

Effect of foot orthoses on the standing balance of a child with spastic diplegic cerebral palsy: A case report

Pornsuree Onmanee¹, Yoshihiro Ehara², Sumiko Yamamoto³, Piyavit Sorachaimetha¹, Kazuhiro Sakai²

¹Sirindhorn National Medical Rehabilitation Centre, Muang Nonthaburi, Thailand

²Department of Prosthetics & Orthotics and Assistive Technology, Niigata University of Health and Welfare, Niigata, Japan

³International University of Health and Welfare Graduate School, Tokyo, Japan

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Abstract

The purpose of this study was to evaluate the effect of foot orthoses on the standing balance of a child with spastic diplegic cerebral palsy (CP). Subject: A 12-year-old girl was diagnosed with spastic diplegic CP with level II of the gross motor function using the gross motor function classification system. Her feet were pronated while walking and crouching. Foot orthoses were prescribed to correct the foot alignment. Standing balance was evaluated in a three-dimensional motion analysis lab at Sirindhorn National Medical Rehabilitation Centre (SNMRC). The movement of the centres of pressure (CoP) of left foot, right foot and whole body, and the centre of mass (CoM) were investigated by using a dual forceplate approach. The subject was instructed to stand barefoot and with the foot orthoses, with her eyes open, for 30 s. This procedure was repeated until there were 3 successful trials for each standing condition. The patient had a crouching posture throughout the trials, irrespective of the standing condition. Moments at the hips, the knees, and the ankles, observed during two conditions were similar; the range of the internal rotation of the hip alone was slightly smaller when standing with the foot orthoses than when standing barefoot. The magnitude of vertical component of the ground reaction force

on the left foot was greater than that on the right foot. The anteroposterior sways of both limbs were smaller when standing with the orthoses than when standing barefoot; however, the mediolateral sways were larger when standing with the orthoses than when standing barefoot and tended toward the left foot. In addition, the 95% confidence ellipse areas of CoP and CoM when standing with the insoles were smaller than those observed when the patient stood barefoot. Although the 95% confidence ellipses for whole body CoP and CoM were reduced when wearing insoles implying better balance, the CoP of the right leg moved in wider area than the CoP of the left leg. Also the left side became dominant, as indicated by the higher proportion of the total vertical load over this limb. The reason for the asymmetrical weight distribution cannot be deducted from the available data. The use of orthoses reduces the range of internal hip rotation improving the stability during quiet standing in the case. Also, asymmetrical weight distribution over the feet may result in better balance.

Introduction

Impaired balance is one of the primary problems encountered by children with spastic diplegic cerebral palsy (CP)[1]. Children with spastic diplegia who can walk independently

Corresponding author: Pornsuree Onmanee

Sirindhorn National Medical Rehabilitation Centre 88/26 Tivanon Rd. MuangNonthaburi, Thailand

Telephone: +66-2591-5455, Ext 6720 Fax: +66-2591-5455 E-mail: p_onmanee@hotmail.com

often cannot stand on 1 leg for 10 s, a task that can be accomplished by children with normal development. In addition, these children usually fail to elicit the stepping response to a disturbance, e. g. pushing. Because of deficiencies in lateral reaction, which maintains balance, these children tend to shift their upper body to the stance side. This shift results in a posture similar to the one adopted in case of a weak hip abductor [2].

A good base for support has proved beneficial to maintain balance. The orthoses may influence balance in this regard; however, the evidence is insufficient. The orthoses available for children with CP who walk independently, ranges from foot orthoses to ankle-foot orthoses (AFOs), including flexible, rigid, and ground reaction force (GRF) AFOs.

To evaluate postural control, many studies have investigated kinematic and kinetic data and the ability of CP patients to perform upper limb functions, e.g., reaching, while standing [3,4]. Furthermore, the movement of the centre of pressure (CoP), as calculated from the GRF, can be an indirect indication of the standing ability [5]. Many studies have used a single force platform to assess the movement of the CoP [6-8], although the forces exerted by each limb could not be separated from each other. A dual force platform was developed to investigate the motor mechanisms underlying static standing balance control [5,9-11].

This study aims to identify the kinematic effects of a current orthotic device on the standing balance in a child with spastic diplegic CP. For the further development of a device to help children with spastic diplegic CP, identifying the advantages of the current device in clinical settings might be useful.

Methods

The patient data was used with the permission of the director of Sirindhorn National Medical

Rehabilitation Centre (SNMRC). The method was explained to the patient and her parents before they agreed and signed the consent forms that were approved by the ethical committees of SNMRC.

Case Description

The patient was a 12-year-old girl with spastic diplegic CP. She was referred as an outpatient to the orthotic clinic at SNMRC. She had level II gross motor function, determined using the gross motor function classification system (GMFCS), which is a simple 5-level ordinal grading system used to describe gross motor function in children with CP. She walked with some limitation, although she was independent during community ambulation. Toe drag, toe out, and pronated feet (calcaneal valgus and flat feet) were signs noted during observational gait assessment.

According to the conditions presented, the patient should have been prescribed with AFOs to control ankle and foot alignment. However, she declined the use of AFOs because of its appearance. Further, she believed that it would offer little advantage. Subsequently, she was prescribed with insoles to correct foot alignment and indirectly control the ankle joints.

