COMBINATION OF IMPEDANCE CARDIOGRAPHY (ICG) AND ELECTROCARDIOGRAPHY (ECG) IN DETECTING CHARACTERISTIC POINTS ON ICG CURVE

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ABSTRACT

Detecting precisely positions of the characteristic points on the Impedance Cardiography (ICG) signal's curve has been being of interest for decades, since it directly affects the accuracy of the calculations of some important Hemodynamics. Recently, a new algorithm called Car. algorithm was proposed using solely ICG signal to find the exact positions of the most two important points, B and X points. These positions were also re-defined by the same authors by synchronizing ICG with Echocardiogram. Based on those results, this paper aims to propose a combination of ICG and Echocardiography (ECG) to overcome the drawbacks of the above algorithm to determine those B's and X's positions precisely even in ICG signal with respiration. In order to evaluate the performance, 10 healthy volunteers aging from 18 to 23 were measured in non-breathing condition to obtain 10 ICG signals without respiration (called "clean" signals). Then, the proposed algorithm was applied in 10 "clean" ICG signals and the same 10 ICG signals with simulated respiration to evaluate the positions of detected points, in which Left Ventricular Ejection time (LVET), which is B-X duration, and Pre-Ejection Period (PEP), which is Q-B duration (Q of ECG and B of ICG), were calculated for comparison. The final results proved the capability of the developed method in identifying the main characteristic points B and X with significantly smaller error in breathing ICG signals compared to Car. algorithm.

1. INTRODUCTION

Impedance Cardiography (ICG) is a non-invasive method to continuously monitor the cardiac function and is an effective tool to evaluate the Hemodynamic parameters. Important diagnostic parameters are related to characteristic points on the ICG signal, such as B and X points marking the opening and the closure of the aortic valve respectively. Therefore, detecting precisely positions of those characteristic points on the ICG curve is vital since it is one of major factor affecting the accuracy of calculated Cardiac Output (CO) and other Hemodynamic parameters. Many traditional methods were proposed to find B and X points based on three signals including ICG, electrocardiography (ECG) and Phonocardiography (PCG), which is summarized in Chapter 2 of G. Cybulsky's book (2011). Recently, an experiment of redefining the positions of B and X points was done convincingly by P. Carvalho et. al. (2011) by synchronizing ICG and echocardiogram (Ultrasound method), proving that the conventional locations of B and X points were not precise the opening and closing of the aortic valve. P. Carvalho et. al. (2011) also proposed an algorithm to determine new B and X points using solely ICG and its high-order derivatives, called Car. algorithm (Figure 1).

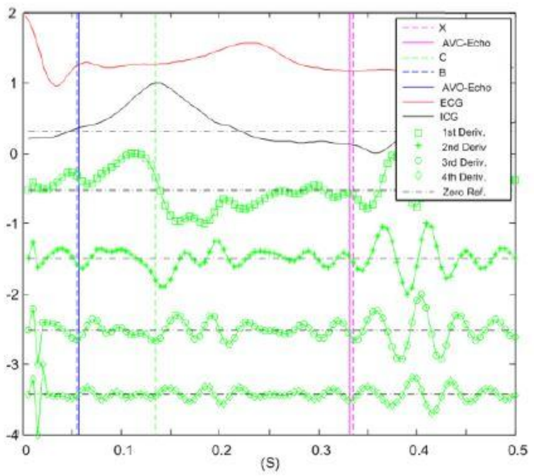

Figure 1. Car. algorithm. High order derivatives of ICG are used to find the positions of B and X points that are nearest to the results of Echocardiography

However, this algorithm only works well in ICG signals without wandering baseline noise, mostly created by respiration. We made a small experiment of applying Car. algorithm in a stable ICG signal and in the same ICG signal but being added with simulated respiration. The positions of B and X found in signal with respiration were compared to those in stable signal. Results showed that the B and X points were drifted heavily due to the effect of respiration (Figure 2). The reasons is that the old positions of B and X are used in algorithm as indications in detecting new B and X points, and these conventional positions are hardly determined precisely when the baseline noise occurs.

