Muscle Sound: Bases for the Introduction of a Mechanomyographic Signal in Muscle Studies

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ABSTRACT: Muscular sound is a mechanical phenomenon detectable at the surface of an active muscle, which has been known and described since 1800. Only recently, because of the availability of reliable transducers and sophisticated analysis techniques, has this signal become attractive for monitoring the mechanical aspects of muscle contraction. The muscular sound characteristics were investigated both during electrically elicited and voluntary contractions. In the first case, the influence of the biophysical and mechanical properties of the muscle on this signal was studied. During voluntary efforts the summation of the mechanical activity of the recruited motor units was analyzed. The results indicate that the muscular sound may be an adjunct tool to the force, the physiological force tremor, and electromyogram to obtain information on the muscle mechanical model as well as on muscle motor control.

This review focuses on the following aspects of the signal: (1) recording problems; (2) muscle sound properties during stimulation of isolated and *in vivo* muscle; (3) time and frequency domain analysis during non fatiguing and fatiguing contractions; (4) comparison with other signals related to muscle activity; (5) discussion on the origin; and (6) possible practical applications.

KEY WORDS: acoustic myography, mechanomyogram, muscle sound, soundmyogram, vibromyogram, muscle activity, motor unit recruitment, muscle fatigue, muscle mechanics.

I. INTRODUCTION

> "vera itaque ratio experimenti praedicti est, quia in digito et brachio totoque corpore continuato fiunt multimotus ac tremores, ob spiritum agitationem huc illuc perpetuo accurrentium"

> "the explanation of the previous experimental result is that, because of the continuous hurried motion of the spirits, tremors and movements occur in finger, arm and in the whole body".

This was the first interpretation, given by Grimaldi in 1665 and cited in Reference 74, of the sound that a subject can hear when he stops his ears with his thumbs and clenches his fists.

Before the twentieth century the muscle sound was only sporadically studied. Wollaston[74] in 1810 described the phenomenon and suggested to relate it with the

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existence of "a great number of contractions repeated at extremely short intervals" per each muscle effort, with the tone of the sound dependent on the frequency of the contractions. In 1860 Collongues (cited in Reference 16) described the "digital buzzing" detected by means of a dynamoscope. The author noted that the properties of the buzzing were influenced by the age, the sex, and the presence of illness. In the same period Helmholtz[35] suggested that the muscular sound was related to quasi-periodic oscillations of the muscle fibers. The similarity of the muscle sound during voluntary contraction and electrically elicited contraction, for a stimulation rate below 30 Hz only, was emphasized by Herroun and Yeo.[37] Moreover, these authors denied the existence of the muscular sound per se, but suggested that the "regular or irregular motions" during muscle contraction induce the membrana tympani to resonate at its own frequency, providing the consequent auditory perception. Even if a rather incomplete description of the phenomenon can be found in the papers of the last century, mainly because of technological impairment in signal detection and analysis, two fundamental points were already understood: *muscle sound is related to muscle activity and its properties are related to the properties of the contraction.*

On this basis the study of muscle sound grew in this century, in particular after 1980, in order to clarify its reliability and its role in the investigation of muscle physiology. This task was made possible because of the availability of electronic sensors (piezoelectric transducers, condenser microphones, piezoceramic membranes, or accelerometers) and of computerized signal processing techniques (both in time and frequency domain). Actually the degree of knowledge of muscle sound has begun to provide notions about many specific aspects of the signal.

Although the terminological problem will be considered in Section VI, it is necessary to underline here, in order to avoid confusion and misunderstandings, that, at this time, the signal detected at the muscle surface has been given different terms by different authors. These terms are acousticmyogram, phonomyogram, or soundmyogram. All of them refer to the recording by a "myograph" of the sound emitted by the active muscle. Moreover, some authors use the term vibromyography, intending the recording of the vibrations generated, within the muscle, by the active fibers. These vibrations may be considered as the source of the skin oscillations detected as sound by air-coupled and contact microphones or as mechanical events by accelerometers. In this review the different terms will be used according to the form chosen by the authors in the cited papers.

II. RECORDING MUSCLE SOUND

A. Technical Problems

The isolated frog muscle, mounted in a sound-insulated chamber, was an extensively used experimental setup for studying muscular sound from the biophysical point of view (see Figure 1). The transducers used for the detection of sound pressure waves in the baths were hydrophones.

During *in vivo* recordings of muscular sound, different transducers have been used. The most important feature of the recording apparatus is the frequency response of the transducer. In particular its low frequency cut-off has to be around 1 to 2 Hz.[19] The upper cut-off has to be chosen so that the greater part of the power is well below 100 Hz.[12,14,19,32,42,49,58,60,72] An important guideline for the choice of transducer concerns the ratio between its mass and that of the muscle investigated. The transducer mass must not induce distortion in the muscle surface. From this point of view the use of very light accelerometers (0.5 g, Entran EGAY-25D or Dytran 3115 a) can be the better solution, allowing, as a consequence, the study of very small muscles.[13,14,75] Another advantage in the use of accelerometers is that the measurement is made in "physiological units (m/s^2) rather than in transducer-dependent units" (mV).[13] The use of larger piezoelectric contact sensors (HP 21050-A, about 44 g)[19,56–63,68] is allowed for muscles with greater mass. It is important to note that piezoelectric contact sensors vary their response with the force applied for mechanical coupling to the muscle surface. The recommended applying force is below 200 g.[12,60] Under these conditions the transducer frequency response was flat in the bandwidth of interest for the soundmyogram (SMG). The minimal applying force has to avoid any relative movement of the device with respect to the muscle surface.[68] On these bases care must be taken when the sensor or the microphone is "strapped" over the muscle. If the band used is rigid, too much pressure may act on the sensor during contraction, particularly at high levels of effort. The use of a compliant band for the piezoelectric contact sensor or of a

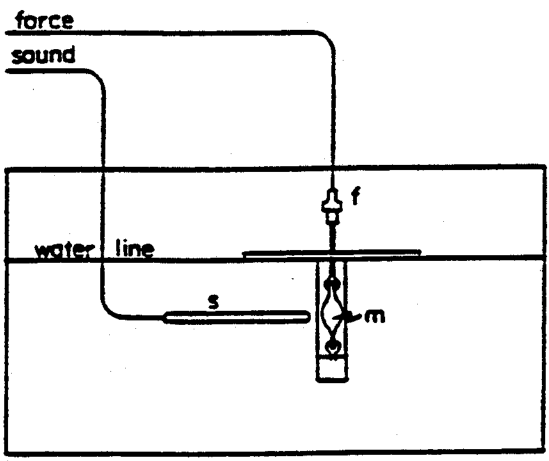

FIGURE 1. Apparatus set up for soundmyogram recording from isolated muscle. m, muscle; f, force transducer; s, hydrophone. (From Frangioni, J. V., Kwan-Gett, T. S., Dobrunz, L. E., and McMahon, T. A., *Biophys. J.,* 51, 775, 1987. With copyright permission of the Biophysical Society.)

double-stick tape for light mass accelerometers is strongly recommended. The use of an omnidirectional electret microphone with flat response between 2 and 90 Hz has been suggested.[32] In this case a closed-cell foam containing the microphone was attached by means of surgical cement. Thus, problems due to possible compression of the muscle during contraction were avoided. The only drawback in using wideband air-coupled microphones may be related to the formation of a skin diaphragm oscillating in a chamber and transmitting the muscle transverse mechanical activity, through an air column to the sensor. This may contribute to some distortions in the detected signal. However the muscular sound from thenar eminence was not different when detection was made by an air-coupled microphone or a piezoelectric contact sensor.[19] On the contrary, great differences in signal shape are evident when signals from accelerometers and other devices are compared (Figure 2). This may be due to their peculiar characteristics and to the diverse ways used to fix the transducers to the muscle.

When the goal is a comparison between electromyogram (EMG) and SMG, a correct device for the detection of both signals is needed. The first description of such a tool was made by Gordon and Holbourn.[33] More recently a composite probe[1] able to pick up both signals was developed, coupling a piezoceramic transducer with a printed circuit carrying three conductive traces, i.e., ground (centrally) and two exploring strips. The frequency response of the sensor was not indicated. A composite probe was developed by our group and has been used since 1988 in our experiments. Figure 3 shows the common piezoelectric transducer with an isolated plate carrying two silver bars 1 cm long and 1 cm spaced for EMG detection.

B. Physiological Problems

Cross-talk is a well-known problem when surface EMG spatial resolution is estimated. The contamination of the EMG from one muscle by the contemporary activity of neighboring (agonists or antagonists) muscles is not negligible. The same phenomenon may take place during recording of the muscular sound. Only one paper has dealt specifically with this problem, recording sounds from the biceps brachii and from the triceps.[73] The task was to support 2.3 kg, keeping as constant as possible the 90° angle between arm and forearm, allowing the forearm the freedom to move. The two contact sensors were fixed to the muscle by adhesive tape. Similar frequency content and a "mean amplitude ratio" of .46 were found by the comparison of the signals from the two sensors. The mechanical coupling between the two muscles probably operated by the adhesive tape, used to fix the contact sensors, as well as the transmission of forearm oscillations through the elbow joint to the whole arm segment, may have enhanced the consistency of the phenomenon. The cross-talk between the extensors and the flexors of the wrist was judged of little importance when the SMG was used for practical application,[5] see Section VII.

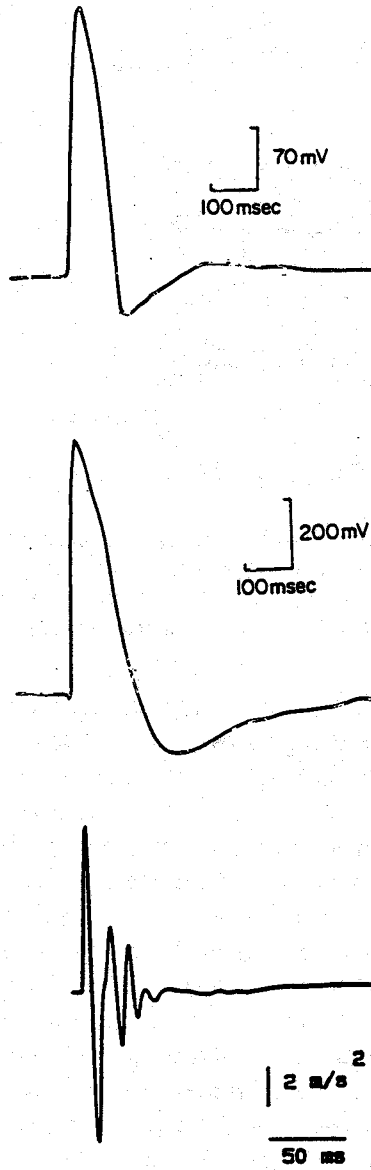

FIGURE 2. Single supramaximal twitches sound pressure waves from air-coupled microphone (upper), piezoelectric contact sensor (middle), and accelerometer (bottom). Differences in the waveforms may be related to the different properties of the transducers as well as their diverse mass, maybe affecting the muscle dynamics to a different extent, and finally to the way in which they have been mechanically coupled to the muscle. (Upper and middle panels from Bolton, C. F., Parkes, A., Thompson, T. R., Clark, M. R., and Stern, C. J., *Muscle Nerve,* 12, 126, 1989. Lower panel from Barry, D. T., *EMG Clin. Neurophysiol.,* 32, 35, 1992. With permission.)

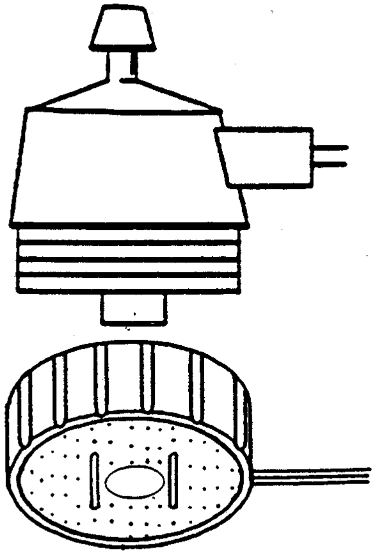

FIGURE 3. The modified piezoelectric contact sensor (HP 21050 A) for contemporary recordings of SMG and surface EMG. The two silver bars were 1 cm spaced and close to the hole allowing the tip of the sensor to reach the muscle surface.

Another important factor that may affect the characteristics of the recorded muscle sound pressure wave is represented by the features of the tissue between the muscle and the transducer. Indeed "the magnitude of the mechanical vibrations is determined by the net force acting upon the tissue between the muscle and the skin surface and the mechanical properties of this layer of tissue".[41] In other words, this tissue layer may act as a low pass filter for the mechanical waves traveling from the muscle to the skin surface. A proof of the existence of this problem is given by the comparison of the amplitude of the recorded sound from the muscle belly and from the fascia. In the first case greater peak-to-peak amplitude during electrically elicited muscle sound from thenar muscles[19] and greater value of the integrated acousticmyogram during voluntary isometric contraction of the quadriceps[70] were reported. On these bases future studies comparing the signal recorded directly at the muscle surface with the signal detected over the skin are needed to settle the filtering role of the interposed tissue on the muscle sound time and frequency domain properties.

III. ELECTRICALLY ELICITED CONTRACTIONS

A. Isolated Muscle

During muscle shortening of about 20 to 30% of its resting length, "discrete sound bursts" were recorded.[20] A piezoelectric transducer was used for sound detection. The burst duration was in the order of 400 μsec and the interburst interval was influenced by the saline bath temperature. According to the AA[20] the fact that the recorded sound was a sequence of bursts, instead of a continuous tone, "indicates that contraction may occur in a discrete, synchronous manner" following a sequence of high-velocity shortening periods (steps) and constant sarcomere length periods as described by optical techniques.

