

Factors Influencing Quadriceps Femoris Muscle Torque Using Transcutaneous Neuromuscular Electrical Stimulation

*Quadriceps femoris muscle torque was measured in 40 subjects during transcutaneous neuromuscular electrical stimulation (NMES). Three different electrode types (carbonized rubber, sponge, and adhesive) were used on each subject, permitting determination of the factors that influenced the magnitude of quadriceps femoris muscle torque induced by NMES. This goal was accomplished by entering the various factors into a multiple-regression model. The electrodes differed significantly in their characteristics. The carbonized-rubber electrode delivered the greatest current with the lowest impedance, resulting in the highest knee extension torque. We found that the most important factor in determining torque generation level was the quadriceps femoris muscle's intrinsic ability to be activated (as opposed to electrode size, current, current density, or skin impedance). These data suggest that NMES efficacy is primarily determined by the intrinsic tissue properties of the individual (defined in this study as "efficiency") and is not dramatically changeable by using high stimulation currents or large electrode sizes. The precise physiological basis for interindividual differences in efficiency is not known. [Lieber RL, Kelly MJ. Factors influencing quadriceps femoris muscle torque using transcutaneous neuromuscular electrical stimulation. *Phys Ther.* 1991;71:715-723.]*

Richard L Lieber
M Joanne Kelly

Key Words: *Electrotherapy, electrical stimulation; Muscle performance, lower extremity.*

Neuromuscular electrical stimulation (NMES) is often used to strengthen atrophied muscle. A number of physiological studies¹⁻⁵ have demonstrated that muscle strengthening is strongly influenced by the tension imposed on the muscle. One of the clearest dem-

onstrations of the effect of tension was seen in an immobilization study in which rat muscles were immobilized at different lengths.¹ When the muscles were immobilized in the stretched position (under tension), immobilization itself did not necessar-

ily cause atrophy and even caused some initial hypertrophy in some rats. When the identical muscle was immobilized in the shortened position (under no tension), however, significant atrophy resulted. A second example of the influence of muscle tension on muscle strength was recently presented by Kernell et al,² who stimulated the cat peroneus longus muscle with various stimulation patterns. They found that the stimulation pattern that caused the greatest strengthening was the one that produced the greatest average muscle tension. These types of experiments highlight the importance of muscle tension in maintaining muscle strength.

RL Lieber, PhD, is Associate Professor, Department of Orthopaedics (V-151), Veterans Administration Medical Center, and University of California, San Diego, School of Medicine, 3350 La Jolla Village Dr, San Diego, CA 92161 (USA). Address all correspondence to Dr Lieber.

MJ Kelly, MA, is Research Physiologist Consultant, Preferred Medical Products, 4172 Pacific Coast Hwy, Ste 111, Torrance, CA 90505.

This research was generously supported by Preferred Medical Products and the Veterans Administration Rehabilitation Research and Development Program.

The testing protocol was approved by the University of California, San Diego, Committee on the Use of Human Subjects in Research.

This article was submitted January 29, 1991, and was accepted June 4, 1991.

Recently, we measured the time course of tension in human quadriceps femoris muscles in order to determine optimal stimulation factors required for strengthening skeletal muscle.⁶ An unanticipated result of the study was that muscles were stimulated to a level equivalent to only 20% to 25% of the individual's maximal voluntary contraction (MVC) level. Although the majority of NMES studies do not actually measure the torque generated during stimulation, reported values for relative MVC levels typically range from 40% to 60% of MVC (see citations in Table 3 of the article by Lieber⁷) and in one study⁸ even reached 110% of MVC.

Different clinical investigators use a variety of electrodes of different types and sizes, as well as a variety of stimulators with various current outputs. Presumably, the variety of electrode and stimulator designs available reflects the general lack of agreement on the precise conditions under which NMES elicits its greatest response. Thus, the purpose of this experiment was to vary electrode size and design in order to determine the factors that most strongly influenced the relative stimulation torque achieved.