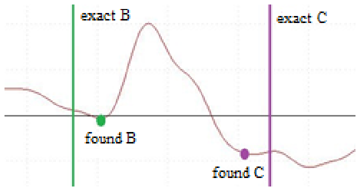

Figure 2. Results of determining B and X points on ICG signal with simulated respiration using Car. algorithm

However, the respiration signal's frequency often ranges from DC to 2 Hz. Hence, removing respiration signal from ICG also often leads to the loss of some low-frequency components of ICG and affects the Hemodynamics calculations to some extents.

In order to overcome this problem, a solution based on Car. Algorithm is proposed in this paper that combines ICG signal and Electrocardiography (ECG) signal to precisely detect the positions of B and X points even in ICG signal containing respiration signals without any noise removal processing. Detail of the proposed method is given in section 2. In section 3, we survey the algorithm's performance with two types of ICG signals: stable signals without respiration (called "clean" ICG) and signals with simulated respiration. A discussion about the results is presented at section 4 and, finally, a conclusion about our work is made in the last section.

2. THE PROPOSED ALGORITHM

2.1. The overall of the proposed algorithm

The overall summarization of the proposed algorithm is presented in Figure 3. Because ECG is also included, there are some pre-processing needed for both ECG and ICG signals before finding B and X points. According to B. E. Hurwitz et.al. (1993) [, the full bandwidth of ICG could expand to 50Hz with high heart rate (above 150 beats/min). Nevertheless, if the amplitude of B, C and X points is not a concern, their positions are still reserved well within the bandwidth of 10Hz (containing main power of ICG). Therefore, idea of this algorithm is to find B, C and X points on filtered ICG with the bandwidth of 20Hz, than marking the same positions on the original ICG signal. Filtering ICG is to help remove high frequency noises that could lead to wrong indications in its derivatives used for detecting B and X, and a 20Hz FIR low-pass filter is implemented here. After filtering the signal, because C point also takes part in finding B and X, the peak detection method of Eli Billaurer's will be used to detect C. Finally, the 1st, 2nd and 3rd derivatives of the signal are calculated using backward derivative method:

$$dy/dt = (\Delta y_i - \Delta y_{i-1}) / (t_i - t_{i-1}) \qquad (1)$$

In terms of ECG, because it is only used for supporting ICG signal, a stable ECG with clear positions of R peaks and Q points is needed rather than reserving its full information. As a result, ECG is filtered by a 1Hz FIR high-pass filter and a 20Hz FIR low-pass filter in order to remove all unrelated noises such as respiration and power-line noise. R peak is then detected also using Eli Billaurer's algorithm, and Q point is finally found as the first valley before R peak.

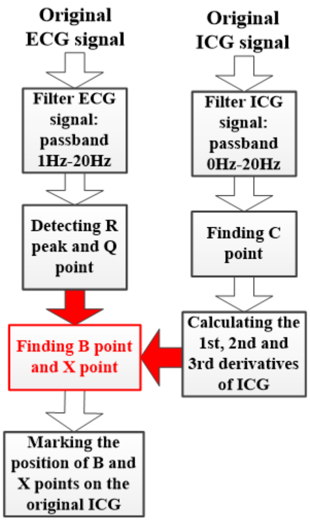

Figure 3. The overview of the proposed algorithm

2.2. B and X detection algorithm

The proposed algorithm uses the same method as Car. algorithm in detecting B and X by detecting zero-crossing events in the 1st, 2nd and 3rd derivatives of the ICG. However, this algorithm tends to add some limitations to limit the scope of B and X detection. According to A. M. Weissle et.al. (1968), the Pre-ejection period (PEP), also the duration between Q point of ECG and B point of ICG, was proved to be smaller than a specific value calculated as below:

$$PEP(min) = -0.0004 \times HR + 0.133 - 0.013 \qquad (2)$$

This limitation is applied for both normal people and people with heart failure. Hence, a new point, labelled as S_1 point, will be the left limitation of the B point with QS being the PEP(min). The detection of B point is presented as followed: (1) Determining S point, (2) checking in the 2nd derivative whether two zero-crossing events (+,-,+,-) occur within C-S or not, (3) if yes, the second zero-crossing point will be chosen as the right limitation, labeled as S_2, and B point will be the first valley after S_2 in the 3rd derivative, and (4) if not, C peak will be the right limitation and B point will be the first zero-crossing before C peak in the 1st derivative.

