# The Effect of Random Modulation of Functional Electrical Stimulation Parameters on Muscle Fatigue

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***Abstract*—Muscle contractions induced by functional electrical stimulation (FES) tend to result in rapid muscle fatigue, which greatly limits activities such as FES-assisted standing and walking. It was hypothesized that muscle fatigue caused by FES could be reduced by randomly modulating parameters of the electrical stimulus. Seven paraplegic subjects participated in this study. While subjects were seated, FES was applied to quadriceps and tibialis anterior muscles bilaterally using surface electrodes. The isometric force was measured, and the time for the force to drop by 3 dB (fatigue time) and the normalized force-time integral (FTI) were determined. Four different modes of FES were applied in random order: constant stimulation, randomized frequency (mean 40 Hz), randomized current amplitude, and randomized pulsewidth (mean 250 $\mu$s). In randomized trials, stimulation parameters were stochastically modulated every 100 ms in a range of $\pm 15\%$ using a uniform probability distribution. There was no significant difference between the fatigue time measurements for the four modes of stimulation. There was also no significant difference in the FTI measurements. Therefore, our particular method of stochastic modulation of the stimulation parameters, which involved moderate (15%) variations updated every 100 ms and centered around 40 Hz, appeared to have no effect on muscle fatigue. There was a strong correlation between maximum force measurements and stimulation order, which was not apparent in the fatigue time or FTI measurements. It was concluded that a 10-min rest period between stimulation trials was insufficient to allow full recovery of muscle strength.**

***Index Terms*—Functional electrical stimulation (FES), isometric contraction, muscle fatigue, spinal cord injury (SCI), stimulation frequency.**

## I. Introduction

FUNCTIONAL electrical stimulation (FES) is a means of evoking contractions in paralyzed muscles by passing small electrical impulses through nervous tissue. It can be used to induce coordinated movements such as walking or grasping [1], [2]. FES has been shown to improve impaired function, to slow down or stop bone and muscle deterioration, and to improve circulation in paralyzed limbs of spinal cord injury (SCI) and stroke patients [3]. However, one of the major limitations is that stimulated muscles tend to fatigue very rapidly, which limits the role of FES in applications such as standing and walking.

Although the exact cause of muscle fatigue is not known, it has been attributed mainly to failure at the synaptic junction, a decrease in transmitter release, and metabolic exhaustion of the contractile mechanism. In the context of SCI, the problem of fatigue is exacerbated by several physiological changes that result from paralysis, including hypertonia and disuse atrophy [4]. In the case of hypertonia, affected muscles are generally overactive, which can put them in an almost constant state of fatigue. Long-term inactivity due to SCI is associated with chronic changes in muscle metabolism, blood flow, and fiber composition [5]–[8]. The bulk of the transformation in muscle fiber type (from slow to fast twitch) due to disuse atrophy occurs during the first ten months after injury. A muscle has greater fatigue resistance in acute paraplegics (less than 10 months postinjury) compared to chronic paraplegics (greater than 10 months postinjury) [9].

One reported solution to the muscle fatigue problem, and the basis for this study, is to apply stochastic modulation to the inter-pulse interval, which is equivalent to randomly modulating the pulse frequency [10]. It was reported that the amount of time that a leg could be extended against gravity was increased by 37% when the inter-pulse interval of stimulation was varied in a range of $\pm 12\%$ (compared to constant frequency stimulation). This was a significant result, but it was limited to a single subject.

Other methods of fatigue reduction have practical limitations. Muscle conditioning is time consuming, requiring several weeks of intense training, and it can lead to a decrease in muscular strength due to the increase in slow fatiguing muscle fibers [5]–[8]. Doublet stimulation, although promising, has demonstrated both a positive and negative effect on the fatigue time depending on the test conditions and protocol [4], [11]–[14]. Sequential stimulation of multiple motor points is not suitable for clinical use on humans since it is invasive, requiring insertion of multiple needle electrodes for each muscle [15]. Intermittent high frequency stimulation has been shown to result in greater contractile forces with less fatigue than intermittent low frequency stimulation in able-bodied and paraplegic subjects [16]. However, due to the extended periods of rest required between pulse trains, intermittent stimulation is limited to cyclic applications such as hybrid orthotics [17]. There remains a clear need for practical solutions to the problem

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of FES-induced muscle fatigue, as well as an understanding of the underlying mechanisms of fatigue.

