Functional Electrical Stimulation of Ankle Plantarflexor and Dorsiflexor Muscles

Effects on Poststroke Gait

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Background and Purpose—Functional electrical stimulation (FES) is a popular poststroke gait rehabilitation intervention. Although stroke causes multijoint gait deficits, FES is commonly used only for the correction of swing-phase foot drop. Ankle plantarflexor muscles play an important role during gait. The aim of the current study was to test the immediate effects of delivering FES to both ankle plantarflexors and dorsiflexors on poststroke gait.

Methods—Gait analysis was performed as subjects (N=13) with chronic poststroke hemiparesis walked at their self-selected walking speeds during walking with and without FES.

Results—Compared with delivering FES to only the ankle dorsiflexor muscles during the swing phase, delivering FES to both the paretic ankle plantarflexors during terminal stance and dorsiflexors during the swing phase provided the advantage of greater swing-phase knee flexion, greater ankle plantarflexion angle at toe-off, and greater forward propulsion. Although FES of both the dorsiflexor and plantarflexor muscles improved swing-phase ankle dorsiflexion compared with noFES, the improvement was less than that observed by stimulating the dorsiflexors alone, suggesting the need to further optimize stimulation parameters and timing for the dorsiflexor muscles during gait.

Conclusions—In contrast to the typical FES approach of stimulating ankle dorsiflexor muscles only during the swing phase, delivering FES to both the plantarflexor and dorsiflexor muscles can help to correct poststroke gait deficits at multiple joints (ankle and knee) during both the swing and stance phases of gait. Our study shows the feasibility and advantages of stimulating the ankle plantarflexors during FES for poststroke gait. **(*Stroke*. 2009;40:00-00.)**

Key Words: functional electrical stimulation ■ variable-frequency trains ■ ankle plantarflexors

The combination of functional electrical stimulation (FES) and treadmill training is a novel and effective intervention for poststroke gait rehabilitation.[1,2] In a recent randomized, controlled trial, the FES approach of stimulating multiple muscles provided a significant therapeutic benefit when added to a comprehensive poststroke gait training program.[1] However, in poststroke individuals, FES is typically delivered only to ankle dorsiflexors to correct "foot drop" during the swing phase,[3–5] thereby failing to address other important stance and swing-phase poststroke gait deficits at the hip and knee. We recently showed that the traditional approach of delivering FES to ankle dorsiflexor muscles during gait successfully corrects swing-phase foot drop but also produces "adverse effects" on poststroke gait, such as decreased swing-phase knee flexion and decreased ankle plantarflexion at toe-off.[6] Also, decreased propulsive force generation at the transition from paretic stance to swing has recently been shown to represent a critical deficit in poststroke gait, contributing to a greater energy cost of walking, and has been shown to be correlated with hemiparetic severity, walking speed, and gait asymmetry in individuals after stroke.[7,8] Dorsiflexor FES does not address this critical deficit of decreased propulsive force generation during paretic terminal stance.[7–9]

The ankle plantarflexor muscles, which play an important role in the gait cycle,[10] demonstrate decreased force-generating ability in individuals with hemiparesis after stroke. Decreased force generation by the ankle plantarflexors is related to deficits during both the swing and stance phases of gait[9,11] and most notably, to slow walking speeds after stroke.[9] Forward dynamic simulations of healthy gait predict that decreased forward propulsive force generation by the ankle plantarflexors during terminal stance can lead to decreased ipsilateral leg kinetic energy at toe-off, thereby resulting in decreased swing-phase knee flexion.[10,12] Although predictions from gait simulations[10,12] purport the relations between ankle plantarflexor muscle activation, propulsive force generation, and swing-phase knee flexion, no

