Novel Patterns of Functional Electrical Stimulation Have an Immediate Effect on Dorsiflexor Muscle Function During Gait for People Poststroke

Trisha M. Kesar, Ramu Perumal, Angela Jancosko, Darcy S. Reisman, Katherine S. Rudolph, Jill S. Higginson, Stuart A. Binder-Macleod

Background. Foot drop is a common gait impairment after stroke. Functional electrical stimulation (FES) of the ankle dorsiflexor muscles during the swing phase of gait can help correct foot drop. Compared with constant-frequency trains (CFTs), which typically are used during FES, novel stimulation patterns called *variable-frequency trains* (VFTs) have been shown to enhance isometric and nonisometric muscle performance. However, VFTs have never been used for FES during gait.

Objective. The purpose of this study was to compare knee and ankle kinematics during the swing phase of gait when FES was delivered to the ankle dorsiflexor muscles using VFTs versus CFTs.

Design. A repeated-measures design was used in this study.

Participants. Thirteen individuals with hemiparesis following stroke (9 men, 4 women; age=46–72 years) participated in the study.

Methods. Participants completed 20- to 40-second bouts of walking at their self-selected walking speeds. Three walking conditions were compared: walking without FES, walking with dorsiflexor muscle FES using CFTs, and walking with dorsiflexor FES using VFTs.

Results. Functional electrical stimulation using both CFTs and VFTs improved ankle dorsiflexion angles during the swing phase of gait compared with walking without FES ($\bar{X}\pm SE=-2.9°\pm1.2°$). Greater ankle dorsiflexion in the swing phase was generated during walking with FES using VFTs ($\bar{X}\pm SE=2.1°\pm1.5°$) versus CFTs ($\bar{X}\pm SE=0.3\pm1.3°$). Surprisingly, dorsiflexor FES resulted in reduced knee flexion during the swing phase and reduced ankle plantar flexion at toe-off.

Conclusions. The findings suggest that novel FES systems capable of delivering VFTs during gait can produce enhanced correction of foot drop compared with traditional FES systems that deliver CFTs. The results also suggest that the timing of delivery of FES during gait is critical and merits further investigation.

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Stroke is a leading cause of long-term adult disability.[1] Regaining walking function is one of the primary concerns for individuals who experience stroke.[2] Even after rehabilitation, residual gait deficits are prevalent in individuals with stroke.[2] Foot drop is a common post-stroke gait impairment estimated to affect 20% of survivors of stroke.[3] Foot drop is caused by total or partial paresis of ankle dorsiflexor muscles,[4] makes ground clearance difficult during swing, and can lead to inefficient gait compensations such as circumduction and hip hiking (increased hip abduction in the unaffected limb during stance, with simultaneous elevation of the affected side of the pelvis during swing).[5,6] Residual gait deficits such as foot drop contribute to increased energy expenditure during gait, decreased endurance, and an increased incidence of falls.[2,5–8]

Ankle-foot orthoses (AFOs) are widely prescribed for the management of foot drop.[9,10] Functional electrical stimulation (FES) is another intervention that is used to deliver electrical stimulation to the ankle dorsiflexor muscles during the swing phase of gait to correct foot drop.[11–14] In contrast to AFOs, FES promotes active muscle contractions, can help improve muscle strength (force-generating capacity),[15–18] prevents disuse atrophy,[19–21] reduces muscle tone (velocity-dependent resistance to stretch) and spasms,[22] produces a more energy-efficient use of proximal limb muscles,[23] and aids in motor relearning.[24]

