Non-Invasive Detection of Moment of Activation Using Laplacian Cardiac Electrogram Body Surface Mapping System

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Summary

An isochronal mapping system was developed with our DOLCE (Directly Obtained Laplacian Car-diac Electrograms) system. It employs minimal signal processing to facilitate diagnosis and possibly accelerate the response to arrhythmic emergencies. Men and women were among the subjects, many with cardiac pathologies. This noninvasive, presently experimental system, promises isochronal maps with high temporal and spatial resolution in quasi-real-time, demonstrating the activation sequence over the heart. Special signal conditioning and processing circuits were integrated with tripolar concentric sensors into "active sensors" for placement on the chest, without skin preparation. Up to seven DOLCE sensors were recorded simultaneously along with a Lead II ECG. Signal averaging over 30-sec epochs, with respect to the automatically detected R-wave in Lead II (L2pk), improved the SNR. The moment of activation (MoA) was also detected automatically from the zero crossing of the signal-averaged DOLCE signal. The process is safe and, in the opinion of the authors, it has shown promise for various clinical applications.

Keywords

Laplacian ECG, active sensors, isochronal maps, moment of activation

Introduction

The Laplacian surface ECG (LECG) is the second spatial partial derivative of the body surface potential. The Laplacian, ($\tilde{N}^2\Psi(r,t)$, where $\Psi(r,t)$ is the scalar potential in the body) at a specific point on the surface of a spherical volume conductor should be rotationally invariant, omni-directional, with respect to the normal vector at that point on the surface. The Laplacian is a function of ***time*** at a point on the surface. It arises from a moving current dipole element within that sphere. Its amplitude depends on the directional relationship between the point of concern on the spherical surface and the normal vector of the current dipole element, as shown schematically in Figure 1.

We obtain the Laplacian ECG with a set of three concentric electrodes. The vector, **r**, spans from the point where the Laplacian is obtained on the surface to the cen-ter of a moving current dipole. The amplitude of the Laplacian is proportional to r^{-N} (N = 3 for tripolar sensors) and cosφ, φ being the angle between **r** and $\mathbf{n}_{dipole}$. In general, $\mathbf{n}_{dipole}$ and $\mathbf{n}_{sensor}$ are not in the same plane. However, it seems to be a fair simplification to assume that the di-pole propagates almost, if not perfectly along $\mathbf{n}_{dipole}$. If the domain of movement for the dipole element is confined to a bounded volume within the sphere representing the heart within the body, then as the dipole travels in this cardiac domain, the maximum of the Laplacian signal will be bounded by the minimum distance between the sensor and the nearest point on the heart. As the heart is depolarized, the maximum and minimum of the Laplacian at a point on the surface should be related to the equivalent dipole of the depolarization wavefront, as it travels past the sensor at the shortest distance. The dipole's trajectory is constrained within the heart, hence the signal will usually feature a zero crossing that corresponds to the instant when $\mathbf{n}_{dipole}$ and **r** are orthogonal. Such a zero crossing is usually observed between two extreme values and corresponds to the steepest rate of change in

the LECG with time. Such a zero crossing may be called the Moment of Activation (MoA) of the heart muscle nearest the sensor.

This introduction began with the assumption that the chest may be modeled as a sphere. However, as the chest surface is slightly deformable, a sensor may be attached to it with a little pressure to form a tangential plane. The sensor's orientation, $\mathbf{n}_{sensor}$, is expected then to point toward that part of the nearest region of the heart muscle, whose depolarization, represented by a current dipole element, would produce the dominant part of the LECG at that point. Therefore, we began our exploration on the as-sumption that the zero crossing of the LECG at a specific point corresponds to the MoA of the nearest cardiac tissues.

While He and others [1] concentrated on producing the LECG at specific points on the chest by computing the second spatial derivative from differences in simultaneously occurring potentials at adjacent points within a regular array of points on the chest surface and then creating activation maps from these computed values, our approach was different in two significant ways. Geselowitz and Ferrara [2] recently stated their concern over the validity of the computational approach to the LECG on theoretical grounds, as well on empirical data from sensors consisting of 5 contact points. They observed significant differences with a 45° rotation of their sensor on the chest surface. As we employ ring sensors, as long as we have good contact around the circumference, the ring sensors provide a rotationally invariant LECG signal. Geselowitz and Ferrara also expressed their concern regarding errors due to the chest surface not being spherical. Our approach is based on the LECG sensor searching for the MoA in the vicinity of its own axis, hence sphericity is not essential.