On these bases the recorded sound may be due to the brief shock waves generated by the synchronous stepwise changes of the radius of several active muscle fibers, i.e., by their thickening, as previously suggested.[33]

This conclusion has to be considered with caution. In fact an important drawback in considering the phenomenon described by Brozovich and Pollack[20] as muscular sound, and therefore in including their paper in the muscle sound bibliography, can be found in Barry's work.[6] In the Introduction a personal communication to the author by F. V. Brozovich (1985) suggests that the recorded bursts may have been due to sliding steps between different muscle fibers. As a consequence these bursts seem not related to the low-frequency sounds reported by Gordon and Holbourn.[33]

During isometric contractions the muscle tendons were sutured, as usual, to a fixed post and to a force transducer. The muscle nerve supply was intact, allowing muscle indirect electrical stimulation.[6,7,10,26,29,61] The overall experimental setup is similar to the one represented in Figure 1. In Figure 4 the acoustic signal recorded during a single twitch is reported. It can be noted that the sound begins after the compound muscle action potential but before the development of tension.[6,29] The ringing sound is nearly completed by the time force has risen to its maximum level;[9] moreover, the larger sound pressure is reached on the second half cycle.[6,29] Only for a particular azimuthal orientation of the hydrophone in respect to the muscle long axis, i.e., "when the muscle is oriented with the primary mode of lateral movement aligned with the sound transducer",[10] can the first half cycle be the largest.[29] After the maximum pressure peak the amplitude of the half cycles decays gradually.[6,29] As a consequence, in the sound recorded during a single twitch an initial increase in amplitude followed by decaying oscillations may be noted.[6]

When stimulated at frequencies generating an unfused tetanus, the amplitude of the recorded sound was proportional to the force ripple[6,29] (see Figure 5, upper panel). At tetanic stimulation frequencies the sound presented a single event after the first stimulus and its amplitude was proportional to the force increment. This correlation was verified also for the stimuli after the first one. In fact throughout the tetanic stimulation, when extremely low force changes occur, the sound pressure did not differ from the resting values[6,29] (see Figure 5, lower panel).

1. Sound Propagation

The investigation of the "pattern of the pressure field"[29] can be made by changing the relative position of the hydrophone with respect to the contracting muscle. When the transducer is moved along the major axis of the muscle, the

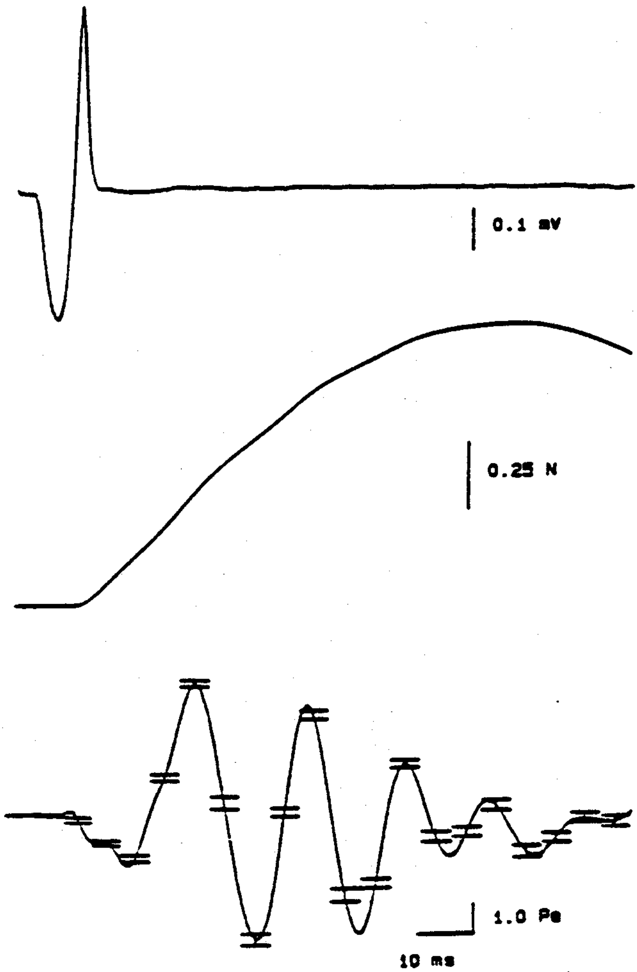

FIGURE 4. Recordings of compound muscle action potential (top), force (middle), and acoustic (bottom) signals. The traces correspond to the average of ten consecutive stimulations. The error bars are reported for the sound signal. Note that the ringing sound is completed by the time force has risen to its maximum. (From Barry, D. T., *Biophys. J.*, 51, 769, 1987. With copyright permission of the Biophysical Society.)

amplitude of the first half cycle of the recorded signal is maximum at the muscle belly and decreases towards the tendon insertions.[29]

If two hydrophones are 180° apart on the transverse plane of the muscle, the recorded sound pressure signals are 180° out of phase.[6] This means that the hydrophone orientation around the long axis of the muscle influences the first half cycle amplitude. If the maximum positive pressure is detected at an arbitrary angle of 180°, the maximum negative one is detected at 0°. The signal tends to zero at 90° and 270°.[26,29] These results suggest that *lateral movement,* towards a direction and away from the opposite one, can be implied in sound generation.

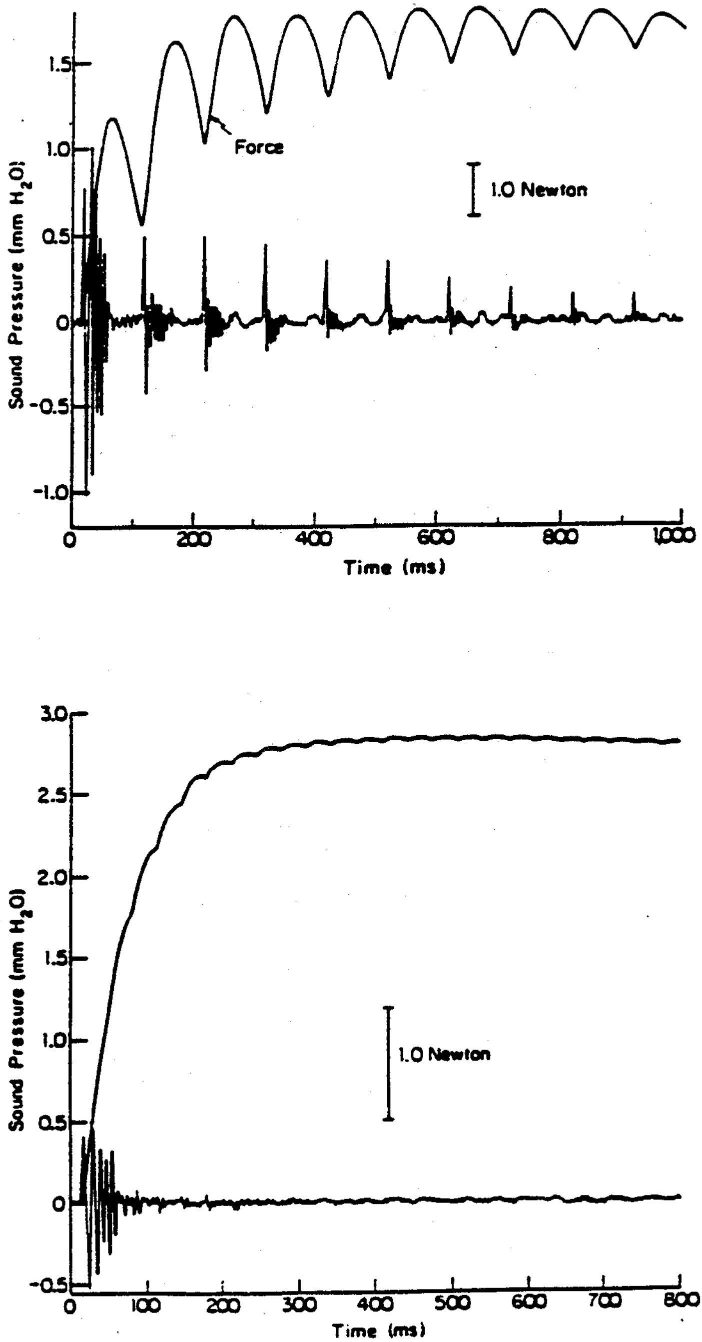

FIGURE 5. The direct relationship between the force ripple and the sound amplitude is shown. At tetanic stimulation frequency the sound occurs only at the first stimulus. (From Frangioni, J. V., Kwan-Gett, T. S., Dobrunz, L. E., and McMahon, T. A., *Biophys. J.*, 51, 775, 1987. With copyright permission of the Biophysical Society.)

2. *Influence of the Muscle Length*

The length of a muscle affects the ringing sound. In fact, during single twitches, the sound peak-to-peak amplitude vs. length relationship is similar to the classic length-tension relationship. As shown in Figure 6, the length at which the sound is maximum is less than L_0.[6,26,29] The sound reduction for muscle lengths >90% L_0 is consistent with the lateral movement sound etiology. In fact, this movement is reduced in those situations in which the muscle internal compliance is reduced, as during stretching. For lengths <90% L_0 the sound amplitude reduction may be attributed to changes in the muscle-microphone system geometry altering the sound energy transmission.[6]

The longer the muscle the higher the number of the half cycles.[6] At lengths near the plateau of the length tension curve the number of the elicited half cycles varies from 7 to 15.[29]

3. *Influence of Temperature*

The duration of the acoustic signal in respect to the force twitch is dependent on the temperature. Indeed, at 7°C the sound event is completed "by the time the force twitch had risen to 50% of the twitch force", while at 25°C it is still present

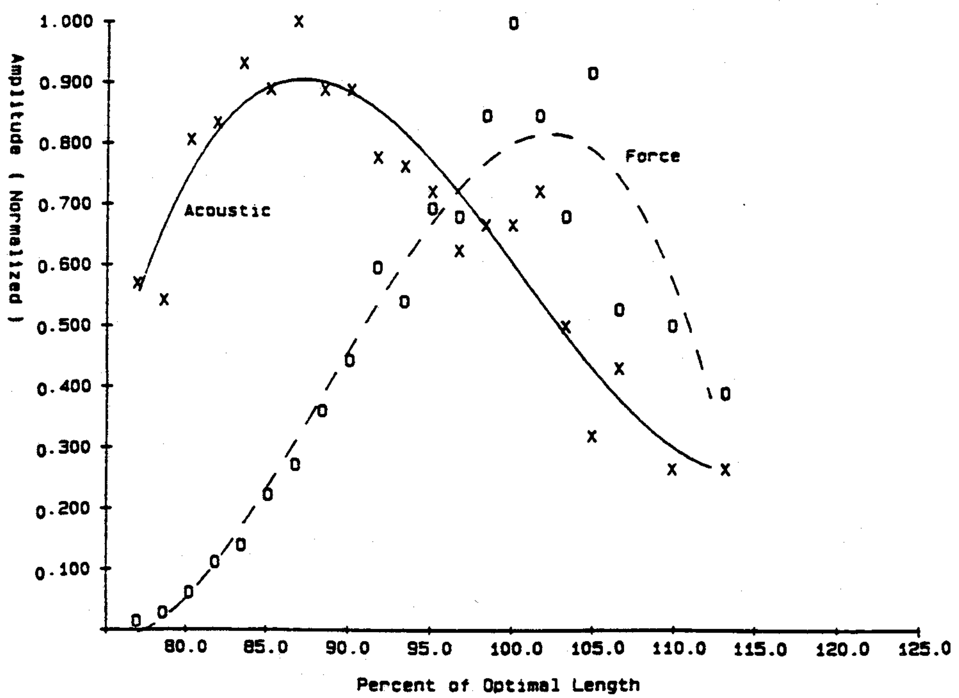

FIGURE 6. Peak-to-peak acoustic amplitude as a function of the muscle length. The maximum is reached at about 80% L_0. (From Barry, D. T., *Biophys. J.*, 51, 769, 1987. With copyright permission of the Biophysical Society.)

in the down-going phase of the twitch.[6] The amplitude of the peak-to-peak sound pressure during isometric twitches is strongly influenced by temperature. An amplitude defined at 25°C as 100% reduces to 15% at 7°C.[6] Moreover, the latency of sound with respect to the compound muscle action potential decreases as the temperature increases, from 7 msec at about 7°C to about 2 msec at 25°C.[6]

The behavior of the muscle sound with temperature parallels that of the force.[6] This suggests that it may be used to get information on the mechanical aspects of contraction.

