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Analogue automated analysis of small intestinal electromyogram

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Abstract—*The recording of intestinal electrical activity is used to study digestive motility. This activity consists of slow waves occurring at a frequency of 13–20 cycles per minute. The slow waves are sometimes superimposed with spike potentials. Different patterns of distribution of spike potentials on the slow waves have been shown to occur in physiological and pathological conditions, so that longlasting recording sessions are increasingly required. This presents the problem of analysis of large amounts of data that has not yet been resolved satisfactorily. We present herein an analogue automated system of analysis of the intestinal electrical activity* in dogs. *This system works in real time and provides online data recorded on a graphic recorder. A microcomputer controlled printer and tape recorder were also used. Slow waves are characterised both by the amplitude and the timing of occurrence of relative minima. The spike bursts are detected on the slow waves; their distribution on the slow waves is given, and their energy is measured. Thus, our system allows an easy analysis of long duration chronic recordings, qualitatively (distribution of the spike bursts on the slow waves) as well as quantitatively (time of occurrence of slow waves and number and energy of the spike bursts).*

Keywords—*Automated analysis, Intestinal electromyography, Online measurements*

1 Introduction

AT THE beginning of the century, Alvarez and Mahoney showed that the intestinal smooth muscle generates two kinds of electrical potentials (ALVAREZ and MAHONEY, 1926). They consist of slow waves, always present, that constitute the basic electrical rhythm (b.e.r.); these slow waves are sometimes superimposed with spike potentials (BUNKER *et al.*, 1967; SZURSZEWSKI *et al.*, 1970). Spike potentials have been shown to correspond to intestinal contractions (BASS *et al.*, 1961), making intestinal electromyography a convenient method for studying intestinal motility. Recent studies have reported that the distribution of the bursts of spike potentials on the slow waves followed a rhythmic pattern (the Migrating Myoelectric Complex) in the fasting state. The duration of the MMC is about 90 min in dogs (SZURSZEWSKI, 1969; CODE and SCHLEGEL, 1974). This pattern changes in various pathophysiological situations, so that long lasting recording sessions of the intestinal electromyogram are increasingly needed. However, the complexity of these electrical signals, and the considerable amount of data accumulated over long periods of recording (up to several tens of hours), lead to the problem of their analysis that has not yet been satisfactorily resolved. Several methods are currently used for analysing the electrical signal of the intestine: among them, digital all computerised analysis or spectrum analysis are the most commonly used. The method that is described in this paper is fundamentally an analogue threshold crossing method, which works on line, allowing inexpensive instant analysis of some important characteristics of the intestinal electromyogram.

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2 Signal treatment

2.1 *Signal recording*

The electrical activity of the intestinal smooth muscle was chronically recorded in dogs by means of Ag–AgCl monopolar contact electrodes sewn onto the serosa of the jejunum during a surgical operation. The lead wires from the electrodes were brought out to the surface through a connector inserted into the abdominal wall. When very long duration recordings were performed, a wireless radio transmitter was implanted to provide the electrical signal without inconvenience for the dog (CRENNER *et al.*, 1978; CRENNER *et al.*, 1979). The electromyogram as well as the counted impulses were recorded on a Beckman Dynograph recorder. The bandwidth of the recording line was 0·03 Hz to 100 Hz (-3 dB).

2.2 *Filtering considerations*

The analogue automated system had first to differentiate two kinds of signals: the slow waves and the spike potentials. We characterised the signals by their rise time Tr (time taken by the signal to go from 10% to 90% of the extreme values). Measurements were made on a storage oscilloscope. Fig. 1 shows a slow wave recorded on the jejenum of dogs. During the fastest potential variation (segment AD) the rise time could be as short as $Tr = 50$ ms, the 'flat' intervals (segment DA′) had rise times approaching $Tr \simeq \infty$. The spiking signal had a rise time ranging from 4 ms to 80 ms. Because of the overlapping of the rise time ranges between 50 and 80 ms, it is theoretically impossible to differentiate the slow waves and the spike potentials with analogue filters (lowpass for the slow waves and highpass for the spikes). However in practice the problem could be satisfactorily solved by some circuit artifices and by accurately choosing the filter types and characteristics.

