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The Respiration Rate Estimation Method based on the Signal Maximums and Minimums Detection and the Signal Amplitude Evaluation

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Introduction

An exclusive attention is recently paid to a comfort in the design of the physiological parameter monitoring systems. The comfortable systems could be applied in all areas where physiological parameter monitoring in the human daily activities is essential.

One of the main problems in the systems design is a detection of the distinctive points in the physiological parameter signals during a motion.

Two of the key physiological parameters used for a human state and emotional condition evaluation are the heart rate and the respiration rate [1].

There is the method, applied for the distinctive points of the respiration rate signal detection and a respiration rate evaluation presented in the article. The distinctive points are inductance plethysmography sensor signal minimums and maximums caused by human inhalation and exhalation. The method does not need a lot of calculation resources and could be implemented in the fixed point arithmetic processors and used in a real time.

The sensor of a respiration rate and the features of the respiration rate signal

The respiration inductance plethysmography (RIP) method was chosen for a respiration signal extraction. The method is based on a human abdomen and chest circumference measurement [2]. The sensitive part of the sensor is an elastic strap with a woven sine wave isolated conductor on it (Fig. 1). The two straps are used for the signal extraction. The user puts one strap on the abdomen and another on the chest.

The changes of the strap perimeter induce the changes of the conductor inductance. The inductance of the conductor is applied in the scheme for a rectangular form signal generation. The impulse width depends on the conductor inductance. Further the frequency signal is transformed to the amplitude signal, filtered by a band pass filter (BPF), amplified and passed to analog-to-digital converter (Fig. 2).

Fig. 1. The inductive strap of the respiration rate sensor

The signal of the inductance plethysmography sensor varies round the isoline during rest. The respiration rate evaluation during rest (Fig. 3) is not complicated and could be even based on the isoline crossing events calculation in a period of time.

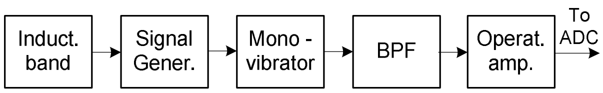

Fig. 2. The structure of the respiration rate sensor

The inductance plethysmography sensor signal deviates from isoline because of the sudden circumference change of the inductance strap. This situation occurs in the daily activities when a human stands up, sits down, stoops down or does relative movements, which changes the circumferences of his/her abdomen and chest (Fig. 4).

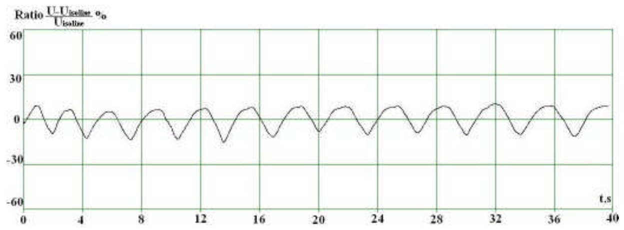

Fig. 3. The respiration sensor signal during rest

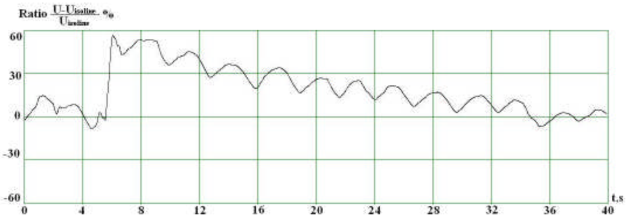

Fig. 4. The respiration sensor signal before and after human stands from the sitting position

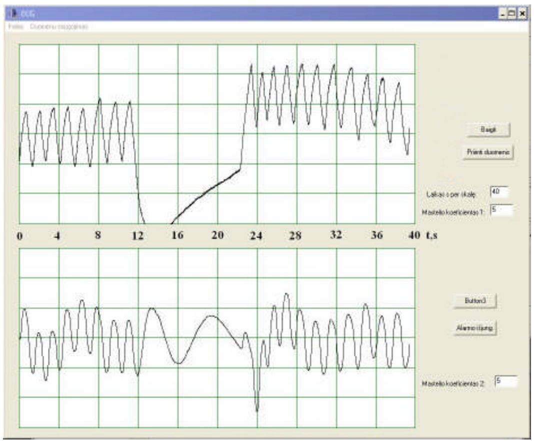

Fig. 5. The respiration sensor signal not processed (above) and processed with the IIR filters (below)

The infinite impulse response high pass filter application for a deviation from isoline suppression is not acceptable. The high pass filters cut off frequency is 0,01-0,05 Hz for the respiration rate signal [3]. The filter oscillations occur and they could be in the same amplitude range and in the same frequency band as a respiration rate signal. Such example is presented in the Fig. 5. The filtered signal oscillations with relation to respiration rate signal amplitude and frequency occurs while a respiration is stopped.

