Comparing the Induced Muscle Fatigue Between Asynchronous and Synchronous Electrical Stimulation in Able-Bodied and Spinal Cord Injured Populations

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***Abstract*—Neuromuscular electrical stimulation (NMES) has been shown to impart a number of health benefits and can be used to produce functional outcomes. However, one limitation of NMES is the onset of NMES-induced fatigue. Multi-channel asynchronous stimulation has been shown to reduce NMES-induced fatigue compared to conventional single-channel stimulation. However, in previous studies in man, the effect of stimulation frequency on the NMES-induced fatigue has not been examined for asynchronous stimulation. Low stimulation frequencies are known to reduce fatigue during conventional stimulation. Therefore, the aim of this study was to examine the fatigue characteristics of high- and low-frequency asynchronous stimulation as well as high- and low-frequency conventional stimulation. Experiments were performed in both able-bodied and spinal cord injured populations. Low frequency asynchronous stimulation is found to have significant fatigue benefits over high frequency asynchronous stimulation as well as high- and low-frequency conventional stimulation, motivating its use for rehabilitation and functional electrical stimulation (FES).**

***Index Terms*—Asynchronous stimulation, fatigue, functional electrical stimulation (FES), neuromuscular electrical stimulation (NMES), spinal cord injury.**

I. Introduction

NEUROMUSCULAR electrical stimulation (NMES) has been shown to impart a number of health benefits such as increased bone mineral density [1], improved muscular strength [2], [3], improved motor control [4], increased lean muscle mass and sensory ability [5], increased range of motion [6], and improved cardiovascular parameters [7], [8]. However, one limitation of NMES is the rapid onset of NMES-induced fatigue. Fatigue limits the duration that NMES can be applied. Therefore, the aforementioned health benefits of NMES may be enhanced by utilizing stimulation methods which reduce NMES-induced fatigue. NMES can be used to produce functional outcomes such as grasping [9], [10], walking [11], reaching [12], stair climbing [13], and cycling [14], [15] where it is termed functional electrical stimulation (FES). However, NMES-induced fatigue limits the duration that functional tasks can be performed, motivating researchers to examine alternative stimulation methods that may reduce fatigue such as doublets [16], [17], N-let pulse trains [18], and modulation of the stimulation parameters [19]–[21].

One suggested cause of NMES-induced fatigue is that, in contrast to physiological contractions, conventional single-channel stimulation exhibits a nonselective, spatially fixed, synchronous recruitment of motor units [22], [23]. Researchers have developed two methods to address this suggested cause of fatigue, namely, sequential stimulation and asynchronous stimulation. During sequential stimulation, multiple stimulation channels are utilized to either segregate the desired muscle into multiple groups of motor units or to segregate multiple synergistic muscles. Pulse trains are then delivered sequentially to each stimulation channel thereby allowing motor units to rest when the corresponding stimulation channel is not active. Similar to sequential stimulation, asynchronous stimulation utilizes multiple stimulation channels to segregate motor units or synergistic muscles. However, during asynchronous stimulation, the stimulus pulses are delivered in an interleaved manner so that lower stimulation frequencies are achieved at each stimulation channel while retaining a high composite stimulation frequency. An illustrative comparison of sequential stimulation, asynchronous stimulation, and conventional stimulation is provided in Fig. 1.

Both sequential stimulation and asynchronous stimulation have been shown to reduce NMES-induced fatigue [24]–[38]. One study [31] found that a shorter on-time (i.e., the time to keep one stimulation channel activated before switching to the next channel) resulted in less fatigue, motivating the use of asynchronous stimulation over sequential stimulation. However, only a limited number of studies has examined asynchronous stimulation in human subjects [26]–[29], [33], [35].

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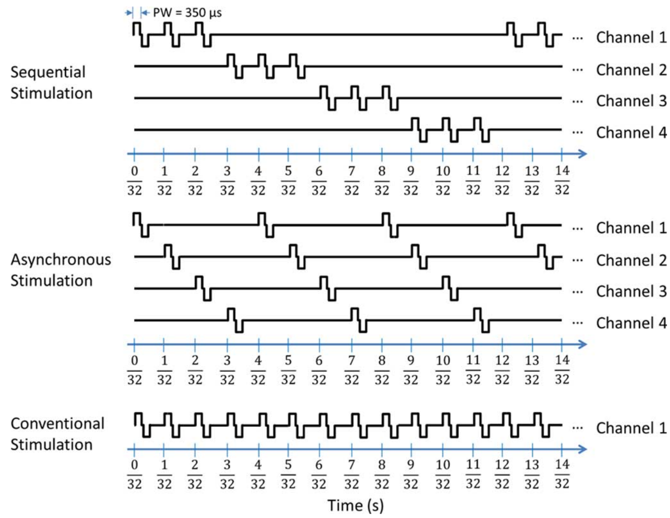

