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A Biomedical Signal Segmentation Algorithm for Event Detection Based on Slope Tracing

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Abstract—**In this paper a simple signal segmentation algorithm is introduced. The algorithm determines the epochs of signal components of interest based on signal characteristic such as amplitude, slope, deflection width, or distance between neighboring deflections. The epochs are segmented indirectly by means of a slope trace wave that traces a signal with its average slope and predetermined delay. The algorithm is applied to ECG and electrogram to show its practical applicability and efficiency. It is found that the algorithm can be used to choose particular signal components appropriately without significant signal preprocessing or complexity.**

I. Introduction

BIOMEDICAL signals carry information on physiological events of both normal and pathological processes, and the information usually resides in waveform deflections. Because noises or artifacts also cause signal deflections, any deflection has a chance to be either the event we are looking for or noises we are trying to eradicate. The part of a signal related to a specific event of interest is often referred to as an epoch of which identification is an important step for biomedical signal analysis. Analysis of a signal for monitoring or diagnosis requires the identification of epochs and investigation of the corresponding events [1].

Usually, the event of interest is determined by various methods of signal processing or algorithms using known signal characteristics of its waveform such as signal amplitude, morphology, time duration, and frequency contents. For QRS complex detection of ECG, the derivative based methods have been suggested [2],[3]. These methods are based on the fact that the QRS complex has the highest slope and amplitude in ECG, and enhance the amplitude within the epoch of QRS complex while suppressing the other signal components such as P-wave, T-wave, noises, artifacts, or interference. In the implantable pacemakers or ICDs (implantable cardioverter defibrillator), the depolarization wave detections of atrium or ventricle are based on amplitude threshold [4]-[6], after an electrogram passes through a band pass filter designed particularly for p-waves or r-waves. A biomedical signal may include more than one events of interest. Examples are P-wave or T-wave detection along with QRS complex in ECG, or far-field or after potential detection in electrograms for automatic sensing or capture verification in pacemakers or ICDs [7]. Sometimes, monitoring noise level or interference amplitude along with physiological events are also desirable for proper amplitude threshold setting between noises and event amplitudes for automatic sensing threshold determination. In addition, some signal analysis requires to preserve the signal morphology or the average signal level that can be easily distorted by signal processing. Examples include the ST segment, T-wave alternant, or intra-cardiac pressure signals [8].

Therefore, it is desirable to detect the events of interest not only from physiological causes but also from artifacts or interference while minimizing the change of signal morphology including a DC component. In doing so, the signal preprocessing should be minimized to prevent considerable frequency or phase distortion.

In this paper, we introduce a simple but interesting signal segmentation algorithm that separates the interval of signal component based on such signal characteristic as amplitude, slope, deflection width, or distance between neighboring deflections. The epochs are segmented indirectly by means of a slope tracing wave that traces a signal with its average slope and predetermined delay. When the slope tracing wave crosses the original signal, the algorithm determines the boundary between neighboring signal deflections. By choosing two heuristically determinable parameters for the period of averaged slope and delay, the algorithm can identify the epoch of deflection that has particular morphological characteristic.

In order to examine the practical applicability of the algorithm, it has been applied to several biomedical signal analysis cases such as QRS complex segmentation, T- and P-wave segmentation in ECG, or p- or r-wave segmentation in a raw electrogram for pacemakers or ICDs.

