Optimal Coding of Vectorcardiographic Sequences Using Spatial Prediction

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***Abstract*— The paper discusses principles, implementation details and advantages of sequence coding algorithm applied to the compression of vectocardiograms. The main novelty of the proposed method is the automatic management of distortion distribution controlled by the local signal contents in both technical and medical aspects. As in clinical practice, the VCG loops representing P, QRS and T waves in the three-dimensional space are considered here as three simultaneous sequences of objects. Because of the similarity of neighboring loops, encoding the values of prediction error significantly reduces the data set volume. The residual values are de-correlated with the DCT and truncated at certain energy threshold. The presented method is based on the irregular temporal distribution of medical data in the signal and makes advantage of variable sampling frequency for automatically detected VCG loops. The features of the proposed algorithm are confirmed by the results of numerical experiment carried out for a wide range of real records. The average data reduction ratio reaches a value of 8.15 while the PRD distortion ratio for the most important sections of signal does not exceed 1.1%.**

***Index Terms*— Data compression, Signal processing, Vectorcardiography**

I. Introduction

THE vectorcardiography (VCG) is a conceptual and methodological extension of the ECG and records the cardiac electrical field from the body surface for the purpose of three-dimensional imaging and analysis. In absence of clear interpretation guidelines, the VCG was underestimated for years, but recently, due to the use of numerically supported spatial transforms, it becomes more and more appreciated in clinical practice. Moreover, the vectorcardiogram provides full three-dimensional representation of body-surface electrical field. In consequence, the VCG is often considered for implementation in clinical and ambulatory equipment and concerned by the data compression requirements as far as the conventional 12-lead ECG. Up to the author's best knowledge, however, no VCG-dedicated coding method was proposed until today. Despite long history, the medical data compression is currently a frequently revisited research field especially in context of telemedicine, pervasive monitoring and home care.

Two recent approaches are interesting in aspect of the proposed vectocardiogram sequence coding (VSC). Quality-on-demand algorithms [1-6] are certainly useful tools, however the necessity to maintain the same quality for all electrocardiogram sections may be questionable. The VSC assumes variable medical relevance of particular sections and respectively applies different coding strategies. The regular ECG shows high redundancy between adjacent samples and adjacent heartbeats, thus considering it as a two dimensional image [7-10] in order to make use of both types of correlation yields good efficiency as long as no irregularities occur. However, the vectocardiogram is a spatial phenomenon by its nature and the heartbeats or their physiological parts may be represented in three dimensions without normalization of whole beats' length. Although most of the algorithms use QRS detection as a heartbeat reference, the proposed method goes further in employing medical data to the signal processing and uses automatically detected wave borders to distinguish signal sections of various relevance.

Unlike a regular ECG record, the three-dimensional VCG imaging of the cardiac cycle focuses the doctor's attention on the loops representing P, QRS and T waves [11]. The baseline is not expected to contain any electrodiagnostic data. Following this approach, we consider the VCG signal as composed of a continuous low-bandwidth baseline and three sequences of 3D objects (one for each wave type) synchronized by a time code attribute. Thanks to the temporal prevalence of the sinoatrial rhythm, the correlation of consecutive heartbeats is very high. Since in VCG 3D loops are considered high compression ratio may be achieved with use of the sequence coding methods originally developed by MPEG for video transmission [12-13].

The optimal VCG loops superposition in space precedes the prediction stage and retaining only the cardiac-originated differences selectively narrows the statistical distribution of local error values during the forecasting of neighboring loops. In this meaning we call the proposed algorithm 'optimal', because the extra-cardiac components of inter-loops variability are represented in separate parameters and have limited influence on the sequence coding efficiency.

The clinical practice and many features observed in the VCG signal justify the approach to the consecutive heartbeats or their parts as to a sequence of events:

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- In the vectocardiogram all diagnostic results are derived from analysis of the spatial loops representing the P, QRS and T waves.
- The role of the signal segments located outside the wave area is thus reduced to a continuous, medically irrelevant connection between the loops.
- In consequence, the extra-wave sections open the area to extensive data reduction, while within the waves the signal should be processed carefully in order to maintain the medical contents as close to the original as possible.
- The correlation of consecutive heartbeats is very high except for the occurrence of serious disorders (e.g. T-wave alternans), providing a good background for compression algorithms based on "long term prediction" successfully applied to the conventional electrocardiogram [14-15].

