Estimation of T-Wave Alternans from Multi-Lead ECG Signals Using a Modified Moving Average Method

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Abstract

T-wave alternans (TWA) manifests on the surface ECG as a pattern of alternating amplitude T-waves, typically in the microvolt range. Consequently, TWA is usually invisible at standard ECG display scales and must be detected using signal processing methods. There is a predictive relationship between TWA and sudden cardiac death, so proper detection and estimation may contribute to clinical decision-making. The objective of the 2008 Physionet/Computers in Cardiology Challenge was to estimate the magnitude of TWA in a dataset of 100 multi-lead ECGs. We used a modified moving average method (MMA) to detect and estimate TWA. We found that the TWA magnitude ranged from to 1.3 to 256.9 μV (43.9 ± 49.7 μV). The Kendall rank correlation coefficient obtained for the challenge was 0.451. In conclusion, the MMA method can be used for detection and estimation of TWA, though additional work could further improve the method.

1. Introduction

T-wave alternans (TWA) is characterized in the surface ECG by alternating T-wave amplitudes, usually at heart rates in the range of 90 to 120 beats per minute. TWA magnitude is typically in the microvolt range, so it is not evident at normal ECG display scales [1]. TWA was considered to be rare finding until in 1981, Adam et al. demonstrated that it occurred much more frequently than previously assumed; however, since it occurs at the microvolt scale, it is usually invisible using the standard ECG display.

It has been shown that there is a predictive relationship between TWA and the risk of sudden cardiac death [2,3]. Consequently, proper detection and estimation of TWA is potentially important in determining proper and cost effective patient care, particularly regarding the decision to implant a prophylactic ICD. A wide variety of signal processing methods have been presented to accomplish this task, but further investigation is needed to determine which method provides the most accurate performance. Currently, there is no generally agreed upon method which may be treated as the "gold standard" [4].

The objective of the Physionet/Computers in Cardiology Challenge 2008 was to estimate the peak magnitude of TWA for a specified dataset of multi-lead ECGs using a fully automated method. This allowed for comparison between a variety of signal processing techniques for TWA estimation using the same dataset. Each entry was compared with a reference ranking and assigned a Kendall rank correlation coefficient (1 = perfect agreement, -1 = perfect disagreement) in order to determine which method performed the best relative to all of the entries.

We chose to investigate the modified moving average (MMA) method because it has been shown to be a robust signal processing method capable of measuring TWA in the presence of noise and artifact [4]. It has been shown to be a capable method for risk stratification in low-risk post myocardial infarction patients [5]. It has also been utilized in the evaluation of risk in patients with implantable cardioverter defibrillators [6]. The MMA method has been successfully used in commercial equipment such as the CASE 8000 (GE Medical Systems, Milwaukee, WI).

In the MMA method, a minimum of signal averaging is utilized in order to avoid the assumption that data is stationary over long periods of time, which is an assumption made by several other methods. Spectral methods, for instance, often require the assumption that the heart rate must be stable for at least two minutes [7]. This requirement of data stationarity may be particularly problematic when applying TWA measurements to clinical applications such as exercise treadmill testing and ambulatory electrocardiography, because in these cases the heart rate cannot be considered to be stationary by the very nature of these methods. Another important advantage of the MMA method in comparison with other methods for TWA estimation is its straightforward implementation.

2. Methods

This paper presents a detailed algorithm for detection and estimation of TWA using a MMA method on the Challenge dataset. The data consisted of 100 multi-

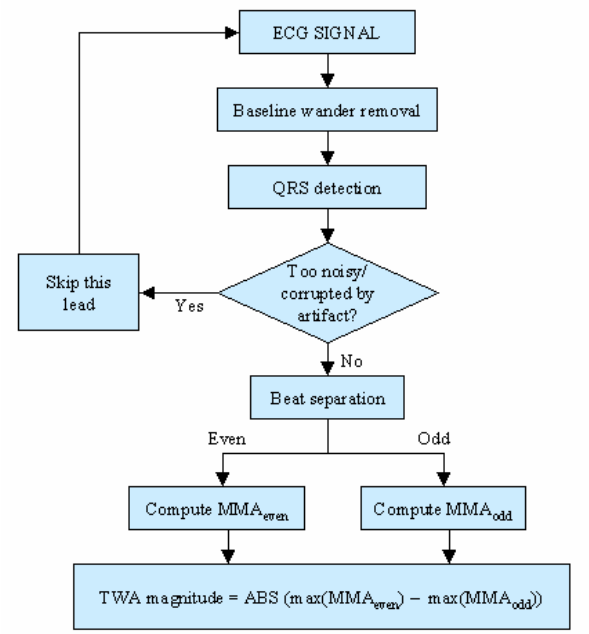

Figure 1: Flow chart demonstrating the major steps in the implementation of the MMA method to estimate TWA magnitude.

