

Non–Velocity-Related Effects of a Rigid Double-Stopped Ankle-Foot Orthosis on Gait and Lower Limb Muscle Activity of Hemiparetic Subjects With an Equinovarus Deformity

Stefan Hesse, MD; Cordula Werner, MS; Konrad Matthias, MD;
Kirker Stephen, MD; Michael Berteau, MD

Background and Purpose—This study investigated the non–velocity-related effects of a 1-bar rigid ankle-foot orthosis on the gait of hemiparetic subjects, with particular emphasis on the muscle activity of the paretic lower limb.

Methods—Twenty-one hemiparetic subjects who had been using an ankle-foot orthosis for equinovarus deformity for <1 week participated. Patients walked cued by a metronome at a comparable speed with and without the orthosis. Dependent variables were basic, limb-dependent cycle parameters, gait symmetry, vertical ground reaction forces, sagittal ankle excursions, and kinesiological electromyogram of several lower limb muscles.

Results—The use of the caliper was associated with more dynamic and balanced gait, characterized by longer relative single-stance duration of the paretic lower limb, better swing symmetry, better pivoting over the stationary paretic foot, and better ankle excursions ($P<0.05$). The functional activity of the paretic quadriceps muscles increased, while the activity of the paretic tibialis anterior muscle decreased ($P<0.05$).

Conclusions—The orthosis led to a more dynamic and balanced gait, with enhanced functional activation of the hemiparetic vastus lateralis muscle. The study further supports the functional benefits of a rigid ankle-foot orthosis in hemiparetic subjects as an integral part of a comprehensive rehabilitation approach. However, the reduced activity in the tibialis muscle may lead to disuse atrophy and hence long-term dependence on the orthosis. (*Stroke*. 1999;30:1855-1861.)

Key Words: gait ■ hemiplegia ■ motor activity ■ orthotic devices ■ rehabilitation

An ankle-foot orthosis (AFO) is frequently prescribed in the rehabilitation of hemiparetic patients suffering from an equinovarus deformity. It should assist foot clearance during the swing phase, improve the mode of initial contact, prevent ankle inversion injuries, and help in advancing the body during midstance.¹

With an AFO, hemiparetic patients walk faster, with a longer stride, diminished plantar flexion, and larger active ankle moment, and they consume less oxygen when walking compared with walking without a caliper.^{1–8} Beyond that, Lehmann and coworkers, who investigated double-stopped AFOs, described how the dorsiflexion stop created a bending moment and the plantar flexion stop created an extensor moment at the knee in hemiparetic patients.^{3,9}

The present study investigated orthosis-related effects on hemiparetic gait that could not be explained by an increase in walking velocity alone. Patients usually walk more quickly with an AFO, and walking speed affects most gait variables.¹⁰

Furthermore, the study addressed the question of the muscular activity of the lower limb muscles in hemiparetic subjects walking with an AFO and barefoot. This question is relevant to subjects' daily clinical routine since therapists often hesitate to prescribe an AFO despite the obvious functional gains. They fear that the use of an orthosis might result in untimely overactivity of the plantar flexor and nonuse of the tibialis anterior muscle.

Subjects and Methods

Subjects

After providing informed consent, 21 hemiparetic patients (10 women, 11 men; mean age, 58.2 years; age range, 30 to 79 years) participated in the study. Twelve patients had a right hemiparesis and 9 a left hemiparesis. The etiology was ischemia in the region of the middle or anterior cerebral artery in 17 cases, intracranial hemorrhage in 3 cases, and tumor surgery in 1 case. Seven patients suffered from sensory impairment, and 3 showed behavioral signs of a sensorimotor neglect syndrome. The mean interval between the lesion and admission to the

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From the Klinik Berlin, Department Neurological Rehabilitation, Free University Berlin (Germany) (S.H., C.W., K.M.); Disablement Services Center, Addenbrookes Hospital NHS Trust, Cambridge, UK (K.S.); and Department of Physical Medicine and Rehabilitation, University of Bucharest (Romania) (M.B.).