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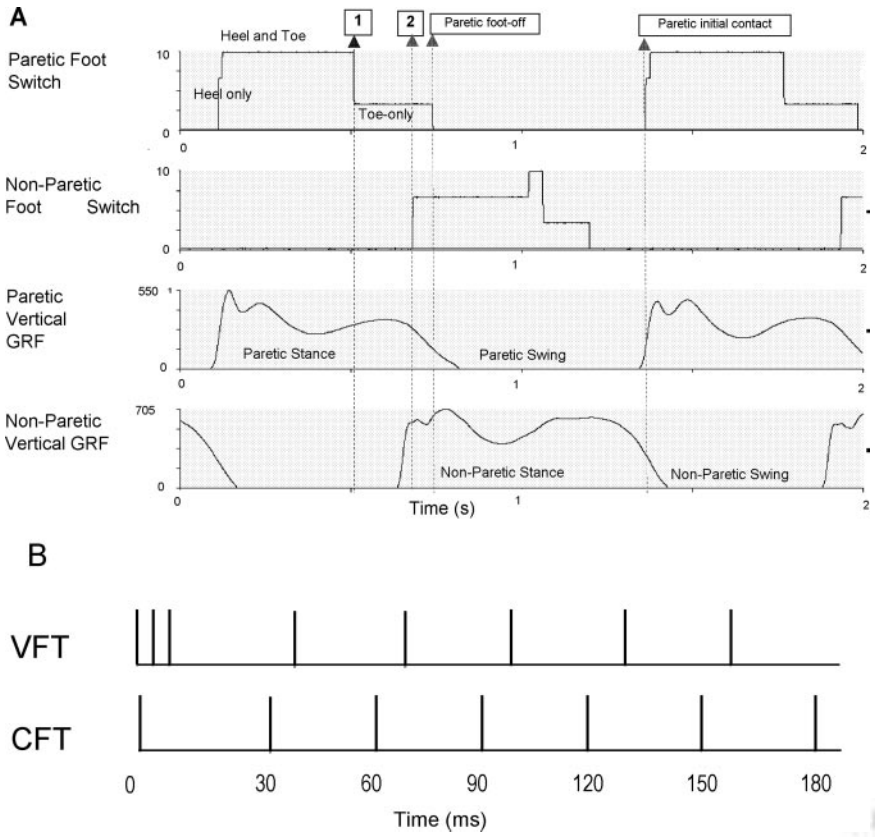

Figure 1. A, Footswitch and vertical GRF data from the paretic and nonparetic leg of 1 subject. There was considerable variability in the relative timing of the events shown in this figure across subjects. Vertical GRFs were used to identify swing and stance phases of gait. Footswitch signals were used to trigger FES during gait. During each gait cycle, dorsiflexor FES was started at paretic foot-off and terminated at paretic initial contact. Plantarflexor FES was started at paretic heel-off (arrow 1) for logic 1 and at nonparetic initial contact (arrow 2) for logic 2 and terminated at paretic foot-off. B, Stimulation train patterns used for FES. VFTs consisting of a 3-pulse, 200-Hz burst at the start of a 30-Hz CFT were used in this study. Traditionally used CFTs are shown for comparison.

previous study has provided direct experimental evidence demonstrating these relations. Also, in the current study, we used novel stimulation patterns called variable-frequency trains (VFTs), which have been shown to enhance isometric muscle performance,[13] and, more recently, gait performance,[6] compared with typically used stimulation patterns (constant-frequency trains, CFTs; Figure 1).

The aim of this study therefore was to compare poststroke patterns during walking with FES delivered to both the plantarflexor and dorsiflexor muscles versus walking with FES delivered to the ankle dorsiflexors only.

Patients and Methods

Thirteen patients (age range, 49 to 72 years; 9 men) with poststroke hemiparesis participated in this study (Table). Inclusion criteria were as follows: 6 months elapsed time after a stroke involving cerebral cortical regions, ability to walk for 5 minutes at their self-selected walking speed, and sufficient passive ankle dorsiflexion range of motion to enable the paretic ankle joint to reach either neutral ankle angle (0°) or a minimum of 5° of plantarflexion with the knee flexed. Exclusion criteria were substantial cognitive deficits, severe aphasia, cerebellar involvement, and any preexisting conditions affecting walking function. All subjects signed informed-consent forms approved by the human subjects review board of the University of Delaware.

