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Enhancement of the non-invasive electroenterogram to identify intestinal pacemaker activity

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Abbreviated title

Enhancement of the non-invasive EEnG to identify intestinal pacemaker activity

Keywords:

Electroenterogram, empirical mode decomposition, artefact reduction, intestinal activity, non-invasive recording.

Abstract

Surface recording of electroenterogram (EEnG) is a non-invasive method for monitoring intestinal myoelectrical activity. However, surface EEnG is seriously affected by a variety of interferences: cardiac activity, respiration, very low frequency components and movement artefacts. The aim of this study is to eliminate respiratory interference and very low frequency components from external EEnG recording by means of empirical mode decomposition (EMD), so as to obtain more robust indicators of intestinal pacemaker activity from external EEnG signal.

For this purpose, 11 recording sessions were performed in an animal model under fasting conditions and in each individual session the myoelectrical signal was recorded simultaneously in the intestinal serosa and the external abdominal surface in physiological states. Various parameters have been proposed for evaluating the efficacy of the method in reducing interferences: the signal-to-interference ratio (S/I ratio), attenuation of the target and interference signals, the normal slow wave percentage and the stability of the dominant frequency (DF) of the signal.

The results show that the S/I ratio of the processed signals is significantly greater than the original values (9.66±4.44 dB vs. 1.23±5.13 dB), while the target signal was barely attenuated (-0.63±1.02 dB). The application of the EMD method also increased the percentage of the normal slow wave to 100% in each individual session and enabled the stability of the DF of the external signal to be increased considerably. Furthermore, the variation coefficient of the DF derived from the external processed signals is comparable to the coefficient obtained using internal recordings. Therefore the EMD method could be a very useful tool to improve the quality of external EEnG recording in

the low frequency range, and therefore to obtain more robust indicators of the intestinal pacemaker activity from non invasive EEnG recordings.

1. Introduction

1.1. Intestinal Motility

Intestinal motility is essential for the transport and segmentation of the chyme poured by the stomach, and also to mix food with enzymes. Under fasting conditions, intestinal contractions are also necessary to empty the contents of the intestines and to prevent the oral migration of germs and non-pathogenic micro-organisms. The study of intestinal motility is of great clinical relevance, as many pathologies, such as intestinal ischemia, irritable bowel syndrome, mechanical obstruction, bacterial overgrowth or paralytic ileus are associated with intestinal motor dysfunctions (Camilleri et al. 1998).

The difficulty of anatomical access is the principal problem in the monitoring of intestinal activity. Traditionally, intestinal motility measurement has been performed using manometric techniques. However, this method poses a series of technical and physiological problems (Camilleri et al. 1998). Myoelectrical techniques have been validated as an alternative for monitoring intestinal motility, given that there is a correlation between the myoelectrical signal generated by the smooth muscle of the small intestine and its mechanical contraction (Martinez-de-Juan et al. 2000). The relationship between intestinal pressure and the myoelectrical signal is also widely accepted (Martinez-de-Juan et al. 2000).

Intestinal myoelectrical activity (IMA) has two components: slow waves (SW) and spike bursts (SB) (Martinez-de-Juan et al. 2000). The former is a pacemaker activity; its function is to determine the maximum rhythm of muscular contraction. SWs spread from the duodenum to the ileum, and their frequency is almost constant at any particular point in the intestine, although it diminishes as we move along the intestine (Lammers and Stephen 2008). In dogs, it ranges from approximately 19 cycles per

minute (cpm) in the duodenum to 11 cpm in the ileum respectively (Bass and Wiley 1965). Meanwhile, SBs are rapid action potentials, which only appear in the slow-wave plateau when the small intestine contracts, indicating the presence and intensity of intestinal contraction. This can be seen in figure 1, in which the presence of SBs (see Fig. 1b) is directly linked to increases in intestinal pressure (see Fig. 1a). It can also be seen that the SW is always present, even when there are no contractions.

However, the application of the internal myoelectrical technique for clinical diagnosis is limited because it requires surgical intervention in order to implant electrodes.

1.2. Surface Recording

Abdominal surface electroenterogram (EEnG) recording could be an alternative method for determining intestinal motility non-invasively. Recent studies show that both IMA components can be detected on the abdominal surface: the dominant frequency (DF) of the external myoelectrical signal coincides with that of the internal recording in normal physiological (Chen et al. 1993; Garcia-Casado et al. 2005) and pathological situations (Bradshaw et al. 1997; Seidel et al. 1999); and in an animal model it has been demonstrated that the SB activity of the small intestine, which is directly related to mechanical contraction, can also be recorded on the abdominal surface (Garcia-Casado et al. 2005).

However, the clinical application of surface EEnG recording still poses a series of difficulties. First of all, surface recorded myoelectrical signals are very weak (Bradshaw et al. 1997; Chen et al. 1993), especially in the SB frequency range (Garcia-Casado et al. 2005), owing to spatial filtering and the insulating effects of the abdominal layers (Bradshaw et al. 1997; Bradshaw et al. 2001). In addition, external EEnG recording is contaminated by strong physiological interferences: cardiac activity (see

fig. 2a), respiration, very low frequency components and movement artefacts (see fig. 2a). Cardiac activity chiefly affects high frequency components of EEnG, i.e. SB activity (Garcia-Casado et al. 2005).

