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Automated EEG Processing for Intraoperative Monitoring:

A Comparison of Techniques

Warren J. Levy, M.D., Harvey M. Shapiro, M.D.,† Gary Maruchak,‡ Edward Meathe, M.S.§*

THE USEFULNESS of the EEG as an intraoperative monitor has been debated almost since the report by Gibbs[1] of the occurrence of EEG changes during the administration of general anesthetics. EEG monitoring is presently recommended for cerebrovascular surgery,[2–5] cardiopulmonary bypass,[6–8] and deliberate hypotension,[9] and as a measure of anesthetic depth.[10–12] However, practical considerations such as the size and complexity of the equipment, the need for a technician to run it, and the difficulty of making intraoperative interpretations of the EEG have limited the popularity of the EEG as an intraoperative monitor. Because recent advances in electronic technology have shown promise in reducing the magnitude of some of these difficulties, there has been a resurgence of interest in the intraoperative EEG, particularly when the EEG is preprocessed by one of a variety of automated techniques. The purpose of this review is to discuss and evaluate some recent developments in automated EEG processing and to delineate their potential as practical intraoperative monitors.

The usefulness of the EEG in cerebrovascular surgery is well documented.[2,3,13,14] The temporal relationship between ipsilateral EEG slowing and occlusion of the carotid artery is almost incontrovertable evidence of cerebral ischemia,[2] and correlates well with changes in cerebral blood flow and with CNS changes in awake patients,[13] although it does not imply irreversible damage.[14] Although automated EEG processing techniques do not improve the diagnostic sensitivity of the EEG under these conditions, automated techniques possess simplicities of use and clarity of display which may make them advantageous.[15] Automated gain adjustment[16] and artifact rejection can reduce or eliminate the need for adjustment during anesthesia. The display of several minutes of the processed EEG in a concise form simplifies detection of the changes that may occur as a result of surgical occlusion of the carotid artery. By comparison, the preocclusion baseline and postocclusion ischemic changes may be separated by 20 or more pages of paper tracings when an unprocessed EEG is being recorded.

Evaluation of the EEG during cardiopulmonary bypass is considerably more complex. First, many of the events that routinely occur during extracorporeal perfusion—including deep hypothermia and hypocarbia—can cause EEG changes that mimic those of hypoxia.[17] Second, cerebral hypoxia during bypass may result from global events, *e.g.*, severe hypotension, or focal events, *e.g.*, air or particulate embolism, or a combination such as occurs when mild hypotension compromises cerebral perfusion distal to a critical stenosis. While global effects can be detected with only a single EEG lead, the identification of randomly distributed focal events (emboli, stenoses) requires an extensive EEG montage and the ability to monitor many leads simultaneously. In this situation, automated EEG processing techniques again might be advantageous because they present trends in the EEG succintly. Thus, they require less attention from the busy anesthesiologist than does a standard EEG. Whether automated techniques will be useful in evaluating complex situations—for instance, hypoxic changes in hypothermic patients—remains to be seen.

Monitoring depth of anesthesia using the EEG is fraught with problems. Early work in this area concentrated on the EEG changes resulting from large variations in anesthetic concentrations.[18–23] The wide-

* Assistant Professor of Anesthesia, University of Pennsylvania.
† Professor of Anesthesiology and Neurosurgery, University of California, San Diego.
‡ Chief of Monitoring Services, Johns Hopkins Hospital, Baltimore, Maryland.
§ Principle Development Engineer, University Hospital, San Diego, California.

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Address reprint requests to Dr. Levy.

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spread use of muscle relaxants and much lighter levels of anesthesia has largely changed this picture, and the usefulness of the EEG in identifying small changes in anesthetic depth is much less accepted.[18] Despite this, numerous reports describing relationships between the processed EEG and anesthetic depth have been published.[10–12,24,25] These suggest that information contained in the EEG may be sufficient to identify even small changes in anesthetic depth, but that these changes are too small to be identified by routine analytic techniques, *i.e.*, visual inspection. Only through automated analysis can these changes be identified and displayed for clinical use.

Excluding evoked potentials, which are beyond the scope of this review, at least six techniques of automated EEG analysis have been described in recent years. Of these, only the Cerebral Function Monitor® (CFM)¶ is commercially available, although both a spectral analysis system** and an aperiodic waveform analysis system†† are expected shortly. Unfortunately, only an abstract describing the aperiodic analysis system has been published,[26] preventing further evaluation of this technique. The remaining three techniques, multiple differential analysis,[27] period-amplitude analysis (PAA),[28] and augmented delta quotient analysis (ADQ),[29] all require special-purpose electronic equipment not generally available. Period-amplitude analysis has been used for manual quantification of the EEG for many years, and thus warrants further consideration as an automated technique. Because no further work on the multiple differential analysis technique has been reported and because of support by its developers for the less complex technique of period-amplitude analysis, no further discussion of the multiple differential technique is undertaken here. The newness of the ADQ and its lack of general availability also make critical evaluation of this technique difficult at present.

Cerebral Function Monitor (CFM)

By manipulating a single channel of the EEG in successive stages, this device produces an output voltage that varies with changes in the EEG and is supposed to represent cerebral function. (It should be noted that the concept of cerebral function is undefined except in reference to this device.) After the input EEG is filtered to remove frequencies below 2 Hz (largely baseline drift) and above 15 Hz, the signal is passed through another electronic filter to de-emphasize its lower frequencies. This frequency weighting is necessary because the low-frequency components of the EEG are quite resistant to pharmacologic and hypoxic depression, and therefore they are poor indicators of these conditions. The final step in the manipulation of the waveform is a standard electronic method for estimating the amplitude of a signal, and consists of rectification and averaging of the peak voltages. The display of this average peak voltage on a strip chart recorder results in a band of activity. "Cerebral function" is estimated as the distance from the bottom of this band to the baseline (fig. 1). Time compression for trend analysis is facilitated by the use of very low paper speeds (30 cm/h for the CFM, compared with 3 cm/sec for a standard EEG); however, this does hamper the detection of acute changes.

The CFM was originally designed to monitor EEG activity in intensive care units.[30] The low recording speed results in a reasonable amount of recorded output, making it easy to detect trends occurring over prolonged periods. The device is easy to operate, and the inclusion of a monitor of electrode impedance permits detection of electrode artifact. It is also capable of monitoring electrical activity over a very wide range of voltage without readjustment.

There is little doubt that the CFM recording shows dramatic changes when global ischemia occurs as a result of cardiovascular collapse.[31–32] Cucchiara *et al.*[3] have recently reported a comparison of the CFM and the 16-lead EEG in the identification of hypoxic changes during carotid endarterectomy. In two of 11 cases in which EEG changes occurred, the CFM failed to demonstrate significant variation. They also point out that these changes of ischemia can be interpreted only when the anesthetic level is constant, that changes in anesthetic depth can mimic changes of ischemia, and that the CFM changes of ischemia were observed only when the electrodes were placed on the same side as the carotid artery being occluded. Placement of the electrodes bilaterally (as recommended by the manufacturer) produced few changes in response to the focal hypoxia produced by carotid occlusion. Thus, as a routine monitoring technique for the detection of focal ischemia, the single-channel CFM appears less useful than a multichannel EEG.

The routine use of the CFM as a monitor of cerebral perfusion during cardiopulmonary bypass has been supported by a number of investigators.[31–34] Branthwaite[33] found a 61 per cent reduction in neurologic complications after instituting aggressive therapy for CFM evidence of hypoxia during extracorporeal circulation. Unfortunately, the control group was selected retrospectively, and variations in surgical technique and skill, patient population, and other

¶ Devices Limited, Welwyn Garden City, Hertforshire, England.
** OTE Biomedica, Florence, Italy.
†† Diatek, Inc., San Diego, California.

uncontrolled variables may have influenced these results. In addition, the control group had an unusually high incidence of major neurologic complications (19.2 per cent). No other controlled study has been reported to support or refute the value of the CFM for this use.

