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Prosthetics and Orthotics International

Effects of Ankle-Foot Orthoses on acceleration and energy cost of walking in children and adolescents with cerebral palsy

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Abstract

Background: Impaired postural control is a key feature of cerebral palsy that affects daily living. Measures of trunk movement and acceleration have been used to assess dynamic postural control previously. In many children with cerebral palsy, ankle-foot orthosis ~~is~~ used to provide a stable base of support, but ~~its~~ effect on postural control is not yet understood.

Objectives: The objectives of the current study were to investigate the effects of ankle-foot orthosis on postural control and energy cost of walking in children with cerebral palsy.

Study design: Clinical study with controls.

Methods: Trunk accelerometry (amplitude and structure) and energy cost of walking (J/kg/m) were recorded from five-minute walking trials with and without ankle-foot orthosis for children with cerebral palsy and without orthosis for the reference group.

Results: Nineteen children with unilateral spastic cerebral palsy and fourteen typically developed children participated. The use of ankle-foot orthoses increased structure complexity of trunk acceleration in mediolateral and anterior–posterior directions. The use of ankle-foot orthoses changed mediolateral-structure toward values found in typically developed children. This change was not associated with a change in energy cost during walking.

Conclusions: The use of ankle-foot orthosis does affect trunk acceleration that may indicate a beneficial effect on postural control. Using measures of trunk acceleration may contribute to clinical understanding on how the use of orthosis affects postural control.

Keywords

gait, postural control, ankle-foot orthoses, AFO, cerebral palsy, acceleration, trunk

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Introduction

Cerebral Palsy (CP) is the most common motor disorder in children,[1] and impaired control of posture is one of the main components associated with the condition.[2] Controlling posture, defined as the relationship between the different parts of the body and between the body and a reference frame, is essential for obtaining balance.[3] Hence, impaired postural control limits activities of daily living and participation in social activities for children with CP. Although around 70% of children with CP are able to walk, they experience difficulties related to walking and as many as 35% of ambulatory children with CP may fall daily.[4] Changes in trunk movement and postural control also affect energy cost of walking[5,6] (ECW). Methods to assess and interventions aiming to improve postural control are therefore important in the clinical management of children with CP.

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Trunk movement reflects aspects of postural control, and increased movement amplitude of the trunk has been described as an indicator for impaired dynamic postural control.[7,8] Measures of trunk acceleration have also been used to describe aspects of postural control during walking, and there are a growing number of studies using this method in children with CP.[9] Iosa found increased amplitude of trunk accelerations in anterior–posterior (AP), mediolateral (ML), and vertical directions in a group of children with CP.[10] Sæther later verified this finding and reported that more severely affected children, reflected as higher levels of the Gross Motor Function Classification System (GMFCS),~~[11]~~ had increased trunk acceleration compared with children with lower GMFCS level. In all these studies, the amplitude of the acceleration signal has been reported. Another characteristic of the acceleration signal is the structure of the signal that describes its complexity and adaptability.[12] This may be an important characteristic to assess because children with CP show an excess of antagonistic coactivation during external perturbations and a reduced capacity to modulate postural adjustments.[13]

Despite the importance of impaired postural control in children with CP and its effect on walking,[14] little research has addressed the effect of ankle-foot orthoses (AFOs) on postural control in ambulatory children with CP. AFOs are commonly used to provide a stable base of support, to correct or prevent deformities, and to improve economy of walking in children with CP.[15,16] AFOs have been shown to increase walking velocity, step length, and single support[17-19] and decrease cadence and ECW.[20,21] To our knowledge, only three studies have investigated the effect of AFOs on postural control without consistent findings. Swinnen reported increased trunk movement in all planes,[7] whereas Degelean found only increased trunk frontal angular velocity[8] and Meyns[6] found increased trunk lateral flexion, trunk rotation, and ML gait instability. Meyns also highlighted the importance of gait stability in the context of ECW. We were not able to identify any studies addressing the effect of AFOs on postural control assessed by accelerometers.

