FEATURES OF VOLTAGE PULSE PLETHYSMOGRAPHY

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ABSTRACT
Measurement of tissue volume changes by voltage pulse plethysmography is being studied. In voltage pulse plethysmography short voltage pulses activate the bio-tissue and the characteristics of the transient process are measured. There are two possible implementations: with grounded bio-tissue and with grounded current sampling resistor. Measurement was carried out for both cases and the obtained wave-shapes for blood pulsations in the arteries and for respiration are presented. From the characteristics of measured voltage the parameters of equivalent circuit for electrode-skin interface with three elements are calculated for dry and gel-wetted electrodes.

KEY WORDS
voltage pulse plethysmography, current pulse plethysmography, electrode-tissue interface, tissue resistance variation

Introduction

Impedance plethysmography is mostly used for longer time clinically. Pulse plethysmography is the object of newer researches. Šantić et al [1], [2] mention advantages of current pulse plethysmography compared to impedance plethysmography. Main advantage of applying short current pulses in current pulse plethysmography instead of sinusoidal current in the impedance plethysmography is that in the pulse plethysmography the amplitude for sensations threshold is much higher then in impedance plethysmography implying that better signal-to-noise ratio could be achieved.

The purpose of this paper is to test characteristics of voltage pulse plethysmography, and to compare it with current pulse plethysmography which has already been examined [1], [2].

Measuring instrumentation

There are two possible implementations of voltage pulse plethysmograph, with grounded bio-tissue and with grounded current sampling resistor, R_0 (Fig. 1). Measurement for both cases was done in order to investigate advantages and disadvantages of both cases.

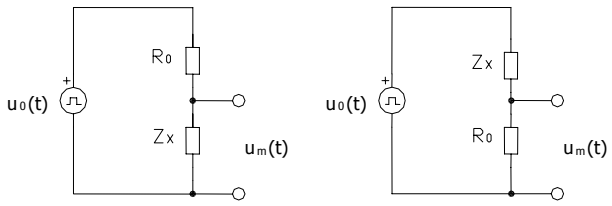

Fig. 1. Implementation with one electrode connected to bio-tissue grounded (left) and with current sampling resistor grounded (right)

Fig. 2 shows the circuit diagram of the voltage pulse plethysmography measurement set-up. The voltage pulse plethysmography set-up consists of the variable power supply (voltage source) U_0, the transistor switch T1, the switch control digital circuit (A.M., M.M.1 and M.M.2), reset transistor switch T2, S&H circuit and the a.c. coupled amplifier with incorporated notch filter.

