Respiration Detection Implemented in Multichannel ECG Front End Module: A Preliminary Study

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Abstract – **A new generation single-chip low-power ECG analog front-end module (ADAS1000) is used to develop a system for 12-lead high-resolution ECG and impedance-based respiration acquisition. The developed respiration rate (RR) measurement algorithm is validated on a test respiration database, including more than 650 reference 15s-window sliding of 5s with annotated respiration periods and assigned to apnea, normal, deep breathing episodes acquired in leads I and II. The mean RR error is found within 0.7 to 4.7 bpm, significantly inversely related with the respiration amplitude.**

Keywords – **impedance, respiration, analog front end ECG**

I. Introduction

Continuous monitoring of respiratory activity is mandatory in clinical high-risk conditions, and appropriate monitoring equipment is life-saving [1]. The approaches for respiration measurement are categorized as applying direct or indirect methods. In direct methods, a sensor is coupled to the airway to quantify different properties of the air transported into and out of the lungs, e.g. the temperature changes in the air (nasal thermistors), the change in carbon dioxide in inhaled and exhaled air (carbon dioxide sensors), etc. Although the direct measurement methods tend to be more accurate, they could interfere with the normal respiration and could cause discomfort for the patient.

The most commonly employed indirect methods are impedance and inductance plethysmography [2]. In inductance plethysmography, sensors are placed around the chest and abdomen. They are excited by a low-current, high-frequency (300 kHz) electrical oscillator circuit. Movement of the chest and abdomen during respiration causes the sensors to generate magnetic fields, which are measured as voltage changes over time. The impedance plethysmography, applies low amplitude, high frequency (50 to 500 kHz) alternating current between two surface electrodes to record thoracic movements or volume changes during a respiratory cycle. The voltage drop between the electrodes is proportional to the impedance, which increases during inspiration and decreases during expiration. A common problem for the above two indirect respiration detection methods is that blood flow generates cardiac-synchronous changes in the acquired respiratory signal which could be a significant source of error [3].

The big advantage of the impedance-based respiration detection is the signal acquisition via standard body surface electrodes without additional sensors. This option for thoracic impedance measurement is recently embedded in a new generation of a single chip low power Analog Front End (AFE) modules [4,5] which simplify the task of acquiring and ensuring high quality ECG signals in monitor and diagnostic multichannel ECG applications, such as bedside patient monitoring, portable telemetry, holters, defibrillators, ambulatory monitors, patient transport, stress testing, etc. The need for fast and reliable impedance-based respiration detection module becomes a challenging task.

This paper aims to provide a module for respiration rate measurement which is implemented in a high-resolution multichannel ECG monitoring system. The validation on the test respiration database is also presented.

II. Hardware

This work presents a realization of a standard 12-lead ECG and respiration monitoring system using two cascaded AFE chips ADAS1000 (Analog Devices) [4], connected in master and slave mode according to the block diagram in Fig. 1. Ten electrodes, placed on the chest, are used for ECG signal acquisition. Two of them (lead I, II or III) are also employed for respiration signal measurement. All electrodes are directly connected to the AFE input via first-order low-pass filters to remove high-frequency noises and anti-aliasing effect before the 19-bit analog-to-digital conversion (ADC) in ADAS1000. ADAS1000 implements in one low power AFE chip, all of the features that are commonly required for a complete ECG module:

- High-resolution measurement of 5 ECG channels.
- Internal pace detection algorithm.
- Detection of bad electrode contact.
- Generation of Wilson Central Terminal.
- Driven right leg.
- Patient cable shield driving for suppression of the common mode interferences.

ADAS1000 integrates the option for measurement of one thoracic impedance channel by injecting high-frequency current with fixed amplitude between two ECG electrodes.

The parameters of the current pulses are programmable:

- Frequency setting (46.5 to 64 kHz).
- Current amplitude (8 to 64 μA_{p-p}).
- Selection of the reference ECG lead for the thoracic impedance measurement (i.e. selection of internal respiration paths in leads I, II or III).
- Programmable input gain (10 states) providing measurement resolution of 0.02 to 0.2 Ω.

