Torque History of Electrically Stimulated Human Quadriceps: Implications for Stimulation Therapy

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Summary: The time course of knee extension torque was measured in human quadriceps muscles during 30 min of transcutaneous neuromuscular electrical stimulation (NMES). Ninety subjects were divided into six experimental groups (n = 15 per group), which received stimulation at one of the following frequency/duty cycle combinations: 10 Hz/50%, 30 Hz/50%, 50 Hz/50%, 10 Hz/70%, 30 Hz/70%, and 50 Hz/70%. Two-way analysis of variance revealed that the magnitude of the relative torque decrease (the percentage of decrease in torque relative to the initial value) was significantly different between frequencies (p < 0.005) and duty cycles (p < 0.02), with no significant interaction (p > 0.6). Increasing either frequency or duty cycle caused a greater decrease in torque. In spite of this result, there was no significant difference between groups in the total activity (torque-time integral) achieved during the 30 min treatment session. The magnitude of this activity corresponded to only about 7-14 maximum voluntary contractions. Finally, the average torque during the treatment session was significantly different among groups (p < 0.001), being greatest for the 50 Hz/50% group and least for the 10 Hz/70% group. Taken together, these data suggest that a smaller number of longer duration contractions produces the greatest muscle tension. They also suggest that the absolute torque levels achieved with NMES are relatively low compared with voluntary muscular activity.

The use of functional neuromuscular electrical stimulation (NMES) is widespread in the current practice of physical and occupational therapy (2). The benefits reportedly gained from NMES therapy include reduced postoperative rehabilitation time, increased range of motion of joints, reduced severity of spasticity, and decreased or delayed disuse atrophy (see references in Lieber [10]). Indeed, strengthening muscle through NMES remains a primary objective in its therapeutic application. Interestingly, the results of most animal studies that applied electrical stimulation to neurologically intact muscles have contrasted sharply with the results of human studies (10). For example, in numerous laboratories, in different animal species, and in a variety of muscles, chronic electrical stimulation dramatically alters muscle structure, metabolism, and force-generating capacity. Specifically, chronic stimulation increases the oxidative capacity of skeletal muscle and dramatically decreases force-generating capacity after only 3-4 weeks (6-8).

We suggested recently that some of the differences between animal and human studies may be due to the tension generated by the stimulated muscle (10). Most animal studies permit "normal cage activity" during stimulation treatment, while most human studies maintain a fixed joint angle. Clearly, based on the well known skeletal muscle
recent experiments by Gorza et al. (6) and Kernel et al. (8,9) provided support for this concept — namely, that the absolute tension imposed on a muscle is of critical importance in the determination of the nature of its adaptation to stimulation. These investigators demonstrated that high-tension contractions produced different types of muscular adaptations than low-tension contractions.

Further evidence that supports the concept that muscle tension is important for the maintenance of muscle strength comes from the classic experiments of Tabary et al. (15), Williams and Goldspink (16), and Simard et al. (13). These authors demonstrated that immobilization of a muscle in a lengthened position (for example, immobilization of the gastrocnemius with the ankle fully dorsiflexed) greatly decreased the atrophy observed compared with immobilization of the same muscle in a shortened or neutral position. A related finding was presented by Lieber et al. (12), who demonstrated (in the rabbit) that, by stimulating the tibialis anterior (TA) muscle, the ipsilateral unstimulated soleus (SOL) muscle also received a significant strengthening stimulus, simply due to repetitive passive stretch. This response was even graded to the TA stimulation frequency, higher stimulation frequency resulting in greater SOL strengthening. Taken together, these experiments highlight the importance of muscle tension to maintain or increase muscle force generating capacity.

If the objective of stimulation therapy is to increase muscle strength, it may thus be important to maximize the muscle tension achieved during treatment. While many guidelines are currently available regarding stimulation parameters needed for NMES (2), there has been no systematic study of the relationship between these parameters and the production of muscle tension during treatment. The purpose of this study was to measure the torque-time history of the human quadriceps in response to three stimulation frequencies: 10, 30, and 50 Hz, and two duty cycles: 50 and 70%. These parameters were chosen to represent the nominal range of stimulation parameters applied clinically. A portion of this work has been presented previously (7).

