Real-time System for High-resolution ECG Diagnosis Based on 3D Late Potential Fractal Dimension Estimation

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Abstract

A real-time high-resolution ECG (HRECG) system was developed to study ventricular late potentials (VLP). In the denoising method, the SFP alignment technique was embedded in a pre-processing micro-controller to support the signal averaging process. This system is intended for screening subjects at cardiac risk in the out-of-hospital environment. It uses the fractal dimension of VLP (LP_δ) as the diagnostic parameter. Performance of the SFP technique was tested by systematic 50Hz and EMG additive noise onto a noiseless ECG signal model of known spectrum. Spectral degradation due to the SFP based averaging, indicated a negligible beat alignment jitter SD under worst case of noise levels: ±2.6ms SD for $340\mu V_{RMS}$ of 50Hz noise, and ±1.3ms SD for $71\mu V_{RMS}$ of EMG noise type. In the LP_δ verification method, the mean value of LP_δ in five healthy volunteers was of 1.204 ±0.0526. A synthetic, coherent, and orthogonal LP signal injection at the body surface (with LP_δ >1.36) indicated satisfactory LP_δ parameter reproducibility.

1. Introduction

In order to prevent sudden cardiac death (SCD), risk classification and stratification of potential subjects or communities is an important preventive healthcare undertaking. In this research work we developed a system device intended to be used for effective electrocardiographic screening and provide reliable evidence for cardiac healthcare strategies and decisions.

Of particular interest in this paper is the study of ventricular late potentials (VLP). These are small amplitude ECG signals (below 40μV), appearing at the terminal portion of the QRS complex, and they contain frequency components from 40Hz up to 300Hz [1]. A particular algorithm commodity for assessing VLP provides a dimensionless VLP waveform metric defined as the fractal dimension of the VLP attractor (LP_δ). Escalona et al. [2], have investigated a diagnostic threshold value for this; it states that a value of LP_δ below 1.3 indicates that the patient is not at risk, otherwise the patient can be considered at risk of SCD.

2. Methods

In order to analyse VLP a real-time HRECG system was implemented. It is formed by two parts: the hardware and the software. The hardware itself is formed by power supply, indicators, amplification, filters, analogue to digital converters (ADC), isolator, RSR-232 USB converter and a personal computer. A block diagram is shown in Figure 1.
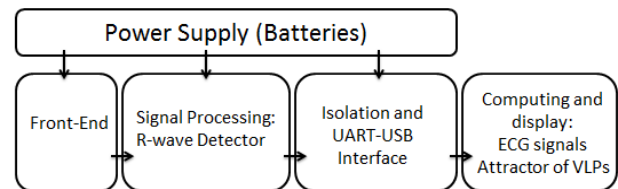

Figure 1. Real-time high-resolution ECG block diagram.

The ECG signals are obtained by recording the orthogonal bipolar XYZ lead system [1].

The ECG amplification section contains a couple of gain stages with overall gain fixed to 2000. The analogue filter stage is composed by a first-order, 3Hz high-pass filter and by an antialiasing fifth-order low-pass filter, set to a high cut off frequency of f_h= 360Hz. The sampling frequency (f_s) was set to 2kHz. Thus, the acquisition front-end has a bandwidth from 3 up to 360Hz, and the dynamic range was set to ±10V. 16-bit resolution ADCs were used in the system and thus the minimum input voltage change per bit was 305μV. Digital isolation was fully implemented to protect the following laptop/PC stage.

Firmware and signal processing were implemented using a micro-controller. It computed a highly accurate real-time alignment reference in the QRS complex, used for signal averaging (SAECG). For this real-time task, coding was implemented to carry out the Single-Fidutial-Point (SFP) aligment technique algorithm [3].

Data output for the three ECG channels (XYZ) plus the QRS alignment reference pulse was sent to the laptop/PC via the USB-port. Operator console interface and high level computational processes were implemented at the

laptop/PC stage, using LabVIEW. Real-time ECG display of the three channels and the QRS reference pulse was provided. The developed LabVIEW application also processes these four signals to compute the fractal dimension of the VLP attractor. This last process involves several steps that are described below.

2.1. Filtered SAECG vector magnitude

For computing the SAECG frame, the SFP alignment algorithm [3] was coded in the micro-controller. The reference channel and the number of beats to be averaged are decided by the operator. The SAECG was computed for the three channels (X, Y, Z). As the frequency spectrum of VLP is mainly between 40 and 300Hz, a 40Hz, bi-directional, 4th order Butterworth high pass filter was applied to generate the filtered SAECG frames [1].