4. Models to Explain the Muscle Sound Properties

The behavior of a vibrating string has been proposed to model the sound pressure generated by the isolated muscle. Indeed, the plucking of the isolated muscle generates a decaying pressure oscillation in which the first half cycle is the largest, "just as would be the case if a taut string was pulled to one side and released".[29] Figure 7 reports the relationship between the resonant frequency (f), the tension (T), and the length of the muscle described by the equation:

$$f = k \cdot T^{1/2}/L$$

where k is a constant taking into account for the actual mass and the shape of the muscle.[29] This relationship, f vs. $T^{1/2}/L$, can be studied by plucking the muscle at

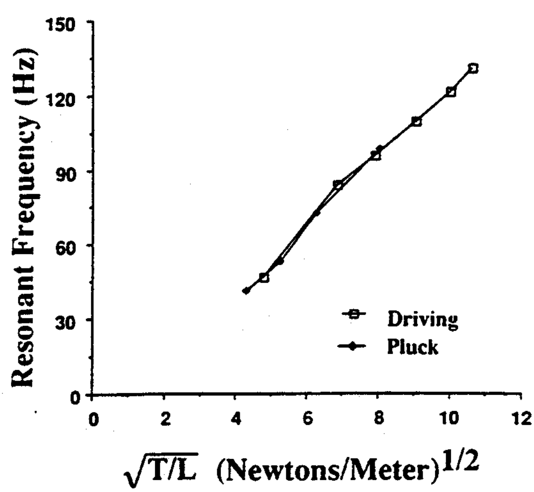

FIGURE 7. Resonant frequency of a passive muscle after mechanical driving or plucking. At a given length and tension the resonant frequency is the same with the two methods. (From Dobrunz, L. E., Pelletier, D. G., and McMahon, T. A., *Biophys. J.*, 58, 557, 1990. With copyright permission of the Biophysical Society.)

different lengths. The resonant frequency was reported to be about 20 and 140 Hz for $T^{1/2}/L$ of 5 and 30 respectively[29] or about 40 and 130 Hz for $T^{1/2}/L$ of 4 and 10 respectively.[26]

According to Fragioni et al.[29] this model is not fully adequate to explain the fact that during electrically elicited isometric twitches the larger half cycle was not the first one, that the pressure sound begins before the tension development and finally that the resonant frequencies are not proportional to the square root of the tension over length, in particular f is about 60 Hz when tension is near zero or still increases when the term $T^{1/2}/L$ is rather stable (beyond L_0, when both the total tension and length increase). The AA[29] concluded that muscle ringing during a twitch occurs in "one or more transverse modes" sustaining vibrations with a different physical mechanism than that of the taut string.

Following these considerations, in order to have a better fitting of the properties of the muscle sound detected during the isometric muscle twitch, an ax-handle model was proposed. This model has the peculiarity to present different frequencies of vibration when plucked in directions corresponding to the two principal axes of its elliptical cross section. If the ax-handle is plucked in a direction in between the two principal axes different signals will be recorded depending on the relative position of the hydrophone. When it is on the same radial of the pluck the maximum amplitude of the sound pressure will be detected at the first half cycle. On the contrary, if the hydrophone is 90° from the pluck direction then the signal recorded in the first half cycle will be nearly zero, because of the reciprocal cancelling of the two modes.[29] On this basis the AA[29] suggested the use of the vibrating string or the ax-handle model to describe muscle sound properties when the muscle is plucked passively or excited into ringing by its own nerve stimulation, respectively.

Another model suggested to explain the muscle sound generation and propagation is described by the fluid mechanics equations for a vibrating sphere, or better, for a vibrating cylinder.[7] In their work, Barry and Cole,[7] with the aid of high-speed cinematography underlined that the lateral motion implied in sound pressure generation has two components. The first one, large and slow, is due to the bulk movement of the muscle related to the nonsimultaneous activation of the fibers. "If one side of the muscle pulls harder or earlier then lateral motion will occur."[7] This slow bulk movement does not give a large contribution to the sound signal because the muscle is "too small to be an efficient generator of very low frequency pressure waves".[9] The second one, smaller and faster, is due to the "mechanical response of the muscle to a step function input";[7] it appears to be related to the muscle lateral acceleration and occurring at the resonant frequency of the muscle itself.

The resonant frequency is influenced by many factors, such as tension, stiffness, mass, length, viscosity of the muscle and of the surrounding medium. The hypothesis that the muscle sounds are emitted at the resonant frequencies of the skeletal muscle during isometric contractions has been tested carefully.[10,26] Indeed, in the first work[10] the muscle resonant frequency (f) and the peak acoustic frequency (PAF) were not measured at the same time because of the impossibility of achieving the muscle resonance, by mechanical stimulation, during the fast rising segment of the force. f was determined from the hydrophone signal during the rising phase of a tetanic contraction, by means of the time-frequency Wigner transformation. PAF was determined during the force plateau of the same tetanic stimulation when the mechanically imposed sinusoidal length changes were able to generate transverse standing waves in the muscle. The overlapping of the f and PAF values estimated at the end of the force rising phase and at the beginning of the force plateau (about 90 Hz for the frog gastrocnemius at 20°C) "strongly suggests that peak instantaneous acoustic frequencies occur at the resonant frequencies of the muscle".[10]

A convincing demonstration of this concept was given by Dobrunz et al.[26] They measured f and PAF during isometric single twitches. f was estimated by the analysis of the pressure sound due to plucks delivered to the muscle at different times during the single twitch. PAF was determined by mechanically driving the muscle at different frequencies during the twitch. For driving frequencies ranging from 40 to 90 Hz the maximum amplitude of the transverse standing waves was reached at different times during the twitch. In Figure 8, upper panel, the relationships between tension, f (from plucking), and PAF (from mechanical driving) vs. time are shown. Moreover, in the same figure the behavior of the muscle stiffness during the force rising and decaying phases is reported. The stiffness calculation was made by a mathematical model (see the equation) once the tension and the resonant frequency were known.

The results from isolated muscle studies suggest that thickening as an etiology of muscle sound production[20,33,39] is consistent with the amplitude vs. frequency relationship or with the influence of the muscle length upon the first half cycle amplitude. On the contrary several findings do not fit with this etiology. In particular they are the following: (1) the presence of signals 180° out of phase when two hydrophones are 180° apart,[6] (2) the dependence of the maximum pressure wave on the azimuthal angle,[6,10,29] (3) the fact that, as monitored by the sound pressure signal, when stimulated the muscle begins to move laterally in the same direction, even if to different extents, both at the belly (where it should enlarge) and near tendon insertions (where it should reduce its thickness).[29] As a consequence the "lateral motion" of the muscle has been suggested as the origin of the muscle sound. The gross lateral movement may be due to the fact that the contractile tissue is not homogeneously distributed in the muscle and when activity begins some regions contract more than others and an initial lateral displacement takes place.[6,26,29] Subsequent lateral movements are resonant frequency vibrations of the muscle[7] and seem to give the main contribution to the sound generation.

B. *In Vivo* Muscle

The recording of the muscle sound from *in vivo* stimulated muscle is intended to get information about the properties of this signal in a very near-physiological

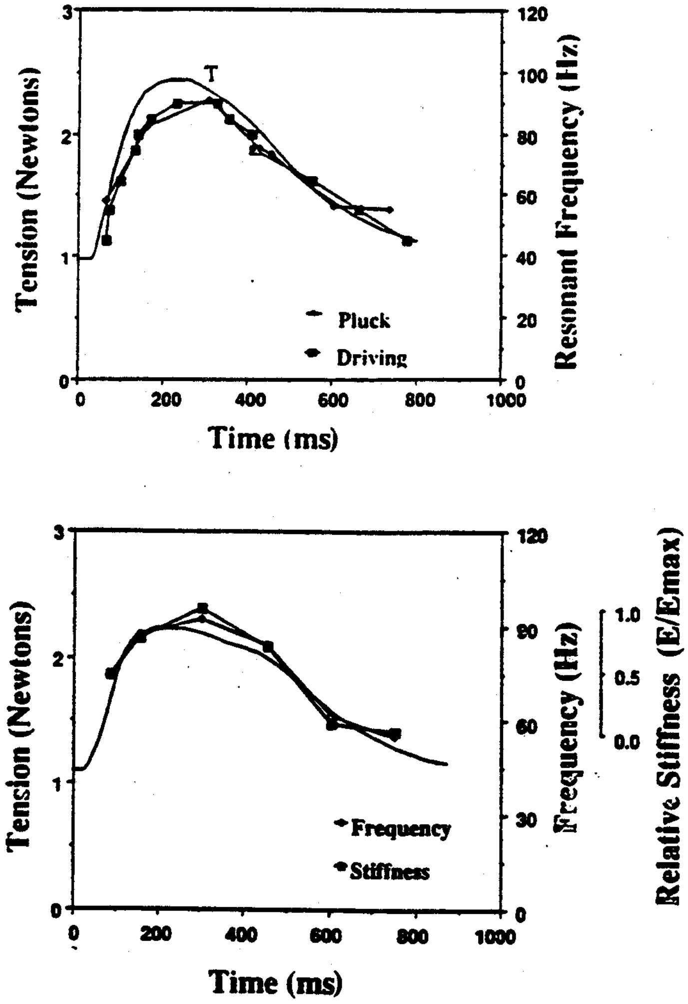

FIGURE 8. Upper panel: Resonant frequency of the muscle throughout a single isometric twitch. It can be noted that the resonant frequency is a function of the tension and as a consequence it changes over time. Lower panel: The stiffness, calculated by a mathematical model, is compared with the resonant frequency. It can be noted that the last parameter may be used to monitor muscle stiffness. (From Dobrunz, L. E., Pelletier, D. G., and McMahon, T. A., *Biophys. J.*, 58, 557, 1990. With copyright permission of the Biophysical Society.)

situation and, at the same time, by means of a well-controlled experimental situation in which problems related to motivation, ambience vibrations, or physiological tremor (see Section V) are completely avoided. During muscle contraction, shortening of the muscle produces movements of the muscle surface. This displacement and its acceleration are well described by means of piezoelectric contact sensors or accelerometers.

1. Single Twitch Recordings

In Figure 2 muscular sounds detected during an isometric single twitch are reported. It can be noted that the shapes of the sound pressure are biphasic curves or oscillating signals when detected by an air-coupled microphone and a piezoelectric contact sensor or an accelerometer, respectively. An analysis of these differences has been made in Section II.

In Table 1 the timing properties of the sound pressure during an electrically elicited isometric single twitch are reported. The duration of the twitch is greatly variable, ranging from about 70 msec in first dorsal interosseus[13] to 110 msec for the biceps brachii[43] when measurement by accelerometer is considered. For piezoelectric contact sensor measurements the duration of the sound pressure wave was 200 msec in abductor pollicis[12] and vastus lateralis,[49] about 250 msec for soleus,[49]

TABLE 1
Timing Properties of the Electrically Elicited Sound Pressure Wave

Sound pressure wave duration (msec)	SMG to stimulus delay (msec)	SMG to EMG delay (msec)	EMG to stimulus delay (msec)	Transducer	Supramaximal nerve (+) or direct (·) stimulation	Muscle	Subjects	AA
		2.2 ± 0.3		Wideband microphone	+	Lateral gastrocn.	6	4
		1.7 ± 0.3				Tibialis anterior	6	
		1–5		Air-coupled microphone	•	Tibialis anterior	?	39
800	9.3 ± 0.1	5.5	3.8 ± 0.2	Piezoelect. contact sensor	+	Thenar muscles	6	19
110		3		Accelerom.	•	Biceps brachii	1	43
≈200	6.9 ± 0.8	≈3.6	3.3 ± 0.5	Piezoelect. contact sensor	+	Abductor pollicis	27	12
	6.6 ± 1.0	≈3.9	2.7 ± 0.3			Abductor digiti minimi	27	
	5.7 ± 0.6			Accelerom.	+	Abductor pollicis		14
≈70	5.1 ± 0.6					Abductor digiti minimi		
≈70				Accelerom.	•	First dorsal inteross.		13
≈200				Piezoelect. contact sensor	•	Vastus lateralis	3	49
≈250						Soleus	3	

and about 800 msec in thenar muscle.[19] It has to be noted that, probably because of a different ratio in fast- and slow-twitch fibers, the time to peak in the vastus lateralis (39 to 55 msec) and soleus (99 to 115 msec) was significantly different.[49] In spite of the large differences in the event duration, a good stability in the time interval between the stimulus and the sound onset was observed. The reported sound delay was about 5 msec in abductor digiti minimi,[14] as a minimum, and about 9 msec in the thenar muscle[19] as a maximum. Even narrower is the range of the delay of the sound onset with respect to the EMG, spanning from 1.7 to 5.5 msec in the tibialis anterior[4] and thenar muscle,[19] respectively. Differences in the muscle sound vs. EMG delays have been found to be significant when comparing data from lateral gastrocnemius and anterior tibialis. The differences were attributed to the prevalence in fast-twitch fibers in anterior tibialis.[4] According to Hufschmidt,[39] during transcutaneous muscle stimulation, the delay between EMG and the sound onset is dependent on the stimulus amplitude; on the contrary, no dependence on the stimulus intensity was noted by Bolton et al.[19] using transcutaneous nerve stimulation. The discrepancy may be attributed to the different stimulation patterns in the two experimental conditions.

As for the force of the twitch, the sound pressure wave amplitude is dependent on the amplitude of the electric stimulus.[13,19,43] In the sound pressure wave response to an electrical stimulus, an initial deflection, opposite to the main one, has been reported.[19,39,43] This phenomenon has been attributed to a precontractile muscle elongation, which is followed by the shortening of fibers, fiber bundles, and of the whole muscle, with an increase in diameter and force generation.[43]

It has to be underlined that the data about the biceps brachii[43] have been recorded by means of a special "piezoelectric EMG needle device". This device is essentially an EMG needle coupled with a very light accelerometer fixed on its top. In this way the electrical and the mechanical activity of a little bundle of muscle is contemporarily detected from a "nearly identical intramuscular site".[43] This device promises to allow a very detailed electromechanical characterization of the muscle activity.

