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8 Unique applications of mechanomyography

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Abstract

Recent investigations have examined some unique applications of surface mechanomyography (MMG). For example, several studies have shown different MMG amplitude and frequency responses for muscles that differed in fiber type composition. Other investigations have used MMG for biofeedback to enhance muscle relaxation. Typically, surface electromyographic (EMG) signals are used for biofeedback, but there are several disadvantages to this method, including changes in skin impedance and a high sensitivity to electrode placement. Previous studies have also used MMG to examine the effects

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of static and dynamic stretching on the neural and mechanical aspects of muscle function, as well as the effects of hypothermia. Generally speaking, these studies have shown that MMG is a very sensitive indicator of changes in muscle function, and, in many cases, provides information that is unique from EMG and force.

Introduction

There are several recent studies that have examined some unique applications of mechanomyography (MMG). For example, Shima et al. (2006) investigated the effects of postactivation potentiation on the MMG signal. The experimental protocol involved a supramaximal electrical stimulation of the tibial nerve to measure peak twitch torque, the peak acceleration of twitch torque development, as well as peak-to-peak electromyographic (EMG) and MMG amplitude values for the gastrocnemius muscle. This initial twitch was followed by a 10-second isometric maximum voluntary contraction (MVC) of the plantar flexors and supramaximal twitches that were elicited 2, 15, 30, 60, and 180 seconds after the isometric MVC. The results showed that the peak twitch torque, peak acceleration of twitch torque development, and peak-to-peak MMG amplitude values were greater 2, 15, and 30 seconds after the MVC when compared to the corresponding values recorded before the MVC. In addition, the peak twitch torque remained elevated 60 and 180 seconds after the MVC, and the peak acceleration of twitch torque development remained elevated 60 seconds after the MVC. Furthermore, the peak-to-peak MMG amplitude was positively correlated with both the peak twitch torque ($r = 0.79$) and the peak acceleration of twitch torque development ($r = 0.78$).

The 10-second isometric MVC had no effect, however, on the peak-to-peak EMG amplitude values. Thus, it was suggested that the peak-to-peak amplitude of the MMG signal from an electrically-stimulated isometric twitch may provide useful information regarding mechanical changes in the muscle that are associated with increased force production after an isometric MVC. It has been hypothesized that this phenomenon is related to facilitation of excitation-contraction coupling mechanisms through the Ca^{2+} - dependent process of phosphorylation of the regulatory light chains of myosin. Thus, MMG may be a useful technique for investigating the mechanisms that underlie post-activation potentiation (Shima et al. 2006). Yoshitake et al. (2005) investigated the relationship between changes in MMG amplitude and alterations in fascicle length (measured with ultrasound) during electrical stimulation of the gastrocnemius muscle. Specifically, the posterior tibial nerve was stimulated at frequencies that increased linearly from 1 to 20 Hz over a 4 second time period. During the stimulation, the surface MMG signal was measured from the gastrocnemius muscle, and ultrasound was used to detect changes in fascicle length. The results showed that with increases in stimulation rate, both MMG amplitude and the changes in fascicle length decreased. These decreases followed similar patterns, and the changes in MMG amplitude were positively correlated (r = 0.94) with the changes in fascicle length. Thus, it was concluded that the origin of the MMG signal is the pressure wave resulting from architectural changes in contracting muscle, and these pressure changes cause mechanical movement of the skin surface (Yoshitake et al. 2005).

Dahmane et al. (2006) investigated the possibility of using MMG as an indicator of muscle fiber type composition. Specifically, MMG was used to measure the contraction time of the biceps femoris muscle during a supramaximal electrically-stimulated twitch. The subjects that participated in the study included 15 sedentary men and 15 high level sprinters. In addition, muscle biopsies were taken from 15 sedentary men that had died due to various causes (e.g., suicide, traffic accident, etc.). The results showed that the contraction time was less for the sprinters than the sedentary men, and was negatively correlated (r = -0.72) with running speed during a flying 20 meter running trial. In addition, the muscle biopsies showed that the biceps femoris consisted of approximately 49% Type I, or slow-twitch fibers. Thus, it was suggested that the biceps femoris may have a large capacity for changing its fiber type composition with training (i.e., it could be a very "plastic" muscle), and that these changes could potentially be tracked with MMG (Dahmane et al. 2006). Beck et al. (2007) also examined the potential for MMG to be used in estimating muscle fiber type composition. The subjects included 5 resistance-trained and 5 aerobically-trained men, and all subjects were required to perform a 30-second sustained isometric muscle action of the leg extensors at 50% MVC as the surface MMG signal was detected from the vastus lateralis. In addition, immediately after the sustained muscle action, a biopsy was taken from the vastus lateralis and analyzed for myosin heavy chain isoform composition. The results from the myosin heavy chain analyses showed that the vastus lateralis of the resistance-trained subjects consisted primarily of fast-twitch muscle fibers, while that of the aerobically-trained subjects contained mainly slow-twitch fibers. In addition, the mean MMG amplitude and MPF values for the resistance-trained subjects were greater than those for the aerobically-trained subjects at all time points during the sustained muscle action. Thus, it was suggested that the MMG amplitude and MPF responses for the resistance-trained subjects could have been due to a greater percentage of fast-twitch muscle fibers in the vastus lateralis when compared to that of the aerobically-trained subjects (Beck et al. 2007). This study was followed up by a second investigation (Beck et al. 2008) that compared the patterns of responses for MMG amplitude versus isometric torque for the vastus lateralis among resistance-trained, aerobically-trained, and sedentary subjects. Like the previous investigation, the resistance-trained subjects had mostly fast-twitch muscle fibers in their vastus lateralis, while the aerobically-trained subjects had primarily slow-twitch fibers. The vastus lateralis of the sedentary subjects had roughly equal proportions of fast- and slow-twitch fibers. Despite these differences in fiber type composition, however, there were no consistent patterns of responses for MMG amplitude versus isometric torque. Some subjects showed increases in MMG amplitude throughout the entire force production range, while others demonstrated a plateau at high force levels. Thus, it was concluded that differences in fiber type composition and training status did not explain the unique torque-related patterns of responses for MMG amplitude (Beck et al. 2008).

