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The role of rigid and hinged polypropylene ankle-foot-orthoses in the management of cerebral palsy: a case study

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Abstract

Ankle-foot orthoses are commonly used in the treatment of spastic cerebral palsy to hold the foot in a position conducive to a more functional gait. This study, utilizing quantitative biomechanical techniques, evaluates the effects of a rigid ankle-foot orthosis and a hinged ankle-foot orthosis on spastic cerebral palsy gait. The subject was a 4.5 year old female diagnosed as spastic diplegic cerebral palsied shortly after birth. Testing involved collection of kinematic coordinate data employing a WATSMART video system and ground reaction force data using a Kistler force plate. Jensen's (1978) photogrammetric method was used to estimate body segment inertial parameters. The hinged ankle-foot orthosis was found to be more effective than the rigid ankle-foot orthosis. The subject exhibited a more natural ankle motion during the stance phase of gait, greater symmetry of segmental lower extremity motion, and decreased knee moments during stance while wearing a hinged ankle-foot orthosis.

Introduction

In spastic diplegic cerebral palsy, the upper motor neuron lesion produces weakness and increased muscle tone which results in varying degrees of disability. The aim of orthotic management in spastic cerebral palsy is to produce a more normal gait pattern by positioning peripheral joints in a way that reduces pathological reflex patterns or by blocking pathological movement of the joint. Analyses of cerebral palsy gait, employing qualitative (Mann, 1983) and quantitative methods (Hershler and Milner, 1980; Skrotzky, 1983; Lee et al., 1985), have been reported in the literature. However, little research has been done evaluating the effects of orthotic devices on the gait of the cerebral palsied child. The studies that have been reported use subjective (Rosenthal, 1984) and qualitative (Taylor and Harris, 1986) means of evaluation. Bertoti (1986) investigated the effect of inhibitive casting on cerebral palsy gait, however, the subjectiveness of evaluating ink blots renders the interpretation of the results questionable. The purpose of this paper is to evaluate the effects of a rigid ankle-foot orthosis (AFO) and a hinged AFO on spastic cerebral palsy gait utilizing quantitative biomechanical techniques.

Spastic cerebral palsy is characterized by abnormal reflex patterns, retention or surfacing of primitive reflexes, and an abnormal increase in muscle tone (Pederson, 1969). Spasticity develops when the inhibitory effects of higher brain centres are removed as a result of damage to the upper motor neurons. The result is abnormal posture and movement patterns (Bobath, 1971). In an attempt to improve balance and normalize movement patterns of spastic individuals, a variety of orthotic devices are used.

The majority of spastic cerebral palsy patients exhibit high tone in the extensor muscles of the lower extremity. Spastic plantarflexors and invertors pull the foot into an equinovarus position. The effective use of inhibitive plaster casts to position the spastic equinus ankle and foot in a functional plantigrade position has been documented (Sussman and Cusick, 1979; Cusick and Sussman, 1982). Extension of the metatarsalphalangeal joint by a raised toe plate may aid in reducing the positive support reaction. The casts position the patient's feet and ankles in a functional alignment allowing more normal movement patterns to be established. Inhibitive casting has been

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reported to aid in reducing foot deformity and muscle tone in the affected extremity (Duncan and Mott, 1983) and in improving stride symmetry and length (Bertoti, 1986).

Tone-reducing ankle foot orthoses (TRAFO's) were developed as removable splints that function in a manner similar to the inhibitive cast. More specific guidelines pertaining to the positioning of the ankle and foot accompanied the introduction of this AFO (Jordan et al., 1984). Fabrication utilizes lightweight fibreglass casting wrap in place of standard plaster bandages. TRAFO's are used as an adjunct to physical therapy to reduce spasticity and aid in unassisted standing and walking (Jordan et al., 1984).

