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Method of digital processing of photoplethysmogram signals for non-invasive monitoring of physiological parameters in real time

S.N. Glebov, Yu.V. Lyamina, M.S. Mazing

Institute of Analytical Instrument Making, Russian Academy of Sciences, St. Petersburg, Russia
E-mail: julia.liamina@gmail.com

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A method for processing photoplethysmogram signals relying on window analysis (10 s) and a combination of a fourth-order Butterworth filter and a modified peak detection algorithm is proposed. It is demonstrated that the average absolute error of heart rate determination is 1.42 bpm with a signal processing time in a single window below 20 ms. The method is optimized for low-performance microcontrollers of wearable devices.

Keywords: photoplethysmography, digital filtering, physiological state monitoring, real-time signal processing.

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The demand for non-invasive physiological monitoring systems has increased in recent years. Photoplethysmography is of particular interest as an optical method for heart rate (HR) measurement, since it is rather simple from a technical perspective and non-invasive.

A specialized hardware and software platform was designed in order to implement this method. The hardware part (Fig. 1) includes an optical sensor based on LEDs with wavelengths of 537, 660, and 880 nm and a photodiode element. The sampling frequency was set to 100 Hz, which provides sufficient temporal resolution for accurate HR measurement.

A software application with graphical user interface, which allowed one to visualize input data from the photoplethysmogram (PPG) sensor and parameters calculated by the developed algorithm and monitor the dynamics of indicators (Fig. 2), was written.

a b

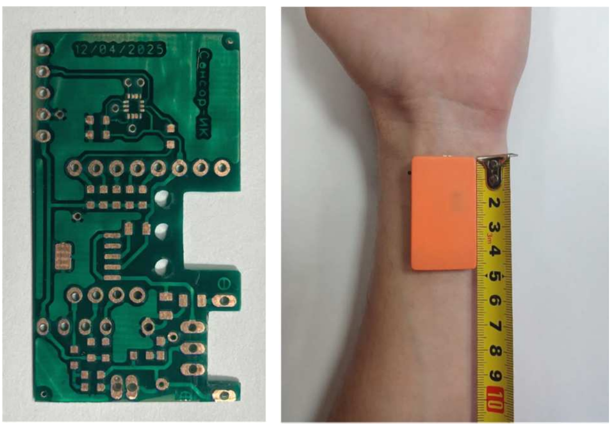

Figure 1. *a* — Printed circuit board of the sensor fabricated at the Institute for Analytical Instrumentation of the Russian Academy of Sciences; *b* — position of the sensor during monitoring of physiological parameters with dimensions indicated by a tape measure.

The heart rate is displayed in a separate information field. This software supports both wireless (Bluetooth) and wired (USB) connections and provides an opportunity to export data in the CSV format for further analysis.

An algorithm based on the analysis of sliding windows with a duration of 10 s and a step of 1 s, which is a fairly common approach, was developed for PPG signal processing. The HR calculation algorithm included the following steps: software vertical mirroring of a 10-second-long section of the 537 nm LED signal, filtering, calculation of the quality index of the filtered signal, identification of peaks, filtering of the peaks, and calculation of the heart rate.

PPG signals are subject to distortion in the form of high-frequency noise, which is associated largely with sensor operation, and various low-frequency noises induced by breathing, movement, and other sources of interference. The main method for suppression of such noise is filtering of the signal frequency response with band filters. These filters suppress signal components outside the passband by a specified number of decibels. A passband commonly used for PPG is 0.5−5 Hz, which corresponds to a heart rate of 30–300 bpm. This was precisely the band that was set in the discussed sensor discretization. Two digital filters were chosen based on the data from [1], where the methods for PPG signal filtering were compared: a fourth-order Butterworth filter and a fourth-order Chebyshev type II filter. The two-pass method (i. e., forward and backward application of the filter) was chosen, since this ensures higher quality of the resulting signal.

However, filtering and subsequent calculation of parameters based on a low-quality signal section may result in large errors in assessment of physiological parameters. A special index (signal quality index, SQI) was introduced

