Active flexible concentric ring electrode for non invasive surface bioelectrical recordings

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Abstract:

Bioelectrical surface recordings are usually performed by unipolar or bipolar disc electrodes even though they entail the serious disadvantage of having poor spatial resolution. Concentric ring electrodes give improved spatial resolution, although this type of electrode has so far only been implemented in rigid substrates and as they are not adapted to the curvature of the recording surface may provide discomfort to the patient. Moreover, the signals recorded by these electrodes are usually lower in amplitude than conventional disc electrodes. The aim of this work was thus to develop and test a new modular active sensor made up of concentric ring electrodes printed on a flexible substrate by thick-film technology together with a reusable battery-powered signal-conditioning circuit. Simultaneous ECG recording with both flexible and rigid concentric ring electrodes were carried out on 10 healthy volunteers at rest and in motion. The results show that flexible concentric ring electrodes not only present lower skin-electrode contact impedance and lower baseline wander than rigid electrodes but are also less sensitive to interference and motion artefacts. We believe these electrodes, which allow bioelectric signals to be acquired non-invasively with better spatial resolution than conventional disk electrodes, to be a step forward in the development of new monitoring systems based on Laplacian potential recordings.

1. Introduction

Bioelectrical activity is a spatio-temporal process which is spatially distributed over the three dimensions of the organ system and evolves in time. It is therefore of great importance to interpret bioelectrical signals not only in time but also in the space domain, which consists of localizing any bioelectrical sources which may be picked up by non-invasive electrical recordings such as electrocardiograms (ECG) and electroencephalograms (EEG). Bioelectrical signals are generally recorded non-invasively by disc electrodes in a bipolar or unipolar configuration. In the former method the potential difference is measured between a pair of electrodes and in the latter the potential of each electrode is compared either to a neutral electrode or to the average of all electrodes [1]. 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 [2-5]. To improve spatial resolution in the field of electrocardiography, body surface potential maps have been obtained by extending the standard 12-lead ECG to include a larger number of recording points covering the body surface [6]. However, the spatial resolution of body surface potential mapping is still limited due to the smearing effect caused by the torso volume conductor, which could not be resolved by simply increasing the number of body surface recording electrodes [7]. In this respect, considerable efforts have been made to study the feasibility of body surface Laplacian electrograms to localize bioelectrical sources [8, 9, 10].

Theoretically, the Laplacian of surface potential is proportional to the derivative of the component of current density orthogonal to the body surface and can be interpreted as a filter that allocates more weight to the bioelectrical dipoles adjacent to the recording points [11]. The 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 [8, 11, 12, 13]. Laplacian methods were first applied to bioelectrical recordings by Hjorth in 1975 in order to increase the spatial resolution achieved in electroencephalographic (EEG) recordings [14]. In this study the Laplacian of the EEG signal was estimated using the surface potential picked up by five disc electrodes arranged in the form of a cross. This technique is known as the five-point method. Since then, other discretization techniques have been developed to estimate Laplacian bioelectrical potentials, including the nine-point method [12]. In the late 80s, analytic solutions were proposed to estimate the Laplacian of the surface potential in order to reduce discretization errors [15-17]. However, these are complex discrete computational techniques and are generally not suitable for real-time applications.

In addition to discrete and analytic methods, the Laplacian potential can also be directly recorded from the body surface using concentric ring electrodes [11, 12], which act as spatial filters reducing low spatial frequencies and increasing spatial selectivity [18]. This reduces mutual information and alleviates orientation problems better than disc electrodes [9]. Bioelectrical signals recorded by concentric ring electrodes on the body surface are very weak in amplitude and present high output impedance, so it is advisable for them to be conditioned (preamplified and filtered) as close as possible to the recording surface [9, 10, 18]. Lu and Tarjan developed an active Laplacian ECG sensor with a rigid tripolar concentric ring electrode in quasi-bipolar configuration (TCB, in which the outer ring and the centre disc were electrically shorted) to study the feasibility of applying this type of electrode to detecting cardiac arrhythmia in real time [19]. Furthermore, activation patterns have also been detected by Laplacian potential maps using ring electrodes [20-21]. Active TCB electrodes have been used to estimate the Laplacian potential of other bioelectrical signals such as the electroencephalogram (EEG), electroenterogram (EEnG) and the electrohysterogram (EHG), so as to increase the spatial resolution of conventional surface potential recordings [9, 10, 22].

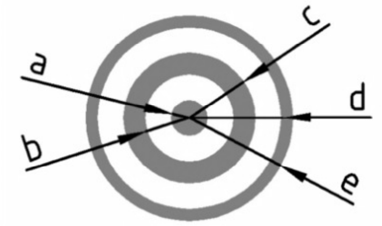
However, in spite of its numerous advantages the clinical application of Laplacian techniques based on concentric ring electrodes is still limited, mainly due to the fact that these electrodes are implemented on rigid substrates, which neither adequately adapt to the body surface contours nor offer an acceptable comfort level in out-patient applications. On the other hand flexible substrates, which are widely used in electronic engineering, would be more adaptable to body contours than rigid electrodes and would potentially improve and stabilize skin-electrode contact impedance [23-25]. Flexible concentric ring electrodes would thus combine the comfort of existing disposable electrodes with high spatial sensitivity.

The goal of this work was therefore to develop and test a new modular active sensor containing disposable concentric ring electrodes printed on a flexible substrate connected to a battery-powered signal conditioning circuit and to compare it to the performance of rigid conventional concentric ring electrodes.