METHODS

Human Subjects

Ninety human subjects were recruited from the University of California, San Diego, population. The subjects ranged from 21-35 years old, had no history of neurological, muscular, or skeletal disease, and had never experienced NMES. The physical characteristics of the subjects are shown in Table 1. Each subject was randomly assigned to one of six stimulation groups (n = 15 per group): 10 Hz stimulation at 50% duty cycle (5 s stimulation, 5 s rest), 30 Hz stimulation at 50% duty cycle, 50 Hz stimulation at 50% duty cycle, 10 Hz stimulation at 70% duty cycle (5 s stimulation, 2 s rest), 30 Hz stimulation at 70% duty cycle, or 50 Hz stimulation at 70% duty cycle. For convenience, these groups will be referred to by frequency/duty cycle: the 10 Hz, 50% duty cycle group will be referred to as the 10/50 group, and so on.

The stimulation waveform that was used was a constant current, bipolar, charge-balanced, asymmetrical waveform with a maximum pulse duration of 250 μs and maximum current intensity of 300 mA. This waveform was chosen on the basis of the studies of Baker et al. (1), who demonstrated its association with the greatest patient comfort and stimulation efficacy. Several other waveforms were tested during pilot studies, but they did not differ substantially from the one ultimately used. Each subject was familiarized with the testing protocol, which was approved by the University of California, San

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**TABLE 1. Characteristics of subjects (mean and SEM)**

<table>
<thead>
<tr>
<th>Variable Studied</th>
<th>10 Hz/50%</th>
<th>10 Hz/70%</th>
<th>30 Hz/50%</th>
<th>30 Hz/70%</th>
<th>50 Hz/50%</th>
<th>50 Hz/70%</th>
</tr>
</thead>
<tbody>
<tr>
<td>Body fat (%)</td>
<td>13.1 ± 1.2</td>
<td>13.3 ± 1.4</td>
<td>14.9 ± 1.5</td>
<td>16.8 ± 1.4</td>
<td>14.6 ± 1.7</td>
<td>10.6 ± 1.6</td>
</tr>
<tr>
<td>Moment arm (cm)</td>
<td>36.5 ± 0.7</td>
<td>35.8 ± 0.8</td>
<td>35.5 ± 0.8</td>
<td>34.5 ± 0.5</td>
<td>36.0 ± 0.7</td>
<td>34.7 ± 0.6</td>
</tr>
<tr>
<td>Age (yr)</td>
<td>24.6 ± 1.1</td>
<td>25.0 ± 1.0</td>
<td>26.4 ± 1.1</td>
<td>26.0 ± 1.2</td>
<td>25.5 ± 1.0</td>
<td>25.6 ± 1.0</td>
</tr>
<tr>
<td>Height (inches)</td>
<td>67.9 ± 1.3</td>
<td>68.6 ± 1.5</td>
<td>67.8 ± 1.2</td>
<td>66.4 ± 0.8</td>
<td>68.8 ± 1.0</td>
<td>68.6 ± 0.9</td>
</tr>
<tr>
<td>Weight (lb)</td>
<td>149.4 ± 8.1</td>
<td>144.8 ± 8.1</td>
<td>149.7 ± 6.9</td>
<td>134.7 ± 5.4</td>
<td>156.3 ± 9.1</td>
<td>154.5 ± 6.2</td>
</tr>
<tr>
<td>Thigh circum. (cm)</td>
<td>48.5 ± 1.2</td>
<td>49.1 ± 0.9</td>
<td>50.4 ± 1.0</td>
<td>47.2 ± 1.0</td>
<td>50.3 ± 1.0</td>
<td>50.9 ± 1.0</td>
</tr>
<tr>
<td>Pre-MVC (Na)</td>
<td>181.0 ± 20.3</td>
<td>168.0 ± 12.0</td>
<td>156 ± 17.0</td>
<td>136.2 ± 12.3</td>
<td>169.8 ± 15.2</td>
<td>200 ± 11.7</td>
</tr>
<tr>
<td>Post-MVC (Nm)</td>
<td>164.6 ± 20.7</td>
<td>143.3 ± 10.3</td>
<td>134 ± 19.2</td>
<td>110.5 ± 10.6</td>
<td>137.4 ± 13.6</td>
<td>176 ± 11.4</td>
</tr>
</tbody>
</table>

MVC = maximum voluntary contraction.
Diego, Committee on the Use of Human Subjects in Research.

