ECG automatic delineation using a wavelet-based multilead approach

R. Almeida[1], J. P. Martínez[2],A.P. Rocha[1], P. Laguna[2], S. Olmos[2]

[1] *Departamento de Matemática Aplicada, Faculdade de Ciências, Universidade do Porto, Portugal*
[2] *Communications Technology Group, Aragon Institute of Engineering Research, University of Zaragoza, Spain*

Introduction

The *electrocardiogram* (ECG) has been extensively used as a diagnostic tool in clinical medicine for more than 80 years. Electrocardiographic signals are measured by means of two electrodes placed at the body surface, recording the electrical field from active heart as a function of time. The simultaneous recording of the ECG on different axes (leads), allows a spatial perception of the cardiac events. The most widely used lead set in clinical practice is the standard 12-lead system, with only 8 truly independent leads. Other alternative lead sets have been proposed and used for clinical and investigation purposes: from orthogonal 3-lead systems, as the corrected Frank system based on the body axes, to the very redundant body surface mapping using as many as 80 or 120 leads. The theoretical basis for using a 3-lead system is the dipole hypothesis, stating that the electrical activity of the heart can be approximated by a time-variant electrical dipole, called the *electrical heart vector* (EHV). Thus, the voltage measured at a given lead would be merely the projection of the EHV into the unitary vector defined by the lead axis. The 3 dimensional record obtained by an orthogonal system is itself a canonical representation of EHV known as *vectocardiogram* (VCG). Although the electrical activity has some non-dipolar components, this model is widely used to interpret spatially the ECG. According to this hypothesis, any hypothetical lead can be synthesized by an adequate projection of the EHV. Moreover, it provides the basis for linear transformations between the standard 12-lead system and the orthogonal systems (Levkov 1987) .

The delineation of ECG characteristic (P, Q, R, S and T) waves associated with each cardiac beat consists of detecting their peaks and boundaries (Fig. 1), supplying fundamental features for extracting clinically useful information (duration and amplitude of electrical phenomena). Delineation of low-amplitude, smooth waves, like P and T waves, is a particularly challenging task due to the low signal-to-noise ratio. Specially problematic is the delineation of flat boundaries as it is usually the case of T wave end. Developing accurate and robust methods for automatic ECG delineation is a topic of main interest, in particular for the analysis of long records. When multiple leads are available, some delineation approaches proposed a rule to select one of the single-lead measurements as a multilead strategy (de Chazal & Celler 1996, Laguna, Jané & Caminal 1994, Martínez, Almeida, Olmos, Rocha & Laguna 2004). In this work, we hypothesize that a robust delineation can be performed by attending to the spatial characteristics of the different available leads. We propose a 3D approach for a multilead extension of a previously validated single-lead delineator based on the Wavelet Transform (Martínez et al. 2004).

Methodology

Single lead delineation

The Wavelet Transform (WT) decomposes a signal $S(t)$ as a combination of a set of basis functions, obtained by means of dilation (a) and translation (b) of a single prototype wavelet:

$$W_a^S(b) = \frac{1}{\sqrt{a}} \int_{-\infty}^{+\infty} S(t)\, \psi\left(\frac{t-b}{a}\right) dt,\ a > 0. \tag{1}$$

Lower frequency components of the signal are characterized by the coefficients corresponding to wider basis function resulting of higher scale factor a, and vice versa. This way, the WT provides a time-scale domain description of the signal, allowing the representation of temporal features at different resolutions, according to their frequency content. For a discrete signal $S(n)$, the Discrete WT (DWT) is usually obtained following a dyadic grid on the time-scale plane: $a = 2^k$ and $b = 2^k l$, $k \in N, l \in Z$. The DWT is equivalent to an octave filter bank (Mallat 1989) and we implemented it using the *algorithm à trous*: a

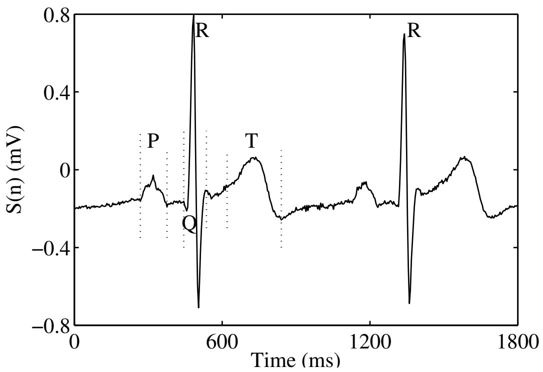

Figure 1: Characteristic wave delineation of a cardiac beat including P,Q,R S and T waves.

