COMPRESSION AND DENOISING OF THE ECG USING STANDARD BANDWIDTH VARIABILITY FUNCTION

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this work is supported by State Committee for Scientific Research grant no: 8 T11E 007 17

Summary

In one of our previous work we invented the function of local bandwidth variability of an electrocardiogram. This function represents local density of diagnostic information, and thus expresses the level of importance of any section in the recorded signal. After a year of validation and improvement, we use the standard bandwidth variability function (SBVF) of an ECG for serious modification of the signal in time-frequency domain: denoising and compression. The obtained results, however only partially verified till now, are quite promising. An example compression ratio is of order of 5.9 with distortion not exceeding 5%.

1. INTRODUCTION

The initial conception of our work was the non-equal distribution of information in an electrocardiogram. It seems obvious, but is not frequently considered, that some segments of the signal are more important for a doctor than the remaining parts. That may be true for a wide range of biomedical signals. In case of electrocardiograms, the developed technology of automated extraction of segments and waves, at least P, QRS and T waves, motivated us to work out the function representing the local importance of the signal with reference to these waves start- and endpoints.

Finding the right representation of information distribution in electrocardiogram was our main scientific interest, and at present several methods are implemented and used to estimate the local diagnostic importance of an ECG signal.

- local bandwidth of a typical electrocardiogram [1],
- local susceptibility to distortion caused by random cancelling of time-frequency coefficients [2],
- average time devoted by an eyeball of an interpreting person to analyse each segment of the electrocardiogram.

Unfortunately, the last method needs a series of experiments on doctors with use of an eyeball positioning system and is very time consuming. The first results are thus expected not earlier than in June 2000. Therefore, as soon as we stated the convergence of the first two methods above and confirmation from our medical co-workers, we tried to verify their usefulness in practical implementation.

As we studied the methods of signal de-noising or compression in time-frequency domain (i.e. those proposed in the Matlab Wavelet Toolbox), we stated that the temporal information is always neglected. The cut-out levels differ in individual frequency bands, but their values are always constant in time. That approach is adequate only when assuming the equal or unknown distribution of information along the time axe. Otherwise, if there are some parts of signals more important than the rest, compression or de-noising can be improved when using the time-variable threshold function for every frequency band. In more important sections, where information is more susceptible to distortion, only slight modifications of signal are allowed. In consequence, the output signal is locally very close to the original, and the information remains unchanged. In the remaining part of signal, considered as less important, more invasive data reduction is performed, that results in significant compression ratio. The only signal-specific problem is the correct distinction of important segments or the event-based function of information distribution.

Fortunately, in case of electrocardiograms, the signal can be temporally divided into components representing consecutive stages of heart's beats. Consequently, the function of information distribution should rather refer to the sections start- and endpoints than to time, because of heart rate variability. Another advantage is the ability of automated P, QRS and T wave detection offered by contemporary ECG-dedicated processing technology.

2. MATERIALS AND METHODS

2.1. Standard bandwidth variability function

Having considered all features of both functions of information distribution mentioned above: local bandwidth of a typical electrocardiogram and local susceptibility to distortion, we finally decided to use the instantaneous, or local, bandwidth of ECG as source of a priori knowledge about the expected nature of an ECG. The main advantage of wavelet-based local bandwidth function is its definition in time-frequency domain, while the concurrent local susceptibility function is defined in time domain.

When comparing wavelet-based to wavelet packet-based local bandwidth function we stated that in case of wavelet decomposition, thanks to exponential frequency scale, the number of time-frequency plane coefficients essential to correct reconstruction of the signal is much lower. That results directly in larger compression ratio, of order of 5 obtained with wavelets, while only 3 with wavelet packets.

Figure 1a displays the function for an average P-QRS-T segment on the wavelet time-frequency plane, and figure 1b, the function for an average P-QRS-T segment on the wavelet packet time-frequency plane. Figure 2 displays the function of local susceptibility to distortion caused by random cancelling of time-frequency coefficients.