Fig. 1. During sequential stimulation, multiple stimulation channels are utilized where high frequency pulse trains are delivered sequentially to each channel, resulting in a lower average stimulation frequency per channel. In the present example, each stimulation channel briefly receives pulses at 32 Hz; however, the average stimulation frequency is 8 Hz per channel (i.e., a 25% duty cycle). The stimulation channel changes after every three pulses in the present example for illustrative purposes; however, the number of pulses allowed before switching channels varies in literature. During asynchronous stimulation, multiple stimulation channels are utilized where high composite stimulation frequencies are achieved by interleaving the pulses. Depicted is asynchronous 8 Hz stimulation with four channels where each channel receives pulse trains at 8 Hz, but the composite stimulation frequency is 32 Hz. During conventional stimulation, only a single stimulation channel is utilized and higher stimulation frequencies are required to achieve a strong and smooth force output, resulting in the onset of fatigue. Depicted in the present example is 32 Hz conventional stimulation. Note that the width of the pulses is not drawn to scale for illustrative purposes.

It is well known that higher stimulation frequencies increase the rate of fatigue during conventional single-channel stimulation [39]–[41]; therefore, it is reasonable to assume that the same frequency-fatigue relationship exists with asynchronous stimulation. However, in each of the aforementioned studies in man, one asynchronous stimulation protocol is compared to one conventional stimulation protocol. In other words, the effect of stimulation frequency on fatigue was not examined. Thus, during asynchronous stimulation in man, the extent that NMES-induced fatigue can be further reduced by reducing the stimulation frequency is not presently clear. Two studies in cats indicate that reduced stimulation frequencies may lead to reduced fatigue during asynchronous stimulation [24], [37]. However, it was previously shown that low-frequency asynchronous stimulation may result in a significant force ripple (i.e., contractions that are not fully fused, thus exhibiting non-smooth force tracings) [42]. Thus, it is not presently clear if the expected fatigue benefits of low-frequency asynchronous stimulation outweigh the potential problem of force ripple. Asynchronous stimulation has been examined in able-boded individuals [29], [35], individuals post-stroke [33], and individuals with spinal cord injury (SCI) [26]–[28]. However, asynchronous stimulation has not been examined in both able-bodied and SCI populations simultaneously. The aim of this study is to characterize the ability of asynchronous stimulation to reduce NMES-induced fatigue at high- and low-stimulation frequencies compared to conventional stimulation in both able-bodied and spinal cord injured populations.

TABLE I
DEMOGRAPHICS OF SCI STUDY GROUP

Subject	Age	Sex	Injury	Months Since Injury
A	55	M	C6	28.7
B	36	M	C7	65.6
C	63	F	T10	77.8
D	35	M	C4	179.3

II. METHODS

A. Subjects

Asynchronous and conventional stimulation were examined in both able-bodied and spinal cord injured populations to better understand the NMES-induced fatigue characteristics of the stimulation protocols. Four individuals with SCI (three male, one female, aged 35 to 63) participated in the study at the Medical University of South Carolina. Prior to participation, written informed consent was obtained from all participants, as approved by the institutional review board at the Medical University of South Carolina. All participants were medically stable, but a physical therapist was present during the study to monitor vital signs as needed and to monitor for signs of autonomic dysreflexia. Demographics are listed in Table I for the four individuals with SCI. Four able-bodied individuals (three male, one female, aged 20 to 27) also participated in the study at the University of Florida. Prior to participation, written informed consent was obtained from all participants, as approved by the institutional review board at the University of Florida.

B. Apparatus

All testing was performed using an apparatus that consisted of the following: 1) a current-controlled eight-channel stimulator (RehaStim, Hasomed GmbH, Germany); 2) a data acquisition device (Quanser Q8-USB); 3) a personal computer running Matlab/Simulink; and 4) a dynamometer to measure the isometric knee-joint torque. At the University of Florida, the dynamometer is a modified leg extension machine (LEM) fitted with force transducers while a Biodex System 4 Pro dynamometer was utilized at the Medical University of South Carolina and is depicted in Fig. 2. The LEM and Biodex allow for seating adjustments to ensure that the center of rotation of the knee joint could be aligned with the center of rotation of the dynamometers. In both apparatuses, the thigh was parallel to the ground and the shank was in a gravity-eliminated position. In general, the hips were flexed approximately 75 degrees, though a reclined (i.e., more extended) position was utilized for SCI individuals that did not have adequate trunk control (i.e., those with cervical level injuries) and straps were used to stabilize the torso.