For VLP isolation method consists, the X_{VLP}, Y_{VLP} and Z_{VLP} vectors were selected. In order to select the VLP vectors, a vector magnitude of the filtered SAECGs was computed. It follows the next equation:

$$M = \sqrt{X_f{}^2 + Y_f{}^2 + Z_f{}^2} \qquad (1)$$

In equation (1) X_f, Y_f and Z_f are the 40 Hz high-pass filtered SAECG signals. On the computed vector magnitude frame (M), the end of VLP (t_e) is located when the amplitude of M is equal to the lowest noise level plus three times its standard deviation in the ST region; the start time (t_s) is located when the amplitude of M is equal to 40µV. The amplitude is obtained by measuring the average of a 10ms window and moving it by steps of 5ms towards the QRS complex.

The segment of each vector (X,Y,Z) between those two time limits ($T = t_e - t_s$), will be the X_{VLP}, Y_{VLP} and Z_{VLP} vectors, in other words, the VLP isolated in each channel.

2.2. Fractal dimension

Once the VLP are isolated, each X_{VLP}, Y_{VLP} and Z_{VLP} vector is numerically scaled into µV units. There are only two parameters that need to be computed to calculate the LP_δ parameter which determines whether or not the patient is at risk of SCD. The method may include an estimation of the fractal dimension of the attractor (LP_δ), as the quotient of:

$$LP_\delta \approx \frac{\log(L)}{\log(DD)} \qquad (2)$$

where L is the total length of the attractor (3-D curve) and DD is the spheric extent diameter of the attractor. Parameters L and DD are measured in the microvolts scale to properly compute the parameter LP_δ. The total length of the trajectory can be calculated as follows:

$$L = \sum_{i=2}^{i=N} \sqrt[2]{\left(X_f(i) - X_f(i-1)\right)^2 + \left(Y_f(i) - Y_f(i-1)\right)^2 + \left(Z_f(i) - Z_f(i-1)\right)^2} \qquad (3)$$

In where N is the number of time steps in the interval

T used to record the ventricular late potential, that is, the number of samples taken by the digitisation process in the interval T. The computation of DD diameter involves more calculations than the one carried out by the total length. However it is not complicated. It states that for each couple of values (X,Y,Z), the distance has to be calculated and placed in a matrix; then the maximum diameter DD, will be the maximum value of the matrix D.

$$D_{ij} = \sqrt[2]{\left(X_f(i) - X_f(j)\right)^2 + \left(Y_f(i) - Y_f(j)\right)^2 + \left(Z_f(i) - Z_f(j)\right)^2} \qquad (4)$$

The matrix representation of all D_{ij} elements has a symmetric form due to the square functions in the equation above, besides when i = j the result is zero (0). Then, only half of the elements are needed to determine the DD parameter.

$$D_{ij} = \begin{bmatrix} 0 & D_{12} & D_{13} & .D_{1j} & .D_{1N} \\ D_{21} & 0 & D_{23} & .D_{2j} & .D_{2N} \\ D_{31} & D_{32} & 0 & .D_{3j} & .D_{3N} \\ . & . & . & . & . & . \\ D_{N1} & D_{N2} & D_{N3} & .D_{Nj} & . & 0 \end{bmatrix} \qquad (5)$$

Afterwards, taking the maximum value of the matrix elements the DD will be obtained.

$$DD = \max(D_{ij}) \qquad (6)$$

Previous studies have found that a fractal dimension LP_δ in excess of 1.3 may be selected as the value that indicates a positive condition for risk of SCD [2, 4].

3. Testing methods

In order to bench test the system, a QRS signal model (a 60ms width periodic pulse with 750ms period) was utilised as an input in channel X, and as the front-end has a band pass from 3 to 360Hz the QRS signal model becomes as Figure 2 shows.
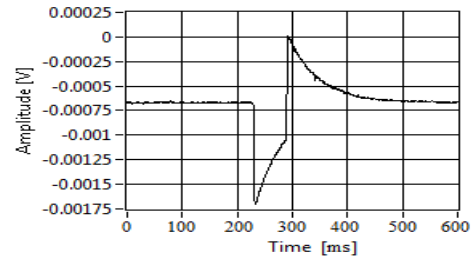

Figure 2. Filtered QRS signal model.

3.1. System noise testing

QRS signal model was applied at the bipolar input of each channel (XYZ), while at the output the system noise was recorded and measured. As the system noise includes interference noise pick up; particularly 50 Hz mains interference. Hence, for different test occasions, different levels of noise were found in the system.

3.2. SFP testing

Performance of the SFP (single-fidutial-point) alignment technique can be evaluated by measuring the alignment jitter. That is, how accurate the algorithm can be while noise conditions increase. For this, QRS signal model was set as an input in channel Z, while the same QRS signal model with added 50Hz or simulated EMG noise was set as an input in channel X. The microcontroller detects the QRS complex using channel X as reference; after, this detection was stored into a vector called *qrs*. Then, a detection of QRS complex of channel Z (without added noise) was ran using Matlab. The evaluation algorithm computed and saved the time difference of the *qrs* and the detection made by Matlab, then the standard deviation is calculated on this vector, the result will be the value of the jitter SD. For our 2kHz sampling operation, if the jitter SD value is closer to 0.5ms when a remarkable noise level is present in channel X, then the SFP algorithm implementation is reliable..

