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The Effects of Ankle Foot Orthoses on Energy Recovery And Work During Gait In Children With Cerebral Palsy

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Abstract

Background—Studies suggest that 50% of children with cerebral palsy are prescribed ankle foot orthoses. One of the aims of ankle foot orthosis use is to aid in walking. This research examined the effects that ankle foot orthoses have on the energy recovery and the mechanical work performed by children with cerebral palsy during walking.

Methods—Twenty-one children with spastic diplegia walked with and without their prescribed bilateral ankle foot orthoses. Ten of the subjects wore articulated (hinged) orthoses and 11 subjects wore solid orthoses. Three dimensional kinematic data were collected and between and within group repeated measures ANOVAs were applied to the dependent measures.

Findings—The results were similar for both groups. There was an increase in stride length, energy recovery, and potential energy and the kinetic energy variation. There was no change in the mechanical work performed to walk or the normalized center of mass vertical excursion. Unfortunately, the increase in energy recovery did not alter the external work, as it was offset by increased variation in the potential and kinetic energies of the center of mass. There was a great deal of variability in the measured work, with both large increases and decreases in the work of individual subjects when wearing orthoses.

Interpretation—These results suggest that current ankle foot orthoses can reduce the work to walk, but do not do so for many children with cerebral palsy. This research suggests that ankle foot orthosis prescription could be aided by measuring the mechanical work during walking.

INTRODUCTION

Individuals with walking disabilities, as a result of cerebral palsy (CP) are often prescribed ankle foot orthoses (AFOs) to aid in their walking. United Cerebral Palsy reports that an estimated 764,000 people in the United States have one or more symptoms of CP and it has been reported that more than 50% of these individuals are prescribed orthoses (Knutson & Clark 1991). While AFOs are prescribed for several reasons, one goal is to improve the walking of the AFO wearer. Unfortunately, the improvements in walking mechanics and metabolic costs of current AFOs are small. Even with AFOs, children with CP commonly expend two to three times as much energy to walk as typically developing children (Brehm et al. 2008). Hence, the goal of this research was to provide insights into the effects of AFOs

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on the gait of children with CP by examining the mechanical work performed and the center of mass (CoM) mechanics.

Several studies have investigated the effects of AFOs on the gait of children with CP. Many of these studies find a small but significant improvement in gait with the use of AFOs. Studies of both spastic di- and hemiplegic subjects have found that wearing AFOs, compared to walking barefoot, resulted in increases in stride length (Abel et al. 1998; Balaban et al. 2007; Buckon et al. 2001; Buckon et al. 2004; Carlson et al. 1997; Dursun et al. 2002; Lam et al. 2005; Romkes & Brunner 2002; Smith et al. 2009; White et al. 2002), and increases in comfortable walking velocity (Abel et al. 1998; Balaban et al. 2007; Brehm et al. 2008; Dursun et al. 2002; Franceschini et al. 2003; Smith et al. 2009; Van de Walle et al. 2009; White et al. 2002). Studies have also found AFO use can reduce the metabolic cost of walking (Balaban et al. 2007; Buckon et al. 2004; Franceschini et al. 2003; Maltais et al. 2001). However, other studies, some by the same researchers, have found no change in stride length (Carlson et al. 1997; Rethlefsen et al. 1995; Smiley et al. 2002), velocity (Buckon et al. 2001; Buckon et al. 2004; Carlson et al. 1997; Lam et al. 2005; Mossberg et al. 1990; Rethlefsen et al. 1995; Romkes & Brunner 2002; Smiley et al. 2002) or metabolic cost (Brehm et al. 2008; Buckon et al. 2001; Maltais et al. 2001; Suzuki et al. 2000). In addition, several studies (Buckon et al. 2004; Radtka et al. 1997; Radtka et al. 2005; Rethlefsen et al. 1999; Smiley et al. 2002) found the type of AFO worn had little effect upon gait. Thus there remains a need to better understand the effects of AFOs on the gait of children with CP.

