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Physiological Parameters Monitoring System for Occupational Safety

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***Abstract*—A wireless physiological parameters monitoring system designed for workers to observe their health conditions during daily activities was proposed and analyzed in our work. This system meant to protect people from accidents and sudden health impairments at work. The key part of the system is wearable sensors that follow human respiratory function and cardiovascular system work. Remote computer evaluates one's health condition by tracking the abnormal deviations of the two main physiological parameters – heart rate and respiratory rate. Attention in this paper was focussed on suppression of the signal noise caused by body movement artefacts. Some new algorithms of the data processing were developed to this end. The proposed monitoring system was assembled and calibrated against standard medical monitor. Results of the system accuracy test are presented in this paper.**

***Index Terms*—ECG, heart rate, inductive plethysmography, respiratory rate, remote monitoring.**

I. INTRODUCTION

Real time patient health monitoring system is a common mean for clinical observations at hospitals or rehabilitation treatment at home. However, continuous health condition monitoring of healthy people required for worker's safety or sportsmen's training purposes is rather new approach of a growing interest in recent years [1]–[3]. The main difficulties of implementation of such systems arise due to nature of worker's daily activities and specifically of the signal noise, related to body movements. The noise superimposed to the signal often significantly exceeds the signal itself. This situation requires more sophisticated equipment and methods of signal processing to get reliable data of the physiological parameters we are seeking for. Another demand is the equipment used for monitoring must be non-invasive and body non-interventional. Most of today's monitoring systems are designed for stationary or reduced movement patients and don't meet requirements for healthy and active people monitoring. In this paper we present non-invasive physiological parameters monitoring system (PPMS) designed for workers (i.e., dispatchers, plant operators, miners, etc.) to observe their health conditions in terms of safety secure at workplaces. A real-time observation and evaluation of the physiological parameters enables us to provide an urgent help in case of sudden health impairment (paralytic stroke, cardiac arrest, intoxication and etc.) and both to minimize the risk of fatal accidents at work.

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II. STRUCTURE OF THE SYSTEM

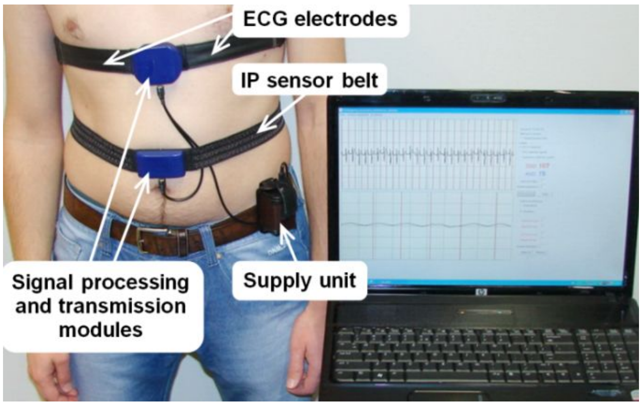

Fig. 1. View of the physiological parameter monitoring system.

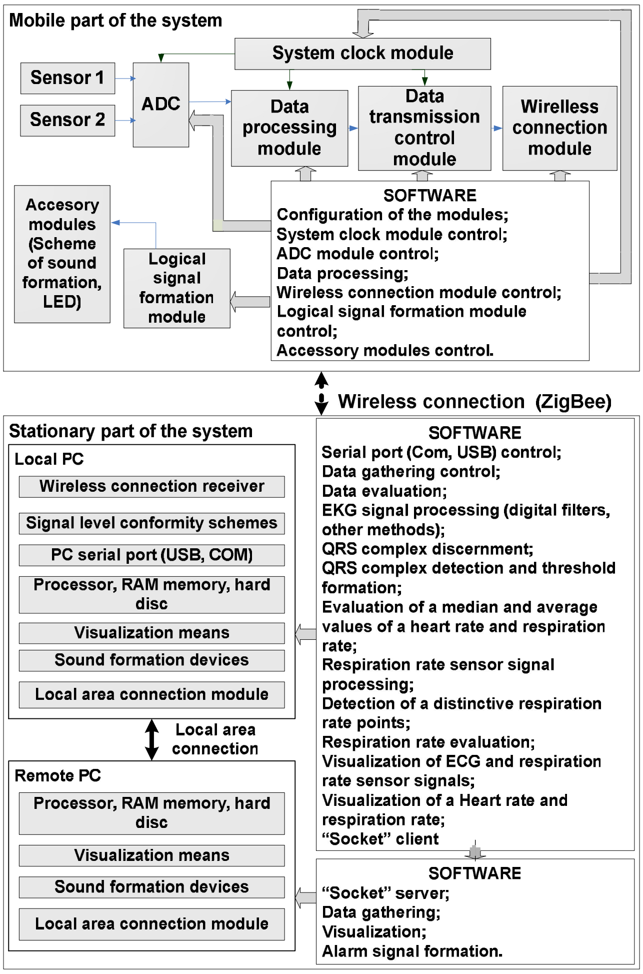

Fig. 2. Structure of the physiological parameter monitoring system.