3.3. SAECG testing

To evaluate denoising performance of the signal averaging (SA) process, two types of noises (50Hz and the simulated EMG) were considered. For this, two analogue noise generators were implemented. The QRS model was corrupted with noise and, also was set as input signal in channel X. Seven different levels of both noises were considered. Then, each level of noise recording, was passed through the SA process and the level of noise in the signal was measured when the number of averaged beats was 1, 10, 20, 50, 100, 200 and 400. For Gaussian noise, it is known that the noise level is inversely proportional to the square root of the number of averaged beats, and this is the case for our simulated EMG noise.

3.4. Analysis of VLP

The cardiac activity of five healthy volunteers were recorded and processed through the VLP algorithm to observe for expected results for healthy subject. ECG coherent signal models of LP for each channel (X, Y, Z), was analogically generated with first order band-pass filter circuits. The idea was to inject rather complex orthogonal, small amplitude LP signal components from the body surface, so they could appear coherently within the cardiac signal activity. The LP signal model consists of three different bipolar pulse signals: resonant wave at frequencies of 50Hz, 70Hz and 90Hz with 20Hz of bandwidth. Each bipolar wave pulse was injected on the body surface, as close as possible to the bipolar XYZ lead system electrodes.

4. Results

4.1. System noise

The magnitude of the highest system noise recorded was 5.1µV. Thus, the system introduces a background noise level that needs to be reduced by special denoising techniques: such as signal averaging.

4.2. SFP

The jitter SD of the QRS alignment technique was computed under certain noise levels for 50Hz and simulated EMG noise types. The results are exposed in Figure 3. The closer to 0.5ms the value of SD jitter, the more accurate is the SFP algorithm. According to the results obtained, an extreme case of 50Hz noise level of 340µV (r m s), was found to be the worst type of noise to handle and produced a jitter level of 2.6ms. The SFP algorithm includes a 30Hz cut-off frequency low-pass filter, but still the 50Hz interference has its influence on the QRS alignment precision. A simulated EMG noise (Gaussian with 300Hz BW) level of 71µV (r m s) yielded a 1.3 ms jitter (SD). With EMG type of noise, spectral degradation can be deduced from the measured jitter by the relation BW = (0.13/(SD jitter)) [3]; several result values are presented in Table 1.
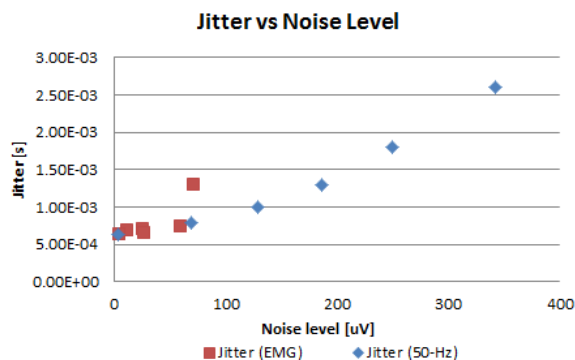

Figure 3. Jitter vs. Noise level added to a clean signal

Table 1. Measured jitter SD of QRS alignment at different noise levels and estimated bandwidth limitation due to jitter.

EMG Noise (µV)	Measured SD Jitter (s)	Bandwidth Limit (Hz) BW ≈ [0.13/(SD jitter)]
70.87	1.30E-03	100.0
59.14	7.32E-04	177.7
25.8	6.96E-04	186.8
26.44	6.51E-04	199.8
12.1	6.91E-04	188.0
4.08	6.38E-04	203.8

4.3. SAECG

The denoising performance of the implemented SA technique, for the SAECG frames generation, is depicted in Figure 4. Noise reduction trends as a function of increased number of averaged beats is evidence for both types of noise (50Hz and simulated EMG). For example, with 400 beats, 26.02dB attenuation on simulated EMG noise can be delivered.

Theoretically, for Gaussian noise, the attenuation factor is $\frac{1}{\sqrt{N}}$, where N is the number of averaged beats.

