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Tutorial

Surface Electromyography for Speech and Swallowing Systems: Measurement, Analysis, and Interpretation

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Purpose: Applying surface electromyography (sEMG) to the study of voice, speech, and swallowing is becoming increasingly popular. An improved understanding of sEMG and building a consensus as to appropriate methodology will improve future research and clinical applications.
Method: An updated review of the theory behind recording sEMG for the speech and swallowing systems is provided. Several factors that are known to affect the content of the sEMG signal are discussed, and practical guidelines for sEMG recording and analysis are presented, focusing on special considerations within the context of the speech and swallowing anatomy.
Results: Unique challenges are seen in application of sEMG to the speech and swallowing musculature owing to the small size of the muscles in relation to the sEMG detection volume and the present lack of knowledge about innervation zone locations.
Conclusions: Despite the challenges discussed, application of sEMG to speech and swallowing has potential as a clinical and research tool when used correctly and is specifically suited to noninvasive clinical studies using between-condition or between-group comparisons for which detection of specific isolated muscle activities is not necessary.

Key Words: voice, speech, swallowing, respiration, electromyography

There is an accumulating body of research in which surface electromyography (sEMG) is used for assessment and rehabilitation of speech and swallowing disorders. sEMG has been used to study and rehabilitate respiration and speech breathing (e.g., Maarsingh, Oud, van Eykern, Hoekstra, & van Aalderen, 2006; Tamplin et al., 2011); voice (e.g., Allen, Bernstein, & Chait, 1991; Andrews, Warner, & Stewart, 1986; Hocevar-Boltezar, Janko, & Zargi, 1998; Stemple, Weiler, Whitehead, & Komray, 1980; Yiu, Verdolini, & Chow, 2005); swallowing (e.g., Crary & Groher, 2000; Huckabee & Cannito, 1999); and speech articulation (e.g., Deng et al., 2009; McClean & Tasko, 2003; Ruark & Moore, 1997). The attraction is clear—sEMG is noninvasive, is seemingly simple to apply, and can provide real-time information about muscle activations. However, sEMG is a technique that can be easily abused due to a lack of knowledge of the factors affecting the signal, inherent technical limitations (e.g., De Luca, 1997), and the anatomy and physiology of the head and neck musculature. Lack of understanding of these issues may explain the inconsistencies in clinical adoption of sEMG for assessment and treatment of voice, speech, and swallowing.

For instance, sEMG signal differences could result from variations in recording methodologies. Surface electrodes intended to detect muscle activation can be placed in nearly any configuration on the body and still detect electrical activity of some kind, including cardiac activity and electrical line noise. Several protocols have been developed for electrode placement to avoid the potential signal misinterpretation and to increase detection reliability (e.g., De Luca, 1997; Hermens, Freriks, Disselhorst-Klug, & Rau, 2000; Hermens et al., 1999). It is imperative that both investigators and consumers of sEMG research understand the appropriate methodologies and limitations, to avoid both wasting time with poor study design and misinterpreting data, which could lead to reduced quality of patient care.

Electrophysiology and the technical aspects of sEMG recording methodology are not commonly included in the standard educational preparation in the discipline of

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speech-language pathology; thus, expertise and experience in this area varies widely among researchers and clinicians in our field. Improved understanding of sEMG and a consensus among our community as to appropriate methodology will improve future research and clinical applications. Although Cooper reviewed electromyography (EMG) for speech research in 1965 and Gay and Harris updated the review in 1971, both of these were primarily focused on needle and hooked wire EMG. Needle and hooked wire EMG are a viable source of information about muscle activity during speech and swallowing, providing information that surface recordings cannot. However, because of its noninvasive nature, sEMG is also now a methodology of choice for a variety of clinical and research applications and provides complementary information to that acquired through invasive techniques. Thus, this tutorial provides an updated review of the theory behind recording sEMG for the speech and swallowing systems. First, details about the generation of the underlying signal are presented. Then, signal detection, the signal composition, data analysis techniques, and possible forms of signal degradation are reviewed. The article closes with a focus on special considerations for speech and swallowing anatomy and future directions.

Motor Unit Physiology and EMG

During muscle contraction, nerve impulses from alpha motor neurons reach motor end plates at the neuromuscular junction. These pulses cause all muscle fibers innervated by that motor neuron's axon to discharge nearly synchronously to create a motor unit action potential (MUAP). MUAPs then propagate along all innervated muscle fibers, away from the motor end plate longitudinally in both directions toward the ends of the muscle fiber. Thus, the electric potential field generated by the depolarization of the extrafusal fiber membranes is essentially an amplified version of the alpha motor neuron activity. The EMG is a representation of this "myoelectricity" as detected at some distance (see Moritani, Stegeman, & Merletti, 2004, for a more detailed review).

The tissues separating the EMG signal sources (depolarized zones of the muscle fibers) from the EMG sensor are referred to as a *volume conductor*. The volume conductor consists of resting muscles, subdermal fat, other soft tissues, and the skin. The volume conductor acts like a spatial low-pass filter on the electrical potential distribution, smoothing each MUAP and decreasing the amplitude of the signal. The distance between the EMG signal sources and the sensors changes the qualities of the volume conductor and, thus, the effects of the spatial low-pass filtering. Greater distances constitute lower signal amplitude and increased smoothing. The EMG may be measured intramuscularly or at the surface of the skin (sEMG), yielding different information based on the distance of the observation site from the active muscle fibers immediately beneath the skin and whether or not other muscles are within the region either immediately beneath the skin or beneath the target muscle. For surface detection particularly, the effect of the separating tissues can become significant, with more than 1–2 cm of subdermal fat at a site precluding the usefulness of sEMG (Merletti, Botter, Troiano, Merlo, & Minetto, 2009).

The sEMG signal is a collection of the multiple MUAPs within the range of the sensor, providing a polyphasic signal of superimposed MUAPs from one or more muscles in the region. The amplitude and frequency content of each of the constitutive MUAPs in the measured sEMG signal is directly related to the distance of each motor unit from the electrode. A MUAP measured from a more superficial muscle fiber will have a larger amplitude and higher frequency content than one measured from a deeper muscle fiber (see Kamen & Caldwell, 1996, for a more detailed discussion). As the central nervous system drives the muscle to generate increased force, more motor units are recruited, and the firing rates of all recruited motor units increase.

