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The Effects of Different Ankle-Foot Orthoses on the Kinematics of Hemiplegic Gait

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INTRODUCTION

One of the primary goals in the rehabilitation of a stroke patient is to develop a safe and efficient gait. A common gait problem exhibited by these patients is a dropfoot caused by spasticity of the calf musculature and/or weakness of the pretibial muscles. This problem is frequently corrected by means of an orthotic device, commonly, an ankle-foot orthosis (AFO). Currently, there are several types of AFOs to select from, the most frequently prescribed for the hemiparetic patient being:

- klenzak-AFO (KL-AFO)
- plastic shoehorn AFO (PL-AFO)
- spring wire AFO (SW-AFO).

These AFOs have similar biomechanical functions and contraindications.[1,25] The prescription of an AFO is usually based upon a physical assessment of the patient, including an assessment of gait, and the biomechanical characteristics of the AFO.

Often the initial gait assessment and the evaluation of the effect of the orthotic device during ambulation are by subjective means. The quality of assessment is, to a large degree, dependent upon the experience of the clinician,[9] but at best must be regarded as an unreliable clinical skill.[24] Therefore, it is not surprising that subjective assessment often does not detect subtle changes, due to the orthosis, or resulting from the long rehabilitation process. Nor does subjective assessment allow for accurate comparison, which is essential if the clinician is to determine improvement or deterioration requiring remedial action. It is for these reasons that there has been a trend towards implementing objective measurement of gait to improve subjective assessments in clinical facilities.[29] To quantify the gait pattern of a patient in a clinical setting, the measuring devices must be easy to apply and non-restrictive, and the results must be immediately available and easy to interpret.[31] Two gait variables which have been shown to be useful in the assessment of hemiplegic gait and which are fairly easy to measure are the temporal (T) and distance (D) parameters, such as stride time and step length. One technique that measures the T/D parameters simultaneously is a specialized walkway.[7,30] More commonly, however, these parameters are measured independently. Ink pads, or some similar technique may be used to leave imprints on the floor indicating the positions of foot/floor contact, from which measurements of the spatial parameters of gait can be made.[5] Electrical switches of various forms can be used to determine the duration of foot/floor contact, thereby allowing the temporal parameters of gait to be calculated.[10] By utilizing a number of switches distributed on the sole of the shoe, footswitches can provide information on the sequencing pattern of foot/floor contact.[33]

Many studies have analyzed hemiparetic gait using different techniques, such as the electromyographic patterns of the muscles of the lower limb and stride kinematics.[11,21,22,28] The results from the stride kinematics indicate that hemiplegic subjects walk significantly slower and take shorter stride lengths than able-bodied persons. The hemiparetic subjects spend less time in stance on the affected limb and thus demonstrate a greater degree of asymmetry in their gait pattern.[29] Gait asymmetries within the various component parts of the support phase have also been shown in patients with residual hemiplegia.[32] The rehabilitation process[18] and the degree of motor recovery[3] have been shown to improve the hemiparetic gait. With rehabilitation and increased motor recovery, using the Brunnstrom method with Fugel-Meyer scoring, hemiparetic subjects have been shown to walk faster with improved symmetry (using single support time as a measure of symmetry). Single support time and stride length increased on the affected side, while double support time (period between heelstrike of the sound leg and toe-off of the paretic leg) decreased.[3] The foot/floor contact patterns for hemiparetic subjects without orthoses have been studied using heel and toe footswitches.[8,11,28] In one of these studies,[28] the subject progressed from making initial contact with the toe only to a flat-footed contact in which both heel and toe made contact with the ground simultaneously. Other studies have shown the foot-floor patterns for hemiparetic subjects using various orthoses.[16,27] Lehmann, et al.[16] found that all the AFO designs they studied restored a normal heel to toe foot contact pattern, although differences were found in the durations of the various temporal phases of the gait cycle. In a comparison of orthoses used by subjects with severe plantarflexion spasticity, it was found that plastic AFO users made initial contact with the heel, whereas those using the BICAAL orthosis made contact with the forefoot, the implication being that, for this type of patient, the use of plastic rigid AFO results in a more normal sequence of foot-floor contact.[27]

