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Variable stimulation patterns in younger and older thenar muscle

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ABSTRACT

Neuromuscular electrical stimulation (NMES) is typically used with older adults receiving rehabilitation therapies, but little is known about the stimulation patterns that maximize force output and minimize fatigue in this population. The purpose of this study was to apply variable patterns of stimulation to the thenar muscles of the hand in younger and older adults to determine if force production and neuromuscular fatigue effects were similar. Three submaximal stimulation patterns were administered: A 20 Hz constant frequency pattern, a pattern that increased from 20 to 40 Hz, and a pattern that incorporated two closely spaced (5 ms) doublet pulses. The doublet stimulation produced significantly higher average forces and force–time integrals (FTIs) than the constant frequency and increasing frequency patterns in both age groups. Additionally, older adults showed less fatigue than the younger group during isometric contractions performed after the fatiguing stimulation patterns. These results suggest that variable pulse NMES patterns enhance force production in the hand in both younger and older individuals better than constant frequency patterns, which are typically used in clinical applications. Also, greater fatigue resistance to electrical stimulation protocols may exist in the older population; this is critical information for the design and application of NMES rehabilitation regimens used with older adults.

1. Introduction

Neuromuscular electrical stimulation (NMES) is frequently used in clinical settings to facilitate motor recovery in muscles following neurological or orthopedic injury. However, NMES imparts rapid fatigue due to simultaneous firing of motor units and has the propensity to alter normal motor unit recruitment order (Vanderthommen and Duchateau, 2007). Stimulation patterns that reduce fatigue and optimize force output have been identified in several studies, but most tested young, able-bodied adults (Chou et al., 2008; Dreibati et al., 2010; Kaczmarek et al., 2010; Kesar and Binder-Macleod, 2006; Maladen et al., 2007). Only a limited number of studies have tested whether the motor response to stimulation found in young healthy adults are comparable to those seen in older adults (Allman and Rice, 2001,2004; Allman et al., 2004; Klein et al., 1988; Narici et al., 1991; Stevens et al., 2001). Because disabling conditions such as stroke, Parkinson's Disease, and osteoarthritis primarily impact persons in middle to older ages (Hughes et al., 2010; Lloyd-Jones et al., 2009; Parkinson's Disease Foundation, 2010), it is critical that NMES rehabilitation regimens be customized to the unique motor behavior of older muscle. Data from younger counterparts can potentially be used as a model for these efforts.

Younger and older individuals may respond differently to electrical stimulation due to the inherent physiology of the muscle tissue. Aged muscle shows reduced muscle mass, strength and power (Macaluso and De Vito, 2004; Visvanathan and Chapman, 2010). Additionally, motor unit remodeling occurs whereby fast twitch fibers are lost, creating a more slow-twitch, fatigue-resistant composition further compromising the ability to generate forceful muscle action (Raj et al., 2009). Age-related metabolic processes such as declines in muscle protein synthesis can further impact muscle function (Boirie, 2009). However, an increased resistance to neuromuscular fatigue has frequently been observed in older muscle, but this phenomenon may be dependent on the muscle used and the type and duration of contraction performed (Lanza et al., 2004; Rawson, 2010; Rubinstein and Kamen, 2005; Russ and Binder-Macleod, 1999).

Individuals may also respond differently to stimulation depending on the parameters of stimulation used. Pulse pattern, pulse duration, stimulation frequency, duty cycle and intensity will all impact the motor response obtained. We have previously studied the effects of maximal and submaximal stimulation patterns when used in a young healthy population (Doucet and Griffin, 2008). We found that variable patterns that transitioned from low to high frequency and patterns that incorporated doublets (two short pulses separated by a 5 ms interval) produced greater force–time

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integrals (FTIs) than 20 Hz constant stimulation patterns. In our subsequent study, we investigated the same stimulation patterns in older healthy and post-stroke individuals. This investigation yielded similar outcomes when the variable stimulation patterns were again more effective in enhancing force output than the constant 20 Hz pattern (Doucet and Griffin, 2009). However, we did not compare the responses of younger and older adults in the same study; this is the objective of the present investigation.

Our purpose was to examine the effects of variable stimulation patterns in healthy aged muscle and healthy younger muscle, comparing data from previous separate studies (Doucet and Griffin, 2008, 2009) so as to define differences and similarities that might exist. Obtaining a greater understanding of how neuromuscular electrical stimulation patterns affect motor performance in younger and older individuals is critical for designing appropriate rehabilitation interventions for these populations. These findings can be used to inform clinical practice.