Following improvement in motor performance the casts or TRAFO's may be replaced with rigid polypropylene AFO's. As with inhibitory casts, the AFO's rigid posterior section locks the ankle in a functional position. Rigid AFO'S are lightweight and more cosmetic than plaster casts and are designed to be worn inside regular shoes. Stabilization of the foot and ankle aids in proper heel-strike and enhances standing balance by blocking all motion at the ankle (Rosenthal, 1984; Harris and Riffle, 1986).

Recently, the custom-moulded polypropylene hinged AFO, with a plantarflexion stop (Figure 1), has become a popular device used to inhibit pathological plantarflexion yet allow some range of normal dorsiflexion to occur at the ankle joint. The potential for ankle dorsiflexion will allow

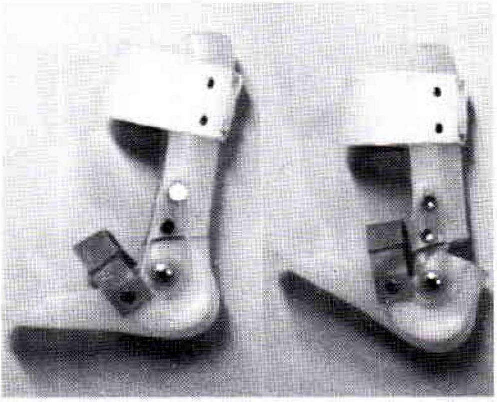

Fig. 1. Custom moulded polypropylene hinged ankle-foot orthosis. Left, plantarflexion stop. Right, dorsiflexion range

stretching of the Achilles tendon which may result in reduced spasticity of the triceps surae muscle. It is presumed that bracing with the hinged AFO produces a more normal gait pattern, however, there is no information examining the effects of the hinged polypropylene orthosis on spastic gait.

Methodology

The subject was 4.5 years of age and diagnosed as spastic diplegic cerebral palsied shortly after birth. Physical assessment revealed full range of plantarflexion and up to 5 degrees of dorsiflexion at the ankle bilaterally. Knee range of motion showed approximately 3 degrees of hyperextension but was otherwise normal. Within the limitations of the spasticity grading scale and subjective gait analysis, both lower extremities exhibited equal degrees of spasticity and were classified as moderately spastic. The subject's first orthoses were bilateral TRAFO's (age 2.3 years and worn for 10 months). These were replaced by rigid custom-moulded polyproplyene AFO's at 3.2 years of age. The rigid AFO's were worn for one year and 4 months. Hinged custom-moulded polypropylene AFO's were dispensed 3 months prior to testing and were being used at the time of the testing. The subject was a voluntary participant and informed consent was obtained from the parents prior to testing.

Fabrication technique

Two pairs of custom-moulded polypropylene AFO's were fabricated for the subject. A negative cast was produced by casting the subject with plaster-of-Paris wrap. The subtalar joint axis was positioned in neutral and the ankle joint was cast in 5 degrees of dorsiflexion. A positive mould was made by filling the negative mould with plaster. This was then modified for the production of a hinged AFO. The axis of the mechanical ankle joint was placed at the distal tip of the medial malleolus, in the line of progression, reflecting normal toe-out. Two Ortholen plates were fabricated and moulded to the positive cast at the ankle axis. The polypropylene was drape-moulded, trimmed and cut into two sections approximately 1.5 centimetres above the ankle axis leaving the Ortholen intact. The calf section of the orthosis was permanently attached to the upper area of the Ortholen plates and the foot section was attached to the

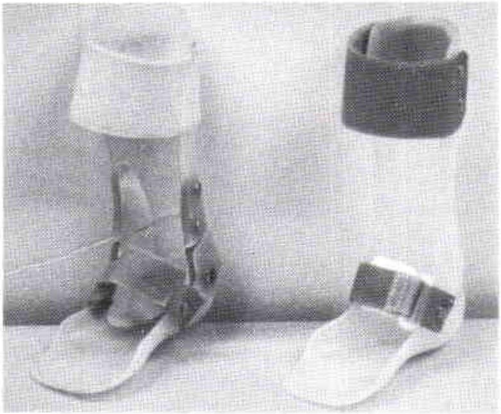

Fig. 2. Hinged and rigid AFO's fabricated for the subject.