Reprint requests to S. Hesse, MD, Klinik Berlin, Kladower Damm 223, 14089 Berlin, Germany. E-mail bhesse@zedat.fu-berlin.de

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TABLE 1. Gait Variables and Vertical Ground Reaction Force

			<i>P</i>	
	Barefoot	With AFO	Multivariate	Univariate
Basic cycle parameters				
Gait velocity, m/s	0.32 (0.17)	0.33 (0.15)		
Stride length, m	0.62 (0.17)	0.65 (0.18)		
Cadence, steps/min	62 (17)	63 (16)		
Relative limb cycle parameters				
Stance				
Affected, % cycle	69.1 (5.9)	67.0 (6.3)		
Nonaffected (−)	79.2 (4.0)	73.7 (3.4)		
Swing				
Affected (−)	30.8 (5.9)	33.0 (6.3)	*	
Nonaffected (−)	20.6 (3.9)	27.1 (4.2)		*
Double support, affected leg				
Initial (−)	21.6 (8.1)	19.6 (6.2)		
Terminal (−)	27.0 (5.2)	21.3 (6.4)		*
Symmetry				
Stance symmetry (−)	0.87 (0.10)	0.89 (0.12)		
Swing symmetry (−)	0.68 (0.12)	0.81 (0.14)	*	*
Double support symmetry (−)	0.70 (0.16)	0.76 (0.20)		
Vertical ground reaction forces				
Gait line				
Affected leg, %	48.2 (8.1)	69.5 (11.9)		*
Nonaffected leg, %	64.3 (8.6)	68.0 (9.8)	*	
Loading rate				
Affected leg, N/s	1.60 (0.5)	1.62 (0.4)		*
Nonaffected leg, N/s	1.48 (0.3)	1.93 (0.5)		

Values are mean (SD). Gait line indicates length of trajectory of the force point of action in percentage of anatomic foot length.

*Significant difference ($P < 0.05$) according to multivariate and univariate tests.

study was 4.9 months (range, 1.5 to 16 months). All patients suffered from marked plantar flexor spasticity. The mean modified Ashworth score,¹¹ which tested for ankle dorsiflexion while the subject was supine, was 3.6 points (range, 3 to 5). Achilles tendon cloni while the subject was walking barefoot occurred in 7 patients.

Inclusion criteria for the study were as follows: the ability to walk 20 m barefoot without physical help by a therapist; use of a Valens AFO for <1 week; minimum modified Ashworth grade of 3 (tested for ankle dorsiflexion while supine; grades 0 to 5); no obvious ankle contracture (plantigrade posture after at least 10 minutes of standing

