PEAK-DETECTION ALGORITHM FOR EEG ANALYSIS

RONALD E. BARR

Mechanical Engineering, University of Texas, Austin, Texas 78712 (USA)

JAMES J. ACKMANN

Marquette University, Milwaukee, Wisconsin (USA)

and

JOSEPH SONNENFELD

Texas A&M University, College Station, Texas (USA)

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SUMMARY

A peak-detection method is described for computer analysis of the electro-encephalogramme (EEG). The technique consists of measuring the amplitude and time interval between successive maxima (peaks) and minima (troughs) in the signal. A critical feature of the peak-detection algorithm is the inclusion of an amplitude threshold criterion which eliminates the registration of low-voltage activity riding on EEG waves. The peak-detection procedure permits the formulation of a variety of intra-band and inter-band EEG statistics which can be useful in on-line computer applications. The peak-detection algorithm has been successfully applied to a number of normal and clinical EEG recordings. Although no computer procedure for EEG analysis has yet been universally adopted, the peak-detection algorithm reported in this paper presents a standardised approach which can be used between EEG clinics.

SOMMAIRE

On décrit une méthode de détection des pics en vue de l'analyse d'électro-encéphalogrammes (EEG) par ordinateur. La technique consiste en la mesure de l'amplitude des maxima (pics) et des minima (creux) successifs du signal ainsi qu'en la détermination des intervalles temporels qui les séparent. Un trait caractéristique de

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l'algorithme de détection des pics est l'usage d'un critère de seuil d'amplitude qui permet d'éliminer l'enregistrement des ondes électro-encéphalographiques de bas voltage. Ce procédé de détection des pics permet d'envisager des statistiques variées, concernant un seul EEG ou des EEG différents, qui peuvent être utiles quand on emploie un ordinateur. L'algorithme de détection des pics a été utilisé avec succès dans de nombreux enregistrements d'EEG normaux ou cliniques. Bien qu'aucune méthode d'analyse d'EEG n'ait encore été universellement adoptée, l'algorithme de détection des pics, décrit dans cet article, offre une méthode d'approche normalisée qui peut être utilisée dans les cliniques électro-encéphalographiques.

INTRODUCTION

Current trends in electroencephalography are toward methods for automatic analysis (Kellaway and Petersen, 1973; Gevins *et al.*, 1975). Two well-known computational procedures that have been applied to the electroencephalogramme (EEG) are the Fast Fourier Transform or FFT (Dumermuth and Fluhler, 1967) and period analysis (Saltzberg and Burch, 1971). The FFT is a standard procedure that expresses the power of a signal as a function of frequency. A major feature of the FFT is the ability to conveniently display data in compressed spectral arrays (Bickford *et al.*, 1971; Huber, 1975). Period analysis or zero-crossing has attained popularity in EEG work due to programming efficiency on small laboratory computers (Legewie and Probst, 1969; Carrie and Frost, 1971; Cohen, 1976).

A third procedure that has received less attention in the EEG field is peak-detection. Leader *et al.* (1967) devised a pattern reading method for analysing the EEG based on the time and amplitude between absolute maxima and minima. Stigsby *et al.* (1973) combined both baseline crossings and peak–valley detection for automatic EEG acquisition and analysis. In addition, our group has successfully used peak-detection to determine clinical EEG trends in trauma patients (Ackmann *et al.*, 1975) and during exchange transfusion in Reye's Syndrome (Barr *et al.*, 1977).

Although no standard algorithm for computer EEG analysis has yet been universally adopted, peak-detection does afford several advantages that can be outlined as follows.

(1) Peak-detection permits efficient measurement of both the amplitude and the time interval using the same two detection points. This is an advantageous feature in on-line, real-time situations.
(2) The algorithm is sufficiently immune to baseline drift, an inherent problem in zero-crossing, and permits registration of activity above and below the baseline.
(3) Peak-detection is a measurement technique that heuristically concurs with standard manual EEG interpretation (Sulg, 1969; Fox, 1973).
(4) Since peak-detection is essentially a wave-by-wave analysis, the procedure allows for harmonious inclusion of waveform detectors for spikes, triphasic waves and other transients.