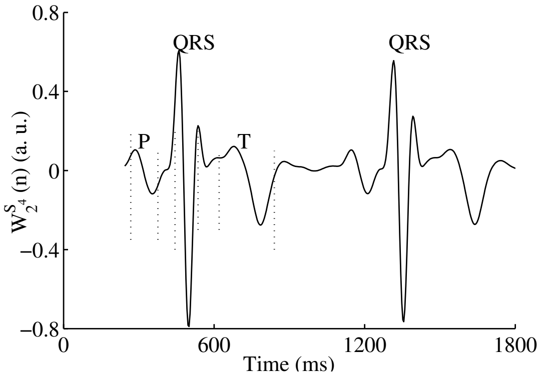

Figure 2: WT of the cardiac beat in Fig. 1: local maxima and zero crossings result of wave peaks and slopes, respectively.

cascade of identical cells (low-pass and high-pass FIR filters) with an interpolation of the filter impulse responses at each stage, keeping the same sampling rate at different scales. A single-lead ECG delineation system based on the WT was developed and evaluated (Martínez et al. 2004) and proven to be quite robust against noise and morphological variations, outperforming other published approaches. The QRS complex location is the first and the most straightforward stage of any delineation system and single lead performs very well in its determination. We used as prototype wavelet a quadratic spline that corresponds to a derivative of a low-pass smoothing function, resulting that $W_a^S(n)$ is proportional to the derivative of the filtered version of the signal $S(n)$ with a smoothing impulse response at scale a (Mallat 1989). Regarding our application, this property is very convenient as we are interested in detecting ECG waves, which are composed of slopes and local maxima (or minima): wave peaks correspond to zero crossings and slopes in ECG correspond to maxima and minima of the WT, as illustrated in Fig. 2.

For each beat i, each wave is delineated within a specific search window relative to the previously detected QRS complex waves. The wave morphology is reflected in the number and polarity of relevant local maxima of $\left|W_a^S(n)\right|$. The wave onset (end) occurs before (after) the first (last) significant slope in the search window, corresponding to the last $\left|W_a^S(n)\right|$ maximum, p_{S_i}. The wave onsets (ends) are detected by selecting the nearest sample to the peak where the WT satisfies one of the following criteria: **i)** the sample where $\left|W_a^S(n)\right|$ is below a threshold ξ relative to the amplitude of p_{S_i}; **ii)** a local minimum of $\left|W_a^S(n)\right|$ before (after) p_{S_i}. Peaks and boundaries detection was carried out across the adequate WT scales according with the dominant frequency components of each ECG wave. For example, QRS waves were found as a simultaneous effect in the scales 2^1 to 2^4, while T wave detection is in the scale 2^4 or 2^5. Noise and artifacts were avoided considering their different contribution at various scales.

Multilead delineation

The multilead extension of the WT based system consists in considering simultaneously the orthogonal Frank leads X, Y and Z; orthogonal leads can also be synthesized from the standard lead system, if they are not available. In this preliminay work, we concentrated in the more problematic T wave end delineation, but analogous strategies can be adapted for all the other points. For each beat i, a T wave search window T_{w_i} was defined taking into account the cardiac frequency (Martínez et al. 2004). The loop in space (Fig. 3) defined by $\mathbf{L}_i(n) = [W_{2^4}^X(n), W_{2^4}^Y(n), W_{2^4}^Z(n)]^T$, $n \in T_{w_i}$, with lead $E \in \{X, Y, Z\}$ corresponds to the T wave derivative and describes the EHV evolution. Taking $\mathbf{U1}_i = [U1_i^X, U1_i^Y, U1_i^Z]^T$ as the director of the best line fit in Total Least Squares sense to all points in the loop, we define a derived signal $D1_i(n)$ by projecting all the points the loop over the direction of the vector $\mathbf{U1}_i$:

$$D1_i(n) = \frac{\mathbf{L}_i(n)^T.\mathbf{U1}_i}{||\mathbf{U1}_i||}. \tag{2}$$

This way we construct a new wavelet signal (Fig. 4) that corresponds to the ECG lead defined by the axis $\mathbf{U1}_i$, the main direction of EHV in T_{w_i}. As the T end occurs after the last WT modulus maximum (p_{D1_i}), the direction $\mathbf{U2}_i$ given by the best line fit to the loop points after p_{D1_i} corresponds to the main

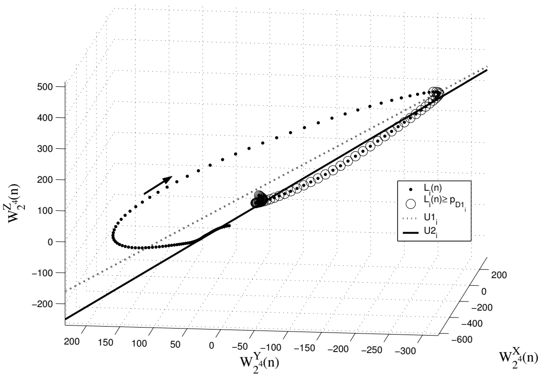

Figure 3: T wave VCG loop and chosen directions (arrow indicates time direction).

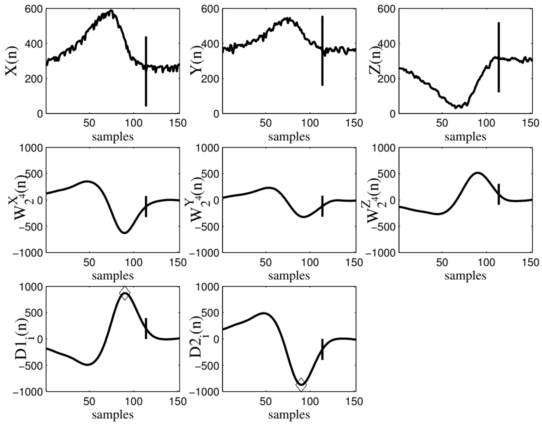

Figure 4: ECG in orthogonal leads (X, Y, Z), correspondent WT and the new derived signals ($\diamond$ indicates p_{D1_i} and p_{D2_i}, line stands for *median referee* T end mark).

direction of EHV on the latter part of the T wave. Defining a new derived signal $D2_i(n)$ by projecting all the points in the loop over the direction of the vector $\mathbf{U}2_i$, we obtained a wavelet signal corresponding to the lead parallel to the T wave end. Therefore, $D2_i(n)$ is likely to present a single slope, very steep, after the last WT modulus maximum p_{D2_i}, optimal for T end detection. Finally T wave end is detected applying the single lead delineator to the lead defined by $\mathbf{U}2_i$.

Results and Discussion

The multilead system extension was evaluated on manually annotated records from CSE multilead measurement database, developed for ECG wave recognition and measurement testing (Willems et al. 1987). The signals include the 12 standard and the 3 Frank leads, at 500 Hz sampling rate. It provides for 27 beats *median referee* T end annotations from 5 cardiologists after outlier rejection.Taking the error (ε) as *median referee* minus the automatically detected T end, we evaluated its mean (m_ε) and standard deviation (s_ε), and also the mean ($m_{|\varepsilon|}$) and standard deviation ($s_{|\varepsilon|}$) of the absolute error ($|\varepsilon|$).

The 3D multilead approach found the T end for 25 out of 32 beats, while in each lead separately it was found for between 25 and 27 beats. As illustrated in Fig. 5, 3D multilead (M) outperformed single-lead based delineation for any of the 15 available leads, with exception of m_ε in 3 leads. The mean error was found to be quite sensitive to the threshold ξ. Best performance was found increasing slightly ($\xi = 3.5$) the value used in single lead delineation ($\xi = 2.5$). In fact, single lead delineation parameters need to be adequated to any possible lead, while in multilead approach T wave end is always detected in a potentially optimal signal, with one very steep slope after p_{D2_i}, what allowed a specific tunning of ξ. Basing the lead direction choice in the $W_{2^4}^E$ loop, instead of the VCG, allowed to avoid the higher frequency noise contamination, allowing a more accurate selection.

