INDIVIDUAL LIMB WORK IS INFLUENCED BY ANKLE-FOOT-ORTHOTICS WORN BY CHILDREN WITH CEREBRAL PALSY

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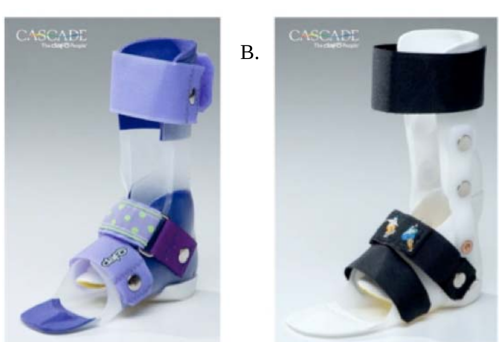
INTRODUCTION

Cerebral palsy (CP) is a neurologic disorder that results from a defect or lesion in the immature brain. Children with CP develop muscular impairments and skeletal deformities of the ankle-foot complex as they mature. Consequences of these impairments may include an equinus foot posture, and inconsistent modulation of the ankle joint throughout the gait cycle. To treat these ankle joint abnormalities, clinicians frequently prescribe an Ankle-Foot-Orthosis (AFO) that prevents excessive plantarflexion during the stance and swing phase (Figure 1). Quite often the prescribed AFO results in positive improvements in gait kinematics during single support, and an improvement in the child's walking efficiency [3].

The external work performed by the limbs during the double support phase of gait has recently received considerable attention because it requires a coordinated effort by the trailing and leading legs to redirect the center of mass [2]. For this redirection to occur properly, the trailing leg must produce a sufficient amount of positive external work to balance the amount of negative external work produced by the leading leg on the center of mass. An imbalance in the amount of work performed by the legs may alter the redirection of the center of mass [2]. Limited efforts have been made to determine the influence of current AFOs designs on the double support phase. It is possible that they may influence the external work performed by the individual legs on the center of mass. The specific aim of this investigation was to evaluate the influence of AFOs on the external work performed by the individual legs of children with CP.

A.



B.

Figure 1. Exemplary solid AFO with a pre-tibial strap (A) and a hinged AFO (B).

METHODS

Eleven children that were diagnosed as having CP with spastic diplegia (Age = 9.1 $\pm$ 2; Mass = 31.6 $\pm$ 13 kg) participated in this investigation. The participants were independent walkers, and had Gross Motor Function Classification levels between 1 and 2 [4]. Three of the children wore solid AFOs with a pre-tibial strap (Figure 1A), and eight wore hinged AFOs (Figure 1B). The hinged AFO design limited ankle plantarflexion, but allowed for full dorsiflexion throughout the gait cycle. The solid AFO limited ankle plantarflexion and dorsiflexion.

Each participant walked at a self-selected pace along a 16-meter walkway while barefoot, and while wearing his or her prescribed AFOs. We collected individual limb ground reaction forces from four AMTI force platforms (120 Hz) that were mounted in series, and evaluated three steps from each participant. The individual limb ground reaction force components were summed to determine the acceleration of the center of mass. Each component of the acceleration was integrated to determine the instantaneous velocity of the center of mass [1]. We calculated the external mechanical work performed by the lead and trail limbs from the dot products of the respective individual limb forces and the velocity of the center of mass [2]. The amount of external work performed by each of the respective limbs was quantified by integrating the power curve, and the work values were normalized by the participant's body mass. We evaluated the amount of positive (W^{+}_{DS}) and negative limb work (W^{-}_{DS}) performed during double support, the difference in the amount of work performed by the individual limbs during double support (W_{DIFF}), and the amount of positive work (W^{+}_{SS}) performed during single support. Repeated measures ANOVA was used to discern differences in the amount of external limb work performed while barefoot and while wearing an AFO.

We additionally used the center of pressure profiles from the respective force platforms to determine changes in the step length and width while barefoot and while wearing AFOs.

RESULTS AND DISCUSSION

The children with CP walked at a significantly faster speed while wearing AFOs (AFO = 1.1$\pm$0.04 m/s; Barefoot = 1.0$\pm$0.04 m/s; p=0.0046), and used a significantly (p=0.004) longer step length (AFO = 53.4 $\pm$ 6 cm; Barefoot= 48.6 $\pm$ 5 cm). Although the mean step width was larger while walking barefoot, there were no significant differences (p = 0.13) between the two conditions (AFO= 11.1 $\pm$ 5 cm; Barefoot= 12.7 $\pm$ 6 cm).

The children with CP performed significantly more negative work by the lead leg during double support ($p < 0.0001$), and significantly less positive work by the trail leg (p = 0.01) while wearing AFOs (Figure 2). The work performed by the individual legs was more balanced while walking barefoot (W_{DIFF} = 0.004 $\pm$ 0.1 J/kg), and had a significantly ($p<0.0001$) more work performed by the lead limb while walking with AFOs (W_{DIFF} = -0.17 $\pm$ 0.1 J/kg). These results imply that the AFO may influence the external limb work that is necessary for the redirection of the center of mass during double support. This notion is further supported by the fact that the children did not have a larger amount of positive work performed by the trail leg while wearing AFOs even though they selected a longer step length.

Significantly (p = 0.02) less positive work was performed on the center of mass during single support while wearing AFOs. We speculate that the reduced amount of work performed during single support may be the reason that children with CP have an improved metabolic cost while wearing AFOs [3]. Taken together, we suspect that the metabolic cost of walking in children with CP may have a large dependence on the amount of limb work performed for the support and control of the body's weight during single support.

CONCLUSIONS

AFOs influence the amount of external limb work performed on the center of mass of children with CP. Although AFOs appear to influence the redirection of the center of mass during double support, they reduce the amount of work performed on the center of mass during single support. The results presented here suggest that new AFO designs are necessary to improve double support phase dynamics. It is possible that the different types of AFOs worn by the children may have had different effects on the amount of external limb work performed on the center of mass. We are currently evaluating if the results presented here extend to the various AFO designs (*i.e.*, hinged, solid, leaf-spring, *etc*.).

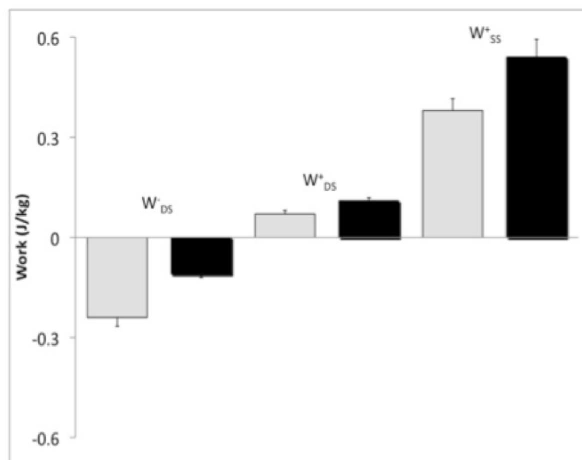


Figure 2. Mean ($\pm$ SEM) external limb work values for double and single support. The black bars are for walking barefoot, and the grey bars are for walking with an AFO.

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