Although there were no consistent fiber type-related differences in the patterns of responses for MMG amplitude, that does not necessarily mean that MMG cannot be used to estimate fiber type composition. Beck et al. (in press) used multiple regression in an attempt to estimate the percentage of fast-twitch muscle fibers in the vastus lateralis based on the MMG median frequency and maximal isometric leg extension strength values. The results showed that neither isometric leg extension strength nor MMG median frequency alone were significantly correlated with the fast-twitch fiber type content. The combination of these two variables, however, explained a significant proportion (59.8%) of the variance in fast-twitch fiber type content, with a multiple correlation of $R = 0.773$ and a standard error of the estimate of 15.4%. Thus, it was concluded that a simple, time-efficient, and noninvasive test that simultaneously measures isometric strength and MMG median frequency could be useful for estimating the fast-twitch fiber type content in well trained men (Beck et al. in press). Mealing et al. (1996) also conducted an interesting study that examined the MMG frequency responses from muscles that had a different fiber type composition. Surface MMG signals were detected from the soleus and biceps brachii during an isometric muscle action at 50% MVC. The results showed that the MMG signal for the soleus muscle was dominated by power between 5 and 10 Hz, whereas the MMG power spectrum for the biceps brachii had a much larger bandwidth, with power not only between 5 and 10 Hz, but also between 10 and 25 Hz (Figure 1).

Thus, it was concluded that the discrepancies between the two muscles for MMG power spectra were likely due to differences in fiber type content. It was also suggested that a possible application of MMG is in sport, where individual differences in fiber type composition could be important for determining performance (Mealing et al. 1996). Marchetti et al. (1992) compared

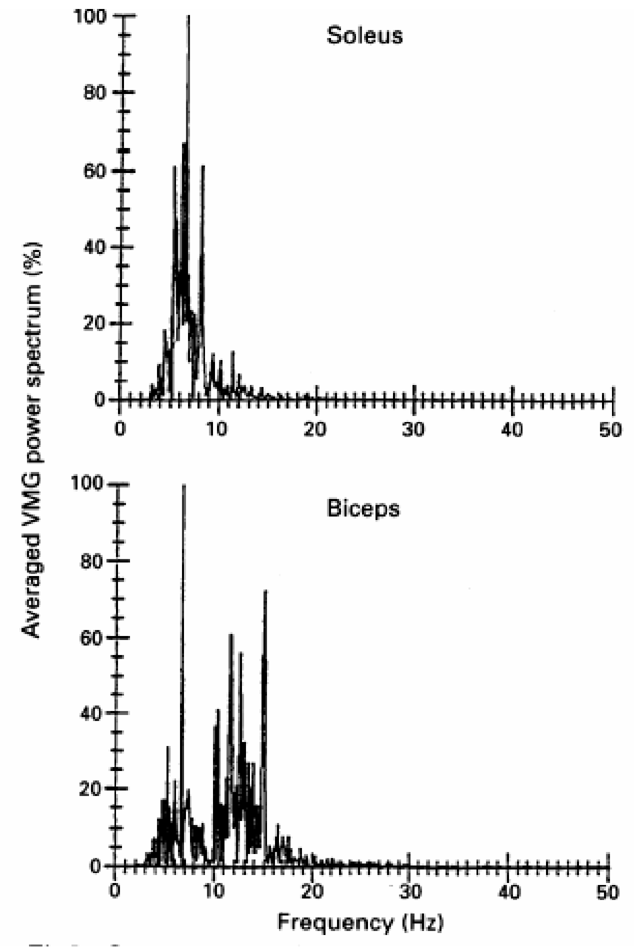

Figure 1. Average (i.e., across subjects) mechanomyographic (MMG) power spectra for the soleus (top graph) and biceps brachii (bottom graph) muscles. The signals were detected during submaximal isometric muscle actions of the plantar flexors or forearm flexors at 50% of the isometric maximum voluntary contraction (MVC). Notice that the spectrum for the soleus was dominated by power below 10 Hz, while the spectrum for the biceps brachii showed most of its power above 10 Hz. *Reprinted with permission from Mealing et al. (1996).

the MMG frequency responses of the vastus lateralis and soleus during supramaximal electrically-stimulated isometric twitches. The results showed that the MMG median frequency values from the vastus lateralis were significantly greater than those for the soleus in all subjects that participated in the study. In addition, the time required to reach peak MMG amplitude was significantly less for the vastus lateralis than the soleus. Thus, it was concluded that the different MMG responses for the two muscles were most likely due to differences in fiber type composition. In addition, the authors recommended that future studies should compare the MMG amplitude and frequency responses from electrical stimulation in both endurance and power athletes (Marchetti et al. 1992).

