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T-wave Vector Alternans Detection based on Holter ECG Recordings

Kaoru Kaeriyama[a], Masaki Nozawa[a], Hiroyuki Shichiku[a], Kazuo Yana[a,b],
Hirohisa Mizuta[c] and Takuya Ono[d]
[a]Department of Applied Informatics, Hosei University, Tokyo 184-8584, Japan.
[b]IT Research Center, Hosei University, Tokyo 102-8160, Japan.
[c]Corporate Headquarters, Josai University, Saitama 350-0295, Japan.
[d]Department of Cardiology, Nippon Medical School, Tokyo 113-8602, Japan.

Correspondence: Kazuo Yana, Depatment of Applied Informatics, Hosei University, 3-7-2 Kajino, Koganei City, Tokyo 184-8584, Japan. E-mail: yana@hosei.ac.jp, phone +81-42-387-6188, fax +81-3-3368-8165

Abstract. This paper proposes a noble method for detecting T-wave alternans (TWA) based on 3-channel Holter ECG data. The method is intended to be the core of realizing an accurate and efficient TWA detection system based on the data acquired under non-controlled common daily activity. The method consists of two steps, i.e. coarse and fine TWA detection. The first coarse TWA detection has been made by detecting the triplet waveform error alteration for the purpose of high speed screening. The second fine detection step examines the change in T-wave morphology based on singular value decomposition (SVD). SVD has been adopted for detecting the precise T-wave morphology change in the noisy Holter recording condition. We call T-wave morphology change found by the SVD, T-wave Vector Alternans (TWVA), which is the key notion in the proposed method. The method has been applied to ten normal subjects and ten subjects with TWA to confirm its validity. The method will be a useful tool to extract an extra important diagnostic index from the popular Holter ECG recordings.

Keywords: T-wave Alternans, ECG signal analysis, Spline Smoothing, Sudden Cardiac Death, Holter ECG

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1. Introduction

T-wave alternans (TWA) has been known as a useful prognostic index of assessing the risk of occurrence of sudden cardiac death (SCD) [Kaufman *et al.*, 2006][Rosenbaum, 2008]. The current standard method of TWA detection enables us to detect the microvolt order of the small alteration of T-wave amplitude. However, the method needs to be based on the controlled ECG measuring environment with an elevated mean heart rate[Rosenbaum *et al.*, 1994]. It would be desirable if accurate TWA detection could be performed in the non-controlled environment. This paper proposes a method to detect TWA based on a twenty four hour Holter ECG record without setting any controlled measuring environments. The method consists of two steps. The first step, termed coarse TWA detection, utilizes the correlation among three consecutive T waves. This step scans data comparing relative mean squared errors between the initial T-wave and two consecutive adjacent T waves to see if there are any successive alternate changes in them. TWA candidate data segments screened by the first step are forwarded to the second step of the precision analysis of T-wave morphology. Singular Value Decomposition (SVD) is applied to three lead (X, Y and Z) Holter ECG signals. The phase diagram obtained by two major orthogonal signals, called the T-wave loop, is utilized to detect the alternate changes in T-wave morphology. We call the alternative change in T-wave morphology based on the decomposed orthogonal signals, T-wave Vector Alternans (TWVA). Since the second step of TWVA detection is time consuming, the two-step procedure proposed in this paper aimed at giving a practical solution for the efficient T-wave alternans detection. The method was applied to twenty subjects (ten normal and ten with TWA) for its validation.

2. Methods

A method of TWA detection based on the Holter ECG recordings is described in this section. Let us denote the original Holter ECG records from X, Y and Z leads as $x^{(m)}[n](=x^{(m)}(n\Delta t))$, m=1,2,3; n=1,…,N. Each lead number m(=1, 2 or 3) designates one of the leads X, Y or Z. Δt is the sampling interval set at 5 (mS). The method consists of two steps, coarse and fine TWA detection.

Step 1. Coarse TWA detection

The *T*-wave starting time T_s of each beat as defined by RR interval percent time at 10%, i.e. T_s is set at the *R* peak time plus 10% of the succeeding *RR* interval. The *T*-wave duration is set at 250 *ms* from T_s. Let us denote the kth *T*-wave data segment from the lead *m* as: $y^{(k,m)}[n], n=1,...,N_T, k=1,...,K$. Since the original Holter ECG signals have significant drift in their base line, the baseline correction is necessary, preceded by the TWA detection. In the *step 1,* for the purpose of screening, a simple linear regression correction is applied, assuming that the ECG values at both T-wave starting and ending points are zeros. Corrected ECG data are denoted by $\tilde{y}^{(k,m)}[n]$. Now, let us form the set of *K-2* triplets of consecutive *T*-wave records for each lead *m*. The kth triplets are shown in the following *T*-wave data set.

