Myoelectric Signal Processing: Optimal Estimation Applied to Electromyography—Part II: Experimental Demonstration of Optimal Myoprocessor Performance

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Abstract—This paper (Part II of two) presents an experimental demonstration of the performance achieved by implementing the mathematically derived optimal myoprocessor described in Part I. Almost an order-of-magnitude improvement over the common myoprocessor is obtained. Excellent agreement of the experimental results with the analytical predictions verifies the mathematical analysis. The relative contributions of each stage of the optimal myoprocessor are examined. A discussion and comparison of several existing and proposed techniques for myoprocessor improvement are presented.

NOMENCLATURE

B_s	Statistical bandwidth.
$G(f)$	Power spectral density.
$\Gamma(\ )$	Gamma function.
$E\{\ \}$	Expected value.
SNR	Signal-to-noise ratio.
F	Muscle force.
$\hat{F}$	Estimated muscle force.
N	Number of degrees of freedom of estimate.
T	Available processing time.
τ	Time constant.
σ	Standard deviation of surface myoelectric activity.
λ	Eigenvalue.
Λ	Matrix of eigenvalues.
Φ	Matrix of eigenvectors.

INTRODUCTION

THE electrical activity of muscle is potentially a very useful biological signal, but attempts to use it as a proportional indicator of muscle activity have been plagued by the presence of an apparent "noise" component of large amplitude and low frequency superimposed on the desired information. At present, the general consensus of opinion is that it is impractical to try to obtain a consistent estimate of muscle contraction from myoelectric activity. In Part I of this paper a novel myoprocessing technique was presented, and a mathematical statement of the optimal myoprocessor was derived. In Part II an experimental demonstration of the effectiveness of this optimal processor—almost an order-of-magnitude improvement in performance over the common myoprocessor—is presented.

Manuscript received August 29, 1979; revised February 8, 1980. This work was supported in part by the Rehabilitation Services Administration, U.S. Department of Health, Education, and Welfare, under Grant 23-P-55854/1 and the Whitaker Professorship of Biomedical Engineering at the Massachusetts Institute of Technology, Cambridge, MA.

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EXPERIMENTAL EQUIPMENT

Myoelectric activity and muscle force were recorded simultaneously from the biceps of normal and amputee subjects maintaining isotonic (constant force), isometric (constant muscle length) contractions. To measure amputee muscle force, we enlisted the aid of an amputee with cineplasty. Cineplasty is the surgical construction of a skin-lined tunnel through an otherwise healthy muscle which is disconnected at its distal end from the skeleton [1].

The experimental arrangement for normal subjects is shown in Fig. 1.

Subjects were seated in a sturdy, armless, wooden chair and grasped a handle attached to a load cell (Statham force transducer) mounted on a ball-joint pivot. The handle was grasped with the hand supine so that the biceps would contribute maximally to the torque about the elbow which was proportional to the measured force at the wrist. To minimize errors due to biceps contributions to wrist supination, the subject was instructed to pull upwards on the handle without twisting. Unfortunately other synergistic flexor muscles, such as the brachialis and the brachioradialis, also contribute to the torque about the elbow. Thus, while the load cell output gave an accurate measure of total elbow torque, we cannot guarantee that it correlates equally highly with biceps force. However, this method proved to be quite successful, although our only unequivocal measure of muscle force was obtained from the amputee.

To measure the force in the amputee's muscle tunnel, a specially designed isometric load cell was mounted directly on the prosthesis which the amputee normally uses (see Fig. 2). The load cell was an aluminum cantilever beam instrumented with four strain gauges used in bridge configuration. The bridge amplifier and the cantilever beam were mounted on a fully adjustable jig (see Fig. 2). Cables from the

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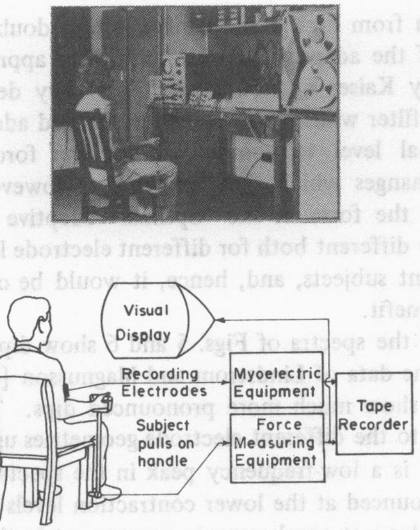

Fig. 1. Experimental arrangement for the able-bodied subject. Except for the force-measuring instrumentation, the same arrangement was used for the amputee subject.

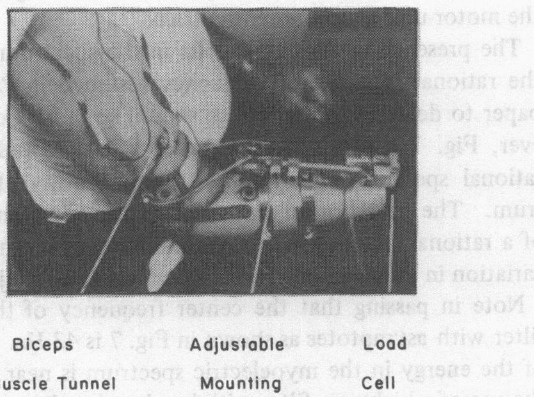

Fig. 2. Instrumentation for measuring the force in the amputee subject's biceps. Note the cineplastic muscle tunnel through the biceps.

muscle tunnel were connected to the load cell using standard Hosmer/Dorrance fittings. Following standard procedure the cables were pretensioned to 1 lb to keep them taut.

Four pairs of differential electrodes with preamplifiers mounted directly on top of the electrodes (obtained from Motion Control, Inc.) were applied over the belly of the biceps (see Fig. 3). The electrodes were 12.7 mm-diameter disks of stainless steel spaced longitudinally at 35 mm centers and laterally at 14 mm centers and were applied without electrode paste and without prior skin preparation. A single, separate reference electrode was applied over the bony prominence of the acromion. Preamplifier characteristics are shown in Fig. 4.

The outputs of the myoelectric preamplifiers and the load cell were recorded on FM magnetic tape (Ampex FR-1300) with a bandwidth of 0–2.5 kHz. These data were subsequently digitized and processed using the hybrid computation facility of the Mechanical and Civil Engineering Departments at M.I.T.

Experimental Procedures

The subject was first instructed to exert maximum force for a short period of time–3 to 5 s. All subsequent recordings during a given experimental session were taken at 25, 10, and

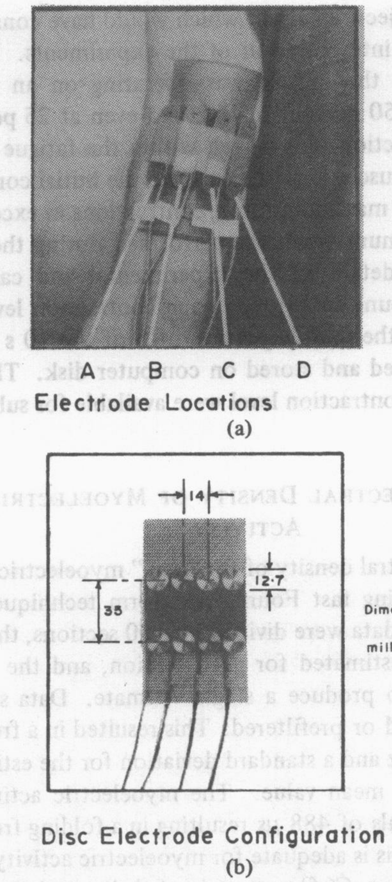

Fig. 3. (a), (b) Location and configuration of the myoelectric instrumentation.

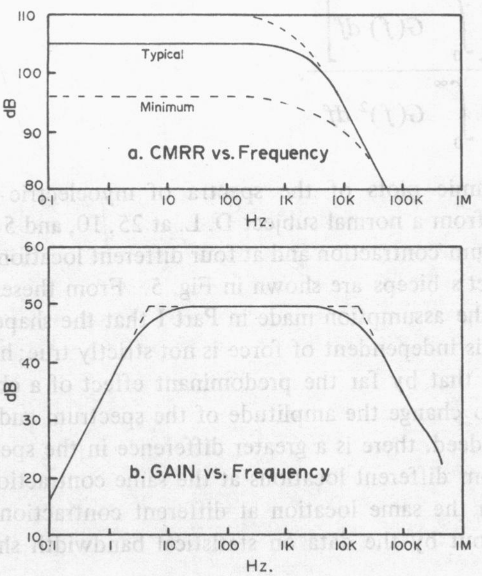

Fig. 4. Myoelectric preamplifier characteristics.