We decided that the idea of stochastically modulating the inter-pulse interval deserves more attention, and, furthermore, that it is a valuable exercise to extend the notion of stochastic modulation to amplitude and pulsewidth. FES recruits motor units in a highly biased manner unlike natural, physiological contractions, which recruit motor units in a cyclic, asynchronous manner. FES pulses delivered with a constant frequency, pulsewidth, and amplitude induce contractions in the same motor units all the time, whereas natural contractions allow motor units to rest occasionally. As a result, FES causes certain motor units to become overworked while others remain completely inactive. The overworked motor units never rest, and, therefore, fatigue more rapidly than if they were activated by the central nervous system. Gradation in muscle force may be achieved in two ways: by varying the number of motor units recruited and by modulating the firing rate. Recruitment can be increased or decreased by modulating either the pulse amplitude or the pulsewidth. Unfortunately, it is not possible, using surface electrodes, to control which specific motor units are recruited; one can only modulate the number. The firing rate of motor neurons can be increased or decreased by modulating the pulse frequency.

The goal of this study was to reduce the rate of muscle fatigue by randomly modulating FES signal parameters. We hypothesized that by randomly modulating the pulse frequency, amplitude, and width, the resulting firing rate and level of recruitment of motor units would vary over time. A constantly changing firing rate and recruitment level should increase and decrease the total number of active motor units, allowing some motor units on the margin of stimulation brief periods of rest, and, thereby, increase the fatigue resistance during isometric contractions. We proposed two mechanisms by which this may occur. First, variations that exist in the threshold (intensity and duration of stimulus needed to generate an action potential in axons) among motor neurons due to their differences in size and depth could be exploited. Varying the amplitude and pulsewidth of stimulation could excite nerve fibers of differing size and location in the nerve bundles and cause quasi-stochastic contractions of motor units with different contractile properties. Secondly, variation in the frequency of stimulation affects the frequency of action potentials and the amount of neurotransmitter released at the synaptic gap. This could lead to variation in the number of muscle fibers recruited and the level of tetany of each fiber.

## II. Procedure for Paper Submission

### A. Subjects

Nine SCI subjects were recruited from the inpatient and outpatient population at the Toronto Rehabilitation Institute, Toronto, ON, Canada. All participants signed a consent form that was approved by the hospital's Research Ethics Board. Two subjects were excluded from the study because no response could be elicited from the selected muscles using FES, which was most likely due to denervation of the lower limb muscles. Of the remaining seven subjects, one was female and six were male (mean age of 31.2±6.2) and their level of injury ranged from C6/C7 to T8 (see Table I). The time since injury ranged from 3 months to 13 years with a mean of 7.2±4.3 years. Of the seven subjects, all had complete motor paralysis, and six had suffered from traumatic SCI while the seventh had suffered transverse myelitis. Four of the subjects were first time FES users, while three of them had previous training ranging from three months to over a year. Two muscle groups were tested bilaterally for each subject: the tibialis anterior and the quadriceps.

TABLE I
Summary of Subject Data

| Subject | Age (years) | Level of Injury | Injury duration (years) | Prior FES Training |
|---|---|---|---|---|
| 1 | 26 | T2/T3 | 9 | Surface Stim 3 months |
| 2 | 27 | T7 | 0.25 | none |
| 3 | 24 | C6/C7 | 8 | none |
| 4 | 31 | C6/C7 | 7 | none |
| 5 | 29 | T4 | 3 | FES bike 1 year |
| 6 | 38 | C7 | 13 | none |
| 7 | 39 | T8 | 10 | Surface Stim 1 year |
| **Average** | 30.6 | | 7.2 | |
| **Standard Dev** | 5.9 | | 4.3 | |
| **Minimum** | 24 | T8 | 0.25 | none |
| **Maximum** | 39 | C6/C7 | 13 | 1 yr |

We were not able to induce measurable contractions for the right tibialis anterior of one subject and the right and left tibialis anterior of another subject, probably due to peripheral nerve damage. Other results were rejected if the muscle exhibited clonic contractions. In these cases, it was not possible to determine a fatigue time or FTI that was comparable to other results. Results that were excluded include those obtained from the left and right tibialis anterior muscles of one subject and the right tibialis anterior of another subject. In such cases, all trials for the muscle were rejected, so data was analyzed for an equivalent of 22 muscles.

