MICROCOMPUTER-BASED SYSTEM FOR THE DETECTION AND QUANTIFICATION OF PETIT MAL EPILEPSY*

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Abstract—A petit mal seizure detector totally implemented in a 16 bit microcomputer and capable of analyzing on-line at least one channel of EEG data is described. The system uses the repetition period of the wave complexes as the primary parameter for the detection and performs well in clinically significant seizures. Besides characterizing the paroxysms in duration and time of occurrence, the system also evaluates on-line the mean values and variances of the detection parameters, yielding more quantitative information about the seizure data than previously described systems.

Microcomputers Automated detection Petit mal epilepsy Abnormal waveform quantification

INTRODUCTION

Petit mal (PM) is a commonly occurring form of epilepsy. During the PM seizures the EEG displays a well-defined pattern known as the spike and wave complex. An example of a classical PM paroxysm is presented in Fig.1. One of the most striking characteristics of the PM pattern is the rhythmic appearance of the spike and wave complexes. Most previously described automated detection schemes do not utilize this feature, rather they rely primarily on the amplitude information of the PM wave components [1, 2] or on the EEG amplitude integrated in certain frequency bands [3]. The system of Smith et al. [4] requires a minimum number of waves per unit time in addition to an amplitude criteria. Carrie and Frost [5] presented a system that detects spikes and slow waves, but which requires artifact information to distinguish PM paroxysms from high energy, mixed frequency artifacts. The system described by Johnson [6] also detected spikes and slow waves and required the occurrence of spikes followed by slow waves. His results showed that the requirement of spikes followed by slow waves was too restrictive for most PM paroxysms. In a large group of PM seizures (the PM variant) the spikes may be absent or substituted by polyspikes. Figure 2 presents some examples of PM variant seizure patterns. Detection of PM epochs by feature extraction following segmentation of the EEG using linear prediction has also been reported possible [7], but no results presented.

The system described here departs in two main respects from the surveyed work. First, it incorporates a two-stage detection, namely the detection of individual components (spikes and slow waves) as well as pattern rules to distinguish random sequences of spikes and slow waves (as present in artifacts) and the orderly pattern of these elements observed during absence seizures. Second, the characterization of PM paroxysms goes beyond their time of occurrence (with respect to a fixed time origin) and their duration, i.e. the detector output is more than a binary waveform. Each sample of the EEG is analyzed and its information condensed in higher structures, like waves and patterns. So the microcomputer, besides detecting and sorting the seizures, evaluating their duration and obtaining total time, is able to compute on-line for each seizure, the mean and variance of the detection parameters. This new information is much more descriptive of the EEG activity and can be utilized in drug

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studies as well as to improve the detector performance (i.e. customize the detection parameters for each patient).

The PM detector is totally implemented in a 16 bit microcomputer (Texas Instrument TI 9900). The microcomputer-based design has highly desirable features including reproducibility, sophistication, small size, low cost and effectively establishes the link between minicomputer and hardware systems.

The system was validated during the evaluation of an anticonvulsant. The drug effectiveness was assessed in a single placebo experiment by counting, during prolonged EEG recordings, the number of seizures (sorted in 3 duration groups) and the total seizure time [8]. The system was able to answer these particular clinical questions reliably and without human intervention.

**A PETIT MAL SEIZURE MODEL**

Any automated waveform detection scheme is built around a synopsis of the discriminant properties of the signal and the background noise. In biological data, due to the high degree of variability of the signals, analytically-derived models are rarely applicable. A productive method is to define the EEG waves ostensively and formulate heuristic models. The PM model discussed here was arrived at following this routine.

In PM epilepsy the paroxysms are composed of spikes and slow waves, recurring rhythmically with approximately a 3 Hz repetition rate. Therefore, the detection can be divided into two stages: the detection of the individual elements and the detection of the repetitive pattern. In this work, the slow waves and spikes are modelled as half sine-waves and triangular waves, respectively, with amplitude and period requirements which will subsequently be described in detail.