As indicated previously, the EMG signal can be detected from the surface of the skin (sEMG) or through an inserted electrode (wire or needle). Intramuscular recordings achieved through the use of needles or wires have the advantage of greater spatial and temporal specificity and can usually provide reliable information about the activation of a select number of motor units and about the overall shape of individual MUAPs. However, intramuscular recordings are relatively invasive, which could cause potential changes in behaviors of some clinical populations of interest. In addition, because the information sampled comes from only a select few motor units, it is not representative of the action of the entire muscle when using bipolar needle electrodes or concentric needle electrodes. However, some intramuscular electrodes can record from much larger regions within a muscle, such as when there is a larger field between two hooked wire electrodes or when a monopolar needle electrode and a distant ground are used. Also, depending on the muscle of interest, it can be difficult to reliably place electrodes within the muscle body of interest due to the lack of direct visualization. However, wire or needle electrodes are the only way to measure EMG from deep muscles and can ensure more selective recordings from single muscles immediately beneath the skin surface.

For muscles that are relatively superficial, sEMG can be effective for detecting muscle activation. Because sEMG detects signals from a larger area, the sEMG signal samples from many motor units. This means that a sEMG signal may be representative of the overall activation of the muscle of interest. However, because of the larger detection area, sEMG is also more prone to detection of signals from nearby muscles (*cross-talk*), and conventional sEMG cannot discriminate among regional differences in activation patterns within a muscle (e.g., Wentzel, Konow, & German, 2010).

How to Detect sEMG

Although sEMG can detect muscle activation, electrical potentials much larger than those produced by muscles can contaminate the sEMG signal. Thus, sEMG signals should always be collected relative to a common reference, referred to as the *ground*. Bioelectrical noise is assumed to be common to both the ground and sensors, allowing the "common mode voltages" to be rejected from the detected signal. The European Union sponsored a project termed Surface Electromyography for the Noninvasive Assessment of Muscles (SENIAM) to collect recommendations on sEMG methodology. In general, SENIAM suggests the wrist, the spinous process of C7, or the ankle as appropriate locations for ground electrodes (Hermens et al., 1999). For individuals recording sEMG from the muscles of speech and swallowing, the use of sites closer to sEMG sensors is recommended, such as the spinous process of C7, the acromion process (bony prominence of the shoulder), forehead, nose, or earlobe. If monopolar electrodes are used, care should be taken not to use a ground that is too low (i.e., with the heart placed between the ground and the electrode) to minimize the effects of cardiac electrical activity in the signal.

The ability to measure high-fidelity sEMG is dependent on the impedance of the electrode–skin interface (the opposition of current flow from the skin to the sensor). Lower impedance of the interface corresponds to improved signal propagation to the electrode and better signal detection. In its natural state, the top layer of the epidermis is electrically insulated, resulting in high electrode–skin impedance. Depending on electrode characteristics and skin state, contact impedance ranges from a few kΩ to a few MΩ, with larger electrodes generally having lower impedance and noise (Merletti et al., 2009). This impedance can be reduced using a variety of methods. The first choice of the researcher is whether to use passive or active electrodes.

Passive electrodes are made of conductive materials that sense electrical current on the skin through the electrode–skin interface, the most simple of which are made of silver. Silver–silver chloride electrodes are also used. These electrodes allow a reversible chloride exchange interface between the electrode and skin and help to minimize motion artifact produced by skin potentials (Webster, 1984). Passive electrodes are often referred to as *wet electrodes,* as they require conductive gel or paste between the electrode and skin to improve the quality of the detected signal. However, Roy and colleagues have shown that the use of conductive gel in the face of perspiration and mechanical perturbations can lead to an increase in the artifacts measured (Roy et al., 2007).

Active electrodes are also referred to as *dry electrodes* or *preamplified electrodes*. These electrodes have signal amplification circuitry embedded near the electrode–skin interface. Active electrodes can increase the signal-to-noise ratio by minimizing source and contact noise, can be used in situations with otherwise unacceptably high electrode–skin impedances, and do not require the use of a conductive gel or paste. These electrodes are preferred in terms of signal quality but are often more expensive and more bulky than passive electrodes.

Regardless of the choice of electrode type, the signal detected can be improved by further reducing the skin–electrode impedance through treatment of the skin. Techniques to remove the dead (top) surface layer of skin and its protective oils can enhance skin–electrode contact, resulting in a reduction of artifacts and noise. SENIAM recommends shaving the skin surface if it is covered with hair and cleaning the skin in question with alcohol (Hermens et al., 1999). Although alcohol treatment has been recommended and has been widely adopted by clinicians and researchers, this practice has been shown to reduce skin–electrode impedance by only roughly 40% (Merletti & Hermens, 2004). Rubbing the skin with medical abrasive paste causes the greatest reduction (roughly 90%) in skin–electrode impedance (Merletti & Hermens, 2004). Because abrasive paste is often unpleasant due to messiness and discomfort, combining the use of alcohol with skin "peeling" is suggested as a less invasive compromise. The practice of light skin abrasion or "peeling" with adhesive tape is known to reduce skin–electrode impedance by roughly 70% (Merletti & Hermens, 2004), and it can be well tolerated by participants, even on delicate skin of the neck and face. It involves repetitive placement and removal of adhesive tape on the skin surface.

As previously discussed, all signals should be recorded relative to the ground. However, in addition, various electrode configurations can be used to further isolate electrical activity of interest. A single electrode that is placed over a muscle and is recorded relative to ground is referred to as a *monopolar configuration*. Monopolar configurations are associated with the largest detection volume and are most susceptible to cross-talk from adjacent muscles. In order to remove interference sources and to compensate for the low-pass filtering effect of the tissue, surface signals are typically detected using a linear combination of different electrodes, the simplest of which is a differential electrode (Farina, Merletti, & Stegeman, 2004).

Differential recording configurations amplify the difference between multiple electrodes placed over the muscle of interest. The most common of these is the *bipolar configuration* (*single differential*). Differential recordings take advantage of "common-mode" rejection, such that potential noise (biological or otherwise) that is sensed at both electrodes is rejected from the amplified signal. In addition, differential recording configurations are more spatially sensitive (smaller detection volume) than monopolar schemes. *Double differential* recording strategies refer to three electrodes linearly configured over the muscle of interest, with three differences among the electrodes used for the resultant signal. This configuration results in a further increase in spatial selectivity relative to bipolar configurations (Merletti & Hermens, 2004). Regardless of whether the differential configuration is single or double, differential electrodes should be aligned so that the electrode axis is parallel to underlying muscular fibers, detecting MUAPs as they travel down the muscle fibers. When electrodes are not aligned parallel to muscle fibers, the amplitude of the detected signal can be reduced by as much as 50% (Vigreux, Cnockaert, & Pertuzon, 1979).