Self-adhesive surface electrical stimulation electrodes were attached over the ankle dorsiflexor (2″×2″; TENS Products, Grand Lake, Colo) and plantarflexor (2″×5″; ConMed Corp, New York, NY) muscles. A Grass S8800 stimulator in combination with a Grass model SIU8TB stimulus isolation unit was used to deliver electrical stimulation (Grass Instrument Co, Quincy, Mass). For both the dorsiflexor and plantarflexor muscles, stimulation amplitude was set at a 300-ms long, 30-Hz train with a pulse duration of 300 μs.

For the ankle dorsiflexor muscles, stimulation amplitude was set with the subjects seated to achieve a neutral ankle joint position (0°) with minimal ankle eversion or inversion. For the ankle plantarflexor muscles, stimulation amplitude was set with the subjects standing in a position similar to terminal double support of the paretic leg during gait to achieve lifting of the paretic heel off the ground or until the subject's maximal tolerance was reached, whichever occurred first. Two compression-closing footswitches (25-mm diameter MA-153; Motion Laboratory Systems Inc, Baton Rouge, La) were attached bilaterally to the soles of each subject's shoes, 1 under the forefoot and the second under the hindfoot.

A customized, real-time FES system (CompactRIO, National Instruments, Austin, Tex) was used to control the Grass stimulator and deliver stimulation during the gait cycle. The FES system delivered stimulation to the ankle dorsiflexor muscles from the time the forefoot footswitch of the paretic leg was off the ground until the hindfoot footswitch of the paretic leg contacted the ground (Figure 1A). Plantarflexor FES was triggered by the use of 2 different timing logics (logics 1 and 2) to stimulate the ankle plantarflexors during the paretic terminal double-support phase of gait. Logic 1 delivered stimulation from the time the paretic hindfoot switch was off the ground until the time the paretic forefoot switch was off the ground. Logic 2 delivered stimulation from the time the nonparetic forefoot switch contacted the ground until the time when the paretic forefoot switch of the paretic leg was off the ground. The FES stimulation pattern comprised a high-frequency (200-Hz) 3-pulse burst[13] followed by a lower-frequency (30-Hz), constant-frequency train (Figure 1B).

Gait analysis was performed as subjects walked on a split-belt treadmill instrumented with two, 6° of freedom force platforms (AMTI, Watertown, Mass). For safety, subjects held onto a front handrail during walking, and all subjects wore a harness that was attached to an overhead support. No body weight was supported by the harness. Marker data were collected at 100 Hz with an 8-camera motion-analysis system (Vicon 5.2, Oxford, England) that was synchronized with the force plates that measured ground reaction forces (GRFs) from each leg separately at 2000 Hz.

Table. Demographic and Clinical Information for the 13 Subjects Tested in the Present Study

Subject No.	Sex	Age, y	Time Since Stroke, y	Side of Hemiparesis, L/R	Gait Speed, m/s	Fugl-Meyer Score (Maximum=34)	PDF Logic No.
1	M	66	2.4	L	0.9	24	L1
2	M	52	6.3	L	0.6	20	L1
3	F	58	21.3	L	0.2	23	…
4	F	51	1.9	L	0.3	20	L2
5	M	49	9.3	R	0.9	28	L2
6	M	72	6.1	R	0.5	18	L2
7	M	57	2.7	R	0.7	22	L1
8	M	58	9.9	R	0.7	21	L2
9	M	60	5.8	R	0.8	25	L1
10	M	74	4.7	R	0.7	31	L2
11	M	56	9.8	R	1.2	25	L2
12	F	46	2.2	L	0.9	23	L2
13	F	66	1.4	R	0.3	18	
Mean		58.8	6.4		0.7	23	
SE		2.5	1.6		0.1	1	

Data for subject 13 were not included in the results because of technical problems during gait analysis. The last column states whether logic 1 (L1) or logic 2 (L2) was used for plantarflexor stimulation during gait (see Methods for details).

