Changes in muscle activity in children with hemiplegic cerebral palsy while walking with and without ankle–foot orthoses

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Abstract

We compared the electromyographic (EMG) signals of lower extremity muscle groups in 10 children with hemiplegic cerebral palsy (CP) while walking barefoot and in a hinged ankle–foot orthosis (HAFO). All children had excessive plantarflexion and initial toe-contact on the affected side when walking barefoot, a typical gait pattern for hemiplegic patients. The patients walked with a physiological heel–toe gait pattern when wearing the HAFO. The peak activity of the tibialis anterior muscle was reduced by 36.1% at initial contact and loading response phase and by 57.3% just after toe-off when using a HAFO. The decrease in activity was thought to result from the change in gait pattern from a toe-gait to a heel–toe gait as well as the use of a HAFO. The HAFO also slightly decreased muscle activity in the proximal leg muscles mainly during swing phase, improved stride length, decreased cadence, improved walking speed, increased peak hip flexion, improved kinematics in loading response phase at the knee, and reduced the excessive ankle plantarflexion.

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1. Introduction

A typical finding in the gait pattern of patients with hemiplegic cerebral palsy (CP) is toe-walking on the affected side. Previous studies involving three-dimensional gait analysis [1–3] showed that an ankle–foot orthosis (AFO) could successfully correct the excessive plantarflexion angle and thereby improve the children’s gait pattern. Data on changes in muscular function, apart from changes in joint angles, joint forces, and moments or energy consumption are required to investigate the effects of AF0s on gait.

A widely recognized classification system for hemiplegia based on joint kinematic data was proposed by Winters et al. [4,5] and classifies the patterns into four different types. A principle finding in type I (the mildest form) is equinus in swing secondary to a relative over activity of the triceps surae compared to the anterior tibial musculature. The main feature of type II is ankle plantarflexion throughout the gait cycle, in both swing and stance from either static or dynamic contracture of the triceps surae. The knee is often forced into slight hyperextension in middle or late stance. Patients with type III show, in addition to the gait anomalies of type II, limited knee motion in the swing phase of gait. Patients can either show reduced knee extension during stance or hyperextension in stance and reduced flexion in swing. The patients in group IV show all features of type III with involvement of the hip musculature.

Equinus may be present in swing or stance phase. Anterior tibialis weakness, dysfunction, or premature gastrocnemius contraction causes swing phase drop foot with resultant dragging of the foot, tripping, and proximal joint kinematic compensations [6]. Compensatory mechanisms, however, can cause secondary problems and alter the gait pattern further.

Children with hemiplegia can benefit from the use of an AFO to improve their gait pattern. If an orthosis has been correctly designed, constructed and applied to the patient, it may act to resist the abnormal pattern of motion and hence...
restore more normal function [7]. The use of AFOs has remained a standard treatment because of their usefulness in preventing equinus contractures [8]. Type I hemiplegic patients require an AFO primarily to assist the dorsiflexors in preventing foot drop during swing. To improve gait, the AFO aims at correcting the foot-shank angle in swing phase to improve pre-positioning of the foot at initial contact and allow heel-strike [3]. However, those with type II hemiplegia require an AFO capable of controlling the equinus through stance and swing. The most common orthotic solution is the solid AFO. The effect of the design can be optimized by the addition of ankle articulations permitting free dorsiflexion, i.e. a hinged AFO (HAFO), and by the addition of some shoe adaptations [2,7].