2. Methods

2.1. Participants

Ten younger (age 23.81 ± 2.71 years; 5 females, 5 males, age range 20–30 years) and 10 older (age 63.60 ± 12.88 years; 4 females, 6 males, age range 45–80 years) healthy individuals participated in the study. Participants were recruited from the University of Texas student population and from the Austin, TX area through newspaper advertisements. In a questionnaire administered upon enrollment, no physical limitation was reported by any participant and all had medical histories free of neuromuscular disorder. None of those participating were engaged in specific strengthening exercises for the hand at that time and no trained athletes participated in the study. Each participant signed a consent form prior to testing and was fully oriented to study procedures. All procedures were approved by our institutional review board.

2.2. Experimental setup

The experimental setup has been described previously (Doucet and Griffin, 2008, 2009). A custom-designed forearm apparatus made of thermoplastic material immobilized the forearm and maintained it in supination. The thumb was extended, abducted, and placed against the force transducer. The force transducer was mounted to a height-adjustable horizontal arm made of two narrow aluminum surfaces that measured the evoked forces of thumb adduction in the vertical (y) and horizontal (x) directions. The resultant force, R (R = square root of $x^2 + y^2$) was recorded and displayed on the monitor using commercially-available software (Spike 2, Version 5.14; Cambridge Electronics Design [CED], Cambridge, UK).

A stimulating electrode was secured over the median nerve at the wrist. All electrical impulses were 50 μs in duration and delivered from a constant-current stimulator (Digitimer, Ltd., Model DS7A; Welwyn Garden City, UK) using custom-written scripts constructed through the Spike 2 software. The electromyographic (EMG) signal was recorded through two 5 mm Ag/AgCl bipolar surface electrodes (Danlee, Syracuse, NY) placed over the thenar eminence slightly medial to the MCP joint, targeting the adductor pollicis muscle. A similar ground electrode was placed over the pisiform bone at the wrist.

2.3. Experimental protocol

All 20 individuals participated in three experimental sessions on three different days with each session being separated by at least 48 h. Participants were tested with one of three fatiguing stimulation patterns applied to the median nerve of the right hand: (1) a 3-min pattern of 20 Hz stimulation; (2) a 3-min pattern of 90 s of 20 Hz followed by 90 s of a gradual increase from 20 to 40 Hz; and (3) a 3-min pattern of 90 s of 20 Hz followed by 90 s of 20 Hz doublet stimulation. Doublet stimulation consisted of two pulses separated by 5 ms and delivered at a 20 Hz frequency.

Prior to administration of the experimental protocol, the stimulating electrode was placed at different locations over the median nerve at the wrist to determine the site at which an optimal motor response occurred. At each location, a 1 Hz pulse was delivered at a low intensity to elicit a compound muscle action potential (M-wave). The location where the largest M-wave amplitude (measured peak to peak) and twitch force occurred was used and maximal M-wave was then measured.

A series of 1 s, 20 Hz trains separated by 20 s of rest were administered until the force obtained was 70% of the force obtained in the 1 s maximal stimulation train. This intensity setting was further verified by administering a longer duration 4 s, 20 Hz train and monitoring the output to insure a 70% force value. We chose a submaximal stimulation level of 70% because it is similar to what is used in clinical settings and produces favorable motor responses while still remaining tolerable for participants. The submaximal and maximal stimulation intensities were recorded and used as the baseline at which to begin intensity of stimulation at subsequent sessions.

The experimental protocol began with 5, 1 Hz pulses at the predetermined maximal stimulation intensity, followed by 5, 1 Hz pulses at the predetermined submaximal stimulation intensity. These were administered before and after the fatiguing pattern to monitor stability of the stimulating electrode and maintenance of the optimal placement site throughout the experiment. The 4 s 20 Hz train was again administered to validate the 70% stimulation level. This was followed by performance of 3 MVCs of 3 s duration, each separated by 3 s. MVCs were collected to determine if voluntary force was impacted by electrical stimulation.

This was followed by five varying frequency stimulation trains, four 4 s trains at 10, 20, 30, and 40 Hz, and one 2 s train at 50 Hz. These were used to assess the presence of low frequency fatigue (LFF). LFF is a phenomenon first described by (Edwards et al., 1977) where significant force loss can be observed in fatigued muscle when stimulated with low frequency trains (<20 Hz).

One of the three fatiguing patterns was then administered. Based on neuromuscular fatigue work done by Burke et al. (1973), we used intermittent stimulation with a duty cycle of 300 ms on and 700 ms off. This ratio of stimulation allows for full tetanization but provides rest intervals to avoid muscle fiber failure. All fatiguing patterns and LFF test trains were administered in random order. The fatiguing pattern was followed by the same 5 varying-frequency trains, 3 MVCs, and 5 submaximal pulses that were delivered prior to the fatiguing pattern. Identical procedures were followed for all sessions except for the fatiguing pattern delivered. The entire protocol is graphically displayed in Fig. 1.