Several studies have been performed to evaluate the usefulness of the CFM as an indicator of anesthetic depth. Dubois *et al.*[12] described the changes in the CFM recording occurring in response to the use of an assortment of intravenous agents. All of those tested showed biphasic changes in the CFM recording, with increasing activity in the early stages progressing to depression of the cerebral activity at higher doses. A CFM recording at or near the preinduction level did not necessarily signify wakefulness, and awakening often occurred at levels of activity rather different from the baseline levels. Although these investigators did document a decrease in the dose of Althesin® necessary for a procedure when the CFM was used to control the infusion rate, many uncontrolled variables, including unconscious bias in administration of the agent, could produce the same results. Sechzer and Ospina[35] had poor results when attempting to correlate the CFM recording and anesthetic depth, and they concluded that such a correlation did not exist. Prior *et al.*,[10] on the other hand, reported good correlation between depth of Althesin anesthesia and a modified CFM in primates. The applicability of these data to man remains to be demonstrated. Thus, the value of the CFM in estimating depth of anesthesia appears to be limited.

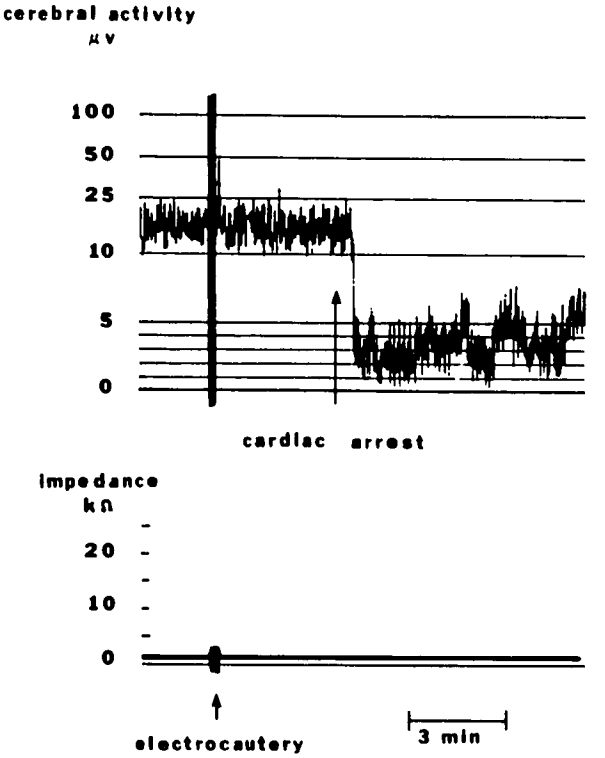


FIG. 1. Cerebral function monitor recording. This example of a recording from a cerebral function monitor demonstrates an abrupt reduction in cerebral activity occurring immediately after an intraoperative cardiac arrest. The low impedance (500 ohms) recorded during the abrupt change in cerebral activity indicates that the decrease in EEG activity is not due to a failure of the leads or monitor. Notice also that electrocautery destroys the tracing, but that changes in the recording of electrode impedance identify this as an artifact and not true cerebral activity. (Modified from a recording, courtesy of J. D. Levitt, M.D.).

Power-spectrum Analysis

The CFM represents one extreme of the spectrum of devices that process the EEG, because it compresses all of the frequency and amplitude information in the EEG into a single value. Power-spectrum analysis (fig. 2) represents the other extreme, because it retains almost all of the information present in the original EEG. The first step in the process of power-spectrum analysis consists of digitizing the EEG at frequent intervals for a period of time known as an epoch (usually 2–16 sec). Next, the epoch of data is manipulated using the complex mathematical technique known as Fourier analysis.[36] This separates the EEG into a number of component sine waves of differing amplitudes whose sum is the original waveform. Thus, a complex non-standard waveform is converted into a number of standard ones, and comparisons between complex waveforms are greatly simplified. Alternatively, Fourier analysis may be thought of as a process that makes explicit the information about frequencies which was implicit in the original EEG. The information in the original EEG is not lost, only its form has been changed.

Finally, the power spectrum is calculated by squaring the amplitudes of the individual frequency components. (The use of power as the measure of amplitude is a convention reflecting the origins of Fourier analysis in radio engineering). After it is calculated, the power spectrum for each epoch is displayed (usually in graphic form), and patterns are identified by examining a number of epochs in succession (see below). If the epochs are kept sufficiently short (2–4 sec), the process approaches a continuous monitoring technique, even though the mathematical analysis is a discontinuous process.

Despite its complexity, spectral analysis possesses several distinct advantages over simpler analytic methods for evaluating changes in the EEG. First, because the EEG data are only transformed, more information is retained and the identification of small changes in a complex EEG is simplified. Because each

frequency band is considered independently, changes in one part of the spectrum cannot balance out changes elsewhere, as they do in the CFM. Thus, for example, a CFM recording of an EEG composed of 6-, 8-, and 12-Hz waves might show no change when the 12 Hz waves disappeared, if there were a simultaneous increase in the 6- or 8-Hz activity. The spectral analysis of such an EEG would clearly demonstrate these changes as they occurred. In addition, the process of generating the power spectrum converts into a single point all of the low-frequency components that comprise baseline drift (0.05–0.5 Hz), thereby minimizing the interference of this drift in evaluation of the EEG.

Another advantage of spectral analysis is that changes in the spectrum may be predicted based on the present knowledge of EEG changes in a particular situation. For example, deepening halothane anesthesia is well known to shift the EEG to lower frequencies; as anesthesia is deepened, a spectral analysis of the EEG shows that less power is present at the higher frequencies, and more at the lower ones, just as predicted. Finally, computer technology has reached the point where the simultaneous, online analysis and separate display of four or more channels of EEG data can be easily performed, allowing the localization of EEG changes in a way that cannot be obtained with a single-channel device like the CFM.

One misconception about computerized EEG analysis is that it will necessitate computer training

