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# The mechanisms underlying the surface mechanomyogram

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## Abstract

*Muscle sounds were first discovered in the 1660s, when Francesco Grimaldi observed a rumbling sound as he placed his thumbs over the opening to his ears and clenched his fists. In 1810, William Hyde Wollaston reported that the sounds from a contracting muscle were very much like those from a carriage that was being pulled on the cobblestone streets of London at a speed of 8 miles per hour. By calculating the number of cobblestones that the carriage wheels contacted every second, Wollaston was able to conclude that muscle sounds were dominated by frequencies around 23 Hz. Until*

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1948, however, research on muscle sounds was limited by the inability to accurately detect them and store the data for further analysis. Thus, the development of piezoelectric transducers and condenser microphones was important because it sparked a new interest in identifying the mechanisms that generate muscle sound and the information that it provides. To avoid confusion in the literature, the term “surface mechanomyogram” (MMG) was proposed to describe muscle sounds because it highlighted the fact that they are generated by the mechanical activities of motor units. In the late 1980s, carefully-controlled electrical stimulation studies with isolated muscle preparations showed that the MMG signal is generated by lateral muscle fiber oscillations, the amplitude of which were dependent on the length of the muscle and the intensity of the stimulation. Furthermore, the frequency content of the MMG signal was closely related to the muscle’s resonant frequency, and it was hypothesized that this property may make MMG useful for examining muscle stiffness. Additional research in the mid-1990s indicated that during most voluntary muscle actions, the activities from individual motor units are summated nonlinearly to form the MMG signal. Recent studies have also shown that the spike-triggered averaging technique allows the activities from individual motor units to be extracted from the MMG signal recorded during a voluntary muscle action. Furthermore, the amplitude and frequency contents of the signals produced by these motor units were influenced by their morphology. This is important because it provides a promising future for work in the area of examining muscle fiber type composition with MMG.