METHODOLOGY

Mathematical formulation

Peak-detection is a derivative form of period analysis in which the detection points of interest are the relative signal extrema (peaks and troughs). The two fundamental measurements consist of the amplitude and time interval between a consecutive peak–trough combination, called a count. These measurements are illustrated in Fig. 1.

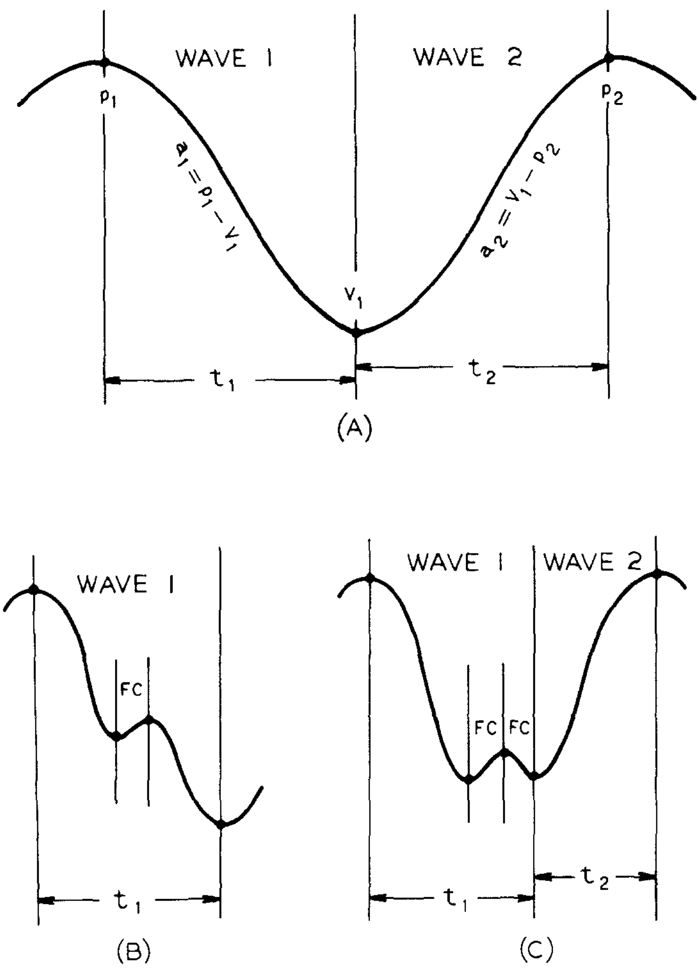

Fig. 1. Wave formulation for various configurations in peak-detection analysis. (A) The fundamental measurements for each count are the time interval t and the amplitude a. (B) When a false count FC or an odd sequence occurs, the two surrounding good counts are incorporated into the same EEG wave. (C) When an even sequence of false counts occurs, the follow-up good count starts a new wave.

If each count is assumed to approximate a sinusoidal shape, then the frequency can be calculated by the inverse relationship:

$$f = \frac{1}{2t} \tag{1}$$

It should be noted that in order to recover the frequency, the time interval must be doubled, since each count is actually a half-wave. This frequency conversion is needed since EEG activity is commonly classified into standard frequency bands, as indicated in Table 1.