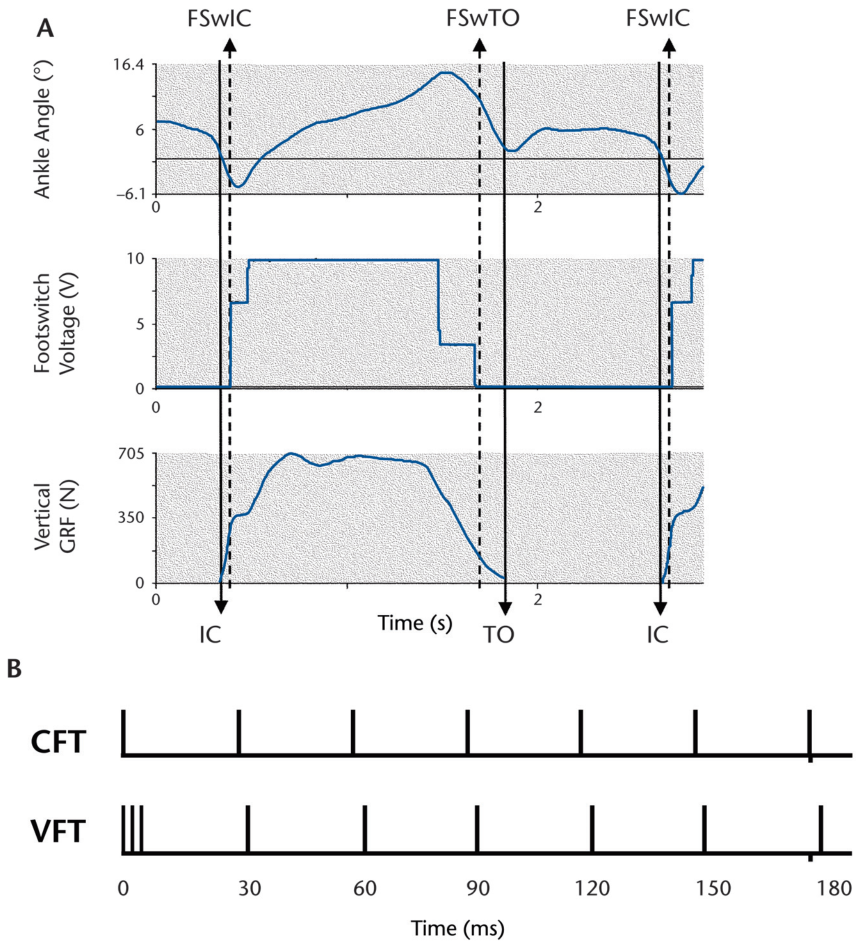

Figure 1.
(A) Example of the ankle angle, footswitch, and vertical ground reaction force (GRF) data from the paretic lower extremity of one representative participant. Data shown are for one complete gait cycle (ie, first initial contact [IC] to toe-off [TO] to second IC for the same leg, as determined using the vertical GRFs). Initial contact and toe-off, as determined by the footswitches, are depicted by FSwIC and FSwTO. Dorsiflexor stimulation was delivered during the swing phase of the paretic limb (ie, from FSwTO to FswIC). (B) A schematic depicting the 2 types of stimulation train patterns used for functional electrical stimulation in this study: constant-frequency trains (CFTs) and variable-frequency trains (VFTs). The CFTs consisted of single pulses (300-microsecond pulse duration) separated by 33-millisecond interpulse intervals. The VFTs consisted of a 200-Hz high-frequency burst at the start of a CFT with 33-millisecond interpulse intervals.

Functional electrical stimulation also has been shown to reduce the energy cost of walking poststroke.[17]

Although FES offers many advantages compared with AFOs, it has been well documented that stimulation parameters traditionally used during FES can contribute to limitations such as imprecise control of force and rapid muscle fatigue and prevent FES from gaining widespread clinical application.[25–30] Stimulation frequency (number of pulses per second) and intensity (amplitude or duration of individual pulses) are the 2 primary parameters that are modulated to control movements during FES. The stimulation pattern, or the arrangement of pulses within

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a stimulation train, is another parameter that can be varied during FES[27,31] (Fig. 1). However, typically, FES applications use only one type of stimulation pattern: stimulation trains consisting of stimulation pulses separated by constant interpulse intervals (*constant-frequency trains* [CFTs]).[32,33] Our laboratory[27,28,31,34] and others[35–37] have shown that novel stimulation patterns known as *variable-frequency trains* (VFTs) have several advantages compared with traditional CFTs.[31]

Variable-frequency trains can take advantage of the catchlike property of human muscles.[38] The catchlike property of skeletal muscle, first discovered in 1970 in mammalian motor units, is the force augmentation produced when an initial, brief, high-frequency burst of 2 to 4 pulses is included at the onset of a subtetanic low-frequency stimulation train.[31,38] Variable-frequency trains have been shown to enhance isometric[31,34] and nonisometric[39–41] muscle performance in healthy human quadriceps femoris muscles compared with CFTs of similar frequency, especially when muscles are fatigued. Variable-frequency trains also have been shown to produce greater knee joint excursions using fewer pulses than CFTs in healthy human quadriceps femoris muscles.[41] In addition to providing enhanced skeletal muscle performance, VFTs are a more physiologically based stimulation pattern compared with CFTs.[42] High-frequency doublets and triplets, such as those included at the onset of VFTs, have been reported to occur during animal and human muscle contractions.[38,42] Interestingly, no previous study has systematically compared the effects of delivering VFTs versus CFTs during FES to correct foot drop in individuals poststroke.

Global measures of walking performance, such as walking speed and physiological cost, typically are used as outcome measures in FES studies.[3,13,43–45] Although important to justify inclusion of FES in rehabilitation protocols, such measures do not provide a detailed biomechanical understanding of how specific aspects of gait are modified during walking with FES. There is a dearth of data in the literature about the immediate effects of FES on gait kinematics, kinetics, and gait compensations.[4,46] We posit that using instrumented gait analysis to study the immediate effects of FES on poststroke gait patterns can help develop better FES strategies to maximize the immediate (orthotic) effects of FES, which can enable the design of better neuroprostheses and potentially help to increase therapeutic benefits to patients. Thus, the purpose of this study was to compare the immediate effects of dorsiflexor FES using CFTs versus VFTs on gait kinematics in individuals who display gait impairments as a result of stroke.

Method

Setting and Participants

Thirteen individuals (9 men, 4 women; age=46–72 years) with poststroke hemiparesis participated in this study (Table). All participants had experienced a stroke involving cerebral cortical regions more than 6 months previously, were able to walk continuously for 5 minutes at their self-selected speed, and had sufficient passive ankle dorsiflexion range of motion to enable their paretic ankle joint to reach at least 5 degrees of plantar flexion with the knee flexed. Exclusion criteria included evidence of moderate to severe chronic white matter disease on magnetic resonance imaging, more than one previous stroke, congestive heart failure, peripheral artery disease with claudication, uncontrolled diabetes, shortness of breath without exertion, unstable angina, resting heart rate outside of the range of 40 to 100 bpm, resting blood pressure outside of the range of 90/60 to 170/90 mm Hg, substantial cognitive deficits (Mini-Mental State Exam score=22), inability to communicate with the investigators (severe aphasia), orthopedic conditions or pain in lower limbs or spine, cerebellar involvement (eg, ataxic hemiparesis), neglect (as assessed with the Star Cancellation Test), hemianopia, and absence of sensation on the skin of the calf or leg of the paretic limb.

Participants completed an initial clinical evaluation conducted by a licensed physical therapist comprising clinical tests for characterizing deficits following a stroke, including the lower-extremity portion of the Fugl-Meyer Assessment of Motor Recovery[47] (Table). Each participant's self-selected overground walking speeds was determined using the 6-meter walk test. All subjects signed informed consent forms approved by the Human Subjects Review Board of the University of Delaware.

Design Overview

Electrical stimulation. Surface electrical stimulation electrodes (5.08 × 5.08 cm [2 ×2 in])* were attached to the ankle dorsiflexor muscles. A Grass S8800 stimulator† in combination with a Grass Model SIU8TB stimulus isolation unit was used to deliver electrical stimulation. With participants seated and the foot hanging freely in a plantar-flexed position, the stimulation amplitude was set by gradually increasing the amplitude of a 300-millisecond-long, 30-Hz train with a pulse duration of 300 microseconds until a neutral ankle joint position (0°), or at least to 5 degrees of plantar flexion in participants with deficits of range of motion, was achieved. Electrode placement was adjusted to minimize ankle

* TENS Products Inc, PO Box 2089, Grand Lake, CO 80447.