Spectral analysis of the sound response to electrically elicited isometric twitch reveals that power is mainly distributed in the low frequencies. In abductor pollicis brevis or digiti quinti hand 70% of the power was below 100 Hz.[14] In the thenar muscle the main peaks of the spectra were between 2 and 3 Hz and 13 to 14 Hz, while 90% of the power was distributed below 20 Hz.[19] In muscles with different fiber type distribution, different spectra resulted. Indeed, the mean frequencies of the sound spectra from vastus lateralis and soleus ranged from 5.34 to 6.26 Hz and from 3.74 to 4.70 Hz, respectively.[49] In comparison with the previous data, a completely different spectrum was obtained from anterior tibialis and gastrocnemius muscles.[4] The spectra presented a clear bimodality, with the first peak between 70 and 90 Hz and the second one between 120 and 140 Hz. The ratio between the power in the first and in the second frequency band was about 4 and 1.2 in the lateral gastrocnemius and anterior tibialis, respectively. A direct comparison of the spectral analysis results cannot be made because only two papers[19,49] indicated the transducer used and the frequency bandwidth of the signal filtering. Confirmation that the greater part of the signal power is distributed in the lower part of the spectra, below 30 Hz, was given by filtering the muscle sound from abductor pollicis brevis.[12] High pass filter set at 30 Hz has shown only a little oscillation, canceling the greater biphasic component of the signal.

The possibility of monitoring fatigue by the analysis of muscle vibrations detected by an accelerometer has been suggested.[13] In fact, evoked muscle vibration decreases with fatigue and the degree and the rate of the surface acceleration reduction with time is strongly influenced by the intensity of effort. Intermittent isometric exercise at 70% of the maximal voluntary contraction (MVC) determines a nearly exponential reduction of the vibration amplitude to about 30% of its initial value within 5 min. On the contrary, at 30% MVC the decrement within 10 min is only to about 50% of its initial value and is linear and slower. The behavior of the sound pressure wave parallels that of the force, presenting also the post-tetanic potentiation phenomenon. One of the factors supporting the acceleration amplitude decrease is the contraction velocity reduction taking place at fatigue.[13]

2. Repetitive Muscle Stimulation

The stimulation of the ulnar nerve at 70 Hz generated a muscular sound having a tone similar to the one detected during voluntary contraction.[64] In the paper no spectra were presented and a quantitative evaluation was impossible. More recently, Stokes and Cooper[71] supramaximally stimulated the ulnar nerve at the wrist and recorded sound from the adductor pollicis. Both the force ripple amplitude and the sound amplitude decreased dramatically from low to high stimulation rates. At 100 Hz the amplitudes of the two signals were about 5% of the reference value related at 10-Hz stimulation frequency. The overall relationships (force, sound, and force oscillation vs. stimulation rate) are reported in Figure 9. "The frequency of the acoustic myogram (AMG) matched stimulation frequency." As a consequence, the reduction of the integrated AMG was attributed to a reduction of the signal amplitude.[71]

The comparison between the isolated frog muscle and the human *in vivo* muscle stimulation data reveals that (1) at tetanic stimulation rates the AMG amplitude is really low for both *in vivo* and *in vitro* muscles. In particular, *in vitro* muscles show that once the resonant vibrations have dampened out the sound is absent because no additional lateral movement takes place throughout stimulation. (2) The large low-frequency component, 3 to 5 Hz, is present only in human muscle recordings.[8] This fact suggests that this slow phenomenon, related to the previously described slow bulk movement of the muscle, can be recorded only when the transducer is in contact with the muscle surface and moving with it, and not by a sound detector.[12] This is because low-frequency sounds can be generated only by objects that are much smaller than the wavelength of the sound itself. This is not the case for the muscles investigated.

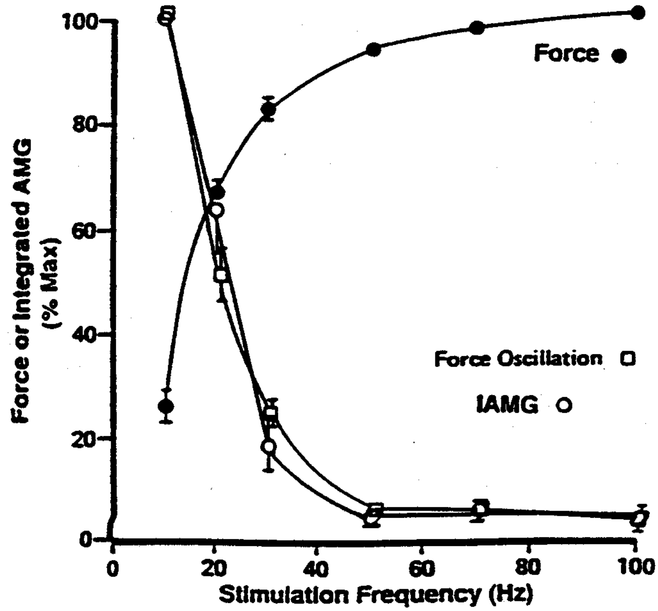

FIGURE 9. Relationships between the average values of force, force oscillation, and integrated AMG (IAMG) vs. stimulation frequency. The similar trend of the force oscillation and IAMG confirms the data from the isolated muscle, see Figure 3. (From Stokes, M. J. and Cooper, R. J., *J. Appl. Physiol.,* 72, 1908, 1992. With permission.)

According to some authors, the similarity between the acceleration signal and the hydrophone signal,[13] as well as the integrated AMG vs. stimulation frequency relationship,[71] supports the hypothesis that *in vivo* the muscle sound pressure wave is generated by the "lateral movement" of the active muscle.[6,29] On the other hand, the thickening of the muscle, due to its shortening, as previously suggested,[33] was not discarded because of the biphasic shape of the vibration recorded during single isometric twitch[19,43] or directly indicated as the muscle sound source.[39]

Whatever is the sound pressure wave origin, the properties of the signal, recorded during single twitch isometric stimulation, were related to the properties of the active muscle, i.e., the performance of the contractile elements and the features of the elastic tissue in parallel to them.

This conclusion is supported by the data on sound signal timing (Table 1), the influence of fatigue,[13] the frequency content when contraction was electrically elicited from different muscles,[4,49] and, finally, by the fact that the force twitch peak is delayed in respect to that of the sound pressure, probably because of the influence of the series elastic elements not affecting muscle sound.[19]

Following some AA[71] acoustic myogram detects the lateral movement produced, together with the longitudinal movement recorded by a strain gauge at the tendon, during muscle contraction shortening. On this basis the similarity between the AMG and the force oscillation during stimulation could be explained and it can be hypothesized that "AMG characteristics are determined by motor control mechanisms rather than intrinsic contractile processes".[71] This view is strongly supported by a recent work[61] aimed at evaluating the influence of motor units recruitment (REC) and firing rate (FR) on the SMG characteristics. By means of a system capable of controlling the REC level and the FR from (2.5 to 50 Hz), the AA demonstrated that at full REC the maximum value of SMG peak-to-peak amplitude and of the SMG root mean square (RMS) were measured at 2.5 and 5 Hz, respectively. An exponential decrease was seen for the two parameters beyond these FRs. Moreover, at a chosen FR the SMG amplitude increases together with the number of active motor units (MUs) even if the extent of the increment was strongly conditioned by the FR itself. The AA[61] concluded that both REC and FR affect the SMG time domain parameters and as a consequence data on MUs activation strategy may be retrieved from SMG analysis.

In conclusion, some evidence derived from the processing of muscular sound recorded during repetitive stimulation indicates that this signal can be used to obtain information about single muscle motor control.

IV. VOLUNTARY CONTRACTIONS

A. Nonfatiguing Contractions

1. Single Motor Unit Sound Spike

Single motor unit sound spike (MUSS) was recorded during contraction of different muscles. The duration of the spike was in the range of 5 to 15 msec in orbicularis oculi,[33] about 46 msec in interosseus dorsalis,[22] and about 40 msec in vastus lateralis after traumatic lesion of the femoral nerve.[3] In Figure 10 the MUSSs are reported together with the motor unit action potentials (MUAPs). The great difference in the reported MUSS duration may be attributed to differences in the muscle mechanical properties, to their innervation ratio, and to the detecting technique. The importance of the last factor can be evaluated considering that when the piezoelectric crystal was directly coupled to the muscle surface, rather than by

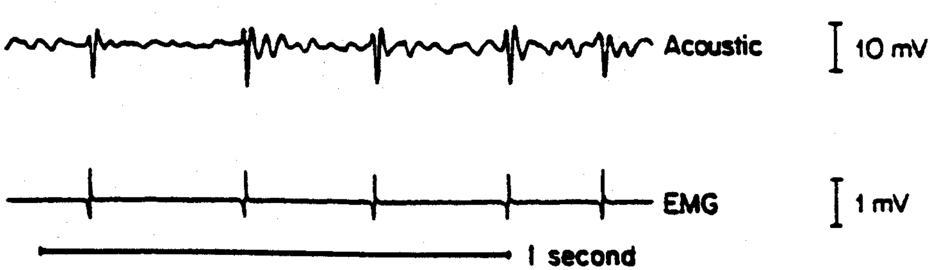

FIGURE 10. Motor unit electrical and acoustic activity recorded from the vastus lateralis muscle. (From Barry, D. T., Geiringer, S. R., and Ball, R. D., *Muscle Nerve,* 8, 189, 1985. With permission.)

a chamber with a diaphragm inducing deformations in a Rochelle salt crystal, Gordon and Holbourn[33] also recorded events lasting 30 to 40 msec. They named these signals muscle movements. It can be concluded that MUSS appears to be shorter than the sound event recorded during isometric single twitch elicited by motor nerve supramaximal stimulation (>60 msec, see Table 1). This indicates that when the whole muscle moves the mechanical event is slower than when a small part of it is activated. It cannot be disregarded that in the former situation the influence of the noncontractile elements of the muscle model, in the determination of the temporal evolution of the mechanical event, is much greater.

2. Time Domain Analysis

Even by the simple experimental condition reported by Wollaston[74] it is possible to recognize that the muscle sound increases with the strength of contraction. The signal amplitude vs. force relationship has been investigated in several muscles and in different ranges of output force.

In biceps brachii the relationship was found linear from a few grams to 10 kg.[64] In the same range Barry et al.[3] showed good linearity of the AMG RMS increase only from 2.5 to 7.5 kg. We have to take into account that, when the force output is measured by a load cell strapped perpendicularly to the wrist, 10 kg is about 50% or less of the MVC. For this reason more recent works investigated the muscular sound vs. force relationship from 10 to 20% MVC to 100% MVC (a representative set of SMG signals is reported in Figure 11). Results showed: (1) an S-shaped AMG RMS vs. force relationships,[46] (2) a parabolic increase of integrated SMG from 10 to 80% MVC followed by a decrease at 90 and 100% MVC,[56] (3) a quadratic increase of the AMG RMS up to 100% MVC,[52] and (4) a linear correlation of integrated vibromyogram (VMG) up to 100% MVC.[76]

In the quadriceps the acoustic myographic activity shows a linear increase with force up to 80% MVC[75] or up to 100% MVC.[69,70] In reality, from the amplitude vs. percent MVC relationships reported in some papers,[69,70] it seems that in the quadriceps the rate of AMG increment decreases from 75 to 100% MVC.

In lumbar erectores spinae during a ramp isometric contraction reaching 100% MVC in 10 s, the AMG RMS showed a quadratic relationship with force. Out of lower levels of effort in the decreasing 10 s ramp the AMG RMS presented greater values than at the same force during the increasing ramp, showing a "reversed hysteresis of the AMG signal".[68]

In human jaw elevator muscles (masseter and anterior temporalis) investigated up to 30% MVC, the AMG RMS increased to a maximum at 5 or 10% MVC, and *remained constant or decreased for higher forces.*[67] *In the adductor pollicis muscle* the integrated AMG activity increased in a curvilinear fashion up to 100% MVC.[71] The proposed SMG amplitude vs. force relationships are summarized in Figure 12.

As discussed by Stokes and Dalton,[69] the integrated EMG or EMG RMS vs. force relationship is different in different muscles. It may be linear or quadratic,

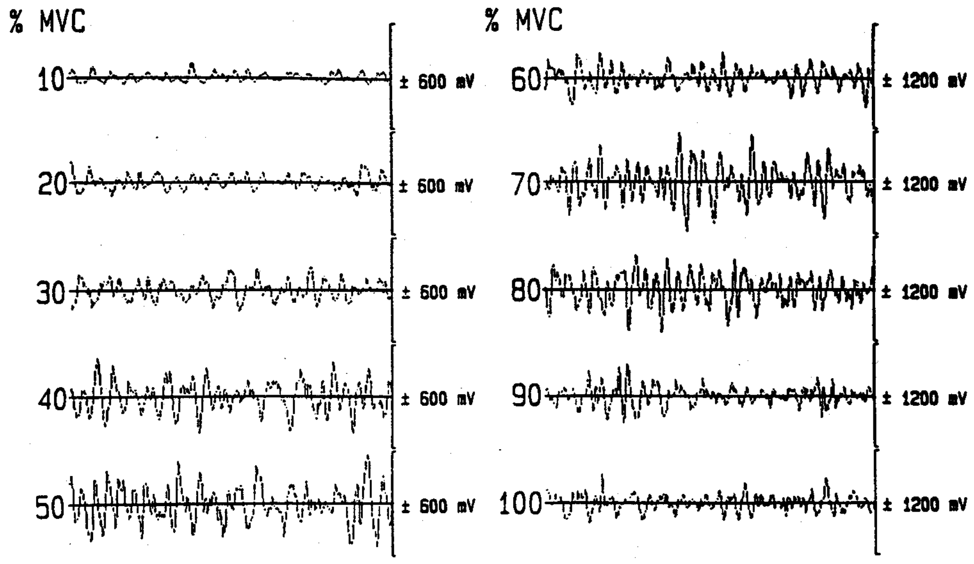

FIGURE 11. Raw soundmyogram recorded from the biceps brachii during isometric contractions at different effort levels. Note the amplitude reduction beyond 80% MVC. (From Orizio, C., Perini, R., Diemont, B., Maranzana Figini, M., and Veicsteinas, A., *J. Appl. Physiol.*, 68, 508, 1990. With permission.)

