Measurement of Heart Rate and Respiratory Rate Using a Textile-Based Wearable Device in Heart Failure Patients

F Chiarugi[1], I Karatzanis[1], G Zacharioudakis[1], P Meriggi[2], F Rizzo[2], M Stratakis[3], S Louloudakis[3], C Biniaris[4], M Valentini[5,6], M Di Rienzo[2], G Parati[5,6]

[1]Foundation for Research and Technology - Hellas, Institute of Computer Science, Heraklion, Crete, Greece
[2]Don Carlo Gnocchi Foundation, Biomedical Technology Department, Milan, Italy
[3]FORTHnet SA, Heraklion, Crete, Greece
[4]Virtual Medical World Solutions Ltd, Romsey, Hampshire, United Kingdom
[5]Istituto Auxologico Italiano, San Luca Hospital, Department of Cardiology, Milan, Italy
[6]University of Milano Bicocca, Department of Clinical Medicine, Cardiology Unit, Milan, Italy

Abstract

Changes in heart rate (HR) and respiratory rate (RespR) may be used as markers of early decompensation in chronic heart failure (CHF) patients monitored at home. Aiming at improving quality of care and at reducing hospitalization rate and health care costs in CHF, progress in technology has led to the development of small portable and even wearable devices for the acquisition and transmission of relevant vital signs to a remote monitoring centre. This paper describes a signal acquisition and processing system, based on a wearable textile-based device with sensors for the measurement of one-lead ECG and chest movement, and focuses on the algorithms for HR and RespR evaluation. An electronic board collects and transmits these signals to a PDA, which sends them via Wi-Fi to a home gateway where the HR and the RespR time series are produced. The home gateway packs the data with other vital signs collected by using different devices and sends them in XML format to a central repository where a clinical decision support system can use them for the detection of early decompensation episodes. The system has successfully overcome a preliminary test phase and is ready for more extensive tests in a real clinical environment.

1. Introduction

During the last years an increasing demand for smart non-invasive devices capable to detect vital signs while the subject is free to continue with his/her everyday habits (at work, at home or in a clinical environment) has emerged. This demand has been generated by the need of the health care institutions to be relieved of the long-term patient stays in situ by providing telemonitoring services for the patients. One of these smart devices is the MagIC system (Maglietta Interattiva Computerizzata, Computerized Interactive T-shirt), which can offer a substantial support in improving quality of care and reducing hospitalization time and health care costs in chronic heart failure (CHF). MagIC is a prototype wearable device that allows the unobtrusive monitoring of some vital signals and hence tracking changes in heart rate (HR) and respiratory rate (RespR) which may be used as markers of early decompensation in home-monitored CHF patients.

In this paper, a service based on the MagIC system, with which a health institute monitors patients at home, is described with main focus on the algorithms for the evaluation of HR and RespR. The cardio-respiratory signals acquired using the MagIC system are recorded and wireless transmitted to a PDA and then, via Wi-Fi, to a home gateway. In the home gateway the HR and RespR time series are extracted by pre-filtering and processing the acquired signals. Finally the resulting data are packed in XML format at the home gateway and transmitted to a central repository where they can be used by the clinical decision support system (CDSS) for the detection of early decompensation episodes.

2. Methods

2.1. The MagIC system

The MagIC system [1] (see figure 1) is composed of a vest including textile sensors and a portable electronic board. At the thorax level the vest includes two woven electrodes made by conductive fibres so to obtain an ECG lead. The contact between textile electrodes and thorax is guaranteed by the elastic properties of the garment. The vest includes a textile-based transducer, obtained by a patented processing of the conductive fibres, that provides a chest movement signal that can be used for the assessment of the respiratory frequency.

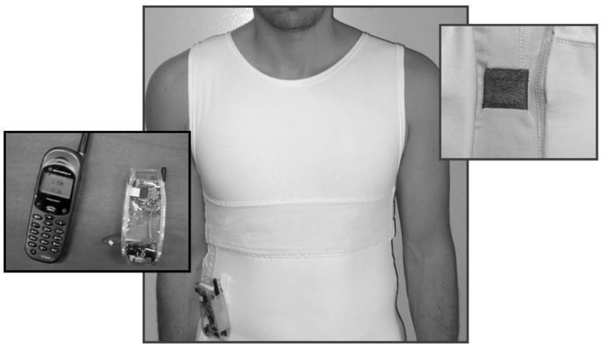

Figure 1. The MagIC system. As it can be seen the MagIC vest is very easy to be worn.