The main sources of interference in the SW range are respiration and very low frequency components. Respiration chiefly affects the SW band, owing to its similar frequency (Chen et al. 1993; Lin and Chen 1994). The intestinal SW repetition frequency is approximately 0.3 Hz (18 cpm) in dogs, while their respiration frequency is from 0.15-0.6 Hz in physiological states and it can, in rare cases, reach 1.8 Hz. This interference may be the result of the variation in the distance between the surface electrodes and the intestinal myoelectrical signal sources, and the variation in contact impedance between the electrodes and the skin (Ramos et al. 1993). Respiratory interference is a common problem in surface myoelectrical recordings. The presence of respiratory interference depends to a large extent on recording conditions: the way the contact electrodes are fitted, and the position of the electrodes and the body position of the subject being studied. On the other hand, the appearance of very low frequency components may be due to variations in contact impedance between the surface electrodes and the skin or to the bioelectric activity of other organs which are dynamically slower i.e. gastric activity (Chen et al. 1993). To sum up, all these interferences must somehow be removed from external EEnG signals to avoid erroneous interpretation of the results.

1.3. Empirical Mode Decomposition (EMD)

Recently, the EMD method has obtained very promising results both with respect to the characterization of biomedical signals and to the removal of the interferences embedded in these signals (Liang et al. 2000; Maestri et al. 2007). This method does not require the signal under study to be known beforehand and consists of

expanding any complicated signal into a finite number of oscillatory functions, known as intrinsic mode functions (IMFs). An IMF is defined as any function which has the same number of extrema (local maxima and minima) and zero crossings and which has a local mean of zero (Huang et al. 1998). The IMF defined by Huang et al. (Huang et al. 1998) can only be obtained empirically and there is no explicit equation for estimating IMFs, so the decomposition procedure is defined by an algorithm.

IMFs can be interpreted as basic adaptive functions which are extracted directly from a signal, so the EMD method is suitable for analyzing non-stationary signals obtained from non-linear processes (Huang et al. 1998). This is the principal advantage of the EMD method with respect to the Fourier transform, where the basic functions are linear combinations of sine waves. In comparison with Wavelet, the IMFs obtained by means of the EMD method, which represent the masked dynamic processes within the original signal, usually offer a better physically meaningful interpretation of the underlying processes (Huang et al. 1998).

The aim of this study is to reduce interferences in the slow wave frequency range of the surface EEnG recording in dogs using the EMD method. Precisely, respiration and very low frequency components (less than 0.15 Hz) will be attenuated. The idea behind this approach is to identify EEnG SW activities recorded on the abdominal surface and to provide more robust non-invasive indicators of intestinal pacemaker activity.

2. Materials and Methods

2.1. Data Acquisition

Eleven recording sessions were performed on fasting (>12 hours) Beagle dogs in conscious state inside a cage. The animals were previously trained to be kept in this

cage for a period of 2-3 h every day during one week. In the signal recording day, the data acquisition did no start until the animal became accustomed to the laboratory conditions. In each session, internal and external EEnG recordings as well as respiration were simultaneously acquired. Three Ag-AgCl bipolar electrodes were sutured to the small bowel serosa in order to conduct the internal recordings. The internal electrodes were distributed as follows: one in the duodenum, one in the jejunum (75 cm from the Treitz angle) and one in the ileum (250 cm from the angle of Treitz) respectively.

In this study, six monoplar (Ag-AgCl) contact electrodes placed on the abdominal surface were used to obtain four bipolar recordings of surface signals. The contact electrodes were distributed in a 2x3 matrix array and positioned symmetrically respect to the longitudinal axis of the animal, although only one external channel was used in this work. The distance between the electrodes was 3 cm. A reference electrode was also placed on the right hind leg of each animal. The area where the electrodes were placed was shaved and conductive gel was used to reduce the effect of contact impedance on the surface recordings.

Both the internal and external intestinal myoelectrical signals were amplified and conditioned using a [0.05, 35] Hz band-pass filter. The respiration signal was amplified and conditioned using a [0.05, 2] Hz band-pass filter. Finally, all the signals were simultaneously recorded being the sampling frequency of 100 Hz.

2.2. Signal Pre-processing

In this study we focused on interferences in the SW frequency range. Previous studies have shown that the energy associated with SWs is concentrated below 2 Hz (Garcia-Casado et al. 2005; Martinez-de-Juan et al. 2000). So each minute of the surface recorded EEnG signal was digitally filtered using a low-pass filter with a cut-off

frequency of 2 Hz. For our research purposes, saturated signal segments were rejected. Segments with a respiration recording which did not exhibit a clearly defined power spectral density (PSD) peak were also rejected.

Unlike other authors who reject signal segments with movement artefacts (Garcia-Casado et al. 2005; Liang et al. 1997; Martinez-de-Juan et al. 2000), agglomerative hierarchical clustering (Gordon 1987) was used to detect them automatically. Once detected, these segments were reconstructed by cubic spline interpolation.

Figure 2a shows a minute of an abdominal-surface myoelectrical signal recording, in which the presence of a movement artefact can be seen at second 47. The signal filtered using a low-pass filter is shown in figure 2b. The signal pre-processed using cluster analysis is shown in figure 2c.

2.3. EMD Algorithm

Given a signal $x(t)$, which could be the surface EEnG recording in our application, the EMD algorithm can be summarized as follows:

Step 1: Identification of all the local maxima and minima for $x(t)$.

Step 2: Calculation of the upper envelope $e_{max}(t)$ and the lower envelope $e_{min}(t)$ of the signal, combining the local maxima and minima respectively by means of cubic spline interpolation.

