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THE RELATION BETWEEN THE MYOELECTRIC SIGNAL AND PHYSIOLOGICAL PROPERTIES OF CONSTANT-FORCE ISOMETRIC CONTRACTIONS

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In the past, various parameters of the myoelectric (ME) signal have been investigated to gain insight into the physiological state of muscle or its output. Two methodologies have prevailed in obtaining these parameters. The first approach has been to calculate various parameters of experimentally obtained ME signals (Lippold 1952; Currier 1969; Kuroda et al. 1970; Vredenbregt and Rau 1973). The second approach has employed mathematical modelling of the ME signal under various specific assumptions of the properties of the constituent motor units (Bernshtein 1967a, b; Moore 1967; Person and Libkind 1967; Libkind 1968, 1969; Coggshall and Bekey 1970; Lindström et al. 1970; Graupe et al. 1973; Brody et al. 1974; Kreifeldt and Yao 1974; Agarwal and Gottlieb 1975; Le Fever and De Luca 1976).

De Luca (1975) differed in his approach to modelling the signal obtained from an individual motor unit recorded during a sustained constant-force isometric contraction. In this model an empirically derived expression for the generalized firing rate of motor units (De Luca and Forrest 1973a) was employed. De Luca and Van Dyk (1975) extended the model for signals recorded from single motor units to the interference ME signal. The expressions were used to generate theoretical curves for these parameters. Some assumptions had to be invoked. However, the empirical result for the firing rates of motor units observed by De Luca and Forrest (1973a) was used. Hence, these curves reflected the effect of firing rate on the ME signal.

Several physiological properties of contracting muscle are known to affect the ME signal. These correlates can be associated with the terms in the expressions which reflect their effect on the parameter. In this study, a comparison between mathematical expressions and empirically obtained parameters was used to determine the relative magnitude of effects of the physiological correlates on the ME signal.

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Model and physiology

General considerations

The expressions derived by De Luca and Van Dyk (1975) are based on the model which represents the ME signal as a linear, spatial and temporal summation of the signals generated by the constituent motor units. A schematic representation of the model is shown in Fig. 1. De Luca (1975) modelled the signal generated from a motor unit as a Dirac delta impulse train with random inter-pulse intervals that is passed through a linear filter. The impulse response of the filter $h_i(t)$, represents the motor unit

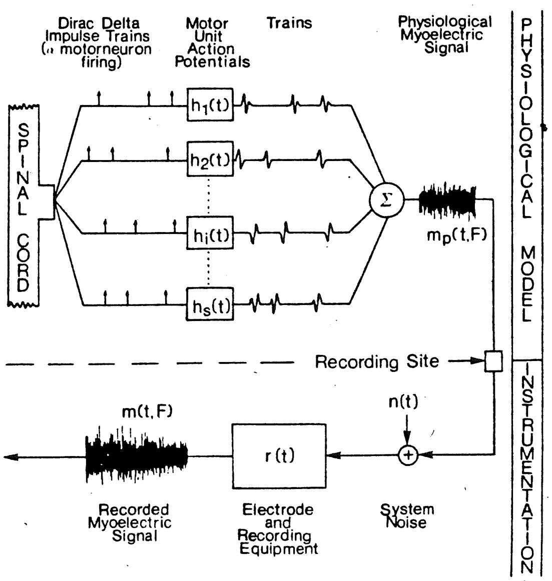


Fig. 1. Schematic representation of the model for the generation of the myoelectric signal.

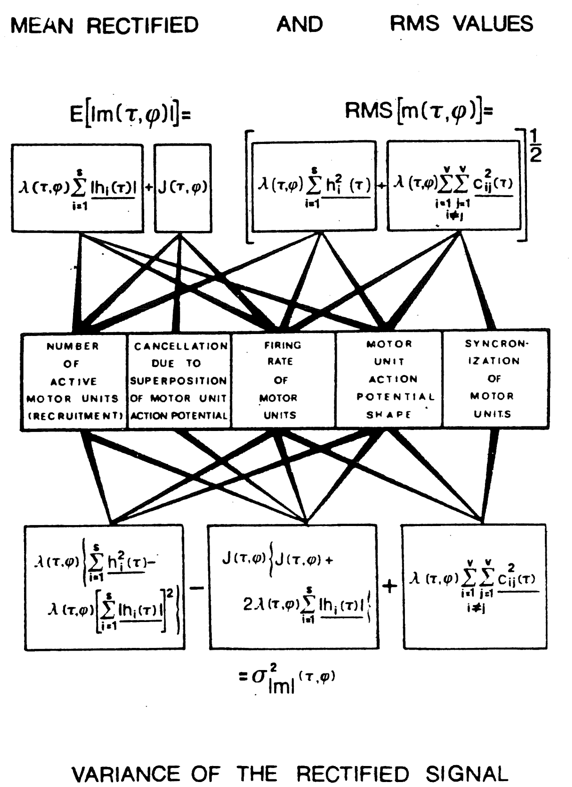


Fig. 2. Theoretical expressions for parameters of the myoelectric signal and their relation to physiological correlates of a contracting muscle. The solid lines indicate that the physiological correlate has an explicit effect on the boxed set of terms in the parameter.

action potential (MUAP) generated by the i^{th} motor unit. The output of the filter is a sequence of MUAPs and represents a motor unit action potential train (MUAPT). These signals generated from all the active motor units superimpose at the recording site to form the physiological ME signal, $m_p(t, F)$ which is a function of both contraction time, t, and force, F. When the signal is detected, an electrical noise, n(t), is added to the signal. This noise is that signal which appears when no motor units are firing. The signal, composed of $m_p(t, F)$ and n(t), is then passed through a filter, r(t), which accounts for the filtering effects of the recording electrode and the instrumentation. The resulting signal, m(t, F), is the observable ME signal. By using this concept, De Luca and Van Dyk (1975) derived expressions for the mean rectified and root mean squared (rms) values of ME signals recorded during a sustained constant-force isometric contraction.

