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6 Technical aspects of surface mechanomyography

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Abstract

The technical aspects of detecting surface mechanomyographic (MMG) signals are important, particularly when a laboratory first decides to conduct MMG research, or when detecting MMG signals from muscles in new applications, such as for prosthesis control. Of particular importance is the amount of contact pressure applied over the sensor, since high contact pressures can attenuate the muscle fiber vibrations that generate the MMG signal. Recent studies have also shown that the MMG amplitude and frequency responses may be different, depending on whether an accelerometer,

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piezoelectric contact sensor, or condenser microphone is used to detect the MMG signal. Several investigations have also used a laser displacement sensor to detect the MMG signal, since there is no contact pressure with this sensor, and it directly measures skin displacement in physiological units (i.e., millimeters or micrometers), rather than transducer-dependent units (i.e., volts or millivolts).

Introduction

The technical aspects of detecting surface mechanomyographic (MMG) signals are important, particularly when a laboratory first decides to conduct MMG research, or when detecting MMG signals for a new application (e.g., controlling an externally-powered prosthesis). The first study to directly examine the technical aspects of MMG was Bolton et al. (1989). Specifically, the authors investigated the MMG responses from the thenar muscle group during supramaximal electrically-stimulated isometric twitches. The MMG signals were detected with either an electret condenser microphone or a piezoelectric crystal contact sensor. The results showed that the MMG responses for the electret condenser microphone were highly variable, as the sound wave amplitude varied 25% when different microphones of the same model were used. This finding highlighted the importance of accurately calibrating each MMG sensor prior to measuring any signals from contracting muscle. In addition, the authors found that contact pressure was very important in determining the response of the piezoelectric crystal contact sensor, as was the location of the sensor over the thenar muscles. Specifically, the investigators moved the electret condenser microphone over the center of the thenar muscle group, as well as in 1.5 cm increments that were parallel or at right angles to the long axis of the thumb. The results showed that MMG amplitude was greatest when the sensor was over the middle, or belly of the muscle, and then decreased rapidly as it was placed closer to the border of the muscle. In addition, the pressure of the microphone on the skin had a large effect on the amplitude of the MMG signal, as increases in pressure tended to result in greater MMG amplitude values. The authors also suggested that the entire MMG frequency range (i.e., 0-100 Hz) should be used, rather than high-pass filtering the signal at an arbitrary cutoff of, for example, 5 or 10 Hz. In fact, the investigators found that even when frequencies below 1 Hz were eliminated, the resulting MMG signal was distinctly different from that when this frequency was passed. Thus, it was concluded that the contact pressure of the sensor and its location over the muscle, as well as the bandwidth of the MMG signal, are all important factors when collecting MMG data (Bolton et al. 1989). Accornero et al. (1989) were among the first to develop a single probe

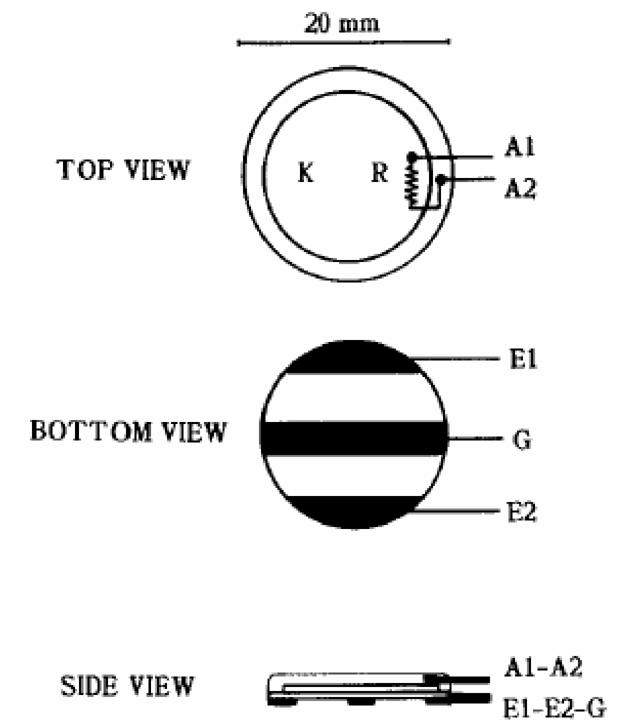

Figure 1. Example of the sensor designed by Accornero et al. (1989) to simultaneously detect mechanomyographic (MMG) and electromyographic (EMG) signals. The sensor is a piezoceramic transducer with two active electrodes (E1 and E2) at its border and a ground electrode in the center. *Reprinted with permission from Accornero et al. (1989).