Method

Subjects

Forty subjects who were recruited from the student population and staff of the University of California, San Diego, participated in this study. All subjects ranged from 21 to 35 years of age and had no history of neurological, muscular, or skeletal disease. Each subject was familiarized with the test-

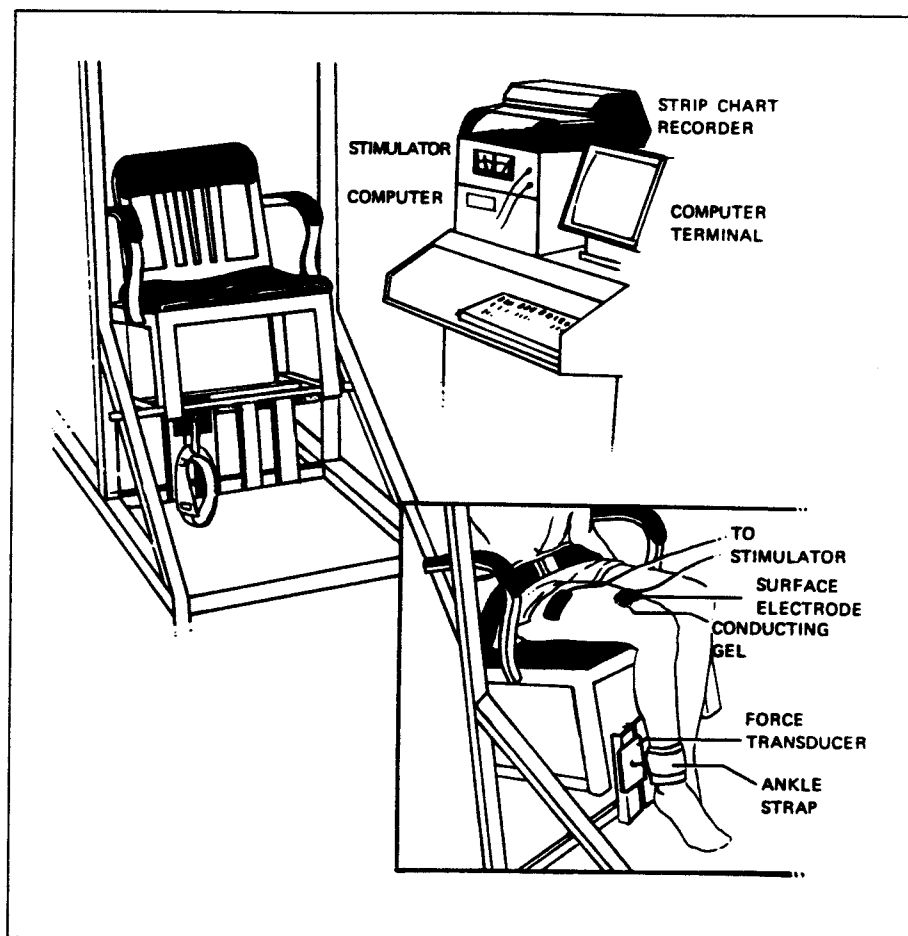


Figure 1. Testing apparatus used in study. Each of three pairs of surface electrodes were placed over subject's quadriceps femoris musculature, and a strap connected to a strain gauge was attached to the ankle (inset). Stimulation variables, timing, and data acquisition were computer-controlled.

ing protocol and signed an informed consent form.

Testing Apparatus

Subjects were seated in the testing apparatus, which consisted of a chair with a seat belt installed to minimize hip flexion during quadriceps femoris muscle stimulation, a strain gauge and

strain gauge conditioner, a computer and terminal, a stimulator, and a strip chart recorder (Fig. 1). All items were custom-developed,* with the exception of the strain gauge,[†] the strain gauge conditioner,[‡] and the strip chart recorder.[§] The strain gauge, strain gauge conditioner, and computer analog input-output interface were calibrated with known masses and shown to be accurate to 2.3% and reproducible to 1.1% over the range of 10 to 300 N.

The data-acquisition system was based on the PDP-11/73 microprocessor^{||} in a portable chassis* running the TSX+ operating system.** Data-acquisition software, display, and analysis routines were simple modifications of the program previously described by Lieber et al⁹ using the Data Translations Ana-

*Preferred Medical Products, 4172 Pacific Coast Hwy, Ste 111, Torrance, CA 90505.

†Statham Model UC3, Gould Instruments, 16310 Arthur St, Cerritos, CA 90701.

‡Daytronic Model 3270, Daytronic Inc, 2589 Corporate Pl, Miamisberg, OH 45342.

§Fisher Model 5000, Fisher Scientific, 2761 Walnut Ave, Tusin, CA 92681.

||Digital Equipment Corp, PO Box CS2008, Nashua, NH 03061.