Orthoses

Semi rigid insoles made of 2 layers of 5-mm ethyl vinyl acetate (EVA) foam sheets with reinforcement plastic, which was length $\frac{3}{4}$ of the foot length, between them were prescribed (Figure 1). These insoles were customized using the foot-casting models. The forefoot and hind foot were corrected to neutral position during manual casting procedure. The medial arch, longitudinal arch and metatarsal dome were created during modification of positive casting model. The patient was scheduled for a follow-up after 1 month of using the orthoses and standing data was captured on the same day.



Figure 1. Prescribed insoles

Instrumentation

Kinematic data were collected by using a Vicon three-dimensional (3D) motion analysis system (Oxford Metrics, Oxford, UK) at the SNMRC. This system consists of 8 MX cameras, 3 AMTI force plates (AMTI Inc., Watertown, MA), 2 high-speed VDO cameras, and the Vicon nexus software. The sampling rate of the force plate was 1000 Hz and that of the cameras was 100 Hz.

Procedure

The testing for the subject was completed during 1 session. The subject was instructed to hold the standing position with and without the shoes fitted with the customized insoles on both legs.

The subject performed the first trial barefoot and the subsequent trials with the orthoses. The subject was instructed to stand for 30 s and then step off the force plates for 120 s of resting a chair between trials. GRF and kinematic data were collected during the trial periods.

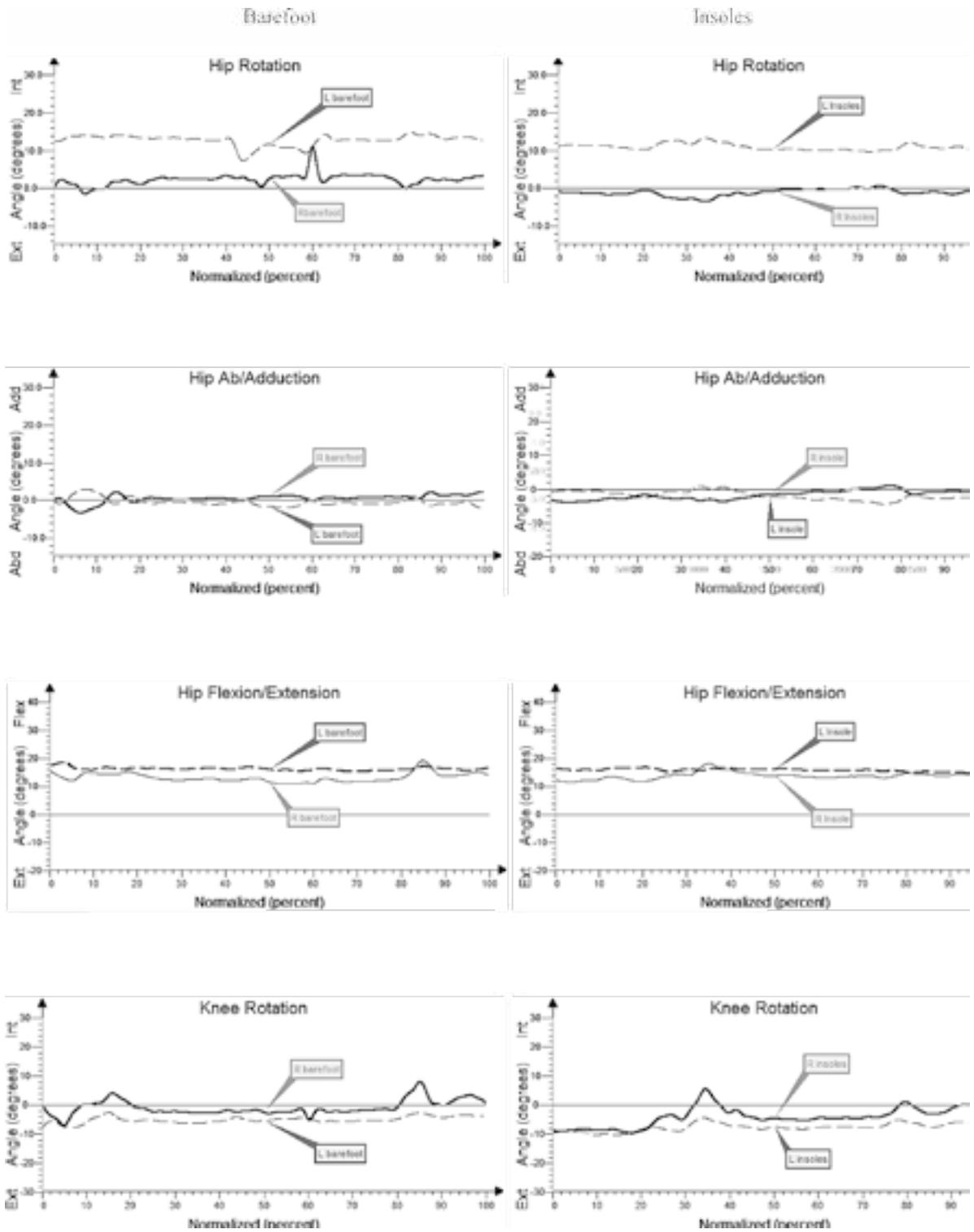
The GRF was captured by the forceplates, and the kinematic coordinate data were captured by the Vicon system. The variables chosen for analysis during the trials included the angles and moments at the hip, the knee, and the ankle and the magnitude of movement of the CoP. The CoP on the left foot (CoP L) and right foot (CoP R) were used to calculate the combined CoP (CoPC). The whole body centre of mass (CoM) was calculated by BodyBuilder using Plug-In gait model described by Vicon (Oxford Metrics,