Through connections still obtained by using the same conductive fibres, ECG and chest movement signals feed a portable electronic module - having the typical size and weight of a small cell phone - which is placed on the vest through a velcro strip. The electronic board detects also the subject's movement through a 3-axis accelerometer and transmits all signals via a wireless connection to a PDA for data visualization, storage and a reply transmission to a remote computer. In case of interest, other external devices (e.g. Pulseoximeters) might be connected to the board as well.

2.2. MagIC system and home gateway communication

In our application, ECG and chest movement signals from the MagIC vest are transmitted via Wi-Fi in XML format to the home gateway. The communication between the monitoring station of the MagIC system and the home gateway is socket based. A server process runs on the home gateway and has a socket that is bound to a specific port. The server waits and listens to the socket for the monitoring station to make a connection request. When a connection is established the server process stores the received data locally as an XML file using the timestamp as filename.

2.3. Computation of the heart rate time series

The MagIC system acquires one-lead ECG signal at 200 Hz sampling frequency. The HR time series is obtained through a time domain processing using a QRS detector with an adaptive threshold applied to a QRS enhanced signal. The QRS enhanced signal is obtained prefiltering the ECG lead in order to select the band with the typical main frequencies of the QRS complexes. The applied algorithm is a down-scale, to the case of one-lead ECG, of the same algorithm used for the two-lead ECG whose results on the MIT-BIH Arrhythmia Database were previously reported in [2].

2.4. Computation of the respiratory rate time series

The number of breaths per minute is defined by counting how many times the patient's chest rises in a minute in normal conditions [3]. The chest movement is expressed in terms of the thorax circumference that is max at the end of the inspiration phase and min at the end of the expiration phase. The MagIC system samples the chest movement with a 50 Hz sampling frequency. The RespR time series is obtained with a frequency domain processing in order to minimize the effect of signal artefacts that are usually more evident in the chest movement signal than in the ECG lead.

2.4.1. Noise removal

In order to clean the chest movement signal from possible movement artefacts, a series of filters is applied to it before the processing for the calculation of the respiratory rate time series.

First of all, the linear trend from the sampled time series of the chest movement is removed. Then, the signal is filtered with a median filter, a non-linear filter that uses a small 3-point sliding window to smooth the data removing possible artefacts. Finally, a window-based finite impulse response filter is applied. It consists in a low-pass filter based on a Hann window, which removes the high frequency noise that may be present (see figure 2). In order to avoid the "edge effect" - that usually appear with the use of moving window filters - the sampled time series has been expanded by mirroring the values half a window size on the left at the beginning of the recording and half a window size on the right at the end of the recording.

2.4.2. Calculation of the respiratory rate in the frequency domain

After the prefiltering, the respiratory rate is calculated in the frequency domain using a moving window of 60 sec. The 60 sec time width has been selected because it perfectly matches the definition of respiratory rate as the amount of breaths in a minute [3], offers a satisfactory frequency resolution and accuracy, and remains sufficiently sensitive to time changes of the respiratory rate while rejecting short-in-time artefacts. The window advances with a step of one second.

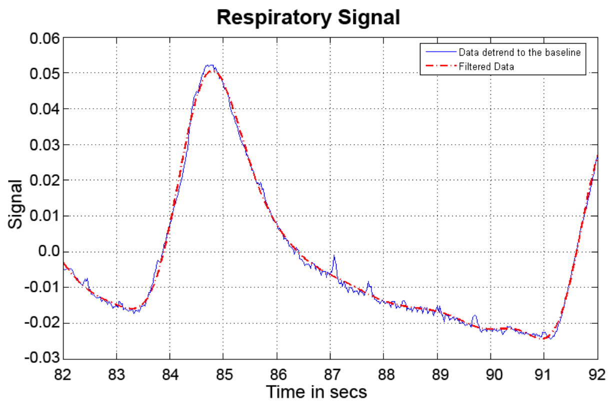

Figure 2. The detrended time series of the original respiratory data received from the MagIC vest (blue line) compared to the filtered time series (red line).

In each window the power spectrum of the signal is computed using a Discrete Fast Fourier transform. A peak picking algorithm - based on derivative changes - locates all the peaks at the power spectrum and picks the best candidate for the dominant frequency in the current 60 sec window. After locating the frequencies for all the steps, a small-window averaging filter is applied in order to smooth the results against computational artefacts and the respiratory rate time series is finally obtained.