**Testing Apparatus**

The subjects were seated in the testing apparatus, which consisted of a chair with a seat-belt installed to minimize hip flexion during quadriceps stimulation, a strain-gauge and strain-gauge conditioner, a computer and terminal, a stimulator, and a chart-recorder (Fig. 1). All items were custom-developed (Preferred Medical Products, Torrance, CA, U.S.A.), with the exception of the strain-gauge (Statham model UC3; Gould Instruments, Cerritos, CA, U.S.A.), strain-gauge conditioner (model 3270; Daytronic, Miamisberg, OH, U.S.A.), and chart-recorder (model 5000; Fischer Scientific, Tustin, CA, U.S.A.). The strain-gauge, conditioner, and computer analog I/O interface were calibrated with known masses and shown to be accurate to 2.3% and reproducible to 1.1% over the range 10-800 N.

The data-acquisition system was based on a PDP-11/73 microprocessor (Digital Equipment Corporation, Nashua, NH, U.S.A.) in a portable chassis (Unbound, Huntington Beach, CA, U.S.A.) running the TSX+ operating system (S&H, South Nashville, TN, U.S.A.). The data-acquisition software, display, and analysis routines were simple modifications of the program previously described by Lieber et al. (11) with use of the Data Translations Analog I/O board set (models 2761/2781; Data Translations, Marlborough, MA, U.S.A.). Specifically, data-acquisition timing and stimulator control were interrupt-driven, with a real-time display of stimulation status. For each NMES-induced contraction, peak torque (in Nm) and integrated torque for that contraction (measured in Nm-min) were calculated, displayed, and stored on disk, which permitted subsequent quantitative determination of the entire torque-time history for each subject.

**Stimulation Treatment Protocol**

The experiment was performed in four phases. In the first, after anthropometric measurement of skinfold thickness, body mass, height, and knee-extension moment arm (measured from the lateral femoral condyle to the middle of the ankle strap), the subject

![FIG. 1. Apparatus used in the present study. Surface electrodes are placed over the quadriceps, and the ankle is attached to a strap which is connected to a strain-gauge (inset). The stimulation parameters, timing, and data acquisition are controlled by a computer.](image-url)
performed a series of maximum voluntary contractions (MVCs). The subject was permitted visual feedback from the chart-recorder in order to achieve a true maximum. The greatest voluntary effort achieved during this series was defined as the MVC regardless of the order in which it was achieved. During MVCs, subjects were instructed not to aid knee extension by elevating the torso or by bracing with the arms. Interday reliability coefficients obtained during pilot tests of 15 subjects exceeded 0.88. Next, following random assignment into one of six groups, stimulation intensity was slowly increased at the consent of the subject (and at the appropriate duty cycle for that group), until the subject no longer wished to increase the intensity. The nominal time for this setup phase was 10 min (Fig. 2).

Stimulation electrodes were placed over the proximal and distal quadriceps motor points (2). The proximal electrode was placed over the lateral border of the rectus femoris at approximately 2/3 of its length as measured from the superior border of the patella (Fig. 1, inset). The distal electrode was placed on the belly of the vastus medialis, approximately 5-7 cm from the superior patellar border. Pilot experiments revealed that this placement resulted in the greatest absolute torque values for knee extension.

Once the stimulation intensity had been selected, the subject was then cyclically stimulated for the 30 min treatment period, during which the computer recorded the “isometric” (fixed joint angle) contractile properties of the quadriceps (Fig. 2). Integrated contractile torque was recorded for each contraction (190 contractions for the 50% duty cycle group and 280 contractions for the 70% group), with raw contractile records stored every 10 s for the first 5 min, every 1 min for the next 5 min, and every 5 min for the remaining 20 min. Thus, a total of 40 raw contractile records and 190 or 280 integrated records was acquired for each subject. Following the 30 min treatment period, the subject again performed MVCs (Fig. 2).

It should be emphasized that the data presented refer to stimulation of naive subjects—those with no previous experience with NMES therapy. Subsequent experimentation has demonstrated a degree of accommodation to the stimulation treatment, and this will be reported separately.

