Flexible Array of Active Concentric Ring Electrodes for Surface Bioelectrical Recordings

Application to Non-invasive Recordings of EEnG

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Abstract: The estimation of laplacian potential on the body surface obtained by means of concentric ring electrodes can provide bioelectrical signals with better spatial resolution and less affected by bioelectrical interferences than monopolar and bipolar recordings with conventional disc electrodes. These ringed electrodes are usually implemented on rigid substrates which cannot adapt to body surface curvature. In this paper an array of flexible concentric ring electrodes for non-invasive bioelectrical activity recordings is presented. The array contains three tripolar electrodes in bipolar configuration (TCB, inner disc and external ring are shorted) which is suitable for body surface mapping. A preconditioning circuit module is directly connected to the electrode array to perform a first stage of filtering and amplification as close as possible to the recording area. Simultaneous recordings of intestinal myoelectrical activity (electroenterogram, EEnG) by means of the flexible array of ringed-electrodes and by disc electrodes with gel were carried out in healthy volunteers in fast state. Signals from the developed array of electrodes presented lower electrocardiographic and respiratory interference than conventional bipolar recordings with disc electrodes. The small bowel's slow wave myoelectrical activity can be identified more easily in the ringed-electrodes recordings.

1 INTRODUCTION

1.1 Biolectrical Laplacian Recordings

Surface recordings of bioelectrical signals are usually recorded by means of disc electrodes in bipolar or unipolar configuration. In the first method the potential difference between a pair of electrodes is measured. In the latter method the potential of each electrode is compared either to a neutral electrode or to the average of several electrodes. One drawback of using conventional disc electrodes in bioelectrical surface recordings is their poor spatial resolution which is mainly caused by the blurring effect of the different conductivities of the volume conductor (Bradshaw et al., 2001). In this respect, Laplacian has been shown to reduce the smoothing effects caused by the volume conductor and to increase the spatial resolution in localizing and differentiating multiple dipole sources (Wu et al., 1999); (Besio et al., 2006).

There are different approaches to estimate the laplacian potential on the body surface. The first ones to be used were discretization techniques like the one introduced by Hjorth as early as in 1975 (Hjorth, 1975). In that study, the laplacian of the EEG signal was estimated as the difference between the average potential of four disc electrodes in the form of a cross and the potential of a fifth disc electrode placed in the center of the cross. In the late 80s, analytic solutions to estimate the laplacian of the surface potential were proposed in order to reduce discretization errors (Perrin et al., 1987). These are complex discrete computational techniques, generally not suitable for real-time applications. Nevertheless, laplacian potential can also be directly estimated by means of concentric ring electrodes in tripolar, bipolar or tripolar in bipolar configuration (TCB, where the outer ring and the center disc were electrically shorted) (Lu and Tarjan, 1999); (Besio et al., 2006); (Koka and Besio, 2007).

TCB ring electrodes have already been used to estimate the Laplacian potential of bioelectrical signals such as the electrocardiogram (ECG),

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electroencephalogram (EEG), electroenterogram (EEnG) and the electrohysterogram (EHG), so as to increase the spatial resolution of and the signal quality of conventional surface potential recordings (Li et al., 2005); (Prats-Boluda et al., 2011); (Koka and Besio, 2007). These electrodes are usually active since the signals sensed by concentric ring electrodes in laplacian configuration are weaker than the ones obtained my conventional monopolar or bipolar recordings, and the output impedance is bigger. Nevertheless, the ringed-electrodes used in these studies were developed on rigid substrates like printed-circuit boards, what can provoke a poor skin-to-electrode contact and discomfort to the patient.

Therefore, the aim of this study is to develop concentric ring electrodes on a flexible substrate to join the advantages of laplacian recordings with the comfort and better adaptation to the body surface curvature of conventional disposable disc electrodes.