TABLE 1
EEG FREQUENCY AND HALF-PERIOD CLASSIFICATION CRITERIA USED IN PEAK-DETECTION ANALYSIS

EEG band	*Frequency range (Hz)*	*Half-period (msec)*
ARTLO	0–0·5	greater than 1000
DELTA	0·5–4·0	1000–125
THETA	4·0–8·0	124–63
ALPHA	8·0–13·0	62–38
BETA	13·0–35·0	37–14
ARTHI	greater than 35	less than 14
ARTPK	large amplitudes exceeding A/D range	

The amplitude is determined by measuring the voltage difference between the relative extrema for each count. The convention adopted in this application is for the amplitude to be positive when going from peak to trough and negative when going from trough to peak. This convention is illustrated in Fig. 1(A) and can be mathematically expressed as:

$$a_1 = p_1 - v_1 \tag{2}$$

$$a_2 = v_1 - p_2 \tag{3}$$

Software development

Development of the peak-detection software routine can be divided into three phases: (1) peak–trough detection, (2) filtering and (3) data storage. A prior consideration for peak-detection is the analogue-to-digital conversion of the EEG signal. For this particular application, the sampling rate was 300 samples per second. During conversion a scale factor must be established to accurately relate digital data to actual EEG microvolts.

The digital data stream is examined for extrema by continuously comparing each sample point with surrounding sample values. A peak is defined to exist if a middle sample point is greater than the two adjacent values. Once a peak is detected, the procedure is repeated to find a trough, with the exception that the middle sample value must now be less than the two surrounding values.

As each peak–trough combination is registered, the time and amplitude are measured. The time interval is determined by counting the number of samples passed between the peak and the trough and then multiplying by the sample interval (3·333 milliseconds in this case). The amplitude of the count is measured in accordance with eqns. (2) and (3). The peak–trough is then immediately subjected to a filtering subroutine. This filtering process classifies the count into one of three possible categories: (1) sub-threshold activity, (2) acceptable EEG activity or (3) artifact. A flow chart of the filter subroutine is shown in Fig. 2.

The initial filtering consideration is to determine if the amplitude fulfills a pre-established threshold level. For this particular application, a threshold criterion of 3 microvolts (μV) is imposed. If the threshold level is not met, this false count is ignored and the mainline program continues searching for the next extremum. An important problem associated with sub-threshold activity is to determine when the formulation of the existing wave should be terminated and a new wave started. To resolve this, an even–odd condition is employed. If the number of contiguous false counts sandwiched between two good counts is odd, then all data are assigned to the current wave. Conversely, if zero or an even sequence of sub-threshold counts exists between two good counts, the latter good count starts a new wave. This even–odd principle of wave formulation is illustrated in the various configurations of Figs. 1(B) and 1(C).

Once the formulation of a new wave is initiated, the filtering process is performed on the preceding wave. Each wave is classified into standard EEG frequency bands or into one of several artifact categories. Frequency classification is based on the inverse relationship with the half-period measurement, as conveyed in eqn. (1). Standard terminology (Chatrian *et al.*, 1974) for EEG frequency and half-period ranges has been adopted in the peak-detection as shown in Table 1. The EEG bands are delta (0·5 to 4 Hz), theta (4 to 8), alpha (8 to 13) and beta (13 to 35).

Provisions for artifact discrimination are also incorporated into the filter subroutine. Artifact is classified into low frequency (less than 0·5 Hz), high frequency (greater than 35 Hz) and large amplitudes secondary to body motion. Frequency artifact is registered when the half-period falls outside the acceptable EEG ranges as outlined in Table 1. Amplitude artifact is registered when the signal saturates the A/D conversion levels (typically $\pm 100\ \mu$V). Due to the noisy nature of most EEG recordings, analogue or digital filtering of the data prior to analysis is recommended to ensure that an inordinate amount of sub-threshold activity is not registered by the peak-detection process. In this particular application, all data were first subjected to a recursive digital filter with 3 dB cutoff points at 50 Hz.