The integration of the individual waves into the pattern criteria should not follow the simple model of spikes followed by slow waves, since in certain forms of PM epilepsy (PM variant) the spike can be absent and/or replaced by polyspikes. The PM seizure criteria implemented here require three slow waves with a prescribed repetition rate (approximately 3 Hz) plus the existence of at least one spike in the same time period. This choice is a compromise between artifact rejection and detection of PM variant patterns. During the paroxysm, the requirement of the spike occurrence is dropped because in the ictal event the
spike is often not well-defined or even absent. Only the periodicity of the wave complexes is checked. Desynchronization of the pattern up to 0.6 s is allowed. The end of seizure is recognized if two slow waves appear consecutively without the 3 Hz repetition rate. The other condition for terminating the seizure is a 1 s epoch without slow waves; the last slow wave detected is taken as the end of the paroxysm.

The implementation in real time of such a complex pattern was facilitated by selecting a microcomputer-based design and choosing easily implementable time domain parameters.

IMPLEMENTATION OF THE DETECTION SCHEME

The block diagram of the PM detector is shown in Fig. 3. It comprises two parallel processing channels (one for the spike the other for the slow wave) and the pattern recognizer which implements the PM model just described.

The detection of the slow waves and spikes was accomplished using time domain properties of the waveforms. Broadband linear bandpass filtering was utilized to attenuate out-of-band activity (high and low frequencies). The period of the individual components of the filtered data was measured. The use of broadband filters is necessary, otherwise, the smoothing produced by the filtering (the filter output is the convolution of the input with the system impulse response) will distort any further processing based on period measures. This methodology has also been successfully used in automated sleep studies [9].

Two fourth-order bandpass filters, one in the slow wave range (0.8–6 Hz) and another in the spike fundamental frequency (10–25 Hz), were used. The bandwidth reflects the compromise between low Q and the desired attenuation of undesirable components (spikes for the slow wave filter and muscle for the spike filter) and was found adequate.

The sample frequency chosen was 240 Hz in order to obtain enough resolution in the spike period measurements (±5%), with simple, zero-crossing techniques. More sophisticated data reconstruction methods, like linear interpolation could be employed with lower sampling rates. However, for this application where a simple microcomputer and its primitive arithmetic facilities are used, and more than one parameter is needed for the signal detection, the faster sampling seems to be less time consuming. With this sampling frequency, the slow wave filter was impossible to realize with the hardware available due to the inherent high gain of narrowband designs [10]. So the sampling frequency for the slow wave channel was reduced to 80 Hz. (The resolution of the slow wave period measurements remained better than ±5%) A 25 Hz analog lowpass filter, (18 dB/octave stop band attenuation rate) was required to reduce the aliased high frequency components (more than 40 db).

The detection of the slow waves and spikes was performed on the filtered data. A set of seizures from four patients was used to determine the detection parameters. Care was taken to choose the seizure patterns used as the training sample so that the detector would be as universal as possible.

The slow waves were defined as waves that have: (1) a peak to valley half period between
0.083 and 0.200 s, which corresponds to sinusoidal activity of 2.5–6 Hz; (2) a maximum time between zero crossings of 0.25 s (to discriminate between in-band activity and fast, high amplitude artifacts); and (3) a filtered peak to peak amplitude of at least 75 µV. The spikes were defined as waveforms that have: (1) a zero crossing period between 0.045 and 0.025 s, which corresponds to sinusoidal activity between 11 and 25 Hz; and (2) a filtered peak to peak amplitude of at least 75 µV.

The allowable repetition period, measured between peaks of the slow waves, is 0.22–0.53 s which corresponds to sinusoidal activity between 1.8 and 4.5 Hz.

