MONITORING OF PSYCHOPHYSIOLOGICAL PROCESSES BY SKIN CONDUCTIVITY MEASUREMENTS USING MICROELECTRODE ARRAYS

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Abstract: This work describes the new modification of non-invasive biomedical monitoring of psychophysiological processes based on skin conductivity measurements by interdigitated array (IDA) of microelectrodes. From electrical model of IDA microelectrode/skin interface and simulations the important outcome has arisen: the electric field distribution and depth of penetration into the outer skin layers (epidermis laminar structures) depend on the configuration and size of electrode system [1]. This knowledge provides a possibility to examine different layers of epidermis by electrical impedance method and we used it for the analysis of electrophysiological processes in skin when the person is under the stress. Standard psychotests showed that the response signals of commercial macroelectrodes measured with GSR method and of our microelectrodes were similar – microelectrode signals were more stabile with shorter response time.

Introduction

Measurements of the electrical conductivity, resistance or impedance of skin surface have a 120 years history. The way how physiological changes in living tissue are reflected in electrical impedance parameters is still not very clear [2]. The simple psycho-galvanometer was one of the earliest tools of psychological research [3].

The psycho-galvanic reflex results in a change of the skin conductivity during periods of stress, excitement or shock. Under these conditions skin conductivity increases, whereas during periods of relaxation the conductivity declines to a minimum [4].

We have developed a novel modification of biomedical monitoring of psychophysiological processes based on skin conductivity measurements by interdigitated array (IDA) of microelectrodes. Our sensors were designed to investigate electrophysiological aspects of human physiology in a completely safe and non-invasive manner. This technique also has no influence on natural physiological processes.

For correct investigation of human stress it is important to observe psychogalvanic reflex, heart pulse and respiration frequency. Our sensors as you will see are able to monitor psychogalvanic reflex and hearth pulse simultaneously.

Theory

The novel principle of our method is based on a special configuration of the thin-film microelectrode system which generates an electric field $\bar{E}$ with intensity lines across (transversely) planar structures of skin. From electrical model of IDA microelectrode/skin interface and simulations the important outcome has arisen: the electric field distribution and depth of penetration into the outer skin layers depend on the configuration and size of electrodes (Figure 1) [1].

In case of macroelectrodes when the distance d between coupled electrodes is greater than the thickness of stratum corneum with potential barrier h $(d >> h)$, the vector intensity lines of the electric field $\bar{E}$ are enclosed across the planar structures of skin (Figure 1a,b).

But if we use microelectrode pairs when the distance of the electrodes pairs is less than the electric thickness of the skin (stratum corneum with potential barrier) ($d < h$, $d > s$), then the lines of electric field are enclosed in longitudinal circuit relative to laminar skin structures of epidermis (in stratum corneum). From deeper layers of skin are these vector intensity lines embossed to the surface (to the area with lower conductivity) by the instrumentality of potential barrier (Figure 1c). But under stress stimulus is the potential barrier narrow down and the electric field can reach deeper layers of skin with higher conductivity and therefore is summary conductivity increasing (Figure 1d). Such configuration is ideal for analysis of electrophysiological processes in skin when the person is under stress.

The results of analytical analysis also showed that in case of non-symmetric coplanar electrodes the electric field is more enclosed in top layers of laminar structures of skin (stratus corneum). This system consists of electrode periodical structure with different areas. In non-symmetric structure the density of the electric field intensity lines along the planar structures of the skin is 30 % higher in top layers[1].

