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ECG Enhancement by Adaptive Cancellation of Electrosurgical Interference

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***Abstract*—A technique utilizing a combination of adaptive noise canceling and conventional signal processing is developed to enhance electrocardiographic monitoring in the operating room by reducing the noise interference that is created by an electrosurgical instrument. Significant amounts of interference are eliminated by radio frequency shielding, passive and active low-pass filtering, and optical isolation. A digital adaptive canceler using the least mean-square algorithm of Widrow and Hoff is used to reduce the remainder of the interference, yielding an improvement in signal-to-noise ratio of approximately 110 dB. Clear electrocardiograms have been obtained with electrocautery equipment in operation.**

I. Introduction

THE electrosurgical unit (ESU), often called the "Bovie,"[1] is a useful and necessary surgical tool without which many operative procedures would be difficult or impossible to perform. Electrocardiographic (ECG) monitoring during surgery is also indispensable. However, ECG monitors are impaired by electrical interference generated by the ESU. A method for removing such interference is described in this paper. The method is based on a combination of adaptive noise canceling and conventional signal processing techniques.

The ESU generates a radio frequency (RF) signal modulated at twice the power line frequency (120 Hz in the U.S.A.). The resultant 100-200 W of broad-band spectral power is delivered to the point of the surgeon's knife to aid in cutting tissue and coagulating severed blood vessels. The ground return is accomplished by a large-area grounding (dispersive) plate that is placed in contact with the patient's back or thigh. During ESU activation, extraordinarily large transient voltages (100-400 V) are generated ubiquitously over the patient's skin surface. These are caused by the ESU currents passing through tissues of varying shapes, densities, and conductivities in reaching the grounding plate [1]-[4]. Additional surface voltages are established through parasitic capacitive coupling from the ESU generator to the patient. The magnitude of these voltages is significantly influenced by the power of the ESU output, and by the spatial relationship of the ESU and the patient. In addition, in some ESU's, a small but constant low-frequency current is passed through the ground plate to assure grounding integrity. Small amounts of patient motion, which change spatial relationships, also induce random low-frequency modulation to the entire ECU current.

The ECG electrodes applied to the patient are generally of the silver/silver-chloride type [5]. The large ESU noise voltages that appear at the patient's skin surface can cause rectification at these electrodes, and can result in demodulation of the burst-type envelope. Thereby, additive nonstationary interference can combine with the ECG waveform in the ECG passband. When the ECG electrodes are placed to obtain the maximum ECG electrical vector, a signal-to-noise ratio of roughly -90 dB exists during periods of ESU activation.

In developing a technique to eliminate the ESU interference, several methods were combined to deal with each of the component problems identified above. Classical circuit design was used for the preliminary signal conditioning, but the final solution made use of an adaptive filter to cancel the nonstationary interference in the passband of the ECG.

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[1] Bovie was original inventor; see [2].

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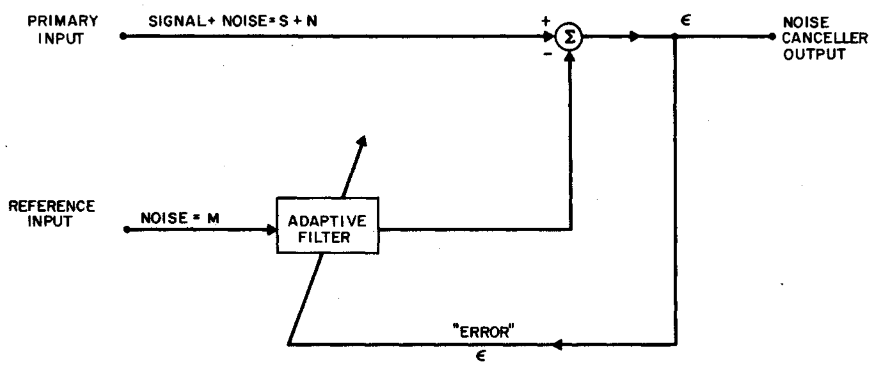

Fig. 1. The adaptive noise canceler.