2.4. Data collection and analysis

The force output signal was amplified by 100 (Bridge 8 Amplifier System, Model 74030, World Precision Instruments, Sarasota, FL), sampled at 1000 Hz and low-pass filtered at 1 kHz. EMG was amplified by 100, high-pass filtered above 8 Hz (Coulbourn Instruments, Allentown, PA), sampled at 2000 Hz, and digitally converted (Micro 1401, Cambridge Electronics Design, Cambridge, UK). All data were recorded and analyzed using Spike 2 software (v. 5.14, CED, Cambridge, Cambridge, UK).

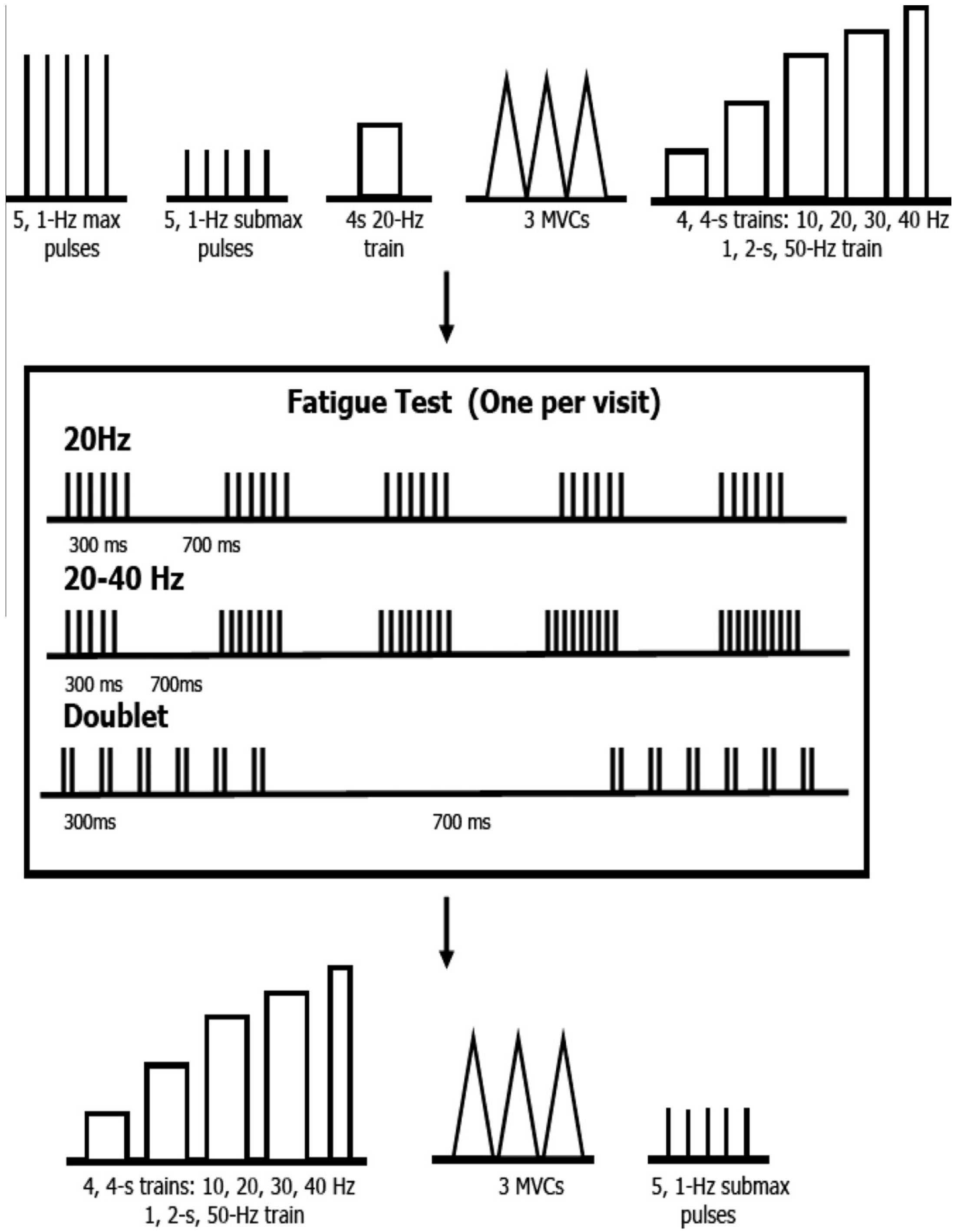

Figure 1. Graphic representation of experimental protocol administered to the thenar muscles of 10 young and 10 older participants. Each fatigue test began with a constant 20 Hz intermittent stimulation pattern; at the 90 s mark, the stimulation switched to each of the patterns shown and continued through 180 s.