C. Stimulation Protocols

Four stimulation protocols were examined: 8 Hz asynchronous stimulation (A8), 16 Hz asynchronous stimulation (A16), 32 Hz conventional stimulation (C32), and 64 Hz conventional stimulation (C64). Conventional stimulation consists of a single stimulation channel with a pair of 3″ by

Fig. 2. Individuals with SCI were seated at the Biodex dynamometer to measure isometric knee joint torque as the muscle fatigues during four stimulation protocols. Protocol order was randomized and electrode positions were marked so that the positioning could be replicated when switching between asynchronous and conventional electrode configurations. Pictured above is single-channel conventional stimulation on the individual's left leg and four-channel asynchronous stimulation on the individual's right leg. Seating adjustments were made to ensure that the center of rotation of the knee joint could be aligned with the center of rotation of the dynamometer.

5″ Valutrode® surface electrodes placed over the quadriceps femoris muscle group, while asynchronous stimulation consists of four channels of stimulation utilizing four electrodes placed distally (1.5″ by 3.5″ Valutrode®) and two electrodes placed proximally (2″ by 3.5″ Valutrode®).[1] For asynchronous stimulation, each channel utilized the same current amplitude, but the stimulation pulses were interleaved across the stimulation channels. In other words, asynchronous stimulation of 16 Hz with four channels results in a composite stimulation frequency of 64 Hz. The electrode configuration utilized during asynchronous stimulation is depicted in Fig. 3, and the method of interleaving the pulses across the stimulation channels is the same as previously depicted for asynchronous stimulation in Fig. 1.

D. Determining Desired Initial Torque

In the subsequently described fatigue trials, the current amplitude is adjusted before the start of each fatigue trial to match the initial torque to a predetermined level. To account for variability in each individual's strength, the desired torque level was determined specifically for each leg in a pretrial (i.e., before any fatigue trials were conducted) test with conventional 64 Hz stimulation. The pretrial test also served as a warm-up session and allowed for individuals to become accustomed to the sensation of the electrical pulses. During the pretrial test, pulse trains were delivered 5 seconds at a time with 25 seconds of rest between pulse trains. The resulting torque was analyzed immediately following each contraction and the current amplitude was adjusted during the rest period preceding the following contraction. Since all four stimulation protocols were examined on the same day, it was expected that there would be some layover effect of fatigue, even with rest between fatigue trials. Therefore, the subsequently described criteria were used to determine the desired initial torque so that the layover effect of fatigue would not later preclude torque matching.

[1]Surface electrodes for the study were provided compliments of Axelgaard Manufacturing Company.

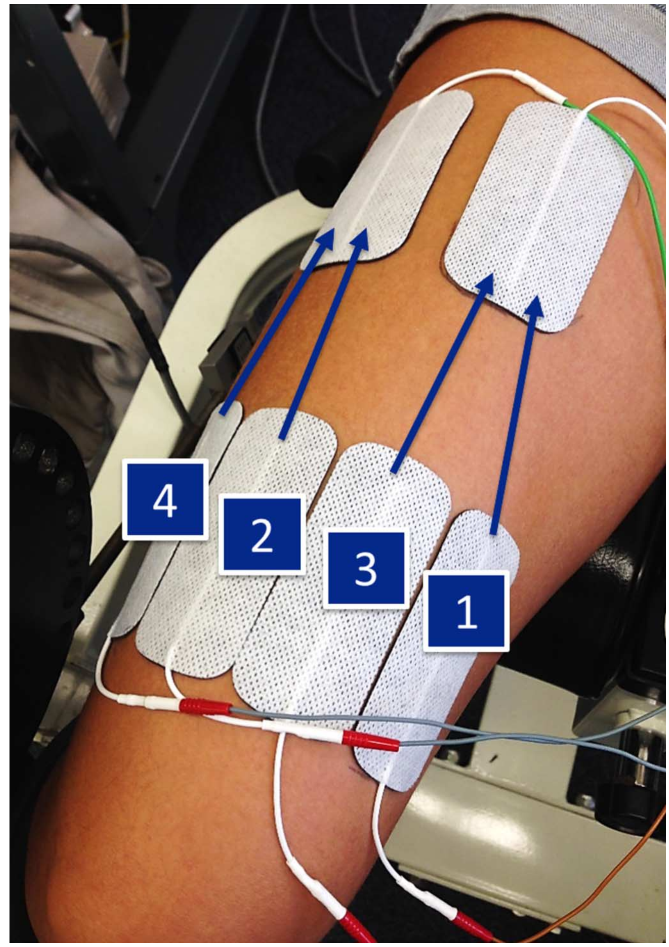

Fig. 3. Electrode configuration utilized for four-channel asynchronous stimulation with two electrodes placed proximally and four electrodes placed distally. Stimulation channels 1 and 3 share the most medial and proximal electrode, while stimulation channels 2 and 4 share the most lateral and proximal electrode.