Mathematical expressions

The expressions for the mean rectified and rms values are shown in Fig. 2. The explanation of some of the terms is presented in Fig. 3. The derivation of these expressions assumes that (a) the noise, n(t), is negligible and (b) the effect of the recording electrodes and instrumentation remain constant with time. These two considerations can be realized with proper experimental procedures. In the expressions, time is normalized by the duration of a constant-force isometric contraction which is sustained until a subject can no longer maintain the force level. Normalized contraction time is denoted by τ. The contraction force is normalized by the force of the maximal voluntary contraction (MVC) which could be elicited by a subject and is represented by ϕ. These normalizations are

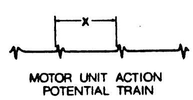
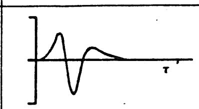
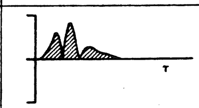
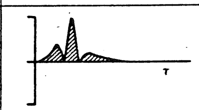
TERM	DIAGRAM	EXPRESSION
a) GENERALIZED FIRING RATE OF A TYPICAL MOTOR UNIT		$\lambda(\tau,\phi) = \frac{1}{E(X)}$
b) MOTOR UNIT ACTION POTENTIAL		$h_i(\tau)$
c) AREA UNDER THE RECTIFIED MOTOR UNIT ACTION POTENTIAL		$\underline{\lvert h_i(\tau)\rvert} = \int_0 \lvert h_i(\tau)\rvert d\tau$
d) AREA UNDER THE SQUARE OF A MOTOR UNIT ACTION POTENTIAL		$\underline{h_i^2(\tau)} = \int_0 h_i^2(\tau)\, d\tau$

Fig. 3. Extended representation of some of the terms of Fig. 2.

necessary so that results obtained from different subjects can be compared meaningfully. The recorded ME signal can be expressed as a function of normalized contraction time and the normalized force by denoting it as $m(\tau, \phi)$. The generalized firing rate, which is defined as the average firing rate of a typical motor unit (De Luca and Forrest 1973a), is represented by $\lambda(\tau, \phi)$, as shown in Fig. 3a. The number of active motor units is denoted by s. The term $\lvert h_i(\tau)\rvert$, shown in Fig. 3c, represents the area under the rectified MUAP, $h_i(\tau)$, in Fig. 3b. The term $\underline{h_i^2(\tau)}$, shown in Fig. 3d, represents the area under the square of a MUAP.

In Fig. 2, the mean rectified value is shown to be a sum of two sets of terms. The first boxed set of terms is the sum of the individual mean rectified values of the constituent MUAPTs. By the Triangle Inequality, the mean rectified value must be less than or equal to the sum of the individual mean rectified values. The second boxed term, $J(\tau, \phi)$, is defined as the superposition term because it accounts for the cancellation in the signal due to the superposition of opposite phases of the MUAPTs.

The rms value is the square root of the two sets of terms as seen in Fig. 2. The first boxed set of terms is the sum of the individual mean-squared values of the MUAPTs. The second box contains a double sum of terms, $\underline{c_{ij}^2(\tau)}$, which accounts for the cross-correlation between any two (i^{th} and j^{th}) correlated MUAPTs. This set of terms is collectively referred to as the synchronization term. The number of correlated MUAPTs is denoted by υ which must be less than or equal to s. Note that in this study, synchronization is defined as dependence between the firings of different MUAPTs. Hence, cross-correlation is a sufficient, but not exclusive condition for synchronization. Thus, if the second boxed set of terms is non-zero, synchronization is present; but if it is zero, there is no correlation, though synchronization may exist. When cross-correlation is not present, the rms value of the interference ME signal is the root of the sum of the individual mean-squared values of the constituent MUAPTs.

Working with the described model, Stulen[1] derived a third parameter, the variance of the rectified ME signal. As seen in Fig. 2, this parameter is represented by 3 boxed sets of terms. The first is the variance of the rectified signal if there is neither cancellation nor cross-correlation of MUAPTs present in the ME signal. The second boxed set of terms accounts for cancellation of signal. If the superposition term, $J(\tau, \phi)$, is zero then this set of terms is zero. The third box contains the synchronization term.

Physiological correlates

Five physiological correlates of contracting muscle are presented in Fig. 2. These are: (a) the firing rate of motor units, (b) the number of active motor units, (c) the synchronization of motor units, (d) the cancellation of signal due to the superposition of MUAPTs

[1] Stulen, F.B. Analysis of the time dependence of myoelectric signal parameters based on a stochastic model of motor unit properties. M. Sc. Thesis, Massachusetts Institute of Technology, 1975, Cambridge, Mass., U.S.A.

when opposite phases superimpose, and (e) the shape of the MUAP. The terms that these properties affect are indicated by the tapered lines. These properties account for the major physiological effects that are presently known to affect the ME signal recorded during a sustained constant-force isometric contraction. The behavior of other intrinsic properties of motor units will be reflected in these correlates.

De Luca and Forrest (1973a) showed that the generalized firing rate is a monotonically decreasing function of contraction time for a sustained constant-force isometric contraction. The generalized firing rate was also shown to be a monotonically increasing function of the force at which the contraction is performed. Gilson and Mills (1941), Masland et al. (1969) and De Luca and Forrest (1973a) have indicated that a motor unit which is active at the beginning of a sustained constant-force isometric contraction appears to remain active throughout the contraction. Hence, the number of active motor units, s, is not likely to decrease during a contraction, but, may conceivably increase. An increase in the number of active motor units during a contraction is known as recruitment. The issue of recruitment during a constant-force isometric contraction is not yet resolved. Edwards and Lippold (1956), Vredenbregt and Rau (1973) and others have postulated that time-dependent recruitment of motor units should occur during a sustained constant-force isometric contraction in order to maintain the force as the muscle fatigues. However, irrefutable proof is yet to be presented.

Evidence of synchronization, as defined earlier, has been presented by several investigators (Buchthal and Madsen 1950; Dietz et al. 1976 and others). Other investigators (Lippold et al. 1960; Missiuro et al. 1962) have claimed to observe synchronization by noting the appearance of large periodic oscillations in the ME signal as the muscle fatigues.

Cancellation of signal due to the superposition of opposite phases of MUAPs is an inevitable process that occurs before the ME signal is detected. The superposition term, $J(\tau, \phi)$ accounts for this loss of signal in the expression for the mean rectified value.