1.2 Intestinal Myoelectrical Activity

The study of intestinal motility is an outstanding field in gastroenterology due to the fact that abnormal motility patterns are related with several intestinal pathologies (Quigley, 1996). This is the case in irritable bowel syndrome, intestinal obstruction, paralytic ileus, and bowel ischemia. The main problem in monitoring intestinal activity is the difficult anatomic access, hence most methods of studying this activity are considered to be invasive. One possible solution would be the recording of intestinal myoelectrical activity on abdominal surface. This signal is named Electroenterogram (EEnG) and it is composed of two waves: slow waves (SW) and spike burst (SB). The former are periodical, omnipresent electrical potentials that regulate the maximum rate of intestinal muscle contractions. The latter are fast action potentials which are located in the plateau of the SW. They are only present when contractions appear. Whereas SW are related to the frequency and propagation velocity of the contractions (Weisbrodt, 1987), SB determine the presence and the intensity of the contractions. The frequency of the SW changes along the small intestine from about 12 cpm at duodenum to 8 cpm at ileum (Fleckenstein and Oigaard, 1978)

There are few studies about abdominal surface recordings to identify the EEnG in humans (Chen et al., 1993); (Chang et al., 2006); (Prats-Boluda et al., 2007); (Prats-Boluda et al., 2011). The main reason is that human EEnG is a very weak signal, which is severely attenuated especially in the SB frequency range, because of the insulation effects of the abdominal layers and spatial filtering (Garcia-Casado et al., 2006). Surface EEnG is also very sensitive to physiological interferences such as ECG and respiration, being difficult to identify the SW component of the EEnG by visual inspection of abdominal surface recordings. The ECG spectral frequency range overlaps the SB frequency range, therefore it is necessary to eliminate it from abdominal recordings to identify the SB component of the EEnG (Garcia-Casado et al., 2006). As regards to respiration interference, the typical breathing frequency range (12cpm to 24cpm) is very close to the frequency of the SW (8cpm to 12cpm), so it is not possible to use conventional filters to remove this interference.

Laplacian recordings of the EEnG by means of active concentric ring electrodes on rigid substrates have proven to enhance signal quality in comparison to conventional monopolar and bipolar recordings with disc electrodes. Therefore, a second objective of this work is to test and study the possible benefits of the flexible concentric ring electrodes to be developed in this study, on the surface recordings of the EEnG.

2 MATERIAL AND METHODS

2.1 Active Electrode Array Design and Implementation

2.1.1 Sensing Part

In this work it has been decided to design an array of electrodes rather than an individual electrode for surface bioelectrical recordings, since this kind of recordings are usually multichannel and moreover laplacian recordings are often used for body surface mappings. Specifically an array of three concentric ring TCB active electrodes was developed. The sensor is made out of two parts: a disposable sensing part with three TCB electrodes and a reusable battery-powered signal conditioning circuit. Each of the three sensing electrodes consist of an inner disc and two concentric rings in bipolar configuration i.e. the disc and the outer ring are shorted together (TCB).

The outer diameter of the external ring was set to 24mm which is a compromise between bigger electrodes that would yield signals of higher amplitude and smaller electrodes that would provide better spatial resolution. The rest of dimensions of the electrode are designed considering the following criteria:

- The sum of the areas of the outer ring (Aout) and the inner disc (Ain) should be equal to the area of the middle ring (Amid) so as to provide similar input impedances, improving the common mode rejection ratio.
- The distance between the inner disc and the middle ring should be the same as the distance between the middle and outer rings to reduce common mode interferences.

Other issues such as a minimum recording area of $50mm^2$ and the limitations of the implementation technique were considered. The dimensions of each ringed-electrode of the array are shown in figure 1. As it can also be appreciated in this figure, an opened-ring version has been designed in order to avoid shorts in the layout of the traces from the electrodes to the connectors. The flexible electrode array was implemented by screen-printing technology on polymer substrates. Specifically, a biocompatible silver paste was printed (Dupont 5064 Silver conductor, thickness 17 µm) on Polyester Melinex ST506 substrate (thickness 175µm). The serigraphy was made by using an AUREL 900 High precision screen stencil printer. Cured period of inks was 130ºC for 10 minutes.

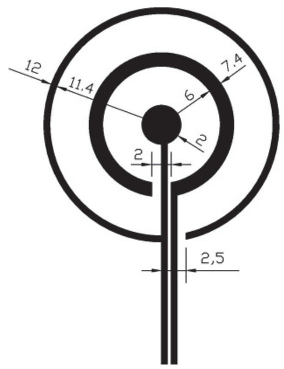

Figure 1: Dimensions (mm) of the concentric ring electrode to be implemented in the array.

2.1.2 Signal Conditioning Part

As stated before, signals from concentric ring electrodes are of very low amplitude, especially in the cases that the bioelectrical signal to be recorded on the body surface is weak. Therefore it is highly recommended to include an amplification stage as close as possible to the sensing electrode.