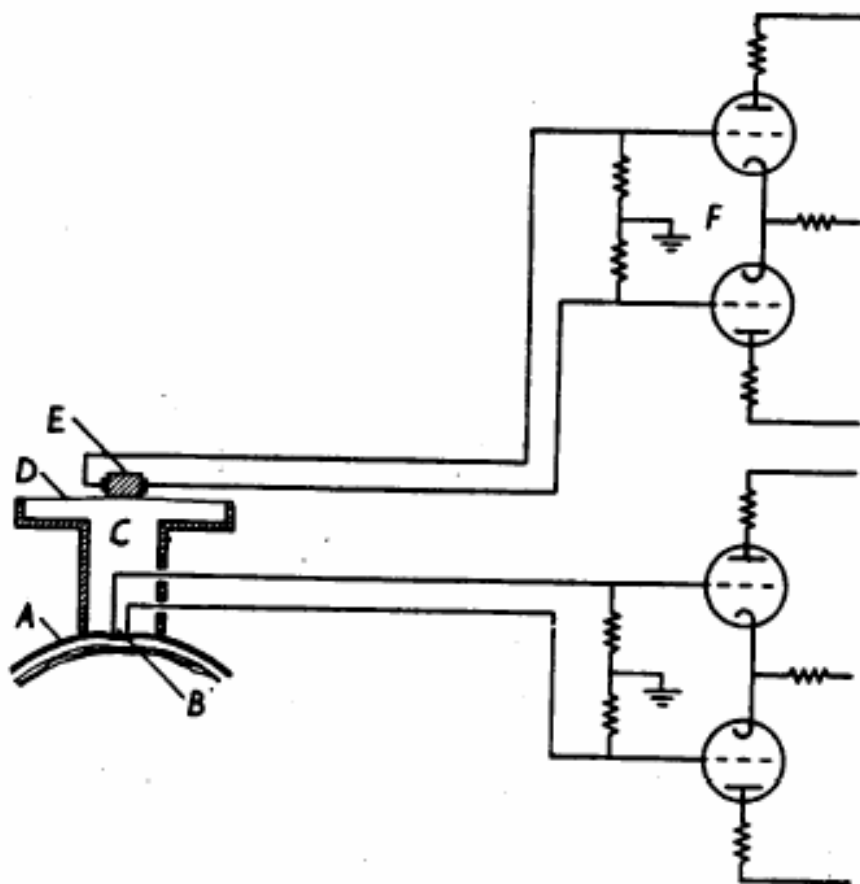
## **Introduction**

A discussion of the mechanisms underlying muscle sound [hereafter referred to as the surface mechanomyogram (MMG)] would not be complete without first providing a historical background of the field. The first acknowledgement of MMG was provided in 1663 by Francesco Grimaldi, who reported a rumbling noise when he placed his thumbs over the opening to his ears and clenched his fists. It was hypothesized that the rumbling was due to the “...continuous hurried motion of the spirits...” that caused tremors in the fingers, arms, and in the whole body (Orizio 1993). At the time, it was thought that muscle contraction was due to fluid movement that emanated from the brain and caused bulging of the muscles. Thus, even though Grimaldi’s explanation for MMG was incomplete, it is clear that he understood that muscle sound was related to muscle activity, and its properties were related to the properties of the contraction.

The first true experiment with MMG was conducted in 1810 by William Hyde Wollaston. Wollaston was a physicist, chemist, and physician, and was particularly interested in the frequencies of muscle sounds. Interestingly, he found that muscle sounds were similar in frequency to the sounds produced when a carriage was pulled on the cobblestone streets of London at a speed of 8 miles per hour. In 1810, the cobblestones were 6 inches in length. Thus, at a speed of 8 miles per hour (11.7 feet per second), the carriage wheels were hitting approximately 23 cobblestones per second. This allowed Wollaston to conclude that the dominant frequency of MMG was 23 Hz, which was a very accurate observation, considering the relatively primitive equipment that he had available. After 1810, studies on MMG were relatively scarce. In 1885, however, Herroun and Yeo reported that the muscle sounds produced during voluntary contractions were similar in frequency to those from electrically-stimulated contractions, but only for stimulation rates less than 30 Hz. Although it was not immediately known at the time, this was an important finding because it was the first study to determine that a muscle does not produce any sound (i.e., it is silent) when its twitches become completely fused. This finding would eventually be supported by the results from future studies that used isolated muscle preparations and very sensitive MMG sensors. An important step was also taken by Gordon and Holbourn (1948), who were the first to create a record of the MMG signal with a microphone (Figure 1).

Although relatively primitive by today's standards, this instrument allowed the authors to simultaneously record the surface electromyographic (EMG) and MMG signals from the orbicularis oculi muscle. As a result, they were able to conclude that the surface MMG signal is the mechanical counterpart of the motor unit electrical activity (Gordon and Holbourn 1948) as measured by EMG.

Despite the advancements made by Herroun and Yeo (1885) and Gordon and Holbourn (1948), very little research was conducted in the area of muscle sounds until 1980, when Oster and Jaffe sparked a new interest in MMG that was at least partially due to the development of electronic sensors capable of detecting the signal. One of the most important aspects of the study by Oster and Jaffe (1980) was that the authors concluded that the MMG signal was actually generated by contracting muscle. Specifically, it had been argued that muscle sound could be caused by rubbing of the sensor on the skin surface during contraction, or even blood flow through vessels near the muscle. However, these hypotheses were refuted by Oster and Jaffe (1980), who found that an MMG signal was still detectable when the muscle was contracted under water, and the sensor was 1 cm away from the skin surface. In addition, muscle sounds were generated by the biceps brachii muscle when an inflatable cuff



**Figure 1.** The microphone used by Gordon and Holbourn (1948) to detect mechanomyographic (MMG) signals. Wire or surface electrodes (indicated by (B) in the figure) are optional. (A) shows the skin surface, (C) the microphone chamber with its end closed by a diaphragm (D). Movements of the skin surface cause pressure changes inside the microphone, that, in turn, result in movement of the diaphragm and deform a Rochelle salt crystal (E). The Rochelle salt crystal has piezoelectric properties, which allows it to create potentials that are transmitted through a screened cable to the amplifier (F). \*Reprinted with permission from Gordon and Holbourn (1948).

was used to restrict blood flow through the arteries of the arm, thereby disproving the blood flow hypothesis. Perhaps the most important aspect of the Oster and Jaffe (1980) study, however, was that the authors used autocorrelation to determine that MMG signals were periodic with a dominant frequency of approximately 25 Hz. In addition, the amplitudes of the MMG signals increased with the intensity of the contractions, but the MMG signals were uncorrelated both in frequency and in phase with the corresponding surface EMG signal. The overall importance of the Oster and Jaffe (1980) investigation was reflected in the number of MMG studies conducted shortly

thereafter. Now that researchers knew the MMG signal was generated by muscle activity, they could focus on the mechanisms underlying it.