Oxford, UK). This model with the corresponding marker placement defines segments of the body. The sum of all the centres of mass of these segments was calculated as the whole body centre of mass (CoM). The model requires some anthropometric information such as height, weight, length of the legs, knee widths, and ankle widths. The CoPC and CoM variables—peak sway velocity and range of sway—were calculated in both the anteroposterior (A/P) and the mediolateral (M/L) directions. In addition, the 95% confidence ellipse areas around the CoP C and CoM movements along both the A/P and M/L axes were obtained using the method presented in previous studies [13]. Further, the vertical component of the GRF (vGRF) exerted by the right and left legs was compared. The Vicon Plug-In gait model and software were used to analyze 3 successful standing trials for each condition.

Results

During the 30 s of standing, the patient had a crouching posture, irrespective of the insole application; the hips and knees were in flexion and the ankles were in dorsiflexion throughout the trials. The magnitude of the hip and knee flexions and the ankle dorsiflexion increased for a short time. Kinematically, the range of the internal hip rotation was only slightly smaller when wearing the insoles than when barefoot. The moments at the main joints of the lower limbs were similar in both conditions (Figure 2).

The CoM moved in the same direction as the CoP C, but it moved slower than the CoP



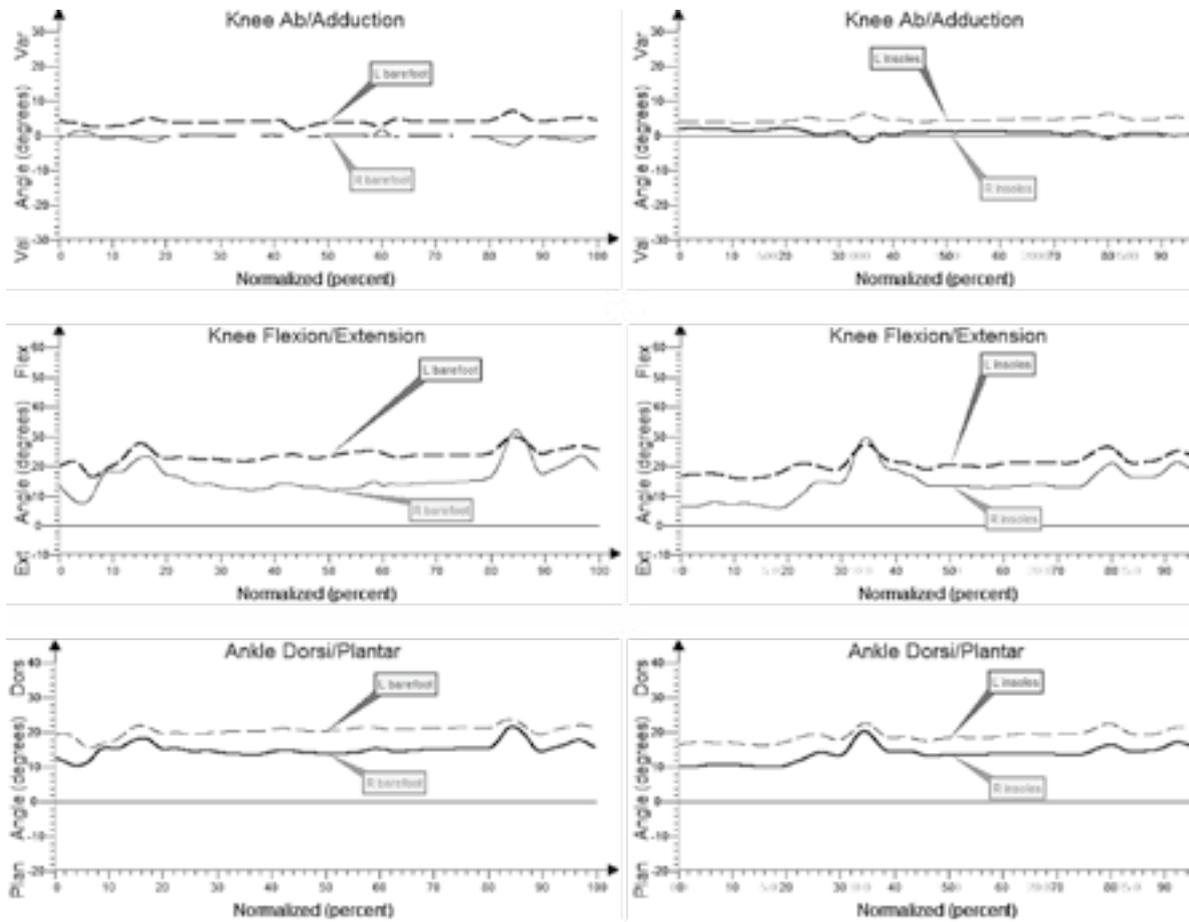


Figure 2. Kinematic graphs : the hip, knee and ankle

C, irrespective of the insole application. The mean velocity or path length (the average travel distance of net body CoP per second in an entire trial) of the CoP C (1.11 mm/s) was approximately 10 times as fast as the CoM (0.11) (Table 1, Figure 3, 4).