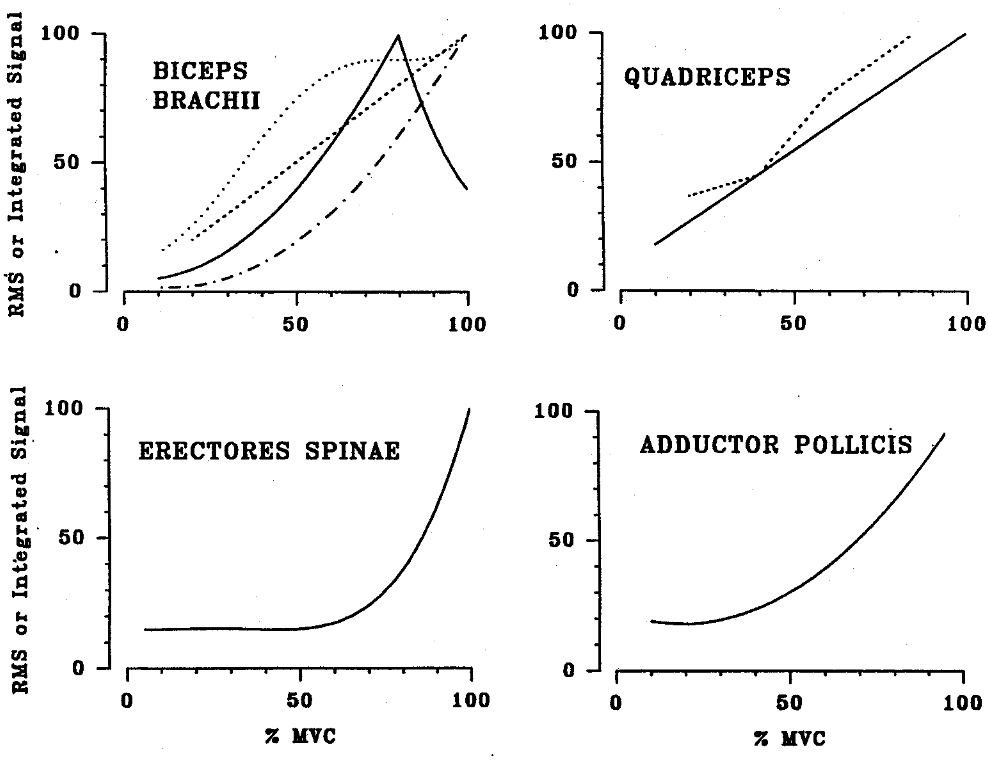

FIGURE 12. Relationships between SMG time domain parameters and force in different muscles as found in the literature; for explanation see text.

depending on a uniform or mixed fiber composition and distribution within the muscle, respectively. The same factor may be responsible for the different SMG vs. force relationships described in the different muscles. Indeed, the large increase in the sound wave amplitude above 30% MVC in biceps brachii was indicated to be due to the recruitment of fast-twitch fiber MU having a more superficial distribution and a more dramatic effect on the muscular sound recorded.[56] On the contrary, the fitting curves for the AMG RMS vs. force relationships were quadratic in the erectores spinae,[68] where the fiber type distribution "was not found to have consistent proportions either in the deep or in the superficial layers".[69] As a consequence, future studies devoted to investigate and clarify the role of the several factors that possibly affect the muscle sound time domain parameters (Section II), are needed.

At this time the data in the literature allow speculation on the role that two important factors, motor unit recruitment and firing rate, can play in determination of SMG properties. It is well known that the output force of a muscle can be controlled by changing the number (recruitment, REC) and the firing rate (FR) of the active motor units. The level of voluntary contraction at which REC finishes and FR is the sole tool to increase the force is a specific characteristic of each muscle.[15,45] In the biceps brachii this level of effort is about 80% MVC and in the adductor pollicis about 50% MVC.[45]

Taking into account that when all the MUs are active an increase of the firing frequency determines a reduction of the sound amplitude,[29,61,71] a reduction of the integrated SMG or SMG RMS can be expected in the biceps brachii or in the adductor pollicis beyond 80 and 50% MVC, respectively. A clear decrease in the integrated SMG at high levels of effort, when high MUs FR are expected, was seen only in the biceps brachii[56] and in the jaw elevator muscles.[67] The reduction of the SMG amplitude may be attributed to the fact that for high MUs FR the muscle fiber dimensional changes are little and, as a consequence, the related pressure waves detected at the muscle surface are weaker.[56] This hypothesis is supported by the already described reduction of the AMG from a single MU when activated at a high frequency rate.[41] Moreover, at high contraction levels, increases in the intramuscular pressure and in the muscle stiffness may limit the cited muscular sound generation mechanism.[56]

On these bases the monotonic increases of the AMG vs. force relationship in adductor pollicis, erectores spinae, and quadriceps may be due to a failure in the cited muscular sound limiting factors. In particular a more favorable relationship between the maximum MUs FR and the contraction times of the muscle fibers may exist.

In the biceps brachii the findings of Orizio et al.[56] were not confirmed in different studies. Zwarts and Keidel[76] presented both reductions and increases of the integrated VMG from 80 to 100% MVC. On the contrary, Maton et al.[52] showed quadratic increase from 10 to 100% MVC. It has to be considered that in this last work different time windows, at a submaximal level of effort (10 s) and at 100% MVC (1 s), were used for the signal integration. This may mask the real values of SMG, especially at 80% MVC, where, within 10 s, changes in SMG amplitude have been described because of fatigue.[57]

In conclusion, during voluntary contraction of the fresh muscle linear or quadratic relationships of the integrated signal or of its RMS vs. force were described, mirroring mainly the number of the active MUs. Finally in some muscle the high MU firing rate, at a high level of effort, may be monitored by a decrease or by a leveling off of the integrated SMG.

The properties of the AMG during dynamic concentric and eccentric contractions in biceps brachii showed that the greater the weight to be moved during flexion and extension of the elbow, the greater the amplitude of the integrated AMG.[24] The flexion and the extension of the forearm was performed in 3 s. The lesser SMG amplitude during eccentric work, when the same weight is supported, was attributed to a reduced number of active motor units in respect to the concentric work phase of the exercise. This result suggests that AMG may be a tool for the evaluation of the viscoelastic properties of the muscle, which influence the difference between the motor units activation strategy during concentric and eccentric contraction.

3. Frequency Domain Analysis

That the frequency of the muscular sound was varying "according to the degree of force exerted in the experiment" was already known in the past.[74] One of the first estimations of the muscular sound frequency content was carried out by Cerquiglini et al.[22] They processed the data by a frequency analyzer and found that the greater part of the signal power was below 100 Hz. This result was reported for the flexor digitorum superficialis, trapezius, biceps brachii, and first interosseus dorsalis. Moreover, they ascertained that increasing the intensity of contraction shifted the "phonomyogram" towards the higher frequencies. The frequency range of the phonomyogram, estimated by the same method, was confirmed later for gastrocnemius and quadriceps.[23]

A great deal of data about SMG frequency content can be found since the 1980s. Oster and Jaffe[64,65] indicated a spectrum of the muscular sound having a "dominant" frequency of 25 ± 2.5 Hz in biceps brachii, the power spectrum distribution being unaffected by the intensity of contraction. The results were the same when autocorrelation[64] or the Fast Fourier Transform (FFT)[65] were applied. In subjects younger than 60, during a maximal-contraction-like effort with no objective force measurements, the peak frequency of the biceps brachii SMG spectrum was indicated as 15 ± 4.2 Hz and no power beyond 100 Hz was found.[66] Considering that the muscle contraction may take place "in a stepwise fashion, each step would pull on the elastic connective tissue and give rise to oscillations of the muscle body".[66] According to Rhatigan et al.,[66] this phenomenon, together with dimensional changes of the active muscle fiber, may contribute to the sound generation.

With the aim to check the adequacy of FFT and maximum entropy spectral estimation (MESE) and to define the most reliable spectral parameter (among mean, median, and peak frequencies), two well-defined physiological situations, 20 and 80% MVC, were investigated during isometric contraction of the biceps brachii.[25] These authors found that both algorithms were useful in SMG analysis, although MESE was faster, and that from low to high contraction level the spectrum shifted towards the higher frequencies. This phenomenon was described by the shift of the mean and median frequency from about 11 to about 15 Hz. The difference was statistically significant. The mean frequency presented the lowest variability, in respect to the different algorithm used, and its changes were only dependent on the level of contraction.

During contraction of the biceps supporting a 2.3-kg weight (about 10% MVC) Wee and Ashley[72] showed that subjects having difficulty in maintaining a steady contraction showed a spectrum of the sound vibrations with more peaks than others. The spectral analysis was carried out by FFT and a dominant peak at 11.3 Hz was clearly found; moreover, the greater part of power was found below 20 Hz. Other peaks between 20 and 50 Hz "can be expressed as exact harmonics" of the peaks found in the 5- to 20-Hz lower bandwidth.[72]

The VMG was recorded by a piezoelectric accelerometer in masseter, biceps brachii, wrist extensors, and tibialis anterior muscles.[42] During rest two major peaks at 10 and 12 Hz can be observed in all the muscles. Other peaks were seen at 3, 7, 20, and 23 Hz. During contraction (maximal isometric for masseter and wrist extensor or 10-N submaximal isometric for biceps brachii and tibialis anterior) an abrupt increase in the power of the 1- to 20-Hz band of the VMG spectrum was seen. A power increase was noted also in the 20- to 80-Hz band.

In the biceps brachii at rest, considering the power in the 1- to 49-Hz band as 100%, the power distribution in different bands was 24% (1 to 7 Hz), 45% (7 to 14 Hz), 17% (14 to 19 Hz), and 14% (19 to 49 Hz). During an isometric contraction producing a 40-N output force, the power in the same bands was 12%, 35%, 23%, and 30%, respectively. According to the AA[42] these results do not support the concept "*of just one stable frequency of, e.g., 10 or 25 Hz*".

The SMG frequency content vs. force relationship was later investigated in biceps brachii by means of an accurate and simple experimental design.[58] The spectral analysis was performed, by means of FFT and MESE, on 2-s SMG time windows. The SMG was recorded for 4 s during 10 isometric contractions from 10 to 100% MVC. The first and the last second of activity were discarded in order to avoid the analysis of transient phenomena. Increasing the intensity of contraction resulted in the enlargement of the SMG spectrum (Figure 13). In particular, above the 30% MVC, when both the slow- and the fast-twitch fiber MUs are active,[34] it tends to be bimodal. This aspect is clearer beyond 70% MVC. The mean frequency ranged from about 11 Hz at 10 to 20% MVC to 15 and 22 Hz at 80 and 100% MVC, respectively. The ratio, HF%, of the power in the 15- to 45-Hz band (firing rate range of fast-twitch fibers MUs) in respect to total power from 6 to 45 Hz (slow-twitch fibers MUs FR range + fast twitch fibers MUs FR range), changes from 20

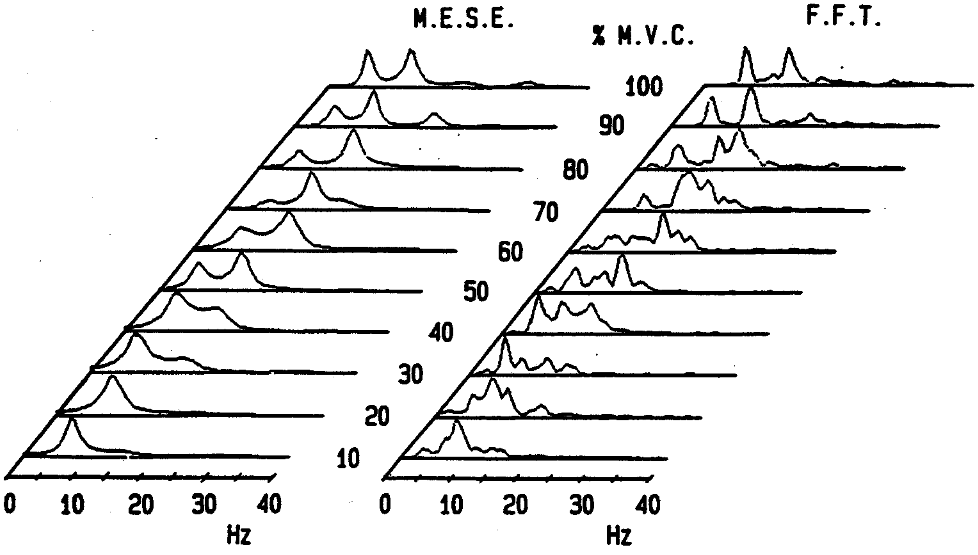

FIGURE 13. Spectral analysis of the SMG detected from the biceps brachii, during isometric contractions of the elbow flexors. Increasing the intensity of the effort shifts the spectrum toward the higher frequencies and tends to distribute the power in two bands. This phenomenon is better described by the MESE method. (From Orizio, C., Perini, R., Diemont, B., Maranzana Figini, M., and Veicsteinas, A., *J. Appl. Physiol.,* 68, 508, 1990. With permission.)

to 55% passing from 10% MVC to 100% MVC, respectively. The FR ranges of the two main groups of MUs have been derived from the literature.[21,30] In Figure 14 it has to be noted that the steeper increase in the MF or HF% vs. %MVC relationships takes place from 70 to 80% to 100% MVC, i.e., the same range in which the integrated SMG reduces.[56]

Following analogous experimental design, Maton et al.[52] did not show a similar relationship between SMG mean frequency and force. On the contrary, in the 30 to 100% MVC force range they found a leveling off of the phonomyogram mean frequency.