2. Material and methods

2.1 Active flexible electrode design & development

In order to analyze the effect of the electrode dimensions on its response, tripolar electrodes were designed in bipolar configuration (TCB) in two different sizes: 18 mm (TCB18) and 24 mm (TCB24) external diameter. The other electrode dimensions were calculated to comply with the following requirements (see Figure 1):



	Radius				
Type	a (mm)	b (mm)	c (mm)	d (mm)	e (mm)
TCB24	2	5.5	7.53	11.03	12
TCB18	2	4.5	5.9	8.4	9

Figure 1. Tripolar concentric ring electrode dimensions.

a) Considering that the common mode input impedance of the signal conditioning system is balanced, it is advisable to ensure that the electrode impedance is also balanced, which is possible if the areas of the electrode leads are equal. In a TCB electrode, in which the internal disc and the outer ring are shorted, the area of the inner disc plus the area of the outer ring may be approximately equal to the area of the middle ring (1). A minimum area of 50 mm^2 was set to ensure signal acquisition.

$$A_1 = \pi \cdot \left(c^2 - b^2\right) \cong A_2 = \pi \cdot a^2 + \pi \cdot \left(e^2 - d^2\right) \geq 50mm^2 \quad (1)$$

b) The distance (D) between the inner disc and middle ring must be equal to the distance between the middle and outer rings in order to attenuate common mode interference:

$$D = b - a = d - c \quad (2)$$

c) The minimum width of the rings was set at 0.6 mm to avoid continuity problems during implementation. For the same reasons a minimum radius of 2 mm was chosen for the central disc.

d) Considering that the greater the distance between registration points, the greater the amplitude of the differential electric potential, the dimensions will be selected that optimize the value of the product of the square of the distance between conductors (D^2) by the conduction area (A_1).

According to these criteria, the possible designs of TCB24 (24 combinations) and TCB18 electrodes (8 combinations) were generated, these being the dimensions that optimised product ($D^2 \bullet A_1$), as shown in Figure 1. The rigid and flexible electrodes used for the ECGs recordings were given these dimensions.

The flexible electrodes were produced by screen-printing technology by means of a three layer design as shown in Figure 2. The first layer was made up of a silver conductor in which the central dot and outer ring were short circuited, with the additional function of containing the two connection lines. The area of both poles was matched by adjusting the semi-arc length. The second layer is insulating and prevents the intermediate disc from being short-circuited by the connection lines that join the inner disc and the outer ring. The third layer was made of a silver conductor and consisted of three concentric rings in contact with the subject’s skin. A 200 mesh screen was used for each layer. The serigraphy was made by using an AUREL 900 high precision screen stencil printer. The curing period of inks was 130ºC for 10 minutes. Conductive and dielectric biocompatible screen-printing pastes were chosen that presented good adherence to polymeric substrates (Dupont 5064 Silver conductor with resistivity $\leq$ 10 mΩ/sq/25μm and Dupont 5036 Heat seal with Insulation Resistance > 10 GΩ/sq/mil and Dielectric Constant (@ 1kHz ASTM D150) < 5).

Tests were carried out on different polymeric substrates (Valox FR-1, Polyester MelinexST506 and Ultem R16SG00) to select the one that offered the best results relative to electrode flexibility and adhesion to conductive and insulating pastes, as well as low skin-electrode impedance. The strength of the adherence of the pastes to the different polymeric substrates (Valox FR-1, polyester MelinexST506 and Ultem R16SG00) was assessed by means of sticky tape (8915 Filament APT 3M). The skin-electrode contact impedance of each electrode pole was measured in 10 volunteers using the General Devices EIM 105 impedance meter, which is based on three electrode impedance measurement techniques. For this, two additional disposable electrodes were placed on both shoulders of the subjects.