1) SCI: For individuals with SCI, the current amplitude was incremented between contractions until one of the following three conditions were met: 1) the isometric torque reached 20 N·m; 2) the torque output began to plateau with increases in current amplitude; or 3) the current amplitude reached 100 mA, whichever occurred first. If the torque reached 20 N·m during this phase of the experiment, then the desired torque was set to 20 N·m for the subsequent fatigue trials. Otherwise, the desired torque was set to 75% of the maximum torque achieved during the pretrial test in an effort to account for any layover effect of fatigue between trials. The torque reached 20 N·m for only one leg of one individual in the pretrial test. After determining the desired torque based on the aforementioned criteria, this value was used as the targeted initial torque for all subsequent fatigue trials on the same leg.

2) Able-Bodied: For able-bodied individuals, the current amplitude was incremented between contractions until one of the following three conditions were met: 1) the isometric torque reached 10% of their maximal voluntary contraction (MVC); 2) the torque output began to plateau with increases in current amplitude; or 3) the individual experienced discomfort in response to the stimulation, whichever occurred first. If the torque reached 10% of the MVC during this phase of the experiment, then the desired torque was set to 10% of the MVC for the subsequent fatigue trials. Otherwise, the desired torque was set to 75% of the maximum torque achieved during the pretrial test in an effort to account for any layover effect of fatigue between trials. The torque reached 10% of the MVC for all able-bodied individuals in the pretrial test. After determining the desired torque based on the aforementioned criteria, this value was used as the targeted initial torque for all subsequent fatigue trials on the same leg.

E. Fatigue Trials

After the desired torque level was determined for each leg, fatigue trials were conducted for each of the four stimulation protocols. Fatigue trials consisted of 5 minutes of intermittent stimulation where pulse trains were delivered for 5 seconds and then the muscle was allowed to rest for 5 seconds. To increase subject comfort during delivery of each 5-second pulse train, the current amplitude was increased as a ramp from 0 mA to the desired current amplitude over the course of 1 second. The current amplitude then remained constant for 3 seconds before returning to 0 mA over the course of 1 second.

1) Precautionary Measures: Since all four stimulation protocols were examined on the same day, the protocol order was randomized and participants were allowed to rest between trials. A minimum of 20 minutes of rest was given between trials although participants were allowed to rest longer if they so desired. Electrode positions were marked so that the placement could be replicated when switching between asynchronous and conventional electrode configurations.

2) Setting Current Amplitude: Before the start of each fatigue trial, the current amplitude was adjusted in order to match the initial value of torque to the desired initial torque that was determined previously (see Section II-D). During this phase of the experiment, pulse trains were delivered 5 seconds at a time with 25 seconds of rest between pulse trains. The resulting torque was analyzed immediately following each contraction and the current amplitude was adjusted during the rest period preceding the following contraction. Fatigue trials were initiated immediately after determining the appropriate current amplitude.

An alternative approach to the study would have been to utilize the same current amplitude for all stimulation protocols in an effort to recruit the same number of fibers for each protocol. However, there is no guarantee that matching the current amplitude across asynchronous and conventional stimulation would recruit the same number of fibers. Given the nonselective recruitment patterns for NMES [22], [23], the relative drop in force should be consistent across stimulation intensities (i.e., current amplitudes). Therefore, similar to [24], [26], [28], [35], the current amplitude was adjusted for each stimulation protocol in order to reach a desired value of torque.

F. Data Analysis

The mean isometric torque was calculated for each contraction (30 contractions per fatigue trial). To account for intersubject variability in strength and intrasubject variability in the initial contraction, the torque was then normalized by the mean torque of the first contraction. The four stimulation protocols were compared according to the following metrics: fatigue time and fatigue index. Fatigue time denotes the time elapsed between the first contraction and the contraction at which the torque decreased below 80% of the initial contraction. Fatigue index is the ratio of the mean torque produced in the final three contractions to the torque produced in the first contraction. To account for intersubject variability in terms of fatigability of the muscle, fatigue times were normalized by the mean fatigue time of all stimulation protocols for each leg. The same normalization process was applied to the fatigue index. Analysis of variance (ANOVA) was performed on the normalized fatigue time and fatigue index of the group data at a significance level of $\alpha = 0.05$. *Post hoc* analysis (Tukey-Kramer method) was used to determine differences between individual stimulation protocols at a significance level of $\alpha = 0.05$.

