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Movement Artefact Resistant Photoplethysmographic Probe

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***Abstract*—Photoplethysmogram (*PPG*) is an exceptionally informative diagnostic tool, but the signal registered by wearable devices usually is corrupted, because the probe is affected by the external force and moves with respect to the point of the measurement during the physical activities of the human. This problem is solved usually applying some techniques for the *PPG* extraction from the corrupted (noisy) signal. The authors of this paper are suggesting the solutions, which prevent corruption of *PPG* instead of noise filtering or other *PPG* extraction from corrupted signal techniques. One of such solutions is described in this paper, by proposing special photoplethysmographic signal probe (*PPGP*). This device emits the amplitude modulated light beam with special code sequence, which like Barker code, can be easily recognized using correlation method. The experimental evaluation of proposed method showed, that even in case of a very high noise level SNR=0dB correlation between reference *PPG* and photoplethysmogram got using proposed method is remarkable, reaching corr = 0.985, while standard deviation of error is small and equal** $std(err) = 0.33$ **.**

***Index Terms*—Photoplethysmography, signal to noise ratio, wearable device, Barker code, optical code sequences.**

I. INTRODUCTION

Wearable devices for the observation of vital signs are already cardinally changing situation in the fields of home healthcare and sport's medicine [1]. Non-invasive – non-occlusive measuring methods are needed for the implementation of such devices. Besides, the important criteria include the size of the device, the place and the fastening of the probe, as well as the measurement independence from the physical properties of the user. These criteria are met by the optical method for the measurement of the human body part volume changes, which is called photoplethysmography.

When the light source is emitting a constant flow of the light, the amplitude of the output current of the receiver, which is called photoplethysmography signal or the volume pulse wave, changes proportionally to the volume changes of the artery. This measuring method is non-invasive – non-occlusive, hence it is usually used for the wearable vital signals observation devices.

Volume pulse wave is a product of the interaction between heart and arterial system; hence it is very informative for the evaluation of the state of the heart and the circulatory system. The heart rhythm, the oxygen saturation [2], and the breathing frequency [3] can be measured by analysing photoplethysmography signal. There are few papers proposing methods for the arterial blood pressure measurements using analysis of the photoplethysmography signal.

Because photoplethysmography signal measurement device is wearied 24 hour per day, photoplethysmography signal is affected by the noise. Exceptionally high noise amplitude is observed during physical activities, when probe is moving with respect to the measurement point, because of the external force influence. Although the probe displacement is small and the changed point of measurement almost does not influence the interpretation of the signal, but as a result the movement artefact (noise) amplitude is noticeably higher than signal amplitude. Movement artefact (*MA*) is an exceptional property of the signals, which are registered and processed by the personal wearable devices. In this case, the signal separation from the noise is complicated process, because probe movement frequency band is overlapped with measured signal frequency band. There are two groups of the methods for solving this problem: 1) methods for the minimization of the influence of the movement artefacts; 2) methods for the signal extraction from the signal with movement artefacts.

In the first case special constructions of the probes are proposed [4], which limit the noise influence on the measured signal or composite probes are used [5], where one of the probes registers signal with noise and other probe registers movement signal. In the second case it is suggested to use non-linear digital filters, which are based on the neural networks [6], support vector machines [7], linear and non-linear forecasting, blind deconvolution, etc.

Application of composite probes and non-linear methods for the signal extraction from the signal with movement artefacts is not suitable for the wearable systems, because the energy resources as well as computational resources are limited in these systems. The use of composite probes requires additional energy consumption. Having in mind, that energy consumption is directly proportional to the processing volume and speed, the complex signal processing methods are not feasible for such kind of systems.

The authors of this paper propose innovative photoplethysmographic probe, which reduces the noise

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influence to the *PPG*. The probe consists of the light emitter, which generates amplitude modulated optical code sequence, and light receiver.

II. Optical Code Sequence

The sequence of values $\{x_0, x_1, \ldots, x_i, \ldots, x_{N-1}\}$, where $x_i = \pm 1$, satisfying condition

$$\circ(k) = \sum_{i=0}^{N-1-k} x_i x_{i+k} = \begin{cases} N, & \text{when } k = 0, \\ 0 \text{ or } \pm 1, & \text{when } k = 1,2,\ldots,N\text{-}1, \end{cases} \quad (1)$$

is called Barker binary code sequence. There are few known sequences [8], [9], which satisfy (1) condition with lengths $N = 1, 2, 3, 4, 5, 7, 11, 13$. These code sequences are used to detect the signal in the noise. Barker code sequences differ by the ratio