The A/P sways of both CoP C and CoM when standing with the insoles were smaller than those observed when standing barefoot, but the M/L sways of them when standing with the insoles were greater than that when standing barefoot and tended toward the left foot. The root-mean-square distance or RMS derived from the resultant distance of the CoM was reduced from 22.28 mm to 11.61 mm when wearing insoles. The 95% confidence ellipse areas of the CoP C

and the CoM when wearing the insoles were also smaller than those observed when the patient was barefoot (Table 1, Figure 5, 6).

The distance parameters of CoP L were significantly decreased when wearing insoles while only slight reduction of CoP R movement was noticed. Also the CoP L tended to move closer to the longitudinal foot axis (from 528.78 mm² to 377.19 mm²) whereas the CoP R moved further away from the right longitudinal foot axis (from 485.86 mm² to 501.86 mm²). The mean velocity of CoP of both feet were maintained (Table 1).

The average vertical component of ground reaction force (vGRF) of the left leg was greater than that of the right (left limb index: the ratio of

Table 1. Sway parameters in barefoot condition

Sway parameters	CoP L	CoPR	CoP C	CoM
A/P Range (mm)	50.88	52.79	48.64	37.24
M/L Range (mm)	20.61	17.99	41.22	31.06
Mean velocity (mm/s)	1.1	1.39	1.11	0.11
RMS (mm)	15.38	12.91	15.93	22.28
95% Confidence ellipse (mm ²)	279.47	296.12	1394.1	895.92
Area along the longitudinal foot axis (mm ²)	528.78	485.06	--	--
Limb index	0.50	0.50	--	--

Note: The origin is the mid-point between two feet on the floor where value in z axis is zero.

Table 2. Sway parameters in insole condition

Sway parameters	CoP L	CoPR	CoP C	CoM
A/P Range (mm)	43.01	51.62	36.33	25.29
M/L Range (mm)	14.21	27.72	49.94	43.93
Mean velocity (mm/s)	1.1	1.36	1.15	0.13
RMS (mm)	9.07	11.04	13.86	11.61
95% Confidence ellipse (mm ²)	147.76	288.34	1108.41	735.91
Area along the longitudinal foot axis (mm ²)	377.19	501.86	--	--
Limb index	0.51	0.49	--	--

Note: The origin is the mid-point between two feet on the floor where value in z axis is zero.

the overall load and the load on the left foot is 0.51) when the insole was worn (Figure 7). In contrast, distribution of the vGRF was 0.5 for both legs in the barefoot condition (Figure 7). Subjectively, the patient reported that she could stand longer with these insoles than when barefoot.

Discussion

Kinematically, the range of the internal hip rotation when wearing the insoles was only slightly smaller than that observed when standing barefoot. Ferdjallah and his team proved that limb rotation altered the GRF forces in both M/L and A/P directions [13].

According to the 3D data obtained from this subject, the movements of the CoM and CoP in the child with spastic diplegic CP were similar to

those observed in healthy subjects. These observations implied that a similar control strategy, i.e., the control of the A/P displacement mainly by the plantarflexors and dorsiflexors and the control of the M/L displacement mainly by the hip abductor and adductor, was employed in both cases. However, the ranges of these movements in the patient were greater than the normal values.

The pattern of the CoP movement in the A/P direction, controlled by the plantarflexor and dorsiflexor, can be explained using the inverted pendulum model. The inverted pendulum model describes how the difference between the positions of the CoP and CoM controls the direction of the angular acceleration of an inverted pendulum. Winter and his team extensively analyzed this model in both sagittal