At this point, it is important to acknowledge the fact that various terms have been used to describe MMG, including soundmyography, phonomyography, acoustic-myography, accelerometryography, and vibromyography. Although the use of these terms was heavily influenced by the type of sensor used to detect the signal, it created confusion in the literature with regard to what was actually being measured. Thus, in 1995, the term “surface mechanomyogram” was suggested at a CIBA Foundation Symposium to distinguish the MMG signal from other mechanical signals that are unrelated to muscle activity (Orizio et al. 2003).

After the Oster and Jaffe (1980) study, three very important investigations into the mechanisms that generate the MMG signal were conducted in the laboratory of Dr. Daniel T. Barry (Barry 1987; Barry and Cole 1988, 1990). In the first study (Barry 1987), isolated frog gastrocnemius muscle was placed in a water bath and MMG signals were recorded during electrically-stimulated isometric twitches with two hydrophones that were placed on opposite sides of the muscle and perpendicular to its long axis. When the muscle was stimulated to contract with a single twitch, the resulting MMG signals from the two hydrophones were very similar in shape, but reversed in phase by 180°. In addition, each isometric twitch demonstrated an MMG signal with oscillations that had maximum amplitude at the beginning of the twitch, followed by oscillations that decayed in amplitude over time. The number of oscillations that were produced with each twitch was also influenced by muscle length, with longer muscle lengths resulting in more oscillations. Muscle length also affected the peak-to-peak amplitude of the MMG signal, since the maximum amplitude occurred at a muscle length that was approximately 90% of the optimal length for force production. Thus, it was suggested that muscle sound is likely generated by lateral muscle fiber oscillations, since the MMG signals from the two hydrophones were reversed in phase, thereby indicating side-to-side movement of the muscle, as opposed to bulging in all directions. In addition, changes in muscle length can affect MMG amplitude, independent of the muscle activation level. The conclusions of Barry (1987) were supported by the results from a second study (Barry and Cole 1988) that used high speed cinematography to track the movement of isolated frog gastrocnemius muscle during electrically stimulated isometric twitches. Specifically, the high speed images showed lateral displacement of the muscle during each twitch that was consistent with the oscillating pressure waves detected by the hydrophones in the previous study. In addition, the frequency content of the MMG signal was closely related to the muscle’s resonant frequency, which suggested that MMG could potentially be used as a noninvasive measure of muscle stiffness (Barry