that could be used to simultaneously detect surface MMG and electromyographic (EMG) signals. Specifically, the probe consisted of a piezoceramic disc that was glued to a flexible printed circuit board with three copper strips that contacted the skin surface. The piezoceramic disc served as the MMG sensor, and the flexible printed circuit board with the three copper strips was used to detect the surface EMG signal. In addition, the surface EMG electrodes were preamplified, which reduced the need for skin abrasion prior to recording (Figure 1).

The authors found that this device worked well for detecting MMG and EMG signals, and suggested that the probe could be useful when conducting studies on muscle fatigue, where MMG and EMG signals often show different responses (Accornero et al. 1989). Smith and Stokes (1993) also performed a very important study to examine the influence of contact pressure on the MMG amplitude responses at different submaximal isometric force levels. Specifically, the subjects were required to perform separate isometric muscle actions of the leg extensors at 10%, 25%, 50%, 75%, and 100% of the maximum voluntary contraction (MVC), and an MMG signal was detected from the rectus femoris during each muscle action. In addition, three different contact pressures (180, 790, and 1200 Pascals) were applied to the MMG sensor to determine if contact pressure affected the patterns of responses for MMG amplitude versus isometric force. The results showed that the MMG amplitude versus isometric force relationships were very similar for the 180 and 790 Pascal contact pressures. When the contact pressure was increased to 1200 Pascals, however, the mean MMG amplitude values increased significantly at all force levels, and the patterns of responses became more curvilinear. Thus, it was concluded that contact pressure is a very important factor when examining the patterns of responses for MMG amplitude versus isometric force.In addition, there may be a threshold contact pressure that should not be exceeded when recording MMG signals, but this pressure is likely to be different for different subjects due to discrepancies in subcutaneous skinfold thickness (Figure 2).

Finally, the authors recommended that absolute MMG amplitude should not be used to predict force production by the muscle, since contact pressure and

