Published in final edited form as:

Muscle Nerve. 2002 November ; 26(5): 673–680. doi:10.1002/mus.10245.

WITHIN-TRAIN NEUROMUSCULAR PROPAGATION VARIES WITH TORQUE IN PARALYZED HUMAN MUSCLE

Ya-Ju Chang, PhD¹ and Richard K. Shields, PhD²

¹Department of Physical Therapy, Chang Gung University, Taiwan, ROC

²Graduate Program in Physical Therapy and Rehabilitation Science, Roy J. and Lucille A. Carver College of Medicine, 2600 Steiner Building, University of Iowa, Iowa City, Iowa 52242, USA

Abstract

Electromyographic (EMG) recordings may serve an important role in predicting torque during repetitive activation of paralyzed muscle. We compared the initial M-wave to the subsequent M-waves of the same train under fatigued and recovered conditions in the paralyzed human soleus muscle. Sixteen individuals with chronic ($n = 13$) or acute paralysis ($n = 3$) had the tibial nerve activated before and after a repetitive supramaximal stimulation protocol. The mean within-train M-wave amplitude and median frequency increased ~20%, whereas the duration decreased ~15% compared with the initial M-wave of each train. During fatigue, there was a linear decrease in the difference between the initial M-wave amplitude and subsequent train (~20% to 8%). Following fatigue, this difference recovered to ~12%. The difference between the M-wave train average and the initial M-wave for amplitude, duration, and median frequency closely followed torque (Pearson correlations = 0.99, 0.94, and 0.98, respectively) during fatigue. We conclude that the difference between the later-occurring M-waves (average of the train) and initial M-wave is large when muscle torque is high and less when torque is low and, therefore, predicts torque during activation of paralyzed muscle. This difference in the within-train M-wave amplitude, duration, and median frequency may reflect a mechanical change, such as muscle shortening and increased muscle cross-sectional area during isometric contractions. Electromyographic feedback may assist in the optimization of neuromuscular electrical stimulation of paralyzed muscle.

Keywords

electromyography (EMG); excitation–contraction coupling; low-frequency fatigue; plasticity

Fatigue is a major limitation to neuromuscular stimulation of paralyzed muscle after spinal cord injury.^{24,25} Both histochemical and physiological studies support a slow-to-fast conversion of muscle from chronic paralysis.^{30–32} One approach to offset the rapid muscle fatigue and eliminate the instrumentation required for force feedback⁷ is to use electromyographic (EMG) feedback control to optimize the electrical stimulation parameters to paralyzed muscle. However, EMG and force are dissociated in paralyzed muscle, most notably during recovery when the muscle is in low-frequency fatigue (LFF).³² Such fatigue,

attributed to excitation–contraction coupling impairment, has been well described following repetitive activation of paralyzed muscle,^{30–33} and is characterized by a prolonged depression of force despite full recovery of the M-wave. Thus, the absolute M-wave amplitude, duration, or median frequency are poor predictors of muscle force during LFF.

Rather than examine the absolute change in M-wave parameters across time, we explored the relative change within the M-wave train during repetitive activation. This approach considers muscle changes associated with contraction, because the first M-wave of a train occurs before muscle force has been generated, whereas the later M-waves occur after muscle force has been generated. Recent data support the idea that, during in vivo isometric contractions of the soleus, the tendon lengthens and the muscle shortens.^{20,21,23,28} Ultrasound and magnetic resonance imaging (MRI) studies indicate that the Achilles tendon elongates during isometric plantar flexion with the ankle in a neutral position, and the elongation is related to force.^{20,28} The cable theory supports the notion that the action potential conduction velocity of a muscle fiber is proportional to the square root of the muscle diameter.¹ During isometric plantar flexion, an estimated 30% increase in fiber diameter would cause a change in the action potential conduction velocity and, presumably, influence the summation of signals contributing to the compound muscle action potentials (M-waves).²² Moreover, if architectural changes are related to force,^{20,28} then analyzing the first M-wave, with respect to the rest of the train, may indirectly provide information about the force-generating properties of the muscle.