Most of the studies that have compared different AFO designs have utilized able-bodied subjects.[13,14,15,23,26] The problem with using able-bodied subjects is that they would be capable of using compensatory patterns that individuals with extensive motor changes would be unable to perform. In a comparison of three different plastic AFOs, using both able-bodied and hemiparetic patients, the relative durations of stance and swing were consistent in both groups.[16] There were significant differences in the duration of midstance (defined as the period from heel-strike to heel-off) and pushoff phase (the period from heel-off to toe-off). The hemiparetic patients spent more time in midstance and less time in push-off. Due to the limited plantarflexion provided by all the PL-AFOs, knee instability occurred during heel-strike, but adequate toe clearance was provided by all the orthoses.[16] At heel strike, with the plastic rigid AFO, normal plantarflexion of the ankle is eliminated and thus restricts the progression of foot flat. This may even occur with the foot locked in slight plantarflexion. In order for the hemiparetic patient to obtain foot flat, while the ankle is locked, compensatory mechanisms, particularly at the knee, have to be utilized. In two studies comparing conventional AFOs (metal and leather orthoses) and plastic AFOs,[6,27] both types improve the walking speed of the hemiparetic. In the first study,[6] there was no significant difference in the oxygen consumption using either orthosis. In the second study[27] using rigid orthoses, the plastic AFO users had heelstrike occurring normally in 15 to 16 patients, while the BICAAL had 9 of 16 normal occurrences. The plastic AFOs demonstrated the normal loading pattern of metatarsal five/metatarsal one, or metatarsal five/metatarsal one/great toe, while the users of the BICAAL orthosis did not consistently demonstrate this characteristic pattern.

The purpose of the study was two fold: to quantify if there were any differences in the temporal sequencing of foot/floor contact or in the temporal stride kinematics of the hemiplegic patients walking with and without an orthosis, and to evaluate the feasibility of utilizing footswitches and a telemetry system as assessment aids in the clinical setting.

METHODOLOGY

Seven hemiparetic patients volunteered to participate in the study, of whom five were male and two female all between the ages of 50 and 75 years. Subject selection was based upon the following criteria:

- a stroke that had occurred more than 18 months prior to the study;
- no limiting soft tissue contractures of the lower limb;
- no severe fixed joint deformities;
- mild to moderate increased tone of the gastrocnemius-soleus;
- no receptive aphasia;
- nonapraxia;
- minimal superficial sensory loss;
- capable of walking eight to ten meters with or without an ambulatory aid;
- had been previously prescribed either the KL-AFO, PL-AFO, or SW-AFO.

Only one subject (C) required an accessory; a lateral T-strap to control a moderate lateral ankle instability (varus) during the weight-bearing stage. A summary of the characteristics of the subjects is presented in Table I.

The KL-AFO consisted of four basic components:

- a steel stirrup, which was attached to the shoe;
- a mechanical joint, which included springs to provide dorsiflexion assist and a 20 degree plantarflexion stop;
- bilateral aluminum uprights;
- an aluminum calf band, covered with leather and a leather strap for the closure, located approximately ¾″ below the fibular head.

All the KL-AFOs utilized a conventional ankle alignment, as opposed to the anatomical method, where the mechanical joints are aligned with the malleoli and tibial torsion is accounted for.

The PL-AFO was fabricated from 3/16″ polypropylene molded to a plaster positive model of the subject's leg. The PL-AFOs were of the semi-flexible type rather than the flexible type. The flexibility of a given orthosis depends primarily upon the trimlines around the ankle joint. The orthosis was made to fit inside the subject's shoe, and the closure at the proximal end was by means of a Velcro™ fastener. The orthosis was fabricated to hold the foot in approximately 0°-5° of dorsiflexion.

The SW-AFO orthosis was made of spring wire and an aluminum calf band with a leather covering and closure. The dorsiflexion assist was provided by a loop in the wire, located at the point of attachment to the shoes. The spring wire was not firmly attached to the shoe, but allowed easy removal for transfer from one shoe to another.