The program occupies approximately 1 K words of memory. The longest loop takes 1.82 ms. Figure 4 shows a typical detection. The first channel is the raw EEG, the second shows the slow wave filter output. The down (positive) pulse in channel two is the beginning of seizure flag (seizure beginning is defined to occur 3 slow waves prior to this flag). The up-going (negative) pulse means that a slow wave which did not meet the period requirements was found (the slow wave preceding it is taken as the end of paroxysm).

THE DATA SET

The EEG data used for system evaluation was obtained from a drug study (Valproic acid—Depakene) conducted at the Neurology Service of the Veterans Administration Hospital in Gainesville. Twenty-five patients with uncontrolled PM seizure disorders, most of them under heavy medication, were admitted to the study. From this group of patients, the subset of 6 patients which showed the highest number of paroxysms was selected to test the detector. Three, six (consecutive)-h EEG recordings were obtained for each patient with a 4-channel telemetry unit (Datel Model 1000); the patients were free to sleep, eat, talk, watch television and move around in a 3 × 5 m room. The data set is therefore representative of the condition which will be found in patient monitoring in a home environment. It is worth stressing that the recording obtained in a constrained environment like in a typical EEG session cannot be used to extrapolate the performance of an automated detector for recordings under unrestrained conditions, due to the presence of high energy, mixed frequency artifacts in the latter.

Besides the paper output, the EEG was recorded on a SANBORN 3900 FM tape recorder at a speed of 15 in s⁻¹ (bandwidth 0–250 Hz). Some clipping of the EEG occurred since the maximum allowable signal for linear operation at the telemetry unit input was 200 µV. This affected the amplitude measure of the PM components, but did not have any detrimental consequences for the detection algorithm. Monopolar EEG montages from the frontal, central and occipital regions (International 10–20 system) were utilized. A 50 µV, 10 Hz calibration sinusoid was recorded at the beginning and end of each session, and recorded on tape.
RESULTS AND DISCUSSION

A group of 3 neurologists scored the paper records of the EEG sessions with respect to seizure (Sz) occurrence, duration and sorting into the categories 1 < Sz < 3 s, 3 < Sz < 10 s, Sz > 10 s. The detection of Sz > 3 s will be emphasized since they are considered the most clinically significant [11].

The tapes were played back at real time and one EEG channel (always one of the frontals) was analyzed by the microcomputer. Table 1 shows the man/machine agreement during 70 h of EEG obtained from the 6 patients. The agreement of the computer vs the human with respect to seizure detection is 100% for Sz > 10 s, 83% for 3 < Sz < 10 with an average of 91% for Sz > 3 s; probably more important clinically is a measure of agreement on seizure durations. It can happen that the computer will correctly detect seizures, but will poorly determine their duration. The total duration of the paroxysms obtained by the computer and by the neurologists, is not a first-order measure of the agreement, since positive and negative contributions may cancel out (i.e. computer overscores the duration of one Sz and underscores another). Taking the time sorting into consideration the agreement is 86% for Sz > 10 s, 73% for 3 < Sz < 10 and 77% compound for Sz > 3.

Table 1 also presents the total number of seizures detected by the computer, along with the misses and false detections. There are no misses nor false detections in the group Sz > 10 s. In

<table>
<thead>
<tr>
<th>Human</th>
<th>Computer</th>
<th>1 &lt; Sz &lt; 3 sec</th>
<th>3 &lt; Sz &lt; 10 sec</th>
<th>Sz &gt; 10 sec</th>
</tr>
</thead>
<tbody>
<tr>
<td>Agreement in Seizure Detection (# of Sz)</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>475</td>
<td>844</td>
<td>195</td>
<td>221</td>
<td>93</td>
</tr>
<tr>
<td>Agreement in Seizure Duration (# of Sz)</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>439</td>
<td>162</td>
<td>80</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Total Time in Agreement (sec)</td>
<td></td>
<td></td>
<td></td>
<td></td>
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<tr>
<td>--</td>
<td>1057</td>
<td>1122</td>
<td>1420</td>
<td>1640</td>
</tr>
<tr>
<td>Computer Counts</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>632</td>
<td>225</td>
<td>86</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Computer Misses</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>369</td>
<td>26</td>
<td>0</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Computer False Detects</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>168</td>
<td>12</td>
<td>0</td>
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</tr>
</tbody>
</table>
the group $3 < Sz < 10$ there are 26 misses, more than half of this number coming from a single patient. Figure 5 shows a seizure recorded from this patient who did not present well-defined paroxysmal spikes. As misses are very costly in terms of performance this probably calls for an improvement in the spike detection and/or a modification of the pattern criteria.