Step 3: Estimation of the local mean $m(t)$ of the signal:

$$m(t) = \frac{e_{\max}(t) + e_{\min}(t)}{2} \tag{1}$$

Step 4. Calculation of the difference between the original signal and the local mean $m(t)$:

$$h_1(t) = x(t) - m(t) \tag{2}$$

Step 5: Replacement of the $x(t)$ signal by $h_1(t)$ and repetition of steps 1-4 until the signal which is obtained meets the two criteria for an IMF. The signal obtained as a result of this sifting process is known as $IMF_1(t)$, which is the first IMF function of the $x(t)$ signal. In this study a criterion based on two thresholds levels (Rilling et al. 2003), was used to check whether a signal had a zero local mean. To do this, the evaluation function $f(t)$ was defined as:

$$f(t) = \left| \frac{e_{\max}(t) + e_{\min}(t)}{e_{\max}(t) - e_{\min}(t)} \right| \tag{3}$$

where $e_{\max}(t)$ and $e_{\min}(t)$ are the upper and lower envelopes respectively.

A signal was considered to be an IMF when the evaluation function $f(t) < 0.05$ for 95% of the data and $f(t) < 0.50$ for the remainder (5%).

Step 6: Replacement of the $x(t)$ signal by the residual signal $r_1(t)$ defined in (4) and repetition of steps 1 to 5 in order to extract the successive IMFs:

$$r_1(t) = x(t) - IMF_1(t) \tag{4}$$

The process ends when the residual signal contains less than 14 extremas in one minute (7 cycles/min=0.12 Hz), given that the myoelectrical signal does not contain information below this frequency.

At the end of the iterative process, a collection of n components $IMF_j(t)$ (j=1…n) and an $r_n(t)$ residue will have been obtained. In this way the original $x(t)$ signal can be reconstructed as a linear combination:

$$x(t) = \sum_{j=1}^{n} IMF_j(t) + r_n(t) \tag{5}$$

2.4. *Identification of Interferences*

The next step is to identify which of all the IMFs obtained is associated with interference, with the aim of removing it from the signal. In order to do this, first of all the dominant respiration frequency DF_{resp} must be estimated from the recorded respiration signal. For this purpose, the periodogram was calculated using the Hamming window (PSD_{resp}) on 60 s segments. The DF_{resp} must be sought in the [0.15, 2] Hz frequency range.

Then each IMF is analyzed in its spectral domain in order to verify whether it corresponds to interference or a target signal. Once the PSD_j has been obtained, in the same way as its respiration equivalent, the dominant frequency in the 0-2 Hz range (DF_j) and the mean frequency (MF_j) must be calculated for each IMF. In this study an IMF is considered to be respiratory interference if its DF and MF fall within the DF_{resp}±0.03 Hz range. If both spectral parameters are below 0.15 Hz, then it is assumed that the IMF in question corresponds to very low frequency interference. Finally, the processed signal $y[k]$ is reconstructed as the sum of all the IMFs which are not interferences. The signal processing procedure using the EMD method has been summarized in the block diagram shown in figure 3.

2.5. *Parameters for Performance Quantification of the Method*

In this study the efficacy of the method was quantified by evaluating the improvement in the quality of external EEnG recording. External recording quality in the SW frequency range was evaluated by means of the S/I (signal-to-interference) ratio parameter. In order to do this, the energy associated with interference E_{interf} (eq. (6)) and the energy associated with the target signal E_{signal} (eq. (7)) were calculated in the spectral domain:

$$E_{interf} = E_{verylow} + E_{respiration}$$
$$= \begin{cases} T \cdot \sum_{0Hz}^{0.15Hz} PSD \cdot \Delta f + T \cdot \sum_{DFresp-0.03}^{DFresp+0.03} PSD \cdot \Delta f & \text{if } DF_{resp}\text{-}0.03 \geq 0.15 \text{ Hz} \\ T \cdot \sum_{f_l=0Hz}^{f_l=DFresp+0.03} PSD \cdot \Delta f & \text{otherwise} \end{cases} \quad (6)$$

$$E_{signal} = E_{total} - E_{interf} = T \cdot \sum_{0}^{2Hz} PSD \cdot \Delta f - E_{interf} \quad (7)$$

where PSD is the power spectral density of the original EEnG signal $x[k]$ and/or the processed signal $y[k]$, and T is the time of the segment (60s). When the DF_{resp} is detected at approximately 0.15 Hz, the calculation of the energy associated with very low frequency components partially overlaps with the energy associated with respiratory interference. The definition of E_{interf} in equation (6) enables any errors in estimation due to the problem of overlap to be eliminated.

The S/I ratio is defined as the ratio between the energy associated with the target signal (E_{signal}) and the energy corresponding to the interferences (E_{interf}):

$$\text{S/I}=10 \cdot \log_{10}\left(\frac{E_{signal}}{E_{interf}}\right) \quad (8)$$

Where E_{signal} and E_{interf} are the energy associated with the target signal and with the interferences defined in eq. (6) and eq. (7) respectively.

Afterwards the paired t-test was employed to analyze any statistically significant differences between the S/I ratio of the original signal $x[k]$ and the processed signal $y[k]$ ($p<0.05$).

Other parameters, such as the attenuation of the energy associated with the target signal (A_S) and the attenuation of the energy associated with the interferences (A_I), were also defined in order to verify the contribution of the target signal and interferences energy to the variation of the S/I ratio. The A_S parameter is calculated as the ratio

between the E_{signal} estimated from the processed signals ($E_{signal}\{y[k]\}$) and that of the original signals ($E_{signal}\{x[k]\}$); the A_I parameter is defined in a similar way from the E_{interf} values:

$$A_S = 10 \cdot \log_{10}\left(\frac{E_{signal}\{y[k]\}}{E_{signal}\{x[k]\}}\right) \tag{9}$$

$$A_I = 10 \cdot \log_{10}\left(\frac{E_{interf}\{y[k]\}}{E_{interf}\{x[k]\}}\right) \tag{10}$$

In addition, any improvement achieved in the robustness of the parameters which permit to characterize the intestinal SW activity, was also evaluated after the application of the proposed method. For this purpose, the dominant frequency (DF) of the external EEnG signal was compared before and after application of the EMD method, as well as being compared with DF of internal recordings. In the present study, the DF of the external and/or internal EEnG signals was estimated in the 0-0.5 Hz range. This frequency range was established in order to localize the DF in the intestinal SW repetition frequency range ([0.15, 0.35] Hz), thus eliminating the effect of the first harmonic peak, which can at times be even greater than the fundamental SW peak. The paired t-test was employed to evaluate whether the mean of the differences in DF between the internal and external recordings was zero ($p<0.05$).