The data presented in this report are a subset of the data collected during 1 testing session for each subject; multiple walking trials were collected within the session.[6] Rest intervals of 5 to 10 minutes were provided between consecutive walking trials; each trial was 20 to 40 seconds long.[6] The current study reports data from 3 walking trials tested at the subject's self-selected walking speeds: (1) walking without FES (noFES), (2) walking with FES delivered to the ankle dorsiflexor muscles during the swing phase with VFTs (DF), and (3) walking with FES delivered to both the ankle dorsiflexor and plantarflexor muscles with VFTs (PDF). The noFES walking trials were collected at the beginning, middle, and end of the testing session. The order of testing of each of the 2 walking conditions with FES was randomly assigned to each subject.

Data Processing

Marker trajectories and GRF data were lowpass filtered (Butterworth fourth order, phase lag) at 6 and 30 Hz, respectively, with the use of commercial software (Visual 3D; C-Motion, Rockville, Md). Vertical GRFs were used to determine gait events. The noFES data were obtained by averaging the 3 noFES trials from the beginning, middle, and end of the session. For the PDF condition for each subject, the timing logic (logic 1 vs logic 2) that generated greater peak anterior GRF (AGRF) was included in the analysis.

Dependent Variables

Peak AGRF during paretic terminal stance was defined as the maximum AGRF between the onset of the propulsion (anteriorly directed) phase of anteroposterior GRFs and the end of the stance phase. Percent paretic propulsion was defined as the ratio of the integral of the AGRF from the onset of propulsion through the end of the stance phase for the paretic leg versus the total AGRF integral for the paretic and nonparetic legs.[7] Peak knee flexion during the swing phase was determined for the paretic leg. Ankle angle at paretic toe-off and peak ankle angle during the paretic swing phase were also measured for the paretic leg.

Statistical Analysis

Paired *t* tests were performed to detect differences between noFES versus PDF, DF versus PDF, and when applicable, noFES versus DF. Gait data for the DF walking condition have been reported previously[6] but are included in the present study for comparison with PDF. A 1-way ANOVA with post hoc paired comparisons for peak AGRF and peak swing-phase knee flexion angles during the noFES walking condition at the beginning, middle, and end of the session was performed to assess the presence of muscle fatigue or potentiation in ankle plantarflexor muscles. All statistical analyses were performed with SPSS 16.0 (SPSS Inc., Chicago, Ill). Descriptive statistics are presented as mean±SE.

Results

Of the 13 subjects tested in this study, 1 subject's data (subject 13) were excluded from analysis because of technical problems during data collection. Results are presented for the remaining 12 subjects (see Table 1). GRF data for 1 of the remaining 12 subjects (subject 4) were not included in the analyses because the subject showed no forward propulsion (anteriorly directed GRFs) during gait. We found that accurate timing of delivery of plantarflexor FES during gait was challenging; having 2 timing logic options for plantarflexor FES enabled us to deliver plantarflexor FES to all of the subjects included in our study. For each subject, data for all outcome variables represent the mean of data from 6 consecutive strides.

Compared with walking without FES (8.9±1.6% body weight), the peak AGRF for the paretic leg showed an ≈18% increase during walking with PDF (P=0.025, Figure 2). There was no difference in peak AGRF between DF and PDF (P=0.25). Peak AGRF value for the nonparetic extremity was 15.9±2.2%.

During walking without FES, the average contribution of the paretic leg to total propulsion was 28.8±5.4%. In normal gait, each leg would contribute equally (ie, 50%) to total propulsion. Compared with noFES, the percent paretic propulsion increased to 33.1±5.2% during PDF (P=0.02; Figure 2). There was no difference in percent propulsion between PDF and DF (31.1±4.2%, P=0.047). The values for paretic push-off integrals during noFES, DF, and PDF were 2.1±0.4,% 2.3±0.4,% and 2.7±0.5% body-weight–seconds,