5 percent of this maximum contraction. The contraction was controlled by displaying the target level and the output of the load cell to the subject via a dual beam CRO. The subject was instructed to pull on the load cell to match its output to the target, maintain this condition for 10 s, and subsequently relax and rest for at least 10 s. This sequence was repeated six times at each contraction level. The procedure was chosen to avoid

fatiguing the subject, an effect which would have considerably complicated the interpretation of the experiments. With the above procedure the muscle was operating on an effective "duty cycle" of 50 percent or less, and even at 25 percent of maximum contraction, this is well within the fatigue limits of normal human muscle [2]. Aside from the initial contraction to determine the maximum force, contractions in excess of 25 percent of maximum were not performed during the experiment. Further details of the experimental and calibration procedures are found in [3]. At each contraction level 5 s of data taken from the middle portion of 5 of the 10 s contractions were digitized and stored on computer disk. Thus, 25 s of data at each contraction level were available for subsequent processing.

Power Spectral Density of Myoelectric Activity

The power spectral density of the "raw" myoelectric activity was estimated using fast Fourier transform techniques. The available 25 s of data were divided into 50 sections, the power spectral density estimated for each section, and the 50 estimates averaged to produce a single estimate. Data segments were not weighted or prefiltered. This resulted in a frequency resolution of 2 Hz and a standard deviation for the estimate of 14 percent of its mean value. The myoelectric activity was sampled at intervals of 488 μs resulting in a folding frequency of 1.024 kHz. This is adequate for myoelectric activity. After the power spectrum $G(f)$ was estimated, its statistical bandwidth B_s was estimated using [4]

$$B_s \triangleq \frac{\left[\int_0^\infty G(f)\, df\right]^2}{\int_0^\infty G(f)^2\, df}.$$

Logarithmic plots of the spectra of myoelectric activity recorded from a normal subject D. L. at 25, 10, and 5 percent of maximum contraction and at four different locations across the subject's biceps are shown in Fig. 5. From these we can see that the assumption made in Part I that the shape of the spectrum is independent of force is not strictly true; however, it is clear that by far the predominant effect of a change in force is to change the amplitude of the spectrum and not its shape. Indeed, there is a greater difference in the spectra obtained from different locations at the same contraction force than from the same location at different contraction forces, as borne out by the data on statistical bandwidth shown in Table I.

Fig. 6 shows logarithmic plots of the spectra of myoelectric activity recorded from the amputee subject E. N. Table II shows the statistical bandwidth as a function of electrode location and contraction force. These data indicate that there is no major difference between the myoelectric activity from an amputee's muscle and normal muscle. Such differences as are observed are most likely to be due to individual differences between subjects (e.g., physique, thickness of subcutaneous fat, etc.).

The data from Figs. 5 and 6 raise serious doubts about the validity of the adaptive prewhitening filter approach recommended by Kaiser and Petersen [5]. They devised a prewhitening filter whose characteristics changed adaptively with mean signal level to compensate for the force-dependent spectral changes which they observed. However, our data show that the form of the "optimal" adaptive filter would need to be different both for different electrode locations and for different subjects, and, hence, it would be costly and of dubious benefit.

Some of the spectra of Figs. 5 and 6 show dips which corroborate the data of Lindstrom and Magnusson [6], although their data show much more pronounced dips. This is most likely due to the different electrode geometries used. In addition, there is a low-frequency peak in the spectrum which is most pronounced at the lower contraction levels (see Fig. 5). This indicates a strong harmonic component in the signal and has been predicted on the basis of a mathematical model of the generation of myoelectric activity by Le Fever and De Luca [7] as being due to the predominant firing frequency of the motor-unit-action potential train.

The presence of dips and peaks in the spectrum means that the rational function of frequency assumed in Part I of this paper to describe the spectrum should be of high order. However, Fig. 7 shows the asymptotes of a two pole, one zero rational spectrum superimposed upon the myoelectric spectrum. The deviation of the shape of the spectrum from that of a rational function is of the same order of magnitude as the variation in shape among forces, locations, and subjects.

Note in passing that the center frequency of the bandpass filter with asymptotes as shown in Fig. 7 is 43 Hz. Thus, most of the energy in the myoelectric spectrum is near 60 Hz, and the use of a high-pass filter with break point above 60 Hz in an attempt to combat 60 Hz interference should be avoided if at all possible.

Optimal Myoprocessor Performance: Comparison of Smoothers

To verify the performance of the optimal myoprocessor derived in Part I, we implemented each stage of it digitally and processed the recorded data. To compare the derived optimal smoother with the more common first-order, low-pass filter, we normalized their response speeds such that the step response rise time to 95 percent of the final value was 0.25 s for each smoother. This requires an averaging time of 0.25 s for the optimal running averager and a time constant of 79.28 ms for the low-pass filter. The performance of either smoother can be predicted (see Part I of this paper) using

$$\mathrm{SNR} = \left[\frac{\Gamma(N/2 + 1/a)\,\Gamma(N/2)}{\Gamma(N/2 + 1/2a)^2} - 1\right]^{-1/2} \qquad (1)^1$$

if we recognize that the linear transfer characteristic of the standard myoprocessor is equivalent to the assumption that $a = 1$ in the equation

[1] A convenient approximation to (1) is $\mathrm{SNR} = \sqrt{2N} \cdot a$. This is accurate for large values of N.

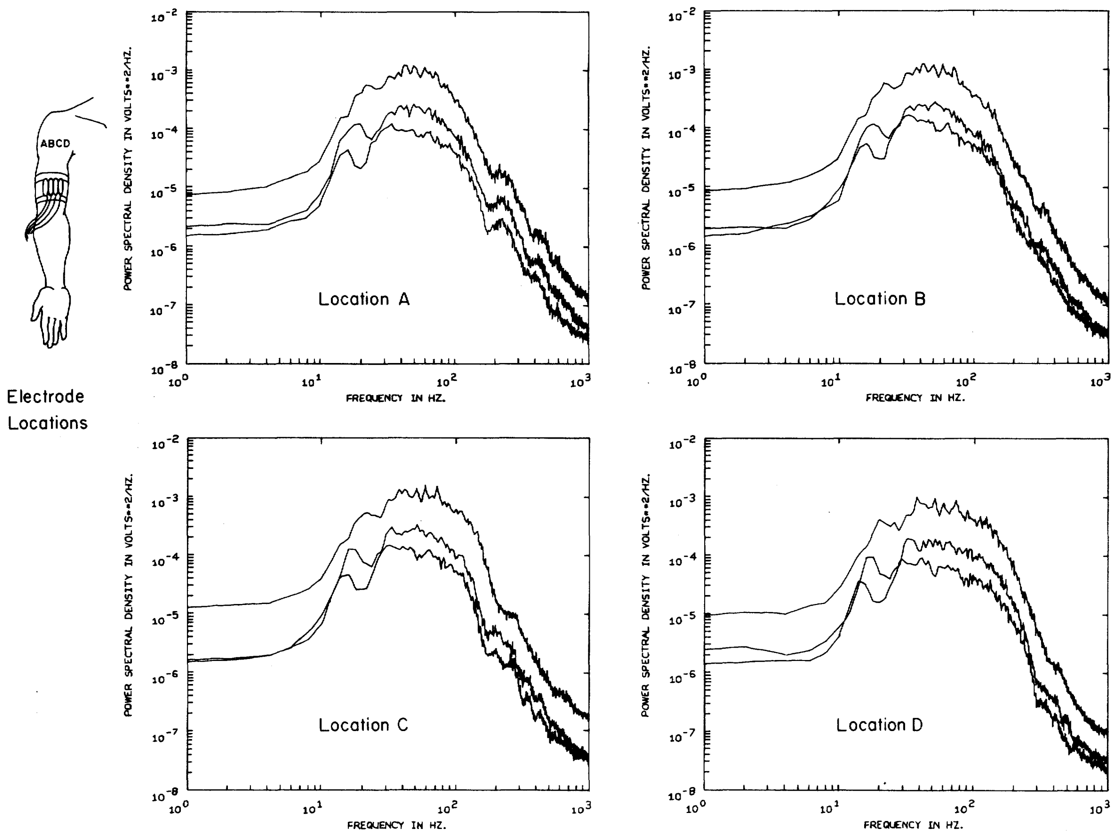

Fig. 5. Logarithmic plots of myoelectric signal power spectra obtained at three contraction levels (5, 10, and 25 percent of maximum voluntary contraction) from four locations across the biceps brachii of able-bodied subject D. L.

TABLE I
STATISTICAL BANDWIDTH OF MYOELECTRIC ACTIVITY RECORDED FROM NORMAL SUBJECT D. L.

		Electrode Location			
		A	*B*	*C*	*D*
Contraction Force (Percent of Maximum)	25%	103.90	114.91	119.43	146.29
	10%	107.34	112.23	106.69	146.94
	5%	113.41	114.89	109.08	146.86
			← Lateral	Medial →	

$$\sigma = kF^a \tag{2}$$

relating muscle force to the standard deviation of myoelectric activity (see Part I). To use (1) we need to know the equivalent number of independent samples N which contribute to the estimate. This is given by [4]

$$N = 2B_sT$$

where B_s is the statistical bandwidth and T is the total length of data contributing to the estimate. For the running averager T equals 0.25 s; for a first-order, low-pass filter, it is shown in [4] that the equivalent data length is given by

$$T_e = 2\tau$$

where τ is the filter time constant.