The VSC software, developed in our laboratory is the main issue of this paper. Our previous algorithm [16] has been improved by considerations following from new research results.

II. Materials and Methods

A. VCG pre-processing

The proposed VSC method consists of two stages: the pre-processing and the sequences coding algorithm. The goal of the pre-processing is a transformation of the VCG signal to three sequences extracted on the medical basis with elimination of the extra-cardiac phenomena like baseline wander and respiratory loops variations. Most procedures used at this stage were originally developed for automatic interpretive electrocardiographs and consider the input signal as a regular ECG, recorded without the assumption of the leads orthogonality.

The pre-processing algorithm begins with the heartbeats detection, calculations of the P, QRS and T segment borders (fig. 1) and determination of the heartbeat's morphology.

Accordingly to the medical guidelines, the sinoatrial beats are identified by the occurrence of the unique P-wave in a given interval of time before the QRS-onset. All heartbeats of other morphology are marked, excluded from the sequences and coded as Intermediate Loops (*I-loops*, described later) for which the data stream is expected for 3600 bps yielding compression ratio close to 5.

Since the efficiency of loops spatial prediction is susceptible to baseline wander, the reversible baseline removal and separate processing is highly recommended. Regular baseline removal algorithms were found to be inappropriate and therefore the orthogonal filter bank was applied to split the signal spectrum into two frequency ranges: the baseline below 2 Hz and the cardiac loops above 2 Hz.

The spectrum split performed after seven decimation steps of the lifted wavelet decomposition (LWD) [17]. The transform results in an orthogonal time-frequency representation in the fixed-point values domain (fig. 2).

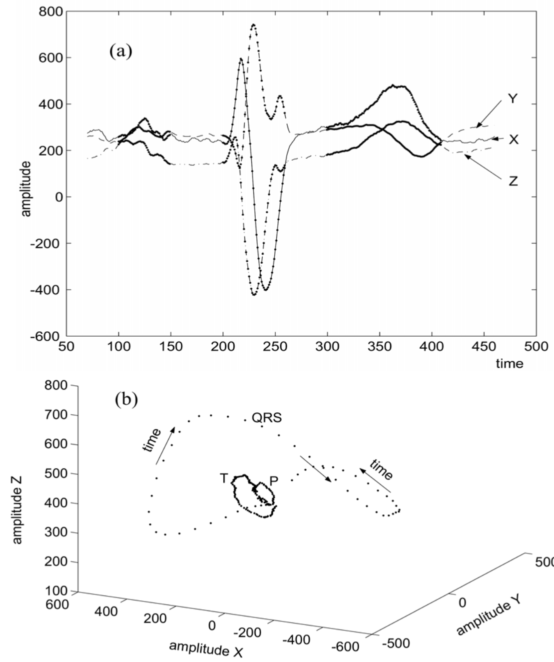

Fig. 1. Segmentation of a heart beat - a standard diagnostic procedure is used to compute the P, QRS and T wave borders. a) time-domain segmentation, b) spatial representation of corresponding loops.

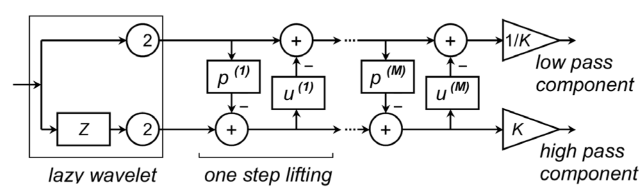

Fig. 2 The principle of lifting wavelet decomposition

To simplify the VSC application, Haar functions were used for *p* and *u*. Accordingly to (1), the first difference acts as high-pass filter, and the average acts as low-pass filter:

$$d_{1,l} = s_{0,2l+1} - s_{0,2l}$$
$$s_{1,l} = \frac{1}{2}(s_{0,2l} + s_{0,2l+1}) \qquad (1)$$

The lifting scheme is a reversible process, thus the components $s_{1,l}$ and $d_{1,l}$ contain complete original information. Thanks to perfect reconstruction property, the residual t-f representation contains all the information necessary to restore the original signal [18].

