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Ankle-foot orthoses in stroke: Effects on functional balance, weight-bearing asymmetry and the contribution of each lower limb to balance control

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ABSTRACT

Background: Ankle-foot orthoses are often provided to improve walking in stroke patients, although the evidence of effects on walking and balance control is still inconsistent. This could be caused by a lack of insight into the influence of orthoses on the underlying impairments. These impairments can be assessed with dual plate posturography to determine the relative contribution of each lower limb to balance control and weight-bearing. This study examined the effects of ankle-foot orthoses on functional balance, static and dynamic weight-bearing asymmetry and dynamic balance control of the paretic and non-paretic lower limbs.

Methods: Twenty stroke subjects (time since stroke 5–127 months) completed the study. Subjects were assessed with and without ankle-foot orthosis. Functional balance was assessed using the Berg Balance Scale, Timed Up & Go test, Timed Balance Test, 10-m walking test and Functional Ambulation Categories. Weight-bearing asymmetry and dynamic balance control were assessed with force plates on a movable platform.

Findings: No significant effects of ankle-foot orthoses were found for weight-bearing asymmetry and dynamic balance control, but significant differences in favour of ankle-foot orthosis use were found for most functional tests.

Interpretation: Although ankle-foot orthoses had no effect on weight-bearing asymmetry or dynamic balance contribution of the paretic lower limb, functional tests were performed significantly better with orthoses. Apparently, improvements at functional level cannot be readily attributed to a greater contribution of the paretic lower limb to weight-bearing or balance control. This finding suggests that ankle-foot orthoses influence compensatory mechanisms.

1. Introduction

Impaired balance control is a major problem for many patients who have sustained a stroke, as postural imbalance greatly restricts activities of daily life (ADL) and gait (Geurts et al., 2005). In clinical practice, ankle-foot orthoses (AFOs) are often used to restore a more normal and safe walking pattern in people with stroke (Geboers et al., 2002). According to Leung and Mosely (2003), AFOs can provide lateral stability to the ankle in stance phase, facilitate toe clearance in swing phase and promote heel strike. However, the literature is generally inconclusive about the effects of AFOs on gait and balance control.

Positive effects of AFOs have been found for various gait parameters (de Wit et al., 2004; Lehmann et al., 1987; Leung and Mosely, 2003; Tyson and Thornton, 2001; Wang et al., 2005, 2007), with improvement in walking speed being reported most frequently. There have also been reports of improvements in functional tests like the Functional Ambulation Categories (Tyson and Thornton, 2001), the Timed Up & Go test and the stairs test (de Wit et al., 2004) when using an AFO. However, other studies did not find beneficial effects of AFOs on gait parameters (Churchill et al., 2003; Gök et al., 2003) or functional tests (Wang et al., 2005). The contradictory results regarding the effect of AFOs might be explained by differences in subject inclusion, as the stroke subjects included show a wide variety in the time since stroke, stroke severity and AFO types used. Furthermore, study designs differed in that different AFO designs were compared, with each other (Gök et al., 2003; Lehmann et al., 1987) as well as with no AFO (de Wit et al., 2004).

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The studies all evaluated the effects of AFOs at a functional level. The use of functional scales means that factors other than the purely mechanical effects of an AFO, such as adaptive strategies and balance confidence, can influence the performance. To isolate the effects of AFOs on the mechanisms underlying balance control, we need to assess their effects at the impairment level. Weight-bearing asymmetry is often used as a measure of impairment in balance control, and restoring symmetry in weight-bearing is traditionally considered an important goal of rehabilitation (Geurts et al., 2005; Marigold and Eng, 2006). Recent studies showed that weight-bearing asymmetry is not a mere reflection of the asymmetry in the contribution of each individual leg to balance control (de Haart et al., 2004; Genthon et al., 2008; Marigold and Eng, 2006; van Asseldonk et al., 2006). Van Asseldonk et al. (2006) introduced a method using dual plate posturography to assess the relative contribution of each individual lower limb to balance control. They found that the contribution of the paretic lower limb to dynamic balance in stroke patients was significantly smaller than the contribution of the same limb to weight-bearing, and did not find a clear relation with weight-bearing asymmetry. In contrast, the contribution to balance control and weight-bearing were closely related in healthy subjects.

The present study applied this posturographic method to examine the effects of AFOs on balance control and weight-bearing asymmetry in patients with stroke. The assessments were used to study the effects at the impairment level. The effects of AFOs at a functional level were interpreted by including functional tests. We hypothesized that an AFO would provide stability in stance in stroke patients (Leung and Mosely, 2003; Wang et al., 2005). Hence, weight-bearing and relative contributions to balance were expected to be distributed more evenly between the paretic and non-paretic lower limb when an AFO was used. The use of an AFO was expected to result in higher scores on the functional tests.