Therefore, the overall aim of the current study was to investigate the effect of AFOs on postural control in children with CP, assessed by trunk worn accelerometers. Because AFOs are intended to provide a more stable base of support, we hypothesized that the use of AFO would improve postural control, reflected as a change in both the amplitude and the structure of the acceleration signal toward values found in typically developing children. A second objective was to investigate whether potential changes in postural control are associated with changes in ECW. A better understanding of how the use of AFOs influences postural control and ECW will be valuable for clinicians when prescribing AFOs to children with CP.

Methods

This study includes baseline data from an ongoing randomized AU2 controlled trial ~~[XXXXX]~~ and clinical data obtained from regular follow-up at ~~XXXX~~.

Study group and design

Children ~~with unilateral spastic CP~~ using AFOs were included. Exclusion criteria were fixed contractures in the ankle joint and major cognitive impairments. Patients who reported problems with fitting or function of their AFOs were not included. Fourteen typically developed (TD) children were included as reference. Ethical approval was granted by the regional ~~XXXX~~ ethics committee for medical and health research in ~~XXXX~~. Parents signed a written consent before participation.

Protocol and equipment

All data used in the study were collected during five-minute walk tests (5MWTs). The 5MWTs were performed twice for children with CP, first with shoes only and then with shoes and AFO. TD children performed the test once with shoes. The children walked back and forth a corridor of 25 m at comfortable speed. Walking distance was measured with a measuring wheel handled by one of the study staff who walked right behind the child. Start and stop times were recorded manually.

Trunk acceleration was measured with a triaxial accelerometer (Axivity AX3, Axivity, Newcastle, UK) fixed to the skin on the lower back (central at the lumbar vertebrae 3). The Axivity AX3 monitor (23.0 × 32.5 × 7.6 mm; 11 g) measures acceleration in vertical (V), AP, and ML directions. The AX3 software (Omgui version 1.0.0.28) was used for configuration and downloading of the logged data. The sampling frequency was 200 Hz, and the sample range was ±8 g. The Axivity AX3 monitors logged raw acceleration data in a binary packed format (Continuous Wave Accelerometer).

Concurrent gas exchange measurements (Metamax MMX II; Cortex Biophysik GmbH, Leipzig, Germany) were performed. The VO_2 and VCO_2 gas analyzers were calibrated before each test day using high-precision gases (16.00 ± 0.04 O_2 and 5.00 ± 0.01 CO_2, Riessner-Gase GmbH & Co, Lichtenfels, Germany), whereas the inspiratory flow meter was calibrated with a 3 L volume syringe (Hans Rudolph Inc., Kansas City, MO). The participants wore the mobile gas analyzer in a neoprene harness on their back. The mobile gas analyzer was worn by the participants in a neoprene harness and had a total weight of 980 g.

Data analysis

Raw acceleration data were downloaded onto a computer, and the raw data were converted into ASCII files using AX3 software (Omgui version 1.0.0.28). The separate periods for each 5MWT were exported into Matlab (MATLAB, R2018a, Natick, MA). Two outcome variables were used for further analysis: (1) the amplitude of the acceleration signal was given as SD of the signal (because of the oscillating nature of the acceleration signal, SD is synonymous with the root mean square, in such signals[22]) and (2) the structure of the acceleration signal was analyzed by Sample Entropy (SE). In studies of trunk stability in other populations, the structure of the acceleration signal is commonly analyzed because changes in the structure of the acceleration pattern are thought to reflect changes in gait stability.[23] SE has been shown to be robust and does not rely on step detection, a potentially critical source of error in related measures.[23] SE is computed by first creating two vectors $X_m(t) = [x(t), x(t+1),\ldots, x(t+m)]$ and $X_{m+1}(t) = [x(t), x(t+1),\ldots, x(t+m+1)]$ of the acceleration signal $x(t)$ in AP, ML, or V direction. Next, the SE is defined by the following equation[24]:

$$SE = -\log\frac{A}{B},$$

where A and B are the number of points where the Chebyshev distance $d[X_{m+1}(i), X_{m+1}(j)] < r$ and $d[X_m(i), X_m(j)] < r$ for $i \neq j$, respectively. In this study, parameter $m = 5$ and $r = 0.3$*SD as used in the previous study on gait in older adults.[25] SE is independent of the acceleration amplitude because r is defined relative to the SD of the individual acceleration signal. A signal that is periodic and regular will have SE≈0, whereas a random noisy signal will have SE approaching infinity. Thus, SE quantifies the irregularity of the structure of the trunk acceleration signal reflecting the pattern of postural adjustment during the 5MWT.

ECW (J/kg/m) is acknowledged as a precise indicator of walking economy in children with CP and can reliably be determined using a 5MWT protocol.[20] ECW was calculated by dividing energy consumption by walking speed. Energy consumption was calculated during the steady state, defined as the 2 minutes where walking speed, oxygen uptake (VO2), and carbon dioxide production (VCO2) showed the least fluctuations (inspected visually) and the respiratory exchange ratio (VO2/VCO2) was below 1.0. The following formula was used:

$$Energy\ cost\ (J/kg/m) = \frac{(4.960x\ RER + 16.040)x\ VO_2\ /body\ weight}{walking\ speed\ [m/min]}.$$

Using the start and stop times for each 5MWT and distance from the measuring wheel, gait speed was calculated for the entire 5MWT and, separately, for the past 2 minutes. Speed was recorded as meters per minute (m/min). Speed values from the entire 5MWT were used to compare conditions and groups.

Statistics

Statistical analysis was performed using IBM SPSS Statistics for Windows, version 25 (IBM Corp, Armonk, NY). Normal distribution of the data was verified by visual inspection of Q-Q plots. To assess the internal validity of the data, we compared children with CP walking without AFOs with TD children. Student's *t*-tests were used to assess group difference for ECW, amplitude, and structure of the acceleration signal.

The first objective was to investigate the effect of AFOs on postural control, assessed by changes in amplitude and structure of the acceleration signal between conditions ~~notAFO/AFO~~. A general linear model, repeated measures ~~ANOVA~~, on amplitude and structure was used. GMFCS level was used as factor and speed as covariate to control for variations caused by these variables. Linear regression was performed on variables with interaction effect.

The second objective was to investigate whether there was an association between AFO-related changes in postural control and ECW. Spearman correlation was used for this assessment. Statistical significance was set to $P < 0.05$ for all statistical tests.

Results

Nineteen children with unilateral spastic CP and GMFCS I-II[1] using AFOs participated in this study. Characteristics of the participants are presented in Table 1. Figure 1 shows an overview of the AFOs used in this study. The time since the past visit to their orthotists, either for adjustments or fitting of the current AFO, was less than 6 months for 14 of the children and between 6 and 12 months for five of them.

There was no difference between children with CP walking without AFOs and TD in gait speed, whereas ECW was higher in children with CP (5.21 vs 4.28 J/kg/m, $P = 0.05$; Table 1). Acceleration amplitudes (V, ML, and AP) were higher in children with CP, and the structure complexity of the acceleration signal was significantly greater in the TD children in the ML direction (Figure 2).

The use of AFOs had a significant main effect on the structure of the acceleration signal and increased its complexity in V and ML directions (Table 2). The main effect of amplitude (V, ML, and AP) could not be interpreted because of the significant interaction effect with speed. The constant for the regression line of each amplitude measure (Figure 3) shows some reduction in ~~AFO~~ condition when no change in walking speed occurs. However, most of the variation in amplitude is explained by changes in walking speed in V = 79% ($P < 0.001$), ML=70% ($P = 0.001$), and AP = 57% ($P \leq 0.001$). No interaction was found between the use of AFO and GMFCS level for any of the variables. No significant association between AFO-related changes in the structure of trunk acceleration and energy cost was found with the Spearman correlation.