2.2. Choice of the mother wavelet

Provided the instantaneous bandwidth is a function in t-f domain, the use of a reversible t→t-f transform is necessary. Although many t-f transforms are widely known, for this

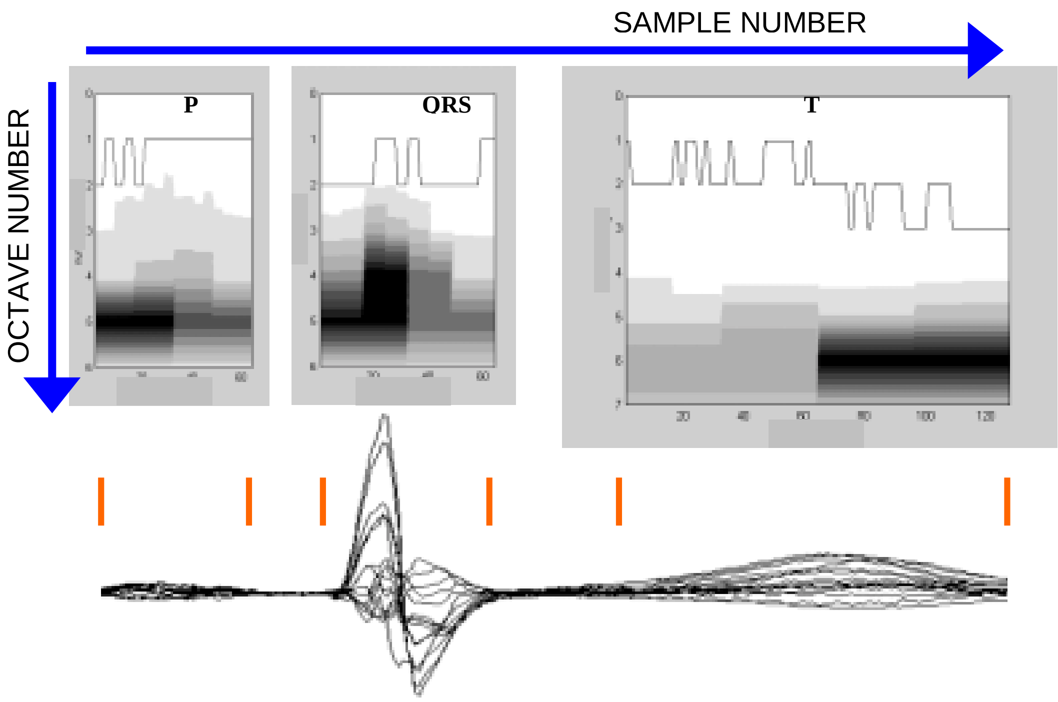

Fig. 1a Averaged normalized time-frequency planes (wavelets) of main components of heart beat along with multilead signal in time domain; black lines separate coefficients representing less than 5% of instantaneous energy; sampling frequency is here 500 Hz.