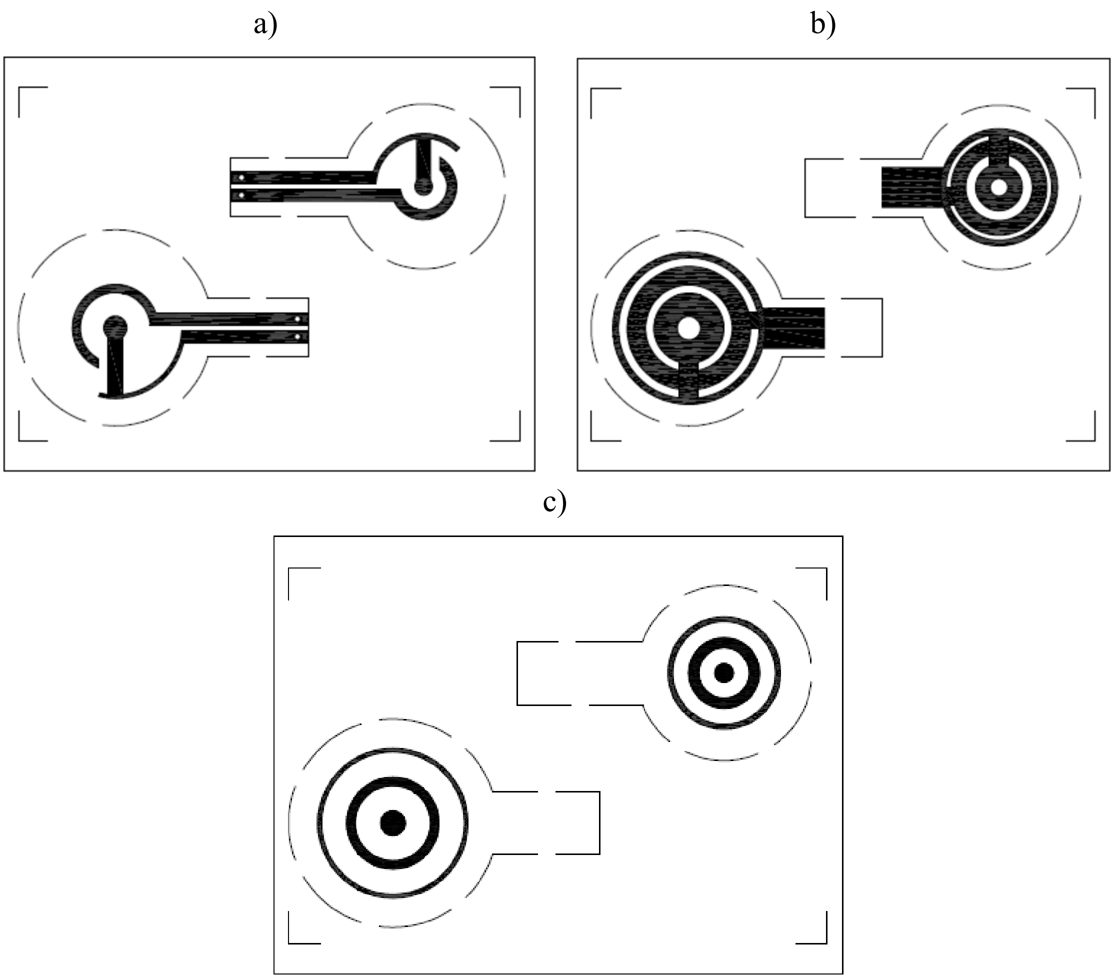


Figure 2. Screen patterns used in the final printing of 2 TCB electrodes. a) 1st layer: composed of conductive Ag paste; b) 2nd layer: composed of dielectric paste; C) 3rd layer: composed of conductive Ag paste. This is the outer layer of the electrode and is in contact with the subject's skin.

Fifteen TCB24 and fifteen TCB18 flexible electrodes, developed using the substrate that offered the best results, were assayed in the ECG recordings as detailed in the following section. The flexible concentric ring electrodes response was also compared to that of rigid ring electrodes with the same dimensions. Fifteen rigid ring electrodes of each size (TCB24 and TFCB18) were implemented on a conventional FR4 substrate.

As has been previously mentioned, flexible and rigid ring electrodes were directly connected to a battery-powered conditioning circuit with a low cut-off frequency of 50 mHz and a total gain of 3384 V/V.

This circuit is similar to that presented in a previous paper of the present research group [10]. The main characteristics of the signal-conditioning circuit of the sensor were experimentally checked, as shown in Table 1. It can be observed that the battery life (>12 h) was adequate for the recording sessions and the CMRR -118 dB at medium frequencies, 100 dB at 50 Hz- and output noise values -0.186 mVrms- were also appropriate for bioelectrical applications.

Table 1. Conditioning circuit main parameters

Parameter	Value
Cut-off frequency of the high pass filter	0.05 Hz
Differential gain at medium frequencies	3384 V/V
CMRR at medium frequencies	118 dB
CMRR at 50 Hz	100 dB
Battery life	750 min
Output noise	0.186 mVrms

The flexible and rigid electrodes were tested before carrying out the protocol described in the next section. Firstly, the electrical continuity of all electrodes was checked. After that, a preliminary inter-electrode reproducibility test was performed on one volunteer: chest skin was prepared and 1 minute ECG signal was monitored for the 15 electrodes of each type (TCB24 flexible, TCB24 rigid, TCB18 flexible and TCB18 rigid. For the same electrode type changes in ECG morphology were not found and ECG amplitudes were also similar.

2.2 ECG signal recording

Ten recording sessions were carried out on 10 healthy volunteers in a supine position. A Laplacian ECG (LECG) recording protocol was established so as to minimize the high dependence of the electrode position on the bioelectrical activity acquired: firstly, the two chest locations on which the electrodes were to be placed, 6 cm over and 2 cm to the right of the left nipple (upper position) and 6 cm below the nipple and aligned with the other electrode (lower position), were exfoliated and cleaned to remove dead skin cells and reduce contact impedance. A flexible TCB24 sensor was then placed in the upper position (U) and a rigid TCB24 sensor in the lower position (L). The contact impedance between electrodes and skin was then measured. Next, ECG was recorded simultaneously during a 5-minute period with the subject at rest. ECG was then recorded during lateral head movement, with a one-minute rest, vertical arm movement, one minute rest, vertical leg movement, one minute rest and laugh (Protocol 1). Afterwards, Protocol 1 was repeated changing the position of the TCB24 electrodes, rigid (U) and flexible (L) (Protocol 2). Next, similar tests were carried out using TCB18 electrodes (Protocols 3 and 4). See Figure 3. For each recording session a new set of electrodes (one TCB24 flexible, one TCB24 rigid, one TCB18 flexible and one TCB18 rigid) was used and the same TCB24 flexible and TCB24 rigid electrodes were changed from upper to lower position and *vice versa* (protocol 1 and 2) and the same TCB18 flexible and TCB18 rigid electrodes were used in protocol 3 and 4.