Discussion and Conclusions

The aim of this study was to investigate the effect of AFOs on postural control in ambulatory children with CP. The use of AFOs increased structure complexity, and the changes in the ML direction represent a normalization toward values found in TD.

To confirm internal validity, we compared gait characteristics of children with CP with those of TD children. We found that children with CP walked with increased amplitude of trunk acceleration (V, ML, and AP), reduced complexity of structure in the ML direction, and elevated energy cost compared with TD children while walking at the same walking speed. These results show a difference in gait characteristics between children with CP and TD in accordance with the literature.[11,20,21]

Previous studies have shown that ambulatory children with CP have impaired trunk control and walk with increased trunk

Table 1. Characteristics of the participating children with CP and TD children.

	Children With CP, n = 19		TD Children, n = 14	
	n	**%**	**n**	**%**
Male gender	7	36.84	8	57.14
Female gender	12	63.16	6	42.86
GMFCS I	14	72.72	—	—
GMFCS II	5	26.32	—	—
	Mean	**SD**	**Mean**	**SD**
Age (y)	9.60	2.69	10.06	2.19
Height (cm)	141.29	16.80	139.85	14.03
Weight (kg)	38.24	12.46	35.10	10.71
Gait speed ~~not~~AFO (m/min) [a]	65.41	10.77	66.71	12.23
Gait speed AFO (m/min)[b]	68.76	10.96	—	—
ECW (J/kg/m) ~~not~~AFO[c]	5.21[d]	1.63	4.28[d]	0.83
ECW (J/kg/m) AFO[c]	5.02	1.24	—	—

Abbreviations: AFO, ankle-foot orthosis; CP, Cerebral Palsy; ECW, energy cost of walking; GMFCS, Gross Motor Function Classification System; TD, typically developed
[a]*Gait speed for children with CP and TD children walking with shoes/without AFOs.*
[b]*Gait speed and for children with CP walking with shoes and AFOs.*
[c]*Energy cost of walking (J/kg/m) during five-minute walk test.*
[d]*Near significant difference in mean between CP and TD and independent t-test of P = 0.05.*

Figure 1. Types of AFOs used: (A) dynamic carbon composite with flexible foot plate, used by four children (ToeOFF Allard); (B) thermoplastic AFO with free dorsiflexion and plantar flexion stop and flexible toe plate, used by 12 children; (C) carbon composite AFO with functional joint and semirigid foot plate, used by one child; and (D) rigid thermoplastic AFO with semiflexible foot plate, used by two children. AFO, ankle-foot orthosis.

accelerations, which is reflected in the amplitude of the signal.[11] When fitting AFOs, further increase of movement excursion of the upper body was reported[7,8] and related to diminished trunk control. In addition, a possible trade-off has been described between increasing AFO stiffness and increased movement of the trunk and relative increase of ECW.[6]

In our study, the use of AFOs showed a tendency to reduce the amplitude (V, ML, and AP directions, Figure 3) in contrast to the study by Degelean[8] who found increased amplitude of trunk acceleration. The strong relationship between changes in walking speed and changes in acceleration signal has been described before.[27] It has also been pointed out that this relationship can be different in the presence of gait impairments[28] and might therefore be difficult to predict. The use of AFOs did not result in a uniform change in waking speed (Figure 3) because some children walked faster and others slower. Our measures of amplitude seemed more sensitive to changes in walking speed than changes in trunk control. This indicates that direct measures of amplitude may not be a reliable indicator of trunk control in situations where walking speed is not controlled.

