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PROCEDURE FOR CORRECTION OF THE ECG SIGNAL ERROR INTRODUCED BY SKIN-ELECTRODE INTERFACE

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Abstract

The paper presents a procedure for correction of the error of an ECG signal, introduced by the skin-electrode interface. This procedure involves three main measuring-calculating stages: parametrical identification of the mathematical model of the interface, realized directly before the diagnostic measurements, registration of the signal at the output of electrodes as well as reconstruction of the input signal of the interface.
The first two stages are realized in the on-line mode, whereas the operation of signal reconstruction presents a numerical task of digital signal processing and is realized in the off-line mode through deconvolution of the registered signal with the transfer function of the skin-electrode interface.
The aim of the paper is to discuss in detail the procedure of parametric identification of the skin-electrode interface with the use of a computer system equipped with a DAQ card and LabVIEW software. The algorithm for error correction introduced by this interface is also presented.

Keywords: Identification of skin-electrode interface, signals reconstruction.

1. Introduction

Disorders of the circulatory system still present a serious medical problem. Appropriate prophylaxis and early diagnosis of heart rhythm disorders are classified as the most important challenges of modern cardiology. Effective diagnosing of atrial and ventricular arrhythmia and precise assessment of heart damage, being a result of the illnesses, are possible thanks to very accurate ECG measurements.

In the systems intended for ECG registration, analogically to the majority of systems dedicated for measurements of quantities variable in time, particular attention should be paid to measuring sensors. This is because, in comparison with other components of the system (amplifiers, filters, A/C converter), they are usually a source of significant errors. In ECG measurements, the error is introduced by the electrode and also by the skin-electrode interface. Depending on individual derma-galvanic reactions of the patient, the interface has different dynamic properties which have a significant effect on the accuracy of the registered signals. As a result, such divergence in measurement accuracy can hinder a fast and suitable diagnosis [1-3].

In the present paper a procedure which allows identification of the skin-electrode interface directly before taking diagnostic measurements is proposed. This task includes only parametric identification, because the non-parametric model of the interface is commonly known and is the same for every patient [4, 5].
Fig. 1 presents a block diagram of the procedure for skin-electrode interface identification.

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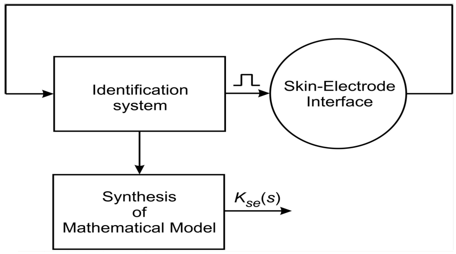

Fig. 1. Block diagram of procedure for identification of skin-electrode interface
$K_{se}(s)$ - transmittance of skin-electrode interface.

In the identification process, the square-wave stimulating signal connected between the diagnostic electrodes is applied [2, 6]. The maximum value of this signal must be adjusted in such a way that the specific value of the current flowing between the electrode is not exceeded. The root mean square value of the current equals $100\,\mu\text{A}$. The result of the identification process is presented by the transmittance $K_{se}(s)$, which can be easily transformed to another form, e.g. to the state equation.

The algorithm for correction of the error of the signals $y(t)$ registered at the interface output is presented as well. This algorithm concerns reconstruction of the signal $x(t)$ generated by the heart muscle, which is the input signal of the skin-electrode interface.

Fig. 2 presents a block diagram for correction of the error of the signals registered on the interface output.

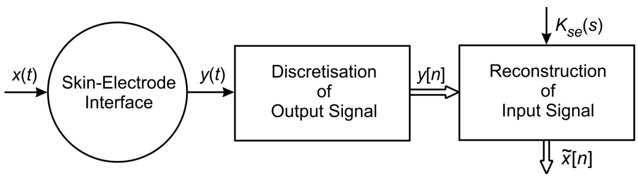

Fig. 2. Block diagram for correction of the error of the signal registered at the interface output
$x(t)$ - input signal of skin-electrode interface, $y(t)$ - output signal of interface, $y[n]$ - discretised output signal, $\tilde{x}[n]$ - reconstructed input signal.