The peak-detection process continues in a flip-flop fashion, alternating between peak-to-trough and trough-to-peak detecting routines, until one complete epoch of data has been analysed. The epoch length can be somewhat arbitrarily selected but for this application was approximately 3·4 sec. For each epoch the following data are stored for subsequent statistical calculations:

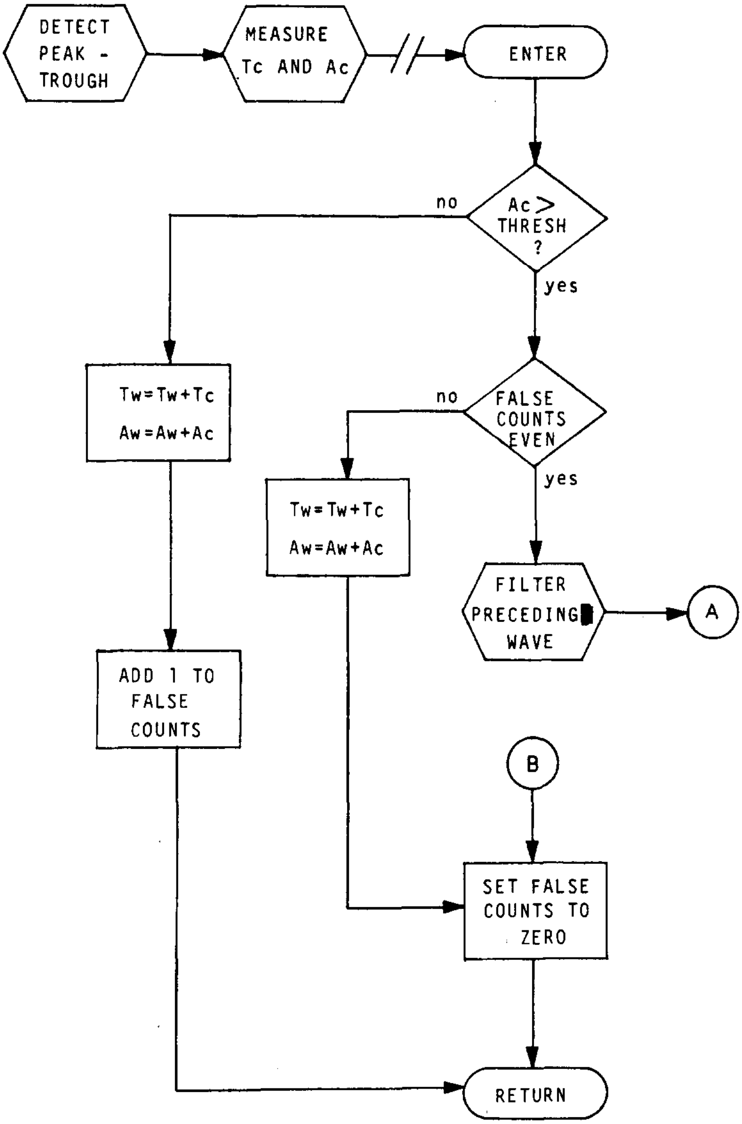

Fig. 2. Flow chart of the FILTER subroutine used in peak-detection analysis. Key: Tc = count time (msec); Tw = wave time (msec); Ac = count amplitude (μV); Aw = wave amplitude (μV).

