Non-Invasive Cardiac Imaging Based on Just the Standard 12-Lead Signals?

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Abstract

Recently, we presented progress in the development of an inverse procedure based on the equivalent double layer at the surface of the heart for non-invasive imaging of the activation sequence on this surface. The number of signals used was 64. In this paper we compared the activation sequences estimated from fewer signals from those obtained when using the 64 signals (the activation times estimated from the 64 signals are taken as the gold standard). The lead systems studied involved 32 (Lux), 23 (Barr), eight (standard 12-lead) or six independent (un-weighted Frank) signals. The inverse procedure proved to be remarkably robust with respect to this reduction in the number of leads, the 12-lead ECG yielding the poorest results. We attribute this robustness to the fact that the initial estimate of the solution was based on the general electrophysiology of activation propagation.

1. Introduction

The non-invasive estimation of cardiac activation and recovery has clinical relevance. Currently, methods exist to estimate non-invasively the electrical activity of the heart from up to 265 measured ECG signals [1]. A lower number of ECG leads will facilitate the clinical application of these non-invasive methods.

In our previous research we have been using 64 leads, a number tuned to the number of independent ECG components observed in large data bases. The non-invasive imaging method used, and recently described by van Dam et al. [2], is based on a double layer at the myocardial surface as the equivalent source of cardiac activity [3].

In this study we investigated the effect of a reduction in the number of leads on the quality of the estimated activation sequences, while taking as the gold standard the solution found by using 64 leads. The performance of the leads systems was tested on seven differing beats in recorded ECGs obtained from one healthy subject (atria and ventricles) and two patients (ventricles only).

2. Materials

Body surface potentials (64-lead ECG) were recorded on: a healthy subject, a WPW patient and a Brugada patient during an Ajmaline provocation test. For each subject, MRI-based geometry data were recorded, from which individualized volume conductor models were constructed, incorporating the major inhomogeneities in the conductive properties of the thorax, *i.e.* the lungs, the blood-filled cavities and the myocardium.

The first subject (NH) is a healthy individual [2] for whom both the atria and the ventricles were reconstructed. The signals of a single beat were analyzed.

The second subject (WPW) is a WPW patient for whom previously estimated activation times have been published [2,4]. The beats analyzed included episodes in which the QRS displayed the typical WPW pattern, *i.e.*, a fusion beat in which the activation is initiated at both the AV node and the Kent bundle. The location of the latter was determined invasively. The ECGs were also analyzed for beats recorded after an AV-nodal block had been induced by a bolus administration of adenosine, which resulted in an activation sequence solely originating from the Kent bundle. Additionally, an ectopic beat originating from the right ventricle was included.

The third subject was a Brugada patient (BG) in whom ECG data were recorded during infusion of a sodium channel blocker (Ajmaline) [5], which changes the activation and/or recovery sequence. Ten bolus infusions of 10 mg were administered at one-minute intervals [6]. The beats selected for the analysis presented here were: the beat 5 minutes prior to infusion and the beat after the last bolus had been administered.

In all, seven beats having a different origin, and thus morphology of the ECG waveforms, were analyzed.

3. Methods

The inverse procedure used in our research is based on the equivalent double layer (EDL) source model. The local source strength is the transmembrane potential (TMP) at the surface of the myocardial surface. The TMP

waveform is described analytically, including a parameter specifying the local timing of activation [7].

The associated parameter estimation problem is non-linear, and consequently requires the specification of an initial estimate. This estimate was based on the fastest route algorithm (FRA), while taking into account the anisotropic nature of propagation [2]. First, activation sequences are generated for all nodes on the myocardial surface with the corresponding node acting as focus. Next the node for which the ECG corresponding to the activation sequence correlates best with the measured ECG is selected. Subsequent break-throughs are added until no further improvement of the correlation between simulated and measured ECG is obtained.

Based on this initial estimate, the activation sequence was obtained by using a dedicated Levenberg-Marquardt algorithm [2]. For the atria the activation sequence was estimated from the signals during the P segment, for the ventricles from those during the QRS segment up to the J-point as observed in the RMS curve.

For each of the lead systems studied, the inverse procedure used the corresponding ECG leads, both in the initial estimation procedure as well as in the optimization procedure subsequently carried out.

The results obtained by using the 64 independent signals of the Nijmegen/Amsterdam lead systems [8] were taken as the gold standard. The results reported on in this paper are those of the following four lead systems: the one published by Lux et al. including electrodes on the back [9], the one published by Barr et al. [10], the standard 12 leads and the Frank lead system. The numbers of independent components involved are listed in Table 1.

For each of the seven beats and each of the four lead systems, the linear correlation coefficient was computed between the estimated timing (activation sequence) at each of discrete nodes on the heart surface (either atria or ventricles) and those of the reference solutions, the ones found using 64 signals. The corresponding ECGs were computed for all five lead systems.

Table 1. The number of independent signals involved in the five lead systems.