Kaufer [3] showed that, as a ring is decomposed into k elements, the average of the computed potentials for those elements rapidly approaches an asymptotic value when $k \geq 8$. This was found useful for dipoles in a region within one or two diameters of the sensor, measured from the sensor's center. Kaufer's simulation also showed the dominance of the closest of several contiguous current dipole elements for a three-ring sensor.

Our sensors, whose performance and technical details were described elsewhere [4], may be summarized: the outer ring, 36 mm ∅, is shorted to a central dot, forming one pole. The areas of the inner and outer

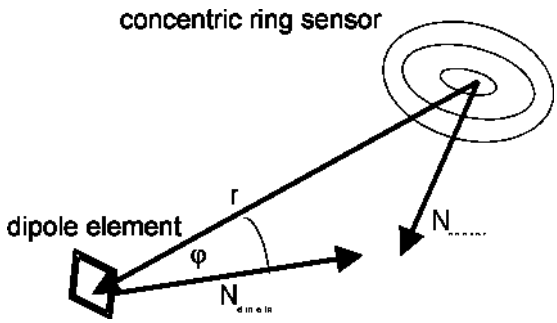

Figure 1. Dipole and sensor relationship.

electrode elements match the area of the inner ring that serves as the second pole. The electrodes are etched on a printed wiring board (PWB). The signal processing electronics is on its flip side: an amplifier and band-pass filter, powered by 2 lithium cells within the sensor. A miniature cable carries the sensor's output. The signals are A/D converted with a card in a battery-powered laptop computer. Data storage, processing and other functions, such as L2pk and MoM detection, are carried out with LabVIEW software (National Instruments). The sensors were designed to meet the following objectives:

1. Pronounced sensitivity to activity within the sensor's vicinity and rejection of far-field events. The sensitivity of the tripolar sensor corresponds to an inverse cube relationship with distance when the inner and outer elements are shorted to form one terminal vs. the middle element:
   $V_{3\text{-ring}} = 0.5(V_{out}+V_{in}) - V_{mid}$
   $= 0.5[(V_{out}-V_{mid})- (V_{mid}-V_{in})] \propto \sigma^2 V_{sfc}/\sigma s^2$
   where s represents an incremental distance on the surface from the center of the sensor.
2. High input impedance to reduce contact problems.
3. High common mode rejection ratio.
4. Processing electronics and batteries integrated with the sensor to reduce inductively coupled interference and power supply noise.

With these criteria met, we attempted to assess the clinical utility of LECGs, especially by detecting activation sequences in 2-D in a novel way [5].

Earlier data from chronically implanted epicardial tripolar concentric sensors provided the basis of our hypothesis that LECGs from three or more points may provide sufficient information for the classification of arrhythmias in an individual subject [6].