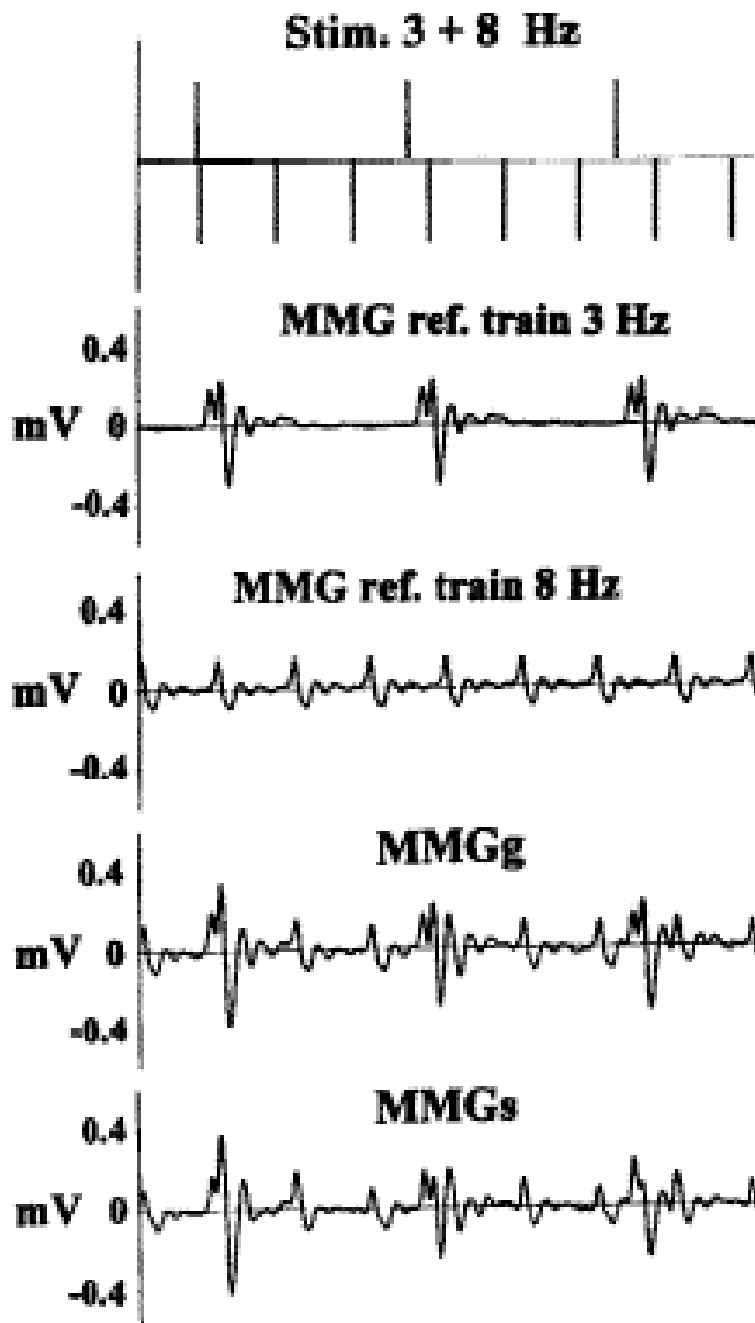
and Cole 1988). This hypothesis led to a third study (Barry and Cole 1990) that used time-frequency signal processing techniques to examine changes in MMG frequency during an electrically-stimulated isometric tetanic contraction. Specifically, isolated frog gastrocnemius muscle was electrically stimulated at a frequency of 150 Hz to induce tetanus, while an MMG signal was detected with a hydrophone. The results showed that the resonant frequency of the muscle was in the same range as the peak instantaneous MMG frequency. Thus, it was concluded (Barry and Cole 1990) that the frequency content of the MMG signal was heavily influenced by the resonant frequency of the muscle, which, in turn, was affected by muscle stiffness.

At roughly the same time that Dr. Barry's laboratory was conducting research to examine the mechanisms underlying the MMG signal, similar research was being done in the Division of Applied Sciences at Harvard University. Specifically, Frangioni et al. (1987) were interested not only in the mechanisms underlying muscle vibration, but also in the directions that the vibrations were transmitted. The authors (Frangioni et al. 1987) electrically stimulated isolated frog gastrocnemius muscle in a saline bath and detected MMG signals with a hydrophone that was placed perpendicular to the long axis of the muscle. Like Barry (1987), Frangioni et al. (1987) found that during an isometric twitch, the MMG signal was generated by lateral movement of the whole muscle, and its maximum amplitude occurred when the muscle length was slightly less than that required for maximum force production. The authors (Frangioni et al. 1987) also hypothesized, however, that when stimulated to contract, the muscle vibrates like an axe handle, or any rigid object having a long elliptical shape. This was an important hypothesis because it was the first time that a model had been proposed to describe muscle sound. The finding that MMG amplitude was greatest when the muscle length was slightly less than that required for maximum force production was supported by Dobrunz et al. (1990) for isolated frog gastrocnemius muscle. In addition, it was hypothesized that the frequency content of the MMG signal may be closely related to muscle stiffness. Around the same time as the studies by Frangioni et al. (1987) and Barry (1987), Brozovich and Pollack (1983) performed an interesting investigation to determine if muscle sounds occurred in a discrete versus continuous manner. Specifically, isolated frog sartorius muscle was electrically stimulated to shorten under lightly loaded conditions while an MMG signal was detected with a piezoelectric sound transducer. The results showed that for each contraction, the MMG signal occurred in discrete bursts, rather than a continuous tone. Thus, the explanation provided by the authors was that since a muscle is a constant volume system, stepwise changes in fiber length should also cause stepwise changes in fiber radius. These stepwise changes in fiber radius, in turn, were thought to

generate the MMG signal (Brozovich and Pollack 1983). The results from these studies (Barry 1987; Barry and Cole 1988, 1990; Frangioni et al. 1987; Dobrunz et al. 1990; Brozovich and Pollack 1983) provided some very useful information regarding the mechanisms underlying the MMG signal during electrically stimulated isometric contractions.