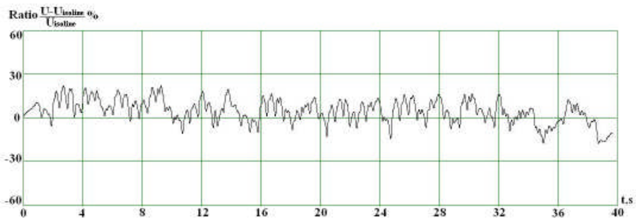

Fig. 6. The respiration sensor signal during walking

A detection of the signal minimums and maximums is one way for the problem solution. The method must be applied deliberately because some motions also induce the peaks in the respiration signal (Fig. 6).

Signal preprocessing

The method for detection of the distinctive points based on the peak detection is used. Peaks could be induced by the human motions also (fig. 6). The moving average filter [4] is used for suppression of these peaks (fig. 7):

$$y(n) = \frac{1}{N}\sum_{k=0}^{N-1} x(n-k) \ , \tag{1}$$

where y is the output value; x is the input value, N is the order of the moving average filter; n is the index of the recent value.

The frequency f response $H(f)$ of the moving average filter is [5]:

$$H(f) = \frac{\sin(\pi f N)}{N \sin(\pi f)} \ . \tag{2}$$

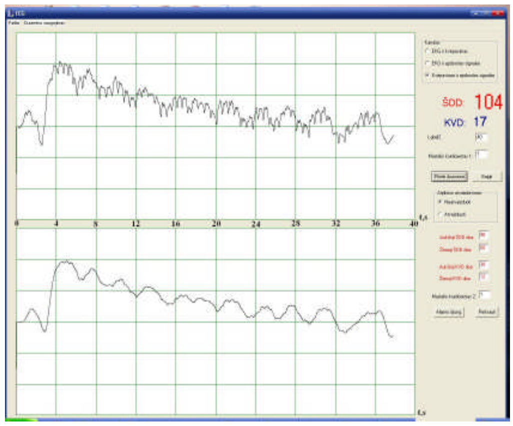

Fig. 7. The respiration sensor signal not processed (above) and processed by moving average filter (below)

The filter filters the noises, but some small noise peaks are left. Suppression of the noise peaks could be increased by increasing the filter order. The filter order must not be too high, because such filter suppresses the respiration signal especially if the respiration rate is relatively high. The alternative solution is application of the adaptive filter, which order depends on a respiration rate: if the respiration rate increases, the filter order is decreased and vice versa.

Respiration rate evaluation algorithm

The algorithm (Fig. 8) for the distinctive points of the respiration rate (inhaling and exhaling maximums and minimums) detection was created, evaluating the specifics of the respiration rate signal. The algorithm could be applied for the respiration rate detection in a real time.

The peak detection is based on the difference Δ change from positive to negative or vice versa detection:

$$\Delta(n) = y(n) - y(n-j), \tag{3}$$

where $\Delta(n)$ is the recent difference of a the marginal peak detection window values, $y(n)$ is the current value of the respiration rate signal amplitude and $y(n\text{-}j)$ is the signal amplitude value obtained before the time interval t_w:

$$t_w = T \cdot j \ , \tag{4}$$

where T is the period of discretization, j is the length of peak detection window, expressed by a number of discretized signal values.