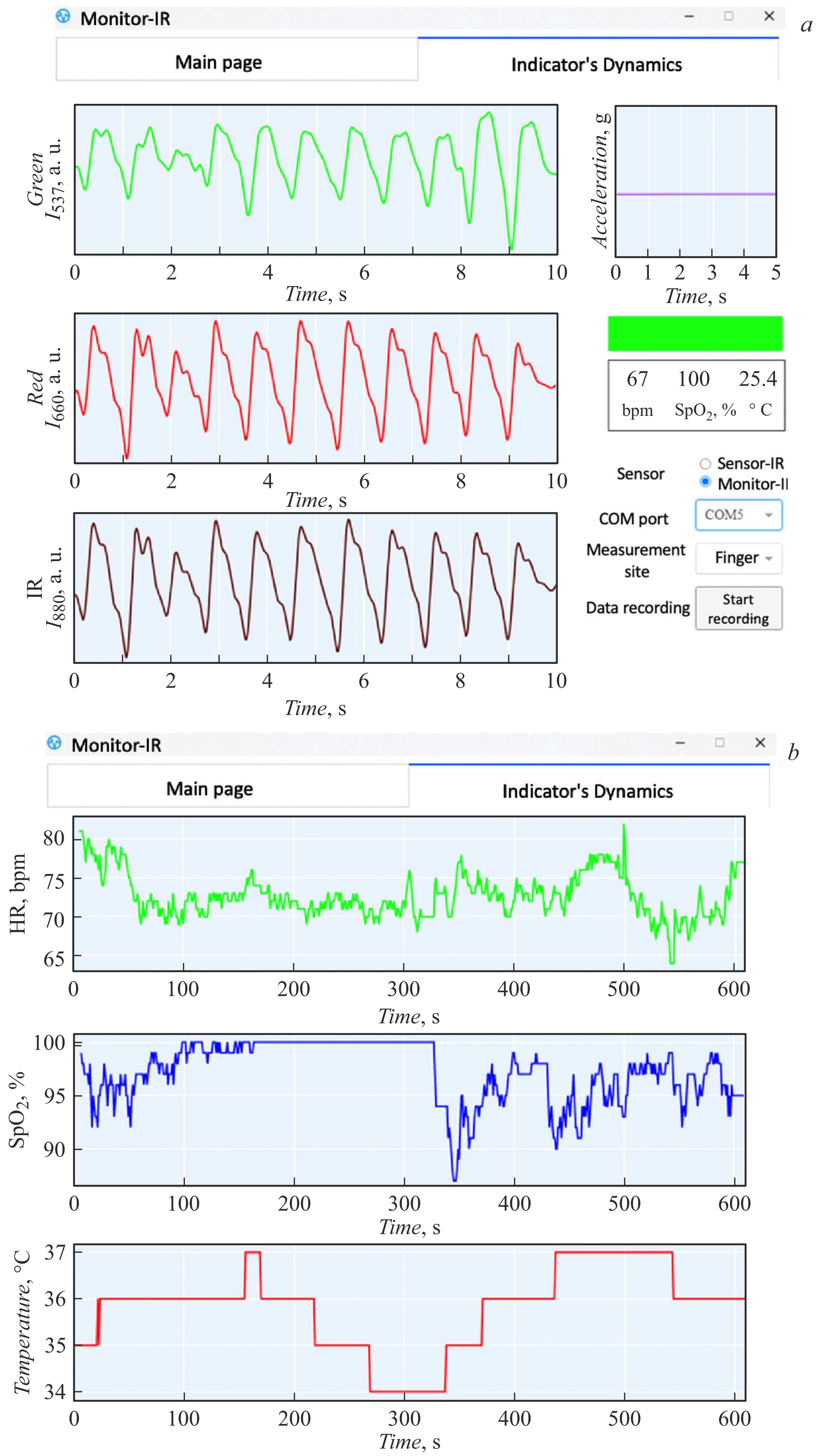

Figure 2. Sensor software. *a* — Main page; *b* — indicator dynamics page.