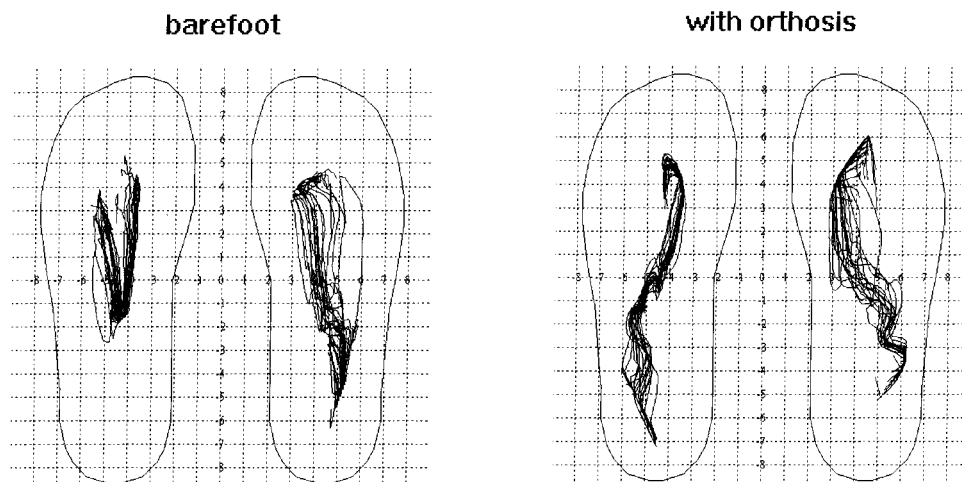
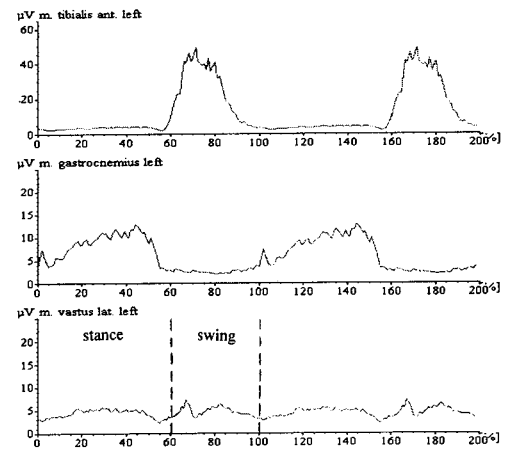
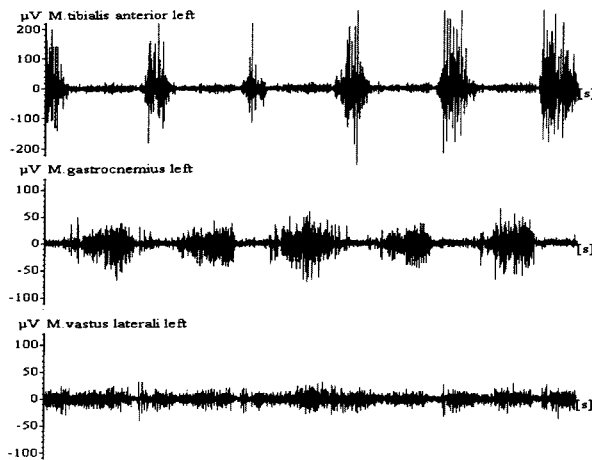


Figure 1. Trajectories of the force point of action under the affected (left) and nonaffected (right) foot of a left hemiparetic subject walking barefoot (left panel) and with an AFO, the Valens caliper, at similar walking velocities (right panel).

barefoot



with orthosis

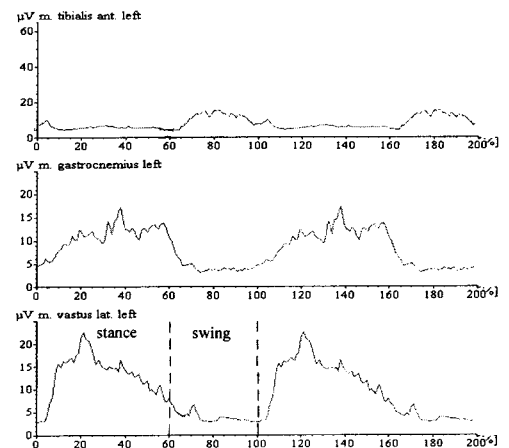
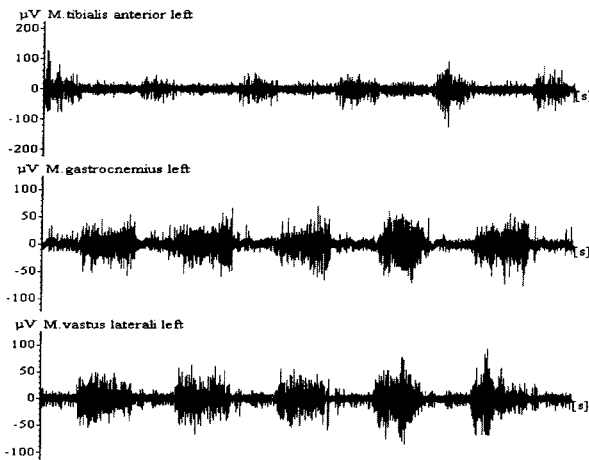


