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Multichannel Non-Contact System for Breathing Detection and Respiratory Rate Measurement Using Capacitive Electrodes

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Abstract – **The breathing and respiratory rate are one of the important parameters in vital signs monitoring systems. There are various types of methods and sensors used for biomedical and vital signs parameters measurements. Non-invasive and non-contact of them are preferred due to the patient comfort, but they require a new methods and processing algorithms due to high noises level or another reasons. In this work we present a breathing detect and respiratory rate calculating algorithm for the monitoring system based on the microcontroller with CTMU and sheet with a matrix of capacitive electrodes.**

Keywords – **capacitive electrodes; breath detection, respiratory rate measurement, vital parameters monitoring; CTMU.**

I. Introduction

The sensing systems for respiratory monitoring [1] use different sensors, hardware, channels number, sensors positions and algorithms. There are various methods for the detection of breathing. Some of them use different types of wearable devices which are not very comfortable [2, 3]. The authors have presented a multichannel system for measuring of the vital parameters that uses a matrix of capacitive electrodes. [4]. The system is based on the microcontroller with CTMU (Fig.1) and sheet with the matrix of 3x3 capacitive electrodes – Fig.2.

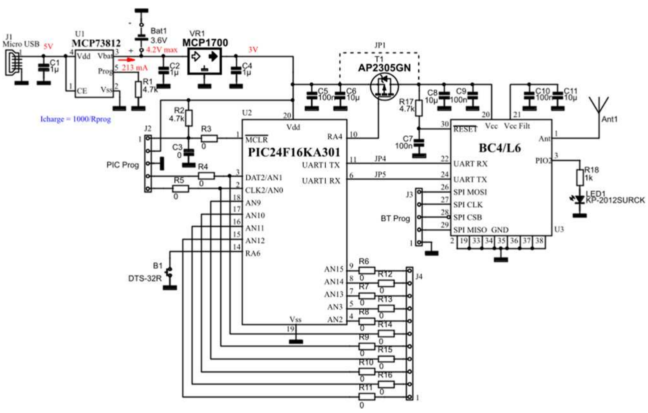

Fig. 1. Capacitance measurement unit circuit diagram

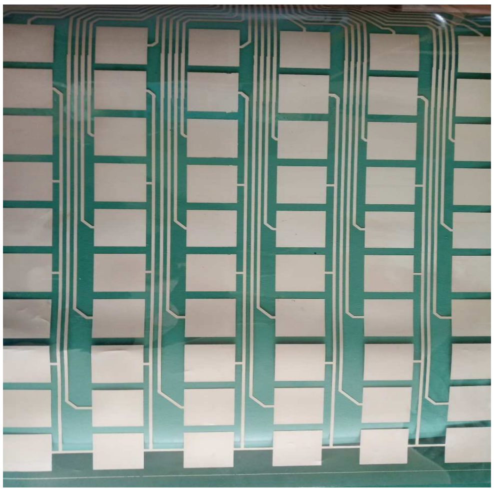

Fig. 2. Construction of the electrode sheet.

The typical signals from the electrodes are presented on the Fig.3. The signal has a sample rate of 10 samples per second for each channel and 12-bit resolution.

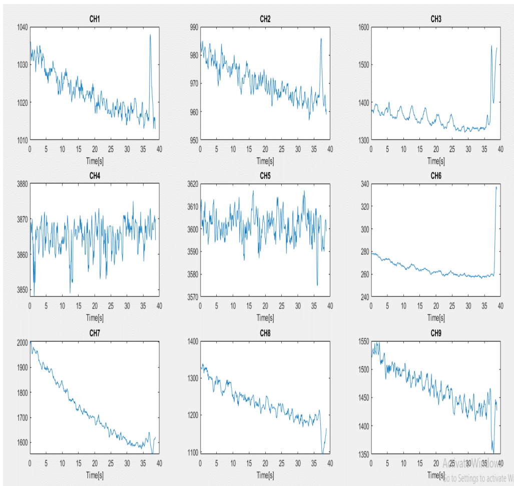

Fig. 3 Typical signals from electrodes.

Some of electrodes are full covered from the patient body, some of them are not covered, and some of them are not full covered. A various artefacts and noises present in the signal – heart beating, muscle tremors, baseline drift and another high frequency noises. This makes it difficult to detect the breathing signal.

II. METHOD

The detection algorithm consists of two sequential procedures – a preprocessor and a peak detector with decision rules.