Accordingly, our purpose in this study was to examine the difference between a train of M-waves and the initial M-wave during repetitive activation and recovery of the paralyzed soleus muscle in humans. We hypothesized that this difference would co-vary with muscle force. Some of these results have been published previously in abstract form.³⁴

METHODS

Thirteen individuals paralyzed for over 1 year and three paralyzed for less than 8 weeks participated in this study. All subjects had complete lesions above the T-11 level, as confirmed by a neurological examination. The means and standard deviations for age, years after injury, and soleus fatigue index (percent of initial) were 35 ± 8 , 6.2 ± 3.8 , and 0.34 ± 0.06 , respectively, for subjects with chronic paralysis, and 32 ± 6 , 5.6 ± 1.2 , and 0.87 ± 0.07 , respectively, for subjects with acute paralysis. The acutely paralyzed soleus muscle has been shown to fatigue minimally.³⁰ The subjects with acute paralysis were included to provide a descriptive comparison with a chronic group that was highly fatigable. The specific instrumentation has been described previously.^{30–33}

Mechanical Recordings

The study was conducted with the subjects seated facing the oscilloscope. The right knee was flexed to 90° and the foot was placed in a neutral position on a plate connected to a force transducer (AWU-250; Genisco Technology, Cambridge, MA). The foot was tightly secured to the footplate via an ankle cuff that directed a force through the heel and into the footplate, which prevented the heel from moving during activation of the plantar flexors. The force transducer was aligned perpendicular to the first metatarsal head. A strap attached

to the torque measurement system was secured over the femur, which further secured the leg and foot to the torque measurement system. Torque was calculated as the product of the external moment arm and the perpendicular force measured from the transducer.

Electrical Recordings

Surface electrodes were used to record M-waves from the soleus muscle. The electrodes consisted of two silver–silver chloride electrodes with an 8-mm-diameter center, with an interelectrode distance of 20 mm (Therapeutics Unlimited, Iowa City, IA). Each electrode contained an on-site preamplifier and could be further amplified (500–10,000 times) at the mainframe. The amplifier used a high-impedance circuit, greater than $15\text{ M}\Omega$ at 100 Hz, with a common mode rejection ratio of 87 dB at 60 Hz and a bandwidth of 15–4000 Hz (Therapeutics Unlimited).

Electrical Stimulation

The soleus muscle was activated by electrical stimulation of the tibial nerve using a custom-designed constant-current stimulator with a range of 1–400 V and a constant current up to 200 mA. The stimulator delivered a square wave with a pulse width that ranged from 250 to 1000 μs . The stimulator was triggered by digital pulses from the data-acquisition board housed in a microcomputer under customized software control. The tibial nerve was stimulated in the popliteal fossa with a double-pronged surface electrode placed so that the cathode was distal to the anode. The stimulation intensity was supramaximal and was about twice the intensity needed to produce the largest M-wave.

Experimental Procedures

All subjects provided informed consent and the protocol was approved by our internal review board. The soleus muscle was stretched for several minutes prior to testing. Supramaximal stimulation intensity and EMG amplifier gain settings were established during several single twitches delivered at 1 Hz. Next, four 15-Hz trains were delivered over 8 s to serve as warm-up and to potentiate the muscle. The testing protocol consisted of six 15-Hz tetanic contractions (each containing 12 pulses) delivered before the fatigue protocol, 10 s after the fatigue protocol, and at 5, 10, 15, and 20 min of recovery. The duty cycle was 726 ms on and 1452 ms off. In the subjects with acute paralysis, the testing protocol was performed before the fatigue protocol and 10 s and 5 min after the fatigue protocol.³⁰ The fatigue protocol consisted of 120 contractions using the same work–rest cycle.

Data Analysis

Force was digitally sampled at 1000 samples per second and EMG was sampled at 5000 samples per second and analyzed using customized software. A fatigue index (FI) was calculated by taking the torque at the 120th contraction and dividing it by the torque generated by the first contraction. The amplitude, area, median frequency, and duration were measured for each M-wave. Amplitude was measured from the maximum positive peak to the minimum negative peak of each M-wave. Duration was measured from the start of the M-wave to the minimum negative peak. The M-wave area was calculated from the start to the end of the M-wave (defined as the first zero baseline crossing following the minimum

negative peak). A fast Fourier transform was carried out on the M-waves and the median frequency was calculated from the transformed curve.

