# Non-invasive Laplacian Electrocardiography and Moment of Activation Mapping

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*Abstract*—**Laplacian Electrocardiogram (LECG) is a non-invasive approach providing high spatiotemporal distributed information of cardiac electrical activity. Recently researchers have recorded surface potentials from monopolar disc electrodes to estimate the Laplacian using finite difference algorithms or spline surface Laplacian estimators. Bipolar and quasi-bipolar electrodes have also been used to record LECG directly. Here based on the "Nine Point Method" (NPM), Laplacian tripolar concentric ring electrode active sensors were designed. High spatiotemporal resolution body surface LECG using the new tripolar active sensors was recorded. Moment of activation (MOA) isochronal maps used to detect cardiac electrical activation patterns exhibited on the body surface were also generated.**

*Keywords*—**ECG, Laplacian, Active sensor, Tripolar concentric electrode (TCE), Nine Point Method (NPM), Instrumentation amplifier (IA).**

## I. INTRODUCTION

HEART disease is the No. 1 killer in the United States. Early stage diagnosis of heart disease is very important. Traditionally used noninvasive 12-lead electrocardiogram (ECG) provides global temporal assessment, but has a limited ability in locating the origin of electrical events in the patients' body. Body surface potential mapping (BSPM) improved the spatial resolution of detection of potentials due to the heart by using a larger number of recording leads over the body surface [1]. However, BSPM can only reveal limited information about multiple concurrent sources because the volume conductor between the heart and the body surface smoothes the potential distribution [2]. Due to these limitations further research has been conducted in attempts to improve the spatial resolution of body surface cardiac signal detection.

Body surface Laplacian mapping (BSLM) has been shown to be an enhancement to body surface potentials mapping with better spatial resolution and improved capability of localization by using the estimated or recorded Laplacian of body surface potentials to create maps [3-8]. The Laplacian is the second spatial derivative of the potentials on the body surface which has been shown to circumvent some smoothing effects of the torso volume conduction and allows more detail in localizing and differentiating multiple concurrent dipole sources [3]. Studies [4-6] have utilized a large number of unipolar electrodes placed on the chest surface to measure the body surface potentials and derived the Laplacian Electrocardiogram (LECG) by using a 5-point finite difference algorithm or spline surface Laplacian estimator. He and Cohen [3, 7] developed a bipolar concentric electrode to directly measure the body surface LECG and demonstrated that LECG has better spatial resolution in resolving and imaging spatially distributed cardiac electrical activity than body surface potentials.

Lu and Tarjan [9] developed an active LECG sensor and Besio [8] demonstrated the efficacy of using this quasi-bipolar electrode (QBE) for detecting atrial activation patterns by recording from 35 locations on the chest surface. However, the QBE was designed with the outer ring and center disc electrically shorted. This short may cause signal distortion on the body surface causing equal potentials on both sides of the middle ring. To overcome these limitations, recently Besio [10] applied the "Nine Point Method" (NPM), a finite difference algorithm, to approximate the Laplacian potentials. Based on the NPM, a new Laplacian tripolar concentric ring electrode active sensor without the outer ring and center disc electrically shorted was designed and used to acquire LECG. Instead of the BSLM, this study will focus on Laplacian moment of activation (MOA) isochronal mapping [8], a newly developed mapping method which analyzes the cardiac activation propagation patterns as measured on the body surface.

## II. METHODOLOGY

### *A. Tripolar Active Laplacian Sensor and Signal Pre-processing*

In the present study the active sensor and signal pre-processing were developed based on a new tripolar concentric electrode (TCE), derived from the NPM, for acquiring surface LECG from the body surface. In the new design the TCE included a center conductive disc and two concentric conductive rings which are on one side of the active LECG sensor and active circuits on the other side. Compared to the QBE [8, 9], the new Laplacian TCE active sensor does not have the center disc and outer ring electrically shorted.

The following equation (1) shows the new algorithm for approximating the surface Laplacian with the TCE [4, 10].

$$L_{tripolar} \cong -\frac{1}{3r^2}[16\times(V_m - V_c) - 1\times(V_o - V_c)] \quad (1)$$

where $V_o$, $V_m$, and $V_c$ refer to the average potentials on the outer, middle, and center electrodes, r is the inter-electrode distance between the outer and middle electrode and between the middle and center electrode. In this design r is set to 0.9cm.

