Mechanomyogram and Electromyogram Responses of Upper Limb During Sustained Isometric Fatigue with Varying Shoulder and Elbow Postures

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Abstract To investigate the behavior of mechanomyogram (MMG) and electromyogram (EMG) signals in the time and frequency domains during sustained isometric contraction, MMG and surface EMG were obtained simultaneously from four muscles: upper trapezius (TP), anterior deltoid (DL), biceps brachii (BB), and brachioradialis (BR) of 10 healthy male subjects. Experimental conditions consisted of 27 combinations of 9 postures [3 shoulder angles (SA): 0°, 30°, 60° and 3 elbow angles (EA): 120°, 90°, 60°] and 3 contraction levels: 20%, 40%, and 60% of maximum voluntary contraction (MVC). Subjective evaluations of fatigue were also assessed using the Borg scale at intervals of 60, 30, and 10 sec at 20%, 40%, and 60% MVC tests, respectively. The mean power frequency (MPF) and root mean square (RMS) of both signals were calculated. The current study found clear and significant relationships among physiological and psychological parameters on the one hand and SA and EA on the other. EA's effect on MVC was found to be significant. SA had a highly significant effect on both endurance time and Borg scale. In all experimental conditions, significant correlations were found between the changes in MPF and RMS of EMG in BB with SA and EA (or muscle length). In all four muscles, MMG frequency content was two or three times lower than EMG frequency content. During sustained isometric contraction, the EMG signal showed the well-known shift to lower frequencies (a continuous decrease from onset to completion of the contraction). In contrast, the MMG spectra did not show any shift, although its form changed (generally remaining about constant). Throughout the contraction, increased RMS of EMG was found for all tests, whereas in the MMG signal, a significant progressive increase in RMS was observed only at 20% MVC in all four muscles. This supports the

hypothesis that the RMS amplitude of the MMG signal produced during contraction is highly correlated with force production. Possible explanations for this behavioral difference between the MMG and EMG signals are discussed. *J Physiol Anthropol* 21 (1): 29–43, 2002 <http://www.jstage.jst.go.jp/en/>

Keywords: mechanomyogram, EMG, muscle fatigue, Borg scale, shoulder and elbow flexion

Introduction

Several investigations have shown that the voluntary contraction of muscles is accompanied by vibratory signals that can be detected at the muscle surface by a variety of transducers. The vibratory signal is considered to be associated with the mechanical activities of a number of muscle fibers and to reflect the motor unit (MU) activation strategy relating to force generation (Barry et al., 1985; Gordon and Holbourn, 1948; Orizio, 1993). The detection and measurement of this signal has been referred to variously as accelerometry (Lammert et al., 1976), muscle sound (Oster and Jaffe, 1980), acoustic myogram (Barry et al., 1985), soundmyogram (Orizio et al., 1989b), vibromyogram (Keidel and Keidel, 1989) or phonomyogram (Maton et al., 1990). More recently, in order to indicate clearly the nature of the signal, the term surface mechanomyogram (MMG) has been suggested (Orizio, 1993). This is a more accurate and global term than the others, because it emphasizes the mechanical nature of the phenomenon independently of the transducer used to detect it. Accordingly, we adopted this term in our study. Piezoelectric contact sensors (Barry, 1991; Bolton et al.,

1989; Orizio et al., 1994), condenser microphones (Bolton et al., 1989; Maton et al., 1990; Stokes and Dalton, 1991b) and accelerometers (Barry, 1992; Lammert et al., 1976; Orizio et al., 1996) have been widely used to detect MMG. The MMG waveform recorded by a piezoelectric contact sensor was similar to that recorded by a condenser microphone (Bolton et al., 1989), but it was markedly different from the signal recorded by an accelerometer (Barry, 1992).

In contrast to electromyographic (EMG) signals, the MMG appears to more directly reflect contractile properties of muscles, so it may also aid in examining various aspects of muscle function including muscle fatigue (Herzog et al., 1994; Koleini et al., 2000; Orizio et al., 1989a), muscle fiber type pattern (Mealing et al., 1996), age-related changes in muscular performance (Esposito et al., 1996), muscle atrophy (Marchetti et al., 1974), and neuromuscular disorders (Akataki et al., 1996). Several studies have examined muscle fatigue responses in MMG signals and fatigue-related changes in MU recruitment and firing rates (Goldenberg et al., 1991). It has been reported that the MMG frequency components reflect active MU firing rates, and that the MMG amplitude represents mainly the mechanical properties of contraction, muscle fiber composition, and firing rate during voluntary contraction (Yoshitake and Moritani, 1999). Moreover, the changes in the activation of MU due to fatigue strongly influence the MMG amplitude (Barry et al., 1985; Orizio et al., 1989a; Zwarts and Keidel, 1991).

A factor of subjective fatigue could also be added to the concept of fatigue (Simonson and Weiser, 1976). As a subjective assessment of fatigue, the Borg category ratio 10 (CR-10) scale (Borg, 1990) has become an accepted way to measure subjective estimates of fatigue in a range of exertion studies. This scale's mathematical properties permit statistical calculations. Few studies have dealt with subjective assessment correlated with objective findings (Kankaanpää et al., 1997). Thus, in this study Borg's category ratio scale (Borg, 1982) was also used to subjectively evaluate muscle fatigue.

Muscle fatigue related to changes in EMG parameters occurs during sustained contractions, as various studies have noted. On the other hand, the study of MMG signals has been renewed in recent years with advances in sensing devices and computers (i.e., the possibility of applying signal processing techniques to a large number of signal samples), but characterization and interpretation of the signals still need further attention. Since both the MMG (Barry, 1987; Mealing et al., 1990; Orizio et al., 1990; Oster and Jaffe, 1980; Zhang et al., 1990) and EMG (De Luca, 1979; Nagata et al., 1990) picked up over the surface of active skeletal muscle would be expected to measure the state of muscle activation, we searched the literature for studies of MMG with EMG simultaneously recorded on the same muscles to

determine their relative properties with levels of contraction, but we found few articles.