The average M-wave amplitude, duration, and median frequency and its variation were used to represent the relative change in the M-wave within each train. As the variation (standard deviation) can be affected by the size of the mean, we used the coefficient of variation (standard deviation/mean) as a relative measure of within-train variation of the M-wave. The coefficient of variation of each of the M-waves (amplitude, duration, and median frequency) was calculated from all of the M-waves generated within each contraction.

The average of pulses 2–12 was calculated for the amplitude, duration, and area. The average amplitude, duration, and area of the M-wave train were normalized to the initial M-wave values of the prefatigue test train to form a ratio. Likewise, the initial M-wave of each train was also normalized to the initial M-wave of the prefatigue test train to form a ratio. The percentage difference between the average M-wave train and the initial M-wave of each train was determined by the following equation: $[(\text{M-wave train average} - \text{initial M-wave}) / (\text{initial M-wave})] \times 100$.

We used a two-way repeated-measures analysis of variance to determine whether the initial M-wave amplitude, duration, and median frequency were different from the remaining M-wave train average and whether these M-wave properties changed during the fatigue protocol and during recovery. The percentage difference between the average train M-waves and the initial M-wave was correlated with the torque using regression analysis.

RESULTS

After the 2-min fatigue protocol the group with chronic paralysis showed a significant reduction in torque (mean fatigue index $\pm$ SD; 34% of initial, $\pm 6\%$). The torque remained decreased (mean $\pm$ SD; $\sim 55\%$ of initial, $\pm 5\%$) during all phases of recovery. The subjects with acute paralysis showed minimal torque reduction supported by a nonsignificant change in the fatigue index (mean 87% of initial, $\pm 7\%$).

Figure 1 shows an example of one subject's train of M-waves before fatigue (Fig. 1A), 10 s after fatigue (Fig. 1B), and at 10 min following fatigue (Fig. 1C) of the chronically paralyzed soleus muscle. Before the fatigue protocol, the M-wave amplitude increased within the train whereas the M-wave duration decreased within the train. As indicated in Figure 1A, the greatest within-train change in the M-wave occurred at the start of the protocol.

The initial M-wave and average of the train of M-waves are presented in Figure 2. The M-wave train average was always higher than the initial M-wave for amplitude and median frequency ($P < 0.05$). Likewise, the average M-wave train duration was always less than the duration of the initial M-wave of the train ($P < 0.05$). The difference between the train of M-waves and the initial M-wave was greatest during the prefatigue condition (Fig. 2). The difference between the average of the M-wave train and the initial M-wave of the train for amplitude, duration, and median frequency became less at 10 s after the fatigue protocol.

(Figs. 2 and 3). During recovery (5, 10, 15, and 20 min), the difference between the M-wave train average and initial M-wave began to increase.

The percent difference between the M-wave train average and the initial M-wave of each train is presented with the torque in Figure 3. Before fatigue (1 s), when the muscle torque was high, the within-train M-wave average amplitude and median frequency were 20% and 24% greater, respectively, than the initial M-wave amplitude of the same trains. The within-train M-wave duration was also 15% shorter than the initial M-wave duration of the same trains. Conversely, at 120 s of the fatigue protocol, when the force was low, the difference between the train average and the initial M-wave of the same train for the amplitude and median frequency was less than 5%. During recovery (5, 10, 15, and 20 min), when the torque partially returned (~55% of initial), the difference between the average of the train and the initial M-wave for the amplitude and median frequency increased to between 10% and 15%, respectively (Fig. 3). The difference between the M-wave train average and the initial M-wave for amplitude and median frequency was significantly less ($P = 0.05$) during fatigue (60, 90, and 120 s) and 10 s after fatigue than before fatigue (Fig. 3). The difference in amplitude and median frequency at 5, 10, 15, and 20 min of recovery was greater than the difference at the 1-s condition ($P = 0.05$). Interestingly, the difference between the train of M-waves and the initial M-wave for amplitude and median frequency followed the average change in torque.