The same procedure is applied in X point detection, with C peak being the right limitation and the conventional X point, labeled as X_0, being the left one. However, unlike Car. algorithm, X_0 is determined as followed: (1) using a 5Hz low-pass filter for ICG signal to smooth the signal, and (2) X_0 will always be the third zero-crossing after C peak in the 3rd derivative. Hence, X point will be the first valley before X_0 in the 3rd derivative (two zero-crossing events exist in the 2nd derivative) or the first zero-crossing after C peak in the 1st derivative (in another case).

3. EXPERIMENTS AND RESULTS

3.1. Experimental Apparatus

10 healthy volunteers aging from 18 to 22 were measured using our system (N. M. Duc et.al., 2015) . Those 10 corresponding 30-second-ICG signals were received at lying posture and non-breathing condition, so that they were relatively stable and free of baseline noise. In this experiment, they were labeled as “clean” ICG. With “clean” ICG, there were not any differences in detection results between Car. algorithm and the proposed algorithm. Hence, those “clean” ICG were added with a simulated respiration, which was a sine wave with frequency ranging from 0.1Hz to 0.5Hz (corresponding with respiratory frequency) and amplitude ranging from 10% to 100% of dZ/dt_{max} (average value of all B-C differences in amplitude found in “clean” ICG), and labeled as “noise” ICG. Then, both Car. algorithm and the proposed algorithm were applied in all cases of “noise” ICG signals. Finally, the detection results were compared to the found positions of B and X in “clean” ICG signals, which were considered as the standard. The evaluation was expressed through a proposed index, called the Error Detection Ratio (EDR):

$$EDR = \frac{\text{Number of wrong detected points}}{\text{Number of all points}} \times 100\% \quad (3)$$

There are EDR(B) for assessing B points detections and EDR(X) for assessing X point detections.

Besides that, Pre-ejection Period (PEP) (Q-B interval, Q of ECG and B of ICG) and Left Ventricular Ejection Time (LVET) (B-X interval) were also calculated based on found B and X by both algorithms on the worst “noise” ICG signals (amplitude of 100% of dZ/dt_{max} and frequency of 0.5Hz) and compared to those found on “clean” ICG signals.

3.2. Results and discussion

Figure 4 and 5 exhibited the ERD(B) and ERD(X) respectively. In B detections, a significant error recorded using Car. algorithm in signals with respiration (Figure 4a), with ERD(B) > 10% even in the lowest case of simulated sine (amplitude of 10% of dZ/dt_{max} and frequency of 0.1H). This error ratio increased approximately linearly when the sine’s amplitude increased and reached around 75% at the maximum amplitude of 100% of dZ/dt_{max} in all frequencies. In contrast, ERD(B) of the proposed algorithm (Figure 4b) showed a much more positive results with a completely precise detection with the added sine’s amplitude ≤ 30% in all cases of frequencies. Even at the worst condition of simulated respiration (amplitude of 100% and frequency of 0.5Hz), EDR(B) of the proposed algorithm was only more than 5%.

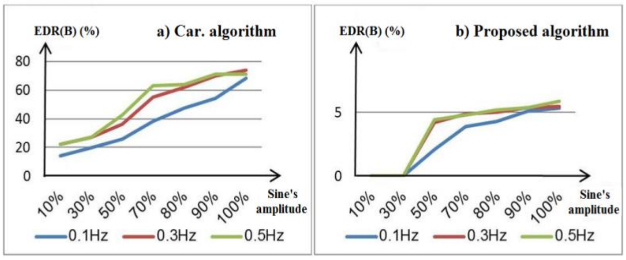

Figure 4. EDR(B) results of Car. algorithm (a) and the proposed algorithm (b) when adding a sine wave with three different frequencies: 0.1Hz, 0.3Hz and 0.5Hz and amplitude ranging from 10% to 100% of dZ/dt_{max} into “clean” ICG