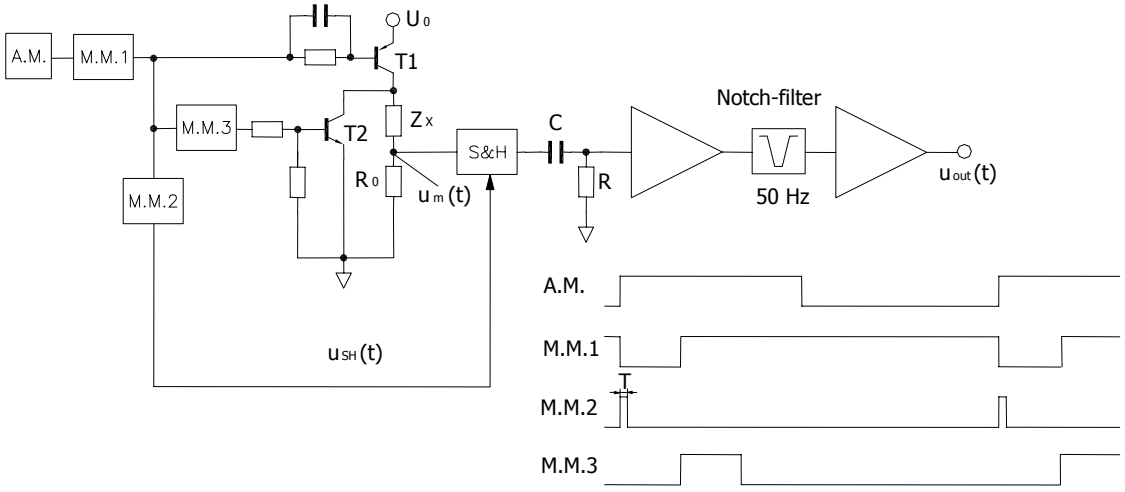

Fig. 2. The block diagram of voltage pulse plethysmography

The voltage pulses are generated by astable multivibrator A.M., monostable multivibrator M.M.1, and PNP transistor that works as a switch which connects voltage source U_0 with very little internal resistance to the bio-tissue and the current sampling resistor. Transistor T2 is used as a switch that discharges the capacitive component (C_X, Fig. 3) of electrode-skin interface impedance. Monostable multivibrator M.M.3 is controlling ON and OFF transistor T2.

The frequency of pulse repetition is defined by astable multivibrator A.M. switching at 10 kHz. As Fig. 2 shows monostable multivibrator M.M.1 is triggered by the positive edge of astable multivibrator pulse. The ON time of M.M.1 controls the duration of the voltage pulse (the ON time of transistor T1). The negative edge of the monostable multivibrator M.M.1 pulse triggers monostable multivibrator M.M.2 which controls S&H circuit. Positive edge of monostable multivibrator M.M.1 pulse triggers monostable multivibrator M.M.3, which controls transistor T2 and which discharges bio-capacitance C_X.

The measurement chain consists of S&H, high pass filter, notch filter for 50 Hz and amplifiers. The S&H samples and holds measured voltage u_m after sampling time T, which is determined by M.M.2 monostable multivibrator pulse duration. The high pass CR filter removes DC component of the voltage $u_m(t)$. High pass filter cutoff frequency is 0.05 Hz. The 50 Hz noise from power lines is reduced by the notch filter. The total gain of the amplifier chain is A=390. Also it should be mentioned, that voltage pulse generator is much simpler than current pulse generator.

Electrode-skin interface

The electrode-skin interface can be modeled by a five-element circuit that can be simplified to a three-element circuit due to the circuit symmetry (Fig. 3) [2]. The impedance of the electrode-skin interface is frequency dependent [3]-[5] and at high frequencies [2], [6] can be replaced by a serial RC-circuit, while resistance value R_{SX} is variable due to the change of the bio-tissue volume (blood pulsation, respiration, etc.). The resistance variations are rather small, typically 0.1-0.5 % [1], [6].

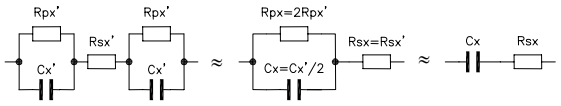

Fig. 3. Equivalent circuits for electrode-skin interface [2]

Wave-shapes corresponding to blood pulsations and respiration

It must be pointed out that maximal sensitivity $\Delta U_X/U_X$ is obtained when series resistance of the sampling resistor is equal to the bio-resistance. It is valid for both cases: with bio-tissue grounded and sampling resistor grounded.

Simple calculation can give us when maximal voltage value between systolic pressure voltage U_S and diastolic voltage U_D pressure $\Delta U_X=U_S-U_D$ can be obtained in comparison to $U_X=U_D$ i.e. $\Delta U_X/U_X$. Systolic pressure voltage is $U_X+\Delta U_X$ and diastolic voltage is U_X. The difference is:

$$\Delta U_X = U_S - U_D = \frac{R_0 U_0}{R_0 + (R_{SX} - \Delta R_{SX})} - \frac{R_0 U_0}{R_0 + R_{SX}}. \quad (1)$$

Looking for extreme value it can be found that it occurs when bio resistance R_{SX} is equal to current sampling resistance R_0.