II. Adaptive Noise Canceling

The subject of adaptive noise canceling is introduced and is treated extensively by Widrow *et al.* [6]; a brief discussion of the subject is presented here. Fig. 1 is a block diagram of an adaptive noise-canceling system. There are two inputs and one output. The "primary" input contains signal plus additive noise. The "reference" input contains noise alone. The primary and reference noises must be uncorrelated with the primary signal, but correlated with each other if some benefit is to result from use of the noise canceler. The reference noise is filtered to create a waveform which is a best least squares estimate of the noise in the primary input. The filtered reference noise is subtracted from the primary input to produce the system output, which in turn is a best least squares estimate of the primary signal.

An adaptive filter is needed to perform the function just described, since the relationship of the primary and reference noises is unknown and can be time varying. The adaptive filter produces an output signal that is based on the reference noise input and on a second input, the "error" signal. It automatically adjusts its own internal parameters (which directly control its impulse response) to find the best combination which minimizes the power (mean square) of the error. In the adaptive noise canceler of Fig. 1, the error is obtained from the system output itself. Thus, the entire system adapts in such a way as to minimize its own output power. This causes the output to be a best least squares estimate of the primary signal, as will be demonstrated below.

In Fig. 1, the primary signal is s, the primary noise is n, and this noise is uncorrelated with s. The reference noise is m and is uncorrelated with s but correlated with n. The adaptive filter output is y, and since it is generated from m, it is correlated with n but uncorrelated with s. The output is the error ϵ, given by

$$\epsilon = s + n - y. \tag{1}$$

The mean square of ϵ is of interest. It can be obtained as follows:

$$\epsilon^2 = s^2 + (n - y)^2 + 2s(n - y)$$
$$E[\epsilon^2] = E[s^2] + E[(n - y)^2] + 2E[s(n - y)]. \tag{2}$$

Since s is uncorrelated with both n and y,

$$E[\epsilon^2] = E[s^2] + E[(n - y)^2]. \tag{3}$$

Adapting the filter minimizes the mean square of ϵ. This has no effect on the input signal s. Accordingly,

$$\min E[\epsilon^2] = E[s^2] + \min E[(n - y)^2]. \tag{4}$$

Minimizing the mean square of ϵ, therefore, causes the filter output y to be a best least squares match to n.

Now, (1) can be rewritten as

$$\epsilon - s = n - y. \tag{5}$$

Minimizing the mean square of ϵ results in

$$\min E[(\epsilon - s)^2] = \min E[(n - y)^2]. \tag{6}$$

Minimizing the mean square of ϵ, therefore, causes ϵ to be a best least squares match to the primary signal s. This result is obtained by the adaptive process without requiring any prior knowledge of the signal and noise statistics except that the signal is uncorrelated with the noises.

In [6], many applications of adaptive noise canceling are shown. Biomedical applications that are presented there illustrate how 60 Hz power-line interference can be removed from ECG signals, how interference from the maternal heart can be removed from fetal ECG signals, and how electrical signals from a transplanted heart can be separated from the signals generated by the independent beating of the residual atrium of the original heart.

III. Electrosurgical Interference

The present problem of electrosurgical interference rejection is another application for an adaptive noise canceler, but it has certain special attributes that have not been previously encountered. Primary electrodes can be placed on the body to receive the maximum ECG signal and these electrodes will, of course, receive interference from the ESU when it is energized. Reference electrodes can be placed on the body to receive the ESU interference while at the same time receiving a minimal amount of ECG signal. It would seem simple then to connect the primary and reference signals to the adaptive filter as in Fig. 1, and thereby solve the problem. Unfortunately, this solution, as is, will not work.

The primary leads receive ECG signals in the microvolt range, but as reported above, they also receive high-voltage RF noise. Likewise, the reference leads derive high-voltage RF noise when the Bovie is energized. It was necessary to first isolate these high-voltage RF noises from any and all electronic devices in the system, to make sure that the ECG leads were only connected to high-impedance loads so that essentially no current flowed in them (to prevent nonlinearity or rectification effects at the skin-to-electrode contact points), and to make sure that no new ground paths were possible for Bovie currents through our equipment. These requirements posed some interesting electronic problems.