The main purposes of the study reported here were to examine changes in the frequency and amplitude characteristics of the MMG signal with its EMG signal as objective measures of muscle fatigue, and to apply Borg scale as a subjective estimate of fatigue at different levels of voluntary contraction at different joint angles. Thus our experiments were designed 1) to describe the changes in both signal parameters during sustained isometric contraction at constant force until exhaustion, 2) to study the relationship between MMG and EMG signal parameters with force, and 3) to study the similarities and differences between MMG and EMG characteristics under the same muscle activation state. An additional aim of the current study was to assess whether the changes in characteristics of the MMG and EMG signals depended on joint angle or muscle length.

Methods

Subjects and experimental procedures

Ten young healthy male subjects participated in this study. Their mean (SD) age, height, and body mass were 24 (1.2) years, 171 (5) cm, and 61.3 (5.8) kg, respectively. All of the subjects were right-handed. Experimental conditions consisted of 27 combinations of 9 postures [3 shoulder angles (0°, 30°, 60°) and 3 elbow angles (120°, 90°, 60°)] and 3 contraction levels, namely 20%, 40%, and 60% of maximum voluntary contraction (MVC).

The experimental device was similar to that used and illustrated in our previous study (Koleini et al., 2001) and is only briefly described here. Each subject was seated in a straight-backed chair with both the knees and hips flexed at 90°. In the sagittal plane, shoulder angle (SA) was defined as the angle between the midline of the upper arm and the line joining the shoulder and hip joints. Elbow angle (EA) was defined as the angle between the midlines of the upper arm and lower arm. An electrogoniometer (Penny & Giles Co.) was attached to the lateral side of the shoulder, and another was attached to the elbow. The experiment began with the determination of the individual MVC according to the Caldwell protocol (Caldwell et al., 1974) at designated postures using a strain gauge dynamometer with a load cell (Model 1269, Takei Kiki Kogyo Co.). The highest of three MVC measurements was considered the MVC. This procedure was repeated during each experiment. Each subject was instructed to hold a weight (equal to the designated %MVC) suspended by a strap from the weight handle, and was asked to maintain this position for as long as possible until exhaustion. The electrogoniometer signal output was displayed on a computer screen positioned directly in front of the subject. Two horizontal lines, indicating the required angle, were superimposed

on the screen to provide feedback to the subject. The degree of subjective feeling of fatigue was also assessed using the Borg scale every 60, 30, and 10 sec during the 20%, 40%, and 60% MVC tests, respectively, with the scale ranging from 0 to 100 (Borg, 1982).

Signal acquisition and processing

MMG and EMG signals were recorded from four muscles: 1) the upper trapezius (TP) as an important stabilizer of the shoulder joint, 2) the anterior deltoid (DL) as the prime mover of forward flexion of the arm, 3) biceps brachii (BB) as a primary muscle at elbow flexion, and 4) brachioradialis (BR) as a particular or controversial participation muscle at elbow flexion. EMG was recorded using disposable surface electrodes (Vitrode Y, Nihon Kohden), and the MMG was recorded by an accelerometer (Model MT-3T, diameter: 23 mm, thickness: 6.5 mm, weight: 3 g, Nihon Kohden), which was secured between EMG electrodes. The EMG and MMG signals were amplified by AC bio-amplifiers (Model AB-621G, Nihon Kohden). The signals were also low-pass filtered with a cutoff frequency of 300 Hz (time constant of 0.03 sec) and 100 Hz (time constant of 0.1 sec) for EMG and MMG, respectively. The raw EMG and MMG signals were simultaneously and continuously stored on a personal computer after analog-to-digital (A/D) conversion at a sampling rate of 1000 Hz.

Data analysis

In this study, endurance time was defined as the duration in which the subject maintained the target angle without deviation of more than $\pm 10^\circ$. Because the endurance time varied between subjects, we normalized the time in the contraction by setting the endurance time as 100%. The endurance time of each subject was divided into 10 equal segments. At the first second of each 10th percentile of endurance time (as a sample of recorded data) as well as at the last second of contraction, the frequency spectrum and mean power frequency (MPF) of each of the EMG (MPF_{EMG}) and MMG (MPF_{MMG}) signals were calculated using a Fast Fourier Transform (FFT) with a Hanning window of 1024 points. To represent the amplitude changes, the root mean squared (RMS) values of the EMG (RMS_{EMG}) and MMG (RMS_{MMG}) signals were computed from the digitized data in each sample in the time domain. To enable data to be compared between subjective evaluation (Borg scale) and objective assessment (EMG and MMG signals) of fatiguing contractions, the Borg scale data were normalized by the following method. A Lagrange polynomial function was used to interpolate data. The corresponding point of endurance time at this interpolation curve was considered as the maximum on the Borg scale (100%). Normalized Borg scale data were calculated at every 10th percentile of endurance time, so that a sample of 11

points (from 0 to 100%) was obtained. Best fitting curve was obtained using the nonlinear power function regression. Then, the time constant of each regression curve was computed for use in the analysis of Borg scale data. This method was repeated for each subject, in all experimental conditions.

Repeated measures analysis of variance (ANOVA) was used for statistical analysis. This analysis was performed using StatView 5.0. Four-way repeated measures ANOVAs ($SA \times EA \times \%MVC \times \text{time}$) were used to analyze the EMG and MMG frequency (MPF) and amplitude (RMS) data. Follow-up for the endurance time and Borg scale data three-way ($SA \times EA \times \%MVC$) ANOVAs were used, and for MVC data two-way ($SA \times EA$) repeated measures ANOVA were used. The Pearson correlation coefficient was used to correlate parametric data. A probability level of $P < 0.05$ was considered statistically significant for all comparisons.

Results

MVC and endurance time

The MVC and endurance time results are reported in Table 1. The mean values of MVC increased with increased SA or with decreased EA. The greatest MVC occurred at $SA60^\circ$ – $EA60^\circ$. The ANOVA results showed a significant main effect of EA on MVC, whereas SA was not significant. However, a significant interaction was observed between the SA and EA on MVC. Table 1 shows that endurance time clearly declined with increased SA and %MVC. As expected, the effect of %MVC on endurance time was significant. In regard to the effect of joint angle on endurance time, there was no significant effect of EA, but the effect of SA was highly significant ($p < 0.0001$). A significant interaction was observed between the SA and %MVC on endurance time.

