Effect of Dynamic AFOs on Three Hemiplegic Adults

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ABSTRACT

Three hemiplegic adults with cerebral vascular accidents (CVAs) were fitted with plantarflexion-stop dynamic ankle-foot orthoses (DAFOs). Explanations of the rationale, modifications, trimlines and fabrication techniques of the DAFO, including cited references, are provided.

A single-session design with multiple trials was carried out using a randomly assigned order to compare stride characteristics in each of the following conditions: barefoot, with prefabricated AFOs and with DAFOs. Footswitch stride analysis was used to collect, measure and analyze the time-distance parameter data. The results of the data collection were gathered, analyzed and documented retrospectively, and the results displayed an increase in walking velocity, stride length, cadence and single-limb support on the affected side when using the DAFO. These results may explain the DAFO's current popularity as an alternative treatment to conventional thermoplastic orthoses.

Keywords: *Cerebral Vascular Accident, Dynamic Ankle-Foot Orthosis, Hemiplegic Gait, Neurophysiological Control, Orthosis, Posterior Leaf Spring, VA-Rancho Stride Analyzer*

Introduction

Pathological gait patterns associated with cerebral vascular accidents (CVAs) often require orthotic intervention. Orthotic management has undergone a renaissance that began with the introduction of molded thermoplastic orthoses (1,2). Concurrently, neurodevelopmental techniques were initiated to control sensory motor output, thereby improving sensory motor input (3-5). As neurophysiological approaches evolved to include tone-inhibiting casts, the role of orthotic intervention was more defined (6-10). By incorporating certain tone-inhibiting characteristics of casting into AFOs, long-term treatment was improved (11-15).

The DAFO initially was prescribed for the cerebral palsy pediatric population (12,15-17). However, the DAFO also has indicated possible benefits for hemiplegic adults (18,19). Many descriptive orthotic studies exist, but few experimentally instrumented studies have been published (18-22). Woolley *et al.*, using modifications described by Hylton, compared supramalleolar DAFOs with traditional, custom-molded thermoplastic AFOs and found no statistically significant differences between the two designs (22). The study described in this article examines three hemiplegic adults while attempting to provide quantifiable results to warrant the use of DAFOs.

The orthoses tested in this study and the method of data collection are similar to those used in the Diamond *et al.* study (19). Diamond *et al.* concluded increased support and alignment of the foot may have contributed to the DAFO's success. The purpose of this study was to analyze and compare five stride characteristics of three hemiplegic subjects. Subjects were evaluated under three conditions: barefoot, with prefabricated AFOs and with DAFOs. The stride characteristics measured were walking velocity, stride length, cadence, stance time, swing time and single-limb support. Plantar pressure patterns also were recorded but were not used in this study. The VA-Rancho Stride Analyzer[a] was used to collect, measure and analyze the stride characteristic data. The authors hypothesized

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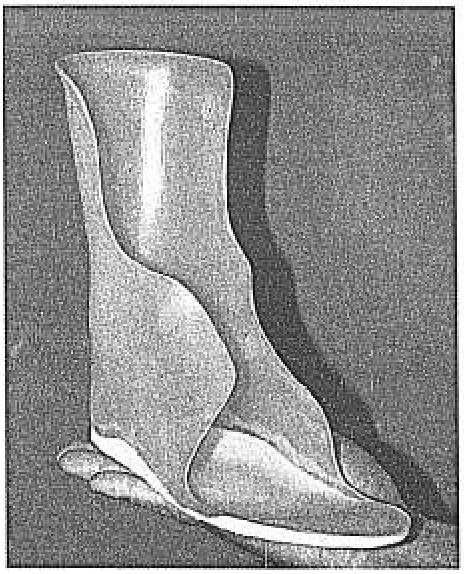

