Effect of ankle-foot orthoses on gait in typically developing children: Developmental trend in segmental coordination

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Abstract. Objective: As orthoses, and particularly ankle-foot orthoses, are widely used in the management of children with motor disorders, including cerebral palsy, we aimed to study their effect in normal children in order to add to normative gait data, which are essential for diagnosing, understanding and treating abnormal gait patterns.

Design: We analyzed the effect of ankle-foot orthoses on classical gait parameters and lower limb segments coordination patterns in typically developing children in two age groups reflecting different neuromaturational/developmental situations. We recorded 3D kinematic gait patterns in 9 children (4–5 years) and 11 children (9–10 years) walking barefoot or wearing bilateral solid ankle-foot orthoses maintaining the ankle joint angle at a neutral position.

Results: Ankle-foot orthoses induced little change in cadence, step length, step width or walking velocity in younger children, though they altered intralimb coordination through the gait cycle. In older children, walking velocity was reduced, shank elevation amplitude increased, while lower limb coordination changed less significantly. In this age group, ankle-foot orthoses significantly reduced the variability of coordinative strategies.

Conclusion: Ankle-foot orthoses affect the gait pattern in children with a typical development at different levels in younger and older subjects, but the resulting changes are minimal.

Keywords: Gait, motor control, intersegmental coordination, ankle-foot orthoses, cerebral palsy

1. Introduction

Ankle-foot orthoses (AFO) are commonly used to improve motor functioning of children with cerebral palsy, particularly in walking [1]. The most commonly used AFO in children with bilateral cerebral palsy are solid, posterior orthoses that are rigid at the ankle level, thereby restricting the ankle joint movement [2–6]. In children with cerebral palsy, AFO have been shown to affect a number of gait parameters. Stride length, single foot support time and velocity tend to be increased [2, 4,7]. Less consistent effects have been reported on cadence and other classical kinematic gait parameters [2, 4,8]. A differential effect has also been noted on energy cost associated with walking, with a decrease noted in more severe types of cerebral palsy, contrasting with a lack of change seen in children with hemiplegia or...
diplegia [9]. The effect on lower limb intersegmental coordination, a parameter whose importance has been increasingly recognized in the physiology of gait control and suggested as a target variable in intervention studies in several motor disorders [10–18], has not been evaluated. The coordination of lower limb segments can be studied by principal component analysis [10–14]. In this approach, elevation angles of the thigh, shank and foot relative to the vertical are considered, as their variation has been shown to be much more stereotypical across trials, speeds, and subjects than the corresponding waveform of either the joint angles [10–12] or electromyographic patterns [15]. Moreover, the temporal changes of these elevation angles are tightly coupled together [10]. When the elevation angles of the thigh, shank and foot are plotted one versus the others, they describe a regular gait loop, which lies close to a plane. This plane orientation and the shape of the loop reflect the phase relationship between the different segments and therefore the timing of intersegmental coordination [15,16], on which postural stability with respect to gravity and dynamic equilibrium for forward progression depend. The plane orientation shifts in a predictable way with increasing speed of walking [16]. Plane orientation shifts reliably correlate with the mechanical energy expenditure [16]. The covariation of lower limb segments can be regarded as an important physiological parameter reflecting neural control of walking [10,11,13]. It has been used to study the emergence of functional gait strategies in children from the very first walking steps on [13,17]. It has also been used to document gait patterns in various motor disorders and assess the effect of treatment [12,14,18].

In contrast to populations with cerebral palsy, the effect of AFO on the gait of typically developing children has hardly been reported. Here we investigated whether wearing solid posterior AFO has a short-term effect on the gait of normal children, with special attention on classical gait parameters and planar covariation of the elevation angles of lower limb segments, in two age groups reflecting different neuromaturational/developmental situations.

### 2. Materials and methods

#### 2.1. Subjects

Twenty healthy children with typical neurological development participated in the study. One group consisted of 9 children (1 girl, 8 boys) aged 4 or 5 years; the other group consisted of 11 children (4 girls, 7 boys) aged 9 or 10 years. Inclusion criteria were as follows: age (4,5,9 or 10 years), full-term delivery (37–42 weeks’ gestation), good health, typical neurological development, reported onset of independent walking between 10 and 18 months, no history of orthopedic surgery and absence of significant asymmetry in lower limb length (< 15 mm). The anthropometric data are shown in Tables 1 and 2.