The high-power RF current to the ESU comes from an oscillator whose power supply is not pure dc but instead is unfiltered, full-wave-rectified, power-line ac. The result is a suppressed-carrier RF signal, modulated at twice the power-line frequency. The ESU circuit is simple and was developed in its basic form at least 50 years ago. The RF current waveform was designed to be effective for cutting and for cauterization. The power density is greatest between 200 kHz and 50 MHz [1]. However, there is significant power below 200 kHz and it is this component of the interference that creates the greatest difficulty.

IV. Electrosurgical Interference Rejection

The basic approach for eliminating the ESU interference is outlined in block diagram form in Fig. 2. The Bovie power unit is shown connected to the patient's ground plate and to the knife itself. Typical placement of the primary and reference electrode pairs are also shown. Signals from each of these pairs are first passively filtered to eliminate the gross RF interference. Passive filters in this application were designed to present a high-impedance (megohms) load to the patient electrodes. Passive filters were chosen for this "front-end" function, since active filters would overload from the high RF voltages present on these electrodes. After passive filtering, the voltage loads were sufficiently reduced in magnitude and spectral extent that battery powered active amplification and buffering were then possible.

Optical couplers were used in the next stage to provide isolation, to remove remaining common mode RF interference, and to prevent new connections from developing between the patient and earth ground. Low-pass active filters followed the optical couplers, and were used to eliminate all remaining spectral components above approximately 600 Hz in the primary and reference channels. These filters were powered from the ac line and shared a common ground with the usual ECG monitoring equipment.

At this point, the majority of the ESU noise was removed. However, there still remained strong interference components at 60, 120, and 180 Hz plus other, random, low-frequency signals. The effects of higher harmonics were negligible. The magnitudes and phases of the 60, 120, and 180 Hz interference components varied substantially and rapidly as the surgeon worked with and moved the knife. An adaptive noise canceler was used to remove these remaining nonstationary interference components from the signal.

The adaptive noise canceler delivered performance superior to that of fixed, tuned notch filters. This resulted from the fact that if the power line frequency changed, the adaptive process sustained notches in the frequency domain exactly where they were needed. In fact, an adaptive system designed for use with a 60 Hz power frequency need not be changed for use with a 50 Hz power supply.

Some of the details of the system design and the electronic design are important and they are presented next.

V. System Design

In this section, the component subsystems of Fig. 2 are described in more detail. The primary and reference paths were processed identically and in a linear fashion to preserve the linear relationship between their ESU noise components. The initial filtering of the primary and reference channels was accomplished using three passive, single pole, low-pass filters in cascade with 3 dB cutoff frequencies at 100, 30, and 1 kHz, respectively. To prevent high-frequency currents from leaking through these filters, each of the three filter stages was isolated from the other filter stages. The isolation was achieved by placing each stage in a separate copper-clad compartment of a larger copper box. The passive filters were followed by high-input impedance, balanced, FET differential amplifiers to ensure a minimum of low-frequency current loading at the output of the passive filters, which in turn ensured only small low-frequency currents through the ECG electrode pads attached to the patient. Thus, the linear coupling characteristic of the pads was retained and rectification at the pads was minimized. Additionally, all signals were referenced to a common point (ECG pad) on the patient's body. This reference point floated with respect to earth ground; thus, it maintained patient isolation. Such isolation eliminated the possibility of accidental skin burns at the ECG electrodes.

The isolated signals were coupled to the remaining earth-ground referenced circuits by using optical coupling. Hewlett-Packard 5082-4354 optical isolators were the specific devices employed. The 5082-4354's, normally nonlinear devices, were used in a circuit configuration that employed optical feedback to achieve 1 percent nonlinear distortion [10]. The optical isolator and differential amplifiers were also shielded in separate copper-clad compartments to prevent transfer of RF common mode leakage currents to the output of the patient isolation stage.

The optical isolators were followed by three-pole, active, linear phase (Bessel) filters with a cutoff frequency of about 600 Hz. The linear phase characteristic was maintained at low frequencies to preserve the important phase relationships of the ECG. The primary and reference low-pass output signals were then fed through analog-to-digital converters (ADC's) into an LSI 11/03 minicomputer where the adaptive filtering was performed. The resulting output was then fed through a digital-to-analog converter (DAC) to a conventional analog ECG monitor.