† Grass Technologies, Div of Astro-Med Inc, 600 E Greenwich Ave, Warwick, RI 02893.

Table.
Participant Demographic and Clinical Information[a]

Participant No.	Sex	Age (y)	Time Since Stroke (y)	Side of Hemiparesis	Gait Speed (m/s)	Fugl-Meyer Score (Maximum Score=34)
1	M	66	2.4	L	0.9	24
2	M	52	6.3	L	0.6	20
3	F	58	21.3	L	0.2	23
4	F	51	1.9	L	0.3	20
5	M	49	9.3	R	0.9	28
6	M	72	6.1	R	0.5	18
7	M	57	2.7	R	0.7	22
8	M	58	9.9	R	0.7	21
9	M	60	5.8	R	0.8	25
10	M	74	4.7	R	0.7	31
11	M	56	9.8	R	1.2	25
12	F	46	2.2	L	0.9	23
13	F	66	1.4	R	0.3	18
Average		58.8	6.4		0.7	23
SD		8.6	5.4		0.3	3.8

[a] Data for participant 13 were not included in the results due to technical problems during data collection. M=male, F=female, L=left, R=right.

eversion and inversion during dorsiflexion.

Two compression-closing footswitches (25-mm diameter MA-153)[‡] attached bilaterally to the soles of each shoe, one on the forefoot under the fifth metatarsal head and the other on the hindfoot under the lateral portion of the heel, were used to control the timing of FES during gait.

Timing of FES during gait. A customized, real-time FES system (CompactRIO)[§] consisting of a real-time controller (NI cRIO-9004),[§] analog input module (NI 9210),[§] and digital input/output module (NI9401)[§] was used to deliver stimulation during gait.[48] The FES system delivered FES to the paretic ankle dorsiflexor muscles during the swing phase of each gait cycle, as detected by the footswitches (ie, from the time when the forefoot footswitch was off the ground to the time when the hindfoot footswitch contacted the ground) (Fig. 1). For FES using CFTs, a 30-Hz constant-frequency train was delivered. For FES using VFTs, a high-frequency (200-Hz) 3-pulse burst was delivered followed by a lower-frequency (30-Hz) CFT (Fig. 1B). All stimulation parameters for the VFTs and CFTs were identical except that the 200-Hz 3-pulse burst was included at the onset of the VFTs.

Marker placement. Retroreflective markers (14-mm diameter) placed bilaterally over the iliac crests, greater trochanters, lateral and medial femoral condyles, lateral and medial malleoli, and fifth metatarsal heads were used to define the joint centers of the lower limb. Elastic bands (SuperWrap)[‖] were tightly wrapped around the bilateral thigh and shank segments. Four-marker clusters attached to rigid thermoplastic shells were affixed to the elastic wraps and used to track the bilateral thigh and shank segments. A 3-marker cluster on the sacrum and 3 additional markers on the shoe were used to track pelvis and foot movements, respectively.

Gait analysis. During gait analysis, participants walked on a split-belt treadmill[#] instrumented with two 6-degree-of-freedom force platforms. Participants held on to a handrail during walking. All participants wore a harness that was attached to an overhead support for safety. No body weight was supported by the harness. Marker data were collected using an 8-camera motion analysis system (Vicon 5.2).[**] Video data were sampled at 100 Hz, and analog data (force platforms, footswitches, and stimulation channel) were sampled at 2,000 Hz.

‡ Motion Lab Systems Inc, 15045 Old Hammond Hwy, Baton Rouge, LA 70816.
§ National Instruments Inc, 11500 N Mopac Expwy, Austin, TX 78759.
‖ Fabrifoam, 900 Springdale Dr, Exton, PA 19341.
\# Advanced Mechanical Technology Inc, 176 Waltham St, Watertown, MA 02472-4800.
** Vicon, 14 Minus Business Park, West Way, Oxford, United Kingdom OX2 OJB.

The data presented are a subset of the data collected during one testing session. The complete testing session comprised ~18 trials, and each trial was 20 to 40 seconds in duration. Rest intervals of 5 to 10 minutes were provided between consecutive trials (Fig. 2). In this article, we report data for 3 walking trials or conditions: (1) walking without FES (noFES), (2) walking with dorsiflexor muscle FES using CFTs (FES-CFT), and (3) walking with dorsiflexor muscle FES using VFTs (FES-VFT).

The noFES data presented in this article were collected during the beginning (1st trial), middle (8th trial), and end (17th trial) of the session (Fig. 2). Data for different types of walking trials with FES of the ankle muscles at 2 different gait speeds were collected either during the 2nd through 7th walking trials (first block) or the 9th through 15th walking trials (second block) (Fig. 2). Within any one trial, the gait speed and the stimulation condition were kept constant. The 2 FES conditions presented in this study (ie, FES-CFT and FES-VFT) were randomly distributed across the first and second blocks within the testing session. All 3 walking conditions presented in this study were tested at the participants' self-selected overground walking speed, determined during a separate clinical testing session.