*Unbound Inc, 15235 Spring St, Huntington Beach, CA 92649.

**S&H Inc, 1027 17th Ave, South Nashville, TN 37212.

log I/O board set.^{††} 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 newton-meters), relative stimulation torque (in percentage of maximal voluntary contraction [%MVC]), and integrated torque for that contraction (in newton-meter-seconds) were calculated, displayed, and stored on disk, enabling subsequent quantitative determination of the entire torque-time history for each subject.

Procedure

An ankle strap attached to a strain gauge was placed around the subjects' distal tibia, and the distance from the position of the strap to the knee joint was measured and used as the moment arm. The knee was fixed at a 90-degree angle.

Subjects were asked to maximally extend the knee joint while verbal encouragement was supplied. This procedure was repeated three to five times until an MVC was achieved. Pilot studies on 15 separate subjects tested on two different test days revealed that MVCs were highly repeatable between days ($r^2 = .91$), in spite of small variations in testing time and ankle strap placement. During the MVCs, the subjects were permitted to view the strip chart recorder displaying the torque record, because visual feedback is known to enhance performance. During MVCs, subjects were instructed not to aid knee extension by elevating their torso or by bracing their arms.

Electrodes and Electrode Placement

Three different sets of electrodes were each placed in random order over the proximal and distal motor points of the subjects' quadriceps femoris muscle. The proximal electrode was placed over the lateral bor-

Table 1. Characteristics of Electrodes Used

Electrode Type	Electrode Area (cm ²)	Material
Carbonized rubber	109	Carbonized rubber
Adhesive	104	Self-adhesive gel
Sponge	54	Sponge/water

der of the rectus femoris muscle at approximately two thirds of its length, as measured from the superior patellar border (Fig. 1, inset). The distal electrode was placed on the belly of the vastus medialis muscle, approximately 5 to 7 cm from the superior patellar border. Electrode characteristics are shown in Table 1. Note that electrode size varied by a factor of two, and electrodes differed with respect to construction material. These size and material variations were chosen to isolate factors that most strongly influenced relative stimulation torque.

Electrode Testing Procedure

The computer-controlled stimulator was attached to the electrode, and stimulation intensity was slowly increased over 10 on-off cycles to the subjects' maximum tolerance. Subjects were strongly encouraged to increase stimulation intensity to a maximum tolerable level. The stimulation waveform used was a constant-current, bipolar, charge-balanced, asymmetrical waveform with a maximum pulse duration of 250 microseconds and a maximum current intensity of 300 mA. Several other waveforms were tested during pilot studies, but they did not differ substantially from the waveform used in this study. Stimulation was administered, 10 seconds on and 10 seconds off, at a frequency of 50 Hz (intensity was ramped on over a 2-second period, held constant for 6 seconds, and ramped off over a 2-second period). The first electrode

pair was then removed, and the procedure repeated with the remaining two electrode pairs. The subjects' skinfold thickness was also measured using calipers to estimate the relative body fat under the stimulating electrodes.

Data Analysis

Measurements of maximum voluntary torque (in newton-meters), skinfold thickness (in millimeters), and thigh circumference (in centimeters) were obtained for each subject. Electrode area (in square centimeters), maximum stimulation current (in milliamperes), and maximum stimulation voltage (in volts) were recorded for each of the electrodes. Skin impedance during maximum stimulation (in kilohms) and relative stimulation torque (stimulated torque/voluntary MVC) were both calculated from the raw data. Skin impedance was calculated from the current-time and voltage-time records acquired during stimulation (Fig. 2). Because voltage rise time was relatively fast (nominally 20 microseconds so that voltage was nearly constant during the 250-microsecond pulse), skin impedance was calculated as the simple ratio of peak voltage to peak current according to Ohm's law (see example in legend to Fig. 2). Skin impedance was extremely consistent within subjects, even if stimulation continued for a long time. We defined stimulation "efficiency" as extension torque/stimulation current (in newton-meters per milliamperes) to represent the intrinsic tissue property relating current input to knee torque output. Data were entered into a PDP 11/73+ computer^{‡‡} and analyzed using the BMDP statistical software package.¹⁰ Differences among

^{††}Data Translations Models 2761/2781, Data Translations Inc, 100 Locke Dr, Marlborough, MA 01752-1192.