### B. Equipment and Apparatus

Biphasic, bipolar, current controlled stimulation pulses were administered using a stimulator and adhesive surface electrodes (Compex Motion, Ecublens, Switzerland) [18]. A push button was used to trigger the onset of electrical stimulation with a linear ramp-up time of 0.5 s. As shown in Fig. 1(a), a pair of $5 \times 5$ cm electrodes were attached over the proximal (active electrode) and distal (reference electrode) ends of the tibialis anterior muscles. A $5 \times 10$ cm electrode was attached to the skin of the proximal (active electrode) end of the quadriceps and a $5 \times 5$ cm electrode to the distal (reference electrode) end of each of the quadriceps [see Fig. 1(c)]. All tests were performed while subjects were seated in an upright position on a padded bench, as shown in Fig. 1(b). Participants were secured in position with waist and leg straps. Isometric joint force was measured using a strain gage based, tension/compression pancake load cell (Honeywell Sensotec, Columbus, OH) [Fig. 1(d)] with a range of $-1100$ to 1100 N. The signal from the load cell

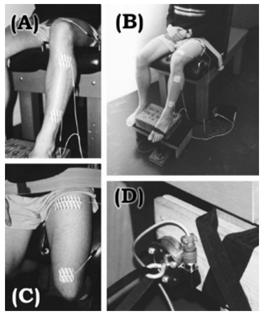


Fig. 1. Photos of the electrode placement on the (A) tibialis anterior and (C) quadriceps, showing the active electrodes located proximally and the reference electrodes located distally. (B) shows the upright sitting position of each subject on the padded bench. (D) shows the strain gauge load cell used to measure force. Person in this figure is able-bodied and was only used to demonstrate the experimental setup.

was amplified using a strain gage conditioner (Daytronic Corp., Dayton, OH) and then passed through an analog to digital converter. Data was sampled at 1000 Hz using data acquisition software written in Labview (National Instrument Co., Austin, TX).

The load cell was mounted on the base of the apparatus in one of two positions. When knee extension moment was being measured, the load cell was mounted anterior to the subject's ankle, and a strap connected in series to the load cell was fixed to the ankle. In the second configuration, the load cell was mounted below the foot rest, and the strap was attached to the foot over the metatarsus. The strap was inelastic, and the load cell was thus used to measure isometric knee extension and isometric dorsiflexion moments.

## *C. Protocol*

Subjects were transferred from their wheelchair to the bench. Before each test, the electrodes were tested for proper placement on the muscle. Stimulus was applied with no randomization while manual resistance was applied to the joint. The pulse amplitude was increased until a level was reached where no further increase in amplitude increased the muscle force or muscle contour as perceived by the investigator. 75% of this value was used as the mean pulse amplitude for all four tests for that muscle. Then, the subject was positioned appropriately and the straps applied accordingly. Four trials were performed on each muscle group; no random modulation of any parameters (control trial), random modulation of pulse amplitude (amplitude trial), random modulation of pulse frequency (frequency trial), and random modulation of pulsewidth (pulsewidth trial). A 10-min rest time was administered between each test, which was considered to be adequate for repeatable results [4], [13], [14]. The order of the trials was randomized. The isometric force, frequently used as an indicator of fatigue in previous studies [4], [12], [13], was measured and recorded during each test. The quadriceps and tibialis anterior muscles were selected as the target muscles since they are easy to stimulate with surface electrodes and easy to measure for force output using a load cell. Each trial was recorded until the muscle force had dropped well below 50% of its initial maximum value.

A mean stimulation frequency of 40 Hz was used. This frequency has been used in previous fatigue tests [4] and was chosen for this study because we wanted to explore fatigue of muscle under tetanic conditions. True tetanization occurs in the range of 30–50 Hz [1]. In the range of 21–27 Hz, which we use in some of our applications, the muscles tend to vibrate noticeably at the stimulation frequency. Randomization of frequency in this range will have less to do with random properties of the signal and more with the tetanization characteristics.

A mean pulsewidth of 250 $\mu$s was used. The pulse amplitude was set between 34 and 110 mA, varying with each subject and muscle group and selected as described above. All three parameters when randomized were varied above and below the mean by 15% using a uniform probability distribution. Values for pulse amplitude, width, and frequency were refreshed every 100 ms. To select the range of randomization, we considered a previous study in which the inter-pulse interval was varied and fatigue reduction was achieved when random modulation was approximately $\pm 12\%$ [10]. We expanded on this experiment by varying the pulsewidth and amplitude as well as the frequency, and to maintain consistency, these values were also modulated by 15%.

