WAVELET TIME ENTROPY AND T WAVE MORPHOLOGY

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ABSTRACT

It is believed that one can identify myocardial ischemia from an ECG. A widely accepted indicator is based on a significant shift of the ST segment [1]. This indicator doesn't take into account possible changes in the morphology of the ECG signal during the ventricular depolarization or depolarization and therefore might be a less sensitive measurement. During controlled coronary artery occlusions on five pigs, data were collected including an orthogonal ECG and hemodynamic signals. From these experiments where the heart rate was kept constant by atrial pacing, a good model is available to compare normal states from ischemic ones. Wavelets allow to efficiently represent a signal in a time-frequency plane [2]. Using such a representation, one can compute the entropy of the signal at various frequency levels (*Wavelet Time Entropy*) and thus find an optimal measure to differentiate normal states from ischemic ones. This new indicator is *independent* from the ST segment amplitude and yet provide a conclusive detection of the ischemic states in the collected data.

ENTROPY AND CELLULAR SYNCHRONISATION

Application of the entropy concept to ST segment / T wave analysis can be stated as follows. From a cellular point of view, the T wave is a summation of *out-of-phase localised* myocardial action potential. If time dispersion of T waves is increased, as it is assumed to be the case during ischemia, synchronization between these superposed waves is lost. It is then hypothesized that a good indicator of synchronization (and ischemia) would be the entropy of the position in time of each superposed wave. Ischemia would then be detected by a significant increase in this entropy.

Unfortunately, it is highly improbable that we will ever be able to directly measure this entropy by non-intrusive methods because i) the myocardial action potential waveforms are unknown; ii) even if we knew the waveforms (from eventual numerical computations for example), they will most probably not constitute a mathematical basis even if a weak sense (a *frame* for example), hence they could not be used directly to decompose mathematically and analyze the signal. Therefore, we have to approximate this entropy. In practice, it is often observed that in a time-frequency representation (as computed with wavelets) localization increase along with the frequency up to a level of noise saturation. Selecting this frequency level makes *time entropy* computations optimal.

One can compute the *wavelet coefficients entropy* at different scales (*time entropy*). Through a *multiscale* approach like the wavelet transform, one is able to automatically and easily detect the relevant scales (and frequencies) without *a priori* choices. Moreover, it allows to compare various scales (or frequencies) in order to differentiate various cases.

DATA PROCESSING

Data processing steps are presented in Figure 1. The data was efficiently compressed and denoised using the *Discrete Karhunen–Loève Transform*. Indeed, on the orthogonal ECG's used for this study, keeping only the R_{off}-T_{off} interval, an average of 95 % of the energy was projected onto the principal vector. It is assumed that the remaining energy comes mostly from noise in this particular experiment.

Only the values between R_{off} and T_{off} are kept and each heartbeat is processed separately. Once the signal has been properly denoised using the FFT and moving median filtering, the Fast Wavelet Transform (FWT) is applied on the last 226 ms. The FWT is computed using Cohen-Daubechies-Feauveau (CDF) wavelets adapted to short signals. The result is 4 different sets of wavelet coefficients; each one, corresponding to a different scale. On the first scale corresponding to the highest frequencies, there are 56 coefficients, then on the following scales, there are respectively 28, 14 and 8 coefficients for a total of 106 wavelet coefficients. The remaining 7 coefficients resulting from the repeated application of the CDF dual lowpass filter are discarded (information such as the ST segment amplitude is mostly contained in these low frequency coefficients). The entropy of each set of coefficients at each scale (see below) is then computed and corresponds respectively to at most 1/2, 1/4, 1/8, and 1/16 the sampling frequency (approximately) so that the corresponding frequency range are

approximately 125 Hz to 250 Hz, 60 Hz to 125 Hz, 30 Hz to 60 Hz, and 15 Hz to 30 Hz respectively. We define the *Wavelet Time Entropy* (*WTE*) by

$$WTE_s = \sum_k \left|w_{s,k}\right| / \sum_i \left|w_{s,i}\right| \times \ln\left(\left|w_{s,k}\right| / \sum_i \left|w_{s,i}\right| \right) \quad (1)$$

where the wavelet coefficients at scale *s* are noted $\left\{w_{s,k}\right\}_k$.