The motor unit action potential shape depends on anatomical and physiological properties of the muscles, and the electrodes used to record the signal. The duration of the action potential is dependent on the conduction velocities of the muscle fibers which compose the motor unit. The potential propagates through the interstitial fluid and intervening tissues such as fascia, fat and skin. These tissues have electrical properties that affect the shape of the action potential. Lindström [2] has modelled the dependence of the shape and amplitude of the action potential on the distance between the fiber and the recording site. The action potential of a muscle fiber near the recording site has more high frequency components than that of a more distant fiber. In the same study, Lindström [2] showed that the action potential is essentially low-pass filtered when the signal is recorded differentially with the recording contacts of the electrode pair located parallel to the muscle fibers. The amount of filtering is mainly determined by the separation of the recording contacts of the differential electrode and the conduction velocities of the muscle fibers. Because action potentials from all the muscle fibers that compose the motor unit superimpose to form the MUAP, the shape of the MUAP is influenced by all the above effects. The shape is further modified by the frequency characteristics of the electrodes and possibly those of the instrumentation.

Method and material

Experiment

Ten males and one female volunteered for

[2] Lindström, L. On the frequency spectrum of EMG signals. Thesis, Res. Lab. Med. Electronics, Chalmers University of Technology, Göteborg, Sweden, 1970.

the experiment. Their ages ranged from 20 to 27 years with an average and S.D. of 24 ± 2 years. Each subject was placed in a horizontal supine position on a table. A cuff was secured proximal to the right elbow. The cuff was attached to a force gauge by a flexible steel cable with a variable-length section of chain. The force gauge was mounted on a foot platform attached to one end of the table. The length of the chain was adjusted so that the angle between the upper limb and the intersection of the coronal and medial sagittal planes was in the range of 45—56°. De Luca and Forrest (1973b) showed that in this position the force output of an abduction of the upper limb is approximately a linear sum of the force contributions of the anterior, middle and posterior fibers of the deltoid and the supraspinatus when the arm is medially rotated. A strap was secured around the legs proximal to the knees. This kept the knees slightly hyperextended to prevent any flexions and extensions about the knees. These movements could have changed the stress on the force gauge as the subject pushed against the foot platform.

Since the parameters of the ME signal inherently depend on the type of recording electrodes, two types of differential pairs were used during each contraction. Two Beckman surface electrodes were attached on the skin above the middle fibers of the deltoid by adhesive stubs 20 min prior to the contraction. In this time, the impedance properties of the electrodes stabilized. These electrodes were placed parallel to the middle fibers and slightly posterior to the center of the muscle. The surface electrodes were separated by approximately 2 cm. The second differential electrode was a DISA (13K80) bipolar needle electrode. The needle was inserted into the middle fibers to a depth of approximately 1 cm. The location of the electrode was equally distant from the two surface electrodes and slightly anterior to the center of the deltoid. The wires of the electrode were separated by approximately 0.5 mm. The differential electrode pairs were connected to two preamplifiers. The cannula of the bipolar needle served as common ground for both types of electrodes. The amplifier connected to the surface electrodes was DC coupled and the amplifier connected to the bipolar needle electrode was AC coupled with the 3 dB point of the filter set at 4 c/sec. The high-frequency 3 dB point of both preamplifiers was set at 10 kc/sec. The two amplified ME signals and the force gauge output were recorded on FM tape.

The force gauge output was amplified and connected to one channel of a dual-trace oscilloscope. The sweep speed of the oscilloscope was relatively fast so that the force gauge output appeared as a line across the screen.

The experimental procedure required that each subject perform sustained constant-force isometric contractions at 25%, 50% and 75% MVC. Each contraction was performed on separate days, with at least 3 days between each session. In order to establish the absolute value of the required force levels for each session, it was necessary to measure the force output of the MVC. Thirty minutes prior to each sustained contraction, the subject was instructed to perform a short (less than 3 sec), maximal abduction of the upper limb in the coronal plane with the arm medially rotated. The subject was urged to contract with as much force as possible. The peak force was considered to be the value of the MVC for that session.

For each session, the trace of the second channel of the dual-trace oscilloscope was grounded and positioned to indicate the required force level, while the first channel displayed the output of the force gauge. Each subject was instructed to abduct the upper limb in the coronal plane while keeping the arm medially rotated, and to align the two traces on the oscilloscope as quickly as possible. The subject was required to maintain the force output at a constant level as long as possible. When the force output declined or had large variations, the contraction was terminated. At the end of the contraction, the electrodes and the force gauge cuff were removed.

Analysis of data

Although each subject was required to maintain the force output constant, each force record had an initial jump from zero to the desired level, with occasional overshoot. As the duration of the contraction increased, the subjects experienced increasing difficulty in maintaining the force output stable. A representative and comparable measure of the beginning of the contraction was obtained by noting the time that the actual force level entered the range of 95—105% of the desired force. The termination of the contraction was designated as the time when actual force value consistently crossed the 95—105% range. The difference between these two measurements supplied a representative and comparable measure of the duration of the contraction. The average durations of the contractions performed at 25%, 50% and 75% MVC are presented in Table I.

Once the duration of the signal was defined, the recorded ME signal was passed through a 48 dB per octave bandpass filter. If any movement artifacts were present in the signal, then the low-frequency cutoff was set to remove these relatively low frequencies. The high-frequency cutoff of the filter was determined by the type of electrode used to record the signal. Kadefors et al. (1968) showed that a coaxial needle electrode records measurable signal frequency components up to 1500 c/sec. Their data show that the amplitude of this frequency component is only 2% of the greatest frequency components. Hence, the signal energy above 1500 c/sec is negligible. In this study, the high-frequency 3 dB point of the filter was set at 2000 c/sec. De Luca[3] showed that a differential pair of Beckman surface electrodes separated by approximately 2 cm would record measurable signal frequency components up to 300 c/sec. For signals recorded with surface electrodes, the high-frequency 3 dB point was set at 333 c/sec.

The filtered ME signal was sampled at a rate twice the high-frequency 3 dB point of the filter and stored on a computer disk. Nonrecursive linear digital filtering was used to calculate the mean rectified and rms values of the sampled ME signal. Each point in the averaging window was equally weighted. The amplitudes and contraction times of both parameters were normalized by their respective maxima in order to compare results obtained from different subjects. Since contraction time was normalized, the frequency components of the normalized parameters were scaled by the reciprocal of the duration of the contraction. To avoid this frequency scaling, the averaging window of the digital filter was determined by the duration of the contraction. The number of points in the window was 9% of the total number of samples of an ME signal. This limited the greatest apparent frequency in the normalized parameters to 5 cycles per duration of the contraction. Hence, the parameter could be calculated from 9% to 100% of normalized contraction time. The initial 9% of a parameter was less than the actual value. The values of the parameter in the initial window were obtained by multiplying them by 9% and dividing by the point in normalized contraction time

TABLE I

Average durations of sustained constant-force isometric contractions. The duration of the contraction is defined from the time the force initially entered 5% limits about the desired constant force until the time the force output consistently crossed these limits.