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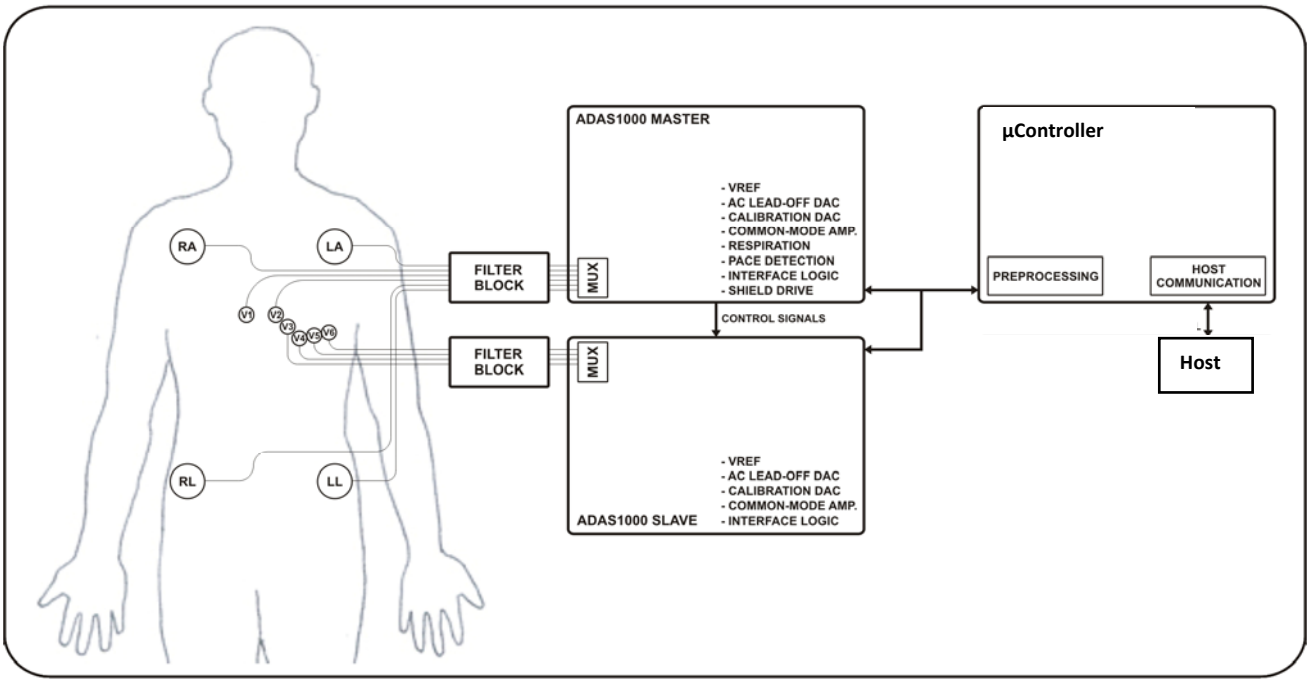

Fig. 1. Block diagram of 12-channel ECG and respiration monitoring system based on two cascaded analog front-end ADAS1000 chips driven by a microcontroller.

The impedance sampling is done synchronously with the ECG channels, returning magnitude and phase information. The change of the thoracic impedance magnitude over time is representative for the respiration activity and rate.

The output ADAS1000 signals are digitally preprocessed by a microcontroller with embedded DSP instructions and a floating point operations module. This extends the mathematical complexity of the build-in signal processing algorithms which include the following basic functions:

- Detection of pacemaker artefacts by analyzing samples in 128 kHz/16-bit ECG data stream.
- High-pass filtering of all ECG leads at 0.05 Hz or 1 Hz cut-off frequency depending on the setting – diagnostic or monitoring ECG mode.
- Low-pass filtering of all ECG leads at 150 Hz cut-off frequency.
- Suppression of power-line interference – 50/60Hz.
- QRS detection and calculation of the heart rate.
- Measurement of thoracic impedance magnitude variation over time that represents the respiration activity output.
- Respiration signal analysis and calculation of the respiration rate. This algorithm is further trained and validated with respiration databases on PC.
- Real-time data transfer to the Host via RS232.

The developed module is designed in compliance with all electro-medical product safety standards and the standards for ECG monitoring and analysis, i.e. AAMI EC11, EC13, EC38 and IEC60601-2-25, IEC60601-2-27, IEC60601-2-51 are fulfilled.

III. Respiration Database

The training and test respiration databases are collected from 25 healthy volunteers by means of the ADAS1000 based module with impedance measurement option as described in the previous section. The excitation current through the chest is set with amplitude about 50 % of the maximal possible current, with high-frequency setting of 50 kHz. The impedance signal is sampled at 125 Hz.