1. PREPROCESOOR

The preprocessor performs filtering of the signal in order to remove the low and high frequency noises - drift of the zero line, motor artifacts, heartbeat. The normal respiratory rate for an adult is 10 to 20 breaths per minute, so the bandpass filter form 0.17Hz to 0.33Hz is needed.

The filtering procedure is based on three filters. The first one is Filter DxN [5] with coefficients D=5, N=9. The magnitude response of the filter is shown on Fig. 4.

The filter uses difference equation given by (1):

$$Y(i)=X(i)- [X(i-D*(N-1)/2) + X(i-D*(N-3)/2) +\dots X(i) + X(i+D*(N-3)/2) + X(i+D*(N-1)/2)] \quad (1)$$

That filter is used to cut-off low-frequency artefacts and noises below 0.17Hz.

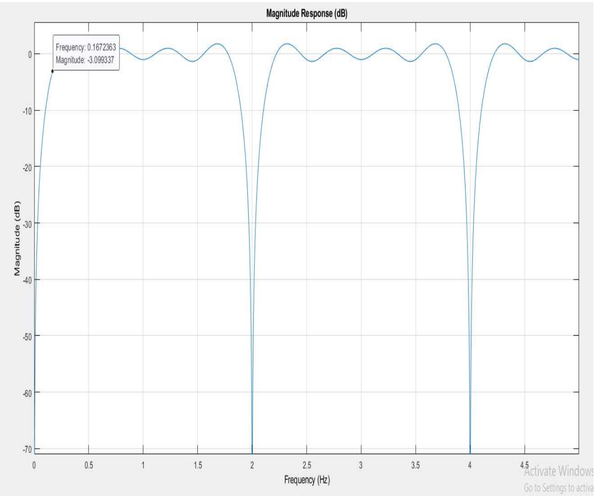

Fig. 4. The magnitude response of Filter DxN(D=5, N=9) Fc=0.15Hz

There are another zeroes on the filter characteristic – respectively on the 2Hz, and 4Hz, but they are out of the breathing frequencies range and don't affect the breathing signal.

The second filter is moving average(MA). It is used to reject the high frequency noises and artefacts. The filter use 10 equal coefficients and its magnitude response is shown on the Fig.5. The cut-off frequency of the filter is around to 0.45Hz.

For better filtering is used third filter which is also MA with 15 equal coefficients(Fig. 6). The cut-off frequency of the filter is around to 0.3Hz.

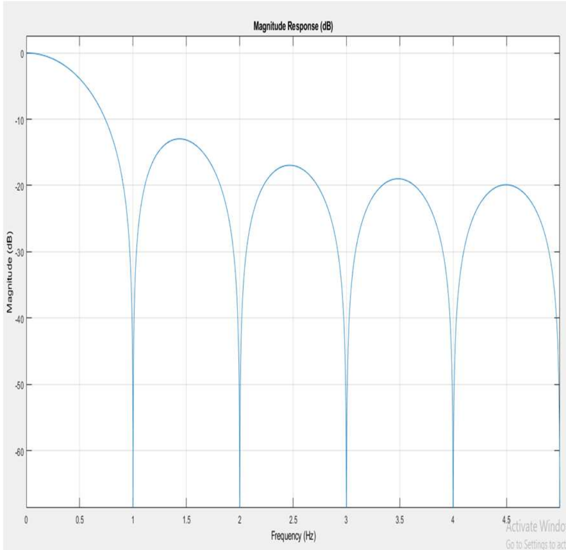

Fig.5 MA Filter (10) Fc=0.45Hz

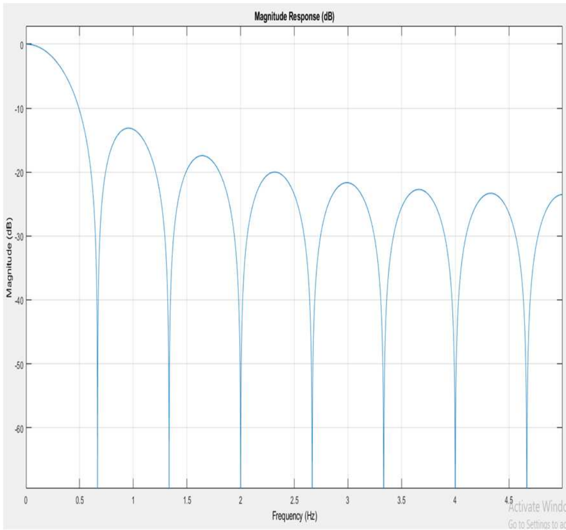

Fig.6 MA Filter (15) Fc=0.3Hz

In all three filters are not used floating point operations. The selection of the filters was made based on the possibility of their use in systems with low computing resources, such as those using microcontrollers.