The final step of the processing consists in the creation of a new XML file containing the original data coming from the MagIC system plus the average Heart Rate, the heart rate time series, the average Respiratory Rate and the respiratory rate time series.

2.4.3. Algorithm validation

Special attention has been dedicated to the selection of the window size for the evaluation of the instantaneous respiratory rate. The computation of the respiratory rate of a constant signal is not affected by the used window size, but, in case the signal has fluctuations, the computed respiratory rate can depend on the selected window size.

In order to see how different window sizes affect the results, a series of artificial signals was created. Considering the "almost" sinusoidal forms of the acquired signals in ideal conditions, sinusoidal functions were used in the creation of the artificial signals used for testing purposes. The sinusoidal function used to create each artificial signal was $f(x) = \sin(2 \cdot \pi \cdot freq \cdot t)$ where $freq$ is a constant selected from 10/60 (10 breaths per minute) to 45/60 (45 breaths per minute) and incremented with a 1/60 step from a signal to another. Figure 3 shows the percent error between the actual and the measured values on the different signals using window sizes of 30 sec, 45 sec, 60 sec and 70 sec (other window sizes were also tested).

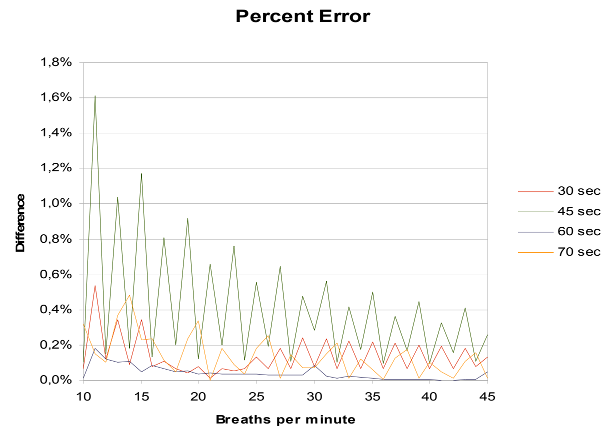

Figure 3. Percent error between actual and measured values with window sizes of 30, 45, 60 and 70 sec.

In relation to the evaluation of the RespR time series the following problems have been faced: a) limited amount of published papers on algorithms for the estimation of the RespR time series; b) lack of annotated chest movement signals with the corresponding RespR time series. These problems make very difficult the verification of the computed results especially in cases where significant noise or artefacts are embedded in the signal. In order to test the performances of the algorithm in presence of noise or artefacts, a set of artificial sinusoidal signals were generated. Such signals were produced without any noise but also with the addition of random noise with a max amplitude equal to 20% of the signal amplitude. For each artificial signal the differences between the real respiratory rate and the computed one on the "clean" and the "noisy" signals were evaluated as:

$$\%\,difference = \frac{|\text{real value} - \text{computed value}|}{\text{real value}} \cdot 100\%$$

The performances of the algorithm were very satisfactory and only very small differences were noticed; in the worst case the difference was still lower than 0.2%.

3. Results

In figures 4 and 5 the results of the algorithm on signals acquired from real persons are shown. In these figures the blue line is the received original signal, while the red one is the computed rate (time series).

In the home gateway the average Respiratory Rate and Heart Rate are collected by the Nurse@Home service from the processed XML file (see figure 6) along with the Minnesota questionnaire and other vital signs acquired using different devices (e.g. systolic and diastolic blood pressure, oxygen saturation, body weight). All the data are finally packed in XML format and transmitted to a central repository where they can be used by the CDSS for the detection of early decompensation episodes.

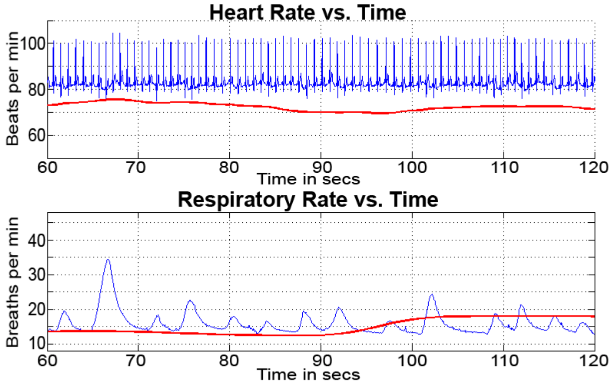

Figure 4. Heart rate and respiratory rate computed in an excerpt of data acquired by the MagIC system. This case has some variability in the chest movement signal.