Figure 2. Raw (left) and averaged and, with respect to the gait cycle (100%), normalized (right) EMG of the tibialis anterior, gastrocnemius, and vastus lateralis muscles of the affected side of left hemiparetic subjects walking barefoot (top) and with an AFO, the Valens caliper, at similar walking velocities (bottom). Note the reduction of the swing activity of the tibialis anterior muscle and the increased stance activity of the vastus lateralis muscle when subjects walk with the caliper.

in the standing bar); and no additional orthopedic or neurological deficits impairing ambulation.

The Valens caliper (a 1-bar rigid AFO, with a medial stainless steel upright, padded, and leather-covered posterior steel calf band and an anterior soft closure, with an outside T strap correcting varus position) was attached to a firm shoe (Bally, Oxford-type shoe with a leather outer sole and rubber heel, with touch-and-loop fasteners). The caliper had a double-stopped ankle, which was set by the technician to the range of 80° (−10° dorsiflexion) to 90° (neutral position). In addition to the plantar flexion stop, dorsiflexion was assisted by a spring, the tension of which could be adjusted with a screw. The Valens caliper recommended by Davies¹² is frequently prescribed in Germany and Switzerland. The Valens caliper is comparable to a thermoplastic model by biomechanical principles. A thermoplastic AFO is rigid, encasing the ankle, and therefore is a biomechanical equivalent to an anterior and posterior pin stop as provided by the Valens caliper.¹³

Gait Analysis

Gait velocity, cadence, and mean stride length were calculated after patients walked 10 m at maximum speed 3 times.

Angle of ankle dorsiflexion, vertical ground reaction forces at heel-on and at toe-off, and onset and durations of stance, swing, and double support were measured by biaxial goniometers (Penny & Giles, type 180) and overshoe slippers with 8 insole force sensors and recorded at 100 Hz by a portable Datalogger worn by the patient (Infotronic System).¹⁴

Surface electromyography (EMG) was recorded from the subjects' hemiparetic and unaffected tibialis anterior, medial head of gastrocnemius, vastus lateralis, and gluteus medius, with the use of 8-mm self-adhesive electrodes. The impedance was checked and kept below 5 kΩ. These signals (1000-Hz sampling rate) were recorded by the same portable Datalogger for later offline analysis.

Limb-dependent cycle parameters (stance, swing, and double-support durations), vertical ground reaction forces at heel-on and at toe-off (Fz1 and Fz2), trajectories of the force point of action under both feet, and the maximum ankle plantar flexion and dorsiflexion of the affected side were averaged over at least 15 strides. The limb-dependent cycle parameters were normalized with respect to the gait cycle. Symmetry ratios for stance, swing, and double-support durations (duration of the shorter side divided by that of the longer side) were calculated. The lengths of the

TABLE 2. Muscular Activity During Gait

			<i>P</i>	
	Barefoot	With AFO	Multivariate	Univariate
Muscles, affected leg (μ V)				
M tibialis anterior (60%–110%)	20.2 (7.0)	10.5 (6.7)		*
M gastrocnemius (20%–50%)	22.2 (6.7)	25.9 (4.6)	*	
Premature activity (90%–110%)	16.5 (6.1)	17.1 (7.4)		
M vastus lateralis (90%–130%)	20.3 (19.2)	27.4 (21.3)		*
M gluteus medius (90%–140%)	35.6 (8.5)	32.1 (12.2)		
Muscles, nonaffected leg (μ V)				
M tibialis anterior (60%–110%)	57.2 (21.1)	46.2 (22.9)		
M gastrocnemius (20%–50%)	37.1 (27.0)	43.3 (27.1)		
Premature activity (90%–110%)	24.7 (23.6)	24.9 (21.5)		
M vastus lateralis (90%–130%)	17.4 (14.0)	18.8 (15.3)		
M gluteus medius (90%–140%)	31.3 (18.7)	35.6 (20.2)		

Values are mean (SD) muscular activity calculated in the given ranges (%) of the normalized gait cycle.