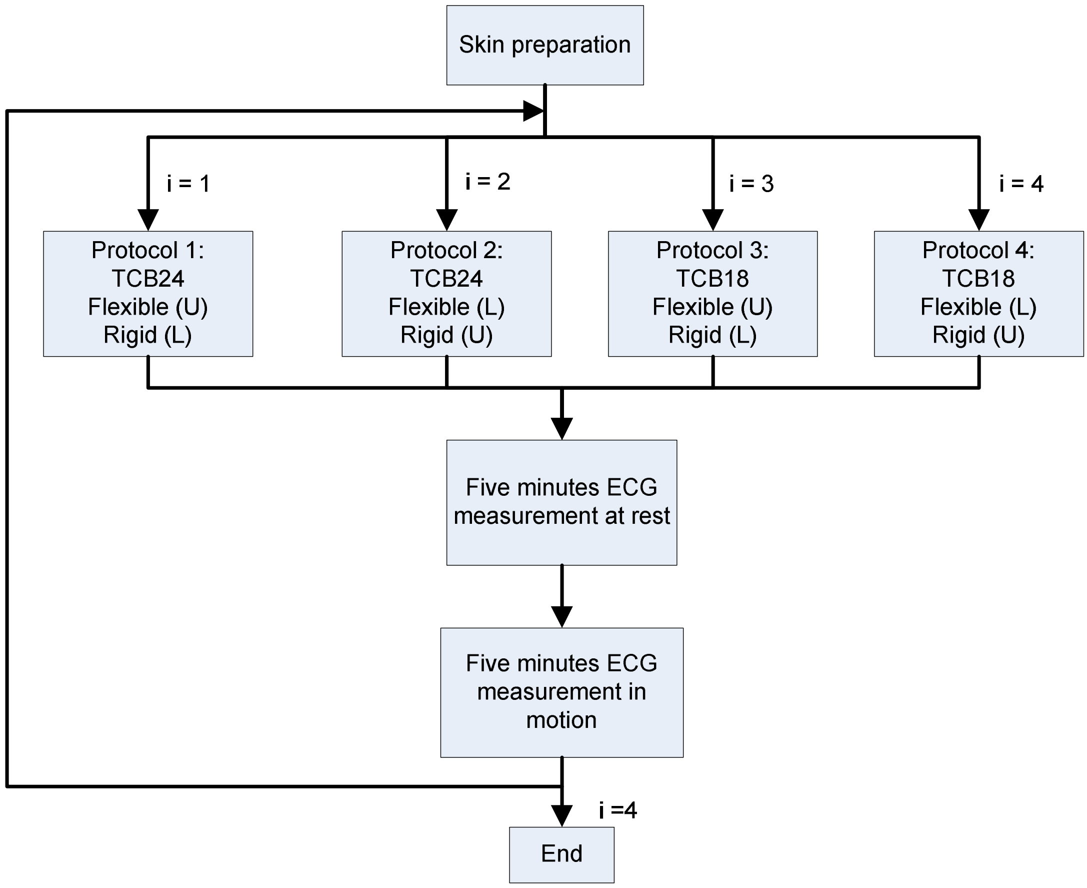


Figure 3. Diagram of ECG recording protocol.

The ECG signals picked up by the ring electrodes, and preamplified by the above mentioned battery-powered signal conditioning circuit, were connected to commercial instrumentation amplifiers (Grass Technologies P511). P511 amplifier gain was set to 10 and its low and high cutoff frequencies were set at 0.05 Hz and 100 Hz, respectively. A third electrode (virtual ground) was placed on the subject's right hip and connected to the ground of the commercial and the battery-powered amplifiers. The sampling rate was 1 kHz for all signals.

2.3 ECG parameters

ECG fiducial points were obtained by detecting the R-wave of the ECG signal with the algorithm proposed by Pan and Tompkins (1985) and slightly modified by Hamilton and Tompkins (1986) [26]. A fifth-order Butterworth high pass filter with a cut-off frequency at 0.1 Hz was applied to surface signals, giving rise to a filtered signal x(t). Then the averaged beat, $\overline{ECG}_{rest}$, extending from 275 ms prior to the R-wave to 450 ms after it, was calculated in analysis windows of 60 s.

In order to compare the surface signals recorded by rigid and flexible active concentric ring electrodes at rest, the following parameters were defined and worked out.

- The root mean square voltage of the averaged beat at rest ($\overline{ECG}_{rest}$): this value shows the amount of energy contained in the acquired ECG signal.
- Signal to noise ratio (SNR_{rest}). ECG is contaminated by different types of interference such as power line interference and muscle noise. Once the $\overline{ECG}_{rest}$ has been worked out, then an estimated target ECG signal ($ECG_{rest}(t)$) is generated by assigning the $\overline{ECG}_{rest}$ to each detected beat, considering a window length about the R wave, as mentioned above. Zero value was assigned outside this window. The noise embedded in the recording was therefore calculated as the difference between the filtered surface signal, x(t), and the estimated target signal $ECG_{rest}(t)$. Then SNR_{rest} was calculated as:

$$SNR_{rest}(dB) = 20 \cdot \log\left(\frac{Vrms\left(\overline{ECG}_{rest}\right)}{Vrms\left(x_{rest}(t) - ECG_{rest}(t)\right)}\right) \tag{3}$$

The parameters studied for motion ECG recordings were:

- SIR_{motion}. Defines the relationship between the Vrms value of the averaged beat at rest $\left(\overline{ECG}_{rest}\right)$ and the effective voltage of the raw surface signal during movement (ECG_{motion}(t)) expressed in decibels (dB) and quantifies the change in the energy content of the ECG signal between rest and motion.

$$SIR_{motion}(dB) = 20 \cdot \log\left(\frac{Vrms\left(\overline{ECG}_{rest}\right)}{Vrms\left(ECG_{motion}(t)\right)}\right) \tag{4}$$

- Saturation percentage. Motion can cause loss of contact between electrodes and skin, which may entail the saturation of the amplifiers. This parameter is defined as the proportion of time during motion during which the amplifiers were saturated.