Pearson correlation coefficients between torque and the differences between the trains of M-waves and the initial M-wave were 0.99 (amplitude), 0.94 (duration), and 0.98 (median frequency) (Fig. 4). When 20 min of recovery was included in the analysis, the correlations decreased to 0.94, 0.75, and 0.68, respectively. These correlations are in contrast to 0.44, 0.52, and 0.60, which were the correlations with torque when the initial M-wave alone (amplitude, duration, and median frequency) was considered during fatigue and recovery.

The coefficient of variation for the within-train M-wave amplitude, median frequency, and duration decreased after 1 min of the fatigue protocol in the chronically paralyzed group ($P = 0.05$) (Table 1). During recovery the coefficient of variation for the M-wave amplitude, duration, and median frequency was less than the prefatigue condition ($P = 0.05$). The within-train M-wave variation was always less when the soleus muscle torque was low. In the acute group, the within-train average was significantly different from the initial M-wave both before and after the stimulation protocol. The average train amplitude, duration, and median frequency were 16% greater than, 14% less than, and 17% greater than the initial M-wave. The acute subjects did not fatigue and therefore showed effects similar to those of chronic subjects before they were fatigued. Thus, when muscle torque was high, regardless of whether it was in acutely or chronically paralyzed muscle, there were large differences between the first M-wave of a train and the remaining M-waves.

DISCUSSION

The amplitude, duration, and median frequency for the trains of M-waves were different from the amplitude, duration, and median frequency of the initial M-waves before and after repetitive stimulation in chronically paralyzed muscle. The magnitude of the difference

between the within-train M-wave average and the initial M-wave (amplitude, duration, and median frequency) decreased during repetitive activation as torque declined with muscle fatigue in the chronic group. When force did not decline, because of minimal fatigue in the acute group, the difference between the within-train M-wave average and the initial M-wave remained high. The magnitude of the within-train M-wave variation (amplitude, duration, and median frequency), expressed as the coefficient of variation (CV), decreased with muscle torque during fatigue. The M-wave within-train variation decreased to a lesser extent when the torque was partly recovered but still depressed from the prefatigue condition (~55% of initial). These findings suggest that muscle torque was associated with the within-train M-wave variation during repetitive activation of paralyzed muscle. Moreover, by assessing the difference between the train of M-waves, when torque is produced, relative to the initial M-wave, when torque is not produced, the overall predictability of torque was improved when compared with using absolute between-train EMG measures.

Within-Train Changes in M-Wave

The within-train M-wave variation for all parameters was greatest when the muscle torque was high. The within-train M-wave variation decreased with fatigue or as the force generated by the muscle was reduced. This finding is contrary to the increased within-train variation described by others for the primarily fast rat hindlimb muscle.^{12,13} Enoka and colleagues^{12,13} suggested that fatigue increased the probability that a motor unit would drop out, which would increase the within-train variability during repetitive activation of fast muscle. In contrast, our findings show that the within-train M-wave variation decreased, along with torque, during the fatigue protocol.

The reduced M-wave variation during fatigue in our study may be attributable to several factors. The stimulation protocol used previously was an intermittent 40-Hz frequency,¹³ whereas our protocol used a 15-Hz intermittent frequency. Thus, it may take a higher frequency of stimulation to induce a within-train transmission failure that causes increased variation. If motor unit dropout occurred, we would have expected an increase rather than decrease in the within-train M-wave variation during repetitive activation.^{3,8,9,26,27,36} Enoka and colleagues¹³ also studied the rat hindlimb with fine-wire electrodes rather than surface EMG electrodes. The muscle geometry as viewed by a surface electrode is quite different from that viewed by an indwelling wire electrode. The potential use of surface EMG as a “more feasible” feedback control method to control neuromuscular stimulation in humans underscores the importance of understanding this relationship between surface-recorded within-train M-wave changes and torque during repetitive activation of paralyzed muscle.