The error of such signals depends on the difference between signals $y(t)$ and $x(t)$. This error is minimized by determination of signal $\tilde{x}(n)$, which presents an estimation of the discretised input signal $x(n)$.

2. Electrical Equivalent Circuit of Skin-Electrode Interface

Potential differences, resulting from the circulation of the electromotive force of the heart muscle, flow through the tissues to the skin surface. Then they are transferred via electrodes to the measuring system and later registered in the form of the ECG diagram.

The structure of the electrical equivalent circuit of the skin-electrode interface was proposed in [4, 5] and there it was proved that the high accuracy of the representation of the dynamic properties of such interface has the double time constant model. This model consists of series-parallel connections of resistors and capacitors.

The skin-electrode interface and its electrical equivalent circuit as the double time constant model are presented in Fig. 3.

The electrical equivalent circuit includes the input resistance R_i of the measurement amplifier, as the exterior resistance connected to the interface terminal. In the equation describing the transmittance of the skin-electrode interface consideration, resistance R_i is necessary.

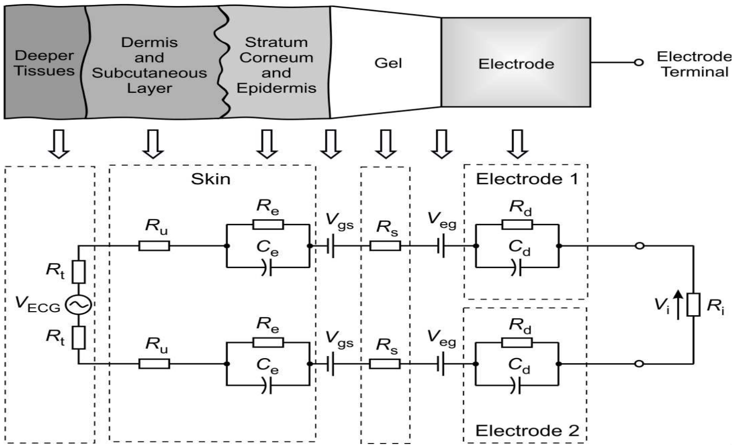

Fig. 3. Skin-electrode interface and its electrical equivalent circuit.

In Fig. 3 the following notations are used: V_{eg} - half cell potential of electrode-gel interface, V_{gs} - half cell potential of gel-skin interface, V_{ECG} - ECG potential, R_t - tissues resistance, R_u - dermis subcutaneous resistance, (R_e, C_e)- outer layer of skin dielectric properties, R_s - gel series resistance, (R_d, C_d)- electrical double layer of electrode-gel interface, R_i- input resistance of the measurement amplifier.

The transmittance of the system shown in Fig. 3 is

$$K_{se}(s) = \frac{V_i(s)}{V_{Ecg}(s)} = R_i \frac{\tau_{ed}s^2 + \tau_{de}s + 1}{as^2 + bs + c} = \frac{R_i \tau_{ed}}{a} + R_i \frac{\left(\tau_{de} - \frac{b}{a}\tau_{ed}\right)s + \left(1 - \frac{c}{a}\tau_{ed}\right)}{as^2 + bs + c},$$

where: (1)

$$\tau_{ed} = \tau_e \tau_d,\ \tau_{de} = \tau_e + \tau_d,\ \tau_e = R_e C_e,\ \tau_d = R_d C_d,\ a = R_p \tau_{ed},$$
$$b = R_p \tau_{de} + 2(R_e \tau_d + R_d \tau_e),\ c = R_p + 2(R_e + R_d),\ R_p = R_i + 2(R_t + R_u + R_s).$$