The approach applied in this study was to examine the center of mass (CoM) mechanics and energetics of subjects walking with and without their AFOs. This approach, which requires measuring full body kinematics, can reveal the effects of AFOs on the mechanical work performed to walk (W_{ext}) as well as the structure of gait. Bennett et al. (2005) used this approach to compare the gait of children with diplegic CP to that of typically developing children. They found that the children with CP performed more than twice as much mechanical work on their CoM as the typically developing children. This was a result of both a greater vertical excursion of their CoM and a poorer phasic relationship between their kinetic and potential energies. These factors resulted in the children with CP being able to recover/transfer only 44% of the energy available while the age matched controls were similar to adults and recovered 66% of the available energy. These results are in line with the earlier two dimensional work of Olney et al (1987) who reported that in children with hemiplegic CP a “poor pattern of exchange between potential and kinetic energy of the HAT (head, arms, torso) segment contributed to high total energy costs.” (Olney et al. 1987) (p1354). Researchers have also used COM analysis in studying patients with lower-limb amputations, (Tesio et al. 1998) patients with gait disorders, (Detrembleur et al. 2000) and those requiring total hip replacements. (Iida & Yamamuro 1987)

The present study examined the CoM mechanics and energetics to gain greater insight into how walking is affected by the use of AFOs. Specifically, this study investigates the effects of both articulated (hinged) and solid AFOs on the mechanical work performed, the CoM vertical movement, the transfer of energy between potential and kinetic forms of the CoM and the character of the energies of the CoM in children with diplegic cerebral palsy. We hypothesized that the gait would be more pendular in nature resulting in an improved exchange between potential (PE) and kinetic energy (KE) (recovery factor) and a reduction of mechanical work performed when children wear their AFOs. We also hypothesized, based on prior work (Abel et al. 1998), that the normalized CoM excursion would not be changed by AFOs. We hypothesized that these changes would occur with both styles of AFOs.

METHODS

Subjects

3-D kinematic data were collected from a convenience sample of 21 children with spastic diplegic CP that wore AFOs. Eleven of the subjects wore hinged AFOs with a plantarflexion stop (HAFOs) and 10 children wore solid, AFOs (SAFOs). The subjects consisted of 16 males and 5 females. Their average height, age, and Gross Motor Function Measure (GMFM) (Russell et al. 1993) were 1.39 ± 0.21 m, 9.6 ± 4.12 yrs, and 93 ± 7 , respectively. The two groups were similar with no statistical difference between them in their anthropometrics or GMFM scores. None of the subjects had undergone surgery or other significant treatments within the last twelve months. All tests were conducted in the Motion Analysis and Motor Performance Laboratory at the University of Virginia. Subject assent and parental consent was approved by the University of Virginia's Human Investigation Committee and was obtained for all subjects. All subjects walked at their preferred walking speed. CoM position was computed with using the full body model validated by Eames et al (1999). To allow comparison between gaits of different stride lengths CoM vertical excursion for each trial was normalized by the vertical excursion of a compass gait model with matching leg and step lengths (Bennett et al. 2005). The kinematic data were used to compute kinetic energy (KE), potential energy (PE), and external work on the CoM. The relative phase and the peak-to-peak ratio of the PE to KE were calculated as by Bennett and colleagues (2005). The energy recovery factor, the percentage of mechanical energy recovered via exchange between KE and PE in the CoM movement, was computed as described by Winter (1990).

Test Protocol and Data Analysis

Procedures and Gait Parameters—A full body marker set (modified Helen Hayes) of 38 markers was attached to the subjects after anthropometric measurements were taken. After a static trial subjects walked at their self-selected speed and three-dimensional kinematic data were collected at 120 Hz using an eight camera Vicon Motion Analysis System (ViconPeak, Denver, CO).

The measurement volume of the Vicon System allowed the capture of two to five strides in each trial. Subjects performed multiple trials. The average velocity of all trials was computed and the two trials closest to the mean were analyzed. Gait analysis and inverse dynamics were performed with VICON Bodybuilder software.

CoM parameters—CoM position was computed from the kinematic data with a model validated by Eames et al. (1999). The model employed 13 segments and used the regression equations to define segment properties for children developed by Jensen (1986). The model has been validated for children with CP by comparison to force plate computations of CoM movement (Bennett et al. 2005).

CoM vertical excursion is a function of both leg length and step length. As step length could be expected to change with AFO use, CoM vertical excursion for each trial was normalized by the vertical excursion of a simple compass gait model (Kerrigan et al. 1996) with matching leg and step lengths. The simple compass gait model represents each lower limb as a single segment. This model assumes the COM follows the path of the hip joint, an arc with the radius of the leg length.

By normalizing the CoM excursion we can gain insight into whether the form of the movement changed. That is, if one were to walk with longer steps but with the same vertical excursion, all other factors remaining the same, the work per distance traveled would decrease. Normalized CoM values would reflect this, where the measured values would not.

