Cost-efficient Accurate Monitoring of Respiration Rate using ECG

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Abstract

Different techniques are proposed to indirectly measure respiration rate (RR) from ECG. Such techniques are useful when ECG, but not respiration, is monitored. Examples include wearable patches that record one channel of ECG to monitor heart rate and rhythm, but have no way of monitoring respiration rate other than maybe through impedance. Measuring impedance using the ECG electrodes, however, adds to the hardware cost, increases the battery usage, and is prone to motion artifact. This study compares the performance of monitoring RR using ECG beat-to-beat variations in either peak time intervals or QRS amplitude in one or two ECG leads.

We show that RR estimation may be more accurate by analyzing changes in QRS amplitude, even in only one lead, rather than the changes in heart rate.

1. Introduction

Monitoring respiration rate (RR) as a critical vital sign can help detecting both subtle and acute changes in patient condition. Respiration monitoring plays an important role in patient care both inside and outside hospital.

There exist several respiration acquisition techniques, but the commonly used method in medical devices such as patient monitors is measuring the respiratory impedance. Such method uses a specialized circuitry to detect changes in a low-voltage high-frequency oscillating signal sourced and measured through ECG electrodes without using the ECG signal itself. However, measuring impedance using this specialized circuitry adds to the hardware cost, increases the battery usage, and is prone to motion artifact. These may be limiting factors for wearable monitors.

Since ECG is routinely monitored in many settings, researchers have also pursued methods for extracting respiration signal, or just the rate, directly from the acquired ECG waveform. A good review of such ECG-derived respiration (EDR) techniques can be found in a recently published paper by Helfenbein et al [1].

2. Methods

In this work, we evaluate and compare three of the methods described in [1], as summarized below.

1. Measure the peak to trough amplitude in each normal QRS complex in a single ECG lead I or II (EDR_I and EDR_II)
2. Use arctangent of the ratio of the measurements done in method 1 for two ECG leads to get the angle of the mean heart axis with respect to one of the lead axes (EDR_I&II)
3. Analyze beat-to-beat variations in peak time intervals, primarily due to respiratory sinus arrhythmia (EDR_RSA)

A hybrid time-frequency domain method was developed to process the EDR waveforms to calculate one respiration rate value for each 30-second window sliding at 5-second steps. In this method, first peaks and troughs of the EDR waveform are detected in time domain in order to estimate the RR. Then the final RR is selected as the maximum frequency of the spectrum within a search windows based on the time-domain RR estimation. Figure 1 shows an example.

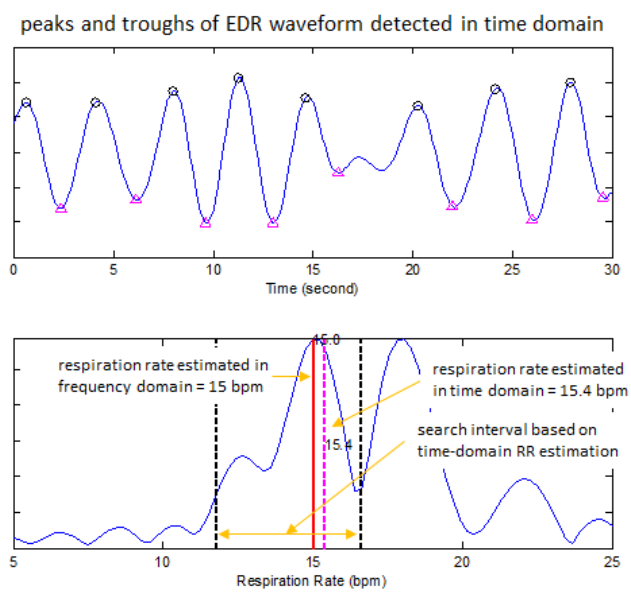

Figure 1. Top plots shows 30 seconds of EDR waveform on which peaks and troughs are detected in time domain. Bottom

plots shows the spectrum of EDR waveform, search window based on time-domain RR estimation, and the final RR selected as the maximum of spectrum within the search window.