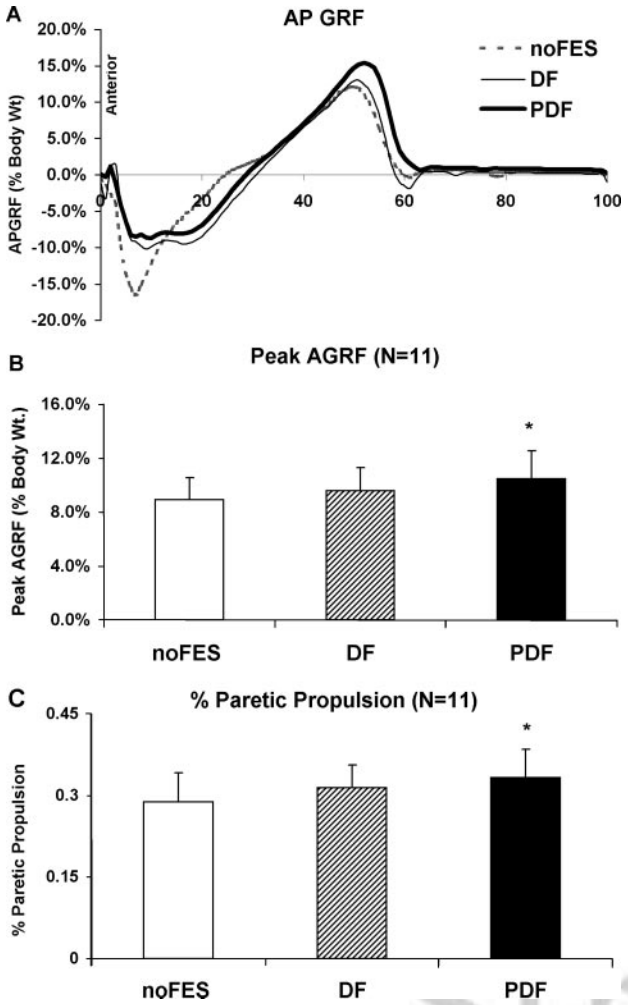

Figure 2. A, Anteroposterior GRF (APGRF) during the gait cycle for 1 representative subject. Means (n=11) and SEs for (B) peak AGRF and (C) percent paretic propulsion. Walking conditions presented are noFES, DF, and PDF. *Significant difference from noFES ($P \leq 0.05$).

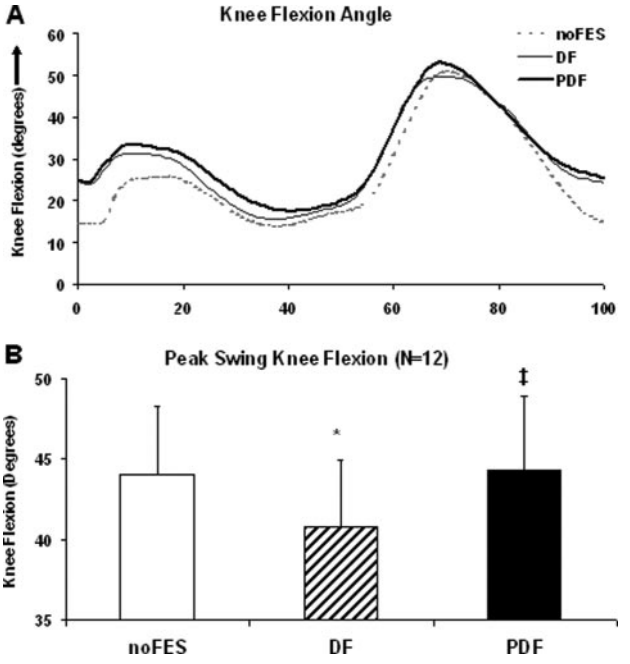

Figure 3. A, Sagittal plane knee angles during the gait cycle for 1 representative subject. B, Means (n=12) and SEs for peak swing-phase knee flexion angles during noFES, DF, and PDF. *Significant difference from noFES ($P \leq 0.05$). ‡Significant difference from DF ($P \leq 0.05$).

respectively. The values for the nonparetic push-off integrals during noFES, DF, and PDF were 5.1±0.7,% 5.0±0.8,% and 5.4±0.9% body-weight–seconds, respectively.

We detected a significant reduction in peak swing-phase knee flexion angles during DF (40.8±4.2°) compared with each of the remaining 2 walking conditions (both probability values ≤0.01, Figure 3). There were no differences in peak swing-phase knee flexion between noFES (44.1±4.2°) and PDF (44.3±4.6°, $P=0.81$). The peak swing-phase knee flexion angle for the nonparetic extremity was 68.3±2.0°.