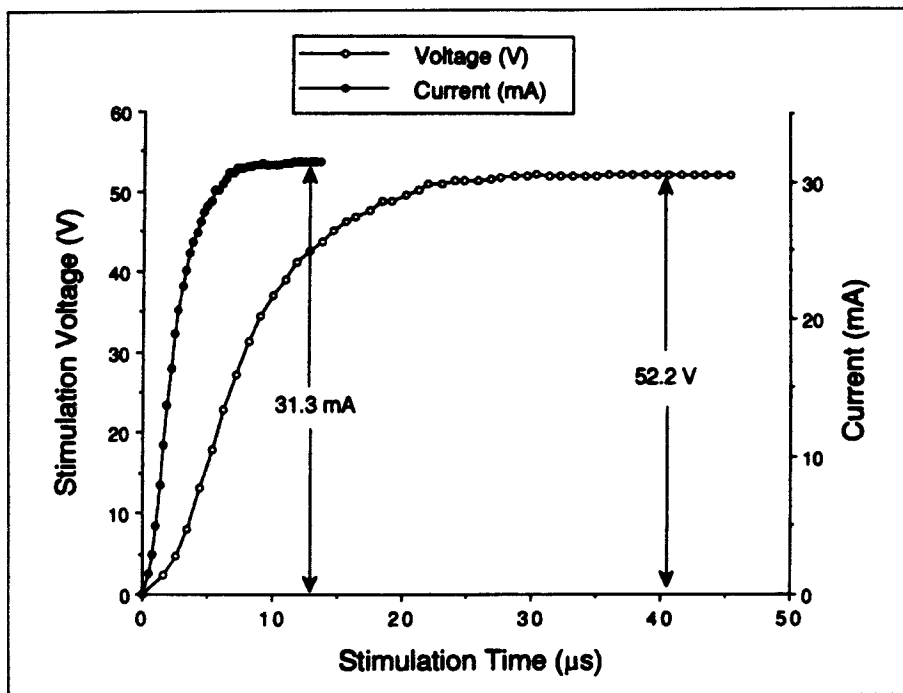


Figure 2. Method for skin impedance calculation. Graph of voltage (left vertical axis) and current (right vertical axis) versus time of digitized data acquired during initial portion of a 250-microsecond pulse via adhesive electrodes. Note that current rise time is faster than voltage rise time because of "charging" of the skin's capacitance. Because voltage rise time was relatively fast (30 microseconds in this example), impedance was calculated as the ratio of voltage to current according to Ohm's law. Skin impedance for this subject was $52.2V/31.3mA=1.67\text{ k}\Omega$. Subject's skin temperature was 33.3°C . Data for current shown only for initial 15 microseconds for clarity. Voltage and current axes scaled to be approximately equal simply for presentation clarity.

the measured variables for the three different electrode types were examined using a one-way analysis of variance (BMDP Program P7D). Pair-wise comparisons were performed using the Bonferroni approximation for three groups. All data sets were explicitly tested for normality and skew prior to the analysis of variance using the BMDP diagnostic software.

The seven independent variables were also entered into a multiple-regression equation and a simple linear-regression equation using relative stimulation torque (%MVC) as the dependent variable. In this way, the multiple-regression model extracted the most important determinant(s) of relative stimulation torque from the variable list. Stepwise linear regression was performed with *F*-to-

enter=4.000 and *F*-to-remove=3.996. These levels were chosen in order to enter only highly significant covariates. The level of statistical significance (α) was chosen as .05, and statistical power (β) exceeded 95% for most variables. All results are presented in the text as mean \pm standard error unless otherwise stated. The 95% and 99% confidence intervals were calculated as mean \pm 2 standard deviations and mean \pm 3 standard deviations, respectively.

Results

General Electrode Characteristics

Significant differences were observed among electrodes. The carbonized-rubber electrode produced the maximum absolute and relative torques

compared with the other two electrodes ($P<.01$, Tab. 2). Apparently, these results were due to the carbonized-rubber electrode's significantly higher current ($48.7\pm 2.2\text{ mA}$) and lower impedance ($0.53\pm 0.02\text{ k}\Omega$) compared with the other two electrodes. Note that these results were not simply due to the large area of the rubber electrode (ie, 109 cm^2), which was almost identical to the area of the adhesive electrode^{††} (ie, 104 cm^2). The only significant difference between the adhesive and sponge electrodes was that the current density of the sponge electrode (ie, $0.62\pm 0.04\text{ mA/cm}^2$) was significantly greater than that of the adhesive electrode ($0.30\pm 0.02\text{ mA/cm}^2$) ($P<.001$). Interestingly, the adhesive electrode was the only electrode for which the subjects complained of a "burning" sensation. They were able to localize this sensation as being beneath the electrode's metallic connector. This increased current density, however, apparently had no functional consequence, because %MVC did not vary significantly between the two electrode types.