The low frequency baseline (fig. 3) is coded in the output data stream using the delta modulation algorithm. The delta modulation is based on the prediction of next data value from the current value and codes the prediction errors instead of absolute signal values [19]. To avoid error accumulation, in every Group of Loops, first baseline data is coded as absolute value of its native 12-bits resolution.

The high frequency part of the signal is reconstructed from the details coefficients within the wave area only, and the extracted P, QRS and T sections are subjects to further 3D processing.

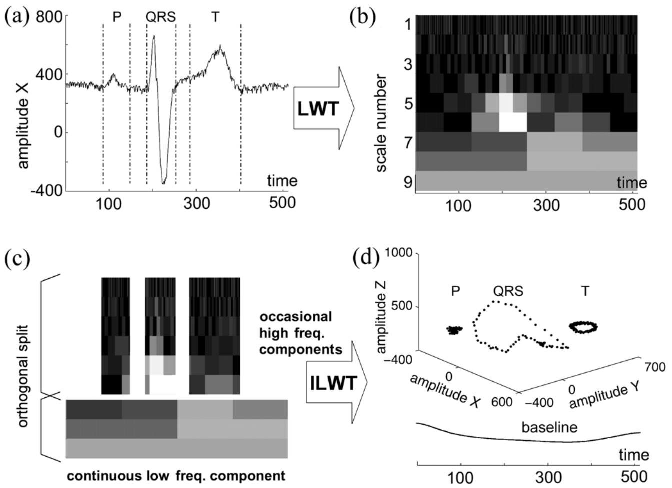

Fig. 3. Splitting the signal in the time-frequency plane. a) time-domain discrete signal; b) the equivalent representation on a time-frequency plane, c) split of the time-frequency representation to baseline and loops components, d) reconstruction of loops and baseline for separate processing.

B. *Principles of VCG sequences coding*

Three sequences of loops P, QRS and T (fig. 4) are processed independently using the same set of parameters. For practical reasons, the length of a sequence is limited to 15 objects unless the occurrence of a heartbeat marked as excluded terminates the sequence prematurely. Each sequence is delimited by two "intra-coded" or *I-loops* shared with the neighboring sequences coded directly in the output data stream. These loops are main reference points for the spatial prediction of all loops lying in between them in a sequence.

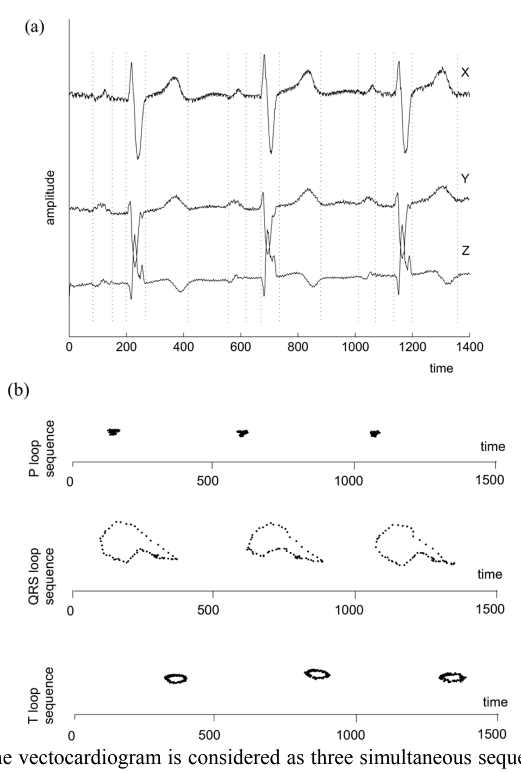

Fig. 4. The vectocardiogram is considered as three simultaneous sequences of three-dimensional objects. High degree of similarity between the neighboring objects may be noticed in this plot. a) time-domain VCG; b) corresponding sequences of loops.

The length of all loops is normalized to 64 samples, using the cubic spline interpolation. In result, the wave duration variability does not influence the representation and the time-corresponding samples may be identified by their numbers. In case of the T-wave, usually much longer than the P or QRS, the length normalization to 64 samples provides a significant reduction of the data volume. The local signal bandwidth within the area of T-wave is far below a half of the reduced sampling frequency and therefore no remarkable alteration of the signal is observed.