## *D. Data Analysis*

Two main indexes of muscle fatigue were considered in this study. First, the "fatigue time" was defined as the duration between the onset of stimulation (time zero) and the point where the force decreased to below 70% of the maximum force. In signal processing and control theory, the following rule exists: when the system's output decreases below $-3$ dB (roughly 70%), it is considered that the system's performance is severely altered and that it cannot be used for practical applications [19]. The same rule was applied to the system in this study consisting of an external stimulator with electrodes (control center), the peripheral motor nerve and the muscle (actuator), and the force of the muscles (output). Since this threshold was chosen arbitrarily, we also conducted the same analysis using thresholds of 60% and 80%.

We also considered the normalized fatigue time integral (FTI), which is defined as follows:

$$\mathrm{FTI} = \frac{\int_0^T F(t)dt}{F_{\max}} \tag{1}$$

where $T$ is the fatigue time for that trial, $F(t)$ is the force over time, and $F_{\max}$ is the maximum force. In this measure, the shape of the curve is taken into account. A gradual decrease of force would yield a lower FTI value than a force that was sustained over the same period of time then dropped off suddenly. The latter case was considered in this study to be indicative of superior fatigue resistance.

Both fatigue time and FTI were calculated from the onset of stimulation, not the point in time when the maximum force occurred. The latter method was not considered to be an accurate measure of muscular fatigue for two reasons. First, muscle activation occurs precisely at the instant that the stimulation begins and, therefore, the muscle begins to fatigue at this instant not at the point of maximum force. Second, the time of maximum force was extremely variable, which caused deviation in the fatigue time and FTI results. Fig. 2 shows a force-time curve

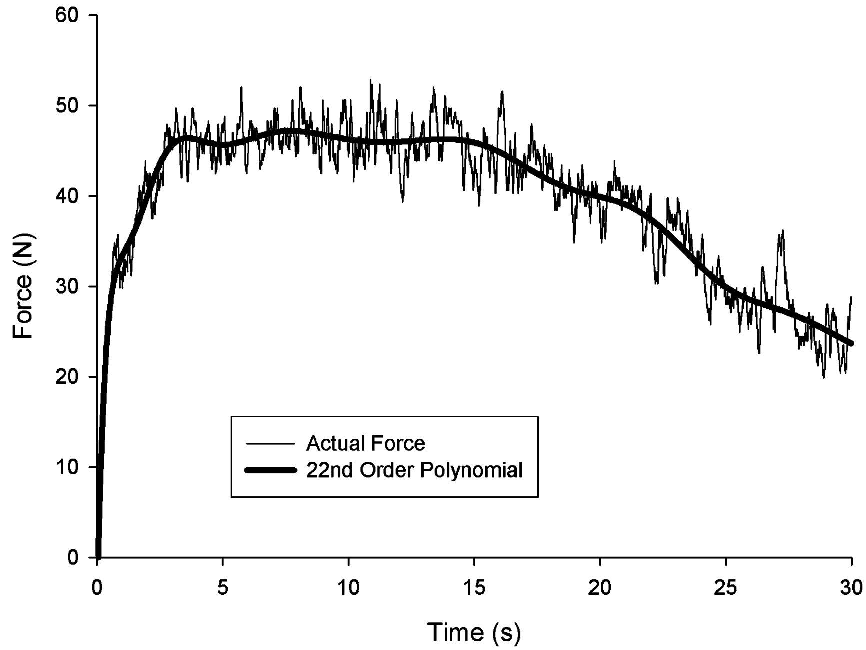


Fig. 2. Force-time curve for stimulation of subject three's right tibialis with amplitude randomization. There are peaks at approximately 3.6, 7.6, and 13.8 s.