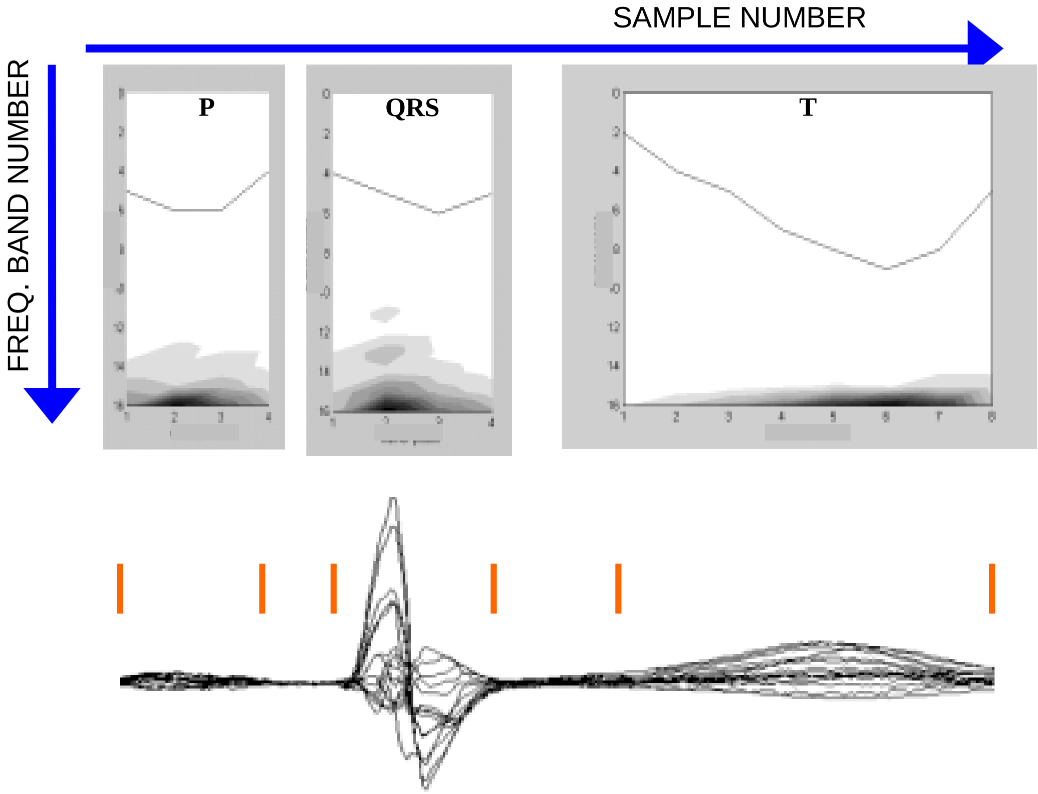

Fig. 1b Averaged normalized time-frequency planes (wavelet packets) of main components of heart beat along with multilead signal in time domain; black lines separates coefficients representing less than 5% of instantaneous energy; sampling frequency is here 500 Hz.

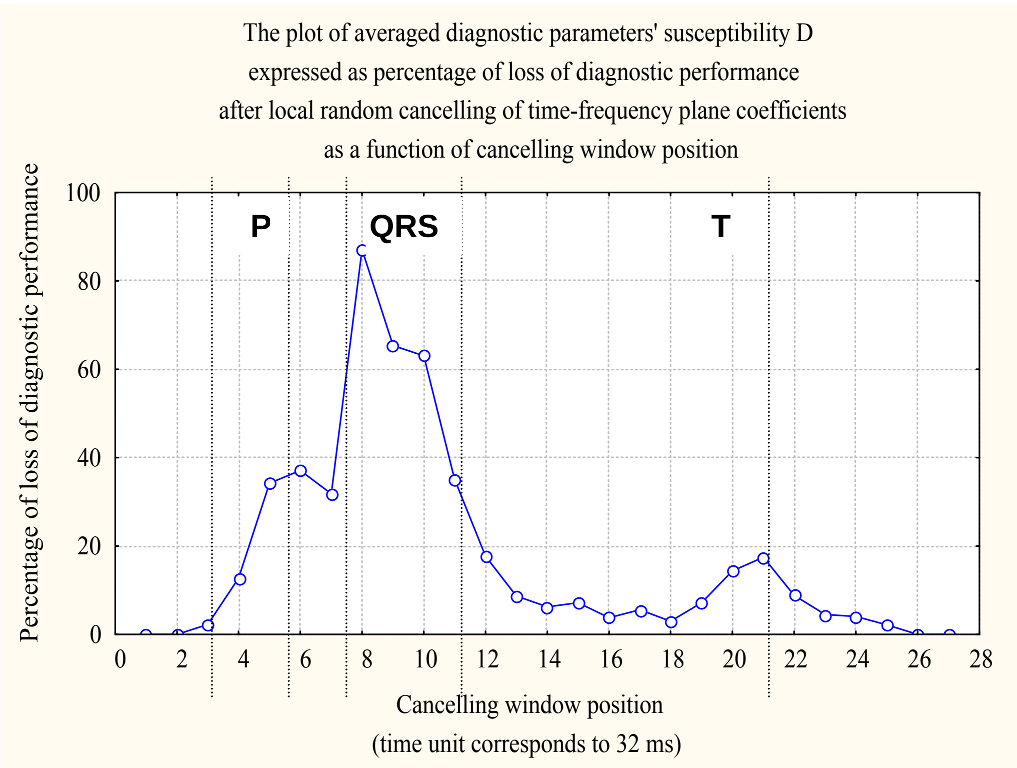

Fig 2 The function of diagnostic parameters' susceptibility to distortion caused by local random cancelling of time-frequency coefficients. Additionally average wave borders are marked.

particular task, the transform is expected to meet the following criteria:

- the condition of losslessness, that is necessary to control all signal properties in t-f domain and guarantees that every changes in the output signal result from the manipulations done in t-f domain,
- the use of filters with support compact and relatively short.

From a big choice of transforms widely used, we selected the 5-th order Daubechies wavelet transform [3]. The use of 5-th order filters is a compromise between the support length and frequency bands separation. The decomposition using pyramid decimation scheme is executed down to level 3. That extracts three high frequency octaves and the coarse approximation signal. The effective sampling frequency for the coarse approximation signal equals to 1/8 original sampling frequency that means 64 Hz.