Features of differential sEMG depend on the size of and space between the electrodes, referred to as the *interelectrode distance* (Roeleveld, Stegeman, Vingerhoets, & Van Oosterom, 1997). SENIAM recommends a maximum electrode size of 10 mm in the muscle fiber direction, with an interelectrode distance of approximately one fourth the length of the muscle fiber or 20 mm, whichever is smaller (Hermens et al., 1999). For speech musculature, one fourth the length of the muscle fiber is often smaller than 20 mm. For instance, muscles such as the sternocleidomastoid (SCM) can be roughly 20 cm in length, so 20 mm is easily a smaller distance than 5 cm. However, many muscles such as the mentalis and depressor labii inferior could be 2–4 cm in length, which would require maximum electrode distances of 0.5–1 cm.

In general, the larger the interelectrode distance, the wider the area sampled and the higher the amplitude of the resultant signal, but the less spatially specific (Fuglevand, Winter, Patla, & Stashuk, 1992; Roeleveld et al., 1997). Simulation has shown that larger interelectrode distances can moderately increase detection depth, but the detected sEMG signal is dominated by MUAPs from muscle fibers located within 10–12 mm of the recording electrode (Fuglevand et al., 1992). Further simulation has shown differences in the relationship between interelectrode distances and detection amplitude for superficial and deep fibers (Farina, Cescon, & Merletti, 2002). For superficial fibers, the amplitude detected at the surface is reduced when the interelectrode distance is less than 15 mm, whereas this cutoff is at 25 mm for deeper fibers (Farina et al., 2002). Although increases in the interelectrode distance may increase the activity detected from deeper fibers, activity from superficial fibers will still dominate the signal. Although simulation has indicated that the size of electrodes used does not cause substantial effects on the detection volume (Fuglevand et al., 1992), it has been asserted that smaller electrodes (diameter less than 5 mm) are preferred for sEMG, as the larger electrodes introduce temporal low-pass filtering (Merletti & Hermens, 2004).

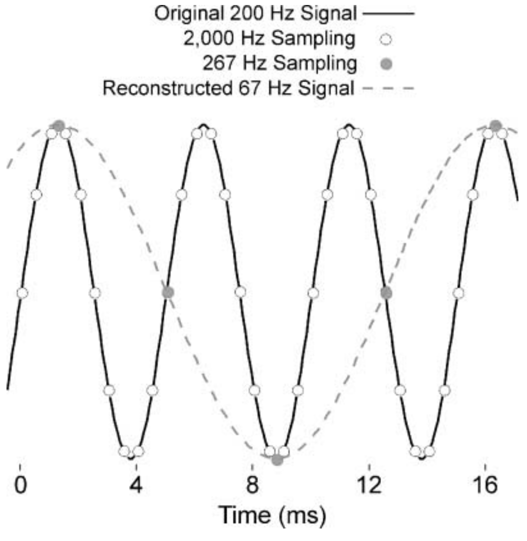
sEMG Signal Composition and Recording Recommendations

Recording EMG requires preamplification of the signal, hardware or software filters (hardware recommended), and an analog to digital (A/D) converter. Preamplification amplifies the original detected signal prior to A/D conversion. This is done to maximize the fidelity of the recorded signal. In active electrodes, preamplification takes place directly at the electrode head, allowing amplification to take place before the possible introduction of noise through electrical cables.

The majority of the power of a typical sEMG signal is in the frequency range 0–450 Hz (Merletti & Hermens, 2004). The remaining power at higher frequencies is mostly electrode and equipment noise, the sources of which are described in more detail in later sections (e.g., see the *Sources of Noise* section). For this reason, it is common to filter the sEMG signal prior to A/D conversion. Movement artifacts create signals in the 0–20 Hz range and can be attenuated by a high-pass filter with a cutoff around 10–20 Hz (De Luca, Gilmore, Kuznetsov, & Roy, 2010). Additionally, the sEMG signal should be low-pass filtered with a cutoff point around 500 Hz to remove high-frequency noise. Investigators differ in their opinions as to whether this filtering should be accomplished prior to digitization using hardware or postdigitization in software. Hardware filtering is preferred by some, as it allows the signal to be digitally acquired without the possibility of aliasing. Software filtering is preferred by others because it allows investigators to see the raw data, which may allow them to identify possible signal contamination that filtering may hide.

Given analog low-pass filtering with a cutoff of roughly 500 Hz (implemented prior to digitization), EMG data should theoretically be acquired with a minimum sampling rate of 1000 Hz to prevent aliasing. *Aliasing* occurs when an analog signal is undersampled, such that the reconstructed digital signal is distorted by high-frequency information from the original signal occurring in the low frequencies of the digitally acquired signal (see Figure 1). Aliasing can be prevented by using a sampling frequency at least 2 times greater than the highest frequency present in the original signal. Analog filtering prior to digitization allows attenuation of unwanted frequencies (> 500 Hz), effectively preventing aliasing for appropriate sampling frequencies. Analog filter cutoffs indicate a transition between the passband and stopband, but they are not infinitely steep: Even after filtering, some energy may remain in the signal at higher frequencies. Thus, to ensure signal integrity of sEMG low-pass filtered at 500 Hz, data acquisition of at least 2000 Hz is recommended. If low-pass filtering is to be accomplished after digitization, the risk of aliasing high-frequency noise to signal-bearing lower frequencies is vastly increased. *Oversampling* can be used to avoid aliasing; this refers to the technique of using sampling frequencies many times higher than the hypothetical Nyquist frequency. For instance, given that the majority of energy in sEMG is less than 500 Hz, investigators may choose to oversample the signal by sampling at 10000 Hz. However, electrical activity other than sEMG may be present in the signal, such that the total frequency profile in the measured signal is usually not under experimental control. For this reason, digitization prior to antialiasing filtering is always risky, even with oversampling, and therefore is not recommended.

Figure 1. Example of potential signal aliasing with a sampling frequency less than twice the signal frequency. The original 200-Hz signal is shown in the solid black line. Sampling at 2000 Hz (unfilled circles; sampling period of 0.5 ms) would provide information necessary to reconstruct the original signal. Sampling at 267 Hz (filled circles; sampling period of 3.75 ms) results in a reconstructed signal (dashed gray line) that is equivalent to a 67-Hz signal. Thus, undersampling can result in aliasing of signal energy from the original signal frequency to a lower frequency.



sEMG Signal Parameters

The raw acquired sEMG signal is often referred to as the *interference sEMG* or the *interference pattern*. The interference sEMG can be used itself to provide gross information about muscle activity; however, a variety of parameters can be estimated from the raw signal that are commonly used to gain more reliable and specific information.