A block diagram of the adaptive filter is shown in Fig. 3. The reference input was fed into a tapped delay line where successive inputs were individually weighted and summed to form a best least squares estimate of the ESU noise that remained in the primary. The resultant ECG estimate that was obtained from subtraction of the two signals was used to update the weighting values in the tapped delay line. The changes were determined by the least mean-squares (LMS) algorithm:

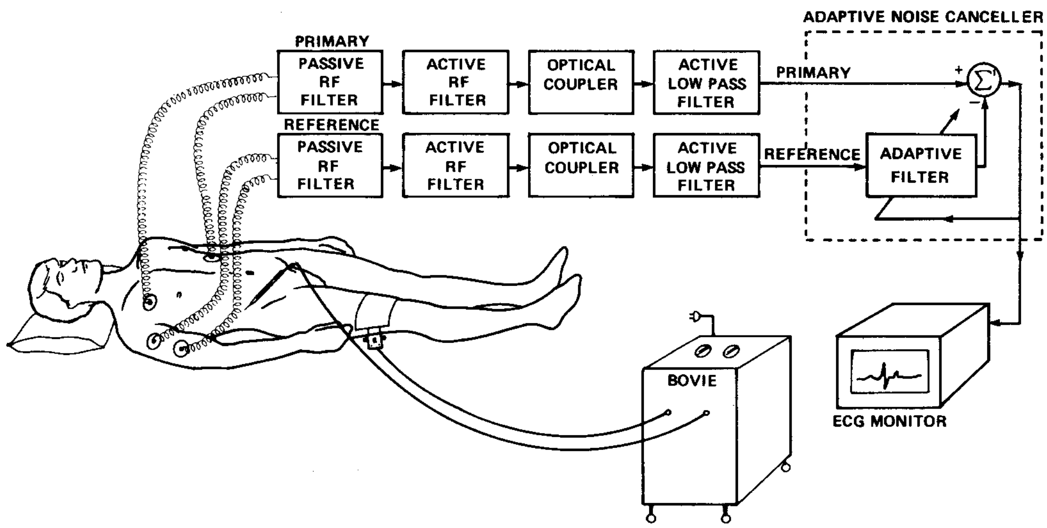

Fig. 2. ECU interference rejection system.

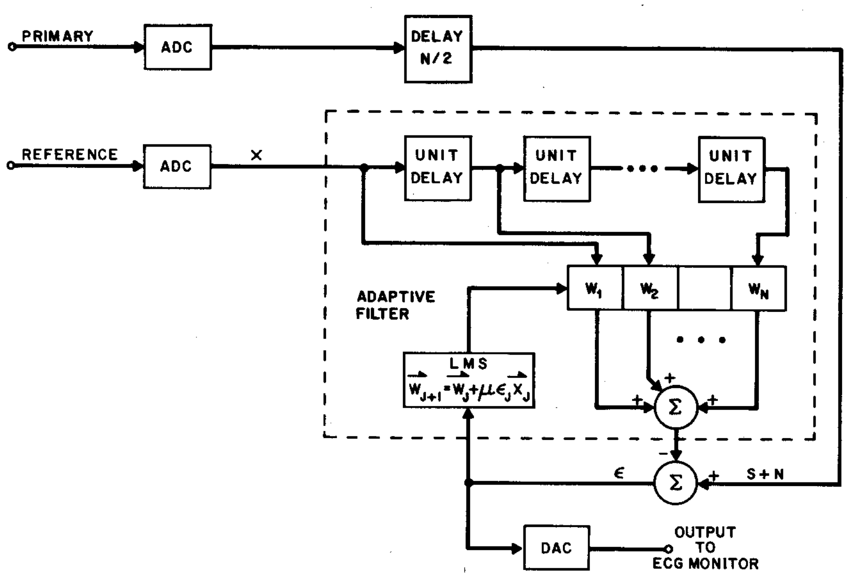

Fig. 3. Detailed block diagram of a digital adaptive noise canceler.

$$W_{j+1} = W_j + \mu\epsilon_j X_j \tag{7}$$

which is described in [6]. The delay in the primary path permits a causal adaptive impulse response to behave very much like a delayed version of a noncausal impulse response in the event that a noncausal impulse response would be advantageous [6].