To make use of our observations of actual muscle force, we defined the measured signal-to-noise ratio as

$$\mathrm{SNR} = \left[\frac{E\{F\}^2}{E\{(\hat{F} - F)^2\}}\right]^{1/2}. \tag{3}$$

Table III shows the predicted and measured values of the signal-to-noise ratio obtained by processing the myoelectric activity of subject D. L. at each contraction level using a first-order, low-pass filter and ideal running averager. Note the excellent agreement between predicted and observed signal-to-noise ratios. Because (1) was derived from the assumption of an amplitude-modulated Gaussian process, this excellent agreement is ample proof that these assumptions are more than adequate for the design of a myoprocessor. The percentage improvement provided by using an averager instead of a low-pass filter is predicted by the analysis to be 26 percent, and our experiments verify this. Note that we are in substantial disagreement with Kreifeldt [8] who reported an improvement of 44 percent. This may be due to the method Kreifeldt used for measuring the signal-to-noise ratio.

DEMODULATION

The analysis of Part I dictates the use of a square-law demodulator. We processed the same segment of data using several values of the demodulator exponent around this value. Note that the "relinearizer" in the optimal myoprocessor maintains a linear transfer characteristic whatever the demodulator exponent. The signal-to-noise ratio obtained by taking the absolute value is not significantly different from that obtained by squaring. This is fortuitous because the former is

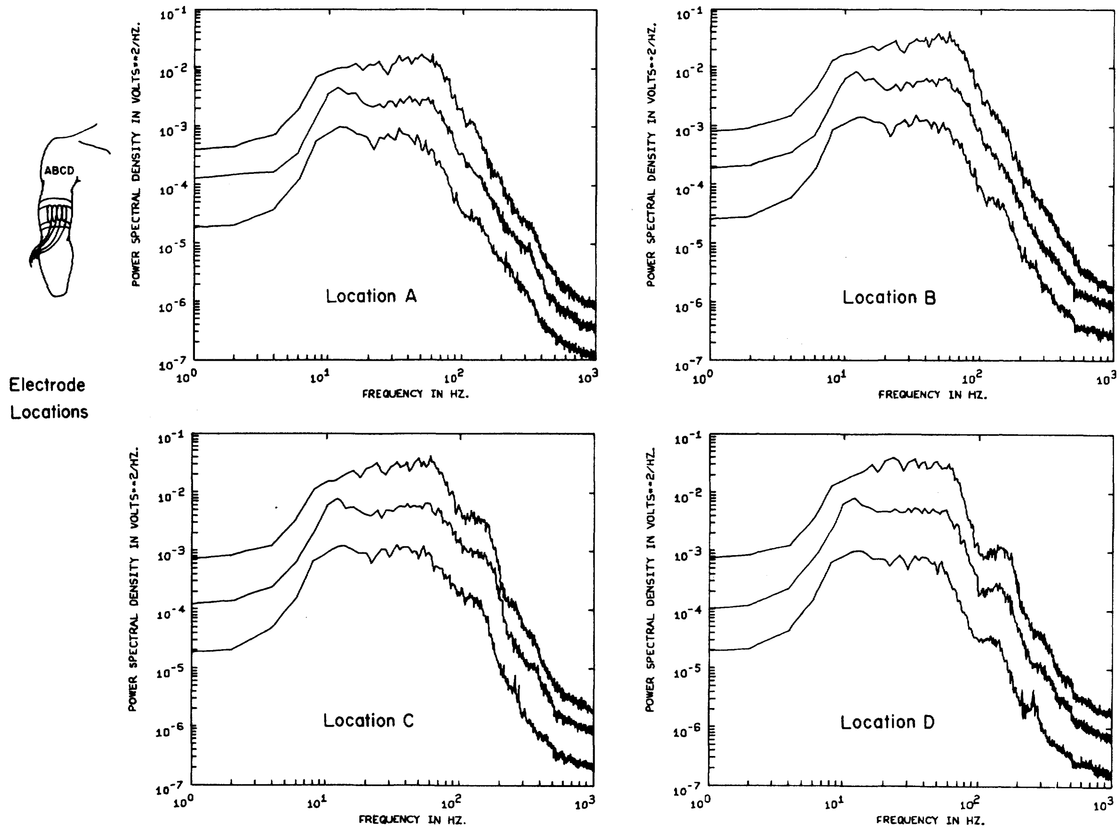

Fig. 6. Logarithmic plots of myoelectric signal power spectra obtained at three contraction levels (5, 10, and 25 percent of maximum voluntary contraction) from four locations across the biceps brachii of amputee subject E. N.

TABLE II
STATISTICAL BANDWIDTH OF MYOELECTRIC ACTIVITY RECORDED FROM AMPUTEE SUBJECT E. N.

		Electrode Location			
		A	*B*	*C*	*D*
Contraction Force	25%	87.53	89.39	100.64	76.45
(Percent of Maximum)	10%	83.72	88.05	113.97	79.84
	5%	72.39	80.66	100.33	70.34
			← Lateral	Medial →	

much simpler to implement in analog hardware requiring only a simple rectifier.

PREWHITENING VIA ELECTRODE GEOMETRY

As discussed in Part I a reduction of interelectrode spacing should increase the bandwidth of the myoelectric signal and improve the signal-to-noise ratio. This matter was investigated briefly using rectangular electrodes. Dimensional analysis of the situation prompted the use of the configuration shown in Fig. 8. Results are tabulated in Table IV.

As predicted a decrease in interelectrode spacing results in an increase in bandwidth and a concomitant increase in the signal-to-noise ratio of the single-channel myoprocessor output. The shape of the electrodes appears to influence the shape of the spectrum, but the important dimension is the outer edge spacing which controls the predominant effect—the increase in bandwidth.

A further experiment was performed using an array of four electrodes of identical geometry shown in Fig. 9. Spectra obtained using these electrodes are shown in Fig. 10. As observed previously with disk electrodes, the bandwidth and shape of the spectrum varies with location, the electrode pair on the medial border of the biceps showing significantly higher bandwidth than the other three. Experimental results are shown in Table V. These experiments confirm that a reduction of the interelectrode spacing can increase the signal-to-noise ratio of the myoprocessor.

Some problems were encountered using this electrode configuration. Because of the reduced interelectrode spacing, build up of perspiration under the electrodes could easily form a short-circuit path between them. The result was a random fluctuation of the effective preamplifier gain and dc bias. This problem of electrode shorting will become particularly acute if the electrodes are inside a prosthesis socket, a location particularly conducive to the build up of perspiration. Thus, electrode shorting may limit the effectiveness of this technique of myoprocessor improvement. However, the problem need not be insurmountable, and further research is required in this area.

COMBINATION OF MULTIPLE CHANNELS

A crucial test of our spatiotemporal sampling artifact hypothesis (see Part I) is the improvement afforded by the use of multiple channels of myoelectric activity. The expected improvement can be predicted as follows: after the spatial prewhitening transformation, the m channels of myoelectric

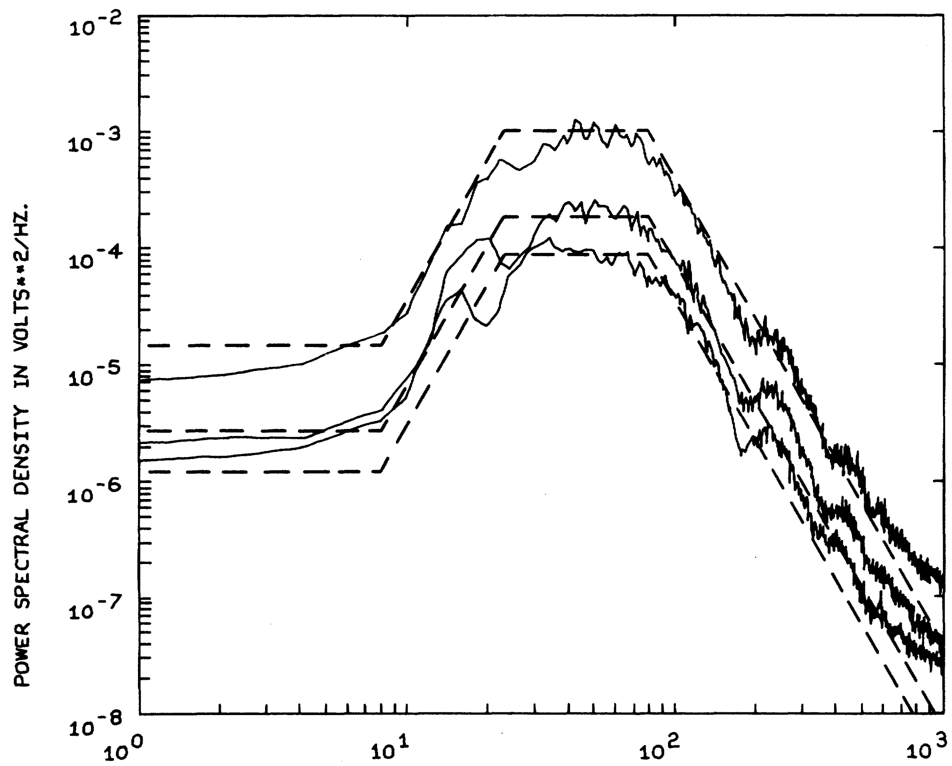

Fig. 7. Asymptotes of a rational spectrum (break points at 8, 23, and 80 Hz) superimposed on myoelectric spectra of able-bodied subject D. L. at three contraction levels (5, 10, and 25 percent of maximum voluntary contraction).