Borg scale

Figure 1 shows the average values of Borg scale rating as a function of time (normalized to endurance time) throughout sustained isometric contraction in the 27 experimental conditions. In these conditions, the Borg curves clearly differed from the beginning to the end of the test. The difference was particularly obvious between the low and high levels of contraction and between different SA, while no difference was observed for EA except for the conditions of $SA0^\circ$ –60% MVC. The results of Borg rating values indicate that the time constant of curves declined with increased %MVC; in contrast, it increased with increased SA. The smallest and the biggest means (SD) of time constant values were 14.1 (11.1) sec and 40.1 (12.0) sec at the condition of $SA0^\circ$ –60% MVC and $SA60^\circ$ –20% MVC, respectively. Moreover, the three-way ANOVA revealed a significant main effect of SA and %MVC on the time constant of Borg rating

Table 1 Mean $\pm$ SD values of maximum voluntary contraction (MVC) at 9 different postures (3 shoulder angles $\times$ 3 elbow angles) and endurance times of sustained isometric contraction under 27 experimental conditions (n=10)

Shoulder angle (SA)		0°			30°			60°		
Elbow angle (EA)		120°	90°	60°	120°	90°	60°	120°	90°	60°
MVC (N)	%MVC	162 $\pm$ 26	168 $\pm$ 24	170 $\pm$ 23	130 $\pm$ 28	155 $\pm$ 21	197 $\pm$ 42	123 $\pm$ 25	155 $\pm$ 36	218 $\pm$ 39
Endurance time (S)	20	562 $\pm$ 187	631 $\pm$ 290	600 $\pm$ 283	371 $\pm$ 101	389 $\pm$ 174	524 $\pm$ 301	290 $\pm$ 130	345 $\pm$ 268	346 $\pm$ 187
	40	138 $\pm$ 37	115 $\pm$ 28	117 $\pm$ 40	81 $\pm$ 26	82 $\pm$ 34	88 $\pm$ 22	69 $\pm$ 25	81 $\pm$ 27	91 $\pm$ 20
	60	61 $\pm$ 28	49 $\pm$ 18	48 $\pm$ 20	40 $\pm$ 23	37 $\pm$ 12	42 $\pm$ 16	37 $\pm$ 16	35 $\pm$ 9	46 $\pm$ 19

Fig. 1 Changes in Borg scale rating as a function of time (expressed as a percentage of the endurance time) during sustained isometric contraction under 27 experimental conditions, provided by a combination of 3 shoulder angles (SA): 0°, 30°, and 60°, 3 elbow angles (EA): 120° (●), 90° (■), and 60° (○), and 3 levels of maximum voluntary contraction (MVC): 20%, 40%, and 60% MVC. Each point is the average value (mean and SD, n=10) at every 10% endurance time.

curves, but the effect of EA was not significant.

Mean power frequency (MPF)

Table 2 shows the mean $\pm$ SD of MPF_{EMG} and MPF_{MMG} values at the onset of contractions and the point when each of the four muscles became exhausted. It is seen that the MMG signal is two or three times lower than the EMG signal. The EMG signal decreased by more than 20 Hz between onset and exhaustion, but no such decrease was found for the MMG signal.

MPF throughout sustained contraction

The MPF_{EMG} tended to decrease continuously

throughout the sustained isometric contraction, whereas the corresponding MPF_{MMG} typically remained almost constant from the onset to the end of contractions. Pearson correlation coefficients between changes in MPF and the corresponding time are reported in Table 3.

In all four muscles, MPF_{EMG} of the power spectrum shifted toward lower frequencies in almost all experimental conditions. The ANOVA results showed a significant main effect of time (in all four muscles) on MPF_{EMG} . Neither the SA nor the EA factor was causally significant as an effect on MPF_{EMG} in TP, DL, or BR, while in BB the effect of SA and that of EA (as well as %MVC) were both highly significant ($p < 0.0001$). Figure 2 shows

Table 2 Electromyogram (EMG) and mechanomyogram (MMG) mean power frequency (MPF, Hz) values (mean $\pm$ SD) of four muscles at onset and exhaustion of sustained isometric contractions (averaged from each of 9 postures, n=10)

MPF	Muscle	20%MVC		40%MVC		60%MVC	
		Onset	Exhaustion	Onset	Exhaustion	Onset	Exhaustion
EMG	TP	107.9 $\pm$ 3.5	87.7 $\pm$ 2.8	105.4 $\pm$ 2.6	83.3 $\pm$ 2.8	104.5 $\pm$ 2.9	81.4 $\pm$ 2.7
	DL	104.0 $\pm$ 2.9	80.8 $\pm$ 3.8	102.3 $\pm$ 3.5	74.7 $\pm$ 4.8	103.2 $\pm$ 5.1	74.5 $\pm$ 5.0
	BB	107.3 $\pm$ 8.8	89.8 $\pm$ 9.6	106.1 $\pm$ 6.6	80.7 $\pm$ 9.4	105.1 $\pm$ 7.0	78.2 $\pm$ 8.8
	BR	116.4 $\pm$ 15.6	103.4 $\pm$ 15.5	108.8 $\pm$ 11.8	97.9 $\pm$ 12.7	110.1 $\pm$ 11.4	95.8 $\pm$ 10.8
MMG	TP	55.2 $\pm$ 6.5	61.0 $\pm$ 9.0	55.7 $\pm$ 6.6	63.7 $\pm$ 8.5	65.2 $\pm$ 8.9	70.9 $\pm$ 9.0
	DL	44.7 $\pm$ 5.1	38.6 $\pm$ 3.4	42.7 $\pm$ 3.4	39.8 $\pm$ 2.5	42.8 $\pm$ 4.5	40.8 $\pm$ 4.3
	BB	32.3 $\pm$ 2.6	33.3 $\pm$ 4.7	35.1 $\pm$ 3.6	35.9 $\pm$ 3.4	36.3 $\pm$ 3.7	37.4 $\pm$ 4.2
	BR	32.4 $\pm$ 4.9	31.5 $\pm$ 5.4	35.2 $\pm$ 6.4	33.7 $\pm$ 4.8	35.8 $\pm$ 5.9	37.5 $\pm$ 5.7

The muscles, TP: upper trapezius, DL: anterior deltoid, BB: biceps brachii, BR: brachioradialis.

the changes in MPF_{EMG} in BB during sustained contraction at 27 different conditions. A significant effect of EA on MPF_{EMG} value is clearly seen in this figure. In BB, throughout the experiments, MPF_{EMG} values of test contraction at EA120° were smaller than those at the other EA, and values of test contraction at EA90° were smaller than those at EA60°.