lead ECGs sampled at 500 Hz with 16 bit resolution over a ± 32 mV range. All of the signal processing was accomplished using MATLAB (The Mathworks Inc., Natick, MA). Pre-processing of the data included removal of baseline wander by high-pass filtering using a fifth order elliptical filter with a cutoff frequency of 0.5 Hz. QRS detection was accomplished by a modified version of the Pan and Tompkins algorithm [8]. The signals were filtered with a low-pass filter (cutoff frequency = 50 Hz) to remove high frequency noise; the square of the derivative of the filtered signal was then calculated. Peaks were extracted from this signal using thresh-holding. Leads in which QRS detection was poor because they were noisy or corrupted by artifact following preprocessing were eliminated from additional analysis.

Following QRS detection, individual beats were identified and isolated for comparison using a windowing scheme. Even and odd beats were separated into two different streams of beats for analysis [7]. The algorithm initialized the MMA_{even} array to the values which comprised the first even beat. Then the next even beat was compared with the MMA_{even} array on a point-by-point basis. For each point comparison, if the value of the next even beat at that particular point was greater than the corresponding value in the MMA_{even} array, then the element of the MMA_{even} array was increased by a given amount (Δ_{even}). Otherwise, if the value of the next even beat was smaller than the corresponding value in the MMA_{even} array, then the value in the MMA_{even} array was decreased by a given amount (Δ_{even}). This procedure was repeated for all beats in the even beat stream to produce MMA_{even}. The same procedure was followed with the odd beat stream to produce MMA_{odd}. The scheme used for computing Δ_{even} or Δ_{odd} in this study was the same as the one used by Nearing et al [7] and is duplicated here in Equations 1 and 2 for reference:

Computed beat $Even_n(i)$
= Computed beat $Even_{n-1}(i) + \Delta_{even}$

$$\begin{aligned}
\Delta_{even} &= -32 && \text{if } \eta \leq -32 \\
\Delta_{even} &= \eta && \text{if } -1 \geq \eta > -32 \\
\Delta_{even} &= -1 && \text{if } 0 > \eta > -1 \\
\Delta_{even} &= 0 && \text{if } \eta = 0 \\
\Delta_{even} &= 1 && \text{if } 1 \geq \eta > 0 \\
\Delta_{even} &= \eta && \text{if } 32 \geq \eta > 1 \\
\Delta_{even} &= 32 && \text{if } \eta \geq 32
\end{aligned} \qquad \text{Eq. 1}$$

where n is the n^{th} beat of the even beats and
η= [ECG beat $Even_{n-1}(i)$ – computed beat $Even_{n-1}(i)$]/8.

Computed beat $Odd_n(i)$ =
computed beat $Odd_{n-1}(i) + \Delta_{odd}$

$$\begin{aligned}
\Delta_{odd} &= -32 && \text{if } \beta \leq -32 \\
\Delta_{odd} &= \beta && \text{if } -1 \geq \beta > -32 \\
\Delta_{odd} &= -1 && \text{if } 0 > \beta > -1 \\
\Delta_{odd} &= 0 && \text{if } \beta = 0 \\
\Delta_{odd} &= 1 && \text{if } 1 \geq \beta > 0 \\
\Delta_{odd} &= \beta && \text{if } 32 \geq \beta > 1 \\
\Delta_{odd} &= 32 && \text{if } \beta \geq 32
\end{aligned} \qquad \text{Eq. 2}$$

where n is the n^{th} beat of the odd beats and
β = [ECG beat $Odd_{n-1}(i)$ – computed beat $Odd_{n-1}(i)$]/8.

The T-waves in each of the templates (MMA_{even} and MMA_{odd}) were isolated for comparison. The magnitude of TWA was computed as the absolute value of the difference between the peak amplitudes of the T-waves in MMA_{even} and MMA_{odd}. This method is depicted graphically in the flowchart found in Figure 1. The TWA magnitude was computed by this method for each of the available leads, and the largest of the TWA magnitude estimates for all of the recorded leads was used to produce the final overall estimate of TWA peak magnitude.

In addition to the MMA method described, we also computed the magnitude of the TWA by finding the standard mean (not weighted) and median rather than the modified moving average. The procedure involved removal of baseline wander and QRS detection in the same way as the MMA method. For the standard mean calculation, aberrant beats were removed from analysis by template matching. Specifically, if a beat had at least a 90% correlation with the median beat in its stream (even or odd), then it was included in the group of beats used to compute the mean. Otherwise, that particular beat was eliminated from further analysis. Aberrant beat rejection was not necessary for the median since it does not suffer from the influence of outliers. We computed both the mean and median even and odd beats for each lead. The TWA magnitudes were computed as the absolute value of the difference between the peaks of the T-waves of the two mean beats and the absolute value of the difference between the peaks of T-waves of the two median beats. Flowcharts demonstrating these procedures used for determination of the TWA magnitudes are shown in Figures 2 and 3.