#### 2.2. Task

Children were instructed to walk as naturally as possible over an 8-meter walkway. Eight trials of barefoot walking were followed by 8 trials of walking with bilateral solid AFO inserted in the child’s own shoes. The subjects were wearing tight pants or light cloth that would not cover the markers or interfere with the movement. Both AFO (provided by Orthodiffusion, Brussels, Belgium) were fabricated from 4.8 mm-thick polypropylene extending distally under the toes and on the mediolateral border of the foot, and proximally on
the posterior aspect of the leg to about 2.5–5.0 cm below the knee with trim lines anterior to both malleoli and straps across the front of the ankle and anterior upper tibia. The AFO maintained the ankle at 0° of dorsiflexion and prevented plantar flexion. Three pairs of AFO of different sizes were available. All children could fit comfortably in one of these. The children were given 15 minutes to adapt to the AFO, during which they could be freely active around the laboratory. We noted, however, that most children adapted very quickly to the new condition (around 5 minutes).

2.3. Gait recording

The task was recorded using the optoelectronic ELITE system (BTS, Milan, Italy) following a standard protocol [12]. This system consists of six charge-coupled device cameras detecting retro-reflective markers using a sampling rate of 100 Hz. The cameras were placed 1.5 m above the floor, and 3 m apart. After calibration, two-dimensional data were corrected for optical distortion and converted to 3D coordinates according to Borghese et al. [10].

The 22 markers were passive retroreflective pedunculated spheres (15 mm in diameter). They were placed on the subjects’ skin overlying the following anatomical landmarks: C7, the acromion process, sacral marker at the midpoint on the line between both posterior superior iliac spines, both anterior superior iliac spines, greater trochanter bilaterally, thigh markers at the mid-point on the line between the greater trochanter and the lateral head of the femur, knee markers at the lateral head of the femur and the fibula, bilateral shank markers at the mid-point on the line between the lateral head of the fibula and the lateral malleolus, ankle markers at the lateral malleolus, bilateral foot markers at the lateral malleolus, bilateral foot markers at the second metatarsal head and the heel.

The position in space of the 22 markers was recorded and processed for real time recognition. Temporospatial measures were computed for each of the eight trials using the Eliclinic software (BTS, Milan, Italy). For each child, cadence, velocity, step length and step width were calculated. The elevation angles of the thigh, shank, and foot in the sagittal plane are noted as, and respectively (Fig. 1B). Thigh elevation angle (\(\alpha_t\)) was calculated as the angle between the vertical and the trochanter-knee marker segment, shank elevation angle (\(\alpha_s\)) as the angle between the vertical and the knee-malleolus marker segment and foot elevation angle (\(\alpha_f\)) as that between the vertical and the malleolus-metatarsal marker segment. When the elevation angles are plotted one versus the others, they describe regular loops constrained close to a plane, which represents an attractor plane common to both the stance and swing phase of the gait cycle. In this graph, paths progress in time in the counter-clockwise direction, heel contact and toe-off phases corresponding roughly to the top and bottom of the loops, respectively. The specific orientation of the plane of angular covariation reflects the phase relationships between the elevation angles of the lower limb segments and therefore the timing of the intersegmental coordination. Because the degrees of freedom of limb angular motion in the sagittal plane are reduced to two by the planar constraint, they match the corresponding degrees of freedom of linear motion of the center of body mass. In other words, the coupling of the angular motion at the different limb segments dictates the spatiotemporal trajectory of the center of body mass.

Mathematical and statistical analysis was performed using the Statistica Software (StatSoft, Tulsa, OK, USA).

The method for analyzing the planar covariation of elevation angles consisted of principal component analysis and was the same as previously described in adult subjects [10] and children [13].

Two sample t-test was used to evaluate the differences in means between two groups (dependent and independent) with \(p < 0.01\). For comparisons between the two groups we used the independent samples t-test and the dependent samples t-test for comparisons within the group.

The study protocol was approved by the institutional ethics committee and informed consent obtained from each subject and legal guardian.

3. Results

3.1. Standard gait parameters

Standard gait parameters are shown in Table 3. Whether with or without AFO, the younger children had significantly smaller step length, step width and velocity than the older children, but cadence did not differ significantly between the two groups. No significant differences were recorded in the group of younger children between barefoot walking and walking with AFO. In the older children, a significant decrease in velocity was noted with AFO, but this was not associated with a significant decrease in either step length or cadence, although the cadence tended to decrease.
Table 3  
Gait kinematic parameters