Dynamic electromyography (EMG) provides valuable information with respect to muscle activity. Surface electrodes applied to the skin overlying the muscles provide an indirect measure of muscle-generated potentials. Errors in neurological control, muscle weakness, voluntary substitutions and obligatory posturing lead to abnormal EMG during walking [1]. Investigation of EMG data on the improvement of gait function using AFOs has been limited. Radtka et al. [9] compared the effects of two types of AFOs on the gait of 10 children with CP four of whom were hemiplegic. The orthoses increased stride length, decreased cadence, and decreased excessive ankle plantarflexion. No changes in lower extremity muscle timing were found compared to barefoot walking. The study of Radtka et al. [9] looked at the timing of muscle activity and did not investigate change in level of activity. Muscle recruitment of spastic muscles in children with CP is characterized by disordered onsets and offsets of muscle contraction. Loss of selective control may result in abnormal co-activation of agonist–antagonist muscles and the presence of co-contracture due to poor selective control may be the reason for the disordered timing of muscle activity [7,10,11]. Knowledge of the change in activation of different muscle groups should help to understand further the function of AFOs.

The purpose of this study was to detect changes in EMG activity of the tibialis anterior muscle in patients with hemiplegic CP when walking with and without a HAFO. We wished to determine if the pattern of muscle activation changed when the movement pattern at the ankle joint level was changed by a HAFO. We also considered the influence of a HAFO on selected upper leg muscles.

2. Methods

2.1. Patients

The study group consisted of 10 children (four girls and six boys) with hemiplegic CP (six right sided, four left sided, mean age 9.7 ± 1.6 years, weight 30.2 ± 6.2 kg, and height 135.2 ± 7.9 cm). None had had orthopaedic surgery. The
the ankle. The orthosis blocked ankle plantarflexion, but allowed at least 20° dorsiflexion through a hinge. Generally, the HAFOs were fitted by the orthotist so that the tibial axis was positioned perpendicular to the floor surface to avoid a forward or backward lean. A control group of 10 healthy children (six girls, four boys mean age 8.7 ± 0.8 years, mean height 133.4 ± 7.7 cm, and mean weight 28.4 ± 6.6 kg) provided comparative kinematic data. The patients and controls did not differ statistically in age, height and weight. The normal reference data were collected during barefoot walking.

2.2. Study protocol

The study protocol consisted of 3D-gait analysis and lower extremity muscle activity measured under two conditions, barefoot and wearing a HAFO with shoewear. All subjects walked at a self-selected walking speed. Testing continued until a minimum of six trials with clear data sets had been collected under each test condition.

2.3. Gait analysis

3D-gait analysis was performed using a six-camera, 50 Hz movement analysis system (VICON 370, Oxford Metrics Ltd., UK). Fifteen 25 mm diameter retro-reflective markers were affixed to the subjects according to the marker protocol of Davis et al. [13] (Fig. 2). When walking with the HAFO and shoes, the heel and toe markers were placed on the shoes at the positions best projecting the anatomical landmarks. All other markers remained at the same positions throughout the testing protocol. Height, weight, leg length, widths of the ankles and knees, and tibial torsion were measured clinically for appropriate anthropometric scaling. Joint angle data were expressed as percentage of gait cycle using Polygon software (Oxford Metrics Ltd.). Specific points and values selected from the movement curves were compared between conditions.

2.4. EMG measurements

Surface EMG was recorded simultaneously with the 3D-gait analysis. Bipolar Ag/AgCl surface electrode pairs with an electrode diameter of 10 mm and an inter-electrode spacing of 22 ± 1 mm were placed on the clean shaven skin overlying the vastus medialis, vastus lateralis, long head of the biceps femoris, semitendinosus, and tibialis anterior muscles of the hemiparetic leg. The SENIAM [14] guidelines for electrode placements in surface EMG were followed. The reference ground electrode was placed overlying the tuberosity of the tibia. All electrodes were left in position between conditions and were connected to single differential amplifiers with a band path of 10–700 Hz (Biovision AG, Wehrheim, Germany). The data were collected at a sampling frequency of 2500 Hz using a Zebris system (Zebris, Tübingen, Germany). EMG signals were visually inspected for artefacts and noise. Initial foot contact and toe-off were determined using a force plate (Kistler Instrumente AG, Winterthur, Switzerland) embedded in the floor. The following ipsilateral initial foot contact was determined with a small 1-directional accelerometer (Biovision, Wehrheim, Germany) attached to the heel placed vertically. Force plate and accelerometer data were sampled at the same frequency as the EMG data. Determining the second foot strike for EMG analysis with an accelerometer proved to be more precise than using the 50 Hz kinematic data and equally good as using a force plate.