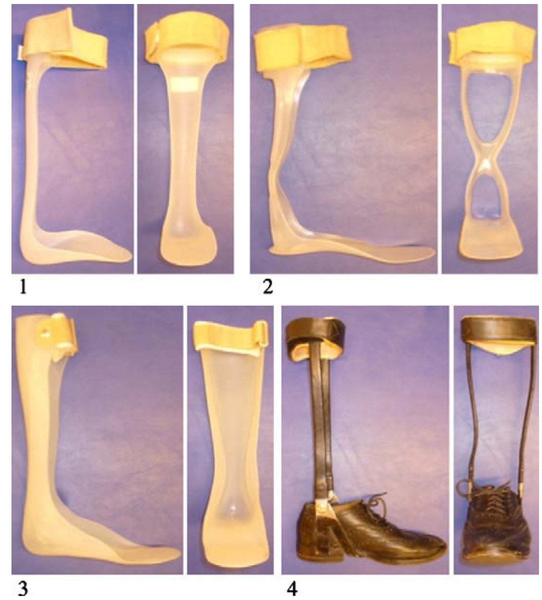

Fig. 1. Types of AFO. Types 1–3: polypropylene, non-articulated AFOs with (1) small posterior steel; (2) two crossed posterior steels and an open heel; (3) large posterior steel. Type (4): articulated metal AFO with double bars attached to the outsole of a normal shoe.

2. Methods

2.1. Participants

Subjects were recruited from a rehabilitation centre and a local hospital in the Netherlands. Participants were subjects with a single and first-ever unilateral ischemic or hemorrhagic supratentorial stroke leading to hemiparesis. They had to be at least 18 years of age and 3 months post stroke. In addition, they had to have used one of the preselected types of AFO regularly (daily) for at least 2 months before the start of the study. Other inclusion criteria were: being able to maintain independent unsupported stance with and without an AFO for at least 90 s, being able to walk for 10 m with or without an assistive device, and being able to follow simple verbal instructions. Subjects suffering from severe comprehensive aphasia or neglect and subjects using medication or with non-stroke related disorders that could affect balance were excluded. The Rivermead Mobility Index (Collen et al., 1991) and Motricity Index (Collin and Wade, 1990) were scored to characterize the subject population. Gender, age, time since stroke, affected hemisphere (left or right), type of stroke (ischemic or hemorrhagic) and the time during which an AFO had been worn were recorded. The study was approved by the local medical ethical committee. All subjects gave their written informed consent.

Since many different types of AFO can be prescribed to stroke patients, various types of AFO were included in this study (see Fig. 1). All subjects were using one of the following preselected types of AFO: (1) polypropylene, non-articulated AFO with a small posterior steel (Dynafo, Ortho-Medico, Herzele, Belgium); (2) polypropylene, non-articulated AFO with two crossed posterior steels and an open heel (Ottobock, Otto Bock Benelux bv, Son en Breugel, The Netherlands); (3) polypropylene, non-articulated AFO with a large posterior steel (Camp, Basko Healthcare, Zaandam, The Netherlands), all three worn inside a normal shoe; and (4) articulated, metal AFO with double bars attached to the outsole of a normal shoe (custom-made, components from Otto Bock Benelux bv, Son en Breugel, The Netherlands). We included equal numbers of AFOs of the more flexible type (1 and 2) and the more rigid type (3 and 4).

2.2. Experimental design

The effects of AFOs on balance control and functional balance were measured using two types of assessment: posturographic tests and functional tests. Both types of assessment were performed with and without subjects wearing an AFO on the paretic side, while all subjects were wearing their own shoes. The order of testing with and without AFO was randomized in a cross-over design. This order was used for both tests, which were performed on separate days within one week.

2.2.1. Posturographic tests

2.2.1.1. Procedure. The posturographic tests were performed on a movable platform. The subjects stood on a dual force plate embedded in a computer-controlled 6-degrees-of-freedom motion platform (Caren, MOTEK, Amsterdam, The Netherlands). Subjects stood upright, with arms folded in front of their chest or hanging by their sides, with both feet slightly apart, such that the distance between the medial sides of the heels was approximately 20 cm. This specific posture was maintained during all of the posturographic tests. Subjects faced a light grey background. A safety harness fixed to the ceiling prevented falls, without restricting movements necessary to maintain balance. The custom-made dual force plate incorporated four 6-degrees of freedom force sensors (ATI-Mini45-SI-580-20, Schunk, 's-Hertogenbosch, The Netherlands) in a 2×2 configuration (two sensors for each force plate). The experimenter ensured that the left foot was placed over the two left (forefoot and hindfoot) sensors, and that the other foot was placed over the two right (forefoot and hindfoot) sensors. The forces recorded by each sensor were sampled at a frequency of 360 Hz. Kinematics of the body were measured at a sampling rate of 120 Hz with a 3-dimensional passive recording system, consisting of six infrared cameras and a control unit (Vicon, Oxford, United Kingdom). The system recorded the position of 27 passive reflective markers, attached to the heel, big toe, lateral malleolus, tibia, knee and femur of both legs, as well as to the sacrum, top of the sternum, vertebra C8, head and both shoulders. A cluster of three markers was attached to the anterior superior iliac spine on both sides, and three markers were attached to the platform to reconstruct platform displacement. Body weight and height of each subject were measured.