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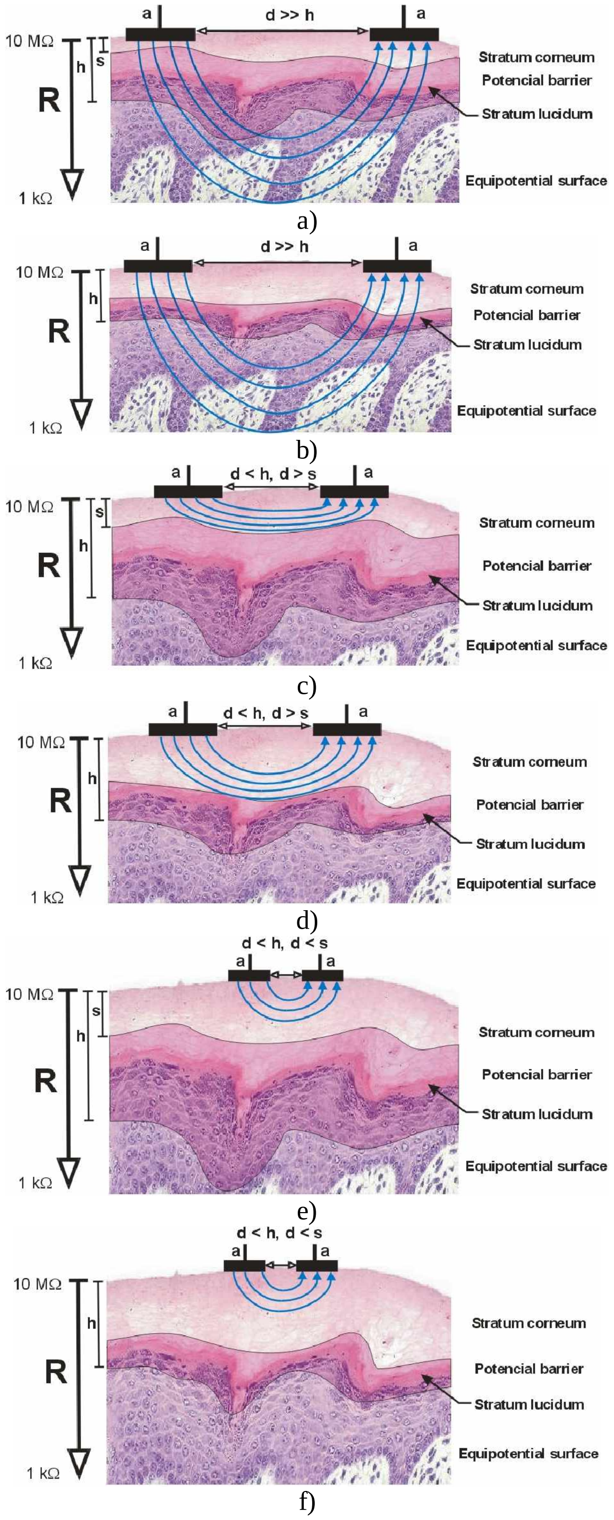

Figure 1: The vector intensity lines of the electric field in human skin by using:
- macroelectrodes:
  a) in relaxation time
  b) under stress stimulus
- microelectrodes:
  c) in relaxation time
  d) under stress stimulus
- non-symmetric microelectrodes:
  e) in relaxation time
  f) under stress stimulus

In this case (non-symmetric electrodes) ($d < h$, $d < s$) the vector intensity lines of the electric field $\bar{E}$ are enclosed in top layers of stratum corneum and the flow of electric lines is independent on thickness of potential barrier (Figure 1e, f). Such electrodes are more ideal for analysis in cosmetic.

The greatest degree of conductivity change occurs in the skin of palms and bottom parts of fingers, as this is a measure of the large volume of the motor cortex involved with hand and finger movement, the delicacy of touch and sensation required for manual skills and pain reception and therefore we put our microelectrodes on these parts. [3].

Experimental techniques

The electrodermal response (EDR) to stress stimulus was detected by variation of IDA microelectrode conductivity ΔG (relative change of current ΔI (Agilent 34970A - Data Acquistion / Switch Unit) at constant supply voltage U and frequency f (Agilent 33120A – Function / Arbitrary Waveform Generator)) placed on forefinger of non-dominant hand (left for right-handers). At the beginning of experiments we found optimal supply voltage/frequency – 3 V/1 kHz.

For non-invasive biomedical monitoring of psycho-physiological processes based on skin conductivity measurements we used 3 types of interdigitated array (IDA) of microelectrodes:

- non-symmetric configuration
  - 15 μm/25 μm /50 μm (finger/gap/finger)
- symmetric configuration
  - 100 μm/100 μm (finger/gap dimensions)
  - 200 μm/200 μm
  - 400 μm/400 μm

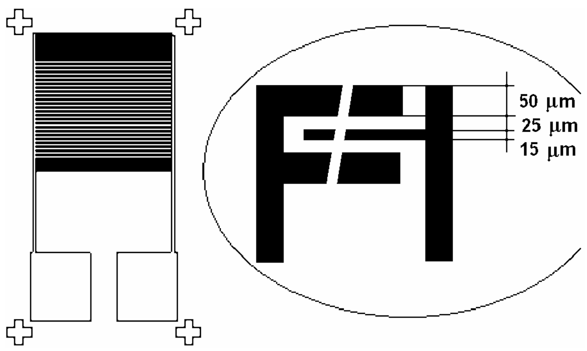

Figure 2: Design of non-symmetric electrode chip

The total dimensions of microelectrode chip were 10 x 15 mm. To minimize the polarization effect electrodes were made from Pt thin film (Figure 2). The microelectrodes were fabricated by standard thin film technology: Pt films (150 nm in thickness) underlaid by Ti film (50 nm) were deposited by rf sputtering on alumina substrates and microelectrodes were lithographically patterned by lift-off technique.