The electromyographic data of six trials were first re-sampled according to the trial with the shortest data length, then full wave rectified for each patient. The data were then time normalised by dividing data for 1 gait cycle into 16 equally spaced intervals (∆1–∆16). Root mean square values
for each muscle signal were calculated for each of these time intervals. The maximum value of the EMG was calculated for all trials of each patient when walking barefoot and all other EMG data were expressed as percentage of the average maximum value over the six trials. The MATLAB software package (The MathWorks Inc.) was used.

2.5. Data analysis

Data of six trials under each condition were averaged for each patient. Paired t-tests were applied to the kinematic parameters and EMG data of the impaired body side to determine changes between the two conditions during walking barefoot and with a HAFO. The level of significance was set at $p < 0.05$.

3. Results

3.1. Temporal-spatial parameters

Table 1 shows the mean values of the temporal-spatial outcome measures for the patients walking barefoot and walking with a HAFO. With a HAFO, patients walked significantly faster by increasing their step and stride length. Cadence was decreased. Stance phase duration, single support time and double support time did not show any significant alterations.

3.2. Joint angles

Fig. 3 shows the mean sagittal plane kinematic data for walking with and without a hinged AFO compared to data from healthy children. All patients had an initial toe contact when walking barefoot due to excessive ankle plantarflexion ($15.0 \pm 9.7^\circ$ at initial contact) followed by a movement in the direction of dorsiflexion. When wearing a HAFO, this angle was within the normal range ($-0.1 \pm 6.3^\circ$) and a physiological heel-strike was followed by a movement in the direction of plantarflexion. Patients showed a significantly increased knee flexion position ($22.8 \pm 8.9^\circ$) at initial foot contact compared to the control group ($10.0 \pm 3.3^\circ$) that did

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Table 1

<table>
<thead>
<tr>
<th>Temporal-spatial parameters</th>
<th>Patients barefoot</th>
<th>Patients hinged AFO</th>
<th>p-Value $^*$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Walking speed (m/s)</td>
<td>1.05 (0.18)</td>
<td>1.14 (0.14)</td>
<td>0.018</td>
</tr>
<tr>
<td>Cadence (steps/min)</td>
<td>119.4 (7.5)</td>
<td>112.5 (7.3)</td>
<td>0.006</td>
</tr>
<tr>
<td>Stride length (m)</td>
<td>1.06 (0.15)</td>
<td>1.22 (0.11)</td>
<td>&lt;0.001</td>
</tr>
<tr>
<td>Step length (m)</td>
<td>0.52 (0.07)</td>
<td>0.61 (0.06)</td>
<td>&lt;0.001</td>
</tr>
<tr>
<td>Stance (%)</td>
<td>57.5 (2.2)</td>
<td>57.6 (1.9)</td>
<td>N.S.</td>
</tr>
<tr>
<td>Single limb support (%)</td>
<td>38.8 (3.0)</td>
<td>38.4 (4.2)</td>
<td>N.S.</td>
</tr>
<tr>
<td>Double limb support (%)</td>
<td>17.7 (2.8)</td>
<td>18.2 (3.1)</td>
<td>N.S.</td>
</tr>
</tbody>
</table>

$^*$ Paired student t-test with $p < 0.05$; N.S. = not significant; values are for the hemiparetic leg.
not alter with a HAFO (21.0 ± 7.3°). Although the increased angle at initial contact could not be reduced with the HAFO, the loading response phase improved by becoming more dynamic and a flexion-extension pattern could be observed. A significant increase in hip flexion at initial contact was seen in the patient group when wearing a HAFO (barefoot: 44.8 ± 6.7°; HAFO: 48.1 ± 5.4°). However, both values were not significantly different from the control group (44.5 ± 4.5°). There were no significant differences at the level of the pelvis or in other selected kinematic parameters.

