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A comparison of gait with solid and hinged ankle-foot orthoses in children with spastic diplegic cerebral palsy

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Abstract

This study compared the effects of solid and hinged ankle-foot orthoses (AFOs) on the gait of children with spastic diplegic cerebral palsy (CP) who ambulate with excessive ankle plantar flexion during stance. Twelve children with spastic diplegic CP wore no AFOs for an initial 2-week period, solid AFOs for 1 month, no AFOs for 2 weeks, and hinged AFOs for 1 month. Lower extremity muscle timing, knee and ankle joint motions, moments and powers, and temporal-distance characteristics were measured during ambulation for an initial barefoot baseline test, and with solid and hinged AFOs for the other two tests. Both orthoses increased stride length, reduced abnormal ankle plantar flexion during initial contact, midstance and terminal stance (TST), and increased ankle plantar flexor moments closer to normal during TST. Hinged AFOs increased ankle dorsiflexion at TST and increased ankle power generation during preswing (PSW) as compared to solid AFOs, and increased ankle dorsiflexion at loading compared to no AFOs. No other significant differences were found for the gait variables when comparing these orthoses. Either AFO could be used to reduce the excessive ankle plantar flexion without affecting the knee position during stance. The hinged AFO would be recommended to produce more normal dorsiflexion during TST and increased ankle power generation during PSW in children with spastic diplegic CP.

Keywords: Cerebral palsy; Gait analysis; Orthosis

1. Introduction

A common gait deviation in children with spastic diplegic cerebral palsy (CP) is dynamic equinus or excessive ankle plantar flexion during stance in ambulation without fixed contracture of the triceps surae muscle group [1]. Premature plantar flexion moments, and early onset and prolonged firing of the triceps surae muscle group are present during stance [2–4]. Excessive knee flexion and abnormal knee extensor moments during stance often accompany the equinus positioning [1,5].

Various ankle-foot orthoses (AFOs) have been used to correct the equinus gait pattern in children with spastic CP [6]. The solid or fixed polypropylene AFO has been traditionally used to decrease equinus positioning and prevent ankle plantar flexor contractures [7]. A disadvantage of the solid AFO is its limitation of normal movement of the tibia forward over the weightbearing foot resulting in decreased ankle dorsiflexion and early heel rise in stance [8,9]. The hinged or articulated polypropylene AFO with a plantar flexion stop has been increasingly recommended by clinicians to decrease equinus positioning [6]. Unlike the solid AFO, the hinged AFO allows the tibia to move forward over the weightbearing foot during stance resulting in more normal ankle dorsiflexion [6,8].

Few published studies have examined the differences between these two types of orthoses during ambulation. Middleton et al. [10] compared the solid and hinged orthoses in a case study of one child with spastic diplegia and found reduced knee extensor moments during early stance and more normal ankle dorsiflexion motion after midstance with hinged AFOs. Rethlefsen et al's. [11] study comparing gait

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with shoes, solid and hinged AFOs in children with spastic diplegic CP showed that dorsiflexion was greatest at terminal stance with the hinged AFO, but no differences in stride length or walking velocity were found.

The purpose of this study was to compare the effects of solid and hinged AFOs on ambulation in children with spastic diplegic CP who demonstrate excessive ankle plantar flexion motion during stance as measured while barefoot during the baseline test. It was hypothesized that the hinged AFO as compared to the solid AFO would produce more normal ankle and knee joint motions, moments and powers, increased walking velocity and stride length, reduced cadence, and more normal timing of the triceps surae, pretibial, quadriceps femoris and hamstrings muscle groups during ambulation.

2. Methods

2.1. Subjects

Six females and six males with an average age of 7.5 years (S.D. 3.83; range 4–16) with spastic diplegic CP were recruited to participate in the study from the outpatient clinic at Shriners Hospital for Children in San Francisco, CA. Written informed consent was obtained from the parents of all subjects, and from children who were 12 years and older prior to participation in the study. The Institutional Review Board at the University of California at San Francisco approved the study.

All subjects were community ambulators who demonstrated the following characteristics: (1) ankle dorsiflexion to 0° in weightbearing during static standing; (2) excessive ankle plantar flexion of 5° or more during stance in gait; (3) passive ankle dorsiflexion to 5° with knee extended; (4) passive hip extension to -10° or less as measured by the Thomas test; (5) passive hamstring length of 50° or more as measured by a straight leg raise; and (6) mild spasticity of the triceps surae, hamstrings and quadriceps or a score of 1 on the Ashworth scale indicating minimal resistance at the end range of passive motion. None of the subjects had ever undergone Achilles tendon or gastrocnemius lengthening surgical procedures in the past or any other orthopedic surgery during the year prior to the study. Ten subjects ambulated without assistive devices, one subject used bilateral quad canes, and one subject used loftstrand crutches. Nine subjects wore solid AFOs and three subjects used hinged AFOs for at least 1 year prior to participation in this study.