The false-positive detections were spread out evenly all over the records and were all $<5$ s duration, occurring primarily during periods when the patient was eating.
Chewing produces electrical potentials in the frontal leads which resemble the PM waveforms, i.e. muscle spikes and slow waves, the repetition rate of the wave complexes being the only noticeable difference. Figure 6a shows a portion of one EEG channel during chewing. The second channel presents the detections of waves that meet the individual wave requirements, however no paroxysm is detected due to the repetition rate requirement. Figure 6b shows an example where even this requirement was met and 2 false seizures (underlined) were detected. Nevertheless, the duration of the false detection is generally very short, since there is a repetition rate requirement during the seizure which is seldom met. For example, one of the patients was eating for 22 min and only two “seizures” >3 s were detected. The other artifacts which cause problems are the body movements and telemetry artifacts (saturation of the FM discriminator produces high amplitude spikes superimposed on a low frequency component). The existence of all these high energy, mixed frequency artifacts, degraded enormously the performance of the detector for seizures <3 s, as Table 1 shows (agreement 36%). One possible way to overcome some of the false detections would be to introduce a muscle artifact detector. However, in our data set the long PM paroxysms were frequently coupled with facial automatism (eye blinking and teeth grinding). Consequently the artifact information would have a detrimental effect on the scored duration of the seizures, so its inclusion was not judged appropriate.

With the PM detector described, it was possible to present the time of occurrence and the duration of each paroxysms detected. Figure 7 shows the distributions of Sz duration (y axis) vs time of occurrence (x axis) for one of the patients. It illustrates a remarkable difference in Sz characteristics with changes in physiological state, i.e. long and sparse Sz during awake and short but dense Sz occurrence in sleep.

With this system it is possible to obtain more detailed information about the PM wave patterns than with previously described systems. At the end of each session, the microcomputer has in memory, for every seizure >3 s: (1) the mean and variance of the PM recruiting period as measured by the time between peaks of the slow waves; (2) the mean and variance of the half period of the slow waves as measured between the peak and valley of the slow wave filter output; (3) the mean and variance of the delay between spike and slow wave measured between slow wave peak and spike peak; (4) the mean and variance of the amplitude of the filtered slow wave; and (5) the mean and variance of the amplitude of the filtered spike.

Table 2 gives these parameters for one of the patients in two different 6 h sessions taken one month apart. The mean values of the period measures (repetition period and half period of the slow waves) are relatively constant from session to session. Also the standard deviation of the period parameter is very small, indicating the constancy of the PM period parameters for the same patient, under the same medication. The amplitude measures vary much more, from session to session and display a much larger standard deviation. This corroborates the experimental observation that EEG amplitude measures vary widely, even for the same patient. It also indicates that the weight given to amplitude parameters in automated detection should be minimized for consistent results.

