Active Laplacian Electrode for The Data-acquisition System of EHG

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Abstract. EHG (Electrohysterogram) is the recording of uterine electromyogram with external electrodes located on the abdomen of pregnant woman. Derived from the electrical activity generated at the muscle fiber lever, it provides complementary information from the muscle, and appears to be a very promising technique for clinical or physiologic investigation of uterine activity, compared with current monitoring which can’t give us complementary phase information of uterine activity. In this article we have shown the disadvantages of the conventional electrodes for EHG data-acquisition system and put forward a new type of electrode that is called active Laplacian electrode. It integrates concentric rings electrode with a bioelectricity preamplifier and is capable of acquiring localized information. We can localise the EHG signals source more easily by using this new electrode.

1. Introduction

During the last 60 years, uterine activity has been extensively investigated. Conventional clinical diagnosis of uterine contraction activity includes two methods. In many situations, the doctors examined the patients only by touching. This method based on the experience of doctors can’t form a criterion. Pressure measurement is another clinical diagnostic method of uterine activity using medical instruments such as TOCO and IUP. It can show us the frequency and pressure of the whole uterine contraction, not including complementary phase information of uterine activity.

A known attempt to record electromyograms of uterine contraction activity was made by Veit [1]. Then Steer and Hertsch named the uterine electromyograms as EHG in 1950 [2]. There are straightforward relationships between EHG signal and uterine contraction activity. EHG (Electrohysterogram) is the recording of uterine electromyograms with external electrodes located on the abdomen of pregnant woman. Derived from the electrical activity generated at the muscle fiber lever, it provides complementary information from the uterine muscle, and appears to be a very promising technique for clinical or physiologic investigations of uterine activity, compared with current invasive monitoring which can’t give us complementary phase information of uterine activity.

Until now, most of EHG study is based on only single-lead recordings. Since EHG signal has no fixed pace-making areas, single-lead recordings will lose phase information of uterine activity. So
multi-lead EHG data acquisition system is often used in recent research. Unipolar electrode and biopolar electrode are often used in multi-lead recordings. A signal recorded by unipolar electrode can cover the whole frequency stage, but is badly affected by noise from many other signals. While signal recorded by biopolar electrode is the difference voltage of the two electrodes, it would reject power-line interference, baseline excursion and common-model signal in the EHG signals.

Otherwise, the transmission distance of EHG signals is limited and not more than 2 centimeters [3]. It is possible to lose EHG signals of pace-making points because of the long distance between the record electrode and the common electrode. Moreover, every lead affects other leads, so it is more difficult for us to detect the positions of pace-making points. Only changing the distance and size of the conventional electrodes can’t resolve the proposed problems, but application of a Laplacian electrode that can acquire localized information would fulfill the requirement.

2. Laplacian electrode
Assuming the physiologic item to be a linear, isotropic and piecewise homogeneous conductor, the current density \( J \) in it is given by

\[
J = \sigma E + J_s
\]

(1)

where \( \sigma \) is the electrical conductivity, \( E \) is the electric field and \( J_s \) is the impressed current. Under quasi-static conditions

\[
\nabla \cdot J = 0
\]

(2)

\[
E = -\nabla \varphi
\]

(3)

where \( \varphi \) is the electrical potential, Maxwell’s equations require that

\[
\nabla \cdot E = \frac{\rho}{\varepsilon}
\]

(4)

where \( \rho \) is the charge density and \( \varepsilon \) is the electric constant. Collectively, equation (1)-(4) require that

\[
\nabla^2 \varphi = -\frac{\rho}{\varepsilon} = \nabla \cdot \frac{J_s}{\sigma}
\]

(5)

Consider a local Cartesian coordinate system \((x, y, z)\) with origin at point P on the body surface where the z axis is orthogonal to the surface (Figure 1). Just inside the body surface, Laplacian’s equation can be expressed:

\[
\frac{\partial^2 \varphi}{\partial x^2} + \frac{\partial^2 \varphi}{\partial y^2} = \frac{\partial^2 \varphi}{\partial z^2} = \left( \frac{1}{\sigma} \right) \left( \frac{\partial J_s}{\partial x} + \frac{\partial J_s}{\partial y} \right) = \left( \frac{1}{\sigma} \right) \left( \frac{\partial J_s}{\partial z} \right) = \frac{\rho_{eq}}{\varepsilon}
\]

(6)

The left-hand side of equation (6) is the two-dimensional Laplacian of the electrical potential on the body surface. The surface Laplacian of the potential is negatively proportional to the two-dimensional divergence of the tangential components of the current density at the body surface, and is proportional to the normal derivative of the normal component of the current density at the body surface. Notice that the normal component of current vanishes at the body surface but the normal derivative of it does not necessarily vanish. The surface Laplacian is also negatively proportional to an equivalent charge density \( \rho_{eq} \) a charge distribution which depends only on \( x \) and \( y \) [4].