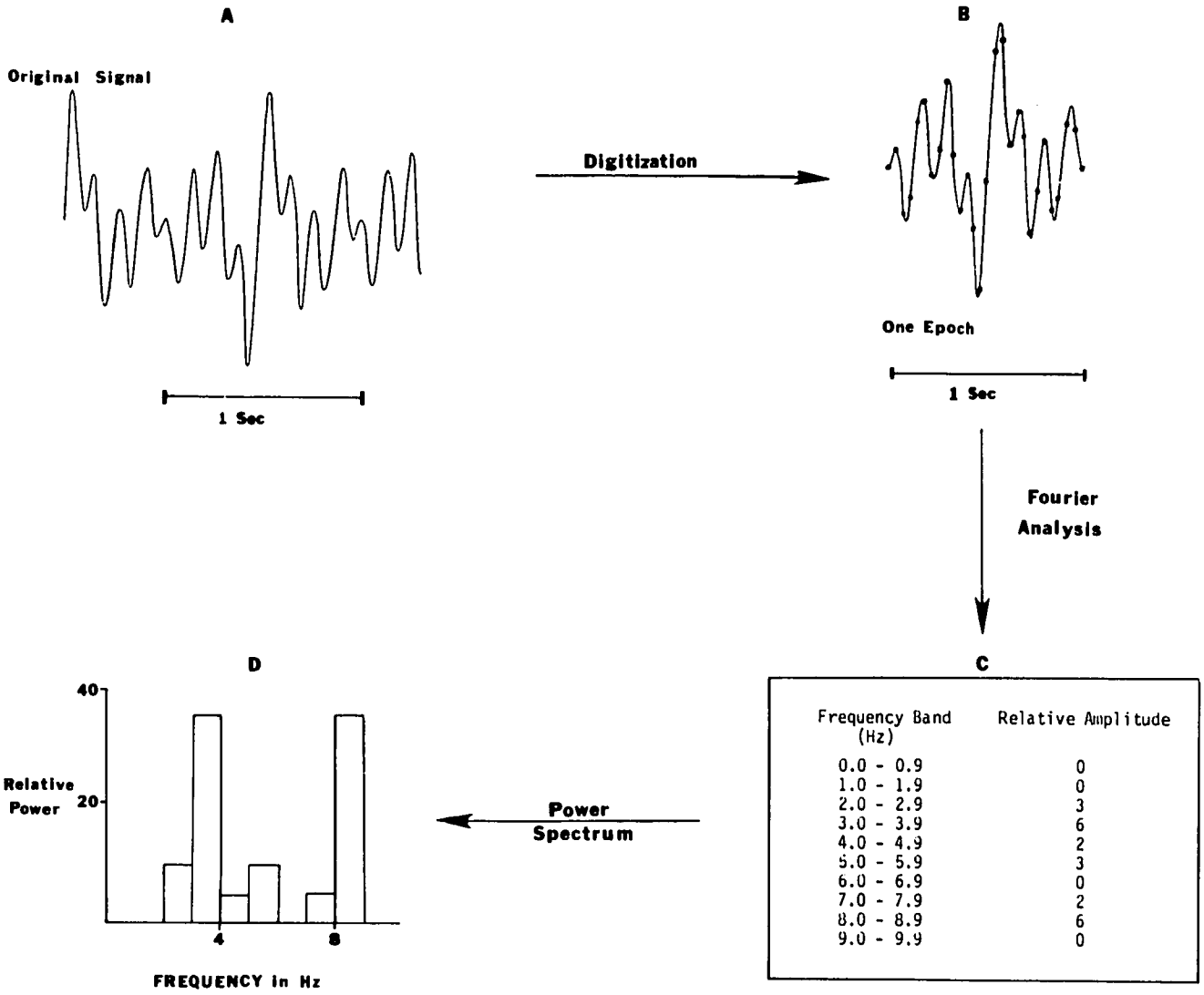


FIG. 2. A schematic representation of the process of power spectrum analysis. The original continuous waveform (*A*) is put in digital form by sampling it repeatedly at small intervals, here 1/28th of a second. One epoch of these data (*B*) is then passed to a computer program to perform a Fourier analysis. The result of this analysis, shown in *C*, is a table giving the amplitude of the activity in each frequency band. Finally, the amplitudes are squared (to give power) and plotted as a histogram (*D*).

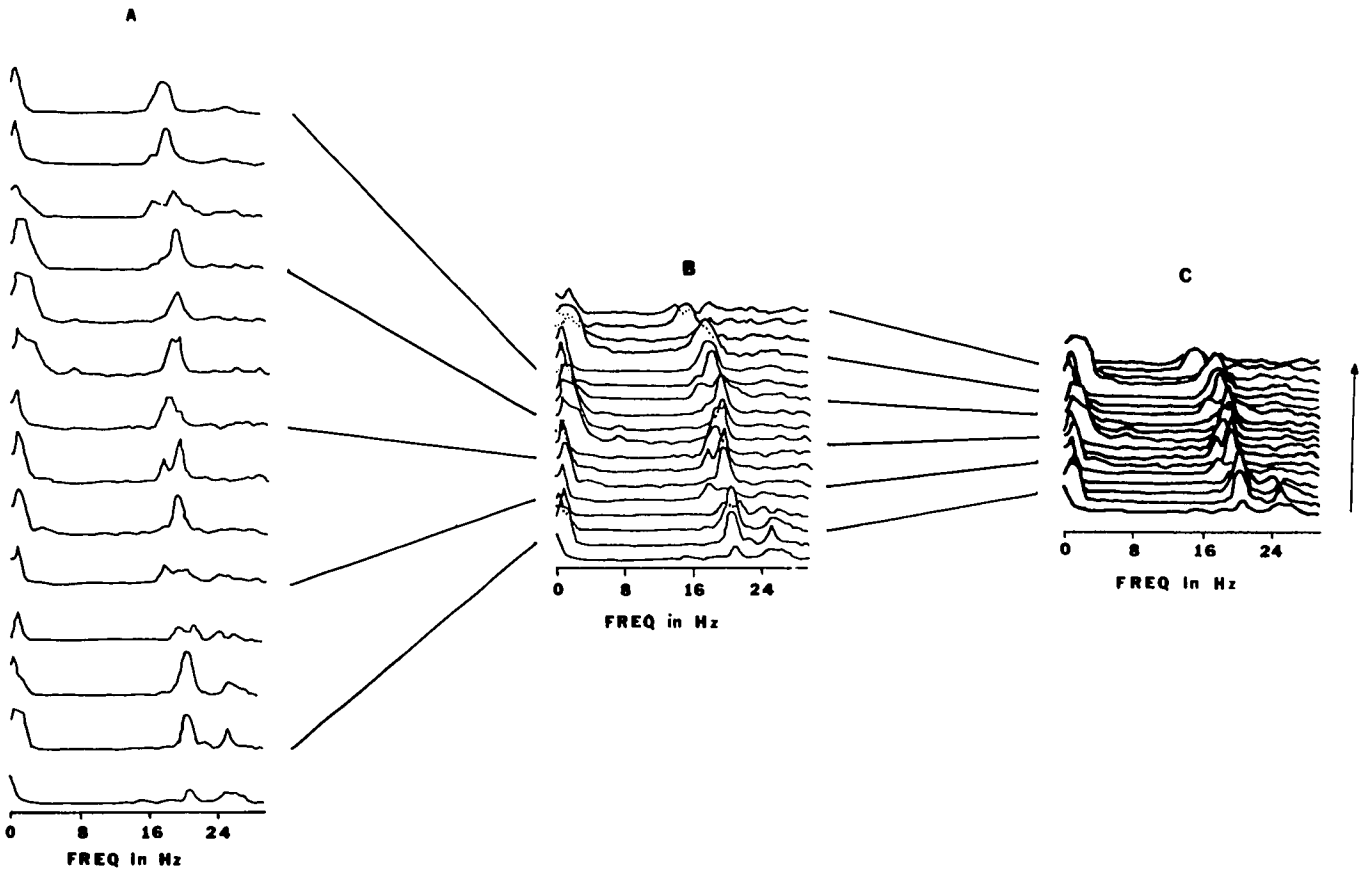


FIG. 3. Formation of the compressed spectral array (CSA). Individual epochs of analysis (*A*) are plotted closely above one another (*B*), and a "hill-and-valley" pattern begins to emerge. This is enhanced by the suppression of lines that would be behind a "hill." In the center panel these lines are dotted, while in the fully compressed picture (*C*), they have been omitted entirely. (This and all subsequent figures based on EEG recordings were made by the authors using the techniques described in the Appendix.)

for the anesthesiologist. The anesthesiologist needs to know no more about the working of a computer than he does about the electronic circuitry of the amplifiers and oscilloscopes that presently comprise the bulk of his monitoring systems. Fleming[16] and colleagues have developed a small multipurpose microcomputer that can perform spectral analysis and fits into the chassis of a standard multichannel strip-chart recorder. This device is constructed from components costing less than $1,000, demonstrating, in addition, that such a device need not be prohibitively expensive.