Lead system	**# ECG signals**
Nijmegen/Amsterdam	64
Lux (*anterior/ posterior leads)*	32
Barr	23
12-Lead	8
(independent) Frank leads	6

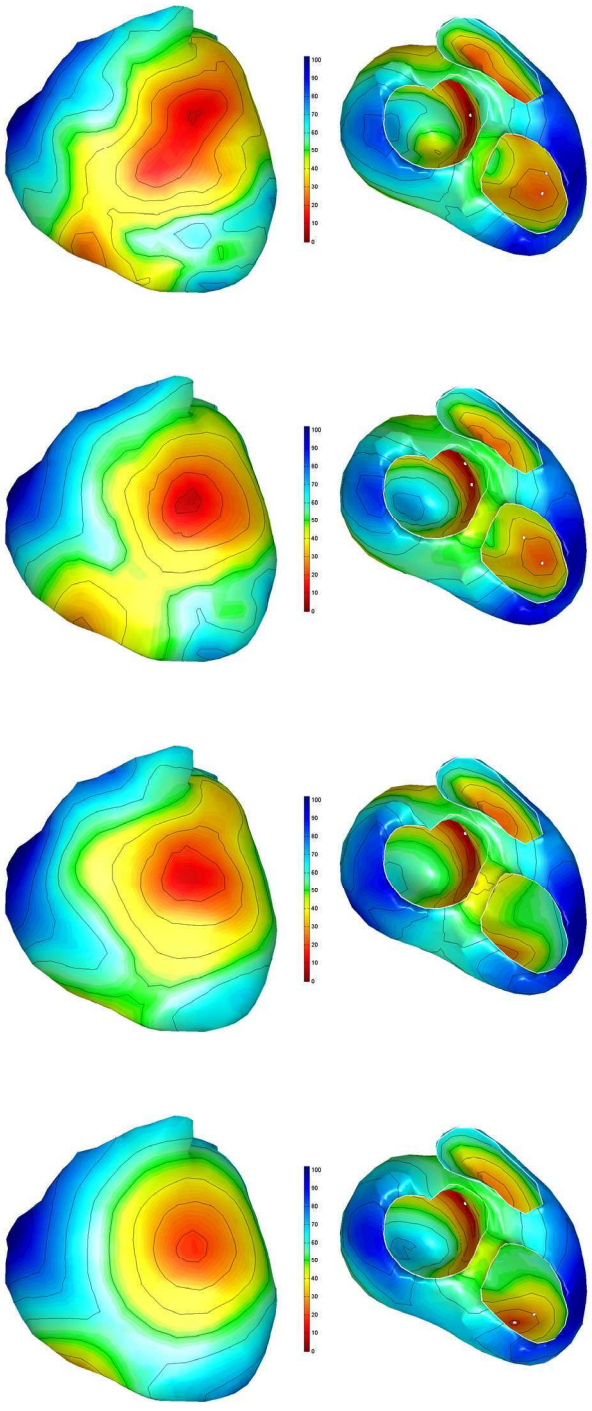

*Figure 1. Normal subject ventricular activation sequences as obtained using **(top)** 64lead, **(second)** Lux lead system, **(third)** the eight independent signals of the 12-lead system, and **(bottom)** the six independent signals of the Frank system. Isochrones are drawn at 10 ms intervals. The ventricles are shown in a frontal view (left) and basal view (right). White dots indicate the early breakthroughs found by the initial estimation procedure.*

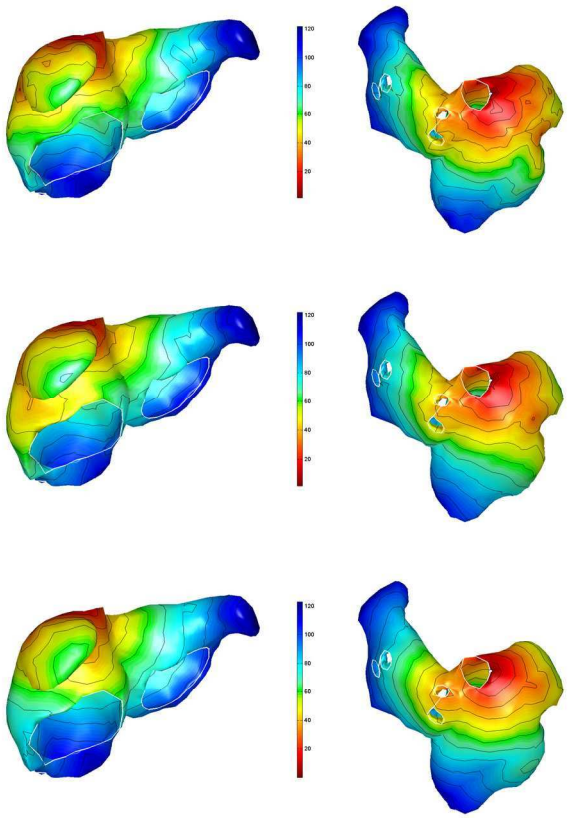

Figure 2. Normal subject atrial activation sequences as obtained using (top) a 64-lead system, (middle) the Barr lead system, and (bottom) the six independent Frank signals. Isochrones are drawn at 10 ms intervals. The atria are shown in a frontal view (left) and oblique posterior view (right). White dots indicate the first early breakthrough found by the initial estimates.