$$\{\tilde{y}^{(k,m)}[n], \tilde{y}^{(k+1,m)}[n], \tilde{y}^{(k+2,m)}[n]\}, k=1,...,K-2, m=1,2,3, \qquad \text{…(1)}$$

Two consecutive data matching errors $\{e_1^{(k,m)}, e_2^{(k,m)}\}$,setting the first *T*-wave data $\tilde{y}^{(k,m)}[n]$ in the triplet as the reference pattern, are calculated such that:

$$e_i^{(k,m)} = \sqrt{\frac{\sum_{n=1}^{N_T} (\tilde{y}^{(k+i,m)}[n] - \tilde{y}^{(k,m)}[n])^2}{\sum_{n=1}^{N_T} (\tilde{y}^{(k,m)}[n])^2}}, i=1,2 \qquad \text{… (2).}$$

Then, these errors are concatenated to form the error sequence, *i.e.*

$$\{e_1^{(1,m)}, e_2^{(1,m)}, ..., e_1^{(k,m)}, e_2^{(k,m)}, ..., e_1^{(K-2,m)}, e_2^{(K-2,m)}\} \qquad \text{….(3).}$$

The error sequence should have an alternate amplitude change in case the TWA presents. The error sequence is segmented by 128 consecutive error values. The rate of alternate changes in each segment is calculated and the segments whose alternate rate exceed a certain threshold are forwarded to the next step of Fine TWA detection.

Step 2. Fine TWA detection

The set of X, Y and Z lead segments are screened by the above first step as potential TWA candidates are further analyzed utilizing singular value decomposition (SVD). Let us denote ECG raw data in the screened segment from the mth lead as, $x_s^{(m)}[n](=x_s^{(m)}(n\Delta t))$, m=1,2,3; n=1,…,N: s=1,…S.

Here, the new index *s* denotes the segment number screened by the first course TWA detection. Baseline correction in this step is done by the smoothing spline interpolation. The spline function $\bar{x}_s^{(m)}(t)$ to minimize the following function, is regarded as the baseline to be corrected.

$$(1-p)\sum_n w_s^{(m)}[n](x_s^{(m)}[n] - \bar{x}_s^{(m)}(n\Delta t))^2 + p\int \left(\frac{d^2\bar{x}_s^{(m)}(t)}{dt^2}\right)^2 dt \qquad \text{…(4).}$$

Here, $w_s^{(m)}$, m=1,2,3; n=1,…N ; s=1,…S are weights which take the value one in the portion where the signal values are assumed to be zero, *i.e.* the time between the *T*-wave end time (T_e) and *P* -wave starting time (TP_s) . Instead of determining T_e and TP_s exactly, here we set the target period to be 60-70% of the time within *RR* intervals. Weight values are set to be zeros elsewhere to avoid the ECG signals from the base line estimation. The smoothing factor *p* in Eq. (4) has been empirically set at 0.01 for realizing the light smoothing effect. The baseline corrected data are obtained as:

$$\tilde{x}_s^{(m)}[n] = x_s^{(m)}(n\Delta t) - \bar{x}_s^{(m)}(n\Delta t),\ m=1,2,3;\ n=1,\ldots,N;\ s=1,\ldots,S \quad \ldots (5).$$

For all segment *s*, *T*-waves in each beat are extracted in the same manner as in *step 1*. The k^{th} *T*-wave data from the lead *m* in the segment *s* is denoted as:

$$\tilde{y}_s^{(k,m)}[n], n = 1,\ldots, N_T, k = 1,\ldots, K, s = 1,\ldots, S \quad \ldots(6).$$

Singular value decomposition (SVD) [Yana *et al.*, 2006][Acar *et al.*, 1999] is applied to a set of average *T*-waves $\tilde{y}_s^{(m)}[n]$, *m*=1,2,3 over the segment *s* and a set of *T*-waves from different leads for each beat in the segment *s*. Two major orthogonal signals obtained by SVD for sets of average *T*-waves and each beat are utilized for the fine TWA detection in the *step 2*. Here, average *T*-waves $\tilde{y}_s^{(m)}[n]$ and *T*-waves of each beat $\tilde{y}_s^{(k,m)}[n]$ are represented by obtained orthogonal signals as:

$$\tilde{y}_s^{(m)}[n] = \sum_{r=1}^{2} a_{mr}^{(s)} u_r^{(s)}[n],\ \ n = 1,\ldots, N_T \quad \ldots(7)$$

$$\tilde{y}_s^{(k,m)}[n] = \sum_{r=1}^{2} a_{mr}^{(k,s)} u_r^{(k,s)}[n],\ \ n = 1,\ldots, N_T \quad \ldots(8)$$