Force level % of MVC	Average duration (sec)	S.D. (sec)
25	476	184
50	102	55.9
75	42.4	19.1

[3] De Luca, C.J. Myo-electric analysis of isometric contraction of the human biceps brachii. M. Sc. Thesis, University of New Brunswick, Fredericton, N.B., Canada, 1968.

where they occurred. This then allowed the parameter values to be calculated from 2% to 100%. There was too much variation in the value of the parameter when this technique was used for contraction time less than 2%. Between 0% and 2% the parameters were set equal to the value at 2% of normalized contraction time.

The normalized mean rectified and rms values of the ME signals, recorded with the same type of electrode and at the same force level but from different subjects, were averaged and renormalized. The normalized average rms and mean rectified values are plotted as a function of normalized contraction time in Figs. 4 and 5 respectively. The heavy lines in these figures represent the average and the

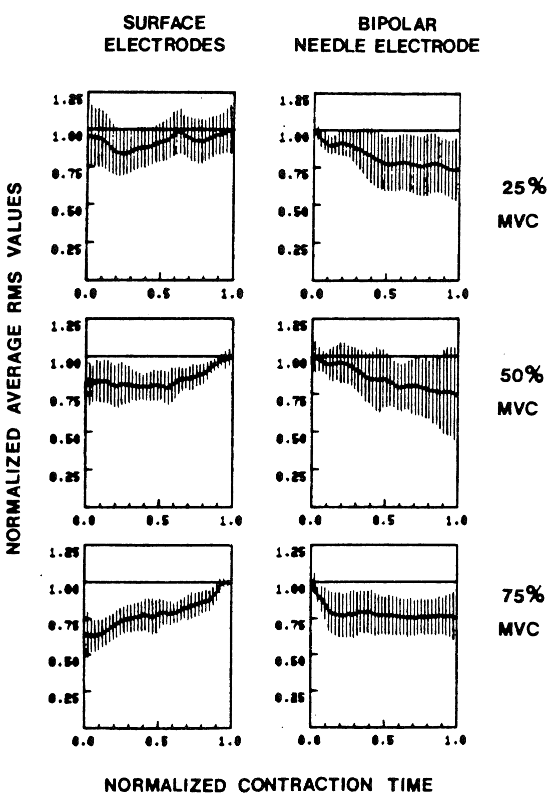


Fig. 4. Average of the normalized root mean squared (rms) values from each subject, plotted as a function of contraction time. Both the amplitude and time duration are normalized to their respective maxima. The vertical lines indicate 1 S.D. about the average.

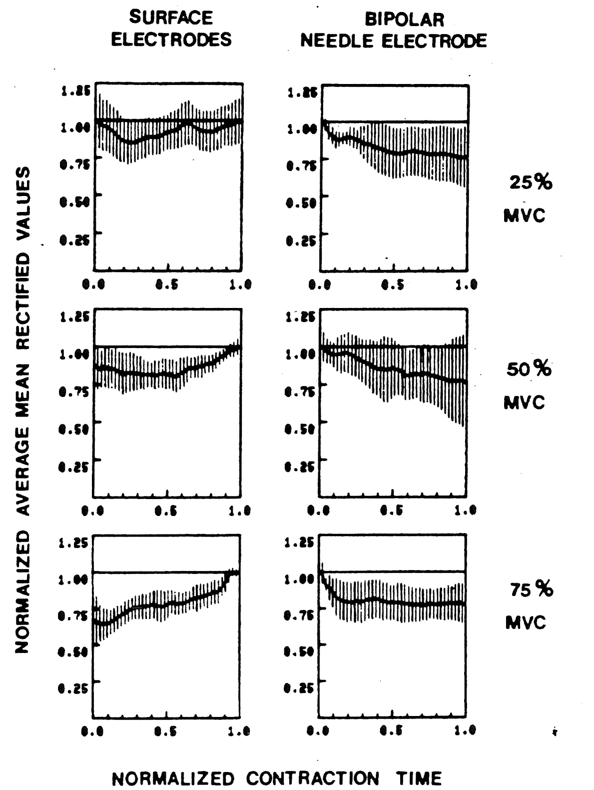


Fig. 5. Average of the normalized mean rectified values obtained from each subject, plotted as a function of contraction time. Both the amplitude and time duration are normalized to their respective maxima. The vertical lines indicate 1 S.D. about the average.

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The ratio of the square root of the variance parameter to the mean rectified value was calculated and averaged for signals recorded with the same type of electrode and at the same force level but from different subjects. This parameter represents the ratio of the AC voltage to the DC voltage of the rectified signal and is referred to as the voltage ratio. The average of the voltage ratio is plotted as a solid line in Fig. 6. The vertical lines in this figure represent 95% confidence limits. The horizontal dashed line indicates what the value of this parameter would be if the instantaneous amplitude of the ME signal was