It is convenient to present it in the form of a state equation

$$K_{se}(s) = \mathbf{C}[s\mathbf{I} - \mathbf{A}]^{-1}\mathbf{B} + \mathbf{D}, \tag{2}$$

where: **A**, **B**, **C**, **D** are real matrices of corresponding dimensions, and in order to reconstruct the skin-electrode interface input signal they should be presented in the observer canonical form presented by (3) [10, 11].

$$\mathbf{A}=\begin{bmatrix}-\frac{b}{a} & 1\\ -\frac{c}{a} & 0\end{bmatrix},\ \mathbf{B}=\begin{bmatrix}\frac{R_{in}}{a}\left(\tau_{\mathrm{de}}-\frac{b}{a}\tau_{\mathrm{ed}}\right)\\ \frac{R_{in}}{a}\left(1-\frac{c}{a}\tau_{\mathrm{ed}}\right)\end{bmatrix},\ \mathbf{C}=\begin{bmatrix}1 & 0\end{bmatrix},\ \mathbf{D}=\left[\frac{R_{in}\tau_{\mathrm{ed}}}{a}\right]. \tag{3}$$

It can be easily checked that the frequency characteristics of the skin-electrode interface are described by the following equations:

$$A_{\mathrm{se}}(\omega)=R_i\,\frac{\sqrt{\left[\omega^4 a\tau_{\mathrm{de}}+\omega^2(b\tau_{\mathrm{de}}-c\tau_{\mathrm{ed}}-a)+c\right]^2+\left[\omega^3(b\tau_{\mathrm{ed}}-a\tau_{\mathrm{de}})+\omega(c\tau_{\mathrm{de}}-b)\right]^2}}{\omega^4(a^2-2ac)+\omega^2b^2+c^2}, \tag{4}$$

$$\varphi_{\mathrm{se}}(\omega)=\mathrm{atan}\left[\frac{\omega^3(b\tau_{\mathrm{ed}}-a\tau_{\mathrm{de}})+\omega(c\tau_{\mathrm{de}}-b)}{\omega^4 a\tau_{\mathrm{de}}+\omega^2(b\tau_{\mathrm{de}}-c\tau_{\mathrm{ed}}-a)+c}\right]. \tag{5}$$

Examples of the determined parameters of model (1) for a representative group of patients of various ages are presented in [1, 6], while the frequency characteristics for various types of electrodes (Ag/AgCl, Ag/AgCl gel, dry Ag, Ag-SiO_2), as test results for patients with hairy and non-hairy arms are presented in [6]. Depending on the values of the parameters of the electrical equivalent circuit these characteristics assume various shapes, often significantly different from the shapes of the characteristics of non-distortion systems. In this case the non-distortion system should be comprehended as a system which has constancy of the amplitude characteristic and phase linearity in the range of the processing frequencies. In [1, 6] it was assumed that in the range of frequencies of ECG signals, the characteristics (4-5) can differ significantly from the characteristics of a non-distortion system.

Consequently, on the basis of (4-5) it can be confirmed that for different values of the parameters of model (1), different values of ECG signals error are obtained. In order to correct this error, it is possible to use specialized computer techniques that enable:
- carrying out the skin-electrode interface identification directly before ECG registration,
- realization of numerical processing of the registered ECG signal in order to reconstruct the signal at the interface input.

According to the requirements of the European Standards and American Heart Association [7, 8], the measurement amplifier should not introduce a greater phase shift than the linear single-pole filter with the time constant equal to $0.05\,\mathrm{Hz}$ as well as transmittance and phase-frequency characteristic described by means of the following equations:

$$K_{\mathrm{sp}}=\frac{s}{s+\frac{1}{\tau}},\ \text{where: } \tau=RC, \tag{6}$$

$$\varphi_{\mathrm{sp}}(\omega)=\mathrm{atan}(\omega_{\mathrm{c}}/\omega),\ \text{where: } \omega_{\mathrm{c}}=0{,}1\pi. \tag{7}$$

These requirements can be satisfied by an appropriate selection of the amplifier input resistance.