2. PEAK DETECTION AND RESPIRATORY RATE MEASUREMENT

The respiratory rate measuring algorithm is based on the finding of the maximums of the breathing signal.

The maximums are defined as discrete with an amplitude higher than that of the previous one and the next one, or with an amplitude higher than the previous one and equal to that of the next.

When the system work not of the all electrodes are in good position to measure breathing signal. Some of them are full covered from the parts of the body that didn't make breathing movements, and some of them are not covered by the body.

The first stage of the algorithm clear all of the two neighbor peaks in each channel, that have a distance between them lower than 30 discretes and higher than 60 discretes.

At the same time, a timer is started with a time twice the maximum normal period. If this timer expires before a new valid peak is detected, the normal breathing indicator (trigger) is reset.

The respiratory rate RespR is calculated on the base of the sample frequency Fs and the number of the discretes between two adjacent peaks N (2):

$$RespR = Fs*60/N \qquad (2)$$

For the better accuracy the algorithm use the average value of the last three measured values. The final result is truncated to the integer value.

III. Results

A single channel signal with added noise – HF, LF, baseline drift and heart beating artifacts was synthesized (Fig. 7) for the testing of the algorithm - Fig. 8.

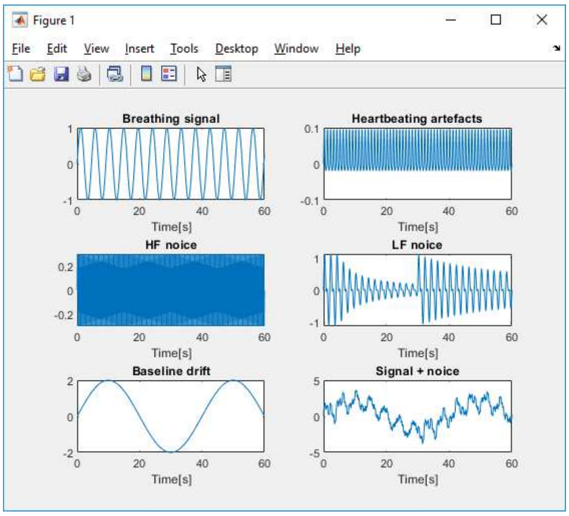

Fig. 7. Single channel breathing signal with added noise

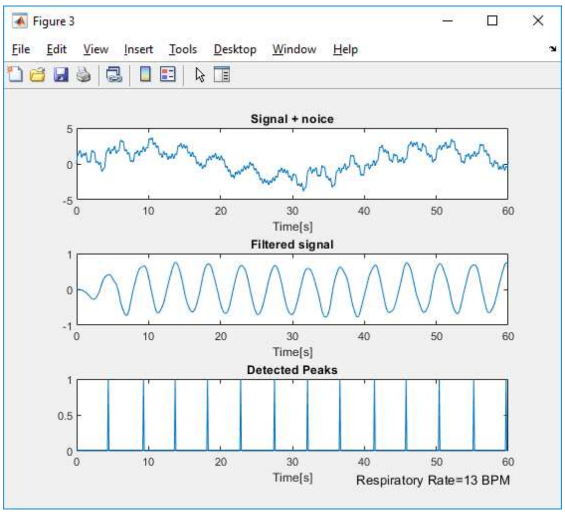

Fig. 8. Single channel filtering and breathing detection

The real setup includes 9 channels. Since one electrode may increase in capacitance and another may decrease during respiration, the absolute value of the signal in the nine channels is taken then the signals in the nine channels are summed to obtain a single channel for the algorithm to work on.

Figure 9 shows real signals from a patient who stops breathing between 15 and 20 seconds into the recording.