The results obtained with our 3D multilead delineation are also presented in the Table 1, together with multilead delineation systems based on post processing rules. A single T wave end location was obtained from the possible different annotations considering 12 or 15 leads, by applying the following rule: single lead annotations are sorted and the end of the wave selected as the last annotation whose k=*4* nearest neighbors lay within a δ = 12 ms interval (de Chazal & Celler 1996, Laguna et al. 1994, Martínez et al. 2004). To make a similar comparison considering X, Y and Z leads, T wave end

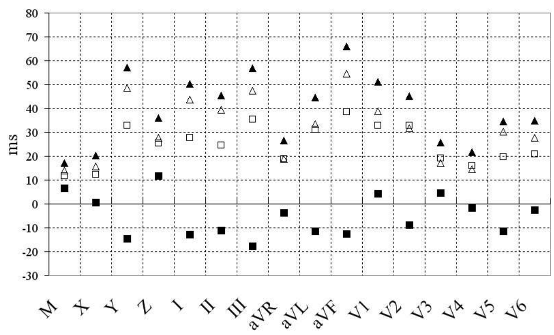

Figure 5: Errors in T wave end delineation: 3D multilead approach (M) versus single lead, leads X to V6 ($\blacksquare$ m_ε, $\blacktriangle$ s_ε, $\square$ $m_{|\varepsilon|}$, $\triangle$ $s_{|\varepsilon|}$).

location was taken as the median mark single delineation. In spite of the lower m_ε obtained using 12 or 15 leads, the lower $m_{|\varepsilon|}$ in multilead results represents higher accuracy. The multilead approach presented also lower values in both s_ε and $s_{|\varepsilon|}$, clearly denoting a performance improvement. According to (The CSE Working Party 1985) recommendations *"the standard deviation of the differences [of a program results] from the reference should not exceed certain limits as listed in Table 2"*, which for T wave end case is $s_{CSE} = 30.6$ ms. This value was obtained as two standard deviations of the differences (in ms) between the median of the individual readers and the final referee estimates; therefore is not consensual if an algorithm should accomplish $s < 2s_{CSE}$ (loose criterion), or $s < s_{CSE}$ (strict criterion). Loose criterion is fulfilled by all the strategies in Table 1 and the multilead approach is the one nearest to accomplish the strict criterion. Remark that the proposed method requires the WT calculation of 3 leads, with delineation procedures involving 2 signals (D_{1_i} and D_{2_i}) what, even considering fitting and projecting features, is likely to be more efficient that applying the complete process to 12 or 15 leads.

	3D multilead	selection rule: 3 leads	selection rule: 12 leads	selection rule: 15 leads
(#)	(24)	(27)	(24)	(26)
$m_\varepsilon \pm s_\varepsilon (ms)$	6.7 ± 17.3	8.5 ± 25.5	0.9 ± 22.4	1.3 ± 21.8
$m_{\|\varepsilon\|} \pm s_{\|\varepsilon\|}$ (ms)	11.8 ± 14.1	18.9 ± 18.9	13.11 ± 18.1	13.2 ± 17.2

Table 1: Delineation results in the CSEDB: Comparison between 3D multilead and multi lead based in selection rules (# stands for number of annotations).

Concluding Remarks

We have proposed a multilead strategy for ECG automatic delineation, based on 3 orthogonal leads and using Wavelet Transforms. This approach represents an alternative to multilead delineation systems based on post processing rules over the locations obtained independently from each of the leads. A preliminary version for detection of the T wave end was validated over a standard manual annotated database. The results pointed out that the proposed methodology is adequate for ECG waves boundaries detection, providing more robust and accurate locations than any electrocardiographic lead by itself. Moreover, it outperformed other multilead strategies based in rule selection after single lead delineation.

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