Statistically important differences between the loops are often caused by diagnostically meaningless extra-cardiac sources (e.g. respiration). Because the VSC is based on loops differences, it first applies the methodology for optimal comparison of serial VCG originally developed by Paul Rubel and Jocelyne Fayn [20], [21] to align every loop within the Group of Loops with the initial *I-loop*. The superposition algorithm has been simplified, because, due to the normalization of wave segments (performed at the precedent stage), time synchronization is no longer necessary. The assumption of negligible evolution between loops S_i and S_j [22] leads to the transformation (2) which contains only the component Φ representing three basic geometrical transforms: translation, rotation and homothety.

$$S_j = \Phi(S_i) \tag{2}$$

These transforms are iteratively applied to minimize the global dissimilarity between I-loop S_i and every other loop in the group S_j (fig. 5). The measure of difference is defined as a quadratic deviation function (3):

$$E_Q^{i,j}(\vec{d},\phi,\mu) = \sum_{k=0}^{N=63} \left\| \vec{d} + \mu R_\phi \vec{s}_k^{\,j} - \vec{s}_k^{\,i} \right\|^2 \tag{3}$$

where d is the translation vector, μ is the homotetia ratio, k is the sample number, i and j are the loop numbers and s is the vector originating at zero and pointing to the signal sample given by corresponding X, Y, Z coordinates. R_ϕ is the rotation matrix of the form (4):

$$R_\phi = \begin{bmatrix} \cos\phi & -\sin\phi & 0 \\ \sin\phi & \cos\phi & 0 \\ 0 & 0 & 1 \end{bmatrix} \tag{4}$$

Once $E^{i,j}$ value stops falling in consecutive iterations, the maximum loops similarity is achieved and the correction coefficients are stored in the output data set.

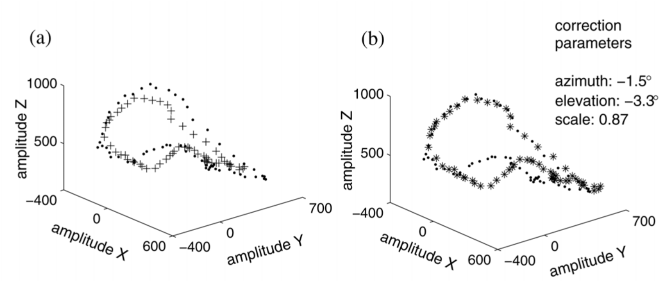

Fig. 5. The spatial synchronization uses basic geometrical transforms: translation, rotation and homothety to minimize the global dissimilarity between two adjacent loops. a) original loops superposition b) optimized loops superposition and the correction parameters.

Next processing stage transforms the Cartesian domain VCG signal (XYZ) to the spherical domain in which it is represented by the magnitude A and two angular variables: azimuth ϕ and elevation λ (5). Further data reduction is gained from the polar transform because of the high regularity of angular variables. The angular data are rounded to the 8-bit representation [21] and decimated to the half of their original sampling rate without measurable distortions of cardiac information.

Each triplet of simultaneous sequences, representing the trains of P, QRS and T-waves, is fed into a double prediction-correction procedure (fig. 6). The bi-directional prediction algorithm refers to two additional loops: S_{N-k1} preceding the current loop by $k1$ loops, and S_{N+k2} delayed by $k2$ loops. Assuming the linear temporal distribution of their differences in the A, ϕ, λ domain, it predicts the values of current loop $\hat{S}_N$ (5).

$$\hat{S}_N^{\{A,\phi,\lambda\}}(i) = S_{N-k1}^{\{A,\phi,\lambda\}}(i) + \frac{S_{N+k2}^{\{A,\phi,\lambda\}}(i) - S_{N-k1}^{\{A,\phi,\lambda\}}(i)}{t_{N+k2} - t_{N-k1}} \cdot \left(t_N - t_{N-k1}\right) \quad (5)$$

The S_{N-k1} and S_{N+k2} loops are not necessarily adjacent to the S_N, however for increasing distances $k1$ and $k2$ the linear interpolation of $\hat{S}_N$ loop may be questionable.
The correction of loops (fig. 7) computes the differences in the A, ϕ, λ domain between the predicted $\hat{S}_N$ and the actual S_N loop samples (6).