Fig. 1 depicts the circuit configuration on the other side of the new LECG active sensor which has a radius of 1.8cm. Two ultra high input impedance instrumentation amplifiers (IA) were used for the first stage amplification of signals from the TCE. These two IAs performed the two differences between each concentric ring and the disc. The gains of the first stage amplification were both set to 10. The output signals from these two IAs were connected to a Grass amplifier system (Grass Telefactor, 15LT) with the filter bandwidth set from 1Hz to 500Hz and the amplification set to 2,000 which gave a total gain of 20,000. The recorded signals from the Grass amplifier were pre-processed to get the tripolar LECG based on equation (1).

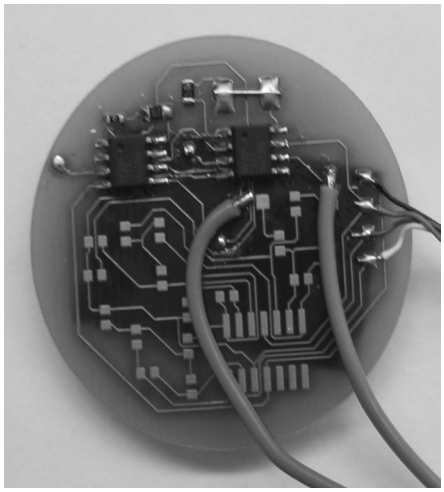

Fig. 1 New LECG TCE active sensor. Two shielded wires used for connecting the signals from the instrumentation amplifiers to a Grass amplifier system.

*B. LECG MOA Algorithm*

The MOA is defined as "the instant the dipole that represents the depolarization wavefront crosses the vector normal to the active sensor's surface" [8]. It is determined by calculating the time when the dipoles' activation wavefront is directly below the sensor. A new algorithm for detecting MOA [11] was automated. This algorithm detects the QRS wave for windowing, performs the cross correlation between the LECG and lead II ECG windows to find the best resemblance between waves and calculates the time offset between these two waves. The cross correlation acts like a pattern matching filter.

*C. Data Acquisition System*

An interface board was designed and assembled to connect the Grass amplifier to the LECG TCE active sensors. Six LECG TCE active sensors and one Lead II ECG sensor which served as a time reference were used to record the cardiac body surface data. A 16-bit A/D converter Dataq Instruments DI-720 Series was used to digitize the analog signals from the Grass amplifier at a sampling rate of 2,000 Hz for 30 seconds.

*D. LECG Acquisition from Human Subjects*

All signal acquisition was performed in accordance with the Louisiana Tech University IRB approved protocol. Signals were recorded from six healthy male subjects ranging between the ages of twenty to twenty-five years old. The six sensors were configured in a 3 row × 2 column matrix and attached to the inside of a wide elastic strap as shown in Fig. 2. While recording, the strap was wrapped around the body to accommodate different torso curvatures to improve the contact between the skin and the electrodes. A thin coat of 10/20 electrode paste (Grass Telefactor) was spread uniformly on the electrodes to ensure good contact between the electrodes and the skin. All these factors help limit the influence of the boundary condition.

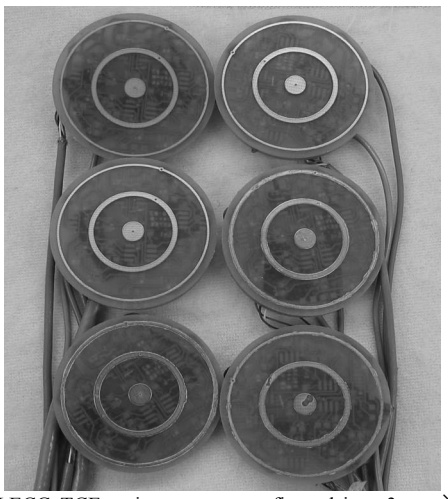

Fig. 2 Six LECG TCE active sensors configured in a 3 row × 2 column matrix and attached to the inside of a wide elastic strap.

Each subject layed in a stationary position to avoid the influence of fluctuations of the heart position on the body surface ECG and LECG [12]. The actual recordings were repeated three times with the sensors moved to each of the preplanned locations with 12mm horizontal spacing on the chest. The total 6 row × 12 column matrix signals were recorded for each subject from 72 locations.