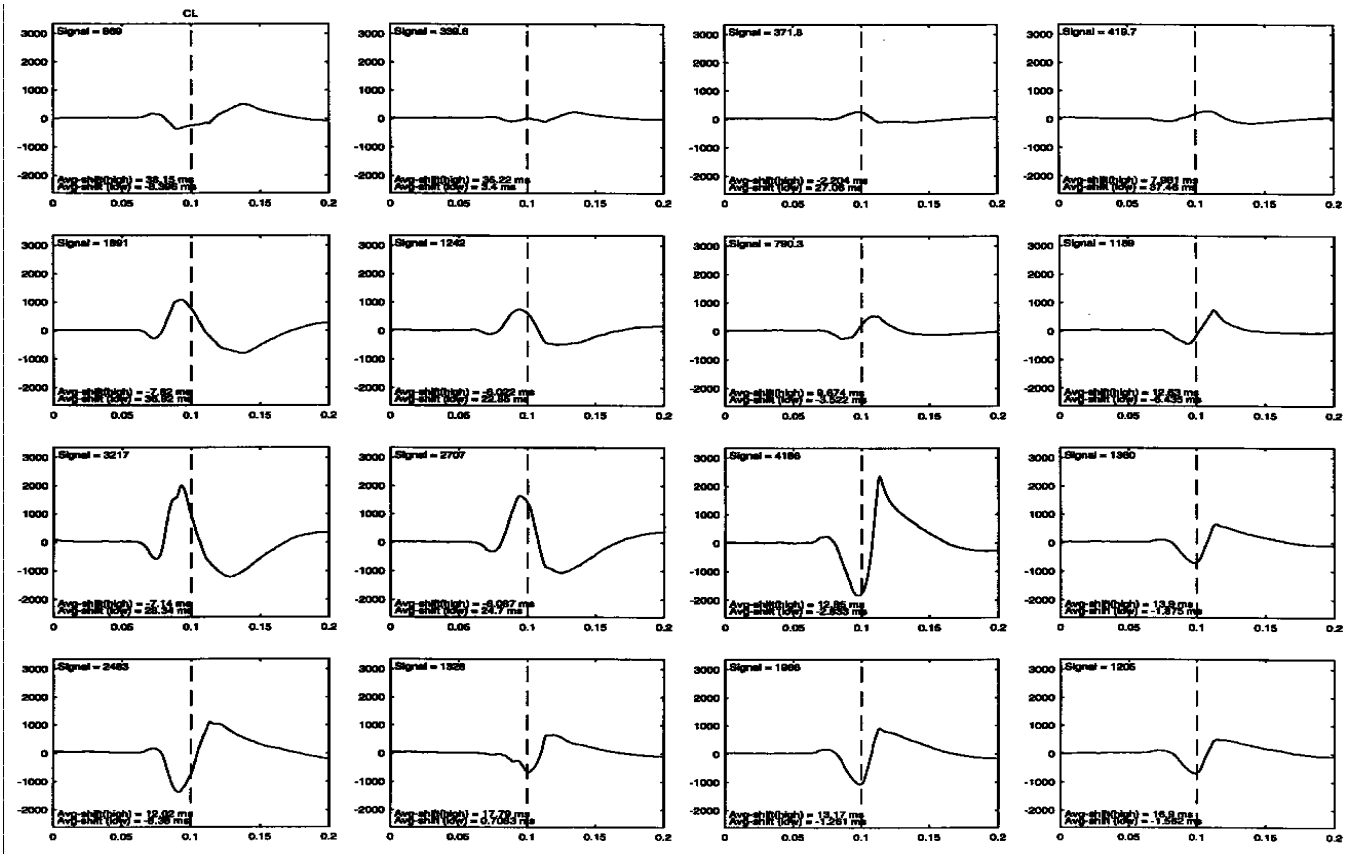

Figure 2. Signal-averaged LECG signal for all 16 sites (G, 36 y).

Methods and Materials

The exploratory work with DOLCE began with a single active sensor with special signal conditioning and pro-cessing circuits for use on the chest. Each 36 mm Æ sensor has its on-board dual power supply with 370 mA drain, high input impedance (»10 GW), high gain (1000), high CMRR (117dB min). The high input impedance allows the active sensor to function without skin preparation. The low-noise amplifier (0.7 mVrms at the output) allows the weak, site-specific DOLCE signals to stand out in real-time. The signal from each active three-element sensor was amplified through a quasi-AC coupled instrumentation amplifier and a band-pass filter, mounted on the back of the substrate of the electrodes which were etched on a relatively rigid PWB. Each sensor was secured to the chest either with woven medical adhesive tape, or with a woven belt around the chest. After a typical session of 20 to 30 minutes, a visible indent was left on the skin that would disappear within a few minutes. We made no attempt to normalize the applied pressure. The amplified and filtered LECG was digitized at 1000 samples/second and stored in a Pentium laptop computer, using LabVIEW software (National Instr.). Repeatability and stability were demonstrated by signal-averaging 30 seconds of data with respect to the automatically detected L2pk. The MoA was also detected automatically from the zero crossing of the DOLCE signal and its delay from the L2pk computed. In the later phases 7 DOLCE sensors were simultaneously used to record from the chest surface, along with a Lead II ECG. The sensor sites were measured with respect to a mid-sternal line and one that crossed the centers of the subject's nipples.