PPMS mobile part is implemented as set of wearable belts with control modules attached to them (Fig. 1). It consists of one Electrocardiography (ECG) and one or two Inductive Plethysmography (IP) sensors with signal formation and amplification circuits, microcontroller, wireless transmitter and software. Obtained by the sensors, amplified and processed analogue signals are sent to microcontroller, and then after discretization and pre-processing they are sent to wireless transmitter.

Stationary part consists of wireless receiver, local and remote personal computer (PC), and software. Signals from a wireless receiver through a USB port are sent to local PC, where further data processing, heart and respiratory rates calculation, visualization and alarm signals formation is implemented. Next, parameter values and alarm signals are transmitted by a computer network (LAN, Internet) to a remote PC in remote monitoring centre. Detailed structure of the developed PPMS is shown in Fig. 2.

III. HEART RATE SENSOR HARDWARE

Heart rate sensor of the system is based on biopotential (ECG) measurement. The method is more robust to human body motions in comparison with competitive methods. This is important taking in mind the system is applied for industrial safety.

ECG analog signal processing module proposed in this work is presented in Fig. 3.

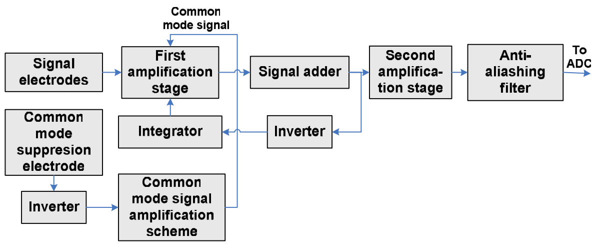

Fig. 3. Structure of the analog ECG signal processing module.

As a passive heart rate sensor part was chosen "Polar" textile electrodes ("Polar Wearlink electrode belt") [4]. Active sensor part was developed and implemented. Signal is amplified 500 times by two amplification stages in the circuit. Low frequency noise is suppressed by applying integrator and inverter in negative closed loop. 3 dB cutoff frequency of the filter is 0,5 Hz. Discerned, amplified and inverted common mode signal is transmitted back to human body by a third electrode mounted on electrode belt to suppress common mode noise. Anti–aliasing analog filter is applied before signal passes to ADC.

IV. HEART RATE EVALUATION ALGORITHM

Modified Pan and Tompkins algorithm [5] is applied for heart beat detection and heart rate evaluation in a local PC. Band-pass filtering, derivative, squaring and digital integration procedures are applied for a QRS complex discernment.

Digital IIR Band-pass filter with the 3 dB cut-off frequencies at 8 Hz and 20 Hz for motion artifacts P, T wave, power lines noise and high frequency noise suppression is applied. Band-pass filter is implemented as series 8^{th} order high-pass filter and 8^{th} order low-pass filter connection.

Digital derivative of the 14–th order is used for further QRS complex emphasis. Implementation of the derivative

$$H(z) = \frac{\sum_{i=0}^{K} a_i \cdot z^{-i1}}{K}, \quad (1)$$

where K is number by one lesser as an order of the derivative, a_i – coefficients of the derivative.

Squaring is next signal processing stage

$$h(nT) = [x(nT)]^2, \quad (2)$$

where n is number of the last sample, T is sampling period. Squaring emphasizes higher frequencies, which are mainly due to QRS complex in the signal. The last ECG signal processing stage is moving window integration

$$y(nT) = \frac{\sum_{j=1}^{N} h(nT - (N-j)T))}{N}. \quad (3)$$

N is integration window length in samples. It is chosen according to maximal possible QRS complex duration time i.e. 150 ms.


Fig. 4. ECG signal (above) and processed signal (below) during rest.