2.2. Procedures

A repeated measures design with subjects serving as their own controls was used for the study [12]. Each subject wore (1) no orthoses for an initial 2-week baseline period, (2) solid AFOs for 1 month, (3) no orthoses for 2 weeks, and (4) hinged AFOs for 1 month. The order of wearing either solid or hinged AFOs was randomly assigned. The orthoses were worn for 1 month so that the subjects could adapt to wearing them for the entire day. The 2-week period with no orthoses allowed the subject to adapt to ambulating without orthoses, but prevented fixed contractures of the triceps surae from developing.

The solid and hinged AFOs were custom-made for each subject from the same positive mold after casting the lower extremity with the subtalar joint aligned in the neutral position by the orthotist at Shriners Hospital for Children in San Francisco, California. Both orthoses were fabricated from 4.8 millimeter thick copolymer extending distally under the toes and on the mediolateral border of the foot, and proximally on the posterior leg to about 2.5–5 cm below the knee with trimlines anterior to both malleoli and straps across the front of the ankle and anterior upper tibia [7]. The solid AFO fixed the ankle at 0° of dorsiflexion and prevented plantar flexion. The articulated AFO with a Gillette hinge located at distal tip of the malleoli allowed free dorsiflexion, but had a plantar flexion stop at 0° of dorsiflexion [6].

All subjects were tested in the Orthopaedic Biomechanics Laboratory at Shriners Hospital for Children in San Francisco, CA at the end of the initial 2-week period with no orthoses for a baseline measurement, the 1-month period wearing solid AFOs, and the 1-month period wearing hinged AFOs. The subjects were tested barefoot for the baseline measurement, and with shoes and orthoses for the other two tests. Subjects were tested barefoot after the initial 2-week period because other comparative AFO studies examined the barefoot condition [9,13]and gait studies conducted in this Laboratory and in similar gait laboratories typically examine children ambulating without shoes for baseline studies to determine gait deviations [14].

The following gait measurement devices were used in the study: (1). Telemetered surface electromyography (EMG) to determine timing of the hamstrings, quadriceps femoris, triceps surae, and pretibial muscle groups during the stance phase; (2) three-dimensional motion analysis to determine sagittal joint motions of the knee and ankle during the stance phase and temporal-distance characteristics including walking velocity, stride length and cadence; and (3) force plates to determine knee and ankle sagittal joint moments and powers during the stance phase.

Active surface EMG electrode pairs (B. & L. Engineering, Santa Fe Springs, CA) requiring no skin preparation and measuring 1.14 cm in diameter with a fixed inter-electrode distance of 1.97 cm were placed on the skin over the muscle belly. The electrode pairs were applied to four muscle groups of one of the lower extremities that had the greatest amount of excessive ankle plantar flexion while barefoot during the stance phase of gait. The electrode sites were based on procedures described by Delagi et al. [15] for the hamstrings (long head of biceps femoris), quadriceps femoris (rectus femoris), triceps surae (lateral head of gastrocnemius), and pretibial muscle groups (tibialis anterior).

Contact-closing footswitches were placed and taped along the entire plantar surface of both feet for the barefoot baseline test and to the shoes for tests with solid and hinged AFOs. Footswitch on-off signals for initial contact and toe-off were used as event markers to indicate stance and swing phases of the gait cycle. EMG and footswitch data were recorded simultaneously with CODAS data collection software (Dataq Instruments Inc., Akron, OH) as the subject ambulated on a 10-m walkway at a self-selected speed for two walking trials for each condition. One quiet resting EMG trial was also collected for each condition. Telemetered EMG signals were amplified with a 5,000 gain factor, sampled at a rate of 2500 cycles per second, and collected with a passband of 30–500 Hz. The EMG signals were processed using the EMG analyzer software program QUANT (B. & L. Engineering, Santa Fe Springs, CA) to determine onset and cessation times for each muscle. A noise level was first established by quantifying the quiet resting trial. For all walking trials, the EMG signal was rectified and summed only when the signal exceeded the noise level and was at least 5% of the maximum EMG voltage recorded during gait. EMG muscle timing, defined as the duration of muscle firing, of the four muscle groups during the stance phase was calculated with QUANT for two walking trials that included 10–12 gait cycles per condition that were averaged for each subject.