Phase 1: Initially 6 healthy volunteers (4G, 2E) were recorded at rest from a 4X4 matrix over the left chest (Figure 2). The SNR was improved by averaging 25 beats with L2pk serving as the common time reference, using 100 samples before and 99 after the L2pk. Propagation of the MoA was evident, as shown in Figure 2 from one subject. Subsequent studies were based on these results.

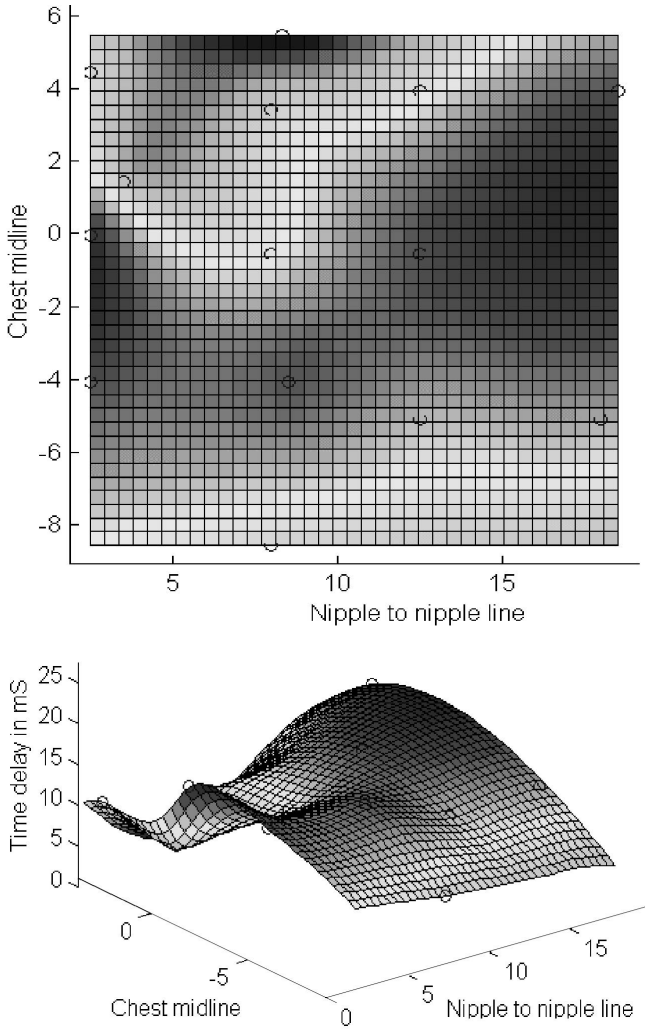

Figure 3a and b. Body surface LECG map:Normal volunteer, female, 57 y.

Phase 2: Three channels were simultaneously recorded from patients undergoing electrophysiological evaluations at the University of Miami School of Medicine (UMSM).

Phase 3: Three to seven channels were recorded simultaneously from a series of 40 cardiac patients and a few healthy volunteers in Ferrara. The sites were chosen to avoid thick layers of adipose tissue areas. The sensors were placed below and around the left breast of each woman. In some instances sites were also chosen below the left clavicle to observe whether atrial activity could be detected. To achieve that goal, the averaging window was widened from 200 to 300 samples and the L2pk time reference was shifted from the center toward the trailing edge of the time window.

Phase 4: Sequential recordings with 7 simultaneous channels were carried out to obtain more than 7 LECGs. As many as 22 sites were used, while at least 2 sites were retained when the other sensors were moved to new sites. The results were first plotted by hand as ***isochronal*** maps for two subjects, a healthy woman and a man with an acute anterior infarct. Subsequently the data were processed into maps with MATLAB as shown in Figure 3 and 4. The automatically detected MoA was visually verified for each site by inspection of the averaged signal in EXCEL.

Results

The system was used extensively to study normal subjects and heart patients, in Miami and primarily in Ferrara, Italy; many with arrhythmias. A few illustrations are presented to show the novelty and progress with DOLCE:

In Phase 1, an asymptomatic subject showed a consistent bigeminal rhythm. Consistent MoA time shifts occurred on the average 8.1 ms earlier for the bigeminal beats than for normal sinus beats.