Figure 1. **Plantarflexion-stop DAFO (heel retainer strap not shown).**

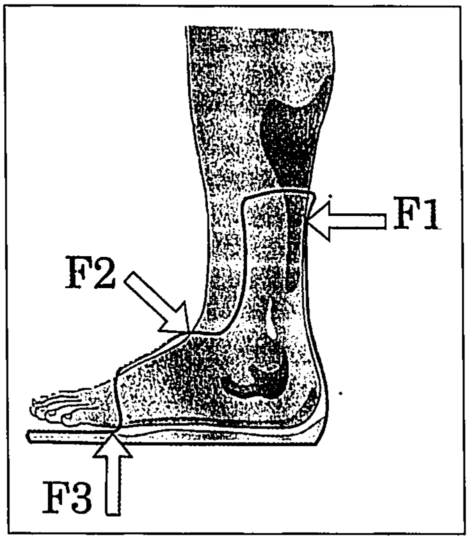

Figure 2. **Sagittal plane of the three-point pressure system.**

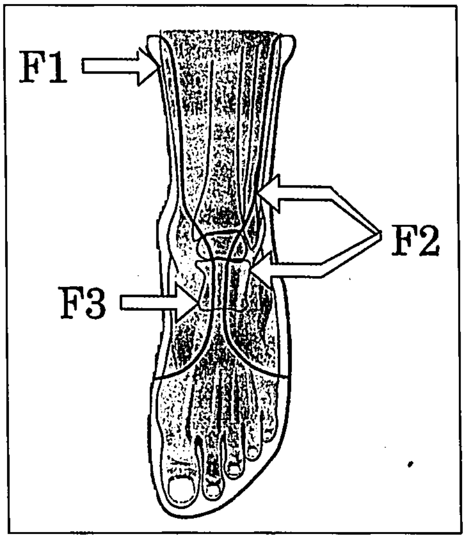

Figure 3. **Coronal plane of the three-point pressure system.**

a quantifiable difference in stride characteristics among the barefoot, prefabricated AFO and DAFO conditions. Based on the literature review, the DAFO was anticipated to demonstrate the most normal stride characteristics.

Characteristics and Design

Multiple variations of the DAFO are prescribed depending on the patient's diagnosis (15,19). The design used herein will be referred to as the plantarflexion-stop DAFO (see *Figure 1*). As a CVA patient progresses, the foot and ankle complex may assume a talipes equinovarus position (23-25).

The orthosis uses three-point pressure systems to counteract this deformity. The first three-point pressure system is designed to limit plantarflexion, which in theory prevents foot slap at initial contact and allows for clearance during swing (see *Figure 2*). The second three-point pressure system attempts to stabilize the subtalar joint, which in theory prevents calcaneal varus and inhibits tone (24) (see *Figure 3*). The third three-point pressure system blocks forefoot adduction and stabilizes the midtarsal joint with assistance from the shoe (see *Figure 4*). The fourth three-point pressure system is designed to prevent the talus from translating anteriorly within the ankle mortise through circumferential containment (see *Figure 5*). This pressure system theoretically uses the coronal and sagittal planes by directing one force (F1) posteriorly toward the calf, a second force (F3) toward the plantar surface of the foot and a third opposing force (F2) anteriorly toward the talus.

The orthoses were constructed of 1/8-inch polypropylene and elongated to approximately 1/32-inch over the dorsum of the foot. A flat crepe base then was applied to the outer plantar surface, thus establishing stabilization within the shoe. The proximal and posterior supramalleolar trimline height is distally located to the insertion of the gastrocnemius. A heel retainer strap across the front of the ankle should help secure the heel and assist with maintaining subtalar neutral. A second strap originally was used at the calf; however, the subjects complained of restricted ankle dorsiflexion and excessive force directed anteriorly at the tibia during midstance. By stabilizing the joints and arches of the foot and ankle and allowing free dorsiflexion, the DAFO may encourage balance reactions, possibly enabling a more normalized gait (12).

The DAFO is a total contact orthosis requiring precise modifications to accomplish the desired neurological and biomechanical objectives (24) (see *Table A*). Stabilization of the joints and arches in the foot is achieved through plaster removal and additions. By combining the medial heel grasp and peroneal notch, a heel lock mechanism is created. The forefoot is stabilized by sinking the metatarsal heads, extending the metatarsal phalangeal joints and circumferentially containing the forefoot. The severity of the modification fluctuated within the study depending on the vascular and/or spastic anomalies. Although DAFO modifications vary among practitioners, their goals remain the same (see *Figures 6* and *7*).

Methodology

Three hemiplegic subjects were evaluated and fitted at the PACT Gait Laboratory, Long Beach VA Medical Center[b]. The subjects were fitted with plantarflexion-stop DAFOs. A fourth subject with excessive knee flexion during stance rejected the DAFO and subsequently was excluded from the study. All subjects were post-CVA males diagnosed within the past 19 months who were assessed as community ambulators with cane assist gait.