The differences between simulated and recorded ECGs were quantified by using the *rd* measure: the root mean square value of the differences between all matrix elements involved relative to that of the recorded data.

4. Results

When using 64 ECG signals, the estimated the activation sequence was in full qualitative agreement with the knowledge available from invasive data [2].

An overview of the correlation between the activation sequences obtained from 64 leads and ones obtained with the reduced lead systems is presented in Table 2.

For subject NH, some of the estimated activation sequences found using different lead systems are shown in Figures 1 and 2, for the ventricular and atrial activation, respectively. The first breakthrough in the atria was invariably found in an area associated with the sinus node, irrespective of the lead systems used. The patterns found in the subsequent optimization procedure are almost the same (first column of Table 2).

Table 2. The correlation coefficients between the activation sequences, estimated by the entire procedure, from 64 leads and the other lead systems (see Table 1).

	Atria NH	**Ventricles NH**	**Baseline BG**	**Peak BG**	**Fusion WPW**	**Kent bundle WPW**	**Ectopic WPW**
Lux	0.97	0.93	0.89	0.89	0.80	0.97	0.95
Barr	0.98	0.88	0.67	0.59	0.90	0.96	0.94
12-leads	0.98	0.57	0.70	0.74	0.80	0.95	0.93
Frank	0.98	0.65	0.70	0.63	0.19	0.85	0.86

For the ventricles, the similarity between the activation patterns and the reference solution (second column of Table 2) decreased with the number of signals involved. The poorest result was found for the 12-lead system. Recall that this involves eight independent signals.

For the Lux lead system all correlation values were above 0.8 for all beat morphologies. Single focus activations, i.e. in the NH Atria, WPW Ectopic and Kent Bundle beats only, were estimated similarly with all tested lead systems. For the cases where the His-Purkinje most likely played a role (beats: NH ventricle, BG and WPW fusion), matching activations were only found for the Lux lead system.

The *rd* values between measured and simulated ECGs are illustrated in Figure 3. It shows a clear rise when using fewer leads. For the Lux and Barr systems the *rd* is still relatively close to the values found with 64 leads.

5. Discussion and conclusions

The inverse procedure identified highly similar activation patterns for the lead systems with 64, 32 and 23 independent signals. For the more simple activation patterns, e.g. the normal atrial activation or the ectopic ventricular activation, all lead systems showed similar activation patterns (Table 2).

For the atria the inverse procedure identified similar activation sequences for all lead systems ($\rho > 0.97$; Table 2), even when only the six independent Frank lead signals or the eight independent signals from the 12- lead system were used. This result is explained by the 'simple' activation originating from the sinus node and propagating more or less with uniform velocity through the atria [11]. The same holds true for 'simple' ventricular activations, such as for the ectopic beat, or the beat originating at the Kent bundle only.

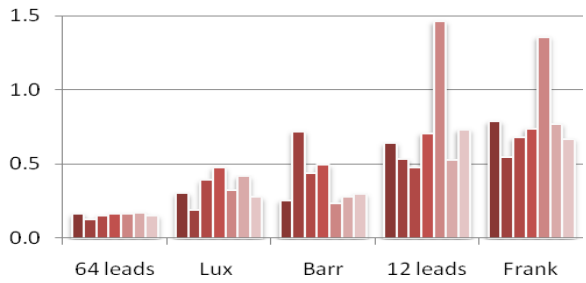

Figure 3. The rd values between values of the measured 64-lead ECG (QRS inverval) and those based on the estimated activation sequences. For each lead system (Table 1) seven cases are shown. From dark to light (left to right): a) Normal atrial, and b) normal ventricular activation, c) Brugada baseline, d) Brugada peak Ajmaline, e)WPW fusion beat, f) WPW Kent bundle only activation and g) WPW patient ectopic activation.

For more complex activation patterns, such as those of normal ventricular activation (involvement of the His-Purkinje system) and – in particular – of the fusion beat, the reconstruction of the ventricular activation requires more leads.

The differences between the reference and the estimated sequences became larger when using the signals of the standard 12-lead system (eight independent signals) or the Frank lead system (six independent signals). This indicates that these lead systems do not contain enough information to recapture the complex activation sequences that involve the His-Purkinje system. Moreover, the ventricular NH beat resulted in a rather low correlation for the 12-lead system, which might be attributed to the fact that no posterior electrodes are incorporated in the 12-lead system.

For the Brugada patient the typical changes in the ECG during the Ajmaline provocation test were visible in the leads close to the right outflow tract (RVOT). This Brugada patient is most likely associated with defects in the electrical functioning in the myocardial tissue near the RVOT [12]. Consequently leads close to the RVOT have to be incorporated in the reduced leadset in order to be able to estimate an activation sequence similar to the reference activation. Obviously this is only the case for the Lux lead system.

The results in this paper show that the method used is robust for the reduction of the number of leads This robustness can be attributed to the use of the initial activation estimate based on the general electro-physiology of activation propagation [2].

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