Frequency analysis of the vibromyogram recorded from the rectus femoris by means of two miniature accelerometers revealed an increase in the peak frequency of the VMG from about 11 to about 19 Hz from 20 to 80% MVC, respectively.[75]

From the results reported above, the qualitative evaluation of Wollaston[74] indicating an increase in the frequency with the effort can be confirmed. In fact, only in the papers by Oster and Jaffe[64] and by Oster[65] was it stated that the SMG spectrum was independent of the contraction strength with a fixed dominant frequency at 25 ± 2.5 Hz. We have to take into account that this observation is relatively valid in the narrow range of muscle output force investigated, up to about 30% MVC. Indeed, the greatest changes in the SMG spectra were detected beyond 30% MVC.[58] The reported dominant frequency, 25 ± 2.5 Hz,[64,65] induced the AA[65] to suggest the dynamic of the cross-bridge attachment and detachment, depending

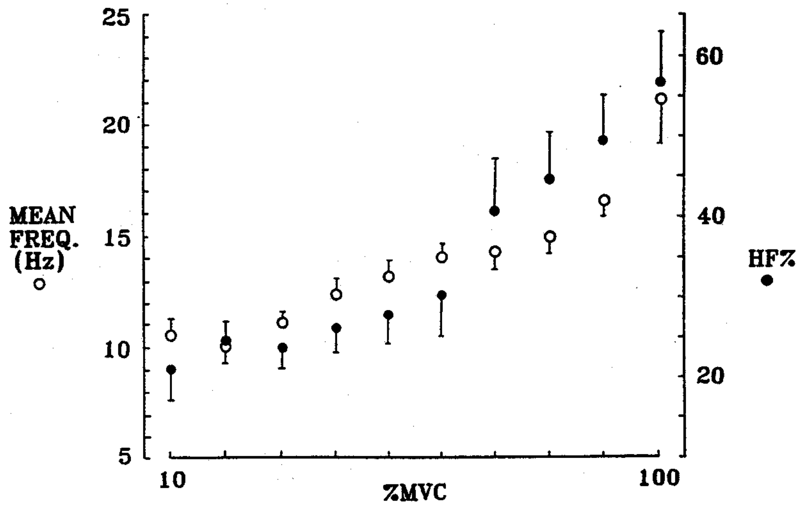

FIGURE 14. Average mean frequency and HF% parameter values (i.e., the relative weight of the power distributed in the 15- to 45-Hz band in respect to the 6- to 45-Hz band); trends as a function of the %MVC. Note the steeper portion of the relationship beyond 70 to 80% MVC. MESE method data. (Redrawn from Reference 58.)

on the 40-msec adenosine triphosphate (ATP) turnover duration, as a major determinant in the SMG generation.

The information about the SMG frequency content provides material for more solid hypotheses on the muscle sound origin. According to Orizio et al.,[56] the decrement in the integrated SMG at 90 and 100% MVC was, partly, related to the high MU firing rate in the biceps brachii which, at those levels of contraction, limits the sound generation. The confirmation of this point of view is provided by the frequency domain analysis that showed the greatest increases in the spectral parameters at those effort levels in which increasing FR is the sole tool to develop force.

Noting that during muscle activity the greatest power increase was in the 20- to 80-Hz range and that the mean MU firing rate reported for several muscles lies in this range, Keidel and Keidel[42] also indicated MU FR as a significant part of the information contained in the SMG. The power in the lower portion of the SMG spectrum was attributed to MUs discharging near the recruiting threshold.[42,58] This view led to underlining the role of the muscle fiber activity in the generation of the muscle sound vibration. On this basis Wee and Ashley[72] explained the power in the lower part of the spectra as large and slow subharmonics related to movements of the whole muscle due to the summation of "smaller and faster movements generated by asynchronous contractions of different groups of muscle fibers".

It is well known that in a compound signal such as the EMG the frequency content is determined by the frequency of the different MUs spike trains and, to a greater extent, by the shape of the single MUAP.[15,36] The SMG can be regarded as a compound signal too, when it is considered as the summation of the different MU mechanical activities, i.e., motor unit sound spike, MUSS[63,75] (see also Figure 18); the morphology of MUSS being determined by the fiber contraction time. The striking difference in the MUAP and MUSS morphology determines the great difference in the spectrum mean and peak frequencies of EMG and SMG.[75] The increase of the MU spike train frequency, with increasing effort, may be responsible for the linear increment of the SMG mean frequency (MF) from 20 to 80% MVC in the rectus femoris.[75] Moreover, it cannot be disregarded that with increased effort larger MUs, with shorter contraction time, and hence shorter MUSS, are recruited. This fact may induce a shift towards the high frequencies of the SMG spectrum, as happens for the EMG spectrum, because of the shorter MUAP of the same MUs.

As for the time domain analysis, the discrepancy in the frequency content observed in biceps brachii at high levels of effort[52,58] can be explained on the basis of a different length in the signal time window considered (see the following section). In fact, fatigue strongly influences the frequency content of the muscle sound and, consequently, the duration of the signal epoch is crucial in the determination of spectrum properties.

In conclusion, the frequency domain analysis suggests that both the motor units firing pattern and contractile properties are reflected in the SMG spectrum. The existence of a direct influence of the FR on the SMG amplitude and on SMG frequency content, which may make it possible to get data on the muscle activation strategy, has to be explored and clarified in all its aspects. Definitive data will be provided by future investigations in which needle EMG will be coupled to the SMG recordings. In this view the use of sophisticated EMG processing methods, such as the decomposition technique,[15] together with the SMG time and frequency domain analysis, will be determinant.

B. Fatiguing Contractions

1. Time Domain Analysis

The first paper[3] dealing with the fatigue influence on the AMG properties investigated the biceps brachii. The AA[3] used two experimental designs consisting of (1) five sequences of four contractions at 2.5, 5, 7.5, and 10 kg (each weight was held for 20 s with 10 s rest between contractions; between one sequence and the following, 15 s of recovery were allowed); (2) a 75% MVC was sustained as long as possible. After the force began to decrease the exercise was stopped when 35% MVC was reached. From the experimental design (1) it resulted that from the first to the last sequence the AMG RMS vs. force relationship became steeper. On the other hand during sustained isometrics at 75% MVC the AMG amplitude reduced together with the force. At the same time EMG did not change its amplitude. The conclusion was that AMG correlates better than EMG to the generated force during sustained isometric contraction, the electromechanical dissociation due to fatigue being evident by the comparison of the two signals. On the contrary at the same level of effort when fatigue develops slowly (experimental design 1), the acoustic signal increases. This was attributed to increased physiological tremor, to a better sound transmission "as muscle swells secondary to osmotic phenomena during prolonged exercise", or to recruitment of new motor units.

The relationship between integrated AMG and force (from 10 to 75% MVC) has been found unaltered after fatigue of the quadriceps muscle.[70] Fatigue was induced by intermittent contractions (10 s on, 10 s off) at 75% MVC until only 40% MVC could be sustained. After 15 min the relationships between integrated AMG and EMG vs. percent MVC were reinvestigated at the same prefatigue absolute forces. These relationships were found the same or steeper for the AMG and the EMG, respectively. The lack of AMG increase at the same force after fatigue, as described for biceps,[3] may be due to the different experimental protocol allowing recording without interference of fatigue tremor in quadriceps.[70] This result indicates that in this case the AMG is well correlated with the force changes even when fatigue develops slowly. Moreover, the electromechanical dissociation was easily described.

The behavior of the SMG during really precise isometric contractions was studied both in the biceps brachii[57] and in abductor digiti minimi.[32] In the first study, in eight subjects, the SMG was recorded during 20, 40, 60, and 80% MVC sustained up to exhaustion, i.e., when the subject was no longer able to maintain the force within ±5% of the target value. From the onset to exhaustion the integrated SMG increased five times at 20% MVC, was nearly the same at 40% MVC, and decreased about five times at 60 and 80% MVC. These results confirm that when fatigue develops slowly, as at 20% MVC lasting 480 ± 110 s, the SMG increases its amplitude, while when it develops rapidly, as at 60% (68 ± 6 s) and 80% MVC (39 ± 6 s), the SMG reduces drastically. This finding is well described in Figure 15 where the electromechanical dissociation is evident, "not only when the force decreases with time as previously shown[3] but also when the force output is maintained constant".[57] During contraction of the biceps brachii at about 10% MVC for 4 min[42] or during 50% MVC up to exhaustion[76] results similar to those reported at 20 and 40% MVC[57] were obtained. Also in these cases the amplitude of the SMG increased clearly during a low-level effort or did not change significantly at 50% MVC.

In the second study on the abductor digiti minimi[32] the AMG RMS increased significantly at 15 and 25% MVC while a clear reduction was seen at 50% MVC. At 75% MVC the AMG RMS presented only a slight decrement.

In order to explain the changes in amplitude throughout sustained constant force isometric contraction a reference to the changes in the MU activation strategy due to fatigue is needed. With reference to the muscular sound this topic has been dealt with in several papers.[3,32,57,60,63] Briefly, during isometric low-level efforts the recruitment of new, larger, and more superficial MUs, with a more direct effect on the muscle sound, and the increase in the firing rate of the already active MUs

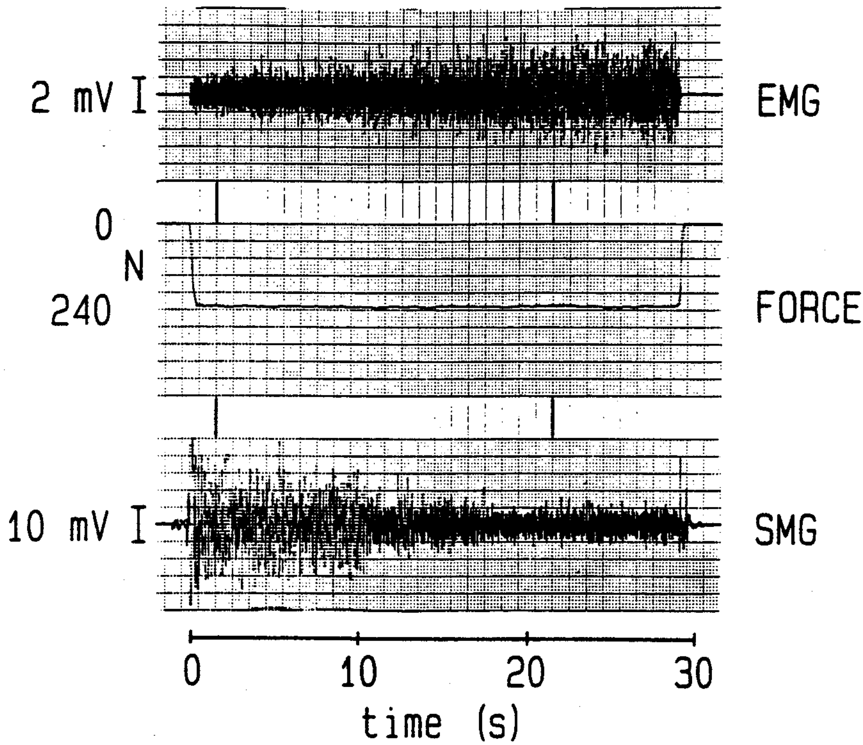

FIGURE 15. Original tracings of electromyogram (EMG), force, and soundmyogram (SMG) during exhausting 80% MVC contraction of the elbow flexors. The signals were detected from the biceps brachii. Despite the constancy of the output force a clear diverse behavior of the EMG and SMG is shown. The electromechanical dissociation and the impossibility to monitor the force by the SMG during this type of effort are demonstrated. (From Orizio, C., Perini, R., and Veicsteinas, A., *J. Appl. Physiol.,* 66, 1593, 1989. With permission.)

produce the required output force over time.[28] Moreover, near exhaustion at very high contraction levels, the muscle fiber relaxation time has been shown to increase and determine, via a "proprioceptive feedback", a reduction of the MUs FR.[18] The possibility for the fast twitch-fibers MUs to be derecruited has been considered by Goldenberg et al.[32] For the maintenance of a constant output force the MUs recruitment may be more important during low-level contractions while an increase in FR may play a major role at high levels of effort,[28] i.e., beyond 50% MVC in the adductor digiti minimi,[32] or 80% MVC in the biceps brachii.[57]

Recruitment of new MUs, and increasing tremor, may be responsible for SMG amplitude increase at 20% MVC (biceps brachii) or at 15 and 25% MVC (adductor pollicis minimi). The reduction of the SMG at high contraction levels in biceps brachii, from the onset (80% MVC) or after 10 to 20 s (60% MVC), can be related to the development of a fusion-like situation in which the relaxation time of the muscle fibers is elongated because of fatigue and the MUs firing frequency may be not proportionally reduced by the proprioceptive feedback. The reduction of

AMG RMS in abductor digiti minimi at 50% MVC was attributed to a drop out of the fast MUs and to a fusion-like situation occurring in the slow MUs for the high firing rates needed to maintain force.[32] As can be suggested from the results reported during sustained isometric contraction, recruitment of new MUs and an increase of MUs FR may determine an increase or a decrease in the integrated SMG or AMG RMS, respectively.[32,57] When the SMG time domain parameters do not change throughout contraction it can be argued that the influence of the two mechanisms is balanced.[32] This may occur at 40 or 50% MVC in biceps brachii.[57,76] The stability of the AMG RMS at 75% MVC in abductor digiti minimi was attributed to a very short duration of the exercise due to the impossibility of the slow-twitch fibers MUs to sustain the effort, even by increasing their firing rate, when the fast-fatiguing larger MUs are derecruited.[32]

During 63 s of sustained MVC the reduction of force is well paralleled by the integrated VMG signal.[76] This phenomenon may be related to derecruitment of more fatigable MUs, reduction of MUs FR, and changes in the intramuscular pressure and stiffness.[63]

The effect of the fatigue tremor on AMG RMS changes in abductor digiti minimi was not taken into account because of a truncation, below 14 Hz, made during FFT processing.[32] Moreover, the influence of the intramuscular pressure and stiffness on the SMG amplitude at fatigue, in particular at 60 and 80% MVC, was suggested to contribute to the SMG reduction throughout contraction.[57]

In conclusion, the time domain analysis of muscular sounds can show that the changes of the calculated parameters depends on the rate at which fatigue is induced in the muscle, by different effort intensities or repetitions frequency. Moreover, it seems dangerous to use the muscular sound as a tool to quantify muscle force at fatigue (see Figure 15), because this is possible only in particular cases, probably when MUs REC and increase in MUs FR effects counteract one another. As a consequence it may be safer to regard muscular sound changes as a result of variations in MUs activation strategy and muscle physical milieu (osmotic pressure, intramuscular pressure, stiffness) due to fatigue.