Amplitude Estimation

The overall amplitude of sEMG is of interest to researchers as the sEMG amplitude generally increases with increases in muscle activation and force. However, this relationship is typically not linear and is affected by a number of factors such as muscle length and fatigue (Disselhorst-Klug, Schmitz-Rode, & Rau, 2009).

Although observation of interference sEMG can provide information about the amplitude of sEMG, the peak amplitudes seen in the raw signal should not be used to estimate amplitude because they can be due to a single motor unit and may not be representative of the overall activity in the detection area. Commonly used amplitude estimators are the average rectified value (ARV) and root-mean-square (RMS; see Equation 1). The sEMG signal is generally well described by a Gaussian distribution, leading to the assumption that the preferred estimator of sEMG amplitude may be RMS, which has a smaller variance when predicting amplitude for Gaussian distributions (Clancy & Hogan, 1999). However, results using experimental data indicate that the ARV may provide smaller variance for amplitude estimation (Clancy & Hogan, 1999). Both techniques require the use of time windowing. When choosing a time window, there is a trade-off between time sensitivity and reliability of the estimate; smaller windows are more sensitive to rapid changes in the signal but inherently result in less reliable amplitude estimates. SENIAM recommends windows of 250–500 ms for contraction levels above 50% maximal voluntary contraction (MVC), or 1,000–2,000 ms for contraction levels below 50% MVC for amplitude estimation. These window lengths are far too great to provide information about dynamic movement. However, in previous work, Norman and colleagues examined the coefficient of variation of sEMG of the biceps integrated over varying window lengths and found that integration times greater than 75 ms resulted in greater reliability (Norman, Nelson, & Cavanagh, 1978). Thus, optimal window lengths will vary on the basis of task, and investigators must choose a compromise between time sensitivity and the quality of the amplitude estimate. Speech articulatory movements tend to be fast with brief bursts and interburst intervals; thus, depending on the specific task and the goal of the amplitude estimation, window lengths of 75–1,000 ms are generally recommended. Before any amplitude estimation technique is applied, investigators should remove any direct current (DC) offset from the signal such that the mean of the raw signal is zero.

$$\mathrm{RMS} = \sqrt{\frac{1}{N}\sum_{i=1}^{N} x_i^2} \tag{1}$$

Frequency Content

The frequency content of the sEMG signal is dependent on the individual frequency characteristics of the constitutive MUAPs as well as their relative distance from the recording electrodes. Individual MUAPs have a specific size and shape, which determine their frequency characteristics. Also, as the recruitment of the motor units changes, rate coding determines the number of MUAPs per second. As muscle excitation increases, more individual MUAPs will sum due to a larger number of recruited motor units and an increase in the rate of MUAPs from each individual motor unit. The frequency content of the EMG is affected by both factors.

If individual MUAPs can be detected from the sEMG signal, two parameters of interest are the firing rate of individual MUAPs and the mean value of the firing rates of the MUAPs detected (Basmajian, 1978). The frequency spectrum of sEMG may also provide information about muscle activation. One of the most commonly used frequency-based measures is the median frequency. The median frequency of the sEMG signal has been shown to strongly correlate with localized muscle fatigue in a variety of physiological systems (e.g., Lindstrom, Kadefors, & Petersen, 1977) and has been assessed in the speech/voice system using intramuscular EMG (Boucher, Ahmarani, & Ayad, 2006) and sEMG of facial musculature (van Boxtel, Goudswaard, van der Molen, & van den Bosch, 1983). The median frequency is the point at which the spectral power of the signal is equally divided into low- and high-frequency halves. As muscles fatigue, the median frequency of the power spectrum shifts to lower frequencies.

Although it is affected by the degree of volume conduction and size of muscle fibers, frequency content is related to firing rate and can be used to assess information about the coactivation of multiple muscles. Muscle is thought to be driven by a number of different physiological oscillations at varying frequencies (see Grosse, Cassidy, & Brown, 2002, for a review), and these oscillations may be characteristic of the function of distinct neural circuits. Frequency bands such as alpha (8–13 Hz), beta (15–35 Hz), and gamma (30–70 Hz) have hypothesized sources within the central nervous system. *Coherence* is a frequency domain measure of the linear dependency or strength of coupling between two processes (e.g., Halliday et al., 1995) and can be used to capture these physiological oscillations. The coherence function, $|R_{xy}(\lambda)|^2$, can be defined as in Equation 2 below, where f_{xx} represents the autospectra of a time series x(t), f_{yy} represents the autospectra of y(t), f_{xy} represents the cross-spectra x(t) and y(t), and λ represents the frequency of interest.

$$|R_{xy}(\lambda)|^2 = \frac{|f_{xy}(\lambda)|^2}{f_{xx}(\lambda)f_{yy}(\lambda)}. \tag{2}$$

Coherence between multiple EMG signals (intermuscular coherence) can be used to measure the common presynaptic drive to motor neurons (Brown, Farmer, Halliday, Marsden, & Rosenberg, 1999) but has not yet been widely adopted in speech research. Smith and Denny have provided the most information about intermusuclar coherence during speech production (Denny & Smith, 1992, 2000; Smith & Denny, 1990), characterizing the activation of the chest wall and masseter muscles during a variety of speech, chewing, and breathing tasks (Smith & Denny, 1990). They further examined the intermuscular coherence of lip, jaw, chest wall, and anterior neck musculature of persons who stutter during speech and speech breathing (Denny & Smith, 1992, 2000). Goffman and Smith (1994) further investigated coherence between different quadrants of the perioral region, finding a lack of functional coupling during speech and chewing tasks. More recently, anterior neck intermuscular coherence has been used in the study of hyperfunctional voice production (Stepp, Hillman, & Heaton, 2010, 2011).

Raw Versus Normalized Amplitude

As mentioned, the tissues separating signal generation and signal detection have the effect of low-pass filtering the sEMG signal such that increases in the separation result in detection of a smoothed signal with lower amplitude (Farina & Rainoldi, 1999). Increases in so-called skin fold thickness can decrease the selectivity of the sEMG signal and result in more rapidly attenuated signals (De la Barrera & Milner, 1994). In fact, models of varying levels of complexity have shown that a majority of the amplitude of the sEMG signal can be lost with skin fold thicknesses from 0.1 to 10 mm (e.g., Andreassen & Rosenfalck, 1978; Kuiken, Lowery, & Stoykov, 2003). Thus, for some body areas of some participants, signal amplitudes are effectively too small to measure with reasonable S&R. Calipers should be used to determine fat layer thickness of participants to ensure that appropriately strong signals can be measured.