Orizio et al. (1999) also examined the MMG responses for the tibialis anterior during electrical stimulation. The experimental protocol included six single supramaximal twitches, followed by a 5-second ramp in which the stimulation frequency was increased from 1 to 50 Hz. After the 1 to 50 Hz ramp, a fatiguing protocol was induced that consisted of continuous stimulation at 35 Hz for 40 seconds. The 6 single twitches and 1 to 50 Hz ramp were then repeated immediately after the fatiguing stimulation, as well as 0.5, 1, 2, 3, 4, and 6 minutes after the fatigue protocol. The results showed that the fatigue protocol caused reductions in peak twitch force, the peak rate of force development, and the peak of the acceleration of force development, while both the contraction time and half-relaxation time increased. In addition, the peak-to-peak MMG amplitude decreased with fatigue. With the exception of half-relaxation time, all of the force and MMG parameters were restored to normal values within two minutes of recovery. Furthermore, the peak-to-peak MMG amplitude was highly correlated with the peak of the acceleration of force development in fresh muscle and during recovery.

It was concluded that MMG can be used to track changes in the mechanics of individual muscles with fatigue. This is an important advantage when compared to the force signal because there are very few joints that are crossed by just one muscle. Thus, examination of the force signal during fatigue provides information regarding fatigue-induced changes in the properties of several muscles, whereas the MMG signal is muscle-specific, and, therefore, can be used to examine fatigue of individual muscles (Orizio et al. 1999). Jaskólska et al. (2004) conducted an interesting study that examined the effect of skinfold thickness on MMG frequency. Specifically, surface MMG signals were detected simultaneously from the biceps brachii, triceps brachii, and brachioradialis during maximal isometric muscle actions of the forearm flexors and extensors, and skinfold thicknesses were measured in the same locations that the MMG sensors were placed for each muscle. The results showed that skinfold thickness had a large effect on the median frequency of the MMG signal and a smaller influence on the peak frequency. Specifically, greater skinfold thicknesses were associated with lower MMG median frequency values. Thus, it was concluded that studies that report absolute MMG frequency values should consider using skinfold thicknesses as a covariate (Jaskólska et al. 2004). Mamaghani et al. (2002) investigated the influence of changes in joint angle on the MMG amplitude and MPF responses of the upper trapezius, anterior deltoid, biceps brachii, and brachioradialis during sustained submaximal isometric muscle actions at 20%, 40%, and 60% MVC. The results showed that MMG amplitude followed force production, except at 20% MVC. In addition, MMG MPF remained relatively constant for all muscles during the sustained muscle actions. Thus, it was suggested that MMG amplitude may be useful for tracking changes in force production during fatiguing isometric muscle actions (Mamaghani et al. 2002).

Nonaka et al. (2006) examined gender differences in the MMG amplitude and MPF versus isometric torque relationships for the biceps brachii. The subjects were required to perform an isometric ramp muscle action of the forearm flexors from 5-80% MVC at a rate of 10% MVC/second while surface MMG signals were detected from the biceps brachii. In addition, the cross-sectional area of the biceps brachii was estimated with ultrasound imaging. The results showed that the mean isometric forearm flexion strength and cross-sectional area values for the men were 56.3% and 56.1% greater than those for the women, respectively. Furthermore, the mean MMG amplitude values for the men were greater than those for the women at all force levels. Even when the MMG amplitude values were expressed relative to the isometric MVC, the men showed greater values than the women at all force levels, and the mean difference became larger at higher force levels. Finally, the mean MMG MPF values were significantly greater for the men when compared to those for the women. Although muscle biopsies were not taken from the biceps brachii, the authors hypothesized that the differences between the men and women for the patterns of responses for MMG amplitude versus isometric torque were probably due to discrepancies in muscle fiber type composition. Specifically, it was suggested that since women generally have a greater percentage of slow-twitch fibers than men, the different torque-related patterns of responses for men and women were likely due to fusion of twitches from slow-twitch motor units at a lower force level for women versus men (Nonaka et al. 2006). Yamamoto and Takano (1994) investigated the possibility of using MMG to assess muscle tonus during functional electrical stimulation. The authors cited the fact that in many cases, it is difficult to directly measure muscle strength during electrical stimulation. Thus, they recorded MMG signals from the tibialis anterior during electrical stimulation, and found that the amplitude of the MMG signal was linearly related to dorsiflexion force production. Therefore, it was concluded that the MMG signal may be a useful method for estimating force production from individual muscles during electrical stimulation (Yamamoto and Takano 1994). Harba and Chee (1997) conducted an interesting investigation that examined the propagation velocities of MMG and EMG signals. The subjects were required to perform a sustained "light" isometric muscle action of the forearm flexors, and two separate surface MMG and EMG signals were detected simultaneously from the biceps brachii muscle. The MMG and EMG sensors were placed in line with the long axis of the muscle, and were separated by the same distance. Cross-correlation was used to determine the time delay between the signals (both MMG and EMG) detected at the two different sensor locations, which allowed for calculation of propagation velocity. The results showed that the propagation velocities of the MMG and EMG signals were nearly the same. These findings have important practical implications in terms of identifying the mechanisms that generate the MMG signal. Specifically, they indicated that the mechanical twitches that underlie the MMG signal may propagate along the muscle fibers in a wave-like fashion, rather than occurring simultaneously at all points along the muscle fiber (Harba and Chee 1997).