3.3. EMG data

Data of the muscle activities are shown in Fig. 4. Most pronounced changes were noted in the tibialis anterior muscle. Compared to barefoot walking, the hemiplegic children showed an average decrease in peak tibialis anterior muscle activity of 36.1 ± 20.6% at the first 12.5% (average of Δ1 and Δ2) of the gait cycle, i.e. initial contact and loading response phase. Individual EMG profiles of the tibialis anterior muscle for all patients are shown in Fig. 5. Nine out of the 10 patients showed a reduction of activity varying between 13.6% and 57.5% whereas one subject stayed equal (reduction of 0.7%). Peak swing phase activity showed, on average, a reduction of 57.3 ± 20.2% when walking with the HAFO. In all the patients activity was reduced varying from 30.1% to 87.5%. Slight significant changes during swing phase could be detected in the semitendinosus, biceps femoris, vastus medialis and lateralis muscles. On all occasions the average activity over the
whole group of patients was reduced when walking with an HAFO compared to barefoot walking. Results were consistent within the group. A reduction of EMG was also found for the rectus femoris during swing phase.

4. Discussion

The main aim of this study was to investigate tibialis anterior muscle activity when patients with hemiplegic CP walked with and without a HAFO. Results showed that a change of movement pattern (toe-gait to heel–toe gait) when wearing a HAFO resulted in a decrease of peak activity of the tibialis anterior muscle at initial stance, loading response, and initial swing phase.

In healthy persons, the tibialis anterior muscle shows a large burst of EMG activity from initial contact to the end of loading response phase which controls foot placement in gait. At initial contact, the ankle is in a neutral position and the forefoot is tilted upward. The tibialis anterior provides support of the foot by a dorsiflexion pull [15]. During loading response the tibialis anterior, together with the long toe extensors, decelerates the rate of ankle plantar flexion as the foot drops to the floor [15]. A second burst of activity is seen starting in pre-swing lasting for the whole swing phase. In pre-swing the ankle rapidly plantar flexes and the onset of tibialis anterior helps to decelerate the rate of foot fall [15]. Peak activity is reached in initial swing phase just after toe-off. Ankle motion is now changed to dorsiflexion towards neutral position to allow floor clearance by the toes. The children with hemiplegic CP walking barefoot with an initial toe-contact on the affected side, showed, similar to healthy persons, a first burst of electromyographic activity at initial foot contact and loading response and a second burst with peak activity after toe-off during swing phase. However, in the hemiplegic CP patients this first burst of activity at initial stance phase was lower compared to the second burst. While making initial contact with the toes instead of the heel, a greater internal plantarflexor moment is generated around the ankle joint [3] and the role of tibialis anterior activity in this phase can be associated with stabilisation and control of the ankle position. The second burst of activity shows normal function at pre-swing but activity diminishes in mid-swing in contrast to healthy subjects, in which this second burst is followed by further EMG activity to secure foot clearance during the rest of swing phase. Thus, no activity of the tibialis anterior muscle is present during the end of swing phase. With a HAFO, the children achieved a physiological heel–toe gait pattern. Peak tibialis anterior muscle activity decreased by 36.1% at initial foot contact and by 57.3% just after toe-off. The HAFO controls the foot position relative to the shank and prevents a foot-drop and the dorsiflexors are no longer required as their function has been replaced by the orthosis. This is in agreement with a study by Geboers et al. [16] who presented data of healthy adult subjects wearing a stiff AFO when walking on a treadmill. They reported that the first burst of tibialis anterior muscle activity, i.e. first 15% of the gait cycle, decreased by 20% when walking with an AFO. Activity of the tibialis anterior muscle did not change during the gait cycle as a whole. However, in this study [16] the healthy subjects walked in a heel–toe gait pattern both with and without an AFO and the AFO did not have a hinge to allow ankle dorsiflexion. This study [16] showed that the use of an AFO immediately reduced muscle activity of the ankle dorsiflexors in healthy people. Patients in the present study and normal individuals seem to reduce muscle activity when this is no longer necessary.