Average relative stimulation torque was $15.2\%\pm 16.8\%$ ($\bar{X}\pm\text{SD}$) across all electrodes and subjects. Thus, the upper limit for the 95% confidence interval was approximately 50% of MVC, which is in agreement with our previous measurements.⁶ When considering the carbon-rubber electrode alone, the 95% confidence interval extended up to about 60% of MVC (Tab. 2).

Determinants of Relative Stimulation Torque

Comparison across electrodes using simple correlation analysis revealed that many variables were significantly correlated with relative stimulation torque. The highest correlations were between efficiency and %MVC ($r^2=.76$; Fig. 3, graph D) and absolute torque and %MVC ($r^2=.76$). The correlation between absolute stimulation torque and %MVC was considered trivial (as it followed directly from calculation of %MVC) and was not permitted to enter the stepwise re-

^{††}Jelrode Electrode, TMO Enterprises, Escondido, CA 92075.

Table 2. Values ($\bar{X} \pm SEM$) Obtained Using Different Electrodes ($N=40$)

Variable	Electrode Type			Statistical Comparisons ^a
	Adhesive	Sponge	Carbonized Rubber	
Torque (N·m) ^b	17.9±3.8	18.4±3.0	33.5±4.6	R>A=S
Current (mA) ^b	31.3±2.2	33.2±2.4	48.7±2.2	R>A=S
Voltage (V)	29.3±2.0	27.4±2.0	26.0±1.7	NS
Impedance (kΩ) ^b	0.98±0.05	0.86±0.05	0.53±0.02	R<A=S
Efficiency (N·m/mA)	0.54±0.13	0.53±0.09	0.66±0.08	NS
Current density (mA/cm ²) ^b	0.30±0.02	0.62±0.04	0.45±0.02	S>R>A
%MVC ^{b,c}	12.0±2.7	12.2±2.3	21.2±2.8	R>A=S

^aPaired *post hoc* comparisons among three electrode sets using Bonferroni approximation for multiple comparisons. (R=carbonized-rubber electrodes, A=adhesive electrodes, S=sponge electrodes.)

^bSignificant differences between groups revealed by one-way analysis of variance.

^c%MVC=percentage of maximal voluntary contraction.

gression model. Other variables such as voltage, current, and impedance were weakly correlated with %MVC (Fig. 3, graphs A–C). During the stepwise regression, variables entered the model in the following order: efficiency, current, voltage, and impedance. The remaining variables did not add sufficient information to enter the model. The multiple-regression model was dominated by efficiency, which accounted for 76% of the model variability. The remaining variables—current, voltage, and impedance—accounted for only 9%, 1.9%, and 0.7% of the variability, respectively. Taken together, the variables in the multiple-regression equation accounted for over 87% of the experimental variability ($P<.001$):

$$\%MVC = 21.6E + .613C - .494V + 9.12I - 14.1$$

where E, C, V, and I signify efficiency, current, voltage, and impedance, respectively. These regression coefficients, extracted for the variables (Tab. 3), permitted prediction of %MVC, given the independent variables using the model equation. A graph of predicted %MVC (using the model equation) and actual %MVC revealed a strong linear correlation with a slope greater than unity (Fig. 4). Also of note was that the three different electrodes were not grossly regionalized to any portion of

the graph, suggesting that the electrode design itself was not the major limiting factor of relative stimulation torque.

Subjectively, 38 subjects reported the greatest comfort using the standard carbonized-rubber electrodes and gel, especially at high stimulation intensities. The remaining 2 subjects had no preference as to electrode type.

Discussion and Conclusions

The main result of this study was that the traditional carbonized-rubber electrode was most effective of the three electrode types in generating quadriceps femoris muscle extension torque

using transcutaneous NMES. The carbonized-rubber electrodes apparently produced the greatest torque by applying the highest current through the lowest impedance (Tab. 2). Activation level was limited by subject tolerance; thus, the relative stimulation level was probably limited by discomfort. Because the carbonized-rubber electrode operated at the highest current (>45 mA), discomfort was clearly not simply a consequence of direct sensory nerve activation.