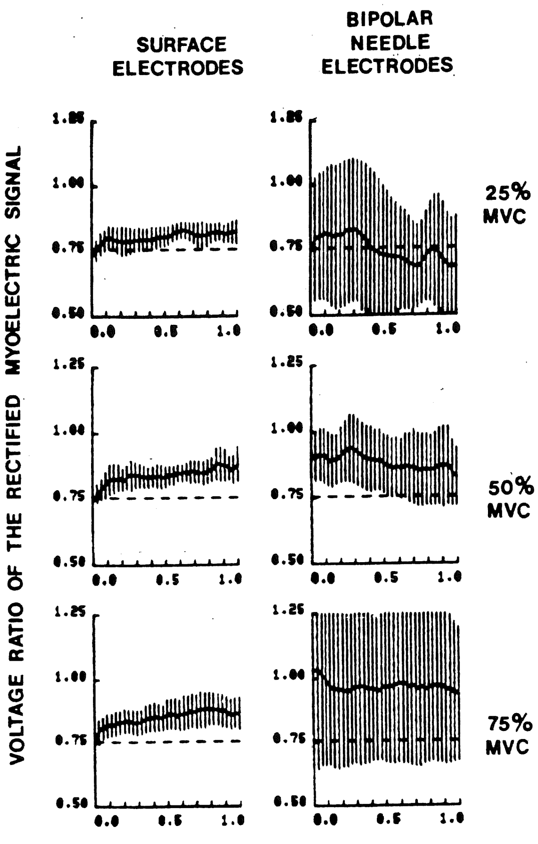

Fig. 6. Average voltage ratio of the rectified myoelectric signal from each subject plotted as a function of normalized contraction time. The voltage ratio is the ratio of AC voltage to DC voltage of the rectified signal. The vertical lines represent the 95% confidence limits. Note that the lines representing the confidence limits are clipped at 1.25 and 0.50. The horizontal dashed line represents the ratio value for a signal having a Gaussian-distributed amplitude.

distributed as a Gaussian variable. The value is 0.75551 and its derivation is presented in the Appendix. The significance of this value is discussed in the next section.

Third order polynomial regression analyses were performed on the normalized-average mean rectified and rms values, and the average voltage ratio. The results are presented in Table II.

Results and discussion

The following discussion examines the time and constant-force dependence of the empirical parameters of the ME signal. Changes in the parameters will be interpreted as time and constant-force dependent effects of the 5 physiological correlates presented earlier.

If the time and constant-force dependence of these parameters were quantitatively known, then their effects on the parameters could be determined by examining the equations of Fig. 2. These equations could then be employed to generate theoretical curves which would reflect the actual time and constant-force dependence of these physiological correlates. The difference between the theoretical curves and the empirical curves would then be determined by the effect of the remaining physiological correlates. This approach allows the elimination of the effects of the physiological correlates which are quantified; hence it further determines the effects of the remaining correlates. Unfortunately, in the field of electromyography, few of the effects of the physiological correlates are explicitly known. However, qualitative descriptions of the effects of these physiological correlates can be found in the literature. This additional information allows the time and constant-force dependence of the difference between the theoretical and empirical parameters to be discussed in terms of the magnitude of the effect of the remaining correlates.

Rms and mean rectified values of the myoelectric signal

Bipolar needle electrode recordings. The average normalized rms values for signals recorded with a bipolar needle electrode decrease with normalized contraction time, as seen in Fig. 4. De Luca and Forrest (1973a) showed that the firing rate of motor units monotonically decreases as a function of contraction time. De Luca (1975) used the empirical measurements of the firing rate and the equation in Fig. 2 to obtain a theoretical

TABLE II

Third order polynomial regression coefficients for the average data of Figs. 4, 5 and 6 as a function of normalized contraction time, τ.

$$\text{Parameter} = \sum_{i=0}^{3} a_i \tau^i \qquad 0 \leqslant \tau \leqslant 1.$$

Parameters	Surface electrodes				Bipolar needle electrode				Force level
	Regression coefficients				Regression coefficients				
	a_3	a_2	a_1	a_0	a_3	a_2	a_1	a_0	
Normalized mean rectified value	−1.062	1.859	−0.8303	1.000	−0.2164	0.5536	−0.5399	1.000	25% MVC
Normalized rms value	−1.098	1.900	−0.8199	0.9986	0.0560	0.1474	−0.4294	1.000	
Voltage ratio of the rectified signal	−0.04601	0.01554	0.07401	0.7735	0.7171	−1.042	0.2522	0.7948	
Normalized mean rectified value	0.1937	0.2074	−0.2629	0.8618	*	0.1376	−0.3689	1.000	50% MVC
Normalized rms value	0.3282	−0.0176	−0.1323	0.8217	0.2082	−0.1653	−0.2797	1.000	
Voltage ratio of the rectified signal	0.3235	−0.5363	0.3182	0.7794	0.4561	−0.7190	0.2332	0.8932	
Normalized mean rectified value	1.207	−1.696	0.9105	0.5784	−0.4456	0.9352	−0.5163	1.000	75% MVC
Normalized rms value	1.156	−1.651	0.9342	0.5609	−0.5425	1.087	−0.5922	1.000	
Voltage ratio of the rectified signal	−0.1582	0.1381	0.07721	0.8075	−0.6674	1.075	−0.5044	1.029	

* The third order polynomial regression did not significantly improve the fit of the curve over second order regression.

curve for the rms value. To generate this curve the following assumptions were made: (a) no recruitment occurred during a constant-force contraction, (b) the area under the MUAPs did not change, and (c) there was no cross-correlation of motor units. With these assumptions, the rms value is proportional to the square root of the firing rate.

The average rms values of Fig. 4 and the theoretical rms values are shown in Fig. 7. For force levels of 25% and 50% MVC, the two values are in close agreement throughout the complete contraction. From this comparison, it can be seen that the average firing rate of the constituent motor units has a significant and dominant effect on the amplitude of the signal when recorded with a bipolar needle electrode. Furthermore, the comparison strongly indicates that during sustained constant-force isometric contractions of 25% and 50% MVC, there is no significant time-dependent recruitment or synchronization with cross-correlation. For contractions performed at 75% MVC, other physiological correlates affect the signal.

The force of contraction was maintained at a constant level throughout the duration, even though the average firing rate of motor units decreased as the contraction progressed. Thus, a compensatory mechanism must have occurred to maintain a constant-force level. One mechanism which could account for this observation is potentiation of twitch tension of the motor units as a contraction progresses. Recently, evidence for potentiation of twitch tension caused by sustained repetitive stimulation has been presented by Gurfinkel' and Levik (1976) in situ in the human forearm flexors, and by Burke et al. (1976) in vivo in the cat gastrocnemius.