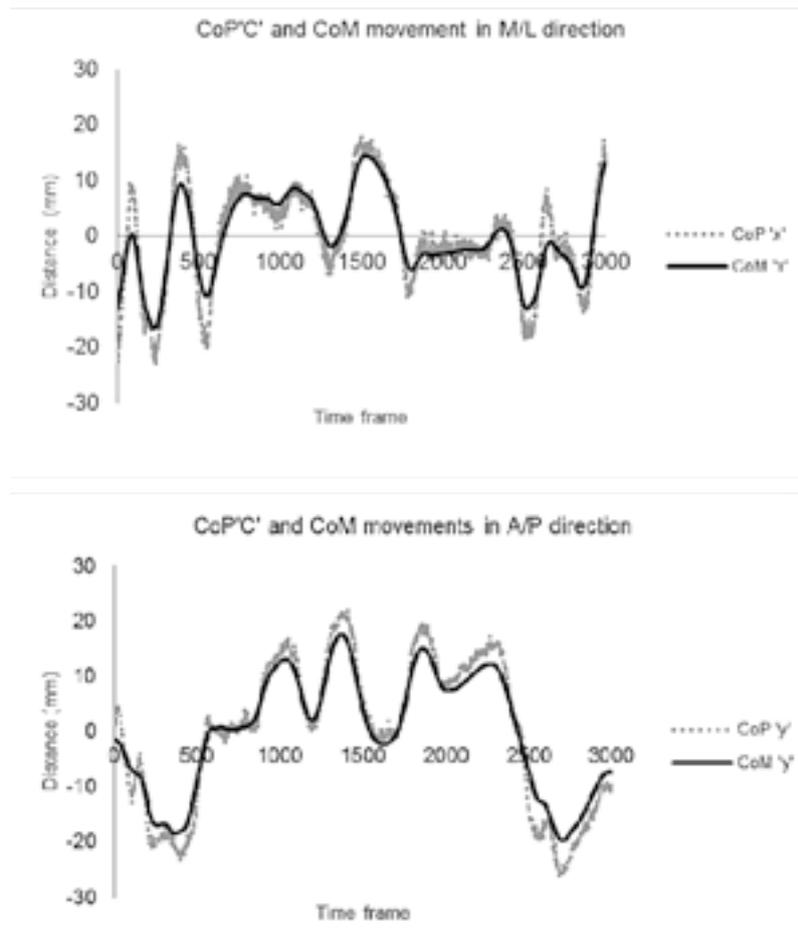


Figure 3. The CoP C and CoM movement in X (mediolateral direction) and Y (anteroposterior direction) during standing 'Barefoot'

and frontal planes[10]. However, the A/P sways of the CoP C in CP children (48.64 mm) was larger than the normal values (approximately 16 mm) [10]. In our study, when the patient wore the insoles, this value reduced to 36.33 mm. The large sway implied impaired balance in the corresponding direction, which can be reduced slightly by the use of foot orthoses. For this child, the A/P sway of the CoP R was reduced by only 1.17 mm, but that of the CoP L was reduced by approximately 6 mm when wearing the insoles. The difference between these values may be due to the fit of the orthoses, the support and correction they provided, and the individual's

physical condition.

The M/L sways of the CoP C, which was 41.22 mm when barefoot and 49.94 mm with the insoles, were also greater than the normal value, which is approximately 14 mm [9]. In the M/L direction, the CoP C moved in a relatively wide range when the patient wore insoles and tended toward the left foot. The M/L sway of the CoPR with the insoles was greater than that observed when the patient was barefoot. However, the M/L sway of the CoP L with the insoles was less than that observed when the patient was barefoot and less than the CoP R when standing with insoles. This implied that the left side was better balanced

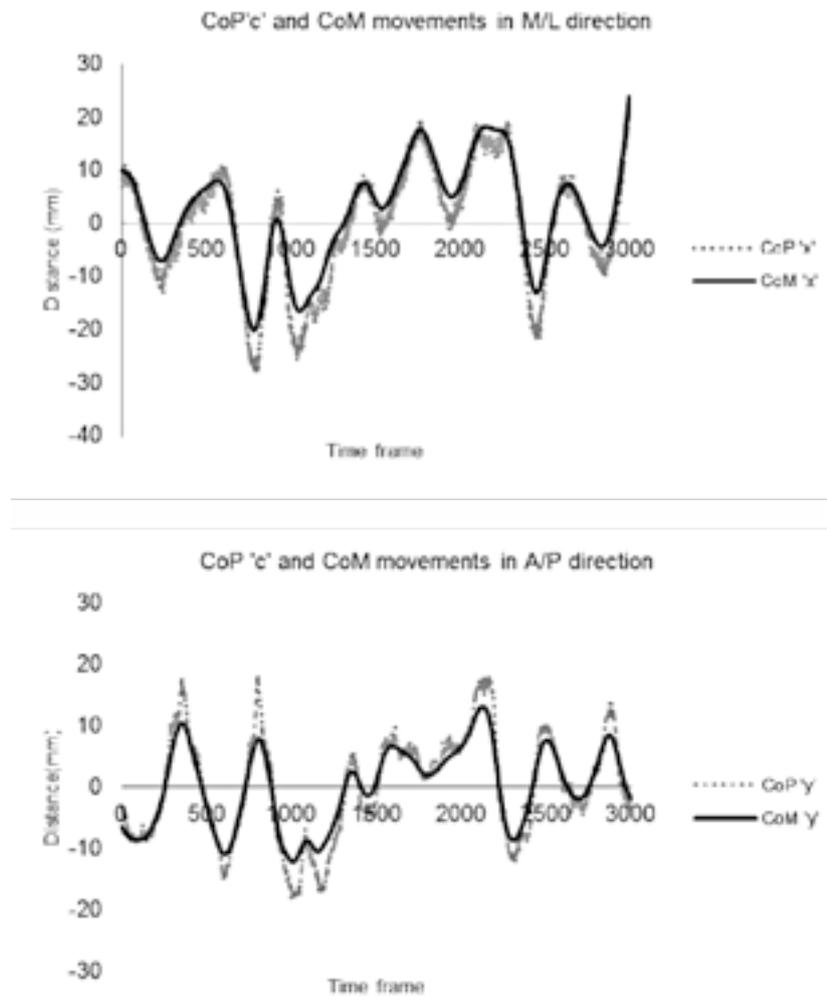


Figure 4. The CoP C and CoM movement in X (mediolateral direction) and Y (anteroposterior direction) during standing 'with insoles'

than the right when wearing the insoles.