To determine how robust the parameters obtained for the external signal were, the DF variation coefficient derived from the external signal was compared to that of internal recording. The variation coefficient was defined as $\sigma/\mu \cdot 100\%$, where μ is the mean and σ the standard deviation. In addition, the normal SW percentage (%NSW) of the external and internal signals was also calculated. This parameter is defined as:

$$NSW\% = \frac{NR}{N} \cdot 100\% \tag{11}$$

Where NR is the number of windows of signal in each recording session which presents a DF in the expected intestinal SW repetition frequency range ([0.15, 0.35] Hz), and N is the total number of windows analyzed in the recording session.

3. Results

3.1. Application Examples

Figure 4 shows an example of the application of the EMD method to a minute of the external EEnG signal. Figures 4a and 4b depict the respiration signal and the pre-processed external EEnG signal respectively, and their corresponding PSDs between 0 and 1 Hz are shown on the right. The PSD of the original EEnG signal has three clearly defined peaks: a peak at 0.04 Hz, which is associated with the very low frequency components contained in the original signal; a peak at 0.26 Hz, which could be associated with intestinal SW frequency; and another peak at 0.41 Hz, which possibly corresponds to respiratory interference, as it coincides with the dominant frequency of the respiration signal and, moreover, it could not correspond to the intestinal SW activity harmonic. Furthermore, the peaks associated with respiration and very low frequency components are of similar magnitude to the peak associated with intestinal SW frequency. The presence of strong interference resulted in a relatively low S/I ratio (0.25 dB).

The decomposition of the signal using the EMD algorithm gave rise to 4 IMFs (see figures 4c-4f) and a residual signal $r_4[k]$ (see figure 4g); their corresponding PSDs from 0 to 1 Hz are shown in figures 4k-4o respectively. In these figures it can be seen that each IMF has different spectral components. In particular, the first extracted IMF corresponds to the most rapid variation of the original signal. As the decomposition process advances, the MF of the IMFs gradually decreases. This is true for all the

analyzed segments and is explained by the fact that, owing to the way the algorithm is constructed, the number of local extrema decreases from one residual signal to the next. The first component, $IMF_1[k]$, corresponds to the most rapid SW component, a finding which might be related to the depolarization and repolarization of cell potentials. The second component, $IMF_2[k]$, could be associated with the first SW activity harmonic. The spectral analysis indicates that the third component, $IMF_3[k]$, corresponds to respiratory interference, given that its DF and MF are 0.41 Hz and 0.39 Hz respectively (see Figure 4m). SW activity is concentrated in $IMF_4[k]$, which has a DF and MF at 0.26 Hz. The residual signal $r_4[k]$ was identified as very low frequency components, given that their DF and MF are 0.05 Hz. The processed signal depicted in figure 4h is therefore the sum of $IMF_1[k]$, $IMF_2[k]$ and $IMF_4[k]$, and its PSD between 0 and 1 Hz is shown in figure 4p. A comparison of the PSDs of the original and the processed signal (figure 4j vs. 4p) clearly shows that the EMD method has enabled to reduce considerably both respiratory and very low frequency component interference, while intestinal SW activity has been successfully obtained. After applying the EMD method, the S/I ratio increased to 13.14 dB.

Figure 5 shows another example which illustrates the application of the EMD method to reduce low frequency interference of the external EEnG signal. The original external EEnG signal is shown in figure 5b and its corresponding PSD is on the right of the figure. In these figures it can be seen that the original signal is strongly affected by very low frequency component interference. The PSD of the original signal (figure 5k) exhibits multiple peaks in the expected SW frequency range ([0.15, 0.35] Hz). The peak at 0.43 Hz might be associated with the respiratory interference contained in the original signal, since it coincides with the respiration frequency for this signal segment. In this case the strong interference even produced a negative S/I ratio (-5.35 dB).

After applying the EMD method, 5 IMFs (see figures 5c-5g) and a residual signal $r_5[k]$ (see figure 5h) were obtained; their corresponding PSDs are shown in figures 5l-5q. Once again, the first $IMF_1[k]$ is the most rapid component contained in the original signal, a finding which might be related to the depolarization and repolarization of cell potentials. The second component, $IMF_2[k]$, presents a DF at 0.43 Hz, which coincides with the DF_{resp}, and its MF is 0.46 Hz, so it was identified as respiratory interference. Meanwhile, $IMF_5[k]$ and $r_5[k]$ correspond to the very low frequency components contained in the original signal, as both the DF and the MF in these signals are below 0.15 Hz. Intestinal SW activity is concentrated in the third IMF, IMF_3, given that its DF and MF are 0.24 and 0.29 Hz respectively. The DF and MF of IMF_4 are 0.17 Hz, so it was not identified as interference and will be included in the signal reconstruction process. Therefore the processed signal shown in figure 5i is the sum of $IMF_1[k]$, $IMF_3[k]$ and $IMF_4[k]$. If the PSD of the original and processed signals are compared (fig. 5k vs. 5r), we can see how the application of the EMD method has correctly revealed the intestinal SW activity submerged in the original signal and that both interferences have been notably reduced. This is also reflected in a significant improvement in the S/I ratio (9.56 dB).