The footswitch and telemetry system used in this study was a commercially available system designed for clinical use.† The footswitches, which were incorporated into an insole, monitored the floor contact times of the following foot regions: heel (HL), metatarsal 5 (M5), metatarsal 1 (M1), and greater toe (GT). The footswitches were attached to the sole of the subject's shoes with tape. Each footswitch produced a different voltage output when closed, and hence a different pen deflection on the polygraph recorder. All combinations of switch contacts were unique.[4] The footswitches were connected to a transmitting box, located around the subject's waist, by thin wires. The footswitch data were picked up by a receiver, the output of which was recorded on a polygraph. This system was initially checked, calibrated, and tested for reliability, using two normal subjects.

The subjects were required to walk a distance of ten meters. Two optical switches were used to detect when the patient entered and left the central six meters of the walkway. Four channels were utilized on

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the polygraph. A timing marker confirmed the paper speed (50mm/second) and provided a time component for the measurement of the stride kinematics. Distance markers (output from the optical switches) delineated the data to be analyzed from which to calculate walking speed. The two remaining channels were used to record the footswitch data from both feet independently.

Each subject was required to walk at a self-selected comfortable speed under two conditions (where possible). The first condition was with the orthosis on, and the second condition was without the orthosis. A training period was permitted to allow the subject to become familiar with the experimental setup and to become familiar with the imposed conditions. A minimum of three walks were recorded for each condition. Two subjects (C and E) could not walk without an orthosis.

Two walking trials were analyzed for each condition performed. Four consecutive strides from the central six meters of the walkway were analyzed for each trial. The selection of the strides to be analyzed was based on the consistency of the raw printout. The following points were digitized[††] from the foot/floor output on the polygraph: heel on, off; M5 on, off; M1 on, off; toe on, off; and swing phase foot contact on, off, indicating scuffing during the swing phase. The output from the optical switches, indicating when the subject entered and left the central six meters of the walkway, was also digitized. The X-Y coordinates of these points were input to a mainframe computer[†††] for processing and storage. The temporal gait variables calculated from these data were the walking speed, stance period, swing period, double support periods, symmetry (single support time ratio), and the onset and duration of heel, M5, M1, and toe contact for both feet.

[††] High Precision Co-ordinate Digitizer, Gentian Electronics Ltd., Kanata, Ontario, Canada.

[†††] Cyber 150/580, Control Data Canada, Mississauga, Ontario, Canada.

RESULTS

Table II shows the temporal kinematics for each subject with and without an orthosis. Two subjects (C and E) could not walk without their orthosis (KL-AFO) and therefore, were, only tested under one condition. The measurements included stride time (ST), braking double support (BDS) (defined as the period of double support following initial foot/floor contact), single support (SS), thrusting double support (TDS) (defined as the period of double support immediately preceding the final foot/floor contact), total support (TS), and swing (Sw). These are presented for the paretic limb only; however, with these data, the values for the unaffected side can also be determined. Thus, swing on the paretic side is equal to single support on the contralateral side. Note also that total support for the unaffected side will be longer than that for the affected side if single support is greater than swing on the paretic limb. Also included in this table are data from healthy able-bodied males aged 60 to 65.[19]

Table III shows the results for the with and without orthosis conditions of the onset and duration of floor contact of the various parts of the foot. Only the results from the paretic side are shown. The normative data shown in this table are estimated values obtained from a representative pattern.[4]

DISCUSSION

Stride Kinematics

All of the subjects walked slower than able-bodied individuals for both conditions. The walking speeds ranged from a low of 0.07m/s to a maximum of 0.6m/s, this latter value being less than half that of the speed adopted by an able-bodied elderly man walking at a self-selected normal speed. The differences seen in velocity between the two conditions are no greater than one might expect from two traverses of the walkway even for an able-bodied subject, the exception to this being subject G, who showed a marked increase in walking speed while wearing an orthosis.