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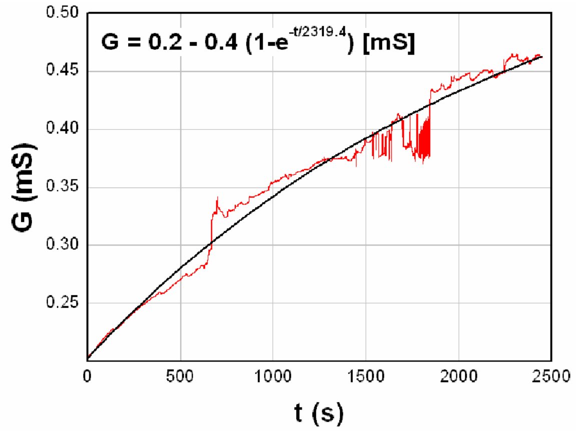
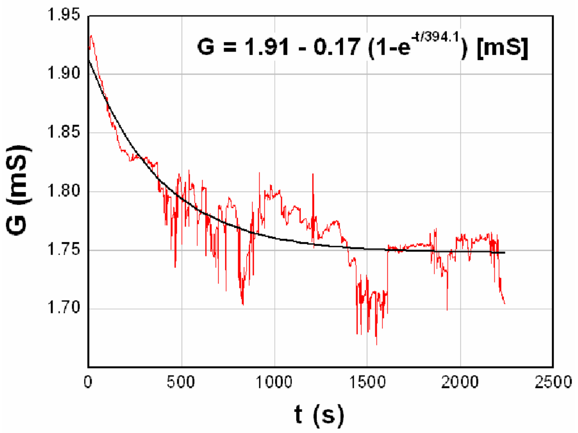
Results

During conductivity measurements a drift of output signals occurs due to polarization effects in skin – electrodermal phenomenon (EDF). We investigated the drift of output signals due to EDF and sweat hydration of skin upper layers (stratus corneum and lucidum). These experiments were done by 200 μm/200 μm microelectrodes for dry skin (Figure 3a) and for wet (sweaty) skin (Figure 3b).

Measured curves were interpolated by exponential function $G(t) = A + B(1-e^{-t/C})$, where G is skin conductivity, t is time and A, B, C are constants. This function seems to be very suitable for description of polarization and hydration effects in skin.



a)

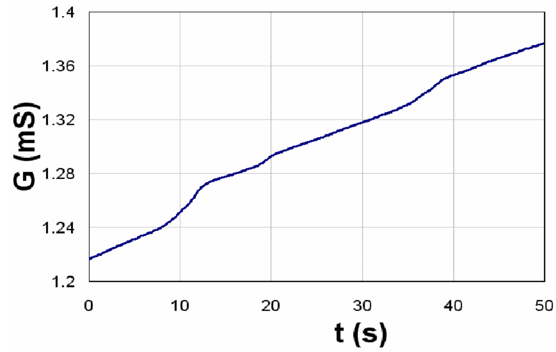


b)

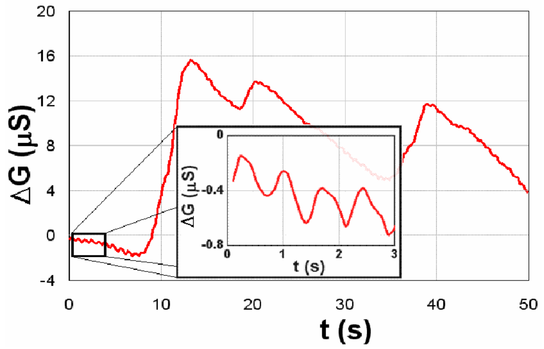
Figure 3: Electrodermal phenomenon and effect of sweat hydration
a) dry skin
b) sweatily (hydrated) skin

These experiments also showed that the stabilization of signal of our microelectrodes occurs faster for hydrated skin (10 - 40 minutes) than for dry skin (40 minutes and more). Increase of conductivity due to EDF of typical macroelectrodes is ended in 30 – 40 minutes [5].