We found a significant main effect of AFOs on the structure of the acceleration signal (ML and AP directions) that was independent of walking speed. Using AFOs increased structure complexity in trunk acceleration, seen as an increase in SE. The more irregular structure of the acceleration signal may indicate more frequent adjustments of trunk posture. Changes in structure regularity of the acceleration signal have been described as reflecting changes in gait stability.[23] A more regular structure (i.e. decrease in SE) has been related to an increased risk of falling[23,29,30] and to a reduced adaptive capacity to maintain balance in elderly patients.[12] In rehabilitation of stroke patients, a more irregular and complex structure was found with improved postural control.[31] Thus, the significant increase in SE (i.e. more irregular structure) found in ML and AP directions may indicate more adaptable adjustments of the trunk posture to external perturbation and improved postural control and stability when walking with AFOs. The more irregular structure in the AP direction may also indicate more pronounced foot-strike spikes when using AFOs. This might occur when an AFO corrects a flat-foot gait to initial contact with a heel strike.

Further studies should investigate in more detail the relationship between AFO-related changes in ankle and trunk kinematics, structure regularity of the trunk acceleration during walking, and reported incidence of falling.

Some considerations have to be made for this study. By using small portable accelerometers, participants were able to walk uninterrupted for 5 minutes. Such long data captures have been shown to be robust and do not rely on step detection, a potentially critical source of error in related measures.[23] This is a strength of our study. Another advantage of such sensors is that they are cheap, have long recording times, and can be used in real-life environments. Parameters used for the calculation of SE were based on previous studies in older adults. Further studies could both look into the effect of m parameter and also extend the concept of SE to multiscale SE, which to a greater extent will quantify the influence of different acceleration frequencies.

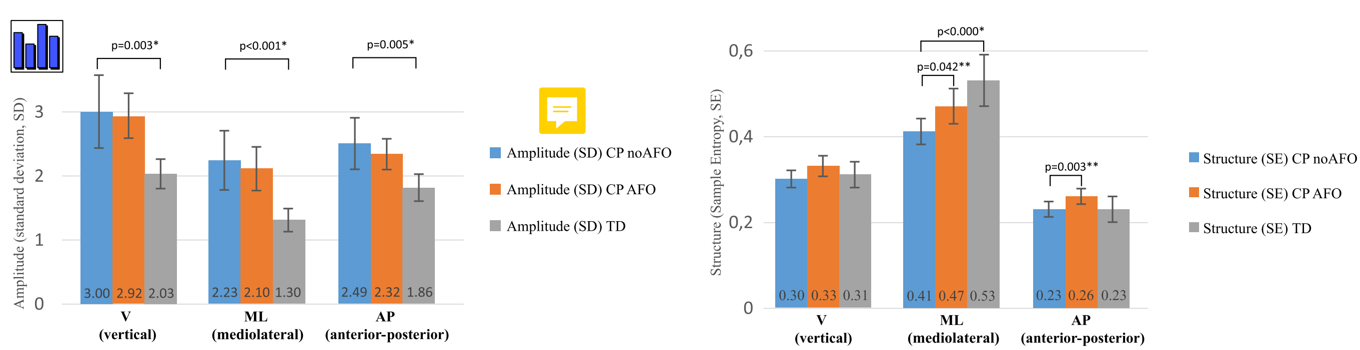

Figure 2. Left: Amplitude of trunk acceleration (SD) in vertical (V), mediolateral (ML), and anterior–posterior (AP) directions for condition ~~not~~AFO, condition AFO, and TD with 95% confidence interval. Right: Structure of trunk acceleration (Sample Entropy, SE) in V, ML, and AP directions for condition ~~not~~AFO, condition AFO, and TD with 95% confidence interval. *Significant p-values for Student's t-test that was used to test differences in mean between conditions AU8 ~~not~~AFO and TD. **Significant main effect of AFO from ANOVA test. AFO, ankle-foot orthosis; TD, typically developed.

AU11 **Table 2.** Statistical details for 2-way general linear model repeated measures ANOVA (without AFOs; with AFOs) on amplitude and structure of trunk acceleration with GMFCS as between-subjects factor and speed as covariate.