Between-Train Changes in M-Waves during Fatigue

The M-wave median frequency and duration are correlated with the conduction velocity of the muscle fiber membrane.^{22,37} The fatigue protocol utilized in this study also induced changes in the between-train duration and median frequency consistent with a slowing of the muscle fiber conduction velocity. Changes in the M-wave duration and median frequency during fatigue may be attributable to a prolongation of individual motor unit action potentials, a selective dropout of the fast motor units, or unsynchronized firing of motor units. Selective dropout of the fast motor units should be associated with a decrease in M-

wave amplitude.³⁵ Instead, the amplitude of the initial M-wave potentiated early during the fatigue protocol, suggesting that units did not drop out. Surface EMG electrodes measure the sum of the individual motor unit action potentials.^{11,12,19} Increasing the nerve conduction time (or decreasing the nerve conduction velocity) of some motor units may result in less synchronized muscle fiber action potentials, prolonging the duration of the summed M-wave. However, minimal evidence exists that these phenomena occur during fatigue. Previous investigators reported that fatigue has little effect on axonal conduction velocity.^{14,15} Therefore, altered arrival times of the axonal action potentials at the endplates do not appear likely. Moreover, a decrease in the conduction velocity of the fastest fibers,^{32,39} which are primarily found in chronically paralyzed muscle, should result in more synchronous muscle fiber action potentials.¹⁴

Potential Mechanical Contributions to the within-Train M-Wave

Isometric contractions in humans are typically associated with some muscle shortening.^{20,23,28} Surface EMG electrodes measure the summed motor unit action potentials within the detection field. However, varied compliance in the tendon may cause uneven movement of the signal source (muscle) with respect to the surface electrode. Consequently, the detection field may be altered as the muscle shortens slightly under the electrode. This would assist in explaining the differences between the initial and the remaining M-waves within a train. After fatigue, less torque may result in reduced muscle movement and the difference in within-train M-waves becomes less. Recent reports support the notion that tendon elongation is linear to the magnitude of the triceps-surae torque during isometric contractions in vivo.²⁰ Indeed, tendon elongation was linearly increased when the muscle was generating from 20 to 80 Nm of torque.²⁰ Accordingly, the difference between the M-wave train average and the initial M-wave for the amplitude should be greater when the muscle force is high. Conversely, when the muscle force is low, as occurs with fatigue, less sarcomere movement would be expected and the difference in within-train amplitude would be less.

The distance between the core of the muscle fiber and the recording electrode increases with muscle shortening as the muscle fiber diameter increases.²⁹ Shortening of the muscle during isometric contraction would be expected to change fiber diameter and therefore the fiber's conduction velocity.¹ The cable theory predicts conduction velocity will increase as the square root of the fiber diameter.^{1,2} A 35% increase in physiological cross-sectional area may be realistic for the plantar flexors during maximal isometric contraction,^{22,28} and therefore the muscle fiber action potential conduction velocity should increase accordingly. An increased action potential conduction velocity indirectly causes an increase in the median frequency of the M-wave and a decrease in the M-wave duration.^{2,4,6,17,38} If (1) the estimated 35% increase in cross-sectional area of the plantar flexors is accurate, (2) the conduction velocity increases as the square root of the fiber diameter, and (3) the median frequency and duration are related to conduction velocity, then it is reasonable to expect an increase in the median frequency and a decrease in the duration within the M-wave train during high force contractions. However, we did not measure the associated mechanical changes in the muscle during these experiments and, therefore, can only speculate that the changes observed in the within-train M-waves were attributable to mechanical effects.

Clinical Implications

Absolute changes of the initial M-wave of the train, which occur before any force, reflect the physiological changes of the muscle membrane with fatigue. However, the within-train M-waves, which occur after force is produced, appear to be influenced by muscle shortening. Therefore, the M-wave train changes, relative to the initial M-wave of the train, should provide meaningful information that assists in predicting torque. This issue is important because the sarcolemmal membrane is known to be dynamic. At times, the membrane becomes hyperpolarized due to changes in the sodium–potassium pump causing M-waves to potentiate, independent of force, during both fatigue and recovery.^{16,18,31} As importantly, the source of peripheral muscle fatigue can be quite distant from the sarcolemma,^{4,5,10,30–32} which is the reason that a clear dissociation occurs between the absolute M-wave measurements and force during low-frequency fatigue.

Because potentiated M-waves occur during various times of repetitive activation, we adjusted for these changes in the sarcolemma by taking the difference between the average of the train of M-waves relative to the initial M-wave of each train. Thus, by subtracting the average M-waves of the train (during muscle force) from the initial M-wave (without muscle force), we explored the potential effects of muscle force on the M-waves. Accordingly, the M-wave differences, as measured in this study, were found to provide an indirect measure of muscle torque. This approach to predicting torque is novel and has not been examined previously (to our knowledge) in paralyzed human muscle. A limitation of this method is that it does not distinguish among the mechanisms (i.e., sarcolemmal membrane changes and mechanical influences) that can contribute to within-train M-wave changes. Moreover, the mechanisms that contribute to within-train M-wave variability probably vary according to the stimulation frequency and duration of the activation protocol.