3.2. Quantitative Parameters for the Recording Sessions

The mean and standard deviation ($\mu \pm \sigma$) of the different quantitative parameters calculated on the basis of the original and processed signals in each session are shown in Table 1: S/I ratio, signal attenuation (A_S) and interference attenuation (A_I). The last row shows the mean and standard deviation of these parameters for all the minutes that were analyzed.

For the total number of minutes analyzed, the S/I ratio of the original signals is 1.23±5.13 dB and there is also wide variation between different sessions (ranging from

-5.94 dB for session 11 and 7.93 dB for session 5). After applying the EMD method, the S/I ratio increased to 9.66±4.44 dB and there was much less variation between different sessions (7.67 dB for session 6 and 13.88 dB for session 4). The S/I ratio variation coefficient ($\sigma/\mu \cdot 100\%$) was 417% for the original signals and 46% for the processed signals. Furthermore, the paired t-test indicates that the difference in the S/I ratio before and after applying the EMD method is statistically significant for all the sessions ($p<0.05$). This enables us to affirm that the S/I ratio of the processed signals is less dispersed and significantly greater than that of the original signals, which represents a significant improvement in the quality of the external EEnG signals.

For the total number of minutes analyzed, the A_S and A_I parameters obtained were -0.63±1.02 dB and -9.07±6.04 dB respectively. This indicates that the energy corresponding to the signal was barely attenuated after applying the EMD method, and consequently that the improvement in the S/I ratio is mainly due to the reduction of the energy associated with interference.

The statistical parameters of the DF for the original external EEnG signals and the signals processed using EMD, as well as the internal recordings, are shown in Table 2. The last row shows the mean and standard variation of the DF for all the minutes that were analyzed. Table 3 shows the DF variation coefficient and the %NSW parameter obtained from the internal and external myoelectrical signals, and the last row shows the mean and standard deviation of these parameters for the 11 recording sessions.

Firstly, the DF mean for the internal recording obtained in the duodenum is greater than that of the jejunum and this, in turn, is greater than in the ileum. This confirms that there is a gradient in intestinal SW frequency from the duodenum to the ileum. Furthermore, the variation coefficient of the DF derived from the internal

intestinal signals is small (4% in the duodenum, 6% in the jejunum and 7% in the ileum).

For the total of minutes analyzed, the DF of the original external EEnG signals is 0.18±0.11 Hz. For the 11 recording sessions, the variation coefficient ($\sigma/\mu \cdot 100\%$) of the DF is 54±27%. The NSW percentage is only 64±27%, which suggests to us that approximately 36% of the analyzed signals have a DF outside of the intestinal SW repetition frequency range. After applying the EMD method, the DF of the external signals is 0.26±0.04 Hz, resulting a variation coefficient of 10±3%, which means the variation obtained was much less than that of the original signals. The NSW percentage increased to 100% in all the sessions. Consequently, the DF of the processed external signals is much more stable and robust than that of the original signals.

When compared with the internal recordings, the DF mean for the processed external signals falls between the average value for the duodenum and the ileum in all the sessions. Furthermore, the variation coefficient of the DF and the NSW percentage from the processed external signals and the NSW percentage are similar in magnitude to that of the internal recordings.

Finally, table 2 shows the results of the statistical analysis, performed using the paired t-test, of the DF obtained from each of the internal recordings with the corresponding DFs derived from external processed signals. In sessions 1, 4, 5 and 10 the statistical test indicates that there are no differences between the DF of the surface processed signals and that of any internal recording. In the remaining cases, the difference between the DF derived from the processed external signal and the internal signal has a mean value which deviates statistically from zero for the 3 internal recordings.

4. Discussion

4.1. Removal of Interferences in Surface EEnG

Respiration and very low frequency components are the main sources of interference in the SW range of EEnG recordings (Chen, Schirmer, & Mccallum 1993d;Garcia-Casado, Martinez-de-Juan, & Ponce 2005a). Conventional frequency domain filters could be used to remove these interferences from surface EEnG signals. However they may distort the waveforms of the myoelectrical signal by filtering out harmonics of the fundamental frequency of the intestinal signal and even removing part of the fundamental component, given that the respiration frequency may be very close to the intestinal SW frequency. Conventional filters could neither be used for reducing interferences in surface EEnG recordings in pathological situations such as intestinal ischemia, given than in these conditions the intestinal SW frequency decreases and could fall into the very low frequency range (Bradshaw et al. 1997a). In addition, the use of conventional filters is restricted by the linearity of the system and by the stationarity of the data. It has been shown that external myoelectrical recording is non-stationary (Lin & Chen 1994a;Lin & Chen 1995) and also is non-lineal (Lindberg 1996). Therefore, the applicability of conventional filters for reducing the interferences embedded in surface myoelectrical recordings has been refuted by diverse authors (Chen & Lin 1993;Irimia & Bradshaw 2005c;Liang 2001c;Liang, Lin, & Mccallum 2000a;Lin & Chen 1994b).

In this study we tested the ability of the EMD method to reduce the principal interferences in the SW range of external EEnG recording: respiration and very low frequency components. Since the EMD decomposition is based on the local characteristic time scale of the data to yield the adaptive basis, this method is applicable to nonlinear and non-stationary processes (Huang et al. 1998a). Another advantage of

this approach is that it does not require any prior knowledge of the nature of the signal under study. The resulting IMFs which represent the masked dynamic processes within the original signal, usually offer a better physically meaningful interpretation of the underlying processes (Huang et al. 1998b). This phenomenon can be seen in the fig.4 and fig.5 in which the distinct components embedded in the original signal have been clearly revealed after the decomposition of the EMD method. The experimental results show the effectiveness of the EMD method for reducing interferences, owing to the basic adaptive functions of the EMD approach, used in conjunction with the proposed identification procedure. This can be seen in the S/I ratio before and after applying the proposed method (9.66±4.44 dB processed vs. 1.23±5.13 dB original signal). Moreover, the energy associated with target signal was barely attenuated after the application of the EMD method. Our results agree with other authors who have used the EMD method to remove a variety of interferences from surface EGG signals (Liang, Lin, & Mccallum 2000b). Unfortunately, it was impossible to compare quantitatively the results found in both studies, given that no parameters which permit to quantify the improvement in the quality the EGG recordings were provided in that study (Liang, Lin, & Mccallum 2000c).