The respiration signals of the first 11 volunteers are included in the training database, acquired via chest electrodes in lead II position (preferred in monitor-defibrillator applications) and including the following respiration episodes in one respiration recording:

- 1 minute normal breathing
- up to 1 minute apnea period
- 1 minute deep breathing
- 1 minute rapid-shallow breathing

After the learning phase has been finished, the test dataset is collected from other 14 persons. It includes thoracic impedance recordings in both lead II and lead I positions. For each person the following respiration attempts are recorded:

- 1 minute normal breathing, apnea period, 1 minute normal breathing
- 1 minute rapid-shallow breathing, apnea period
- 2 minutes deep breathing

The respiration databases are annotated by manual identification of all respiration periods at synchronous instants during maximal peaks. The annotator, however, experienced difficulties in reliable visual identification of all respiration waves during rapid-shallow breathing due to their low amplitudes resembling apnea episodes, as well as their interference with the pulse waves (see the example in Figure 2). Therefore, reference manual annotation of rapid-shallow breathing cycles is banned. In addition, apnea, normal and deep breathing attempts are reliably annotated, collecting in the test database a total number of more than 650 reference 15s-windows sliding of 5s (see Table1).

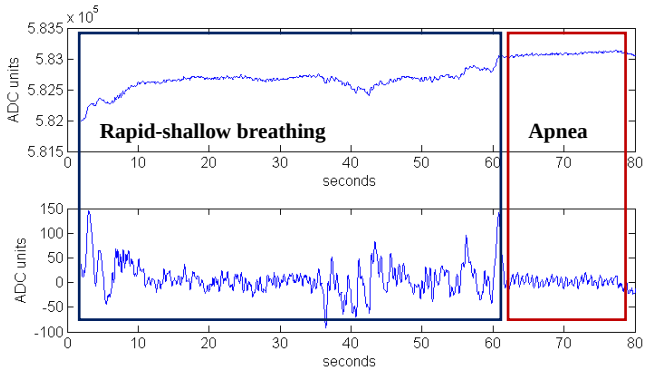

Fig. 2. Example of rapid-shallow breathing followed by apnea period. First trace – raw impedance signal recorded by ADAS1000 module. Second trace – respiration signal after the preprocessing stage. The manual annotation of the respiration cycles between 10s and 35s is impossible, since pulse waves are mostly visible. The high-amplitude waves are either motion artifacts or real normal respiration cycles.

IV. Measurement of the Respiration Rate

The presented algorithm is designed with sample-by-sample architecture for analysis of the respiration signal, thus providing real-time tracing of the respiration rate (RR). It includes the following steps:

1) Preprocessing
   - Subtraction of the mean signal value calculated for the preceding 1 second segment;
   - Low-pass filtering at 2 Hz – allows measurement of RR up to 120 breaths-per-minute (bpm).
2) Setting of initial amplitude threshold (THR_{INIT}) for detection of the respiration activity: Its value should be set above the maximal peak-to-peak amplitude measured during the apnea episodes in the training database. THR_{INIT} is related to the amplitude of the pulse waves which are sensed in the impedance channel and disturb the respiration signal.
3) Detection of the respiration periods based on zero level crossing:
   - Detection of up-slope zero level crossing based on change of the sign of the respiration signal;
   - Validation of the detected zero level crossing according to preset criteria for:
     - minimal duration of the preceding negative wave which is set to preserve against high-frequency noises;
     - minimal duration of the actual respiration period;
     - minimal peak-to-peak amplitude of the actual respiration period;
4) Update of the amplitude threshold after validation of the zero level crossing
5) Calculation of the actual respiration rate *RR* after validation of the actual respiration period (*RP*):
   - If the number of validated zero level crossings within a preset time interval is less than or equal to 1, *RR* is set to 0;
   - Otherwise, *RR* is inversely proportional to *RP*, converted to breaths-per-minute (bpm):

$$RR = 60/RP \quad (1)$$

V. Respiration Rate Error

The respiration rate error is evaluated as the absolute difference between reference and measured RR median values within reference sliding windows of 15s:

$$ErrRR = abs(RRref - RRmeas), \quad (2)$$

- *RRref* is the reference respiration rate proportional to the median reference respiration period *(RPref)* within the sliding window:

$$RRref = 60/median(RPref), \quad (3)$$

- *RRmeas* is the measured respiration rate proportional to the median measured respiration period *(RPmeas)* within the sliding window:

$$RRmeas = 60/median(RPmeas), \quad (4)$$

VI. Results

The accuracy of the presented method for respiration rate measurement is reported in Table 1 on the test database. The mean value of the respiration rate error *ErrRR* is found from 0.7 to 4.7 bpm for deep, normal breathing and apnea reference windows of 15 s, slide by 5 s. Figures 3-6 show examples of the respiration detection performance.