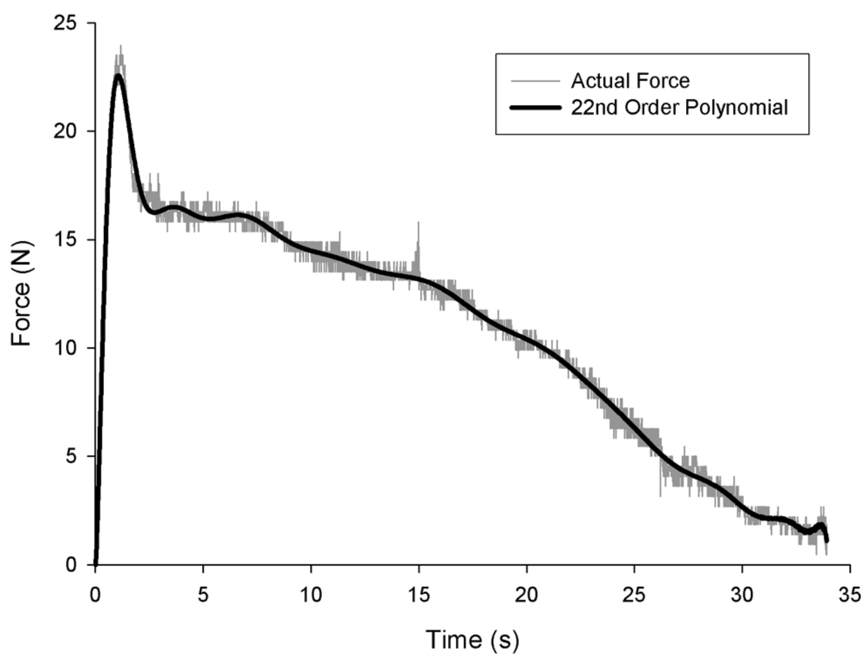


Fig. 3. Example of a curve with a sharp initial peak. This trial was rejected.

with three peaks, all very close to the maximum. Depending on which peak was chosen as the maximum, the fatigue time and FTI values changed drastically.

In order to remove noise and smooth the data, a 22nd order polynomial was fitted to each curve using a least squares algorithm to facilitate data analysis (Fig. 2). The order of the polynomial was determined using an iterative method on a representative sample of curves by increasing the order until the $R^2$ value remained the same (to three significant digits) for consecutive iterations. The polynomial was used to find the instant in time when the force dropped below threshold and the FTI. To approximate how much the stimulation order biased the results, the stimulation order for each muscle was compared to the order in the magnitude of maximum force, fatigue time, and FTI for each test.

Some of the force-time curves exhibited a sharp peak before leveling off (e.g., Fig. 3), which was unexpected. These peaks are considered artifacts, and, therefore, were eliminated from the analysis. We introduced the following criterion to determine such cases: if the fitted curve exhibited a maximum in the first 2.5 s that was more than 25% greater than the maximum value exhibited after 2.5 s, then the trial was eliminated. The peaks were an anomaly only present in a minority of the force-time series (six out of 88) and were, therefore, not characteristic of

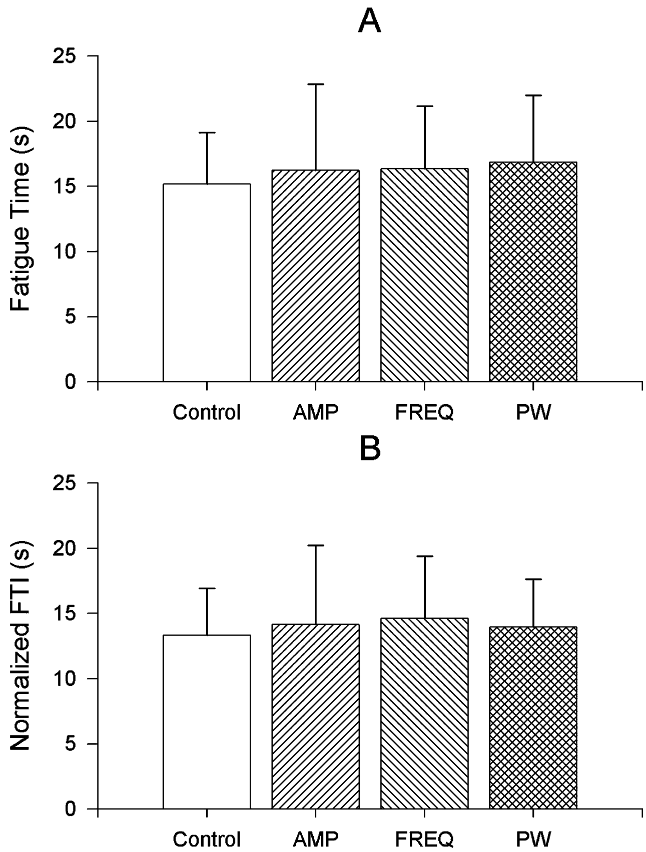


Fig. 4. Average values over all muscles of fatigue time and FTI for the four treatment groups: Constant stimulation (Control), amplitude modulation (AMP), frequency modulation (FREQ), and pulsewidth modulation (PW). $P$ value $> 0.05$ for all tests.

most isometric contractions. Four of these instances involved the tibialis anterior muscles of Subject 6; the other two involved the tibialis anterior muscles of two other subjects.