During walking without FES, the average ankle plantarflexion angle at paretic toe-off was −9.2±1.2°. There was significant reduction in ankle plantarflexion at toe-off during DF stimulation[6] (−3.1±1.5°) compared with each of the other 2 walking conditions (both probability values ≤0.01, Figure 4). There were no differences in ankle angle at toe-off between noFES and PDF (−8.3±1.3°, $P=0.41$). The ankle angle at toe-off for the nonparetic extremity was −13.4±2.1°.

Without FES, subjects walked with their paretic ankle angles in a slightly plantarflexed position during the swing phase (−2.9±1.2°, Figure 4). Dorsiflexor FES brought the paretic ankle into a dorsiflexed position during the swing phase (2.1±1.5°). Compared with noFES, the paretic ankle demonstrated significantly greater peak swing-phase ankle dorsiflexion during walking with PDF (−1.2±1.0°, $P=0.04$). PDF produced a significant reduction in ankle dorsiflexion compared with DF (2.1±1.5°, $P \leq 0.01$). The peak swing-phase ankle angle for the nonparetic extremity was 5.2±2.0°.

Effects of fatigue were tested by comparing noFES trials collected at the beginning, middle, and end of the testing session. There were no significant differences in peak AGRF among the 3 noFES walking trials collected at the beginning, middle, and end of the testing session (8.5±1.3%, 8.3±1.2%, and 8.7±1.3%, respectively; $F=0.26$, $P=0.77$). Similarly, there were no significant differences in peak swing-phase knee flexion angles among the noFES walking trials collected at these 3 time points (48.7±3.2°, 49.7±3.6°, and 50.3±3.6°, respectively; $F=0.96$, $P=0.41$).

Discussion

Our results showed that delivering FES to both the paretic ankle plantarflexor muscles during terminal stance and to the ankle dorsiflexor muscles during the swing phase provided the advantage of increased swing-phase knee flexion and increased plantarflexion angles at toe-off compared with the traditional approach of stimulating the ankle dorsiflexors alone. Stimulating both the plantarflexor and dorsiflexor muscles also resulted in greater peak AGRFs, percent contribution of the paretic leg to total propulsion, and swing-phase ankle dorsiflexion angles compared with noFES. Although FES of both the dorsiflexor and plantarflexor muscles was able to improve ankle dorsiflexion during the swing phase compared with noFES, stimulating both muscles produced