Irrespective of insole application during this study, the oscillatory patterns of the vGRF and M/L movements of the CoP C, CoP L, and CoP R were similar to those observed in healthy subjects explained by the “load/unload” mechanism of the body that is mainly controlled by the hip abductors and adductors (Figure 7). The maximum oscillating forces of the left and right sides are approximately half of the body weight. The fluctuations of these forces were virtually equal in magnitude and 180° out of phase[9].

The insoles improved stability during quiet standing by reducing the range of the internal hip rotation because the CoP C and CoM were moved in the smaller area, indicating more effective balance control. However, the child still had an apparent crouching posture.

Moreover, the insoles reduced the CoP L movement in this subject, showing improved stability on the left side, while the CoP R had a wider range of movement than the CoP L; therefore, the left side became dominant, as indicated by the higher proportion of the total

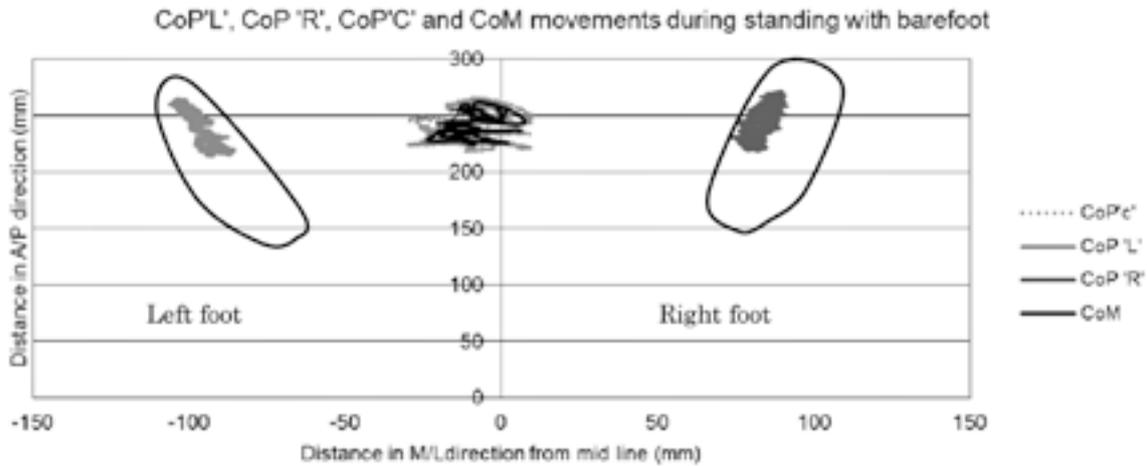


Figure 5. CoP and CoM Trajectories in 'Barefoot condition'. The feet were drawn from toe, heel and ankle markers.

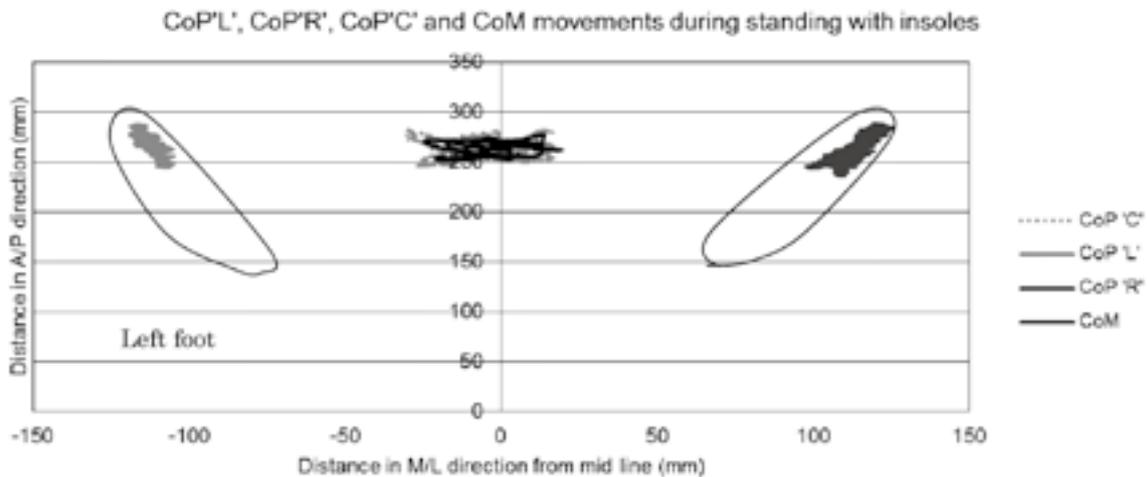


Figure 6. CoP and CoM Trajectories in 'Insole condition' The feet were drawn from toe, heel and ankle markers. Note: Due to the CoM calculation errors, the mean values of CoM coordinations were offset to the same values of CoP C, then the instantaneous coordination were recalculated.

vertical load over this limb. Even though the M/L sways of the CoP R, CoP C, and CoM increased, they were moving toward the dominant side—the left side. The reason for the asymmetrical weight distribution cannot be deduced from the available data.