A pair of orthogonal signals characterizes the associated *T*-wave. Hence, the TWA may result in the alternate changes in these signals. We term TWA detected by these orthogonal signals as, *T*-wave vector alternans (TWVA). Characteristics of the orthogonal signals can be captured in many ways. We adopted the norm of the center of gravity of the closed region, specified by the *T*-loop obtained from the two orthogonal signals. The *T*-loop is a phase plot of two orthogonal signals. Let $G^{(k,s)} = (g_1^{(k,s)}, g_2^{(k,s)})$ and $\overline{G}^{(s)} = (\bar{g}_1^{(s)}, \bar{g}_2^{(s)})$ be the center of gravity of the k^{th} beat in the s^{th} segment and that of averaged *T*-wave over the s^{th} segment. The norm difference sequence $\alpha^{(s)}[k]$ relative to the averaged *T*-wave norm :

$$\alpha^{(s)}[k] = \left\| G^{(k,s)} \right\| - \left\| \overline{G}^{(s)} \right\| \quad \ldots(9)$$

is constructed for each segment *s* to detect the alternate changes in the orthogonal signals. Angular fluctuation:

$$\theta^{(s)}[k] = \left\| \tan^{-1} \frac{g_2^{(k,s)}}{g_1^{(k,s)}} \right\| - \left\| \tan^{-1} \frac{\bar{g}_2^{(s)}}{\bar{g}_1^{(s)}} \right\| \quad \ldots(10)$$

could also be worth examining to detect the alternate changes. The rate of alternate changes in those sequences, as is used in the first step of coarse TWA detection, or the periodgram intensity value of the above sequence at the Nyquist frequency will be the quantitative index to reveal the presence of the TWVA. We adopted the periodgram intensity at Nyquist frequency to detect the TWVA in the second step of fine TWA detection.

s.

3. Results

To validate the proposed method of TWA detection, real data analysis has been carried out. Three channel Holter ECG recordings (*Spiderview: ELA Medical, Cedex France*) were made on ten normal subjects and ten subjects with various cardio vascular disorders (cardiac sarcoidosis, ventricular tachycardia, cardio-pulmonary arrest, angina pectoris, Brugada syndrome myocardial infarction, hypertrophic cardiomyopathy, acromegalic cardimyopathy). Data were digitized at the sampling rate of 200Hz. The results of the fine TWA detection for screened ECG data are shown in this section. Fig. 1 shows typical examples of the *T*-wave loops. The normal *T*-wave pattern is shown on the left side of the figure. Large beat-to-beat deviations in the loop patterns are seen for the data with TWA (right).

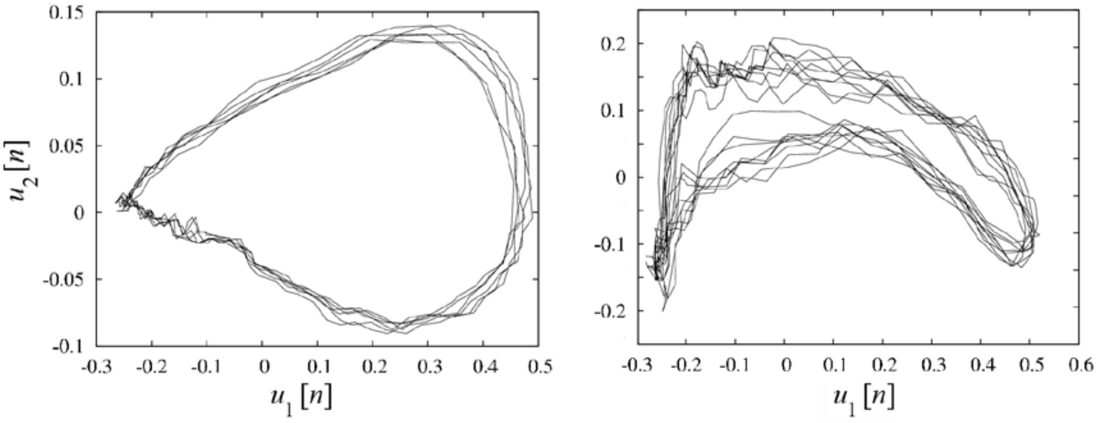

Figure 1. *Typical examples of the T-wave loop: phase plot of two orthogonal signals obtained by SVD (Left: normal subject; Right: subject with TWA).*

Fig. 2 shows typical examples of the norm difference sequence (left: normal subject; right; subject with TWA). Alternate changes are clearly seen in the sequence from the subject with TWA (right).