When expressed in terms of percentages of stride time, the durations of each of the double support phases, as well as the total support phase, all increase with a decrease in walking speed, whereas single support and swing both decrease. This certainly helps to explain some of the differences noted between the data shown for the hemiplegic subjects and the normal values. However, the most noticeable differences in the temporal kinematics are the patient to patient variability and the asymmetrical nature of the walking pattern. All the subjects, in both conditions, spent more time in total support on the sound leg than on the paretic leg, a finding that is in agreement with a number of other studies on hemiplegic gait.[29,32] The favoring of the paretic limb is also reflected in the shorter single support phase on that side compared to the sound limb (i.e., the swing time of the paretic limb). It has been stated by Brandstatter, et al.[3] that hemiparetic gait is best characterized by single support symmetry. The duration of the two double support phases is very similar in some subjects (A, B, C, and D) and very marked in others (E, F, and G). This variability in asymmetry, both in extent and position within the support phase is a feature that has been discussed by Wall and Turnbull[32] with respect to a group of residual stroke patients.

For all subjects except F, the durations of the phases remain virtually unchanged between the two conditions, even for subject G in whom a marked increase in walking speed was noted while wearing an orthosis. Although there is very little difference in walking speed or stride time between the two conditions for subject F, the duration of the braking double support phase increases as did the total support phase when using an orthosis. Swing time decreases also and the result is a more symmetrical walking pattern.

Sequencing Pattern of the Foot/Floor Contact

Table III shows the normal sequence of foot/floor contact using data from another study.[4] Here initial contact is made with the heel followed by M5, M1, and finally the great toe. A feature commonly seen in hemiplegic gait patterns is a drop foot in which initial contact is made with the forepart of the foot and this is often associated with toe contact during part or all of the swing phase (scuffing). If there is also varus, then initial contact is with the M5 region rather than with the toe. Subjects A and F demonstrated this pattern when walking without an orthosis. The fact that the earliest contact made by subject A in this condition is shown as occurring at 1.6% of the stride, rather than zero, can be accounted for by the data shown: mean values of four strides are for each of two walks. Subject A never had the heel of the paretic foot in contact with the ground, but was still able to clear the ground during the swing phase. However, subject F demonstrated scuffing. The application of an orthosis eliminated this gait problem in both subjects and resulted in a relatively normal sequencing pattern, as characterized by initial HL contact, followed by M5, M1, and finally GT.

The paretic leg exhibited a fairly normal sequencing in both conditions in subjects B, D, and G. There were no major alterations in the gait with the use of an orthosis.

For subject C, the paretic leg only had floor contact by the HL and M5 regions. This strongly suggested that the orthosis maintained the ankle/foot complex in a varus position, indicating that the weight was borne only on the lateral aspect of the foot. The possible reasons for this pattern were either a poor alignment of the mechanical joints, the ankle may have assumed a varus orientation within the orthosis during weight bearing, the knee may have gone into varus, or the sole of the shoe may have been worn on the lateral aspect (which is usually the result of one of the previous factors).

The paretic leg of subject E exhibited a flat-foot gait, with almost simultaneous contact of the HL, M5, and M1 regions. Heel-rise occurred prematurely, and there was a prolonged contact of M5. The flat-foot and premature heel-rise patterns may be related to the angle that the orthosis was set in and to the rigidity of the AFO

Characteristics of Subjects Used In Sample							
Subject	Age	Sex	Height (m)	Paretic Side	Years Post CVA	Orthosis Used	Walking Aid
A	67	M	1.68	R	15	SW-AFO	Regular Cane
B	62	F	1.56	R	6	KL-AFO	Regular Cane
C	73	M	1.69	R	5.5	KL-AFO	Fourpost Cane
D	73	M	1.58	R	5	PL-AFO	Regular Cane
E	71	M	1.65	R	4	KL-AFO	Regular Cane
F	51	F	1.60	L	4	PL-AFO	Regular Cane
G	58	M	1.67	R	7	PL-AFO	Regular Cane

Table I.

