ABSTRACT: Neuromuscular electrical stimulation can improve motor function in those affected by paralysis, but its use is limited by a high rate of muscular fatigue. Variable stimulation patterns have been examined in young adults with and without spinal cord injury, but much less investigation has been devoted to studying the effects of variable stimulation patterns administered to older adults or those paralyzed by stroke. Significant changes occur in the neuromuscular system with age that may affect the response to variable stimulation patterns. We administered three, 3-min intermittent stimulation patterns to the median nerves of 10 individuals with hemiplegia from stroke and 10 age-matched able-bodied adults: (1) constant 20 Hz, (2) a pattern that began at 20 Hz and progressively increased to 40 Hz in the latter half of the task, and (3) a 20-Hz pattern that switched to a 20-Hz doublet pattern after 90 s. In the able-bodied group the doublet pattern produced significantly higher force time integrals (FTI) (1409.72 ± 3.15 N · s) than the 20–40-Hz pattern (1067.46 ± 1.15 N · s) or the 20-Hz pattern (831 ± 1.87 N · s). In the poststroke individuals the doublet pattern also produced the highest FTI (724.04 ± 2.02 N · s), and there was no significant difference between the 20–40-Hz (636.42 ± 1.65 N · s) and 20-Hz (583.64 ± 3.02 N · s) patterns. These results indicate that protocols that incorporate doublets in the later stages of fatigue are effective in older adults and in older adults with paralysis from stroke.

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VARIABLE STIMULATION PATTERNS FOR POSTSTROKE HEMIPLEGIA

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Stroke is a leading cause of serious, long-term disability in the United States, and the resultant upper extremity hemiplegia that occurs is a major factor that contributes to overall disability. Neuromuscular changes such as weakness, increased spasticity, and pathological synergistic muscle activation are typically present following stroke[37]; however, the primary source of impaired motor function is the inability to generate and produce force from paralyzed muscle.

Application of neuromuscular electrical stimulation (NMES) to facilitate force generation in paralyzed muscle is an intervention commonly used in clinical settings. NMES programs that elicit functional movement patterns have been incorporated into neuroprosthetic devices[18,26,47] and implanted systems[27] for the upper extremity. Several studies have shown NMES to be effective in improving motor control,[19] increasing muscle strength,[41] and reducing spasticity[1,41] following stroke, but the specific parameters of NMES used to optimize force production while simultaneously minimizing fatigue are unknown.

Many studies have investigated modulation of stimulation frequency to maximize force output over time in healthy young adults.[5,7,25,32] A few studies have investigated the use of variable-frequency patterns for young adults paralyzed by spinal cord injury (SCI).[4,31,34,49,50,52] Only one study has examined variable-frequency patterns in able-bodied older adults,[3] and none have investigated the use of variable-frequency patterns for individuals paralyzed by stroke.

Abbreviations: ANOVA, analysis of variance; FTI, force-time integral; LFF, low-frequency fatigue; MANOVA, multivariate analysis of variance; MVC, maximal voluntary contraction; NMES, neuromuscular electrical stimulation

Key words: electrical stimulation; doublet; fatigue; stroke; variable frequency patterns

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Muscles paralyzed by stroke may have properties that are different than those paralyzed by SCI. Stroke commonly occurs in older adults, whereas SCI typically occurs in younger individuals. As people age, significant type II muscle fiber cell loss occurs along with a reduction in maximal voluntary force capacity.[28] Similar muscle atrophy occurs with paralysis from stroke.[15] The predominance of slower type I fibers results in a leftward shift in the force-frequency curve.[46] In general, low-frequency stimulation (20 Hz) induces greater fatigue in older compared to younger adults,[16,17,40] whereas fatigue rates are similar during evoked contractions with higher frequencies (40 Hz) of stimulation.[16,36,44] It is not known whether muscles paralyzed by stroke in older adults would show the same responses to variable patterns of stimulation as those of young adults paralyzed by SCI.