Filters were chosen while considering that the analogue system following the filtering works on signal levels: the amplitude-frequency diagram of the filters had to be rather regular, causing us to use a filter with a polynomial transfer function. The Bessel filters, presenting a very regular amplitude-frequency curve, have a too slow attenuation, even with high-order filters. Legendre or Tchebischeff filters, showing a better discrimination, have a very irregular group propagation time: the different frequency components of the signal will suffer different delays through the filter. This causes important transient deformations and distorts temporal relations. Butterworth filters were finally chosen as a practical compromise because of their intermediate characteristics. They are further optimised for a flat response in the transmission band. A sixth-order filter provides a suitable attenuation slope of -36 dB/octave.

Slow waves were recognised using a lowpass filter with a cutoff point (-3 dB point) at 5 Hz. This cutoff frequency was too low to preserve the highest components of the slow wave signal; however when these highest components were present in the signal, the shape of the slow wave was evident enough for reliable recognition of the cycles. To avoid undesirable slow potential variations due to electrode polarisation, a highpass filter with a slope of -6 dB/octave was used to eliminate the frequencies below 0·03 Hz (equivalent time constant: 5 s).

Spike potentials were selected with a complex combination of three sixth-order Butterworth filters. To measure the energy of the signal, an analogue squaring method was used for rectifying the filtered signal before integrating.

3 Circuit design for slow waves detection

Slow wave detection has several difficulties owing to the nonconstant characteristics of the signal: they vary regarding to the location of the electrode; and even at the same place, they vary with time. The slow wave frequency in the upper small intestine of the dog is 14 to 21 cycles per minute (pseudoperiod: 2·9–4·3 s).

An important problem was that the amplitude of the slow wave (mean value $\simeq 2$ mV p–p) varied with time on a same electrode over a ration of 5 to 1. These variations occurred at irregular and apparently random intervals during long recording sessions.

3.1 *Principle of detection*

The limits of a slow wave were defined as the minima C, C', C'' of the slow wave (Fig. 1). The potentials at points $A, B, C, A', \ldots$ are noted $a, b, c, a', \ldots$, respectively. The system is running until point A'. The potential difference $(a-c)$ is held in an analogue memory; the potential $V = a' - K(a-c)$ is calculated (with $K = 0{\cdot}5$), and held in a memory. A slow wave will be recognised at B', where $b = V$; the wave will be counted at the next minimum C' following B'. After B', the detector is inhibited till point A'', where the same

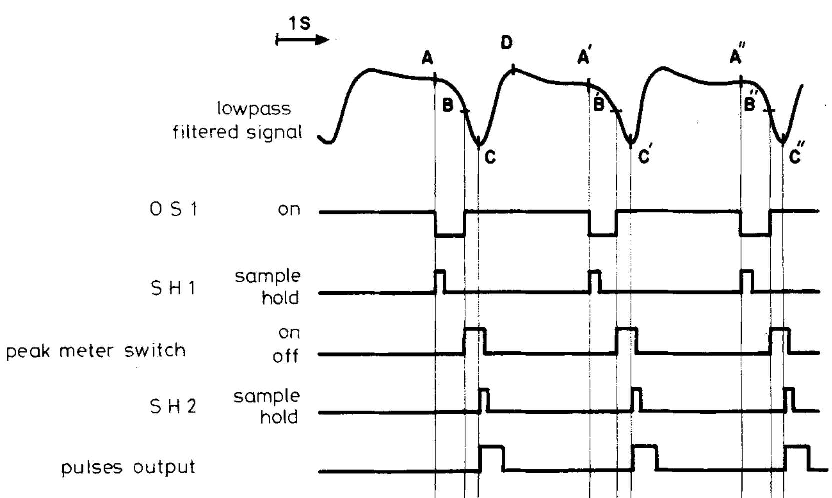

Fig. 1 *Slow waves and timing diagram of the detector*

sequence will recur. The time C' A'' is approximately equal to 4/5 of the minimal possible pseudoperiod of the slow wave. The value given to K sets the maximal possible amplitude variation between two consecutive waves. A large value of 0·8 for K provides good precision in recognition even of complicated waves, but it does not allow the amplitude to vary. A small value of 0·2 for K tolerates large amplitude variations, but increases the risk of erroneous recognitions. The value $K = 0{\cdot}5$ seems to be a suitable compromise.