Improved use of neuromuscular electrical stimulation of paralyzed muscle relies on feedback, such as EMG, to optimize torque generation of paralyzed muscle. Placing an EMG electrode on a limb is more feasible than instrumenting a limb with force transducers. A normalizing procedure that focuses on relative changes within the train of M-waves over time, rather than absolute changes between contractions, appears to have promise in predicting changes in muscle torque. More studies that explore the relative importance of factors predicting torque during fatigue and recovery are essential if function is to be restored by electrical stimulation of paralyzed muscle.

Acknowledgments

This work was supported by grants from the National Institutes of Health (NIH R01HD39445) (R.K.S.) and the Paralyzed Veterans of America Spinal Cord Research Foundation (R.K.S.).

Abbreviations

EMG	electromyography
EC	excitation–contraction
LFF	low-frequency fatigue

MRI magnetic resonance imaging

REFERENCES

1. Aidley, DJ. The physiology of excitable cells. 4th edition. Cambridge, UK: Cambridge University Press; 1998.
2. Arendt-Nielsen L, Zwarts M. Measurement of muscle fiber conduction velocity in humans: techniques and applications. *J Clin Neurophysiol.* 1989; 6:173–190. [PubMed: 2708517]
3. Bellemare F, Garzaniti N. Failure of neuromuscular propagation during human maximal voluntary contraction. *J Appl Physiol.* 1988; 63:1084–1093. [PubMed: 3366731]
4. Bigland-Ritchie B, Donovan EF, Roussos CS. Ion velocity and EMG power spectrum changes in fatigue of sustained maximal efforts. *Respir Environ Exerc Physiol.* 1981; 51:1300–1305.
5. Bigland-Ritchie B, Jones DA, Woods JJ. Excitation frequency and muscle fatigue: electrical responses during human voluntary and stimulated contractions. *Exp Neurol.* 1979; 64:414–427. [PubMed: 428516]
6. Bigland-Ritchie B, Thomas CK, Rice CL, Howarth JV, Woods JJ. Muscle temperature, contractile speed, and motoneuron firing rates during human voluntary contractions. *J Appl Physiol.* 1992; 73:2457–2461. [PubMed: 1490958]
7. Chen J, Yu N. The validity of stimulus-evoked EMG for studying muscle fatigue characteristics of paraplegic subjects during dynamic cycling movement. *IEEE Trans Rehabil Eng.* 1997; 5:170–178. [PubMed: 9184903]
8. Clamann HP, Robinson AJ. A comparison of electromyographic and mechanical fatigue properties in motor units of the cat hindlimb. *Brain Res.* 1985; 327:203–219. [PubMed: 3986499]
9. Cope TC, Webb CB, Yee AK, Botterman BR. Nonuniform fatigue characteristics of slow-twitch motor units activated at a fixed percentage of their maximum tetanic tension. *J Neurophysiol.* 1991; 66:1483–1492. [PubMed: 1765789]
10. Duchateau J, Hainaut K. Electrical and mechanical failures during sustained and intermittent contractions in humans. *J Appl Physiol.* 1987; 58:942–947. [PubMed: 3980395]
11. Edman, KAP. Myofibrillar fatigue versus failure of activation. In: Gandevia, SC.; Enoka, RM.; McComas, AJ.; Stuart, DG.; Thomas, CK., editors. *Fatigue: neural and muscular mechanisms.* New York: Plenum Press; 1995. p. 29-43.
12. Enoka RM, Rankin LL, Joyner MJ, Stuart DG. Fatigue-related changes in neuromuscular excitability of rat hindlimb muscles. *Muscle Nerve.* 1988; 11:1123–1132. [PubMed: 3226428]
13. Enoka RM, Trayanova N, Laouris Y, Bevin L, Reinking RM, Stuart DG. Fatigue-related changes in motor unit action potentials of adult cats. *Muscle Nerve.* 1992; 15:138–150. [PubMed: 1549136]
14. Fuglevand, AJ. The role of the sarcolemma action potential in fatigue. In: Gandevia, SC.; Enoka, RM.; McComas, AJ.; Stuart, DG.; Thomas, CK., editors. *Fatigue: neural and muscular mechanisms.* New York: Plenum Press; 1995. p. 101-108.
15. Fuglevand AJ, Zackowski KM, Huey KA, Enoka RM. Impairment of neuromuscular propagation during human fatiguing contractions at submaximal forces. *J Physiol.* 1993; 460:549–572. [PubMed: 8387589]
16. Garner S, Hicks A, McComas A. M-wave potentiation during muscle fatigue and recovery in man and rat. *J Physiol Lond.* 1986; 377:108P.
17. Hagg GM. Interpretation of EMG spectral alterations. *J Appl Physiol.* 1992; 73:1211–1217. [PubMed: 1447061]
18. Hicks A, McComas AJ. Increased sodium pump activity following repetitive stimulation of rat soleus muscles. *J Physiol.* 1989; 414:337–349. [PubMed: 2558169]
19. Jones DA, Bigland-Ritchie B, Edwards RH. Excitation frequency and muscle fatigue mechanical responses during voluntary and stimulated contractions. *Exp Neurol.* 1979; 64:401–413. [PubMed: 428515]

