Automatic Detection of Chest Compression Pauses Using the Transthoracic Impedance Signal

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Abstract

The recommended treatment for out-of-hospital cardiac arrest (OHCA) is immediate cardiopulmonary resuscitation (CPR) and early electrical defibrillation. During CPR, chest compressions and ventilations should be provided with a compression-ventilation ratio of 30:2. Chest compressions and ventilations induce fast and slow fluctuations, respectively, on the transthoracic impedance (TTI).

In this work we present a method for the automatic detection of pauses in chest compression using the TTI. The fluctuations induced by chest compressions were first isolated and emphasized and then, using an adaptive threshold, the intervals without chest compressions were identified. The method was adjusted and evaluated using a dataset of 3596 pauses corresponding to OHCA episodes. The mean duration of the pauses was 7.0 ± 6.2 *s. For the test set, the sensitivity and the positive predictive value were 93.9% and 96.2%, respectively.*

This method could be used for both online and offline CPR quality evaluation or to detect artifact free ECG intervals in which a rhythm assesment could be launched.

1. Introduction

During out-of-hospital cardiac arrest (OHCA) two interventions are key for a successful outcome of the patient: early cardiopulmonary resuscitation (CPR) and early defibrillation, usually delivered by an automated external defibrillator (AED). Current resuscitation guidelines [1] emphasize the importance of early, high quality chest compressions (CC) with a compression-ventilation ratio of 30:2. However, CC during CPR induce an artifact on the ECG signal which compromises the AED diagnosis.

The automatic detection of pauses in CC is relevant for two main reasons. First, in the field of quality CPR, it enables the AED to provide online feedback to the rescuer in order to avoid unnecessary interruptions of the CC. Second, it allows the detection of CPR artifact free ECG intervals (for instance, pauses for ventilation or rescuer switch) in which the AED could assess the rhythm [2].

Current commercial AEDs have the capacity to record the transthoracic impedance signal (TTI). The TTI acquisition process consists on injecting a high frequency current between the defibrillation pads and measuring the resulting voltage. This signal is used to verify that the pads are correctly attached to the patient's chest and also to adjust the defibrillation voltage prior to delivering a shock. During CPR, CC induce low amplitude fluctuations on the base TTI of the chest, which is in adults 70-80 ohms in average [1, 3].

In this study we present a method for the automatic detection of chest compression pauses during CPR episodes using the TTI signal. The performance of the method was evaluated with a dataset extracted from OHCA episodes.

2. Materials and methods

2.1. Database description and annotation

The dataset used in this study is a subset of a larger database recorded during a prospective study of OHCA episodes between 2002 and 2004 [4, 5]. The surface ECG and several reference channels were acquired using Heartstart 4000 defibrillators (Philips Medical Systems, Andover, MA, USA) equipped with an additional chest pad. During these episodes, CPR was delivered with a compression-ventilation ratio of 15:2 and 5:1. Intervals with and without CC were annotated based on the compression depth (CD) signal. The CD was calculated from the acceleration and the force recorded by the chest pad.

For this study, records containing pauses in chest compressions, such as for rhythm assessment or ventilations, were extracted and resampled at 250 Hz. Each record comprised three channels: the TTI, the CD and the surface ECG. Additionally, each record had annotations indicating pauses in CC. Chest compression intervals with only 2 to 4 compressions were discarded. We extracted a total of 600 records containing 3596 pauses with a mean duration of 7.0 ± 6.2 s. All the records and annotations were manually reviewed. The CD and the surface ECG were used to correct the CC intervals annotations, if necessary.