In contrast to the EMG signal, MPF_{MMG} did not shift toward lower frequency. Changes in the MPF_{MMG} of TP (as a typical muscle among the four) were selected to show the behavior of MMG signal frequency during sustained contraction (Fig. 3). There were significant main effects of SA ($p < 0.01$ to $p < 0.0001$), EA ($p < 0.01$, except in the TP and the BB), %MVC ($p < 0.05$ to $p < 0.001$), and time ($p < 0.0001$, except in BB) on MPF_{MMG} for all four muscles.

MPF as a function of force

Figure 4a shows the relationship between the level of contraction and MPF_{EMG} and MPF_{MMG} . In this figure, the average MPF of both signals of a typical muscle (BB) are represented as a function of %MVC. Consistent with previous findings, MPF_{EMG} decreased, at both the onset and the end of contraction, almost linearly with increased contraction levels. In addition, the difference between the onset and exhaustion values of each contraction level increased as the force level increased. In contrast to the findings with EMG signal, it is clear from Fig. 4a that the MMG displayed entirely different behavior (except for the DL). At the onset of contraction in DL, MPF_{MMG} decreased (as it did with EMG), but the relationship was the inverse of that with EMG: at exhaustion, MPF_{MMG} increased with increased contraction level.

Root mean square (RMS)

Table 4 presents the mean $\pm$ SD of RMS_{EMG} and RMS_{MMG} values at the onset of contraction and at exhaustion of each of the four muscles. It should be noted that RMS_{MMG} values magnified 200 times their original

output voltage of accelerometer. However, these values essentially indicate the same meaning of arbitrary units. As shown in Table 4, the smallest and the largest values of RMS for EMG were found for TP at 20% MVC (onset: 0.03 mV, exhaustion: 0.07 mV) and DL at 60% MVC (onset: 0.63 mV, exhaustion: 1.32 mV), whereas for MMG it was found for TP at 20% MVC (onset: 0.10 V, exhaustion: 0.55 V) and BB at 60% MVC (onset: 4.84 V, exhaustion: 6.82 V), respectively.

RMS throughout sustained contraction

The correlation coefficients found in this study between the duration of contraction and the RMS_{EMG} and RMS_{MMG} values are reported in Table 5. A significant increase of RMS_{EMG} throughout sustained isometric contraction was found for each of the four muscles under almost all experimental conditions. The correlation coefficients between RMS_{MMG} and time showed a highly significant increase only in the 20% MVC experiments.

Changes in RMS_{EMG} of BB during the test contraction are shown in Fig. 5. The results of repeated measures ANOVA revealed that %MVC and time (except for BR) each had a causally significant effect on RMS_{EMG} . There was a significant main effect of SA on RMS_{EMG} for all four muscles, while EA was highly significant for BB only. During the test contraction in BB, the RMS value at EA120° was bigger than that at EA90° or EA60°; also, the value at EA90° was bigger than at EA60°. Only one muscle, BB, showed this kind of effect under all experimental conditions.

Figure 6 represents the changes in RMS_{MMG} of BR under 27 experimental conditions throughout sustained isometric contraction. This figure shows clearly that RMS_{MMG} behaves differently at a low level of contraction (20% MVC) than at the other contraction levels (40% and 60% MVC). At 20% MVC, RMS_{MMG} increased progressively from the beginning to the end of the test in all four muscles. At this 20% contraction, the RMS_{MMG} values significantly increased with increased EA. The ANOVA

Table 3 Pearson correlation coefficient between time of contraction and mean power frequency (MPF) values of electromyogram (EMG) and mechanomyogram (MMG) signals of four muscles during sustained isometric contractions under 27 experimental conditions (n=10)