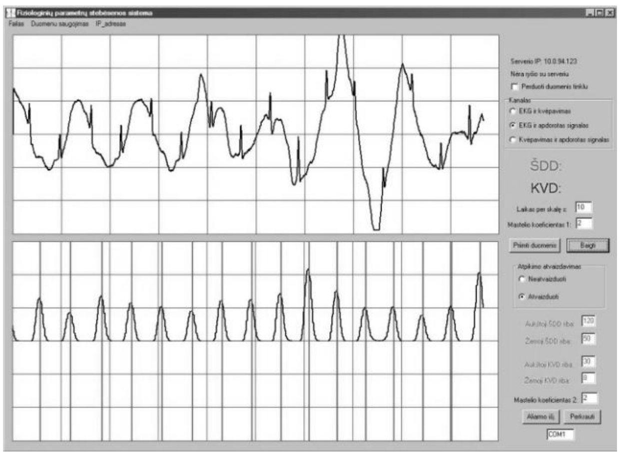
Fig. 5. ECG signal (above) and processed signal (below) during motion.

ECG signal and the processed signal obtained by the system sensors during rest are shown in Fig. 4, using a PPMS visualization tool. The same signals during motion are shown in Fig. 5. As we can see here, an intensive physical work causes a significant signal distortion, and thus, requires more complex signal processing algorithms than those used in analogous systems for patient treatment [3].

QRS complexes are detected as maximum points of the processed ECG signal by applying moving difference window. The principle is described in detail further in section about the respiratory rate evaluation. Fixed window length conversely to respiratory rate evaluation method is used in QRS complex detection.

V. RESPIRATORY RATE SENSOR HARDWARE

A respiratory rate was selected to be measured using Inductive Plethysmography (IP) method [6]. Compared with other techniques, IP has the advantages of greater resistance to noise and higher safety, and unlike Impedance Plethysmography it is fully un-invasive method [7].

A common Respiratory Inductance Plethysmograph (RIP) consists of two wavy wire coils attached to the soft belts and placed around the rib cage under the armpits and around the abdomen. Both areas are measured by connecting the coils as the inductance in variable frequency LC oscillators, which frequency is proportional to the self-inductance of the coils and thus the radius of surrounded area. Those areas expand with each inhalation. The output signal is produced by combining two oscillator signals together. This so called dual-band system is more resistive to motion artefacts than a single-band system with only one inductive coil. However, our designed PPMS can realize both methods.

Respiratory signal conditioning circuit proposed in this work is made of variable frequency oscillator, respiratory signal demodulator, band-pass filter and amplifier (Fig. 6.).

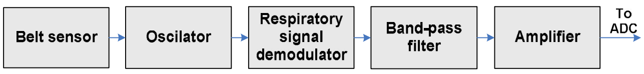

Fig. 6. Structure of the one channel RIP signal processing module.

The nominal frequency of the signal generated by the oscillator is set to 300 kHz. Demodulator and filter together present frequency to voltage converter. Demodulator transforms the oscillator signal into a pulse signal with a constant pulse width. Frequency of the pulse signal is the same as generated by oscillator. Band-pass filter cuts off high and low frequencies and passes respiratory signal, within frequencies from 0,02 Hz to 180 Hz. The filter is followed by signal amplifier with the gain of 20. Finally, the output signal of required magnitude gets into microcontroller for further processing.

VI. BREATHS DETECTION AND RESPIRATORY RATE EVALUATION ALGORITHM

Developed breath detection method is based on peaks detection [8]. Human body motions induce noise. The noise creates undesirable peaks in the signal. The noise is reduced by a moving average filter

$$y(nT) = \frac{\sum_{k=1}^{L} x[(n+k-L)T]}{L}, \qquad (4)$$

where n is number of the last sample; x is input value of the signal; y is output value of the signal; L is length of the moving average window in samples; T is sampling period.

Peaks are detected by applying moving difference window. Application of the moving difference window makes detection prone to low frequency noise, i.e. deviation from zero signal line. Application of wide difference window makes the method prone to relatively high frequency noise. Adaptive moving difference window is applied. Length of the window is inversely proportional to respiratory rate.

Maximum peak is detected if condition

$$y((n-1)T) - y((n-Q-1)T) \geq 0 \,\&\, y(nT) - \\ - y((n-Q)T) < 0, \qquad (5)$$

is met. The point of peak lies in the window $(n-Q)T \div nT$, where Q is moving difference window length.