Low-frequency (<25 Hz) additive distortion was found to be present in the primary and reference inputs, and was probably caused by variation in RF current flow as the patient was touched. The low-frequency distortion components were particularly troublesome. In some rare instances, these components had rendered overall system performance inadequate. To eliminate these components, a dual reference adaptive noise canceler was used as shown in Fig. 4. By using a digital low-pass filter with a cutoff frequency of 25 Hz, we divided the reference signal into two components. One component consisted of the line frequency distortion at 60, 120, 180 Hz, etc., while the other component consisted of only the low-frequency distortion below 25 Hz. These two reference signals were then applied to the inputs of two separate but simultaneously working adaptive noise cancelers, as shown in Fig. 4. The delay indicated by D in Fig. 4 is the delay introduced by the causal low-pass filter. Typically, it was about 30 sampling intervals. The main reason for the dual reference signal components was to be able to separately choose the μ parameters that controlled the rate of convergence. Automatic gain control (AGC) circuits were used on the primary and high-frequency reference signals to normalize input power, to control dynamic range, and thereby to effect an additional performance improvement.

The sampling rate chosen for this implementation was about 400 Hz. 16 bit fixed point arithmetic with 12 bit analog interfaces was found to be adequate to represent all quantities.

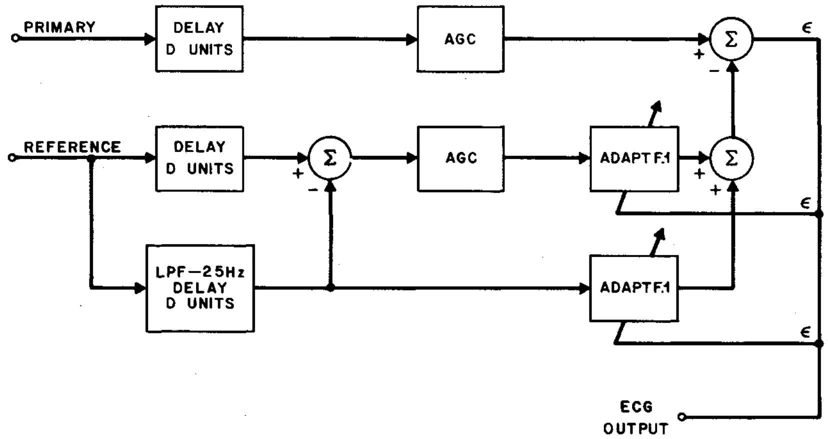

Fig. 4. Dual reference adaptive noise canceler.