3. Results

3.1 Active flexible electrode

Types TCB24 and TCB18 flexible ring electrodes were developed on different substrates (Melinex, Valox or Ultem) and the TCB24 and TCB18 rigid electrodes on FR4 (printed circuit board) substrate. Photographs of Type TCB24 rigid and flexible electrodes can be seen in Figure 4. The flexibility and reduced thickness of the flexible electrodes can be appreciated in Figure 5.

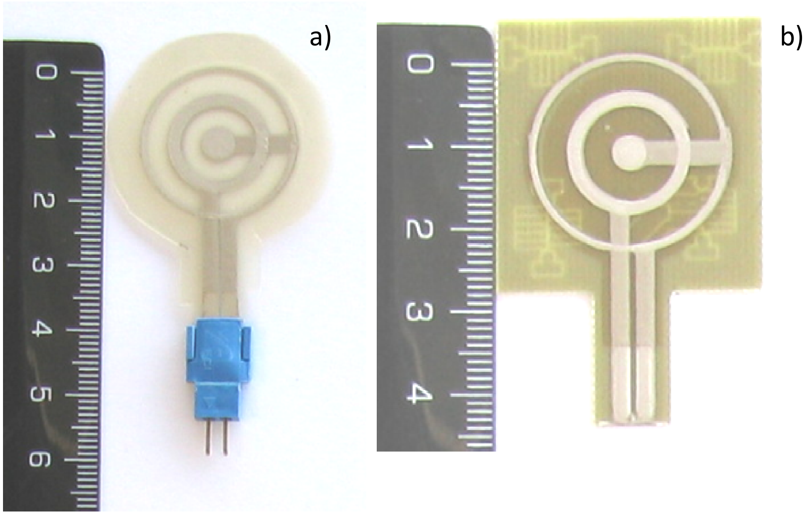


Figure 4. (a) TCB flexible concentric ring electrode and (b) TCB rigid electrode.

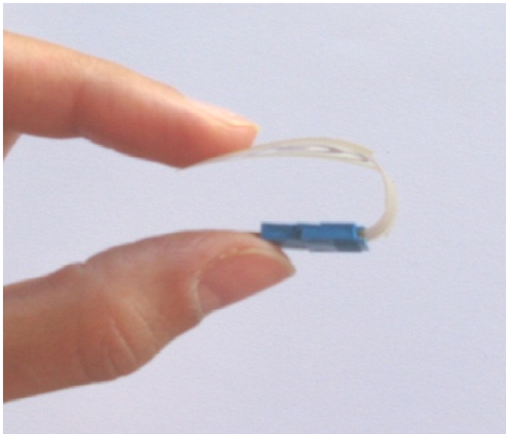


Figure 5. Photograph of the flexible ring electrode showing its thickness and flexibility

Contact impedance and pastes-substrate adherence trials were carried out to select the most suitable flexible substrate for the implementation of the ring electrodes. All the flexible substrates tested showed good adherence with the silver and dielectric pastes: the dielectric paste separated from polyester MelinexST506 after more than 30 cycles, whereas both the conductor and dielectric pastes lasted more than 50 cycles on both UltemR16SG00 and Valox FR-1. Table 1 shows the mean electrode-skin contact impedance and standard deviation for both flexible and rigid electrodes in 10 volunteers. It is noteworthy that all the flexible electrodes presented lower skin-electrode impedance than the rigid ones, which suggests that flexible electrodes give better skin-electrode contact. Based on these tests, flexible electrodes implemented on UltemR16SG00 were selected for further ECG monitoring analysis, due to its ease of production by screen printing and its slightly lower skin-electrode impedance (see Table 2). Nevertheless, it is believed that similar results would be obtained if other flexible assayed substrates were to be used in ECG recordings.

Table 2. Electrode-skin contact impedance values of concentric ring electrodes TCB24 implemented on different substrates measured on the chest. The electrode-skin contact impedance was measured for each substrate in 10 volunteers (6 men, 4 women).

	Electrode-skin Contact impedance (kΩ)
Valox	5.14 ± 2.28
Melinex	5.65 ± 2.79
Ultem	5.09 ± 2.25
Rigid PCB	8.86 ± 3.83

3.2 Laplacian ECG recordings (LECG)

Figure 6 shows 15 s of simultaneous LECG recordings at rest with the rigid ring electrode placed in the upper position and the flexible electrode in the lower position. Heartbeat fiducial points can be clearly distinguished in both LECG recordings, as their morphology is more abrupt than the conventional ECG signal. The LECG amplitude in both recordings is in tens of microvolts (about 30 µV), that considering a mean radius of 6.14 mm for TCB24 electrodes corresponds to 319.2 µV/cm^2), which is similar to the LECG signal amplitude reported by other authors [12, 27].