Perhaps the biggest disadvantage of spectral analysis is the amount of data generated. Evaluating only the component frequencies between 0 and 32 Hz and using 4-sec epochs, spectral analysis generates almost 2,000 data points each minute for each EEG channel processed. Since four or more channels may be processed simultaneously, this mass of data obviously cannot be comprehended in digital form, and sophisticated graphic display techniques are integral to the clinical applicability of spectral analysis.

Linear Display of the Spectral Analysis (CSA‡‡)

The first, and most commonly used, display for the spectral analysis is a linear display, developed by Bickford.[37] In this display, a graph is drawn of relative power versus frequency for each epoch of the analysis. Shifting the origin of the plot vertically with time produces a three-dimensional graph in which there appear to be hills at those frequencies making large contributions to the EEG, and valleys at frequencies that contain less power (fig. 3). To heighten this effect, those points that would lie behind a "hill" are not printed.

A number of authors have used this approach to evaluate intraoperative changes in the EEG. Stockard *et al.*[8,38] have successfully correlated the changes in the spectral analysis of the EEG with the occurrence of hypotension during cardiopulmonary bypass. A similar study by Myers and Stockard[5] demonstrated the applicability of these techniques to EEG monitoring

‡‡ CSA stands for Compressed Spectral Array, and is the term coined by Bickford for this type of display of the spectral analysis.

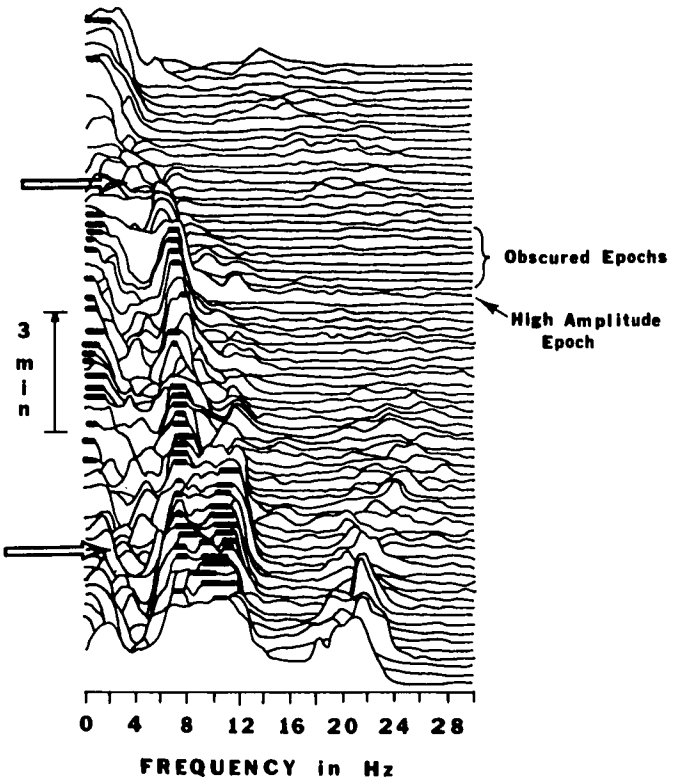


FIG. 4. Problems with the linear display of the spectral analysis (CSA).

1) Loss of data. An epoch of high-amplitude activity is shown obscuring part of the data in subsequent epochs for almost 3 min. Any changes in the 0–7-Hz range of the EEG during this time would not be detectable. In an effort to prevent an even greater loss of data when very-high-amplitude activity occurs, "hills" may be truncated, as indicated by the heavy horizontal lines. Unfortunately, this only changes the nature of the data lost, it does not prevent the loss.

2) Confusion of time and amplitude. Because time and amplitude of the EEG spectrum are displayed on the same axis, an identifiable baseline is needed to estimate the amplitude. The low-amplitude, high-frequency segment of the epoch often serves as such a baseline; however, in areas such as those identified by the hollow arrows, it may be difficult or impossible to identify the baseline and estimate EEG activity.

during carotid endarterectomy, and Myers[39] has used the technique to demonstrate cyclic fluctuations in EEG activity following glutethimide overdose. Findeiss *et al.*[24] and Bart *et al.*[25] have used spectral analysis to characterize the EEG changes produced by an assortment of anesthetic agents, including cyclopropane, fluroxene, methoxyflurane, enflurane, and halothane.

Berezowskyj *et al.*[11] carried the idea of using spectral analysis to evaluate depth of anesthesia a step further. Their study compared the EEG spectrum with a clinical estimate of anesthetic depth based on blood pressure, heart rate, and movement. Statistical methods were used to correlate specific spectral patterns with anesthetic depth. Once these patterns were identified, a computer program was capable of identifying the depth of anesthesia correctly based on spectral analysis of the EEG in 65 per cent of cases. In addition, Smith *et al.*[40] have demonstrated relationships between certain portions of the EEG spectrum and the anesthetizing concentrations of halothane or enflurane. Taken together, the results of these studies suggest that spectral analysis of the EEG has real potential as a technique to estimate depth of anesthesia.

Despite the value of the linear display of the spectral analysis, there are problems with this type of display (fig. 4). High-amplitude activity tends to obscure subsequent low-amplitude activity at the same frequency. In an effort to limit the amount of data hidden behind such large "hills," they are truncated at some arbitrary value. Although only a small amount of information is lost, legibility is impaired considerably, thus increasing the likelihood of misreading or misinterpreting the display. Another problem with this type of display is that both time and power are indicated by vertical displacements. Since they are on the same axis, it may be difficult to determine the precise time a change in power occurred. In addition, this type of display requires a two-dimensional (X-Y) plotter, a complex device not normally found in intraoperative monitoring equipment.

Grey-scale Display of the Spectral Analysis (DSA§§)

Another technique for displaying the power spectrum of the EEG has been developed by Fleming and Smith.[41] Known as "density modulation," it allows display of the spectral analysis without loss of data. With this technique, each epoch of the spectral analysis is displayed as a line of varying density or a series of dots of various sizes. (A display using dots is also known as a dot-matrix display.) The areas of maximum intensity (largest dots) correspond to those frequencies making the largest contributions to the EEG spectrum, while the areas with little contribution are light grey or have only scattered dots. When compared with the CSA display, differences are minor (fig. 5); however, shifts in frequency spectra are observed somewhat earlier and more clearly on the DSA. Another advantage of the DSA is that it remains legible even when read from a distance. The use of colors to emphasize particular frequencies or levels of activity is also possible, although not used at present. Finally, for optimal legibility of the CSA, time must be placed on the vertical axis, which makes it difficult to incorporate the CSA into the anesthetic record (in which time is normally placed on the horizontal axis). Because

§§ Smith refers to this display as the Density-modulated Spectral Array, hence the term DSA.

the DSA is equally legible with time on either axis, its inclusion in the anesthetic record is straightforward. (In figure 5, the DSA is displayed according to CSA convention, *i.e.*, time is on the Y-axis. In figures 8–10, the need to compare the various techniques necessitated the rotation of the CSA and DSA so that time is on the X-axis). Since the analytic technique is that of spectral analysis and only the method of display of the data is different, the observations concerning spectral analysis and display in the form of the CSA should still be valid when the display is in the form of the DSA. The improvements in legibility and ease of incorporation into the anesthetic record are likely to make this the preferred display technique.

Period Analysis

One of the earliest methods for quantitating the EEG was period analysis.[42,43] This could be done manually by counting the number of times the EEG crossed the isoelectric (zero-voltage) line in a given time period, giving an estimate of the mean frequency of the EEG known as the zero-cross frequency (ZXF). Although the technique may be performed by extremely simple, inexpensive, and reliable devices, some theoretical problems are presented by its use.