3.2 *Circuit description* (see block diagram Fig. 2)

The amplified and filtered input signal goes to the negative input of comparator C1. The positive input of C1 is at the potential V. At point B (Fig. 1) the input potential is equal to V and C1 goes high. This activates the peak-meter and causes a 2·5 s pulse to be generated by one-shot OS1. As OS1 is a non-retriggerable monostable, the generation of a pulse is prevented during its on-time; this time determines the inhibition time for the system. At point C (Fig. 1), comparator C2 goes high and causes an output pulse to be generated by OS3. Simultaneously, the peak meter is inactivated and holds the potential c, and sample-and-hold SH2 receives a sample order. SH1 contains a, and the potential held by SH2 is $(c-a)$. When the output voltage of OS1 drops at point A', it causes SH1 to sample the potential a'. SH1 will hold a' till point A''. The positive input of C1 is then at potential $V = a' - K(c-a)$, and the system is ready to recognise another slow wave at point B'. The initial clamping of the system and its fast re-synchronisation when the signal has been disturbed is insured by a missing pulse detector and leak inducers on the sample-and-hold circuits.

4 Circuit design for spike burst detection

The design of the spike limits detector presented a fundamental problem: the precise characteristics of the signals which have to be counted as spikes are hard to define, while visual recognition of the spike potentials is dependent on a global inspection of the electromyogram and on the ability of the operator. The amplitude and shape variations of the b.e.r. waves, whose fast component will sometimes ensure a variable noise level in the spike detector, causes a technical problem.

4.1 *Problem of detection*

After high-pass filtering, the fast electrical signal is squared in an analogue multiplier and goes to an integrator. The integral value, which is proportional to the energy of the signal, goes to a threshold detector that will detect the occurrence of the spike bursts. The

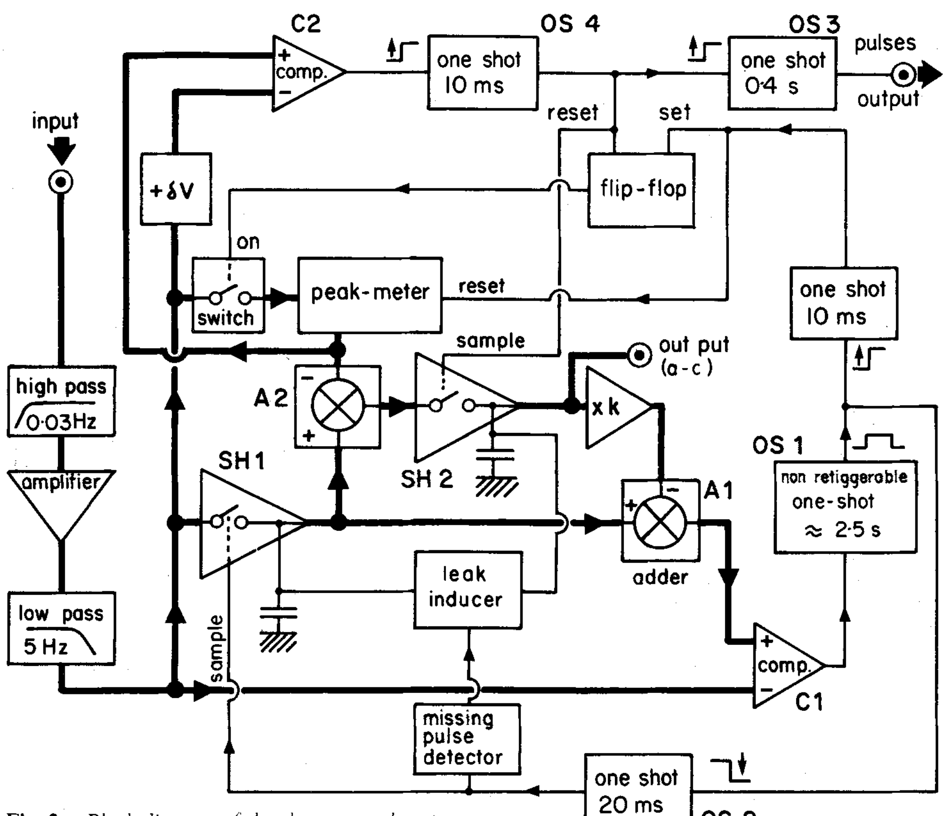

Fig. 2 *Block diagram of the slow waves detector*

integrator is reseted for each slow wave counted by the analyser. The device can run satisfactorily on this basic outline as long as there is no change in the signal amplitude or shape. When recording for long time periods, frequent manual adjustments of the threshold level are necessary, minimising the reliability and the reproductibility of the measurements. Therefore we designed a system based on a pseudo-variable level threshold detector, automatically adjusted with regard to the signal characteristics.

