Are 2 Electrocardiographic Leads Enough for Multilead Wave Boundary Location and QT Measuring?

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Abstract

Different latencies on the cardiac electric phenomena can be found across leads. Thus combining adequately the information provided by multiple leads is essential for the correct location of global wave boundaries. A multilead-VCG strategy for boundaries location, by constructing a transformed spatial lead obtained from 3 orthogonal leads and optimized for delineation improvement, has been previously proposed and validated [1]. The goal of this work was to study and to quantify the performance loss when just 2 orthogonal leads are used. Combination of 2 leads were considered, both using recorded Frank leads, leads synthesised using inverse Dower transformation, and principal component analysis. The errors in QRS onset and T wave end location and in QT interval measurement were evaluated over the PTB database files. Results indicate that 2 leads properly selected are enough to locate QRS onset but possibly not for T wave end.

1. Introduction

According to the dipolar hypothesis, the electrical activity of the heart can be approximated by a time-variant electrical dipole, called the *electrical heart vector* (EHV). The voltage measured at a given lead would be the EHV's projection into the unitary vector defined by the lead axis [2]. Choosing a particular lead for ECG delineation determines a point of view over the cardiac phenomena and different latencies on the waves' onsets and ends are found in different leads. Nevertheless, the onset and end of the cardiac electric phenomena are indeed unique, and therefore a global feature for all the leads. Thus combining the information from different leads is crucial to locate unique lead-independent waves' boundaries, which can be imperceptible in a particular lead.

The need for a multilead (ML) based strategy is frequently supplied by the use of post-processing decision rules that choose global marks from single-lead (SL) based sets of locations [3]. However this approach does not provide truly ML locations and it requires applying the SL methodology to a large number of leads.

A novel automatic ML strategy for ECG boundaries delineation was proposed and validated in [1]. The system obtains, from 3 orthogonal leads, a transformed spatial lead more fitted for the specific boundary delineation. SL delineation is applied to the new synthesized lead, using a wavelet transform (WT) based system [3]. Thus the ML system allows to deal with multiple leads, taking advantage of their availability to further improve the delineation. The ML algorithms developed are general and can be applied to any orthogonal lead set, real or derived. In particular, it was validated considering the vectocardiographic (VCG) system given by the recorded Frank leads (F), leads synthesised from the standard 12-lead system using inverse Dower transformation (D) and a subset of the 12-lead standard system [1]. Using 3D WT loops the ML system provided more robust and more accurate boundaries locations than any ECG lead by itself, outperforming strategies based in selection rules after SL delineation. Best performance was obtained considering lead set F.

Considering WT loops in a 2D plane instead of in a 3D space is also possible, allowing to apply this methodology to any 2 orthogonal leads, if the only available (e.g. in a 2-lead Holter recording). The hypothesis is that the loss in performance will be negligible since normal VCG loops are very planar [4]. The goal of this work is to study and to quantify the performance loss when just 2 leads are used.

2. Methods

The SL and ML based delineation systems are described in detail elsewhere [1, 3] and only general features are referred in the next subsections. The implementation used in this work is a revised version with minor modifications, with respect to the implementations used in those papers.

2.1. Single-lead delineation

The WT provides a description of the signal in the time-scale domain, allowing the representation of its temporal features at different resolutions (scales) according to their frequency content. Thus, regarding the purpose of locating different waves with typical frequency characteristics, the WT is a suitable tool for ECG automatic delineation.

The detection of the fiducial points is carried out across the adequate WT scales, attending to the dominant frequency components of each ECG wave: QRS waves correspond to a simultaneous effect in scales 2^1 to 2^4, while the T and P waves affect mainly scales 2^4 or 2^5.

The prototype wavelet used (a derivative of a smoothing function) allows to obtain a WT at scale 2^m, $w_{x,m}[n]$, proportional to the derivative of the filtered version of the signal $x[n]$ with a smoothing impulse response at scale 2^m. Thus, ECG wave peaks correspond to zero crossings in the WT and ECG maximum slopes correspond to WT's maxima and minima. Depending on the number and polarity of the slopes found, a wave morphology is assigned and boundaries are located using threshold based criteria.