$$X_N^{\{A,\phi,\lambda\}}(i) = S_N^{\{A,\phi,\lambda\}}(i) - \hat{S}_N^{\{A,\phi,\lambda\}}(i) \quad (6)$$

At the first prediction stage, the values of selected internal loops are derived from the values of *I-loops* delimiting the whole sequence. The loops selected for the first prediction are regularly distributed in a sequence and called "coarse" or *C-loops* because their prediction is computed from the distant neighbors and thus not precise. True values of *C-loops* are in turn the starting points for the forecast of all remaining loops that lay in between them. The second prediction stage uses the values of closer neighbors and in general yields quite accurate outcome, therefore we call these loops "fine" or *F-loops*. As a result the whole sequence, except the delimiting *I-loops* is represented by a set of prediction error coefficients having very narrow distribution of values (fig. 8).

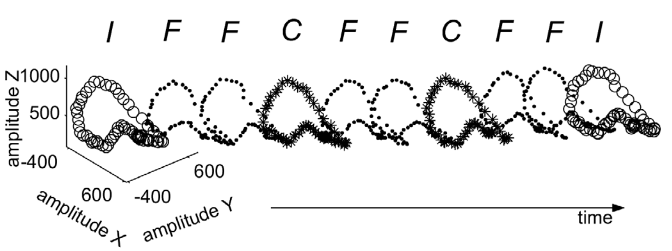

Fig. 6. The QRS loop sequence bordered by *I-loops* and containing *C-loops* and *F-loops* whose values are predicted. The contribution of *F-loops*, predicted with high accuracy is higher than *C-loops*.

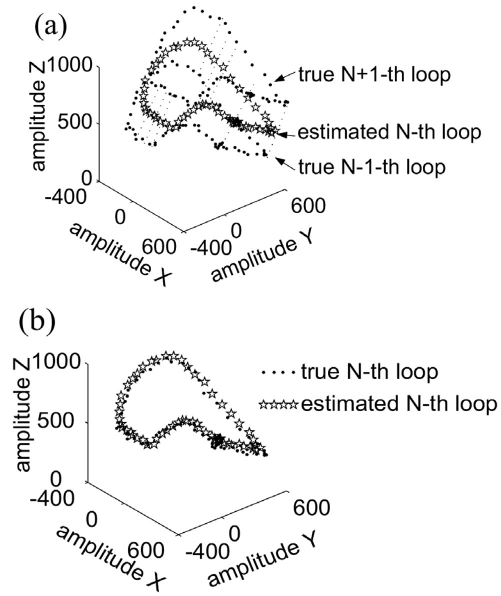

Fig. 7. Prediction and correction of loops in the sequence. a) bi-directional prediction of the N-th loop for $k1 = k2 = 1$; b) computation of prediction error values as the distance between the corresponding predicted and actual samples.

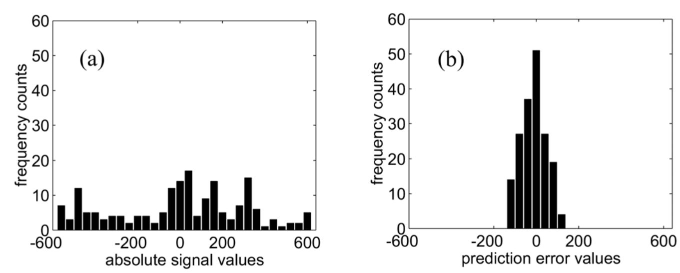

Fig. 8. Comparison of histograms of loop samples magnitudes. a) absolute values; b) values of prediction error

The *I-loops* and all beats excluded from the sequences are coded with a conventional delta modulation [19] and stored in the output data stream. The sequence of $C = 64$ prediction error values $X = [x(0) \cdots x(C-1)]$ is fed to the de-correlation procedure based on the discrete cosine transform (DCT) [24]. For such sequence, the DCT coefficients are defined as (7):

$$G(k) = \sqrt{\frac{\alpha_k}{C}} \sum_{s=0}^{C-1} x(s) \cdot \cos\frac{(2 \cdot s + 1) \cdot k \cdot \pi}{2 \cdot C} \quad (7)$$

where k = 0, 1, 2, ... C-1, α_k = 1 for k = 0 and α_k = 2 otherwise. Although the DCT is a suboptimal transform, its computation is very efficient due to a fast algorithm and the absence of the transformation matrix, which is substituted by the cosine function, and therefore does not require an extra storage.