In this work a battery-powered conditioning circuit was developed and directly connected to the electrode array. Precisely, a 12 bias (only six are used) flexible-flat-cable-to-flexible-printed-circuit connector (TE Connectivity/AMP-1-84953-2 FFC/FPC) was used for the connection. The circuit is composed of a preamplifier (gain 31.9), followed by a coupling circuit (high pass cut off frequency 0.05Hz) and an additional differential amplification stage (gain 106.1) for each of the three TCB electrodes of the array. Specifically, the integrated circuits used were 3 OP747 for the operational amplifiers and 3 AD620 for the instrumentation amplifiers. The signal conditioning circuit weights less than 15 grams. Its main electrical characteristics were experimentally checked and are shown in the next section.

2.2 Signal Recordings

Five recording sessions, of about three hours, were carried out in healthy volunteers in fast state (>8h). Subjects were in a supine position inside a Faraday cage. Firstly the abdominal body surface was exfoliated to remove dead skin cells to reduce contact impedance. The abdominal surface was also shaved in male subjects.

Figure 2, shows the location of electrodes for the EEnG recordings. The developed flexible electrode array was placed horizontally 2.5 cm below the umbilicus, providing three laplacian signals. Three monopolar Ag-AgCl floating electrodes of 8mm of sensing diameter were placed 2.5cm above the umbilicus. Interelectrode distance was also 2.5cm. Two bipolar recordings of EEnG were obtained from adjacent monopolar electrodes.

The main sources of physiological interferences usually present on surface EEnG were simultaneously recorded. Specifically, ECG was monitored by Lead 1 using disposable electrodes; respiration was recorded by an airflow transducer (1401G Grass), and movements were measured by means of 3-axis accelerometer (ADXL 335).

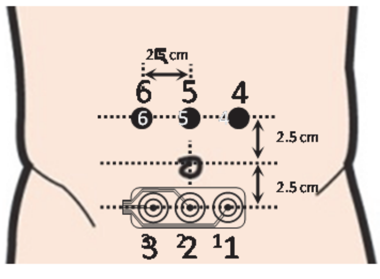

Figure 2: Location of electrodes for EEnG recordings.

All signals, except from acceleration signals, were amplified and band-pass filtered (0.1-100Hz) by means of commercial bioamplifiers (Grass P511). A disposable electrode placed on the left ankle of the subject was used as reference for the bioelectrical recordings. Signals were simultaneously recorded at a sampling rate of 1kHz.

2.3 Signal Analysis

In order to study the activity of the low-frequency component of the EEnG i.e. the slow wave, EEnG and respiratory signals were low-pass filtered (fc=0.5Hz) and resampled at 4Hz.

The power spectral density (PSD) of signals was estimated by means of autoregressive parametrical techniques (AR, order 120). PSD was estimated for moving windows of 120s every 15s of the recorded signals. The dominant frequency (DF) over 8 cpm of the PSD of every window was determined. The parameter *%Resp* was defined as the ratio between the number of windows in which the DF of the surface signal (bipolar or laplacian) is within the DF of respiration ±0.5 cpm and the total number of windows. Similarly, *%TFSW* is defined as the ratio of analysed windows whose DF is inside the typical frequency range of intestinal slow wave (8-12 cpm). The rest of cases are included in the parameter *%Other*.

3 RESULTS

3.1 Active Electrode Array

Figure 3 shows the sensing part of the array of active concentric ring TCB electrodes. It can be appreciated that the substrate is flexible enough to fit the body surface curvature. Moreover the adhesion of the conductor paste to the substrate was checked by means of a sticky tape (8915 Filament APT, 3M). The paste took off after more than 30 cycles proving the good adherence.

Both sides of the signal preconditioning circuit of the active electrode array can be seen in figure 4. The small size and weight of the circuit and the flexible nature of the array makes it possible to place this part above the electrodes. With the proper fixing strategy, the active electrode array could be used for ambulatory monitoring. Table 1 summarizes the main electrical characteristics of the developed signal conditioning circuit. It can be observed that the battery life is adequate for the recording sessions, and the CMRR and output noise are also appropriate for bioelectrical applications.