$$D = \circ(0) / \circ(k)_{\max}. \quad (2)$$

Autocorrelation function peak and maximum value $\circ(k)_{max}$ ratio always is equal to the length of the Barker code sequence $N = D$. It is not suitable to directly change Barker code values from $x_i = \{1, -1\}$ to $x_i = \{1, 0\}$ when using optical signal for the transmission of information. Hence, the new code sequence is required, which consists of $x_i = \{1, 0\}$. The optical code sequence, generated according to the method proposed in [10], is used in this paper. Suppose sequence $\{x_0, x_1, \ldots, x_{m-1}\}$, where values are $x_i = \{1, 0\}$. Value *"1"* means activated light source, while value "0" means deactivated light source. Because the first and the last values of the sequence must satisfy condition $x_0 = x_{m-1} = 1$, then adding zero values to the left and the right side of the sequence do not have sense, because of this the values of the autocorrelation function do not change. Say u_i is the difference between $(i+1)$–th and (i)–th bit values. Then it can be written

$$u_1 + u_2 + \ldots + u_{n-1} = m - 1, \quad (3)$$

where n – the number of *"1"* values in the sequence. It is obvious, that autocorrelation function peak value $\circ(0) = n$ for such sequence (3). All elements of the triangular matrix

$$M = \begin{bmatrix} S_{11} & S_{12} & \ldots & \ldots & \ldots & S_{1(n-1)} \\ & S_{22} & S_{23} & \ldots & \ldots & S_{2(n-1)} \\ & & S_{33} & \ldots & \ldots & S_{3(n-1)} \\ & & & \ldots & \ldots & \ldots \\ & & & S_{(n-2)(n-2)} & & S_{(n-2)(n-1)} \\ & & & & & S_{(n-1)(n-1)} \end{bmatrix}, \quad (4)$$

are different integer numbers from the interval $(1, 2, \ldots, m-1)$, where m is the length of the code sequence. Values (S_{ij}) satisfy equation

$$\begin{vmatrix} S_{i,i} = u_i, \\ S_{i,j} = u_i + u_{i+1} + \ldots + u_j, \quad \text{where} \quad j > i. \end{vmatrix} \quad (5)$$

The values of the autocorrelation function for the code sequence generated in such way will be less than *"1"*, except case, when shift value $k = 0$

$$\circ(k) = \begin{cases} n, & k = 0, \\ 0, 1 & k = 1, 2, \ldots \text{ m-1}. \end{cases} \quad (6)$$

Suppose $u_1 = 5, u_2 = 3, u_3 = 4, u_4 = 2$. All elements of the matrix

$$M = \begin{bmatrix} 2 & 6 & 9 & 14 \\ & 4 & 7 & 12 \\ & & 3 & 8 \\ & & & 5 \end{bmatrix}, \quad (7)$$

are different, while code sequence $B_0(15,5) = 110000001110001$. This code sequence is suitable for the modulation of the light emitter (source).

The produced code sequence has parameter value $D = 5$, when $N = 15$. This sequence is much longer than Barker code sequence to achieve the same parameter D value and to obtain code sequence suitable for the optical modulation.

III. The Principles of the Movement Artefact Resistant Photoplethysmographic Probe

The proposed photoplethysmographic probe Fig. 1 consists of the light source and the light receiver.

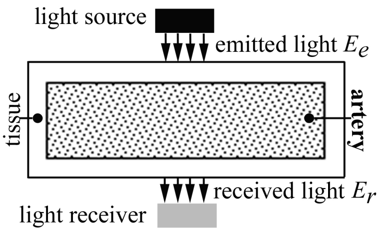

Fig. 1. The structure of the proposed photoplethysmographic probe.

The sequence of the impulses generated by the light source is produced in three stages. During the first stage, as showed in Fig. 2, the discretization period T_d is selected together with impulse width T_w. During the second stage, each discretization impulse is divided into the sequence of K impulses with increasing amplitude Fig. 3.

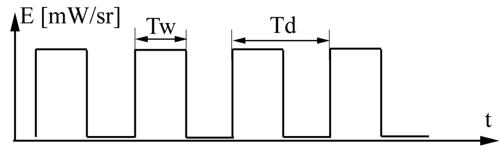

Fig. 2. The code sequence emitted by the probe, where E – radiant intensity, T_w – impulse width, T_d – impulse period.