The CoM potential and kinetic energies per unit mass were computed as described in Bennett et al. (2005). The energy ratio of PE to KE was computed from the average of the peak-to-peak values. The continuous relative phase between the two energy components was computed with a cross-spectral density function (Duarte 2002).

Energy exchange was quantified by the method of Winter (1990). The energy recovery factor, R , the percentage of mechanical energy recovered via exchange between kinetic and potential energy in the CoM movement was computed as (Winter 1990):

$$R = 100 * \frac{(W_{ne} - W_{ext})}{W_{ne}}$$

Where W_{ne} , the work done assuming no exchange, and W_{ext} the work done on the CoM allowing energy exchange, are computed as described in Bennett, et al. (2005). In the ideal case (e.g. a frictionless pendulum) of complete energy transfer $W_{ext} = 0$ and $R = 100$. If there is no energy transfer $W_{ne} = W_{ext}$ and $R = 0$. The dependent measures from the CoM analysis included the CoM vertical excursion, both measured and normalized, the external work on the CoM, recovery factor, relative phase of the two energy curves, and relative amplitudes of the PE to the KE.

A harmonic analysis of both energy curves within each stride cycle was performed to assess the symmetry of the gait and pendular nature of the gait for children with and without AFOs. The energy curves were detrended and normalized so the peak-to-peak amplitudes were one. The coefficients of the first five harmonics were estimated by fitting the data with the following Fourier-series:

$$x(t) = a_o + \sum_{k=1}^6 a_k \cos(2\pi k f t + \Phi_k)$$

where a_o is the mean position over the stride cycle; a_k and Φ_k are the amplitude and phase of the k^{th} harmonic, respectively; and f is the step frequency. Since there are two energy oscillations per stride the terms with even k , which make equal contributions to each oscillation, reflect the symmetric aspects of CoM movement and are referred to as Intrinsic Harmonics (Cappozzo 1984). The term for $k=2$, the step frequency, is known as the Fundamental Intrinsic Harmonic (FIH). Terms with odd k reflect asymmetries in the gait as they make unequal contributions to the oscillations and are called Extrinsic Harmonics. The term at $k=1$, the stride frequency, is known as the Fundamental Extrinsic Harmonic (FEH).

Statistical analysis revealed no differences between the two groups for any of the dependent measures and the data was subsequently analyzed with a within group repeated measures analysis. The main dependent measures were the mechanical work performed, the percentage of recovery recovered (recovery factor), the relative phase between the potential and kinetic energy curves, and the peak to peak energy variation of the potential to kinetic energies and their ratio. In addition the stride length, walking speed, and normalized CoM vertical movement were analyzed. The effect size was calculated by subtracting the means of outcome measures with and without AFOs and dividing them by the standard deviation of difference scores. The Shapiro-Wilk's W test was used to verify that the data were normally distributed. A significance level of 0.01 was used.

RESULTS

The gait parameters and CoM mechanics with and without AFOs are summarized in Table 1. The analysis found no decrease in work performed to walk, despite an increase in energy recovery with AFO use. Stride length, CoM excursion (PE variation), KE variation, and recovery factor all significantly increased with AFO use, with tendencies for improved PE to KE ratio and relative phase and gait velocity. However, as discussed below, there was a great deal of variability in the measures related to the work performed with most coefficients of variation greater than 20%.

The use of AFOs resulted in an increase in stride length of 13% ($p < 0.005$) and there was a tendency for the gait speed to increase. There was a 38% increase in the peak to peak variation of KE and a 21% increase in the vertical CoM excursion (PE variation) with the use of the AFOs ($p < 0.01$), see Figure 1. However, there was no change in the normalized CoM excursion.

Figure 2 shows the amplitudes of the Fourier coefficients of the PE and KE curves for the two conditions. Post Hoc analysis (Tukey's honest significant difference test) revealed that the only difference in coefficients with AFO use occurred in the Fundamental Intrinsic Harmonic (FIH) coefficients ($n=2$) of the PE curves ($p < 0.001$). That is, the use of AFOs made the PE curve more like a pure sine wave. There was a trend ($p = 0.02$) for the kinetic energy variation to be more symmetric between legs (FEH) but otherwise the use of AFOs had little effect upon the shape of the KE curves.