Table 1. Sample size and respiration rate error (Mean ± standard deviation) evaluated for the test dataset.

Breathing:	Apnea	Normal	Deep
Lead I			
Nb 15s windows	64	321	276
ErrRR (bpm)	3.06±6.6	2.03±3.83	0.73±1.46
Lead II			
Nb 15s windows	93	292	297
ErrRR (bpm)	4.68±9.4	2.21±3.96	1.19±1.86

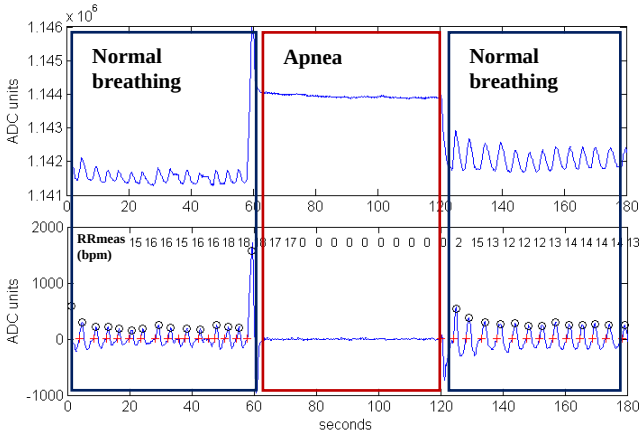

Fig. 3. Example of normal breathing interrupted by apnea period. The traces are the same as in Figure 2 with additional marks of the respiration periods – annotations ('o'), detections ('+') and values of the calculated respiration rate *RRmeas*. All respiration periods are correctly recognized.

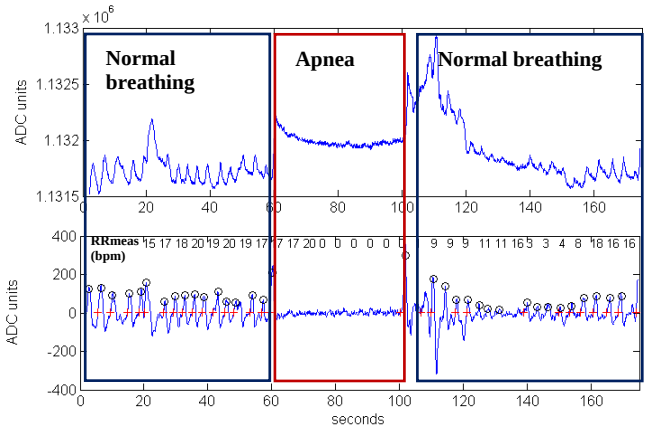

Fig. 4. Example of normal breathing interrupted by apnea episode. The traces and marks are the same as in Figure 3. After the apnea, the amplitude of 5 respiration periods is too low (below THRINIT) and they are not detected. This leads to erroneously low estimation of the respiration rate between 140s - 160s.

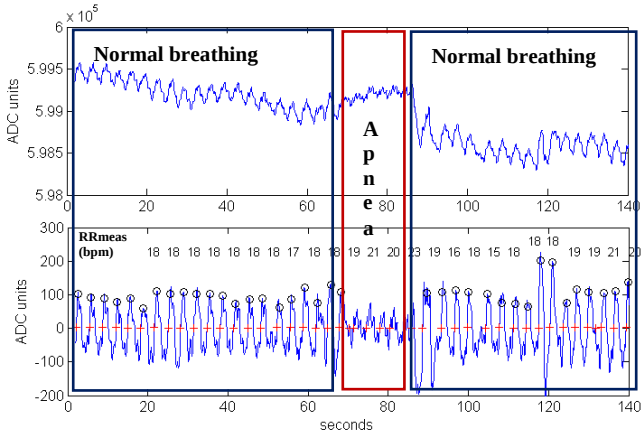

Fig. 5. Example of normal breathing interrupted by short apnea period with visible high-amplitude pulse waves. The respiration detection algorithm has wrongly detected the pulse waves as respiration periods and thus the calculated respiration rate for the apnea period is incorrect.