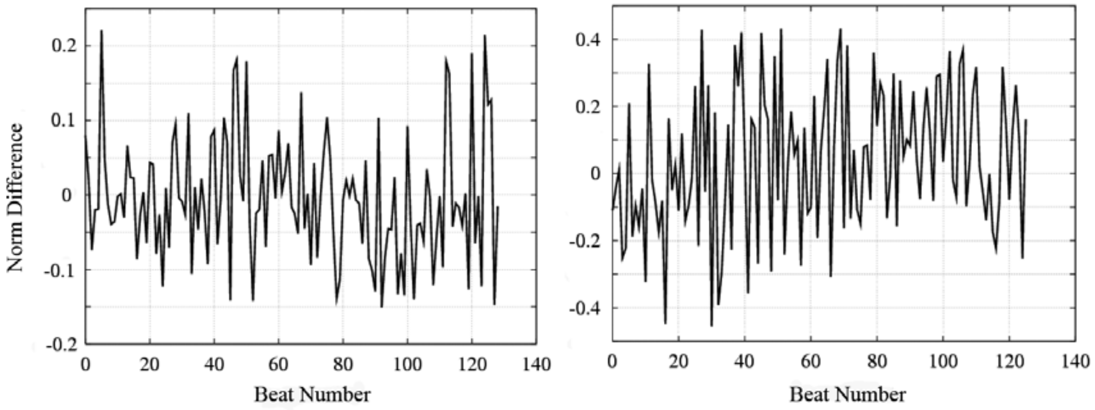

Figure 2. *Typical examples of the norm difference sequence (Left: normal subject; Right: subject with TWA).*

Fig. 3 shows the periodgram of the norm sequence defined by Eq. (9). A prominent peak at the Nyquist frequency is seen for the patient with TWA. The intensity values at Nyquist frequency for the normal subjects and the subject with TWA were and , which showed a statistically significant difference ($p<0.05$).

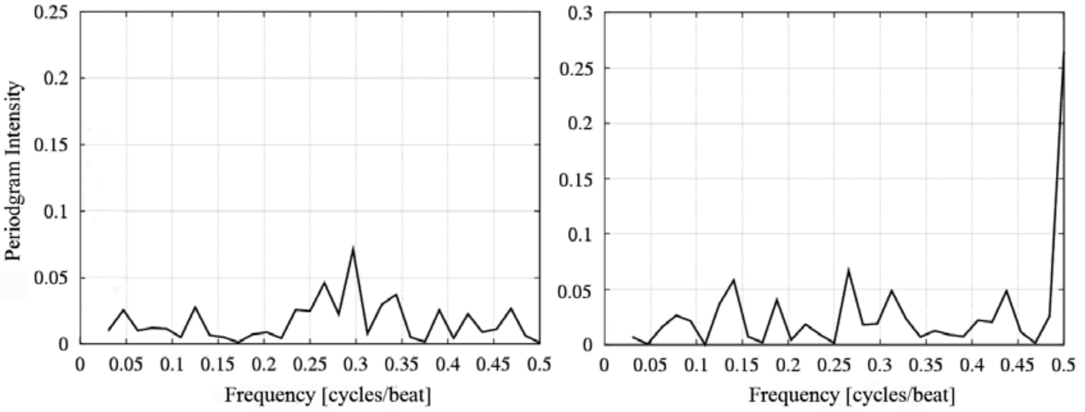

Figure 3. *Typical examples of periodgram of the norm difference sequence.*

(Left: normal subject; Right: subject with TWA).

4. Discussion

A method for detecting TWA from the Holter ECG recordings is presented. The presence of TWA is successfully detected by the proposed method for the validation data. Two orthogonal signals derived from SVD capture the T-wave waveform morphology, resulting in the clear beat-to-beat waveform distinction. It should also be noted that the SVD derived orthogonal signals, hence the T-loop patterns, are consistent, no matter if individual waveforms from X, Y and Z leads are different. This lead waveform free characteristic enables a stable TWA detection independent from the lead signal variations. We have confirmed that the norm sequence of the center of gravity of the T-wave loop is sensitive for detecting TWA. However, angular fluctuations or other quantities, e.g.the area or length of the loop, could be utilized to improve the detection accuracy. Baseline collection is important for accurate TWA detection, since Holter ECG raw data include considerable baseline fluctuations. The smoothing spline function is utilized before the final T-wave analysis using SVD. This procedure and applying SVD to ECG signals, are time consuming. The simplified coarse TWA detection in the proposed method speeds up the entire twenty four hour ECG data processing. However, to ensure the accuracy, it is desirable to apply the fine TWA detection utilizing SVD to entire record. Development of a dedicated high speed hardware computation device may enable researchers to realize this scheme. Although further large scale validation is necessary, the method contributes to progress toward realizing the TWA detection based on Holter ECG recordings under a non-controlled test environment.

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