## III. RESULTS

*A. Tripolar LECG Waves*

The processing involved QRS detection [11], 1600-point window (800ms) Weiner filtering, and ensemble averaging.

Three typical processed tripolar LECG signals from one healthy male subject which represent three main types of LECG waveforms----monophasic negative (A), monophasic positive (B) and biphasic (C) are shown on Panel A, B and C of Fig. 3 as well as the Lead II ECG in Panel D. The monophasic negative wave describes the wave which has a strong downward pulse and a weak upward pulse. The monophasic positive wave represents the wave which has a strong upward pulse and a weak downward pulse and the biphasic wave represents a strong positive/negative pulse followed by a strong negative/positive pulse. The duration was indicated between two dashed lines in each panel. For consistency the first zero crossing before and after the peak determined the start and end of each QRS wave.

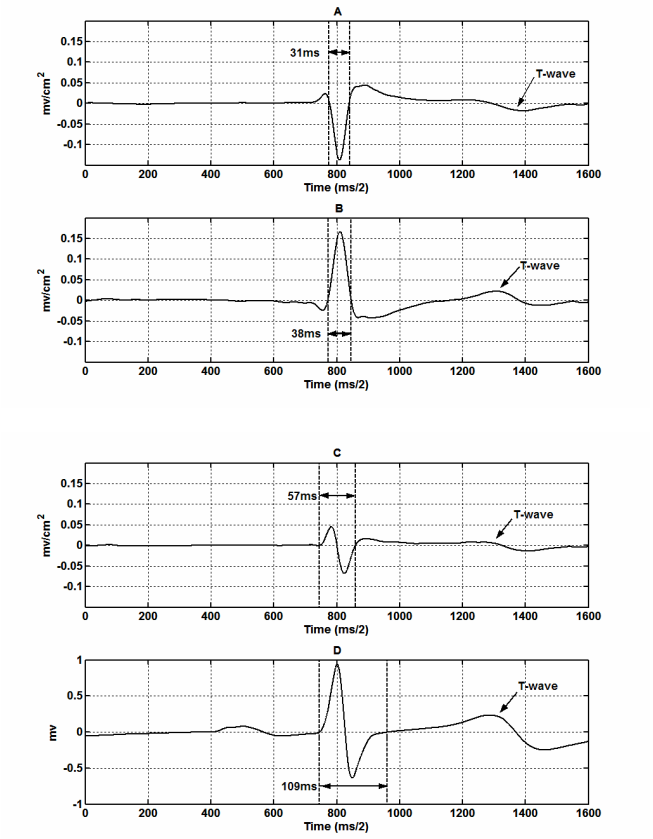


Fig. 3 Tripolar LECGs Panel A (monophasic negative), B (monophasic positive) and C (biphasic), Panel D is the relative Lead II ECG recorded simultaneously. The X coordinate designates the time in 1/2 ms increments. T-waves and QRS duration are marked in the figure.

### *B. MOA Isochronal Mapping*

An LECG MOA isochronal map was created using the calculated MOAs for each subject to reveal the cardiac activation propagation sequences and properties as measured on the body surface.

Fig. 4 shows an MOA isochronal map from the same subject as the signals in Fig. 3. The propagation time sequence of the cardiac activation with respect to the Lead II ECG R-wave peak is shown in isochrone contours with the MOA denoted. Negative MOA indicates earlier activation compared with Lead II ECG and positive MOA corresponds to later activation on the chest surface. The range of MOAs for this subject were -25ms to 26 ms. Two localized earlier activation areas with centers located at (2, -2) and (3.5, 5) respectively were observed with the early MOA isochrones of -10ms. In the top left corner the early MOA isochrone of -25ms was shown at the area with the center located at (-1, 8). Other earlier activation time sequences were observed in the lower left area (-1.8, -8) and (4.5, -9) of the map with the early MOA isochrones of -10ms. Positive time sequences showing later activation were viewed over a broad area.