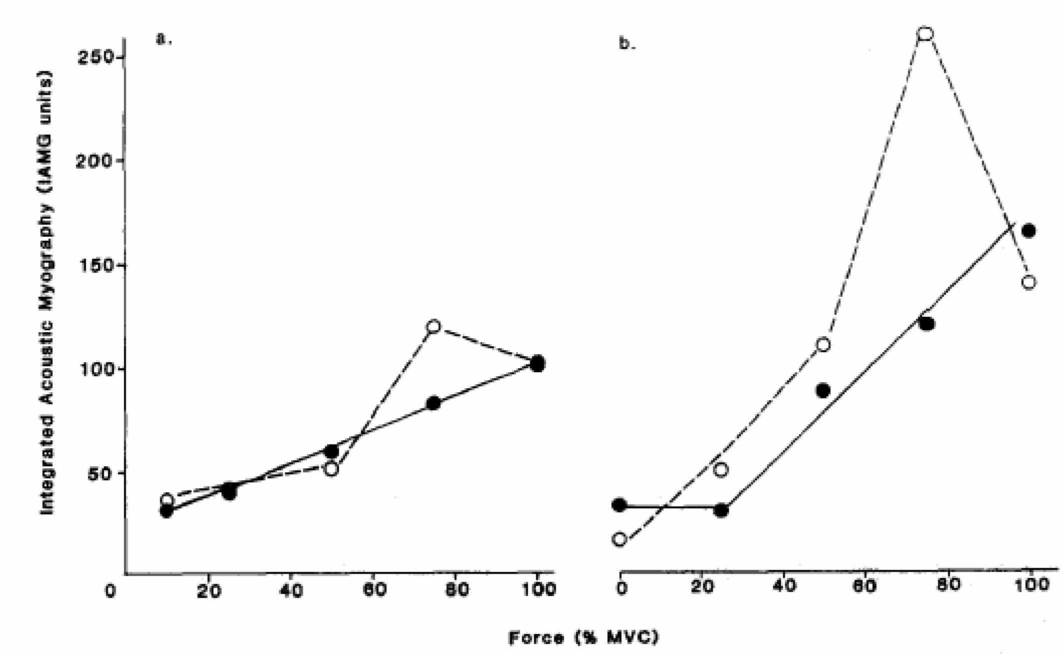

Figure 2. Changes in mechanomyographic (MMG) amplitude (indicated as Integrated Acoustic Myography in this figure) for the rectus femoris muscle with increases in isometric leg extension force for two separate subjects (a and b, respectively). The solid symbols represent the data when the contact pressure of the sensor was 180 Pascals, and the open symbols show the data for a much higher contact pressure (790 Pascals in (a) and 1200 Pascals in (b)). Notice that for the low contact pressure, the increases in MMG amplitude with force were highly linear, but MMG amplitude decreased from 80-100% MVC for the higher contact pressures. *Reprinted with permission from Smith and Stokes (1993).

differences in skinfold thickness can have a large influence on these values (Smith and Stokes 1993). Watakabe et al. (1998) also addressed an important technical aspect of detecting MMG signals with a piezoelectric contact sensor and an accelerometer. Specifically, the primary purpose of the investigation was to determine exactly what physiological parameters (i.e., acceleration, velocity, or displacement) were being measured by a piezoelectric crystal contact sensor and an accelerometer.Thus, both the piezoelectric crystal contact sensor (Hewlett Packard, model 21050A) and accelerometer (Nihon Koden MT-3T) were tested on a sinusoidal vibration system, as well as during voluntary isometric muscle actions of the rectus femoris. The results showed that when the double integral of the accelerometer signal was calculated (i.e., to convert acceleration in $m \cdot s^{-2}$ to displacement in µm), the resulting MMG signal was very similar in shape to that obtained by the piezoelectric crystal contact sensor. Thus, it was concluded that the piezoelectric crystal contact sensor measures skin displacement during muscle contraction. In addition, the amplitude of the MMG signal detected during a voluntary muscle action increased progressively with contact pressures ranging from 0.5-1.5 Newtons, to the point where the MMG amplitude values for a contact pressure of 1.5 Newtons were nearly twice as large as those for a contact pressure of 0.5 Newtons. Thus, it was concluded that both the accelerometer and piezoelectric crystal contact sensor are adequate devices for detecting MMG signals. However, the advantage of the accelerometer is that it provides a measurement in physical units (i.e., $m \cdot s^{-2}$), rather than transducer-dependent units (Watakabe et al. 1998). This study was followed up by a second investigation (Watakabe et al. 2001) that examined the characteristics of a condenser microphone (Matsushita Communication Industrial model WM-034B) and an accelerometer (Kistler Instrument model 8352A2). Specifically, the authors investigated the frequency response of the condenser microphone during mechanical sinusoidal vibration, as well as a voluntary muscle action of the forearm flexors. The results showed that during the sinusoidal vibration, the frequency response of the condenser microphone was highly dependent on the length of the air chamber used in the microphone, such that lengths of 15, 20, and 25 mm resulted in cutoff frequencies of 10, 8, and 4 Hz, respectively. In addition, during the voluntary muscle actions, the MMG signal from the condenser microphone was very similar in shape to the double integral of the corresponding signal from the accelerometer. Thus, it was concluded that with the condenser microphone, muscle vibrations cause skin displacement, resulting in pressure changes in the air chamber of the microphone. In addition, when the limb was moved to purposefully create movement artifact, the resulting amplitude of the MMG signal increased 7.7 to 12.3 times the baseline value. The movement artifact was much more severe, however, for the accelerometer than the condenser microphone. Thus, it was concluded that the condenser microphone is probably a much better sensor for detecting the MMG signal during dynamic muscle actions than is the accelerometer. In addition, the condenser microphone should have a diameter of at least 10 mm and a length of 15 mm in order to accurately cover the frequency range of MMG signals (Watakabe et al. 2001). The last study in the series that examined the responses of different MMG sensors investigated the MMG responses of an accelerometer (Kistler Instrument 8352A2) and a laser displacement sensor (Keyence Corporation LK-080). These devices were compared during sinusoidal mechanical vibration, as well as voluntary isometric muscle actions of the rectus femoris at 20%, 40%, and 60% of the isometric leg extension MVC. The results showed that the double integral of the accelerometer signal was very similar to the corresponding signal from the laser displacement sensor, and, therefore, the accelerometer accurately measured acceleration of the skin surface. In addition, the MMG signal from the accelerometer was gradually distorted when weights of 2, 4, 10, and 50 grams were applied over it. Thus, it was suggested that when detecting MMG signals, the weight of the accelerometer should not be greater than 5 grams (Watakabe et al. 2003).