Two different posturographic tests were performed on the platform: static and dynamic. The static trial was used to asses weight-bearing asymmetry, while the subjects were instructed to stand quietly on the dual force plate for 90 s. Subsequently, balance was perturbed to accurately estimate the relative contributions of the individual lower limbs to balance control (van Asseldonk et al., 2006; van der Kooij et al., 2005). Subjects performed three dynamic trials lasting 90 s, in which they were instructed to maintain balance without moving their feet. Balance was perturbed by continuous platform movements in the forward–backward direction. The random platform movements were produces by a multisine signal which was unpredictable and therefore prevented anticipation of the disturbance. The multisine had a period of 34.13 s. The 90 s perturbation trial consisted of approximately 2.5 multisine cycles, which were repeated time after time. Two synchronized complete cycles were extracted from each perturbation trial, while the rest of the trial was discarded. The results of the six (two cycles $\times$ three perturbation trials) cycles were averaged. The perturbation signal included 112 sinusoidal signals with frequencies in the range of 0.06–4.25 Hz and with mutual random phase shifts (van der Kooij et al., 2007). The zero-to-peak amplitude of the platform movement was 5 cm. Before the start of the assessment, this perturbation was presented to the subjects to familiarize them with the dynamic condition. For some subjects, these 5 cm perturbations were too large to withstand for multiple 90-s trials. In these cases, the supervising physical therapist decided to reduce the amplitude to 4 or 3 cm (see Table 1 for an overview of the perturbation amplitudes used). Subjects were allowed to rest between the trials to prevent fatigue. If a subject had to make a step to prevent a fall, the trial was repeated.

2.2.2. Data processing and analysis

2.2.2.1. Static. Static weight-bearing (SW) asymmetry between the paretic and non-paretic sides was calculated by dividing the average vertical force below the paretic and non-paretic feet, respectively, by the average sum of the vertical forces below both feet during the 90 s. Consequently, the fractions of the paretic and non-paretic sides (SW paretic and SW non-paretic) add up to 100%.

2.2.2.2. Dynamic. Calculation of the dynamic weight-bearing (DW) asymmetry during the perturbations was similar to that for the static trial. The results of the three dynamic trials were averaged. The results of the dynamic trials were also used to calculate the dynamic balance contribution (DBC) of each lower limb (DBC paretic and non-paretic). The DBC expressed the relative contribution of each individual lower limb to body stabilization. The body sway evoked by the perturbations was defined as the angle in the sagittal plane between a "virtual" pendulum connecting the ankle and the Centre of Mass (CoM) and the vertical. This vertical reflects upright stance. To determine the relative contribution of each individual lower limb, body sway needed to be related to the torques generated about each ankle. For this purpose, balance was perturbed and closed-loop system identification techniques were applied (joint input–output approach; van der Kooij et al., 2005). The use of closed-loop system identification techniques allowed the sway to be specifically related to the torques that were generated to counteract the sway.

The data analysis used to calculate the DBC was previously described by van Asseldonk et al. (2006). Fig. 2 shows a schematic representation of the data analysis. First, the body sway was reconstructed from the recorded marker positions (Fig. 2(1)). The individual rotations and positions of the body segments were calculated and subsequently used to determine the CoM as the weight sum of individual segment positions (Koopman et al.,

Table 1
Subject characteristics.

Subject	Sex	Age (y)	Time since stroke (months)	Time wearing AFO (months)	Affected hemisphere	Stroke type	Type of AFO[a]	RMI[b]	MI[c] total	MI[c] leg (hip/knee/ankle)	Disturbance signal platform (cm)[d]
1	M	56	9	3	Right	Ischemic	3	13	14	14 (0/14/0)	5
2	F	61	12	10	Left	Hemorrhagic	4	12	159	75 (25/25/25)	5
3	M	42	6	3	Right	Ischemic	2	13	64	50 (0/25/25)	5
4	M	57	23	21	Right	Ischemic	2	12	53	39 (0/14/25)	3
5	M	63	16	13	Right	Ischemic	1	14	156	91(25/33/33)	5
6	M	65	70	68	Right	Ischemic	3	12	64	39 (0/25/14)	3
7	M	44	55	49	Left	Hemorrhagic	3	14	67	39 (0/14/25)	5
8	M	71	93	87	Left	Ischemic	1	10	28	28 (0/14/14)	5
9	F	63	127	123	Left	Ischemic	1	13	129	64 (14/25/25)	5
10	M	36	43	24	Left	Ischemic	4	14	64	50 (0/25/25)	5
11	M	58	24	18	Right	Ischemic	4	11	70	42 (14/14/14)	4
12	F	53	11	8	Left	Ischemic	2	12	28	28 (0/14/14)	4
13	M	49	18	13	Right	Ischemic	4	13	37	28 (0/14/14)	3
14	M	54	64	60	Right	Ischemic	2	11	73	50 (0/25/25)	3
15	M	58	5	2	Right	Ischemic	3	14	90	56 (9/33/14)	5
16	M	78	86	84	Left	Ischemic	1	12	116	66 (0/33/33)	3
17	F	51	94	90	Left	Hemorrhagic	3	12	96	37 (9/14/14)	4
18	F	71	8	2	Left	Ischemic	3	13	93	39 (0/14/25)	4
19	F	52	17	12	Left	Ischemic	2	13	103	53 (14/25/14)	3
20	M	61	5	3	Right	Ischemic	1	14	129	75 (25/25/25)	5
Mean		57.2	39.3	34.7				12.6	81.7	48.2 (6.8/21.3/20.2)	

[a] Type of AFO: (1) polypropylene, non-articulated AFO with a small posterior steel; (2) polypropylene, non-articulated AFO with two crossed posterior steels and an open heel; (3) polypropylene, non-articulated AFO with a large posterior steel; (4) articulated, metal AFO with double bars.
[b] RMI: Rivermead Mobility Index.
[c] MI: Motricity Index, total score (affected arm and leg) and leg score (hip/knee/ankle).
[d] Maximum platform amplitude of the multisine disturbance signal.