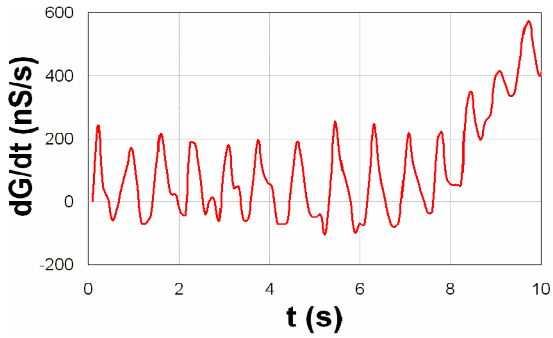
During the first seconds of measurements it was possible to minimize and correct the signal drift (caused by EDF) by software (Figure 4a, b). The software analysis program for skin conductance activity was developed by us and made in development environment HP-VEE 6.0.



a)



b)



c)

Figure 4: Typical time dependences of EDR
a) uncorrected
b) corrected EDR and its zoom comprising the pulse of heart
c) derived uncorrected signal comprising the pulse of hearth

Our experiments led to very important result: our microelectrode probes are able to monitor simultaneously electrodermal response and hearth pulse (Figure 4b). The hearth pulse was easily read out from derivation of signal (Figure 4c).

In next experiments we investigated the influence of size of microelectrodes (symmetric: 200 µm/200 µm, 100 µm/ 100 µm, non-symmetric 15 µm/25 µm/50 µm) on output signals. As mentioned in "Theory" section different types of microelectrodes generate the electrical field enclosed in various structures (depths) of skin. Results were obtained using supply voltage 3 V/1 kHz and EDF correction. It is clearly shown (Figure 5), that output signal of non-symmetric IDA is very low in comparison with symmetric IDA 100 µm/ 100 µm and 200 µm/200 µm arrangement. It is inflicted by the fact that the penetration depth of the electric field generated by non-symmetric microelectrodes is insufficient to reach the potential barrier of skin, which is very sensitive for detection of psycho-physiological processes, e.g. human body stress [3, 5]. In case of non-symmetric microelectrodes most of electrical field intensity lines – about 80 % - are enclosed in the depth of 0 - 25 µm [1], while the most important layer of potential barrier is placed more than 30 µm beneath the surface. The low response for sensor is due to sweat in upper layer - stratum corneum.

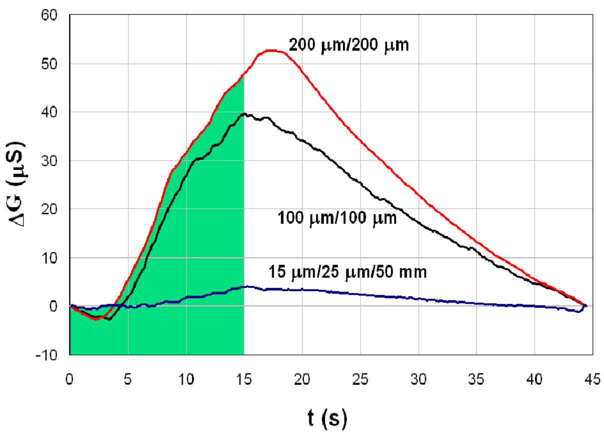


Figure 5: Influence of size and configuration of microelectrodes on amplitude of output signals

We have also analyzed influence of sweat hydration on output signals. For this purpose we moistened the skin surface with NaCl solution of concentration which was multiple times higher than normal concentration of human sweat (which is 0.3 – 0.8 %). The measurements were done by 100 µm/100 µm, 200 µm/ 200 µm and 400 µm/400 µm microelectrodes. Stress responce was very small in case of non-symetric microelectrodes. Measurements showed that sweat hydration causes oscilation of signal which is higher by using small microelectrodes and therefore the read out is more difficult. Finaly we can say that sweat hydration increases conductivity and brings noise to output, but even though the sweat is not dominant for current flow across the skin [5]. This is supplyed mainly by hydrated epidermis (potential barrier) near sweat duct (tubulus) formed around stratum lucidum.

After this experiment we can conclude that skin hydration (sweat) had slight less influence on output signal as it was expected. The skin hydration had an effect on absolute values of conductivity but only slight influence on the course of EDR function.