Ortholen by two Chicago screws. The foot section was free to rotate on the Ortholen hinge through a range from 5 to 35 degrees of dorsiflexion. Two Velcro straps were attached, one over the proximal tibia and one at the proximal foot section. Rigid AFO's were fabricated from the same positive casts (with the ankle re-modified to form a rigid structure) in an attempt to maintain similar structure for both orthoses (Figure 2).

Data collection

Testing involved collection of kinematic coordinate data using a WATSMART video system and ground reaction force data employing a Kistler force plate. A 35mm photograph of the subject was taken in order that segment inertial parameters could be estimated. Force plate and kinematic coordinate data were collected for the right lower extremity and kinematic coordinate data were collected for the left lower extremity. Two independent three segment link systems were used to model the motion of the lower extremities during ambulation.

A two camera WATSMART kinematic data acquisition system was used to acquire the two-dimensional positions of four anatomical landmarks on each lower extremity (50 hertz sampling rate). Three centimetre diameter disks containing 3 infra-red emitting diodes (IREDS) were placed over the anatomical landmarks. These landmarks located the anatomical joint centres of the hip, knee and ankle, and the distal end of the foot segment. The cameras were placed perpendicular to the sagittal motion of the lower extremities. The subject was familiarized with the testing area to promote natural performance during data collection. Walking trials lasted approximately 3–4 seconds and 20 walking trials were carried out for each of the three test conditions (unbraced, wearing rigid AFO's wearing hinged AFO's). The subject wore shoes throughout the testing.

The 35mm still photograph was analyzed by projecting the image on an X-Y digitizer and following procedures consistent with Jensen's (1978) photogrammetric method. Estimation of segment inertial parameters (mass and moment of inertia about the transverse proximal axis) were determined mathematically by modelling body segments as standard geometric shapes. The model employed considers the body segments to be composed of elliptical zones two centimetres wide. Segment densities are assumed (Clauser et al., 1969) and used with the calculated segment volumes to give the segment masses. A computer program was used to calculate inertial parameters, for the thigh, leg and foot segments, from the digitized record.

Subsequent kinematic and kinetic analyses of coordinate data records were performed using the Waterloo Biomechanical Motion Analysis Software Package. Three walking trials from each of the three test conditions were selected for analysis from available trials based on the subject's success in hitting the force plate. Each walking trial was composed of one complete stride and both left and right lower extremities were analyzed. All joint coordinate data were filtered through a low pass recursive second order Butterworth digital filter using a 5 hertz cutoff frequency (Pezzack, 1977). Waterloo Program input parameters were selected and employed in the established manner (Winter, 1979).

Discussion

Quantitative biomechanical analysis was employed to evaluate the influence of the rigid and hinged ankle-foot orthoses on cerebral palsy gait. Ankle relative joint angles (the angle of the foot in relation to the leg) were determined bilaterally so that ankle movement during stance could be evaluated. Absolute angular displacements (the angle of the segment relative to the ground) of the thigh and leg segments were calculated, and bilateral leg/thigh angle-angle diagrams were used in evaluating the degree of symmetry between lower extremity action during the gait cycle. Resultant muscle moments applied to the leg at the knee were calculated during stance for both braced conditions. Parameters reflecting ankle motion, and lower extremity action were selected to determine objectively the effect of the orthoses on the gait pattern. Knee kinetics were used to evaluate the resultant moment acting about the knee with respect to rigid versus hinged AFO gait. These variables are considered clinically relevant in assessing pathological gait.