SA	Muscle	20%MVC		40%MVC		60%MVC	
		EMG	MMG	EMG	MMG	EMG	MMG
0°	TP	-0.26***	0.20**	-0.50***	0.26***	-0.46***	0.10 _{ns}
		-0.32***	0.03 _{ns}	-0.50***	0.08 _{ns}	-0.44***	0.02 _{ns}
		-0.16 _{ns}	0.18*	-0.28***	0.14 _{ns}	-0.39***	-0.04 _{ns}
	DL	-0.38***	-0.21**	-0.44***	0.14 _{ns}	-0.42***	0.26***
		-0.50***	-0.19*	-0.54***	0.07 _{ns}	-0.47***	0.26***
		-0.60***	-0.04 _{ns}	-0.58***	0.10 _{ns}	-0.40***	0.15 _{ns}
	BB	-0.48***	-0.07 _{ns}	-0.64***	0.11 _{ns}	-0.66***	0.05 _{ns}
		-0.42***	0.00 _{ns}	-0.72***	0.11 _{ns}	-0.76***	0.07 _{ns}
		-0.36***	0.15 _{ns}	-0.57***	-0.03 _{ns}	-0.62***	-0.16*
	BR	-0.27***	-0.09 _{ns}	-0.57***	0.18*	-0.48***	0.18*
		-0.22**	-0.13 _{ns}	-0.52***	0.06 _{ns}	-0.51***	0.04 _{ns}
		-0.25**	0.01 _{ns}	-0.50***	-0.26**	-0.49***	-0.13 _{ns}
30°	TP	-0.54***	0.10 _{ns}	-0.62***	0.14 _{ns}	-0.55***	0.20**
		-0.60***	0.16 _{ns}	-0.56***	0.19**	-0.61***	0.10 _{ns}
		-0.46***	-0.02 _{ns}	-0.65***	0.11 _{ns}	-0.59***	0.15 _{ns}
	DL	-0.54***	-0.17*	-0.66***	-0.13 _{ns}	-0.61***	-0.22**
		-0.70***	-0.22**	-0.76***	-0.25**	-0.62***	-0.28***
		-0.57***	-0.04 _{ns}	-0.79***	0.01 _{ns}	-0.72***	-0.39***
	BB	-0.35***	-0.08 _{ns}	-0.45***	0.15 _{ns}	-0.47***	-0.01 _{ns}
		-0.13 _{ns}	-0.15 _{ns}	-0.60***	-0.01 _{ns}	-0.53***	-0.07 _{ns}
		-0.19**	0.04 _{ns}	-0.48***	0.20**	-0.63***	-0.10 _{ns}
	BR	-0.05 _{ns}	-0.03 _{ns}	-0.19**	-0.08 _{ns}	-0.13 _{ns}	-0.05 _{ns}
		-0.25**	-0.19*	-0.23**	0.04 _{ns}	-0.21**	0.09 _{ns}
		-0.22*	-0.11 _{ns}	0.03 _{ns}	-0.06 _{ns}	-0.34***	0.15 _{ns}
60°	TP	-0.53***	0.00 _{ns}	-0.52***	0.19**	-0.66***	-0.07 _{ns}
		-0.70***	0.07 _{ns}	-0.60***	0.18*	-0.61***	0.09 _{ns}
		-0.45***	0.09 _{ns}	-0.62***	0.09 _{ns}	-0.67***	0.13 _{ns}
	DL	-0.67***	-0.31***	-0.57***	-0.27***	-0.76***	-0.15 _{ns}
		-0.59***	-0.08 _{ns}	-0.69***	-0.38***	-0.78***	-0.43***
		-0.59***	-0.13 _{ns}	-0.74***	-0.38***	-0.78***	-0.37***
	BB	-0.10 _{ns}	-0.21**	-0.35***	-0.10 _{ns}	-0.50***	-0.19**
		-0.43***	-0.18*	-0.55***	0.16 _{ns}	-0.53***	-0.06 _{ns}
		-0.53***	0.01 _{ns}	-0.45***	0.15 _{ns}	-0.67***	0.28***
	BR	-0.07 _{ns}	-0.23**	0.04 _{ns}	0.11 _{ns}	-0.17 _{ns}	0.11 _{ns}
		-0.06 _{ns}	-0.02 _{ns}	-0.18*	0.12 _{ns}	0.06 _{ns}	0.30***
		-0.37***	-0.11 _{ns}	-0.34***	0.09 _{ns}	-0.37***	-0.03 _{ns}

SA: shoulder angle; muscles, TP: upper trapezius, DL: anterior deltoid, BB: biceps brachii, BR: brachioradialis. For each muscle, the three rows indicate, from higher to lower, the values for elbow angles 120°, 90°, and 60°, respectively. *: P<0.05; **: P<0.01; ***: P<0.001; ns: not significant.

results showed a main significant effect of SA, %MVC, and time on RMS_{MMG} for all muscles. There was a significant main effect of EA on RMS_{MMG} for DL and BR.

RMS as a function of force

Figure 4b shows the relationship between RMS_{EMG} and RMS_{MMG} with the contraction levels of a typical muscle (BB). As expected, an increase in the isometric contraction level produces increasing EMG and MMG

signal RMS values. At the onset of contraction and at exhaustion, the RMS amplitude of both EMG and MMG was almost linearly related to %MVC. However, as Fig. 4b shows clearly, the differences between the onset and exhaustion values of EMG increased with increased %MVC, while this kind of difference was not observed for the MMG signal. This behavior was noted for all four muscles.

Fig. 2 Changes in the electromyogram (EMG), mean power frequency (MPF) of biceps brachii muscle (BB) as a function of time (expressed as a percentage of the endurance time) during sustained isometric contraction under 27 experimental conditions. Each point is the average value (mean and SD, $n=10$) at the first second of each 10% of the endurance time. SA: shoulder angle; EA: elbow angle; MVC: maximum voluntary contraction.

Discussion

Although recent years have brought a renewed interest in the study of MMG, it is difficult, if not impossible, to compare results because of the diversity of experimental designs that have not employed well-known, reproducible, easy-to-measure physiological conditions or similar signal-processing techniques (Orizio et al., 1990). Previous articles (Rodriquez et al., 1996; Vaz et al., 1996) have suggested that the mechanism of muscle vibrations is not yet understood. The mechanism is complex but known, and it provides a basis for understanding experimental results. It is important to analyze both the frequency domain and the time domain in order to understand muscle contractile properties and their pathological conditions (Diemont et al., 1988; Esposito et al., 1996; Orizio et al., 1990; Wee and Ashley, 1989).

Frequency domain analysis

MPF of the EMG and MMG in this study were found to be within the same range as those reported for different muscles in our previous study (Koleini et al., 2001) and in studies by other investigators (Basmajian and De Luca, 1985; Nagata et al., 1990; Orizio et al., 1990; Pan et al.,

1989). As can be seen in Table 2, the MMG frequency content is approximately two to three times lower than that of the EMG in all muscles. This distinct difference may reveal the difference in the underlying sources of the EMG and MMG signals. A possible explanation was suggested by Zhang et al. (1992). If the muscle fiber is cylindrical, shortening (contraction) could generate a radial expansion, resulting in vibration waves. Muscle fiber contraction is accompanied by the fiber's action potential. When muscle contraction increases, the firing rate and the number of recruited MUs increase following the size principle (Henneman et al., 1965). The EMG and MMG signals detected on the skin over the contracting muscle reflect a spatiotemporal summation of MU action potential and MU vibration signals, respectively. The shape differences between MU action potential and MU vibration signals are responsible for the differences in MPF reported in Table 2. This interpretation is supported by the simultaneous recording of MU action potential and MU vibration signal data obtained by electrical stimulation of the frog gastrocnemius muscle by Barry (1987) and Frangioni et al. (1987).