Fig. 4 shows the measured voltage $u_m(t)$ while measuring pulsation with current sampling resistor grounded. Supposing the equivalent circuit of the electrode-skin surface with two elements, the voltage $u_m(t)$ during the time of low output of the monostable M.M.1 falls exponentially according to the relation:

$$u_m(t) = U_0 \frac{R_0}{R_0 + R_{SX}} \cdot e^{\frac{-t}{C_X (R_0 + R_{SX})}}. \quad (2)$$

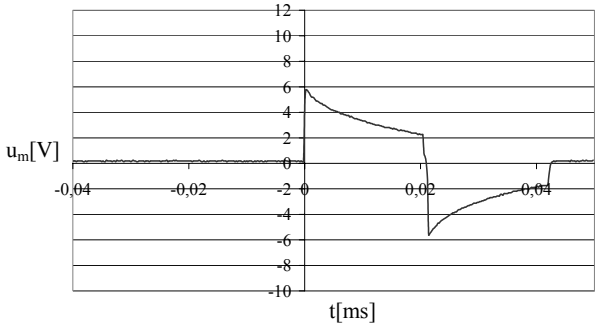

Fig. 4. The measured voltage waveform when pulsation is measured with current sampling resistor grounded

In all of the shown measurement of the blood pulsation and respiration the current sampling resistor R_0 is 682 Ω and the voltage U_0 is 11 V.

The sampling time T is 1.34 μs.

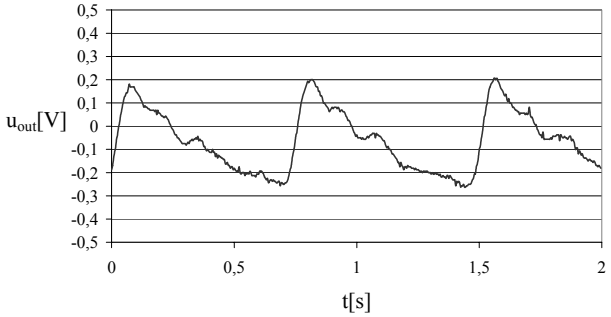

Fig. 5. The output voltage for blood pulsation in the arteries with grounded current sampling resistor

Fig. 5 shows blood pulsation measured in arteries with the current sampling resistor grounded. Although the circuit has tuning ability of the output offset, the output voltage has a variable DC component so it is difficult to adjust the

DC component to zero. Because of it, a high pass filter CR is connected as is shown in Fig. 2.
Fig. 6 shows the measured voltage $u_m(t)$ during pulsations measurement with one electrode grounded. Supposing the equivalent circuit of the electrode-skin interface with two elements, during the low output of the monostable M.M.1 the voltage $u_m(t)$ rises exponentially according to:

$$u_m(t) = U_0\left(1 - \frac{R_0}{R_0 + R_{SX}} \cdot e^{\frac{-t}{C_X(R_0+R_{SX})}}\right). \quad (3)$$

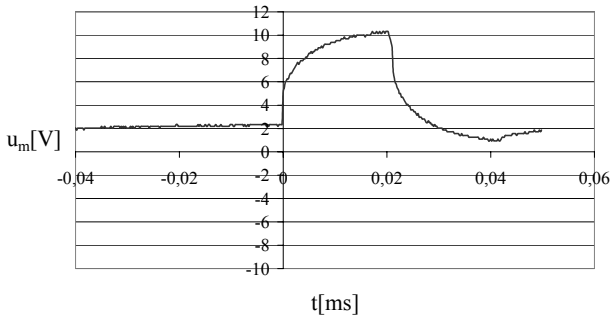

Fig. 6. The measured voltage waveform when measuring pulsation with bio-tissue grounded

This measurements with bio-tissue resistance grounded as his waveform shown on the Fig. 6 has essentially the same exponential wave shape as it in current pulse plethysmography and with similar characteristics.
Although the initial supposition was that the measurement with the electrode grounded will exhibit less noise, after multiple testing on several people, a significant advantage of either of the two methods hasn't been noticed. The reason lies in relatively small measuring object impedance, that is, in the small resistance value of the current sampling resistor. The capacitive coupling with the interference source in that case generates small, actually, the same and negligible signal noise. As no significant differences in measuring with a grounded current sampling resistor and a grounded electrode have been shown, the respiration measurement has been carried out only with current sampling resistor grounded. Fig. 7 shows the output voltage for normal breathing and Fig. 8 shows the output voltage for deep breathing.