**Statistical Analysis**

Data were obtained from 90 subjects who had been divided into six groups. All data were converted into a format readable by the BMDP statistical package (13). Following data screening to test for outliers and transcription errors (BMDP Program P1D), all data sets were screened for normality and skew (BMDP Program P2D) with the intention of performing parametric analysis. In order to determine the multiple correlation between anthropometric measurements and relative stimulation torque, multiple linear regression was used with F-to-enter = 4.000 and F-to-remove = 3.996. Data transformations were made for those data sets whose values for skew (g1) or kurtosis (g2) were significantly different from that expected for a normal distribution. In general, these were the parameters which were expressed as percentages. Following transformation, a two-way analysis of variance (ANOVA) was performed on

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**FIG. 2.** Typical treatment protocol used in the present study. The subject first performed a series of maximum voluntary contractions (Pre-MVC) to determine maximum extension torque. Stimulation intensity was then slowly increased to the subject’s tolerated maximum (Setup). Stimulation treatment at the selected frequency and duty cycle was then performed for 30 min (Treatment), after which MVC was again determined (Post-MVC). Note the time and torque calibration bar.
data grouped by frequency and duty cycle. Simple linear regression was used to determine the relationship between time and cumulative torque for the different stimulation groups, and all results were considered significant for \( p < 0.05 \). Power analysis revealed a statistical power of approximately 85% for most of the parameters measured. Thus, this experiment was performed in the context in which the probability of Type 1 error was 5% and the probability of Type 2 error was about 15%.

RESULTS

Time Course of Torque Change During NMES

The most general finding was that the nature of the torque change during NMES treatment was highly variable among subjects (Fig. 3). While approximately 30% of the subjects displayed a relatively regular monotonic torque decrease with time (Fig. 3B and D), others demonstrated portions of increasing and then decreasing torque (Fig. 3A and C), while still others demonstrated a relatively constant torque-time history showing no clear trend. As a result, it was seen that application of the simple monotonically decreasing curves obtained from animal stimulation experiments to the human situation was not appropriate. We were confident that the large fluctuations observed were not due to periodic voluntary contraction superimposed on the NMES-induced contractions, and that the small fluctuations in hip angle which were possible while seated in the apparatus could not account for the large, unpredictable changes observed. However, in spite of this overall variable response, the averaged data curves from the six groups revealed several regular trends.

All of the averaged curves (with the exception of the 30/50 group) displayed a monotonic decrease in torque with time (Fig. 4). The 30/50 group demonstrated a transient increase and then a decrease, as has been observed in both human and animal muscle stimulated at intermediate frequencies (12).

Two-way ANOVA revealed that the magnitude of the relative torque decrease (the percentage torque...
FIG. 4. Average torque-time histories for the 50% duty cycle group (A) and the 70% duty cycle group (B). Note that in both cases, initial torque was greatest in the order 50 Hz > 30 Hz > 10 Hz, independent of duty cycle. However, the magnitude of the decline of tension depended heavily on duty cycle, with the 70% groups declining more at a given frequency than the 50% groups (see Fig. 5). Data represent mean ± SEM in all figures.
FIG. 5. Relative decrease in torque in the 30 min treatment period for the six experimental groups. The open bars represent 50% duty cycle groups and the hatched bars represent 70% duty cycle groups. The relative torque decrease was significantly greater with frequency (p < 0.05) and duty cycle (p < 0.02), with no significant interaction (p > 0.6).

As a result, the differences between the two duty cycles decreased as stimulation frequency increased (Fig. 5). The added fatigue due to the higher duty cycle decreased at higher frequencies. In fact, at all times during treatment there was a significant difference in relative torque between frequencies (p < 0.01) and duty cycles (p < 0.05), with no significant interaction (p > 0.1).

There was a significant effect of frequency (p < 0.02), but no effect of duty cycle (p > 0.2) or interaction (p > 0.3) on the initial stimulation magnitude relative to the subject’s MVC. As expected, the 50 Hz group began at the highest torque relative to MVC and the 10 Hz group, at the lowest.