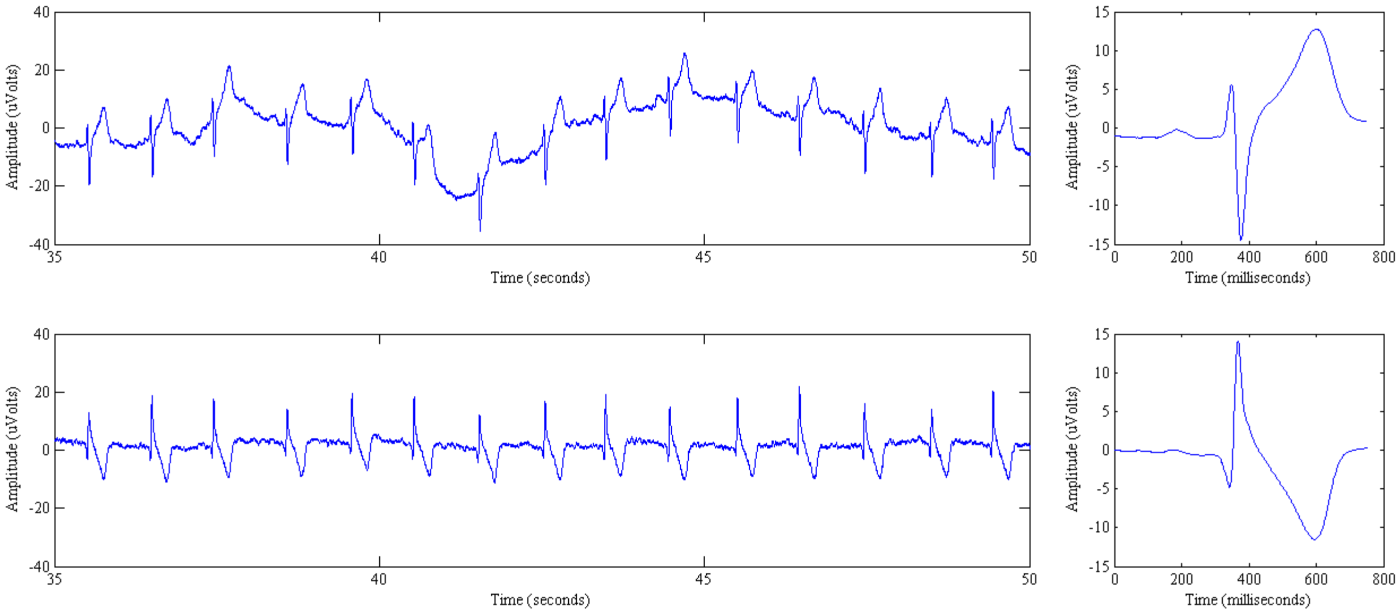


Figure 6. Left: 15 seconds of simultaneous ECG recordings obtained with a concentric rigid TCB24 electrode in the upper position and a flexible TCB24 concentric electrode in the lower position. Right: averaged beat corresponding to 1 minute recording of the chest surface signals shown on the left.

Furthermore, it was observed in all recording sessions that P wave amplitude strongly depends on the electrode position (U or L) no matter the electrode substrate (rigid or flexible) as shown in figure 6: The P wave can be clearly indentified in the averaged beats of the signals picked up by the ring electrodes in the upper position, close to the atrial area, whereas the P wave is much more attenuated in the signals captured by the ring electrodes in the lower position, farthest from the atrial area. It agrees to the fact that laplacian recordings presented high spatial resolution.

It can also be appreciated that the signals recorded by flexible electrodes have lower baseline wander and are less sensitive to low frequency interference than those recorded by rigid electrodes of the same dimensions.

Table 3 shows the average and standard deviation of the signal parameters at rest and in motion obtained from 10 volunteers. Regarding the parameters at rest, In general, both the rms value of the averaged beat at rest $\left(\overline{ECG}_{rest}\right)$and the SNR_{rest} show mean values that are slightly higher for flexible electrodes than that of rigid electrodes of the same size in the same position – except for $\overline{ECG}_{rest}$ of TCB18 electrodes in the lower position –.. However, paired t-tests reveal that these differences are not significant for $\overline{ECG}_{rest}$ (p_{1-2} = 0.59, p_{3-4} = 0.4, p_{5-6} = 0.12, p_{7-8} = 0.42), nor SNR_{rest} (p_{1-2} = 0.29, p_{3-4} = 0.29, p_{7-8} = 0.28), except for TCB24 electrodes in lower position with significant better SNR_{rest} results for flexible electrodes (p_{5-6} = 0.003). As for the influence of electrode size (for the same substrate and recording position) on the parameters during rest, higher mean values of $\overline{ECG}_{rest}$ are obtained in general for TCB24 vs TCB18. Nevertheless, again the differences are not statically significant for this parameter (p_{1-3} = 0.14, p_{2-4} = 0.054, p_{5-7} = 0.31, p_{6-8} = 0.07), neither for the SNR_{rest} (p_{1-3} = 0.63, p_{2-4} = 0.69, p_{5-7} = 0.13, p_{6-8} = 0.13).

Table 3. Values for the rest test parameters ($\overline{ECG}_{rest}$, and SNR_{rest}) and for the motion test parameters (SIR_{motion}, Saturation) corresponding to flexible and rigid electrodes TCB24 and TCB18.