II. The Slope Trace wave

To determine the epoch of particular event of interest effectively, two slope tracing waves, the descending slope trace wave (DSTW) and the ascending slope trace wave (ASTW), are proposed as shown in Fig. 1 and 2, respectively, although only the DSTW is used in this paper. In Fig. 1, the DSTW (thick line) follows the original signal (thin line) in the ascending interval (a-p interval) and traces the signal with t_H a delay and average slopes that are updated continuously using several previously sampled data during time interval

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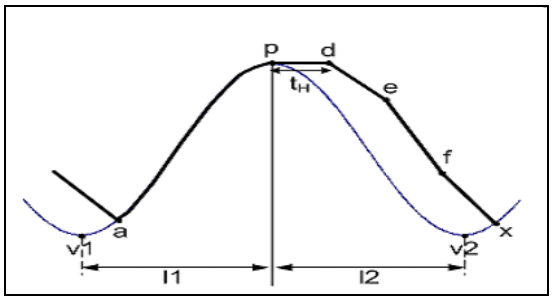

Fig. 1 The behavior of the descending slope trace wave (thick line), the DSTW, tracing sinusoidal wave (thin line).

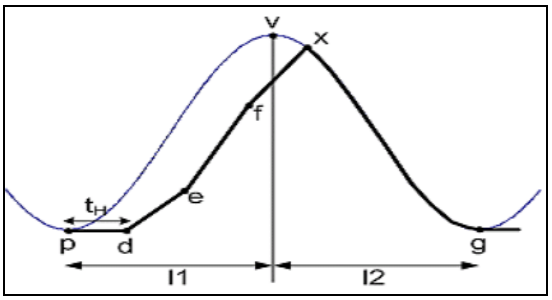

Fig. 2 The behavior of the ascending slope trace wave (thick line), the ASTW, tracing sinusoidal wave (thin line).

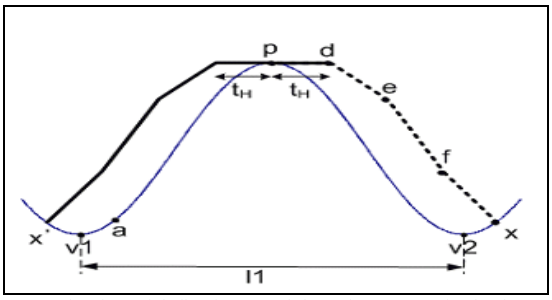

Fig. 3 Determination of deflection epoch by using DSTW in the forward and reverse direction.

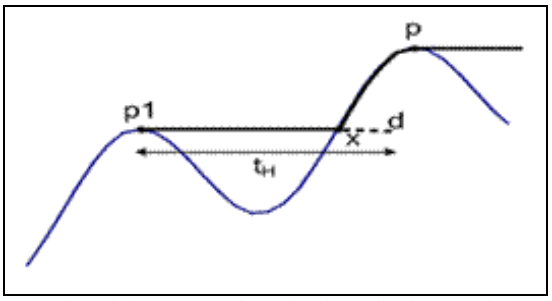

Fig. 4 Behavior of the DSTW when signal amplitude exceeds the DSTW during delay period t_H. The first peak p_1 is ignored because signal exceeds DSTW within the delay.

t_H in the descending interval (p-x interval). The DSTW is det ermined as following:

$$\begin{aligned} &If\ DSTW[n\text{-}1] < x[n] \\ &\quad DSTW[n] = x[n] \qquad (1) \\ &else\ \ (that\ is, DSTW[n-1] \geq x[n]) \\ &\quad DSTW[n] = DSTW[n-1],\ \ for\ N\ sample\ after\ peak\ detection \qquad (2) \\ &\quad DSTW[n] = DSTW[n-1] - [Avg\_S],\ \ elsewhere \qquad (3) \\ &here,\ \ |Avg\_S| = \left| \sum_{i=n-N+1}^{n} \left(x[i] - x[i-1] \right) \right| / N \qquad (4) \end{aligned}$$

where x[n] represents the signal amplitude sampled at n and $|Avg\_S|$ is the average slope of signal during time interval t_H that includes N samples.