Recent studies have shown that increases in frequency during intermittent stimulation produced greater force over time than constant frequency protocols in young able-bodied adults[20] and in young adults paralyzed by SCI.[34,50] Protocols that incorporate doublets (5–10 ms interpulse intervals) from the onset of the task have been shown to increase,[4,6,11,23,33,48,51] decrease,[9,10] or not change[35,52] force production over time compared to constant frequency stimulation in young adults. They have also been shown to exacerbate fatigue in older adults.[3] Protocols that used continuous doublets were more effective than those that used just one doublet at the start of each train in young able-bodied adults.[11,33] Intermittent stimulation protocols that began with constant frequency trains and then switched to trains that incorporated doublets later in the fatigue task were more effective than unchanging constant frequency stimulation in young able-bodied individuals[20,48] and in young adults paralyzed by SCI.[49]

Although many studies have used able-bodied individuals as a model to test variable frequency patterns in young adults, very few studies have compared the effects of variable frequency patterns in individuals with and without paralysis in the same study. Thomas et al.[52] found that neither healthy nor SCI-paralyzed muscle force output was enhanced during a fatigue task that included doublets from the onset of the task compared to constant frequency stimulation. Bickel et al.[4] found that doublets enhanced force output during fatigue tasks to a greater degree in young able-bodied individuals than in those paralyzed by SCI.

Thus, it appears that stimulation trains that increase in frequency over time or include doublets later in the fatigue task may be the best patterns for evoked contractions in young, able-bodied individuals and in young individuals paralyzed by SCI. The goal of this study was to test whether these patterns would also be effective in older able-bodied adults and in older adults paralyzed by stroke.

MATERIALS AND METHODS

Participants. Ten individuals who sustained a stroke with resultant upper extremity hemiplegia (63.80 ± 12.69 years; eight men, two women) and 10 age-matched healthy adults (63.60 ± 12.88 years; six men, four women) participated in this study. One individual who had sustained a stroke had unusually high initial muscular forces that fatigued very rapidly and was excluded from analysis. The average age of the group used for analysis who had sustained a stroke (seven men, two women) was 64.44 ± 12.88 years.

Study participants were recruited from the Austin, Texas vicinity through local newspaper advertisements over a 12-month period. Individuals with hemiplegia were selected if the following criteria were met: (1) status poststroke onset of at least 1 year prior to start of study involvement; (2) full discharge from any inpatient, outpatient, or home health therapies; (3) ability to maintain forearm supination, full digit extension, and active thumb adduction of at least 30° for positioning and testing in the experimental apparatus; and (4) ability to comprehend the objectives of the study and follow study-related directions. The exclusion criteria for the poststroke participants were: (1) presence of extensive spasticity or flaccidity in the affected upper extremity; (2) contraindication for application of electrical stimulation (pacemaker, implanted surgical hardware in the hand or forearms, skin lesions at application site, etc.); (3) confusion or disorientation; or (4) presence of pain syndromes of the upper extremity. Although all participants had recovered some degree of active movement in the hand, fine motor coordination, skilled dexterity, and precise manipulation remained difficult for most. Fugl–Meyer (upper extremity subsection) scores for the poststroke participants averaged 55.00 ± 11.74 out of a maximum of 66. All participants underwent an orientation session during which they signed consent forms prior to testing and performed the Fugl–Meyer upper extremity test. A medical clearance form was signed by physicians who were familiar with the medical history of the poststroke participants to ensure that experimental activities would be safely tolerated. All procedures were approved by the Institutional Review Board, University of Texas at Austin.

Experimental Setup. The experimental apparatus and setup have been described previously.[25,29,31] Participants were seated in a high-back chair with their right forearm supinated and resting on a tabletop. The shoulder and upper arm were positioned parallel with the trunk, and the elbow was positioned at 90°. Hip, knee, and ankle joints were all maintained in a neutral 90° position. A custom-designed apparatus immobilized the forearm and maintained the arm in a position of supination. Straps secured the forearm at the wrist and forearm midpoint; the upper arm was secured with a strap positioned slightly above the elbow that attached to the upper back portion of the chair. The hand was stabilized with therapeutic putty placed underneath and on the volar surface. A thermoplastic plate was positioned on the putty over the digits, and a strap secured the interphalangeal joints and the metacarpal-phalangeal joints in extension. The thumb was extended and abducted and positioned against the force transducer. The custom-designed force transducer device consisted of a mobile, rotating, height-adjustable horizontal arm made of two narrow aluminum surfaces with a strain gauge that measured forces in the horizontal (x), and vertical (y) directions. The contact area spanned from the thumb tip to midway between the IP and MCP joints. The resultant force, $R = \sqrt{x^2 + y^2}$ was displayed on the computer monitor and recorded using commercially available software (Spike 2, v. 5.14, Cambridge Electronics Design, Cambridge, UK).