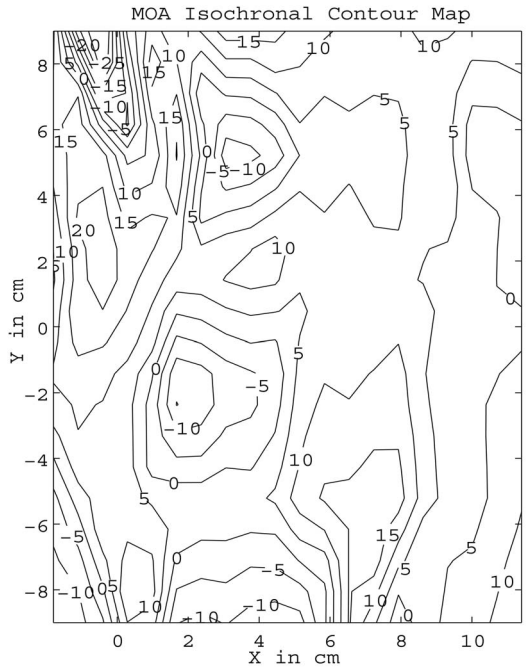


Fig. 4 MOA isochronal map. The X and Y coordinates (x, y) in cm designate the position of the MOA map over the chest. The MOAs are labeled on the isochrones with the units in ms.

Consistent MOA isochronal map patterns across different subjects were obtained with certain minor variations of activation patterns. The range of MOAs for all 6 subjects is -31ms to 29ms. As the cardiac activation spreads over the chest, the earlier activation area located in the center left, (2,-2) in Fig. 4, is viewed in all 6 subjects with slightly different locations and different occurrence times. The location shift is within a range of 2.5-3.2cm. The difference between occurrence times is about 10ms.

## IV. DISCUSSION

This paper reports on the experimental investigation of a new LECG TCE active sensor used on a group of human subjects. Compared to the normal 12-lead ECG system and the BSPM system, the new tripolar LECG system described in the present study has shown promise for recording high-quality, site-specific local LECG, which possesses high spatio-temporal resolution, directly and quickly from any geometrical body surface. With this new technique,

clinicians may locate the heart disease non-invasively. Since the recordings were conducted at 12 mm intervals, the LECGs we acquired had higher spatial resolution compared with the other previous LECG recordings [8], 72 vs. 35 locations over the same chest area.

From Fig. 3 it can be seen that the duration of the LECG QRS wave was less than that of Lead II ECG. The LECG has much less magnitude compared with the Lead II ECG. Both of these features may result from having a much smaller electrode spacing (inter-electrode distance) on the LECG sensor compared with the long distance between the Lead II ECG electrodes which were put on the left leg and right arm. Panel A, B, and C reflect that the T-wave which represents ventricular repolarization can be seen clearly with different polarities in the LECG.

Since the human body is an inhomogeneous conductive distribution, spatial nonuniformities of isochrone density are possible as can be seen in Fig. 4. There are regions with apparent faster spread of cardiac activation where the isochrones are sparse and regions with relatively slow activation spread where isochrones are crowded. For example, in Fig. 4 the area with the center located at (2, -2) has faster spread than the area with the center located at (-1, 8), by reason that the later area has more crowded isochrones.

From this MOA map, we can also see when one cardiac electrical activation spreads out over the chest from the heart more than one early activation was shown. This was probably because some of these MOAs detected are from different subsequent cardiac cycles. There was approximately one second between adjacent heart beats. More investigation, such as using a heart-torso model to control the period of cardiac pacing, will be performed in the future to explain this phenomenon.

According to the timing and locations, the heart may be located beneath the surface area with the center located at (2, -2) which has an earlier activation time sequence, where the original cardiac activation may start from. Despite the slight variability of occurrence time and locations on MOA maps among subjects, possibly due to differences in body conductivity and variation in heart orientation/position within the chest caused by the body position [12], consistent MOA map patterns were obtained. Therefore the MOA isochronal mapping may be associated with the underlying cardiac electrical events and used for diagnosing purposes.

## V. CONCLUSION

The new tripolar LECG TCE active sensor can provide more detailed spatial information than other active Laplacian sensors. Based on the newly developed MOA algorithm, the Laplacian MOA isochronal map reveals the properties of the cardiac activation propagation which may be useful to identify abnormal cardiac activations. Future work will be focused on differentiating the atrial and ventricular activation by MOA mapping.

## ACKNOWLEDGMENT

The authors thank Dr. Wei Jiang for her assistance in hardware testing techniques and all of our lab members for their kindness.

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