2.5. Statistical analysis

Two-way repeated measures ANOVAs with Holm–Sidak post hoc analysis were used to compare MVC forces, the 4 s 20 Hz stimulation trains, FTIs, and the forces at 10 s intervals during the fatiguing patterns; the independent variables used were the three fatiguing patterns (20 Hz, 20–40 Hz and doublet) and condition (younger and older). A mixed effects repeated measures ANOVA with a structured covariance to express differences across time was used to analyze starting, midpoint, and ending forces. M-wave amplitudes measured before and after fatigue were compared with a 3-way repeated measures ANOVA with post hoc analysis using fatiguing pattern (20 Hz, 20–40 Hz, or doublet), condition (younger versus older), and time (pre- or post-fatiguing stimulation pattern) as independent variables. Multivariate ANOVAs (MANOVA) were used to compare rise times and maximum evoked forces for each of the five brief 4 s trains (10, 20, 30, 40, and 50 Hz). Data were compared across factors of pattern (20 Hz, 20–40 Hz and doublet), age group (younger and older) and time level (pre- and post-fatigue); Bonferroni corrections were used for post hoc analysis.

An alpha level of 0.05 was used for all statistical comparisons. All data are presented as mean ± standard deviation in the text and mean ± standard error in tables and figures.

3. Results

There were no significant differences between forces during the 4 s, 20 Hz trains administered before each of the stimulating patterns within or between groups, indicating consistency of stimulation intensity across days and protocols. Peak forces during the 4 s trains for the younger adults were 20 Hz, 13.28 ± 1.42 N; 20–40 Hz, 14.10 ± 1.52 N; doublet, 14.25 ± 1.39 N. Peak forces during the 4 s trains for the older participants 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.

3.1. Maximal Voluntary Contractions (MVCs)

MVC forces before and after all tests are shown in Table 1. There were no significant differences in MVC starting forces across days for the younger and older conditions. There were no significant differences for the factor of age; however, the younger group showed significant differences from pre-fatigue MVCs to post-fatigue MVCs for all three fatigue patterns administered. The older group showed a pre-post difference for the increasing frequency pattern and the doublet pattern.

3.2. Fatigue Test Forces

Starting forces during the fatigue tests were measured at 10 s after the initiation of stimulation to allow time for the force to plateau from twitch potentiation. There were no significant differences between the starting forces across the three fatigue tests within the young group; the older group showed a difference in starting force for the 20 Hz pattern when compared to the doublet pattern ($p = 0.03$).

The initiation of the doublet pattern at 90 s into the fatigue task immediately increased force output upon application for the younger group (55.44% increase) as well as the older group (60.57% increase). Average starting, midpoint (89 and 90 s), and ending forces for both groups and can be seen in Table 2. Significant differences within the three fatigue patterns at these time points were found as noted. There were also differences between patterns in each age group. For both age groups, the 90 s force value in the doublet pattern was significantly larger when compared to the 90 s force value in the 20 Hz pattern and the 20–40 Hz pattern ($p < 0.0001$ for all). Additionally, for the older group, the ending force value at 180 s was significantly larger in the 20–40 Hz pattern ($p = 0.04$) and the doublet pattern ($p = 0.0002$) when compared to the ending force value in the 20 Hz pattern; however, there were no significant differences between the young and the older groups.

For the younger group, average forces measured across all 10 s intervals during the doublet pattern (7.66 ± 1.25 N) were significantly higher than forces in the 20 Hz (5.99 ± 0.86 N, $p < 0.001$) and the 20–40 Hz pattern (avg., 6.66 ± 0.93 N, $p < 0.001$). The 20–40 Hz pattern also produced significantly higher average forces than the 20 Hz stimulation pattern ($p = 0.018$). The older adults showed similar results. The doublet stimulation pattern produced significantly higher average forces (7.81 ± 1.29 N) than the 20 Hz (avg., 4.72 ± 0.87 N, $p < 0.001$) and 20–40 Hz (avg., 5.98 ± 0.52 N, $p < 0.001$) patterns, with the 20–40 Hz pattern being more effective in maximizing forces than the 20 Hz pattern ($p < 0.001$).

There was a statistically significant interaction between age and pattern ($p = 0.004$) with the 20 Hz pattern and the 20–40 Hz pattern showing significant differences in force output between the two age groups ($p < 0.001$ and $p = 0.038$, respectively); forces in both patterns were higher in the younger adults. However, forces in the doublet pattern were highest independent of the group (Fig. 2).

Table 1
Maximal voluntary contraction (MVC) forces (N) in thenar muscle performed before and after fatigue tests ($N = 20$, 9 females, 11 males; values expressed as mean ± SE).