After generation of a present sample of a signal, the algorithm sets DSTW[n] to the amplitude of the present sample x[n] of signal if x[n] is higher than DSTW[n-1] determined at previous sampling moment, as in (1). This is shown in Fig. 1 as the interval between a and p. After the peak of a deflection, the algorithm sets DSTW[n] to the peak amplitude and keeps it for a predetermined time delay t_H that includes N samples (between p and d) as described in (2). At the end of the delay period (point d), the algorithm calculates the average slope of signal, $|Avg\_S|$, by dividing the accumulated amplitude difference between neighboring N+1 samples by N using (4), and then starts to descend the DSTW by the average slope at every following sampling moment for the next same period t_H as in (3). At the end of the first descending period (point e), the average slope is recalculated and updated in the same way as before using (4) and used for next period, and so on.

If the recalculated average slope is less than 75% of the previous average slope, the algorithm chooses the 75% of the previous average slope and uses it for next period, in place of the recalculated average.

When the amplitude of DSTW[n] becomes lower than sampled signal amplitude (point x), the algorithm sets the DSTW[n] to the amplitude of the signal using (1) and determines one side of event deflection epoch. The other side of the epoch is determined after the algorithm reaches the end of delay period (point d), where the algorithm picks the peak amplitude as a temporary DSTW and follows the same procedure explained above but in the backward direction, assuming enough number of previous sampled data is stored, as shown in Fig. 3 (thick solid line). When the temporary DSTW in the backward direction becomes lower than signal amplitude, the algorithm determines the other side of the epoch (point x').

In Fig. 1, the algorithm divide the deflection into two interval l_1 and l_2, from valley to peak and from peak to next valley, respectively, whereas in Fig. 3, two intervals are represented as one. How to divide the intervals may depend on signal characteristic or purpose of epoch determination.

If a sampled signal amplitude x[n] becomes higher than DSTW[n-1] within the delay period t_H as shown in Fig. 4, the algorithm ignores the current peak p1 and trace the signal using Eq. (1) because the sampled amplitude is higher than previously determined DSTW. This behavior is useful because it allows us to consider two neighboring deflections as one or as two different deflections by choosing delay period value (t_H) differently based on the waveform characteristic of interest. This feature can be used to ignore the small deflection

superimposed into a larger deflection and reduce the number of epochs determined to be analyzed.

After epoch of a signal deflection is determined using DSTW, the algorithm also uses an amplitude threshold or an epoch interval threshold to choose deflections of particular morphological characteristics. The amplitude threshold is compared to the amplitude difference between the peak amplitude and the signal amplitude at either right or left epoch boundary whichever is lower. If necessary, two amplitude thresholds can be used to choose event deflections whose amplitudes are between two different levels.

Fig. 5 shows an example of DSTW tracing an ECG (record 101 MIT/BIH database). The delay t_H is 25.0ms (N=9) and amplitude threshold is 20. The solid lines covering both sides of the deflection are the forward and backward DSTW, respectively, and the long vertical line at the right side of a deflection is the right side of the determined epoch and the shot vertical line is the left side of the epoch. As can be seen, the deflections of QRS complex, P-wave, and T-wave are segmented properly while small amplitude and high frequency deflections are ignored.

III. Application of the Signal Segmentation Algorithm

The algorithm has been applied to several ECGs to check the practical applicability and efficacy in signal segmentation. Fig. 6 shows the epochs determined for QRS complexes. After an epoch is determined, an amplitude threshold is compared to the amplitude difference within the epoch, eliminating low amplitude deflections. The amplitude threshold is chosen properly after reviewing the signal and compared to the difference of maximum and minimum in the epochs determined by DSTW.

The algorithm is also applied to determine P-wave and T-wave epochs as shown in Fig. 7. After an epoch is determined by DSTW, the amplitude between two thresholds are used to remove high amplitude deflections like QRS complexes and very low amplitude deflections.

In order to see the behavior of the algorithm during baseline wandering, the algorithm is applied to an ECG shown in Fig. 8. The algorithm determines the QRS complex epochs and T-wave epochs appropriately regardless of baseline wandering. The amplitude threshold is chosen higher than P-wave amplitude difference in this case.