<table>
<thead>
<tr>
<th></th>
<th>4–5 years</th>
<th>9–10 years</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>barefoot</td>
<td>AFO</td>
</tr>
<tr>
<td>Step length (m)</td>
<td>0.82 ± 0.11*</td>
<td>0.84 ± 0.14</td>
</tr>
<tr>
<td>Step width (m)</td>
<td>0.19 ± 0.03*</td>
<td>0.21 ± 0.03</td>
</tr>
<tr>
<td>Walking velocity (m/s)</td>
<td>0.88 ± 0.13*</td>
<td>0.87 ± 0.14</td>
</tr>
<tr>
<td>Cadence (stride/min)</td>
<td>72.9 ± 2.75</td>
<td>74.9 ± 7.7</td>
</tr>
<tr>
<td>Elevation angles thigh</td>
<td>41.57 ± 8.27</td>
<td>43.67 ± 5.17</td>
</tr>
<tr>
<td>Elevation angles shank</td>
<td>69.71 ± 9.93</td>
<td>72.37 ± 9.42</td>
</tr>
<tr>
<td>Elevation angles foot</td>
<td>78.82 ± 11.52</td>
<td>73.75 ± 10.21</td>
</tr>
<tr>
<td>Eigenvalue 3</td>
<td>0.77 ± 0.16</td>
<td>0.71 ± 0.37</td>
</tr>
</tbody>
</table>

* significantly different from the barefoot value between the two age groups (P < 0.01).
** significantly different between barefoot and AFO within each age group (P < 0.01).

3.2. Intersegmental coordination

Maximal $\alpha_t$, $\alpha_s$ and $\alpha_f$ values are shown in Table 3. There were no significant differences in the amplitude of elevation angles between the age groups. In younger children, there were no differences in maximal elevation angles of the lower limb segments in gait with or without AFO. In older children the only significant change noted was an increase in $\alpha_s$ with AFO (Table 3, Fig. 2A,B). Principal component analysis showed a slight decrease in the third eigenvalue in both groups walking with AFO as compared to barefoot (Table 3).

We calculated the angle between the mean orientation of the covariation plane of a normative group of adults (Borghese et al. 1996) and the orientation of the children’s best fitting planes in the barefoot situation and wearing AFO. The orientation of the children’s covariation planes only deviated from the adult’s planes
Fig. 2. Kinograms (stick diagrams) representing sagittal projection of locomotion. In these diagrams, two consecutive segments represent 20 msec. (A) Barefoot condition, 10 year-old child. The arrow points to normal elevation of the thigh. (B) Ankle-foot orthoses condition, 10 year-old child. The arrow points to increased elevation of the thigh. (C) Barefoot condition, 5 year-old child. (D) Ankle-foot orthoses condition, 5 year-old child.

by 6.13 ± 4.01°, with a maximum deviation of 16.39°. There was no significant difference in these angles for different age groups or situations.

We computed the orientation of the mean normal vector in each of the 4 groups, namely: 4–5 years barefoot, 4–5 years AFO, 9–10 years barefoot, and 9–10 years AFO. The angle between the third eigenvector of the covariance plane of each child and this mean vector allowed us to assess within-group variability of coordination strategies. The means and standard deviations of this angle did not differ significantly between the groups, except in the older children wearing AFO (barefoot: 5.8 ± 2.61°, AFO: 3.42 ± 1.42°, p = 0.02).

Finally we calculated for each child the angle σ between the third eigenvector of the covariance plane respectively while walking barefoot and while wearing AFO. The angle σ therefore represents a measure of the difference in lower limb intersegmental coordination with or without AFO. The mean angle σ significantly differed between age groups. In the older children (4.30 ± 3.0°, n = 11), the angle was lower than in the younger children (9.70 ± 5.25°, n = 10), (p = 0.008). This means that the orientation of the plane changed more if we fit AFO to young children, than if we fit them to older children.

4. Discussion

The purpose of the present study was to determine if wearing AFO alters spatiotemporal parameters and intersegmental coordination in typically developing children. We found that wearing AFO induced little change in standard gait parameters in younger children, though it altered the organization of gait; changes were more pronounced in older children. The observed changes were related to age: in 4–5 year-old children the amplitude of elevation angles, orientation of the plane and spatiotemporal parameters remain unaltered; in 9–10 year-old children walking velocity was reduced, shank amplitude increased, while the plane orientation changed significantly less than in younger children. In this age group, AFO significantly reduced the variability of coordinative strategies.

Previous studies reported greater inter-subject and intra-subject variability of gait in children of all ages than of adults. They also showed that measurements of stride-to-stride variability were significantly greater in younger children compared to older ones. These effects persisted after adjusting for height [19–22].