Ankle motion

Figure 3 depicts left and right ankle relative joint angles for each of the three test conditions (unbraced, rigid AFO, hinged AFO) during stance. Dorsiflexion, displayed most dramatically in unbraced and hinged AFO trials, occurs following foot-floor contact in all three test conditions. The ankle dorsiflexes at weight acceptance since the subject strikes the ground with the forefoot rather than the heel. Dorsiflexion following foot-floor contact is less prominent while wearing the rigid AFO. The rigid AFO shows some ankle motion at weight acceptance as the polypropylene is stressed and bends slightly. As the subject progresses through stance phase the polypropylene rebounds to its original position.

While unbraced or wearing the hinged AFO, both ankles dorsiflex following midstance. This dorsiflexion reflects the forward progression of the body over the base of support. This progression is evident in both

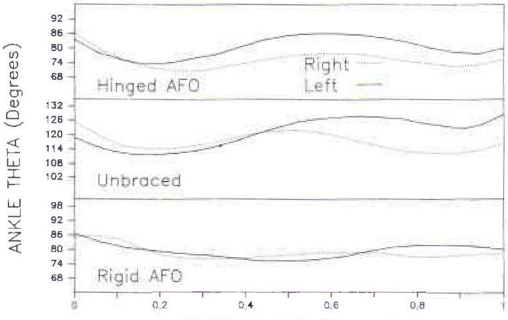

Fig. 3. Ankle relative joint angles, during stance, while wearing rigid AFO's, hinged AFO's and unbraced.

hinged AFO and unbraced conditions, however, while unbraced, the foot remains in a plantarflexed position throughout stance. This plantarflexed foot position is characteristic of unbraced spastic diplegic gait (Lee et al., 1985; Gage, 1983). Plantarflexion following foot-ground contact and during push off, displayed in normal gait (Mann, 1983), is not possible due to the solid structure of the rigid AFO and the plantarflexion stop incorporated into the hinged AFO. The hinged AFO appears to elicit a more desirable ankle joint action since the foot remains in a dorsiflexed position and dorsiflexion occurs during stance. Ankle dorsiflexion during weight-bearing allows stretching of the Achilles tendon which may result in reduced spasticity of the triceps surae muscle.

Lower extremity symmetry

Symmetrical segmental lower extremity motions are characteristic of normal gait patterns. Angle-angle diagrams were produced to illustrate the motions of the right and left lower extremities for one stride. An estimate of congruity or similarity in shape between any two XY patterns may be obtained by chain encoding each pattern and then determining the cross-relation function from the two generated chains (McIlwain and Jensen, 1985; Whiting and Zernicke, 1982). This function, referred to as the recognition coefficient (C), can vary from 0.0 to 1.0. A value of 1.0 indicates perfect congruity or exact symmetry and a value of 0.0 indicates absence of congruity between patterns. This chain encoding technique was employed in determining the degree of symmetry between right and left sides for all walking trials. Furthermore, all trials (both right and left sides) within each of the three test conditions were chain encoded to evaluate between trial variability.

Angle-angle plots of relative joint angles have been used in the assessment of cerebral palsy gait (Hershler and Milner, 1980). Absolute segmental angular displacements for the leg and thigh were selected for this study since the use of ankle relative joint angles was contraindicated due to the lack of ankle motion inherent to the wearing of a rigid AFO. Furthermore, it was felt that absolute angular displacements of the thigh and leg segments better depicted the action of the lower extremity. Figures 4, 5 and 6 illustrate angle-angle plots for unbraced, rigid AFO and hinged AFO trials. Tables 1, 2 and 3 present the results of chain encoding both right and left side angle-angle plots for all trials within each of the three test conditions.