III. Results

A. Torque Matching

Matching the initial torque to the desired level proved to be difficult in the SCI population as the torque mismatch was greater than 25% of the desired initial torque in six of the 32 fatigue trials (note there were 32 fatigue trials in the SCI population as four protocols were tested on both legs of the four individuals). In five of these six instances, the maximum possible current amplitude for the stimulator (126 mA) was reached and in the remaining instance, the torque plateaued with respect to increasing current amplitude. These results suggest that there may have been some layover effect of fatigue in the SCI population, even after 20 minutes of rest. Meanwhile, in the able-bodied population, the torque mismatch was greater than 25% of the desired initial torque in only one of the 32 fatigue trials.

B. Fatigue

1) SCI: Fig. 4 shows the normalized torque for each protocol across all SCI participants as a function of the contraction number. Fatigue times are provided in Table II and the fatigue indices are provided in Table III. ANOVA revealed differences in the fatigue time of the four protocols ($F = 71.41$, $p = 3.110E - 13$) as well as the fatigue index ($F = 22.33$, $p = 1.379E - 7$). *Post hoc* analysis indicated that the mean fatigue time of A8 was significantly longer than that of A16, C32, and C64; A16 was significantly longer than C32 and C64; and C32 was significantly longer than C64. Further, *post hoc* analysis indicated that the mean fatigue index of A8 was significantly larger than C32 and C64; and A16 and C32 were significantly

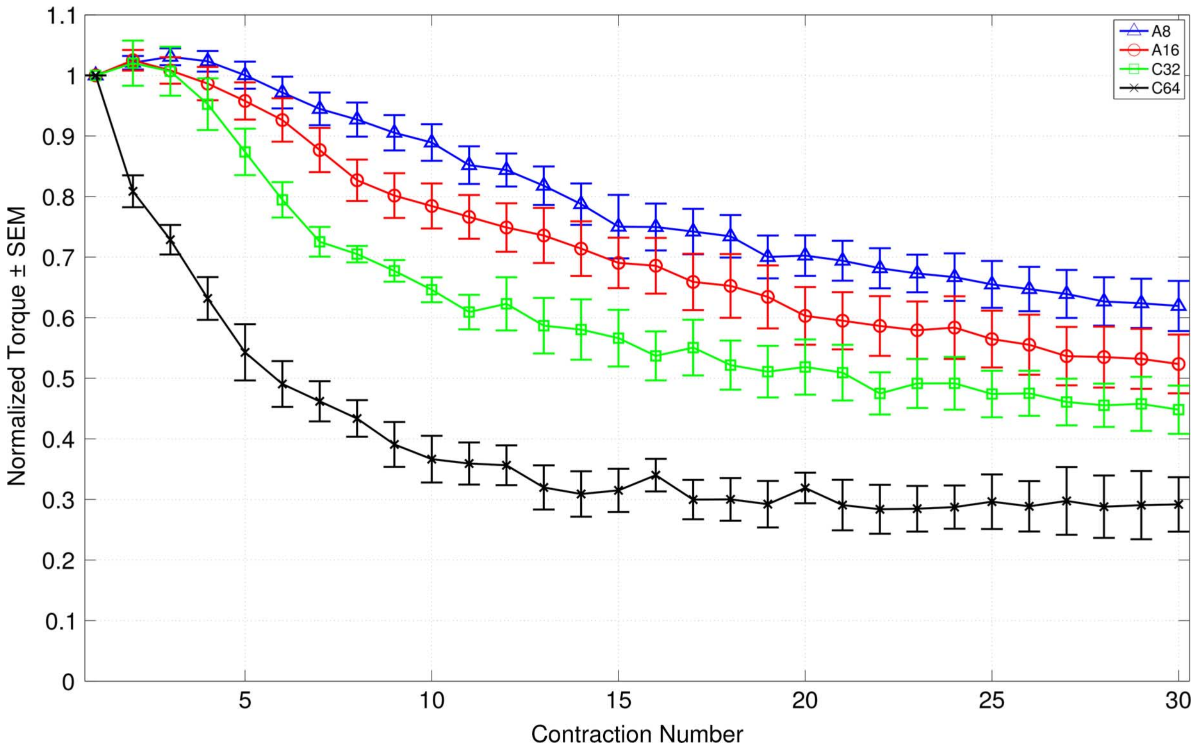

Fig. 4. Fatigue in SCI individuals represented by the normalized torque produced $\pm$ the standard error of the mean (SEM) as a function of the contraction number.