Conclusion

Overall, the insoles could improve stability in this child, even though the weight distribution between the 2 legs was not equal. The dominant leg showed greater stability. Therefore, a symmetrical weight distribution between the 2 legs may not always imply better stability than that afforded by an asymmetrical weight

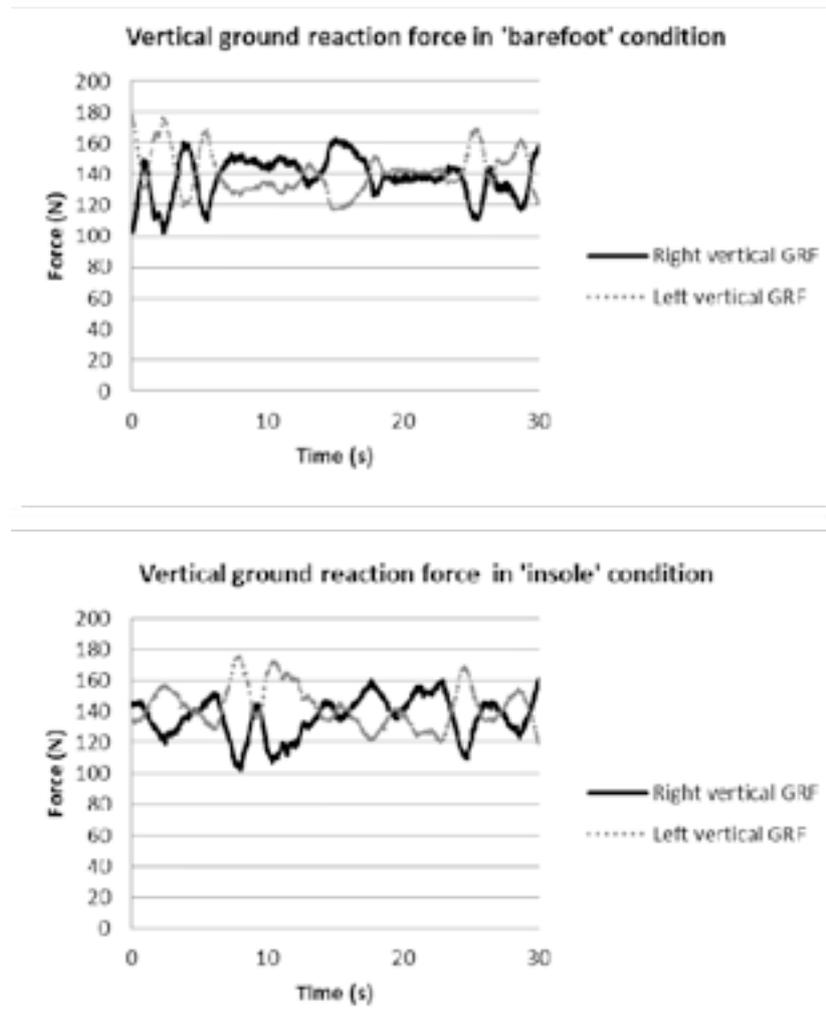


Figure 7. Vertical ground reaction forces in 2 conditions

distribution.

Future work

A better understanding of standing balance could be acquired by using electromyographic (EMG) signals to assess muscular activity. The effect of weight distribution between the 2 feet while balancing should be further investigated.

Furthermore, different types of orthoses can be considered to prevent excessive ankle dorsiflexion and observe the indirect effects on the knee and hip. The fit of the orthoses must be assessed and controlled. The foot positions while standing

should also be controlled when comparing the effectiveness of different orthoses.

Acknowledgements

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References

1. Gage JR. The Treatment of Gait Problems in Cerebral Palsy (Clinics in Developmental

- Medicine). London: Mac Keith Press; 2004.
2. Bleck EE. Orthopaedic management in cerebral palsy(Clinics in Developmental Medicine). London: Mac Keith Press;1987.
 3. Jesinkey K, Naslund A, Hirschfeld H. Initiation of reaching when standing with and without DAFOs in children with spastic diplegia. *Advances in Physiotherapy*.2005;7: 144-153.
 4. Naslund A, Sundelin G, Hirschfeld, H. Reach performance and postural adjustments during standing in children with severe spastic diplegia using dynamic ankle-foot orthoses. *J RehabilMed*.2007; 39: 715-723.
 5. Rha DW, Kim DJ, Park ES. Effect of hinged ankle-foot orthoses on standing balance control in children with bilateral spastic cerebral palsy. *YonseiMedical Journal*. 2010; 51: 746-752.
 6. Riach CL, HayesKC. Maturation of postural sway in young children. *Dev Med Child Neurol*. 1987; 29: 650-658.
 7. Rose J, Wolff DR, Jones VK, Bloch DA, Oehlert JW, Gamble JG. Postural balance in children with cerebral palsy. *Dev Med Child Neurol*. 2002; 44: 58-63.
 8. Wolff DR, Rose J, Jones VK, Bloch DA, Oehlert JW, Gamble JG. Postural balance measurements for children and adolescents. *JOrthopRes*. 1998; 16: 271-275.
 9. Winter DA. Biomechanics and motor control of human movement. Hoboken, New Jersey: John Wiley & Sons, Inc; 2009.
 10. Winter DA, Patla AE, Ishac M, Gage WH. Motor mechanisms of balance during quiet standing. *J ElectromyogrKinesiol*. 2003; 13: 49-56.
 11. Winter DA, Prince F, Frank JS, Powell C, Zabjek KF. Unified theory regarding A/P and M/L balance in quiet stance. *J Neurophysiol*. 1996; 75: 2334-2343.
 12. Bigelow KE. Identification of key traditional and fractal postural sway parameters to develop a clinical protocol for fall risk assessment in older adults. The Ohio State University2008. Available from: <http://etd.ohiolink.edu/send-pdf.cgi/Bigelow%20Kimberly%20Edginton.pdf?osu1227561171> (accessed March 31, 2012)
 13. Ferdjallah M, Harris GF, Smith P, Wertsch JJ. Analysis of postural control synergies during quiet standing in healthy children and children with cerebral palsy. *Clinical biomechanics*.2002; 17: 203-210.