A computerized, three-dimensional Motion Analysis™ system (Motion Analysis Corporation, Santa Rosa, CA) and two force plates (Kistler Instrument Corporation, Amherst, NY) were used to collect kinematic and kinetic data, and temporal-distance characteristics. The subject ambulated barefoot, and with solid and hinged AFOs at a self-selected speed for at least two walking trials per condition on a 10 m walkway with two force plates embedded in the floor. Each walking trial included two to three gait cycles. Twenty-one retroreflective markers using a modified Helen Hayes Hospital marker set [16], were attached directly to the skin with tape or straps over selected anatomical landmarks on the bilateral upper extremities, lower extremities and pelvis as previously published [13]. Six video cameras (NEC Corporation, Tokyo, Japan) recorded the images from the markers in the sagittal, coronal, and transverse planes at a sampling rate of 60 frames per second.

Data were processed with ExpertVision and Orthotrak 2.5 software packages (Motion Analysis Corporation, Santa Rosa, California) to determine the following: (1) joint motions, internal moments normalized to body weight, and powers for the ankle and knee in the sagittal plane; and (2) temporal-distance gait characteristics including walking velocity (distance/time in cm/s), cadence (steps/min), and stride length (distance in centimeters between two consecutive initial contact on one foot). Two walking trials with four to six gait cycles per condition were averaged for each subject.

Data from the lower extremity with the greatest amount of excessive ankle plantar flexion while barefoot during stance were used in the analysis of joint motions at specific gait events, and peak joint moments and powers during gait phases. The gait events for joint motions in the normalized gait cycle were identified as follows: (1) initial contact (IC) = 0%; (2) loading (LD) = 9%; (3) midstance (MST) = 25%; and (4) terminal stance (TST) = 42%. The gait phases used in the analysis of peak joint moments and powers in the normalized gait cycle were identified as follows: (1) LD = 0–10%; (2) MST = 10–30%; (3) TST = 30–50%; and preswing (PSW) = 50–60%.

2.3. Data analysis

Descriptive statistics including group means and standard deviations for the subjects were calculated for the test periods with solid and hinged AFOs and while barefoot for the baseline measurement for the following variables: (1) temporal-distance gait characteristics including walking velocity, stride length and cadence; (2) muscle timing for the four lower extremity muscle groups during the stance phase; (3) sagittal plane knee and ankle joint motions at the gait events of IC, LD, MST, and TST; and (4). sagittal plane peak knee and ankle joint moments and powers during the LD, MST, TST and PSW phases of gait. One-way analysis of variance (ANOVA) with repeated measures [12] was used to examine the barefoot baseline measurement, and the effects of solid AFOs and hinged AFOs on temporal-distance gait characteristics, muscle timing, sagittal plane knee and ankle joint motions, peak moments and powers at an α level of 0.05. For all significant ANOVA tests, three post-hoc pairwise comparisons between solid and hinged AFOs, no AFOs and solid AFOs, and no AFOs and hinged AFOs using Tukey's Honestly Significant Difference (HSD) Test [12] were conducted to determine significant differences at an α level of 0.05. Statistical analyses were performed using the SYSTAT computer software package (SPSS Inc., Chicago, IL).

3. Results

3.1. Temporal-distance gait characteristics

Findings showed no significant differences in walking velocity and cadence when comparing no AFOs, solid and hinged AFOs (Fig. 1). There was a significant difference in stride length ($F = 7.83$, d.f. $= 2, 22$) with post-hoc tests showing significant mean differences between no AFOs and solid AFOs, and between no AFOs and hinged AFOs. The mean stride length was increased with both solid ($x =$ 87 cm, S.D. = 18) and hinged AFOs ($x = 85$ cm, S.D. = 20) when compared to no AFOs ($x = 76$ cm, S.D. = 18). The mean stride length was similar for solid and hinged AFOs as there was no significant difference between the two orthoses.