TABLE III
PREDICTED AND MEASURED SIGNAL-TO-NOISE RATIO FOR THE MYOELECTRIC ACTIVITY OF SUBJECT D. L.

	Contraction Level	Electrode Location A	B	C	D
A. Rectified and Low-Pass Filtered					
Predicted	25%	8.12	8.54	8.70	9.64
	10%	8.25	8.43	8.23	9.65
	5%	8.48	8.54	8.31	9.65
Measured	25%	7.41	7.75	8.56	9.91
	10%	8.07	7.89	8.13	9.86
	5%	8.21	8.91	8.75	9.33
B. Rectified and Averaged					
Predicted	25%	10.19	10.72	10.93	12.10
	10%	10.36	10.59	10.33	12.12
	5%	10.65	10.72	10.44	12.12
Measured	25%	9.20	9.63	10.72	12.72
	10%	10.74	10.07	10.41	13.20
	5%	11.07	11.15	11.20	12.90

activity are uncorrelated, Gaussian distributed random variables of equal variance which are squared and added to form a pooled estimate of the total variance. This pooled estimate is chi-squared distributed with a number of degrees of freedom equal to the sum of the degrees of freedom contributed by each channel; i.e.,

$$N_{\text{TOTAL}} = \sum_{i=1}^{m} N_i = \sum_{i=1}^{m} 2 \cdot T \cdot B_{si}.$$

This number is then used in (1). However, this prediction was obtained by assuming perfect spatial prewhitening, and as discussed in Part I, this is unlikely to be achieved if any of the eigenvalues are too small (i.e., correlations too high). In effect, this prediction sets an upper bound on the expected signal-to-noise ratio.

For the data of the 25 percent contraction level using disk electrodes, a signal-to-noise ratio of 22.01 is predicted for the four channels combined, approximately twice the value predicted for any one channel alone. The signal-to-noise ratio observed experimentally at the 25 percent contraction level was 16.21. Although this is below the theoretical upper bound, it is an average improvement of 47 percent over the single-channel case using the same smoother. The results for the 10 and 5 percent contractions are similar (see Table VI).

The eigenvalues obtained at the 25 percent contraction level are shown in Table VII. Because of the high cross correlation between electrodes, the smallest eigenvalue is 47 times smaller than the largest. This accounts for the deviation between the observed signal-to-noise ratio and the theoretical prediction.

This is borne out by the results obtained by combining the inner pair of channels (cross correlation: 0.89, ratio of eigenvalues: 14) and by combining the outer pair of channels (cross correlations: 0.53, ratio of eigenvalues: 3). In both cases the signal-to-noise ratios obtained experimentally are quite close to the predicted theoretical limits (see Table VIII).

The above results provide strong verification of our sampling artifact hypothesis and the modeling and analysis presented in Part I.

When we compare the performance obtained using four

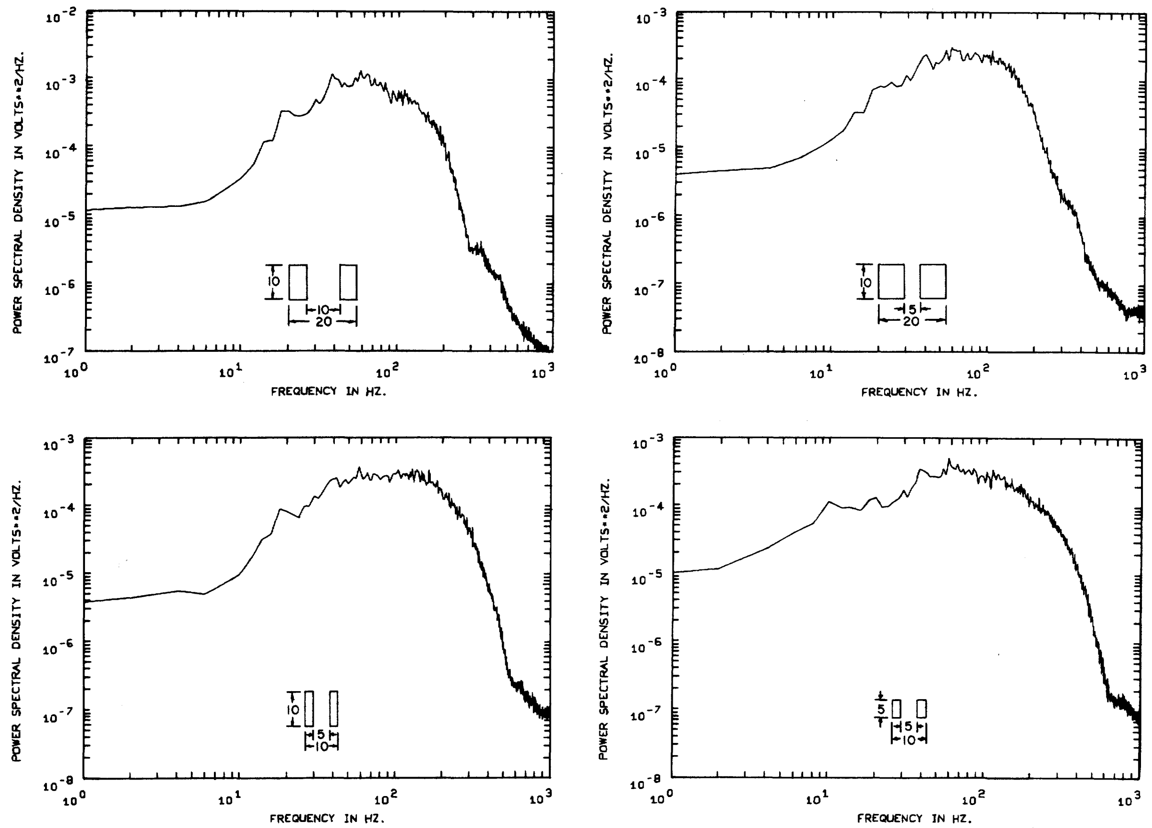

Fig. 8. Logarithmic plots of the myoelectric signal power spectra resulting from different electrode configurations (subject D. L., 25 percent contraction).

TABLE IV
BANDWIDTH AND SIGNAL-TO-NOISE RATIO AS A FUNCTION OF ELECTRODE GEOMETRY; MYOELECTRIC ACTIVITY OF SUBJECT D. L. (b = ELECTRODE WIDTH, c = INNER EDGE SPACING, AND d = OUTER EDGE SPACING, SEE FIG. 8).

b mm	c mm	d mm	Bandwidth Hz	SNR
10	10	20	149.20	12.78
10	5	20	162.16	12.49
10	5	10	250.35	17.42
5	5	10	247.43	16.92

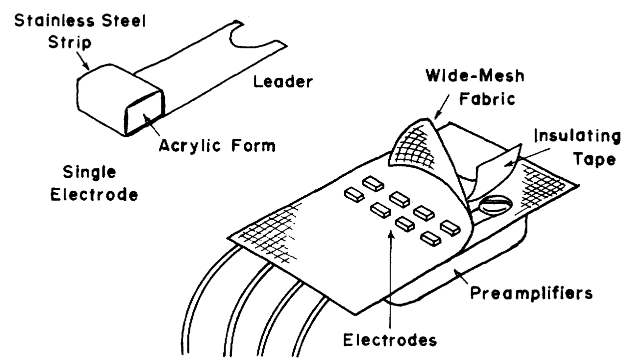

Fig. 9. Construction of electrodes with reduced spacing.

channels—47 percent improvement over the average single channel—with that obtained using the outer two channels alone—35 percent improvement, we see that because of the high correlation, the additional two channels do not add as much to the performance of the myoprocessor as the first two. Indeed, we found that after spatial prewhitening, the channel corresponding to the smallest eigenvalue could be omitted from the combination without incurring more than 2 percent degradation in performance. Consequently, we investigated ways of reducing the cross correlation between channels.

Covariance and Correlation Between Channels

The covariance and correlation matrices for the normal subject D. L. for each contraction level are shown in Table IX. The covariance and correlation matrices obtained from the amputee subject E. N. are shown in Table X. These are quite similar to those of the normal subject. The correlation between electrode activities is plotted against lateral electrode separation for each force in Fig. 11.

These data agree with that of Person and Mishin [9]. Contrary to the amplitude-modulation assumption of Part I, the correlation matrix changes with force, but as with the variation of the frequency spectrum, the changes are not pronounced, and the effect on the myoprocessor performance is insignificant. This is borne out by the data of Table XI, which shows that the spatial prewhitening transformation coefficient matrix $\Lambda^{-1/2} \cdot \Phi^T$ obtained at one contraction level can be used at other contraction levels with little effect on the myoprocessor performance. As with the frequency spectrum, the predominant effect of a change in force is to modulate the amplitude of the myoelectric activity.