TABLE II
FATIGUE TIME (IN SECONDS)—SCI

Subject-Leg	A8	A16	C32	C64
A - Left	170.0	160.0	50.0	20.0
A - Right	190.0	70.0	40.0	20.0
B - Left	170.0	140.0	60.0	10.0
B - Right	130.0	60.0	40.0	10.0
C - Left	70.0	70.0	50.0	10.0
C - Right	100.0	80.0	50.0	30.0
D - Left	180.0	140.0	60.0	10.0
D - Right	110.0	80.0	60.0	20.0
Mean	$140.0^{a,b,c}$	$100.0^{a,b}$	51.3^{a}	16.3
SD	43.8	39.6	8.3	7.4

[a] *Significantly longer fatigue time than C64*
[b] *Significantly longer fatigue time than C32*
[c] *Significantly longer fatigue time than A16*

TABLE III
FATIGUE INDEX—SCI

Subject-Leg	A8	A16	C32	C64
A - Left	0.689	0.617	0.566	0.238
A - Right	0.720	0.746	0.526	0.361
B - Left	0.604	0.413	0.353	0.170
B - Right	0.615	0.300	0.274	0.090
C - Left	0.384	0.444	0.354	0.157
C - Right	0.628	0.523	0.526	0.358
D - Left	0.746	0.609	0.440	0.432
D - Right	0.544	0.541	0.558	0.473
Mean	$0.616^{a,b}$	0.524^{a}	0.450^{a}	0.285
SD	0.115	0.139	0.111	0.140

[a] *Significantly larger fatigue index than C64*
[b] *Significantly larger fatigue index than C32*

larger than C64. Significant differences between the mean fatigue index of A8 and A16 could not be concluded. Significant differences could also not be concluded between the mean fatigue index of A16 and C32.

2) Able-Bodied: Fig. 5 shows the normalized torque for each protocol across all able-bodied participants as a function of the contraction number. Fatigue times are provided in Table IV and the fatigue indices are provided in Table V. ANOVA revealed differences in the fatigue time of the four protocols $(F = 59.61,\ p = 2.814E-12)$ as well as the fatigue index $(F = 37.27,\ p = 6.527E-10)$. *Post hoc* analysis indicated that the mean fatigue time of A8 was significantly longer than that of A16, C32, and C64; and A16 was significantly longer than C32 and C64. Statistical differences could not be concluded between C32 and C64. Further, *post hoc* analysis indicated that the mean fatigue index of A8 was significantly larger than A16, C32 and C64; A16 was significantly larger than C32 and C64; and C32 was significantly larger than C64.

IV. DISCUSSION

Previous studies have suggested that asynchronous stimulation may reduce NMES-induced fatigue compared to