While overall there were small changes in the dependent measures of the groups, there was a great deal of variation within the groups. Figure 3 illustrates the distribution of changes in the dependent measures among the subjects. Here the responses are divided into three groups: Positive (reflecting an improvement in gait/reduction of work), No Change, and Negative (reflecting a worsening in gait/increase in work). A 10% level of change was used as a cutoff. Note that for most measures there were subjects in all three categories and increases in the energy variations were the most consistent changes seen with AFO use.

DISCUSSION

In diplegic CP, spasticity, abnormal muscle activation patterns, and muscle tightness result in an energy expensive gait. Individuals with equinus, exhibited by all subjects in this study, must have increased knee flexion at foot strike to keep the body center of mass over the base of support; the forefoot (Abel et al. 1998). However, this strategy (the knee never straightens completely during stance) lowers the nadir of the CoM vertical path, that occurs during double support, resulting in increased CoM excursion and mechanical work to walk (Bennett et al. 2005). With weight acceptance, the ankle and foot move in the direction of dorsiflexion, but this movement is interrupted by premature plantarflexion. Plantarflexion persists through the remainder of stance tending to increase the peak CoM height (and reducing late cycle power generation). However, this effect is offset by the flexed knee in stance (Abel et al. 1998; Perry 1992). The overall result is an exaggerated vertical excursion of the body center of mass rather than the normal, low-amplitude sinusoidal movement (Bennett et al. 2005). Consequently, the diplegic person with equinus performs more work to walk (Bennett et al. 2005). AFOs are prescribed for these patients, in part, to improve their walking by changing their knee/ankle angles. The present work examined how AFO use modified the CoM mechanics and energetics of walking in patients with diplegic CP.

The measured changes in traditional gait parameters of children with CP due to the wearing of AFOs were comparable to results of previous studies. The increase in stride length with the solid and articulated braces found in this study are with consistent with the 8% to 18%

increases seen in several studies (Abel et al. 1998; Buckon et al. 2004; Dursun et al. 2002; Radtka et al. 2005; Romkes & Brunner 2002). As in the present research these studies compared barefoot walking to walking in AFOs and shoes. We also found a tendency to increase preferred walking velocity with AFO use. Previous research has been equivocal with some researchers finding an increase in preferred walking velocity (Abel et al. 1998; Dursun et al. 2002; White et al. 2002) and others finding no change in preferred velocity (Buckon et al. 2004; Mossberg et al. 1990; Rethlefsen et al. 1999).

Statistically we found little effect of the AFOs on the mechanics or energetics of the walkers' CoMs. The hypothesis that the normalized CoM vertical excursion would not change with AFO use was found to be true, because the increase in CoM excursion was merely the result of a longer stride length rather than a change in movement form. The hypothesis that the gait would be more pendular with an improved recovery factor was also found to be true, but it did not result in a reduction in the external work on the walker's CoM as the improved mechanics were offset by increased variations of the energies.

The increase in energy variation in both the KE and PE and the improved relative phase for the AFO condition can be seen in Fig. 1. It is typical that increased stride length increases PE variation and it is one reason why we are less efficient when we walk faster than our preferred comfortable speed with longer strides. The lack of change in the normalized CoM excursion reveals that the changes in excursion reflect the geometric decrease in CoM height in double support that accompanies an increased stride length. Fortunately, AFO use did offset the increased energy variation with improved energy recovery but the improvements were insufficient to decrease the net work.

The lack of effect of the AFOs on the form of the gait is reflected in the lack of change seen in the Fourier analysis of the CoM movement. Changes in the Fourier coefficients with AFO use reveal the PE curve became somewhat more sinusoidal with an increase in the power of Fundamental Intrinsic Harmonic (step frequency) while there was a trend of the KE becoming more symmetric with an increase in the power of the Fundamental Extrinsic Harmonic (stride frequency), see Fig. 2. These results are similar to those of Bennett, et al. (2005) and suggest that the change in recovery is mostly due to the improved phasing as there was little change in the shape of the energy curves.