4.2 *Circuit description* (see block diagram, Fig. 3)

The filters represented in the diagram are sixth-order Butterworth filters—i.e. 36 dB/octave of attenuation slope. The electromyographic signal first goes through a highpass filter with a cutoff point at 4 Hz (−3 dB point). Two paths bring the signal through a lowpass and a highpass filter with the same cutoff point at 7 Hz. The filtered signals are squared in analogue multipliers. The squared signals are amplified, with a coefficient of 1 for the highpass filtered signal (> 7 Hz), and about 0·2 for the bandpass-filtered signal (4 to 7 Hz). The second signal is then subtracted from the first, and the outcome is integrated. The integral value is compared with a fixed level in a threshold detector, that causes the generation of a pulse on the output when the threshold is reached. The integrator is reseted by impulses coming from the slow wave detector for each slow wave minimum. Before each reset, the value contained in the integrator can be picked up for a measurement of the energy contained in each burst of spike.

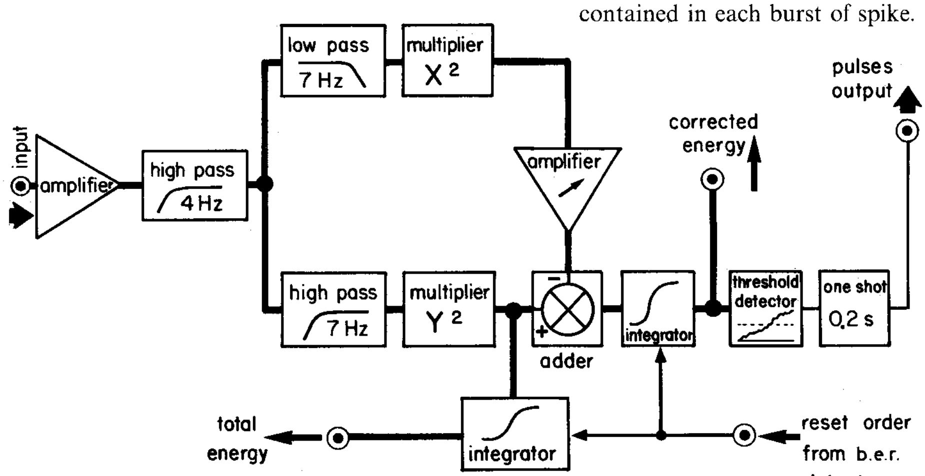

Fig. 3 *Block diagram of the spike bursts detector*

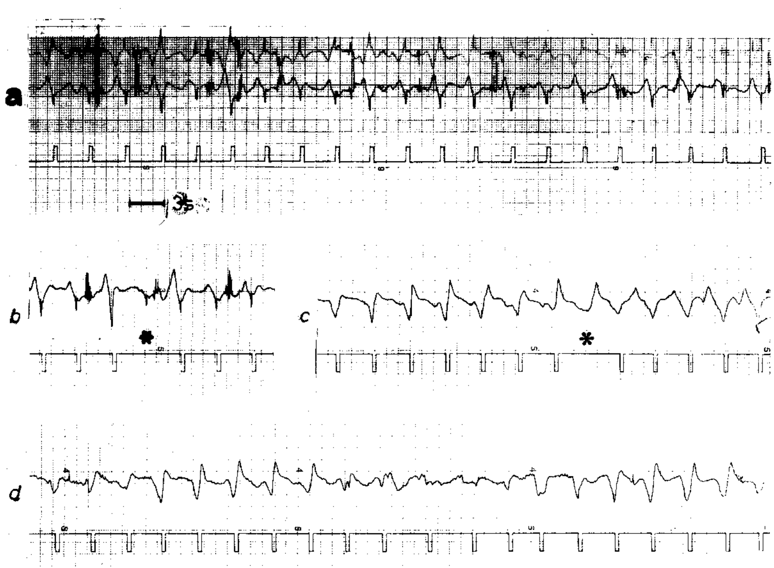

Fig. 4 (*a*) *working of the slow waves detector*
(*b*) * *large amplitude variation brought a failure*
(*c*) * *failure due to signal shape distortion*
(*d*) *working on a distorted signal*