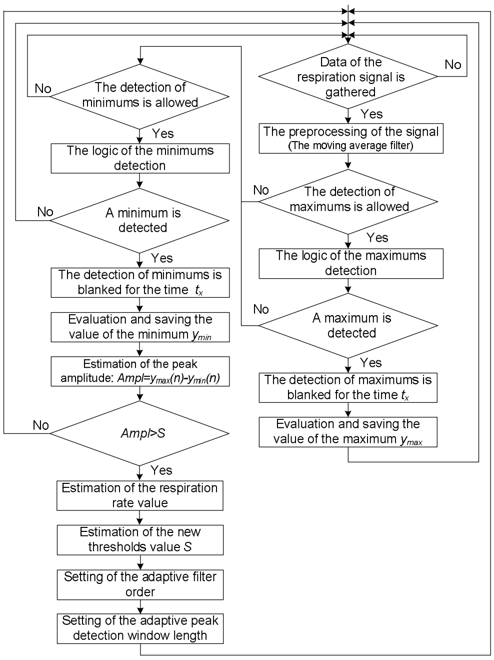

Fig. 8. The algorithm for the respiration rate evaluation

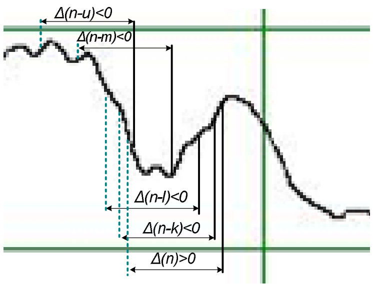

Fig. 9. The logic for the peak detection

The length of the peak detection window (j) is set adaptively and depends on a respiration rate. The number j is decreased if respiration rate increases and j is increased if respiration rate decreases. The initial length is 0.9 s of the peak detection window. The same length is applied for the respiration rate range of 0÷20 breaths per minute. Normal respiration rate of an adult varies from 12 to 20 breaths per minute [6]. The window of 0.9 s could also be applied for detection of the respiration events during the higher respiration rates (even for up to 45 breaths per minute) with lower reliability.

A minimum is detected if Δ value changes from negative to positive:

$$\Delta(n-1) < 0 \text{ and } \Delta(n) > 0. \tag{5}$$

The minimum value could be found approximately:

$$y_{\min} = y(n - \text{floor}(j/2)), \tag{6}$$

where floor is the operation of rounding down to the integer value, or exactly by searching the minimal value in the window n-j ÷ n:

$$y_{\min} = \text{MIN}[y(n-j), y(n-j+1)....y(n-1), y(n)]. \tag{7}$$

A maximum is detected if the Δ value changes from positive to negative:

$$\Delta(n-1) > 0 \text{ and } \Delta(n) < 0. \tag{8}$$

The maximum value could be found approximately:

$$y_{\max} = y(n - \text{floor}(j/2)), \tag{9}$$

or exactly by searching the maximal value in the window n-j ÷ n:

$$y_{\max} = \text{MAX}[y(n-j), y(n-j+1)....y(n-1), y(n)]. \tag{10}$$

The current peak amplitude is evaluated if a minimum is detected:

$$Ampl(n) = y_{\max}(n) - y_{\min}(n). \tag{11}$$

The amplitude is compared with an adaptive amplitude threshold in order to avoid a confusion of the noise peak and the respiration signal peak.

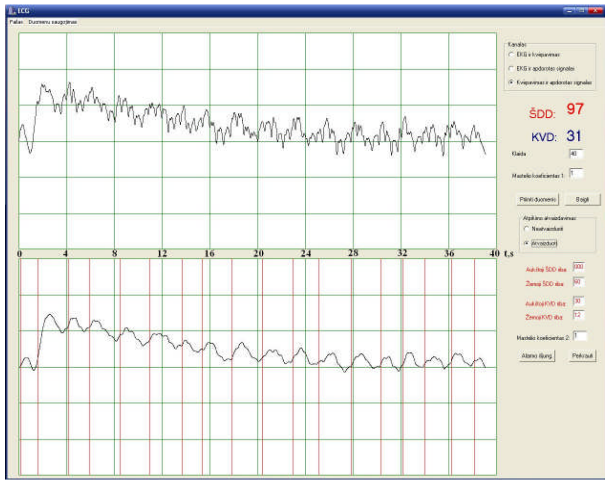

Fig. 10. The respiration rate detection during motion (walking). Not filtered signal (above) and filtered signal (below)

If $Ampl>S$, the detected peak is a respiration signal peak and if $Ampl<S$, the detected peak is a noise peak.

The adaptive threshold S is obtained from the past respiration rate peak amplitude values by applying the median filter:

$$S = C \cdot \text{Median}(Ampl(n-k), Ampl(n-k+1),..., Ampl(n-1)), \tag{12}$$

where S – the threshold value, C – the threshold constant, k – the order of the median filter, Median – the function for the median value estimation.

The adaptive threshold is not applied during the initial stage of the respiration event detection. The moving average filter and the peak detection window show the sufficient performance during the initial stage of the respiration rate evaluation.

The current value of the respiration rate is obtained according to a passed time during the two respiration signal minimums.

The current value of the respiration rate is calculated after each minimum of the respiration signal is detected. According to the respiration rate the moving average filter and the peak detection window are adaptively adjusted:

$$\begin{aligned}
&\text{if } RR < DIAP1;\ j = J(1),\ N = N(1);\\
&\text{if } DIAP1 < RR < DIAP2;\ j = J(2),\ N = N(2);\\
&\ldots\\
&\ldots\\
&\text{if } DIAP(k-1) < RR < DIAPK;\ j = J(k-1),\ N = N(k-1);\\
&\text{if } RR > DIAPk;\ j = Jk,\ N = Nk,
\end{aligned} \tag{13}$$

where RR – the current respiration rate (instantaneous, average or median); $DIAPn$ – the top value of the n-th respiration rate range:

$$DIAP1 < DIAP2 < ...... < DIAP(k-1) < DIAPk; \qquad (14)$$

where j – the length of the peak detection window; $J(n)$ – the particular length of the peak detection window assigned to the particular range of the respiration rate:

$$J1 > J2 > ... > J(k-1) > Jk., \qquad (15)$$

where N – the order of the moving average filter; $N(n)$ – the order of the moving average filter assigned to the n-th range of the respiration rate.