We chose to compute the standard mean in order to determine how weighting the mean (as in the MMA method) affects the calculation of TWA magnitude. We also chose to examine the median because it is simpler to compute than the modified moving average and does not require the use of thresholds, which were empirically determined for the MMA method.

Paired t-tests were performed to compare the MMA method with each of the other methods (mean and median). The magnitudes of the TWA for the MMA method were submitted for scoring to the Challenge.

3. Results

The TWA magnitude ranged from 1.3 to 256.9 µV for the Challenge dataset of 100 ECGs (mean ± standard deviation, 43.9 ± 49.7 µV) using the MMA method. The TWA magnitude ranged from 0.4 to 269.4 µV (17.8 ± 32.9 µV) using the standard mean beat method. The TWA magnitude ranged from 1.0 to 253.2 µV (21.2 ± 43.9 µV) using the median beat method. Of the three methods, the MMA method resulted in the highest mean and highest standard deviation for the peak TWA magnitude. The MMA method also resulted in TWA magnitudes which covered the largest range of the three methods. The standard mean beat method produced the lowest mean and standard deviation of TWA magnitudes. The p-value of the t-test comparing the MMA and mean methods was $4*10^{-7}$; the p-value of the t-test comparing the MMA and median methods was $8*10^{-5}$.

The final score (Kendall rank correlation coefficient) obtained in the Challenge for the TWA magnitude measurement obtained using the MMA method was 0.451.

4. Discussion and conclusions

The MMA method has previously been shown to be

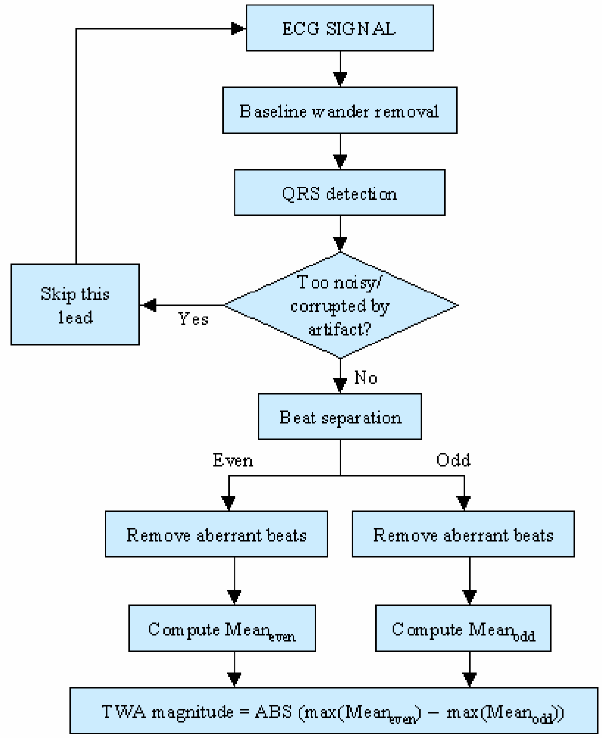

Figure 2: Flow chart demonstrating the major steps in the implementation of the standard mean method to estimate TWA magnitude.

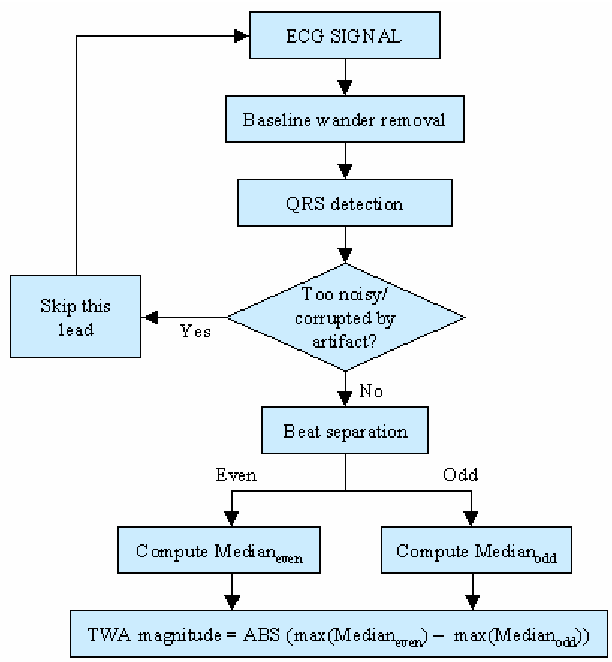

Figure 3: Flow chart demonstrating the major steps in the implementation of the median method to estimate TWA magnitude.