The analysis on the stability of the DF of external signal and the NSW percentage before and after the application of the EMD method, also show that the DF of the processed external signals is more stable and robust than that of original signals. We ought to emphasize that under physiological conditions the NSW percentage does not truly reflect the number of signal segments whose DF is associated with intestinal SW frequency, given that in these conditions, respiration frequency may fall within the expected intestinal SW repetition frequency range. Nevertheless, the DF variation coefficient of the processed external signal is similar in magnitude to the internal ones

and the improvement obtained in the S/I ratio suggests that respiratory interference in the external recordings after the application of the EMD method is minimal. Everything points to the conclusion that the EMD method effectively cancels the low-frequency interferences in external EEnG recordings, and it can be used to improve the quality of surface EEnG recordings. As a consequence, more robust parameters which permit to characterize the SW activity can be obtained from non-invasive EEnG recordings. In short, the EMD method could be a very helpful tool to identify the intestinal pacemaker activity from the non-invasive EEnG recordings.

Besides the above mentioned method, other techniques, such as adaptive filtering (Chen and Lin 1993; Lin and Chen 1994; Prats-Boluda et al. 2007) and independent component analysis (ICA) (Irimia and Bradshaw 2005; Wang et al. 1999), also have been proposed to remove the interferences from external myoelectrical recordings. These studies have demonstrated the ability of these techniques to reduce the interferences in external myoelectrical recordings, and to improve the quality of surface recordings. However, the applicability of adaptive filtering for this purpose has been questioned by some authors (Liang, Lin, & Mccallum 2000d;Wang, Cheung, & Chen 1999c), given that it requires a reference signal that is the comprehensive signal of the various interferences to be removed. Therefore, its ability to reduce interferences strongly depends on the reference signal which is used (Chen and Lin 1993; Lin and Chen 1994; Prats-Boluda et al. 2007). Like the adaptive filter, in this work it has been proved that the EMD method could also be employed to reduce the respiratory interference in external myoelectrical recording, In this context, the advantage of the EMD-based method lies in the fact that there is no need to find an optimal reference signal, since it is only necessary to know the respiration frequency. Moreover, the respiratory interference contained in external recordings is not always present

throughout the recording session i.e it can vary in adjacent segments .In these circumstances, the filter weights need time to adapt and in the meanwhile they can distort the signal. On the contrary, with the EMD approach the interferences are only removed from the signal when they have been proved to be present in the IMFs otherwise the signal remains unaffected.

Additionally, the application of the ICA method to a similar signal to the surface EEnG such as surface EGG has shown its ability to identify the frequency of the SW activity (Estombelo-Montesco et al. 2007a;Irimia & Bradshaw 2005b;Liang 2001b;Wang, Cheung, & Chen 1999b). Although the results are not shown in this paper, we also tested the ICA method to reduce respiration and very low frequency interferences from the four channels of EEnG signals. The processed signals presented better S/I ratio, and lower variability in the DF compared to the original ones. Nevertheless these results were worse than those obtained when applying the EMD method. This agreee with the observations of other authors who defended that the ICA method is not suitable for improving the S/I ratio of each external channel when only a few channels are available (Liang 2001a;Wang, Cheung, & Chen 1999a). Others studies that used 19 channels of simultaneously recorded magnetogastrogram, showed that ICA method was able to separate even frequency-overlapping interferences and wide-band artefacts from gastric signals (Estombelo-Montesco et al. 2007b;Irimia & Bradshaw 2005a). Therefore, the number of recording channels is a key point in the success of ICA-based methods, and future works should check the performance of these methods using a greater number (>4) of surface EEnG channels. Regarding to this, the EMD presents the advantage of not requiring multichannel recordings to decompose the original signal into its different components.

4.2. Detection of the SW Component in Surface Recording

Although SW activity of the IMA is not a direct representation of intestinal contraction (Garcia-Casado et al. 2005; Martinez-de-Juan et al. 2000), some pathologies, like intestinal ischemia, are directly related to pacemaker activity and they reduce SW frequency and amplitude (Bradshaw et al. 1997; Seidel et al. 1999). Other authors defend the view that pathologies which are related in some way to pacemaker activity can be identified by evaluating the percentage of time the myoelectrical signal shows an abnormally high or low DF, absence of the DF or the energy percentage associated with the DF (Chen and Mccallum 1991). Consequently, the detection of SW activity is also of clinical value.

Measurement of IMA has traditionally been performed by placing electrodes on the bowel serosa. Nevertheless, the clinical application of these techniques is limited due to its invasiveness. Surface EEnG recording could be an alternative for monitoring the IMA non-invasively (Chen, Schirmer, & Mccallum 1993c;Garcia-Casado, Martinez-de-Juan, & Ponce 2005b). External EEnG signals are severely attenuated by tissues of low conductivity of abdominal layers and due to the spatial filtering effect associated to the longer distance between the sensing electrodes and the signal origin (Bradshaw et al. 1997b). Previous studies of these effects on the surface EEnG spectrum revealed that the high-frequency components of the EEnG are more severely attenuated than the low-frequency components (Garcia-Casado, Martinez-de-Juan, & Ponce 2003). This fact has led others authors (Bradshaw et al. 1997c;Chen, Schirmer, & Mccallum 1993b), to focus their studies on the low-frequency component of EEnG, i.e. the slow wave. In addition, the presence of respiratory interference and very low frequency components in surface EEnG recording impair seriously the stability of the signal (Amaris et al. 2002), and therefore could induce an erroneous diagnosis. This can be seen in the variation

coefficient of the DF of original external signals which is much higher than that of internal recordings (see table 3). Therefore, the development of a method, which enables to remove these interferences, would be of great importance to extract more robust indicators from these surface signals.