Temporal Stride Kinematics for the Paretic Limb								
Subject	Orthosis/ No Orthosis	VEL (m/s)	ST (s)	Temporal Phases (% Stride Time)				
				BDS	SS	TDS	TS	SW
A	O	0.60	1.65	10.2	30.2	12.7	53.1	46.9
	NO	0.63	1.63	9.6	30.6	10.0	50.4	49.6
B	O	0.17	2.78	31.2	16.2	27.7	75.1	24.9
	NO	0.15	2.98	31.3	14.4	26.3	72.0	28.0
C	O	0.07	3.67	31.8	10.0	36.0	72.8	22.2
D	O	0.19	2.96	26.4	19.1	22.6	68.1	31.9
	NO	0.22	2.64	25.3	21.2	20.8	67.3	32.7
E	O	0.32	2.15	8.6	20.0	33.8	62.4	37.6
F	O	0.39	1.81	27.9	23.4	17.8	69.1	30.9
	NO	0.37	1.82	20.0	22.8	14.2	57.0	43.0
G	O	0.48	1.51	26.4	23.2	16.5	66.1	33.9
	NO	0.36	1.64	29.2	20.1	18.4	67.7	32.3
*	NO	1.51	1.06	11.0	39.0	11.0	61.0	39.0

*Data for able-bodied males ages 20-65 years of age (Murray, et al., 1966)[19]

Table II.

(the orthosis was a PL-AFO). For instance, if the foot was maintained in a slight plantarflexed position, toe clearance and heel strike may still occur, yet M5 and M1 contact would occur earlier. If the orthosis was rigid as well, the dorsiflexion that normally occurs during midstance, will be resisted and the forward progression of the body would be reduced. The hemiparetic subject may compensate by exerting a greater force at pushoff on the sound side in order to "vault" over the paretic limb. This would result in an early heel-rise, if the plantarflexion orientation was maintained.

Walking speed and single support (SS) time are the principle measures of performance.[2,10,17,20] Single support time has also been used to reflect the subject's abil-

Onset and Duration of Contacts of the Various Parts of the Paretic Foot with the Ground as a Percentage of Stride Time

		Support Phase						Swing			
	Orthosis/	Heel		M5		M1		Toe		Toe	
Subject	No Orthosis	On	Dur	On	Dur	On	Dur	On	Dur	On	Dur
A	O	0	32.4	1.5	45.4	7.4	41.4	46.9	6.2	—	—
	NO	—	0	0	46.5	0.7	40.5	41.6	8.8	—	—
B	O	0	53.8	19.5	36.9	23.7	36.6	38.1	37.0	—	—
	NO	0	52.2	3.0	54.5	26.8	32.6	28.4	43.6	—	—
C	O	0	59.5	9.4	68.8	—	—	—	—	—	—
D	O	0	52.1	8.5	46.3	16.9	41.6	56.3	11.8	—	—
	NO	0	54.2	12.7	44.3	16.7	42.7	58.9	8.4	—	—
E	O	0	18.8	1.3	38.8	1.9	38.3	40.1	22.3	—	—
F	O	0	48.6	14.1	49.2	18.2	46.0	63.3	5.8	—	—
	NO	2.1	36.9	1.6	52.7	5.4	47.8	53.3	3.7	89.1	6.6
G	O	0	57.9	17.3	43.2	24.8	36.3	60.6	5.5	—	—
	NO	0	59.4	9.8	51.3	30.7	31.0	41.4	26.3	—	—
*	NO	0	39.0	9.7	47.7	20.6	35.9	43.3	16.4	—	—

*Data determined from Botranger (1977)[4]

Table III.

ity to load the limb and the symmetry of the gait pattern.[10, 17, 20] In this study, three of five subjects increased the walking speed, albeit marginally in two of the subjects, with a concommitant increase in the SS time. Based on the walking speed and single support time variables, only three subjects improved their gait performance with the orthosis. There was a decrease in four of the five subjects for the SS time. This suggested that the orthoses enabled the subjects to increase the time spent on the paretic leg, and/or decrease the time spent on the sound leg. The longer double support periods (BDS and TDS) reflected the insecurity of the subjects, even with the orthosis.