The developed method for the respiration rate evaluation is implemented in the C++ programming language and applied in the system of a physiological parameter monitoring.

Conclusions

1. The respiration rate signal obtained by the inductance plethysmography sensors distinguishes fast and big deviations from isoline. The deviations are induced by the human torso circumference changes. The application of the high pass filters for the deviations suppression is not acceptable, because of the filters oscillations. So, formation of threshold is complicated. Therefore the method of respiration rate evaluation based on the peaks detection was developed.
2. The respiration rate signal could have many peaks induced by some noises. These peaks are suppressed by the adaptive moving average filter according to the method. The order of the filter depends on the respiration rate.
3. The robustness to noises of the method depends on the length of the peak detection window. The adaptive, variable length peak detection window is applied in the method.
4. To improve a reliability of the method, all amplitudes of the detected peaks are compared to an adaptive threshold.

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The respiration rate estimation method is developed. The method is based on the signal maximum and minimum values detection and amplitudes evaluation. The high deviations from isoline of the inductance plethysmography sensor signal are caused by the human torso movements. The application of the common filters for the deviations suppression is not acceptable because of the oscillations caused by filters. Detection of signal minimums and maximums is applied in the method. The noise peaks are suppressed by the adaptive moving average filter. The adaptive peak detection window with the variable width is applied for the peak detection. The detected peaks are compared with an adaptive peak threshold. The method is implemented in the C++ language and applied in the system of physiological parameter monitoring. Ill. 10, bibl. 6 (in English; summaries in English, Russian and Lithuanian).

Р. Лукочюс, Ю. А. Вирбалис, И. Даунорас, А. Вэгис. Метод для вычисления частоты дыхания основан на обнаружении пиков сигнала и оценки амплитуды сигнала // Электроника и электротехника. – Каунас: Технология, 2008. – № 8(88). – С. 51–54.

Создан метод для определения частоты дыхания, основанный на обнаружении пиков сигнала и оценке амплитуд пиков. Большие отклонения от изолинии сигнала датчика индуктивной плетизмографии вызваны движениями туловища человека. Так как рекурсивные фильтры генерируют паразитные колебания, они не приемлемы для подавления этих отклонений. Поэтому в разработанном методе применено обнаружение пиков. Шумовые пики подавляются фильтром скользящего среднего значения. Для обнаружении пиков применено адаптивное окно с переменной шириной. Обнаруженные пики сравниваются с адаптивным порогом. Метод осуществлен посредством языка программирования C ++ и применен в системе контроля физиологических параметров. Ил. 10, библ. 6 (на английском языке; рефераты на английском, русском и литовском яз.).

R. Lukočius, J. A. Virbalis, J. Daunoras, A. Vegys. Kvėpavimo dažnio nustatymo metodas, pagrįstas signalo maksimumų ir minimumų aptikimu ir signalo amplitudės įvertinimu // Elektronika ir elektrotechnika. – Kaunas: Technologija, 2008. – Nr. 8(88). – P. 51–54.

Sukurtas kvėpavimo dažnio nustatymo metodas, pagrįstas signalo maksimumų ir minimumų aptikimu ir jų amplitudės dydžio nustatymu. Sukeliamų svyravimų begalinės delsos impulsiniai filtrai taikomi induktyvinės pletizmografijos jutiklio signalo nuokrypiams nuo izolinijos, susidarantiems dėl žmogaus korpuso judesių, slopinti. Metodas remiasi kvėpavimo sukeliamų signalo minimumų ar maksimumų aptikimu. Triukšmų pikai slopinami adaptyviu slenkančio vidurkio filtru. Pikams aptikti naudojamas adaptyvus kintamo pločio pikų aptikimo langas. Aptikti pikai įvertinami palyginant juos su adaptyviu pikų amplitudės slenksčiu. Metodas realizuotas C++ programavimo kalboje ir pritaikytas fiziologinių parametrų stebėsenos sistemoje. Il. 10, bibl. 6 (anglų kalba; santraukos anglų, rusų ir lietuvių k.).

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