

CLINICAL NOTE

Near-Normal Gait Pattern With Peroneal Electrical Stimulation as a Neuroprosthesis in the Chronic Phase of Stroke: A Case Report

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ABSTRACT. van Swigchem R, Weerdesteyn V, van Duijnhoven HJ, den Boer J, Beems T, Geurts AC. Near-normal gait pattern with peroneal electrical stimulation as a neuroprosthesis in the chronic phase of stroke: a case report. *Arch Phys Med Rehabil* 2011;92:320-4.

In recent years, the use of functional electrical stimulation (FES) of the peroneal nerve has increased as an alternative for an ankle-foot orthosis (AFO) to treat stroke-related drop foot. We present a chronic stroke patient demonstrating an almost normal gait pattern with peroneal FES as a neuroprosthesis. A 60-year-old survivor of a right hemisphere infarction 21 months ago, who regularly used a polypropylene AFO, was provided with a surface-based peroneal FES device for severe drop foot. In a second instance, he received an implanted FES system because of skin problems with the surface stimulator. With both FES devices, the patient achieved an adequate foot elevation. Moreover, his hip and knee flexion angles during walking increased to normal values and his ankle push-off power increased. His gait pattern became almost symmetrical and less variable than with the AFO. Furthermore, his ability to avoid a sudden obstacle improved to normal values with FES. Our patient showed benefits from peroneal FES beyond what can be attributed to improved foot lift alone. With regard to the potential working mechanisms underlying this response to FES, biomechanical benefits related to improved ankle push-off are suggested as the main mechanism.

Key Words: Electric stimulation; Peroneal nerve; Rehabilitation; Stroke; Walking.

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IN STROKE PATIENTS who suffer from a drop foot, an AFO is usually provided to lift the foot during the swing phase and early stance phase of gait in order to prevent the toes from touching the ground and to facilitate heel loading. As an alternative treatment, FES of the peroneal nerve as a form of

neuroprosthesis is gradually becoming more feasible due to advanced and commercially available systems. Peroneal FES activates the muscles that dorsiflex and evert the ankle joint as well as the toe extensors and may lead to significant improvement in the gait pattern¹ and gait speed^{2,3} compared to walking without aids. Although FES has several theoretical advantages over an AFO and patients' preferences often support the use of FES, there is as yet no conclusive evidence for the superiority of peroneal FES over an AFO with respect to walking abilities.^{4,5} The aim of the present report is to demonstrate the potential superiority of peroneal FES over an AFO. To this end, we present a stroke patient who was used to walking with a polypropylene AFO and who showed functional benefits from FES that went beyond what can be attributed to improved foot elevation alone (open access videos on http://www.neurorehab.nl/APMR_english.htm). Secondly, we aim to discuss the potential mechanisms underlying the observed functional improvements.

CASE DESCRIPTION

Patient and Methods

A 60-year-old farmer presented himself at the outpatient clinic of our university hospital 21 months after a right hemispheric infarction. Characteristics of the participant are provided in table 1. The Motricity Index⁶ of his left lower limb was 27% with a 0 score for the ankle dorsiflexors. The strength of his calf muscles was scored Medical Research Council⁷ grade 4. Because of his ankle dorsiflexor paralysis, the patient used an AFO, which was a custom made polypropylene (2.5-mm thickness) posterior splint, trimmed behind the malleoli of the ankle. The AFO was rigid into plantar flexion, but allowed about 15° of ankle dorsiflexion. The patient was hardly able to walk without this AFO due to foot drag during the swing phase of gait, which he was not able to compensate by active hip and knee flexion. At baseline (t0), his gait ability with AFO was assessed, after which he received a surface-based peroneal FES device, the NESS L300.⁴ The L300 stimulated the peroneal nerve and the anterior tibial muscle using 2 electrodes embedded into a lightweight orthosis placed just

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List of Abbreviations

AFO	ankle-foot orthosis
COV	coefficient of variation
FES	functional electrical stimulation
V_{pifl}	maximal ankle plantar flexion velocity (at push-off)
$T_{swing\ asym}$	asymmetry in swing time duration between left and right limb
M_{pifl}	maximal ankle plantar flexion torque (at push-off)