Because small differences in submental fat can so greatly change the amplitude of the sEMG signal measured, the raw amplitude of the signal is not a reliable measure among multiple participants or even as a function of time in a single participant. Thus, sEMG signals should be normalized to some reference contraction before they are compared between conditions and/or participants to reduce the variability caused by differences in surface electrode contact and submental fat levels (Netto & Burnett, 2006). Common references include MVC or some percentage of the MVC (usually 50% or 60%). Studies have shown that submaximal contractions are more reliable for simple, one-joint systems (Allison, Marshall, & Singer, 1993; Yang & Winter, 1983). However, Netto and Burnett (2006) found that for anterior neck musculature, the MVC reference was more reliable both within a given day and between days and speculated that this is likely due to the complex structure and synergistic action of neck musculature. Reference contraction recommendations for muscles of the face and thorax are not presently known.

Unfortunately, identifying tasks to induce MVC for speech musculature is not straightforward. In the anterior neck, maximal neck contraction against manual resistance measured by a dynamometer fit with a chin guard has been used with success (Stepp, Heaton, et al., 2011). Manual resistance of this type is not as well suited for some other speech musculature such as the muscles of respiration, for which more physiological tasks should be devised to prompt MVC behaviors.

MVC normalization at best reduces the effects of differences in skin fold thickness. Great care should be taken when comparing measures of sEMG amplitude across populations or even within single subjects as a function of time because normalization itself may introduce uncertainty due to a lack of reliability in the reference contraction itself: Participants' perception and production of maximal effort are easily affected by their environment and other motivating factors. However, it is imperative to attempt proper normalization prior to making comparisons of sEMG amplitudes in order for amplitude estimates to have physiological meaning.

Potential Sources of sEMG Signal Degradation

Sources of Noise

sEMG signals can be degraded by several types of noise. In the following section, the most common sources of noise, how to recognize the noise sources, and how to minimize their effects are reviewed. When recording sEMG, signals should always be monitored in real time to ensure signal integrity.

Perhaps the most common source of noise in the sEMG signal is power line interference. *Power line interference* is noise resulting from the alternating current used to power electrical devices and is primarily at the line frequency of either 50 or 60 Hz, plus associated harmonics (at integer multiples of the line frequency). This noise is typically larger than the sEMG signal in magnitude. Power line interference in the sEMG signal can be vastly reduced by using active electrodes, proper grounding, differential recording configurations, and properly shielded cables. In addition, performing experiments in environments with limited electrical noise can also reduce the effects of electrical line noise.

Poor signal integrity (low signal-to-noise ratios) due to high levels of subdermal fat can exacerbate contamination of the signal with electrical line noise. As an example, Figure 2 shows sEMG collected from the anterior neck during speech production from a young individual with minimal subdermal neck fat and from a middle-aged participant with significant levels of subdermal fat. Although troublesome, power line interference is typically easy to identify, both in the time and frequency domains (see Figure 2). Postprocessing with notch filters to remove 50 or 60 Hz interference is not advisable, as the power density of the sEMG signal is high in this range, and the associated phase rotation can be introduced to the time waveform (Hermens et al., 1999).

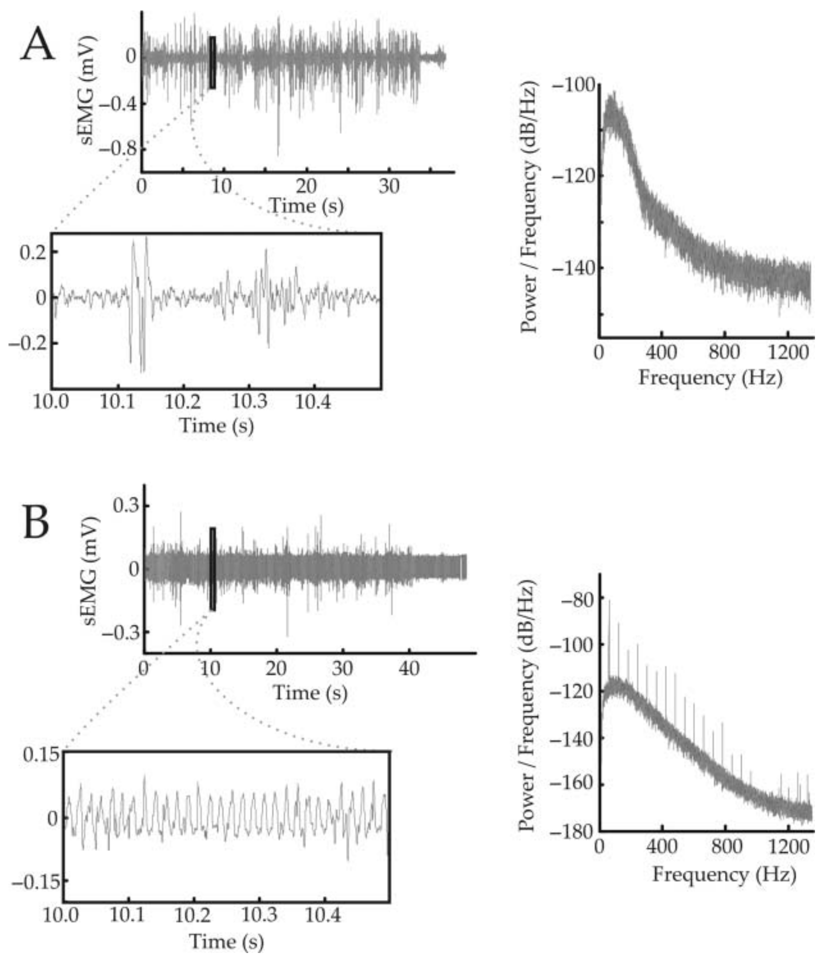
Low-frequency artifact can contaminate the sEMG signal as a result of movement between the electrode and skin or as a result of cable sway. Cable sway is less of an issue with active electrodes because the signal is amplified prior to entering the cable. Movement artifacts can be reduced by maintaining good adhesion of electrodes, using appropriate skin preparation techniques and high-pass filtering. Comprehensive testing of the relationship between high-pass filter corner frequency and movement artifact has resulted in the general recommendation of a Butterworth filter with a corner frequency of 20 Hz, with a 12-dB/oct slope as for general use (De Luca et al., 2010). This result is consistent with the only work in this area specifically using speech musculature, which suggested optimal high-pass cutoff frequencies between 15 and 25 Hz (van Boxtel, 2001).