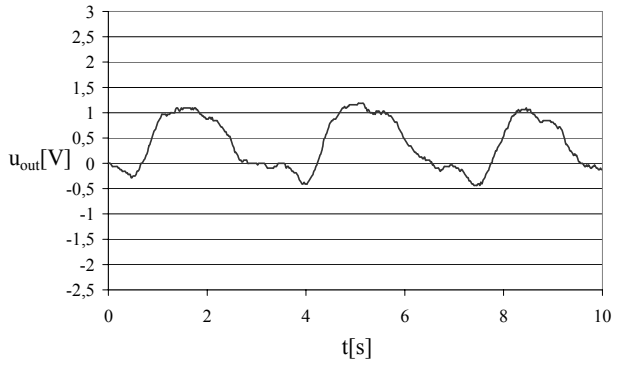

Fig. 7. Output voltage for normal breathing respiration with the current sampling resistor grounded

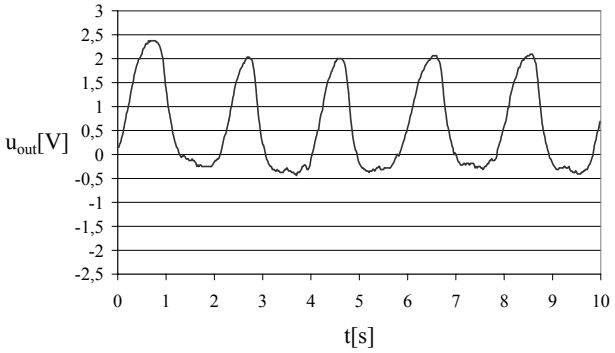

Fig. 8. Output voltage for deep breathing respiration with the current sampling resistor grounded

Determining the equivalent circuit of the electrode-skin interface

When the supposed equivalent circuit of the electrode-skin interface consists of elements and the parameters for it are calculated, the approximate voltage waveform, that is, the calculated voltage waveform based on these parameters differs from the measured voltage. But because the measuring sample has to be taken as soon as possible, after starting moment in the time T (1 μs – 2 μs) and S&H keep its value, the parallel capacitive component C_{PX} and resistive parallel component R_{PX} have no remarkable influence on measuring value.
Fig. 9 shows the measurement circuit for determining the parameters of the equivalent circuit electrode-skin interface.

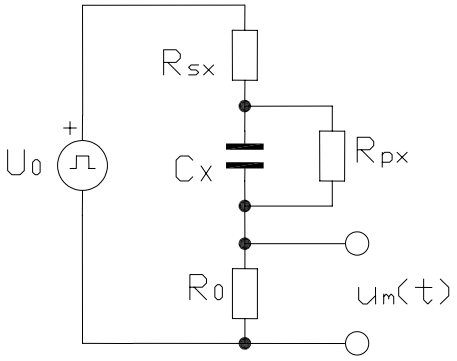

Fig. 9. The measurement circuit for determining the parameters of the equivalent circuit electrode-skin interface

A simple analysis gives the expression for the voltage $u_m(t)$:

$$u_m(t) = \frac{R_0 U_0}{R_0 + R_{SX} + R_{PX}} \cdot \frac{R_{PX}}{R_0 + R_{SX}} \cdot e^{-\frac{t}{\tau_1}} + \frac{R_0 U_0}{R_0 + R_{SX} + R_{PX}}, \quad (4)$$

where is $\tau_1 = \frac{R_{PX}(R_0 + R_{SX})C_X}{R_0 + R_{SX} + R_{PX}}$. (5)