2.2. Multilead delineation

According to the dipole hypothesis, the VCG is an EHV's canonical representation defined by 3 orthogonal leads [2], usually acquired as the corrected Frank leads $\mathbf{s}[n] = [x[n], y[n], z[n]]^T$. As a consequence of the WT prototype used, the spatial WT *loop* in a time window W

$$\mathbf{w}_m[n] = [w_{x,m}[n], w_{y,m}[n], w_{z,m}[n]]^T \tag{1}$$

for a given scale $2^m \big|_{m\in\{1,2,3,\ldots\}}$ is proportional to the VCG derivative and describes the velocity of evolution of the EHV in W. Assuming that the noise is spatially homogeneous, the direction with maximum projection of the WT in the region close to the wave boundary location would define the ECG lead maximizing the local SNR, and thus, the most appropriate for boundary delineation. The main direction $\mathbf{u} = [u_{\mathrm{X}}, u_{\mathrm{Y}}, u_{\mathrm{Z}}]^T$ of EHV variations in a scale 2^m on any time interval W is given by the director vector of the best straight linear fit to all points in the WT loop $\mathbf{w}_m[n]$. By choosing adequately the time interval W it is possible to find the $\mathbf{u}$ corresponding to the lead most suited for delineation purposes.

Considering the VCG loop $\mathbf{s}[n]$ in any time interval I, a *derived* lead $d[n]$ defined by the axis $\mathbf{u}$ and combining the information provided by the 3 leads in $\mathbf{s}[n]$, can be constructed by projecting the points of the VCG loop. Instead, the WT loop ($\mathbf{w}_m[n]$) can be projected and a *derived wavelet* signal $w_{d,m}[n]$, corresponding to the ECG lead defined by the axis $\mathbf{u}$ is constructed, as:

$$w_{d,m}[n] = \frac{\mathbf{w}_m^T[n].\mathbf{u}}{||\mathbf{u}||},\ n \in I. \tag{2}$$

The strategy proposed for ML boundary delineation using WT loops is based in a multi-step iterative search for a *better* spatial lead for delineation improvement (with *steeper* slopes), particularized for each boundary. At each step (i), the vector $\mathbf{u}^{(\mathrm{i})}$ is determined *separately for each beat and boundary* by total least squares fitting. The interval W is defined specifically for each boundary and updated in a way to increase the SNR and ensure steeper slopes in $w_{d,m}^{(\mathrm{i})}[n]$ obtained by eq. (2). The goal is to construct a *derived wavelet* transformed signal well suited for boundaries location, using the same detection criteria as in the SL delineation.

2.3. Data and lead sets

Records from the PTB database (549 files at 1000 Hz, 12 standard leads + 3 Frank leads) were used for validation. A set of reference annotations for this data (one beat per file) was published in [5], consisting in manual annotations based in lead II, for QRS onset and T wave end.

In this work were considered the VCG systems given by the recorded Frank leads (lead set F) and the leads synthesised from the standard 12-lead system using inverse Dower transformation (lead set D). Combination of each pair of orthogonal leads were considered, defining the 2D lead sets: F_{XY}, F_{XZ}, F_{YZ}, D_{XY},D_{XZ}, D_{YZ}.

Principal component analysis (PCA) was used as alternative method for deriving orthogonal leads, as theoretically it is the optimum transform in least square sense. PCA is mathematically defined as an orthogonal linear transformation such that the greatest variance by any projection of the data comes to lie on the new first coordinate, called the first principal component (PC), the second greatest variance on the second PC, and so on [6]. For $\mathbf{\Sigma}$ the covariance matrix of the random vector that generates the signal $\mathbf{s}[n]$, the k^{th} PC is given by $p_k[n] = \mathbf{b}_k^T\mathbf{s}[n]$, where $\mathbf{b}_k$ is the eigenvector of $\mathbf{\Sigma}$ corresponding to the k^{th} largest eigenvalue. For real data, the unknown matrix $\mathbf{\Sigma}$ is replaced by the sample covariance matrix $\mathbf{S}$.

The standard 12-lead system is a redundant under the dipole model assumptions. Eight independent leads out of the 12-lead system are assumed, as the precordial leads are considered to describe also non dipole components. In this work, the first 3 PC, based on the 8 independent leads out of the 12-lead system, were considered to define a VCG system (lead set P). The 2D subsystem given by the first 2 PC define the lead set P_{2D}: $\mathbf{p}_{2D}[n] = [p_1[n], p_2[n]]^T$. In order to better describe the relevant information in the ECG, the matrix $\mathbf{S}$ is calculated over the samples between each QRS complex onset and T wave end, in all beats in the analised segment of signal. The intervals for obtaining $\mathbf{S}$ were based in the locations provided by SL delineation over lead II.