The $\varphi_{\mathrm{se}}(\omega)<\varphi_{\mathrm{sp}}(\omega)$ condition is fulfilled when

$$R_{\mathrm{i}}>\frac{\omega^2\tau_{\mathrm{ed}}[2(R_{\mathrm{e}}\tau_{\mathrm{d}}+R_{\mathrm{d}}\tau_{\mathrm{e}})-\omega_{\mathrm{c}}a]+\omega_{\mathrm{c}}\tau_{\mathrm{ed}}(R_{\mathrm{p}}+c)-\tau_{\mathrm{de}}(b-c)-b}{\omega^2\omega_{\mathrm{c}}\tau_{\mathrm{ed}}{}^2-\omega_{\mathrm{c}}\tau_{\mathrm{ed}}+\tau_{\mathrm{de}}(\tau_{\mathrm{de}}+1)}, \tag{8}$$

where: $\varphi_{se}(\omega)$, $\varphi_{sp}(\omega)$ are phase shifts introduced by the skin-electrode interface and the linear single-pole filter respectively.

From (6-8) it follows that the measurement amplifier resistance depends on the values of the skin-electrode interface parameters [1] and must always fulfil condition (8).

3. Identification of Skin-Electrode Interface

In contemporary measurement theory and practice, identification methods are based on discrete parametric models and computer systems equipped with a data acquisition card and dedicated control-measuring software.

The commonly known parametric models, such as: AR, ARX, ARMAX, Box-Jenkins, output-error and state space, differ significantly in the way the effect of distortions on the functioning of the system represented by those models is approached. The last model has the best properties in the context of realization of the reconstruction algorithms. In comparison with all the other methods, it does not require determination of the initial values of the model parameters and time delay introduced by the identified system.

The computer-aided measurement system for skin-electrode interface identification is presented in Fig. 4.

The computer was equipped with a data acquisition (DAQ) card as well as LabVIEW measurement-control software with the system identification toolkit.

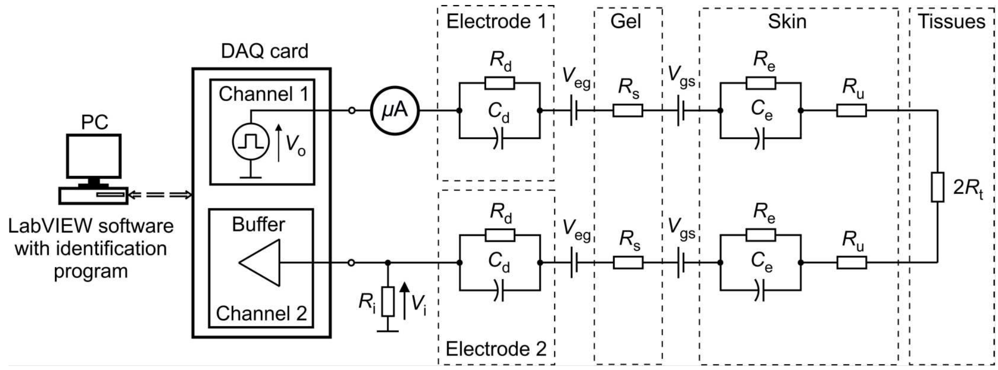

Fig. 4. System for identification of the skin-electrode interface.

The skin-electrode interface is excited by means of square-wave voltage signal V_0 which presents the output signal of the DAQ card, while the voltage signal V_i is the input signal of this card. It can be easily proved that relation $V_i(s)/V_{Ecg}(s)$ presented by means of (1) and resulting from Fig. 3 equals relation $V_i(s)/V_o(s)$ resulting from Fig. 4.

Fig. 5 presents a block diagram of the program for identification of skin-electrode interface.