5 Results

5.1 *Slow waves recognition*

Fig. 4 contains electromyographic tracings obtained on the jejenum of a dog. Fig. 4*a* illustrates the good functioning of the device on signals containing spike bursts. An extreme example of amplitude variation between two consecutive waves is given in Fig. 4*b*: the system, adjusted with $K = 0{\cdot}5$, has failed. Fig. 4*c* shows an extreme disturbance of the shape of the signal, where the machine did not recognise a slow wave limit. It can be seen on these two examples that the device resynchronised itself very fast on the correct relative minimum of the slow wave after an error. Fig. 4*d* illustrates a tracing where the signal passed through a confused sequence, where even a visual analysis could not determine the wave limits. The machine found minima that may be slow wave limits. It can be noticed that the waves of the beginning and of the end of the tracing were out of phase (phase shift $\simeq 180°$), so that nothing could be affirmed about the location of the perturbed wave limits. Nevertheless, the machine did not make obvious counting errors, and did not fall out of step.

The detection of the waves is practically independent of the amplitude of the signal. As long as the amplitude variation is slow enough to conserve the gross global shape of two consecutive waves, the amplitude of the slow waves may vary in a ratio of at least 10 to 1 during a recording without disturbing the detector.

The recognition rate for slow waves of the automatic system is not easy to define because it is sometimes difficult to know whether the system must be blamed or the signal was too much distorted for allowing a recognition of the waves. A controlled trial of visual versus machine scoring for the wave limits recognition, which would be made on a necessarily limited number of waves, would be more representative of the particular tracings chosen for the test than of the machine global performance.

To assess the reliability of the device, five recordings coming from different dogs and containing each 20 000 slow waves were analysed. The machine was charged with failure each time that a pulse was not at its normal location, i.e. when a pulse was missing or when the period suffered a shift of more than $\pm 20\%$ compared with the mean period calculated on the 10 preceding waves. The reliability rate was calculated as $R = 100\left(1 - \frac{A}{B}\right)$, where A is the number of failure of the machine and B the total number of slow waves; B is obtained for each tracing by dividing the duration of the recording by the mean period of the waves. On the 100 000 slow waves controlled, representing 92 h of recording, the reliability rate was $R = 98\%$. This rate was obtained without changing any setting of the apparatus even when the dog under experience was changed.

5.2 *Spike bursts detection*

Fig. 5 illustrates how the spike bursts detector works; the threshold of energy for which a pulse was generated was adjusted in order that it corresponds with visual analysis of the spiking activity. Fig. 5*a* shows the counted bursts of spikes. In Fig. 5*b*, an apparent burst of spikes was not counted: the rise-time of the spikes was too low to present a sufficient energy in the frequency band of interest. Fig. 5*c* contains a very noisy signal, inducing over-recognitions. For this setting of the threshold the reliability rate was estimated in comparison with a visual analysis; agreement of the reader about the bursts detected by the machine was 90 to 95%. The recognition disagreements mostly occurred for low energetic spike bursts. For a firing threshold setted at a higher value, a visual control is no more possible. The work of the spike bursts detector has been observed to present a very good repeatability: successive processings of the same recording reproduced exactly the same levels of energy in each slow wave, and consequently the same locations of the pulses on the tracings.

Fig. 6 illustrates the signals that are provided on line by the machine during our recordings.

6 Discussion

The usual methods of small intestine electromyographic signal analysis are based on three main techniques: digital all computerised analysis, spectrum analysis and analogue threshold crossing.