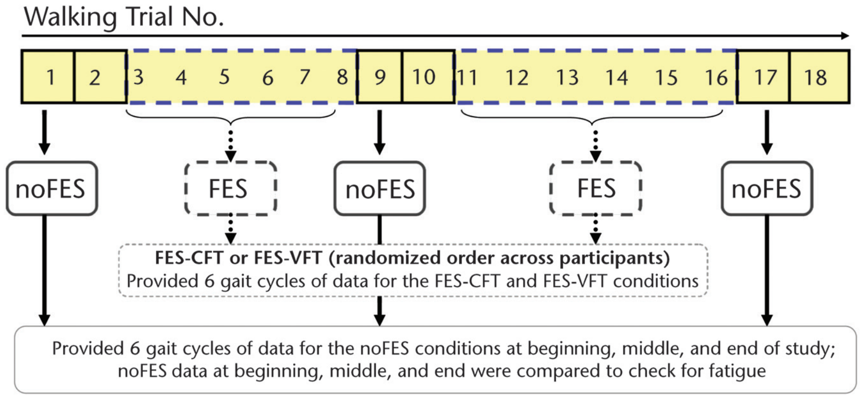

Figure 2.
Schematic showing the walking trials conducted as part of the experimental protocol. Each trial comprised 20 to 40 seconds of treadmill walking with a 5- to 10-minute rest provided between consecutive trials. The data for walking without functional electrical stimulation (noFES) used to compare with the data for walking with dorsiflexor muscle functional electrical stimulation using constant-frequency trains (FES-CFT) and walking with dorsiflexor muscle functional electrical stimulation using variable-frequency trains (FES-VFT) were obtained by averaging the noFES data obtained from the walking trials at the beginning, middle, and end of the session.

Data Processing

Marker trajectories and ground reaction force data were low-pass filtered (Butterworth fourth-order, phase lag) at 6 and 30 Hz, respectively, using commercial software (Visual 3D).[††] Lower-limb kinematics were calculated using rigid-body analysis and Euler angles using Visual 3D software. Vertical GRFs were used to identify initial contact and toe-off for the gait data, using a force threshold of 20 N. Strides were time normalized to 100% of the gait cycle and averaged across trials for each participant and walking condition. The noFES data were obtained by averaging the 3 noFES trials from the beginning, middle, and end of the session. For each of the 3 walking conditions, all outcome variables were computed using Visual 3D software for each stride and averaged across strides using a custom-written program.[§]

Outcome Variables

The 2 primary outcome measures for the effectiveness of FES during gait were peak ankle dorsiflexion angle during swing and ankle dorsiflexion angle at initial contact. Three secondary outcome measures were computed: peak flexion of the paretic knee during the swing phase, circumduction, and dorsiflexion angle of the paretic ankle at toe-off. *Circumduction* refers to a compensatory strategy during the swing phase that involves hip abduction and lateral (external) rotation and often is accompanied by pelvic hiking on the paretic side during the swing phase. Circumduction was defined as the maximum distance between the heel marker position during stance and the greatest lateral position of the heel marker during the subsequent swing phase.[49,50] Based on an exploratory analysis of the data, we included ankle dorsiflexion angle at toe-off as a secondary outcome variable to capture the interesting and surprising decrease in ankle plantar-flexion angle observed at the stance-to-swing transition.

Data Analysis

One-way analyses of variance (ANOVAs) for repeated measures were performed for each dependent variable to test for overall differences across the 3 walking conditions tested (noFES, FES-CFT, and FES-VFT). *Post hoc* pair-wise comparisons were performed to detect differences between noFES versus FES-CFT, noFES versus FES-VFT, and FES-CFT versus FES-VFT conditions. Based on our directional hypotheses, we planned to perform one-tailed paired t tests for ankle angle during swing and ankle angle at initial contact. Two-tailed paired t tests were

[††] C-Motion Inc, 15819 Crabbs Branch Way #A, Rockville, MD 20855.

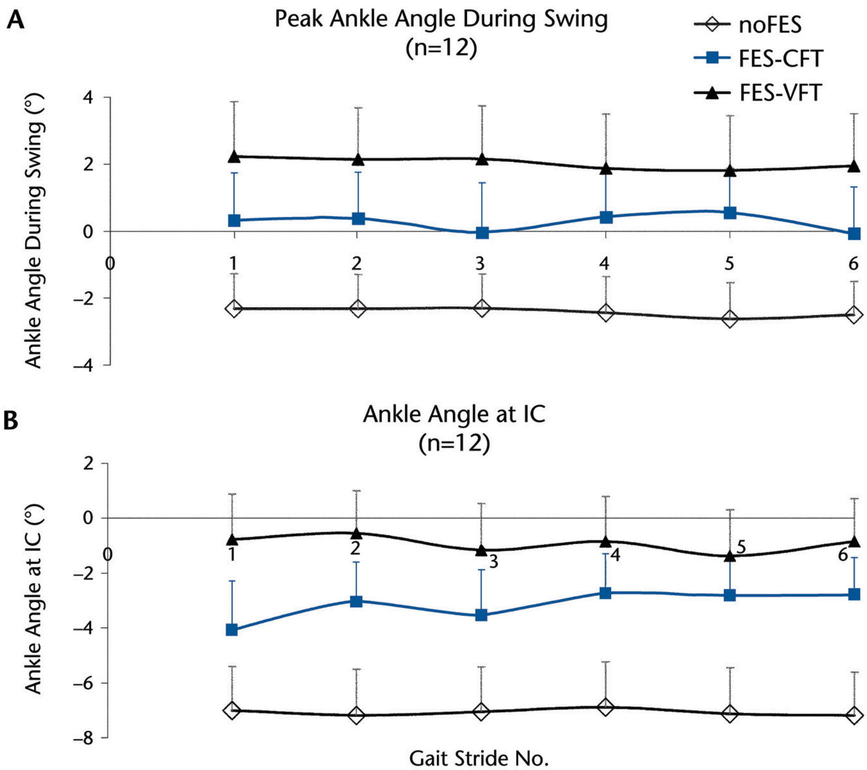

Figure 3.
Average (n=12) stride-by-stride values of the 2 primary outcome variables: (A) peak swing phase ankle angle and (B) ankle angle at initial contact for the 3 walking conditions tested (walking without functional electrical stimulation [noFES], walking with dorsiflexor muscle functional electrical stimulation using constant-frequency trains [FES-CFT], and walking with dorsiflexor muscle functional electrical stimulation using variable-frequency trains [FES-VFT]). Data shown are for the first 6 consecutive strides. Positive standard error bars are shown. Positive angles represent dorsiflexion. IC=initial contact.

planned for circumduction, peak swing phase knee flexion, and ankle angles at toe-off.