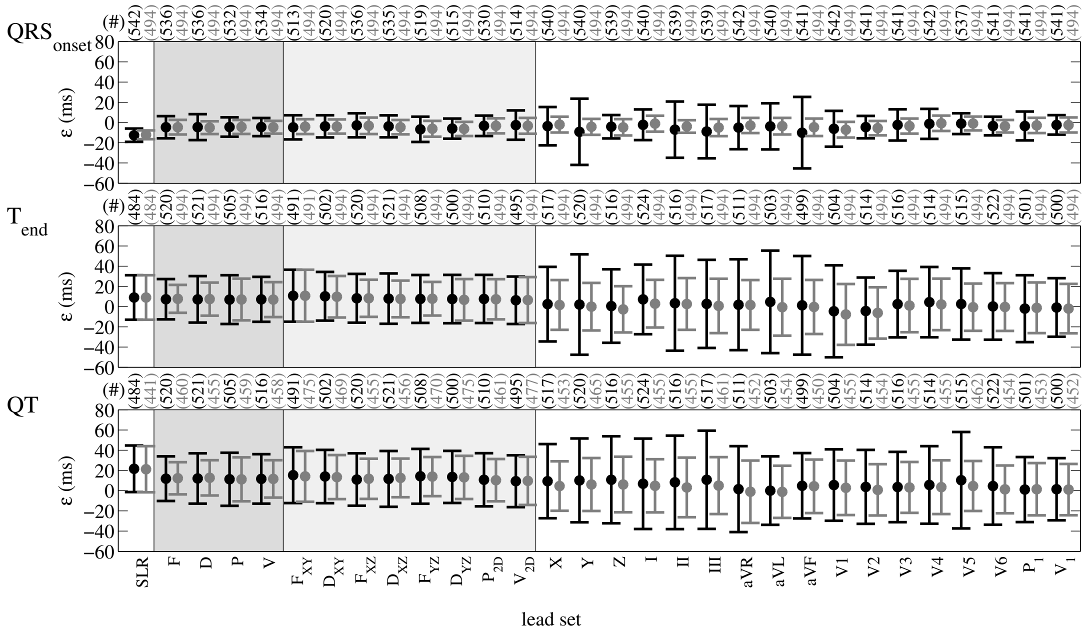

Figure 1. Results in PTBDB: error ϵ on QT boundaries location and QT length. The symbol # denotes the number of TP out of 542 reference marks provided (black) or after excluding extreme cases in each approach (grey).

Additionally, was also considered PCA based only in the precordial leads, together with the $aVl[n]$ lead, which corresponds to a spatial axis orthogonal to the transversal plane, defining the lead sets V and V_{2D}: $\mathbf{v}[n] = \left[p_1^V[n], p_2^V[n], aVl[n]\right]^T$ and $\mathbf{v}_{2D} = \left[p_1^V[n], aVl[n]\right]^T$.

3. Results and discussion

The errors in QRS onset, T wave end location and in QT interval measurement were evaluated over the PTB database files as overall mean (m) and standard deviation (s). The *detection* performance was evaluated calculating the *Sensitivity*: $Se = 100\frac{TP}{TP+FN}$, where TP is the number of true positive detections and FN stands for the number of false negative detections. The error (ε) was taken as the *automatic locations (measures) minus the respective referee mark (measure)*. Se and $m \pm s$ of ε across files were calculated considering all TP detections and also after the exclusion of the 10% most *extreme cases* for each boundary, more likely to be outliers.

Results for lead sets with 3 orthogonal leads (F, D, P, V) and with 2 orthogonal leads (F_{XY}, F_{XZ}, F_{YZ}, D_{XY},D_{XZ}, D_{YZ}, P_{2D},V_{2D}) can be found in Figure 1. For comparison purposes are also presented the values found for using a post processing decision rules over the 12-lead standard system (SLR) [3] and SL delineation over each of the 15 leads available, $p_1[n]$ and $p_1^V[n]$. The values of Se and $m \pm s$ for all 3D approaches and some 2D are also presented in Tables 1 and 2. Notice that for QT length the Se value over all TP detections corresponds to the values for T wave end. In Table 1 are also presented the tolerance values ($2s_{CSE}$) for the standard deviation of the error, according to the recommendations of [7]. All s values found for T wave end error using ML delineation and none for QRS onset are within the tolerance provided.