Prior to entering the study, each participant was required to sign an informed consent form. Initial evaluation

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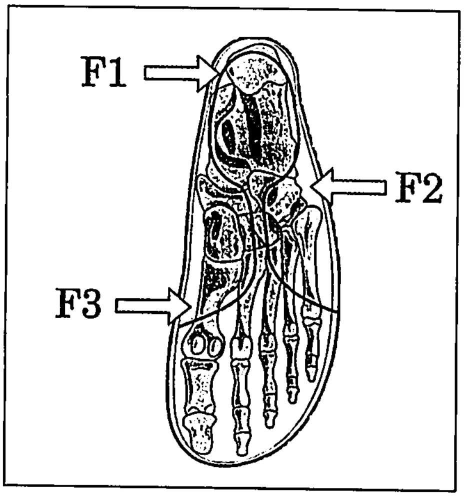

Figure 4. **Transverse plane of the three-point pressure system.**

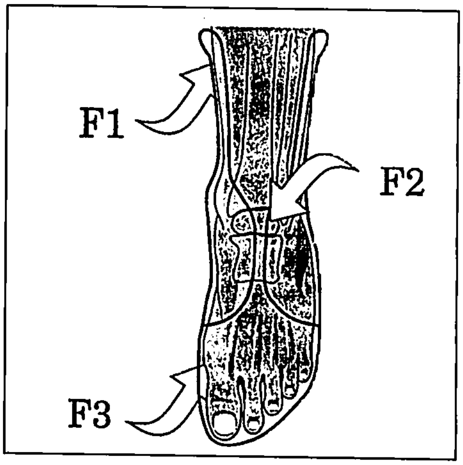

Figure 5. **Coronal and sagittal planes of the three-point pressure system.**

Cast Modifications	
Plaster Removal	**Objectives**
• medial heel grasp inferior to the sustentaculum tali (26,27)	• maintain subtalar neutral, which reduces tone
• transverse arch	• unload metatarsal heads to inhibit toe grasp
• peroneal notch, which is posterior to the base of the fifth metatarsal (4,5,10,12,24)	• facilitate eversion reflex (peroneals) and balance the medial heel grasp
• toe crest distal to the PIP joints (8,12,14,24)	• extend MP joints to inhibit toe grasp and unload metatarsal heads at late stance
Plaster Additions	**Objectives**
• sink metatarsal heads (12,24)	• inhibit toe grasp
• base of the fifth metatarsal	• prevent skin ulcers
• navicular	• prevent skin ulcers
• medial and lateral malleoli	• prevent skin ulcers
• dorsum of foot superficial to the dorsiflexor tendons	• prevent impingement of the dorsiflexor tendons
• proximal calf flair	• prevent skin pinching at initial contact

Table A. **Precise cost modifications are necessary to achieve the desired neurological and biomechanical goals of the DAFO.**