In brief, the power spectrum of the sEMG signal has a specific and fairly consistent profile, whereas sEMG that has been contaminated with noise from motion artifacts and power line interference shows distinct changes to that power spectrum. Even if all precautions for proper recording are attempted and online monitoring of the time waveform does not reveal contamination, before interpreting a signal, one should check for data quality by computing the spectrum.

Effects of Innervation Zones

Neuromuscular junctions are often concentrated in a strip referred to as the *innervation zone* (or *active zone*).

Figure 2. Surface electromyography (sEMG) recordings illustrating high signal-to-noise ratio (A) and poor signal-to-noise ratio (B) with contamination from 60 cycle noise. A: Anterior neck sEMG from a young healthy participant with minimal subdermal neck fat during reading aloud. Left plots show sEMG as a function of time. Right plot shows the power spectral density the sEMG using Welch's method. B: Anterior neck sEMG from a middle-aged participant with significant levels of subdermal fat, leading to a poor signal-to-noise ratio and contamination of the signal with electrical line noise. Electrical line noise is apparent in both the power spectral density (as peaks of energy at 60 Hz and its harmonics) as well as the time trace of the sEMG at close time intervals.



Although the innervation zone is often idealized as a single strip, many muscles have decentralized (diffusely localized) neuromuscular junctions. It is generally recommended that differential electrodes be applied between the innervation zone and a tendon. In the past, sensors have been instead placed over the center of the muscle (the belly) or over the innervation zone (motor end plate zone), as this was the best location to record "large" monopolar sEMG signals. It is now known that this location is not suitable for differential recordings; it is not stable or reproducible because relatively small displacements of the sensors with respect to the innervation zone cause large effects on the amplitude of the sEMG signal (Merletti & Hermens, 2004), particularly on the low-frequency (< 110 Hz) components (Beck et al., 2009). This is because the MUAPs propagate in both directions away from the innervation zone toward the tendons. If bipolar electrodes are arranged on each side of the innervation zone, there is likely to be a substantial level of cancellation of common signals, reducing the amplitude of the recorded signal. In fact, differences in placement along the body of a muscle have been known to mask differences in task conditions (Mercer, Bezodis, DeLion, Zachry, & Rubley, 2006). Thus, for sEMG signals to be as accurate and repeatable as possible, there must be a clear definition of electrode position relative to the innervation zones (Hermens et al., 1999). When the locations of innervation zones are unknown, use of double differential electrodes can reduce the effects of an ill-placed sensor (Farina, Merletti, & Disselhorst-Klug, 2004).

Ideal sEMG recording procedures would first identify the innervation zones and find the optimal electrode position on a subject-by-subject basis using multichannel electrode arrays. Falla, Dall'Alba, Rainoldi, Merletti, and Jull (2002), for example, examined the SCM muscles in this way in 11 healthy individuals in order to determine specific recommendations for electrode placement to optimize sEMG recordings (Falla et al., 2002). Recommendations of this type are not presently available for most speech musculature, but some muscles have been studied. One group has examined innervation zone locations for the muscles of jaw elevation (Castroflorio et al., 2005); they found a large variability within and between participants in the location of major innervation zones of the masseter and could not recommend an optimal position for electrode placement. Conversely, Lapatki and colleagues (2006) found distinct clusters of motor endplates in the depressor anguli oris, depressor labii inferioris, mentalis, and orbicularis oris inferior muscles across participants when topographical locations were spatially warped to correct for anatomical differences between participants. Similarly, the geniohyoid has also been found to have clustered motor endplates, but with distinct clusters located in separate compartments (Mu & Sanders, 1998).

When examining speech musculature, it may not be possible to completely avoid the effects of innervation zones. However, researchers and clinicians should be aware of the possible effects of innervation zones on the resulting signal and should take steps to avoid them when possible, such as using double-differential electrodes.

Cross-Talk

Although researchers and clinicians are often interested in detecting activation from isolated muscles, muscles typically do not act in isolation. In the case of speech musculature, the detection volume of sEMG is often large enough to detect activity from more than one muscle at once, which is referred to as *cross-talk*. The sensor will detect from the nearest muscle as well as those adjacent to it. When multiple signal sources are available, the sEMG interference pattern will contain elements from all sources with the largest amplitudes coming from the sources closest to the sensor.

The effects of cross-talk are minimized by decreasing the detection volume. This can be accomplished by using smaller electrodes with smaller interelectrode distances and double-differential configurations (Koh & Grabiner, 1992, 1993). In the case of recording speech musculature with sEMG, cross-talk may be unavoidable. If isolated muscle activations are of interest, sEMG may not be appropriate. Unfortunately, due to the small size of some muscles and regional differences in activation, even signals recorded using intramuscular electrodes can contain crosstalk (e.g., Blair & Smith, 1986).

Special Considerations for Speech and Swallowing Anatomy

Scientists and clinicians wishing to measure sEMG from speech and swallowing anatomy have special obstacles to overcome in data collection. These muscles are often small and have overlapping fibers. Thus, it is often not possible to isolate the activity of single muscles of speech and swallowing. However, sEMG can still provide a valuable tool to better understand and assess speech and swallowing physiology as long as the limitations in recording are well understood prior to interpretation of data. The following sections contain a review of the work accomplished in the area of speech and swallowing anatomy that may guide investigators with specific recommendations for electrode placement, confirmatory tasks, and signal interpretation.

Orofacial Musculature (Muscles of Articulation and Mastication)

Despite their importance in understanding typical and disordered speech motor control, isolated activations of orofacial musculature are difficult to acquire. A notable exception to this issue would be the masseter muscle, which is large, simple to palpate, and located superficially. The masseter can be easily recorded using sEMG and shows comparable results when studied simultaneously by sEMG and intramuscular EMG (Koole, de Jongh, & Boering, 1991). In fact, both amplitude and frequency parameters of sEMG of the masseter have been shown to be reliable over multiple days (Suvinen, Malmberg, Forster, & Kemppainen, 2009). Similarly, bipolar sEMG over the zygomaticus major region found significant correlations in sEMG amplitude during specific facial poses recorded on different days (Tassinary, Cacioppo, & Geen, 1989).