The results of the present investigation of all four muscles under all experimental conditions (except for BB

Fig. 3 Changes in the mechanomyogram (MMG), mean power frequency (MPF) of trapezius muscle (TP) as a function of time (expressed as a percentage of the endurance time) during sustained isometric contraction under 27 experimental conditions. Each point is the average value (mean and SD, $n=10$) at the first second of each 10% of the endurance time. SA: shoulder angle; EA: elbow angle; MVC: maximum voluntary contraction.

Fig. 4 Mean power frequency (MPF, a) and root mean square (RMS, b) of both electromyogram (EMG) and mechanomyogram (MMG) signals of typical muscle biceps brachii (BB) observed at the onset of contraction and at exhaustion, as a function of percent of maximal voluntary contraction (%MVC). Values are mean and SD (averaged from each of 9 postures [3 shoulder angles $\times$ 3 elbow angles], $n=10$).

Table 4 Electromyogram (EMG, mV) and mechanomyogram (MMG, V) root mean square (RMS) values (mean $\pm$ SD) of four muscles at onset and exhaustion of sustained isometric contractions (averaged from each of 9 postures, n=10)

RMS	Muscle	20%MVC		40%MVC		60%MVC	
		Onset	Exhaustion	Onset	Exhaustion	Onset	Exhaustion
EMG	TP	0.03 $\pm$ 0.03	0.07 $\pm$ 0.04	0.06 $\pm$ 0.05	0.13 $\pm$ 0.06	0.11 $\pm$ 0.06	0.31 $\pm$ 0.10
	DL	0.10 $\pm$ 0.04	0.23 $\pm$ 0.09	0.35 $\pm$ 0.11	0.84 $\pm$ 0.43	0.63 $\pm$ 0.20	1.32 $\pm$ 0.58
	BB	0.07 $\pm$ 0.08	0.16 $\pm$ 0.16	0.22 $\pm$ 0.13	0.49 $\pm$ 0.21	0.45 $\pm$ 0.21	1.00 $\pm$ 0.36
	BR	0.05 $\pm$ 0.06	0.09 $\pm$ 0.12	0.13 $\pm$ 0.15	0.20 $\pm$ 0.21	0.26 $\pm$ 0.24	0.40 $\pm$ 0.30
MMG	TP	0.10 $\pm$ 0.06	0.55 $\pm$ 0.14	0.48 $\pm$ 0.10	0.83 $\pm$ 0.18	0.91 $\pm$ 0.24	1.56 $\pm$ 0.79
	DL	0.12 $\pm$ 0.06	1.04 $\pm$ 0.23	0.96 $\pm$ 0.35	1.72 $\pm$ 0.56	2.07 $\pm$ 0.81	3.29 $\pm$ 1.68
	BB	0.28 $\pm$ 0.09	2.29 $\pm$ 0.73	2.42 $\pm$ 1.01	4.07 $\pm$ 2.08	4.84 $\pm$ 2.29	6.82 $\pm$ 3.85
	BR	0.23 $\pm$ 0.12	1.74 $\pm$ 0.43	1.95 $\pm$ 0.94	4.18 $\pm$ 1.95	4.38 $\pm$ 2.20	6.64 $\pm$ 3.77

The muscles, TP: upper trapezius, DL: anterior deltoid, BB: biceps brachii, BR: brachioradialis.

at SA30°-EA60°-40% MVC, SA60°-EA120°-40% MVC, and SA60°-EA90°-60% MVC) indicate that the MPF_{EMG} signal tended to decrease continuously throughout the sustained isometric contraction, and that it is associated with a loss of the high-frequency contents of the signal (Fig. 2), whereas the corresponding MPF_{MMG} signal typically remained about constant (Fig. 3). During sustained isometric contraction, the EMG signal showed the well-known shift to lower frequencies. The principal reason for this shift is the lowering of the muscle fiber conduction velocity, and central factors such as synchronization could also play a role (Zwarts et al., 1987).

In contrast, the MPF_{MMG} did not shift throughout the test contractions, although the form of MMG spectra changed. The factor of muscle stiffness could be the basis of one of the physiological hypotheses to explain this behavior of MMG frequency over time. The resonant frequency of muscle is a function of several parameters, including muscle mass, length, topology, and stiffness. During an isometric contraction, the change in stiffness is much greater than the change in any of the other parameters. This implies that the change in resonant frequency should be closely related to the change in stiffness during isometric contraction (Barry and Cole, 1990). In our experiment, force was constant (as %MVC) during test contraction. Since muscle vibrations appear to occur at the resonant frequency of the muscle, the MPF_{MMG} signal tends to remain roughly constant throughout sustained isometric contraction. MMG frequency changes found over time in the present study were in agreement with those reported in the literature by Herzog et al. (1994) and Koleini et al. (2001).

Time domain analysis

The EMG amplitude changes found during sustained isometric contraction are well known from the literature. In the present study, increased RMS_{EMG} from the beginning to the end of the test was found (Fig. 5) for all

muscles under all experimental conditions (except for DL at SA0°-EA120°-40% MVC). Our findings were in good agreement with previous investigations (De Luca, 1984; Moritani et al., 1986). The increase in RMS_{EMG} is due to an increase in the firing rate and recruitment of MUs (Freund, 1983).

MMG amplitude has shown to reflect many factors. Orizio (1993) suggested that the MMG amplitude may also be affected by factors such as muscle architecture and the properties of the tissue layers between the muscle and the MMG recording device on the skin's surface. It has been suggested that, during sustained low-level isometric contraction, MMG amplitude increases slowly as fatigue proceeds, and during high-level contraction MMG amplitude shows either a lack of increase or a decrease (Goldenberg et al., 1991; Orizio et al., 1989a; Rodriguez et al., 1993). The response seen in the MMG during low-level contractions could be attributed to the recruitment of new MUs and to an increase in MU firing rate. The MMG responses seen during high-level contractions would result from the recruitment of fast MUs and a reduction in the MU firing rate (Orizio, 1993). This may reflect force-dependent differences in the fatigue process at differing levels of contraction. Generally, it is accepted that only the type I (slow-twitch) fibers are active at 20% MVC (Burke, 1981; Freund, 1983), while at high contraction levels, when both type I and type II (fast-twitch) fibers are activated, the MMG is affected by the individual muscular fiber composition (Diemont et al., 1988). As can be seen in Fig. 6, our investigation also found this change in MMG amplitude. In the test at 20% MVC, the RMS_{MMG} increased monotonically up to exhaustion. At 40% MVC, RMS increased at varying rates during low-level contraction. The test at 60% MVC demonstrated no unequivocal trend in the amplitude (fluctuation form), namely, a progressive increase in power followed by decreases thereafter. This finding supports the hypothesis (Barry et al., 1985) that the RMS amplitude of the MMG signal