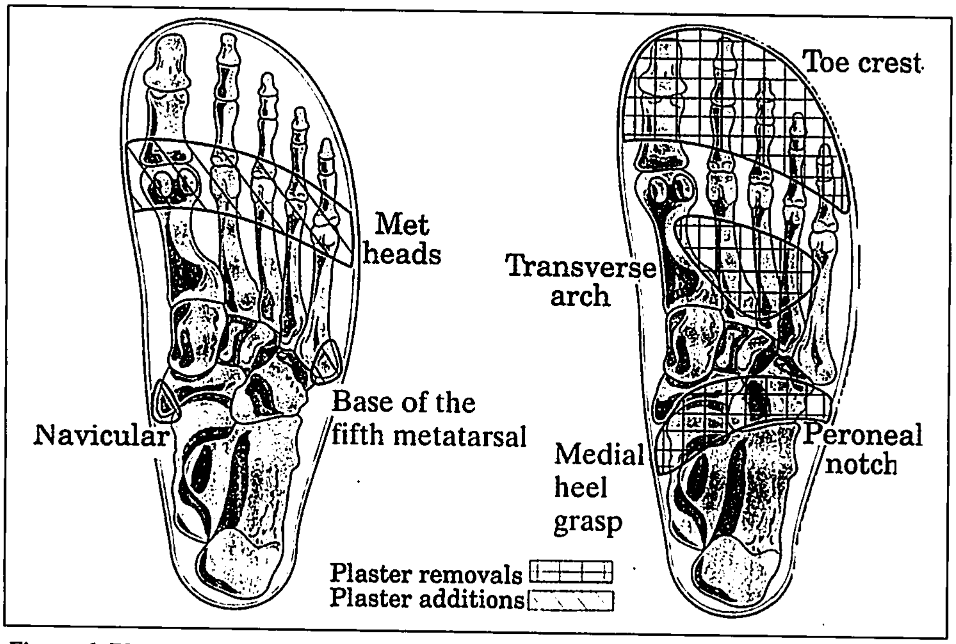

Figure 6. **Plantar view.**

consisted of collecting baseline information to establish prescription indications for the DAFO (see *Table B*). The inclusion criteria required all participants to ambulate a minimum of 10 m without a cane, have passive range of motion (ROM) to 90 degrees dorsiflexion at the talocrual joint, and present with flaccid or minimal to moderate spasticity, which was defined as 1 to 3 on the Ashworth Scale (28).

Due to day-to-day fluctuation, mild spasticity scores may vary (28). Spasticity was tested with subjects sitting in a relaxed position. The affected limb was manipulated through passive ROM at the knee and ankle and recorded as follows: 0 = no increase in tone; 1 = slight increase in tone manifesting a clonospasm when the limb was moved in flexion/extension; 2 = more marked increase in tone but limb easily flexed; 3 = considerable increase in tone and passive movement is difficult; and 4 = limb rigid in flexion or extension.

After fitting the participants with DAFOs, a wearing schedule of two hours the first day with one additional hour per successive day followed until full-time use was achieved. Full-time use was defined as eight to 12 hours' use per day. Participants then were instructed to wear their DAFOs four to six weeks before gait testing; this time period allowed the subjects to wear into the orthosis and develop tactile proprioception.

A single-session design with multiple trials was performed using a randomly assigned order to compare stride characteristics in each condition. The prefabricated AFOs consisted of a standard polypropylene posterior leaf spring (PLS) design with a Velcro® calf strap (see *Figure 8*). The single-session design enabled collection of all of the data on the same day, which was believed to increase the consistency of the methodology and provide a pragmatic approach to complete the study. Before each trial, subjects were permitted to practice ambulating under each given condition. Each subject ambulated a total of 10 m per trial (see *Figure 9*). One trial was performed under each condition. The first and last two meters of each trial were not recorded.

The VA-Rancho Stride Analyzer[a] was used to measure and analyze the stride characteristics (29,30), which included walking velocity, stride length, cadence, stance time, swing time and single-limb support. The stride analyzer is a computerized system for collecting and analyzing data via sole footswitches. The instrumentation consists of footswitches, an automatic start/stop controller, a waist pack memory unit and a calculator. The footswitches are worn as soles outside the shoes or taped to the bare foot and indicate when the foot is weightbearing. Each sole contains compression closing sensors in the areas of the heel, fifth and first metatarsal heads, and great toe. The start/stop controller is automatically operated and used to initiate and stop recording of data for the designated 6-m walking distance. A photoelectric sensor is attached to the deltoid muscle and is triggered by stationary light sources located at the start and end of the 6-m walking distance. A cable from the sole footswitch plugs into the memory unit, which is connected to the subject's waist. The memory unit stores the data so they may be entered into the calculator.

The calculator is preprogrammed with a relationship between absolute distance parameters used and normal stride values (29). The norms were de-