<table>
<thead>
<tr>
<th>Session (EEG Montage)</th>
<th>Repetition Period (sec)</th>
<th>Half Period Slow Waves (sec)</th>
<th>Delay Slow Wave Spike (sec)</th>
<th>Filtered Slow Wave Amplitude (µV)</th>
<th>Filtered Spike Amplitude (µV)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>x</td>
<td>x</td>
<td>x</td>
<td>x</td>
<td>x</td>
</tr>
<tr>
<td>In 1st SFz-A1</td>
<td>0.32</td>
<td>0.07</td>
<td>0.02</td>
<td>0.23</td>
<td>7.16</td>
</tr>
<tr>
<td></td>
<td>0.09</td>
<td>0.11</td>
<td>0.04</td>
<td>0.39</td>
<td>11.80</td>
</tr>
<tr>
<td>In 2nd SFz-A2</td>
<td>0.33</td>
<td>0.08</td>
<td>0.03</td>
<td>0.34</td>
<td>6.69</td>
</tr>
<tr>
<td></td>
<td>0.12</td>
<td>0.11</td>
<td>0.09</td>
<td>0.35</td>
<td>13.04</td>
</tr>
</tbody>
</table>
The constancy of the PM parameters for the same patient has important implications. First, the parameters can be used to customize the detector for each patient pattern, increasing the detection agreement. The present study used one set of parameters for the entire subject population. As the parameter limits have to be large to cope with the intersubject variability, the system was somewhat sensitive to artifacts and false negatives. The availability of the statistics can be used to pinpoint the false detections at the end of each session. A plot of the repetition period of the wave complexes and its variance vs the time of occurrence is presented in Fig. 8. The false detection is easily identified by the anomalous value (with respect to the constancy) of the repetition period as well as its large variance. This example suggests the use of standard statistical signal detection procedures on the detected seizure data to improve the detector performance [12]. The alternative would be to dynamically customize the seizure parameters for each patient and thereby reduce the number of false detections. For instance, as there are no false detections for the \( Sz > 10 \) s group, the parameters of the first two paroxysms detected in this group could be used for the rest of the sessions. Since the parameter window will be considerably smaller, the artifact rejection will be higher. Experiments are under way to compare both methods.

The other implication of the constancy of the PM period parameters is the use of the system for drug evaluation studies. Changes in the pattern may be coupled with the administration of certain drugs (e.g. there is some evidence that the repetition period of the slow wave complexes varies with Depakene). Therefore, the system can be used to test hypotheses about drug actuation and may eventually lead to dosage indicators.

The PM pattern is easily recognized in the EEG. Nevertheless, long recordings are necessary to obtain a significant sample of the brain activity, to establish a diagnostic with confidence, to quantify the patient response to the medication and to arrive at the adequate dosage. One application of this system is to eliminate the need to hand score the EEG records for number of seizures and total seizure time, since the system automatically supplies this information as soon as the session terminates, saving substantial time to the doctor. Presently the system is being implemented as a low power, self contained, portable device to monitor patients in their home environment for long periods of time. When operational and validated the system will not only save time to the doctor during the evaluation of therapy, but will also operate independent of EEG recording equipment.

The ability to characterize the seizure data and store its parameters in microcomputer memory is not yet fully utilized. It has clear advantages from the detection point of view, working in fact as a new design parameter. However, it is at the level of seizure characterization and drug studies that this facility seems to be the most valuable.
SUMMARY

A microcomputer-based PM detector and analyzer, using the repetition period of the wave complexes as the primary parameter, has been described. The system performs well in clinically significant seizure epochs and supplies much additional information about seizure data. This additional data may prove helpful in drug studies and in the general understanding of PM epilepsy. The first-order data descriptors can also be used to dynamically customize the detector parameters for each subject and help improve the detector performance. The system performance deteriorates for seizures of short duration. Accurate detection of short seizures probably requires the processing of an extra channel of EEG data to use the bilateral synchrony property of the PM epilepsy.

This application also illustrates the interesting characteristics of the microcomputer implementation of the system, combining the advantages of minicomputer-based signal processing algorithms (sophistication, reproducibility) and those of hardware systems (cost, size). The constant technological advances in the area of microelectronics make feasible the implementation of the described system as a low power, portable system.

REFERENCES


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