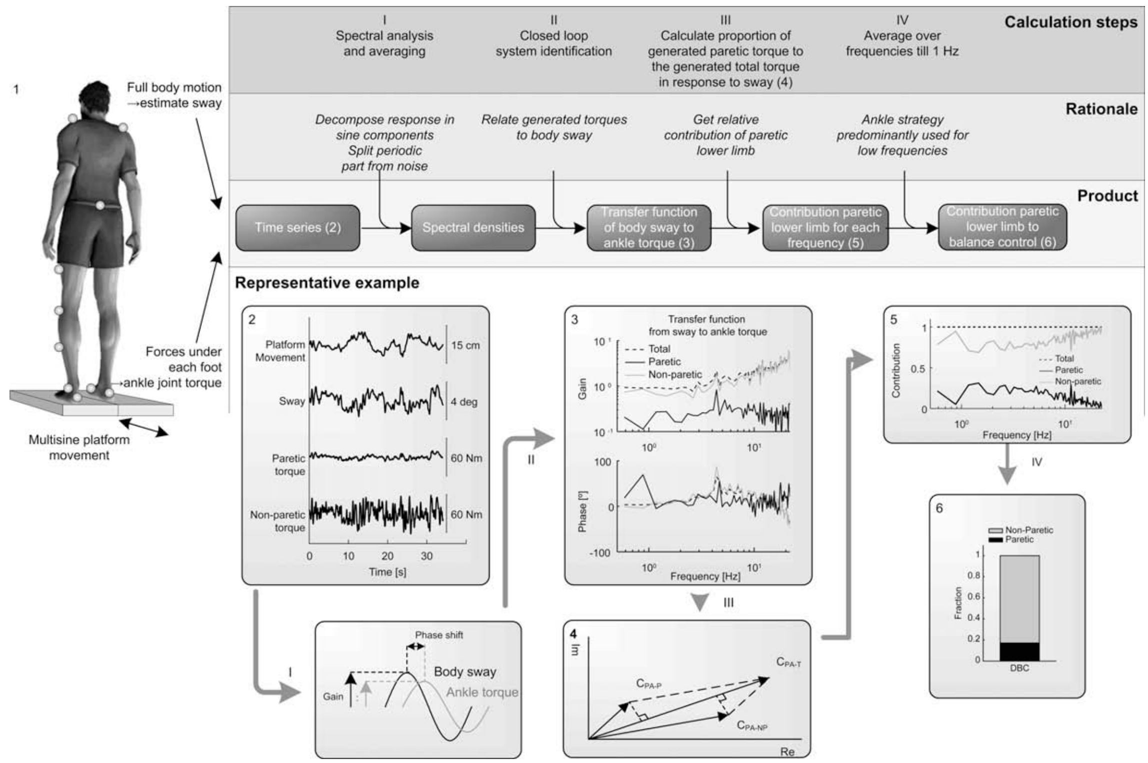

Fig. 2. Schematic representation of the different steps in the data analysis to estimate the dynamic balance contributions of the paretic and non-paretic lower limbs. The calculation steps are indicated at the top, with the rationale for these steps and their intermediate products. The intermediate and final results of the data processing of a typical subject are depicted at the bottom. (1) Subjects stood on a motion platform that moved in forward–backward directions. The platform movements consisted of a multisine with different frequencies and randomized phases. Ground reaction forces under each foot were measured by four force sensors (two under each foot). Full body motion was captured with a passive registration system. (2) Motion of Centre of Mass was reconstructed from the reflective markers using an anatomical model (Koopman et al., 1995). Ankle torques of each foot were calculated with inverse dynamics. (3) The transfer function from body sway to corrective torque was estimated by dividing the cross-spectral density of platform movement and corrective ankle torque by the cross-spectral density of platform movement and body sway, for the frequencies generated. The gain of this transfer function indicates how much corrective torque was generated in response to body sway. When corrective torque preceded body sway, the function had a positive phase, and a negative phase when corrective torques lagged behind the body sway. Both gain and phase were functions of frequency. (4) The gain and phase of the transfer function for a specific frequency can also be represented in the complex plane, in which the length of the vector was equal to the gain while the angle with the real axis equalled the phase of the transfer function. The relative contribution of one lower limb is defined as the projection of the transfer function vector of this individual lower limb on the transfer function vector of both lower limbs together. This method was used to determine the relative contribution for each frequency generated. (5) Relative contributions of both lower limbs as a function of the frequencies generated. (6) Relative contributions of both lower limbs were averaged over the frequencies generated (<1 Hz), resulting in the dynamic balance contributions of the paretic and non-paretic lower limbs. More details can be found in Van Asseldonk et al. (2006).

1995). The forces and torques measured by the sensors below each foot were used to calculate the vertical and horizontal forces below each foot and their point of application, the Centre of Pressure (CoP). The torques about the paretic and non-paretic ankles in the sagittal plane were calculated using inverse dynamics (Koopman et al., 1995). Spectral analysis (Fig. 2, calculation step 1) was then performed to decompose the time series (Fig. 2(2)) into sine signals with frequencies of the perturbation signal. Second, closed-loop system identification was used to determine the transfer functions (Fig. 2(3)) of the sway to the torques generated (van der Kooij et al., 2005). These transfer functions indicated the corrective torque generated about each ankle in response to a deviation from erect stance (body sway) for each frequency in the perturbation signal. Third, the relative contribution of a lower limb was calculated by dividing the transfer function of this individual lower limb by the transfer function of both lower limbs (Fig. 2(5)). Finally, the DBC was calculated by averaging this ratio for all frequencies below 1 Hz (Fig. 2(6)). Only these lower frequencies were taken into account, because the present study focused on the contribution of the ankle, and the ankle is mostly used to counteract low frequency perturbations (<1 Hz) (Creath et al., 2005).