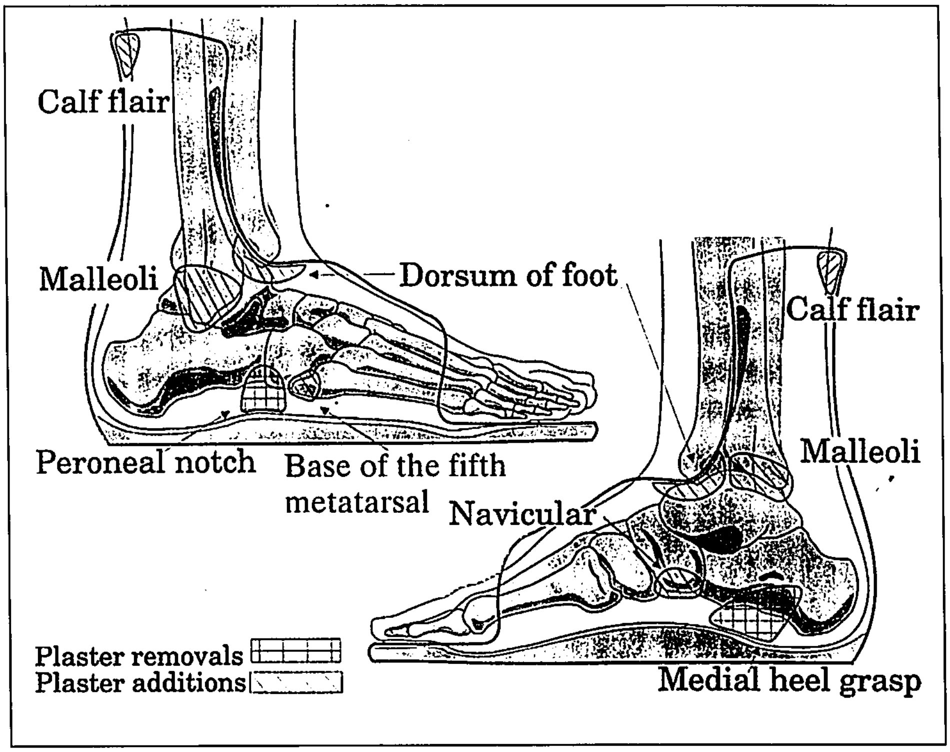

Figure 7. **Lateral and medial views.**

Baseline Information

Patient Name: Age:
Weight: Height:
Diagnosis: Date of Onset:

1. General attitude of subject? ☐ cooperative ☐ indifferent ☐ negative
2. General state of health? ☐ strong/no complications ☐ weak
3. Is the affected limb on the subject's dominant side? ☐ yes ☐ no
4. Is the upper extremity involved? ☐ yes ☐ no
5. Is the subject's proprioception intact? ☐ yes ☐ no
   Proprioception was tested with eyes closed. First, manipulate the subject's affected limb then have the subject replicate the movement on his or her sound side (4).
6. Is the subject using auxiliary walking aids? ☐ walker ☐ cane ☐ none
7. Has the subject worn an orthosis in past? ☐ yes ☐ no
   If so, what type? ____________________
8. What is the subject's ROM?
   talocrual joint: _____
   knee joint: _____
9. What is the subject's Ashworth Score? _____

Table B. **Patients were asked to provide baseline information to establish prescription indications.**

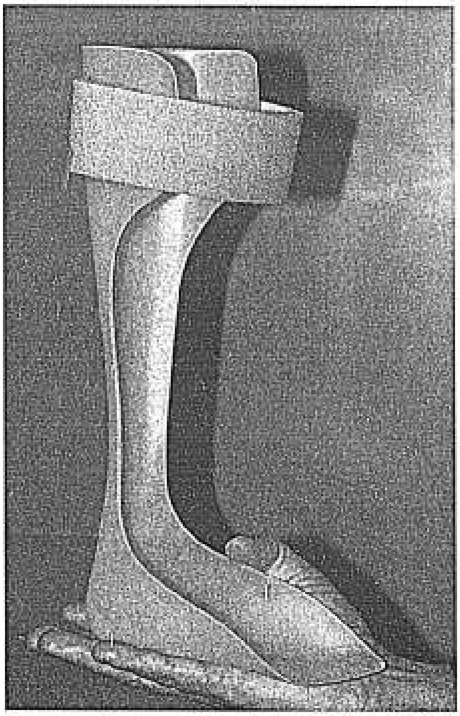

Figure 8. **Posterior leaf spring (PLS) design of a prefabricated AFO.**

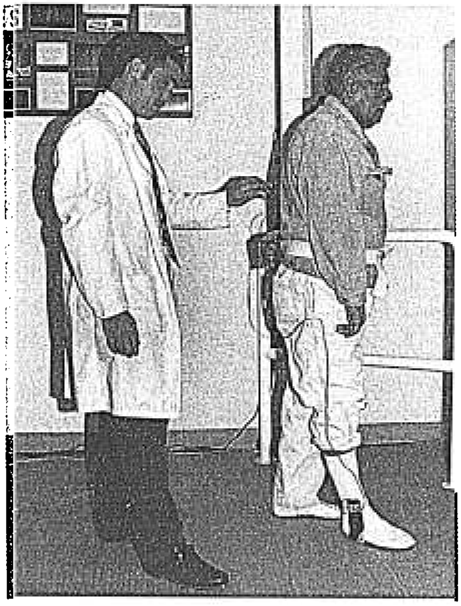

Figure 9. **Subject ambulating.**

termined by a group of 43 normal male and female subjects. For each measured distance parameter, the calculator internally determines the single-stance time for a normal individual and uses that value to normalize the measured single-stance time for the subject's right and left sides following each trial (29).