In addition, we compared the peak ankle angles during swing and ankle angle at initial contact measured during the noFES walking condition at the beginning, middle, and end of the session using a one-way ANOVA for repeated measures to assess the presence of either muscle fatigue or potentiation. A decrease in ankle dorsiflexion angles during swing or initial contact would signify the presence of fatigue within the testing session. A *post hoc* 2-tailed pair-wise comparison was performed to check for differences between noFES data at the beginning versus the end of the session. The alpha level was set at .05. The Shapiro-Wilk test was used to test for normal distribution of data for each of the outcome variables. All statistical analyses were performed using SPSS version 16.0.[‡‡]

Role of the Funding Source

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[‡‡] SPSS Inc, 233 S Wacker Dr, Chicago, IL 60606.

Results

Of the 13 participants tested in this study, one participant's data (participant 13) were excluded from the analyses due to technical problems encountered during the testing session. Results are presented for the remaining 12 participants (Table). For each condition, data for each outcome variable represent the mean of the first 6 consecutive gait cycles for which accurate forceplate data could be analyzed.

A qualitative analysis of stride-by-stride data showed that FES using VFTs consistently produced greater ankle dorsiflexion during swing and at initial contact compared with FES using CFTs and that FES using CFTs produced greater ankle angles compared with noFES for each of the 6 gait strides included in the analyses (Fig. 3). The ensemble gait data for the 12 participants' sagittal-plane ankle angles throughout the gait cycle showed that FES-CFT and FES-VFT shifted the ankle angle toward greater dorsiflexion during swing, at initial contact, and at toe-off (Fig. 4A). The descriptive statistics presented in the text are means±standard errors.

Primary Outcome Variables

Peak ankle angle during swing. Without FES, participants walked with their paretic ankle joints in a slightly plantar-flexed position (ankle angle=−2.9°±1.2°) during swing (Figs. 4A and 5A). The repeated-measures ANOVA detected significant differences in swing phase ankle angles among the 3 walking conditions tested ($F=14.73$, $P\leq.01$). Dorsiflexor FES using either CFTs or VFTs produced significant improvements in ankle dorsiflexion during swing compared with noFES (both $P \leq .01$). Functional electrical stimulation using CFTs brought the paretic ankle joint to an approximately neutral position during swing (0.3°±1.3°), and FES using VFTs produced significantly greater ankle dorsiflexion (2.1°±1.5°) compared with FES using CFTs ($P \leq .05$). The peak swing phase ankle angles for the nonparetic extremities were 5.2±2.0 degrees.

Ankle angle at initial contact. Overall, there were differences in ankle angle at initial contact among the 3 walking conditions tested ($F=16.92$, $P \leq .01$) (Figs. 4A and 5B). Our participants were in a markedly plantar-flexed position at initial contact (−8.5°±1.6°) when walking without FES. Functional electrical stimulation using either CFTs (−3°±1.4°) or VFTs (−1°±1.7°) significantly reduced the amount of ankle plantar flexion at initial contact compared with walking without FES (both $P \leq .01$). The *post hoc* pair-wise comparison detected a trend toward significantly improved dorsiflexion using VFTs versus CFTs ($P=.07$). The ankle angles at initial contact for the nonparetic extremities were 2.8±2.2 degrees.

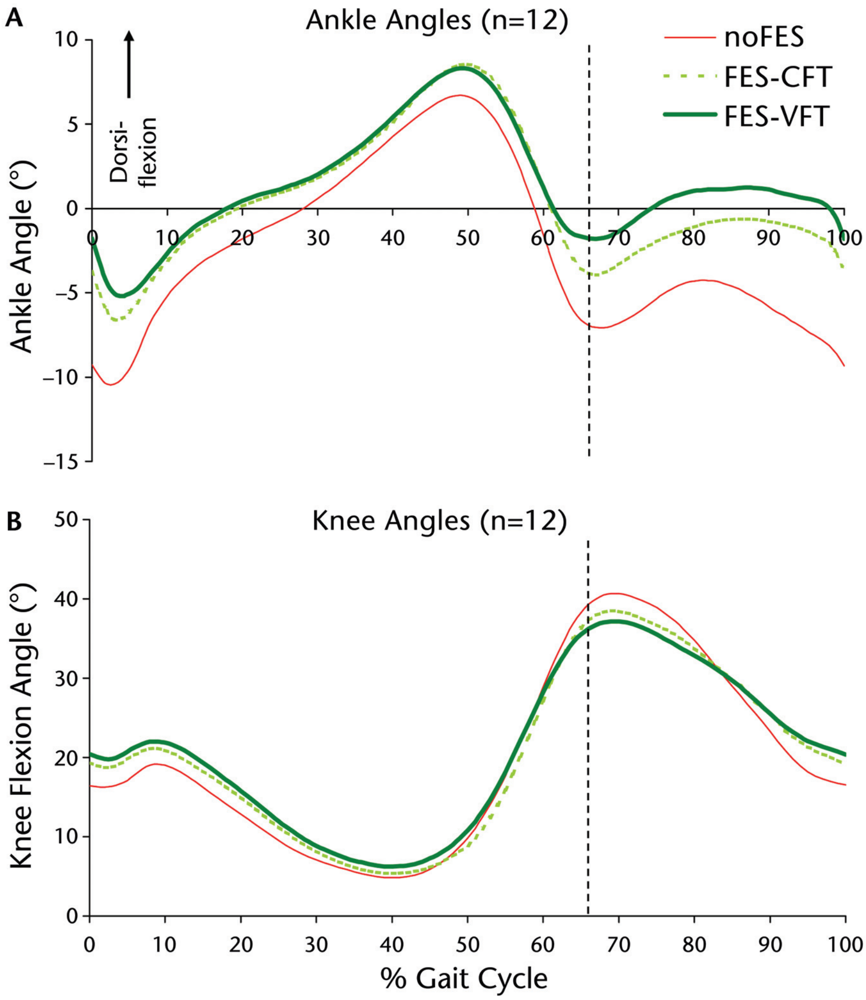

Figure 4.
Ensemble plots showing the average time normalized (0% to 100%=initial contact to initial contact) ankle (A) and knee (B) angles for 12 participants. Ankle dorsiflexion and knee flexion are positive. The 3 lines in each graph denote the 3 walking conditions tested: walking without functional electrical stimulation (noFES), walking with dorsiflexor muscle functional electrical stimulation using constant-frequency trains (FES-CFT), and walking with dorsiflexor muscle functional electrical stimulation using variable-frequency trains (FES-VFT). Vertical dashed line represents toe-off.