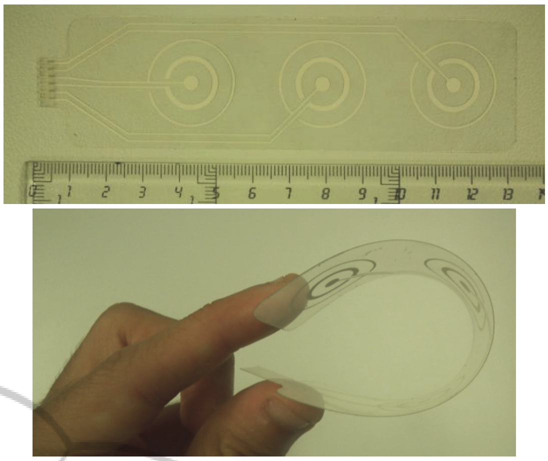

Figure 3: Implemented flexible array of three TCB concentric ring electrodes.

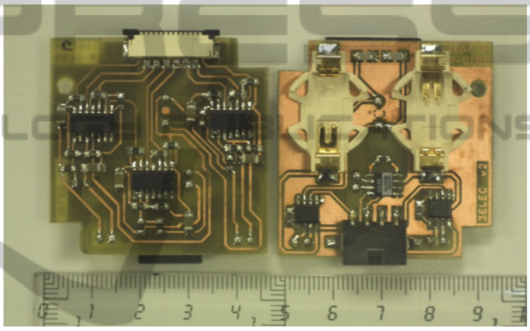

Figure 4: Signal preconditioning circuit: bottom side (left) and top side (right).

Table 1: Main electrical parameters of the signal-preconditioning circuit.

Cut-off frequency of high pass filter	0.049 Hz
Differential gain at medium freq.	3386 V/V
CMRR at medium frequencies	116 dB
CMRR at 50 Hz	103 dB
Output noise	0.195 mVrms
Battery life	280 min.

3.2 EEnG Monitoring

Figure 5 shows an example of the biosignals simultaneously recorded. It can be appreciated that, as expected, the amplitude of the signals picked up by the ringed-electrodes of the array is smaller than that of the bipolar recordings with disc electrodes. Nevertheless, the conventional bipolar recordings present stronger ECG interference as it can be easily observed in this figure. In the signals from concentric ring electrodes the electrocardiographic interference is almost null. It can only be hardly appreciated in the signal corresponding to electrode 1 (Lp1) which is placed on the left side of the subject. Regarding the intestinal SW activity, approximately five waves can be identified on the laplacian recordings. In bipolar recordings this is difficult to identify by visual inspection since it seems they are strongly corrupted by the respiration. This can also be observed in the example of PSD of signals shown in figure 6. In bipolar recordings the dominant frequency (DF) corresponds to the respiratory frequency; whereas the DF of laplacian recordings is around 10.5 cpm which fits the normal SW frequency in human jejunum.

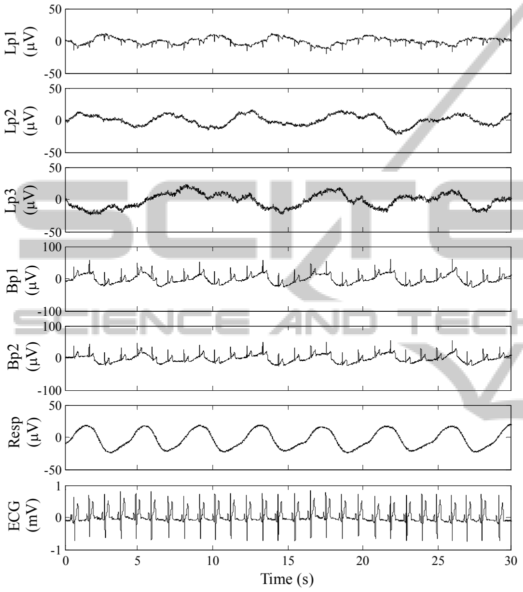

Figure 5: Simultaneous recording of biosignals: Lp1-3: laplacian signals from the electrode array; Bp1-2: bipolar signals from disc electrodes; Resp: respiration; ECG: electrocardiogram

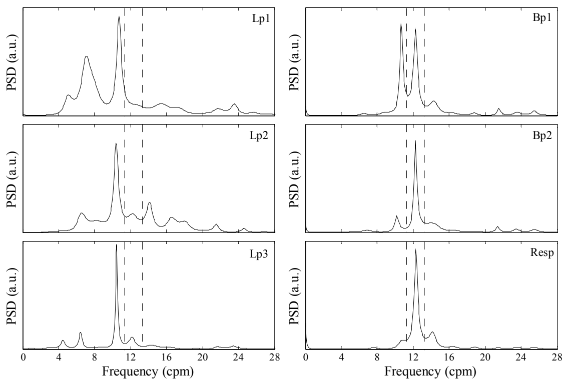

Figure 6: Example of power spectrum density (AR120) of a 120s window of recorded biosignals.