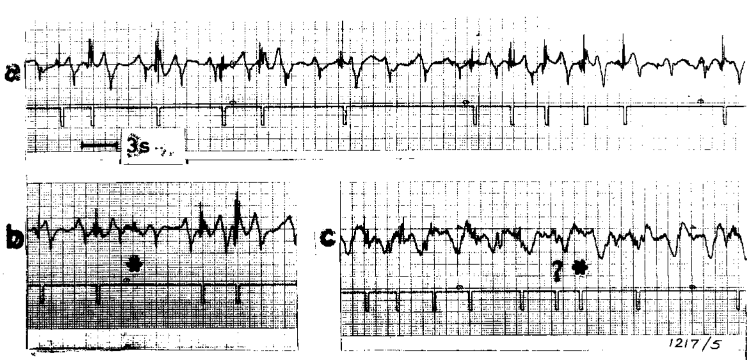

Fig. 5 (*a*) *working of the spikes bursts detector*
(*b*) ** low energetic burst*
(*c*) ** false counting on a noisy signal*
? dubious burst of spikes

A suitable computer program should allow a good scanning of the electromyographic signal, with the ability to put forward any desired parameter concerning individual waves as well as long intervals of time (HIESINGER *et al.*, 1978; POUSSE *et al.*, 1978; POUSSE *et al.*, 1979); however, owing to the sampling rate (at least 100 Hz) necessary for digital direct treatment of the analogue signal, storage requirements are enormous since long duration recordings have to be efficiently treated. Furthermore, if reliable and precise results are desired, the software sophistication requires a high-capability computer for treatment, introducing the problem of systems costs. Furthermore, this method generally does not provide the results online.

Spectrum analysis displays the signal in the frequency domain (STODDARD *et al.*, 1979); although this technique is relatively easy to use and brings interesting results, nothing can be known about individual cycles or time intervals. The duration of the period of analysis is critical to select: too short a period gives imprecise results, while too long a period may bring results too global to be of interest.

The analogue threshold crossing methods are comparable to the previously described system (WINGATE *et al.*, 1977; WINGATE and BARNETT, 1978); however, some characteristics of the intestinal electromyographic signal make the apparent simplicity of that method unmanageable when true and reliable results need to be obtained over long periods of recording. For slow waves detection for instance, use of a fixed threshold is hazardous owing to the variable properties of the signal in amplitude and shape; our system allows for an amplitude variation of more than 10 to 1 without falling out of step. In addition, a simple threshold crossing system provides pulses which are placed on the rising phase of the slow waves, but their exact temporal location depends both on amplitude and shape; this allows measurements of time intervals and cycle durations when each may vary greatly compared to the imprecision of the pulse location (like in stomach) (HIESINGER *et al.*, 1978). But this is not convenient for small but nevertheless physiologically important variations of the slow wave duration in the small bowel (3·1 to 3·5 s at a point in the duodenum). Relative minimum recognition (WODLINGER *et al.*, 1979), if used as definition of the slow wave limit, provides a more accurate temporal location of pulses and allows more legitimate period measurement; furthermore, tracings of interval histograms are then allowed.

Another problem occurs with threshold crossing spikes detection: the fast components of the slow wave, variable in their amplitude and mean slope, should not be recognised as spikes. The usual shunting mean is to set the threshold value higher than the maximum possible noise of the slow wave (HIESINGER *et al.*, 1978; WINGATE *et al.*, 1978). However, numerous weak spike bursts might be lost. In our device, the spikes are defined as having a fundamental frequency up to 7 Hz; the energy contained in the frequency band of 4 to 7 Hz was evaluated and was supposed to be representative of the fast component of the slow wave and of undesirable signals. When the energy in this band