One problem with the ZXF is that it is excessively sensitive to minor changes in the EEG. Figure 6 shows an EEG in which very slow activity of high amplitude predominates. Superimposed on this are higher-frequency components. Notice that in the middle portion of the tracing, a small decrease in the *amplitude* of the low-frequency component has resulted in a very large increase in the zero-cross frequency, even though the underlying rhythms have not changed. This variability can be eliminated by averaging the results of the period analysis over a long interval; however, such averaging delays the response to abrupt changes in the EEG.

Another problem of the zero-cross frequency as a measure of EEG activity is that the ZXF that is generated by an EEG is not uniquely related to that EEG–several different EEG waveforms may give the same zero-cross frequency. Figure 7 shows three waveforms of equal zero-cross frequencies but very different shapes. Since they have the same zero-cross frequency, the implication is that EEGs such as these would have similar electrophysiologic significances. Clearly, a change from one of these waveforms to another would represent a change in the EEG, and this analytic technique lacks the sensitivity to identify it.

In spite of these theoretical objections, Davis *et al.*[28] have suggested that the combination of the zero-cross frequency and a measure of the average EEG amplitude, the mean rectified voltage¶¶ (MRV), be used for the evaluation of anesthetic depth. Such an analysis is known as a period-amplitude analysis (PAA) because it is composed of separate measures of the period (ZXF) and amplitude (MRV). Such a technique might prove more powerful than the CFM and less complex than spectral analysis. Unfortunately, no other investigators have reported data obtained using this technique of EEG analysis.

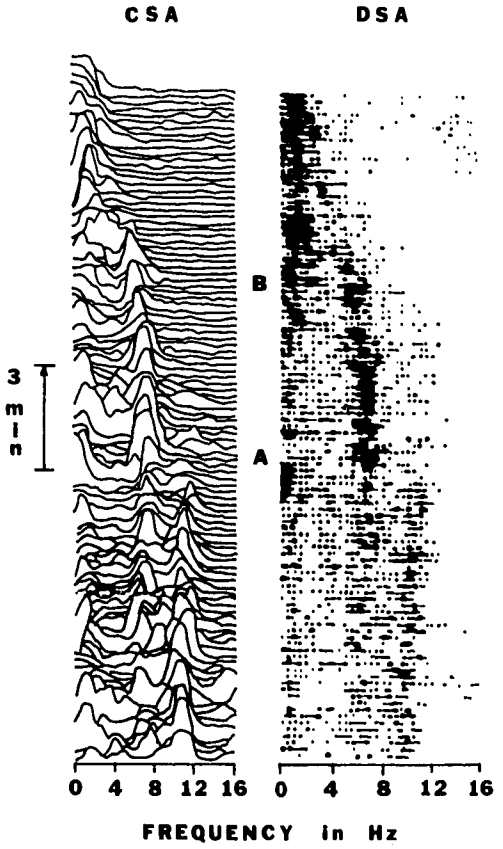


FIG. 5. Comparison of displays for spectral analysis. A comparison of the linear (CSA) and density (DSA) displays of the spectral analysis in a patient during gradually deepening halothane–N_2O anesthesia. The DSA is displayed in dot-matrix form, *i.e.*, large dots signify areas of high activity, small ones, low activity. Changes in components in the 8–12-Hz and 4–8-Hz ranges are present at *A* and *B* and are more easily distinguished on the DSA than on the CSA.

Comparative Analysis

No previous reports have been published showing side-by-side comparisons of any of these techniques of automated EEG analysis. Thus, rational com-

¶¶ The mean rectified voltage is a standard engineering method for estimating the amplitude of a waveform. For purposes of quantitating the EEG, it is no better or worse as a measure of the average amplitude than is the peak rectified voltage that is used in the CFM.

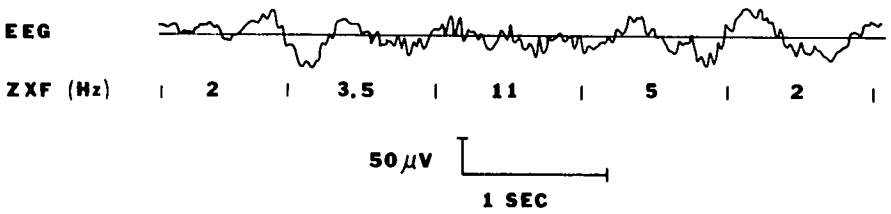


FIG. 6. Variability of the zero-cross frequency. A standard EEG recording is superimposed on the baseline used to determine the zero-cross frequency (ZXF), and the ZXF for each 1-sec period is indicated below the tracing. A basic pattern of slow waves (1–2 Hz) of moderate amplitude, combined with lower-amplitude high-frequency activity (about 16 Hz), is evident throughout. A small decrease in the amplitude of the low-frequency component in the middle of the tracing causes more than a fivefold variation in the ZXF during this 5-sec recording.

parisons among techniques are difficult, and the preference for one technique over another is often based on emotional or other nonscientific considerations. We therefore compared the three types of automated EEG analysis that have been reviewed—the CFM, period-amplitude analysis, and spectral analysis. To do this, a computer program capable of performing all of these analyses was written and used to analyze EEGs recorded intraoperatively during cardiac surgery, carotid endarterectomy, and craniotomy. (See appendix for details of equipment and recording techniques).

It should be noted that some minor differences existed between the simulated analyses and those performed by the special-purpose analog devices being modeled. Specifically, in order to simulate the filtering characteristics of the CFM, it was necessary first to reduce the EEG to its components, *i.e.*, to perform a spectral analysis. The electronic filters in the CFM could then be simulated by decreasing the amplitudes of the frequency components to the same extent they would have been decreased by the CFM; however, phase changes caused by the CFM circuitry could not be simulated. Because the spectral analysis is performed on an epoch of data (4 sec), only one such point could be simulated every 4 sec. The value obtained is the average of the values that would be plotted by the CFM during the 4-sec epoch in which sampling occurred. Its filtering is identical to that of the CFM; thus, the simulated CFM tracing varies in a fashion identical to the CFM; however, since fewer points are plotted, and they are spread farther apart, the graph looks somewhat different from the one generated by the CFM. The term "integrated microvoltage" or "IMV" has been used previously to describe measures of the EEG similar to this,[5] and is used in subsequent discussions here in order to avoid confusion between the simulated analysis and that performed by the Cerebral Function Monitor.