20. Kawakami Y, Nakai K, Maganaris CN, Oda T, Chino K, Fukunaga T. In vivo estimation of human achilles tendon moment arm based on ultrasonography. Twenty-fifth annual meeting of the American Society of Biomechanics. 2001:41–42.
21. Lieber, RL. Skeletal muscle structure and function. Baltimore: Williams & Wilkins; 1992. The production of movement; p. 111–158.
22. Lindstrom L, Magnusson R, Petersen I. Muscular fatigue and action potential conduction velocity changes studied with frequency analysis of EMG signals. Electromyography. 1970; 10:341–356. [PubMed: 5521836]
23. Maganaris CN, Baltzopoulos V, Ball D, Sargeant A. In vivo specific tension of human skeletal muscle. J Appl Physiol. 2001:865–872. [PubMed: 11181594]
24. Marsolais EB, Edwards BG. Energy costs of walking and standing with functional neuromuscular stimulation and long leg braces. Arch Phys Med Rehabil. 1988; 69:243–249. [PubMed: 3258509]
25. Marsolais EB, Kobetic R. Functional electrical stimulation for walking in paraplegia. J Bone Joint Surg. 1987; 69:728–733. [PubMed: 3496340]
26. Miller RG, Giannini D, Milner-Brown HS, Layzer RB, Koretsky AP, Hooper D, Weiner MW. Effects of fatiguing exercise on high-energy phosphates, force, and EMG: evidence for three phases of recovery. Muscle Nerve. 1987; 10:810–821. [PubMed: 3683452]
27. Milner-Brown HS, Miller RG. Muscle membrane excitation and impulse propagation velocity are reduced during muscle fatigue. Muscle Nerve. 1986; 9:367–374. [PubMed: 3012329]
28. Narici MV, Binzoni T, Hiltbrand E, Fasel J, Terrier F, Cerretelli P. In vivo human gastrocnemius architecture with changing joint angle at rest and during graded isometric contraction. J Physiol. 1996; 496:287–297. [PubMed: 8910216]
29. Radicheva N, Gerilovsky L, Gydikov A. Changes in the muscle fibre extracellular action potentials in long-lasting (fatiguing) activity. Eur J Appl Physiol. 1986; 55:545–552.
30. Shields RK. Fatigability, relaxation properties, and electromyographic responses of the human paralyzed soleus muscle. J Neurophysiol. 1995; 73:2195–2206. [PubMed: 7666132]
31. Shields RK, Chang Y-J. The effects of fatigue on the torque–frequency curve of the human paralyzed soleus muscle. J Electromyogr Kinesiol. 1997; 7:3–13. [PubMed: 20719687]
32. Shields RK, Chang Y-J, Ross M. Neuromuscular propagation after fatiguing contractions of the paralyzed soleus muscle in humans. Muscle Nerve. 1998; 21:776–787. [PubMed: 9585332]
33. Shields RK, Law LF, Reiling B, Sass K, Wilwert J. Effects of electrically induced fatigue on the twitch and tetanus of paralyzed soleus muscle in humans. J Appl Physiol. 1997; 82:1499–1507. [PubMed: 9134899]
34. Shields RK, Chang YJ, Littmann A. The effects of potentiation and fatigue on the electromyographic properties of paralyzed muscle. Soc Neurosci Abstr. 1997; 83:177.
35. Sieck, GC.; Prakash, YS. Fatigue at the neuromuscular junction. In: Gandevia, SC.; Enoka, RM.; McComas, AJ.; Stuart, DG.; Thomas, CK., editors. Fatigue: neural and muscular mechanisms. New York: Plenum Press; 1995. p. 83–100.
36. Smith DO. Mechanisms of action potential propagation failure at sites of axon branching in the crayfish. J Physiol (Lond). 1980; 301:243–259. [PubMed: 7411430]
37. Stalberg E. Propagation velocity in human muscle fibers in situ. Acta Physiol Scand. 1966; 70(Suppl 287)
38. Stulen FB, De Luca CJ. Frequency parameters of the myoelectric signal as a measure of muscle conduction velocity. IEEE Trans Biomed Eng. 1981; 28:515–523. [PubMed: 7275132]
39. Zijdwind I, Kernell D. Fatigue associated EMG behavior of the first dorsal interosseous and adductor pollicis muscles in different groups of subjects. Muscle Nerve. 17:1044–1054. [PubMed: 8065391]