A stimulating electrode was placed over the median nerve at the wrist and secured with a Velcro band and tape after optimal placement was obtained. Electrical pulses of 50 μs duration were delivered from a constant current stimulator (Digitimer, Model DS7A, Welwyn Springs, UK) using custom-written scripts constructed through Spike 2 software. The EMG signal was recorded through two adhesive pre-gelled, 5 mm diameter, Ag/AgCl bipolar surface electrodes (Danlee, Syracuse, New York). The active electrode was placed over the thenar eminence slightly below the MCP joint of the thumb and the reference electrode ≈1 cm medial to the active electrode toward the center of the palm. A ground electrode was placed over the pisiform bone.

Experimental Protocol. Three different fatigue tests were administered in random order for a total of three sessions for each of the two groups. Each session was separated by at least 48 h to allow for recovery from possible effects of low-frequency fatigue.[43] All fatigue tasks lasted 3 min and were composed of intermittent 300 ms trains with 700 ms rest based on the Burke et al.[12] experimental protocol used frequently in NMES research. All protocols were identical except for the fatigue tests. A diagram of the experimental protocol is shown in Figure 1.

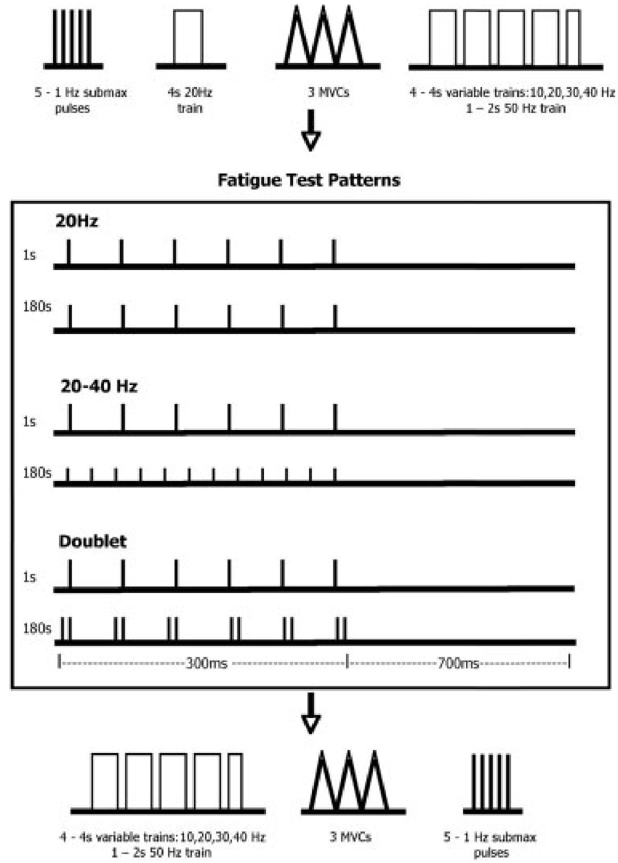


FIGURE 1. Experimental protocol.

Fatigue Test 1 consisted of constant frequency, 20-Hz trains. Fatigue Test 2 consisted of stimulation trains that began at 20 Hz. Midway through the protocol, stimulation frequency was gradually increased from 20 Hz to 40 Hz by 0.22 Hz per s such that 40 Hz was the terminal frequency reached at 180 s. Fatigue Test 3 began similarly at 20 Hz, and midway through the protocol the stimulation was changed to a 20-Hz pattern of doublets (interpulse interval: 5 ms) that continued to the end of the fatigue task.