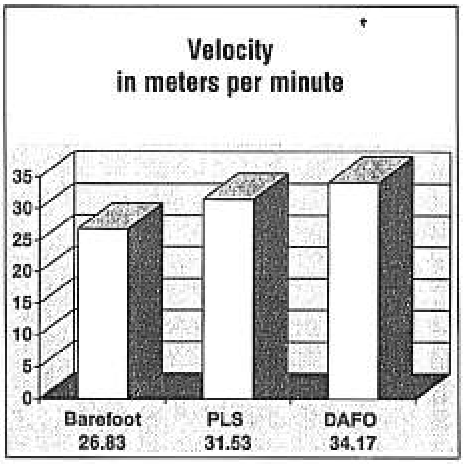

Figure 10. **Subjects' velocities while barefoot, with a posterior leaf spring (PLS) AFO and wearing the DAFO.**

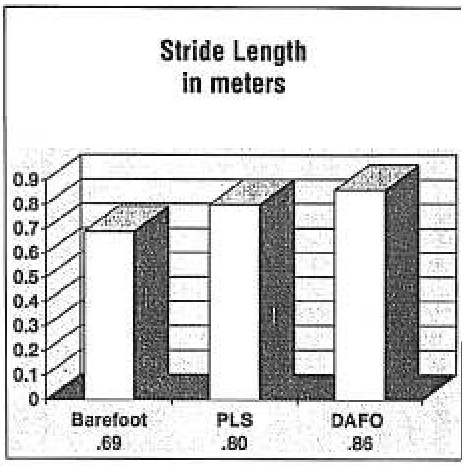

Figure 11. **Subjects' stride lengths while barefoot, with a PLS AFO and wearing the DAFO.**

Results

The mean values of each subject's stride characteristics were calculated over three strides. Each subject's mean value then was added and divided by the total number of subjects. The total mean of the three strides for the three subjects then was graphically displayed. Whenever possible, these figures illustrated values for both the sound and affected sides.

Walking velocity defines the rate of travel by identifying the time required to cover a designated distance (30). Velocity can vary depending on the subject's age and limb length. "Normal free gait velocity on a smooth surface averages 82 m/min for adults" (30). *Figure 10* illustrates the subject's DAFO velocity as 7.34 m/min increased over the bare foot and 2.64 m/min increased over the subject's PLS velocity.

The primary determinants of walking velocity are the length and repetition of the person's stride (19,30,31). Stride represents the distance between two consecutive points of initial contact with the same foot (30). Stride length for a normal person during free walk averages 1.41 m (30). The subject's stride length in the DAFO was .17 m greater than the barefoot trial and .06 m greater than the subject's stride length in the PLS (see *Figure 11*).

Cadence represents the repetition rate of the person's stride or steps (30). Individuals with a shorter stride often compensate by increasing their cadence (30,31). The average adult's cadence during free walk is 113 steps/min (30). The subjects' cadence in the DAFO increased 1.67 steps/min to the bare foot and 2.00 steps/min to the subjects' cadence in the PLS (see *Figure 12*).

Stance is referred to as the period of time when the foot is in contact with the ground (30,31). "The gross distribution of the floor contact periods are 60 percent for stance and 40 percent for swing" (30). This distribution is expressed as a percentage of the gait cycle. A gait cycle is the equivalent of one stride. The stance times on the sound side were increased relative to the affected side under all conditions (see *Figure 13*). The PLS had the longest stance time on the affected side (see *Figure 14*).

Swing applies to the time when the foot is off the ground for limb advancement (30,31). Due to the increased stance times on the sound side, the swing times on the sound side were decreased under all conditions (see *Figures 15* and *16*). The affected side had a more normal distribution between stance and swing time under all three conditions.

Single-limb support is defined as the interval of time when the body's entire weight is resting on one extremity (30). Single-limb support constitutes approximately 40 percent of the gait cycle and occurs during the stance phase (30). The

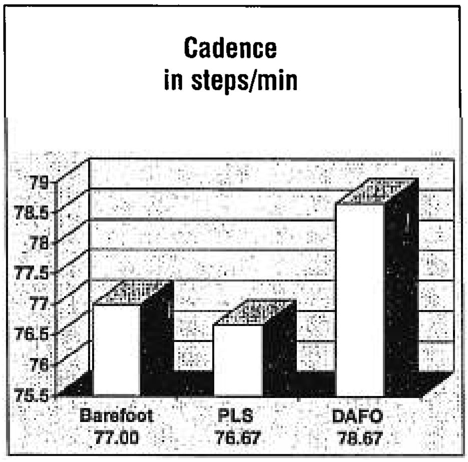

Figure 12. **Subjects' cadences while barefoot, with a PLS AFO and wearing the DAFO.**