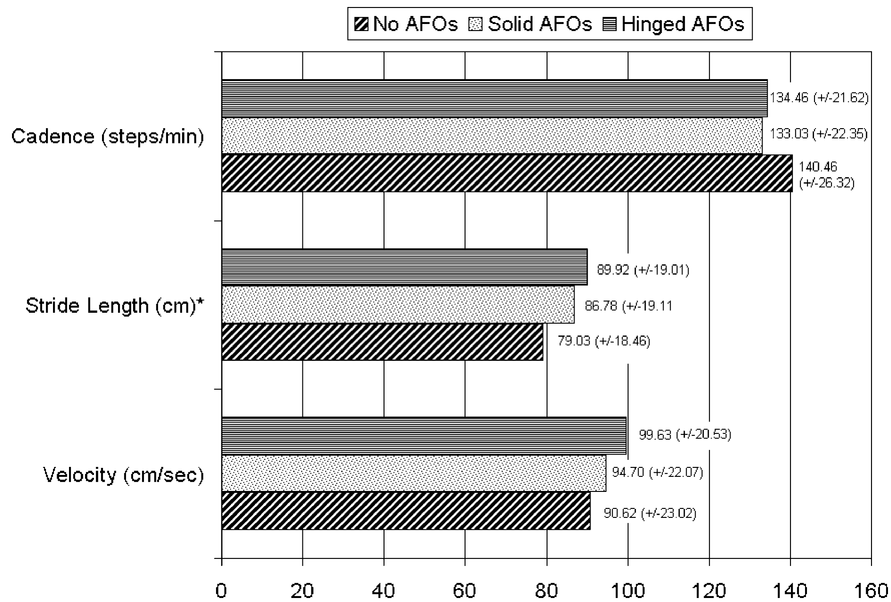

Fig. 1. The mean and standard deviations for walking velocity (cm/s), stride length (cm) and cadence (steps/min) for 12 subjects with and without AFOs. *Stride length is significant at $P < 0.05$.

3.2. Muscle timing

Findings showed no significant differences in timing of the pretibial, triceps surae, quadriceps femoris, and hamstrings muscle groups during the stance phase when comparing ambulation with no AFOs, solid and hinged AFOs (Fig. 2). Muscle timing is expressed as a percentage of the stance phase with all muscles active at initial contact or 0% of the gait cycle. As compared to normal children studied by Sutherland et al. [17], the triceps surae muscle group fired prematurely and all muscle timing was prolonged during stance in the subjects with spastic diplegic CP (Fig. 2).