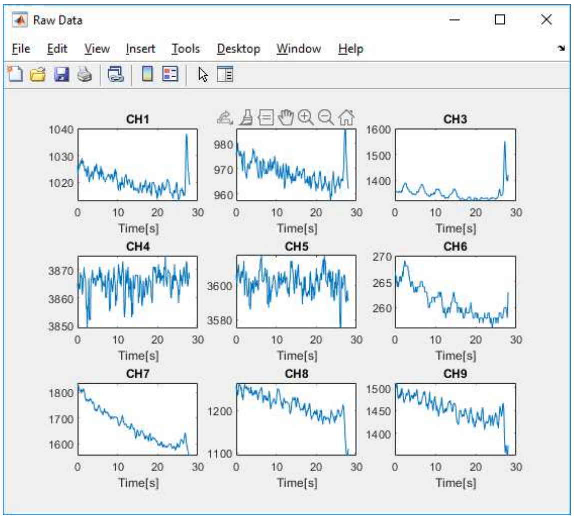

Fig. 9 Signals from a patient before processing

Figure 10 shows the filtered signals after the preprocessor.

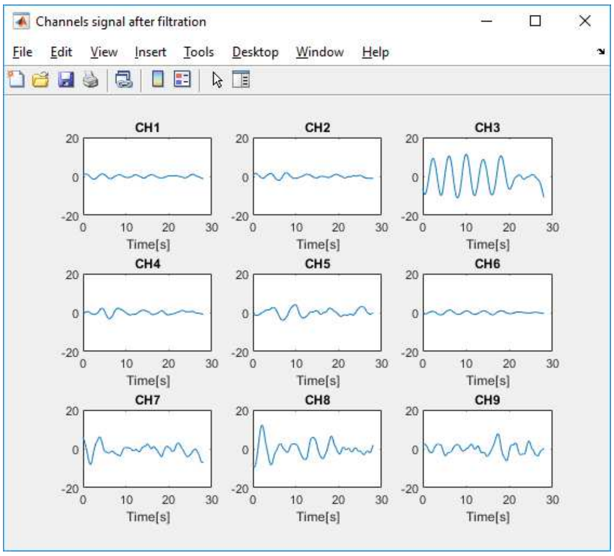

Fig. 10 Filtered signals after the preprocessor

Fig. 11 shows channels taken in absolute value.

Fig. 12 shows in sequence the signal obtained during the summation, the primary detection of peaks, the screening of the incorrectly timed peaks, the respiration rate and the trigger for correct respiration.

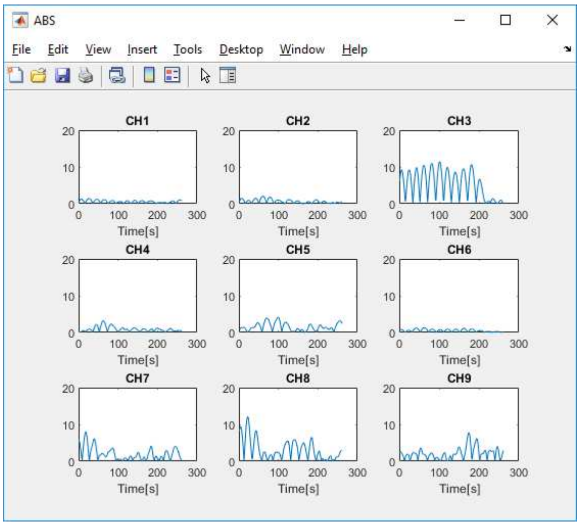

Fig. 11. Absolute values of all channels

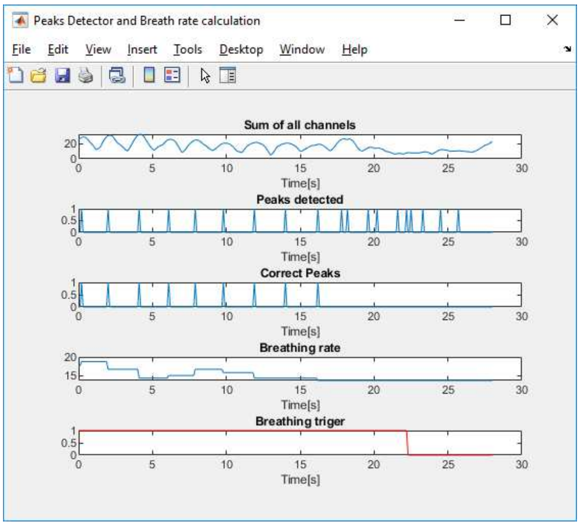

Fig. 12. Peaks detection and BR calculation

IV. CONCLUSION

The results from the measurements show that the developed algorithm can be effectively used to control the status of the patient and the breathing process.

ACKNOWLEDGMENT

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