Despite the small statistical changes in most parameters, the individual changes were often quite large resulting in both improvements and degradation of the subject's gait. Of the 21 subjects 7 (33%) (4 SAFO and 3 HAFO) had changes of less than 10% in the work performed with AFO use. Eight (38%) (3 SAFO and 5 HAFO) of the subjects had more than a 10% decrease in work performed with an average reduction of 31%, an effect size of 1.13, with reductions as large as 46%, an effect size of 1.68. However, there were 6 subjects (29% - 3 SAFO and 3 HAFO) that had an increase in work with an average increase of 24%, an effect size of 0.87, with increases as large as 42%, an effect size of 1.53

These results are similar to those of Brehm et al (2008) who examined changes in walking energy cost with AFO use in 172 children (97 diplegics) with CP. They found no reduction in energy cost in the diplegic or hemiplegic groups. Overall they reported mixed results with a "significant number" of individuals with increased energy costs with AFO use. They also divided their subjects into three groups using a 10% change in metabolic cost to sort the subjects in their study. In their diplegic group they classified 41% as good responders, 25% as non-responders, and 34% as bad responders, very similar to 38%, 29%, and 33% groupings in the current study.

In a study of very high functioning children with posterior leafspring AFOs (hemiplegic and diplegic) Van de Walle, et al. (2009) reported an increase of 30% in mechanical work with

AFO use. Their population of less impaired subjects performed less than 50% of the work to walk than subjects in the present study. They found that with AFOs their subjects had increased CoM vertical excursion combined with a reduced recovery factor.

Limitations of Present Study

The present study had a relatively small sample size. There were several results that might become significant with an increased sample size. A larger sample size might also be helpful in discerning how the different style AFOs affects the CoM mechanics. However, as discussed above, the results of this study, that the use of AFOs has mixed results, are in line with those of Brehm et al. (2008) which had a large number subjects.

Conclusion

The effects of AFOs on the gait on our population resulted in small or no changes in gait as measured by typical metrics. While we found that AFO use promoted a more pendular gait which allowed increased energy recovery, the increased work of longer steps (increased energy variation) meant that the work per meter on the CoM did not change. Finally, we saw that the changes in gait varied greatly between individuals. Our data suggests that current AFOs can improve the gait of an individual patient, but may not be beneficial for the gait of some children with CP. While this finding may not be new, this study describes analysis methods that can lend insight into whether a specific AFO is “improving” the gait of a child.

Additional attention needs to be paid to the gait characteristics of *individuals* and what AFO characteristics would be of benefit. Unfortunately the relationship between gait patterns and AFO and ankle properties has not been established. It is also unclear whether any passive device can provide the assistance needed by most children with CP. Additional research in this area is needed to complement the development of passive and active AFOs to provide larger improvements in walking capabilities.

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References

- Abel MF, Juhl GA, Vaughan CL, Damiano DL. Gait assessment of fixed ankle-foot orthoses in children with spastic diplegia. *Arch Phys Med Rehabil.* 1998; vol. 79(no. 2):126–133. [PubMed: 9473992]
- Balaban B, Yasar E, Dal U, Yazicioglu K, Mohur H, Kalyon TA. The effect of hinged ankle-foot orthosis on gait and energy expenditure in spastic hemiplegic cerebral palsy. *Disability and Rehabilitation.* 2007; vol. 29(no. 2):139–144. [PubMed: 17373095]
- Bennett BC, Abel MF, Wolovick A, Franklin T, Allaire PE, Kerrigan DC. Center of Mass Movement and Energy Transfer During Walking in Children With Cerebral Palsy. *Archives of Physical Medicine & Rehabilitation.* 2005; vol. 86(no. 11):2189–2194. [PubMed: 16271569]
- Brehm MA, Harlaar J, Schwartz M. Effect of ankle-foot orthoses on walking efficiency and gait in children with cerebral palsy. *Journal of Rehabilitation Medicine.* 2008; vol. 40(no. 7):529–534. [PubMed: 18758669]
- Buckon CE, Thomas SS, Jakobson-Huston S, Moor M, Sussman M, Aiona M. Comparison of three ankle-foot orthosis configurations for children with spastic hemiplegia. *Developmental Medicine and Child Neurology.* 2001; vol. 43(no. 6):371–378. [PubMed: 11409825]
- Buckon CE, Thomas SS, Jakobson-Huston S, Moor M, Sussman M, Aiona M. Comparison of three ankle-foot orthosis configurations for children with spastic diplegia. *Dev. Med Child Neurol.* 2004; vol. 46:590–598. [PubMed: 15344518]