2.1. Evaluation database

All three EDR techniques were already developed using databases which had no overlap with the database used in this study [1]. In the current work we only evaluate these techniques with no modification of their original design.

For comparing the three techniques, the publicly available PhysioNet/cebsdb database was used that contains two annotated channels of ECG and one channel of respiratory signal from a thoracic piezoresistive band [2,3]. The 16 presumed healthy volunteers with clean recordings (age: 24±3 years; gender: 9 males, 7 females) were in supine position listening to classical music for approximately 50 minutes.

For the ECG measurement monitoring electrodes with foam tape and sticky gel were used. Each ECG channel was recorded at 5000 samples per second at the bandwidth of [0.05-150] Hz.

The respiratory signal from the thoracic piezoresistive band (RESP) was used as the gold standard.

3. Results

The reference RR from RESP was compared to that from the other four EDR waveforms: EDR_I, EDR_II, EDR_I&II, and EDR_RSA. Comparing a total of 9474 sets of RR values, Table 1 summarizes the results.

Table 1. Table of similarity measures between the RR coming from reference RESP waveform and each of the other four EDR waveforms (n=9474)

RR from RESP and that from …	EDR_I	EDR_II	EDR_I&II	EDR_RSA
mean±std difference (bpm)	0.14 ± 0.92	0.08 ± 1.18	0.13 ± 0.94	0.96 ± 2.50
correlation coefficient	0.96	0.93	0.96	0.66
coefficient of variation CV (%)	5.1	6.5	5.2	14
r^2 measure of goodness of fit	0.9195	0.8666	0.9157	0.4702
sum of squared difference SSE	0.91	1.1	0.94	2.3

The scatter and Bland-Altman plots for the RR from each of the four waveforms versus that coming from RESP waveform is shown in Figures 2-5.

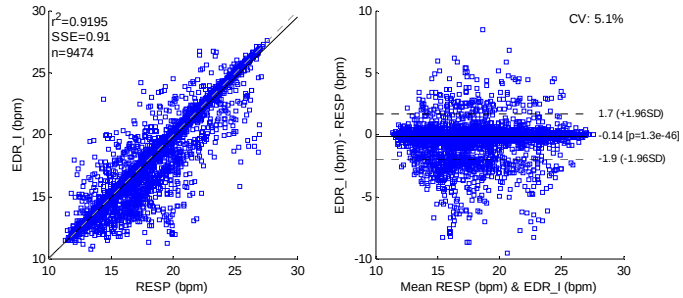

Figure 2. Scatter and Bland-Altman plots for the RR from RESP versus that from EDR_I

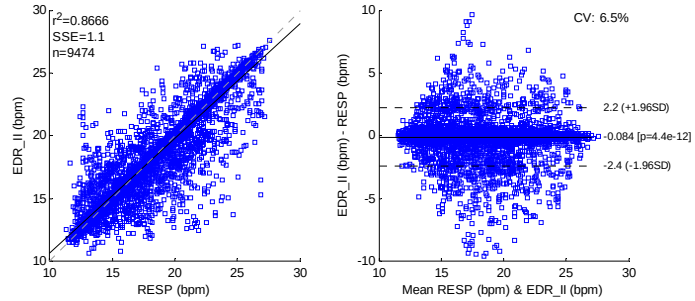

Figure 3. Scatter and Bland-Altman plots for the RR from RESP versus that from EDR_II

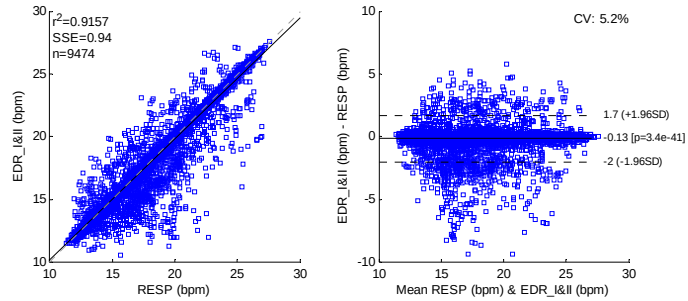

Figure 4. Scatter and Bland-Altman plots for the RR from RESP versus that from EDR_I&II