We also found, based on the stepwise regression model, that the most important determinant of relative stimulation torque was not electrode size or stimulation current, or any other

Table 3. Results of Stepwise Linear-Regression Analysis on Percentage of Maximal Voluntary Contraction (%MVC)

Variable Entered	Change in r^2	Regression Coefficient
Efficiency (N·m/mA)	.76	21.6 ^a
Current (mA)	.09	.613 ^b
Voltage (V)	.019	.494 ^c
Impedance (kΩ)	.007	-9.12 ^d

^a%MVC·N·m⁻¹·mA⁻¹.

^b%MVC·mA⁻¹.

^c%MVC·V⁻¹.

^d%MVC·kΩ⁻¹.

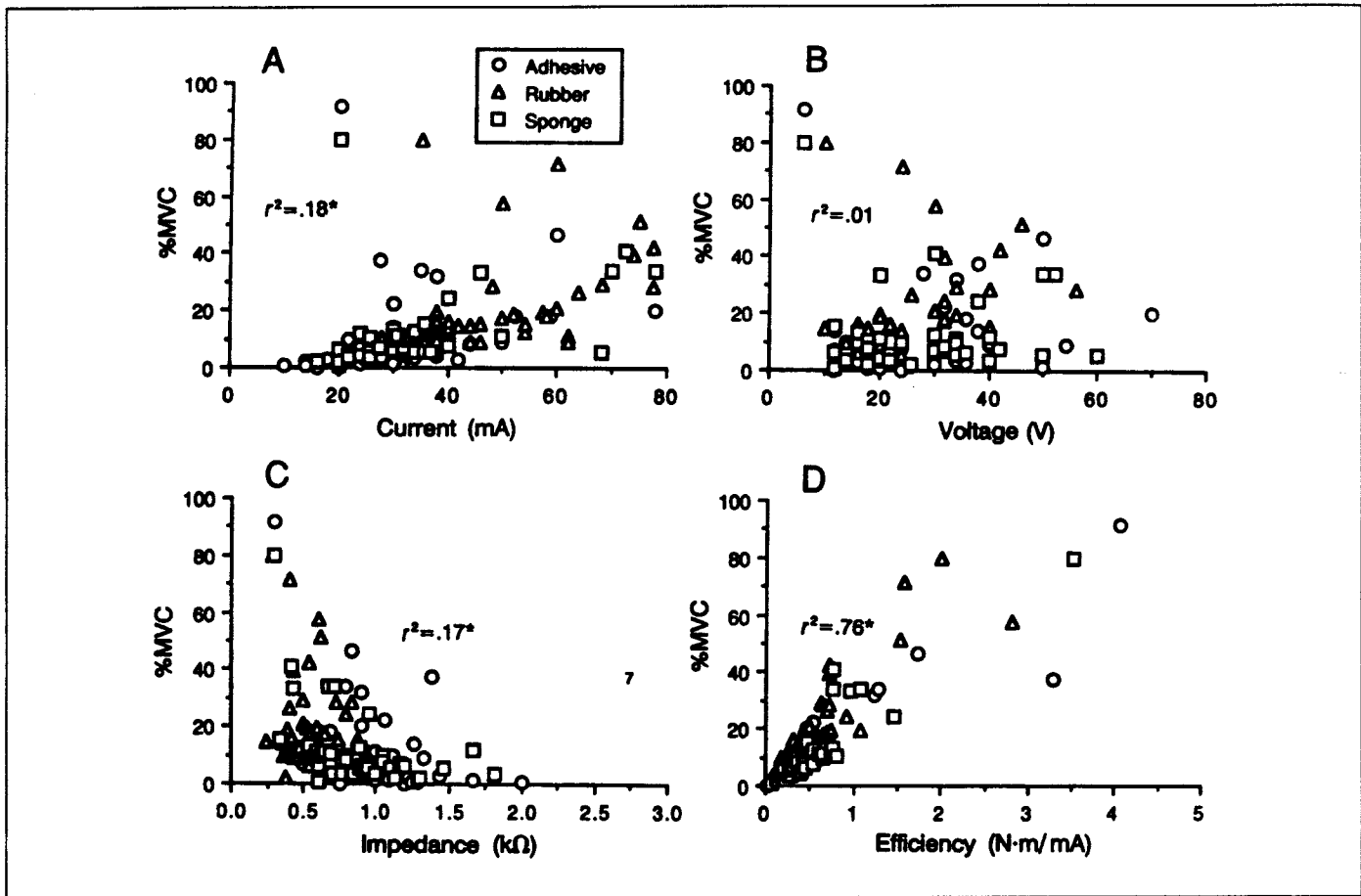


Figure 3. Graphs of simple correlations between stimulation variables and relative stimulation torque (percentage of maximal voluntary contraction [%MVC]). Correlation coefficients (r^2) are shown on each graph. Asterisk indicates significant correlation ($P < .05$). Data are displayed by electrode type (see legend). Note that similar symbols are distributed across all graphs, indicating that electrode type alone was not a major factor in determining %MVC. (A) Stimulation current versus %MVC. (B) Stimulation voltage versus %MVC. (C) Stimulation impedance versus %MVC. (D) Stimulation efficiency versus %MVC.

externally controllable factor, but some intrinsic property of the quadriceps femoris musculature. In spite of the fact that current, voltage, and impedance were correlated with %MVC (Fig. 3), together they accounted for only 12% of the total variability, whereas efficiency alone accounted for 76% of the variability. The implication of this result is that big electrodes (with large areas) and powerful stimulators (with high current outputs) do not necessarily induce high activation levels. For example, both the carbonized-rubber electrode and the adhesive electrode have areas that were approximately twice as great as that of the sponge electrode. Only the rubber electrode, however, allowed the higher relative torque levels, presumably because of its