The results of the *%TFSW*, *%Resp* and *%Other* parameters, which are presented in table 2, confirm this behaviour. It is shown that in around 25% of the signal windows studied, the respiratory interference masks the intestinal SW activity. In contrast, this ratio is only around 10% for the laplacian recordings. Furthermore, around 75% of the cases the DF of laplacians recordings the DF is in the frequency range associated to the intestinal SW, whereas it is about 65% for the conventional bipolar recordings. Finally to say that the number of cases whose DF is associated to SW armonics or other components is slightly higher in laplacian recordings than in conventional bipolar recordings.

Table 2: Percentage of dominant frequency in the bandwidths of the different components (mean±standard deviation); N=3385, Lp1-3: laplacian signals from the electrode array, Bp1-2: bipolar signals from disc electrodes.

Channel	%TFSW	%Resp	%Other
Lp1	76,6±8,6	11,3±5,6	12,0±5,8
Lp2	72,8±8,2	9,5±7,0	17,7±6,2
Lp3	71,2±6,9	11,6±8,0	17,2±3,8
Bp1	63,0±12,1	27,3±12,0	9,7±4,8
Bp2	65,6±13,8	25,2±13,8	9,2±6,0

4 DISCUSSION

To the authors' knowledge, the flexible array of active concentric ring electrodes presented in this paper is the first one of these characteristics. Other authors have developed active concentric ring electrodes but on rigid substrates (Li et al., 2005); (Prats-Boluda et al., 2011); (Koka and Besio, 2007). This new sensor is more comfortable for the subject under study and provides a better contact since it adapts to the body surface curvature. Our group has recently developed other flexible concentric ring electrodes (Prats-Boluda et al., 2012). However such electrodes require the screen-printing of three layers, alternating conductor and dielectric pastes. In flexible substrates it is very complicated to use bias between layers, and the solution proposed in the present work favours an easier manufacturing. Moreover, in contrast to individual electrodes (Prats-Boluda et al., 2012); (Li et al., 2005); (Koka and Besio, 2007), the electrode array developed in this work is a more compact solution that reduces the signal preconditioning cost and space, and it is more suitable for bioelectrical mapping of the body surface. Furthermore, the modularity of the developed sensor permits to reuse the signal conditioning circuit while the sensing part can be disposed for hygienic reasons.

Signal recording experiences of this work show that active concentric electrodes of the flexible array enhance the quality of non-invasive EEnG signals in terms of electrocardiographic and respiratory interferences, in comparison to bipolar recordings with conventional disc electrodes. This is in agreement with previous studies that used this kind of electrodes implemented on rigid substrates (Prats-Boluda, 2011). On one hand, the reduction of respiratory interference permits to identify more easily the activity of intestinal slow wave. On the other hand, the reduction of ECG interference could help the identification of spike bursts activity. This could provide more robust systems to non-invasively monitor intestinal myoelectrical activity which could bring close the clinical application of this technique. Nevertheless, this should be confirmed in future studies.

Moreover, although it has not been tested in this work, according to other authors (Besio and Chen, 2007); (Soundararajan and Besio, 2005) the laplacian potential mapping can enhance spatial sensibility for surface bioelectrical activity. This can be of great importance for the studies of propagation maps of cardiac, electroencephalographic or uterine activity which can provide electrophysiological information of clinical relevance. The developed flexible array of active concentric ring electrodes would be very suitable for these applications.

5 CONCLUSIONS

The flexible array of active concentric ring electrodes developed in this paper joins the benefits of laplacian techniques in terms of enhancing spatial resolution, with the comfort and adaptation to body surface curvature of conventional disposable electrodes.

The non-invasive recordings of intestinal myolectrical signals with this new kind of electrodes provide enhanced bioelectric signals in terms of robustness to physiological interferences such as ECG and respiration, and permit to identify more easily the intestinal slow wave activity.

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