The proposed EMD method proved to be able to reduce the main interferences that affect the intestinal pacemaker frequency band, being the variation coefficient of the DF of processed surface signals similar to those of internal recordings (see table 3). Not only the variability but also the values of the DF of processed surface signals are more similar to those of internal signals (see table 2). Moreover, after processing the signal, a frequency peak within the expected value of intestinal SW (0.15- 0.35 Hz) was found in all the analysed segments of surface EEnG signals (NSW=100%). We believe that the detected peak frequency was of small bowel origin due to a series of basis. Firstly, the detected peak frequency is consistent with that reported in the literature (19 cpm in the duodenum and 11 cpm in the ileum) (Bass & Wiley 1965). Secondly, the average of the DF of external signals on each recording session falls within the average values of the DF obtained from duodenum and ileum. Moreover, as stated before, the DF of the processed external signals is stable and robust. Finally, rejecting the respiration, there is no evidence of the existence of any other omnipresent myoelectrical signal whose frequencies lies within 9-21 cpm in other parts of the human body (Chen, Schirmer, & Mccallum 1993a). Therefore, as earlier stated by other authors (Bradshaw et al. 1997; Chen et al. 1993; Garcia-Casado et al. 2005), the experimental results in this work showed that it is possible to detect intestinal SW activity in external EEnG recordings, and that the application of the EMD method is of great help to cancel the interferences that could lead to misinterpretations of these non-invasive studies.

On the other hand, though the results of the present study suggest that the activity recorded by the surface electrodes corresponds to a bowel segment located between the internal electrodes of duodenum and jejunum (see average DF, table 2), the statistical comparison of the DF of the external and internal recordings failed to identify the point or section of the intestine whose activity was being recorded during external recording. This is one of the difficulties of non-invasive EEnG recording (Chen et al. 1993). In addition, it is a problem which is shared by manometric techniques which do not involve surgery (Quigley 1996) and other non-invasive techniques for recording intestinal activity, such as measuring intestinal sounds (Tomomasa et al. 1999) or surface magnetic recording (Seidel et al. 1999). This difficulty may be due to differences in the size of the area used for external and internal recording and the poor spatial resolution of the bipolar recordings employed in the present study. Some authors have postulated that they failed to identify the propagation of SWs in external bipolar EGG recordings owing to poor spatial resolution (Mintchev et al. 1993). This poor spatial resolution may also provoke that the surface electrodes record the activity of several intestinal segments which are physically close to one another (Bradshaw et al. 1997d). In these cases, multiple peaks can appear in the intestinal SW frequency range (see fig. 5r). Moreover, the contribution of the different internal source signals to the external recording could vary as a result of possible changes in the position of the subject under study. This could explain why the variability of the processed surface signals remains slightly higher than that of internal signals (see table 3). The latest research in this field indicates that the spatial resolution of conventional bipolar surface recordings could be improved by using electrodes based on the estimation of the laplacian of the potential (Prats-Boluda et al. 2007). Theoretically, the laplacian of the potential contains the information which is inherent to the potential recording, but it

provides better spatial resolution than bipolar recordings of the potential (Prats-Boluda et al. 2007). This is because the laplacian of the potential is more sensitive to electrical sources which are perpendicular to the abdominal surface, with the result that there is greater rejection of distant signal sources. However, laplacian recordings of intestinal myoelectrical signals are also affected by respiration and very low-frequency interferences (Prats-Boluda et al. 2007). In respect to this, the EMD method proposed in this study could easily be adapted in order to reduce interferences in the intestinal SW range of both laplacian or bipolar surface EEnG recordings in humans. By doing so, the potential uses of these non-invasive techniques in clinical medicine would be increased.

5. Conclusion

The experimental results show that the EMD method, used in conjunction with the procedure proposed for identifying interferences, permits the two major interferences in the SW frequency range of external EEnG recording to be considerably reduced, i.e.: respiratory interference and very low frequency components. The S/I ratio of the external EEnG signals was significantly improved, as a result of an exhaustive attenuation of the interferences. The frequency parameters of the processed external signals are much more robust than the parameters of the original signals, and the dominant frequency is within the expected range for intestinal SW activity. Moreover, the variation in this frequency in processed signals is similar in magnitude to the variation which occurs in internal recordings.

Therefore the proposed method could be a very useful tool for improving the quality of external EEnG recordings and obtaining more robust indicators of intestinal SW activity, thus potentiating the future clinical use of this non-invasive recording technique.

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Table 1. Mean ± standard deviation of the S/I ratio, A_S and A_I before and after the application of the EMD Method. N is the number of analyzed minutes in each session.