Keidel and Keidel (1989) also performed an important study that simultaneously investigated the MMG and indwelling EMG responses from the biceps brachii, masseter, wrist extensor, and tibialis anterior muscles during relaxation, as well as during graded isometric muscle actions. The results showed that the MMG signal provided important information regarding motor control strategies, and could potentially reflect the activity of central reflex loops. It was also suggested, however, that more studies needed to be performed to identify the mechanisms that cause the various MMG responses in different experimental protocols (Keidel and Keidel 1989). Petitjean and Maton (1995) performed a similar study that applied the spike-triggered averaging technique to MMG signals. The subjects were required to perform submaximal isometric muscle actions of the forearm extensors as surface MMG and indwelling EMG signals were detected simultaneously from the anconeus muscle. The intramuscular EMG signal was used for the spike-triggered averaging, and provided audio feedback to the subjects to help them achieve a steady muscle activation level. The results showed that by using the spike-triggered averaging technique, it was possible to isolate the MMG signal from individual motor units. Furthermore, the MMG signal from each motor unit had a characteristic shape, and, therefore, could be used to identify the contributions of individual motor units to force production. These findings were important because they indicated that the mechanical activities of individual motor units made predictable contributions to the MMG signal and could be extracted. It was recommended, however, that future studies were needed to determine whether or not the spike-triggered averaging method could be used at high, as well as at low force levels (Petitjean and Maton 1995). The spike-triggered averaging technique was also used by Jørgensen and Lammert (1976) to investigate the MMG responses from individual motor units in the rectus femoris during submaximal isometric muscle actions of the leg extensors at 20%, 30%, and 40% MVC. The results showed that the size of the action potential detected with the indwelling EMG electrodes was correlated with the amplitude of the mechanical contribution of the motor unit to the MMG signal. In addition, the mechanical contributions of individual motor units to the MMG signal were very reliable. Thus, it was concluded that the MMG signal is generated by the mechanical activities of individual motor units, and these activities are different, based on the motor unit's size and location within the muscle (Jørgensen and Lammert 1976). The study by Yoshitake et al. (2002) was also very important for determining the mechanisms that generate the MMG signal. Specifically, the authors electrically stimulated the medial gastrocnemius at rates of 5, 10, 15, and 20 Hz, and intramuscular EMG and surface MMG signals were detected from the muscle. The indwelling EMG electrodes allowed eight separate motor units to be identified. The results showed that when each motor unit was stimulated, there was a significant positive correlation between the duration of the MMG signal and that of the force twitch. In addition, both MMG amplitude and the magnitude of the force fluctuations decreased with increasing stimulation rates. Finally, MMG amplitude was negatively correlated with several twitch parameters, including twitch duration and half-relaxation time. Thus, it was concluded that the characteristics of the surface MMG signal were dependent on those of the active motor units during the contraction (Yoshitake et al. 2002).

Celichowski et al. (1998) examined the relationship between force fluctuations and MMG amplitude for the forefinger flexor muscles. All subjects were required to perform triangular or trapeziform contractions at different force levels, and MMG signals were detected from the anteromedial part of the forearm. The results showed that MMG amplitude was dependent on the force that was being produced. Specifically, both the rate and amplitude of changes in force were correlated with those for MMG amplitude. Thus, it was suggested that MMG could be used to examine the force production of individual muscles (Celichowski et al. 1998). Rodriquez et al. (1993) examined the MMG and EMG amplitude and median frequency responses from the rectus femoris during sustained isometric muscle actions of the leg extensors at 20%, 40%, and 80% MVC. Each muscle action was performed until exhaustion, and the results showed that for each force level, EMG amplitude increased and EMG median frequency decreased. The results for MMG amplitude, however, demonstrated increases over time only during the 20% and 40% MVC muscle actions. Thus, it was concluded that MMG and EMG amplitude behave similarly, but only at low force levels. Furthermore, in many situations, MMG provides information that is unique from that of EMG (Rodriquez et al. 1993).

Mitchell et al. (2008) examined the effects of diathermy on muscle temperature, EMG amplitude, and MMG amplitude. This was an important study from a practical standpoint because exercise often results in increases in muscle temperature. The subjects were randomly assigned to one of three groups: (a) a diathermy group that was subjected to a 20-minute treatment, (b) a sham-diathermy group that thought they were receiving the treatment, but actually were not, and (c) a control group that did not receive the treatment. Prior to, and immediately following the diathermy treatment, all subjects were required to perform a 6-second isometric ramp muscle action of the leg extensors from 10-90% MVC, and surface MMG signals were detected from the vastus lateralis. The results showed that the diathermy treatment provided a significant increase (approximately 2° Celsius) in muscle temperature, MMG amplitude, and MMG MPF at all torque levels, but had no effect on leg extension force, EMG amplitude, or EMG MPF (Figure 2).