Prior to each fatigue test, participants performed three maximal voluntary contractions (MVCs) of thumb adduction. Each contraction was held for 3 s, with a rest of 3 s. Single 1-Hz pulses were then delivered at various stimulator placements over the median nerve, and the location where the largest

M-wave amplitude and twitch force occurred at a low stimulation intensity was determined. The stimulator intensity was then progressively increased to obtain maximal M-wave amplitude and twitch force. The intensity was set such that an M-wave of 70% of maximum was used for the remainder of the protocol. This intensity evoked a force of ≈70% MVC during a 1-s, 20-Hz train.

Once the intensity was set, five maximal and five submaximal 1-Hz pulses were administered. The maximal M-waves were collected only at the beginning of the experiment to measure the submaximal intensity (70% of maximum) to be used for the remainder of the protocol. This was followed by administration of a 4-s constant 20-Hz train that was used as a baseline comparison to match the experimental setup across testing days. Three, 3-s MVCs were again performed before and after the fatigue task along with five random, constant-frequency, 4-s trains of 10, 20, 30, and 40 Hz and a 2-s train of 50 Hz. These data were collected to determine the presence of low-frequency fatigue following the fatigue tests. Low-frequency fatigue (LFF) is a phenomenon where greater force loss occurs in fatigued muscle during low frequencies than high frequencies of stimulation.[21] Five submaximal M-waves were delivered at the end of the experiment to verify stability of the stimulating electrode during testing.

Data Collection and Analysis. The force output signal was amplified by 100 and lowpass-filtered at 1 kHz (Bridge 8 Amplifier System, Model 74030, World Precision Instruments, Sarasota, Florida). EMG was amplified by 100 and highpass-filtered above 8 Hz, (Coulbourn Instruments Isolated Bioamplifier with Bandpass Filter, Model V75-04, Allentown, Pennsylvania). The force and EMG were sampled at 1,000 and 2,000 Hz, respectively (Micro 1401 mkII 500 kHz 16-bit Analog-Digital Converter, Cambridge Electronics Design). All data were recorded and analyzed using the Spike 2 software.

Statistical Analysis. Two-way repeated-measures analysis of variance (ANOVA) with Bonferroni posthoc analysis were used with pattern (20 Hz, 20–40 Hz, and doublet) and condition (able-bodied and stroke) as the independent variables to compare the 4-s, 20-Hz constant stimulation trains, force–time integrals (FTIs) during the fatigue tasks, and the forces at 10-s intervals during the fatigue tasks.

MVCs, 20/50-Hz force ratios, and M-wave amplitudes measured before and after fatigue were compared with a 3-way repeated measures ANOVA with Bonferroni correction using pattern (20 Hz, 20–40 Hz, or doublet), condition (able-bodied versus stroke), and time (pre- or postfatiguing stimulation pattern) as independent variables.

Maximum evoked forces from each of the five 2–4-s trains (10, 20, 30, 40, and 50 Hz) were compared across the three patterns (20 Hz, 20–40 Hz, and doublet), the two stimulation conditions (able-bodied and stroke), and the two time levels (pre- and postfatigue) using a multivariate analysis of variance (MANOVA) with univariate ANOVAs and Bonferroni corrections for post-hoc analysis. Starting, middle, and ending forces were also compared using the same model MANOVA. An alpha level of 0.05 was used for all statistical comparisons. All data are presented as mean ± standard error.

RESULTS

Day-to-Day Repeatability. There were no significant differences between the forces during the 4-s, 20-Hz trains administered before each of the protocols in the two conditions. Average forces during the 4-s trains were 11.43 ± 1.62 N, 11.21 ± 1.15 N, and 11.88 ± 1.63 N for the 20-Hz, 20–40-Hz, and doublet protocols, respectively, in the able-bodied group ($P = 0.485$), and 8.91 ± 1.13 N, 8.91 ± 1.09 N, and 8.99 ± 0.76 N, respectively, for the poststroke group ($P = 0.994$). The ICC for the 4-s, 20-Hz trains in the able-bodied group was 0.93; for the poststroke group it was 0.99.