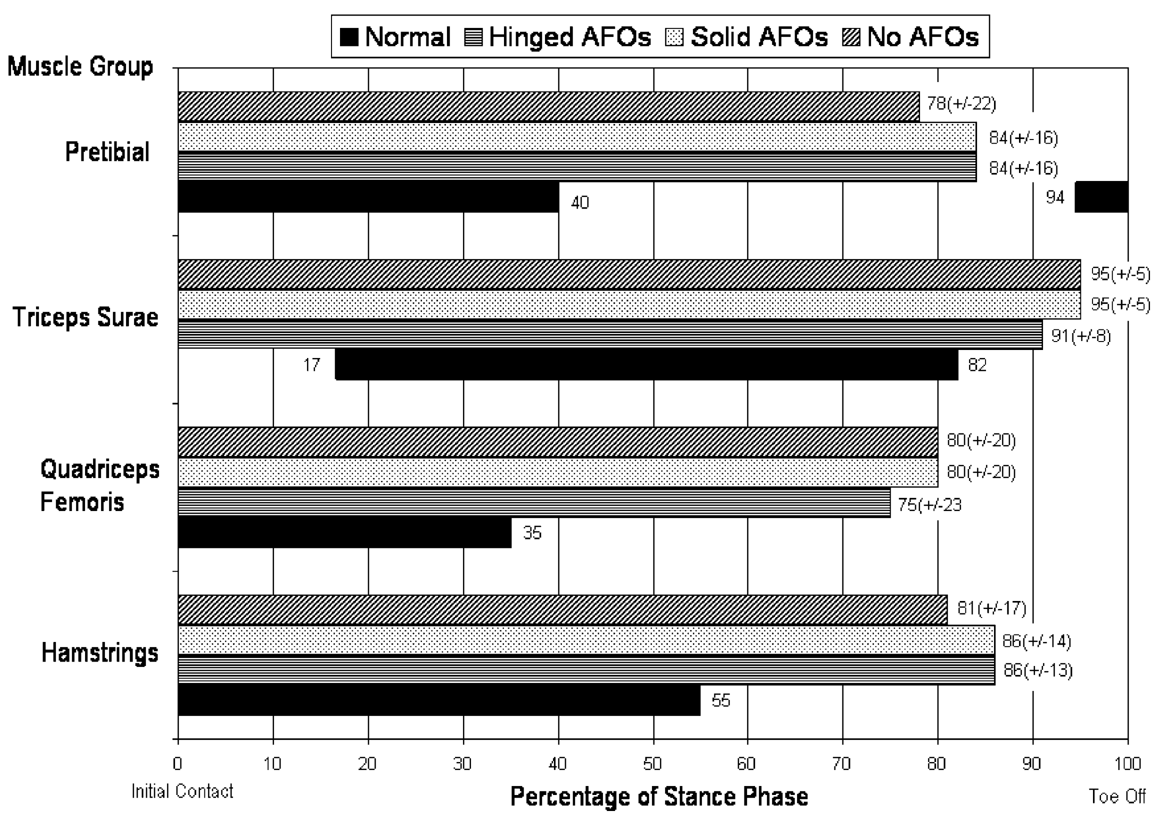

Fig. 2. The mean times (±standard deviation) for onset and cessation of muscles during stance phase of gait for 12 subjects with and without AFOs as compared to normal children [17].

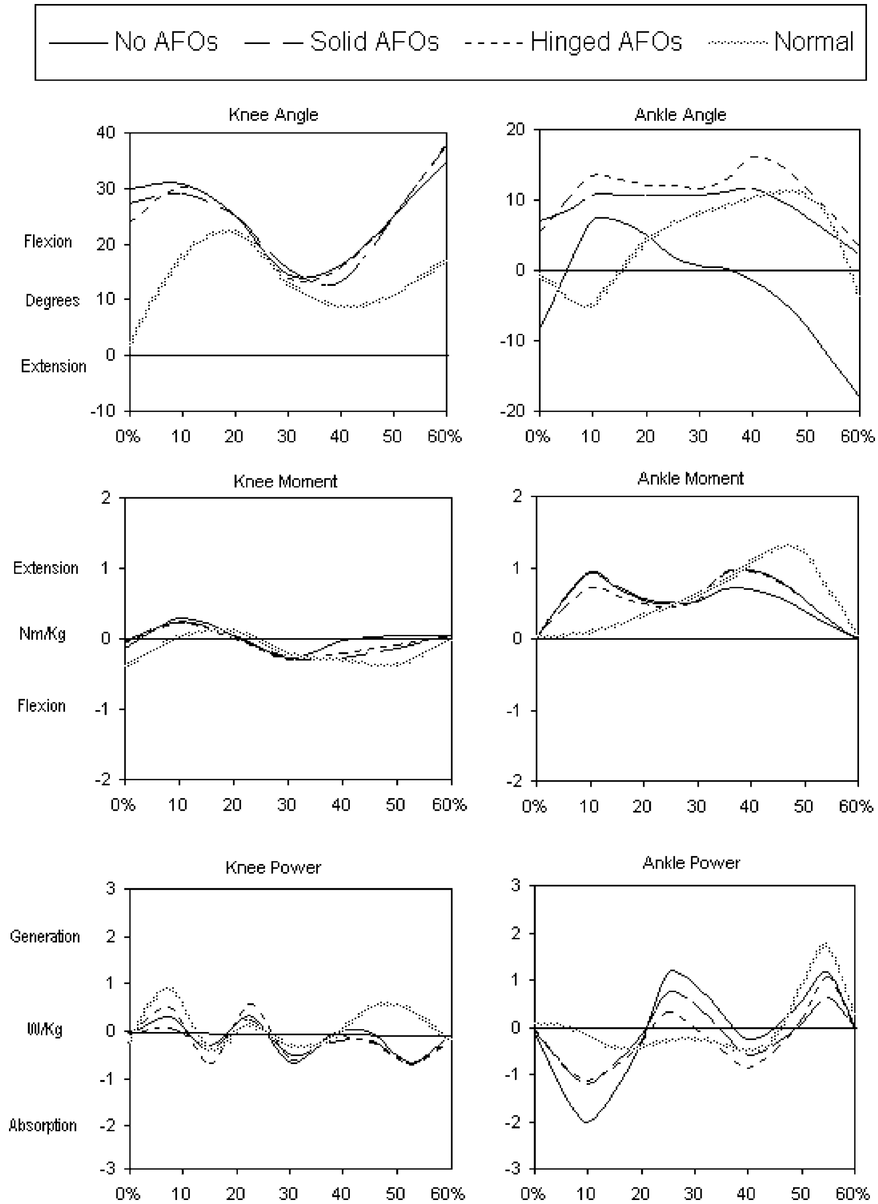

Fig. 3. The mean knee and ankle sagittal joint angles ($N = 12$), internal moments and powers ($N = 10$) during stance phase of gait with and without AFOs as compared to normal children (Orthotrak 2.5). Stance phase represented by 0–60% of gait cycle with initial contact (0%), loading response (0–10%), midstance (10–30%), terminal stance (30–50%), preswing (50–60%) and toe-off (60%).