lower impedance. In our study, across electrodes, electrode area was positively correlated with relative torque, as was stimulation current (Fig. 3). These correlations, although statistically significant, were relatively weak. The fact that stimulation voltage was not significantly correlated with %MVC ($P > .05$) reinforces the concept that stimulation current causes muscle activation, not stimulation potential.¹¹

The data, therefore, suggest that certain individuals are more likely than others to receive effective stimulation therapy based simply on an increased ability to activate their muscles to high tensions. As a result, it would not be surprising that some individuals would respond very well to stimula-

tion treatment, whereas others would not.

Taken as a whole, the literature on NMES is divided as to whether NMES can cause muscle strengthening and prevent atrophy.⁷ An equal number of reports have appeared that support and refute the claim. Our subjective impression, after stimulating over 150 subjects while measuring joint torque in this study and in a previous study,⁶ is that certain individuals are predisposed to relatively high muscle activation levels (see, for example, Delitto et al⁶), probably based on anatomical differences. For example, one of our subjects, who attained 71% of her MVC (94 N·m), did so at a peak current of only 45 mA for a calculated efficiency of 2.1 N·m/mA. A different

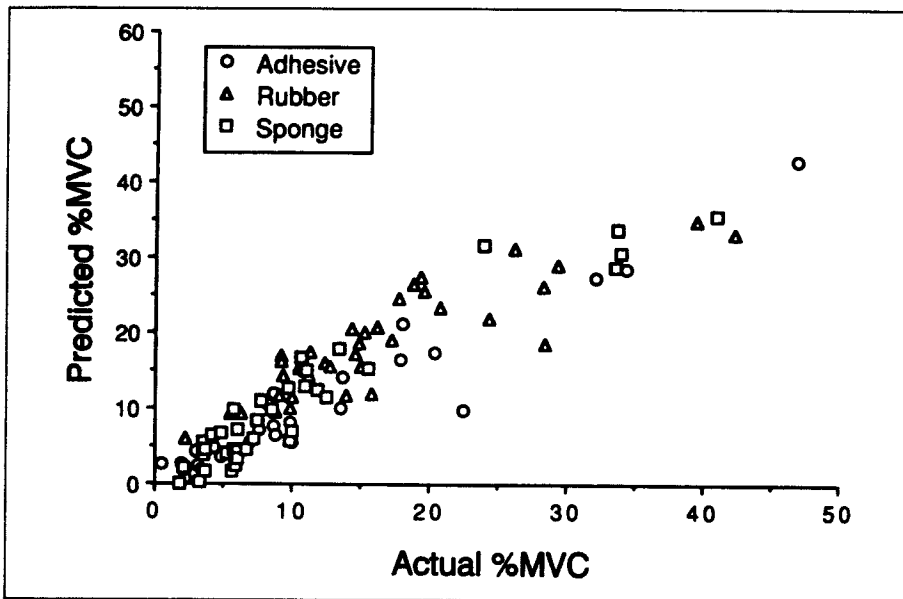


Figure 4. Relationship between predicted percentage of maximal voluntary contraction (%MVC) (vertical axis calculated using model equation) and actual measured %MVC (horizontal axis) for all 120 tests. Data are displayed by electrode type (see legend). Note that all three electrodes are well represented across the entire ranges of relative torque. Note also that the actual data are well approximated by the predicted data. (Data are shown only for %MVC \leq 50% to expand data in lower range. Only three measurements exceeded 50% and were obtained with carbon-rubber electrodes.)