The input parameters for the square function generator are: amplitude, frequency, sampling rate, while for the block of the transfer function model estimation the input parameters are: orders of numerator and denumerator of the transfer function.

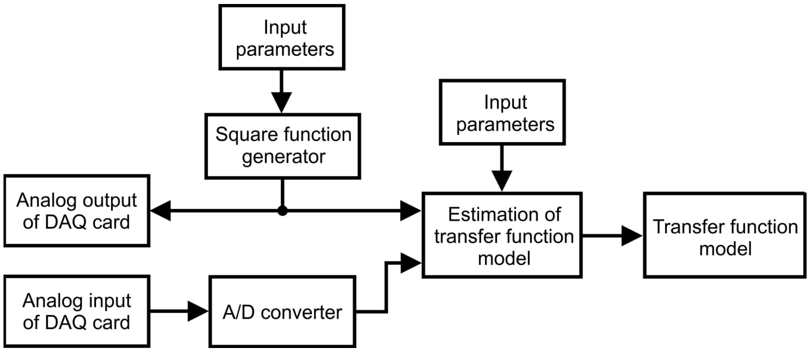

Fig. 5. Block diagram of program for skin-electrode interface identification.

Figs. 6 and 7 present, respectively: the program for skin-electrode interface identification implemented in the LabVIEW software, corresponding with the block diagram presented in Fig. 5, as well as the wave form of the signal across the skin-electrode interface circuit in response to the square stimulus signal. This wave form was obtained as a result of the identification process of skin-electrode interface for an exemplary person.

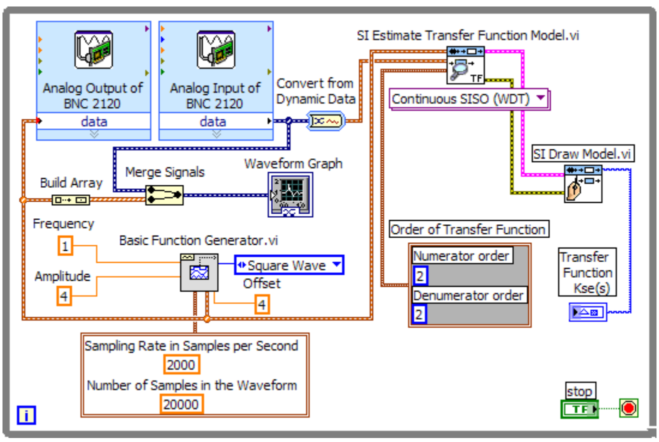

Fig. 6. Program for identification of skin-electrode interface implemented in LabVIEW.

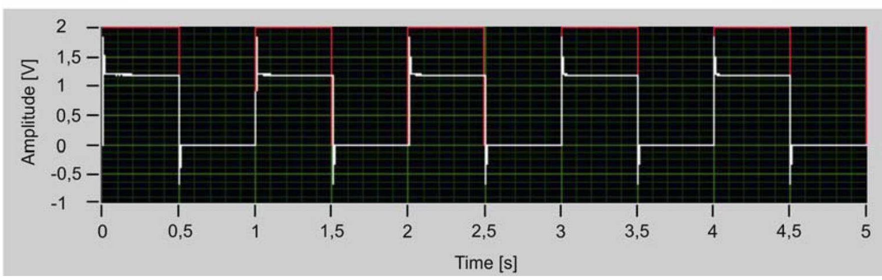

Fig. 7. Wave form of signal across the skin-electrode interface circuit in response to a square stimulus signal.

4. Signals Registration at the Skin-Electrode Interface Output

Fig. 8 presents a system for signals registration at the skin-electrode interface output. The system includes: three electrodes, a measurement amplifier as well as a computer equipped with a DAQ card and LabVIEW software. The two electrodes present the input of the skin-electrode interface while the electrode connected to the right leg presents the reference point of the measurement system. The output signal of the skin-electrode interface is acquired by means of the DAQ card as well as LabVIEW software.