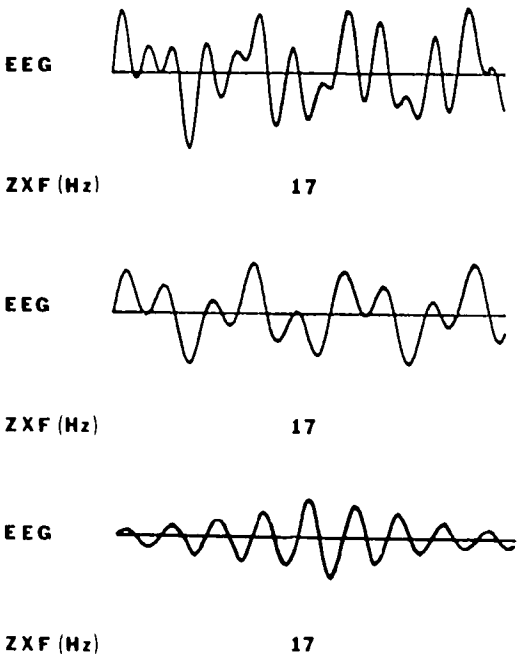
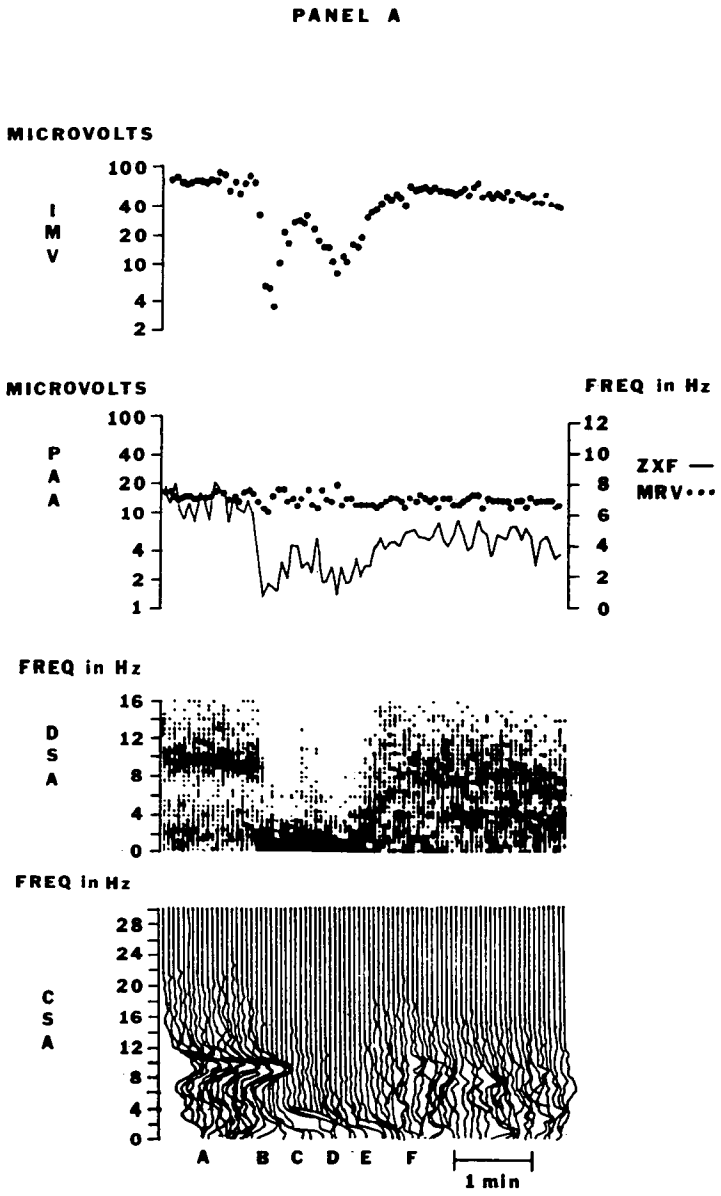


FIG. 7. Nonspecificity of the zero-cross frequency. Three possible EEGs of very different waveforms, but the same zero-cross frequency (ZXF), are shown. Although such different EEGs might imply three different physiologic states, they would be indistinguishable on the basis of the zero-cross frequency. This lack of specificity of the ZXF for the original EEG could lead to misinterpretation of the electrophysiologic state of the patient.

A minor change was also made in the determination of the zero-cross frequency. Since a sine wave changes polarity twice in each cycle, the zero-cross frequency of a wave is always twice the true frequency. In order to provide for simpler comparison between all analyses, zero-cross frequencies were divided by two, thus making them numerically comparable to the values generated by other forms of analysis.

Figure 8 compares the IMV, PAA, spectral analysis (both DSA and CSA displays), and representative portions of the unprocessed EEG for a patient at the start of cardiopulmonary bypass. Notice that an abrupt reduction in EEG activity occurred at point *B*, and was reflected by frequency shifts in both displays of the spectral analysis, by a decline in the zero-cross frequency and a decrease in the IMV. Because the amplitude of the EEG was largely unchanged, the



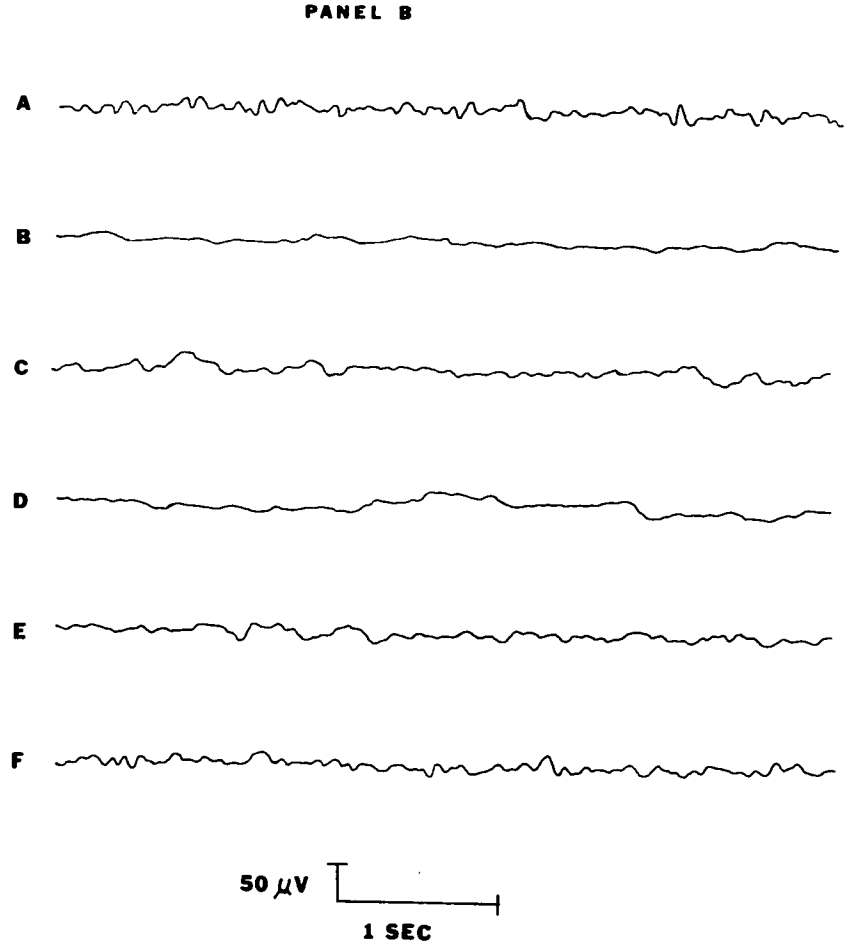


FIG. 8. Comparison of automated analysis techniques: global hypotension. *Panel A* shows the integrated microvoltage (IMV), period-amplitude analysis (PAA), and density-modulated (DSA) and linear (CSA) displays of the spectral analysis for a patient subjected to a brief period of severe hypotension (mean arterial pressure below 50 torr) during initiation of cardiopulmonary bypass. Samples of the unprocessed EEG were obtained at points *A–F* and are shown in *Panel B*. The changes evident in the unprocessed EEG are also evident in each of the analyses displayed. Notice, however, that in the period-amplitude analysis only the zero-cross frequency (ZXF) shows a change; the mean rectified voltage (MRV) is essentially constant.