		Upper position				Lower position			
		TCB24		TCB18		TCB24		TCB18	
		Flexible (1)	Rigid (2)	Flexible (3)	Rigid (4)	Flexible (5)	Rigid (6)	Flexible (7)	Rigid (8)
Rest	$\overline{ECG}_{rest}$ (μV)	4.89 ±1.29	4.63 ±1.34	3.79 ±1.62	2.92 ±1.97	4.33 ±1.98	3.25 ±0.54	3.63 ±1.45	4.43 ±1.84
	SNR_{rest} (dB)	10.79 ±3.93	8.82 ±2.94	12.0 ±5.70	9.39 ±5.55	17.59 ±4.17	7.85 ±4.43	13.54 ±7.56	9.64 ±5.0
Motion	SIR_{motion} (dB)	-2.57 ±1.89	-6.64 ±3.52	-3.58 ±3.95	-14.46 ±4.90	-1.80 ±2.23	-5.42 ±3.66	-8.64 ±7.04	-8.58 ±5.17
	Saturation (%)	0.84 ±1.68	5.96 ±2.81	4.38 ± 6.21	13.10 ±10.23	2.76 ±2.59	1.27 ±2.54	13.20 ±10.26	12.27 ±12.67

Figure 7 shows how the rigid electrode motion recordings (lower trace) are more sensitive to motion artefacts: a strong artefact around the second 9 causes saturation of the amplifiers. However, in the top trace in this figure, corresponding to the simultaneous flexible electrode recording, motion only involves a small change in the baseline.

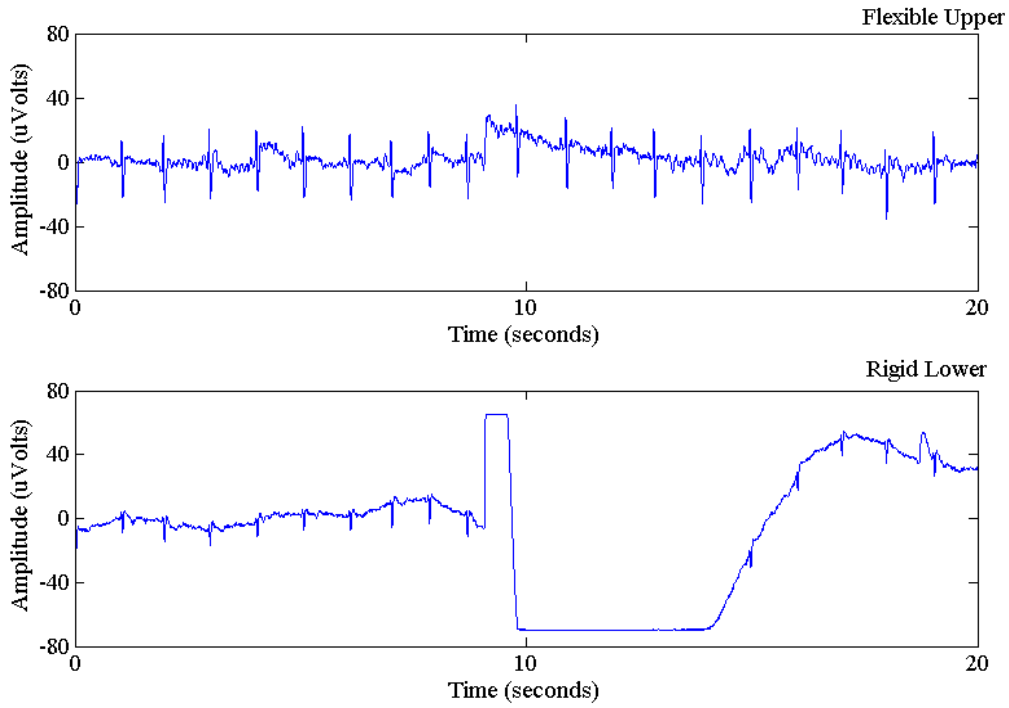


Figure 7. Simultaneous ECG recorded in motion with a concentric rigid TCB24 electrode in the lower position and a flexible TCB24 concentric electrode in the upper position.

This is also reflected in the parameters that quantify the behaviour of ring electrodes in motion recordings (see Table 3): in general flexible ring electrodes have significantly higher values for the SIR_{motion} parameter than rigid ones (paired t-tests, $p_{1\text{-}2} = 0.02$, $p_{3\text{-}4} = 1.48 \cdot 10^{-5}$, $p_{5\text{-}6} = 1.47 \cdot 10^{-4}$, $p_{7\text{-}8} = 0.59$), which indicates that they are less sensitive to motion artefacts. As for electrode dimension, best results correspond to TCB24 electrodes. Likewise saturation percentage in motion, in upper position ECG recordings, is significantly lower for flexible ring electrodes compared to rigid ring electrodes (paired t-test, $p_{1\text{-}2} = 0.003$, $p_{3\text{-}4} = 8.09 \cdot 10^{-4}$) but without significant differences for lower position ECG recordings. Finally, saturation percentage for TCB24 electrodes is significantly lower than that of TCB18 (paired t-tests, $p_{1\text{-}3} = 0.03$, $p_{2\text{-}4} = 3.27$, $p_{5\text{-}7} = 0.02$), except for rigid electrodes in lower position ($p_{6\text{-}8} = 0.12$).

4. Discussion

4.1 Active flexible concentric ring electrode

Active flexible concentric ring electrodes were developed and implemented on flexible substrates including a preconditioning and amplification system. The aim was to develop electrodes that adapt to the patient's body contours, allowing non-invasive surface bioelectric recordings to be obtained with a combination of comfort, high spatial sensitivity and good signal-to-noise ratio. It is noteworthy that other research groups have developed and used active concentric ring sensors to increase spatial resolution in the acquisition of surface bioelectric potential, such as in electrocardiograms, electroenterograms and the electrohysterogram [9, 10, 12, 28]. However all these sensors were developed on rigid substrates and were used in research but are not suitable for clinical or out-patient applications.