3.3. Sagittal plane joint motions

Findings showed no significant differences when comparing no AFOs, solid and hinged AFOs in knee flexion/extension at IC, LD, MST, and TST (Fig. 3). There were significant differences for ankle dorsiflexion/plantar flexion at IC ($F = 39.92$, d.f. $= 2, 22$), LD ($F = 4.73$, d.f. $= 2, 22$), MST ($F = 25.14$, d.f. $= 2, 22$) and TST ($F = 37.36$, d.f. $= 2, 22$). Post-hoc tests showed significant differences in ankle dorsiflexion/plantar flexion between no AFOs and solid AFOs at IC, MST, and TST, no AFOs and hinged AFOs at IC, LD, MST, and TST, and solid AFOs and hinged AFOs at TST (Table 1). As compared to normal children [14; Orthotrak 2.5], the abnormal amount of ankle plantar flexion at IC, MST, and TST in the subjects with spastic diplegic CP was reduced with both orthoses (Fig. 3). The hinged AFO produced more normal dorsiflexion at TST than the solid AFO, and more excessive dorsiflexion during LD than no AFO (Fig. 3).

3.4. Sagittal plane joint moments and powers

Kinetic data were deleted for two subjects who ambulated with canes and crutches as these assistive devices contacted the force plates producing inaccurate data. Findings showed no significant differences when comparing no AFOs (barefoot), solid and hinged AFOs in peak knee moments during LD, MST, TST phases, and peak ankle moments dur-

Table 1
Group means, standard deviations and pairwise comparisons for sagittal ankle joint motions (°) at initial contact, loading, midstance, and terminal stance; $N = 12$

Variable	No AFOs		Solid AFOs		Hinged AFOs	
	Mean	S.D.	Mean	S.D.	Mean	S.D.
Initial contact (0%)						
Ankle dorsi/plantar flex[a]	−8.14[b,c]	5.46	7.09[b]	5.06	5.37[c]	7.00
Loading (9%)						
Ankle dorsi/plantar flex	7.15[c]	6.01	10.70	5.91	13.30[c]	7.54
Midstance (25%)						
Ankle dorsi/plantar flex	0.69[b,c]	4.30	10.59[b]	4.93	11.67[c]	7.00
Terminal stance (42%)						
Ankle dorsi/plantar flex	−1.30[b,c]	6.59	11.50[b,d]	4.28	16.13[c,d]	6.17

[a] (+) Value denotes dorsiflexion; (−) value denotes plantar flexion.
[b] No AFOs and solid AFOs (significant $P < 0.05$).
[c] No AFOs and hinged AFOs (significant $P < 0.05$)
[d] Solid and hinged AFOs (significant $P < 0.05$).

Table 2
Group means, standard deviations and pairwise comparisons for peak sagittal ankle joint moments (in N m/kg) and powers (in W/kg) during terminal stance (40–50%) and preswing (50–60%) phases; $N = 10$

Variable	No AFOs		Solid AFOs		Hinged AFOs	
	Mean	S.D.	Mean	S.D.	Mean	S.D.
Terminal stance						
Ankle moments[a]	0.69[b,c]	0.14	0.96[b]	0.22	0.94[c]	0.25
Ankle powers[d]	−0.26[b]	0.33	−0.60[b,e]	0.24	−0.87[e]	0.42
Preswing						
Ankle powers[d]	1.16[c]	0.39	0.62[e]	0.31	1.07[c,e]	0.46

[a] (+) Value denotes plantar flexor moment; (−) value denotes dorsiflexor moment.
[b] No AFOs and solid AFOs (significant at $P < 0.05$).
[c] No AFOs and hinged AFOs (significant at $P < 0.05$).
[d] (+)Value denotes power generation; (−) value denotes power absorption.
[e] Solid and hinged AFOs (significant at $P < 0.05$).

ing LD and MST phases. There was a significant difference in peak ankle moments during the TST phase ($F = 6.94$, d.f. $= 2, 18$) with post-hoc tests showing significant differences between no AFOs and solid AFOs, and no AFOs and hinged AFOs (Table 2). The plantar flexion moments during TST phase were similar for solid and hinged AFOs. Both orthoses produced increased ankle plantar flexion moments closer to normal during TST phase when compared to no AFOs (Fig. 3). As compared to normal children [14; Orthotrak 2.5], these peak ankle plantar flexion moments during barefoot gait were increased prematurely during LD phase and decreased during TST phase in the subjects with spastic diplegic CP (Fig. 3).