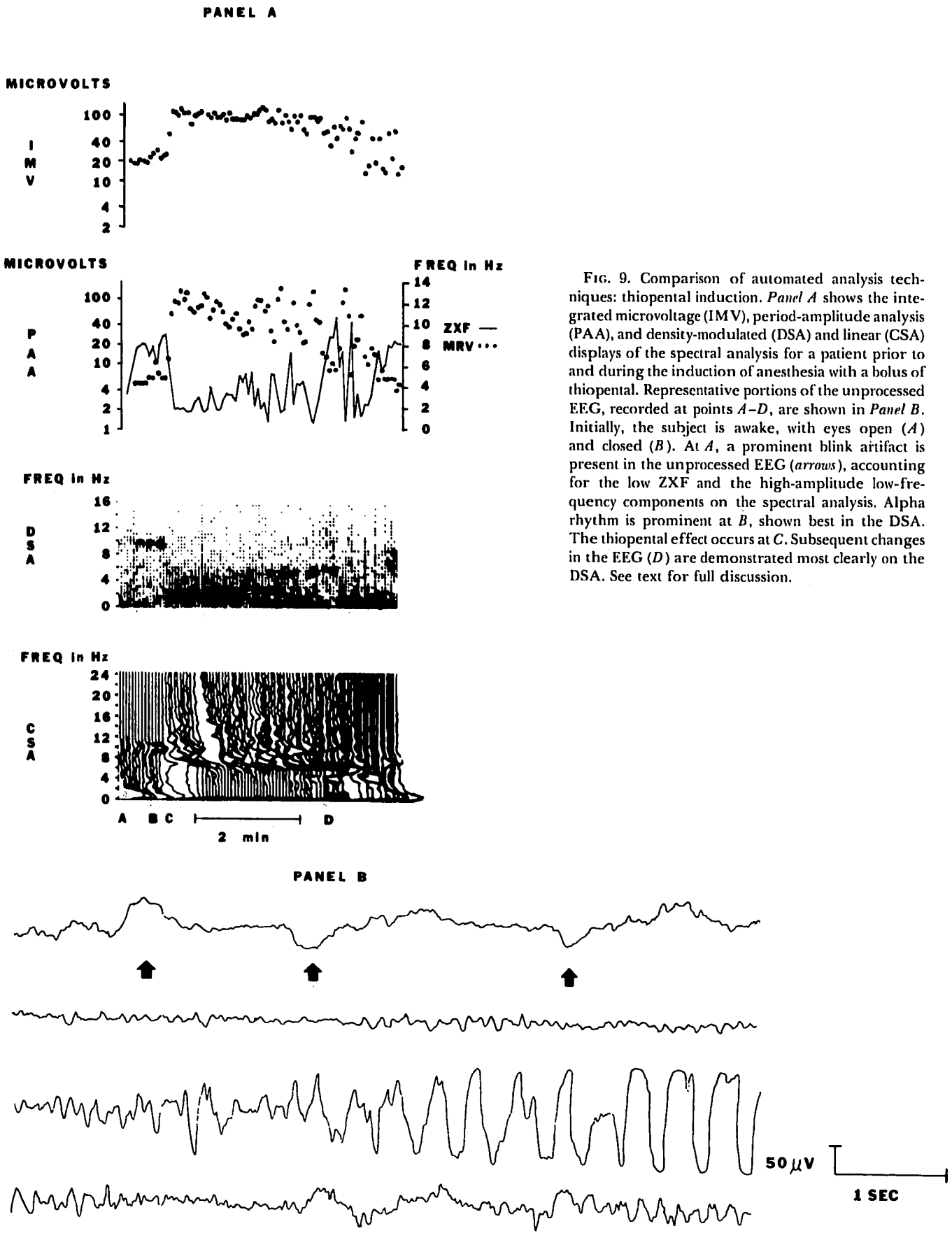


FIG. 9. Comparison of automated analysis techniques: thiopental induction. *Panel A* shows the integrated microvoltage (IMV), period-amplitude analysis (PAA), and density-modulated (DSA) and linear (CSA) displays of the spectral analysis for a patient prior to and during the induction of anesthesia with a bolus of thiopental. Representative portions of the unprocessed EEG, recorded at points *A–D*, are shown in *Panel B*. Initially, the subject is awake, with eyes open (*A*) and closed (*B*). At *A*, a prominent blink artifact is present in the unprocessed EEG (*arrows*), accounting for the low ZXF and the high-amplitude low-frequency components on the spectral analysis. Alpha rhythm is prominent at *B*, shown best in the DSA. The thiopental effect occurs at *C*. Subsequent changes in the EEG (*D*) are demonstrated most clearly on the DSA. See text for full discussion.

MRV was essentially constant. This EEG suppression was caused by a brief period of hypotension at the beginning of cardiopulmonary bypass, and the return of electrical activity coincided with the return of adequate perfusion pressures. It is also noteworthy that these changes were of brief duration and might have been missed entirely if a permanent record of these events had not been made.

Figure 9 compares the various analyses and display systems during the preinduction period and during induction of anesthesia with thiopental (4 mg/kg). At the points labeled *A* and *B* the patient is awake, with eyes open and closed, respectively. Blink artifact (*arrows*) is prominent in the raw EEG, and thus gives an opportunity to observe the performance of each of these techniques under relatively poor recording conditions and while a simple, well-defined, and fairly abrupt change occurs in the EEG. The IMV shows no change whether the signal is mostly artifact (*A*) or EEG (*B*). The period-amplitude analysis shows that there is low-frequency activity first (the blink artifact), which then changes to higher-frequency activity with eye closure. Unfortunately for PAA, any high-frequency information that might have been present at *A* would have been overwhelmed by the artifact. Spectral analysis is more successful at separating noise from signal, showing both a large low-frequency component (noise) and scattered high-frequency activity, as is seen on the unprocessed EEG at *A*. The DSA is easier to read and displays the alpha rhythm at *B* the most clearly of the techniques shown. At point *C*, the EEG effects of the bolus of thiopental occur. As seen on the unprocessed EEG, very-high-amplitude, low-frequency activity predominates. The IMV shows an increase in cerebral activity due to the very high amplitude of the low-frequency components. However, it is important to note that from the IMV alone (without the EEG) it is impossible to tell whether this change occurred due to a shift in amplitude or a change in frequency, or both. The period-amplitude analysis is very successful at defining the EEG effect of the barbiturate, showing that low-frequency (2 Hz), high-amplitude (50–100 microvolt) activity is present. Spectral analysis (both CSA and DSA displays) also shows this well, with the CSA suffering from illegibility due to truncation of the very-high-amplitude "peaks" in the 0–4-Hz band. At point *D* the EEG has reverted to a mixed pattern, with many different frequency components present. The increasing scatter of the IMV suggests that something very erratic is occurring in the EEG, as does the scatter in the zero-cross frequency and mean rectified voltage of the period-amplitude analysis. The spectral analysis separates out the various frequency components, displaying them nicely in the DSA, and emphasizing the presence of a band of activity in the 5-Hz range. In the CSA display, all of this information is hidden behind the very high peaks left over from the earlier low-frequency activity. This example demonstrates the superiority of spectral analysis in dealing with an artifact-laden signal, as well as in emphasizing changes that occur in the EEG. It also demonstrates the improvement in legibility that results when the spectral analysis is displayed in a density-modulated form rather than in a linear form.

Figure 10 depicts the EEG of a patient who was being cooled to 26 C while on cardiopulmonary bypass. Notice that between points *B* and *C*, significant amounts of EEG data are present at frequencies above 16 Hz. Both the period-amplitude analysis and spectral analysis (CSA) clearly reflect the changes seen in the raw EEG. Because only frequencies below 16 Hz are included in the CFM (and also in the IMV), the IMV shows increasing EEG activity at a time when temperature-related EEG suppression is actually occurring. Since only the spectrum below 16 Hz has been displayed on the DSA, it too gives an inaccurate impression of the cerebral electrical activity, and clearly demonstrates the need for the inclusion of frequencies above 16 Hz in any automated processing and display technique.