Secondary Outcome Variables

Although FES was delivered only to the muscles crossing the paretic ankle joint, there were significant differences in the peak knee flexion attained by the paretic leg during the swing phase among the 3 walking conditions ($F=13.9$, $P \leq .01$) (Figs. 4B and 6). Two-tailed pair-wise comparisons showed that participants demonstrated significantly reduced knee flexion during walking with FES using CFTs (42.6°±4.3°) or VFTs (40.8°±4.2°) compared with walking without FES (44.1°±4.2°) (both $P \leq .05$) (Fig. 6A). In addition, FES using VFTs produced a greater decrease in knee flexion compared with FES using CFTs ($P \leq .05$). The peak swing phase knee flexion angles for the nonparetic extremity were 68.3±2.0 degrees.

Participants demonstrated no differences in the amount of circumduction among the 3 walking conditions ($F=0.473$, $P=.63$). Average circumduction was 4.0±0.8 cm for noFES, 3.8±0.8 cm for FES-CFT, and 3.7±0.7 cm for FES-VFT. The circumduction values for the nonparetic extremities were 1.62±0.39 cm.

There were significant differences in the position of the paretic ankle angle at the stance-to-swing transition ($F=16.92$, $P \leq .01$) among the 3 walking conditions (Fig. 6B). The position of the paretic ankle angle at toe-off changed from a more plantar-flexed position during walking without FES (−9.1°±1.2°) to significantly less plantar flexion during walking with FES using CFTs (−5.2°±1.2°; $P \leq .01$) or VFTs

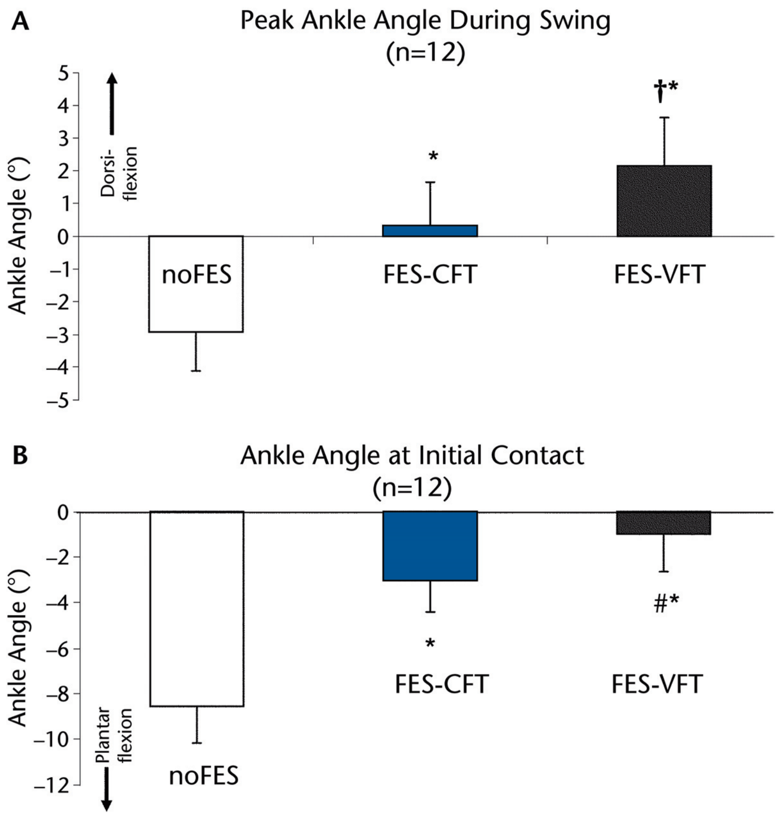

Figure 5.
Graphs showing average values and standard error bars for (A) peak ankle angles during swing and (B) ankle angles at initial contact for the 12 participants tested in our study. Three walking conditions shown are: walking without functional electrical stimulation (noFES), walking with dorsiflexor muscle functional electrical stimulation using constant-frequency trains (FES-CFT), and walking with dorsiflexor muscle functional electrical stimulation using variable-frequency trains (FES-VFT). *Significant difference from noFES ($P\leq.05$). †Significant difference from FES-CFT ($P\leq.05$). #Trend for difference from FES-CFT ($P=.07$).

($-3.1°\pm1.5°$; $P\leq.01$). Functional electrical stimulation using VFTs produced lesser plantar flexion at toe-off compared with FES using CFTs ($P\leq.01$). The ankle angles at toe-off for the nonparetic extremities were -13.4 ± 2.1 degrees.

Analysis of No FES Data

There were significant differences in peak ankle angles during swing among the 3 noFES walking trials at the beginning, middle, and end of the testing session ($F=5.13$, $P\leq.05$). Participants demonstrated significantly reduced peak swing phase ankle angles ($P\leq.05$) during the noFES walking trial at the end ($-3.8°\pm1.3°$) versus the beginning of the session ($-2.2°\pm1.1°$). Similarly, there were significant differences in ankle angles at initial contact among the noFES walk trials collected at the beginning, middle, and end of the session ($F=9.04$, $P\leq.01$). There was significantly less ankle dorsiflexion at initial contact during the noFES walking trials at the end ($-9.5°\pm1.6°$) versus the beginning ($-7°\pm1.8°$) of the session.