Signal value of maximum peak is detected

$$y_{\max} = y(round(n - \frac{Q}{2})T). \qquad (6)$$

Minimum peak is detected if condition

$$y((n-1)T) - y((n-Q-1)T) \leq 0 \,\&\, y(nT) - \\ - y((n-Q)T) > 0, \qquad (7)$$

is met.

The signal value of minimum peak $y_{\min}$ is calculated analogous to (6).

Each time a peak is detected blanking period is applied. Further procedure is amplitude of the peaks evaluation. Amplitude of the recent maximum is difference of the last breath induced maximum y_{max} and minimum y_{min} values. Peaks are considered as breath induced peaks if value of the peak is greater than a threshold. Peaks threshold χ is formed as a median filter value of the last k peaks values

$$\chi = C \times \mathrm{Med}(y_{\max,l-k} - y_{\min,l-k}; y_{\max,l-k+1} - \\ - y_{\min,l-k+1}; \ldots; y_{\max,l-1} - y_{\min,l-1}; y_{\max,l} - y_{\min,l}), \quad (8)$$

where Med is median value evaluation procedure; C is threshold constant; l is number of the last detected minimum or maximum; k is length of the median filter.

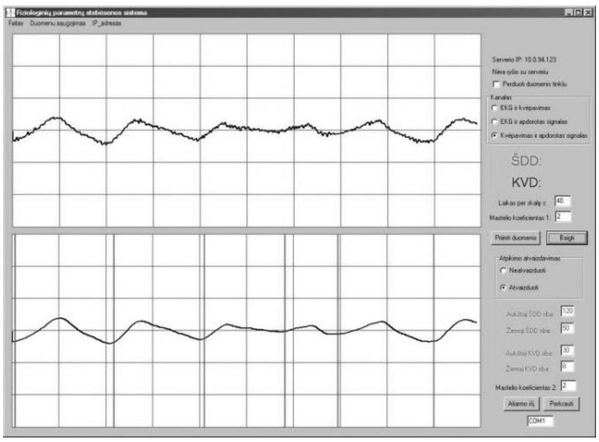

Fig. 7. Signal of the RIP sensor during rest. Original signal (above), processed signal (below).

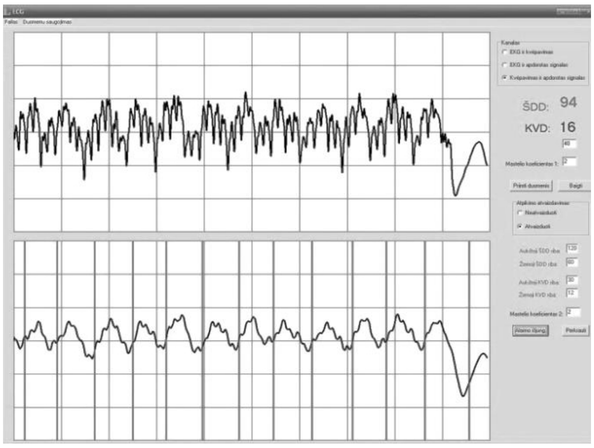

Fig. 8. Signal of RIP sensors during motion (walking). Original signal (above), processed signal (below).

Figure 7 and Fig. 8 illustrate respiratory signals during rest and motion periods.

Respiratory rate is recalculated each time a detected minimum peak is considered as a breath induced peak.

VII. EXPERIMENTAL ANALYSIS

PPMS was analysed experimentally by comparing results of measurements made in real conditions with our designed system and measurements obtained simultaneously by standard patient monitor MEC-1000 (Mindray). This monitor is used for clinical monitoring of the health state of adults and children. MEC-1000 evaluates heart rate from electrocardiographic signal obtained by I, II or III standard leads. In our case, it was an II standard lead used. Respiratory rate is evaluated by Impedance Plethysmography method, using II standard lead electrodes. The basic accuracy of the monitor is 1 % for heart rate estimation and 2 % for respiratory rate estimation.

The aim of the experiment was to measure heart and respiratory rates for the same person using our designed monitoring system and the standard monitor with well-defined parameters, and to compare obtained values. Both devices worked simultaneously and their readings of heart and respiratory rates were taken at the same time. Random time for discrete readings was chosen, holding at least twenty seconds (it’s approximate time of data averaging by using a moving averaging window) pauses between succeeding readings, so that previous measurements haven’t influence on each successive measurement. The experiment lasted till sufficient amount of data for statistical analysis were collected. The differences between corresponding rates obtained with both devices at each measurement point were calculated. In this way, the mean value of total heart rate differences was found to be 0.26 beats per minute (bpm) and standard deviation – 1.77 bpm. For respiratory rate those parameters were 0.24 rpm and 3.84 rpm correspondingly. Those values are rather small as compare to absolute values of heart and respiratory rates (heart rate – 61…93 bpm and respiratory rate – 9…21 rpm) measured in this experiment.