Table 1: Characteristics of the Participant at Baseline

Variable	Patient's outcome
Age (y)	59
Time poststroke (mo)	21
Sex (male/female)	Male
Body weight (kg)	101
Body height (m)	1.78
Hemisphere of stroke	Right
Type of stroke	Infarction
Modified Ashworth Score (0–5)*	
Knee flexors/extensors*	0/0
Ankle plantar flexors/dorsiflexors*	0/0
Lower extremity Motricity Index (0–100) *	27
Ankle score (0–33)*	0
Lower extremity Fugl-Meyer Assessment (0–34)*	6
Calf muscle strength (Medical Research Council 0–5)*	4
Quantitative Vibration Threshold (0=no sensation; 8=normal)*	
First metatarsophalangeal joint*	<2
Lateral malleolus*	2
Berg Balance Scale (0–56)	41
Passive range of motion at ankle (degrees)*	
Dorsiflexion (knee extended/flexed)/	
Plantar flexion*	(0/10)/35
Varus/Valgus*	30/15

*Scores at the paretic body side.

below the knee (symmetrical bipolar 42mA current; pulse rate 30Hz; phase duration 200 μ sec) during the swing phase and the subsequent loading response. Stimulation onset and offset were based on detection of heel-off and heel-on by an insole foot switch that communicated wirelessly with the stimulator that was attached to the orthosis. In the following 2 weeks, the

patient increased the use of FES up to 6 hours a day. Subsequently, the gait assessment was repeated with both the AFO and FES (t1 [measurements after 2 wk]). In the third week, the patient developed an allergic skin reaction to the electrodes, which was dermatologically confirmed. Because no alternative electrodes were available at that time, the patient had to terminate the use of the surface stimulator shortly after he had started. One year later, this patient was the first person to be implanted with the ActiGait^b peroneal stimulator in The Netherlands. With this system, the common peroneal nerve was directly stimulated through 4 distinct electrode arrays embedded in a cuff, which was surgically placed around the nerve about 4cm above the knee joint (asymmetrical bipolar 1.2mA current; pulse rate 20Hz; phase duration 247, 236, 210, and 0 μ sec for channel 1 to 4). The stimulation settings of the 4 channels were individually adjusted such that a balanced eversion and dorsiflexion were evoked during the swing phase and early stance phase of gait. Stimulation onset was timed simultaneously with heel-off, and the offset of stimulation was timed using a ramp-down period of 0.5 seconds following heel strike. Heel contacts were detected by an insole foot switch that communicated wirelessly with the control unit that was worn on a waist belt. The control unit was hard-wired externally to a transmitter coil (antenna), positioned on the skin over a receiver and stimulator, which was implanted subcutaneously on the lateral side of the proximal thigh. The stimulator was hard-wired subcutaneously to the cuff electrode. After he had used this system all day long for 6 months, a third gait assessment was conducted with his AFO and with the implanted stimulator (t2 [measurement after 1.5 y]). In addition, in spite of the difficulties the patient experienced when he walked without an orthosis, the assessment of comfortable walking was also performed without any device at this occasion.

During each gait assessment, 70 gait cycles of treadmill walking at a comfortable speed of 0.56m/s were analyzed. Flexion-extension movements of the hip, knee, and ankle joints were measured with goniometers^c at a sample rate of 1000Hz, and maximal flexion and extension joint angles during the step cycle