2.2.3. Functional tests

Several tests were performed to assess functional balance. Again, all tests were performed with and without subjects using an AFO on the paretic side. The Berg Balance Scale (BBS) measured balance during 14 balance tasks (Berg et al., 1989). Each task was graded from 0 to 4, with higher scores reflecting higher speed, more stability or less assistance required to complete the task. The Timed Up & Go test (TUG) assessed functional mobility (Podsiadlo and Richardson, 1991) by measuring the time needed to stand up from a chair, walk 3 m, turn around, walk back and sit down again. The Timed Balance Test (TBT) evaluated balance during five different positions of bilateral stance, which had to be maintained for one minute each (Bohannon et al., 1984). One point was scored for each position maintained. In the 10-m walking test (10MWT) the comfortable walking speed was measured over a distance of 10 m (Wade, 1992). Walking speed was calculated by averaging the results of three performances. The Functional Ambulation Categories (FAC) evaluated the level of independence during walking (Holden et al., 1984). Performance was graded from 0 (requiring support from two people) to 5 (able to walk indoors and outdoors without supervision).

2.3. Statistical analysis

Descriptive statistics were used to characterize the subject population. Differences in SW, DW, DBC and functional tests between measurements with and without AFO were analysed using the Wilcoxon signed ranks test. The level of significance for all analyses

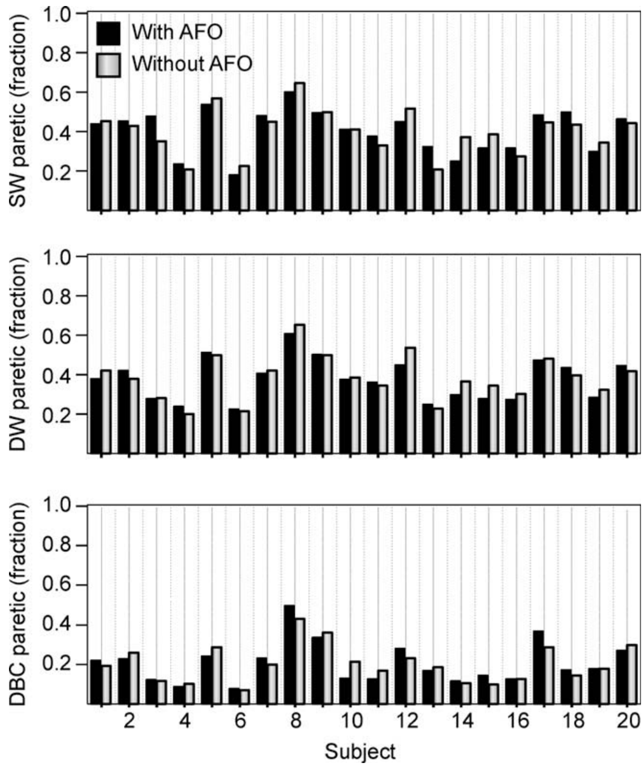

Fig. 3. Static and dynamic weight-bearing asymmetry (SW and DW, respectively) and dynamic balance contribution (DBC) of the paretic lower limb for each individual subject. Each clustered bar represents a subject ($N = 20$), with and without AFO. Only the results of the paretic lower limb are shown, presented as fractions. Consequently, the results of the non-paretic lower limb add up to 1.

was set at $P < .05$. The statistical analysis was conducted with SPSS for Windows.

3. Results

Twenty-six subjects were assessed for eligibility, 23 of whom met the inclusion criteria. Three subjects dropped out: one had an epileptic insult between the two measurement days, one had the sensation that he might suffer an epileptic insult prior to the measurements and one was unable to withstand the platform perturbations during the posturographic tests. Table 1 summarizes the characteristics of the 20 subjects who completed the study.

3.1. Posturographic tests

Fig. 2(2) shows the raw data of a representative subject standing on the dual force plate during a dynamic trial. The topmost line in this figure represents the platform movements, while the second line depicts the body sway. Inertia of the body mass resulted in body sway in the opposite direction to the platform movements. The remaining lines show the ankle torques generated to counteract the body sway. The fluctuations in the ankle torque were higher in the non-paretic lower limb than in the paretic limb. Processing the raw data of Fig. 2(2) resulted in the contributions of the paretic and non-paretic lower limbs to balance, presented as fractions (Fig. 2(6)). The individual values of all 20 subjects for the SW, DW and DBC of the paretic lower limb with and without AFO are presented in Fig. 3.

Table 2 presents the mean weight-bearing of the paretic lower limb for the static and dynamic trials. During quiet stance, mean paretic SW was 40.5% (SD 11.2) with AFO and 40.0% (SD 11.6) without AFO. During the forward–backward perturbations, mean paretic DW values were 37.6% (SD 10.6) and 38.6% (SD 11.4) with and without AFO, respectively. There was no significant effect of using an AFO ($P = .991$ and $P = .380$ for SW and DW, respectively). The mean paretic DBC was 20.1% (SD 9.5) with AFO vs. 20.5% (SD 10.6) without AFO (Table 2). Again there was no significant difference with and without AFO use ($P = .702$).