*Significant difference ($P < 0.05$) according to multivariate and univariate tests.

trajectories of the force point of action under both feet were given as a percentage of the anatomic foot length (referred to as “gait line”). Furthermore, the loading rate of the first force crest was assessed from zero intercept to 80% of Fz1. Additionally, the mode of initial contact of the affected leg was noted (either forefoot, midfoot, or heel).

The EMG data were digitally filtered (bandpass, 10 to 300 Hz), rectified, averaged over at least 10 strides, and time-normalized to the mean cycle duration set to 100%. To quantify the physiological, functional activities of the recorded muscles, mean values of the non-low-pass signals were calculated in the following intervals of the cycle duration, set to 100% (in accordance with the adult muscle phasic activity chart, Shriner’s Hospital, San Francisco, Calif): from 60% to 100% (tibialis anterior), from 20% to 50% (gastrocnemius), from 90% to 120% (vastus lateralis), and from 90% to 140% (gluteus medius). For the gastrocnemius, so-called premature activity was calculated in the interval from 90% to 110%. Two previous kinematic EMG studies of equinovarus after stroke revealed premature calf muscle activation in the terminal swing as an important cause of excessive plantar flexion and varus in stroke patients.^{15,16} All data were gathered within 1 session, so that the comparison of absolute EMG activities within subjects was appropriate if stable recording conditions were assumed.

Experimental Procedure

The gait of all subjects was assessed (1) when subjects were walking barefoot and (2) when subjects were using the Valens caliper and a firm shoe. Any necessary use of a cane (in 5 subjects) was kept constant.

All patients walked cued by a metronome at a comparable walking velocity during both experimental conditions to eliminate the effect of changes in gait speed, so that the effect of the orthosis alone could be analyzed. Before analysis, patients were able to become accustomed to the experimental protocol, and subjects chose a speed at which they could walk with or without the orthosis.

Statistical Analysis

Data were analyzed with the help of a multivariate profile analysis. EMG activity was logarithmically transformed because our own previous experience had shown that the distribution was more normal and variances more uniform after this transformation. Walking without and with orthosis was compared across each type of dependent variable. Additionally, to make analysis more comprehensive, certain types of variables were analyzed simultaneously on a

multivariate basis, namely, basic and limb-dependent cycle parameters, symmetry ratios, vertical ground reaction forces, and muscles of affected and nonaffected sides. In the case of a significant multivariate result ($\alpha = 0.05$), the univariate tests for the different types of variables were additionally considered.

Results

Basic Cycle Parameters

As expected, there was no difference in gait speed, stride length, or cadence between walking with the caliper or barefoot.

Limb-Dependent Cycle Parameters

The use of the caliper resulted in a significant increase (+31.6%) of the relative single-stance period of the affected lower limb (swing not affected) and of the relative terminal double-support duration (–21.1%) with the paretic limb behind. The remaining parameters (stance affected, stance not affected, swing affected, and initial double-support times) did not change significantly (Table 1).

Symmetry Ratios

Swing symmetry improved with the orthosis (+19.1%), while the stance and double-support symmetry remained unchanged (Table 1).