The transform yields the signal-equivalent data string $G(k)$ containing samples ordered by their energy. This string is truncated at the point from which the sum of remaining energy is a non-significant fraction (e. g. 1%) of the total signal energy. Due to the narrow histogram of coded values, the 1% energy threshold is usually met below 30% of the string's length and the abstracting economizes a significant amount of data that do not have to be stored. The retained DCT values are normalized aiming at the optimal fitting into the dynamic range provided by the target representation format (16-bit for magnitude and 8-bit for each angular variable).

At the end of the processing chain, the fixed-point DCT representation of spherical domain loop prediction errors is coded to the output data stream with use of the Huffman algorithm [25]. Both angular variables have similar statistical properties and share the same symbol dictionary economizing the space in the output string.

The complete compression processing diagram and data flow is summarized in figure 9.

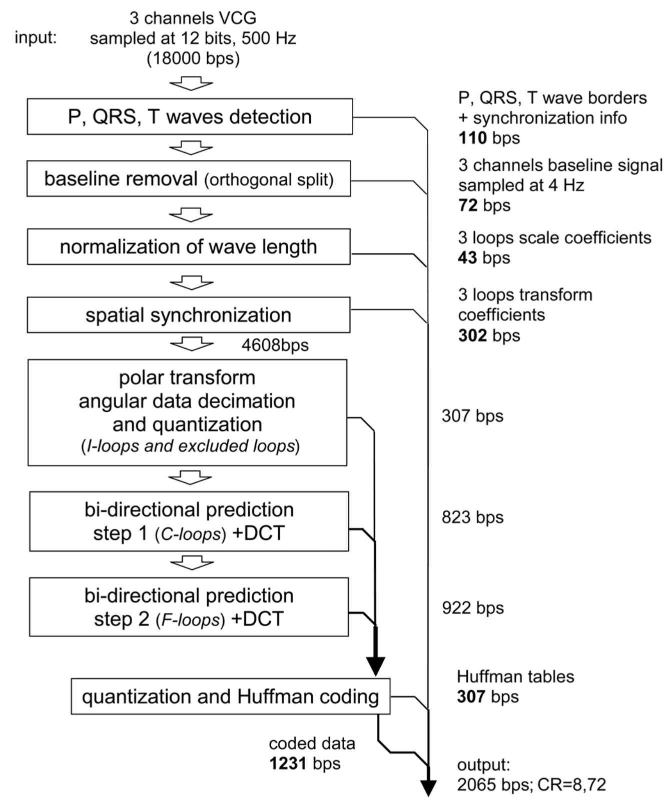

Fig. 9. Complete compression processing diagram and data flow specification assuming HR=72, *C-loops* contribution 4/15 and DCT CR = 1.5, F-loops contribution 10/15 and DCT CR = 3.3, Huffman CR=1,67

C. *Analogies of video and VCG coding*

Except for the baseline removal, the geometrical corrections of extra-cardiac variability and the time-domain normalization, main steps of the proposed VSC method correspond directly to the MPEG-2 coding stages:

- The polar transform $XYZ \rightarrow A\phi\lambda$ is the analogy of color space transform RGB $\rightarrow$ YR_bR_g distinguishing one variable with important high-frequency contents.
- The group of loops structure and the double step bi-directional prediction-correction process is conceptually based on the idea of group of pictures (GOP); for its better performance only bi-directional prediction is applied.
- The DCT is used for decorrelation of error coefficients. Truncating the insignificant tail of the string reduces the data stream volume.
- The Huffman coding benefits from a very narrow distribution of error values.

D. *Numerical tests of VCG sequences coding*

The conceived VSC algorithm was coded in Matlab without optimization of computation costs. The CSE compliant heartbeat detector and segmentation procedures were obtained from the commercial ECG equipment manufacturer as executable files.

The source of test signals was the CSE-Multilead Database (Dataset 3) that contains 125 original 10s traces acquired with use of pseudo-orthogonal XYZ Frank leads [26]. The quantization level is 12 bits (2.44 µV) and the sampling frequency is 500 Hz. Two records (No. 67 and 70) containing pacemaker-triggered evolutions were excluded for the lack of waves annotations.