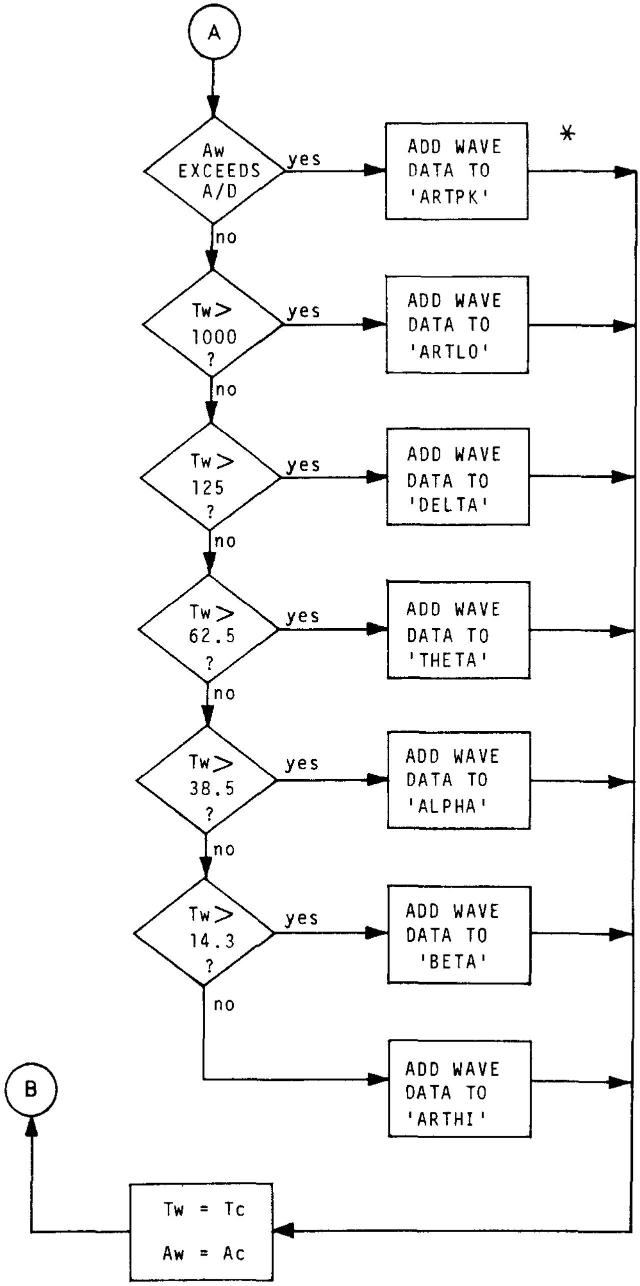

Fig. 2—*cont.* * Wave data are Tw and Aw and the class counter is incremented by 1.

(1) The epoch number and data channel.
(2) The number of waves (W_1, W_2, W_3, W_4) in each EEG band.
(3) The sum of the time (T_1, T_2, T_3, T_4) in each EEG band.
(4) The sum of the amplitudes (A_1, A_2, A_3, A_4) in each EEG band.
(5) The sum of the time (T_5, T_6, T_7) in each artifact category.

Statistics generated

A distinct feature of the peak-detection software is the ability to generate a variety of both time- and frequency-domain EEG statistics. These statistics can be divided into two major categories: intra-band and inter-band. A summary of these statistical categories can be found in Table 2. In addition, right–left ratios, correlation coefficients and other such specifically defined factors can be readily calculated with peak-detection data.

TABLE 2
EEG STATISTICS FROM TEN NORMAL SUBJECTS USING PEAK-DETECTION ANALYSIS

	A. Intra-Band EEG Statistics[a]		
	Activity time (per cent)	*Band frequency (Hz)*	*Band amplitude* (μV)
DELTA	0·77	3·53	21·82
THETA	19·97	6·82	30·09
ALPHA	35·37	10·40	27·44
BETA	41·15	19·18	18·62

B. Inter-Band EEG Statistics[a]

Mean frequency—13·33 Hz
Mean amplitude—22·86 μV
Per cent artifact—2·74 per cent

[a] Grand means for $N = 10$.

The intra-band statistics consist of the per cent activity time, average frequency and average amplitude within each of the four major EEG bands. These can be calculated by using the following formulae for each band:

$$\text{Per cent activity} \doteq \frac{T_i}{\sum_{i=1}^{7} T_i} \times 100 \text{ per cent} \tag{4}$$

$$\text{Band frequency} \doteq \frac{W_i}{2T_i} \text{ Hz} \tag{5}$$

$$\text{Band amplitude} \doteq \frac{A_i}{W_i} \ \mu\text{V} \tag{6}$$

The inter-band statistics consist of wide-band information and yield a representative indication of the overall EEG signal. These statistics can be formulated using the following relations:

$$\text{Mean frequency} \doteq \frac{\sum_{i=1}^{4} W_i}{\sum_{i=1}^{4} 2T_i} \text{ Hz} \tag{7}$$

$$\text{Mean amplitude} \doteq \frac{\sum_{i=1}^{4} A_i}{\sum_{i=1}^{4} W_i} \ \mu\text{V} \tag{8}$$

$$\text{Per cent artifact} \doteq \frac{T_5 + T_6 + T_7}{\sum_{i=1}^{7} T_i} \times 100 \text{ per cent} \tag{9}$$