Integrated Torque During Treatment

The findings presented earlier pose an interesting trade-off for maximization of muscle torque during stimulation treatment. The 50/70 group fatigued the most (~60%), but began at the highest relative torque (25% MVC). The 10/50 group, on the other hand, fatigued very little (~20%) and began at a lower torque (10% MVC). The key question, in terms of muscle strengthening, was which combination of stimulation frequency and duty cycle produced the greatest total torque over the 30 min treatment period. To address this question, we integrated torque records from each contraction and then totaled them over the entire 30 min treatment period to yield the torque-time double integral (referred to as “activity”) (Fig. 6). Interestingly, we found no significant effect of frequency (p > 0.25) or duty cycle (p > 0.5) on integrated activity (Fig. 6). However, the trend was that the greatest integrated activity was obtained in the order 50/70 > 50/50 > 30/50 > 10/50 > 30/70 > 10/70. In other words, at high frequencies, higher duty cycles slightly increased activity, whereas at lower frequencies, the added fatigue resulting from the higher duty cycle decreased activity. Thus, in spite of the fact that use of a 70% duty cycle increased the number of contractions from 190 to 280, the total activity did not change significantly.
In spite of the fact that the total activity was not significantly different among groups, the absolute average torque levels achieved during treatment were significantly different among groups (Fig. 7). A significant effect of frequency (p < 0.001) and duty cycle (p < 0.005) with no significant interaction (p > 0.4) was observed, with the greatest average torque produced by the 50/50 group and the least produced by the 10/70. Interestingly, all of the 50% groups generated greater average torques than any of the 70% groups. This reaction was due to the rapid fatigue rate of the 70% groups. Thus, to the extent that absolute torque increases muscle strength (5), the 50/50 treatment would be most effective in its strengthening stimulus.

MVC Change due to Stimulation Treatment

There was no significant effect of frequency (p > 0.25) or duty cycle (p > 0.8) on the MVC decrease secondary to stimulation treatment. This result is consistent with the observation that the total activity among groups was not significantly different. If we conjecture that the MVC change was related to the amount of muscle fatigue during treatment, apparently the muscles were fatigued to the same extent with the different protocols.

DISCUSSION

This study demonstrated differences in the torque decline of the quadriceps muscles with different stimulation parameters. Some of the current results are in agreement with what would be expected on the basis of extrapolation of previous physiological data obtained on animal muscle—namely, that an increased stimulation frequency and/or duty cycle increases the magnitude of the force decline (Fig. 3). However, what could not be predicted on the basis of previous data was that, independent of frequency or duty cycle, the activity was not significantly different with any of the different stimulation parameters. This result was due to the trade-off between frequency and duty cycle: a higher frequency resulted in higher torque but also in greater fatigue, while a higher duty cycle resulted in more activity, due to the increased number of contractions, but in similar net...
activity, due to the increased fatigue rate.

The results have significant implications for those using NMES for muscle strengthening. If muscle strength results from the total tension placed on the muscle, all groups would be expected to be strengthened to the same extent. However, if strengthening results from the absolute tension imposed, it would be most reasonable to stimulate the muscle at the higher frequency (50 Hz) and lower duty cycle, which was associated with the highest absolute tension (Fig. 7). The data from Kernell et al. (8) suggest that high absolute tension may be the most important factor in the gain of strength, far outweighing total activity. Thus, while the total activity was not significantly different among groups, it is very possible that the efficacy of strengthening would be quite different. Kernell et al. (8) demonstrated that a maximal contraction which occurred as seldom as once per hour was as effective in increasing muscle strength as was a maximal contraction delivered once per minute. Muscle endurance capacity, however, was related more closely to the total quantity of stimulation. It is as if these two muscular properties are regulated via different mechanisms.

The energetic basis for the increased fatigue observed at high duty cycles is probably based on the high energetic cost of activating skeletal muscle. It has been demonstrated in humans (3) and animals (14) that muscle activation can account for nearly one-third of the energetic cost of contraction; it "costs" more to activate muscle for a short contraction than to maintain torque levels for a longer duration. These studies suggest that fewer contractions of long duration would be most energetically favorable in producing the maximum torque for the minimum cost.