The distortion level was computed as the percent root-mean-square difference (PRD) accordingly to (8) for the whole heartbeat and also separately in sections within the confines of each wave.

$$PRD = \sqrt{\frac{\sum_{i=1}^{n}\left[x_1(i) - x_2(i)\right]^2}{\sum_{i=1}^{n}\left[x_1(i)\right]^2}} \cdot 100\% \tag{8}$$

The section-wise PRD computation reveals the algorithm's ability to concentrate the distortions in the diagnostically meaningless sections and partially overcomes the most questioned drawback of this measure, which is equalizing signal distances in sections of different medical significance.

III. Results

The main outcome of the numerical experiment was the estimation of the efficiency and the distortion coefficients for the newly proposed method. The resulting compression ratio (CR) and distortion parameters (PRD) are summarized for the CSE database signals in table 1 and an example of temporal distortion distribution in the reconstructed VCG signal (Mo_001) is displayed in fig. 10.

In addition to the main goal, our experiments confirmed the expected correlation of the compression ratio (CR) and the average heart rate (HR) The derived relationship can be expressed as (9):

$$CR = 12.6336 - 0.036HR \tag{9}$$

The HR-based CR estimation confidence level reaches the value of 85%.

TABLE. 1
SUMMARY OF COMPRESSION PERFORMANCE AND DISTORTION LEVEL OF THE OPTIMAL VCG SEQUENCE CODING.

CSE No	HR	CR	distortions (PRD %) total	P	QRS	T
1	74	8.59	5.72	0.78	0.11	0.39
2	125	5.71	7.47	0.42	0.12	0.37
…	…	…	…	…	…	…
124	73	8.31	8.71	0.72	0.18	0.31
125	79	7.95	9.10	1.07	0.13	0.47
average	78	8.15	6.22	0.94	0.16	1.09
standard deviation	20.9	2.57	1.52	0.26	0.03	1.07

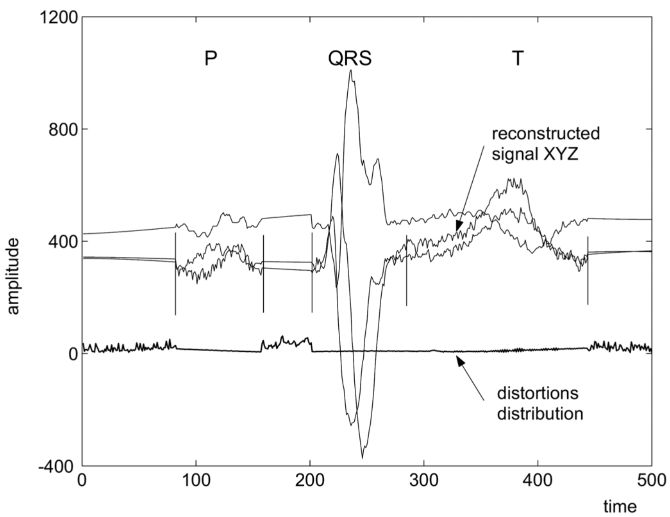

Fig. 10. Reconstructed VCG traces CSE Mo_001 (X, Y, Z) and the residual signal illustrating the distribution of distortions in the reconstructed signal; within the P and QRS waves the compression is practically lossless, within the T wave the distortions are just noticeable due to limited bandwidth..

In the supplementary numerical tests the influence of T-wave normalization and DCT energy threshold value to the algorithm performance. The optimal length of T wave representation was found to be 64 samples and the DCT threshold value should not exceed 1% of total signal energy.

Using the rhythm type criteria, four categories of records were extracted from the CSE database:

- supraventricular regular rhythm supraventricular
- irregular rhythm (ΔRR > 15%)
- ventricular rhythm (similar beats without P wave)
- mixed rhythms (arrhythmias and escape beats)

Separate calculations of the compression statistics in these categories reveal the behavior of proposed VCG sequence coding in case of irregular rhythms. The resulting compression parameters are summarized in table 2.

Processing the records containing supraventricular irregular rhythm is similar to the regular rhythm and thus the difference in compression ratio may be explained by higher heart rate in irregular rhythm.