The multilayer screen-printing technique was chosen to produce the electrodes due to the fact that it provides them with the optimum homogeneous finish, significantly thicker than that obtained by electroplating. Screen-printing is also highly reproducible and facilitates serial production. The use of graphic art printing technologies in the electronic industry is making it possible to develop new types of electrodes on flexible substrates and at present alternative or complementary technologies are being introduced, as the inkjet or gravure printed, which can make the manufacture in mass of this type of electrodes easier [29, 30].

On the other hand, an important feature of the developed sensor is its modularity, which means that the electrodes can be interchanged while maintaining the same signal conditioning system. This feature is especially important in clinical applications with regard to hygiene standards and helps to reduce operating costs.

In addition, the proposed flexible ring electrodes could be used in long-term recordings, avoiding the gel dehydration problems involved in using wet electrodes. In the last decade, research groups have made a considerable effort to develop conductive and capacitive electrodes for dry ECG recordings, most of them focused on the acquisition of unipolar ECG signals [31, 32]. One of the major problems of bioelectrical surface recordings with dry electrodes is the high skin-electrode contact impedance involved, giving rise to surface recordings with low signal-to-noise ratio, high baseline wander and high sensitivity to motion artefacts. In order to reduce the skin-electrode contact impedance in the present study, Ag pastes were used to implement the ring electrodes, since the low frequency impedance of Ag paste is lower and more stable than that of other materials such as Ag/AgCl, aluminium or steel used for dry recording electrodes. The noise level present in the LECG recordings in this study is similar to that obtained in other studies that used Ag dry electrodes for LECG recordings and is lower than ECG recorded by capacitive $SiO2$ electrodes [31].

The LECG recordings obtained in the present study with flexible electrodes also have less baseline fluctuation and greater immunity to motion artefacts than those obtained from rigid electrodes, thanks to their being able to adjust to body contours, thus providing better contact [31]. However, these electrodes are still more susceptible to motion artefacts than commercial wet Ag/AgCl electrodes. The incorporation of a self-adhesive substrate in contact with the skin, similar to that of the presently available disposable wet electrodes, would exert a constant pressure on both rings, avoiding friction with the skin and minimizing any motion artefacts generated by the skin-electrode interface.

4.2 Laplacian recordings of ECG

It has been proven that Laplacian potential mapping can enhance the spatial sensibility of surface bioelectrical activity recording in respect to conventional biopotential mapping [27, 33]. Non-invasive high spatial resolution mapping of cardiac electrical activity can provide valuable information for clinical diagnosis of a wide range of cardiac abnormalities, including infarction and arrhythmia. It has also been found that the QRS complex can be identified in Laplacian ECG recordings carried out with 36 mm external-diameter active rigid ring electrodes [27, 34]. In the present work we tested the possibility of picking up the heart signal with smaller ring electrodes (TCB18 and TCB24) in order to obtain potential Laplacian maps with higher spatial resolution than those obtainable from 36 mm electrodes.

It is known that the greater the external diameter of ring electrodes, the greater their capacity to pick up deep bioelectrical sources and the higher the signal amplitude obtained, even though this involves a reduction in spatial resolution [35]. We found that the fiducial points of the ECG complex can be identified in surface recordings by TCB18 and TCB24 electrodes without significant differences in the amplitude of the LECG signals due to electrode size. This suggests that smaller electrodes (TCB18) could be used for non-invasive high density surface mapping of cardiac signals with the subject at rest. By contrast, TCB24 electrodes significantly improve on the TCB18 response against motion artefacts (lower saturation percentage). The higher resistance of the TCB24 electrodes to motion artefacts could be due to the better skin-electrode contact obtained from their larger recording area. These results highlight the importance of ensuring good skin-electrode contact in surface bioelectrical recordings.

Regarding the influence of the substrate, as mentioned above, in general flexible electrodes give better performance in terms immunity to motion artefacts than recordings made with rigid ring electrodes, which means that flexible TCB24 electrodes would be suitable for out-patient applications. In this work we tested the response of both rigid and flexible electrodes to slight movements of head, arms and legs. The flexible electrodes performed well against sudden and continuous movements typically found in out-patient applications. This aspect will be further developed in future studies.

The LECG morphology obtained from the ring electrodes recordings was sharper than that associated with the standard ECG, but is similar to the morphology reported by other research groups using discretization techniques or ring electrodes to estimate the ECG Laplacian potential [8, 27, 33, 34, 36]. The LECG amplitudes are much smaller (μV) than that of the standard ECG (mV), and are similar to LECG amplitudes obtained from 36 mm external diameter rigid electrode recordings [27]. The noise level is also similar to that reported by other authors in LECG recordings, which are noisier than the standard ECG recordings acquired from disposable wet electrodes.

It is noteworthy that the high spatial resolution of the ring electrodes is evident in the surface recordings obtained from the upper chest position close to the atrial area. The P wave can be clearly distinguished on the averaged beat, $\overline{ECG}_{rest}$,but in the simultaneous recording from the electrodes in the lower position this component is much more attenuated, picking up mainly ventricular activity. LECG recordings have already been shown to give high spatial resolution in detecting atrial activity patterns [20, 21].

5. Conclusions

This work describes the development of active concentric ring electrodes on flexible substrates using printed electronics technology that combine ease of use and the adaptability to body contours of conventional disposable electrodes with the enhanced spatial resolution of concentric ring electrodes. Flexible ring electrodes provide better skin-electrode contact than ring electrodes on rigid substrates, reduce baseline wander, noise and sensitivity to motion artefacts. We believe the proposed electrodes are an advance in Laplacian-based monitoring systems and will allow bioelectrical signals to be acquired non-invasively with higher spatial resolution than those obtained from conventional disc electrodes.

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