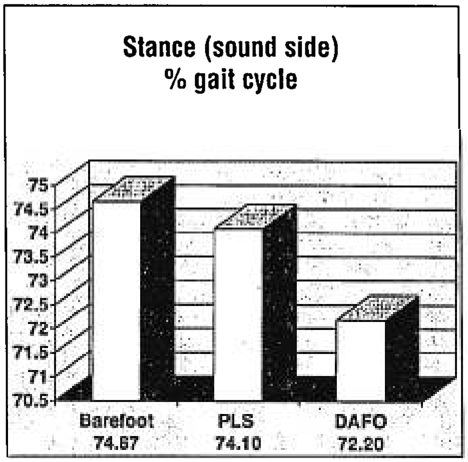

Figure 13. **Subjects' stance (sound side) while barefoot, with a PLS AFO and wearing the DAFO.**

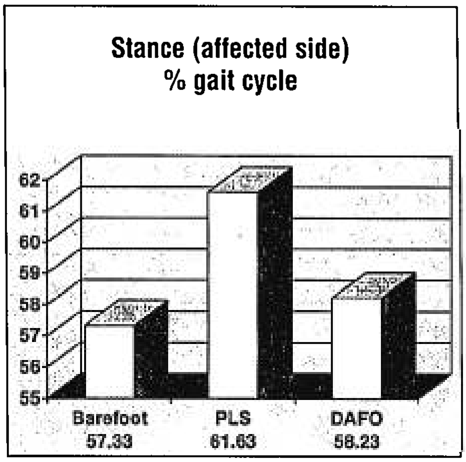

Figure 14. **Subjects' stance (affected side) while barefoot, with a PLS AFO and wearing the DAFO.**

data collected under the barefoot condition had the longest single-limb support time on the sound side (see *Figure 17*). The subject's single-limb support in the DAFO was 2.54 percent over the bare foot and 1.97 percent more than the subject's PLS single-limb support on the affected side (see *Figure 18*).

Discussion

As previously stated, velocity can increase with stride length and/or cadence (19,30,31). This relationship tends to be linear and usually consistent (19,30). This study reported increases in velocity over barefoot ambulation when orthoses were used. The subjects wearing the PLS slightly decreased in cadence but increased stride length enough to increase their velocity. The subjects wearing the DAFO had the highest velocity through a combined increase in cadence and stride length.

Although step length was not measured the authors delineated the following theories from the stride-length data. First, the PLS would appear to have less anterior lever arm support than the DAFO. Anterior support could be a function of either the linear measurement of the foot plate or the level of elasticity derived from the materials used. Second, step length appeared to decrease for the advancing limb with the PLS. This supposition is supported by the increased stride length for the DAFO reflected in *Figure 11*.

Single-limb support is possibly the best indicator of the limb's support capability (30). The results reported a favorable increase with orthotic treatment. The barefoot condition had the longest single-limb support time on the sound side with the shortest time on the affected side. This indicates a prolonged weight transfer period to the sound limb to compensate for the non-orthosis-affected limb. The subjects wearing the DAFO had a relatively normal single-limb support time on the affected side. In summary, the DAFO was associated with a more normal single-limb support time on both the sound and affected sides, indicating a more symmetrical gait.

Study Limitations

This study was not without limitations. The authors' intent was to present a possible alternative approach to the orthotic treatment of hemiplegic patients. First, this study does not have a sufficient sample population to obtain statistical significance. The small sample population may account for the different results founded by Woolley *et al.* Another possible cause for varying results to the Woolley *et al.* study may be explained by the consistent level of the AFO rigidity.

Second, many of the cited references identify excessive pronation and dorsiflexion during terminal stance with hemiplegic gait. However, without kinematic gait analysis, the investigators were unable to measure the amount of motion occurring in the lower extremities. Therefore, control of excessive motion (i.e., dorsiflexion, pronation) could not be determined. Further research using a similar study design and larger sample population with data analysis is required to obtain statistical significance.