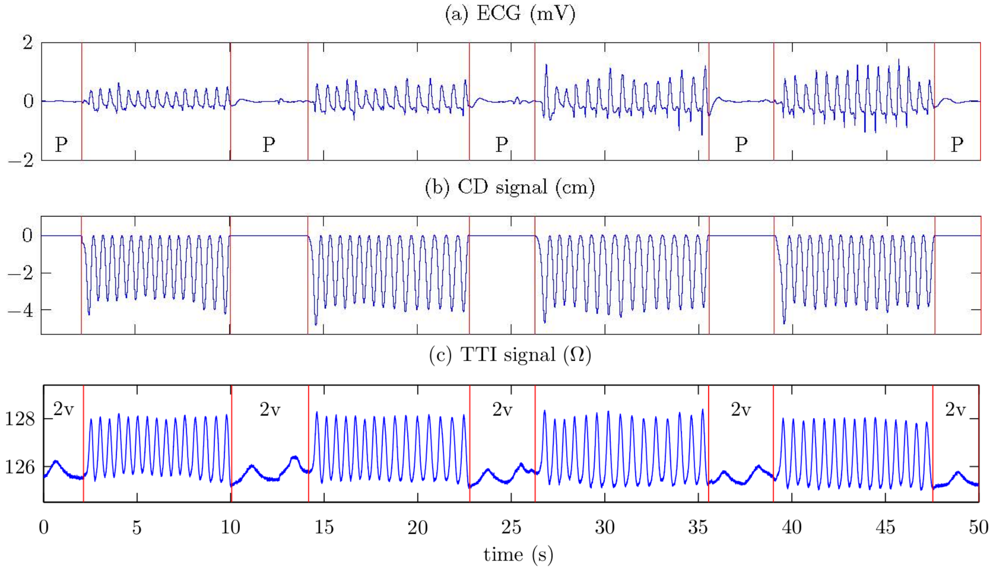

Figure 1. Example of a record for a patient in asystole. (a) Chest compressions (CC) induce an artifact on the ECG. During CC pauses, denoted in the figure with the letter “P”, the underlying rhythm (asystole) is revealed. CC intervals were annotated using the CD (b) and the ECG, and are depicted with vertical red lines. (c) Ventilations are observed as slow fluctuations on the TTI. The number of ventilations in each pause is depicted in (c).

When a single chest compression was present during a pause interval, it was reannotated as part of the pause. The TTI was used to annotate the number of ventilations in each pause. The records were then split in two halves: a training set to optimize the parameters of the pause detection method and a test set to evaluate its performance.

Fig. 1 shows an example record corresponding to a patient in asystole. The record consists of the ECG (panel a), the CD (b) and the TTI (c). During CC intervals the artifact corrupts the ECG, while during the pauses the underlying rhythm (asystole) is revealed. The fast and slow fluctuations induced by the CC and ventilations, respectively, can be observed on the TTI. In the intervals without CC, the slow fluctuations were used to count the number of ventilations.

2.2. Chest compression pauses detection

2.2.1. Method description

Chest compressions and ventilations produce fast and slow fluctuations respectively in the TTI. The proposed method consists on emphasizing the fluctuations induced by CC and then, using an adaptive threshold, discriminate the intervals with and without CC. A minimum duration of 1.5 s was established for the CC intervals. Fig. 2 shows the signals at different stages of the detection method, which is described below.

First, the TTI (panel a) is band-pass filtered to isolate the CC fluctuations using a 3rd-order Chebychev low-pass filter of $f_c = 2.2$ Hz, and a high-pass filter of $f_c = 1.7$ Hz. From the filtered signal, $TTI[n]$ (b), the slope is calculated as the first difference, scaled and squared, as shown in the following equation:

$$sTTI[n] = \left(\frac{TTI[n] - TTI[n-1]}{0.024}\right)^2 \qquad (1)$$

Then, the $sTTI[n]$ (c) is smoothed using a 1st-order low-pass filter of $f_c = 0.6$ Hz. The resulting signal, $pTTI[n]$, presents high amplitude during CC intervals and low amplitude during pauses (d). The onset and offset of the CC intervals are determined by comparing the pTTI signal with an adaptive threshold and applying a correction factor to compensate the delays. The threshold has an initial value of Th_0 and varies between Th_{min} and Th_{max}. The applied corrections are t_{on} for the onset and t_{off} for the offset. Threshold parameters (Th_0, Th_{min} and Th_{max}) and the delay corrections (t_{on} and t_{off}) were optimized to maximize the method performance with the training set. The values obtained after the optimization process were $Th_0 = 0.035$, $Th_{\text{min}} = 0.002$, $Th_{\text{max}} = 0.3$, $t_{\text{on}} = 0.4$ s and $t_{\text{off}} = 1$ s. The threshold was dynamically updated each time a chest compression interval was detected as 0.09 times the average of the pTTI in that interval.