The average normalized mean rectified value for signals recorded with a bipolar needle electrode also decreases with normalized contraction time. The decrease in the mean rectified and rms values is very similar as seen in Figs. 4 and 5. Milner-Brown and Stein (1975) also observed that the mean rectified and rms values were approximately

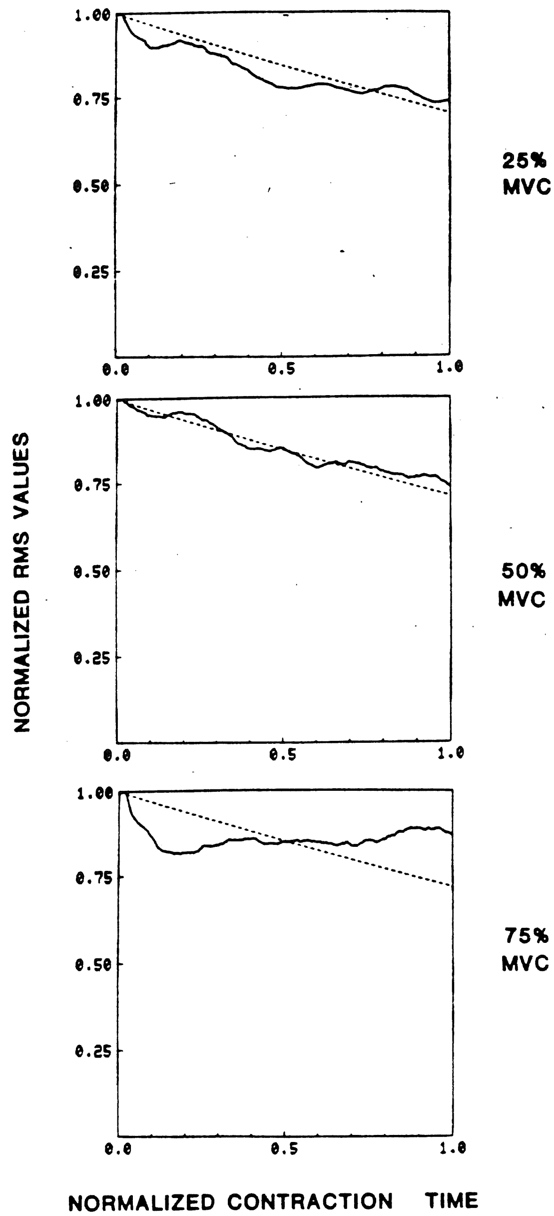

Fig. 7. Comparison of theoretical and average curves for the rms value. The solid line represents the normalized average rms curves shown in Fig. 4. The dashed line represents the theoretically derived rms value. See text for further details.

proportional over a wide range of force for ME signals recorded with surface electrodes from the first dorsal interosseus muscle. Thus, the mean rectified value at 25% and 50%

MVC also decreases as the square root of the firing rate. If the sum of the mean rectified values of the individual MUAPTs dominated, the mean rectified value would decrease with the firing rate, as seen in the first set of terms in Fig. 2. This is not the case. Therefore, the superposition term must have a significant effect, for the mean rectified value to be approximately proportional to the rms value. If the assumptions used to calculate the theoretical expressions of Fig. 7 are invoked, then the magnitude of the superposition term can be expressed as follows:

$$J(\tau, \phi) = K_1\lambda^{1/2}(\tau, \phi)[1 - \lambda^{1/2}(\tau, \phi)K_2] .$$

K_1 and K_2 are factors which depend on the proportionality, the number of MUAPTs and the area under the MUAPTs.

The mean rectified and rms values obtained from the needle electrode for contractions performed at 75% MVC tend to decrease initially (Figs. 4 and 5). This can be accounted for by the effect of the firing rate. During the last 80% of the contraction, these parameters tend to remain constant. However, the firing rate is monotonically decreasing throughout a contraction performed at 75% MVC (De Luca and Forrest 1973a). Hence, in this case there is a relative increase of the empirical parameters values in contrast to the effect predicted by a decreasing firing rate. This implies that other physiological correlates must be considered during a 75% MVC.

Surface electrode recordings. A significant increase in the mean rectified and rms values, in spite of the effect of a decreasing firing rate, is observed when signals are recorded with surface electrodes as seen in Figs. 4 and 5. The decrease of firing rate is a property of motor units that is not dependent on the type of electrode used to record the ME signal. A decrease in firing rate would also tend to decrease the values of these curves. The remaining physiological correlates must counteract the decrease and account for the observed increase. The increase in the curves is similar for both the mean rectified and rms values. The greatest increase is observed for parameters recorded with a pair of surface electrodes at 75% MVC. This increase must be accounted for by the remaining physiological correlates: (a) number of motor units, recruitment, (b) the shape of the motor unit action potential, and (c) synchronization.

Gydikov and Kosarov (1974) showed evidence that minimal recruitment in the biceps brachii occurred above 60% MVC for force-varying contractions. Milner-Brown and co-workers (Milner-Brown et al. 1973a, b, c; Milner-Brown and Stein 1975) observed motor units in the first dorsal interosseus muscle; they found that recruitment is significant only at low force levels. Bigland and Lippold (1954) made a similar conclusion for signals obtained for motor units observed in the abductor digiti minimi brevis and abductor pollicis. Clamann (1970) observed that the largest motor units located near the surface of the biceps brachii are recruited last. The recruitment of these large motor units would be detected by surface electrodes and would account for the increase. Yet, Clamann (1970) did not observe recruitment above 75% MVC for force-varying contractions. Therefore, the amount of time-dependent recruitment that occurs during a sustained contraction performed at 75% MVC is presumably negligible. In contrast, Hannerz (1974) stated that in the anterior tibialis, recruitment occurs up to 100% MVC. However, he did not report the relative amount of recruitment as a function of force. Thus, the large increase in the mean rectified and rms values of ME signals recorded with surface electrodes at 75% MVC cannot be accounted for by recruitment. This conclusion is in conflict with the suggestions of Bigland and Lippold (1954), Vredenbregt and Rau (1973) and others. They suggest that recruitment must occur during a sustained isometric contraction to maintain the force constant and account for the increase in the amplitude of the ME signal. Even if the relative increase in the parameters at 25% and 50% MVC were to be accounted for in part by recruitment, the in Figs. 4 and 5, the increase in the parameters obtained for a 75% MVC is significantly larger than the increase observed for signals recorded for 50% MVC. The only physiological correlate which remains to account for this additional increase is synchronization of motor units.

The signals recorded with surface electrodes are significantly affected by the changes in firing rate and the shape of the MUAPs. Time-dependent recruitment, although possible, was not noted to have any effect at lower force levels. Synchronization of motor units may also conceivably affect the signal, especially at 75% MVC. This point will be elaborated on in the next subsection.