Table 2: Results of the Gait Assessments

Variable	t0		t1		t2			Referent
	AFO	AFO	Surface FES	AFO	Implanted FES	NO		
Hip flexion (deg)	23.2 \pm 1.5	17.9 \pm 1.5	24.2 \pm 1.5	20.5 \pm 3.2	24.5 \pm 1.3	21.8 \pm 2.1	18.6 \pm 5.4 ¹²	
Hip extension (deg)	0.5 \pm 1.6	-7.7 \pm 3.6	-8.1 \pm 1.5	8.9 \pm 5.2	-9.3 \pm 1.5	6.3 \pm 5.0	-11.3 \pm 7.8 ¹²	
Knee flexion (deg)	45.6 \pm 5.9	45.9 \pm 5.5	66.3 \pm 1.6	27.6 \pm 2.4	56.8 \pm 1.2	25.9 \pm 9.5	61.8 \pm 7.0 ¹²	
Knee extension (deg)	1.8 \pm 0.9	4.2 \pm 0.6	2.7 \pm 0.7	6.3 \pm 4.6	4.4 \pm 0.6	5.9 \pm 1.4	1.7 \pm 3.2 ¹²	
Ankle dorsiflexion (deg)	9.2 \pm 1.3	6.8 \pm 1.8	9.5 \pm 0.8	3.2 \pm 2.1	10.8 \pm 1.0	1.0 \pm 4.1	8.6 \pm 3.6 ¹²	
Ankle plantar flexion (deg)	-7.9 \pm 0.9	-8.0 \pm 1.4	-17.1 \pm 2.7	-13.2 \pm 2.7	-16.6 \pm 0.8	-26.9 \pm 4.0	-17.6 \pm 4.7 ¹²	
M_{plfi} (Nm/kg)	ND	ND	ND	0.93 \pm 0.06	1.50 \pm 0.02	1.15 \pm 0.21	1.54 \pm 0.23 ¹²	
V_{plfi} ($^{\circ}$ /s)	76.5 \pm 15.0	88.7 \pm 14.1	135.0 \pm 28.0	31.1 \pm 10.1	181.0 \pm 34.2	117.4 \pm 46.0	136 \pm 53 ^{*12}	
COV_{hip} (%)	16.0	26.8	15.4	22.5	14.6	27.1	8 ⁸	
COV_{knee} (%)	37.1	37.6	23.2	65.0	18.5	71.7	8 ⁸	
$\text{COV}_{\text{ankle}}$ (%)	33.6	34.3	29.4	48.1	26.3	37.6	21 ⁸	
$T_{\text{swing asym}}$ (%)	42.0	38.6	14.8	35.9	18.1	60.1	-0.1 \pm 11.3 ⁹	
OA success (%)	17.0	0.0	68.0	7.0	90.0	ND	89.7 \pm 5.8 ¹⁰	
Gait speed (m/s)	0.99 \pm 0.13	1.06 \pm 0.06	1.07 \pm 0.04	1.01 \pm 0.09	1.00 \pm 0.17	0.41 \pm 0.02	1.36 \pm 0.21 ¹³	

NOTE. Results are displayed as mean \pm (within subject) SD at the paretic body side or as otherwise noted. Referent data are retrieved from literature^{8-10,12,13} and displayed as mean \pm (between subjects) SD.

Abbreviations: $\text{COV}_{\text{ankle}}$, coefficient of variation for the ankle; COV_{hip} , coefficient of variation for the hip; COV_{knee} , coefficient of variation for the knee; deg, degrees; gait speed, 10-m comfortable walking speed; M_{plfi} , maximal ankle plantar flexion torque (at push-off); ND, no data; NO, no orthosis; OA success, obstacle avoidance success score; $T_{\text{swing asym}}$, asymmetry in swing time between left and right leg; V_{plfi} , maximal ankle plantar flexion velocity within 200ms before toe-off.