2.3. Continuous noise estimation

Measurement of the noise level for denoising purpose and the level of background energy for compression can be performed in a nearly continuous way. The simplest approach is to perform these measurements on the baseline (i. e. P-Q section), which is the physiologically documented lack of electrical activity of the heart. All signals at the baseline are of extra-cardiac origin and thus can be considered as background activity. Since the baseline occurs once per beat and lasts for less than 100 ms, the measured value could be extrapolated for the remaining part of signal in time domain. This assumes the background activity to be stationary for a period of time 10 times longer than a measurement period, what is not always true. Another limitation of the relatively short baseline is that the noise measurement is possible in upper frequency bands only, for frequencies higher than the inverse baseline length. Thus for lower frequencies no denoising or compression can be made without estimating of the noise level in frequency domain. And our first proposal of compression was limited to the three highest octaves only, and assumed the background activity to be stationary.

After a year of investigations, we proposed an improved nearly continuous model of all background activities [4] based on:

- measurement performed on the baseline for three upper octaves,
- measurement performed on the remaining part of the signal, except for QRS complex, where the local bandwidth variability function tells that no signal components are expected in two upper octave,
- spectral 1-st order (linear) extrapolation of background activity for lower frequency ranges,
- temporal extrapolation of background activity for QRS complex section.

The principle of the nearly continuous noise model is displayed in the figure 3, and figure 4 contains some more details on spectral extrapolation.

Continuous noise estimation is the fundamental verification tool in both compression and denoising processes. The verification consists in continuous comparing the current signal time-frequency representation with its expected local energy contents. The expectations are expressed by the standard bandwidth variability function with reference to the waves start- and endpoints pre-detected in the ECG signal. The function of noise estimation is twofold:

- correcting of the standard local bandwidth accordingly to the current signal properties, often necessary for abnormal signals containing irregular waveforms (e. g. atrial fibrillation),
- expanding of the noise estimation down to lower frequency ranges since the standard bandwidth variability function was computed with use of a 3-stages decomposition due to the baseline length limitation.

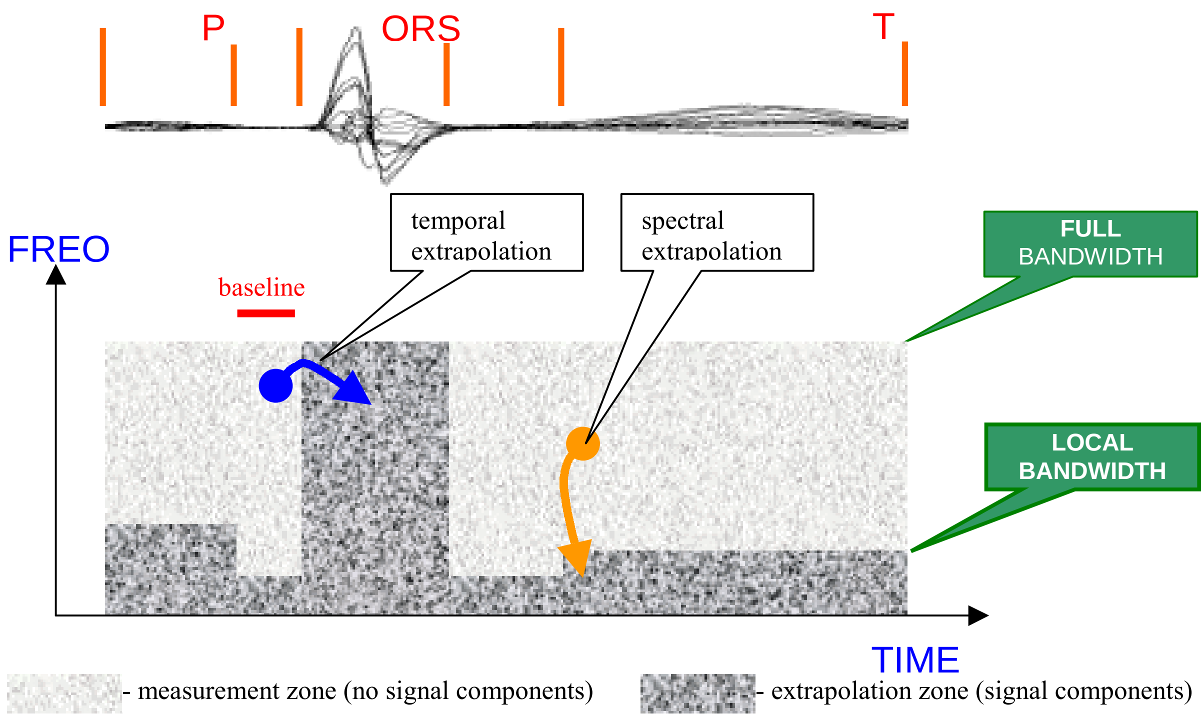

Fig. 3 The principle of the nearly continuous model of noise. Segmented example multilead heart beat is displayed along with the time-frequency plane split by the local bandwidth variability function into two zones: measurements and extrapolations. In the measurement zone (light grey) no signal components are expected, thus the parameters of the background activity model can be measured directly. In the extrapolation zone (dark grey) no measurements can be performed, since the signal components are expected. The parameters of the background activity model are updated by temporal extrapolation (in case of QRS occupying all frequency bands) or by spectral extrapolation compensating the baseline length limitation in frequency domain.