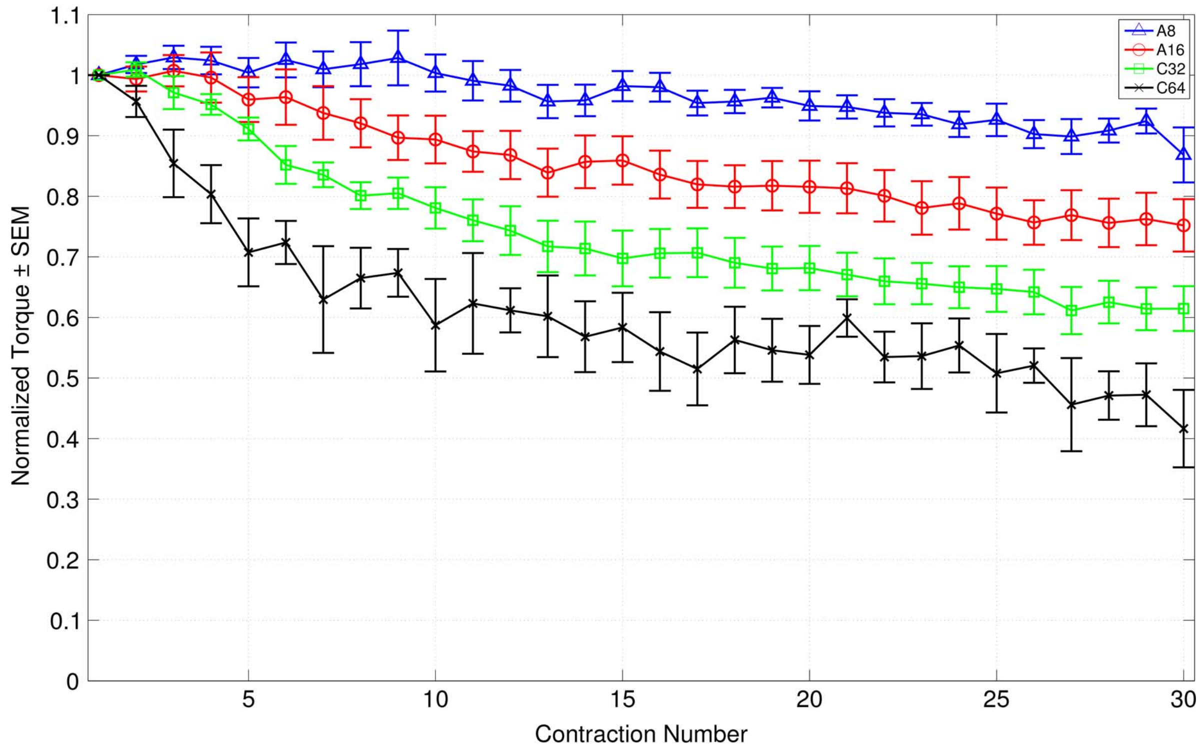

Fig. 5. Fatigue in able-bodied individuals represented by the normalized torque produced ± the standard error of the mean (SEM) as a function of the contraction number.

TABLE IV
FATIGUE TIME (IN SECONDS)—ABLE-BODIED

Subject-Leg	A8	A16	C32	C64
A - Left	300.0	100.0	60.0	20.0
A - Right	300.0	100.0	40.0	30.0
B - Left	260.0	80.0	50.0	40.0
B - Right	300.0	90.0	90.0	70.0
C - Left	240.0	150.0	100.0	40.0
C - Right	300.0	300.0	70.0	40.0
D - Left	300.0	120.0	60.0	30.0
D - Right	300.0	300.0	180.0	50.0
Mean	$287.5^{a,b,c}$	$155.0^{a,b}$	81.3	40.0
SD	23.8	92.0	44.5	15.1

[a] *Significantly longer fatigue time than C64*
[b] *Significantly longer fatigue time than C32*
[c] *Significantly longer fatigue time than A16*

TABLE V
FATIGUE INDEX—ABLE-BODIED

Subject-Leg	A8	A16	C32	C64
A - Left	0.894	0.740	0.617	0.235
A - Right	0.970	0.640	0.658	0.661
B - Left	0.756	0.661	0.612	0.284
B - Right	0.962	0.721	0.681	0.510
C - Left	0.894	0.617	0.488	0.337
C - Right	0.849	0.958	0.683	0.528
D - Left	0.813	0.793	0.451	0.436
D - Right	0.977	0.898	0.736	0.605
Mean	$0.889^{a,b,c}$	$0.754^{a,b}$	0.616^{a}	0.450
SD	0.080	0.123	0.099	0.154

[a] *Significantly larger fatigue index than C64*
[b] *Significantly larger fatigue index than C32*
[c] *Significantly larger fatigue index than A16*

conventional single-channel stimulation. Popovic *et al.* examined isometric knee torque and found that asynchronous 16 Hz stimulation with four channels prolonged the average fatigue interval by 153% compared to conventional 40 Hz stimulation in six individuals with SCI [26]. In a follow up study to [26], Malesevic *et al.* found that asynchronous 16 Hz stimulation with four channels resulted in 26% longer fatigue intervals compared to conventional 30 Hz stimulation in six individuals with SCI [27]. Nguyen *et al.* examined isometric ankle torque and found asynchronous 10 Hz stimulation with four channels resulted in a 280% longer time to fatigue and a 234% higher fatigue index than conventional 40 Hz stimulation in one individual with SCI [28]. Downey *et al.* examined FES cycling and found that asynchronous 16.67 Hz stimulation with six channels almost doubled the fatigue time compared to conventional 30 Hz stimulation in two able-bodied individuals [29]. Maneski *et al.* examined grasping pressure and found that asynchronous 10 Hz stimulation with four channels more than doubled the time interval before the onset of fatigue compared to conventional 40 Hz stimulation in six individuals post-stroke [33]. Sayenko *et al.* concluded that asynchronous 10 Hz stimulation with four stimulation channels is more effective at reducing muscle fatigue compared to 40 Hz conventional stimulation in able-bodied individuals and the reason is that different sets of muscle fibers are activated alternately by the different electrodes [35].