Appendix

Calculations

These are suggested by Bigelow [12].

Pre-calculation

Mean location of the centre of pressure (\bar{x} , \bar{y}) calculated from:

The calculation of anterior-posterior data set

$$y_n = y_i - \bar{y} \quad \text{for } i = 1 \dots N \quad (1)$$

Where y = the anteroposterior CoP displacement

y_n = the desired transform data point of interest

y_i = the desired raw data point of interest

\bar{y} = is the average value of the entire A/P CoP data series;

$$\bar{y} = \frac{1}{N} \sum_{i=1}^N y_i$$

Where N = the total number of sample

The calculation of medial-lateral data set

$$x_n = x_i - \bar{x} \quad \text{for } i = 1 \dots N \quad (2)$$

Where x = the mediolateral CoP displacement

x_n = the desired transform data point of interest

x_i = the desired raw data point of interest

\bar{x} = is the average value of the entire A/P CoP data series;

$$\bar{x} = \frac{1}{N} \sum_{i=1}^N x_i$$

Where N = the total number of sample.

The calculation (1), (2) will be performed for the total N number of x_i and y_i values.

1. Area along the longitudinal foot axis

Assumption : the area along the foot axis which is drawn from toe marker to heel marker is oval shape. Therefore $\text{Area} = \Pi * M/L_{\text{sway}}/2 * A/P_{\text{sway}}/2$

2. The 95% confidence ellipse.

This is a statistical measure to describe the area of sway based on the assumption that there is a normal bivariate distribution of the data. The calculation fits the major axis to be coincident with the primary direction of sway. Then fits an ellipse to the data. So that the center of pressure data points is in the ellipse with 95% confidence. The calculation consists of a number of steps:

First, the co-variance matrix of the data points, transformed to be given with respect to the mean location of the COP, must be calculated according to equations (3) through (8).

$$\text{Co-variance Matrix} = \begin{bmatrix} \sigma_x^2 & \sigma_{xy} \\ \sigma_{xy} & \sigma_y^2 \end{bmatrix} \quad (3)$$

where: σ_x = the RMS of the medial-lateral data center of pressure excursion, calculated using (8)

$$\sigma_x = \text{M/L RMS} = \sqrt{\frac{\sum_{n=1}^N x_n^2}{N}} \quad (4),$$

where: σ_y = the RMS of the anterior-posterior data center of pressure excursion, calculated using (9)

$$\sigma_y = \text{A/P RMS} = \sqrt{\frac{\sum_{n=1}^N y_n^2}{N}} \quad (5),$$

and where: σ_{xy} is calculated using (6)

$$\sigma_{xy} = \frac{\sum_{n=1}^N (x_n * y_n)}{N} \quad (6)$$

The co-variance matrix is then used to calculate the eigenvalues. The eigenvalues can be calculated by substituting in equation (7) for the two eigenvalues, λ_1 and λ_2 .

$$\det \left(\begin{bmatrix} \lambda & 0 \\ 0 & \lambda \end{bmatrix} - \begin{bmatrix} \sigma_x^2 & \sigma_{xy} \\ \sigma_{xy} & \sigma_y^2 \end{bmatrix} \right) = 0 \quad (7)$$

Taking the determinant, results in equation (8):

$$\lambda^2 - (\sigma_x^2 + \sigma_y^2) \lambda + (\sigma_x^2 \sigma_y^2 - \sigma_{xy}^2) = 0 \quad (8)$$

There are two solutions of λ , λ_1 and λ_2 using the two eigenvalues. Multiplying each of these eigenvalues by the desired confidence gives the length of the semi-axes of the ellipse. The F-value for a 95% confidence ellipse is 1.96.

Equation (9) calculates the measure of the major semi-axis of the ellipse.

$$a = 1.96 \times \sqrt{\lambda_1} \quad (9)$$

Equation (10) calculates the measure of the minor semi-axis of the ellipse.

$$b = 1.96 \times \sqrt{\lambda_2} \quad (10)$$

Finally, substitute these values into equation (11)

$$95\% \text{ Confidence Ellipse Area} = \pi \times a \times b \quad (2. 15)$$