Wee and Ashley (1990) investigated the possibility of cross-talk with surface MMG signals. One MMG sensor was placed over the biceps brachii muscle and a second sensor over the triceps brachii muscle. The subjects were then required to perform a sustained isometric muscle action of the forearm flexors in which they supported a load of 2.3 kg, and MMG signals were detected simultaneously from both the biceps brachii and triceps brachii. The results indicated that there was significant transmission of muscle vibrations from the biceps brachii to the triceps brachii during the forearm flexion muscle action. Thus, it was suggested that the transmission of muscle vibrations to distant tissues should be considered when recording MMG signals, particularly when the muscles of interest are in close proximity to each other (Wee and Ashley 1990). Inoue et al. (2000) examined the possibility of using an accelerometer to develop a portable MMG system that could be used to record MMG signals during daily life activities. Specifically, the system incorporated multiple sensors, including sensors for the electrocardiogram, limb acceleration, and angular speed, as well as the MMG signal. The results showed that the portable MMG system accurately measured MMG data during a variety of daily life activities, including walking, quiet sitting, getting on and off of a train, as well as isometric and dynamic training. Thus, it was concluded that the portable MMG system could be used to examine muscle function during various activities of daily living away from the laboratory setting (Inoue et al. 2000). Madeleine et al. (2006) examined the influence of sensor location on the patterns of responses for MMG amplitude and center frequency [both mean power frequency (MPF) and median frequency] versus isometric force for the tibialis anterior muscle. A 5 × 3 grid of accelerometers was placed over the tibialis anterior, and the subjects were required to perform submaximal to maximal isometric muscle actions of the dorsiflexors. The results showed that different sensor locations caused different MMG amplitude and center frequency values, as well as different patterns of responses across isometric force. Thus, it was concluded that it would be difficult to describe motor control strategies from MMG amplitude and center frequency patterns when only one sensor is used to detect the signal. In addition, it may be necessary to record MMG signals from multiple locations, and use a composite, or some other type of weighted response when describing motor control strategies (Madeleine et al. 2006).

Jaskólska et al. (2007) compared a condenser microphone with an accelerometer for examining the MMG amplitude and MPF responses during submaximal concentric, isometric, and eccentric muscle actions. The experimental protocol required the subjects to perform submaximal concentric, isometric, and eccentric muscle actions of the forearm flexors at 10%, 30%, 50%, and 70% of the isometric MVC. During each muscle action, two separate surface MMG signals were detected from the biceps brachii with an accelerometer and a condenser microphone. The results showed that during the concentric, eccentric, and isometric muscle actions, the overall shape of the MMG amplitude versus torque relationships were similar for the condenser microphone and accelerometer. The MMG MPF versus torque relationships were different, however, for the two sensors, particularly during the eccentric muscle actions. Thus, it was concluded that the accelerometer and condenser microphone may provide different MMG amplitude and/or center frequency responses, and, therefore, the type of sensor that is used could affect the interpretation of the patterns of responses (Jaskólska et al. 2007). These findings were similar to those of Beck et al. (2006), who compared a piezoelectric crystal contact sensor with an accelerometer for the patterns of responses for MMG amplitude and MPF versus torque for the biceps brachii during both concentric and isometric muscle actions of the forearm flexors (Figure 3).