R_{SX} and R_{PX} are calculated by the „end points“ method, that is, they are calculated from $u_m(t)$ at the starting moment (t=0+) and the „end“ moment (distant enough

from the starting moment, when the transient process actually dies out, t->∞). The expression for $u_m(t)$ at the starting moment is:

$$u_m(0+) = \frac{R_0 U_0}{R_0 + R_{SX}}. \quad (6)$$

The expression for $u_m(\infty)$ is:

$$u_m(\infty) = \frac{R_0 U_0}{R_0 + R_{SX} + R_{PX}}. \quad (7)$$

Which gives:

$$R_{SX} = R_0\left(\frac{U_0}{u_m(0+)} - 1\right), \quad (8)$$

$$R_{PX} = R_0 U_0\left(\frac{1}{u_m(\infty)} - \frac{1}{u_m(0+)}\right). \quad (9)$$

Inserting the expression (8) and (9) into (5) results in:

$$\tau_1 = \frac{R_{PX}(R_0 + R_{SX})C_X}{R_0 + R_{SX} + R_{PX}} = \frac{R_0 U_0\left(\frac{1}{u_m(\infty)} - \frac{1}{u_m(0+)}\right)\left[R_0 + \frac{R_0 U_0}{u_m(0+)} - R_0\right]C_X}{\left[1 + \left(\frac{U_0}{u_m(0+)} - 1\right) + U_0\left(\frac{1}{u_m(\infty)} - \frac{1}{u_m(0+)}\right)\right]R_0} \quad (10)$$

Further we get:

$$\tau_1 = R_0 U_0 C_X \frac{u_m(0+) - u_m(\infty)}{[u_m(0+)]^2}. \quad (11)$$

Inserting the expression (8) and (9) into (4) results in:

$$u_m(t) = [u_m(0+) - u_m(\infty)] \cdot e^{-\frac{t}{\tau_1}} + u_m(\infty). \quad (12)$$

Further follows:

$$e^{\frac{-t}{\tau_1}} = \frac{u_m(t) - u_m(\infty)}{u_m(0+) - u_m(\infty)}, \quad (13)$$

and expression for τ_1:

$$\tau_1 = \frac{-t}{\ln\left[\frac{u_m(t) - u_m(\infty)}{u_m(0+) - u_m(\infty)}\right]}. \quad (14)$$

For determining the C_X value the point is taken when the voltage u_m falls to one half of its value:

$$u_m(T_1) = \frac{u_m(0+)}{2}. \quad (15)$$

Fig. 10 shows the measured voltage curve and the values are marked which are used to calculate the equivalent circuit parameters of the electrode-skin interface.

τ_1 is calculated as follows:

$$\tau_1 = \frac{-T_1}{\ln\left[\frac{0{,}5 \cdot u_m(0+) - u_m(\infty)}{u_m(0+) - u_m(\infty)}\right]}. \quad (16)$$

C_X is calculated from τ_1 by formula:

$$C_X = \frac{\tau_1}{R_{SX}}. \quad (17)$$

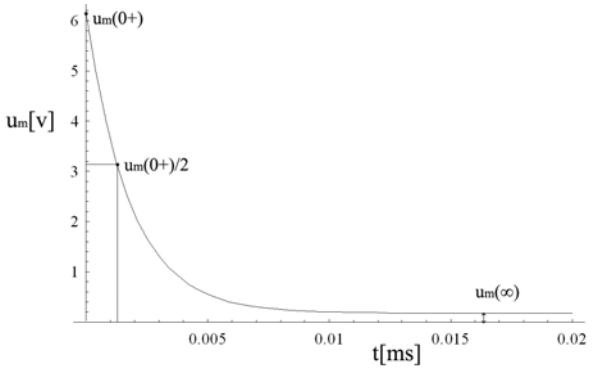

Fig. 10. The measured voltage curve with marked values used in calculation of the equivalent circuit parameters of electrode-skin interface

Fig. 11 shows the measured voltage (average value from three measurements) and the calculated value, which is calculated inserting the values obtained by the „end points“ method and from the point when the voltage drop to one half of its value according to (8), (9) and (17). The curve is given for the excitation voltage U_0=11 V. Fig. 12 shows the discrepancy of the calculated curve from the measured one.