Value K defines the number of quantization levels of the signal as well, because the range of the energy emitted towards the receiver $E_{min} \le E \le E_{max}$ is divided into K parts

$\Delta = (E_{max} - E_{min})/K$, while *i-th* impulse amplitude is equal $E[i] = i \cdot \Delta$, where $1 \le i \le K$. This is analogous to the *b* bits analog-digital converter, where $K = 2^b$.

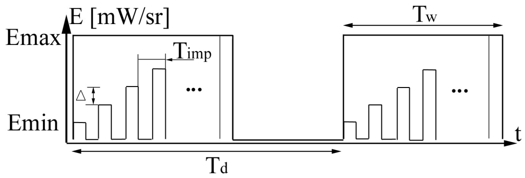

Fig. 3. The impulses with amplitude *E[i]* and period T_{imp} generated by the probe emitter.

In the beginning of the cardio cycle, when the volume of the artery is minimal, all *K* light impulses will pass the artery. The impulses are counted at the receiver. In this manner the first value of the pulse wave is obtained an equal to volume pulse wave *VPW[I = 1] = K*. Next to the last value of the discrete *PW* (pulse wave) is equal to *VPW[i] = K - 1* and will be obtained when receiver will react to the *K - 1* impulses. In the case of the maximum artery volume change, the receiver will react only to the last (maximum amplitude) impulse. When all light signals are absorbed, the value of discrete *VPW* is equal to *VPW[i] = 0*.

Another method can be used for the measurement of the *PW* amplitude. Because in the segment T_w the period of impulses T_{imp} is constant, then *PW* amplitude can be expressed as the delay between the beginning of T_w and the first impulse detected in the receiver. In this case *amplitude-time* converter is got.

In both cases photoplethysmographic probe sensitivity depends on the number *K* of quantization levels. The number of the impulses, which can be sent during the period T_w, depends on the performance of the light emitter and the light receiver. The proposed method implements *amplitude-time* converter. The quantization error of the converter is equal to the quantization step Δ, because value *K* which corresponds to the minimal PW value is got, when all light impulses pass the volume of the artery including the minimal amplitude E_{min} impulse.

In order to ensure photoplethysmographic probe noise resistance, each impulse T_{imp} is coded using optical code sequence B_0. It consists of *K* code sequences with *K* different amplitudes, which are emitted during period T_w . In order to detect the code sequence in the noisy signal correlation between B_0 and received signal is calculated. Analog correlation calculation device is used for this purpose [11]. The number of correlation function peaks in the output of this device corresponds to the *VPW* amplitude.

IV. MODELLING AND RESULTS

Modelling in the Matlab environment was performed in order to evaluate effectiveness of the proposed method. The volume pulse wave, which is registered by the photoplethysmographic device, occurs as a result of the pressure, hence PPG signal and pressure forms are analogous. Because of this reason, modified four elements Windkessel model [12] was used for the generation of *PPG* signal. This model connects the blood debit $Q(t)$ and the arterial pressure $p(t)$ Fig. 4.

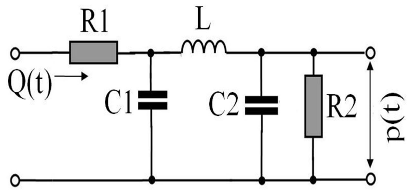

Fig. 4. The modified four elements Windkessel model connecting blood debit and arterial pressure.

The blood debit change is described as flow generator function [13]

$$Q(t) = \frac{q_0 \cdot t}{\tau^2} \exp\left(-t^2 / 2\tau^2\right), \tag{8}$$

where $0 \le t \le T$, q_0 – is the cardiac output and τ – is the time at which the maximal cardiac output is reached, *T* – cardio cycle period (heart contraction period). Model presented in Fig. 4 can be described as equation

$$P(t) = a_2 \exp(-a_3 t) + a_4 \exp(-a_5 t) \cos(a_6 t - a_7). \tag{9}$$

Because the signal form is important for the performed research, $P(t)$ values were normalised so, that amplitude values would satisfy condition $0 \le A_{sig} \le 1$.

The following parameters were selected for the modelling of optical signals produced by the light source: sampling period $T_d = 10^{-3} s$, the number of the code sequences in the sampling period $K = 1000$, the values of the code sequence amplitudes from the interval $0 \le E \le 1$ with step $\Delta = 1/K$. Since the length of the code sequence is known, both the moment of correlation function peak appearance τ and amplitude values $corr_{max}(i), 1 \le i \le K$ (when there is no noise) are also known. The values of the output of the light receiver $\overline{PPG}$ are calculated by adding *PPG* signal and $0 \le A_{noise} \le 1$ amplitude noise

$$\overline{PPG} = \mathrm{PPG} + \mathrm{A_{noise}} \cdot noise. \tag{10}$$

The realizations of the noise sequences are taken from www.physionet.org database. This database contains the random realizations of surrogate stationary time series with varying degrees of correlations, trends of several types, and various types of nonstationarities.