The subjects were required to perform submaximal to maximal concentric isokinetic and isometric muscle actions of the forearm flexors, and two separate surface MMG signals were detected simultaneously from the biceps brachii with an accelerometer and a piezoelectric crystal contact sensor. The results showed that during both the dynamic and isometric muscle actions, there

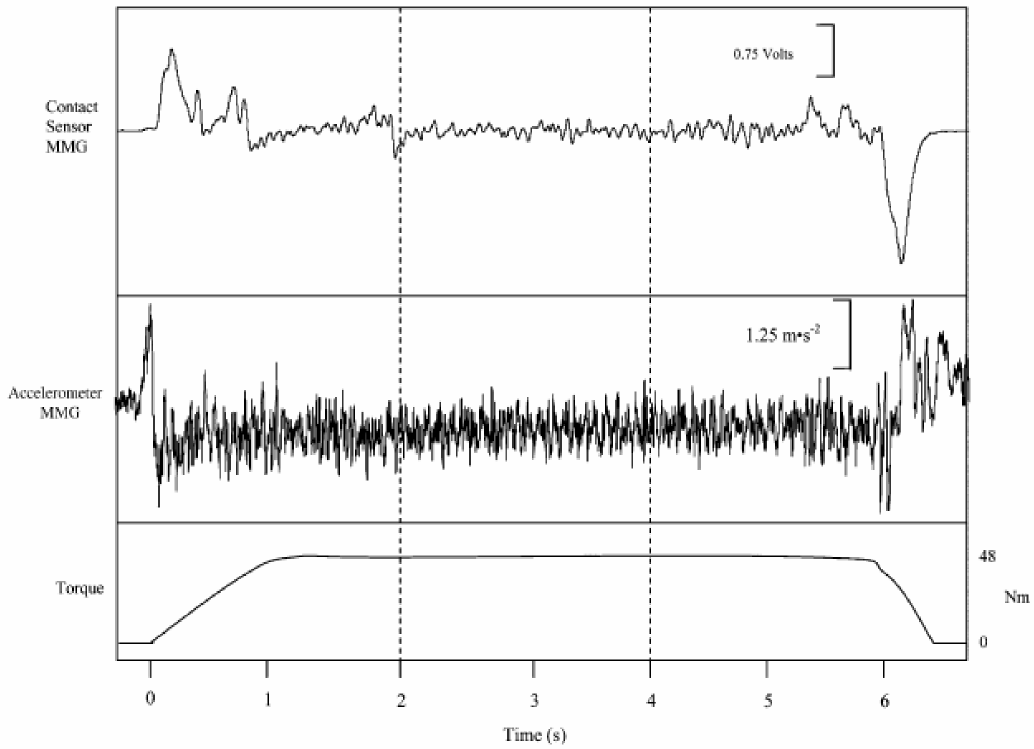

Figure 3. Mechanomyographic (MMG) signals detected from the biceps brachii during a 6-second isometric muscle action of the forearm flexors of one subject at 80% MVC. The top graph shows the signal from a piezoelectric crystal contact sensor, the middle graph is the signal from an accelerometer, and the bottom signal is forearm flexion torque. Notice that the MMG signals from the contact sensor and accelerometer had very different shapes. *Reprinted with permission from Beck et al. (2006).

were linear increases in MMG amplitude with torque for the accelerometer and piezoelectric crystal contact sensor. However, the linear slope for the normalized MMG amplitude versus isokinetic torque relationship for the accelerometer was less than that for the contact sensor. In addition, there was no significant relationship for normalized MMG MPF versus isokinetic and isometric torque for the contact sensor, but the accelerometer demonstrated a quadratic or linear relationship for the isokinetic and isometric muscle actions, respectively. There were also several significant mean differences between the contact sensor and accelerometer for normalized MMG amplitude and MPF values. Thus, it was concluded that in some cases involving dynamic and isometric muscle actions, the contact sensor and accelerometer resulted in different torque-related responses for MMG amplitude and MPF that may affect the interpretation of the motor control strategies involved (Beck et al. 2006). Ouamer et al. (1999) examined the influence of sensor orientation (i.e., parallel versus perpendicular to the long axis of the muscle) on MMG responses from the biceps brachii during a voluntary isometric muscle action.