Results showed no significant differences when comparing no AFOs, solid and hinged AFOs in peak knee powers during LD, MST, and TST phases, and peak ankle powers during LD and MST phases. There was a significant difference in peak ankle powers during PSW ($F = 7.81$, d.f. $=$ 2, 18) with post-hoc tests showing significant differences between no AFOs and solid AFOs, and hinged AFOs and solid AFOs (Table 2). The peak ankle power was similar for no AFOs and hinged AFOs during PSW. The hinged AFOs produced increased ankle power generation during PSW when compared to solid AFOs (Fig. 3). As compared to normal children [14; Orthotrak 2.5], the power generation of the plantar flexors during PSW was reduced in the subjects with spastic diplegic CP (Fig. 3).

4. Discussion

This study supported the benefits of using AFOs in children with spastic diplegic CP who demonstrate a dynamic equinus gait pattern with excessive ankle plantar flexion motion during the stance phase. Children wearing either the solid or hinged AFO showed significant gait improvements including a longer stride length that was closer to normal [17], and reduced abnormal ankle plantar flexion motion at IC, MST and TST. The hinged AFO produced significantly more normal ankle dorsiflexion motion at TST than the solid AFO which has been one of the important benefits purported by clinicians who recommend this orthosis [6,8]. Also, it allowed significantly more power generation to occur during PSW than the solid AFO indicating greater plantar flexor muscle concentric contraction for push-off. No other significant differences in temporal-distance gait characteristics, lower extremity muscle timing, sagittal ankle and knee joint kinetics, and sagittal knee kinematics between the two orthoses were shown, however.

The increased stride length with solid AFOs supported the findings of previous studies on orthotic use in children with cerebral palsy [9,13,18]. However, Rethlefsen et al. [11] found no difference in stride length when comparing shoes, solid and hinged AFOs which contradicts this study's findings of increased stride length with both orthoses. Different methodologies used in the studies could partially account for this discrepancy. Rethlefsen et al. [11] tested subjects for all three conditions in one single session while this study tested subjects after wearing the orthoses for monthly intervals.

Faster walking velocity can result from longer stride length and/or faster cadence [1]. However, the improved stride length that was closer to normal for both orthoses in this study was not enough to produce a significant increase in walking velocity. This finding was consistent with Carlson et al's. [18] study showing that increased stride length for both solid AFOs and supramalleolar (SMOs) orthoses as compared to shoes did not produce a faster walking velocity.

This current study did support previous findings [10,11] that the abnormal ankle plantar flexion motion at IC during gait without orthoses was reduced with both solid and hinged AFOs. However, the excessive ankle dorsiflexion motion at LD while barefoot [14] was not remedied by either orthosis. More ankle dorsiflexion than expected also occurred at MST and TST with the solid AFO due to the deformation of the polypropylene material during weightbearing. Other studies of solid AFOs have also shown ankle dorsiflexion of 8–11.9° during stance [11,13,18] due to polypropylene deformation that occurs even with these rigid AFOs.

The corresponding ankle joint kinetics during the LD phase while barefoot showed excessive and premature peak plantar flexor moments with excessive power absorption as also seen in other studies [1,4,18]. These findings indicate that excessive plantar flexor eccentric contraction is occurring as the ankle excessively dorsiflexes during LD in barefoot gait. The abnormal ankle moments persisted with both orthoses during LD, however, the power absorption was decreased and closer to normal. These results are consistent with Carlson et al's. [18] findings with solid AFOs, but have not been previously reported for hinged AFOs.

More normal dorsiflexion during TST was produced by the hinged AFO when compared to the solid AFO. These results confirmed previous research findings [10,11] and clinicians' observations [6,8] that the solid AFO limits the normal forward progression of the tibia over the weightbearing foot resulting in decreased ankle dorsiflexion and early heel rise. The hinged AFO has the advantage of allowing more normal dorsiflexion to occur during MST and TST as the tibia transitions over the foot [14].

The corresponding ankle joint kinetics while barefoot showed a reduced peak plantar flexion moment during TST and reduced power generation during PSW when compared to normal [14]. Both orthoses produced larger and closer to normal peak ankle plantar flexor moments during TST as also reported by Rethlefsen et al. [11]. The abnormally reduced power generation during PSW while barefoot was decreased more by the solid AFO than the hinged AFO, a result that is consistent with Rethlefsen et al's. [11] findings of higher power generation with the hinged AFO as compared to the solid AFO. This finding indicates that the hinged AFO allows greater plantar flexor concentric contraction for push-off during PSW than the solid AFO.