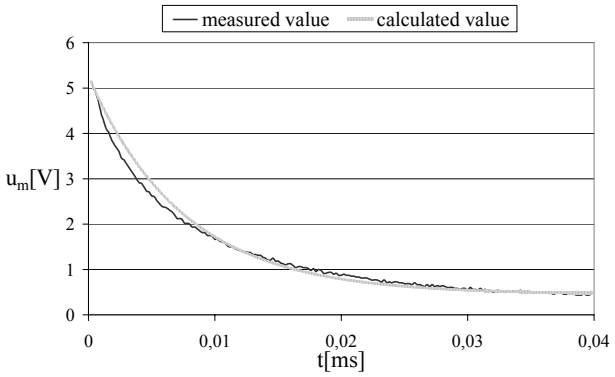

Fig. 11. Measured voltage compared to the calculated value for U_0=11 V, gel-wetted electrodes

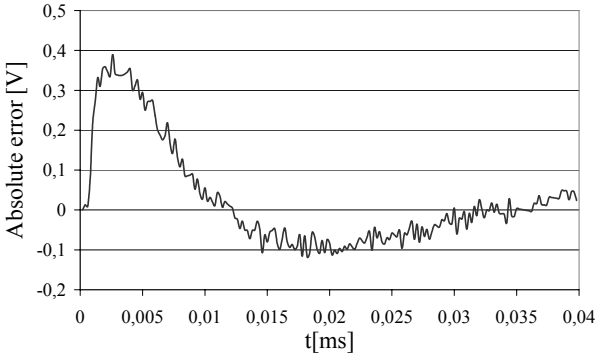

Fig. 12. Absolute error of the calculated curve for U_0=11 V excitation

Since by determining the equivalent circuit parameters for the electrode-skin interface with three elements a certain

variance of the measured from the calculated voltage curve is obtained, it was examined if determining the parameters from some other three points would give less variance. Although the results of determination from three arbitrarily chosen points gave equally good results as the "end-points" method and the point when the voltage drops to the half value, this method is considered more exact (the initial and the final voltage value can be well determined because they don't depend on the capacitance C_X) than the choice of any three pints, so in the study the value parameters of the equivalent circuit were calculated in this way.

There is a noticeable variance of the measured value from the calculated value obtained by calculating from the equivalent circuit parameters of the electrode-skin interface determined by any of the procedures.

Figures 13, 14 and 15 show the dependences of the resistance R_{SX}, the capacitance C_X and the resistance R_{PX} of the equivalent circuit electrode-skin interface with three elements determined by the "end-points" method and from the point when the voltage drops to its half value. The dependence on the voltage U_0 for the gel-wetted electrodes and for dry electrodes is shown.