A similar scene was also recorded in X point detections. Although EDR(X) results of the Car. algorithm (Figure 5a) were much lower than EDR(B)s, accounting for around 10% and 50% with the frequency of 0.1Hz and 0.5Hz respectively at the maximum amplitude of the added sine, the error in detections was still too high that could lead to significant errors in calculating Hemodynamics. Meanwhile, the proposed algorithm produced an extremely low EDR(X)s (Figure 5b) with about 1.2% was the highest EDR(X) recorded.

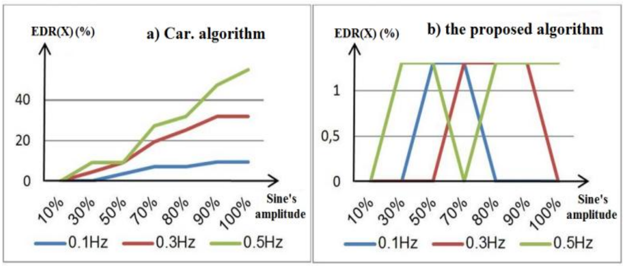

Figure 5. EDR(X) results of Car. algorithm (a) and the proposed algorithm (b) when adding a sine wave with three different frequencies: 0.1Hz, 0.3Hz and 0.5Hz and amplitude ranging from 10% to 100% of dZ/dt_{max} into “clean” ICG

Through EDR(B) and EDR(X) results, it is clearly seen that proposed algorithm can work very efficiently and accurately even when a fast and heavy respiration occurs. This characteristic is very important, since respiration always takes part in all physiological signals during monitoring in reality.

Figure 6 exhibited the calculated PEP and LVET in each beat, which was also converted to instant Heart Rate, of all 10 healthy volunteers. Those PEPs and LVETs were retrieved using detected B and X points found in the worst case of simulated respiration. There is also a standard line which is a five-order polynomial constructed based on PEPs and LVETs calculated in “clean” ICG. Again, while PEP and LVET produced by Car. algorithm had no related to the standard line, those in the proposed algorithm tightly followed the standard line in all ranges of Heart Rate. It continuously proves that the proposed algorithm had successfully minimized the error in calculating PEP and LVET, 2 important Hemodynamics and also taking part in calculating many other Hemodynamics, when respiration is taken into account.

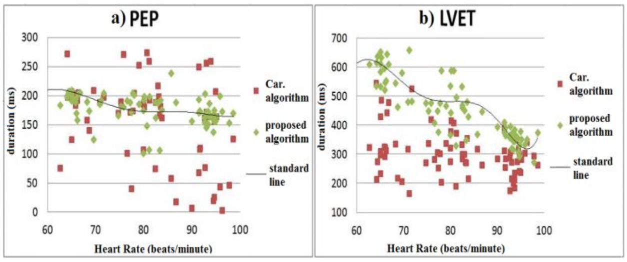

Figure 6. PEPs (a) and LVETs (b) calculated in each beat of all 10 healthy volunteers, based on B and X points found in the worst case of the added sine (amplitude of 100% and frequency of 0.5Hz**)**

CONCLUSION

Detecting accurately characteristic points on the ICG signal, such as B and X, is vital since they all take part in calculating some important Hemodynamics like Cardiac Output (CO), SV, LVET or PEP. Recently, the positions of B and X points have been re-defined by Carvalho. et.al., who also proposed an algorithm to find those positions, called Cal. algorithm. However, Cal. algorithm only works well in ICG signals without wandering baseline noises which is created mainly by respiration. This paper has successfully proposed an algorithm that takes full advantages of Cal. Algorithm as well as makes suitable changes to work stably even when respiration is taken into account. In experiment in ICG signals with different levels of simulated respiration, the proposed algorithm has been proved its effectiveness even at the worst case of respiration, with only around 5% of wrong points detected. Therefore, this algorithm is more realistic for both research and commercial use when respiration is always an unavoidable signal during diagnosis, monitoring and treatment.

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