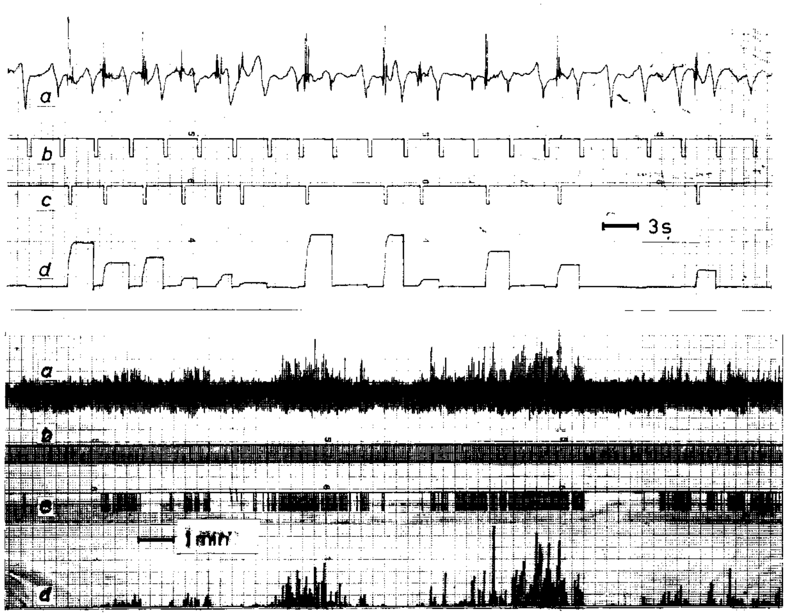

Fig. 6 *Signal obtained online with the machine:*
(*a*) *the intestinal electromyogram (time constant: 5 s, lowpass filtering: 100 Hz)*
(*b*) *pulses provided by the slow waves minima detector*
(*c*) *pulses provided by the spike bursts detector*
(*d*) *energy of the spike bursts, with a linear arbitrary ordinate*

increases, the electromyographic signal should become less 'clean' and the threshold level of energy for recognising a spike burst is augmented. This provides practical results in good accordance with visual examination of the signal; the energy threshold level is automatically maintained just over the level of the noise energy.

The method for evaluating the energy of the spike bursts, calculated as the integral value of the square of the amplitude of the filtered signal, was chosen after an arbitrary definition: a burst of spikes has to be recognised by the counter according to the energy it contains and whether it reaches a determined level in a fixed frequency band. The energy is displayed and taken into account on a linear ordinate, but no numerical scale is given: calculating the value of the energy requires the value of the impedance at which the signal is delivered. This impedance was not monitored and the signal was considered as delivered by a constant impedance source. The threshold level can be set at any desired value, according to the particular study that is done; it is commonly set to a relatively low level for counting the bursts of spikes automatically, so replacing manual counting. To recognise a burst of spikes by visual analysis, a great number of parameters have to be considered, some of them being objective like the amplitude and the number of spikes, some others being more subjective like the general aspect of the tracing and like the slope and the energy of the signal. For our tracings and our conception of the signals, a better agreement with visual inspection was obtained, as far as an analogue system is concerned, by measuring the energy of spike bursts.

The assessment of the reliability rate of the spike bursts detector was a delicate problem, because of the deficiency of objective definitions of a spike burst. No precise quantification of the reliability has been done because nearly 10% of the spike bursts detected by a reader are liable to discussion. The large advantage of the automatic device is that it provides results based on stable and objective characteristics. It seems to be more important that a spike detector can compare precisely and quantitatively different phases of a tracing or different tracings than to reach a top-scoring in a visual versus machine trial. In our experiments, we evaluated the level of energy of each spike burst, which was considered as a fundamental characteristic. The pulses provided for a given level of energy were used for more particular experiments.

Finally, the use of the system can be summarised as follows: it provides online results with easy implementing and comparatively low cost. The relative minimum of each slow wave is detected, thus allowing precise measurements of period and time intervals between pulses. Each spike burst is individually detected and temporally located on the slow wave; an indication of the energy contained in each spike burst and of the slow wave amplitude is provided at the end of each slow wave. The simplest way to work the system needs easy to design digital counters and additional tracks on the graphic recorder. Better results can be obtained if a microcomputer driving a printer and a cassette tape recorder is also used. The storage requirements can be greatly reduced when the collected characteristics of the signal are saved instead of the complete analogue electromyographic signal. Besides it allows renewed further examination.

This signal analysis device is largely used in our laboratory, for example to study the electromyographic response to food intake under physiological or pathological conditions. These responses are characterised by variations of spike appearance on the slow waves (SCHANG *et al.*, 1978). On a clinical point of view, with intent to recognise normal and abnormal patterns of myoelectrical activities, human gastroenterography is now developed: the prolonged recordings require an automated and easy-to-control system, that could be directly derived from the previously described device.

A complete circuit diagram of the device is available from the authors.

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