Conclusions

The Cerebral Function Monitor is the simplest, most readily available automated EEG processor for intraoperative use. It appears to be less sensitive than the unprocessed, multilead EEG for the determination of focal ischemia, although it does successfully demonstrate severe, global cerebral ischemia due to hypoxia or hypotension. It may be of some value in estimating the depth of anesthesia, but the evidence for this is inconclusive, and on occasion, the shift of EEG activity into or out of its limited bandwidth may result in misrepresentation of the changes in cerebral activity that are actually occurring. It is extremely insensitive to changes of small or moderate magnitude in the EEG, and thus appears useful only for the determination of gross anesthetic overdose or global cerebral hypoperfusion of a magnitude that is likely to be recognizable by other means.

Period-amplitude analysis may have potential as an intraoperative monitoring technique; however, it is not readily available at present, and there are theoretical arguments against it. In side-by-side comparisons it appears to be of greater value than the CFM, and its simplicity and potential to be constructed inexpensively may make further study of this technique worthwhile.

Spectral analysis has documented value as a monitor of cerebral ischemia and possible value in the deter-

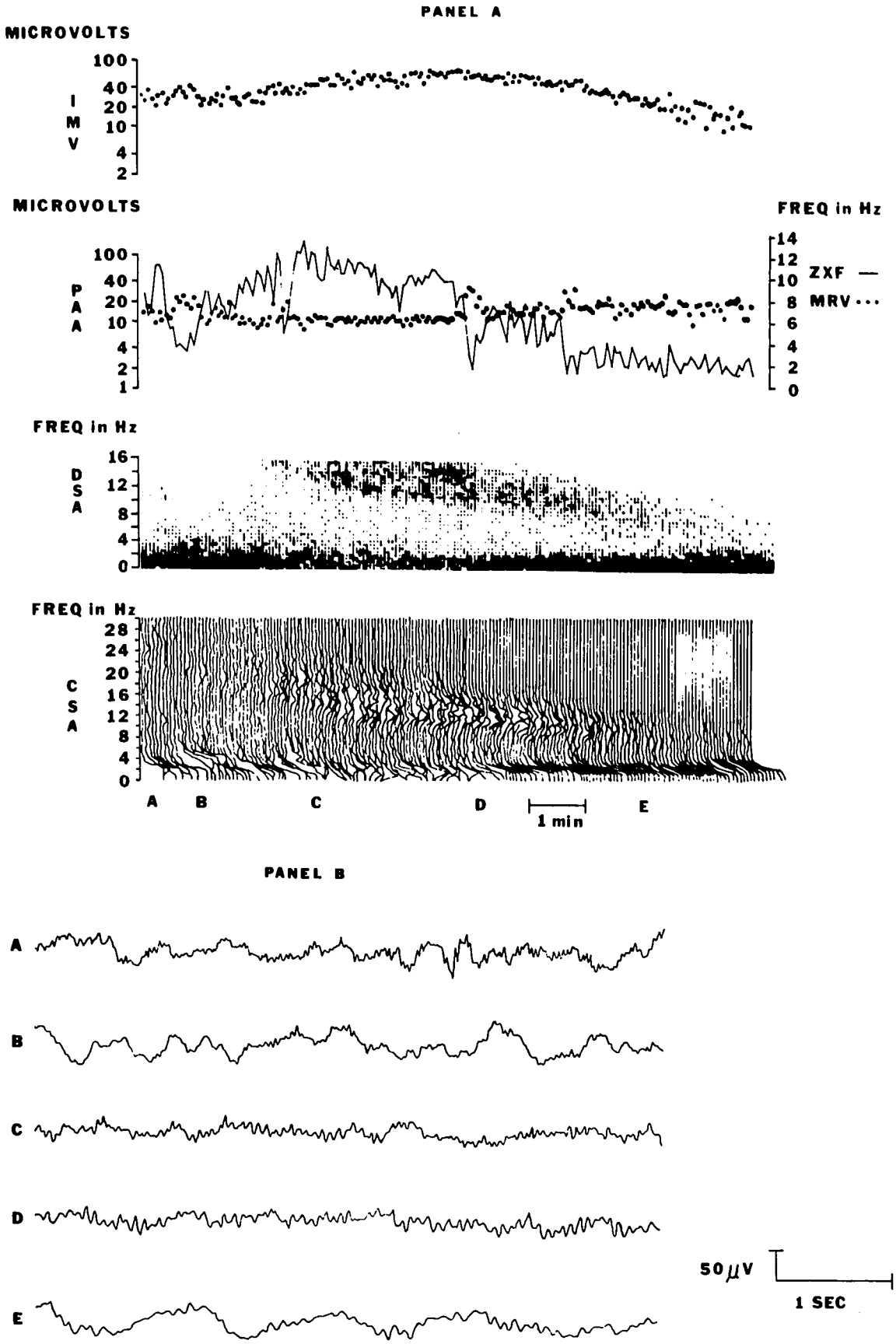


FIG. 10. Comparison of automated analysis techniques: deep hypothermia. *Panel A* shows the integrated microvoltage (IMV), period-amplitude analysis (PAA), and density-modulated (DSA) and linear (CSA) displays of the spectral analysis for a patient being cooled while on cardiopulmonary bypass. Representative portions of the unprocessed EEG, recorded at points *A–E*, are shown in *Panel B*. Nasopharyngeal temperature is 36 C at *A* when cooling begins and 24 C at *E*. The extensive activity above 16 Hz is demonstrated most clearly on the CSA. (The DSA has arbitrarily been limited to 16 Hz.) The IMV shows an increasing cerebral function level between *C* and *D*, even though temperature-related suppression of high-frequency EEG activity is clearly occurring.

mination of anesthetic depth. It gives the information available in the unprocessed EEG and succeeds in doing so even in the presence of considerable artifact. Its disadvantages include its relative expense and the complexities of its displays; however, advances in computer technology are likely to reduce the cost of this technique considerably, and newer display systems such as the density-modulated display (DSA) are a definite improvement over the earlier linear display (CSA). Future developments in spectral analysis and display systems are likely to make this technique the most useful and practical method for the intraoperative evaluation of the EEG.

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APPENDIX

Continuous four-channel EEG recordings were performed as part of the routine monitoring of selected patients undergoing cardiopulmonary bypass, carotid endarterectomy, and transsphenoidal hypophysectomy. A Beckman Accutrace® eight-channel EEG machine was used. It had a sensitivity of 7.5 microvolts/mm and a bandwidth from 1 to 70 Hz (3 db points). Lead configurations varied depending on the surgical procedure. Arterial and central venous pressures were measured from indwelling catheters utilizing Bell and Howell transducers and Hewlett-Packard amplifiers, oscilloscope displays, and recorders. These data were simultaneously recorded on a Vetter® eight-channel FM tape recorder. Analog-to-digital conversion and analysis of the data were performed by a PDP® 11/40 computer, with graphic displays provided by a Tektronix® video terminal. All computer programs were written in FORTRAN except the Fast Fourier Transform and the graphics package, which were programmed in assembly language.

Erratum

An error appeared in the article, "The Neonatal Neurobehavioral Effects of Bupivacaine, Mepivacaine, and 2-Chloroprocaine Used for Pudendal Block" (ANESTHESIOLOGY 52: 309–312, 1980). In the abstract at the beginning of the paper, the bupivacaine concentration in the neonate at 4 hours of age should be 0.015 μg/ml rather than a 0.15 μg/ml. The statement, "Bupivacaine gave higher neonatal capillary blood levels (0.15 μg/ml at 4 hours of age) than previously reported, but the drug still produced no detectable neonatal neurobehavioral effects," should be *deleted*. In place of that statement should be "Bupivacaine levels in the neonate were also low (0.015 μg/ml) at 4 hours of age."