The algorithm is then applied to electrograms (CPI bipolar, 4269 for atrium and 4261 for ventricle), as shown in Fig. 9, which had passed through a bandpass filter of 3-300Hz. Because this pass band is much wider than that of usual pacemaker sensing filter, the signal contains the far-fields and T-waves that are usually eliminated by normal pacemaker filters. Actually, the electrograms shown in the figure are inverted in polarity to apply DSTW in this paper, although ASTW could be used for original non-inverted signals. In the figure, the upper signal is an atrial electrogram with determined p-wave epochs and the lower signal is ventricular electrogram with r-wave and T-wave epochs. By choosing the shorter delay and low amplitude threshold, it is found that the far-filed in the atrial electrogram caused by ventricular depolarization can be segmented.

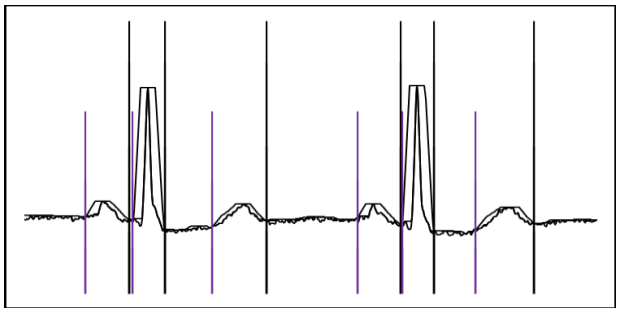

Fig. 5 The DSTW tracing an ECG and the determined epoches (record 101 in MIT/BIH database, delay =25.0 *ms* , amplitude threshold=20).

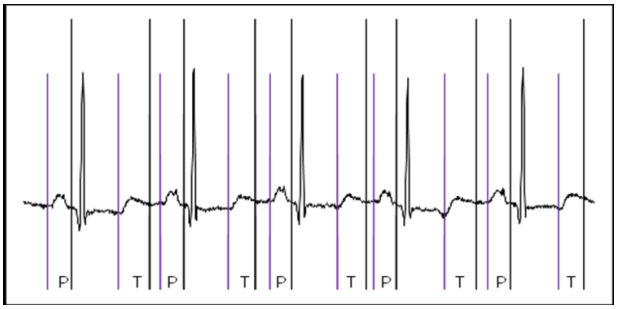

Fig. 7. Epochs determined for P- and T- waves by the DSTW (record 100 in MIT/BIH database, delay=9 *ms* , amplitude threshold1=10, and amplitude threshold=80)

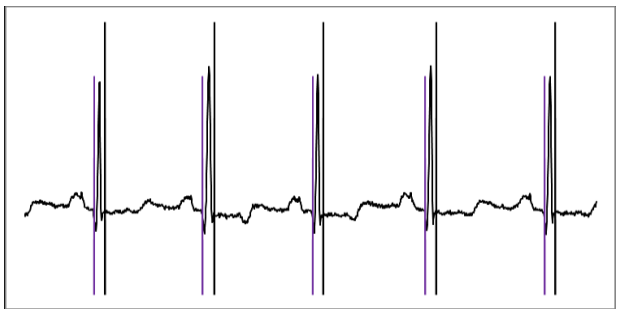

Fig. 6. Epochs determined for QRS complexes by the DSTW (record 100 in MIT/BIH database, delay=7 *ms* , amplitude threshold=50).

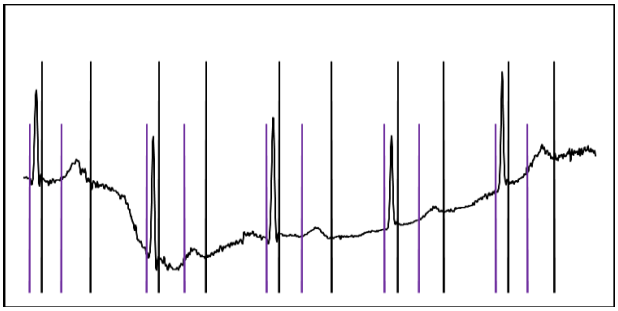

Fig 8. Epoch determination for QRS complexes and T-waves during baseline wandering (record 103 in MIT/BIH database, delay=7 *ms* , amplitude threshold=30).