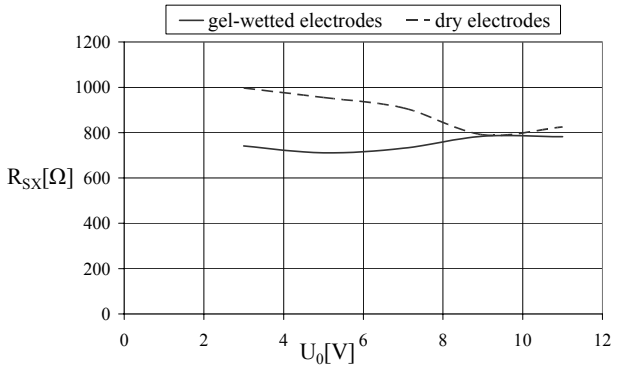

Fig. 13. Dependence of the resistance R_{SX} on the voltage U_0

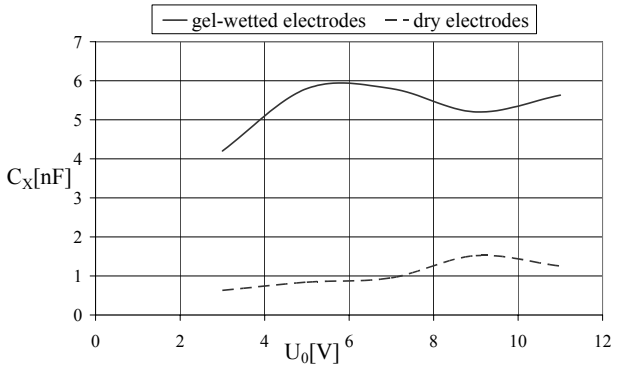

Fig. 14. Dependence of the capacitance C_X on the voltage U_0

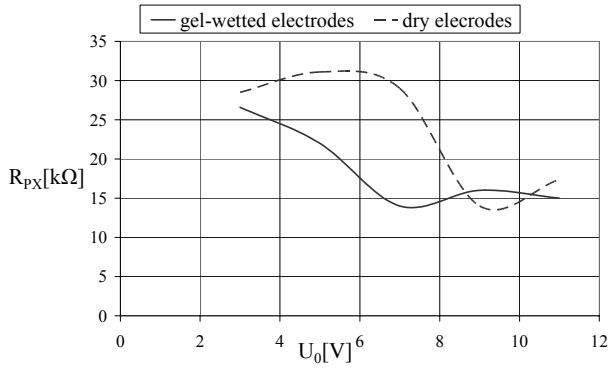

Fig. 15. Dependence of the resistance R_{PX} on the voltage U_0

When using gel-wetted electrodes the R_{SX} values don't vary considerably, the value range is between 710 Ω and 790 Ω. When using dry electrodes there are also no considerable variations, the value range is between 800 Ω and 1000 Ω, that is, the values are somewhat higher than when using gel-wetted electrodes (when the transient skin surface resistance is lower).

Neither when using gel-wetted electrodes nor when using dry electrodes the capacitance C_X values don't vary significantly. When using gel-wetted electrodes the capacitance value range is between 4 nF and 6 nF. When using dry electrodes the value range is between 0.6 nF and 1.5 nF, that is, the values are smaller than when using gel-wetted electrodes. C_X is higher when using gel-wetted electrodes than with dry electrodes because the tissue soaked with gel acts as if the distance between the electrodes has become shorter so the capacitance is larger.

For the R_{PX} dependence on U_0 a negative trend has been noticed. When using the gel-wetted electrodes the values vary between 13 kΩ and 27 kΩ. When using dry electrodes the values vary between 13 kΩ and 32 kΩ. R_{PX} (13 kΩ – 32 kΩ) has no much influence on measurements, because it is much higher than R_{SX} (700 Ω - 1000 Ω).

Conclusion

Two voltage pulse plethysmography implementations are possible – with the current sampling resistor grounded and with one electrode grounded. Blood pulsations measurement in arteries and the respirations for both implementations have been carried out. Although it was supposed that the implementation with one electrode grounded will show less noise, this supposition has not been confirmed. On the contrary, no significant difference in the quality of obtained measured results in the two implementations has been noticed.

In the voltage pulse plethysmography the equivalent circuit parameters of the electrode-skin interface with three elements have been determined from the measured voltage characteristics. The absolute error was given as a measure for the calculated voltage variance that is calculated from the obtained equivalent circuit parameters and the measured voltage. A significant variance has been noticed. That can be explained by impossibility of

substituting the electrode-skin interface with a three linear-element circuit and points out the need to modify the electrode-skin interface with an equivalent circuit with non-linear elements.

But series resistance R_{SX} is significant in this measurements while the capacitive component C_{PX} and parallel resistance R_{PX} have very little influence when sampling time is short (1 μs – 2 μs). Also it is important to know, that maximal sensitivity can be obtained when sampling series resistance and bio-resistance are equal. Besides, it should be mentioned, that voltage pulse generator circuit is simpler than current pulse generator circuit.

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