- Cappozzo A. Gait Analysis Methodology. Human Movement Science. 1984; vol. 3(no. 1–2):27–50.
- Carlson WE, Vaughan CL, Damiano DL, Abel MF. Orthotic management of gait in spastic diplegia. Am J Phys Med Rehabil. 1997; vol. 76(no. 3):219–225. [PubMed: 9207708]
- Detrembleur C, van den Hecke A, Dierick F. Motion of the body centre of gravity as a summary indicator of the mechanics of human pathological gait. Gait & Posture. 2000; vol. 12(no. 3):243–250. [PubMed: 11154935]
- Duarte, M Relphase. Matlab Software web site. 2002 10-11-2003. Ref Type: Electronic Citation.
- Dursun E, Dursun N, Alica D. Ankle-foot orthoses: effect on gait in children with cerebral palsy. Disability and Rehabilitation. 2002; vol. 24(no. 7):345–347. [PubMed: 12022783]
- Eames MHA, Cosgrove A, Baker R. Comparing methods of estimating the total body centre of mass in three-dimensions in normal and pathological gaits. Human Movement Science. 1999; vol. 18(no. 5):637–646.
- Franceschini M, Massucci M, Ferrari L, Agosti M, Paroli C. Effects of an ankle-foot orthosis on spatiotemporal parameters and energy cost of hemiparetic gait. Clinical Rehabilitation. 2003; vol. 17(no. 4):368–372. [PubMed: 12785244]
- Iida H, Yamamuro T. Kinetic analysis of the center of gravity of the human body in normal and pathological gaits. Journal of Biomechanics. 1987; vol. 20(no. 10):987–995. [PubMed: 3693380]
- Jensen RK. Body Segment Mass, Radius and Radius of Gyration Proportions of Children. Journal of Biomechanics. 1986; vol. 19(no. 5):359–368. [PubMed: 3733761]
- Kerrigan DC, Thirunarayan MA, Sheffler LR, Ribaud TA, Corcoran PJ. A tool to assess biomechanical gait efficiency; a preliminary clinical study. Am J Phys Med Rehabil. 1996; vol. 75(no. 1):3–8. [PubMed: 8645436]
- Knutson LM, Clark DE. Orthotic Devices for Ambulation in Children with Cerebral-Palsy and Myelomeningocele. Physical Therapy. 1991; vol. 71(no. 12):947–960. [PubMed: 1946628]
- Lam WK, Leong JCY, Li YH, Hu Y, Lu WW. Biomechanical and electromyographic evaluation of ankle foot orthosis and dynamic ankle foot orthosis in spastic cerebral palsy. Gait & Posture. 2005; vol. 22(no. 3):189–197. [PubMed: 16214658]
- Maltais D, Bar-Or O, Galea V, Pierrynowski M. Use of orthoses lowers the O(2) cost of walking in children with spastic cerebral palsy. Medicine & Science in Sports & Exercise. 2001; vol. 33(no. 2):320–325. [PubMed: 11224824]
- Mossberg KA, Linton KA, Friske K. Ankle-foot orthoses: effect on energy expenditure of gait in spastic diplegic children. Arch Phys Med Rehabil. 1990; vol. 71(no. 7):490–494. [PubMed: 2350218]
- Olney SJ, Costigan PA, Hedden DM. Mechanical Energy Patterns in Gait of Cerebral-Palsied Children with Hemiplegia. Physical Therapy. 1987; vol. 67(no. 9):1348–1354. [PubMed: 3628488]
- Perry, J. Gait Analysis: normal and pathological function. Thorofare: SLACK, Inc.; 1992.
- Radtka SA, Skinner SR, Dixon DM, Johanson ME. A comparison of gait with solid, dynamic, and no ankle-foot orthoses in children with spastic cerebral palsy. Phys Ther. 1997; vol. 77(no. 4):395–409. [PubMed: 9105342]
- Radtka SA, Skinner SR, Elise Johanson M. A comparison of gait with solid and hinged ankle-foot orthoses in children with spastic diplegic cerebral palsy. Gait & Posture. 2005; vol. 21(no. 3):303–310. [PubMed: 15760746]
- Rethlefsen S, Kay R, Dennis S, Forstein M, Tolo V. The effects of fixed and articulated ankle-foot orthoses on gait patterns in subjects with cerebral palsy. Journal of Pediatric Orthopaedics. 1999; vol. 19(no. 4):470–474. [PubMed: 10412995]
- Rethlefsen S, Dennis SW, Forstein M, Reynolds RAK, Tolo VT, Antonelli D. A comparison of the effects of fixed versus articulated ankle foot orthoses on gait in subjects with cerebral palsy. Gait & Posture. 1995; vol. 3(no. 2):90.
- Romkes J, Brunner R. Comparison of a dynamic and a hinged ankle-foot orthosis by gait analysis in patients with hemiplegic cerebral palsy. Gait & Posture. 2002; vol. 15(no. 1):18–24. [PubMed: 11809577]
- Russell, D.; Gowland, C.; Hardy, S., et al. Gross Motor Function Measure Manual. 2 edn. Hamilton, Ontario: Neurodevelopmental Clinical Research Unit, McMaster University; 1993.