(The prewhitened channel with the smallest eigenvalue has been omitted from the combination.)

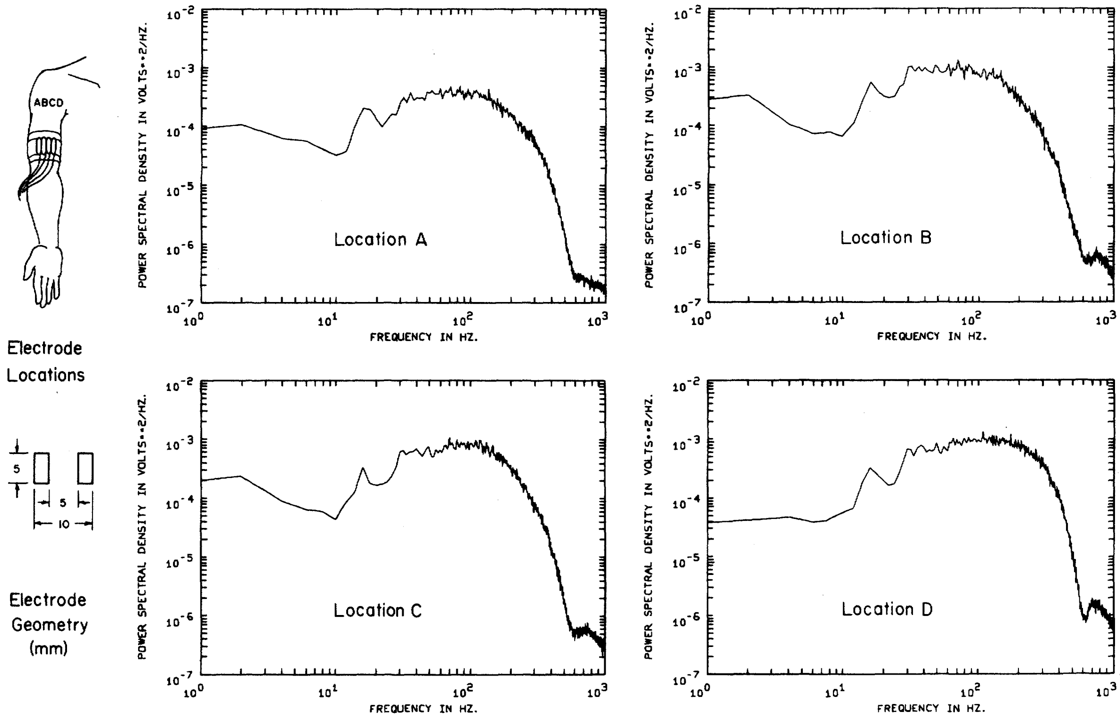

Fig. 10. Logarithmic plots of the myoelectric signal power spectra obtained from the electrodes of Fig. 9 (subject D. L., 10 percent contraction).

TABLE V

BANDWIDTH AND SIGNAL-TO-NOISE RATIO FOR RECTIFIED AND AVERAGED MYOELECTRIC ACTIVITY FROM ELECTRODES WITH REDUCED SPACING; 10 PERCENT CONTRACTION FORCE, SUBJECT D. L.

	$b = 5$ mm $c = 5$ mm $d = 10$ mm Electrode Location				
	A	*B*	*C*	*D*	
Bandwidth	248.39	220.27	230.87	321.27	Hz
Predicted SNR	15.76	14.84	15.19	17.92	
Measured SNR	13.80	17.83	15.19	17.99	

TABLE VI

SIGNAL-TO-NOISE RATIO FOR FOUR CHANNELS COMBINED. MYOELECTRIC ACTIVITY OF SUBJECT D. L. USING DISK ELECTRODES

Contraction Force	25%	10%	5%
Signal-to-Noise Ratio	16.21	16.45	17.82

All of the results of combining multiple channels of myoelectric activity presented so far have been for disk electrodes, which exhibit the high cross correlations between channels seen in Tables IX and X.

To combat the high cross correlation, the spacing between the two electrodes of a differential pair can be reduced. Reduction of interelectrode spacing implies a concomitant reduction of the effective "pickup" region of an electrode pair, thus, reducing the region of overlap between the "pickup" regions of adjacent electrode pairs. However, ultimately this will also reduce the total number of active muscle fibers contributing to the final processor output. According to the spatiotemporal sampling artifact hypothesis, which is strongly supported by our data above, this reduction will ultimately degrade the myoprocessor performance. This indicates the existence of an optimum interelectrode spacing beyond which further reductions will not yield improvements in myoprocessor performance.

The covariance and correlation matrices obtained for a 10 percent contraction using the electrodes of Fig. 9 are shown in Table XII. The correlation is plotted against the lateral spacing in Fig. 12. Comparing this with Fig. 11, we see that the entire curve has been shifted downward; the maximum correlation has been reduced to 0.55, the minimum correlation to 0.14. As shown in Fig. 12 this is in reasonable agreement with results reported by Person and Mishin [9].

Using these electrodes, the predicted upper bound on the signal-to-noise ratio for the four channels combined is 31.95 (assuming perfect spatial prewhitening). The ratio of the largest eigenvalue to the smallest is 7; thus, on the basis of the two-channel experiments with disk electrodes, we would expect the measured signal-to-noise ratio to be reasonably close to this upper bound. In fact, the signal-to-noise ratio obtained is 23.43. Similar results were obtained for pairwise combinations of channels. Measured signal-to-noise ratios fell short of predictions by 18 and 15 percent despite eigenvalue ratios of 2 and 4, respectively. These results may be due to the expected tradeoff between bandwidth and the pickup region mentioned above. It was observed that single-motor-unit action potentials could readily be identified in the raw myoelectric signal from these electrodes, indicating that not enough superimposition was taking place for the central limit theorem to apply and yield a Gaussian amplitude distribution, thus leading to inaccuracies in the spatial prewhitening transformation. In other words, not enough active motor units

TABLE VII
EIGENVALUES AND EIGENVECTORS OF THE COVARIANCE MATRIX OF THE MYOELECTRIC ACTIVITY OF SUBJECT D. L. RECORDED AT 25 PERCENT CONTRACTION FORCE

	Four Channels			
Eigenvalues, λ_i	2.558×10^{-1}	4.024×10^{-2}	1.989×10^{-2}	5.416×10^{-3}
Corresponding Eigenvectors	-0.447	0.610	0.464	-0.461
	-0.533	0.292	-0.119	0.785
Φ	-0.582	-0.231	-0.664	-0.409
	-0.421	-0.699	0.574	0.061

	Inner Two Channels (*B* & *C*)			Outer Two Channels (*A* & *D*)	
λ_i	1.660×10^{-1}	1.206×10^{-2}	λ_i	1.078×10^{-1}	3.545×10^{-2}
Φ	0.663	0.749	Φ	0.707	0.707
	0.749	-0.663		0.707	-0.707

TABLE VIII
PREDICTED AND MEASURED SIGNAL-TO-NOISE RATIO FOR TWO CHANNELS OF MYOELECTRIC ACTIVITY OF SUBJECT D. L. COMBINED

SNR	Contraction Force		
	25%	10%	5%
Predicted			
Inner Pair	15.31	14.80	14.97
Outer Pair	15.32	15.95	16.27
Measured			
Inner Pair	13.35	14.26	15.12
Outer Pair	15.07	14.76	16.30

TABLE IX
COVARIANCE AND CORRELATION MATRICES FOR THE MYOELECTRIC ACTIVITY OF SUBJECT D. L.

	Covariance				Correlation			
Location	*A*	*B*	*C*	*D*	*A*	*B*	*C*	*D*
25% Contraction	7.16×10^{-2}	6.51×10^{-2}	5.58×10^{-2}	3.62×10^{-2}	1.0	0.862	0.665	0.505
		7.97×10^{-2}	7.64×10^{-2}	4.81×10^{-2}		1.0	0.863	0.637
			9.84×10^{-2}	6.15×10^{-2}			1.0	0.732
				7.17×10^{-2}				1.0
10% Contraction	1.62×10^{-2}	1.43×10^{-2}	1.13×10^{-2}	0.683×10^{-2}	1.0	0.844	0.617	0.414
		1.77×10^{-2}	1.61×10^{-2}	0.979×10^{-2}		1.0	0.838	0.565
			2.06×10^{-2}	1.33×10^{-2}			1.0	0.715
				1.68×10^{-2}				1.0
5% Contraction	7.34×10^{-3}	6.48×10^{-3}	4.96×10^{-3}	2.81×10^{-3}	1.0	0.644	0.585	0.392
		10.07×10^{-3}	8.49×10^{-3}	4.77×10^{-3}		1.0	0.744	0.501
			9.89×10^{-3}	5.71×10^{-3}			1.0	0.684
				7.06×10^{-3}				1.0

were contributing to the processor output. However, it is unwise to draw conclusions from a single experiment, and further investigation is required.

Whatever the explanation, the important thing is that a significant improvement in performance was achieved. A signal-to-noise ratio of 23.43 is better than the previous case using disk electrodes by 46 percent.