2.2.2. Performance evaluation

The performance of the method was evaluated using three parameters: the sensitivity (Se), the positive pre-

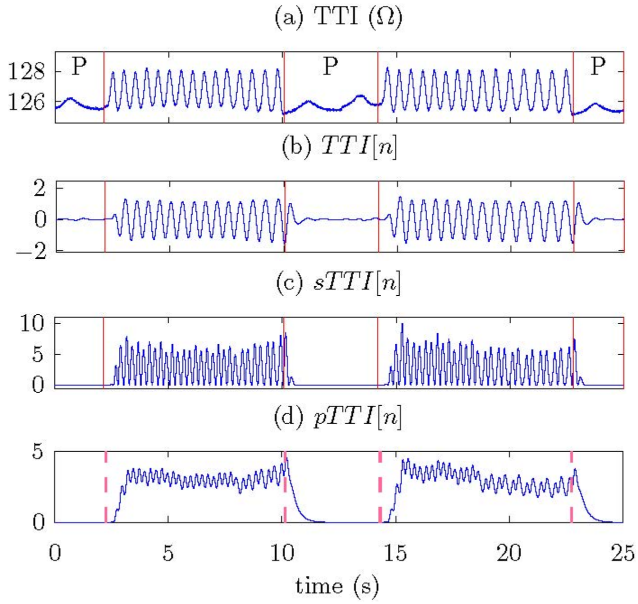

Figure 2. Signals involved in the pauses detection method. The annotated pauses are depicted with vertical solid lines (panels a, b and c). The detected pauses are depicted with vertical broken lines in panel d.

dictive value (PPV) and the mean difference between the duration of the annotated and of the detected pauses. Se is the proportion of annotated pauses which are correctly detected, and PPV is the proportion of detected pauses which are correct. The maximum admissible tolerance for the onset and the offset of the pauses was 0.75 s.

3. Results

3.1. Training and test results

For the training set, the Se and the PPV were 94.3% and 96.5%, respectively. The mean difference between the actual and the detected duration of the pauses was 0.34 ± 0.77 s.

For the test database, the Se and the PPV were 93.9% and 96.2%, respectively. The difference between the durations of the pauses was 0.24 ± 0.93 s.

The 1966 pauses containing 2 ventilations were separately analyzed. The mean duration of these pauses was 5.2 ± 1.8 s and more than 95% of them had a duration above 3.0 s. For these pauses, we obtained a Se=93.6%, a PPV=94.5% and a mean difference of 0.23 ± 0.73 s.

3.2. Graphical examples

This section provides some examples of the operation of the method. Fig. 2 shows a correctly detected pause. In Fig. 3 the fluctuations induced by the CC and by the ventilation are similar, and cannot be distinguished in the pTTI.

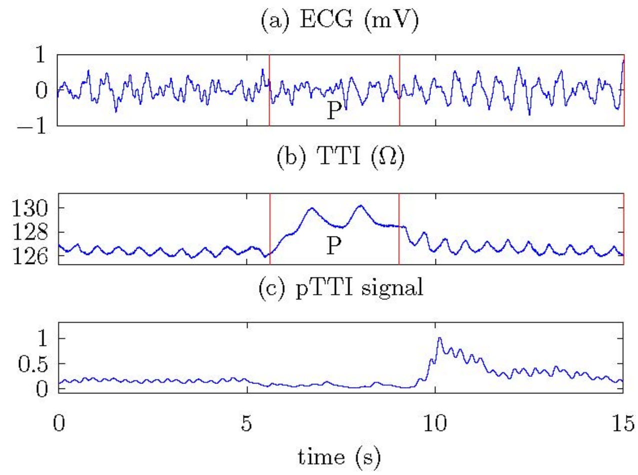

Figure 3. Undetected pause.

In Fig. 4 the detected pause is longer than annotated. The amplitude of the fluctuations induced by the last chest compressions in the first CC interval is very low, and the method identifies them as part of the pause.