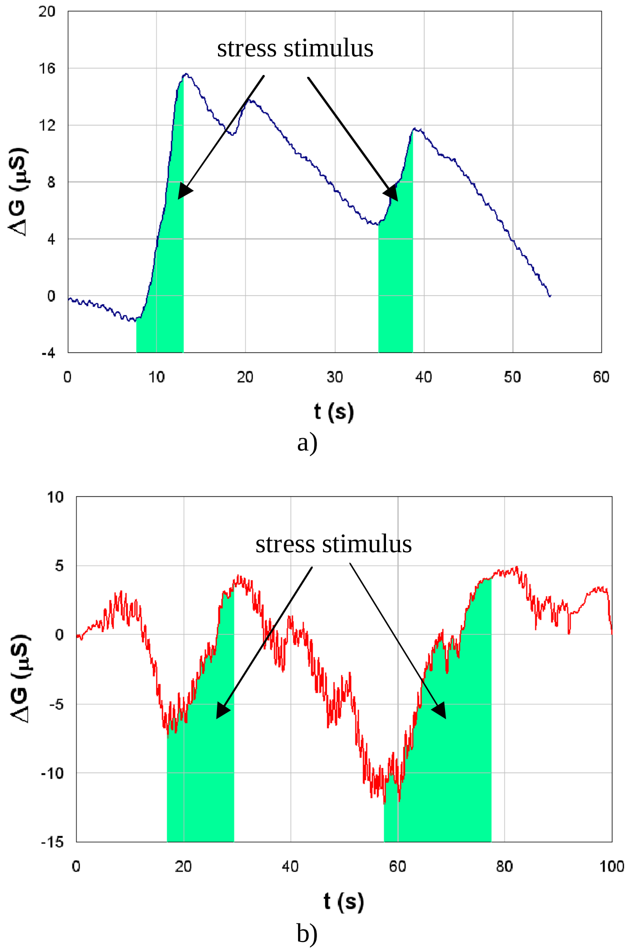


Figure 6: Influence of sweat hydration on output signal noise:

a) dry skin (G$_0$ = 1.22 mS)

b) sweatily skin (G$_0$ = 2.14 mS)

In last experiments we compared results obtained by our microelectrode approach with classical macroelectrode GSR method [6] (Figure 7). Supply signal parameters for each method were different:

o macroelectrode: U=25 mV, f = 75 Hz

o microelectrode: U=3 V, f = 1 kHz

Both output signals were recorded at the same time during the psychotest. Microelectrode (200 µm/200 µm) was placed on forefinger bottom and macroelectrodes between middle finger and ring finger of non-dominant hand.

Standard psychotests showed that the response signals of commercial macroelectrodes measured with GSR method and our microelectrodes were similar – microelectrode signals were more stabile and faster. The response time for microelectrodes was 5 seconds and for macroelectrodes about 10 seconds.

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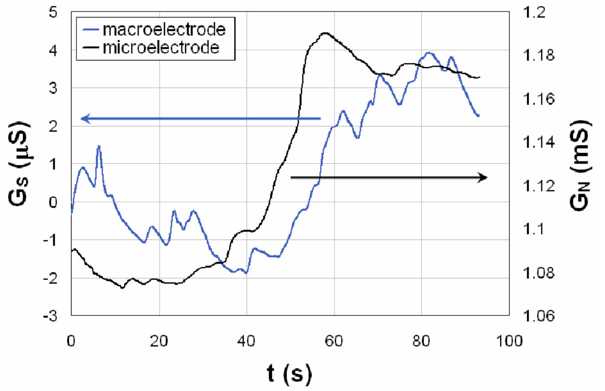
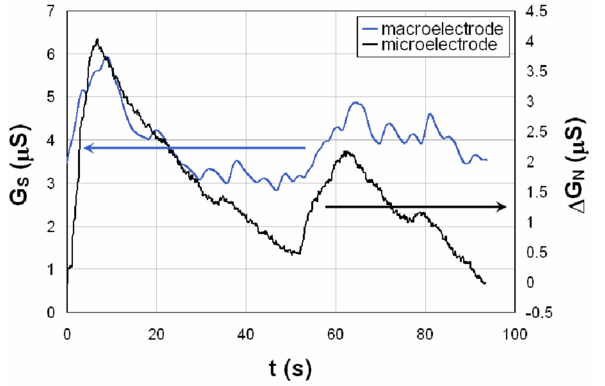


a)



b)

Figure 7: Comparison of our microelectrode and classical macroelectrode GSR methods

Conclusions

A new modification of conductivity EDR monitoring has been developed by using IDA thin film microelectrodes. Experimental results showed that our microelectrode probes are able to monitor simultaneously electrodermal response and pulse of blood caused by the heart rhythm. Software for correction of output signal drift determined by EDF was worked out. We found optimal supply voltage and frequency (3 V/1 kHz) and maximum EDR for IDA 200 µm/200 µm microelectrode. The skin hydration had an effect on absolute values of conductivity but only slight influence on the course of EDR function.

Comparison of commercial macroelectrodes using GSR method and our microelectrode approach confirmed that both response signals had similar shape during standard psychotests – microelectrode signals were more stabile with shorter response time.

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