	Younger		Older	
	Pre	Post	Pre	Post
20 Hz	67.72 ± 8.71	62.93 ± 7.96*	60.47 ± 7.77	57.13 ± 8.11
20–40 Hz	67.73 ± 9.93	60.50 ± 8.21*	69.18 ± 7.33	61.48 ± 7.45*
Doublet	73.11 ± 10.79	66.80 ± 8.74*	64.58 ± 8.08	59.21 ± 8.72*

* Significant at $p < 0.05$.

Table 2
Differences within fatigue patterns for starting, midpoint, and ending evoked forces in thenar muscle ($N = 20$, 9 females, 11 males; values expressed as mean ± SE). For both young and older, the 90 s force value in the doublet pattern was significantly larger when compared to the 90 s force value in the 20 Hz pattern and the 20–40 Hz pattern ($p < 0.0001$ for all). For the older group, the ending force value at 180s was significantly larger in the 20–40 Hz pattern ($p = 0.04$) and the doublet pattern ($p = 0.0002$) when compared to the ending force value in the 20 Hz pattern; however, there were no significant differences between the young and the older groups.

	Fatigue Test Forces (N)		
	20 Hz	20–40 Hz	Doublet
Younger			
Start (10 s)	7.40 ± 0.74	8.25 ± 1.01	8.55 ± 0.81
89 s	5.84 ± 0.54*	5.76 ± 0.77*	6.53 ± 0.72*,†
90 s	5.70 ± 0.55*	5.58 ± 0.78*	10.19 ± 1.10‡
End (180 s)	5.29 ± 0.65*	5.39 ± 0.96	5.64 ± 0.92*
Older			
Start (10s)	6.18 ± 0.74	6.97 ± 1.01	7.66 ± 0.81
89 s	4.58 ± 0.54*,‡	5.18 ± 0.77*	6.30 ± 0.72*,†
90 s	4.51 ± 0.55*,‡	5.22 ± 0.78*	10.51 ± 1.10*,†
End (180 s)	3.79 ± 0.65*	5.43 ± 0.96	6.92 ± 0.92

* Significant when compared to starting force ($p \leqslant 0.05$).
† Significant when compared to 90 s force ($p \leqslant 0.05$).
‡ Significant when compared to 180 s force ($p \leqslant 0.05$).

3.3. Force time integrals

Fig. 3 displays the overall FTIs, or force produced per second over the 3-min fatiguing patterns. The doublet stimulation produced significantly higher FTIs than both the 20-Hz and the 20–40 Hz patterns in both groups (1.37 ± 0.12 N·s younger; 1.41 ± 0.12 N·s older; $p < 0.001$ for both age groups). The 20–40 Hz pattern also produced a significantly higher FTI compared to the 20-Hz pattern in both age groups ($p = 0.022$ younger; $p < 0.001$ older). There were no significant differences in FTIs between the younger and older groups.

3.4. Force–frequency curves and low frequency fatigue

Force data from the five 4 s trains (10, 20, 30, 40, and 50 Hz) delivered to all participants following the 3-min stimulation patterns are shown in Fig. 4. LFF would manifest itself as depressed forces during the 10 and/or 20 Hz trains we administered following the three fatiguing stimulation patterns. LFF is often measured by the percentage difference in force between the 20 Hz stimulation train and the 50 Hz stimulation train (Edwards et al., 1977); however, LFF was not apparent following any of the three programs in either group when 20/50 force percentages were compared: younger, 20 Hz pre, 71.81 ± 3.74%, post, 66.41 ± 5.84%; 20–40 Hz pre, 75.90 ± 5.96%, post, 66.99 ± 4.74%; doublet pre, 72.35 ± 5.13%, post, 76.43 ± 4.55%; older, 20 Hz pre, 71.69 ± 5.46%, post, 65.98 ± 4.16%;