3.2. Functional tests

Table 3 shows the results of the functional tests, with and without using an AFO. The mean BBS score was significantly higher with than without AFO (48.0, SD 4.8 vs. 46.2, SD 5.5, $P = .001$). The mean time needed to perform the TUG improved by 5.7 s from 29.1 s (SD 12.9) without AFO to 23.4 s (SD 9.7) with AFO ($P < .001$). The mean walking speed improved by 0.12 m/s (0.58 m/s with AFO (SD 0.24) vs. 0.46 m/s without AFO (SD 0.21), $P < .001$) and the mean FAC score also improved significantly ($P = .001$). The TBT (4.0 with AFO (SD 1.0) vs. 3.5 without AFO (SD 1.0)) showed no statistically significant difference in favour of AFO use ($P = .051$).

4. Discussion

The present study found that clear functional improvements were obtained by AFO use in stroke subjects, whereas the same subjects showed no effects of AFO use on weight-bearing asymmetry during unperturbed and perturbed stance, or on the contribution of the paretic lower limb to the balance control during perturbed stance.

The review by Leung and Mosely (2003) reported favourable effects of AFOs on walking speed and gait pattern. These authors stated that the significance of these changes for daily functioning remained largely unresolved. The current study found functional improvements in terms of BBS, TUG, 10MWT and FAC. The TBT was the only test not showing a significant difference with and without AFO use, although the difference was nearly significant ($P = .051$). This lack of significance may be due to two subjects using a rigid polypropylene AFO, both of whom had a lower TBT score with than without AFO. The stiffness of the AFO, limiting dorsiflexion in stance, restricted them in performing some of the items of the TBT.

Table 2
Static and dynamic weight-bearing asymmetry and dynamic balance contribution of the paretic lower limb with and without AFO.

	SW (paretic) (fraction)		DW (paretic) (fraction)		DBC (paretic) (fraction)	
	With AFO	Without AFO	With AFO	Without AFO	With AFO	Without AFO
Mean (SD)	0.405 (0.112)	0.400 (0.116)	0.376 (0.106)	0.386 (0.114)	0.201 (0.095)	0.205 (0.106)
Difference with and without AFO (SD)	0.005 (0.062)		0.010 (0.038)		0.004 (0.041)	
P-value Wilcoxon*	0.911		0.380		0.702	

Abbreviations: SD: Standard Deviation; SW: static weight-bearing; DW: dynamic weight-bearing; DBC: dynamic balance contribution. The values represent the contribution of the paretic lower limb (in fractions).
* *P*-value with Wilcoxon signed ranks test for standing with vs. without AFO (2-tailed).

Table 3
Functional tests with and without AFO.

	BBS		TUG (s)		10MWT (m/s)		FAC		TBT	
	With AFO	Without AFO	With AFO	Without AFO	With AFO	Without AFO	With AFO	Without AFO	With AFO	Without AFO
Mean (SD)	48.1 (4.8)	46.2 (5.5)	23.4 (9.7)	29.1 (12.9)	0.58 (0.24)	0.46 (0.21)	4.7 (0.5)	4.0 (0.6)	4.0 (1.0)	3.5 (1.0)
Minimum score	40	37	12	13.3	0.22	0.18	4	3	3	2
Maximum score	55	54	46.3	56.3	1.16	1.06	5	5	5	5
P-value Wilcoxon*	0.001		0.000		0.000		0.001		0.051	

Abbreviations: SD: Standard Deviation; BBS: Berg Balance Scale (max. 56 points); TUG: Timed Up & Go test; 10MWT: 10-m Walking Test; FAC: Functional Ambulation Categories (max. 5 points); TBT: Timed Balance Test (max. 5 points).
* *P*-value with Wilcoxon signed ranks test for performance with vs. without AFO (2-tailed).

In line with our results, various other studies have reported favourable effects of AFOs on the FAC (Tyson and Thornton, 2001), TUG (de Wit et al., 2004) and walking speed (de Wit et al., 2004; Lehmann et al., 1987; Tyson and Thornton, 2001; Wang et al., 2007). Wang et al. (2005) did not find any effects of AFOs on the BBS, which may be explained by ceiling effects. Their mean BBS score without AFO was 51 out of 56 points, compared to 46.2 points in our study. As ceiling effects for the BBS are well-known (Mao et al., 2002), subjects with high functional levels might benefit less from using an AFO than subjects with lower functional levels.

Although AFO use was generally beneficial for functional balance performance in our study, no improvements were found in SW, DW or DBC. Apparently, improvements at a functional level cannot be readily attributed to a greater contribution by the paretic lower limb to weight-bearing or dynamic balance control. When subjects were not using an AFO, the measured torques reflected only the internally generated torques. However, when subjects were using an AFO, the torques reflected the net results of the internally generated ankle torques and the externally generated AFO torques. Only the net result was measured, as the contribution of each of these components cannot be distinguished in the measurement we used. Although one might expect that wearing an AFO would lead to a greater contribution of the paretic lower limb to body stabilization, this presumed greater contribution was not found in our study, as the net torques did not change when subjects used an AFO. This can be explained by less internally corrective ankle torque being generated when using an AFO, because subjects passively rely on the AFO. As a passive structure, the AFO generates corrective torques in response to body sway. As the net ankle torque hardly changed when subjects were using an AFO, this induced less corrective ankle torque when using an AFO, and thus a decreased active contribution of the ankle. Another explanation might be that active ankle torques did increase during the perturbations, but were counteracted by the AFO, resulting in an unchanged net torque, as the corrective torques of the AFO can have a different direction and timing to the torques generated by the ankle. Future research could use electromyography of muscles in the lower limb to further determine the internal and external components of the total ankle torque.