Table 5 Pearson correlation coefficient between time of contraction and root mean square (RMS) values of electromyogram (EMG) and mechanomyogram (MMG) signals of four muscles during sustained isometric contractions under 27 experimental conditions (n=10)

SA	Muscle	20%MVC		40%MVC		60%MVC	
		EMG	MMG	EMG	MMG	EMG	MMG
0°	TP	0.34***	0.79***	0.35***	0.22**	0.45***	0.09 _{ns}
		0.57***	0.60***	0.24**	0.39***	0.53***	0.14 _{ns}
		0.44***	0.67***	0.43***	0.43***	0.20**	0.17 _{ns}
	DL	0.06 _{ns}	0.72***	-0.17 _{ns}	0.26***	0.03 _{ns}	-0.09 _{ns}
		0.27***	0.60***	0.20*	0.38***	0.30***	0.05 _{ns}
		0.27**	0.70***	0.18*	0.36***	0.27**	0.13 _{ns}
	BB	0.59***	0.78***	0.32***	-0.08 _{ns}	0.32***	-0.44***
		0.37***	0.68***	0.46***	0.25**	0.51***	-0.17*
		0.30***	0.68***	0.31***	0.24**	0.30***	0.04 _{ns}
	BR	0.57***	0.84***	0.41***	0.11 _{ns}	0.29***	-0.38***
		0.41***	0.71***	0.54***	0.47***	0.25**	-0.08 _{ns}
		0.35***	0.77***	0.23**	0.50***	0.07 _{ns}	0.25**
30°	TP	0.47***	0.48***	0.34***	0.24**	0.38***	-0.01 _{ns}
		0.59***	0.56***	0.33***	0.20**	0.57***	0.02 _{ns}
		0.45***	0.55***	0.33***	0.38***	0.33***	0.29***
	DL	0.28***	0.52***	0.36***	0.35***	0.26***	0.02 _{ns}
		0.41***	0.54***	0.34***	0.11 _{ns}	0.28***	-0.03 _{ns}
		0.42***	0.57***	0.46***	0.31***	0.39***	0.14 _{ns}
	BB	0.30***	0.57***	0.28***	0.20**	0.27***	-0.09 _{ns}
		0.29***	0.65***	0.21**	0.06 _{ns}	0.15 _{ns}	0.05 _{ns}
		0.37***	0.61***	0.45***	0.40***	0.40***	0.17*
	BR	0.38***	0.53***	0.20**	0.31***	0.14 _{ns}	0.03 _{ns}
		0.42***	0.67***	0.15 _{ns}	0.29***	0.18*	0.11 _{ns}
		0.40***	0.63***	0.49***	0.47***	0.41***	0.27***
60°	TP	0.15 _{ns}	0.54***	0.35***	0.08 _{ns}	0.46***	0.20**
		0.43***	0.45***	0.44***	0.26***	0.48***	0.24**
		0.33***	0.61***	0.48***	0.16 _{ns}	0.49***	-0.06 _{ns}
	DL	0.24**	0.50***	0.32***	0.10 _{ns}	0.40***	0.21**
		0.29***	0.48***	0.34***	0.23**	0.42***	0.24**
		0.46***	0.55***	0.52***	0.07 _{ns}	0.39***	-0.17*
	BB	0.21**	0.60***	0.21**	0.32***	0.27***	0.32***
		0.53***	0.48***	0.24**	0.33***	0.15 _{ns}	0.23**
		0.44***	0.47***	0.58***	0.10 _{ns}	0.44***	-0.20**
	BR	0.17 _{ns}	0.53***	0.02 _{ns}	0.18*	0.13 _{ns}	0.27***
		0.35***	0.38***	0.63***	0.31***	0.24**	0.30***
		0.47***	0.54***	0.33***	0.40***	0.40***	0.14 _{ns}

SA: shoulder angle; muscles, TP: upper trapezius, DL: anterior deltoid, BB: biceps brachii, BR: brachioradialis. For each muscle, the three rows indicate, from higher to lower, the values for elbow angles 120°, 90°, and 60°, respectively. *: P<0.05; **: P<0.01; ***: P<0.001; ns: not significant.

produced during skeletal muscle contraction is highly correlated with force production.

EMG and MMG signal changes with force

Fewer data are available regarding MMG frequency characteristics and force, but MPF has also shown to increase with increases in force production (Dalton and Stokes, 1993; Matheson et al., 1997; Maton et al., 1990; Orizio et al., 1990). Except for DL at the onset of

contraction, the results of the present study for TP, BB, and BR were consistent with those of previous investigations reporting that kind of relationship. Orizio et al. (1990) clearly demonstrated the relationship between MMG frequency content vs. force in biceps brachii muscle. They showed that increasing the intensity of muscle contraction resulted in the enlargement of the MMG spectrum that is, shifts to higher frequencies with increases in intensity as might be the

Fig. 5 Changes in the electromyogram (EMG), root mean square (RMS) of biceps brachii muscle (BB) as a function of time (expressed as a percentage of the endurance time) during sustained isometric contraction under 27 experimental conditions. Each point is the average value (mean and SD, $n=10$) for the period of 10% of the endurance time. SA: shoulder angle; EA: elbow angle; MVC: maximum voluntary contraction.

case with a greater contribution by type II fibers with increasing force generation (Mealing and McCarthy, 1991). Moreover, these force-dependent MPF_{MMG} changes have been suggested to reflect the firing rate of MUs (Shinohara et al., 1998). It has been suggested that increased MPF_{MMG} with increased firing rate may have been due to increased muscle stiffness associated with increased firing rate (Vaz et al., 1997).