	Main effect of AFO		Interaction: effect of AFO and change in walking speed		Interaction: effect of AFO and GMFCS	
	$F_{(1,19)}$	*P*	$F_{(1,19)}$	*P*	$F_{(1,19)}$	*P*
V amplitude	8.559	0.010[a]	36.317	**0.000**	0.233	0.636
ML amplitude	6.721	0.010[a]	16.568	**0.002**	0.006	0.939
AP amplitude	10.346	0.010[a]	22.237	**0.000**	0.000	0.988
V structure	2.389	0.142	2.453	0.137	1.750	0.204
ML structure	4.871	**0.042**	3.218	0.092	1.311	0.269
AP structure	11.738	**0.003**	0.837	0.374	1.625	0.221

AU12

AFO, ankle-foot orthosis; AP, anterior–posterior; GMFCS, gross motor function classification system; ML, mediolateral; V, vertical.
[a]Main effect of AFO for V, ML, and AP amplitude cannot be interpreted because of the significant interaction with walking speed.

ECW was measured in a laboratory setting, reflecting capacity of walking, which is different from walking in the community or at home. However, walking with and without AFOs occurred in identical settings so we feel confident that measures of ECW were appropriate to investigate the effect of AFOs on gait economy.

The children in our study used AFOs of different designs, and the AFOs had been individually fitted to best address gait deviations in each patient. Although this reflects the variety of AFOs seen in clinical patients, it may also affect the results of both, measures for postural control and ECW. A larger study sample could have allowed an analysis on AFO type, which would have been beneficial.

We wanted to find out whether we could identify a general effect of using AFOs on measures reflecting postural control. Our results indicate an improvement in postural control measured in the domain of the structure of the acceleration signal in contrast to Meyns,[6] who found reduced margins of stability in the ML direction. AFOs with larger mechanical constraints might have a different effect on trunk acceleration, and compared with the Meyns[6] study, our participants had better gait function (fewer GMFCS II and no GMFCS III) and the AFOs used were apparently more flexible. In the current study, all children used flexible or semiflexible foot plates and 12 AFOs had free dorsiflexion compared with AFOs in the study by Meyns, which were designed with rigid foot plates and fairly large resistance to dorsiflexion.

~~The improvement in postural control because of AFOs found in our study was not reflected in a corresponding change in ECW, and it is not clear how much change in postural control is required to affect ECW.~~

The effect of AFOs on amplitude and structure of the acceleration signal was not affected by GMFCS level (Table 2). ~~This may indicate that the use of AFOs in children with greater impairment (GMFCS II vs. GMFCS I) did not cause increased deviations of trunk acceleration.~~ The sample size in the two subgroups was low (five and 14 children, respectively), decreasing the power of our analysis. Future studies with larger sample size are therefore needed to make such group comparisons.

The current study has documented that the effect of fitting AFOs goes beyond a mechanical control of the lower limbs. There are changes in structure of the trunk acceleration that occur when using AFOs, which may indicate an overall positive effect on postural control. Although ECW is often used as an overall measure of gait economy, the current study did not find an association with changes in postural control. This study supports further investigation into the analysis of the structure of trunk acceleration as a potentially sensitive measure to changes in postural control caused by the use of AFOs.