It is assumed that the measurement amplifier does not introduce distortion of the registered ECG signal because of the wide range of the band transferring the frequencies it measures.

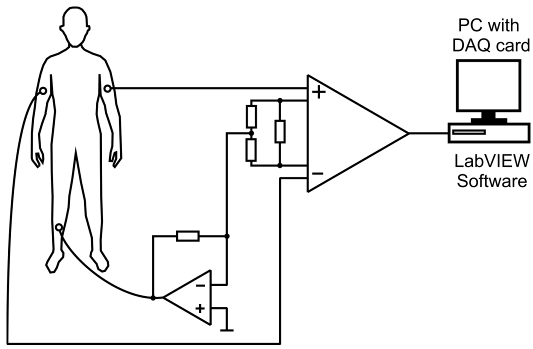

Fig. 8. System for signals registration at the skin-electrode interface output.

Fig. 9 presents the signal registered at the skin-electrode interface output in time T equal 2 s. (for sampling, the rate equals 395 samples per second).

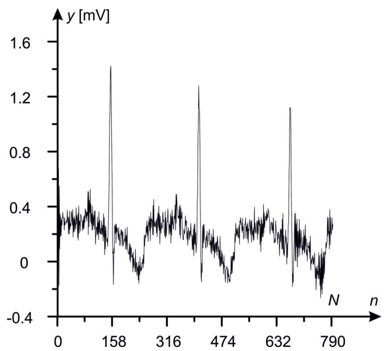

Fig. 9. Signal registered on skin-electrode interface output.

5. Reconstruction of Input Signals of the Skin-Electrode Interface

The model of the state space has the form:

$$y(n+1) = \mathbf{\Phi}\, y(n) + \mathbf{\Psi}\, x(n), \quad n = 0,1,\ldots,N-1, \; N = T/\Delta,$$

Δ - sample interval of continuous time T, $y(\cdot)$ - state vector, $y(\cdot) \in \Re^2$, (9)

$x(\cdot)$ - input (control) vector, $x(\cdot) \in \Re^2$, where:

$$\Phi = e^{\mathbf{A}\Delta}, \quad \Psi = \int_0^{\Delta} e^{\mathbf{A}t} dt\, \mathbf{B} = \mathbf{A}^{-1}(e^{\mathbf{A}\Delta} - \mathbf{I})\mathbf{B}, \quad \mathbf{A}\text{ - state matrix, } \mathbf{B}\text{ - input (control)} \tag{10}$$

matrix, $\mathbf{I}$ - unit matrix.

Writing the full form of equation (9) we obtain

$$\begin{bmatrix} y(n+1) \\ y_2(n+1) \end{bmatrix} = \begin{bmatrix} \varphi_{1,1} & \varphi_{1,2} \\ \varphi_{2,1} & \varphi_{2,2} \end{bmatrix} \begin{bmatrix} y(n) \\ y_2(n) \end{bmatrix} + \begin{bmatrix} \psi_1 \\ \psi_2 \end{bmatrix} x(n), \; n = 0, 1, ..., N-1, \tag{11}$$

where $y(0) = 0$ - state variable measured directly.

Solving the above system of equations with respect to the input quantity $x(n)$, the following system of recurrent equations is obtained

$$\tilde{x}(n) = \frac{1}{\psi_1}\left[y(n+1) - \varphi_{1,1} y(n) - \varphi_{1,2} y_2(n)\right], \; n = 0, 1, ..., N-1, \tag{12}$$

where:

$$y_2(n+1) = \psi_2 \tilde{x}(n) + \varphi_{2,1} y(n) + \varphi_{2,2} y_2(n), \;\; y_2(0) = 0 \tag{13}$$

and $\tilde{x}(n)$ is the estimation of the input quantity at time instant n, determined on the basis of the output quantity $y(n)$ at time instants n and $n + 1$ as well as stored in memory state variable $y_2(n)$ from previous time instant [9].