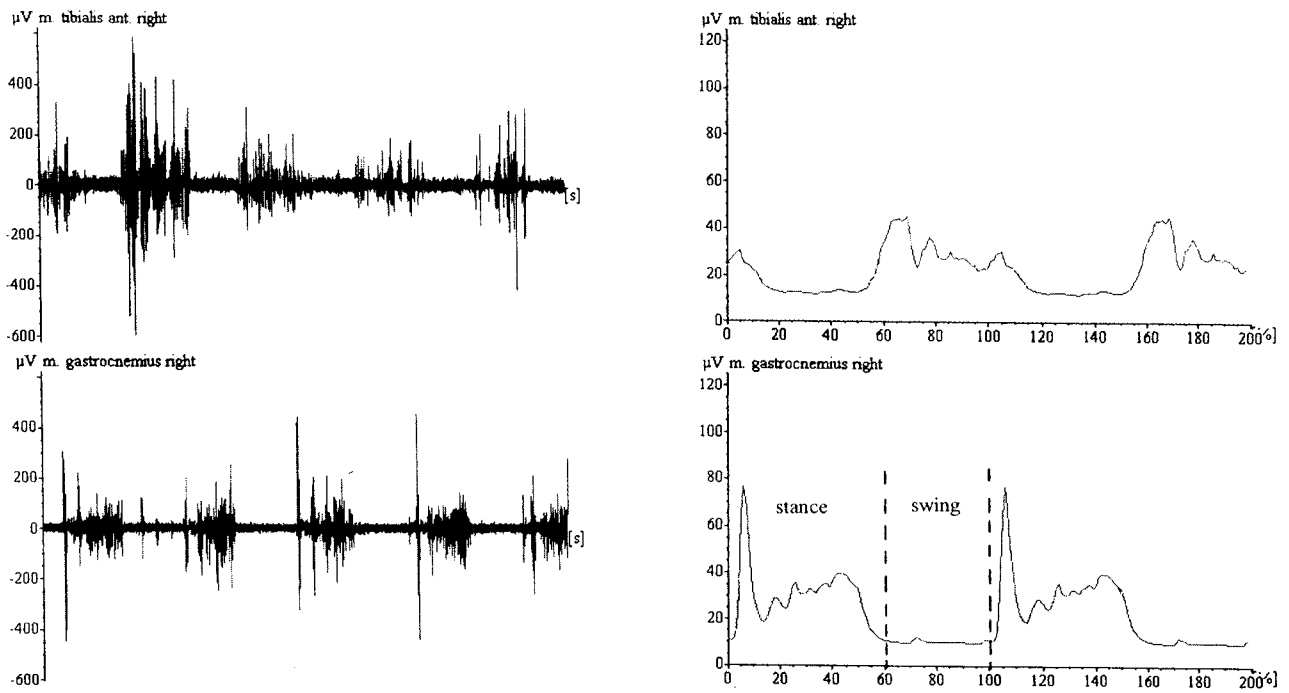
Vertical Ground Reaction Forces

The gait line of the affected lower limb increased with caliper use (+44.2%), and the loading rate of the nonparetic limb was greater (+30.4%) (Figure 1). The peak forces, gait line of the nonaffected lower limb, and loading rate of the affected lower limb did not change (Table 1).

Ankle Excursions

When the subjects were wearing the orthosis, the ankle dorsiflexion during stance became larger (+201.2%) and the plantar flexion during swing became less (–71.2%).