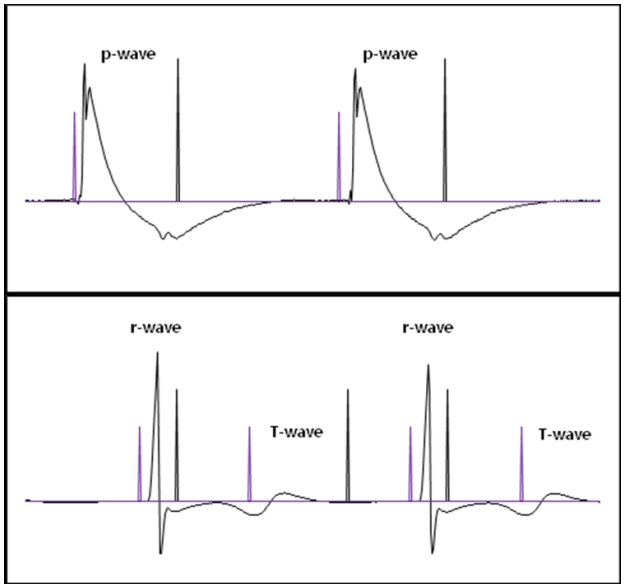

Fig. 9. Epochs determined in a DDD electrogram for p-wave in the atrial electrogram (upper) and for r-wave and T-wave in the ventricular electrogram (lower).

IV. Discussions

Usually, biomedical signal passes through a preprocessing stage before analysis or recognition. Well-processed signal simplifies the following analysis greatly and reduces errors significantly. However, because signal processing involves filtering, the processed signal experiences morphological change caused by amplitude and phase characteristic of the filter.

Epoch determination is another important step in signal analysis because a properly determined epoch can help to reduce analysis time significantly, removing the waveform deflections that do not seem to have the morphological characteristic of event we are looking for. The epoch determination algorithm introduced in this paper is based on morphological characteristic of a signal in time domain and intended to be applied to biomedical signals without preprocessing or with minimized preprocessing. For example, Fig. 10 compares raw (upper signal, 3-300Hz bandpass filtered) and filtered (lower signal, 18-100Hz bandpass filtered) ventricular electrograms. The T-waves after ventricular depolarization wave can be seen clearly in the raw signal while they are removed in the filtered signal. Although the r-wave detection is the prime concern in this case, the T-wave also includes valuable information that can be used for various clinical or operational benefits.

Although the algorithm has performed well in the examples shown in Fig. 6-9, we observed several cases of improper epoch determination, caused mainly by the change of waveform of deflection. The fixed delay and amplitude threshold chosen initially became improper for significantly deformed waveforms.


Fig. 10. Comparison of raw electrogram (upper) and filtered electrogram (lower) from ventricle.

Usually, the algorithm has determined the epochs of signal deflection well when waveform morphology remained unchanged and is able to segment only the deflections of particular characteristic as shown in Fig. 6 and 7. In this paper, we used the heuristically obtained delay and amplitude threshold after visual signal examination. At present, the method of obtaining optimal delay and amplitude is under development.

In addition, we saw the potential that the algorithm can be used as an independent event detector without further analysis. More quantitative and qualitative analysis of QRS complex detection on the standardized data is currently under investigation.

Although only DSTW is used in this paper, both DSTW and ASTW can be applied simultaneously for better deflection segmentation or analysis. Example of the use of both trace waves are shown in the accompanying paper [9] that remove high frequency deflections surgically in ECG for event detection or baseline removal.

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