The error in T wave delineation represents the main contribution for QT measure error, as lower Se and higher standard deviation of the error (s) are found for T wave end than for QRS onset, across all methods, as result of the lower SNR. The best combination of 2 leads out of F or D are obtained for lead sets excluding the lead Y, as expected as this particular lead is likely to be noisy. The loss of performance is small for QRS onset but relevant for T wave end. Using PC based lead sets allows to reduce the s of the error for QRS onset, in accordance with previous results over different data [8], but not for T wave end location. A substantial performance lost is found for lead set $\mathbf{V_{2D}}$, with a much lower Se. Thus, directly recorded Frank leads (or the use of inverse Dower transformation to obtain those) should be preferred.

Globally, the results indicate that 2 leads properly selected are enough to locate QRS onset but possibly not for T wave end. This can indicate that the EHV changes at the beginning of the QRS are mainly along a single plane, while the T wave end comprehends spatial changes that require a 3D description. Nevertheless, the 2D approach allows to locate T wave end better than SL over any lead by itself or over $p_1[n]$ or $p_1^V[n]$. Notice also that an higher Se is achieved, specially compared with SLR.

The s of the error on QT length using 2 leads only increases around 4 ms for Frank leads, and around 2 ms under Dower transformation, compared with 3D approach.

	all files				after extreme exclusion			
	QRS onset		T end		QRS onset		T end	
	Se, %	$m \pm s$, ms	Se, %	$m \pm s$, ms	Se, %	$m \pm s$, ms	Se, %	$m \pm s$, ms
F	98.9	-4.7 ± 11.0	95.9	7.3 ± 20.0	91.1	-4.6 ± 7.3	91.1	7.7 ± 13.8
D	98.9	-4.5 ± 12.8	96.1	7.1 ± 23.0	91.1	-5.1 ± 6.3	91.1	7.4 ± 16.4
P	98.2	-4.4 ± 9.6	93.2	6.9 ± 24.2	91.1	-4.4 ± 6.9	91.1	7.0 ± 20.6
V	98.9	-4.1 ± 13.2	96.1	7.4 ± 21.8	91.1	-4.7 ± 6.2	91.1	7.4 ± 16.7
F_{XZ}	98.9	-2.8 ± 11.9	95.9	8.2 ± 24.1	91.1	-3.3 ± 8.1	91.1	8.2 ± 18.4
D_{XZ}	98.7	-3.8 ± 10.9	96.1	7.9 ± 24.9	91.1	-4.6 ± 7.1	91.1	7.6 ± 18.1
P_{2D}	97.8	-3.3 ± 10.1	94.1	7.6 ± 23.8	91.1	-3.3 ± 7.5	91.1	7.2 ± 19.9
V_{2D}	95.0	-2.8 ± 14.0	91.3	6.2 ± 23.0	91.1	-3.2 ± 7.8	91.1	6.5 ± 22.2
$2s_{\text{CSE}}$		± 6.50		± 30.6		± 6.50		± 30.6

Table 1. QT delineation results: Se and $m \pm s$ of the error for 3D and 2D approaches.

QT length	all files	after extreme exclusion	
	$m \pm s$, ms	Se, %	$m \pm s$, ms
F	11.8 ± 22.1	84.9	12.2 ± 15.9
D	12.1 ± 24.9	83.9	12.7 ± 17.4
P	11.3 ± 26.2	84.7	11.2 ± 21.8
V	12.0 ± 23.8	84.7	11.8 ± 17.6
F_{XZ}	11.0 ± 25.9	83.9	11.7 ± 19.9
D_{XZ}	11.7 ± 27.6	84.1	12.6 ± 19.1
P_{2D}	10.8 ± 26.3	85.1	10.4 ± 20.9
V_{2D}	9.5 ± 25.3	87.6	9.7 ± 23.3

Table 2. QT length measuring results: Se and $m \pm s$ of the error for 3D and 2D approaches.

Similar Se values are found with a bias reduction, still outperforming the performance of SLR. These results indicate than using ML method, even in the case of only 2 available orthogonal leads, is preferable to SL delineation for QT measuring.

4. Concluding remarks

The VCG based ML delineation methods considered in this work allows to take advantage of the spatial information provided by multilple leads, in order to better establish the onset and end of the cardiac electrical phenomena. Using 2 orthogonal leads instead of 3 reduces the stability of the locations provided, specially with respect to the T wave end. Nevertheless it still performs better than SL delineation and allows to measure the QT length over more beats, with more stable measuresand less biased marks than post-processing rules over SL based marks.

Acknowledgements

This work was partially supported by project TEC2007-68076-c02-02 from MCyT and FEDER, Grupo Consolidado GTC from DGA T:30, and *CMUP*, financed by *FCT*, Portugal, through the programmes POCTI and POCI 2010, with national and European Community Structural Funds.

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