It is important to point out that the mechanisms underlying the MMG signal during an electrically-stimulated contraction are very different from those during a voluntary muscle action (dynamic or isometric). Specifically, during an electrically stimulated contraction, all fibers are stimulated to contract simultaneously, and the response of the muscle is dependent on the stimulation frequency and the muscle's ability to contract and relax at a rate that matches the stimulation rate. If the stimulation rate is too fast for the contraction and relaxation times of the muscle (e.g., due to muscle fatigue), then the twitches of the muscle eventually become fused, resulting in tetanus. During a voluntary muscle action, however, the motor unit activities are usually not synchronized, and the twitches from each motor unit are summed to create a complex MMG signal (i.e., complex in the sense that it is generated by many motor units that are firing at different times, as opposed to one or a few motor units that are firing synchronously). An obvious question then, is how are the motor unit mechanical activities during a voluntary muscle action summed to form the MMG signal? Is this summation linear or nonlinear?

A very important study in this regard was conducted by Orizio et al. (1996). To determine how the motor unit mechanical activities are summed to form the MMG signal, the authors electrically stimulated two separate motor units in the extensor digitorum communis muscle at different rates. During the first stimulation protocol, the first motor unit was stimulated at 3 Hz, and the second motor unit was stimulated separately at a frequency of 8 Hz. This same procedure was then performed a second time, but at stimulation rates of 9 and 20 Hz for the first and second motor units, respectively. The MMG signal from the 3 Hz stimulation was then linearly summed with that from the 8 Hz stimulation to create a new MMG signal. The same procedure was also followed for the MMG signals from the 9 Hz and 20 Hz stimulation rates. The second stimulation protocol involved simultaneous stimulation of the first and second motor units at 3 and 8 Hz. The resulting MMG signal was then compared with that from linearly summing the MMG signals from the 3 and 8 Hz stimulation rates. The MMG signal from the simultaneous 9 and 20 Hz stimulation was also compared with that from the linear sum of the separate 9 and 20 Hz stimulations. Figure 2 shows that the MMG signal from the simultaneous 3 and 8 Hz stimulation was nearly identical to that from linearly summing the signals from the separate 3 and 8 Hz stimulations, which suggested that when two separate motor units are active at these stimulation rates,



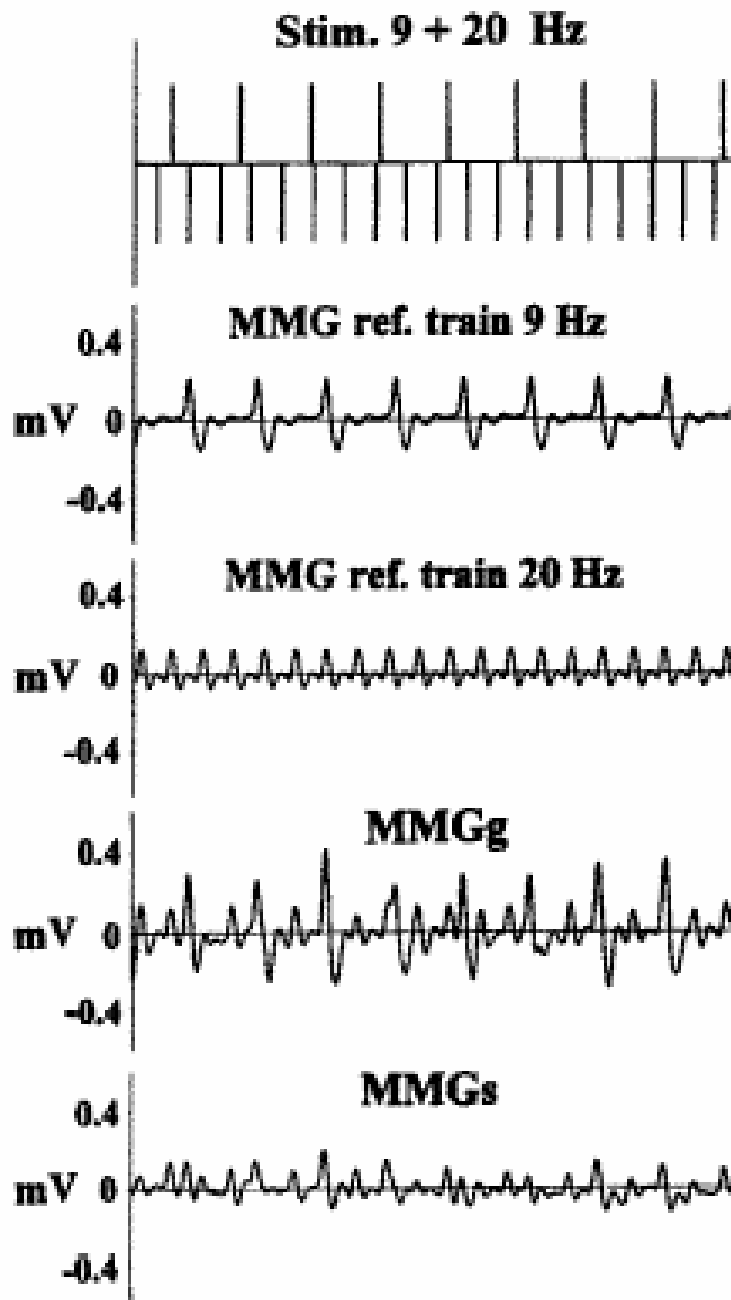
**Figure 2.** One second time windows in which the mechanomyographic (MMG) signal was detected from the extensor digitorum communis during separate electrical stimulations of single motor units at 3 Hz (second graph from top) and 8 Hz (third graph from top). MMGg shows the MMG signal generated from mathematical (i.e., linear) summation of the 3 and 8 Hz stimulation signals. MMGs demonstrates the MMG signal from simultaneous stimulation of the separate motor units at 3 and 8 Hz. Notice that MMGg and MMGs are nearly identical, thereby indicating linear summation of the mechanical activities from the two motor units. The top graph shows the electrical stimulation protocol. \*Reprinted with permission from Orizio et al. (1996).