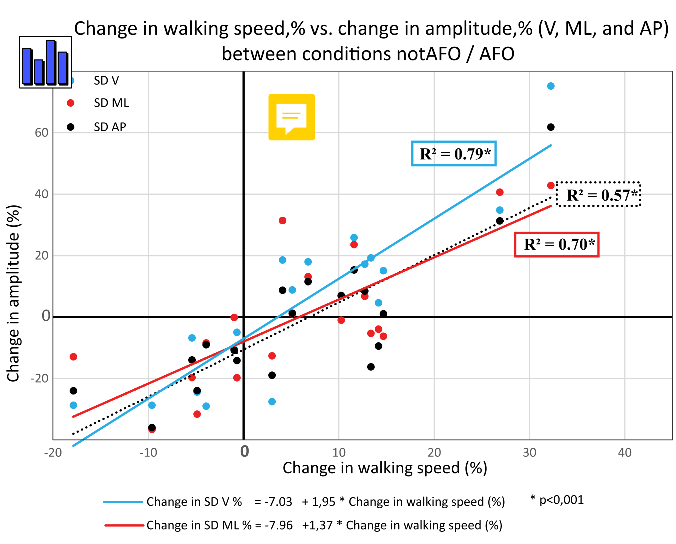

Figure 3. Simple linear regression analysis was performed to estimate the constant for the regression line of each amplitude measure because this shows how much the amplitude reduces (in %) in condition AFO compared ~~with notAFO~~ when there is no change in walking speed. Vertical = −7.03% (95% CI −13.81 to 0.26, *P* = 0.043); ML= −7.96% (95% CI −16.05 to 0.12, *P* = 0.053); and AP = −10.58% (95% CI −18.43 to −3.73, *P* = 0.005). R^2 shows how much of the variation in amplitude is explained by changes in walking speed: V = 79% ($P < 0.001$), ML = 70% ($P = 0.001$), and AP = 57% ($P \leq 0.001$). AFO, ankle-foot orthosis; AP, anterior–posterior, CI, confidence interval; ML, mediolateral.

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Declaration of conflicting interest

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Supplemental material

There is no supplemental material in this article.