![Figure 1. A local orthogonal coordinate system](image)

Considering a set of five unipolar electrodes on the body surface, the two-dimensional Laplacian of the potential at the center point can be given by
where \( \varphi \) represents the electrical potential at electrode \( i \) and \( b \) is the interelectrode distance.

Equation (7) shows that the two-dimensional Laplacian of the potential at the center point \( \varphi_0 \) is approximately proportional to the difference of the average potential of the four neighboring points \( \varphi_i \) \((i=1,2,3,4)\) with the potential at the center point.

Since the two-dimensional Laplacian of the potential should be independent of the coordinate system selected, the relationship established in equation (7) should hold for another coordinate system \( x'-y' \) which can be obtained by rotating the coordinate \( x-y \) around the center point. By averaging the estimate of the two-dimensional Laplacian over a complete circular rotation of the coordinate system, we obtain

\[
\nabla^2_{xy} \varphi \approx \left( \frac{4}{b^2} \right) \left( 1/4 \right) \sum_{i=1}^{4} \varphi_i - \varphi_0
\]

where the integral is taken around a circle of radius \( b \).

Based on the above formulation, an integrated bipolar Laplacian electrode was developed to measure directly the surface Laplacian of the potential, or the equivalent charge density, on the body surface. Figure 2 depicts the basic schematic diagram of the bipolar Laplacian electrode developed in the present study. The electrode consists of two parts: a central conductive disk and a concentric circular conductive ring. The intermediate between them is an electric insulator. The output of the bipolar Laplacian electrode is the potential difference between the potential at the center and the average potential over the ring. As long as the interelectrode distance \( b \) is sufficiently small, the approximation will yield a good approximation of the two-dimensional Laplacian.

Thus, the output of the bipolar Laplacian electrode is proportional to the magnitude of the two-dimensional Laplacian of the potential and opposite in sign. In other words, the output of the electrode is proportional to the equivalent charge density at the center point [5]. The surface Laplacian will give different and perhaps more localized information concerning the sources that give rise to the surface potential distribution, \( V \).

Supposed EHG is to be generated by the electric dipole and Nacl solution is the medium, we do a simple experiment to research the relationship of Laplacian electrode and dipoles position. If the dipole is oriented perpendicular to the surface of the Laplacian electrode, faint signal can be recorded. While if the dipole is oriented parallel to the surface of the Laplacian electrode, distinct signal can be recorded. But it is an especially situation when the dipole is parallel symmetric distribution, the signal recorded by the circular electrode would be counteracted.

It is obvious that the Laplacian electrode is capable of acquiring localized information, so the sources of EHG signals are localized more easily. Since the surface Laplacian involves the second derivative of the potential, it would be anticipated to be much noisier than the potential itself. In this article we put forward a new type of electrode that is called active Laplacian electrode that integrates a Laplacian electrode with a bioelectricity preamplifier to resolve the proposed problem.

### 3. Active Laplacian electrode for the data-acquisition of EHG

The active Laplacian electrode is developed for high efficiency in acquiring accurate bioelectric information from the surface of the body without the use of conductive pastes. The active electrode consists of two layers: 1. a pair of concentric circular electrodes(an outer ring, a middle ring, an inner ring and a center dot)( Figure 2)etched on the backside of the non – conductive substrate, and 2. a high gain signal conditioning amplifier(mounted on the topside of the non-conductive substrate)(Figure 3). The system has four parts: the balanced ac-coupling network, the fully differential amplifier consists of two op amps, the high-pass difference amplifier and the driven-right-leg circuit.
3.1 Concentric circular electrode
A silvering circular electrode etched on the backside of the non-conductive substrate is located on the abdomen of pregnant woman. The geometry of the physiologic item to be measured influences the size of the active electrode. The optimum size would be half the diameter of the measured system. The adult uterus is like an inverted pear, which is 7-8cm long, 2-3cm thick and 40-50g in weight. Because the circular electrode is used to acquire more localized information, the active electrode has a middle diameter of 18mm. Moreover, the contact area of the concentric electrodes (an outer ring, a middle ring, an inner ring and a center dot) are made equal to ensure that the source impedances of the input leads are equal (thereby decreasing the mismatch and improving the CMRR of the amplifier).

![Figure 2. Dimensions of the Laplacian electrode designed. (Unites expressed in millimeters)](image)

3.2 The high gain signal conditioning amplifier
The novel front-end circuit differs from a common front end in biopotential measurement [6]. One is the balanced ac-coupling network in front of the first stage amplifier. The other is the amplifier concentrates its gain in the first stage.