If healthy individuals adopt a different gait pattern by voluntary toe-walking, EMG will change accordingly. Davids et al. [17] compared kinematic, kinetic, and EMG data of healthy children during voluntarily toe-walking to children with CP walking in an obligatory toe-walking gait pattern. Significant deviations common to both normal and CP toe-walking groups were determined to be due, at least in part, to the biomechanical constraints associated with a toe-walking gait pattern. The reduction of tibialis anterior muscle activity in this study could be partly related to the change of movement pattern, i.e. toe-walking to heel–toe

Fig. 5. Individual EMG profiles of the tibialis anterior muscle of 10 patients walking with and without a hinged ankle–foot orthosis (AFO). Each trace is an average of six trials for one patient.
gait pattern, and partly to the use of a HAFO in that the HAFO prevents plantarflexion and hence reduces the need to dorsiflex the foot.

Orthoses may be prescribed to prevent a deformity that is anticipated or might increase without treatment [8] and to allow improved gait function. All children in this study were experienced users, having worn a HAFO for at least several years. None had fixed contractures of the Achilles tendon and calf muscles which might result from wearing a HAFO. While the aims of an AFO in children with hemiplegic CP are well defined, their effectiveness in the long term is less certain. It is possible that a decrease in muscle activity of the dorsiflexors with the prolonged use of an AFO could lead to loss of strength and have a negative effect. In this group of hemiplegic children, the benefits of achieving: (1) a more normal gait pattern (i.e. increased walking speed and step and stride length and improved kinematics) and (2) preventing secondary compensation mechanisms and equinus contracture, were more important than a potential loss of muscle strength for their tibialis anterior muscle.

In the present study, kinematic changes observed at the ankle joint were as expected and in agreement with a previous study comparing the same type of orthosis in a similar group of patients [3]. Other sagittal plane kinematic data showed a significant increase in hip flexion at initial contact when walking with a HAFO. Although within group values were significant, these were not significantly different to normal values. At the level of the knee, loading response phase improved with a HAFO and showed a more dynamic movement pattern. These findings can be the result of increased walking speed as reported by Van der Linden et al. [18]. With increasing walking speed, peak hip flexion and extension increased over the gait cycle in a group of healthy children. In the present study, a higher walking speed was observed when walking with a HAFO due to larger steps, which can explain the increased peak hip flexion in early stance. Changes in knee dynamics in loading response can also be influenced by the increased walking speed [18]. Although small significant changes were seen as a reduction of activity in upper leg muscle groups, these did not translate into relevant functional effects.

A limitation to this study is that the three markers representing the foot had to be placed on the shoes but these were placed as correctly as possible to represent the long axis of the foot. The associated measurement errors are likely to be greater compared to those from skin markers placement. EMG data of the gastrocnemius and soleus muscles was not obtained and might have given some further insights.

5. Conclusion

This study showed that the use of a HAFO reduced activity of the tibialis anterior muscle during walking at initial contact, loading response, pre-swing and initial swing phase. Wearing the HAFO also improved stride length, decreased cadence, improved walking speed, increased hip flexion at initial contact, improved knee kinematics at loading response phase, and reduced the excessive ankle plantarflexion at initial contact and during swing phase. The decrease in activity of the tibialis anterior muscle is thought to result from the change in gait pattern from a toe-gait to a heel–toe gait pattern as well as the use of a HAFO itself which substitutes for the tibialis anterior muscle.

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References