Excessive knee flexion during stance was evident in subjects during barefoot gait [14]. These abnormal knee motions were not changed with either hinged or solid AFOs as also reported by Rethlefsen et al. [11]. Clinicians' concerns regarding the possibility of more knee flexion for a crouched gait pattern as a result of hinged or solid AFOs were not substantiated [11]. The increased knee extensor moments often seen in children with spastic CP [4] were not present in this study's subjects which is possibly due to their lack of knee flexion contractures. Knee moments during stance were not changed in subjects wearing either orthosis. Middleton et al's. [10] findings of decreased excessive knee extension moments occurring during LD with hinged AFOs as compared to solid AFOs in one child with spastic diplegic CP were not fully substantiated by this study.

All four lower extremity muscles were active at IC, which is normal, except for the triceps surae muscle group that fired prematurely [17]. The duration of all four muscles' activity during stance was also excessively prolonged when compared to normal [17]. Although the abnormal ankle plantar flexion motion was reduced by both orthoses, and the hinged AFO produced more normal dorsiflexion at TST with more ankle plantar flexor concentric contraction during PSW, there was no accompanying change in the abnormal timing of the triceps surae muscle group during stance (Fig. 2). This finding that orthotic use does not change abnormal muscle timing was consistent with Radtka et al's. [13] study comparing solid and dynamic AFOs, and no orthoses. Rethlefsen et al. [11] also found no differences in peak EMG amplitude of calf muscles during gait between shoes, hinged and solid AFOs.

Several factors could have affected the outcomes of this study. Subjects were tested barefoot for the initial baseline measurement without orthoses, whereas shoes were worn for all gait tests with orthoses. Ambulation might have been affected by the use of shoes. However, Oeffinger et al. [19] compared gait with and without shoes and found that shoes have a small impact on gait kinetics and kinematics in healthy children. The small sample size also decreased the probability of finding a significant difference between the orthoses. Since nine subjects wore solid AFOs and three subjects used hinged AFOs for at least 1 year prior to the study, the 1-month period of wearing orthoses might not have been long enough to accommodate to the new AFO. However, no published evidence is available that supports the optimal length of time for full adaptation to a new AFO. The mild to moderate amount of excessive ankle plantar flexion during stance, the mild lower extremity muscle spasticity, and the lack of hip and knee flexor, and ankle plantar flexor contractures found in this sample of subjects with spastic diplegic CP limit the generalization of the results to similar children.

This study supports the use of either solid or hinged AFOs to reduce the dynamic equinus without affecting the knee position during stance in children with spastic diplegic CP. The hinged AFO would be recommended instead of the solid AFO to produce more normal ankle dorsiflexion during TST. The hinged AFO might be contraindicated if excessive dorsiflexion occurs during LD in a barefoot baseline test as seen in this study. Additional individual factors such as orthotic costs and the effects on other functional mobility should also be considered when selecting either orthosis. Individual differences in children also need to be addressed when orthotic recommendations are made because children with spastic diplegic CP are a heterogenous group showing variations in gait [1]. Future studies examining the effects of these orthoses are needed that include children with spastic CP having knee flexion contractures and moderate to severe amounts of equinus during ambulation. Additional research should compare the effects of hinged and solid AFOs on other functional activities such as gait during stair climbing.

5. Conclusions

Findings showed that both orthoses increased stride length, reduced the abnormal ankle plantar flexion at IC, MST, and TST, and increased the ankle plantar flexor moment closer to normal in TST. Hinged AFOs increased ankle dorsiflexion at TST and increased ankle power generation during PSW as compared to solid AFOs, and increased ankle dorsiflexion at LD compared to no AFOs. No other significant differences were found in lower extremity muscle timing, sagittal knee and ankle motions, peak powers and moments during stance, temporal-distance gait characteristics when comparing solid and hinged AFOs. Either AFO could be used to reduce the excessive ankle plantar flexion without affecting the knee position during stance in children with spastic diplegic CP. The hinged AFO would be recommended to produce more normal dorsiflexion during TST, but not if excessive dorsiflexion occurs during LD in a barefoot baseline test. This study should be replicated with more children with spastic CP having moderate to severe amounts of dynamic equinus during ambulation, and knee flexion contractures.

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