While exploring the possibility of methodological error in the designed PPMS, the hypothesis about the errors (rate differences) distribution according to normal (Gaussian) distribution law was revised. The Kolmogorov-Smirnov test was applied for this reason. After the data testing with the level of significance $\alpha = 0.05$ (standard case), it was stated that the errors are distributed according to normal distribution law for both heart rate and respiratory rates.

A statistical analysis of data obtained with different devices was performed for a quantitative evaluation of experimental results. Considering the fact, that all data are distributed according to normal distribution law, the t-test was applied to find out if the data fit statistically. In this case the level of significance was taken as standard ($\alpha = 0.05$). The results showed up that differences between the heart rate values measured with the proposed monitoring system and the metrologically verified monitor MEC-1000, are statistically insignificant (p-value 0.358). The same way values of the respiratory rate also differs insignificantly (p-value 1.66). It must be mentioned, that different physical principles applied for respiratory rate measurements, have no significant influence on the results too.

VIII. CONCLUSIONS

Proposed algorithm for heart rate evaluation from ECG signal provided a good QRS detection and heart rate calculation reliability for cases, when the signal is formed during high intensiveness of human body motion.

The IP method was proposed for respiratory rate evaluation due to its un-invasive nature, noise robustness and convenience as compare to other techniques. The proposed respiratory rate calculation algorithm effectively detects distinctive points of breaths from the signal of RIP sensor obtained during body motion.

Experimental results showed up that differences between the heart rate and respiratory rate values measured with the proposed monitoring system and monitor MEC-1000, are distributed according to normal distribution and are statistically insignificant. Those results show that the developed system can be applicable for worker’s health condition monitoring with sufficient accuracy for an intensive physical work demanding occupations.

REFERENCES

[1] E. Teaw, G. Hou, M. Gouzman, K. W. Tang, M. Kane, A. Kesluk, J. Farrell, “A wireless health monitoring system”, in *Proc. 2005 IEEE Int. Conf. Information Acquisition*, 2005, pp. 247–252. [Online]. Available: http://dx.doi.org/10.1109/ICIA.2005.1635090

[2] A. Mikelovic, C. Otto, E. Jovanov, “Wireless sensor networks for personal health monitoring: Issues and an implementation“, *Computer Communications*, vol. 29, pp. 2521–2533, 2006. [Online]. Available: http://dx.doi.org/10.1016/j.comcom.2006.02.011

[3] Y.-D. Lee, W.-Y. Chung, “Wireless sensor network based wearable smart shirt for ubiquitous health and activity monitoring”, *Sensors and Actuators B*, vol. 140, pp. 390–395, 2009. [Online]. Available: http://dx.doi.org/10.1016/j.snb.2009.04.040

[4] L. Rodriguez-Linares, P. Cuesta, R. Alonso, A. J. Mendez, M. J. Lado, X. A. Vila, “VARVI: A software tool for analyzing the variability of the heart rate in response to visual stimuli”, *Computing in Cardiology*, vol. 40, pp. 401–404, 2013.

[5] B. U. Kohler, C. Hennig, R. Orglmeister, “The principless of software QRS detection”, *IEEE Engineering in Megicine and Biology*, pp. 42–57, 2002.

[6] M. Vaitkunas, A. Dosinas, V. Bartkevicius, J. Daunoras, “Respiratory signal estimation using non-invasive measurements”, *Elektronika ir Elektrotechnika*, no. 1, vol. 89, pp. 35–38, 2009.

[7] C. F. Clarenbach, O. Senn, T. Brack, M. Kohler, K. E. Bloch, “Monitoring of ventilation during exercise by a portable respiratory inductive plethysmograph”, *Chest*, vol. 128, no. 3, pp. 1282–1290, 2005. [Online]. Available: http://dx.doi.org/10.1378/chest.128.3.1282

[8] R. Lukocius, J. A. Virbalis, J. Daunoras, A. Vegys. “Respiration rate evaluation method based on signal maximum and minimum values detection and signal amplitude evaluation”, *Elektronika ir Elektrotechnika*, no. 8, pp. 51–54, 2008.