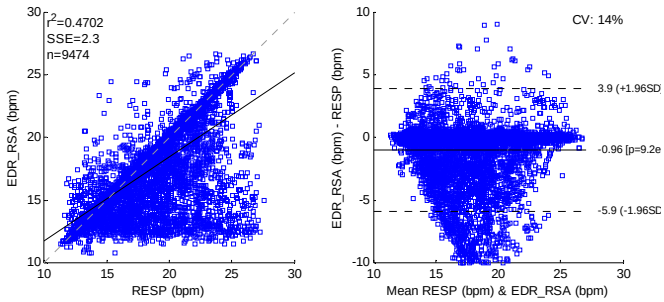

Figure 5. Scatter and Bland-Altman plots for the RR from RESP versus that from EDR_RSA

As seen, in this small database of young and presumed healthy individuals, deriving the respiration rate from QRS amplitude is more accurate than using RSA. For cardiac patients with less pronounced RSA, this difference in accuracy may be even more significant. Lead I seems to be a better choice for EDR than lead II, and adding the second lead may not improve the performance.

3.2. Discussion

We reported the performance of three different EDR methods using one or two regular ECG leads, I and II, on a small database containing young and presumed healthy individuals quietly resting in supine position without much movement. In practice, several factors should be considered to choose and adapt a technique for a particular application and a specific patient population. Below is a list of some such factors:

a) Some wearable devices may not have two ECG leads, so using EDR_I&II would not be possible.
b) A small wearable patch may have two electrodes very close to each other, hence recording a short-vector ECG as opposed to a regular long-vector ECG lead. A short-vector recording will probably not have a negative impact on RSA method. However, more research is needed to evaluate the accuracy of QRS amplitude techniques using a short-vector ECG. This is because the change in QRS amplitude in a short-vector ECG may not be as pronounced as that in a regular long-vector lead.
c) The change in QRS amplitude due to respiration depends on the location of the electrodes on body. Additional research is needed to find the optimal electrode placement for using this method.
d) The RSA technique only needs the location of the beats. Therefore, it may potentially be able to use the beat locations coming from techniques other than ECG, e.g. photoplethysmography. Additional research is needed to evaluate this hypothesis.
e) The underlying premise of RSA method breaks down when respiration-induced change in heart rate has naturally decreased with age or illness, when there are supra-ventricular or ventricular arrhythmias, or when patients are on rhythm- and rate-control medications or devices.
f) In both QRS and RSA methods, respiratory wave samples are available only at normal QRS times, and thus the derived respiratory waveform may be under-sampled if respiration rate is high or hear rate is low. Frequent ectopic beats will also further reduce the respiration signal sampling.

In [1] we reviewed another EDR technique based on extraction and processing of the electromyogram (EMG) signal from the ECG which is based on the premise that ECG recordings contain muscle tremor “noise” from electrical activation of the inter-costal chest muscles and the diaphragm during the respiratory cycle. The respiratory EMG is best found in the frequency band above 250 Hz. Therefore, we were not able to evaluate the performance of that method on the current database which has used a 150 Hz lowpass filter. One advantage of the EMG method is that it has a similar respiratory phase with both impedance and tidal volume. It also reflects respiratory “effort”, a measurement not available with other methods, and thus has potential for use in applications like ventilator triggering [1].

4. Conclusion

Although it may be possible to improve the overall performance of measuring RR from ECG by combining different methods, it is unlikely that ECG-derived respiration techniques will ever be as accurate as using more direct methods such as using an airway airflow sensor. However, the high accuracy of respiration rate measured from a single lead of ECG may be good enough for many practical applications, especially when ECG is monitored for other reasons and the additional cost of specialized hardware solely for respiratory monitoring is of a concern. The simplicity of these algorithms, in addition to their good performance, is an advantage for battery-operated wearable monitors that require algorithms with low computational cost.

References

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