2. Frequency Domain Analysis

Only a few studies deal with the influence of fatigue on the SMG frequency content. In biceps brachii an increase in the power in the 8- to 18-Hz and 20- to 30-Hz bands was described during an isometric contraction of the elbow flexors supporting 3 kg at a fixed angle between arm and forearm.[42] The clearer "demarcation of the peaks" was indicated as a monitor of the MUs synchronization at certain frequencies. In biceps brachii during 50% MVC sustained up to exhaustion the changes of the VMG spectrum were related only to its shape (from bimodal to unimodal) with no changes in the calculated mean frequency. The morphological changes were explained by considering the increased MUs twitch duration and by the presence of synchronization.[76] Some changes in the spectrum MF were reported for 63-s sustained MVC.[76]

From the beginning to the end of 50% MVC sustained contraction of the abductor digiti minimi, the AMG spectra, calculated after the below 14 Hz truncation, showed a clear shift towards the lower frequencies and changes of the MF from 23 to 17 Hz.[32]

A more detailed investigation of the SMG frequency content throughout sustained isometric exhausting contraction was carried out on biceps brachii only recently.[60] The SMG frequency changes were studied at 20, 40, 60, and 80% MVC. It resulted that the frequency content did not change significantly at 20% MVC, while at higher effort levels after an initial shift of the power towards the higher frequencies a compression of the spectra was described. These phenomena are well represented in Figure 16, in which the trends of the average MF and of the average HF%, at the four contraction intensities, from the beginning up to exhaustion, are reported. The changes in frequency content of the SMG are well explained by the changes in MUs activation strategy already described (see the previous section).

For the SMG spectrum changes at high levels of effort it has to be considered that the force output is maintained constant with time increasing the FR of the active MUs.[28] In particular, it has been found that the fast-twitch fiber MUs presented a transient increase, followed by a clear decrease, of their firing rate.[34] Moreover near exhaustion the proprioceptive feedback, monitoring the elongation of the muscle fiber relaxation time, provides a reduction of the MUs FR.

The relative stability of the frequency at 20% MVC fits well with the stability of the mean MU FR[51] and synchronization[44,51] reported for these levels of effort

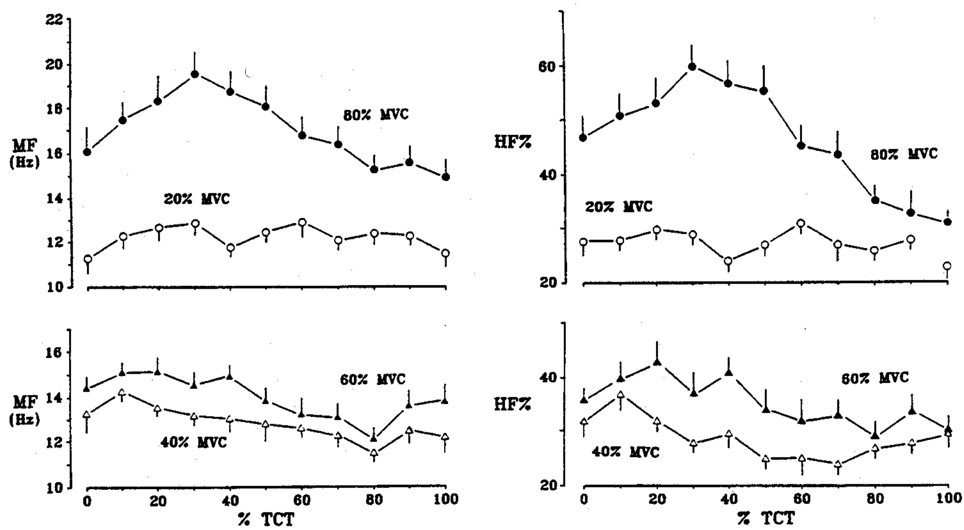

FIGURE 16. Changes of the average values of the mean frequency and HF% during exhausting isometric contractions, of different intensities, of the elbow flexors. Signals recorded from the biceps brachii. The behavior of the spectral parameters fit well with the changes in motor units activation strategy during the contraction. (Redrawn from Reference 60.)

during exhausting isometric contractions. The less defined changes in the SMG spectra at 40 and 60% MVC are in agreement with the result of the needle EMG recorded during 60-s 50% MVC in biceps brachii.[55] In this study an increase in number and firing rate of the active MUs in the first 30 s and a reduction of the intramuscular spike activity at the 60-s time mark was observed even if a significant reduction of the mean MUs FR did not occur.

In conclusion, the modifications of the SMG spectra are strongly influenced by the changes of the MUs FR pattern at fatigue. When SMG amplitude and SMG frequency content are combined throughout isometric sustained effort, more reliable information about the changes in both recruitment and mean firing rate of the active MUs can be retrieved from the analysis of the muscular sound. The results reported suggest that combining the spectra of SMG, monitoring mainly the MUs FR, and EMG, monitoring mainly the muscle fibers conduction velocity,[36,54] the muscle activation strategy at fatigue may be well followed.

V. COMPARISON WITH OTHER SIGNALS RELATED TO MUSCLE ACTIVITY

In addition to the force recording, two other signals have been used to study muscle activity, one reflecting the mechanical (physiological force tremor, PFT) and the other the electrical (electromyogram) aspects of muscular contraction. In this chapter a comparison between the SMG and the two other signals will be made in order to identify the common features and the peculiarities of each one.

The PFT was defined as the inevitable result of the asynchronous discharge of motor nerve fibers "converted to mechanical ripples by the muscles".[50] As a consequence the MUs activity can be mostly retrieved by its power spectrum.[2] Indeed, the variation in force output of a muscle, related to the changes in the net activity of the motoneuronal pool (number and firing rate of the active MUs), is reflected in the frequency bandwidth 0 to 6 Hz, while the "unfused parts of the twitch contractions of single motor units" determine the higher frequency content of the power spectrum up to about 30 Hz.[2] In the second component, in the bandwidth 8 to 12 Hz, the "intense synchronous modulation of the motor units spike trains" was also identified.[27] The PFT generating mechanisms cited overlap in the band 6 to 12 Hz determining a local peak that is still preserved when the intensity of isometric contraction increased.[2] The peak frequency has been shown to increase, although not significantly, with force.[38] The peak was also related to the onset FR of the MUs.[2,38] This may explain its relative constancy, in respect to the force, because no great difference in the onset FR can be found between slow- (already recruited at low level of effort) and fast-twitch MUs (recruited at high level of effort).[38] The changes of the SMG in the time and frequency domains previously reported show a behavior, with respect to the exerted force and fatigue, that is similar to that of the force tremor.

One of the improvements in the use of the SMG as a tool to track the MUs FR is due to the fact that it is detected directly over the muscle surface and its frequency content is less affected by the damping due to the biomechanical chain (muscle tendon, joint, and bone segment length) interposed between the contracting fibers and the force detecting point.[2,38] This observation is confirmed by the fact that the degree of the SMG spectra modification is much greater, when force increases from low to high intensity of contraction, than reported in force tremor spectra.

On these bases both PFT and SMG seem to reflect the mechanical events of the recruited MUs. In particular the SMG may reflect the lateral movements and the PFT the longitudinal movements due to the active muscle fibers.

Another specific feature of the SMG deals with the possibility to retrieve the mechanical aspects of the MUs activity from different regions of a muscle or from one single muscle only, even if several agonists are implied in the contraction (this is obviously not possible by the PFT). This allows very careful study of the correlation between the mechanical and the electrical events detected from nearly the same area.

Considering both the surface EMG and the SMG as compound signals in which the MUs FR information may be contained,[15,42,58] an attempt to retrieve this common information was made by cross-spectral[63] analysis and coherence analysis[59] between the two signals. The first study analyzed the cross spectrum (CS) during exhausting isometric contractions of the biceps brachii at 20, 80, and 100% MVC in order to get data from efforts with different MUs activation patterns. Throughout exhausting isometric 25% MVC of the biceps brachii MUs synchronization has been shown to occur and determine the increase in tremor peak in the range 10 to 17 Hz.[31,44] The tremor peak was clearly retrievable in the same frequency band of the EMG spectrum.[31] The SMG-EMG cross spectrum and the low-frequency band of EMG spectrum behaviors are similar, indicating that at 20% MVC the modulation in MUs FR is well tracked by the CS between the two signals.

The already described MUs action pattern changes during sustained 80% MVC are well described by the CS morphology and mean frequency, too. Indeed, CS MF increases in the first 30% of the total contraction time and then decreases up to exhaustion. During exhausting 100% MVC the number of MUs vs. FR histogram changes its frequency bandwidth from 10 to 60 Hz[17] at the onset to 5–25 Hz[18] at exhaustion. This behavior is well monitored by the SMG-EMG cross spectrum shape and by its MF changes (an exponential decrease of 50% from onset to the end of the effort).

The reported results indicate that the SMG-EMG cross spectrum may be a tool to retrieve the dominant MUs FR and the degree of its modulation.

The SMG-EMG coherence analysis at low and high contraction levels revealed that the frequency ranges, in which the common information about MUs FR was contained, were from 11 to 20 Hz (20% MVC) and 15 to 26 Hz (80% MVC). These results fit well with the expected increase in the MUs firing rate ranges from 20 to 80% MVC.[59]

The usefulness of the comparative analysis between SMG and EMG in order to obtain information about the dominant MUs FR is well represented in Figure 17, in which SMG-EMG CS at the onset of 100% MVC is reported together with the number of MUs vs. FR histogram in the same condition.

VI. OVERALL CONSIDERATIONS ON THE ORIGIN OF THE MUSCLE SOUND

From the experimental evidence presented in the previous chapters it can be concluded that at the basis of the muscular sound there is always a change in the geometry of the muscle structures. The way in which this phenomenon takes place is peculiar with respect to the different experimental situations.

Isolated muscle: During single twitch elicited by supramaximal nerve stimulation the lateral displacement of the muscle surface is due to the following two phenomena that, as a consequence, are involved in sound generation: (1) a slow bulk movement of the muscle related to the different regional distribution of contractile tissue; and (2) the excitation into ringing of the muscle at its own resonant frequency due to the forces associated with (1). The first one is poorly evident while the second one is the overwhelming component in sound generation,[6,13,14,26,29] when the signal is recorded by a *hydrophone*.

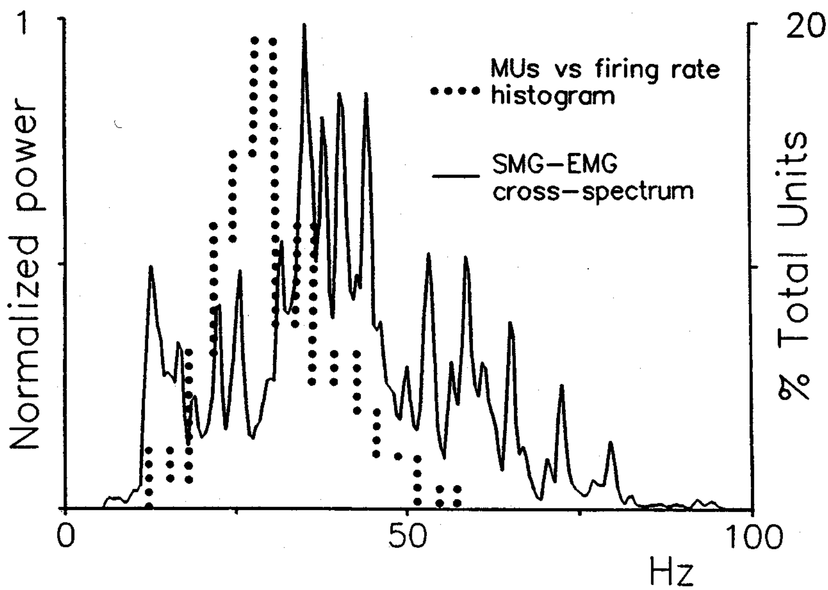

FIGURE 17. Comparison between the biceps brachii number of MUs vs. firing rate histogram (pooled data from four subjects, redrawn from Reference 17) and the normalized average SMG-EMG cross spectrum at 100% MVC onset. The common information of the dominant MUs firing rate contained in SMG and surface EMG, retrieved by the cross spectrum, is close to the one indicated by the histogram derived from the needle EMG. (From Orizio, C. J., *Electromyograph. Kinesiol.*, 2, 141, 1992. With permission.)