barefoot



with orthosis

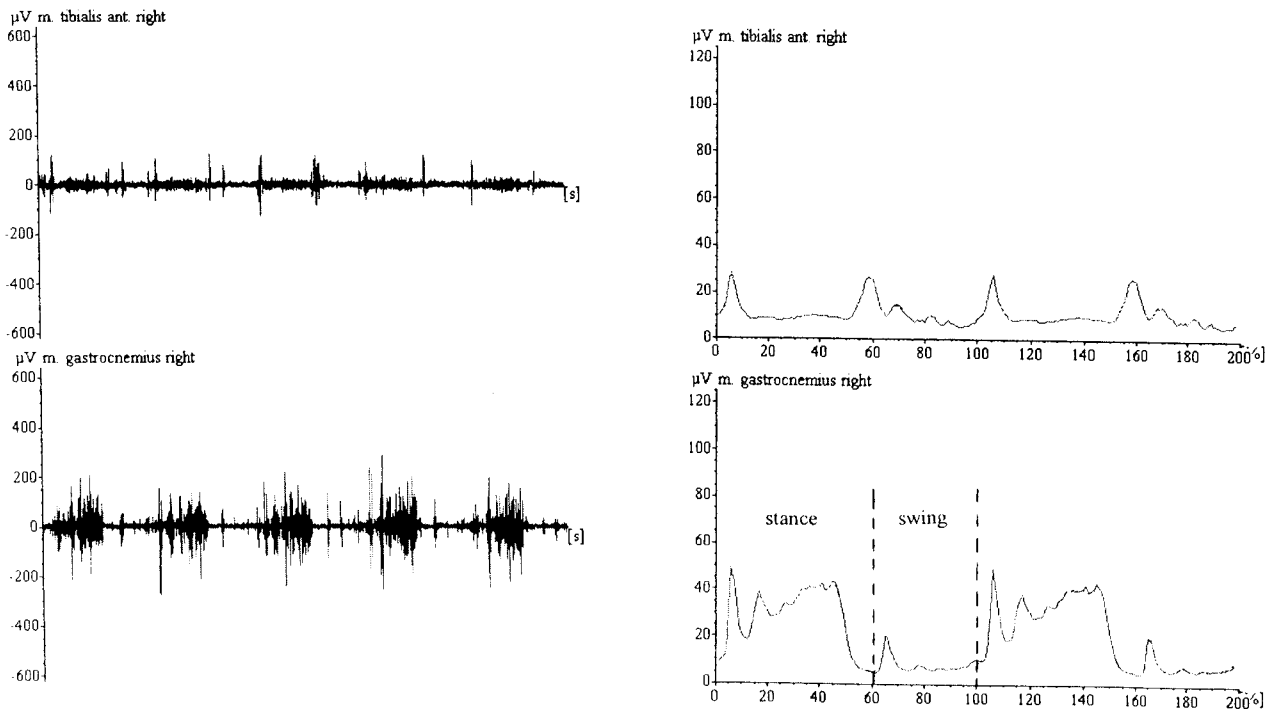


Figure 3. Raw (left) and averaged and, with respect to the gait cycle (100%), normalized (right) EMG of the tibialis anterior and gastrocnemius muscles of the affected side of right hemiparetic subjects walking barefoot (top) and with an AFO, the Valens caliper, at similar walking velocities (bottom). Note the reduction of the swing activity of the tibialis anterior muscle and the reduced premature activity of the plantar flexor when subjects walk with the caliper.

Muscular Activity

Paretic tibialis anterior activity decreased by -48.0% (± 24.2) and the affected vastus lateralis increased by $+35.0\%$ (± 39.1) with caliper use (Figure 2). The activity of the remaining recorded muscles did not change significantly (Table 2). The same applied to so-called premature activity (90% to 110% of gait cycle) of the plantar flexors when subjects walked with the caliper (although 7 of the 21 patients exhibited less premature activity with the AFO) (Figure 3).

Discussion

The use of the caliper effected a more dynamic and balanced gait irrespective of any change of gait velocity. It was characterized by longer relative single-stance duration of the affected lower limb, better swing symmetry, better pivoting over the stationary paretic foot, and better ankle excursions. The use of the orthosis increased activity of the vastus lateralis muscle of the affected side, while the activity of the affected tibialis anterior decreased.

The use of the orthosis reduced double-stance duration and increased single-stance duration on the hemiparetic leg. This was also reflected in an increase in gait symmetry with the AFO. Obviously, the brace could provide a feeling of security, thereby reducing the subjects' need to quickly shift weight to the nonparetic limb.

In confirmation of previous reports,¹⁻⁵ patients wearing the AFO cleared their toes better during swing and pivoted over the stationary foot better. The corresponding changes of the ankle angles were characterized by less plantar flexion during the swing (providing better toe clearance) and a larger dorsiflexion during the stance phase. These effects are mainly attributable to the mechanical properties of the shoes and the preset ankle excursions allowed by the double-stopped brace. A larger flexion of the knee and hip might have also contributed to the observed better toe clearance (as described by Lehmann et al³), but this was not assessed within the present study.