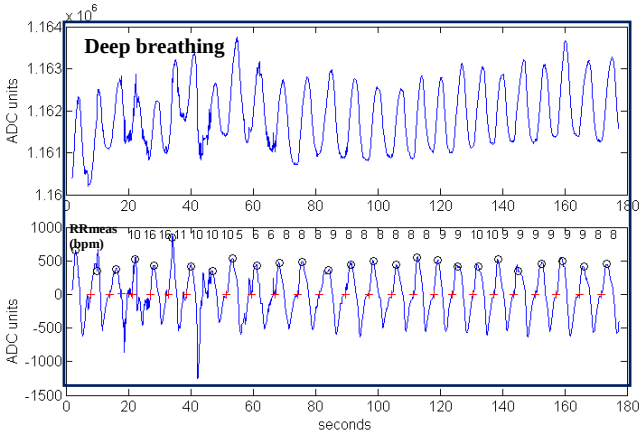

Fig. 6. Example of deep breathing. The traces and marks are the same as in Figure 3. All respiration periods are correctly recognized, except one between 40s and 60s. This breath has relatively low amplitude compared to the previous one and does not cross the updated amplitude threshold. Despite this, the calculated respiration rate is correct, due to the use of median measured RR value.

VII. Discussion and Conclusions

The developed multichannel ECG front-end module is applicable for acquisition of standard 12-lead ECG and synchronous respiration channel using the ECG electrodes in lead I, lead II or lead III position.

The measurement setting with ADAS1000 and 19-bit ADC ensures the acquisition of signals within a large dynamic range which in turn allows for a low amplification of the analog signals – typically 6 to 10-fold. In comparison, the standard ECG amplifiers use more than 250-fold amplification, leading to the well-known practical complication – the mandatory use of high-pass analog filters which result in a slow recovery time of the isoelectric line after peak artifacts or saturation, especially in case of diagnostic 0.05 Hz cut-off frequency. Generally, the ADAS1000-based module relies on digital filtering and complex signal processing algorithms. This is a flexible solution, which is adaptable to the specifics of any target application.

As shown in Table 1, the accuracy of the respiration detection algorithm significantly depends on the breathing amplitude, i.e. deep breathing attempts are more reliably detected than normal and apnea episodes, considering for example *ErrRR* in lead I equal to 0.73 bpm for deep breathing, increasing up to 2.0 and 3.1 bpm for normal breathing and apnea ($p<0.001$). Similar significant trend is also valid for lead II with *ErrRR* in the range 1.2 to 4.7 bpm ($p<0.001$). The electrode lead position does not significantly affect the accuracy for estimation of the respiration rate, although lead I is shown to be preferable than lead II due to about 0.5 to 1.6 bpm lower mean rate error, considering respiration amplitudes from deep to apnea level.

The presented examples show adequate respiration detections for people with regular breathing and smoothed pulse waves (Fig. 3, 6). More complex update of the amplitude threshold could solve the problem observed for variable breathing depth (Fig. 4), while a solution with ECG synchronization could be applicable to manage signal with high pulse waves artifacts (Fig. 5).

The limitation of the study concerns the rapid-shallow breathing episodes for which the respiration rate measurement accuracy could not be reported due to the inability for reliable manual reference identification of all respiration periods (Fig. 2).

References

[1] M. Folke, L. Cernerud, M. Ekstrom, B. Hok. *Critical review of non-invasive respiratory monitoring in medical care*, Medical & Biological Engineering & Computing, vol. 41, pp. 377-383, 2003.

[2] L. Mason. *Signal Processing Methods for Non-Invasive Respiration Monitoring*, PhD Thesis, Department of Engineering Science, University of Oxford, pp. 1-163, 2002.

[3] C. Landon. *Respiratory Monitoring: Advantages of Inductive Plethysmography over Impedance Pneumography*, Ventura County Medical Center, VMLA-039-02.

[4] Analog Devices. *"Datasheet of ADAS1000 Low Power, Five Electrode Electrocardiogram (ECG) Analog Front End"*.

[5] Texas Instruments. *"Datasheet of ADS1293 Low Power Integrated Analog Front End for ECG Applications"*.