**Subjective Observations**

In spite of the relatively low level of muscle activation relative to MVC (about 20%), all subjects in the 30 and 50 Hz groups reported significant levels of muscle soreness 24 and 48 h following the treatment. Similar levels of voluntary contraction over 30 min do not result in such soreness (data not shown). No such muscle soreness was reported by either of the 10 Hz groups. However, the 10 Hz groups did report significant levels of discomfort during stimu-
lation, and this was not echoed by the other groups. Our impression was that the discomfort was due to the vibratory movement resulting from the 10 Hz protocol. Interestingly, in spite of the report of discomfort, the 10 Hz group subjects permitted muscle activation at significantly higher stimulation intensities than the 30 or 50 Hz groups \((p < 0.001)\). These data suggest that tolerance to stimulation intensity was limited more by the muscle tension than by direct activation of cutaneous sensory nerves, as has been suggested \((2)\).

**Limitations in Quadriceps Muscle Activation**

Despite the present findings, which may assist therapists in choosing NMES stimulation parameters, the data that were presented do not provide unqualified support for the use of NMES in muscle-strengthening programs. First, the muscles were rarely activated to levels greater than about 25% maximum. Similar levels of voluntary contraction exercise would not be expected to produce a strengthening effect \((5)\). In addition, the mechanism of NMES-induced muscle activation to a 25% MVC level is probably quite different than that achieved voluntarily. It is possible that, with use of NMES, 25% MVC is achieved by activation of 25% of the fibers to 100% maximum as compared with a physiological contraction which may activate a greater fraction of muscle fibers at a lower level. This hypothesis is supported by the observation that, in spite of these relatively low percentages of MVC, the subjects had the impression that muscle tension was extremely high, and significant muscle soreness resulted. These activated fibers may be distributed as a superficial “shell” of fibers near the electrodes or may be predominantly one fiber type selectively recruited in a reverse manner. The mechanism of activation clearly requires further investigation.

To emphasize the relatively low activity achieved with NMES, it is interesting to express total activity in terms of the subject’s MVC. The overall average MVC of the 90 subjects was 168 Nm. Thus, during a single MVC, the torque-time integral was 14.0 Nm-min \((168 \text{ Nm} \times 0.083 \text{ min})\). The 50/70 group, which had the highest activity level (approximately 200 Nm-min) \((\text{Fig. 6})\), thus experienced the equivalent of about 14 MVCs over the 30 min treatment period. At the other extreme, the 10/70 group, which had the lowest activity level (approximately 100 Nm-min), experienced the equivalent of about 7 MVCs over the 30 min treatment period. Clearly, for the effort expended, these results expressed in terms of the subject’s MVC are not overly impressive.

**Determinants of Relative Activation Level**

In order to determine the anthropometric measures (if any) which were best correlated with the relative degree of muscle activation, we performed stepwise linear regression. We hypothesized that the thickness of subcutaneous fat below the electrodes might insulate motor nerve branches from activation or that the absolute size of the limb might influence the depth of the motor nerves at the motor point. However, there was no significant correlation between percentage of body fat \((p > 0.6)\), thigh skinfold thickness \((p > 0.6)\), height \((p > 0.4)\), weight \((p > 0.8)\), sex \((p > 0.6)\), or age \((p > 0.4)\) and the activation level relative to MVC. In one anomalous subject, activation at more than 60% of the MVC was possible, but the reason for this extraordinary performance was not clear.

Lieber recently reviewed the conditions under which previous human NMES studies have been performed \((10)\) and reported that most studies implemented activation levels corresponding to 40-60% of the MVC. Several of those authors were contacted in an attempt to account for the dramatic differences between their studies and the current study. All stated that the results of the present study were not surprising. One significant point is that the subjects in this study were naive in regard to NMES, and it is clear that higher relative stimulation levels are attainable with repeated NMES bouts. While this might alter the absolute magnitude of the effect, it would not necessarily alter the quadriceps torque history or the relative differences in absolute torque levels.

**Acknowledgment:** We would like to thank the subjects who participated in this study for their pleasant cooperation and enthusiasm. We thank Dr. Peter Francis, Dr. Paul Paolini, Dr. Lindsay Carter (SDSU), and Dr. Roland R. Roy (UCLA) for their helpful comments during the course of this work. We also thank Becky Chamberlain for her expert artistic work.

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