In vivo muscle: During single twitch due to supramaximal nerve stimulation the signal recorded by *contact sensors* contains very low frequencies. This suggests that in this condition, in addition to the resonating phenomenon, the slow bulk movement of the muscle can substantially contribute to the sound signal.[12] Moreover, in this case, the synchronous oscillation of the whole muscle as a single unit[71] led some authors to suggest that *the pressure waves, related to the fibers thickening and travelling through the muscle body, can contribute to sound generation.*[19,39] This last mechanism has been proposed also when small bundles of muscle were stimulated.[43] During repetitive stimulation, after the first second of activity, the signal frequency matches the stimulation rate,[71] indicating that, in this condition, the surface oscillation due to dimensional changes in the synchronous fibers may be the main sound source.

Voluntary contraction: The previously cited sound generation mechanisms are present also during voluntary effort. Indeed, during voluntary contraction, transient slow phenomena can be noted at the beginning and at the end of contraction.[56,71] These phenomena have been attributed to the gross dimensional changes of the muscle when it reduces the slack for tension generation at the onset or it relaxes at the end of the effort, respectively. As a consequence, lateral movement may be considered as a muscular sound component when large dimensional changes take place at the onset and at the end of contraction. The confirmation of this hypothesis needs the recording of the "transient phenomena" from different muscle sites in order to check if the relative surface displacements present the same peculiar phase shifts of those recorded during the stimulation of the isolated muscle (see Section III).

On the analogy of the *in vivo* stimulated muscle, subsequently to the initial slow movement of its surface the muscle resonant frequency should determine the signal characteristics. In reality the contribution of the mechanical activity of each recruited motor unit to the sound signal cannot be disregarded. In fact from the signal reported in Figure 18 it can be seen how the summation process of the single MU twitches takes place. Note that the mechanical activity of each MU influences the morphology of the other MU twitches. Moreover, from this simple figure the "compound" nature of this signal is clearly shown. This explains why the time and frequency domain properties of the muscle sound are clearly related to the number, the type, and the firing rate of the recruited MUs.

It can be concluded that during steady voluntary contraction the main sound generation mechanism is related to the summation of the twitching of each individual motor unit. The contribution of other phenomena in the generation of muscle sound as transducers rubbing on the skin[64,66] and blood movement through the muscle[42,64] have been estimated as not significant.

A. Information Contained

During isometric contraction of an isolated muscle the peak frequency of the spectrum of the signal changes throughout the single twitch, paralleling the changes

FIGURE 18. Motor units electrical (upper trace) and mechanical (lower trace) activities. Recordings from the foot extensor digitorum brevis muscle. The mechanical activity was detected by light accelerometer (0.5 g, Entran 125–50 AD) and the electrical by needle electrode. Note the correspondence of the two signals and the "summation" of the mechanical activity when more motor units are active. In the lower trace 20 mV corresponds to 0.125 m/s^2. (The recordings were made by the author and Prof. D. De Grandis at the Neurological Division of the St. Ann Hospital, Ferrara, Italy.)

in the tension or stiffness of the contractile elements. On this basis the time frequency analysis has been suggested as a tool to track this mechanical property of the muscle.[10]

Different latencies were described between stimulus and force or sound onset. This suggests that information about the elastic tissue in series (acting on the force signal) and in parallel (acting on the SMG signal) to the contractile elements may be retrieved by the comparison of the force and SMG traces during single twitch recordings.[19]

The changes of the signal amplitude with the stimulation rate and recruitment provide data on the muscle "motor control mechanisms".[61,71] This view is confirmed by the results of the time and frequency domain analysis of the SMG recorded during voluntary contraction. In fact, the variation in the RMS, integrated value of the signal and spectral characteristics for increasing muscle output force (or at fatigue), parallel the changes in the MU recruitment and firing rate. This information about the motor unit activation pattern is contained in the SMG,[32,42,56,58,71] as well as in other mechanical signals related to muscle contraction force as physiological tremor[2,38] or tangential acceleration.[31] The frequency content of the SMG and of the force tremor presents similar peaks up to 15 Hz, indicating a common generating mechanism in the low frequencies bandwidth.[32,42] This mechanism operates according to the different features of the motoneuronal pool activity such as recruitment, firing rate, and synchronization.[2,30,38] From this point of view high pass filtering, above 14 Hz,[32] of the SMG must be avoided when the MUs activation pattern is to be investigated by muscular sound. For this purpose the SMG is to be preferred to the physiological force tremor, for the reasons reported in Section IV.

In summary, the muscular sound contains information about the state of the contractile and parallel elastic elements and the muscle motor control mechanism.

B. Terminological Problems

From the reported results it is evident that the terms muscular sound, soundmyogram, phonomyogram, and acousticmyogram should be avoided; their existence is due only to historical heritage. This consideration is based on the fact that whatever is the experimental condition in which the signal is recorded the greater part of the power is below the audible range of the human ear. The name accelerometermyography[46] underlines the transducer used to detect the signal. It should be avoided since it does not indicate the nature of the "myography". The term vibromyography appeared most useful and appropriate. In fact it indicates clearly the kind (vibratory) of the phenomenon detected. Unfortunately the "vibromyogram" is a tool mainly used to elicit spindle activity; as a consequence, confusion can easily be created between the two vibromyograms generated "externally to the muscle" or "by the muscle". Following the above considerations a more appropriate and not misleading term for the definition and the identification of the signal has to be suggested. First of all it has to indicate its mechanical nature. For this purpose the combining form "mechano" has to be considered. The resulting term "mechanomyogram" could be fruitfully used without problems of misunderstanding or lack of accuracy in the identification of its origin.

VII. PRACTICAL APPLICATIONS

As reported in Table 1, the recording of muscular sound can be used to estimate the electromechanical delay. This parameter was used to study the effect of denervation in patients with peripheral nerve or root lesions.[40] "The electromyographic diagnosis of acute denervation relies largely upon the finding of fibrillation potentials", this sign appears only 2 to 3 weeks after the noxious event affected the nerve. In the patients investigated the amplitude of the "lateral sound wave", elicited after stimulation at the motor point, appeared to be reduced when compared with the contralateral unaffected muscles. Such reduction was already evident at the sixth day. The AA[40] suggested that muscular sound "is useful in assessing the functional state of the terminal endings and end-plates".

The effect of muscle atrophy on the properties of the phonomyogram have been studied in patients submitted to joint immobilization because of bone fractures.[48] With respect to the control limb, after cast removal, the frequency content of the SMG during maximal voluntary contraction recorded by a piezoelectric contact sensor showed a clear reduction in the high frequency content with a peak frequency of 13 Hz. A reduction was also observed in EMG spectral content. The AA[48] attributed this result to "some deteriorations of the ability of the motor centers to recruit motor units and/or to fire them at higher frequencies". Changes in the peripheral motoneuron activation pattern and in electrical and mechanical characteristics of the muscle fibers were also suggested to participate in the signals alteration.

During voluntary contractions of the elbow flexors of children affected by muscle disease the ratio between EMG and AMG amplitude, at different effort levels, is statistically increased with respect to the normal subjects,[11] indicating a reduction of the electromechanical efficiency. This may be attributed to "atrophic fibers that generate electrical activity with little mechanical contribution". The AMG allows collection of data on the muscle mechanical activity even in those muscle groups in which force measurements are difficult, i.e., in paraspinal muscles that are often involved early in myopathies, and to estimate their electromechanical efficiency. According to these AA[11] the fact that when the muscle fibers are active at tetanic firing frequency no sound is emitted, can be used to an advantage. In fact the silence of the tetanized MUs offers the possibility to resolve acoustically the late recruited MUs in isolation.

The recording of muscular sounds from the biceps brachii of patients with neuromuscular disorders showed significantly lower frequency content in respect to normal subjects under 60 years. Both aging and myopathies seem to affect the spectra of the SMG with a shift towards the low frequencies.[66]

Recently the reliability of the AMG, compared with the EMG, in the assessment of the lumbar paraspinal muscle function[47] has been investigated. It was shown that AMG can be used as a good indicator of the mechanical aspect of contraction also in patients with chronic low back pain.

Acoustic myography, detected by an air-coupled microphone, was used also as a control signal for externally powered prosthesis.[5] The advantages with respect to the EMG are the following: no need for direct contact with the skin, no influence of the skin impedance changes, high output voltage with fewer amplification problems and a less critical placement of the transducer on the muscle surface. The sensitivity of the microphone to the ambience vibration can be electronically overcome.

The application of the muscular sound in sport medicine may contribute to obtaining data on the muscle fiber typing. This role of the SMG is supported by the fact that its frequency content is shifted toward the higher or the lower frequencies when muscles with a prevalence of fast- or slow-twitch muscle fibers are studied, respectively. Indeed the muscular sound dominant frequency was greater in orbicularis oris (about 22 Hz) than in soleus (about 11 Hz).[53] The direct stimulation of the vastus lateralis and of the soleus resulted in phonomyographic "spikes" with statistically different time to peak and median frequency for vastus lateralis (39 to 55 msec; 5.34 to 6.26 Hz) and for soleus (99 to 115 msec; 3.74 to 4.70 Hz).[49]

A confirmation of the relationship between the muscle fiber typing and the properties of the SMG has been found recently comparing the data from the vastus lateralis of sprinters, long distance runners, and sedentary subjects during maximal voluntary contraction.[62] In particular the HF% (the spectrum relative power in the 15- to 35-Hz band, i.e., the fast-twitch MUs firing rate range) was found to be about 74, 39, and 54% in the three groups, respectively. These values fit well with the relative area of the fast-twitch fiber MUs reported in the investigated groups from biopsies of the vastus lateralis. Power training seems to induce a shift of the phonomyogram power spectrum towards the higher frequencies as an enrichment of the fast-twitch fibers take place.[23]

In conclusion it resulted that the SMG properties are influenced by the ratio between the area of the fast- and slow-twitch MUs. Both the corresponding MUs mechanical twitch and their firing rates can contribute to the spectral changes in SMG.

VIII. FINAL CONSIDERATIONS

The results reported in this review, even if related to a biological signal still in its "infancy", indicate that, besides the electromyogram, another important signal with a mechanical origin can be detected from a contracting muscle. The whole body of published work on the topic up to now can be considered to be the fundamentals on which the physiological and the biophysical meanings of the mechanomyogram are grounded.

On these bases two main research fields are now open to investigation by the mechanomyogram. The first one deals with the biomechanics of the muscle. Classic recordings of the myograms of the single twitch, the clonus, and the tetanus or the definition of the length-tension and the force-velocity relationships can be reinvestigated by this mechanical signal. The most important feature of this new signal is that these studies can be made on *in vivo* muscles and also on those muscles in which force cannot be estimated.

The comparison between the force recording and the mechanomyogram detected at the muscle surface may allow collection of information on the properties of the components of the mechanical model of the muscle, the influence of the muscle fiber orientation (degree of pennation), as well as on the role of the tendon and joint in damping the transmission of the force generated by the contractile elements.

The second research field deals with the necessity to fully understand the mechanism by which the mechanomyographic signal is generated from the summation of the mechanical activity of each recruited motor unit. This aspect will need the recording of the mechanomyogram together with the needle or the surface EMG. This will allow the processing of the mechanical signal on the basis of the muscle motor control scheme. The aim is to provide proofs of the reliability of a noninvasive tool that by comparison of the mechanomyogram and of the surface electromyogram could give information on the motor unit activation pattern. This data will be very useful in the primary or secondary myopathies diagnosis and follow-up as well as in rehabilitation.

Moreover, the possibility of describing electrical and mechanical activity of even the single motor unit will be very useful for estimation of the distribution of the different MU types in skeletal muscle and for the characterization of the MU changes in neuromuscular diseases.

It is evident that, because of the complexity of the task, an interdisciplinary approach is needed in future studies on the mechanomyogram. The physiologist, the clinician, as well as the physicist and the bioengineer, will have to join their competencies in order to reach the indicated theoretical and practical goals.

In conclusion, a large research field has been opened by the earlier studies on "muscle sound", from the data we have at this time it can be expected that future fruitful results will be reached for the knowledge of the physiological and clinical aspects of muscle mechanical activity.

ACKNOWLEDGMENTS

The author is grateful to Prof. M. Solomonow for his helpful suggestions and encouragement throughout the months this work has taken, to Prof. R. Merletti for his rigorous and fruitful objections, to Dr. D. T. Barry for the deep discussion we had on many aspects of "muscle sound", to Dr. M. J. Stokes for her accurate comments, and to Prof. A. Veicsteinas who made possible and promoted "muscle sound" studies at our Institute.

This work was supported by Consiglio Nazionale delle Ricerche (CNR, grant 92.00386.PF40), and by "Centro di Studio e Ricerca di Fisiologia Muscolare e dello Sport", University of Brescia, Italy.

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