In clinical practice, many different types of AFO can be prescribed to stroke patients. To reflect this large variability, various types of AFO were included in this study. Using "type of AFO" as one of the inclusion criteria led to the inclusion of a variety of stroke subjects. The included AFOs were categorized as "flexible" or more "rigid" AFOs, and these two categories were included in equal numbers. Rigid AFOs are usually prescribed to more severely affected patients, as they are meant to provide a relatively high degree of stability. These rigid types are typically used to counteract hyperextension of the knee in stance, or to correct varus deformity at the ankle in stance and/or swing, whereas the flexible types are primarily used to correct dropfoot and small varus deformities at the ankle during swing. Although this was not a primary research question, an additional analysis was performed to study whether the various types of AFO had influenced the results. These additional separate analyses for the flexible and rigid AFO types revealed that "type of AFO" had no influence at all on the outcomes of the study.

Overall, the findings of this study are in line with recent findings about balance and gait recovery after stroke, emphasizing the important role of various compensatory mechanisms for functional recovery (de Haart et al., 2004; Garland et al., 2003; Kwakkel et al., 2004). For instance, compensation through the non-paretic lower limb and through adapted trunk control are considered important substitution mechanisms. In addition, improved sensory control and related improvement of automaticity and confidence levels with regard to balance and gait may very well contribute to functional recovery (de Wit et al., 2004; Geurts et al., 2005). Further research should therefore focus on the possible "non-mechanical" effects of AFOs on balance in patients with stroke, for instance by improving confidence in the passive stability of the paretic lower limb.

4.1. Study limitations

The posturographic tests used in this study strongly suggest that AFO use has no direct beneficial effect on the loading and control characteristics of the paretic lower limb in stroke patients. It must be acknowledged, however, that we only considered balance control during forward–backward perturbations of stance. Possible beneficial effects with regard to ankle control and stability during sideward perturbations and during gait need further study. The model used in this study assessed only the paretic and the non-paretic ankle to determine the contribution of the paretic and the non-paretic lower limb to dynamic balance. Since stroke subjects may adapt the control of their paretic knee and hip when wearing an AFO, future research should incorporate proximal joints in the model as well. Adding electromyography of the lower limb muscles would allow the internal and external corrective torques to be studied in more detail. Finally, the current study used a cross-over design, so future research may benefit from a longitudinal design to study the effects of AFOs over time (Buurke et al., 2008; Kwakkel et al., 2004; Tyson and Kent, 2009).

5. Conclusions

Wearing an ankle-foot orthosis had no significant effect on weight-bearing asymmetry during unperturbed and perturbed stance, nor on the contribution of the paretic lower limb to dynamic balance control in stroke patients. Nevertheless, functional tests were performed significantly better using an AFO. Apparently, improvements at a functional level cannot be readily attributed to a greater contribution of the paretic lower limb to weight-bearing or balance control. These findings suggest that AFOs influence various compensatory mechanisms.

Suppliers

Basko Healthcare (Camp AFO), P.O. Box 2194, 1500 GD Zaandam, The Netherlands.

MOTEK Motion Technology (Caren motion platform), Inc. Nieuwe Hemweg 6A, 1013 BG, Amsterdam, The Netherlands.

Ortho-MEDICO (Dynafo AFO), Mutsaardstraat 47, B-9550 Herzele, Belgium.

Otto Bock Benelux B.V. (Ottobock AFO, and custom-made, articulated metal AFO with double bars using components from Otto Bock Benelux B.V.), P.O. Box 133, 5690 AC Son en Breugel, The Netherlands.

SCHUNK Intec B.V. (force sensors ATI-Mini45-SI-580-20), Speldenmakerstraat 3d, 5232 BH 's-Hertogenbosch, The Netherlands.

Vicon, 14 Minns Business Park, West Way, Oxford OX2 0JB, United Kingdom.

Conflicts of interest

The authors have no conflicts of interest related to the work presented in this manuscript. The funding resource had no involvement in the study design, the collection, analysis or interpretation of the data, nor in the writing or decision to submit the manuscript.