Discussion

Our results showed that during dorsiflexor FES, VFTs produced greater increases in ankle dorsiflexion during the swing phase compared with CFTs. Surprisingly, we also found that dorsiflexor FES decreased ankle plantar flexion at toe-off and decreased knee flexion during the swing phase, and these effects were enhanced by the use of VFTs versus CFTs. Although peak ankle dorsiflexion during swing improved with FES, there were no changes in circumduction during walking with FES. Additionally, analysis of the noFES data suggested that despite the rest intervals provided to the participants and the short (20- to 40-second) durations of the walking trials, the ankle dorsiflexor muscles may have experienced fatigue within one testing session.

This was the first study to compare the effects of using traditional (CFTs) versus novel (VFTs) stimulation patterns for dorsiflexor FES on poststroke gait. Interestingly, dorsiflexor FES using VFTs produced greater peak dorsiflexion angles during the paretic swing phase compared with FES using CFTs. Previous dynamometric studies on human and animal muscles showed greater rates of rise of force in response to VFTs versus CFTs during isometric contractions in nonfatigued muscle.[33,34,51] Furthermore, in fatigued muscles, VFTs have consistently been shown to generate greater rates of rise of force, peak forces, and force-time integrals during isometric contractions[51] and greater joint excursions and power during dynamic contractions.[31,40]

Our testing session was designed with 5- to 10-minute rest breaks between brief (20- to 40-second) walking trials to minimize fatigue. Nevertheless, our noFES walking data showed a significant decrement in ankle dorsiflexor angles from the beginning to the end of the session. The presence of fatigue in the dorsiflexor muscles could partly explain the enhanced ankle dorsiflexion seen with VFTs versus CFTs,[51] along with the enhanced rates of rise of force produced by the VFTs.[31] The observed muscle fatigue may be an important factor limiting gait performance during prolonged walking with FES.[29,30] These results pave the way for future studies investigating the use of VFTs during prolonged bouts of walking to test whether VFTs can help generate repeated foot drop correction for a greater number of steps compared with CFTs. We also found a trend (P=.07) toward greater dorsiflexion at initial contact (at the end of the swing phase) during FES using VFTs, despite the fact that the high-frequency burst that differentiated the VFTs from the CFTs occurred at the beginning of the swing phase. Evidently, the force enhancement produced by the high-frequency burst at the onset of the VFTs lasted long enough to generate some enhancement in ankle dorsiflexion throughout the paretic swing phase, further supporting the advantages of VFTs for FES applications.

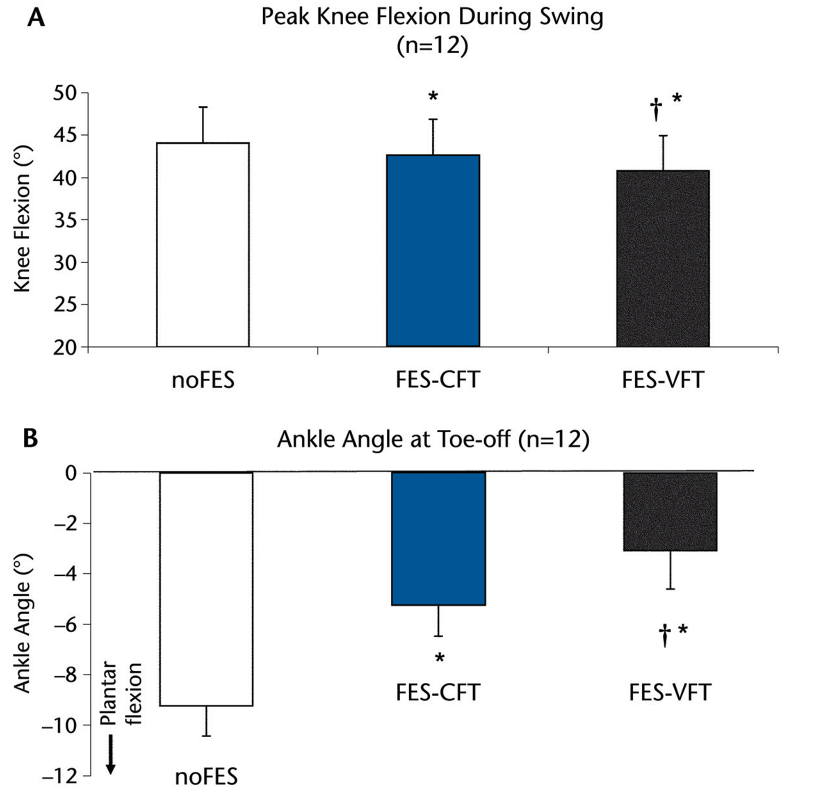

Figure 6.
Graph showing average and standard error bars for (A) peak knee flexion during swing and (B) ankle angle at toe-off for the paretic leg for 12 participants. Three walking conditions shown are: walking without functional electrical stimulation (noFES), walking with dorsiflexor muscle functional electrical stimulation using constant-frequency trains (FES-CFT), and walking with dorsiflexor muscle functional electrical stimulation using variable-frequency trains (FES-VFT). *Significant difference from noFES (P≤.05). †Significant difference from FES-CFT (P≤.05).

Typically, CFTs are delivered during dorsiflexor FES, but there is a dearth of data in the literature quantifying the immediate effects of FES using CFTs on poststroke gait. In our study, as expected, we observed that delivering FES to the paretic ankle dorsiflexor muscles during the swing phase produced greater ankle dorsiflexion angles during the swing phase and at initial contact compared with walking without FES. Increased dorsiflexion during the swing phase and at initial contact with FES using traditional stimulation patterns (CFTs) are not surprising and have been shown previously.[52,53] However, in this study, we also presented an analysis of the effects of FES using CFTs on knee kinematics and circumduction.