MVC forces before and after all tests are shown in Table 1. MVC forces of the able-bodied were significantly higher than for the poststroke participants ($P = 0.002$). For both conditions, changes from prefatigue force values to postfatigue force values were significant (able-bodied, $P = 0.004$; stroke, $P = 0.003$). No significant differences were present between patterns for the two conditions (able-bodied, $P = 0.575$; stroke, $P = 0.265$).

Fatigue. Starting forces during the fatigue tests were measured at 10 s after the initiation of the stimulation to allow time for the force to plateau. Average forces at each of the 10-s intervals over the three fatiguing stimulation patterns for both groups are shown in Figure 2. The starting, midpoint, and final forces for all fatigue tests are shown in Table 2. A significant overall difference was present in the factors of pattern and time. In the able-bodied, 89-s midpoint ($P = 0.021$) and final ($P = 0.018$) forces were significantly lower than the starting forces for the 20-Hz protocol, and only 89-s midpoint forces ($P = 0.007$) were significantly lower than starting forces for the 20–40-Hz protocol. There was no

Table 1. MVC forces before and after all tests.				
	Maximal Voluntary Contraction Forces (*N*)			
	Able-Bodied		Poststroke	
	Pre	Post	Pre	Post
20 Hz	65.22 ± 8.22	60.14 ± 7.13*	48.98 ± 7.36	46.88 ± 7.73*
20-40 Hz	69.39 ± 6.85	62.06 ± 7.65*	50.99 ± 7.26	45.99 ± 7.08*
Doublet	70.86 ± 6.88	61.80 ± 5.79*	47.13 ± 6.64	39.27 ± 6.99*
	Submaximal M-wave Amplitudes (mV)			
20 Hz	6.88 ± 0.37	6.58 ± 0.32	5.32 ± 0.32	5.26 ± 0.34
20-40 Hz	6.94 ± 0.38	6.76 ± 0.37	6.42 ± 0.36	6.38 ± 0.38
Doublet	7.39 ± 0.53	7.20 ± 0.44	5.52 ± 0.49	5.41 ± 0.50

**Significant difference in pre- to postfatigue values,* $P < 0.05$.

significant difference in pairwise comparisons for starting, 89-s midpoint, and ending forces in the doublet pattern. The force at the start of the doublets (90 s) was higher than for the single pulses at the start of the task. For the individuals with hemiplegia, all forces were significantly lower at the end than at the start of the fatigue task and there was no significant change at the start of the doublets.

Fatigue Test Force–Time Integrals. Mean FTIs for all fatigue tasks for both groups are shown in Figure 3.