The effect of the four stimulation modes (control, frequency modulation, amplitude modulation, and pulsewidth modulation) was tested using an analysis of variance (ANOVA) for repeated measures. Quadriceps and tibialis anterior muscles were considered first together, then separately. Separate tests were performed using fatigue time measurements and FTI measurements. We also tested the hypothesis that fatigue time and FTI were sensitive to the order of trials. Similarly, the maximum force measurements were also considered. Statistical significance was set at $P < 0.05$.

## III. Results

Fig. 4 illustrates the average values of fatigue time and FTI for all muscles using the 70% force threshold for the four modes of stimulation. Although random modulation of the amplitude, frequency and pulsewidth produced slightly higher fatigue time measurements than the control trials, the differences were not significantly different ($p$ value $= 0.311$). There was also no significant effect of random modulation on FTI ($p$ value $= 0.436$). Maximum force, however, was clearly affected by the order of stimulation ($p$ value $= 0.0086$). Table II shows the $p$ values resulting from all tests for a randomization effect, and effect due

TABLE II
Summary of Statistical Results From all Hypothesis Tests Using all Dependent Variables on all Muscles. First Column of p-Values Refers to the Test for an Effect of Randomized Stimulation. Second Column of p-Values Refers to the Test for an Effect of the Trial Order

| Repeated measures ANOVA (All muscles) | Threshold (%) | *p-value* Stim. effect | Order effect |
|---|---|---|---|
| Fatigue time | 60 | 0.328 | 0.108 |
| | 70 | 0.311 | 0.255 |
| | 80 | 0.528 | 0.382 |
| FTI | 60 | 0.399 | 0.101 |
| | 70 | 0.436 | 0.257 |
| | 80 | 0.515 | 0.338 |
| Maximum Force | | 0.332 | **0.0086** |

TABLE III
Summary of Statistical Results From all Hypothesis Tests Using all Dependent Variables on Only the Quadriceps Muscles

| Repeated measures ANOVA (Quadriceps only) | Threshold (%) | *p-value* Stim. effect | Order effect |
|---|---|---|---|
| Fatigue time | 60 | 0.403 | 0.255 |
| | 70 | 0.272 | 0.193 |
| | 80 | 0.426 | 0.135 |
| FTI | 60 | 0.305 | 0.303 |
| | 70 | 0.228 | 0.289 |
| | 80 | 0.406 | 0.229 |
| Maximum Force | | 0.840 | **0.0055** |

TABLE IV
Summary of Statistical Results From all Hypothesis Tests Using all Dependent Variables on Only the Tibialis Anterior Muscles

| Repeated measures ANOVA (Tibialis anterior only) | Threshold (%) | *p-value* Stim. effect | Order effect |
|---|---|---|---|
| Fatigue time | 60 | 0.335 | 0.125 |
| | 70 | 0.304 | 0.186 |
| | 80 | 0.498 | 0.522 |
| FTI | 60 | 0.286 | 0.134 |
| | 70 | 0.291 | 0.178 |
| | 80 | 0.388 | 0.369 |
| Maximum Force | | 0.817 | **0.017** |

to trial order. Results of the hypothesis tests when only quadriceps muscles were considered are given in Table III. Table IV gives the results for the tibialis anterior muscles. No significant effects were seen except for the order effect on maximum force.

It was confirmed that the fatigue time and the maximum force measurements were independent. In Fig. 5, all trials are plotted versus maximum force, and a best fit line was determined using least squares regression. There was very little, if any, correlation between the maximum force and the fatigue time at 70% threshold ($R^2 = 0.086$ and $p$ value $= 0.171$). Similarly, Fig. 5(b) shows no correlation between the maximum force and the normalized FTI at 70% threshold ($R^2 = 0.051$ and $p$ value $= 0.134$).

There was no difference seen between subjects with previous FES training and subjects with no previous FES training in terms of fatigue time ($p$ value $= 0.938$) or FTI ($p$ value $= 0.910$) measurements. Furthermore, there was no correlation between length of FES training and fatigue time ($R^2 = 0.027$, $p$ value $= 0.083$) or FTI ($R^2 = 0.019$, $p$ value $= 0.105$).