RELINEARIZATION

The function of this element is to invert the static relation between muscle force and the standard deviation of myoelectric activity. As shown in Part I, this relation may be determined from observations of muscle force versus mean rectified myoelectric activity. Many different forms have been reported in the literature for this relation—linear [10]-[11], nonlinear [12]-[15], and even multivalued [16]. We used the data of Vredenbregt and Rau [15] who showed that if muscle force is normalized about its maximum value at a given muscle length, the variability in the relation is greatly reduced. These authors found a nonlinear relation between muscle force and myoelectric activity of biceps brachii, to which we fit a power-law curve (see Fig. 13). Note that close to the origin the relation does not deviate much from linearity, which may account for the reports of a linear relation. Lippold [10], for example, did not record forces above 10 percent of maximum. Also, the slope of the curve near the origin is clearly different from

TABLE X
COVARIANCE AND CORRELATION FOR MYOELECTRIC ACTIVITY OF SUBJECT E. N.

Location	Covariance				Correlation			
	A	*B*	*C*	*D*	*A*	*B*	*C*	*D*
25% Contraction	0.680	0.883	0.553	0.406	1.0	0.859	0.509	0.429
		1.554	1.175	0.725		1.0	0.715	0.507
			1.739	1.088			1.0	0.719
				1.317				1.0
10% Contraction	0.129	0.156	0.106	0.062	1.0	0.811	0.488	0.367
		0.289	0.227	0.122		1.0	0.697	0.480
			0.367	0.197			1.0	0.689
				0.223				1.0
5% Contraction	0.236×10^{-1}	0.282×10^{-1}	0.170×10^{-1}	0.093×10^{-1}	1.0	0.835	0.471	0.380
		0.479×10^{-1}	0.356×10^{-1}	0.170×10^{-1}		1.0	0.696	0.487
			0.545×10^{-1}	0.259×10^{-1}			1.0	0.697
				0.254×10^{-1}				1.0

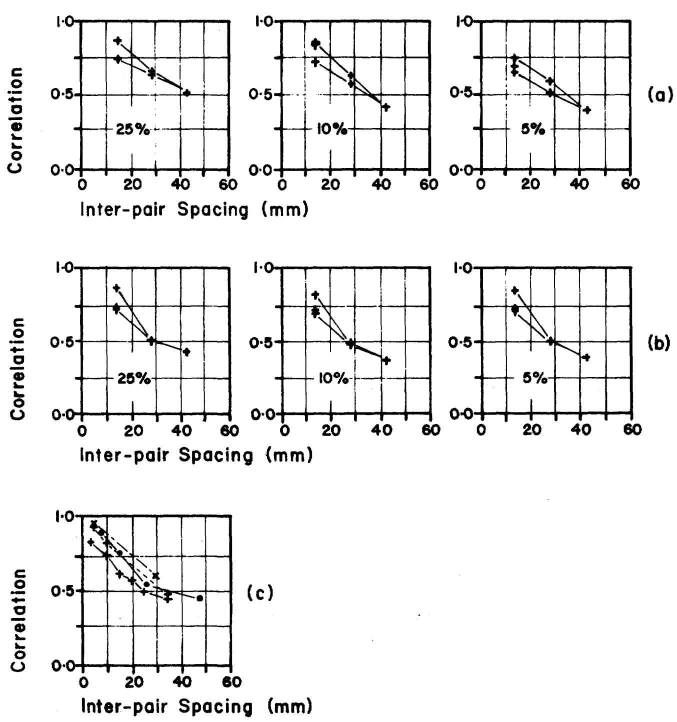

Fig. 11. Cross correlation versus interpair spacing for (a) able-bodied subject D. L. and (b) amputee subject E. N. using the disk electrodes of Fig. 3 at the contraction levels indicated. Data reported by Person and Mishin for biceps brachii at an unspecified contraction level are shown in (c).

zero, indicating that although a power-law curve fits these data with a correlation coefficient of 0.98, this form is not exact. We used the power-law form primarily to keep our analysis simple, but in addition the power-law curve has the advantage that the form of the relationship is independent of arbitrary scale changes in either the input or the output variable. This is of practical value because the amplitude of myoelectric activity for a given force and amplifier gain may change from day-to-day, and it is desirable that the performance of the myoprocessor not be degraded as a result.

Using the power-law relation (2), the relinearizer has the form

TABLE XI
INSENSITIVITY OF MYOPROCESSOR PERFORMANCE TO INTERCHANGE OF PREWHITENING TRANSFORMATION COEFFICIENT MATRICES

SNR		$\Lambda^{-1/2} \cdot \Phi^T$ (25%)	$\Lambda^{-1/2} \cdot \Phi^T$ (10%)	$\Lambda^{-1/2} \cdot \Phi^T$ (5%)
Contraction Level	25%	15.98	15.72	16.05
	10%	16.00	15.88	16.43
	5%	16.99	17.11	17.46

$$\hat{F} = K\{\cdot\}^{1/a}$$

where K is a scaling constant. Prediction of the effect of this relinearizer on myoprocessor performance is a simple matter of substituting the appropriate value of a into (1). From the approximation

$$\text{SNR} = \sqrt{2N} \cdot a,$$

we can see that the predicted signal-to-noise ratio is proportional to a; thus, it is tempting to increase the value of a beyond that dictated by the mathematical analysis. The experimental results of applying this relinearizer to the four-channel case for various values of parameter a are shown in Table XIII.

The agreement between predictions and observations is good. For the four-channel case with reduced-spacing electrodes and a value of a equal to 4, the measured signal-to-noise ratio is 82.32, 10 times better than the performance obtained using a single-channel with disk electrodes and a simple first-order, low-pass filter. However, it must be remembered that for this value of a the condition

$$E\{\hat{F}\} = F$$

has been severely violated. The relation between muscle force and myoprocessor output is now nonlinear in the extreme. This will be detrimental in most applications. However, it is clearly better to overestimate the value of a rather than underestimate it. From Fig. 13 we can see that it would not be unreasonable to use $a = 2$ rather than $a = 1.74$. Using $a = 2$ yields

TABLE XII

(a) COVARIANCE AND CORRELATION OF THE MYOELECTRIC ACTIVITY FROM FOUR ELECTRODES WITH REDUCED SPACING. 10 PERCENT CONTRACTION, SUBJECT D. L.

Location	Covariance A	Covariance B	Covariance C	Covariance D	Correlation A	Correlation B	Correlation C	Correlation D
	6.425×10^{-2}	4.522×10^{-2}	2.612×10^{-2}	1.739×10^{-2}	1.0	0.478	0.285	0.141
		13.957×10^{-2}	7.414×10^{-2}	3.269×10^{-2}		1.0	0.548	0.180
			13.106×10^{-2}	5.114×10^{-2}			1.0	0.291
				23.585×10^{-2}				1.0

(b) EIGENVALUES AND EIGENVECTORS OF COVARIANCE MATRIX

Eigenvalues λ_i	2.926×10^{-1}	1.706×10^{-1}	6.634×10^{-2}	4.117×10^{-2}
Corresponding eigenvectors Φ	0.738	0.666	-0.097	-0.043
	0.469	-0.395	0.758	0.224
	0.443	-0.586	-0.423	-0.530
	0.198	-0.237	-0.488	0.817

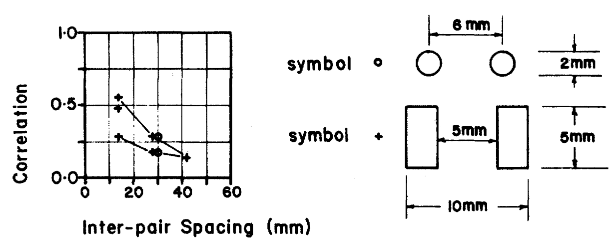

Fig. 12. Cross correlation versus interpair spacing for able-bodied subject D. L. at 10 percent contraction level using the electrodes of Fig. 9 (symbol +). Person and Mishin describe their electrodes as ". . . interelectrode distance being 6 mm, electrode area—about 3 mm². The distance between the leads was 30 mm." A probable geometry for each pair is shown above (symbol o).

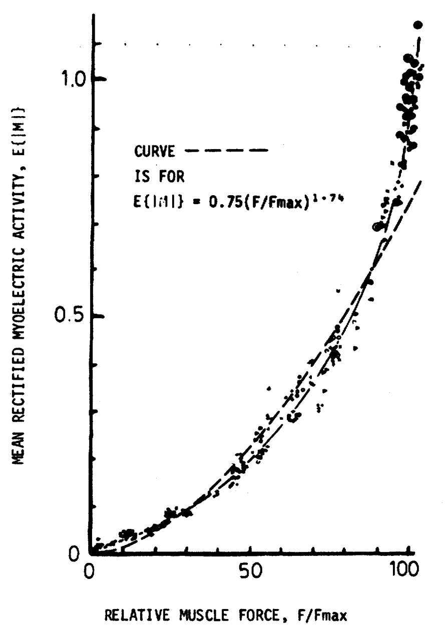

Fig. 13. Relative muscle force expressed as a percentage versus mean rectified myoelectric activity. Redrawn from Vredenbregt and Rau.

performance $5\frac{1}{2}$ times better than that obtained using common myoprocessing techniques.