Session	N (min)	S/I ratio (dB)		Attenuation (dB)	
		Original	Processed	A_S	A_I
1	128	-0.70±3.10	8.14±3.40	-1.01±1.09	-9.81±4.80
2	141	-2.25±3.31	8.60±3.78	-0.90±1.08	-11.76±4.96
3	125	1.76±3.15	9.27±3.94	-0.42±0.67	-7.89±4.27
4	118	3.73±3.68	13.88±4.49	-0.15±0.25	-10.52±4.14
5	121	7.93±4.27	12.12±2.79	-0.14±0.30	-4.33±4.03
6	124	3.82±3.24	7.67±3.65	-0.65±1.02	-4.47±4.45
7	86	2.46±3.68	10.10±3.03	-0.24±0.45	-7.89±4.15
8	64	-0.09±4.50	8.61±4.27	-0.84±1.04	-9.53±6.22
9	124	1.95±3.70	7.25±4.28	-1.06±1.27	-6.36±6.03
10	99	2.27±4.36	12.24±4.63	-0.14±1.23	-10.10±5.18
11	136	-5.94±3.51	9.07±4.72	-1.11±1.19	-16.09±6.31
Total	1266	1.23±5.13	9.66±4.44	-0.63±1.02	-9.07±6.04

Table 2. Mean ± standard deviation of the DF of internal recordings, original surface EEnG recordings (*x[k]*) and processed signals by means of the EMD method (*y[k]*). N is the number of analyzed minutes in each session. Asterisks indicate that the mean of the difference between the DF of each internal recording and of the external processed signal is zero.

Session	N (min)	Internal DF *(Hz)*			Surface DF (Hz)	
		Duodenum	Jejunum	Ileum	*x[k]*	*y[k]*
1	128	0.27±0.01	0.28±0.01	0.24±0.01 *	0.14±0.11	0.25±0.03
2	141	0.31±0.04	0.30±0.04	0.25±0.02	0.13±0.11	0.26±0.03
3	125	0.29±0.01	0.29±0.02	0.25±0.02	0.20±0.11	0.27±0.03
4	118	0.31±0.01	0.29±0.02 *	0.24±0.01	0.21±0.10	0.29±0.03
5	121	0.29±0.01	0.26±0.02	0.22±0.02*	0.21±0.05	0.23±0.02
6	124	0.29±0.01	0.26±0.02	0.22±0.02	0.24±0.04	0.25±0.03
7	86	0.30±0.01	0.28±0.01	0.23±0.01	0.18±0.11	0.26±0.02
8	64	0.30±0.01	0.27±0.02	0.22±0.02	0.17±0.11	0.25±0.03
9	124	0.30±0.01	0.27±0.02	0.22±0.01	0.25±0.04	0.26±0.03
10	99	0.32±0.01 *	0.32±0.01 *	0.28±0.02	0.24±0.11	0.32±0.01
11	137	0.32±0.01	0.31±0.01	0.28±0.02	0.05±0.05	0.28±0.04
Total	1266	0.30±0.02	0.29±0.03	0.24±0.03	0.18±0.11	0.26±0.04

Table 3. Variation coefficient (σ/μ·100%) of the DF and the percentage of normal intestinal slow wave activity (% NSW) obtained from internal recordings, original surface EEnG recordings (*x[k]*) and processed signals by means of the EMD method (*y[k]*).

Session	σ/μ·100 (%)					%NSW				
	Duodenum	Jejunum	Ileum	*x[k]*	*y[k]*	Duodenum	Jejunum	Ileum	*x[k]*	*y[k]*
1	4	4	4	78	12	100	100	100	48	100
2	13	13	8	84	12	100	100	100	43	100
3	3	7	8	53	11	100	100	100	70	100
4	3	7	4	48	10	100	100	100	62	100
5	3	8	9	24	9	100	100	100	93	100
6	3	8	9	17	12	100	100	100	97	100
7	3	4	4	61	8	100	100	100	66	100
8	3	7	9	65	12	100	100	100	58	100
9	3	7	5	16	12	100	100	100	97	100
10	3	3	7	46	3	100	100	100	68	100
11	3	3	7	100	14	100	100	100	3	100
μ±σ (N=11)	4±3	6±3	7±2	54±27	10±3	100±0	100±0	100±0	64±27	100±0

CAPTIONS.

Figure 1. Simultaneous recording of bowel pressure (a) and internal myoelectrical activity (b) in the same bowel loop in a non-sedated dog.

Figure 2. (a) One minute of original surface EEnG recording containing a motion artefact (second 47). (b) Filtered EEnG with cut-off frequency at 2 Hz. (c) Estimated EEnG without motion artefact processed by hierarchical agglomerative clustering.

Figure 3. Block diagram of signal processing using the EMD method, where *x[k]* is the pre-processed surface EEnG signal, DF is the dominant frequency of the IMFs between 0-2 Hz, MF is the mean frequency of the IMFs and *y[k]* is the processed signal. The residual signal $r_n[k]$ was not taken into consideration during the reconstruction process due to its very low frequency.

Figure 4. Application of the EMD method to 1 min of surface EEnG recording. (a) Respiration recording. (b) Original EEnG signal after pre-processing (*x[k]*). (c)-(g) Results after application of the EMD method: four IMFs and one residual signal. (h) Processed signal (*y[k]*): the sum of IMF_1, IMF_2 and IMF_4. (i)-(p) Power spectra of the signals that are depicted on the left-hand side. It should be noted that the PSD of each IMF has a different scale, and the PSD of the original signal (plotted as j) and that of the processed signal (plotted as p) are represented on the same scale so they can be visually compared.

Figure 5. Application of the EMD method to 1 min of surface EEnG recording. (a) Respiration recording. (b) Original EEnG signal after pre-processing (*x[k]*). (c)-(h) Results of the EMD method: five IMFs and one residual signal. (i) Processed signal (*y[k]*): the sum of IMF_1, IMF_3 and IMF_4. (j)-(r) Power spectra of the signals that are depicted on the left-hand side. It should be noted that the PSD of each IMF has a different scale, and the PSD of the original signal (plotted as k) and that of the processed signal (plotted as r) are represented on the same scale so they can be visually compared.

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