![Figure 3. Design of bioelectricity preamplifier](image)

The high gain signal conditioning amplifier includes the novel, balanced ac-coupling network shown in figure 3. This circuit provides ac-coupling for differential signals and a dc path for amplifier bias currents, which drain to ground through the third (common) biopotential electrode. Because the input network is not grounded, if a common mode input voltage is applied, no currents flow through the network, so that all network’s nodes achieve the same potential. This absence of potential difference due to common mode inputs implies an ideally infinite CMRR regardless of component tolerances. The balanced ac-coupling network removes dc input signal voltage and prevents amplifier
saturation. Because the transfer function does not depend on \( R_1 \), one design criterion is to select it as high as practical in order to avoid loading effects on the input signal. Alternatively, selecting \( R_1 = R_2 = R \) simplifies the design. \( R = 4.7 \, \text{M}\Omega \) largely fulfills the requirement that we need an input impedance larger than \( 1\,\text{M}\Omega \) at 10 Hz.

The first stage amplifier is the fully differential amplifier consisting of the two op amps. Because the amplitude of EHG is smaller than 1mv, we need an amplifier with a gain of 1000. A common front-end amplifier can’t concentrate its gain in the first stage because of electrode-offset potentials. Electrode offset potentials can be several orders of magnitude larger than the amplitudes of the biological signals of interest, thus limiting the admissible gain of a dc-coupled front end to prevent amplifier saturation. Moreover, the op amps input offset voltages are amplified as input signals and can significantly reduce the output dynamic range. In fact, if the input stage has a large gain, this circuit achieves a high common mode rejection ratio (CMRR) without any trimmings and reduces system noise. Furthermore, the equivalent input noise only depends on the two op amps constituting the fully differential amplifier. These are coveted features in biopotential amplifiers, but due to electrode-offset voltages, the overall gain is limited to moderate values. In the design of the active lapalciand electrode the balanced ac-coupling network removes dc input signal voltages, but the op amps input offset voltages are amplified as input signals and can significantly reduce the output dynamic range. Moreover, the thermal noise of the resistance and op amps input voltage noise, including \( 1/f \) noise, would be amplified. To remove offset voltage and reduce \( 1/f \) noise, the amplifier itself must reject low frequencies. Figure 3 shows a solution that uses an integrator in a feedback loop around the difference amplifier. In the high gain signal conditioning amplifier the driven-right-leg circuit contributes to reduce the noise of power-line interference.

The proposed design has been implemented using the low-noise OP747 (quad op amps) and OP2177 (dual op amps) of ADI. A high gain signal conditioning amplifier with a gain of 1000 built according to the design rules proposed and tested for CMRR that can get 123 dB at 50 Hz.

3.3 Data acquisition experimentation

The subjects are normal pregnant women in the age range of 27.3 ± 3.6 yr and avoirdupois range of 70.5 ± 6.5kg. The recordings are normal uterine contraction evoked by medicament 24-48 hours before parturition. The multi-electrodes are fixed on the abdomen of pregnant woman (Figure 4), in which 1,5,9,13 according to the position of the uterine bottom and 4,8,12,16 according to the cervix. To pick up multi-channel EHG with the16-lead data acquisition system we can obtain waveforms of harmonious uterine contraction.

![Figure 4. Distributing of the electrodes located on the abdomen of pregnant woman](image)

Baseline of signal waveform is comparative steady. It is obvious that every channel signal changes rapidly simultaneously and gets comparative high amplitude when uterine contraction is harmonious. According to multi-channel EHG signals two new methods can be used to distinguish the initiation of uterine contraction. The first one is based on statistical characteristics of EHG; the second one we developed is a wavelet transform-based method. With the two algorithms, we can estimate the electrical wave propagation along the uterus. Since synchronization and polarity are significant characteristics of uterine contraction, the research will help to find out the mechanism of uterine contraction, and will help to provide a very promising technique for clinical or physiologic investigations of uterine activity.
3.4 Excellences of active Laplacian electrode

The signal recorded by the Laplacian electrode is fainter than by the conventional electrode, so it is more difficult for us to detect the signal in high quality among the noises. Active Laplacian electrode mounts the amplifier directly onto the electrode, eliminating the need for connective wire (to feed the signal into a remote amplifier) which act as antennas and pick up a large amount of power-line interference from the surroundings. The active Laplacian electrode concentrates its gain in the first stage, so the system noise can be smaller. The new type of electrode can almost reduce the common-mode signal including maternal electrocardiogram and breathing, quickening, electrode excursion and power-line interference. The common-mode rejection ratio (CMRR) of the active electrode is up to 123dB. The most excellence of active Laplacian electrode is in acquiring accurate local bioelectric information from the surface of the abdomen without the influence among multi-leads. So the source of EHG signal is localized more easily by using this new electrode. Since EHG signal has no fixed pace-making areas, we can observe the position change of pace-making points and energy transferring from one lead to anther through the recorded signals of different leads. Then we can tell that the uterine contraction is harmonious or unharmonious and monitor the pregnant woman well.

4. Summary

The active Laplacian electrode is applicable not only to the detection of the surface EHG signal developed, but also to many other surface bioelectric measurements (such as ECG or EMG). It can detect local information in high quality. Its minute size and very high noise rejection properties make this electrode very versatile in its application abilities.

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