In the SCI population of the present study, asynchronous stimulation resulted in a longer fatigue time than conventional stimulation with A16 yielding a 513% increase over C64 and A8 yielding a 173% increase over C32. In the able-bodied population, asynchronous stimulation also resulted in a longer fatigue time than conventional stimulation with A16 yielding a 287% increase over C64 and A8 yielding a 253% increase over C32. However, it should be noted that the muscle did not fatigue beyond the 80% threshold in six of the A8 trials and two of the A16 trials. Therefore, the true fatigue times are likely to be longer than the reported values of 300 seconds (i.e., the entire duration of the fatigue trials) in Table IV. Asynchronous stimulation also resulted in less total fatigue at the end of the trials compared to conventional stimulation. In the SCI population, A16 yielded an 84% larger fatigue index than C64 and A8 yielded a 37% larger fatigue index than C32. Similarly, in the able-bodied population, A16 yielded a 68% larger fatigue index than C64 and A8 yielded a 44% larger fatigue index than C32.

While it is well known that asynchronous stimulation reduces NMES-induced fatigue compared to conventional stimulation, lower frequency asynchronous stimulation may also be preferred over high frequency asynchronous stimulation. Wise *et al.* examined six-channel asynchronous stimulation in cats at both 6 and 10 Hz [37]. The authors found that 6 Hz stimulation resulted in less fatigue than 10 Hz during intermittent stimulation, but it did not result in less fatigue during continuous stimulation. However, data was only collected in two cats for 10 Hz continuous stimulation and in three cats for 6 Hz continuous stimulation. Furthermore, the authors stated that the discrepancy between intermittent and continuous stimulation may be due to the fact that there was a large amount of potentiation present at the beginning of the 10 Hz continuous stimulation protocol. Although lowering the stimulation frequency did not reduce fatigue during continuous asynchronous stimulation in [37], a more recent study by McDonnall *et al.* also examined continuous asynchronous stimulation in cats. The authors found that four-channel asynchronous stimulation at 15 Hz resulted in a greater fatigue index (i.e., less fatigue) than two-channel asynchronous stimulation at 30 Hz [24].

In the present study, we found that there is a significant advantage to utilizing lower stimulation frequencies, even for asynchronous stimulation, as *post hoc* analysis determined that A8 resulted in significantly longer fatigue times than A16 (40% longer on average in the SCI population and 85% longer on average in the able-bodied population). *Post hoc* analysis also determined there to be a statistically significant difference in the mean fatigue indices of A8 and A16 for the able-bodied population but a statistical difference could not be concluded for the SCI population. However, the data suggests that A8 leads to less total fatigue as A8 resulted in an 18% greater fatigue index than A16 on average for both populations. While there are obvious differences between the able-bodied and SCI populations in the sense that the SCI population exhibited higher rates of fatigue (cf. Tables II and IV) and more total fatigue at the end of the trials (cf. Tables III and Tables V), both populations exhibited the same general trend in the performance of the four protocols.

While the present results are promising in that the fatigue benefits of asynchronous stimulation can be extended by reducing the stimulation frequency, care should be taken when using low frequency asynchronous stimulation in certain applications because low frequency asynchronous stimulation may result in a significant force ripple [42]. Force ripple is unlikely to be a major concern if asynchronous stimulation is used for muscle strengthening, increasing bone mineral density, or other rehabilitative efforts. However, it may pose a problem to FES where feedback control may be required, although efforts have been made to develop controllers that allow for limbs to track a desired trajectory during asynchronous stimulation [43]. Furthermore, while asynchronous stimulation may reduce NMES-induced fatigue, conventional stimulation may be preferred for muscle strengthening therapy as indicated by one study [27]. However, when the goal of NMES is to increase bone mineral density [1], improve cardiovascular parameters [7], [8], or achieve other health related benefits other than muscle strengthening, asynchronous stimulation is preferred over conventional stimulation since it allows the treatment duration to be extended. Similarly, asynchronous stimulation is also preferred over conventional stimulation in FES applications where desired tasks and movements need to be accomplished for as long as possible. In conclusion, the present work demonstrated that NMES-induced fatigue is significantly reduced with asynchronous stimulation and that low frequency asynchronous stimulation significantly reduces fatigue compared to high frequency asynchronous stimulation. The results are promising both for clinical adoption of asynchronous stimulation and for its use in assistive devices.

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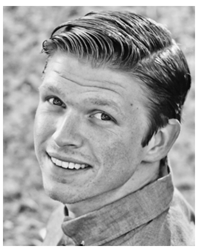

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