The algorithm for the detection of the *PPG* signal amplitude consists of few steps:

1. The correlation between light receiver output signal $\overline{PPG}$ and the reference signal B_0. (B_0 duration is equal to $15 \cdot T_{imp}$) is calculated;

2. The condition is checked: *If* $corr(i,\tau) \ge corr_{max}(i)$ *then cnt = cnt+1; otherwise cnt =0;* value *cnt* means *PPG* amplitude at the discrete time moment $t = i \cdot Td$.

The maximum possible value of $\overline{PPG}$ signal is *K*, while *PPG* is equal to one, hence in order to compare $\overline{PPG}$ and

PPG signals the $\overline{PPG}$ signal values were divided by *K*.

In order to evaluate the proposed method effectiveness the error (11) and the error standard deviation *std(err)* was calculated (Fig. 5)

$$err = PPG - \overline{PPG}, \tag{11}$$

as well as the level of the noise Fig. 6(a) in decibels using signal to noise ratio (*SNR*) (12)

$$SNR = 20\log_{10}\left(A_{sig} / A_{noise}\right), \tag{12}$$

where A_{sig} – is the *PPG* amplitude and A_{noise} – is the noise amplitude.

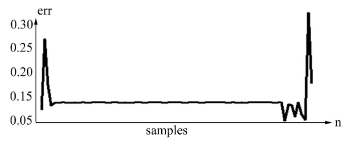

Fig. 5. Error *err* values, when *SNR = 0 dB.*

Medics when interpreting physiological signals (ECG, EEG, EMG) are analysing these signals by searching distinctive patterns, which characterize the state of the organism. This shows, that the major part of information is carried by the form of the signal, hence correlation values were calculated in order to evaluate the proposed photoplethysmographic probe effectiveness in the case of noisy signal Fig. 6(b).

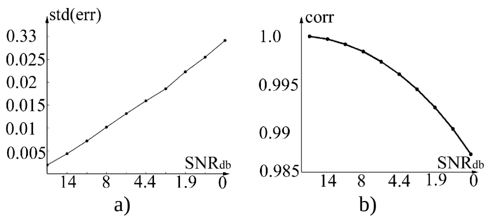

Fig. 6. Standard deviation of error *std(err)* (a), and correlation between *PPG* and $\overline{PPG}$ values *corr* (b), for the different levels of noise.

The values of the output signals produced by commonly used *PPGP* devices Fig. 7(a) and the proposed *PPGP* device Fig. 7(b) when *SNR = 0 dB* are given in Fig. 7.

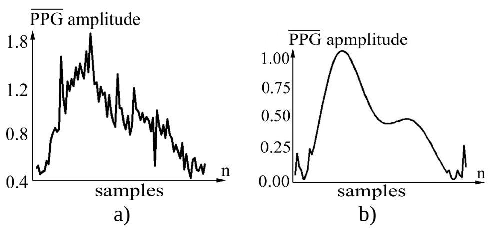

Fig. 7. Common *PPG* probe (a), and proposed *PPG* probe (b), output values, when noise level *SNR = 0dB.*

Measuring signal using proposed method, the output signal preserves the distinctive form of initial signal even in case when signal and the noise levels are equal.

V. Conclusions

The results of the modeling show, that when applying the proposed method, the noise affects the different parts of the PPG signal unequally. The signal parts with smallest amplitude are affected more and give biggest input to the error err deviation as well as affect correlation Fig. 6(a), Fig. 6(b). The proposed method is comparable with other traditional methods, when is used to calculate such diagnostic indexes like pulse transit time or pulse wave velocity, because in order to find them, it is needed to find exact *PPG* signal starting point. During evaluation of the state of circulatory system or diagnosis of some diseases like II type diabetes, Windkessel model Fig. 4 parameter values (9) are calculated by comparing the measured *PPG* signals rear front form comparison with rear front generated by model. As modelling results show Fig. 7, *PPGP* which generates optical amplitude modulated code sequence, preserves the forms of the *PPG* signals fore and rear fronts. The noise influence to the measured signal is remarkably smaller using proposed *PPGP* than using traditional *PPGP* devices. Hence, the proposed method is suitable for the implementation of wearable circulatory system observation devices, because it noticeably improves the quality of the processed signal as well as the reliability of calculations.

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