In Fig. 6, the average force, FTI and fatigue time measurements are shown with respect to the order of the trials. The mag-

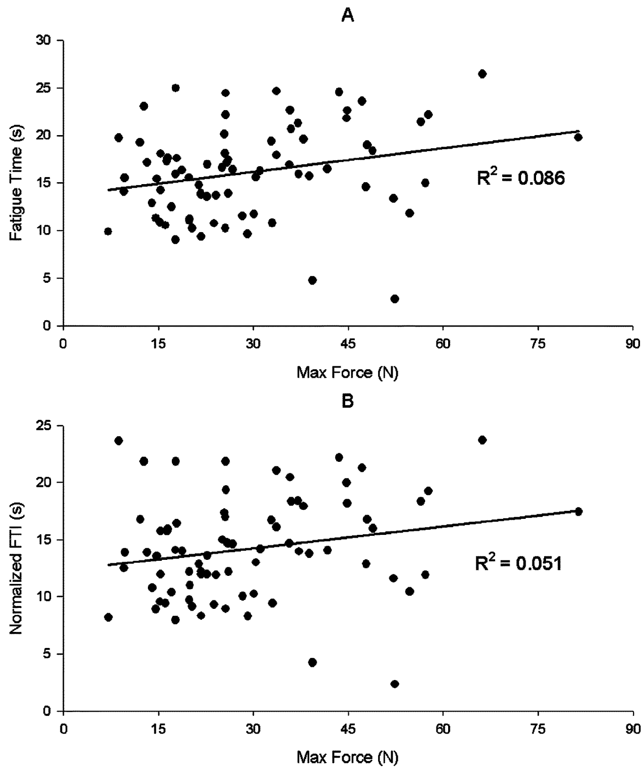


Fig. 5. Linear correlation between (A) fatigue time and maximum force, and (B) FTI and maximum force.

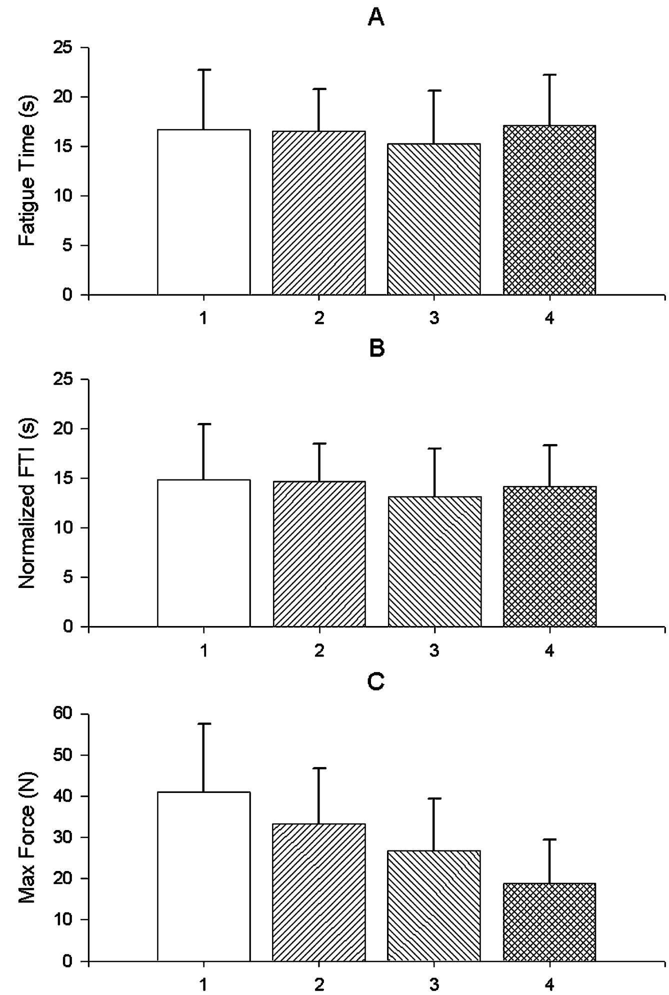


Fig. 6. Fatigue time, FTI and maximum force data averaged for all subjects for the first, second, third, and fourth trial. Stimulation order appeared to have no effect in terms of fatigue time or FTI, however it had a clear, diminishing effect on maximum force ($p = 0.0086$). Error bars indicate ± one standard deviation.

nitude of the maximum force clearly decreases from one test to the next [Fig. 6(a)]. This, however, did not affect the normalized FTI measurements. Were the FTI measurements not normalized, they would have been highly dependent on trial order ($p < 0.0001$). The fatigue time measurements were independent of the trial order ($p = 0.255$).