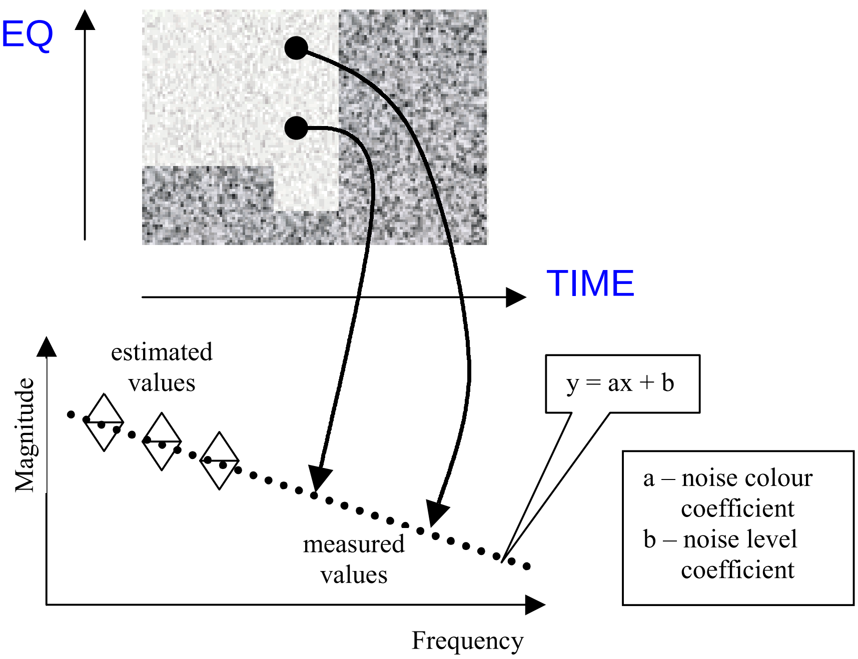

Fig. 4 Details on spectral extrapolation

3. TESTING THE COMPRESSION ALGORITHM

Numerical tests were performed for both: compression and denoising algorithms, but due to the difficulty in expressing the denoising efficiency adequately, only the compression results will be discussed hereby. For testing we used the well-known source of annotated ECG beats of different morphology, the CSE-Multilead Database [5] (sampling parameters: 500 Hz, 16 bit) that provides the exact start- and endpoints of P, QRS and T waves. That information made the results independent on the pre-processor performance. Knowing these points for each signal made possible to retrieve their closest references in the time-frequency (t-f) representation of an ECG. The effective count of tested signals was 99 for P-wave, 123 for QRS-complex and 103 for T-wave. The pacemaker triggered and noisy signals were not considered. Table 1 displays the average waves timing parameters of the test signals.

Table 1 Average lengths of ECG components (in samples, each representing 2 ms)

	RR	**P**	**QRS**	**T**
average	443.62	55.61	54.75	146.27
st. dev.	106.35	6.49	11.56	18.17
% of RR		12.5	12.3	24.0

4. RESULTS AND DISCUSSION

The compression efficiency can be estimated directly on time-frequency plane by counting the percentage of coefficients being essential to reconstruct the signal correctly that means without exceeding a given error level.
The number of essential t-f coefficients is given in table 2 for original and compressed signals. It makes clear that on average only 16.9 % of original sample number are necessary for correct signal reconstruction. Since all other samples are found redundant, the estimated compression ratio is 5.9.
The measurement of distortion is very difficult, since we contested the equal distribution of information in the signal and the commonly used PRD estimator does not represent the local variability of data stream. Although we insist on wrong representation of real distortion in the PRD factor, we used it for the reasons of compatibility with other referred compression methods.
The absolute maximum value of PRD reconstruction error was 5%, but the actual distortion of medical information contained in ECG was fairly inferior, because the compression is practically lossless on the P-QRS-T section.

Table 2. Number of essential samples for P, QRS and T wave.

samples	**P**	**QRS**	**T**	**izoline**	**entire RR**
original normalized	64	64	128	187	443
essential	27	19	23	6	75
% of orig.	42	29.7	17.9	3.2	16.9

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