The muscles of the lower face and submental area represent a greater challenge due to their small size and overlapping fibers. For instance, surface signals detected over the anterior digastric show differences in activation during isometric contractions from intramuscular recordings (Koole et al., 1991). However, work from Lapatki and colleagues (2006) has provided guidance for the muscles of the lower face: Distinct clusters of motor endplates and primary muscle fiber orientations have been shown in the depressor anguli oris, depressor labii inferioris, mentalis, and orbicularis oris inferior muscles, which can lead to more informed placement of electrodes. In their subsequent work, Lapatki et al. have used their methods to suggest optimal placements of bipolar electrodes for muscles of the lower face: depressor anguli oris, depressor labii inferioris, mentalis, and orbicularis oris inferior (Lapatki et al., 2010).

The work of Blair and Smith (1986) has shown that single muscles of human lips cannot be measured with current technologies because of interdigitation of muscle fibers. They concluded that this limitation of recording quality should be acknowledged during interpretation of recordings. Despite this limitation, application of sEMG to the perioral region has resulted in a substantial body of knowledge about the motor control of the muscles of the lips (e.g., Goffman & Smith, 1994; Wohlert & Goffman, 1994; Wohlert & Hammen, 2000). In fact, in this region, sEMG may offer as much discrimination as intramuscular electrodes. For instance, Lapatki and colleagues have developed a small sEMG electrode for use with facial musculature, which has been shown to have similar muscle selectivity as intramuscular recording techniques of the muscles of the lower face (Lapatki, Stegeman, & Jonas, 2003).

Understanding what tasks can activate specific musculature can aid in confirmation of an appropriate electrode position. Placement of electrodes over muscles of mastication such as the buccinators and masseter can be aided by asking participants to clench their teeth and then palpating to find the muscle body. Other smaller muscles can be more difficult. O'Dwyer and colleagues have suggested procedures for the verification of hooked wire electrode placement for a variety of orofacial and mandibular muscles, including gestures used as stimuli for confirmation of activation (O'Dwyer, Quinn, Guitar, Andrews, & Neilson, 1981). The speech-relevant muscles that could be detected with sEMG with satisfactory gestures for confirmation based on O'Dwyer et al. (1981) and Burnett, Mann, Cornell, and Ludlow (2003) are shown in Table 1. The gestures are known to elicit strong activation from each muscle, although not *isolated* activation.

Table 1. Gestures known to elicit strong (not isolated) activation from each muscle (O'Dwyer et al., 1981).

Muscle	Gesture
Levator labii superioris	Unilateral snarl elevating the upper lip
Zygomaticus major	Broad laugh
Buccinator	Puffing out the cheeks with the lips closed
Risorius	Broad smile with the lips closed
Orbicularis oris superioris	Compressing the upper lip against the upper incisors
Orbicularis oris inferioris	Compressing the lower lip against the lower incisors
Depressor anguli oris	Pulling down the corners of the mouth
Depressor labii inferioris	Pulling down the lower lip with the jaw closed
Mentalis	Raising and everting the lower lip while wrinkling the chin
Mylohyoid	Swallowing
Anterior belly of the digastric	Lowering jaw under resistance

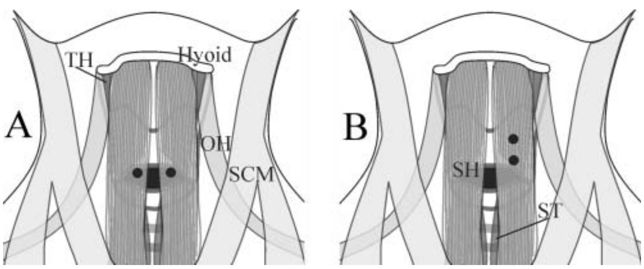
Submental and Anterior Neck Musculature

It is relatively easy to record from infrahyoid musculature using sEMG due to the prominent size and superficial location of the sternohyoid and omohyoid muscles. High-quality recordings can be obtained from the neck surface by placing single- or double-differential sEMG electrodes 1 cm lateral to the neck midline, and located from the gap between the cricoid and thyroid cartilages of the larynx and as far superior as the border of the submental surface. Varying the superior–posterior positions of electrodes can lead to some variation in the activity recorded. With a surface electrode placed directly over the gap between the cricoid and thyroid cartilages, previous authors have hypothesized that it is possible to record from the cricothyroid, an intrinsic laryngeal muscle; however, based on its relatively deep location, it is unlikely to contribute to surface recordings, and past work using simultaneous intramuscular EMG and neck sEMG has shown that surface recordings do not show evidence of CT activation (Loucks, Poletto, Saxon, & Ludlow, 2005). Signals detected at this location are likely largely composed of activations of the sternohyoid, and researchers interested in cricothyroid activation should use intermuscular EMG. More superior placements are more likely to also include activations of the omohyoid.

There has been considerable interest in recording from the sternohyoid muscle for voice and swallowing applications; however, the electrode locations and configurations used have not always been optimized. One suboptimal configuration that has been used is to use a bipolar recording configuration with each electrode located on opposite sides of the neck. This configuration results in a recording of the difference in activation between the two sides, which is essentially noise in the bilateral activation pattern. If a difference between the two sides is of interest, this should be examined after differential signals from the two sides have been recorded. The bipolar configuration has been developed to record reliably when electrodes are placed longitudinally to the body of the muscle (see example configurations in Figure 3). If the sternohyoid is of particular interest, spatial selectivity can be improved by using a double-differential electrode, which limits the contribution of deeper muscles to the detected signal.

Suprahyoid and submental musculature represent more of a recording challenge for using sEMG. When measuring from the anterior neck, the thyrohyoid is deep to the sternohyoid and, thus, likely does not

Figure 3. Diagram of neck muscles as seen from the front illustrating examples of bipolar electrode configurations. A: Incorrect bilateral configuration. B: Suggested configuration with electrodes placed parallel to the longitudinal axis of the muscle body, in line with the fibers of the muscle. TH = thyrohyoid; OH = omohyoid, SCM = sternocleidomastoid; SH = sternohyoid; ST = sternothyroid.



contribute substantially to surface recordings. Recordings of the submental surface have the potential to detect activation of the mylohyoid and the anterior belly of the digastric, but are more difficult in many individuals due to increased subdermal fat in this area. Another issue is the much smaller size of muscle bellies in this area and the overlapping fibers. Table 1 indicates gestures known to elicit strong activation of many of the submental muscles. Although it is arguably of limited importance during speech, the SCM is an accessory respiratory muscle, and is known to activate during speech and singing (Pettersen, Bjorkoy, Torp, & Westgaard, 2005). Due to its large size and superficial location, there is a large body of previous research on recording sEMG from the SCM. In particular, previous research has identified the common location of innervation zones in the SCM and recommended that electrodes should be placed one third of the distance from the sternal notch to the mastoid process, in the direction of the line from the sternal notch to the mastoid process in order to avoid placement near innervation zones (Falla et al., 2002).