Conclusion

In conclusion, the subjects' opinions of the plantarflexion-stop DAFO varied. They described decreased heel-only time at initial contact and difficulty donning and doffing the orthosis. Suggestions to the above remarks were to wear an athletic sneaker with a thick heel cushion and, once weaned into the DAFO, donning and doffing the orthosis just once a day.

Conversely, the subjects were pleased with the intimate fit and low-profile design, and they noticed increased correction of their equinovarus position and decreased fatigue. In addition, they noted an increase in support of the plantar arches.

The results of this study have demonstrated the plantarflexion-stop DAFO has a positive effect on gait with post-CVA hemiplegic adults when compared to the barefoot and prefabricat-

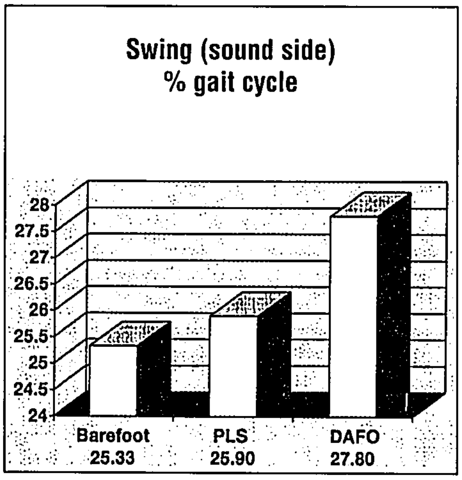

Figure 15. **Subjects' swing (sound side) while barefoot, with a PLS AFO and wearing the DAFO.**

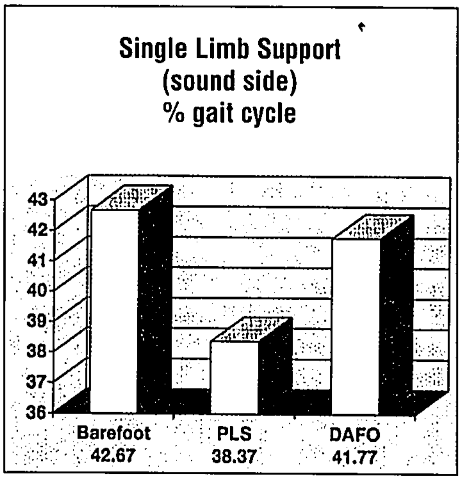

Figure 17. **Subjects' single-limb support (sound side) while barefoot, with a PLS AFO and wearing the DAFO.**

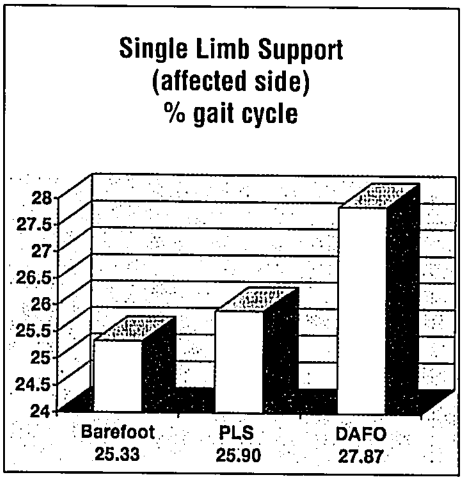

Figure 18. **Subjects' single-limb support (affected side) while barefoot, with a PLS AFO and wearing the DAFO.**

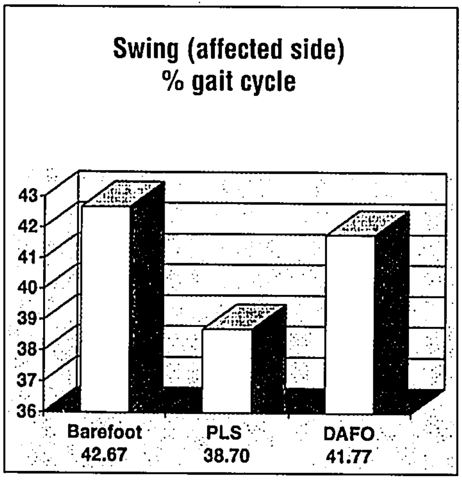

Figure 16. **Subjects' swing (affected side) while barefoot, with a PLS AFO and wearing the DAFO.**

ed-AFO conditions. Further research comparing custom-molded AFOs and DAFOs with the same lever arms and proximal trimlines is needed to eliminate the possible differences resulting in the DAFO's supramalleolar trimlines. By maintaining the same lever arms and proximal trimlines, a future study can isolate differences resulting from hypothetical tone-reducing principles. □

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