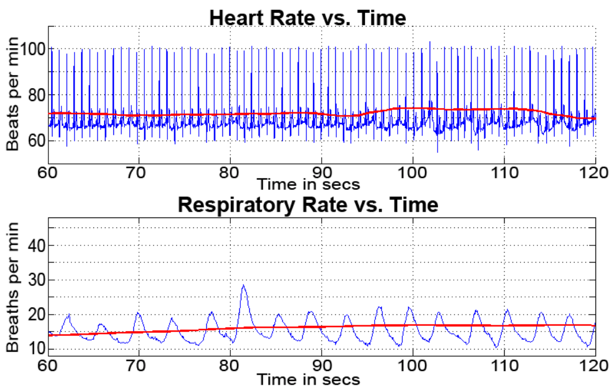

Figure 5. Heart rate and respiratory rate computed in an excerpt of data acquired by the MagIC system. This case has a more regular chest movement signal.

4. Discussion and conclusions

The performances of the QRS detection algorithm have been previously evaluated on an annotated database [2] and the good results obtained on a learning set composed by real acquisitions confirm its robustness.

As to the respiratory signal, an accurate assessment of the performances of the algorithm would require the availability of an annotated database (annotated series of respiration data). In absence of such annotated database, two different ways of acquiring the respiratory signals could be simultaneously used. In such case the chest movement signal (changes in the thorax circumference of the patient) would be available together with e.g. the air intake and these two respiratory signals from totally different sources acquired from the same patient could be more easily correlated for the evaluation of a more accurate respiratory rate and the assessment of the designed algorithms.

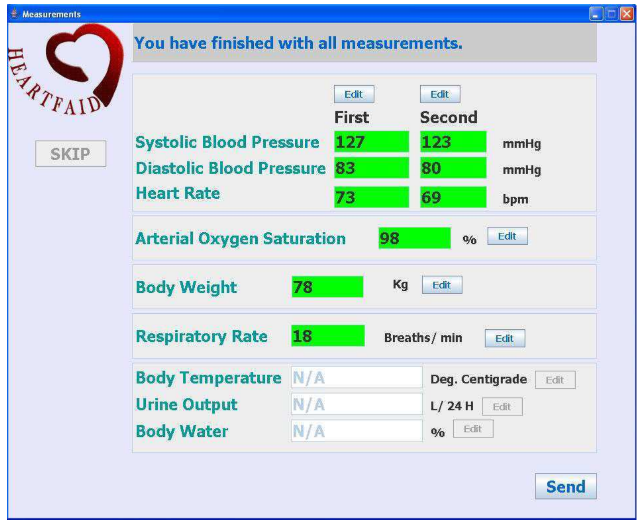

Figure 6. The Nurse@Home service collects all the required measurements for the patient at home.

Acknowledgements

This work was partially supported by the European Project “HEARTFAID: a knowledge based platform of services for supporting medical-clinical management of the heart failure within the elderly population” (IST-2005-027107), 2006-2009.

The authors would like to thank C. Assimakopoulou of FORTHnet S.A. and the entire HEARTFAID consortium for their valuable collaboration and help.

References

[1] Di Rienzo M, Rizzo F, Parati G, Ferratini M, Brambilla G, Castiglioni P. A Textile-Based Wearable System for Vital Sign Monitoring: Applicability in Cardiac Patients. Computers in Cardiology 2005;32: 699-701.

[2] Chiarugi F, Sakkalis V, Emmanouilidou D, Krontiris T, Varanini M, Tollis IG. Adaptive Threshold QRS Detector with Best Channel Selection Based on a Noise Rating System. Computers in Cardiology 2007;34: 157-160.

[3] Simoes EA, Roark R, Berman S, Esler LL, Murphy J. Respiratory rate: measurement of variability over time and accuracy at different counting periods. Arch Dis Child. 1991 October; 66(10): 1199-1203.

Address for correspondence

Franco Chiarugi
Foundation for Research and Technology - Hellas,
Institute of Computer Science,
Biomedical Informatics Laboratory,
P.O. Box 1385, N. Plastira 100, Vassilika Vouton
GR 70013, Heraklion, Crete, Greece

E-mail address: chiarugi@ics.forth.gr