The above equations present the algorithm for the reconstruction of the input signal of the skin-electrode interface.

In order to carry out the calculations according to (13), it is necessary to know the initial value of the state variable $y_2(n)$. This value can be estimated or assumed as zero.

6. Example of Application

The transmittance generated by the block *Transfer Function* of identification program presented in Fig. 6 equals

$$y(s) = \frac{-269 \cdot 10^{-6} s^2 + 0.12 s + 0.58}{247 \cdot 10^{-6} s^2 + 0.20 s + 1}. \tag{14}$$

According to (3) the obtained transmittance can be presented in the observer canonical form

$$\mathbf{A} = \begin{bmatrix} -722.61 & 1 \\ -3571.43 & 0 \end{bmatrix}, \; \mathbf{B} = \begin{bmatrix} 1118.51 \\ -9453.51 \end{bmatrix}, \; \mathbf{C} = \begin{bmatrix} 1 & 0 \end{bmatrix}, \; \mathbf{D} = \begin{bmatrix} -0.96 \end{bmatrix}. \tag{15}$$

On the basis of (15) as well as the application of recurrent equation (12-13) presented in point 5, the operation of signal numerical reconstruction presented in Fig. 2 was carried out. The calculation was performed using the computer program implemented in MatLab software.

Fig. 10 presents signal $\tilde{x}(n)$ reconstructed at the skin-electrode interface input and scaled in reference to the gain of the measurement amplifier so that it could be compared with signal $y(n)$.

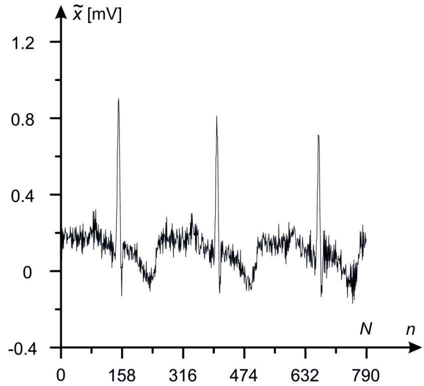

Fig. 10. Signal $\tilde{x}(n)$ reconstructed on skin-electrode interface input.

The difference between signals $\tilde{x}(n)$ and $y(n)$ is presented in Fig. 11.

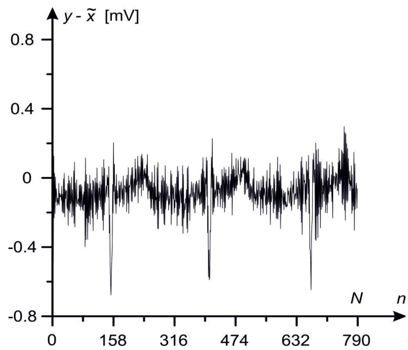

Fig. 11. Difference between signals $y(n)$ and $\tilde{x}(n)$.

6. Conclusions

The procedure of correcting the error of the ECG signal by the reconstruction of the skin-electrode interface input presented in the paper can be used for the reduction of the error introduced by the entire circuit for processing of those signals. For this purpose, apart from the skin-electrode interface identification, a mathematical model of the electrocardiograph system must be developed. While the synthesis of the electrocardiograph model can only be done once, and at any time, the synthesis of the skin-electrode interface model must be done each time, before ECG measurements. For the models obtained in this way, the procedure of reconstruction the registered ECG signal must be followed. Then the operation of mathematical convolution of the reconstructed signal with the discretised impulse response of the ideal band-pass filter must be carried out. The pass-band of this filter is relevant to the frequency band of the ECG signals, i.e. 0.05-100 Hz.

The signal obtained in this way can be further numerically processed, e.g. through wavelet transformation, in order to reduce ECG signal distortion, independent of the dynamic properties of the processing circuit.

The accuracy of error correction of the ECG signal is not considered in this paper. This important problem will constitute the basis for the research works to follow.

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