The interpretations of the results obtained from signals recorded with the two types of electrodes are consistent. Lindström (see footnote 2) theoretically showed that the frequency content of a differentially recorded ME signal is dependent on the depth of muscle fibers from the recording electrode and the distance between the electrode contacts of the differential electrode. The wires which comprise the differential electrode contacts of the DISA (13K80) bipolar needle electrode record most of the significant frequency components of the physiological ME signal ($m_p(t, F)$) as it exists in the vicinity of the active muscle fibers. On the other hand, the electrode separation of the Beckman surface electrodes used in this study was approximately 2 cm. This implies that the signal recorded with surface electrodes contained only the lower portion of the ME signal power spectrum. As the conduction velocity decreases, the spectrum shifts towards lower frequencies. Hence, more of the ME signal is passed through the surface differential electrodes. This is reflected in the parameter amplitude gain as contraction time increases. Hence, signals recorded with a needle electrode are primarily affected by the decrease in firing rate as a sustained constant-force isometric contraction progresses. The signals recorded with surface electrodes are also affected by the firing rate; however, the effect of conduction velocity on the MUAPs counteracts the effects of a decrease in firing rate and causes an overall increase in parameter amplitude. Hence, signals recorded with a differential surface electrode are primarily affected by the changes in the shape of the MUAP.

Another difference in the recording characteristics of the differential surface electrodes and the bipolar needle electrode can be seen in Figs. 4, 5 and 6. The S.D. of the average parameters is much less for signals recorded with surface electrodes than with the bipolar needle electrode at all force levels. Hence, recording with surface electrodes yields more consistent results for signals obtained from different subjects than recording with a bipolar needle electrode. Buskirk and Komi (1970) made a similar conclusion when they compared surface electrodes to indwelling wire electrodes. The bipolar needle electrode records from a much smaller volume than a surface electrode. Thus, the ME signal obtained from the needle electrode is generated by fewer motor units than the signal obtained from surface electrodes. The large population of motor units contributing to the signal recorded with surface electrodes smoothens the results and reduces the S.D. of the average for the normalized parameters obtained from different subjects.

Voltage ratio of the rectified signal

To further investigate the effect of synchronization, the average voltage ratio is examined. This parameter is the ratio of the AC voltage to the DC voltage in the rectified signal. Since the parameter is a ratio, the effects of any changes in signal amplitude are removed. Hence, this parameter is minimally affected by the changes in the shapes of the MUAPs associated with a decreasing conduction velocity.

The horizontal dashed lines in Fig. 6 represent the value (0.75551) of the voltage ratio if the instantaneous amplitude of the ME signal has a Gaussian distribution. For details, see Appendix. Since the interference ME signal is a sum of MUAPTs, one would expect an amplitude histogram for the voltage of the ME signal to be Gaussian. This is a necessary (but not sufficient) condition for the ME signal to be considered a Gaussian random process.

This parameter obtained from signals recorded with a bipolar needle electrode tends to have substantial variation. This occurs because the signal recorded with an indwelling electrode records from relatively few motor units. Hence, the characteristics of a single or several motor units may dominate the signal. Their individual amplitude distributions are not Gaussian random variables because a MUAPT contains relatively long periods when no MUAP is present. Furthermore, there is not a sufficient number of MUAPTs present for the ME signal to approach a Gaussian amplitude variable. In addition, the needle electrodes are susceptible to small movements which would cause the recorded population of the motor units to change. This, too, adds variations in the results.

The voltage ratio obtained from the signals recorded with surface electrodes initially approximates a Gaussian amplitude process. As a contraction progresses, the voltage ratio significantly increases over the value for a Gaussian amplitude variable. This indicates that there is more variation of the rectified signal about the mean rectified value as the contraction progresses. The relative increase in the AC voltage over the DC voltage may be caused by either recruitment or synchronization (accompanied by cross-correlation).

The average voltage ratio obtained from signals recorded at 75% MVC tend to have the largest increase. As discussed earlier, time-dependent recruitment probably has a minimal effect at this force level. Thus, synchronization may account for this increase in the voltage ratio which represents a shift away from a Gaussian amplitude variable.

The synchronization term may be positive, negative or zero (De Luca and Van Dyk 1975). If the relative increase in the voltage ratio obtained with surface electrodes for 75% MVC is considered to be evidence of synchronization, then an observation about the sign and magnitudes of the $\underline{c_{ij}^2(\tau)}$ terms can be made. As seen in Fig. 2, the synchronization term makes a positive contribution to the variance of the rectified value. Hence, the magnitude of the majority of the $\underline{c_{ij}^2(\tau)}$ terms must be positive.

It is tempting to interpret the behavior of the voltage ratio of the rectified signal as being influenced by synchronization. However, the present data only allow a qualitative assessment of the parameter behavior. But, if such an interpretation is chosen, then it is interesting to point out that the effect of synchronization tends to be more prevalent at the higher force levels of 50% and 75% MVC and near the end of a sustained constant-force contraction. Support for this interpretation can be found in the literature. Buchthal and Madsen (1950), Lippold et al. (1960), Missiuro et al. (1962) and others have observed that synchronization increases as a contraction progresses at relatively high-force levels (near 100% MVC). Missiuro et al. (1962) went further and stated that the ME signal is not significantly affected by synchronization at low-force levels.

Summary

Myoelectric (ME) signals were simultaneously recorded with surface electrodes and a bipolar needle electrode from the middle fibers of the deltoid. Eleven subjects performed sustained isometric contractions at constant-force levels of 25%, 50%, and 75% maximum voluntary contraction (MVC). The mean rectified value, root mean squared (rms) value and the ratio of AC voltage to DC voltage in the rectified ME signals were calculated. Theoretical expressions were compared with experimental results of the parameters to determine the effects of 5 physiological correlates of contracting muscle on the ME signal.

The signals recorded with a bipolar needle electrode during a sustained constant-force isometric contraction were shown to be affected primarily by a decrease in the firing rate of motor units. This effect caused the mean rectified and rms values to decrease as a function of contraction time. The ME signals recorded with surface electrodes were also affected by a decreasing firing rate; however, the mean rectified and rms values increased with contraction time. This increase was most likely due to the effect of decreasing conduction velocity of muscle fibers on the shapes of the constituent motor unit action potentials (MUAPs) during a sustained contraction. The decreased conduction velocity causes an increase in the low-frequency components of the MUAPs. Hence, an ME signal of greater energy passes through the low-pass filter effect of the body tissues. The results obtained with surface electrodes were also more consistent than the results obtained with a bipolar needle electrode. Time-dependent recruitment of motor units was also considered.