A summary of all of the improvements in myoprocessing which were achieved by digital processing is shown in Fig. 14. Fig. 14(h) is included to indicate the performance which may be attained if processor nonlinearity can be tolerated.

TABLE XIII

MYOPROCESSOR PERFORMANCE WITH RELINEARIZER. MYOELECTRIC ACTIVITY FROM FOUR ELECTRODE PAIRS WITH REDUCED SPACING; SUBJECT D. L.; 10 PERCENT CONTRACTION FORCE; b = 5 mm, c = 5 mm, AND d = 10 mm

	Parameter a 1	2	3	4
Predicted SNR	–	46.86	70.29	93.72
Measured SNR	23.43	46.27	66.32	82.32

ON-LINE ANALOG OPTIMAL MYOPROCESSOR

All of the experimental results presented up to this point were processed digitally off-line. As a final test, the optimal myoprocessor was implemented using analog hardware so as to provide an on-line, real-time estimate of muscle force. Analog hardware was used rather than a digital microprocessor because the time required to perform the software multiplication necessary for the prewhitening transformation would have severely curtailed the maximum attainable sampling rate. The multichannel myoprocessor is shown in Fig. 15. To avoid problems with electrode shorting, the wider spaced disk electrodes were used. Fig. 16 shows the same section of tape-recorded data processed in real time by:

1) rectifying and low-pass filtering a single channel (τ = 79 ms);
2) rectifying and averaging a single channel (T = 250 ms);
3) combining four channels and averaging;
4) combining four channels, averaging, and relinearizing with a equal to 2;
5) shows the corresponding muscle force.

This figure clearly shows the inadequacy of the common myoprocessing technique and the improvements attained using the techniques presented in this paper. The signal-to-noise ratio of trace D is 32.

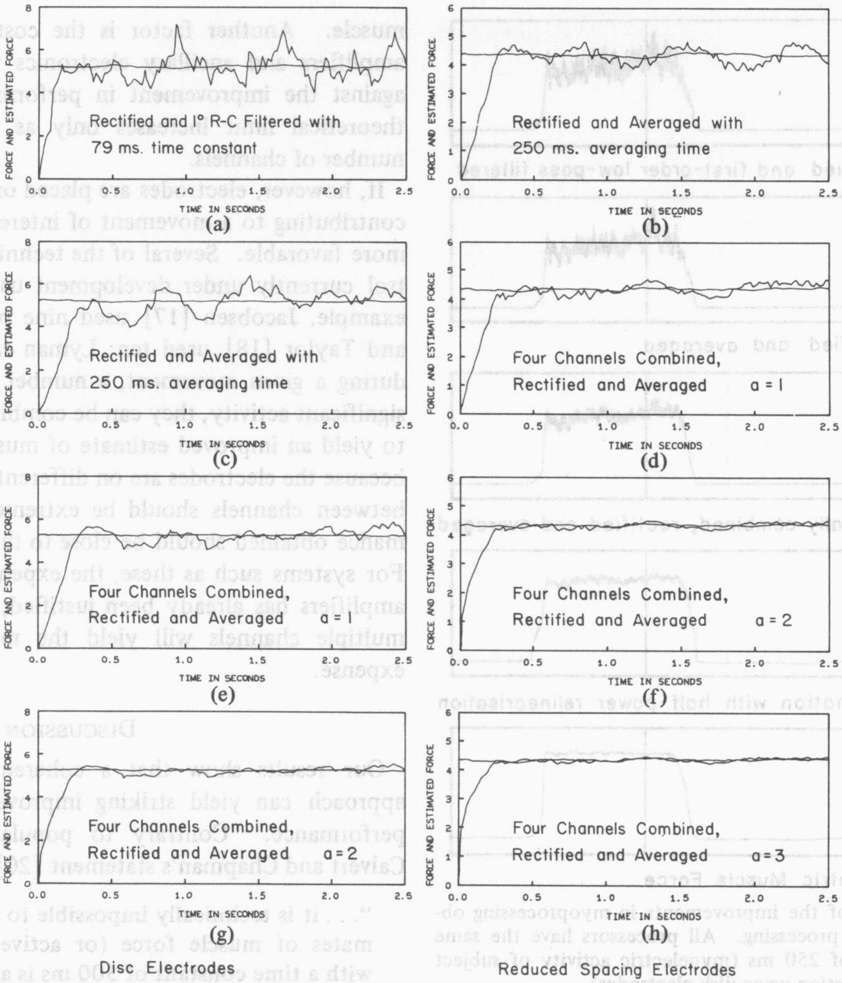

Fig. 14. (a)–(h) Summary of the improvements in myoprocessing obtained by off-line digital processing. All processors have the same step response rise time of 250 ms (myoelectric activity of subject D. L. at 10 percent contraction).

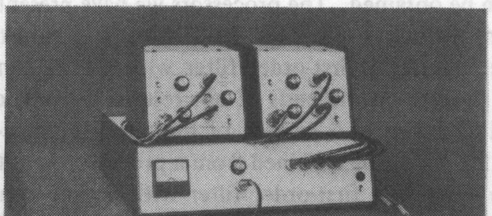

Fig. 15. Analog version of optimal myoprocessor.

Tradeoffs in the Design of Myoprocessors

The final myoprocessor configuration used in a particular application is a design decision which must take into account many factors, such as cost, complexity, reliability, etc. One of the major contributions of the analytical and experimental results presented in these papers is that they permit objective evaluation of the tradeoffs which can be made in the design of a myoprocessor. In summary, these are:

1) Demodulation: the analysis dictates a square-law demodulator; however, our experiments show that no significant detriment in performance results from the use of a demodulator with unity exponent. There is no experimental justification for the added cost and complexity of using anything other than a simple rectifier.

2) Smoothing: Under the assumptions of our analysis, for a given processing time, the best smoother obtainable is a running averager. The only design decision to be made is the choice of the averaging time—the longer the averaging time, the smoother the myoprocessor output. However, longer averaging time means poorer temporal fidelity. In addition, the signal-to-noise ratio only increases with the square root of the processing time.

3) Relinearization: If the myoprocessor output is required to be proportional to muscle force, then the relinearizer must be designed to invert the static relation between muscle force and myoelectric activity. However, in some applications it may be worth "overrelinearizing." This will result in a smoother myoprocessor output, but which is nonlinearly related to muscle force.

4) Prewhitening: Because of the variability of the myoelectric spectrum between persons and with electrode location, electronic prewhitening did not seem to be very practical, and we did not investigate it. However, the myoprocessor performance can be improved by the simple expedient of reducing the electrode spacing. A major problem with this method is the likelihood of electrode shorting due to perspiration

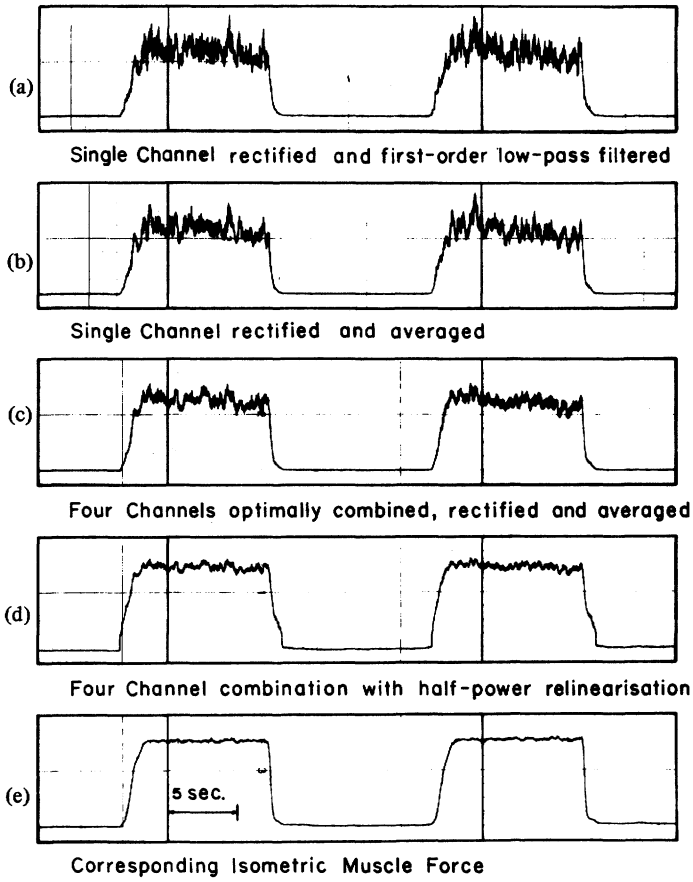

Fig. 16. (a)-(e) Summary of the improvements in myoprocessing obtained by on-line analog processing. All processors have the same step response rise time of 250 ms (myoelectric activity of subject D. L. at 25 percent contraction using disk electrodes).

buildup, a problem which becomes aggravated the closer the electrodes are. This problem is particularly acute when the electrodes are inside the stump socket of an amputee.