An interesting and novel finding of this study was the reduction in swing phase knee flexion during dorsiflexor FES. In their randomized controlled trial assessing the Odstock Drop Foot Stimulator, Burridge and colleagues anecdotally noted that the immediate effect of FES on the tibialis anterior and peroneal muscles was "to bring the ankle into greater dorsiflexion as the foot left the ground and facilitate a flexor withdrawal response in which flexion occurred at both hip and knee joints."[3(pp208–209)] In the present study, we did not have a directional hypothesis about the change in knee flexion with FES. As stated by Burridge and colleagues, it can be argued that swing phase knee flexion might increase during FES because of the flexion withdrawal reflex and the prevalence of flexion synergy in people poststroke. We were surprised to find a decrease in swing phase knee flexion during walking with dorsiflexor FES versus walking without FES. The decrease in knee flexion with dorsiflexor FES is particularly important in individuals poststroke because they show a decreased swing phase knee flexion in the paretic leg compared with the nonparetic leg and compared with control subjects without neurological impairment walking at matched speeds.[5,50] Thus, dorsiflexor FES seems to enhance a typical poststroke gait impairment that can negatively influence foot clearance during swing.

To our knowledge, decreased knee flexion as an immediate effect of dorsiflexor FES has not been reported previously in the literature. We believe that the decreased swing phase knee flexion observed in our study was related to the decreased plantar flexion that we observed at toe-off. The decreased plantar-flexion angles could result in decreased push-off forces at the ankle during the stance-to-swing transition. Forward-dynamic simulations of healthy gait suggest that ankle plantar-flexor force generation during terminal stance helps increase the knee flexion velocity at toe-off, which is a critical contributor to swing phase knee flexion.[54] Interestingly, consistent with our findings, preliminary results from our laboratory of forward-dynamic gait simulations of poststroke gait have predicted that enhanced excitation of ankle dorsiflexor muscles would result in reduced swing knee flexion.[55]

We posit that the timing of dorsiflexor FES during gait could be a factor contributing to decreased plantar flexion at toe-off and decreased knee flexion during the swing phase. Because we triggered the FES using footswitches, dorsiflexor stimulation began before actual toe-off (Fig. 1). Thus, the forces generated by the dorsiflexor muscles at toe-off could have reduced the net plantar-flexor moment at toe-off, potentially resulting in the observed decreased ankle plantar flexion at toe-off, reduced push-off force generation during terminal stance, and reduced knee flexion during swing. The timing of delivery of dorsiflexor FES needs to be systematically manipulated and studied to test this hypothesis further. In addition, the effects of delivering FES to the ankle plantar-flexor muscles to increase force generation during push-off needs to be investigated.

An ideal FES intervention for management of foot drop poststroke would increase dorsiflexion during the swing phase and at initial contact and increase plantar flexion at toe-off. However, we found that although FES helped produce greater dorsiflexion during swing and initial contact, it worsened the already reduced plantar flexion at toe-off prevalent in the paretic leg. Similar to the onset of dorsiflexor FES in our study, previous FES studies have shown that onset of dorsiflexor FES was between heel-off and toe-off and that dorsiflexion increased at toe-off during dorsiflexor FES,[52,53] suggesting that this issue is prevalent in other FES systems and needs to be addressed in future studies.

We did not detect differences in circumduction among the 3 walking conditions tested in this study. The average values of circumduction demonstrated by the individuals tested in our study during walking without FES ($\overline{X}\pm SD=4.0\pm2.9$ cm) were similar to the amount of circumduction shown in individuals with poststroke hemiparesis by Chen and colleagues (4.6 ± 3.2 cm).[50] Because circumduction is a learned compensatory gait strategy, it is understandable that the amount of circumduction did not change as an immediate effect of FES, even though our participants were achieving greater ankle dorsiflexion during the swing phase and, therefore, may not have required circumduction to clear the foot. Perhaps correction of gait compensations such as circumduction could occur if FES is used in conjunction with other interventions during gait retraining under the guidance of a physical therapist to help individuals with poststroke hemiparesis to "unlearn" the compensatory strategies they have developed over time. Nevertheless, a better understanding of the immediate and therapeutic effects (or the lack of effects) of FES on gait compensations, such as circumduction, may help to modify FES-based gait interventions to enable prevention and correction of gait compensations poststroke.

Our findings suggest that novel FES systems capable of delivering VFTs during gait[48,56] can produce enhanced correction of foot drop compared with traditional FES systems that deliver CFTs. The use of VFTs for dorsiflexor FES during poststroke gait, as presented for the first time in our study, is an example of the successful translation of research evidence from animal studies[34,38] to isometric human studies[27,31,33,51] and, finally, in the current study, to a clinical application. The present study also brings forward several interesting issues, such as the reduced swing-phase knee flexion and reduced ankle plantar flexion at toe-off with dorsiflexor FES and the need for more precise timing of dorsiflexor FES, that merit future investigation. One limitation of the present study is that we did not restrict handrail hold during testing. The use of handrails may affect kinematics and kinetics during gait[57,58] and must be investigated adequately in future studies. There also is a need to investigate the effects of delivering FES to multiple muscles to address the many multijoint deficits in poststroke gait. Finally, future studies are needed to investigate whether the use of VFTs during FES can enable the generation of targeted gait performance for a greater number of walking strides.

All authors provided concept/idea/research design. Dr Kesar, Dr Perumal, and Ms Angela Jancosko provided data collection. Dr Perumal provided hardware and software design for the functional electrical stimulation aspect of the data collection. Dr Reisman, Ms Jancosko, and Dr Rudolph assisted with data processing. Dr Kesar provided data analysis and manuscript writing. Dr Binder-Macleod, Dr Perumal, Ms Jancosko, Dr Reisman, and Dr Rudolph provided reviews of the manuscript before submission. The authors thank Ms Margie Roos, PT, NCS, for clinical testing and participant recruitment and Ms Leigh Shrewsbury for scheduling and recruitment. They also thank Dr Andrew Fuglevand for providing helpful reviews of previous drafts of the manuscript.

This study was approved by the Human Subjects Review Board of the University of Delaware.

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