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References

Berg, K., Wood-Dauphinee, S., Williams, J.I., Gayton, D., 1989. Measuring balance in the elderly: preliminary development of an instrument. Physiother. Can. 41, 304–311.
Bohannon, R.W., Larkin, P.A., Cook, A.C., Gear, J., Singer, J., 1984. Decrease in timed balance test scores with aging. Phys. Ther. 64, 1067–1070.
Buurke, J.H., Nene, A.V., Kwakkel, G., Erren-Wolters, V., Ijzermand, M.J., Hermens, H.J., 2008. Recovery of gait after stroke: what changes? Neurorehabil. Neural Repair 22, 676–683.
Churchill, A.J., Halligan, P.W., Wade, D.T., 2003. Relative contribution of footwear to the efficacy of ankle-foot orthoses. Clin. Rehabil. 17, 553–557.
Collen, F.M., Wade, D.T., Robb, G.F., Bradshaw, C.M., 1991. The rivermead mobility index: a further development of the rivermead motor assessment. Int. Disabil. Stud. 13, 50–54.
Collin, C., Wade, D., 1990. Assessing motor impairment after stroke: a pilot reliability study. J. Neurol. Neurosurg. Psychiatry 53, 576–579.
Creath, R., Kiemel, T., Horak, F., Peterka, R., Jeka, J., 2005. A unified view of quiet and perturbed stance: simultaneous co-existing excitable modes. Neurosci. Lett. 377, 75–80.
de Haart, M., Geurts, A.C., Huidekoper, S.C., Fasotti, L., van Limbeek, J., 2004. Recovery of standing balance in postacute stroke patients: a rehabilitation cohort study. Arch. Phys. Med. Rehabil. 85, 886–895.
de Wit, D.C., Buurke, J.H., Nijlant, J.M., IJzerman, M.J., Hermens, H.J., 2004. The effect of an ankle-foot orthosis on walking ability in chronic stroke patients: a randomized controlled trial. Clin. Rehabil. 18, 550–557.
Garland, S.J., Willems, D.A., Ivanova, T.D., Miller, K.J., 2003. Recovery of standing balance and functional mobility after stroke. Arch. Phys. Med. Rehabil. 84, 1753–1759.
Geboers, J.F., Drost, M.R., Spaans, F., Kuipers, H., Seelen, H.A., 2002. Immediate and long-term effects of ankle-foot orthosis on muscle activity during walking: a randomized study of patients with unilateral foot drop. Arch. Phys. Med. Rehabil. 83, 240–245.
Genthon, N., Rougier, P., Gissot, A.S., Froger, J., Pelissier, J., Perennou, D., 2008. Contribution of each lower limb to upright standing in stroke patients. Stroke 39, 1793–1799.
Geurts, A.C., de Haart, M., van Nes, I.J., Duysens, J., 2005. A review of standing balance recovery from stroke. Gait Posture 22, 267–281.
Gök, H., Küçükdeveci, A., Altinkaynak, H., Yavuzer, G., Ergin, S., 2003. Effects of ankle-foot orthoses on hemiparetic gait. Clin. Rehabil. 17, 137–139.
Holden, M.K., Gill, K.M., Magliozzi, M.R., Nathan, J., Piehl-Baker, L., 1984. Clinical gait assessment in the neurologically impaired. Reliability and meaningfulness. Phys. Ther. 64, 35–40.
Koopman, B., Grootenboer, H.J., de Jongh, H.J., 1995. An inverse dynamics model for the analysis, reconstruction and prediction of bipedal walking. J. Biomech. 28, 1369–1376.
Kwakkel, G., Kollen, B., Lindeman, E., 2004. Understanding the pattern of functional recovery after stroke: facts and theories. Restor. Neurol. Neurosci. 22, 281–299.
Lehmann, J.F., Condon, S.M., Price, R., deLateur, B.J., 1987. Gait abnormalities in hemiplegia: their correction by ankle-foot orthoses. Arch. Phys. Med. Rehabil. 68, 763–771.
Leung, J., Mosely, A., 2003. Impact of ankle-foot orthoses on gait and leg muscle activity in adults with hemiplegia. Physiotherapy 89, 39–55.
Mao, H.F., Hsueh, I.P., Tang, P.F., Sheu, C.F., Hsieh, C.L., 2002. Analysis and comparison of the psychometric properties of three balance measures for stroke patients. Stroke 33, 1022–1027.
Marigold, D.S., Eng, J.J., 2006. The relationship of asymmetric weight-bearing with postural sway and visual reliance in stroke. Gait Posture 23, 249–255.
Podsiadlo, D., Richardson, S., 1991. The timed "Up and Go": a test of basic functional mobility for frail elderly persons. J. Am. Geriatr. Soc. 39, 142–148.
Tyson, S.F., Thornton, H.A., 2001. The effect of a hinged ankle foot orthosis on hemiplegic gait: objective measures and users' opinions. Clin. Rehabil. 15, 53–58.
Tyson, S.F., Kent, R.M., 2009. Orthotic devices after stroke and other non-progressive brain lesions. Cochrane Database Syst. Rev. 21 (1), CD003694.
van Asseldonk, E.H., Buurke, J.H., Bloem, B.R., Renzenbrink, G.J., Nene, A.V., van der Helm, F.C., van der Kooij, H., 2006. Disentangling the contribution of the paretic and non-paretic ankle to balance control in stroke patients. Exp. Neurol. 201, 441–451.
van der Kooij, H., van Asseldonk, E., van der Helm, F.C., 2005. Comparison of different methods to identify and quantify balance control. J. Neurosci. Methods 145, 175–203.
van der Kooij, H., van Asseldonk, E.H., Geelen, J., van Vugt, J.P., Bloem, B.R., 2007. Detecting asymmetries in balance control with system identification: first experimental results from Parkinson patients. J. Neural. Transm. 114, 1333–1337.
Wade, D.T., 1992. Measurements in Neurological Rehabilitation. Oxford University Press, Oxford.
Wang, R.Y., Lin, P.Y., Lee, C.C., Yang, Y.R., 2007. Gait and balance performance improvements attributable to ankle-foot orthosis in subjects with hemiparesis. Am. J. Phys. Med. Rehabil. 86, 556–562.
Wang, R.Y., Yen, L., Lee, C.C., Lin, P.Y., Wang, M.F., Yang, Y.R., 2005. Effects of an ankle-foot orthosis on balance performance in patients with hemiparesis of different durations. Clin. Rehabil. 19, 37–44.