5) Combination of Multiple Channels: The technique of combining multiple channels of myoelectric activity can yield a considerable improvement in myoprocessor performance, and in contrast to the above methods, whose scope for further improvement is limited, the combination of multiple channels promises considerable improvement beyond the performance demonstrated in this paper. Of course, the method is not without its drawbacks which will vary with the application. Although further myoprocessor improvements can be obtained by using more than four channels, if all the electrode pairs are to be placed over one muscle, there is a limit to the number of channels which can be used. Firstly, the available surface area of muscle for placing electrodes is restricted, especially if the remnant musculature of an amputee's stump is to be used. Because of atrophy, the remnant muscle can be quite small, and if the muscle was not tied down during surgery, it may move within the stump, and the "window" of surface area under which the muscle can be found under all conditions is smaller still. Secondly, adjacent electrode pairs cannot be placed too close together or their activities will become too highly correlated. Further experimental work is required to determine the practical limit on the number of reasonably uncorrelated channels which can be obtained from a single muscle. Another factor is the cost of the additional preamplifiers and ancillary electronics which must be weighed against the improvement in performance which even at the theoretical limit increases only as the square root of the number of channels.

If, however, electrodes are placed on several distinct muscles contributing to a movement of interest, the situation is much more favorable. Several of the techniques of myoelectric control currently under development use multiple channels; for example, Jacobsen [17] used nine pairs of electrodes; Wirta and Taylor [18] used ten; Lyman *et al.* [19] used nine. If, during a given movement, a number of these channels shows significant activity, they can be combined using our techniques to yield an improved estimate of muscle force. Furthermore, because the electrodes are on different muscles, the correlation between channels should be extremely low, and the performance obtained should be close to the theoretical predictions. For systems such as these, the expense of the additional preamplifiers has already been justified, and the combination of multiple channels will yield the maximum return for this expense.

Discussion

Our results show that a coherent mathematically based approach can yield striking improvements in myoprocessor performance. Contrary to popular opinion, typified by Calvert and Chapman's statement [20] that:

> "... it is technically impossible to obtain consistent estimates of muscle force (or active state) unless a filter with a time constant of 300 ms is applied to the rectified EMG ..."

we have shown that timely and accurate estimates of muscle force can be obtained. The processors we have presented yield consistent estimates (see Figs. 14 and 16) and have the same response times as a first-order filter with a time constant of 79 ms. Indeed, on the basis of our results, we predict that a standard deviation of 5 percent of the mean (a signal-to-noise ratio of 20) could be obtained from a processor with the same response time as a first-order filter with a time constant of 15 ms. We do not recommend this, however, as we feel that such response speeds are unnecessary, and the tradeoffs between speed and performance should be made to yield better signal-to-noise ratios.

Our experimental setup allowed us to measure our subject's ability to maintain steady isometric force with visual feedback. It is interesting to note in passing that despite our early misgivings about the reliability of the measurement of muscle force in normals, the method appeared to be just as dependable as our measurements on the amputee with cineplasty. This indicates that the patterns of synergy of a muscle group are stable and unchanging (at least for one limb position), and the activity of a single muscle may be used to represent the group.

Our subjects could typically maintain a steady force with an rms deviation about the mean of 1 percent with no significant dependence on the mean force level. Similar results have been reported elsewhere [21]. This corresponds to a signal-to-noise ratio of 100 and provides a good benchmark for assessing myoprocessor performance. Under the restriction that myoprocessor outputs vary linearly with force, the best performance we attained was a signal-to-noise ratio of about 45, corresponding to an rms deviation about the mean of just over 2 percent. While this is excellent performance, there is room for improvement.

In the on-line analog multichannel myoprocessor, setting up the prewhitening transformation requires the adjustment of 16 coefficient potentiometers, a troublesome and tedious process. This task could be simplified by using a digital microprocessor to implement the myoprocessor. Furthermore, the covariance between channels may change from application to application or as a result of fatigue. A microprocessor of sufficient capability could be programmed to estimate the covariance between channels continuously and adjust the prewhitening transformation coefficients accordingly. Such a system would be extremely flexible as it would be capable of adapting to different subjects and different electrode locations. At the time of completion of this project in 1976, microprocessors of sufficient speed were not available. However, in view of the recent rate of advance in microprocessor technology, microprocessors of sufficient speed may be available now, or if not, are likely to become available within a matter of years.

An interesting and unexplored possibility of great importance is that the covariance between channels, and hence the prewhitening transformation, may change when the same muscle performs different functions (e.g., biceps flex the elbow and supinate the wrist). If this effect was sufficiently pronounced, it would be possible to obtain signals related to the different movements from the same set of electrodes. This is the spatial equivalent of the frequency domain technique being developed by Graupe *et al.* [22]. These researchers hope to identify different roles of a muscle by observing changes in the myoelectric frequency spectrum. It is not clear that this is practical, because although we did see spectral changes which may be attributable to the action of different muscles (or different portions of the same muscle—see Fig. 5 and Table I), these changes were second-order effects at best. In contrast, we would expect the changes in the covariance between channels with different functions to be more marked as different portions of the group of muscles seen by the electrodes participate in different functions. This has been verified in part by the pattern recognition work of Lawrence *et al.* [23]. Better still would be to combine the spatial- and frequency-domain techniques and use the entire matrix of auto- and cross-power spectra between channels as an information base for distinguishing different functions. If successful, this technique could provide much needed multiple-degree-of-freedom control signals from a set of electrodes arrayed around a single group of muscles. Further research is needed on the change in the covariances with different functions, different electrode locations, different people, etc.

The part of the myoprocessor which consistently gave trouble and which is the most in need of further development is the electrode array. Currently available electrodes for detecting myoelectric activity are highly susceptible to movement artifact. Because of changes at the skin electrode interface, the sensitivity and dc bias of the detector system (electrodes and preamplifiers) change slowly with time and differs from application to application. Changes in the dc bias are easily handled, but changes in sensitivity mean that the myoprocessor has to be frequently recalibrated if it is to provide a quantitative estimate of muscle activity. Although there has been a lot of work on biopotential electrodes in the past few decades, considerable further research is required to understand and eliminate these problems and produce a stable, reliable, artifact-free electrode system. Some interesting approaches are those of Portnoy and David [24] and Hoenig *et al.* [25].

Acknowledgment

The authors wish to acknowledge the valuable counsel of Prof. T. B. Sheridan and Prof. E. Bizzi of the Massachusetts Institute of Technology and Dr. C. J. De Luca of the Children's Hospital Medical Centre. Our sincere thanks to L. Blake who typed the manuscript.

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Neville Hogan, for a photograph and biography, see this issue p. 395.

Robert W. Mann (SM'71-F'79), for a photograph and biography, see this issue p. 395.

Communications

An Automated Measuring System for EMG Silent Period

Y. YAMADA, K. ISHIOKA, AND M. M. ASH, JR.

Abstract—**A small, economical, and accurate system for measuring the silent period duration as well as the latency has been designed. In nine normal subjects electromyographic silent periods in the masseter muscle were measured both manually and by the system. With the comparison of those data, it is concluded that this measuring system is suitable for clinical use.**

I. Introduction

Electromyography has proven to be valuable in the study of the function of the masticatory muscles. In recent years there has been a good deal of attention given to the phenomenon known as the electromyographic silent period (SP) [1]. The silent period is the brief absence of electrical activity of the jaw-closing muscles observed shortly after tooth contact during chewing [2]. It is also seen following mechanical stimulation of the oral receptors during clenching of the teeth [3], [4].

It was reported by Bessette *et al.* [5] that in patients with temporomandibular joint (TMJ) pain dysfunction syndrome, the duration of the silent period in the jaw-closing muscles is increased compared to normal subjects. This suggests a potentially diagnostic value for the silent period in this syndrome in which modality of treatment is a current topic in dental research. The observations of Bessette *et al.* have been supported by other investigators [6]-[8]. These facts served to motivate the research described here.

Presently, SP research is bound to facilities which can bear the expense in money and space of electromyographic equipment. Also, the current measurement techniques are mainly manual, thus requiring a significant investment of time to reduce the data. If SP is to become a diagnostic tool for the dentist in the clinic a technique will be needed which is not too complicated or too expensive to be practical.

A small and economical system for measuring the SP duration as well as the latency of its onset from the time of the causal stimulus has been designed (Fig. 1). Unlike other automated systems that have been developed for research purposes and requiring considerable computer hardware and software, the presented system because of its small size, cost, and ease of use is an ideal instrument for the dental practitioner who is interested in the evaluation of functional disturbances of the masticatory system. The design and performance as well as the practicality for clinical use of the system are discussed.

II. Methods

To evaluate the system, silent period data were collected from nine human subjects with natural dentitions. Ages ranged from 23 to 36 years. Since the problem investigated concerns an evaluation of the new measuring system for the SP, no special effort was made to classify the subjects as normal or dysfunctional. However, no subject complained of symptoms of disturbances of the masticatory system.

Manuscript received July 25, 1979; revised February 20, 1980. This work was supported by the National Institutes of Health under Grant DE 02731-21.

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0018-9294/80/0700-0410$00.75 © 1980 IEEE