The initial deflection at the 16 sites in a 4x4 array over the left chest showed fairly consistent patterns for the 8 subjects: the polarity switched from upward first to downward first along a diagonal over the array, pointing toward the left shoulder.

Conduction delays in Bundle Branch Block were recorded from 8 other subjects. Comparing the MoAs at sites above the right and left ventricles, the delays could be quantified noninvasively.

One of the 40 subjects in Ferrara, a 75 year old man, with a one-year old infarct and a left atrial myxomatectomy 36 years earlier, was in chronic atrial flutter (AFL). Recordings from sensors at the "V6" and "V7" sites showed sharp AFL waves, while frontal sites did not, as the lateral sensors were closest to the left atrium. The high definition of the non-invasively obtained AFL signals was astonishing.

LECGs from more than 20 frontal and subaxillary sites were recorded in overlapping sequences from two subjects using L2pk as the reference. Figure 3a and 4a show planar displays. Figure 3b and 4b show time delays, from L2pk to MoA, along the vertical axis. The female subject (Figure 3) with a normal depolarization sequence shows nearly monotonic progression over the surface. The MoA in the zone that corresponds to the anterior infarct in the male subject (Figure 4) is

delayed and is activated last as shown by the peak on the warped surface in Figure 4b.

Discussion

Earlier attempts, with concentric ring sensors on the chest, failed, as the signals were too feeble and noisy [5]. While we present here only preliminary findings, they demonstrate the feasibility of isochronal MoA maps with active sensors based on concentric electrodes. We do believe that it is feasible to produce isochronal maps with arrays of 16 or even 32 sensors in near-real-time, within a few seconds after the data is collected. Such activation maps are expected to be of service to the clinician in the detection of various time and space dependent events such as the action of thrombolytic agents, anti-arrhythmic drugs which alter propagation velocity, detecting atrial activity as useful diagnostic information in the analysis of various reciprocating arrhythmias involving the atria and the ventricles. Detection and estimation of the size of an infarct and its subsequent evolution may also prove to be useful. Much of the preliminary data has not yet been fully analyzed, but it appears that the data collection method is safe, without discomfort and appears to be repeatable.

A practical concern is the rapid and accurate registration of the topography of the sensors. In Ferrara, the center of each sensor was related to two natural, but somewhat imprecise reference axes: the "vertical" midline running through the xiphoid process and the "horizontal" axis that intercepts the subject's nipples. The latter is not a stable reference line, especially in women. The coordinates were determined with a flexible tape measure and recorded by hand. An automatic and accurate site registration method is yet to be developed, although there are numerous possibilities.

Another recurring problem was related to temporarily noisy signals at some sites when the sensors were first applied. These problems were almost always promptly resolved by wiping the skin and the sensors with alcohol. In other instances a folded bed sheet was placed on top of the sensors whose weight improved the pressure at the contacts and contributed to clearer tracings. In a few instances the operator applied gentle pressure by hand to a single noisy sensor, through an insulating rod, to eliminate capacitive coupling of line-frequency interference from the operator to the subject. (The Lead II ECG used a third electrode as a reference.)