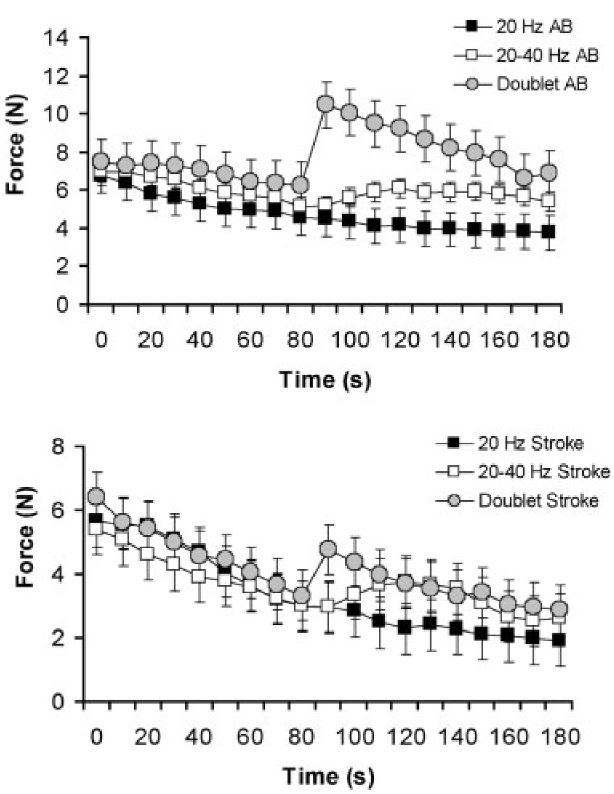


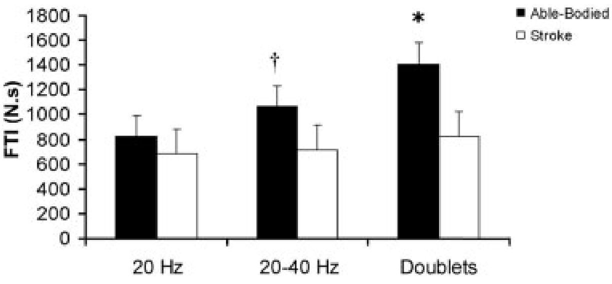
FIGURE 2. Average forces recorded at 10-s intervals during the three fatiguing stimulation patterns for the able-bodied (top) and poststroke (bottom) groups.

The doublet stimulation produced significantly higher FTIs than both the 20-Hz and the 20–40-Hz patterns ($P < 0.001$ for both) for the able-bodied group. The 20–40-Hz pattern also produced a significantly higher FTI compared to the 20-Hz pattern ($P < 0.001$). For poststroke subjects, the doublet pattern also produced significantly greater FTIs than the 20-Hz pattern ($P = 0.007$); however, no significant difference was present between the doublet and the 20–40-Hz patterns or between the 20–40-Hz pattern and the 20-Hz pattern.

Force–Frequency Distributions. Figure 4 displays the force–frequency curves pre- and postfatigue for both groups. Significant overall effects occurred for frequency ($P < 0.001$) and time ($P = 0.03$). Forces at all frequencies decreased significantly for both groups. Reduction in force expressed as percent initial force for all stimulation frequencies are shown in Figure 5. LFF is commonly measured by changes in 20/50-Hz force

Table 2. Starting, midpoint, and final forces for all fatigue tests.			
	Fatigue Test Forces (*N*)		
	20 Hz	20-40 Hz	Doublet
Able-bodied			
Start (10s)	6.80 ± 0.87	7.01 ± 0.52	7.50 ± 1.27
89s	4.58 ± 0.67*	5.18 ± 0.55*	6.29 ± 0.50
90s	4.51 ± 0.22*	5.22 ± 0.23	10.51 ± 0.45*†‡
End (180s)	3.79 ± 0.22*	5.43 ± 0.27	6.62 ± 1.22
Stroke			
Start (10s)	5.58 ± 0.38	5.09 ± 0.30	5.63 ± 0.37
89s	3.02 ± 0.14	3.00 ± 0.26*	3.34 ± 0.25*
90s	2.97 ± 0.13	2.99 ± 0.14*	4.78 ± 0.28
End (180s)	1.93 ± 0.07*	2.62 ± 0.26*	2.89 ± 0.13*

**Significant at* $P \leq 0.05$ *when compared to starting (10 s) force.*
†Significant at $P \leq 0.05$ *when compared to 89 s force.*
‡Significant at $P \leq 0.05$ *when compared to ending (180 s) force.*



∗ Significant at the P<0.05 level when compared to 20-40 Hz and 20 Hz patterns

† Significant at the P<0.05 level when compared to 20 Hz pattern

FIGURE 3. Force–time integrals (FTIs) for the three fatiguing stimulation patterns in both conditions. Asterisks indicate significant differences from the 20-Hz and 20–40-Hz protocols. The cross indicates significant differences from the 20-Hz protocol.