subject, who tolerated over 85 mA of current, generated only 48% of MVC (33 N·m) for an efficiency of only 0.39 N·m/mA. This difference was in spite of the fact that the latter subject had a smaller thigh circumference and skin-fold thickness. This pattern was repeatedly observed and supports the idea that producing high relative torques is not simply a matter of applying high currents to small and lean individuals.

The physiological basis of what we term "efficiency" is not known. Hultman et al¹² clearly demonstrated that transcutaneous NMES indirectly elicits muscle contraction by activating the motor nerves that innervate muscles. When human subjects were curarized,

it was nearly impossible to activate their muscles transcutaneously. We hypothesize that those individuals with high efficiency (as defined in this study) may have relatively superficial patterns of motor nerve branching that render them vulnerable to electrical stimulation. Our data suggest that it would be *extremely rare* to activate quadriceps femoris muscles to a tension level greater than 70% of MVC (99% confidence interval) and *very rare* to activate an individual to a tension level greater than 55% of MVC (95% confidence interval). Clearly, further studies are required to determine the anatomical basis for these observations and to determine whether other measures can be taken to increase stimulation efficiency.

Acknowledgments

We thank the subjects who participated in this study for their pleasant cooperation and enthusiasm. We thank Becky Chamberlain for her expert artistic work. Finally, we thank TMO Enterprises for generously supplying test electrodes.

References

- 1 Simard CP, Spector SA, Edgerton VR. Contractile properties of rat hindlimb muscles immobilized at different lengths. *Exp Neurol*. 1982;77:467-482.
- 2 Kernell D, Eerbeek O, Verhey BA, Donselaar Y. Effects of physiological amounts of high- and low-rate chronic stimulation on fast-twitch muscle of the cat hindlimb, I: speed- and force-related properties. *J Neurophysiol*. 1987;58:598-613.
- 3 Edgerton VR. Neuromuscular adaptation to power and endurance work. *Canadian Journal of Applied Sports Science*. 1972;1:49-58.
- 4 Williams P. Effect of intermittent stretch on immobilised muscle. *Ann Rheum Dis*. 1988;47:1014-1016.
- 5 Lieber RL, Ferro TD, Hargens AR. Differential effects of 10 Hz and 50 Hz stimulation of the tibialis anterior on the ipsilateral, unstimulated soleus muscle. *Exp Neurol*. 1988;100:426-435.
- 6 Kelly MJ, Lieber RL. Human quadriceps muscle fatigue at three frequencies and two duty cycles using electrical stimulation. *Transactions of the Orthopedic Research Society*. 1991;37:41. Abstract.
- 7 Lieber RL. Skeletal muscle adaptability, III: muscle properties following chronic electrical stimulation. *Dev Med Child Neurol*. 1986;28:662-670.
- 8 Delitto A, Brown M, Strube MJ, et al. Electrical stimulation of quadriceps femoris in an elite weight lifter: a single subject experiment. *Int J Sports Med*. 1989;10:187-191.
- 9 Lieber RL, Smith DE, Hargens AR. Real-time acquisition and data analysis of skeletal muscle contraction in a multi-user environment. *Computer Programs and Methods in Biomedicine*. 1986;22:259-265.
- 10 Dixon WJ. *BMDP Statistical Software*. Los Angeles, Calif: University of California Press; 1983.
- 11 Benton L, Baker L, Bowman B, Waters R. *Functional Electrical Stimulation: A Practical Guide*. 2nd ed. Downey, Calif: Rancho Los Amigos Rehabilitation Engineering Center; 1983.
- 12 Hultman E, Sjöholm H, Jäderholm-Ek I, Krynicki J. Evaluation of methods for electrical stimulation of human skeletal muscle in situ. *Pflugers Arch*. 1983;398:139-141.