The increase in the amplitudes of the EMG and MMG signals accompanying increased contraction force was previously attributed to increased temporal and spatial recruitment (Dalton and Stokes, 1993; De Luca, 1984). It was previously postulated that EMG amplitude is affected by recruitment of additional MUs, firing rate, muscle architecture, and muscle fiber conduction velocity (De Luca, 1984). Factors affecting the MMG signal amplitude are less clear. As the contracting muscle becomes fatigued, the increase in signal strength with increasing force suggests that MMG amplitude reflects MU recruitment, although it is unclear how much firing rates, recruitment of new MUs, tremor, or mechanical factors contribute to the increase (Bolton et al., 1989; Dalton and Stokes, 1993).

Several studies have examined the relationship between MMG amplitude and %MVC. At least for

contractions of up to 80% MVC, it has been found that MMG amplitude increases with increased force (Orizio et al., 1989b, 1990). This relationship was found to be linear in the quadriceps (Stokes and Dalton, 1991a), quadratic in the lumbar erector spinae (Stokes et al., 1988), and curvilinear in the adductor pollicis (Stokes and Cooper, 1992). Orizio (1993) and Stokes and Dalton (1991a) suggested that these non-uniform relationships were due largely to the differences in the composition and contractile properties of muscle fibers. In the present study, we observed increases of RMS_{MMG} from 20% to 60% MVC for all four muscles. Our results on MMG amplitude are in good agreement with previous investigations.

Dependence on joint angle

In this study, sustained contraction was performed up to exhaustion at three SA and three EA. As can be seen in Table 1, at three different levels of contraction endurance was briefest at SA60°-EA120° (except for 60% MVC at EA90°). Furthermore, the smallest MVC also occurred at this posture. These results showed that the shortest endurance time occurred at the longest muscle length of BB and BR. Previous studies (Caldwell, 1963; Petrofsky and Phillips, 1980) indicate that endurance time in elbow flexion is not dependent on the elbow angle. Our data

Fig. 6 Changes in the mechanomyogram (MMG), root mean square (RMS) of brachioradialis muscle (BR) as a function of time (expressed as a percentage of the endurance time) during sustained isometric contraction under 27 experimental conditions. Each point is the average value (mean and SD, $n=10$) for the period of 10% of the endurance time. SA: shoulder angle; EA: elbow angle; MVC: maximum voluntary contraction.

support this previous finding, but in contrast to EA, SA showed a highly significant ($p<0.0001$) main effect on endurance time for all postures.

Several studies have shown that muscle length affects the rate of muscle fatigue, based on the rate of force loss (Fitch and McComas, 1985; Sacco et al., 1994) as well as on the rate of changes in the EMG signal (Doud and Walsh, 1995; Weir et al., 1996) and MMG signal (Weir et al., 2000). This study has also noted a dependence of MPF and RMS values of the EMG and MMG signals on joint angles. It is interesting to note that for TP and DL, the largest rate of decrease of MPF_{EMG} occurred at $SA60^{\circ}$ – $EA60^{\circ}$ with change ratios of -27% and -37% , and that the smallest rate of decrease occurred at $SA0^{\circ}$ – $EA60^{\circ}$, $SA0^{\circ}$ – $EA120^{\circ}$ with change ratios of -11% and -14% , respectively. On the other hand, for BB and BR the largest rate of decreases were at $SA0^{\circ}$ – $EA120^{\circ}$ with change ratios of -37% and -25% , and the smallest rate of decreases were at $SA30^{\circ}$ – $EA90^{\circ}$, $SA60^{\circ}$ – $EA90^{\circ}$ with change ratios of -7% and -4% , respectively. This clearly indicates that the most important muscles during flexion at the shoulder joint are TP and DL, while the most important muscles during flexion at the elbow joint are BB and BR. With regard to the results of the present

investigation, the largest rate of increase of RMS_{MMG} values (which was found at 20% MVC), occurred at the muscle length corresponding to $SA60^{\circ}$ – $EA60^{\circ}$ for TP, DL, and BR (1884% , 3746% , 2672%), and for BB at $SA60^{\circ}$ – $EA120^{\circ}$ (2525%), respectively. The ANOVA results also revealed the main effect of SA on changes of RMS_{MMG} for all four muscles. Moreover, at $SA60^{\circ}$ the maximum values of RMS_{MMG} found for TP, DL, BB, and BR were 3.4 V, 6.2 V, 11.7 V, and 12.6 V, respectively. It appears that muscle located far from the shoulder joint has a higher RMS_{MMG} value than muscle closer to the joint.

The Borg scale is a recognized and widely used method for measuring subjective estimates of fatigue (Borg, 1982). We assessed the Borg scale values and calculated the time constant of values in order to produce an index that was comparable with other parameters. The results of the current study indicate that the time constant is directly related to SA. In other words, the feeling of fatigue occurred slowly with increased SA. This effect of SA on the time constant was also found to be highly significant. It can be seen in Fig. 1 that the effect of three different EA on Borg scale values was not so considerable. Statistical analysis also showed no significant effect of EA on this parameter.

Conclusion

We have examined the simultaneous MMG and EMG signals during sustained isometric contraction. MPF, RMS, and %MVC relationships of the MMG and those of the EMG described the basic muscle activation patterns. The present results obtained from both signals combined confirm and extend previous reports on the contractile properties of muscles. The differences in frequency content between the MMG and EMG indicate the shape differences of waveform between the mechanical and electrical responses to muscle. However, since there are some fundamental differences between the MMG and EMG signals from muscle, we would emphasize the need for more research to clarify and better understand the phenomenon of vibratory signal emission from contracting muscle. An additional aim of the current study was to assess the effect of joint angle on MMG and EMG. We have found clear significant relationships between the signal parameters and both SA and EA. In particular, in BB an increase in the EA (or muscle length) produces a) decreasing MPF_{EMG} values at all experimental conditions and b) increasing RMS of both EMG (at all conditions) and MMG signals (only at 20% MVC tests). Furthermore, a significant relationship was also found between SA and subjective evaluation. These results support the findings of our group's previous study (Koleini et al. 2001).

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