Specifically, two separate surface MMG sensors were placed either parallel or perpendicular to the long axis of the biceps brachii muscle as the subjects were required to perform an isometric MVC of the forearm flexors, as well as submaximal muscle actions at 30% and 50% MVC. The results indicated that when the MMG sensors were placed in line with the muscle, the resulting signals were in phase with each other. When the sensors were placed perpendicular to the long axis of the muscle, however, the resulting signals were reversed in phase. Thus, it was concluded that sensor location relative to the long axis of the muscle is an important consideration when detecting MMG signals. It was also hypothesized, however, that the resonant bending vibrations may not be the only movement that takes place in the muscle. Specifically, frequencies between 10 and 16 Hz may reflect non-resonant movements that could be due to local (i.e., at the muscle fiber or motor unit level) factors, rather than factors at the whole muscle level (Ouamer et al. 1999). Courteville et al. (1998) performed an interesting study that described the use of a high sensitivity microphone for detecting MMG signals. Specifically, the authors used a condenser microphone to detect MMG signals from the flexor digitorum superficialis during submaximal isometric muscle actions of the wrist flexors at 15%, 25%, 30%, 35%, and 45% MVC. The results showed that both MMG amplitude and MMG MPF increased with isometric force. In addition, the authors reported that there were several advantages of their MMG sensor when compared to other devices, such as accelerometers and piezoelectric crystal contact sensors. These advantages included a high sensitivity to low frequency vibrations, a value that is expressed in displacement units, and insensitivity to global movements of the muscle and/or limb (Courteville et al. 1998).

Mito et al. (2007) investigated the effects of changes in skin temperature on MMG and EMG amplitude during voluntary isometric muscle actions of the forearm flexors. The subjects were required to perform isometric forearm flexion muscle actions at 20%, 40%, 60%, and 80% MVC, and surface MMG and EMG signals were detected simultaneously from the biceps brachii. This series of muscle actions was performed at a control temperature (34˚ Celsius), as well as at higher (40˚ Celsius) and lower (28˚ Celsius) temperatures that were achieved by heating and cooling the muscle, respectively. The results showed that for all temperatures, both MMG and EMG amplitude for the biceps brachii increased with isometric force. In addition, there were increases in MMG amplitude for the high temperature condition from 20-60% MVC, but then MMG amplitude plateaued from 60-80% MVC. These responses were different, however, under the control and low temperature conditions, and the changes in temperature had no effect on the patterns of responses for EMG amplitude. Thus, it was concluded that the changes in muscle temperature likely altered the mechanical properties of the muscle fibers in the biceps brachii, thereby affecting the MMG amplitude responses (Mito et al. 2007). Kim et al. (2008) examined the influence of force tremor on the MMG signals detected with a condenser microphone and an accelerometer. Specifically, the subjects performed isometric muscle actions of the forearm flexors and extensors at 20%, 40%, 60%, 80%, and 100% MVC, and surface MMG signals were detected from the biceps brachii and triceps brachii with a condenser microphone and an accelerometer. The results showed that during agonist muscle activity (either biceps brachii or triceps brachii), the amplitude of the MMG signal from the condenser microphone was greater for the agonist than for the antagonist. In contrast, however, MMG amplitude for the accelerometer was often similar in the agonist and antagonist muscles. Thus, it was concluded that the MMG signal detected by an accelerometer is more susceptible to force tremor than the corresponding signal from a condenser microphone, since the force tremor was transmitted to the antagonist muscle and detected by the accelerometer. It was also recommended, however, that future studies need to be done to determine the mechanisms that cause an accelerometer to be more susceptible to force tremor than a condenser microphone (Kim et al. 2008). Kim et al. (2008) compared the MMG signals detected with a condenser microphone versus an accelerometer during fatiguing isometric muscle actions of the forearm flexors. Specifically, the MMG signals were detected from both the biceps brachii (i.e., an agonist muscle) and the triceps brachii (i.e., an antagonist muscle) during a sustained isometric muscle action of the forearm flexors at 30% MVC. The results showed that the condenser microphone provided different MMG amplitude and frequency responses when compared to the accelerometer. In addition, the MMG signal detected by the condenser microphone was less affected by physiological tremor than the corresponding signal from the accelerometer. Thus, it was concluded that in situations where physiological tremor could affect the MMG signal, it may be more appropriate to use a condenser microphone to detect the signal than an accelerometer (Kim et al. 2008).