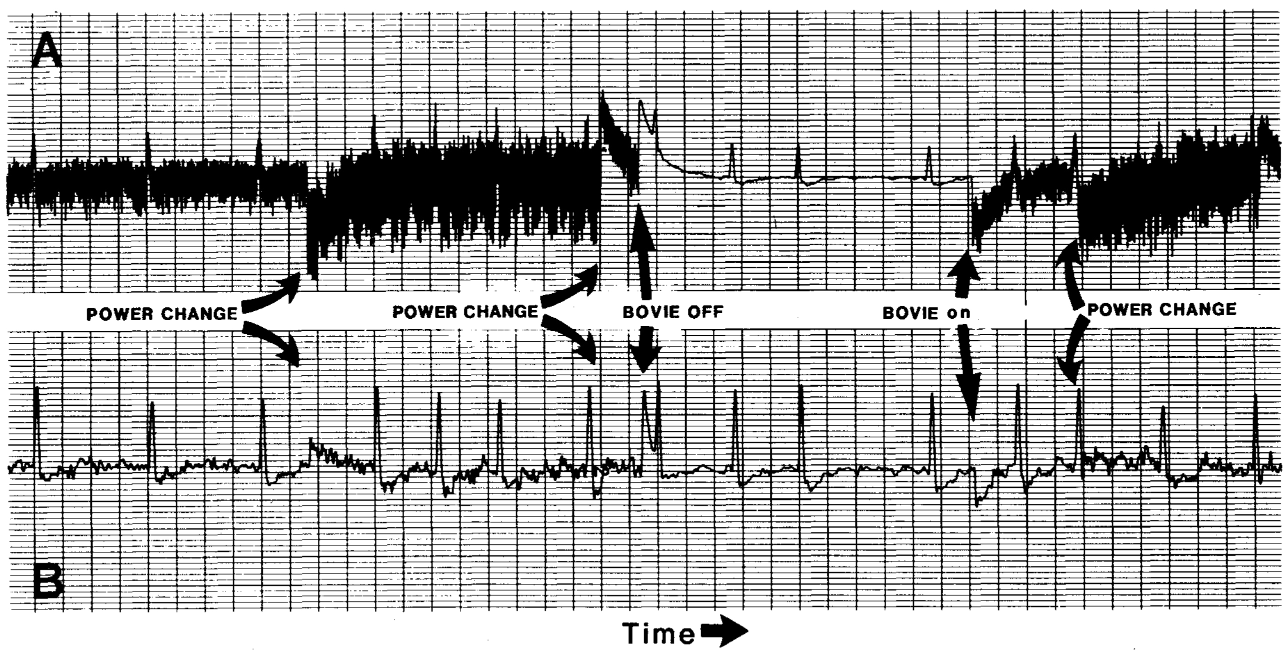

Fig. 5. Cancellation/coagulation.

The convergence parameters μ were generally chosen to be between 0.02 and 0.2 with reference input powers normalized to unity.

VI. Experimental Results

The system that was described above has been successfully used to monitor the ECG during periods of ESU energization. The inital signal-to-noise ratio (SNR) was approximately -90 dB at the ECG pads. The dual reference canceler was used to obtain the most useful results, although the single reference canceler also produced adequate results during most periods of ESU operation. A substantial reduction in input noise power (approximately 80 dB) was accomplished in the preliminary filtering stages of the system. Final noise reduction was accomplished by adaptive cancellation. The adaptive noise canceler provided an additional noise reduction of about 30 dB. The resultant SNR of about +20 dB was more than adequate to discern the heart rate as well as all of the subcomponents of the ECG wave, such as the P and T waves.

Typical experimental results are presented in Figs. 5 and 6. The upper trace (A) in each figure is the primary input to the adaptive canceler after analog preprocessing to eliminate all components above 600 Hz.[2] The noise canceler output, the final system output, is shown in the lower trace (B) in these figures. Note that as the ECU is turned off and on in the upper trace, there is a small transient in the lower plot of duration less than one heart cycle. During periods of ECU energization, the ECG is not visible in the upper trace while it remains virtually undistorted in the canceler output. There are two modes

[2]If one attempted to plot the raw data, the noise level would be so high that it would be impossible to display it on the given scale.