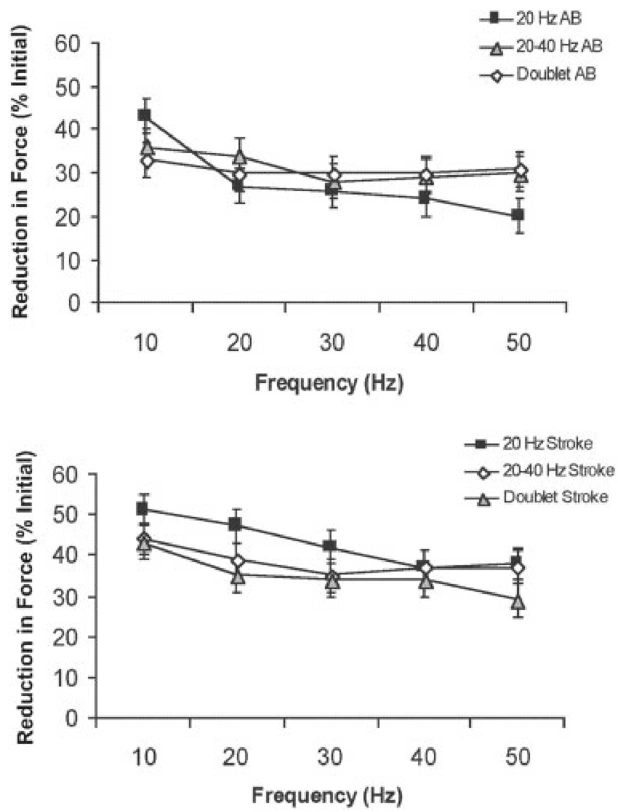


FIGURE 5. Decreases in force at all frequencies are plotted for both groups. The reduction in force is plotted as percent initial ((final – initial) / initial).

ratios.[21] There were no significant differences in the 20/50-Hz force ratios before versus after fatigue or across conditions. Pre- and postfatigue average 20/50-Hz force ratios in the able-bodied group were: 20 Hz pre, 71.69 ± 5.46%; post, 65.98 ± 4.16%; 20–40 Hz pre, 82.12 ± 2.66%; post, 76.96 ± 3.90%; doublet pre, 73.49 ± 4.40%, post, 73.36 ± 2.38%. In the poststroke condition, 20/50 Hz ratios were: 20 Hz pre, 87.01 ± 10.14%, post, 82.76 ± 19.48%; 20–40 Hz pre, 76.12 ± 10.83%, post, 69.06 ± 19.94%; Doublet pre, 86.27 ± 12.14%, post, 78.63 ± 6.87%.

M-Wave Amplitude. Submaximal M-wave amplitudes before and after fatigue are shown in Table 1. In the able-bodied group, maximal M-waves measured prior to testing were 9.92 ± 0.54, 9.08 ± 0.42, and 10.00 ± 0.56 mV in the 20-Hz, 20–40-Hz, and doublet patterns, respectively. On average, submaximal M-waves in the able-bodied group were 73.19 ± 3.54% of the maximal M-waves. In the participants with paralysis from stroke, maximal M-waves for the three patterns were 7.41 ± 0.39, 8.31 ± 0.44, and 8.98 ± 0.49 mV for the 20-Hz, 20–40-Hz, and doublet patterns, respectively. For these individuals, submaximal M-waves averaged 70.17 ± 4.12% of maximal. There were no significant differences in pre/post values, validating consistency in electrode placement throughout the experiments, and no significant differences were present between patterns or conditions.

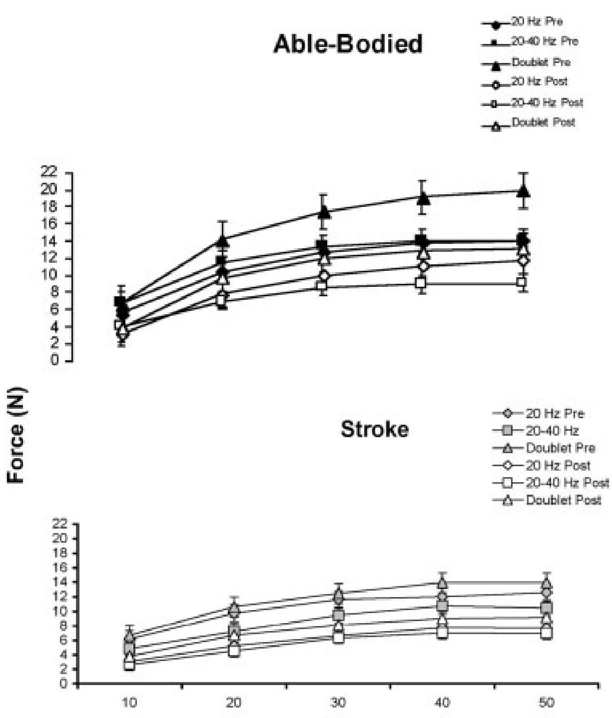


FIGURE 4. Force versus frequency curves before and after each fatiguing stimulation pattern at both intensities.