*Referent SD of V_{plfi} was estimated from the figure of Winter.¹²

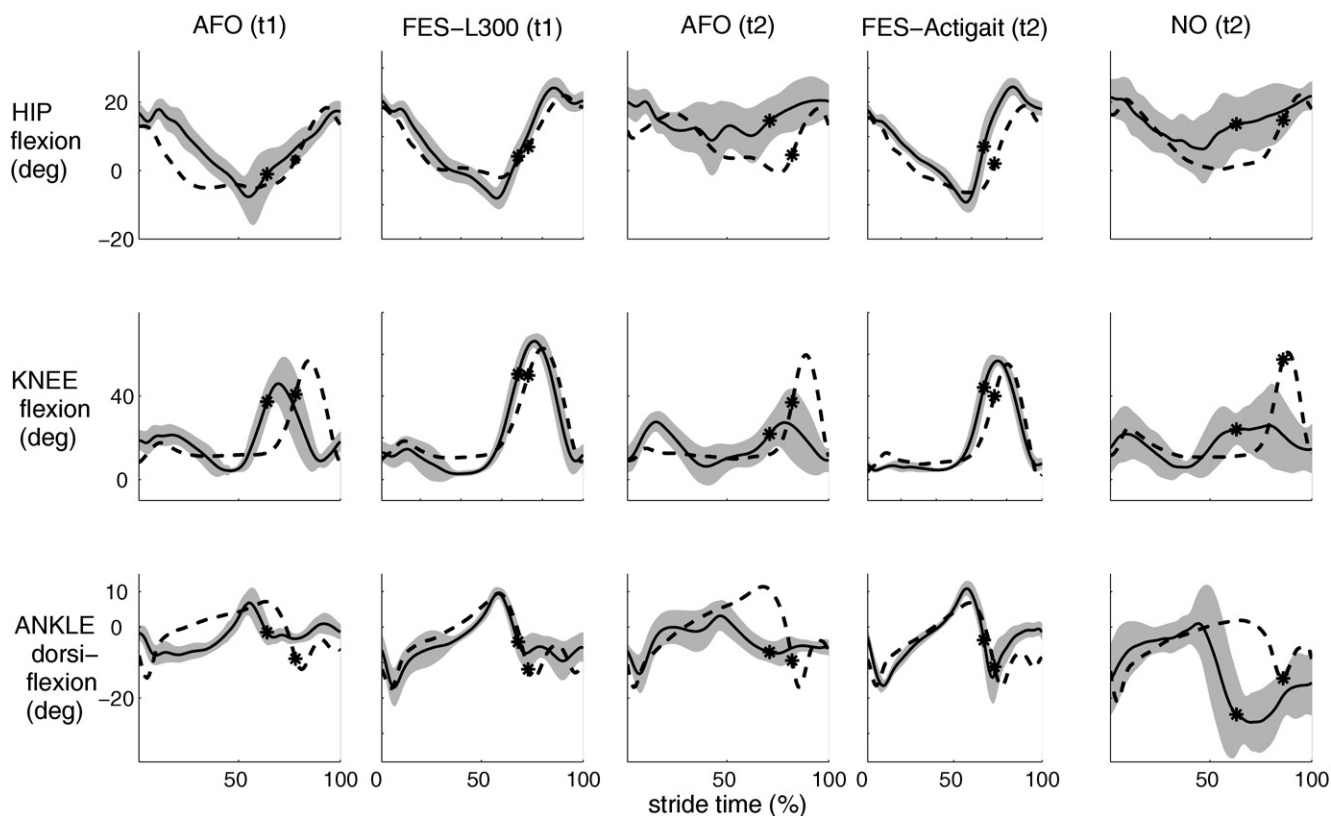


Fig 1. Time-normalized joint angles (from heel strike to heel strike) of the measurements at t1 (with the AFO and surface-based FES) and at t2 (with the AFO, implanted FES, and no orthosis). Note: Movement variability (see variability band) of hip and knee decreased with FES. Broken line, unaffected joint; solid line, affected joint; gray area, variability (2 times SD); Asterisks (*) at the affected and unaffected joint angles indicate toe-off.

were determined. The COV for the hip (COV_{hip}), knee (COV_{knee}), and ankle (COV_{ankle}) joint angles, and maximal ankle plantar flexion velocity (V_{plf}) were computed after low-pass filtering (6Hz; zero lag, 4th order Butterworth filter) of the joint angle signals. The COV was calculated as the root mean square of the SD at each time interval divided by the mean magnitude of joint angle over the stride.⁸ V_{plf} was determined from the differentiated ankle angle signal as the maximum value within 200ms before toe off. With a 6-camera 3-dimensional motion analysis system^d, foot marker data were collected at a sample rate of 100Hz to determine heel strike and toe off. The asymmetry in swing time between left and right limb ($T_{swing\,asym}$) was defined as:

$$\left[\frac{(T_{swing\,left} - T_{swing\,right})}{\max(T_{swing\,left}, T_{swing\,right})} \right] * 100^9$$

where $T_{swing\,left}$ stands for swing time (from toe off to heel strike) of the left (paretic) leg and $T_{swing\,right}$ stands for swing time of the right (nonparetic) leg. Subsequently, to assess advanced gait skills, the ability to avoid sudden obstacles during walking was tested on a treadmill. Thirty obstacles were dropped in front of the paretic foot during each test (for detailed description of the methods see Weerdesteyn et al^{10,11}). Avoidance success rates were determined as the proportion of trials without foot contact to the obstacle. Without any device, walking ability was too poor to perform the obstacle avoidance trials. Finally, comfortable gait speed was measured during overground walking on a 10-m walkway, both with FES and AFO.

In addition, at t2, ground reaction forces were measured for 1 step with each leg during 3 overground trials by means of 2 force

plates^e positioned in the middle of the walkway. Kinematic data were recorded from 16 reflective markers, which were placed on the lower limbs according to the lower-body model (Vicon^d). These data were analyzed using the PlugInGait model in Vicon Workstation^d to obtain the joint torques of the lower limb.

RESULTS

The outcomes of the gait assessments are summarized in table 2. With surface-based FES (t1), maximal hip and knee flexion angles during the gait cycle increased by 6.3° and 20.4°, respectively, compared with the AFO, while maximal hip and knee extension angles were similar. COV_{hip} and COV_{knee} decreased by 11.4% and 14.4%, respectively. V_{plf} increased by 46.3%/s with FES compared to AFO. Finally, movement symmetry increased with FES, as indicated by a 23.8% decrease in $T_{swing\,asym}$.