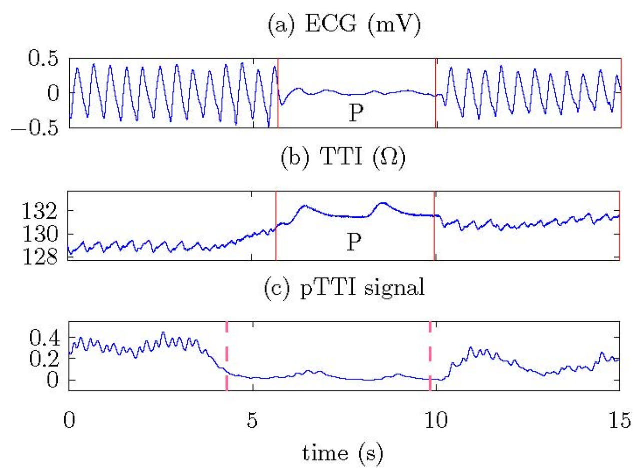

Figure 4. Pause detected longer than it actually is.

Fig. 5 illustrates the opposite case: a pause detected shorter than it is. In the beginning of the pause the amplitude of the fluctuations in the TTI is high and this part is identified as a CC interval.

4. Discussion and conclusions

A method to automatically detect pauses in chest compressions by analyzing the TTI signal has been developed. The proposed method was adjusted and evaluated using data extracted from OHCA episodes. The dataset consisted of 3596 pauses predominantly caused by ventilations,

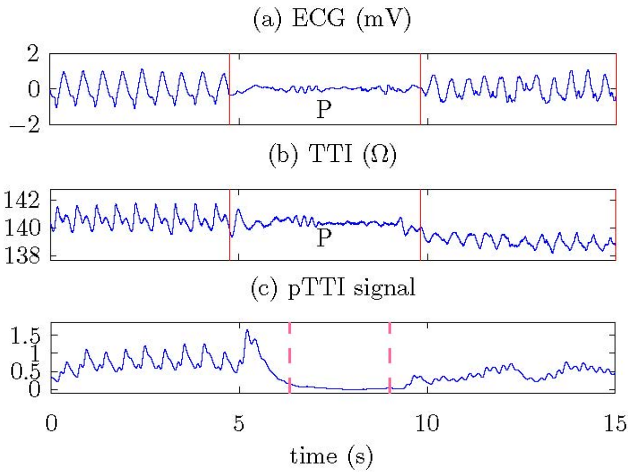

Figure 5. Pause detected shorter than it actually is.

although pauses for other reasons as rhythm analysis, pulse check and rescuer switch were also present.

One of the applications of the presented method is to identify CPR artifact-free ECG intervals in which the AED could launch a rhythm assessment. Using the complete dataset, the time required to provide 2 ventilations was evaluated. A value of 5.2 ± 1.8 s was obtained, consistent with that previously reported by Ødegaard et al. [6]. This time should be long enough to perform a rhythm analysis using a fast shock advice algorithm, such as that proposed by Irusta et al. [7].

The parameters of the method were adjusted to minimize the number of false positives, that is, the CC intervals detected as pauses. A false positive would trigger a rhythm analysis during a corrupt interval, which could result in an erroneous AED diagnosis. After that adjustment, the Se and the PPV obtained for the test set were 93.9% and 96.2%, respectively.

Section 3.2 presents some graphical examples of the operation of the method. In the examples shown in Fig. 3 (a undetected pause) and in Fig. 5 (a pause detected shorter than it actually is), the only possible implication of the failure of the method would be a delay in the rhythm assessment. In Fig. 4, however, the premature detection of the onset of the pause would cause a rhythm analysis during a corrupted interval, which could lead to a wrong diagnosis.

The second application of the proposed method is related to CPR quality. Current guidelines recommend a compression-ventilation ratio of 30:2 and a CC rate of at least 100 compressions per minute. The frequency and the duration of the pauses provide information about the fulfillment of the guidelines, and could be used to give feedback to the rescuer.

The proposed method uses a direct algorithm with low computational requirements and consequently, it could be easily implemented in commercial AEDs.

Acknowledgements

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