AU6
References

1. Rosenbaum PL, Palisano RJ, Bartlett DJ, et al. Development of the Gross Motor Function Classification System for cerebral palsy. *Dev Med Child Neurol* 2008; 50: 249–253.
2. Heyrman L, Feys H, Molenaers G, et al. Three-dimensional head and trunk movement characteristics during gait in children with spastic diplegia. *Gait Posture* 2013; 38: 770–776.
3. Hadders-Algra M and Carlberg EB. *Postural Control: a Key Issue In Developmental Disorders. Clinics in Developmental Medicine*. vol. ix, Mac AU7 Keith; 2008: 331.
4. Boyer ER and Patterson A. Gait pathology subtypes are not associated with self-reported fall frequency in children with cerebral palsy. *Gait Posture* 2018; 63: 189–194.
5. Van de Walle P, Hallemans A, Truijen S, et al. Increased mechanical cost of walking in children with diplegia: the role of the passenger unit cannot be neglected. *Res Dev Disabil* 2012; 33: 1996–2003.
6. Meyns P, Kerkum YL, Brehm MA, et al. Ankle foot orthoses in cerebral palsy: effects of ankle stiffness on trunk kinematics, gait stability and energy cost of walking. *Eur J Paediatr Neurol* 2020; 26: 68–74.
7. Swinnen E, Baeyens JP, Van Mulders B, et al. The influence of the use of ankle-foot orthoses on thorax, spine, and pelvis kinematics during walking in children with cerebral palsy. *Prosthet Orthot Int* 2018; 42: 208–213.
8. Degelean M, De Borre L, Salvia P, et al. Effect of ankle-foot orthoses on trunk sway and lower limb intersegmental coordination in children with bilateral cerebral palsy. *J Pediatr Rehabil Med* 2012; 5: 171–179.
9. Chen X, Liao S, Cao S, et al. An acceleration-based Gait Assessment Method for children with cerebral palsy. *Sensors (Basel)* 2017; 17: 1002.
10. Iosa M, Marro T, Paolucci S, et al. Stability and harmony of gait in children with cerebral palsy. *Res Dev Disabil* 2012; 33: 129–135.
11. Saether R, Helbostad JL, Adde L, et al. Gait characteristics in children and adolescents with cerebral palsy assessed with a trunk-worn accelerometer. *Res Dev Disabil* 2014; 35: 1773–1781.
12. Soangra R and Lockhart TE. Inertial sensor-based variables are indicators of frailty and adverse post-operative outcomes in cardiovascular disease patients. *Sensors (Basel)* 2018; 18: 1792.
13. de Graaf-Peters VB, Blauw-Hospers CH, Dirks T, et al. Development of postural control in typically developing children and children with cerebral palsy: possibilities for intervention? *Neurosci Biobehav Rev* 2007; 31: 1191–1200.
14. Heyrman L, Feys H, Molenaers G, et al. Altered trunk movements during gait in children with spastic diplegia: compensatory or underlying trunk control deficit? *Res Dev Disabil* 2014; 35: 2044–2052.
15. Morris C, Bowers R, Ross K, et al. Orthotic management of cerebral palsy: recommendations from a consensus conference. *NeuroRehabilitation* 2011; 28: 37–46.
16. Rogozinski BM, Davids JR, Davis RB III, et al. The efficacy of the floor-reaction ankle-foot orthosis in children with cerebral palsy. *J Bone Joint Surg Am* 2009; 91: 2440–2447.
17. Davids JR, Rowan F and Davis RB. Indications for orthoses to improve gait in children with cerebral palsy. *J Am Acad Orthop Surg* 2007; 15: 178–188.
18. Rose J, Gamble JG, Burgos A, et al. Energy expenditure index of walking for normal children and for children with cerebral palsy. *Dev Med Child Neurol* 1990; 32: 333–340.
19. Aboutorabi A, Arazpour M, Ahmadi Bani M, et al. Efficacy of ankle foot orthoses types on walking in children with cerebral palsy: a systematic review. *Ann Phys Rehabil Med* 2017; 60: 393–402.
20. Brehm MA, Becher J and Harlaar J. Reproducibility evaluation of gross and net walking efficiency in children with cerebral palsy. *Dev Med Child Neurol* 2007; 49: 45–48.
21. Norman JF, Bossman S, Gardner P, et al. Comparison of the energy expenditure index and oxygen consumption index during self-paced walking in children with spastic diplegia cerebral palsy and children without physical disabilities. *Pediatr Phys Ther* 2004; 16: 206–211.
22. Iosa M, Fusco A, Morone G, et al. Assessment of upper-body dynamic stability during walking in patients with subacute stroke. *J Rehabil Res Dev* 2012; 49: 439–450.
23. Riva F, Toebes MJ, Pijnappels M, et al. Estimating fall risk with inertial sensors using gait stability measures that do not require step detection. *Gait Posture* 2013; 38: 170–174.
24. Richman JS and Moorman JR. Physiological time-series analysis using approximate entropy and sample entropy. *Am J Physiol Heart Circ Physiol* 2000; 278: H2039–H2049.
25. Rispens SM, Pijnappels M, van Schooten KS, et al. Consistency of gait characteristics as determined from acceleration data collected at different trunk locations. *Gait Posture* 2014; 40: 187–192.
26. ~~Hof A. Scaling gait data to body size. *Gait Posture* 1996; 4: 222–223.~~
27. Kavanagh JJ and Menz HB. Accelerometry: a technique for quantifying movement patterns during walking. *Gait Posture* 2008; 28: 1–15.
28. Moe-Nilssen R. A new method for evaluating motor control in gait under real-life environmental conditions. Part 2: gait analysis. *Clin Biomech (Bristol, Avon)* 1998; 13: 328–335.
29. Karmakar CK, Khandoker AH, Begg RK, et al. Understanding ageing effects by approximate entropy analysis of gait variability. *Annu Int Conf IEEE Eng Med Biol Soc* 2007; 2007: 1965–1968.
30. Ihlen EAF, Weiss A, Bourke A, et al. The complexity of daily life walking in older adult community-dwelling fallers and non-fallers. *J Biomech* 2016; 49: 1420–1428.
31. Roerdink M, De Haart M, Daffertshofer A, et al. Dynamical structure of center-of-pressure trajectories in patients recovering from stroke. *Exp Brain Res* 2006; 174: 256–269.