Thus, it was concluded that increases in muscle temperature may decrease musculotendonous stiffness, thereby affecting both MMG amplitude and MMG MPF. The changes in muscle temperature did not, however, affect leg extension strength or the motor control strategies that influence isometric force production (Mitchell et al. 2008). Kimura et al. (2003) examined the effects of decreases in muscle temperature on the MMG amplitude responses for the soleus and medial gastrocnemius muscles. Surface MMG signals were detected simultaneously from both muscles during supramaximal electrically-stimulated

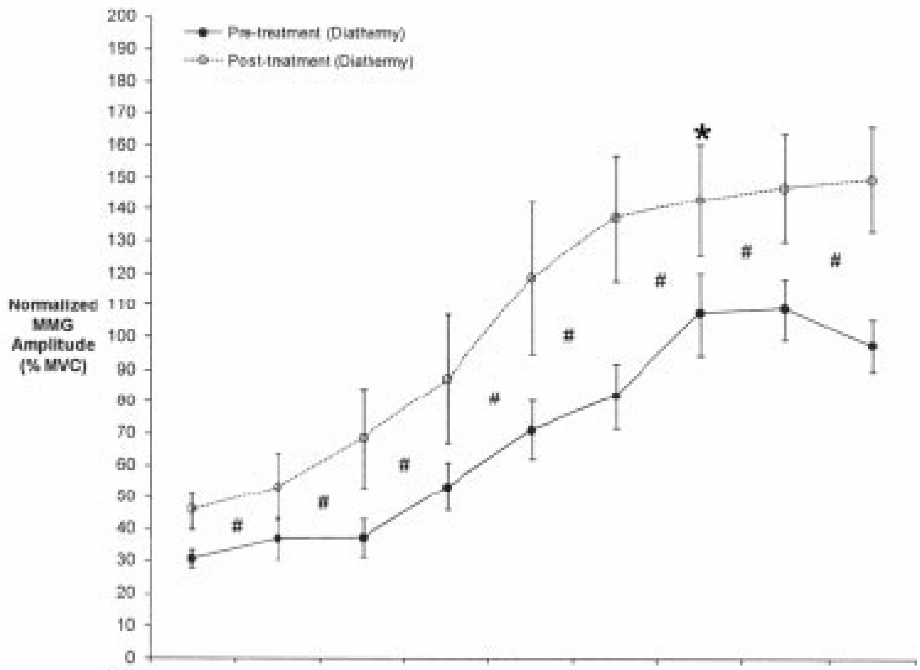

Figure 2. Changes in mechanomyographic (MMG) amplitude for the vastus lateralis with increases in isometric leg extension torque from 10% to 90% of the isometric maximum voluntary contraction (MVC). The closed symbols reflect the values before diathermy treatment, and the open symbols show the data after diathermy treatment. Notice that the diathermy treatment caused an increase in MMG amplitude at nearly all force levels. *Reprinted with permission from Mitchell et al. (2008).

twitches both before, and immediately after manipulating muscle temperature with ice packs. The control temperature was 34° Celsius, and the administration of ice packs allowed data to be collected at temperatures of 25°, 20°, and 15° Celsius. The results showed that the decreases in muscle temperature caused reductions in peak twitch force and the maximal rate of force development, increased contraction time, half-relaxation time, and the maximal rate of force relaxation. The decrease in muscle temperature resulted in increased force production during the sustained 10 Hz stimulation, but the magnitude of the force fluctuations decreased, as did MMG amplitude for both the medial gastrocnemius and soleus muscles (Figure 3).

Thus, it was concluded that the significant decrease in muscle contractile properties caused by the reduced muscle temperature was reflected in the MMG responses to electrical stimulation. Therefore, MMG could be useful for examining muscle contractile properties in a variety of physiological conditions (Kimura et al. 2003).

Evetovich et al. (2007) investigated the use of MMG as a biofeedback method for improving relaxation and delaying fatigue. The subjects were required to perform as many repetitions as possible with 85% of their one-repetition

