–lateral stability during stance and permit toe clearance during swing.

Relative to the energy costs associated with walking while wearing an AFO, Corcoran et al.[33] found that walking with a brace significantly reduced the metabolic energy expenditure, measured by oxygen consumption, relative to a no-brace condition, regardless of whether or not the brace was made of metal or plastic. There was a trend toward lower oxygen consumption with the plastic AFO relative to the metal AFO, although this difference was not significant. The comfortable walking speed for the hemiplegic patients was 46% slower than normal participants without the brace and 39% slower than normal when either brace was used. However, the shape of the energy expenditure–speed relationship was the same for the hemiplegic patients and the normal healthy participants, which suggests that the patients require more energy expenditure for a given increase in walking velocity.[33]

Another issue pertinent to orthotic prescription is the knee moment at initial contact.[9,39,42] It has been suggested that there needs to be a trade-off between the amount of toe pickup provided during swing and the amount of knee stability provided during stance.[39,43] In a group of normal participants wearing an AFO set in 5 degrees of dorsiflexion, the knee was thrust forward after initial contact resulting in an increase in the knee flexion moment.[42] When the orthosis was set in 5 degrees of plantarflexion, there was an increase in the knee extension moment resulting in knee hyperextension during stance.[42] In a similar study performed on hemiplegic patients, the mean total knee flexion moment during midstance was found to be significantly greater with the AFO set in 5 degrees of dorsiflexion relative to the AFO set in 5 degrees in plantarflexion or walking without an AFO.[9] The more dorsiflexion that was provided for toe clearance during swing, the greater the knee instability.[39] Yamamoto et al.[41] supported this idea and suggested that the dorsiflexion corrective moment required to prevent foot drop during swing was small, whereas the dorsiflexion corrective moment need for initial contact was relatively large. These authors suggested that the amount of corrective dorsiflexion moment at initial contact was the most important factor in selecting the function of an AFO suited for a particular patient.[41]

Although it is beyond the scope of this review to examine the literature comparing the efficacy of various types of AFOs on the gait patterns of hemiplegic patients, it has been suggested that consideration should be given to the flexibility of the AFO and the severity of ankle plantarflexor spasticity.[43] The walking patterns of hemiplegic patients were examined as they wore five different types of plastic AFOs. All of the AFOs improved the walking patterns and were associated with limited plantarflexion at initial contact, reduced dorsiflexion at push-off, and reduced dorsiflexion during swing. More rigid AFOs were associated with greater knee flexion moments after initial contact. In addition, it was suggested that more rigid AFOs may be appropriate for patients with weak plantarflexor muscles or insufficient push-off, by providing resistance for dorsiflexion at push-off and thereby simulating ankle plantarflexor muscle function and providing greater plantarflexion resistance during swing to prevent toe drag.[43] Several case reports have suggested that tone-inhibiting AFOs may also improve the walking patterns of hemiplegic patients more than the traditional AFO.[40,44] The tone-inhibiting AFOs were associated with increased walking velocity,[40] increased step length on the involved side,[40] reduced total stance duration,[44] and improved stability in the foot measured by an increase in both the total foot contact area with the ground and the total force generated through the foot[44] relative to a traditional AFO.

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