DISCUSSION

This study was the first to test variable frequency patterns in individuals paralyzed by stroke and the first to test increasing frequencies of stimulation in older adults. It was also the first to compare a pattern of increasing frequency against a pattern that changed to doublets in any population. It is essential to maintain force output over time during NMES programs whether they are used for functional or rehabilitative purposes. The main findings of the present study were that variable frequency patterns were effective in maintaining force output in older able-bodied adults and in older adults paralyzed by stroke. In both groups, a constant-frequency protocol that changed to doublets in the second half of the task was the most effective at maintaining force output. The task with progressively increasing frequency also produced higher FTIs than the constant-frequency task in the able-bodied group, but it was not as effective in the individuals with paralysis from stroke.

Many studies tested variable-frequency trains before and after a standard voluntary or evoked fatiguing contraction in young able-bodied adults. They reported that trains that incorporated doublets produced relatively higher force following fatigue than constant-frequency trains.[19,31,38,39] Others tested the use of doublets during the fatigue task itself in able-bodied younger adults; these studies produced mixed results. Some showed an increase in force over time,[4,6,11,33,51] while others showed a decrease[9,10] or no change.[35,52] Thomas et al.[52] found that fatigue protocols with doublet trains produced the same force output as constant frequency trains in younger adults paralyzed by SCI. Scott et al.[49] found that protocols that incorporated doublets later in the fatigue task improved force output compared to constant frequency protocols and continuous doublet frequency stimulation in younger adults paralyzed by SCI. Similarly, we found that doublets incorporated later in the fatigue task were effective for both able-bodied and poststroke older adults.

Early studies proposed that, during sustained evoked maximal contractions, stimulation frequencies should decrease (from 60–80 Hz to 20 Hz) over the course of the fatigue task to prevent neuromuscular transmission failure[5,30] and excess energy expenditure.[8,42] More recently, however, studies that used frequencies within the range of motor unit firing during voluntary contractions (15–40 Hz) found that a reduction in frequency exacerbates force loss during fatigue.[23,32] Moreover, protocols that increased in stimulation frequency produced greater FTIs than constant-frequency patterns in young able-bodied adults.[20,32] In the present study, increasing stimulation frequency increased force output compared to constant frequency stimulation in older able-bodied adults but not in older adults paralyzed by stroke. It appears that the muscles of nonparalyzed individuals are more responsive to increases in frequency than those affected by paralysis. This may be due to differences in calcium kinetics or to differences in the elasticity of the noncontractile elements of the muscle. Morphological changes such as loss of overall muscle mass, increased adiposity, increased density of connective tissue, and reduced tensile strength of tendons occur during aging,[13] and they can change further with paralysis from stroke.[15] Sarcoplasmic[22] and intracellular[45] calcium concentrations also decrease with age. These processes likely also further degenerate with paralysis due to stroke and result in decreased responsiveness to changes in stimulation frequency.

Neither group displayed significant LFF, which is a phenomenon where repeated stimulation causes calcium to remain in the interstitial spaces of muscle cells for extended time periods. This provokes an overall decrease of calcium release from the sarcoplasmic reticulum, and higher frequencies are needed to maintain force output.[14] LFF is commonly observed in young, able-bodied adults.[21,24] In the present study, an increase in frequency enhanced force generation in the older able-bodied adults but not in the older poststroke adults. It is possible that the higher frequencies of stimulation enhanced interstitial calcium levels to a greater degree in the able-bodied older adults. However, there was not significant LFF in either group. Similarly, Allman and Rice[2] found that LFF did not occur in older adults.

The use of NMES for rehabilitation of individuals with neuromuscular paralysis has a strong potential to be effective, but specific patterns and parameters that maximize force output and delay the onset of fatigue must be defined. The results of our investigation show that the use of electrical stimulation programs that incorporate doublets are more effective in maintaining force output in older adults and in individuals with paralysis from stroke than the constant frequency patterns that are typically used in clinical applications. These data also show that nonparalyzed muscle is more responsive to increases in stimulation frequency and doublets during fatigue protocols than paralyzed muscle in older adults.

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