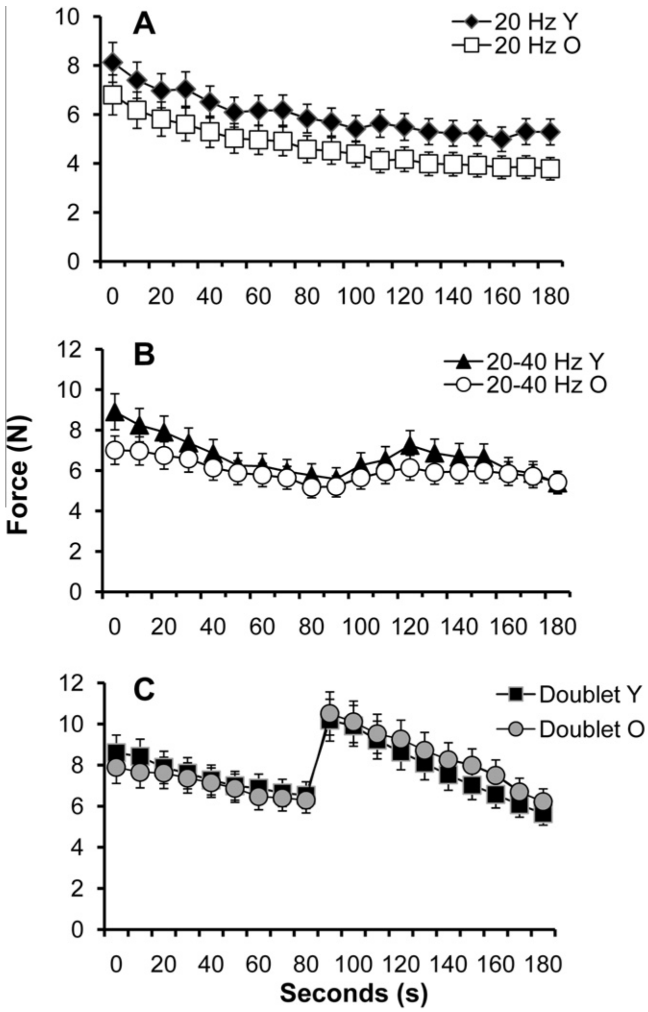

Figure 2. Average forces for 10 younger (Y) vs. 10 older (O) participants during (A) 20 Hz constant frequency, (B) 20–40 Hz increasing frequency and (C) doublet fatiguing stimulation patterns administered to the thenar muscle.

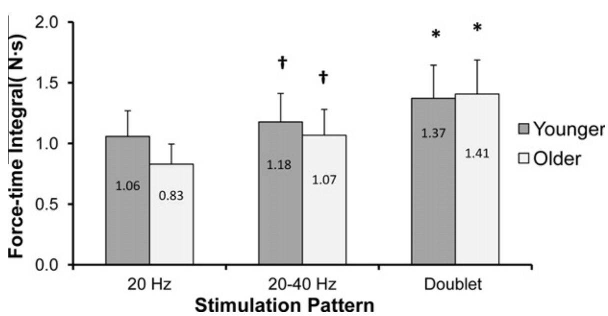

Figure 3. Amount of total thenar muscle force produced per second over the 3-min fatigue program is represented as force–time integral (FTI). * = Significantly different from 20–40 Hz and 20 Hz patterns; † = significantly different from 20 Hz pattern.

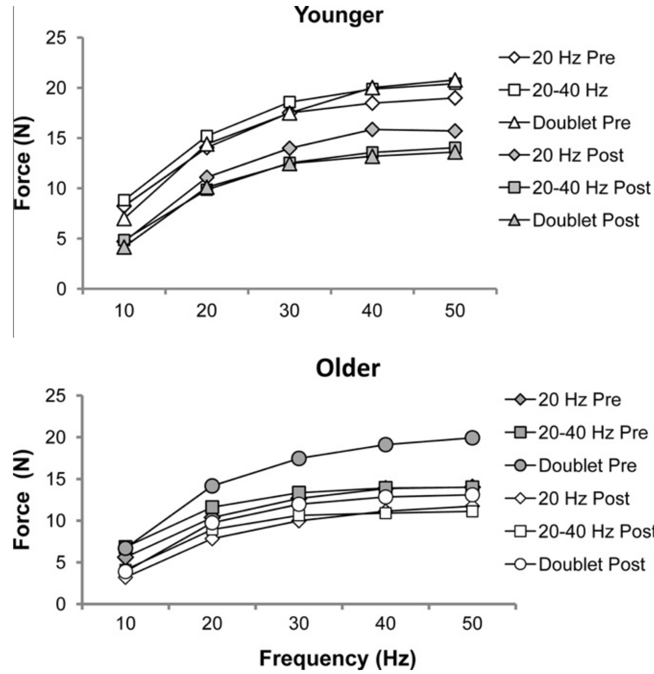

Figure 4. Average forces for 4 s varying frequency trains (10, 20, 30, 40, and 50 Hz) pre and post fatiguing pattern in younger (upper graph) and older participants (lower graph) to assess low frequency fatigue (LFF).

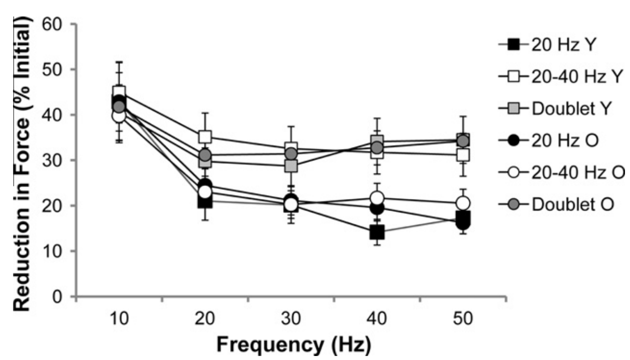

Figure 5. Percentage drop in force pre-fatigue to post-fatigue when five different frequency trains were administered.