The use of the orthosis changed the muscle activity of the lower limbs. When the patients were wearing the AFO, the activity of the tibialis anterior of the affected side became less, confirming therapists' notion of diminished activity of the ankle dorsiflexor during swing. The brace with its plantar flexor stop probably reduced the patients' efforts to clear the toe during the swing. In the long term, this might render the patients' muscles inactive and might therefore prolong the patients' dependency on the mechanical device. On the other hand, therapists may aim at suppression of an abnormal flexor synergy in hemiparetic subjects triggered by the swing effort.¹² Within this synergy, the tibialis anterior muscle is responsible for an unwanted foot inversion due to its medial insertion with respect to the subtalar joint.¹⁵ The resulting varus deformity impedes the initial contact and weight acceptance, with a risk of spraining of the ankle.

In addition, patients wearing the AFO exhibited a larger activity of the quadriceps muscle during the early stance, when it helps with weight acceptance, and during midstance, when it helps with load bearing.¹⁷ The main peripheral drives for this facilitation were most likely the aforementioned faster loading of the paretic limb and its prolongation of the relative

single-stance period. Accordingly, animal and human studies showed that loading of the paretic limb had a potent facilitatory effect on the antigravity muscles.^{18,19}

On the other hand, there was no overactivity of the plantar flexors, particularly not of the so-called premature activity (90% to 110% of the gait cycle) occurring during the terminal swing and initial stance. Two previous kinematic EMG studies of equinovarus after stroke revealed premature calf muscle activation in the terminal swing as an important cause of excessive plantar flexion and varus in stroke patients.^{15,16} The present result of no overall change of plantar flexor activity therefore does not confirm therapists' fear of increased muscle tone because of the caliper. The marked premature gastrocnemius activity of some patients was even reduced when they were wearing the AFO, as previously noted.¹⁵ This may have been due to a reduction in calf muscle stretch when the first contact with the ground is at the heel instead of at the forefoot.

The swing symmetry improved when the subjects used an AFO. In this context, one should keep in mind that a balanced gait is a hallmark of healthy subjects. "Neurodevelopmental" or Bobath therapists, aiming to restore a symmetrical gait pattern, often hesitate to prescribe an "artificial" caliper under this premise. A large outcome study on 156 hemiparetic subjects, however, failed to show any improvement of gait symmetry during a 4-week comprehensive rehabilitation program following the neurodevelopmental technique.²⁰

Potential confounding factors were the use of a cane in 5 subjects and the firm shoes worn by all subjects together with the AFO. According to 2 previous studies, the gait of hemiparetic subjects walking with and without a cane, however, was not shown to differ significantly with respect to gait symmetry and the kinesiological EMG of thigh and shank muscles of the affected side.^{21,22} With regard to the potential effect of shoes, one study compared walking barefoot and with firm shoes. The use of shoes merely improved the gait line of the nonaffected limb, while the gait line of the nonaffected lower limb, relative cycle parameters, and symmetry ratios were comparable in both conditions.⁵ Ankle excursions and muscular activity, however, were not assessed.

In conclusion, the use of the caliper effected a more dynamic and balanced gait irrespective of any change of gait speed. Gait was characterized by better ankle excursions, faster loading of the paretic limb, longer relative single-stance duration of the paretic lower limb, better swing symmetry, and better pivoting over the stationary paretic foot. Furthermore, the functional activity of the paretic quadriceps muscles increased, while there was no overactivity of the plantar flexors. These effects are predominantly beneficial and support the correction of spastic ankle plantar flexion and inversion with the Valens caliper as an integral part of a comprehensive rehabilitation approach; however, the long-term effects, particularly on the hemiparetic tibialis anterior, need further study.

Acknowledgments

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