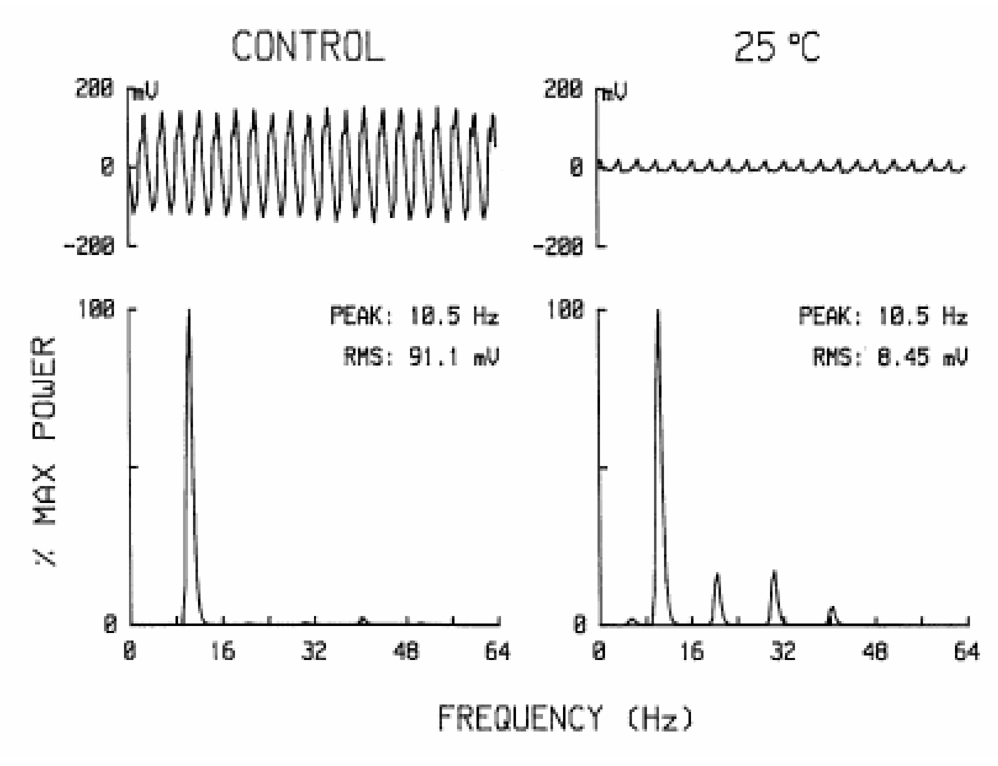

Figure 3. Mechanomyographic (MMG) signals for the soleus muscle during electrically-stimulated contractions at 10 Hz when the muscle was at room temperature (control) and cooled to 25° Celsius. Notice the dramatic decrease in MMG amplitude when the muscle was cooled. *Reprinted with permission from Kimura et al. (2003).

maximum (1-RM) during the bilateral forearm flexion exercise. During all muscle actions, surface MMG and EMG signals were detected from the biceps brachii, but only the MMG signal was used for biofeedback. In addition, the subjects were required to use the MMG signal during separate visits in an attempt to achieve complete muscle relaxation. The results showed that the use of MMG biofeedback did not increase the number of repetitions that the subject could perform with 85% of their 1-RM, but it did help to achieve more complete relaxation of the biceps brachii muscle. Thus, it was concluded that MMG can enhance the development of muscle relaxation, but it is not useful for delaying fatigue (Evetovich et al. 2007).

Silva et al. (2003) have conducted some very interesting work in the area of prosthesis control. Specifically, the authors investigated the use of silicon-embedded accelerometers to detect MMG signals for the purpose of controlling an externally-powered prosthesis. An important limitation of many EMG-powered prostheses is that it is often difficult to keep EMG sensors placed reliably within a silicon-based socket. Thus, the signal-to-noise ratio of embedded accelerometers was compared with that of non-embedded accelerometers, and the authors found that the embedded accelerometers provided a much better signal quality. In addition, it was recommended that the softest or hardest silicon types should be used to embed the accelerometers, since they provided the least variability. It was also suggested, however, that additional work needs to be done before silicon-embedded accelerometers can be used on a widespread basis for prosthesis control (Silva et al. 2003). Silva et al. (2004) then performed a second investigation that examined the use of a pattern classification procedure in which MMG amplitude was used to control an externally powered prosthesis. Three silicon-embedded accelerometers were placed in separate locations on the end of the stump, and the MMG amplitude values from the accelerometers were used to classify the movements of hand extension and flexion. The results showed that for the two subjects that participated in the study, the classification accuracy values were 70.90% and 70.01%. Thus, it was concluded that MMG could be a useful method for controlling an externally-powered prosthesis, and future studies are needed to determine its feasibility for grading different contraction levels (Silva et al. 2004). Silva et al. (2005) continued their work in the area of prosthesis control by examining the possibility of using the conventional 2-site electromyography system, but with MMG sensors in place of the EMG sensors. The results showed accuracy rates of 88% and 71% for the two subjects that participated in the study. The authors also described several advantages of MMG over EMG for prosthesis control, including less sensitivity to sensor placement, robustness to changing skin impedance levels, and reduced sensor costs (Silva et al. 2005). Alves and Chau (2006) continued their work in prosthesis control by using a vision-based segmentation procedure to separate the MMG signals generated during different types of hand grasping sequences. The authors noted that a major challenge with continuous collection of MMG signals for prosthesis control is the subsequent separation of the MMG data stream into segments that reflect individual contractions. Thus, MMG data acquisition was used in conjunction with transverse plane video identification of the gripping activities. The results showed that this system could recognize two different grips with an average accuracy of 97.8% and seven different grips with an average accuracy of 73.0%. Thus, it was concluded that this procedure could be useful for the development of an MMG-based multi-function prosthetic hand (Alves and Chau 2006).

Alves and Chau (2008) also examined the stationarity of MMG signals during different types of gripping activities. Specifically, MMG signals were detected from the wrist extensors during various types of gripping activities, and the signals were tested for weak stationarity using the reverse arrangements test. The results indicated that 20% of the MMG signals recorded during the gripping activities were nonstationary. Thus, it was suggested that time-frequency techniques may be necessary in order to accurately control externally powered prostheses with MMG. In addition, for 47% of the nonstationary signals, the source of the nonstationarity was undetermined, but in 38% of the cases, the nonstationarity was due to a changing variance. The remaining 15% of nonstationary signals were due to time-varying mean, variance and mean, frequency, or frequency and variance values. Therefore, it was concluded that the distribution of the stationary test statistic could reflect the time course of muscle activity used to generate functional grasps (Alves and Chau 2008). Alves and Chau (2008) then conducted a second study that examined vision-based segmentation of MMG signals during gripping activities. Specifically, MMG signals were detected from the wrist extensors and flexors, and a video-based system was used along with MMG to segment the signals. The results showed that the automatic signal segmentation method was capable of tolerating extraneous hand movements, as well as differentiating among different types of grips and estimating their transition times. The accuracy of this technique was 97.8% for two grips and 73.0% for seven grips. In addition, the detection procedure identified contraction and termination times that were within 173 milliseconds of the times measured with manual segmentation. Thus, it was concluded that this technique could be very useful when using large collections of signals for the purpose of training MMG-based prostheses (Alves and Chau 2008).