RESULTS

To test the accuracy of the peak-detection program, selected EEG segments were computer analysed individually. The time interval and amplitude were printed for each contiguous half-wave registered by the program. The same segments were graphed on a digital plotter and, using appropriate voltage and time scale factors, were then scored manually for comparison with the computer results.

A typical example of this signal plotting is shown in Fig. 3. This plot reveals that there was generally good correspondence between the hand-scored and computer results, with most half-period times agreeing within 4 milliseconds. The amplitude comparisons are not depicted but were also in close agreement. Of particular interest in Fig. 3 is the fact that several false counts were correctly not registered, demonstrating the utility of the threshold criterion.

In order to establish baseline data, EEG's from ten normal subjects were analysed with the peak-detection routine. The subjects were in a relaxed, eyes-closed, upright position. Magnetic FM tape-recordings were gathered from C1-O1 and C2-O2 electrode sites over the left and right hemispheres. Typical recording sessions lasted for one minute, with one replication per subject following the administration of a series of mental tasks related to another experiment. Paper strip-charts were also obtained to facilitate visual examination of the data. The FM tape-recordings were digitised on a PDP-11 minicomputer and then processed on an Amdahl 470v/6 computer, generating statistics at epoch lengths of approximately 3·4 sec.

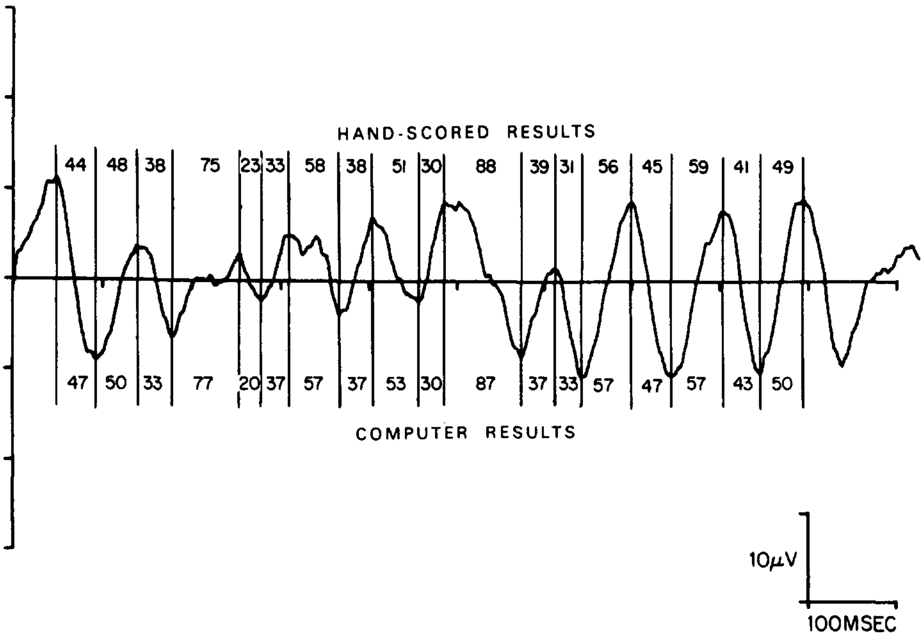

Fig. 3. An EEG segment of peak–trough combinations comparing hand-scored with peak-detection results. Time intervals generally agree within 4 msec.