Unbraced walking trials exhibit the least degree of symmetry between right and left lower extremities. Rigid AFO trials are more symmetrical than unbraced trials with the exception of trial 2 which demonstrates a degree of symmetry comparable to the unbraced trials. All three hinged AFO trials display a greater degree of symmetry than the unbraced trials. Hinged and rigid AFO trials are generally comparable in degree of symmetry. Within test conditions variability is highest in the unbraced condition. Unbraced lower extremity movements in cerebral palsy gait may be characterized as lacking in symmetry and variable between cycles. Braced lower extremity movements tend to have a greater degree of symmetry between cycles. Both hinged and rigid AFO trials have similar between trial variability. The results indicate that bracing spastic diplegic gait increases the degree of lower extremity symmetry.

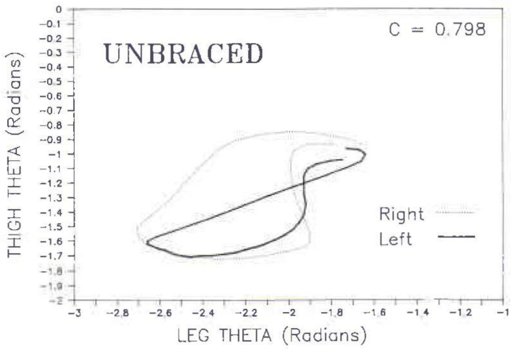

Fig. 4. Angle-angle plots of leg and thigh during an unbraced stride.

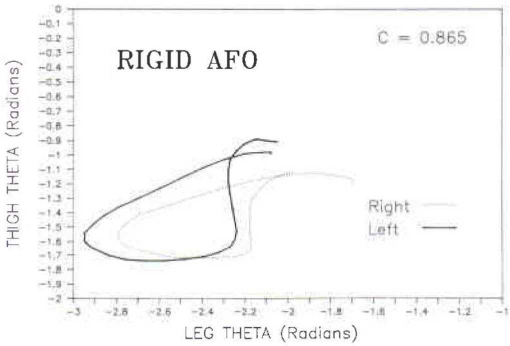

Fig. 5. Angle-angle plots of leg and thigh for one stride while wearing rigid AFO's.

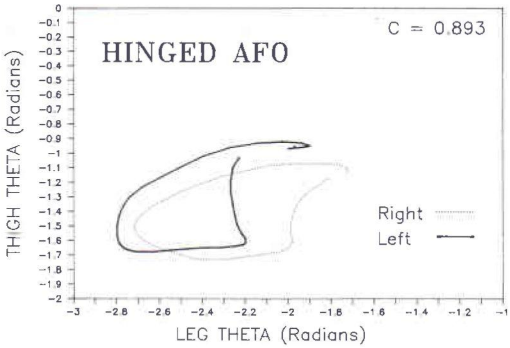

Fig. 6. Angle-angle plots of leg and thigh for one stride while wearing hinged AFO's.

Table 1. Chain encoding results of angle-angle plots for unbraced walking trials.

	Trial 1 Left	Trial 2 Right	Trial 2 Left	Trial 3 Right	Trial 3 Left
Trial 1 Right	**0.831**	0.713	0.799	0.836	0.739
Trial 1 Left		0.751	0.859	0.844	0.819
Trial 2 Right			**0.798**	0.757	0.792
Trial 2 Left				0.839	0.814
Trial 3 Right					**0.799**

Table 2. Chain encoding results of angle-angle plots for rigid AFO walking trials.

	Trial 1 Left	Trial 2 Right	Trial 2 Left	Trial 3 Right	Trial 3 Left
Trial 1 Right	**0.851**	0.806	0.853	0.860	0.841
Trial 1 Left		0.785	0.822	0.854	0.846
Trial 2 Right			**0.796**	0.789	0.792
Trial 2 Left				0.856	0.768
Trial 3 Right					**0.865**

Table 3. Chain encoding results of angle-angle plots for hinged AFO walking trials.