Marusiak et al. (in press) recently investigated differences between the EMG and MMG signals from normal, healthy subjects versus those that suffered from Parkinson's disease. Specifically, the subjects were required to perform submaximal and maximal isometric muscle actions of the forearm flexors while MMG and EMG signals were detected simultaneously from the biceps brachii. The results showed that when compared to the normal subjects, the patients with Parkinson's disease showed higher MMG amplitude, and lower MMG median frequency values. In addition, there were no consistent differences between the normal subjects and patients with Parkinson's disease for EMG amplitude or EMG median frequency values. Thus, it was concluded that the MMG signal can be used to identify differences in muscle function between normal subjects and patients with Parkinson's disease, and, therefore, it could be a useful tool for assessing those with a neuromuscular disease (Marusiak et al. in press). Garcia et al. (2008) investigated the possibility of using MMG to determine arm dominance. The subjects were required to perform submaximal to maximal isometric muscle actions of both the right and left forearm flexors, and surface MMG signals were detected from the biceps brachii. The results showed that there were no significant differences between the dominant and nondominant arms for the MMG amplitude and MPF responses with increases in isometric force. Thus, it was concluded that time and frequency domain parameters from the MMG signal cannot be used to determine arm dominance (Garcia et al. 2008). Orizio et al. (2008) recently investigated the MMG responses for the tibialis anterior during electrically-stimulated isometric muscle actions of the dorsiflexors. The electrical stimulation protocol involved both a short (12.5 seconds) and a long duration procedure (approximately 1 hour). Both procedures used various stimulation frequencies, and MMG signals were detected from the tibialis anterior with a laser displacement sensor. The results showed that detecting MMG signals during electrically stimulated contractions could be used to develop a noninvasive technique for identifying the muscle-tendon-joint transfer function. In addition, there were many similarities between the transfer functions obtained from the torque signal and the laser-detected MMG signal. Thus, MMG could be used to study the properties of the muscle-tendon-joint unit when torque cannot be measured directly (Orizio et al. 2008). Toca-Herrera et al. (2008) also examined the MMG and EMG responses following one training session that involved maximal electrical stimulation. Specifically, the untrained subjects were required to perform a maximal unilateral isometric muscle action of the leg extensors, and surface EMG signals were detected from the rectus femoris and biceps femoris both before and after the electrical stimulation training. The training consisted of 100 Hz cycles with a 300μs pulse duration, with cycles of 10-seconds on and 10-seconds off for a total duration of 10 minutes. In addition, the surface MMG signal was detected from the rectus femoris during each maximal muscle action. The results showed that when compared to a control group that performed no training, the electrical stimulation resulted in significant increases in isometric leg extension strength, EMG amplitude for the rectus femoris, and a decrease in EMG amplitude for the biceps femoris. There were no changes, however, in MMG amplitude for the rectus femoris for the training or control groups. Thus, it was concluded that electrical stimulation training could be useful in rehabilitative settings during the first stages of injury when the affected limb cannot be used to perform work. It was also suggested, however, that future work needs to be done to determine if electrical stimulation training is beneficial for those that already perform resistance training on a regular basis (Toca-Herrera et al. 2008).

Kouzaki and Fukunaga (2008) performed an important study that investigated the MMG frequency responses for the soleus muscle during quiet standing. Specifically, the subjects were required to stand barefoot in the upright position with the eyes open or closed, and MMG and EMG signals were detected simultaneously from the soleus. In addition, a laser displacement sensor was used to detect changes in the location of the subject's center of mass. The results showed that there was significant coherency between the MMG signal and the displacement of the center of mass. Thus, it was concluded that the kinematic and physiological measures of postural control during quiet standing can be examined with the frequency content of the MMG signal (Kouzaki and Fukunaga 2008). Esposito et al. (in press) investigated the MMG and EMG responses from the finger flexor muscles in elite rock-climbers and control subjects. All subjects were required to perform submaximal to maximal isometric muscle actions of the finger flexors, and MMG and EMG signals were detected during each muscle action. The results showed that the rock climbers demonstrated significantly greater MVC values for the finger flexors than the controls. In addition, EMG amplitude increased with force for both groups, but it was significantly greater for the climbers when compared to the controls at 60%, 80%, and 100% MVC. Furthermore, MMG amplitude increased with force up to 80% MVC for the climbers and 60% MVC for the controls, beyond which, it decreased. The mean EMG MPF values increased from 20-80% MVC in both the climbers and controls, followed by a plateau from 80-100% MVC. Finally, MMG MPF increased from 20-100% MVC for both the climbers and controls, but it was significantly greater for the climbers at 60%, 80%, and 100% MVC. Thus, it was concluded that the combined analysis of MMG and EMG signals provided information regarding potential differences in motor control strategies between the two groups. In addition, the strenuous training performed by the rock climbers may have caused conversion of some slow-twitch fibers to fast-twitch fibers (Esposito et al. in press). Herda et al. (2008) recently examined the acute effects of static or dynamic stretching on isometric leg flexion strength, as well as MMG amplitude and EMG amplitude for the biceps femoris muscle. Specifically, unilateral isometric leg flexion strength was measured at knee joint angles of 41°, 61°, 81°, and 101° below full leg extension prior to, and immediately following static or dynamic stretching exercises. The results showed that static stretching caused decreases in strength at the 81° and 101° knee joint angles, but not at the 41° and 61° joint angles. In addition, the dynamic stretching had no effect on isometric leg flexion strength. The findings for EMG amplitude indicated that there was no change after the static stretching, and EMG amplitude actually increased after the dynamic stretching, but only for the 101° and 81° knee joint angles. In addition, the static stretching caused increases in MMG amplitude, but only at the 101° knee joint angle, and the dynamic stretching caused increases in MMG amplitude at all joint angles. Thus, it was suggested that the decreases in strength after static stretching may have been due to mechanical, rather than neural factors. In addition, dynamic stretching could be less detrimental to muscle strength than static stretching (Herda et al. 2008).