20–40 Hz pre, 82.12 ± 2.66%, post, 76.96 ± 3.90%; doublet pre, 73.49 ± 4.40%, post, 73.36 ± 2.38%.

Although the percentage of force loss was greatest during the 10 Hz train for both groups, this did not reach statistical significance (Fig. 5). There were no significant differences in relative force loss between the older and the younger groups as a percentage of initial force for any of the test trains. Overall differences were present for frequency and time, but no interactions were found.

The rise times of the five forces were calculated by measuring the time from which the signal left baseline to the peak amplitude during the selected train. When the rise times were compared before and after each of the three fatiguing patterns, a significant difference was present overall for the factors of frequency and age. Pairwise comparisons indicated that the rise times of the variable forces administered after the 20 Hz constant pattern showed the greatest slowing in the older group, where rise times were 77.85% longer than the young group (avg., 0.29 ± 0.08 s older, vs. 0.06 ± 0.02 s younger, $p = 0.016$). This was especially true at the 10 and 20 Hz trains where older rise times were 40.43% ($p = 0.011$) and 89.66% ($p < 0.001$) longer respectively. Rise times were also significantly longer for the older participants at the 10 Hz level after the 20–40 Hz pattern was administered (avg., 0.70 ± 0.03 s older, vs. 0.34 ± 0.03 s younger; $p = 0.004$). In contrast, rise times for the younger participants were longer for the 10 Hz train (avg., 0.73 ± 0.03 s younger, vs. 0.48 ± 0.05 s older; $p = 0.036$) and the 20 Hz train (avg., 0.36 ± 0.04 s younger, vs. 0.10 ± 0.06 s older; $p = 0.045$) after the doublet pattern was administered.

3.5. M-wave amplitude

There were no significant differences in pre/post M-wave amplitude values, validating consistency in stimulating electrode placement and neuromuscular propagation throughout the experiments. In the younger group, 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, M-waves averaged 70.17 ± 4.12% of maximal. In the older 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, M-waves in the older group were 73.19 ± 3.54% of the maximal M-waves. M-wave values in both groups confirmed data collection at the 70% MVC level.

4. Discussion

Despite possible differences in the overall physiology and the fatigue processes of younger and older muscle, force output as a result of variable stimulation patterns administered to the thenar muscle of both age groups resulted in similar outcomes. A pattern of doublets enhanced force output optimally during a 3-min fatiguing protocol in the thenar muscles of both younger and older adults. The variable stimulation pattern that increased in frequency from 20 to 40 Hz likewise increased force output more effectively when compared to a constant-frequency stimulation pattern. These findings are extremely valuable for the design and prescription of stimulation regimens with the rehabilitation population, as stimulation patterns used in clinical applications are typically constant frequency patterns and rapid fatigue is a major concern (Bracciano, 2008).

The younger participants showed significant force loss of MVCs after all of the fatiguing stimulation patterns, whereas the older group showed notable force loss in two patterns. Fatigue resistance in elders has been attributed to the reduced number of viable motor units present in aged muscle, a selective loss of fast-twitch muscle fibers, and an overall slowing of muscle contractile properties (Allman and Rice, 2002); however, investigations on this topic have produced conflicting results, especially when related to hand musculature.

In particular, the hand thenar group is composed of predominantly Type I fibers, giving it a more slow-twitch fatigue-resistant baseline composition (Johnson et al., 1973). As such, these hand muscles are possibly less affected by the age-related remodeling that has been reported to alter larger fast-twitch muscles to a more slow-twitch functionality (Macaluso and De Vito, 2004). The hand muscles have also been described to be less prone to functional decline and atrophy when compared to larger limb and postural muscles; this may be due in part to the relatively small size and less propensity for fat and restrictive connective tissue to accumulate in this area of the body (Narici et al., 1991). Narici et al. (1991) also observed that strength declines and increases in half relaxation time did not become significantly different when compared to younger counterparts until approximately age 60. This may account for the similarity in starting forces and MVC responses observed in our two groups as our older participants could be considered a "younger old" group, with the average age being about 63 years.

Other investigators studying hand muscles have published similar outcomes as found in our study. For example, older adults showed an increased resistance to fatigue when MVCs and evoked twitch forces were measured in the adductor pollicis muscle; males in both groups demonstrated stronger twitch forces, but elders showed less percentage of force loss from baseline when compared to the younger participants (Ditor and Hicks, 2000). In comparison, some studies indicate that elders can show increased fatigability depending on the task; A recent review of current literature on fatigue effects in younger and older persons indicated that older individuals generally show greater fatigue for dynamic contractions and rapid movements requiring power, but tend to show a resistance to fatigue for most other types of contractions when compared to younger individuals (Christie et al., 2011).