At t2, the gait pattern with AFO had deteriorated compared to t0 and t1, as indicated most clearly by decreased stance phase hip extension and swing phase knee flexion and increased movement variability. With implanted FES, similar improvements as with surface-based FES were observed for hip and knee flexion angles, COV_{hip} and COV_{knee} , $T_{swing\,asym}$, and for V_{plf} . Walking ability without any device was measured, although only 10 gait cycles on the treadmill could be performed. The movement pattern was characterized by large variability and asymmetry, but could not be compared with previous assessments because measurements without a device were lacking at those instants.

The increased joint excursions and decreased joint motion variability with FES are illustrated in figure 1. In addition, this

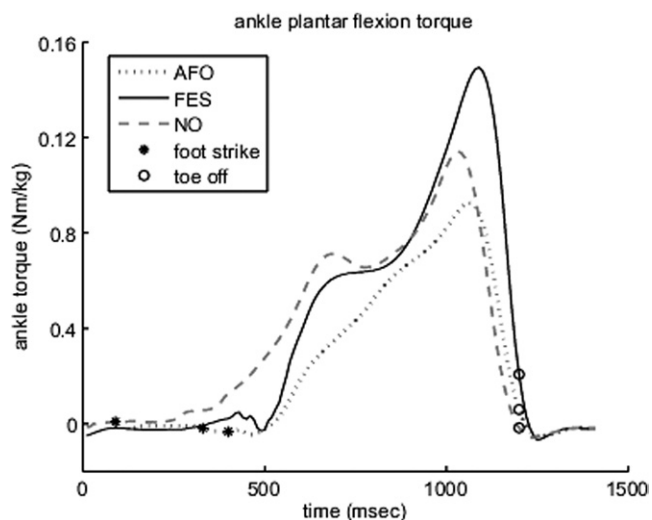


Fig 2. Plantar flexion torque at the affected ankle, averaged over 3 trials in each of the conditions. Abbreviation: NO, no orthosis.

figure shows that with FES the joint angles at the paretic side closely resembled those at the nonparetic side, whereas with the AFO the joint excursions showed substantial differences between the lower limbs with a time shift in the start of the swing phase. These results demonstrate that with FES, the movement pattern became more symmetrical, and that particularly, the ranges of motion at the ankle, knee, and hip were almost normalized.¹²

Kinetic data were only obtained at t2. Profiles of ankle torques with FES, AFO, and without device are provided in figure 2. Consistent with the data for V_{plf} , maximal ankle plantar flexion torques at push-off (M_{plf}) were larger with FES than with AFO or without device (1.50, 0.93, and 1.15 Nm/kg, respectively) (table 2).

The obstacle avoidance success scores were better with surface-based and implanted FES (68% and 90% success, respectively) than with the AFO (0%–17%) (see table 2). In contrast, the comfortable walking speed did not differ between AFO and FES (1.04 m/s with both devices).

DISCUSSION

In our patient, knee and hip movements almost normalized when he walked with FES, which was reflected by increased knee and hip flexion angles, decreased knee and hip motion variability, and increased movement symmetry (see table 2). Remarkably, the active ankle push-off benefited from FES. In addition, the ability to avoid sudden obstacles during walking improved with FES. These results indicate that peroneal FES not only elevates the foot, but may improve the entire gait pattern after stroke, even when walking is affected by severe hemiparesis.

In order to evaluate whether this potential of peroneal FES has been addressed previously, a literature search was conducted. Gait speed appeared to be used most frequently as an outcome measure. However, gait speed does not provide information about the quality of the walking pattern. Indeed, our patient had an almost normal gait speed,¹³ yet a very asymmetric and inefficient gait pattern at baseline (see t0 in table 2). The effects of peroneal FES on knee and hip motion angles during gait have only been reported in 2 preexperimental studies^{14,15} showing inconclusive results. Hence, the possibility