Figure 5 depicts the evidence of this fact. There, (Final Noise) vs (Initial Noise) of seven cases of EMG noise levels is plotted. To understand this more clearly, the equation $N_f = \frac{N_i}{\sqrt{N}}$, will be rewritten as $y = m.x$. Hence, $N_i{}^2 = N.N_f{}^2$; where $y = N_i{}^2$ and $x = N_f{}^2$, in where, the slope of the linear function is meant to be N, which is 400 in Figure 5.
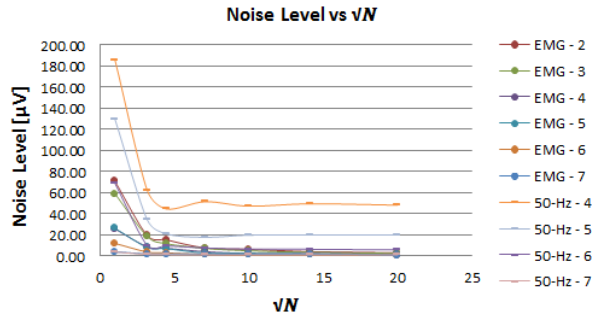

Figure 4. Noise reduction profile with increased number of averaged beats ($\sqrt{N}$), for EMG and 50Hz noise.
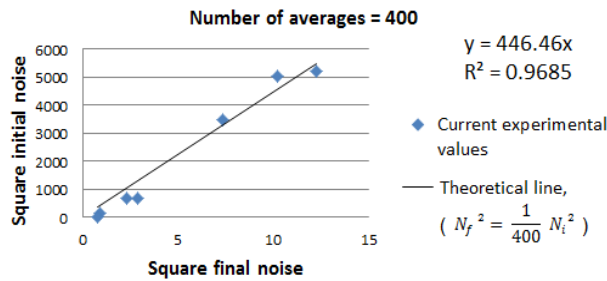

Figure 5. Plot of (Square initial noise) vs (Square final noise).

Table 2. Fractal dimension of VLP attractor (LP_δ) for five healthy volunteers: 1 to 5.

1	1.179
2	1.272
3	1.158
4	1.162
5	1.248
Average	**1.204**
SD	**0.0526**

4.4. Analysis of VLP

The ECG signal of five healthy volunteers were recorded and analysed with the completed system prototype. The average value of the LP_δ parameter was 1.204±0.0526SD, which is under 1.3; as expected for healthy subjects. Table 2 shows the fractal dimension values of the five volunteers.

Furthermore, the orthogonal LP signal model was coherently injected as the body surface ECG signal was recorded in one of the healthy volunteers (number 1). As the synthetic LP model was of remarkable complexity

and had a large DD value of about 63µV, with an expected LP_δ >1.36, it could be detected by our HRECG system. The value of LP_δ in this healthy subject increased from 1.179 (see Figure 6) to 1.382.

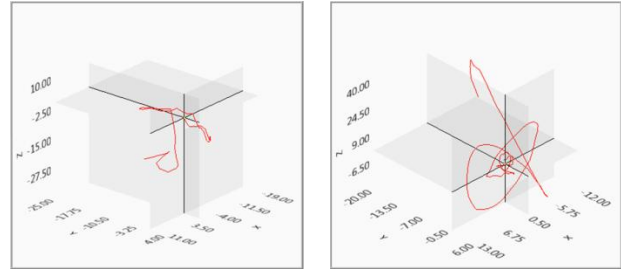**(a) Volunteer #1: LP_δ = 1.179** **(b) With model LP: LP_δ = 1.382**

Figure 6. 3D plots of VLP attractors: (a) baseline recording, (b) with the injected synthetic orthogonal LP signal model.

5. Conclusions and further work

To support healthcare policies in preventing patient from suffering SCD, a real-time HRECG system prototype was implemented and tested at the Centre for Advanced Cardiovascular Research (CACR), in Ulster. By using fractal dimension analysis of the VLPs, a handheld portable and reliable cardiac point-of-care diagnostic device can be provided using these methods. The system device can enable doctors to screen patients at risk in cardiac clinics and out-of-hospital communities. It is important to mention that this method is a non-invasive one, and can be used anywhere. It would require the patient to be at rest for a few minutes while recording. Also, this device may prove useful in future research studies about VLP related cardiac pathologies, such as Brugada syndrome.

References

[1] Breithardt G, Cain ME, et al.. Standards for analysis of ventricular late potentials using high resolution or signal-averaged electrocardiography. European Heart Journal. 1991, vol.1 (12), pp.473-480.
[2] Escalona OJ . Analysis of heart waveforms. United States Patent, 1997, 5,694,942, 1997-12-09.
[3] Escalona OJ, et all. A fast and reliable QRS alignment technique for high-frequency analysis of the signal-averaged ECG. Med. & Biol. Eng. & Comp, 1996, vol. 31, pp. S137-S146.
[4] Mitchell RH, Escalona OJ . Discriminating at-risk post-MI patients by fractal dimension analysis of the late potential attractor". Proceedings of the 20th Annual International Conference, 1998, IEEE-EMBC, pp. 1573-1575.

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