When starting forces were compared to ending forces in the doublet fatigue test, younger subjects showed a significant decrease whereas the older participants did not. Other study results have been inconclusive as to whether younger subjects respond differently than their older counterparts when evoked force protocols are used, as this design has been less frequently studied. Less fatigue was observed in the thumb muscle of older compared to younger adults during a half-minute 30 Hz stimulation protocol (Narici et al., 1991). In contrast, no difference was seen in the fatigue response of older and younger individuals when the quadriceps was stimulated with a 40 Hz, 2.5-min protocol (Stevens et al., 2001) or when the triceps surae was stimulated with a 20 Hz protocol for 10 min (Klein et al., 1988).

Older and younger participants demonstrated similar outcomes in force output per second (FTI) over the course that the protocol was tested. The FTIs produced by the two variable patterns were greater when compared to the constant frequency pattern. The doublet pattern FTI was significantly higher than both the increasing frequency pattern and the constant pattern. At first glance, one might argue that the higher force output in the variable patterns could be attributed to higher overall number of pulses than are present in the constant frequency pattern; however, from the 90 s time point when the pulse patterns changed, the increasing frequency pattern administered 1575 total pulses whereas the doublet pattern administered 1080, and the doublet pattern produced the higher FTI. Early research with animal models indicated that brief high frequency pulses with short interpulse intervals (5–10 ms doublets) used in conjunction with constant lower frequency trains of stimulation could produce a rapid shortening of muscle resulting in an enhancement of force; this was labeled the "catchlike" property of muscle (Burke et al., 1976). This occurrence was subsequently examined in later work and found to augment force (Kebaetse et al., 2001) improve muscle performance (Maladen et al., 2007) and potentially offset the effects of fatigue (Karu et al., 1995; Russ and Binder-Macleod, 1999).

Intuitively, using stimulation patterns that are composed of doublets exclusively would suggest optimal force output, but evidence has shown otherwise. Bentley and Lehman (2005) investigated the effects of doublet stimulation in forearm muscles and determined that doublets have the greatest effect on fatigued muscle within the first few pulses, but doublet force enhancement may be diminished thereafter. This effect was also observed when repetitive doublet-only stimulation was administered to the quadriceps muscle and greater fatigue was present using this protocol when compared to the constant- or variable-frequency protocols (Binder-Macleod and Scott, 2001; Scott and Binder-Macleod, 2003). This study further confirms that variable patterns or closely spaced pulses can be effective in temporarily augmenting force output but this is most effective when used after the muscle has become fatigued or later in a task.

Studies investigating the mechanisms contributing to enhanced muscle performance with variable stimulation suggest that these types of stimulation patterns increase the rate of rise in force; more rapid shortening of muscle fibers has been shown to improve the overall mechanics and efficiency of force production (Lee et al., 1999). Rapid shortening also tends to take up the "slack" in the muscle series elastic component (Parmiggiani and Stein, 1981); this rapid force development can also enhance calcium release from the sarcoplasmic reticulum resulting in more available calcium for use by the muscle cross bridge mechanisms (Slade et al., 2003).

We assessed the presence of LFF by administering the short 4 s trains before and after each fatiguing pattern and measuring comparative force loss following all protocols. Although greater force loss was seen at the 10 Hz train in both groups, this did not reach statistical significance in the younger and older adults. Rise times at the 10 and 20 Hz trains were notably longer in the older subjects after the 20 Hz fatiguing pattern and at 10 Hz after the 20–40 Hz pattern; in contrast, less slowing of rise times were noted after the doublet pattern in the these individuals. The results may again suggest a greater resistance to fatigue in older subjects when higher frequencies or more variable patterns are used.

Our study was limited by the small number of participants and our sample of convenience; therefore, additional studies with randomized controls and a greater number of participants will potentially yield more powerful effects and further substantiate or refute these findings.

5. Conclusions

This study was the first to examine the differences and similarities between younger and older thenar muscle force output when constant and variable electrical stimulation patterns were applied. The results suggest that younger and older thenar muscle respond similarly to variable electrical stimulation patterns and that stimulation effects seen in younger adults can be used as a model for older adults; however, further investigation with larger participant numbers exploring different types of musculature should be undertaken to test this hypothesis.

Even so, the information obtained from this study can be useful in designing appropriate NMES rehabilitation regimens that optimize motor recovery in both younger and older individuals. Older adults may show enhanced fatigue resistance to stimulation regimens applied to hand muscle. Using variable stimulation patterns could enhance force output in muscle and potentially optimize treatment outcomes for elders participating in rehabilitation therapies. This knowledge will also assist in the design and development of therapeutic interventions and in the advancement of rehabilitation technology.

Conflict of interest

The authors declare that they have no conflict of interest.

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