their contributions are summed linearly to form the MMG signal.

As shown in Figure 3, however, the MMG signal from the simultaneous 9 and 20 Hz stimulation was very different from that of the linear sum of the separate 9 and 20 Hz signals. Thus, these findings suggested that when these higher stimulation rates were used, the contributions from the first and second motor units were not summed linearly. The practical importance of these results is that most voluntary muscle actions involve firing rates greater than 8 Hz. Therefore, during almost all voluntary activities, the contributions of each motor unit are summed nonlinearly to form the MMG signal.

Another important question is whether or not individual motor unit activities can be extracted from the MMG signal. Three important studies in this regard were conducted fairly recently. Specifically, Bichler and Celichowski (2001, p. 388) reported that during an electrically stimulated fatigue test of isolated motor units in rat medial gastrocnemius muscle, the motor units that were classified as “fast fatigable” (i.e., based on their MMG responses) demonstrated the greatest signs of fatigue, while the “fast resistant” and “slow-twitch” motor units were the most resistant to fatigue. Similar results were reported by Bichler (2000), who found that during electrically stimulated contractions (both single twitches and a fused tetanus) of isolated rat medial gastrocnemius muscle, the fast fatigable motor units demonstrated higher MMG amplitude values than both the fatigue resistant and slow-twitch motor units. In addition, Bichler and Celichowski (2001) found that during electrically stimulated contractions of isolated motor units in rat medial gastrocnemius muscle at frequencies that resulted in unfused tetanus, MMG amplitude was positively correlated with the velocity of the tension increase at the beginning of the contraction. Thus, it is possible that under these conditions, MMG amplitude could be used to identify different types of motor units (Bichler and Celichowski 2001).

Two interesting studies in this area have also been performed by Kaczmarek et al. (2005, in press). Specifically, the authors used both experimental and model MMG signals to examine the influence of motor unit location on the MMG signals from pennate muscles. In both investigations, the authors found that the characteristics of the MMG signal were influenced by the location of the MMG sensor over the muscle as well as the location of the active motor units within the muscle. These findings (Kaczmarek et al. 2005, in press) were important from a practical standpoint because they were the first to describe the influence of muscle architecture on the MMG signal. In addition, muscle architecture can affect the MMG signal, independent of the twitch properties of the active motor units.



**Figure 3.** One second time windows in which the mechanomyographic (MMG) signal was detected from the extensor digitorum communis during separate electrical stimulations of single motor units at 9 Hz (second graph from top) and 20 Hz (third graph from top). MMGg shows the MMG signal generated from mathematical (i.e., linear) summation of the 9 and 20 Hz stimulation signals. MMGs demonstrates the MMG signal from simultaneous stimulation of the separate motor units at 9 and 20 Hz. Notice that unlike the results shown in Figure 2, MMGg and MMGs are not identical, thereby indicating nonlinear summation of the mechanical activities from the two motor units. The top graph shows the electrical stimulation protocol. \*Reprinted with permission from Orizio et al. (1996).