RESULTS

From a mathematical point of view, these indicators (WTE_1, WTE_2, WTE_3, and WTE_4) are independent of the ST segment amplitude. WTE_3 can serve to differentiate convincingly the occlusion state from the control state even though more data would be needed to study and possibly set a precise threshold. One observes that for some of the pigs, there is a delay (about 10 minutes) between the occlusion and a significant increase in WTE_3. Lower frequencies (WTE_4) also seem to be of interest even though in a less convincing manner than WTE_3. Overall, ST-VM at J+60 ms gives similar results to WTE_2, however ST-VM seems to provide earlier detection. It might suggests that cellular desynchronization, as a consequence of ischemia, is a slower process and may begin only after a change in the ST segment amplitude has taken place (see Figure 2).

CONCLUSION

The Wavelet Time Entropy indicators show that more information in an ECG concerning myocardial ischemia is contained in the ST segment - T wave different than the ST segment amplitude (at J+60 ms). It is shown that information is contained in the distribution of the energy (entropy) at specific frequency ranges (below 125 Hz) independently of possible energy transfers between frequencies.

[1] Badir BF, LeBlanc AR, Nasmith JB, Palisaitis D, Dubé. B, Nadeau R, J Electrocardiol., Vol 30(3):175-187, 1997.
[2] Gramatikov B, Thakor N,. Proc. 15th Intern. Annual Conf. IEEE/EMBS, pp. 731, San Diego, Oct 1993.

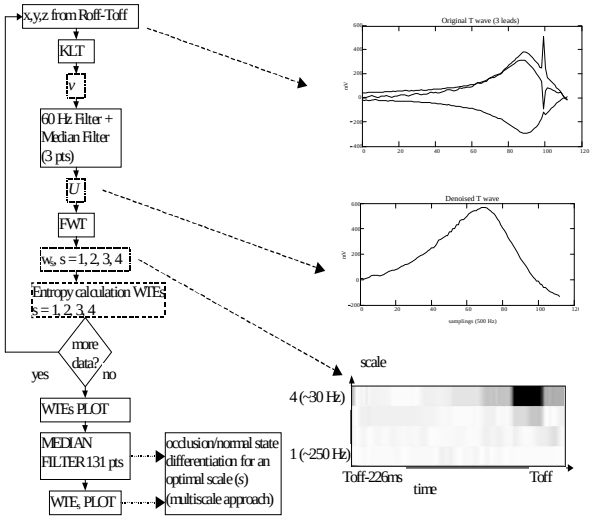

Figure 1 : General algorithm starting from orthogonal ECG data and finishing with a time-frequency representation. Through a multiscale approach, one is able to automatically select a frequency range (scale) to differentiate the ischemic states.

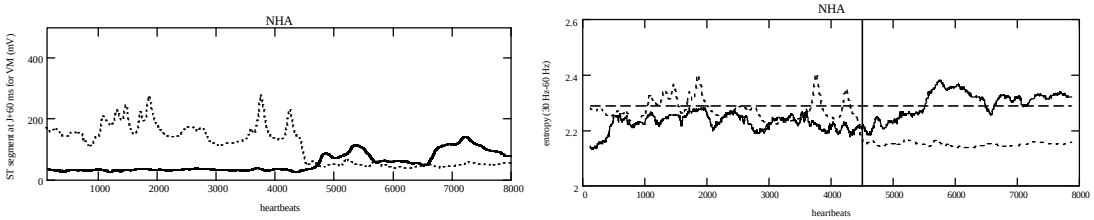

Figure 2 : Comparison between ST-VM at J+60ms and the WTE_2 (30 Hz to 60 Hz). Blood flow is indicated by a dotted line.