TABLE. 2
COMPRESSION PARAMETERS FOR VARIOUS RHYTHM CATEGORIES

rhythm category	available records	heart rate avg. ± std.	compress. ratio avg. ± std.
supraventricular regular rhythm	84	76,6 ± 19,5	8,54 ± 2,63
supraventricular irregular rhythm	12	82,0 ± 27,1	7,99 ± 2,75
ventricular rhythm	11	77,2 ± 27,3	6,29 ± 2,32
mixed rhythms and arrhythmias	16	85,1 ± 18,0	7,47 ± 2,25

The purely ventricular rhythm contains only *I-loops* and consequently no bi-directional prediction is performed affecting significantly the compression performance. However, the absence of P-loop eliminates 33% of data from the storage and the total compression efficiency is acceptable.

In case of mixed rhythms the algorithm tries to assemble consecutive supraventricular beats whenever possible. The compression ratio for this rhythm category highly depends on percentage, type and distribution of irregular beats. The worst case of bigeminy (CSE no. 117) with HR = 79 bpm yields a compression ratio of 5,23.

IV. DISCUSSION

The principles of video sequence coding were adapted to the three-dimensional VCG loops compression. The proposed VSC method features high compression ratio and low level of distortion. From the cardiologist's viewpoint focused on the P, QRS and T-wave loops the VSC method guarantees reliable signal reconstruction (error below 10 μV) close to the digital acquisition standard requirements [27]. Decompressed waves are practically identical to original, representing all important medical information, including low energy components. T-wave processing uses decimation not affecting the cardiac components, but possibly the noise if present in the record.

The reconstruction error appears only in the signal sections of low diagnostic importance, which are the main source of bitrate economy, since in these parts the data reduction is achieved by significant limitation of the bandwidth. The distortions in the sections between the waves were not noticeable as long as the detector performs correctly and experts display 2D or 3D VCG signal. The experts' reception of the linear display similar to figure 9, was in general fair, with acceptance of a noise-free baseline. Although baseline errors represent a noise-canceling side effect and not alterations of the cardiac electrical representation, few experts, probably suspecting a loss of medical information, complained of the total 'silence' on the baseline. Further improvement in this domain can be expected from applying of appropriate medically justified parameters for the sequence processing of each loop type.

Unfortunately in the aspect of reproducibility, the CSE database is not as widespread as the MIT/BIH used in [9] and [10] (tab. 3). Therefore our experimental results are expressed in compression ratio, or bits per sample (bpsmp) in order to compensate differences in sampling resolution and frequency (12bit, 500Hz CSE vs. 11bit, 360 Hz MIT/BIH).

TABLE. 3
COMPARISON OF THE PROPOSED ALGORITHM WITH THE JPEG2000-BASED [9] AND WITH 2D MODIFIED SPIHT-BASED [10] COMPRESSION METHODS

compression ratio	Bilgin [9]	Thai [10] block 32x256	VSC (proposed method)
8	2.18		
1,37 bpsmp	0.86 (file 117)		
8,15 ± 2,57			6,22 ± 1,52
1,47 bpsmp			0,94 ± 0,26 (P)
			0,16 ± 0,03 (QRS)
			1,09 ± 0,07 (T)
10	2,93	1,08 ± 0,44	
1,10 bpsmp	1.03 (file 117)	0,55 (file 117)	
20	6,92	2,38 ± 0,90	
0,55 bpsmp		0.85 (file 117)	

The compression ratio depends on the heart rate because the higher rate shortens primarily the baseline. The variation of cardiac rhythm, by influencing the sequence length, affects the algorithm efficiency. In subjects with high irregularity or arrhythmia the compression ratio is remarkably lower.

The advantageous and innovative estimation of local medical relevance of the ECG signal relies on the automated detection and recognition of waves. Any error in wave detection, missed or irregular events (e.g. U waves) may degrade the reconstructed signal quality (false negative cases) or affect the compression efficiency (false positive cases).

Encoding operations typical for MPEG2 are already supported by the electronic circuitry including portable low-power devices. Despite its apparent complexity, the VSC algorithm is currently considered for implementation in a prototype wireless ECG monitoring system.

V. CONCLUSION

The spatial prediction technique is very efficient in application to the optimal coding of vectocardiographic sequences. Main novelty and advantage lies in the temporal management of the signal distortion distribution controlled by the parameters of processed signal and by the general knowledge about the local signal importance.

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