Another important question is whether or not the contributions from single motor units can be extracted from the MMG signal recorded during a voluntary muscle action. The results from the electrical stimulation studies indicated that fast-twitch motor units demonstrated greater MMG amplitude values than slow-twitch motor units, but the responses during voluntary muscle actions were still unclear. The first study to examine this issue was Yoshitake and Moritani (1999), who used the spike-triggered averaging technique to isolate single motor unit activities in MMG signals. Specifically, the authors (Yoshitake and Moritani 1999) detected intramuscular EMG signals from the biceps brachii muscle to identify individual motor unit action potentials. The time occurrences of these action potentials were then used to trigger the MMG signals, and the resulting peak-to-peak amplitudes of the MMG signals from the motor unit spikes were examined. The authors (Yoshitake and Moritani 1999) found that the peak-to-peak amplitude of the triggered MMG signal was positively correlated ( $r = 0.631-0.657$ ) with that of the indwelling EMG signal. In a second study, Yoshitake et al. (2002) electrically stimulated isolated motor units from the medial gastrocnemius muscle and reported that during single twitches, the duration of the MMG signal was positively correlated with both half relaxation time ( $r = 0.76$ ) and twitch duration ( $r = 0.89$ ). Furthermore, during repeated electrical stimulation of the motor units at frequencies of 5, 10, 15, and 20 Hz, the changes in force fluctuations with increases in stimulation rate were positively correlated ( $r = 0.76$ ) with the changes in MMG amplitude. Thus, it was suggested that the characteristics of the MMG signal are affected by the contractile properties of the activated motor units (Yoshitake et al. 2002). The surface EMG signal has also been used to trigger the motor unit mechanical activities in MMG signals (Cescon et al. 2004). Specifically, surface EMG and MMG signals were detected simultaneously from the abductor digiti minimi or first dorsal interosseous during 60-second sustained isometric muscle actions at 2% or 5% MVC, or with a selected motor unit firing at a rate of 8-10 Hz. The results showed that the peak-to-peak amplitudes of the EMG and MMG spikes were not significantly correlated. In addition, the peak-to-peak amplitude of the MMG signal from a single motor unit was different for the 2% versus the 5% MVC contractions. Thus, it was suggested (Cescon et al. 2004) that at these low force levels, the mechanical activities of the active motor units are summated nonlinearly to form the MMG signal. This study (Cescon et al. 2004) was then followed up with a second investigation (Cescon et al. 2006) that used the surface EMG signal to trigger events in MMG signals from higher force levels (20%, 50%, and 80% MVC). The authors (Cescon et al. 2006) found that the EMG motor unit spike amplitude increased with force, while the mean power frequency (MPF) decreased. Thus, it was concluded that the amplitude and frequency contents

of the single motor unit EMG signals were not correlated with the corresponding contents from single motor unit MMG signals (Cescon et al. 2006).

The spike-triggered averaging technique has also been used to examine the influence of sensor location on the MMG signal. For example, Cescon et al. (2007) used intramuscular EMG signals to trigger the MMG signals recorded from 15 different locations (i.e., a grid of accelerometers with 5 rows and 3 columns) over the tibialis anterior during isometric dorsiflexion muscle actions. The authors (Cescon et al. 2007) reported that for each motor unit action potential, the lateral MMG sensors showed signals that were primarily negative in phase, while the medial sensors demonstrated signals that were primarily positive in phase. Thus, it was suggested that a grid of MMG sensors may be capable of providing more complete information regarding the motor unit mechanical activities than when a single MMG sensor is used (Cescon et al. 2007). A similar study (Cescon et al. in press) was also performed with a slightly different arrangement of accelerometers. Specifically, 6 MMG sensors were placed along the long axis of the tibialis anterior muscle with a 30 mm separation. The purpose of this arrangement was to measure the longitudinal component of the MMG signal. Seven additional MMG sensors were also located around the circumference of the lower leg to measure the transverse vibrations, and intramuscular EMG electrodes were used to detect the firings of individual motor units. The EMG signal from the intramuscular electrodes was used as a trigger to identify the individual motor unit contributions to MMG signals. The authors (Cescon et al. in press) reported that the peak-to-peak amplitude of the spike-triggered MMG signal was the same for all sensors located along the long axis of the tibialis anterior muscle, but it was different for the sensors placed around the circumference of the lower leg. In addition, the MMG sensors that were 180° apart around the circumference of the lower leg demonstrated MMG signals that were opposite in phase. Thus, it was suggested that the dependence of the MMG signal on the transverse, but not the longitudinal location of the sensor supported the hypothesis that lateral displacement of muscle fibers, rather than changes in muscle fiber diameter, is the primary mechanism generating the MMG signal (Cescon et al. in press).