4.1. Spatiotemporal parameters

Our study groups show similar parameters to previous reports. Previous studies [23,24] demonstrated some age-related differences in spatiotemporal parameters, attributed to differences in body size, proportion and walking strategies. Cadence decreases gradually while step length and velocity increase with increasing age. We found a tendency toward decreasing walking cadence with AFO, but this was not significant. This may be due to small sample size. Moreover, variability in spatiotemporal parameters has been shown to be influenced by age between 4 and 8 years, and by
speed [19]. The only significant difference we found between barefoot walking and walking with AFO concerned velocity. In contrast with the reduction in velocity that we found, studies of children with impaired motor control in the context of cerebral palsy found either no change in velocity in all AFO configurations [8] or a significant increase in walking velocity and stride length with AFO [25,26]. In the AFO condition, the children used their own shoes in addition to the AFO; this may have affected the results. A study of the effect of shoes on gait showed that minimal changes resulted in statistically significant differences, due to very tight standard deviations of the data, yet these changes did not appear to be clinically significant [27]. The authors concluded that for most of the clinical studies, it is not necessary to perform assessment with shoes [27].

4.2. Intersegmental coordination

A number of previous studies have demonstrated the physiological relevance of covariation of elevation angles αt, αs and αf in intersegmental coordination in gait. Planar covariation of these angles emerges very early after walking onset in toddlers [13] and holds throughout childhood and adulthood for walking at different speeds [16], forward or backward directions [15], erect or bent posture [12], and different levels of body weight unloading [20]. Our study confirms this coordination rule in the barefoot condition in both age groups. This may be partly explained by limitation of the ankle joint movement by the AFO [28], though other factors than a close foot-shank correlation are likely to be involved, as a reduction of degrees of freedom from three to two would result in a plane orientation parallel to the thigh axis (u3T would be 0) [29], which was not the case in our data, even in the AFO condition.

The global orientation of the covariation plane in 3D-position space changes in different conditions, such as running, hopping or kicking [22,30]. In our results, the mean plane orientation for all conditions was similar to the mean orientation of normal walking in children (beyond the toddler age) and adults [13].

As regards variability of plane orientation, we did not find any differences between younger and older children. This contrasts with the greater variability of spatiotemporal parameters, vertical and anterior-posterior parameters in younger children [19–21,24]. Only when comparing the variability in the older children in the barefoot and AFO conditions, did we find a significant reduction of variability with AFO. AFO thus seems to reduce the choice of coordinative strategies compared to the barefoot situation, but only (significantly) in older children. This may relate to optimized management of energy expenditure, given the known correlation between kinematics represented by plane orientation and mechanical energy output [16]. This suggests that the specific tuning of the law of planar covariation can be used by the nervous system for limiting energy expenditure, an issue that has been studied in reference to AFO in several cerebral palsy types [9]. We hypothesize that older children are more experienced in finding a strategy that limits energy expenditure, which means that in case of an external perturbation like walking with an AFO, they tend towards a more stereotyped solution than young children.

Angle σ, computed between the barefoot and AFO condition for each child, was significantly greater in the younger group compared to the older group. This is another illustration of the difference in dealing with the constraint exerted by the AFO according to age: whereas 4–5 year-old children maintained their velocity but altered covariation plane orientation, 9–10 year-old children walked slower but did not significantly changed the pattern of intersegmental coordination and showed less inter-subject variability in plane orientation. We hypothesize that older children have already achieved a much more robust, efficient coordination, which is not easily perturbed.

AFO not only restrict the ankle joint movement but also provide different sensory information. Sensory information of the movement is an important issue in adapting the movement to the environment [31]. Studies of the development of locomotor balance control in healthy children demonstrated two functional principles for the sensorimotor organization of balance control in humans [31]. Younger children tend to use an ‘en bloc’ mode and older children an ‘articulated’ mode. Generally speaking, many sensorimotor activities change at the age of 6–7 years. This may underlie the differences in adaptations and strategies between the groups [32].

Among the limitations of this study, the small numbers in each group, the lack of recording the subject walking with shoes (without AFO) and of direct energy cost measurement must be noted.

5. Conclusion

We conclude that wearing AFO affects the gait pattern in children with a typical development at different levels in younger and older subjects, but the resulting
changes are minimal. This observation probably points to a high potential for functional reorganization, which children in whom AFO are prescribed generally lack. However, it may appear as an important prerequisite for AFO use for potential improvement of gait in individuals with limited possibilities for developing alternative strategies in face of biomechanical constraints such as those associated with AFO. Our results suggest that younger children change their intersegmental coordination at the lower legs but do not adapt step length, cadence and walking velocity. In contrast, older children maintain similar lower-limb intersegmental coordination pattern to barefoot condition, but reduce walking velocity. This differential effect may reflect different strategies at play. Younger children may take advantage of the distal stabilization provided by the AFO [32], whereas older children favor minimization of the risk of falling over, by reducing walking velocity [33]. Further investigation in large size groups is needed to confirm these results.

Conflicts of interest

None reported.

References