## IV. Discussion

The human neuromuscular system has evolved such that activation of motor units occurs asynchronously [1], [20]. It is likely that this occurs because it is a most efficient way of producing muscular force with less susceptibility to fatigue. One previous study had demonstrated a significant increase in fatigue time using stochastic modulation of stimulation frequency about a mean of 24 Hz [10]. However, it was only demonstrated on a single subject. The improved fatigue resistance could have been a result of recruiting more muscle fibers and distributing the load over more muscle or of varying the level of tetany over time of the muscle fibers. Our results on seven different individuals showed no overall effect on fatigue of randomly modulating stimulation parameters in the range of $\pm 15\%$ about the chosen mean values. It should be noted that the mean frequency of 40 Hz used in our experiments is significantly higher than most FES applications. Therefore, our results are not comparable to those in [10] and cannot be extrapolated to general cases. A post hoc analysis of statistical power revealed that our experiment is capable of detecting differences in fatigue time of 3.03 s in a paired $t$ test.

The phenomenon of FES-induced muscle fatigue is not well understood. Hence, it is difficult to devise solutions to a problem whose underlying mechanisms are not known. In our experiments, a 10-min rest time was chosen due in part to time constraints and in part because repeatable results have been achieved using a 10-min rest time in previous studies [4], [13], [14]. In addition, studies have demonstrated a full recovery in peak force and endurance from short high intensity stimulation after only 10 min [21] and 95% recovery in peak force from continuous maximum voluntary contractions after only 3 min [22]. Our results did not indicate a full recovery in muscles' potential to reach peak force since the peak force was highly dependent on stimulation order.

Non-normalized FTI, which incorporates force, was also highly dependent on stimulation order and was, therefore, not a reliable measure of muscular fatigue. Furthermore, since it is influenced by maximal force, it does not allow comparison between different subjects, different muscles, or different experiments if the electrodes are removed inbetween. Fatigue time was not highly influenced by the order of stimulation. One possible explanation for this may be that during the first stimulation trial, all fibers participated in the contraction, which led to high forces which declined as the fast-twitch fibers fatigued. In latter trials, the contraction is caused mostly by fatigue-resistant slow-twitch fibers, resulting in lower forces, but not much change in fatigue time since the slow fibers maintain a longer capacity to contract.

Isometric muscle force is a critical factor in many daily activities such as standing and grasping and, therefore, effort is justified in trying to reduce isometric fatigue. We chose to investigate fatigue in isometric conditions for several reasons. First, it is the easiest condition to control experimentally. Second, it is desirable to limit the number of factors, such as stretch velocity and different muscle lengths, so as not to confound the results with too many dimensions.

The force-time curves obtained in this experiment did not lend themselves to an obvious method of data extraction for accurate comparison. As mentioned earlier, the shape of the curves varied. For this reason, we tried several methods of data extraction, including two measures of the fatigue time and the FTI. The time from the zero time to the $-3$ dB force was found to be the most appropriate, but was not without limitation. Curves that were more drawn out and had lower maximum forces, were rewarded with longer fatigue times than curves with higher maximum points. To get around this, the FTI was introduced. Significant, sharp peaks had to be ignored for lack of any method of incorporating them into the fatigue time measurement.

Subjects with previous FES training demonstrated no particular resistance to fatigue when compared to subjects with no previous FES experience. This was a somewhat surprising result, but we had no indication of how intensive the subjects' FES applications had been.

## V. Conclusion

Despite significant efforts to reduce and eliminate the problem of muscle fatigue associated with FES, it remains a major limitation for applications of FES, such as walking and grasping. The particular method of random modulation of frequency, amplitude, and pulsewidth used in this study did not appear to have any effect on the fatigue rate of isometric contractions of the quadriceps and tibialis anterior muscles of subjects with complete SCI. Therefore, we conclude that this particular technique is not viable for fatigue reduction in practice. Fatigue reduction may still be possible using other parameters, such as a lower stimulation frequency. Rest periods of 10 min were found to be insufficient to allow complete restoration of muscle strength between stimulation trials. The order of trials significantly affected the maximum force.

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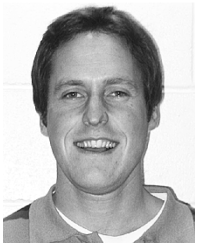

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