Miyamoto and Oda (2003) performed an important study that investigated the effect of changes in knee and/or ankle joint angle on the MMG and EMG responses for the soleus and both the medial and lateral heads of the gastrocnemius muscle during an isometric MVC of the plantar flexors. Specifically, the subjects were required to perform maximal isometric muscle actions of the plantar flexors at knee joint angles ranging from 60° to 180° (full extension) in steps of 30°, and ankle joint angles ranging from 80° (10° of dorsiflexion) to 120° (30° of plantar flexion) in steps of 10°. The results

showed that MMG amplitude for the medial and lateral heads of the gastrocnemius increased with increases in knee joint angle, but there were no changes in MMG amplitude for the soleus. In addition, as the ankle angle increased from 80° to 120°, MMG amplitude decreased for all three muscles. Finally, the changes in ankle and knee joint angles had no effect on the EMG amplitude values for any of the muscles. Thus, it was concluded that the MMG signals detected from individual muscles during an isometric MVC may be useful for providing information regarding the length-tension relationships (Miyamoto and Oda 2003). In addition, these findings supported those from previous *in vitro* studies that showed changes in MMG amplitude with changes in muscle length. This study was followed up by a second investigation (Miyamoto and Oda 2005) that examined the influence of changes in elbow joint angle on MMG amplitude for the biceps brachii during unfused and fused electrically-stimulated contractions. Specifically, the subjects were required to perform an electrical stimulation protocol in which the biceps brachii was stimulated at the motor point with single twitches, as well as at frequencies that caused unfused (10 Hz) and fused (30 Hz) contractions. This stimulation protocol was performed at elbow joint angles of 75°, 90°, 105°, 120°, 135°, and 150°, where 180° represents full extension of the forearm. The results showed that during the 10 Hz stimulation, decreases in force fluctuation were accompanied by reductions in MMG amplitude as the elbow joint angle increased. In addition, for the 30 Hz stimulation, the force fluctuations and MMG amplitude were similar at all elbow joint angles.

Finally, the contraction time and half relaxation time at each elbow joint angle were correlated with the force fluctuation and MMG amplitude during the 10 Hz stimulation, but not during the 30 Hz stimulation. Thus, it was concluded that the 30 Hz stimulation protocol caused fusion of motor unit twitches at all elbow joint angles, but the 10 Hz stimulation allowed for investigation of the motor units in the unfused state. Therefore, MMG could be a useful technique for studying the development of fusion and changes in the contractile properties of the muscle during unfused contractions (Miyamoto and Oda 2005). Farina et al. (2008) also performed an important study that examined the influence of motor unit location on the amplitude of the MMG signal. Specifically, 12 separate MMG signals were detected with a 3 × 4 grid of accelerometers that was placed over the tibialis anterior muscle, and three separate indwelling EMG signals were detected simultaneously with wire electrodes from different locations in the muscle. The subjects were instructed to voluntarily activate three different motor units with feedback from the indwelling EMG signals. The MMG signals from the 12 accelerometers were then averaged, and the indwelling EMG signals were used to spike trigger the MMG signal. The results showed that the peak-to-peak amplitude of the spike-

triggered MMG signal was influenced both by the location of the motor unit within the muscle and the location of the accelerometer. Thus, it was concluded that like surface EMG, the volume conductor has an important influence on the MMG signal, and interpretations of MMG amplitude should take into account this influence. In addition, any relationship between twitch torque and single motor unit MMG responses from only one recording location has validity that is limited to motor units that are in a similar location (Farina et al. 2008).

In summary, the results from the studies described in this chapter showed that the MMG signal is generated primarily by lateral oscillations of muscle fibers. In addition, the MMG signal is highly complex, and reflects the mechanical activities of individual motor units. This information could be useful for describing differences in fiber type composition and identifying the mechanisms that underlie the force generation process. More research is needed, however, to determine the factors that can affect the individual motor unit contributions to the MMG signal.

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