The AC to DC voltage ratio of the rectified signal was analyzed. This parameter eliminated the effects of changes in the shapes of the MUAPs. The initial value for ME signals recorded with surface electrodes was approximately equal to the value of a Gaussian amplitude variable. As contraction time increased, the voltage ratio increased. This indicated that the ME signal could not be considered a Gaussian process throughout the contraction. The increase in this parameter could have resulted from the recruitment of dominant motor units and/or synchronization. The increase in the voltage ratio was greatest for contractions performed at 75% MVC where time-dependent recruitment is least likely to occur, indicating that synchronization may occur at this force level. This deduction is also supported by the behavior of the rms value of the ME signal recorded with bipolar needle electrodes.

The results of this investigation are consistent with the following two hypotheses. In the deltoid, the effects of motor unit synchronization become more pronounced at high-force levels (75% MVC); time-dependent recruitment of motor units is not notable during constant-force isometric contractions performed at less than 50% MVC.

Résumé

Relations entre signal myoélectrique et propriétés physiologiques des contractions isométriques à force constante

Les signaux myoélectriques (ME) ont été enregistrés en même temps avec des électrodes de surface et une électrode-aiguille bipolaire au niveau des fibres médianes du deltoïde. Onze sujets ont réalisé des contractions isométriques soutenues à des niveaux de force constants de 25%, 50% et 75% de la contraction volontaire maximale (MVC). La valeur moyenne rectifiée, la valeur de la racine carrée moyenne (rms) et le rapport du voltage AC au voltage DC des signaux ME rectifiés ont été calculés. Ces expressions théoriques sont comparées aux résultats expérimentaux des paramètres pour déterminer les effets de 5 corrélats physiologiques du muscle en cours de contraction sur le signal ME.

Les signaux enregistrés avec électrode-aiguilles bipolaires au cours de la contraction isométrique soutenue à force constante s'avèrent être principalement affectés par une diminution du taux de décharges des unités motrices. Cet effet provoque une diminution de la moyenne des valeurs rectifiées et des valeurs rms en fonction du temps de contraction. Les signaux ME enregistrés avec des électrodes de surface sont également affectés par la diminution du taux de décharges; cependant, les valeurs moyennes rectifiées et les valeurs rms augmentent avec le temps de contraction. Cette augmentation est plus vraisemblablement due à l'effet d'une diminution de la vitesse de conduction des fibres musculaires sur les formes des potentiels d'action unitaires moteurs (MUAPs) au cours de la contraction soutenue. La diminution de la vitesse de conduction provoque une augmentation des composantes de basses fréquences des MUAPs. A

partir de là, un signal ME de plus grande énergie passe au travers de l'effet filtre passe-bas des tissus corporels. Les résultats obtenus au moyen d'électrodes de surface sont également plus constants que ceux obtenus avec les électrodes-aiguilles bipolaires. Le recrutement avec le temps des unités motrices est également pris en considération.

Le rapport du voltage AC au voltage DC du signal rectifié est analysé. Ce paramètre élimine les effets de modifications de formes des MUAPs. La valeur initiale des signaux ME enregistrés avec électrodes de surface est à peu près égale à la valeur d'une variable d'amplitude Gaussienne. Au fur et à mesure que le temps de contraction augmente, le rapport des voltages augmente. Ceci indique que le signal ME ne peut pas être considéré comme un processus Gaussien tout au long de la contraction. L'augmentation de ce paramètre peut résulter du recrutement d'unités motrices et/ou de synchronisation. L'augmentation du rapport de voltages est très grande pour des contractions réalisées à 75% MVC alors que le recrutement lié au temps est moins susceptible de survenir, indiquant que la synchronisation peut se produire à ce niveau de force. Cette déduction trouve également un argument dans le comportement de la valeur rms du signal ME enregistré au moyen d'électrodes-aiguilles bipolaires.

Les résultats de cette investigation concordent avec les deux hypothèses suivantes: dans le deltoïde, les effets de la synchronisation des unités motrices deviennent plus prononcés à des niveaux élevés de force (75% MVC); le recrutement, lié au temps, des unités motrices n'est pas important au cours de contractions isométriques à force constante réalisées à moins de 50% MVC.

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Appendix

The ME signal is a summation of the MUAPTs generated by active motor units. If the motor units are firing independently (asynchronously) and the number of active motor units increases, the instantaneous amplitude of the ME signal will tend to be distributed as a Gaussian amplitude variable. This statement is a consequence of the Central Limit Theorem of Statistics. In this case, the probability density function, p, of the amplitude of the ME signal would be:

$$p_m(M) = \frac{1}{\sqrt{2\pi}\,\sigma} \exp^{-M^2/2\sigma^2}.$$

The variable M denotes a sample value of the ME signal, m. The mean value of this probability density function is zero. By necessity, this is true whenever the ME signal is recorded through an AC coupled amplifier. The standard deviation is represented by σ and it can be interpreted as the rms value.

The mean rectified value can be represented as:

$$E(|m|) = \int_{-\infty}^{\infty} |M| p_m(M) dM$$

$$= 2 \int_{0}^{\infty} M p_m(M) dM$$

$$= \sqrt{2/\pi}\,\sigma.$$

Thus under the above assumptions, the mean rectified value and rms values are related by a constant of proportionality.

The ratio of the AC voltage to the DC voltage is defined as the square root of the variance of the rectified signal divided by the mean rectified value. This can be expressed as:

$$\text{voltage ratio} = \frac{\sqrt{\sigma^2_{|m|}}}{E(|m|)}$$

$$= \frac{\sqrt{[\text{rms}(M)]^2 - [E(|m|)]^2}}{E(|m|)}$$

If the amplitude of the ME signal is distributed as a Gaussian variable, then

$$\text{voltage ratio} = \frac{\sqrt{\sigma^2 - (2/\pi)\sigma^2}}{\sqrt{2/\pi}\,\sigma} \simeq 0.75551$$

This value is plotted in Fig. 6 as a horizontal dashed line.

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