Silva and Chau (2005) examined the use of a mathematical model for removing movement artifacts from MMG signals for the purpose of controlling an externally powered prosthesis. Specifically, a microphone and accelerometer were coupled into a single sensor that could be used not only for prosthesis control, but also for removing the sounds associated with swallowing and respiration, as well as heart sounds. The results showed that the mathematical model accurately separated the sounds associated with muscle contraction from the movement artifacts, and, therefore, helped to improve the accuracy of controlling an externally powered prosthesis. In addition, it was suggested that future studies should use the model to try to remove other noise sources from MMG signals (Silva and Chau 2005). Gregori et al. (2003) investigated the feasibility of using a differential probe to reject movement artifacts from MMG signals. Specifically,two piezoelectric membranes were spaced 25 mm apart and integrated into a differential circuit board that allowed for detection of the differential MMG signal. The results showed that the MMG signal from the differential probe was very similar in amplitude and frequency to that of the signal from a non-differential, or single MMG sensor. In addition, the differential probe allowed for a much better rejection of movement artifacts, and, therefore, improved the signal-to-noise ratio. Thus, it was concluded that the differential MMG probe could be useful for recording MMG signals in noisy environments (Gregori et al. 2003).

Rafolt and Gallasch (2002) examined the use of a scanner galvanometer to detect the muscle vibrations associated with MMG signals. The precision of the galvanometer was first tested on a piezo-electric disc actuator, which showed that when the contact pressure forces ranged from 0.1-2.0 Newtons, the galvanometer was capable of detecting skin deflections of 1 µm. In the second experiment, the galvanometer was placed over the gastrocnemius muscle, and the posterior tibial nerve was stimulated at progressively higher voltages to elicit different twitch responses. The results showed that for contact pressure forces between 0.1 and 20 Newtons, the amplitude of the skin displacement detected by the galvanometer increased with stimulation voltage. Finally, the responses of the galvanometer were compared with those of an accelerometer during a submaximal isometric muscle action of the plantar flexors. The results showed that when the galvanometer displacement signal was double differentiated, the resulting acceleration signal was very similar to that recorded simultaneously from an accelerometer. Thus, it was concluded that the scanner galvanometer is an accurate instrument for detecting the skin fluctuations associated with the MMG signal. It was also recommended, however, that future investigations should test the responses of the galvanometer under different testing conditions and with various muscles (Rafolt and Gallasch 2002). Akataki et al. (1999) examined the within-day and between-day reliability of MMG and EMG amplitude for the biceps brachii during submaximal to maximal isometric muscle actions of the forearm flexors. The results showed that the coefficient of variation values for within-day and between-day reliability were very similar for the MMG and EMG signals. In addition, the within-day intraclass correlation coefficient for MMG amplitude was $R = 0.95$, with a between-day coefficient of $R = 0.80$. Thus, it was concluded that the repeatability of MMG amplitude for the biceps brachii was similar to that for EMG amplitude (Akataki et al. 1999).

Overall, the results from the studies discussed in this chapter indicated that accelerometers, piezoelectric crystal contact sensors, condenser microphones, and laser displacement sensors can all be used to detect MMG signals. However, contact pressure, sensor location, and an understanding of the measurement units provided by the MMG sensor are important when interpreting both MMG amplitude and frequency responses. These factors become particularly important when using MMG in clinical applications, such as for control of externally powered prostheses. In addition, sensors that provide an MMG signal in volts or millivolts must be calibrated if the signal is to be converted to physiological units, such as micrometers or $m{\cdot}s^{-2}$.

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