of more extended effects of FES on the hemiparetic walking pattern has received little attention. Some mechanisms have been suggested that might explain the extended effects of FES that we observed. First, peroneal FES might reduce lower-limb spasticity.^{16,17} However, in our patient, this mechanism is not likely because the modified Ashworth scores at baseline (see table 1) indicated no signs of spasticity. Particularly in spinal cord injured patients, it has been observed that peroneal FES may facilitate the lower-limb flexion reflex.¹⁸ In our patient, maximal knee and hip flexion increased with FES (see table 2), which might indeed be indicative of such flexion reflex facilitation. An additional observation in our patient might support this notion. When he was sitting on a chair, he was not able to flex his paretic hip to lift his leg from the ground, whereas with FES he was able to do so. On the other hand, the movement sequence for swing initiation, as observed in all conditions, was similar to the sequence in normal gait and started with knee flexion, followed by hip flexion, and finally ankle dorsiflexion.¹² When the flexion reflex would have been triggered in a stationary position, the onset of hip, knee, and ankle movements would have typically occurred simultaneously, which renders this possibility less likely. Nevertheless, because we did not record muscle activations and because gait is influenced by passive mechanical constraints as well, preservation of the normal movement sequence does not preclude the possibility of a facilitated flexion reflex. Another mechanism that has been mentioned in the literature is that, in contrast to most AFOs, FES allows the use of residual active ankle plantar flexion at push-off.¹⁹ Such increased plantar flexion power increases knee flexion during preswing and, thus, facilitates the swing phase of gait.²⁰ Indeed, the increased V_{plf} (see table 2) indicated that with FES our patient had faster plantar flexion movements at push-off. The increase in M_{plf} at t2 showed that with FES, the ankle plantar flexion torques increased to normal values (see table 2), which confirms that the increased ankle plantar flexion velocity reflected an active propulsion. However, the precise mechanism underlying the improved ankle push-off with FES in our patient is not clear. It might be related to less dependency on a compensatory movement strategy that he had developed to overcome impaired swing phase initiation. Both with the AFO and without a device, he showed excessive hip exorotation during the stance phase, which may have enabled him to use the inertia of the trunk to initiate the subsequent swing of the leg (see videos on http://www.neurorehab.nl/APMR_english.htm). This increased hip exorotation might explain the decreased hip extension and ankle dorsiflexion during the stance phase as observed with and without AFO at t2 (see table 2). With FES, there was improved hip extension (without exorotation) and increased ankle dorsiflexion during the stance phase, which may have contributed to an increase in push-off ankle torque due to more passive stretch of the calf muscle. The larger ankle plantar flexion velocity may have further added to the increased push-off power with FES.

With respect to the functional benefits related to FES in our patient, we expected him to have better walking abilities with peroneal FES on regular, but even more so, on irregular terrain. The large improvement in obstacle avoidance success rates confirmed that his advanced gait skill had indeed substantially improved with FES. Such improvements might also have led to safer ambulation during daily life. Indeed, our patient reported to fall about once a week with his AFO, whereas he reported no more falls with FES. With FES, he experienced an improvement of balance and went out for a walk, which he was reluctant to do with his AFO. With respect to comfortable walking speed, we observed no difference between AFO and FES, whereas previous work on the L300 and the ActiGait

systems reported increased speeds (34%¹ and 19%²¹, respectively) for walking with FES in the long term compared with walking without FES at baseline. The difference with the present study can most likely be attributed to the almost normal initial walking speed of our patient, which probably prevented any FES-induced improvement. Previous studies did not evaluate the effects of the L300 or ActiGait on the quality of the gait pattern in terms of kinematics.

In order to convincingly demonstrate the potential superiority of FES over an AFO in our patient, the finding of a deterioration of gait with the AFO at t2 needs to be further clarified. In the year that the patient waited for implantation, he retired from his profession and reduced his activity level. At a gait assessment 2 weeks before implantation, the gait pattern already showed deteriorations similar to those observed at t2, with absent hip extension, decreased knee flexion angles, and increased movement variability. This indicated that the deterioration of gait at t2 was not related to the 6 months of FES use, but was most likely due to the decrease in physical activity before the implantation of FES. Whether a similar deterioration was also present in his gait pattern without any device could not be evaluated because of the lack of walking assessments without any device at t0 and t1.

CONCLUSIONS

The presented case study shows that peroneal FES may have important effects on the entire hemiparetic gait pattern beyond what can be attributed to improved foot lift alone. The effects of FES on knee and hip motion angles as well as on gait symmetry, gait variability, and obstacle avoidance skills call for further research including these outcome measures and the most commonly used walking speed. Furthermore, fundamental studies are needed on the neural and biomechanical mechanisms that may underlie the observed extended effects of peroneal FES.⁴ Finally, because the extended effects of FES in our patient may not be generalized to other patients, future research should try to identify the subject characteristics that are predictive of such a beneficial response in patients with hemiparesis due to stroke.

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Suppliers

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