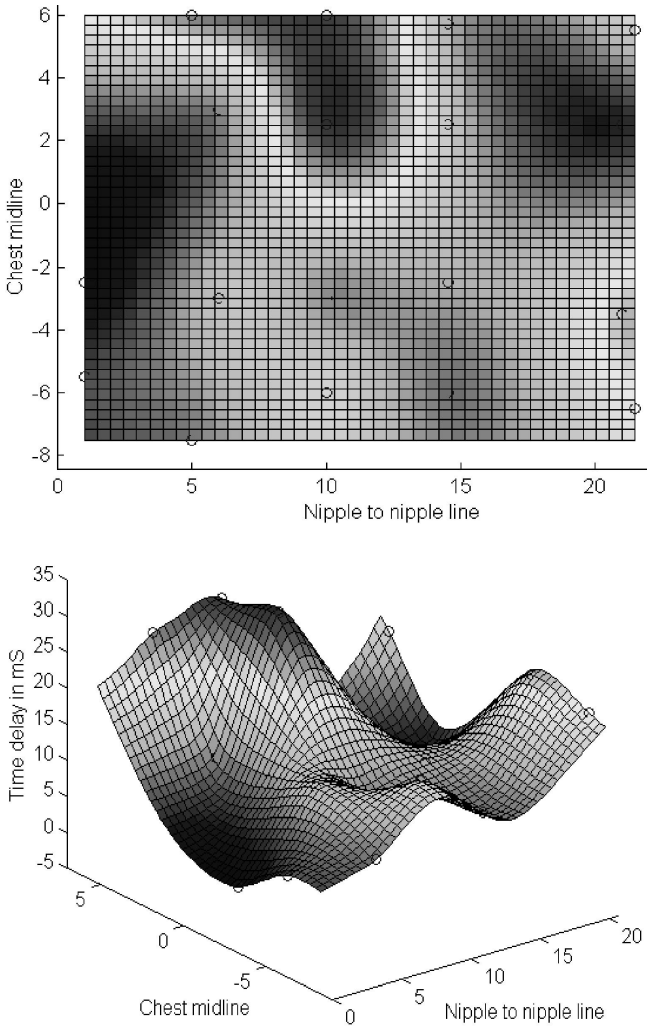

Figure 4. Body surface LECG map:Time delay for male subject with anterior infarct.

Signals coupled to respiration often showed up at some specific sites, presumably contributed by the EMG from costal muscles. The 30 sec averaging process, synchronized to L2pk, did minimize the influence of respiration related EMGs.

It is noteworthy that women could be successfully included in this series of studies in contrast with other reports, which were limited to men.

Conclusions

An isochronal mapping system was demonstrated with Directly Obtained Laplacian Cardiac Electrograms (DOLCE) for rapid arrhythmia diagnosis, with minimal signal processing, to facilitate medical response to arrhythmic emergencies. Our noninvasive, experimen-

tal system generates body surface maps with adequate temporal and spatial resolution in quasi-real-time, showing the activation sequence of the heart.

The portable DOLCE mapping system provides a tool for understanding the propagation sequence of the heart non-invasively, in quasi-real-time. Isochronal activation maps have been presented for two subjects which promise wider applications of DOLCE-based isochronal or even isopotential maps, which may be shown automatically after a brief period of data processing. The active sensors appear to eliminate the need for obtaining the LECGs by computation and eliminate the problems associated with the rotation of the sensors.

Isopotential LECG maps may provide additional information for the clinician about the nature of propagation in pathological cases, but those have not been explored.

Larger numbers of simultaneous channels than 7 may be utilized in the future, however, their rapid and reliable application calls for innovations relying on some vest or other elastic and anatomically correct garment.

References

[1] He B. Theory and applications of body-surface Laplacian ECG mapping. IEEE Engineering in Medicine and Biology Society. 1998; 17(5): 102-109.

[2] Geselowitz DB, Ferrara JE. Is accurate recording of the ECG surface Laplacian feasible? IEEE Transactions on Biomedical Engineering. 1999; 46(4): 377-381.

[3] Kaufer M, Rasquinha L, Tarjan PP. Optimization of multi-ring electrode set. Annual International Conference of the IEEE Engineering in Medicine and Biology Society. 1990.

[4] Lu CC, Tarjan PP. Pasteless, active, concentric ring sensors for Directly Obtained Laplacian Cardiac Electrograms. The 3rd International Workshop on Biosignal Interpretation. Chicago. 1999.

[5] Lu CC. Non-Invasive Laplacian ECG Detection Using Active Concentric Ring Sensors. Ph.D. Dissertation, Univ. of Miami, Coral Gables, FL, 1998.

[6] Rasquinha L, Kaufer M, Tarjan PP. Directional depolarization sensors of body surface ECG. Annual International Conference of the IEEE Engineering in Medicine and Biology Society. 1991.

[7] Rasquinha L, Kaufer M, Tarjan PP. Arrhythmia detection with spatially differentiating epicardial sensors: signal processing. Proceeding of the 7th Purdue conference on Cardiac Defibrillation. 1992.