A particular issue for individuals interested in measuring from the submental surface and anterior neck is contamination from the platysma. The *platysma* is a superficially located thin sheet of muscle in the subcutaneous tissue of the neck. It extends over the anterolateral aspect of the neck from the inferior border of the mandible to the superior aspect of the pectoralis major. Although the activation of the platysma during speech has been studied somewhat less than other laryngeal and orofacial musculature, it is thought to be active during speech production as an antagonist to the orbicularis oris inferior muscle (McClean & Sapir, 1980). In addition, the platysma is thought to be active in individuals during swallow, although this activation is not highly correlated with overall muscle activation during swallow and varies widely between individuals (Palmer, Luschei, Jaffe, & McCulloch, 1999). Although it is an extremely thin sheath of muscle, whenever active, the platysma will be a substantial source of activity detected at the neck surface due to its relatively superficial location compared with surrounding muscles.

Resistance against manual force elicits strong consistent activation of strap musculature for purposes of confirmation of electrode placement and MVC recordings (e.g., Stepp, Hillman, & Heaton, 2011). Resistance against manual force can be achieved by mounting an athletic chin guard or similar apparatus to a dynamometer to allow for simultaneous collection of force data. Collection of force data in concert with sEMG during MVC maneuvers can help investigators to improve reliability of multiday recordings.

Muscles of Respiration

Early work to study the muscles of respiration using sEMG examined the internal and external intercostals during forced respiration and simple speech tasks (Jones, Beargie, & Pauley, 1953). Eblen (1963) reviewed the subject of using sEMG to record speech-related respiratory muscle activity and suggested that sEMG was of limited use due to the inability to isolate activity of individual muscles. However, McFarland and Smith (1989) carefully assessed the ability to use sEMG to study primary and accessory respiratory musculature during speech and nonspeech tasks, finding that sEMG could be used to record respiratory-related activations from the rib cage during speech, but particularly during expiration. They placed electrodes in a bipolar configuration on the medial and lateral rib cage. The medial set of electrodes was placed on the seventh and eighth interspaces, 2 cm lateral to the midclavicular line with the goal of recording from the diaphragm and intercostals. The lateral set of electrodes was placed on the eighth interspace, straddling the anterior axillary line, also with the goal of detecting activity of the diaphragm and intercostals. Although expiratory activations could be consistently detected from the rib cage, they were not able to consistently record from the diaphragm or other muscles of inspiration during speech using electrodes placed on the chest wall. However, inspiratory-related activity was measured from the chest wall when participants were at higher lung volumes than those used during typical conversational speech.

Although sEMG can be used to record from abdominal muscles during respiration and speech (McFarland & Smith, 1989), this area is prone to significant levels of subdermal fat and thus degraded signal quality. McFarland and Smith recorded from the rectus abdominis by placing one electrode 2 cm from the midline and 3 cm superior to the umbilicus, with the second electrode 3 cm directly superior. They sampled activity from the internal and external oblique by placing electrodes over the anterior axillary line midway between the anterior iliac crest and costal margin, again using an electrode separation of 3 cm (McFarland & Smith, 1989).

Future Directions

Most of the research in sEMG has been accomplished in limb musculature. However, there is evidence to suggest that there are multiple differences in the anatomy and physiology between these muscles and the muscles of speech and swallowing. One example is the difference in discharge rate. Both genioglossus and sternohyoid musculature have been shown to have higher and more variable discharge rates than limb musculature (Bailey, Rice, & Fuglevand, 2007; Farina & Falla, 2009). Speech musculature is not load-bearing and differs in many ways from the more studied muscles of the upper and lower limbs. Additional research is needed to understand these differences and to more specifically characterize muscle electrophysiology of the speech and swallowing systems.

Before sEMG may be dependably applied clinically, evidence is needed for the reliability of repeated sEMG measurements. Although issues of reliability of sEMG have been investigated in other systems, there is still much work to do for speech anatomy to understand how methodological factors such as electrode placement affect the accuracy and reliability of the detected activity. Studies in the upper limb have indicated that the most prominent effect on intersession reliability of sEMG is electrode placement (Yang & Winter, 1983). Thus, standardized recommendations for electrode placement and orientation will increase the potential for reliable sEMG recordings of speech anatomy. Digital photography of electrode placement and careful reapplication using anatomical landmarks may also aid in the intersession reliability of these recordings. Use of even the highest quality commercial recording systems will result in inaccurate or unreliable data if sensors are improperly or inconsistently placed.

Current technology in common use does not allow isolated muscle recordings of speech musculature using sEMG. However, future research into flexible high-density arrays of electrodes may allow for noninvasive characterization of the behavior of isolated muscles. High-density sEMG in which 2D arrays of electrodes are used are becoming increasingly common in research (Merletti et al., 2009). These arrays consist of several closely spaced small electrodes and can be used to estimate the locations of innervation zones and muscle fiber orientations (e.g., Lapatki et al., 2010), and in concert with new signal processing techniques to decompose the sEMG signal into multiple single MUAP trains (e.g., Nawab, Chang, & De Luca, 2010). High-density electrodes could be of particular use to speech researchers, providing a potential solution to small, overlapping musculature in which innervation zones are unknown. Although not yet widely used, researchers have developed flexible high-density electrode grids, which have already been shown to be well suited to facial recording applications (Lapatki, Van Dijk, Jonas, Zwarts, & Stegeman, 2004).

Despite the challenges associated with recording from speech and swallowing musculature, the currently available sEMG still has potential as a clinical and research tool when used correctly. In a study examining the viability of sEMG recordings of lip muscles, Blair and Smith (1986) argue that even when isolated muscles cannot be recorded, sEMG can be successfully used for between-condition or between-group comparisons when consistent electrode placements are used. Even with its limitations, sEMG can be used to great effect to provide assistance and rehabilitation for individuals with disordered speech. For example, even without an attempt to isolate individual muscle groups, multichannel sEMG can provide successful recognition of mouthed speech (subvocal) in both healthy speakers and individuals with dysarthia (Deng et al., 2009; Meltzner et al., 2008). If reviewers and consumers of sEMG literature demand high-quality methodology and reporting, we will begin to achieve the full potential of this tool and truly affect patient care through improved assessment and rehabilitation using sEMG.

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