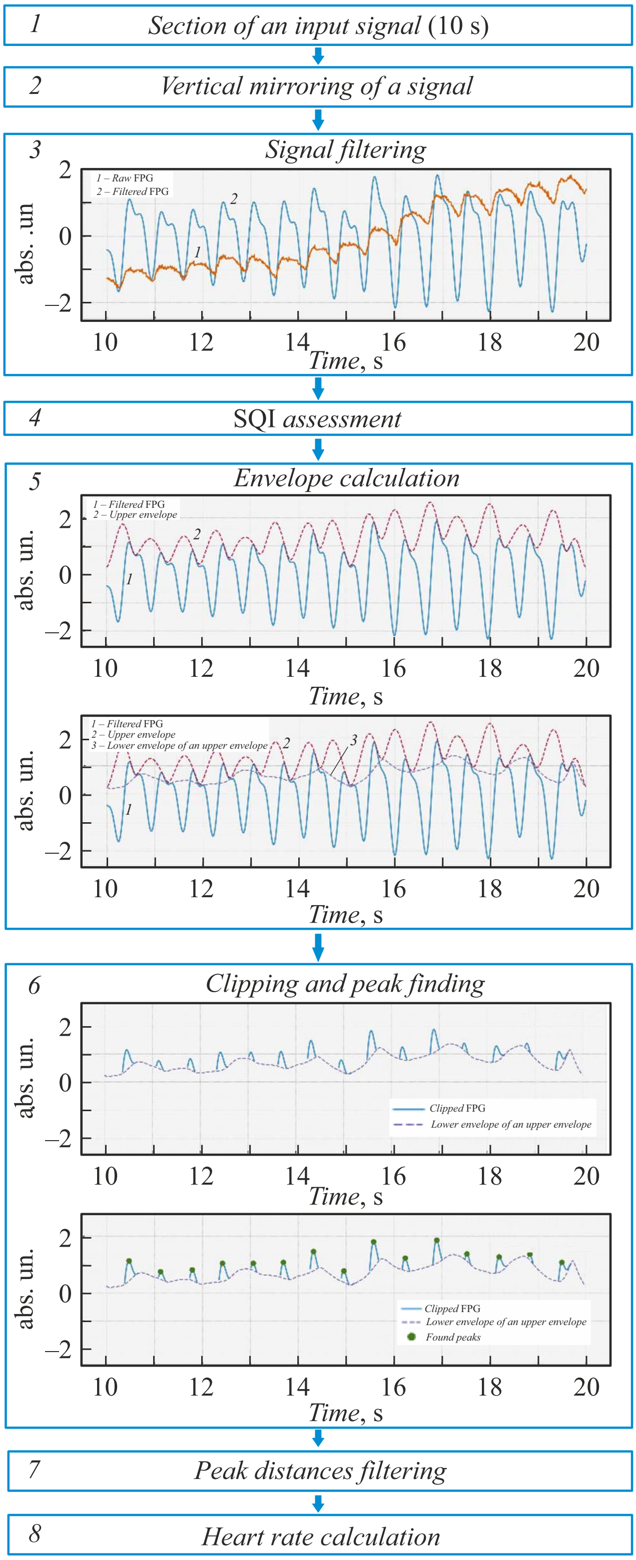

Figure 3. Visual representation of the used HR calculation algorithm. *1* — The input data are 10-s-long signal sections; *2* — vertical mirroring of the signal; *3* — signal filtering; *4* — signal quality assessment; *5* — calculation of envelopes; *6* — signal clipping and identification of peaks; *7* — filtering of distances between peaks; and *8* — HR calculation.

to address this issue [2]. Physiological parameters are not calculated within signal sections marked as low-quality ones. To optimize the operation of the proposed method, relatively undemanding SQIs (asymmetry coefficient, kurtosis, entropy, and relative power) were chosen.

A modification of the method outlined in [3], which is based on clipping the neighborhoods of peaks by the upper envelope of the PPG signal and the lower envelope of the upper envelope (Fig. 3), was used for HR calculation. Only the signal regions located between the two envelopes were considered in the process of peak detection. This approach allows to reduce significantly the number of false peaks found in a signal of imperfect quality without the adjustment of hyperparameters and without the need for excessive computation resources, simplifying and unifying the processing of PPG signals picked up by different sensors from different parts of the body. Several changes were made to the algorithm. First, the lower envelope was shifted by a small amount (1 % of the filtered signal amplitude), which allows for more reliable clipping of regions with potential peaks. Second, the peak search function in clipped regions was altered. It was modified so that the priority of a point as a peak candidate was calculated as the product of the signal value and the negative of the second derivative of the signal at that point. This allows for more accurate peak identification in the case of double peaks, since they may reduce the accuracy of HR determination. It was found that the negative value of the second derivative of the first peak, which corresponds to the systolic peak, exceeds significantly the corresponding value for the second peak, and this fact was used in the development of the algorithm. It should be noted that the peak candidate priority was used only to select a peak among candidates located closer than 1/3 s, which corresponds to 180 bpm.

The distances between successive identified peaks were then calculated. These distances were filtered thoroughly. First, the distances inconsistent with the HR interval of 30–180 bpm were neglected. Second, the distances with an absolute value of the Z-score (distance to 0 after standardization) below 1.96 were neglected. Third, the distances outside of the $(m - b, m + b)$ interval, where m is the average distance, $b = \max(0.25m, 1/3 \cdot f)$, and f is the sampling frequency, were also discarded. If the number of peaks left after filtering was lower than 3 or higher than 30, the corresponding section was considered to be of low quality and the HR value was set as undefined. In a last step, HR was calculated as $60f/m$, where m is the average of the remaining distances.

To evaluate the efficiency of the developed algorithm, it was tested using the open BIDMC database [4] containing 54 PPG signals 480 s in length and the corresponding HR indicators determined with a reference instrument. The method described above was used to calculate the HR values for PPG, and the results were compared with reference HR values. The mean absolute error was calculated for each signal, and the resulting errors were averaged. The fourth-order Butterworth filter and kurtosis turned out to be the best filter–SQI combination with a mean absolute error of HR determination over the entire BIDMC database of 1.4 ± 0.4 bpm.

The obtained test results demonstrated that the proposed method ensures heart rate determination with a processing time below 20 ms on the ESP32S3 platform. This makes it applicable in wearable monitoring devices.

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Compliance with ethical standards

This research did not include any tests requiring approval from an ethical committee.

Conflict of interest

The authors declare that they have no conflict of interest.

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