Average EEG statistics were calculated for each of the ten subjects. These averages were formulated by selecting artifact-free epochs from each one-minute session by visual inspection of the strip-chart records. Grand means of the EEG statistics were then obtained for the entire sample population, as shown in Table 2. The overall EEG composite obtained using peak-detection generally agrees with subjective descriptions of the normal EEG (Kiloh *et al.*, 1972). Specifically, the majority of EEG activity in this awake condition is centred in the alpha and beta frequency bands, with negligible activity existing in the delta band. It should be pointed out that in peak-detection analysis the actual raw numbers, such as those shown in Table 2, are dependent upon the threshold criterion (3 μV), the band limits and the filtering constraints (cut-off at 50 Hz) imposed in the analysis.

DISCUSSION

The use of computer techniques for EEG analysis is becoming common in both research and clinical environments. Nonetheless, a standardised procedure for EEG analysis, which can be used inter-clinically, remains to be established. As pointed out by Fox (1973), this lack of a computer standard may be due, in part, to the difficulty in determining visual classification criteria for EEG's. However, it seems reasonable that a set of computer definitions for basic EEG wave categorisation, such as that offered by the peak-detection algorithm reported in this paper, could be adopted and used between clinics.

Leader *et al.* (1967) were the first to suggest the use of extrema detection as the appropriate procedure for computer analysis of the clinical EEG. As mentioned earlier, peak-detection provides an analytical proximity to manual wave analysis (Sulg, 1969). In peak-detection, each contiguous EEG wave is characterised individually for frequency and amplitude information. In situations that require transient analysis, the wave could be simultaneously subjected to pattern or waveform analysers, although none were employed in this particular application.

A major concern with the peak-detection method is the unwanted registration of low-voltage activity riding on EEG waves. To overcome this problem, an amplitude threshold criterion is employed. Although the 3 μV level imposed in this research is somewhat empirical, it agrees in magnitude with the 2 to 5 μV range reported by others (Sulg, 1969; Stigsby *et al.*, 1973; Jorgensen, 1974). In addition, an even–odd principle for wave formulation, as illustrated in Fig. 1, has been developed to establish the separation point between EEG waves in the presence of sub-threshold activity.

The peak-detection algorithm opportunes the efficient generation of numerous EEG parameters, as shown in Table 2. With peak-detection, the EEG can be classified into time, frequency or amplitude information. When dividing EEG activity into canonical frequency bands, a series of intra-band statistics can be calculated to permit closer scrutiny of particular rhythms. The computational directness of the peak-detection process facilitates EEG analysis and parameter extraction in on-line applications such as neurological monitoring, sleep research, anaesthesia and mental task studies.

The peak-detection algorithm has been tested and proven using a number of normal EEG recordings. A typical EEG composite using peak-detection has been shown in Table 2. In addition, the peak-detection method has been found to be useful in detecting clinical EEG changes during exchange transfusion in Reye's Syndrome. The results of this clinical study have been extensively reported elsewhere (Barr *et al.*, 1977) but can be briefly reviewed here.

The EEG's from seven patients with Reye's Syndrome (hepatic encephalopathy) were monitored using computerised peak-detection techniques. The EEG parameters of primary interest were the delta and theta activities, the mean frequency and the mean amplitude. The major therapy during this period consisted of exchange blood transfusions. In five subjects that survived, noticeable improvement in EEG and clinical profile were observed following therapy. A particularly favourable pattern, consisting of decreasing amplitude and increasing frequency, was observed in the delta wave statistics of these five patients. In the two subjects who expired, no change was observed during exchange treatment and the EEG record gradually became isoelectric. The results of the study strongly suggested that continuous EEG monitoring would be of significant value as a normal clinical adjunct in the treatment of Reye's Syndrome.

The peak-detection software is presently written and documented in FORTRAN. The program is designed for off-line analysis of up to eight channels of digital EEG data using a large-scale computer. The EEG data are entered using an (8,1024) matrix for typical epoch lengths of less than 5 sec. A listing of the program is available from the authors upon request.

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