	Trial 1 Left	Trial 2 Right	Trial 2 Left	Trial 3 Right	Trial 3 Left
Trial 1 Right	**0.836**	0.897	0.825	0.860	0.833
Trial 1 Left		0.816	0.833	0.816	0.821
Trial 2 Right			**0.850**	0.870	0.850
Trial 2 Left				0.810	0.805
Trial 3 Right					**0.893**

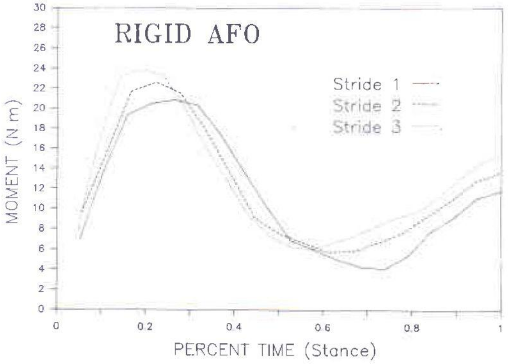

Fig. 7. Muscle moments applied to the right leg, at the knee, during stance while wearing rigid AFO's (positive values indicate extension).

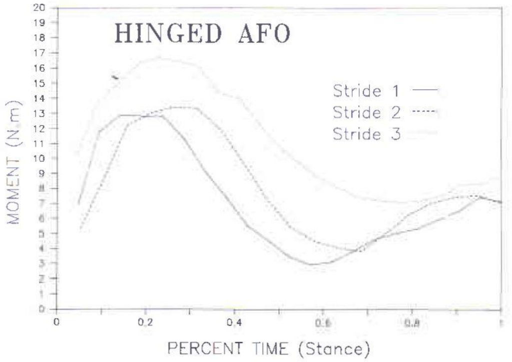

Fig. 8. Muscle moments applied to the right leg, at the knee, during stance while wearing hinged AFO's (positive values indicate extension).

Knee kinetics

The resultant muscle moments applied to the leg, at the knee, during stance for all braced walking trials are depicted in Figures 7 and 8. Resultant muscle moment data estimates the internal moment generated by the muscles and is presented for stance phase only since the magnitude during swing is comparitively small. All knee moment data corresponds to the right lower extremity. Walking velocity for each trial is presented in Table 4.

Table 4. Walking velocity.

	Rigid AFO	Hinged AFO
Trial 1	0.792 ms^{-1}	0.760 ms^{-1}
Trial 2	0.853 ms^{-1}	0.786 ms^{-1}
Trial 3	0.903 ms^{-1}	0.835 ms^{-1}

In all braced trials, peak extension muscle moments occur shortly after foot-floor contact. This peak corresponds to weight acceptance and increases with walking velocity. The magnitude of the resultant extension moments occurring during hinged AFO trials is less than in the rigid AFO condition. Walking velocities between each of the test conditions are comparable suggesting that the differences in magnitude of knee muscle moments between test conditions may not be attributed to differences in walking velocity. In normal gait, ankle dorsiflexion during stance allows smooth forward progression of the body over the base of support. Rigid AFO's prevent dorsiflexion making it necessary for knee extension to be used to advance the body forward and, subsequently, cause greater extension moments about the knee during stance. Decreased knee muscle moments are desirable since they influence energy expenditure and stability during ambulation. This data clearly demonstrates that hinged AFO gait exhibits lower knee resultant muscle moments throughout stance than rigid AFO gait.

Conclusion

Ankle-foot orthoses are commonly used in the treatment of spastic cerebral palsy to hold the foot in a position conducive to a more functional gait. The results of this study demonstrate the effectiveness of bracing in spastic diplegic gait based on increased lower limb symmetry. For the subject evaluated in this study, the hinged ankle-foot orthosis was found to be more effective than the rigid ankle-foot orthosis. The subject exhibited a more natural ankle motion during stance, greater symmetry of segmental lower extremity motion and decreased knee moments during stance while wearing the hinged ankle-foot orthosis. Within the limitations of this study, it appears that the hinged ankle-foot orthosis provides the clinician with a more effective tool for treating spastic cerebral palsy patients than the rigid ankle-foot orthosis.

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