Jaskólski et al. (2007) recently examined the EMG and MMG responses of agonist and antagonist muscles after eccentric exercise. All subjects performed 25 submaximal eccentric muscle actions of the forearm extensors at 50% of their isometric MVC. Prior to, immediately following, and 24, 48, 72, and 120 hours after the eccentric exercise, the subjects were tested for isometric forearm flexion strength and EMG and MMG amplitude and median frequency of the biceps brachii and triceps brachii muscles. The results showed that the eccentric exercise caused a 34% decrease in isometric forearm flexion strength immediately after exercise, and strength levels did not return to their resting values within the 120 hour time period that was measured. In addition, EMG median frequency decreased for both the biceps brachii and triceps brachii after eccentric exercise, and the MMG amplitude values for both muscles were lower 24, 48, 72, and 120 hours after the eccentric exercise when compared to those immediately after the eccentric exercise. Thus, it was concluded that the similar electrical and mechanical changes in the agonist and antagonist muscles were reflective of a common drive that controlled the agonist and antagonist motor unit pools. In addition, the eccentric exercise-induced changes in EMG and MMG amplitude and frequency parameters may have been due to increased tremor and contractile impairments, such as a reduced rate of calcium release from the sarcoplasmic reticulum, changes in motor control of the agonist and antagonist muscles, and increased muscle stiffness (Jaskólski et al. 2007). Madeleine et al. (2006) investigated the possibility of using MMG for biofeedback during standardized computer work. Specifically, MMG and EMG sensors were placed on the upper portion of the trapezius muscle, and the subjects were required to perform work on a computer with either MMG or EMG as a source of biofeedback. The results showed that when MMG was used for biofeedback, muscle activity was significantly lower than when EMG was used. In addition, the MMG-based biofeedback decreased the rating of perceived exertion during the computer work, but it also reduced the total amount of work that could be performed, because the subjects were forced to concentrate on reducing muscle activity, rather than completing their work. Thus, it was concluded that MMG-based biofeedback may be a reliable alternative to the EMG-based methodology, and, therefore, it could be helpful for decreasing the risk of work-related musculoskeletal disorders (Madeleine et al. 2006). Reza et al. (2005) performed an interesting study that investigated the MMG and EMG responses during transcranial magnetic stimulation. Specifically, surface MMG and EMG sensors were placed over the biceps brachii muscle, and the subjects were required to perform submaximal to maximal isometric muscle actions of the forearm flexors at 5%, 10%, 20%, 30%, 40%, 60%, and 100% of the isometric MVC. The transcranial magnetic stimulation was then used to elicit motor evoked potentials at rest and during the submaximal to maximal muscle actions. The results showed that MMG amplitude increased with force up to 60% MVC, and then decreased. In addition, both the onset latency and length of the silent period decreased with an increase in the isometric force level. Thus, it was concluded that MMG could be very useful for examining the mechanical responses of the muscle during transcranial magnetic stimulation, particularly when surface EMG is not feasible, such as when recording signals in environments that are contaminated with electromagnetic noise (Reza et al. 2005). Drake et al. (2003) examined the effects of oral contraceptives on isometric leg extension strength and MMG and EMG amplitude for the rectus femoris. Specifically, submaximal (25%, 50%, and 75% MVC) and maximal isometric muscle actions of the leg extensors were performed either with or without the use of oral contraceptives. The results showed that the oral contraceptives had no effect on leg extension strength, MMG amplitude, or EMG amplitude for the rectus femoris. Thus, it was concluded that the use of oral contraceptives did not influence strength or muscle function during isometric muscle actions. It was also recommended, however, that future studies needed to be performed to determine if the use of oral contraceptives could affect strength in other muscle groups (Drake et al. 2003).

In summary, the results from the studies reviewed in this chapter showed the variety of applications for MMG research. It is important for future investigations to carefully and systematically examine the validity of these applications to ensure the accuracy and reliability of the information provided by the MMG signal. To this end, studies are still needed to determine which factors have the greatest influence on the MMG signal and the limitations that they impose.

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