FIGURE 1.

An example of the train of M-waves (with overlay) obtained from the chronically paralyzed soleus muscle before fatigue (A), 10 s after fatigue (B), and 10 min after fatigue (C).

FIGURE 2.

The average M-wave amplitude (A), duration (B), area (C), and median frequency (D) ratios for the chronically paralyzed group before fatigue and at 10 s, 5 min, 10 min, 15 min, and 20 min after fatigue. All values were ratios obtained by dividing all M-wave values by the initial M-wave of the pre-fatigue trials. Error bars are standard errors.

FIGURE 3.

The difference (%) between the average of the train and the initial M-wave of the train ($M\text{-wave train average} - \text{initial M-wave} / \text{initial M-wave} \times 100$) for amplitude and median frequency during fatigue and recovery for the chronically paralyzed group. The absolute change in torque (Nm) is also represented on the y-axis. Asterisks indicate amplitude and median frequency were significantly less at 1- and 30-s and 5-, 10-, 15-, and 20-min conditions ($P < 0.05$). Plus signs indicate amplitude and median frequency were significantly less than 1-s condition ($P < 0.05$). Error bars are standard errors.

FIGURE 4.

The difference (%) between the train average and the initial M-wave plotted against torque (Nm) during fatigue of the chronically paralyzed soleus muscle. The duration percent difference is the absolute value. Error bars indicate standard errors.

Table 1

Within-train M-wave variation in chronically paralyzed soleus muscle at various timepoints.

	Coefficient of variation		
	Amplitude	Duration	Median frequency
0	7.1 ± 3.9	7.5 ± 2.1	7.0 ± 3.6
30 s	8.1 ± 5.0	5.7 ± 2.8	5.7 ± 2.8
60 s	5.4 ± 2.6	3.8 ± 1.9	3.9 ± 2.0*
90 s	4.3 ± 2.1*	2.7 ± 1.7	3.0 ± 1.8*
120 s	5.1 ± 3.3*	2.6 ± 1.1	2.7 ± 1.6*
Recovery			
10 s	5.5 ± 3.1*	3.8 ± 2.5	3.1 ± 1.7*
5 min	5.4 ± 3.0*	5.1 ± 2.1	4.8 ± 2.9*
10 min	5.0 ± 3.2*	4.9 ± 1.9	5.0 ± 2.3*
15 min	4.7 ± 3.5*	4.8 ± 2.6	4.9 ± 1.9*
20 min	4.8 ± 3.7*	4.9 ± 2.4	4.8 ± 2.0*

Mean ± SD data are shown.

* Significantly different ($P < 0.01$) from before fatigue.