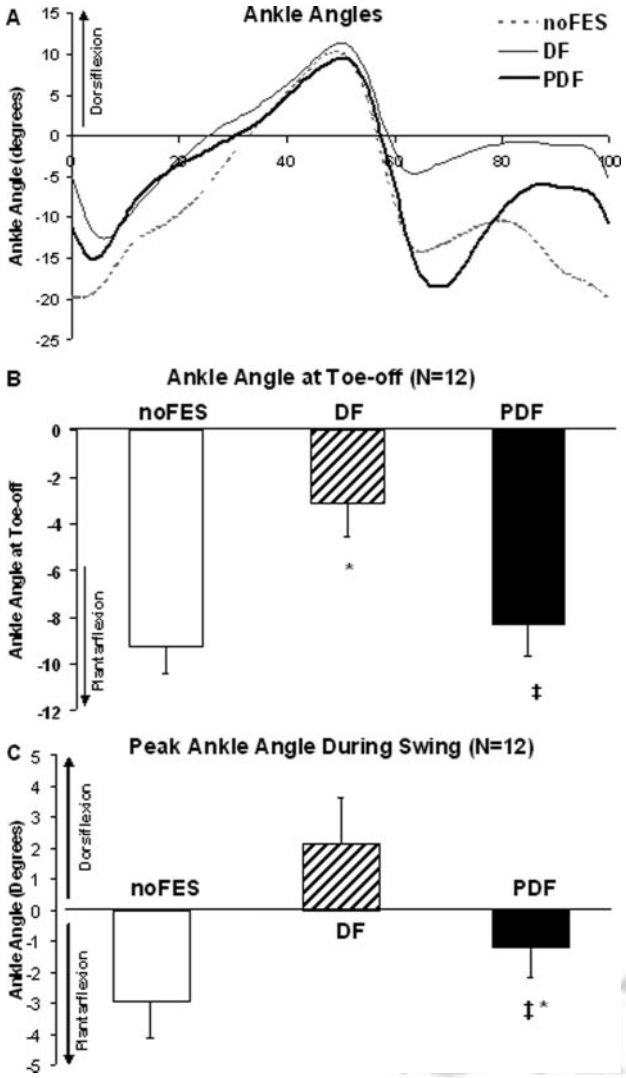

Figure 4. A, Sagittal plane ankle angles during the gait cycle for 1 representative subject. Means (n=12 subjects) and SEs for ankle angles at paretic toe-off (B) and peak swing-phase ankle dorsiflexion (C) during noFES, DF, and PDF. *Significant difference from noFES ($P \leq 0.05$). ‡Significant difference from DF ($P \leq 0.05$). Negative angles represent plantarflexion.

less peak dorsiflexion during the swing phase compared with stimulating the dorsiflexors alone. Additionally, we showed the feasibility of using novel physiologically based VFT patterns[13] during FES.

Forward dynamic gait simulations of able-bodied individuals predict that during terminal stance, the gastrocnemius muscle plays an important role in generating push-off force, thereby providing the critical propulsion needed for initiation of the swing phase of gait.[9,10,14] In support of previous findings by Bowden et al,[7] our current results showed that during walking without FES, the average contribution of the paretic leg toward total propulsion was 28.1±5.5%, indicating marked asymmetry in the amount of propulsion generated by the paretic compared with the nonparetic leg (perfect symmetry would result in 50% contribution by each leg). Previous researchers have suggested that decreased paretic plantarflexion muscle force generation at terminal stance is related to the inadequate paretic leg propulsion during walking after stroke, although this has never been directly tested.[9,15] Thus, we hypothesized that FES delivered to the paretic plantarflexor muscles during terminal stance would increase propulsive force generation. Our results demonstrated that, as hypothesized, adding FES to the ankle plantarflexor muscles in poststroke individuals who were receiving dorsiflexor FES resulted in an ≈18% increase in both the peak AGRF and percent paretic propulsion compared with walking without FES. Our present results provide the first direct evidence in poststroke individuals that forward propulsion produced by the paretic leg during walking can be positively influenced by electrically stimulating the paretic plantarflexor muscles during terminal stance.

On the basis of predictions of forward dynamic gait simulations,[10,12] we hypothesized that delivering FES to the paretic ankle plantarflexors during terminal stance would generate greater forward propulsive forces, which in turn would provide the paretic leg with greater kinetic energy at toe-off, thereby increasing paretic knee flexion during the swing phase.[10,12] No previous studies reported changes in swing-phase knee flexion as a result of plantarflexor or dorsiflexor FES. In the present study, because of foot drop during the swing phase, we were unable to assess the effects of delivering FES to only the plantarflexor muscles on gait performance; both the ankle muscles were stimulated. Much to our surprise, the traditional approach of delivering FES to the ankle dorsiflexor muscles only during swing (DF condition) resulted in an ≈8% (3.3°) decrease in mean swing-phase knee flexion compared with walking without FES.[6] However, delivering FES to both the ankle flexor muscles resulted in an 8.6% increase in average swing-phase knee flexion compared with delivering FES to ankle dorsiflexor muscles alone, thus counteracting the effects of DF stimulation. Thus, the increase in swing-phase knee flexion produced by plantarflexor FES was able to counteract the decrease in swing-phase knee flexion produced by dorsiflexor FES. This is a critically important finding, given that chronic stroke survivors generally show decreased knee flexion in the paretic leg during swing compared with the nonparetic leg and compared with neurologically unimpaired control subjects walking at matched speeds.[16,17]

Another advantage of delivering FES to both the plantarflexor and dorsiflexor muscles instead of the dorsiflexors alone was observed at the ankle joint as the paretic leg transitioned from stance to swing phase. As shown in our previous study[6] and as reported by others,[18] compared with noFES, dorsiflexor FES reduced ankle plantarflexion angles at toe-off. However, in the present study, just as was found for peak knee flexion, delivering FES to both flexors counteracted the decreased ankle plantarflexion caused by dorsiflexor FES alone at toe-off. These results, together with the results for peak knee flexion during swing, suggest that the application of DF stimulation alone may negatively impact important aspects of the gait pattern in chronic stroke survivors, but that this impact may be overcome by the addition of plantarflexion stimulation.