robust in terms of TWA estimation in the presence of noise and artifact [7]; however, its accuracy is highly dependent upon the accuracy of QRS detection. This dependency exists since alternating beats must be correctly assigned as either "odd" or "even" beats in order to measure the differences in the averaged alternating beats. If, for instance, an even QRS complex is missed near the middle of the signal, then all following T-waves would be mis-assigned to the odd stream. The estimated TWA magnitude would be averaged out and would, of course, be inaccurate.

Another consideration in using the MMA method is how the values of Δ_{even} and Δ_{odd} affect the results. Since these values are empirically chosen, they may not perform as well for all datasets. The values used in this study followed those described by Nearing et al. [7], but that does not necessarily imply that they will work well for all realistic datasets. Further study is required to validate these empirical parameters on a wide range of ECG data.

The value of TWA magnitude was also computed using the standard mean and median in order to compare those results with the MMA. It is simpler to compute the mean or median rather than the MMA. In addition, the median does not require aberrant beat rejection, so even less processing is needed. Furthermore, the mean and median are both simply derived from the overall signal, rather than needing to process the signal sequentially as is the case for MMA. Nevertheless, both the mean and median methods make the assumption that the data is stationary, whereas the MMA method is advantageous because such an assumption is not necessary. Nevertheless, as expected, the p-values comparing the results of the methods indicated that there is a statistically significant difference between the MMA method and the other two methods. This implies that neither the standard mean nor the median would be a suitable replacement for the MMA.

A limitation for this work is that using any of the three presented methods to find the TWA magnitude evaluates the magnitude of the entire T-wave. However, it has been shown that during ischemia, the alternation is generally observable in only the first half of the T-wave [9]. Consequently, methods such as MMA may underestimate the degree of TWA [10]. Further work is necessary to investigate this issue.

Further work to improve the MMA algorithm should include enhanced methods of combining data from multiple ECG leads. For this study, we used the largest of the TWA estimates from the leads for each patient. An improvement to this method would involve a sophisticated method to combine information from multiple leads in order to determine the most accurate estimate for TWA.

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References

[1] Adam DR, Akselrod S, Cohen RJ. Estimation of ventricular vulnerability to fibrillation through T-wave time series analysis. *Computers in Cardiology* 1981;8:307-310.

[2] Armoundas AA, Tomaselli GF, Esperer HD. Pathophysiological basis and clinical application of T-wave alternans. JACC 2002;40:207-17.

[3] Nieminen T, Lehtimaki T, Viik J, Lehtinen R, Nikus K, Koobi T, Niemela, Turjanmaa V, Kaiser W, Huhtala H, Verrier RL, Huikuri H, Kahonen M. T-wave alternans predicts mortality in a population undergoing a clinically indicated exercise test. Eur Heart J 2007 Oct:28(19);2332-7.

[4] Martinez JP, Olmos S. Methodological principles of T wave alternans analysis: a unified framework. IEEE Trans on Biomed Eng 2005:52(4);599-613

[5] Verrier RL, Nearing BD, La Rovere MT, Pinna GD, Mittleman MA, Bigger JT Jr, Schwartz PJ, ATRAMI Investigators. Ambulatory electrocardiogram-based tracking of T wave alternans in postmyocardial infarction patients to assess risk of cardiac arrest or arrhythmic death. J Cardiovasc Electrophysiol 2003;14(7):705-711.

[6] Kop WJ, Krantz DS, Nearing BD, Gottdiener JS, Quigley JF, O'Callahan M, DelNegro AA, Friehling TD, Karasik P, Suchday S, Levine J, Verrier RL. Effects of acute mental stress and exercise on T-wave alternans in patients with implantable cardioverter defibrillators and controls. Circulation 2004:109(15);1864-1869.

[7] Nearing BD, Verrier RL. Modified moving average analysis of T-wave alternans to predict ventricular fibrillation with high accuracy. J Appl Physiol 2002:92;541-549.

[8] Pan J, Tompkins WJ. A real-time QRS detection algorithm. IEEE Trans Biomed Eng 1985 Mar;32(3):230-6.

[9] Nearing BD, Huang AH, Verrier RL. Dynamic tracking of cardiac vulnerability by complex demodulation of the T wave. Science 1991:252(5004);437-440.

[10] Verrier RL, Kwaku KF, Nearing BD, Josephson ME. T-wave alternans: does size matter. J Cardiovasc Electrophysiol 2005:16(6);625-628.

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