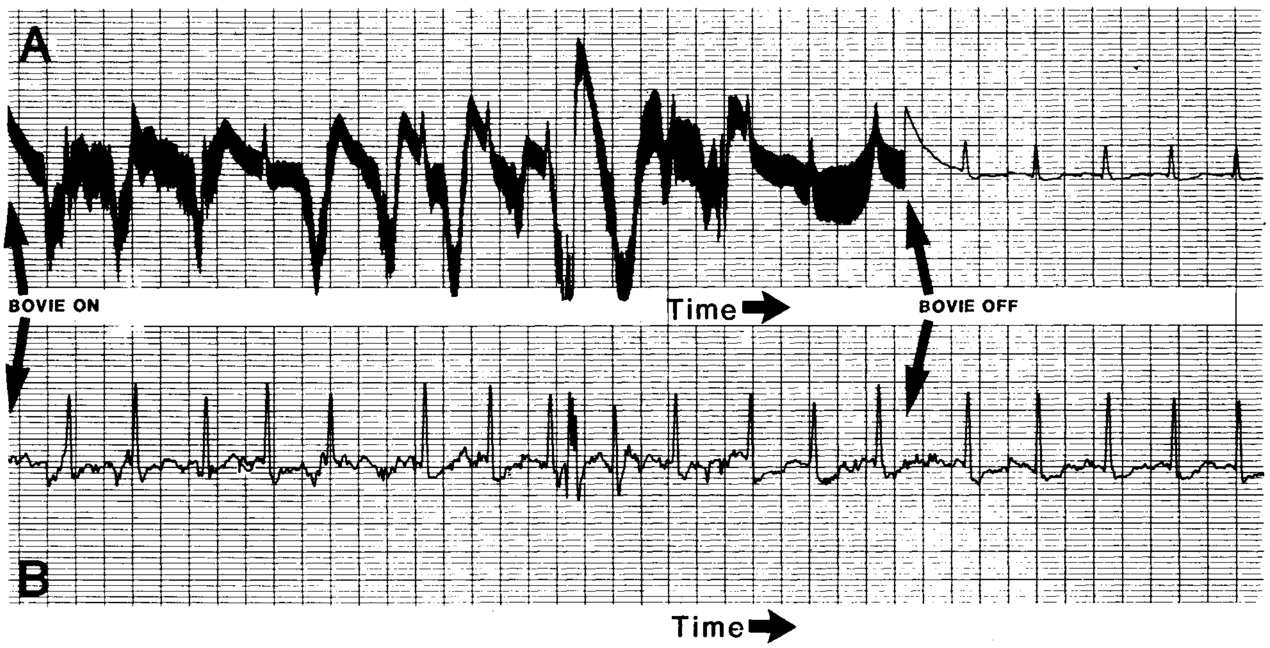

Fig. 6. Cancellation/cut.

of operation for the usual Bovie knife in wide usage, "cut" and "coagulate." These modes involve different waveforms and power levels. Cancellation of Bovie noise during coagulate is shown in Fig. 5, while cancellation of this noise during cut is shown in Fig. 6.

The overall improvement in SNR as compared to using an electrocardiograph without the system of Fig. 2 is approximately 110 dB. This unusually large improvement by a filtering and noise canceling system is due to the fact that most of the noise can be separated into several large components, each of which can be eliminated at a different stage of the system.

The adaptive portion of this system can be easily implemented on a microprocessor. The present implementation uses a PDP 11/03 minicomputer and is programmed in assembly language.

VII. Discussion and Conclusions

There is an increasing demand to continuously and precisely monitor signals with very small SNR's in the operating room. The retrieval of the ECG from a background of electrical interference is a classical example; the ECG is the most commonly monitored of the patient's electrical signals while the interfering ESU is the oldest and most essential, yet most offensive, electrical apparatus in the operating room. A general solution to the ECU noise problem was needed in order that future advances can be made in monitoring patient electrical signals. The system described above has been used successfully on human patients during surgery.

This paper illustrates that a hybrid analog-hardware/digital-software solution to these problems can offer advantages. Classical filtering, isolation, and buffering hardware can eliminate a significant portion of the interfering noise, thus reserving the final and most difficult segment for the adaptive filter. The adaptive filter is then able to improve the final results beyond that which could be obtained previously. However, the adaptive filter does require a reference source of pure interference which is uncorrelated with the desired signal.

With this form of approach, it is now realistic to begin addressing the retrieval of electroencephalographic (EEG) signals, which are several orders of magnitude below the ECG in absolute signal value, from noisy environments for real-time evaluation.

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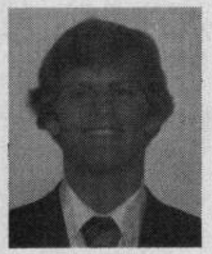

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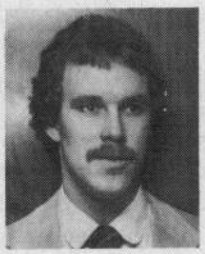

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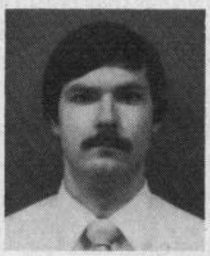

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