In contrast to the improvements in AGRF during terminal stance, knee flexion during the swing phase and ankle

plantarflexion at toe-off with plantarflexor and dorsiflexor FES, the latter (PDF) resulted in less ankle dorsiflexion during the swing phase compared with dorsiflexor FES. The dorsiflexor muscles were stimulated with the same stimulation intensity during PDF and DF walking conditions. Thus, given the increased plantarflexion at toe-off facilitated by the plantarflexion stimulation during PDF versus DF, the same dorsiflexor stimulation would have to generate a greater ankle excursion to achieve the same peak swing-phase ankle angle. Greater dorsiflexor stimulation intensity would therefore be needed to generate the same peak swing-phase dorsiflexion during PDF versus DF. Increasing the dorsiflexor muscle's stimulation intensity when both muscles are being stimulated can help the dorsiflexor muscles to overcome the opposing plantarflexor force and thereby increase the degree of swing-phase dorsiflexion produced during plantarflexor and dorsiflexor FES. Furthermore, the timing of onset of dorsiflexor FES during gait can be modified to minimize plantarflexor-dorsiflexor muscle cocontraction during FES.

In our current study, the participants held onto a handrail during treadmill walking, Only 2 recent studies investigated the effect of handrail hold during treadmill walking.[15,19] In a group of able-bodied individuals, Siler and colleagues[19] showed no effect of handrail hold on sagittal plane gait kinematics during treadmill walking. In a group of individuals with poststroke hemiparesis, Chen and colleagues[15] showed that although handrail hold did not significantly affect paretic swing time, swing knee flexion, and leg kinetic energy at toe-off, it resulted in decreased leg kinetic energy at toe-off for the nonparetic leg.[15] Thus, on the basis of limited evidence, we can infer that percent propulsion, which is computed from push-off integral values for both the paretic and nonparetic legs, may be a variable influenced by handrail hold. However, in the current study, all data were collected within 1 testing session, and we attempted to keep the type and strength of handrail hold constant throughout the testing session. Also, given that all of the data were collected within 1 testing session, it is difficult to envision that changes in the nonparetic leg propulsion due to handrail hold would be differentially affected by FES. Nevertheless, the handrail hold maybe a potential limitation in our study. Future studies are needed to systematically study the effect of handrails on gait kinetics during treadmill walking.[15,19]

Recent randomized, controlled trials and meta-analyses concluded that dorsiflexor FES can produce improvements in walking speed[5,20] and energy efficiency.[21,22] It is noteworthy, however, that with the exception of a recent study by Daly and colleagues,[1] the majority of the studies and controlled trials that reported improvements in poststroke walking speed with FES stimulated only the dorsiflexors during swing[4,5,20,21] and did not investigate the effect of DF stimulation on other swing- and stance-phase poststroke gait deficits. Our results show that the traditional FES approach of stimulating ankle dorsiflexor muscles alone during the swing phase achieves correction of only the ankle dorsiflexor deficit during the swing phase and can result in reduced swing-phase knee flexion, which is already reduced after stroke. In contrast, delivering FES to both the flexor muscles can help to correct poststroke gait deficits at both the ankle and knee joints and during both the swing (knee flexion, ankle dorsiflexion) and stance (propulsive force generation, ankle plantarflexion at toe-off) phases of gait. The immediate effects of FES shown in our study suggest that FES strategies, similar to the one used in the present study, when used as a gait training intervention, may produce even greater improvements in gait performance compared with those obtained by stimulating the dorsiflexors alone.

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Disclosures

None.

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