- Smiley SJ, Johnston R, Jacobsen FS, Park C, Mielke C, Ovaska GJ. A comparison of the effects of solid, articulated, and posterior leaf-spring ankle-foot orthoses and shoes alone on gait and energy expenditure in children with spastic diplegic cerebral palsy. *Orthopedics*. 2002; vol. 25(no. 4): 411–415. [PubMed: 12002212]
- Smith PA, Hassani S, Graf A, Flanagan A, Reiners K, Kuo KN, Roh JY, Harris GF. Brace Evaluation in Children with Diplegic Cerebral Palsy with a Jump Gait Pattern. *Journal of Bone and Joint Surgery-American Volume*. 2009; vol. 91A(no. 2):356–365.
- Suzuki N, Shinohara T, Kimizuka M, Yamaguchi K, Mita K. Energy expenditure of diplegic ambulation using flexible plastic ankle foot orthoses. *Bulletin - Hospital for Joint Diseases*. 2000; vol. 59(no. 2):76–80.
- Tesio L, Lanzi D, Detrembleur C. The 3-D motion of the centre of gravity of the human body during level walking. II. Lower limb amputees. *Clinical Biomechanics*. 1998; vol. 13(no. 2):83–90. [PubMed: 11415775]
- Van de Walle P, Hallemans A, Christiaens K, Molenaers G, Truijen S, Duden I, Demey L, Desloovere K. The effect of ankle foot orthoses on mechanical energy in children with cerebral palsy. *Gait & Posture*. 2009; vol. 30(no. Supplement 2):S71–S72.
- White H, Jenkins J, Neace WP, Tylkowski C, Walker J. Clinically prescribed orthoses demonstrate an increase in velocity of gait in children with cerebral palsy: a retrospective study. *Dev Med Child Neurol*. 2002; vol. 44(no. 4):227–232. [PubMed: 11995890]
- Winter, DA. *Biomechanics and motor control of human movement*. N.Y.: Wiley-Interscience; 1990.

Figure 1.

The mean energy curves for BF and AFO conditions of all subjects. The dotted lines are KE and solid lines are PE. Thick lines are the BF condition and thin lines are the AFO condition. AFO use increased the variation of both energies and moved the relative phase closer to the ideal of 180°.

Figure 2. Fourier coefficients of (A) the PE and (B) the KE for BF and AFO conditions. FIH and FEH are indicated. FIH differed between conditions for PE. * $P < 0.001$.

Figure 3.

A histogram of the frequency of Positive, No Change, and Negative effects of AFO use on dependent measures. No Change was defined as any change of less than 10% of the no brace values. Positive and Negative were defined as changes of more than 10% that would lead to a reduction or increase in work, respectively. For the relative phase a change of 10° was used to group the subjects.

Table 1

Summary of experimental results; means and standard deviations.

	No Brace	Brace	P	Effect Size	TDC
Stride Length (m)	0.94(0.15)	1.06(0.17)	<0.001	1.05	1.2*
Velocity (m/s)	1.01(.16)	1.10(.21)	0.016	0.56	1.3*
W_{ext} (J/kg/m)	1.31(0.52)	1.20(0.44)	0.23	0.28	0.54*
Recovery Factor (%)	40.0(16.2)	48.1(15.9)	<0.001	0.92	66*
CoM Vertical Excursion (cm)	3.4(0.9)	4.1(1.2)	<0.007	0.58	--
Normalized CoM Vertical Excursion	0.86(0.24)	0.82(0.23)	0.2	0.27	0.51*
KE Variation (J/kg)	0.16(0.05)	0.22(0.08)	<0.001	0.90	--
PE/KE	2.14(0.48)	1.94(0.70)	0.054	0.46	1.3*
Relative Phase (degrees)	126(34)	141(26)	0.012	0.63	166*

Values measured for typically developing children (TDC) from Bennett, et al. (2005) (*) are provided for reference.