Two subjects had a decrease in the walking speed and SS time. The decrease suggests a decline in the gait performance, but the stride kinematics do not provide a complete picture of the role the orthoses played. In many pathological conditions, the normal foot/floor sequencing pattern is altered.[17] The altered foot/floor patterns influence the subject's security and safety during ambulation. In two subjects (A and F) the drop-foot pattern was eliminated and a normal sequencing pattern was re-established with the orthosis. Two subjects (C and E) required the orthosis to ambulate, which in itself is indicative of improved performance. In both cases, the orthosis allowed the subject to walk, but there were problems with their gait as reflected in the foot/floor sequencing patterns. In three subjects, although the sequencing was similar, the durations were reduced, approaching normal times. Thus, it appeared that the orthoses improved the foot sequencing patterns and stride kinematics. For some the improvement was minimal, but for others it was quite significant.

There were several limitations to the study:

- the number of subjects and subject trials
- the heterogeneity of the subject group (the subjects were of varying stages of motor recovery and no screening was done)
- the use of only one method of objective gait measurement.

CONCLUSIONS

The AFO is most often prescribed to prevent dropfoot and provide medial/lateral stability to the ankle/foot complex. The orthosis is designed to enable the hemiparetic to walk safely by reducing the risk of stumbling, and efficiently by reducing the need for compensatory movements, such as hip hiking to assist a drop foot in clearing the ground during the swing phase. By controlling the unstable ankle/foot complex and reducing the need for compensatory patterns, walking speed and asymmetry, may be improved. This study found that gait symmetry and walking speed were similar for both conditions in most cases, suggesting that the application of an AFO does not alone lead to improved symmetry and increased walking speed. The role of the AFO was best demonstrated in subjects A and F, who exhibited dropfoot patterns when not using an orthosis. This was eliminated with the application of the AFOs. Three subjects (B, D, G) did not exhibit any major changes in the temporal sequencing patterns or in the measures of gait efficiency. But there are several underlying factors that must be recognized before making any decisions on the function of the orthoses. When the AFO was prescribed, the need for the orthosis may have been greater due to lesser motor recovery level and poorer balance. With improved motor recovery and the constant use of the AFO for several years, the habituated pattern of walking may have been maintained during the short time that the patient was tested without an orthosis. With fatigue and prolonged walking without the orthosis, a footdrop pattern may develop and increase the risk of tripping. All the patients felt insecure without the orthosis, even with the level walking conditions of the laboratory. On uneven terrain, this factor would become far more important. The provision of an AFO does not guarantee improved walking pattern, and in the case of subject C, although the AFO enabled him to walk, the pattern appeared to be unstable due to the contact only on the lateral aspect. There were many possible reasons for this, but optimal anatomical and mechanical joint alignment and periodic checkups, to ensure that footwear and the orthosis are applied and operating properly, will reduce the chance that such a situation will occur.

The footswitches were easy to apply and nonrestrictive. The set up time for the experiment, for each subject required approximately 20 to 30 minutes. The results were not immediately available following data collection in this study, but the system could be automated with on-line data collection and processing performed with a computer. A system which allows this to be done is in fact commercially available through the manufacturers of the telemetry unit. The footswitches do provide useful information which could augment the usually subjective assessment. For example, by quantifying the gait patterns of the hemiparetic patient, unassisted and orthotically-assisted, as has been done in this study, one can demonstrate the effectiveness of the orthosis in attaining a more normal gait pattern. The quantification of the gait pattern may also assist in detecting abnormal gait patterns and may aid in rehabilitation by providing numerical information for future comparison, necessary in monitoring the progress of the patient. However, since the footswitches provide only kinematic data, the underlying reasons for the patterns of foot/floor contact have to be inferred. Supporting analytic methods, such as video, film, or goniometry would provide useful additional information which might, in turn, lead to determining what kinetic measurements should be made to better understand the underlying causes for a given gait deviation. The footswitch system is useful as a first step in the supply of objective data to augment the observational techniques most commonly used to assess gait in the clinic. Indeed footswitches provide one of the few techniques by which one can determine the sequence in which various parts of the foot make contact with the ground and have proved extremely useful in this study where orthotic devices have been employed to overcome abnormalities in this aspect of the gait cycle.

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