Design of Sensor Systems for Long Time Electrodermal Activity Monitoring

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Abstract. *This article describes successive development of electrodermal activity monitoring sensor system. Our aim is to improve existing systems to be more practical and suitable for long-term monitoring. Therefore, compared to conventional devices, our system must be easily wearable, without limiting the examined person in ordinary life, with low power consumption, battery operated and reducing the impact of negative artefacts. Specifically, we describe here three devices. The first is serving mainly to familiarize with the methodology, extensive testing and optimization of measurement parameters. Based on the obtained result, we constructed second system in form of small ring - "EDA ring". Last sensor system is developed with the effort to integrate the monitoring of electrodermal activity in e-health and smart clothes.*

Keywords

EDA-ring, e-Health, electrodermal activity, IDAE electrodes, smart clothes.

1. Introduction

Interconnection of microelectronics and medicine has been particularly in last decade very interesting technical field. The increase of computing power of new microprocessors, production of novel integrated chips and continual development of measurable human health parameters sensing techniques positively affect the expansion of new integrated medical devices. These health monitoring devices [1], [2] and [3] present not only a powerful healthcare assistance for end consumer, but also useful instrument for many biomedical research groups.

2. Theory

Electrodermal Activity (EDA), also known as galvanic skin response, electrodermal response or psychogalvanic reflex is used widely in physiological research due to its low cost and high utility. Typical EDA measuring systems [4] measure skin conductance in macroscopic level. That means that electric field is enclosed perpendicularly to the skin surface. The conductivity is measured through planar structures of epidermis and dermis under the first electrode, across blood vessels to the second electrode and through next skin structures. Imperfection of these systems is obvious: there is a lack of local measurement and the total measured signal includes fluctuating parameters of human skin structure that are insensitive to nervous system.

First, it was assumed that increase in the skin conductivity during a stress stimulus is only caused by the skin perspiration. Later, an important factor of potential barrier existence near the stratum lucidum layer was discovered, analysed and proven. Its thickness changes due to the nervous system activity. The greatest degree of conductivity variations occurs in the skin of palms and bottom parts of fingers [5] and [6].Therefore, our sensors are designed particularly for local measurement of top layers of the skin (stratum corneum and stratum lucidum). For such design the ideal means are interdigital arrays microelectrodes (IDAE).

Figure 1 presents an equivalent substitute model for the measurement of impedance properties of the human skin. R_E and C_E characterize the electrical parameters of IDAE electrode system, R_C, C_C characterizes the electrical parameters of stratum corneum and R_{EQ}, C_{EQ} properties of equipotential area. Electrical parameters were determined and calculated with respect to the published values [7], [8] and [9]. Psychological conditions changes and the composition of the stratum corneum were simulated by R_E and C_E values.

By decreasing the distance between the electrodes (d) the sensitivity to simulated measurement system is increasing. This is due to reduction of the value of R_{C1} with decreasing electrode distance d. If the distance between the electrodes is smaller than the thickness of the upper laminar structures, the importance of the parameter R_{C1} increases, because its value becomes lower than RC and therefore it is more dominant because of parallel connection.

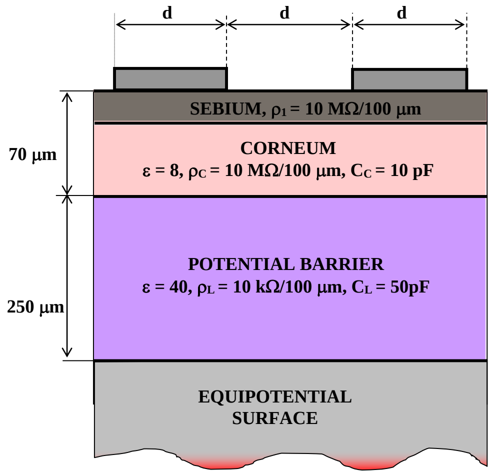
A great help for modelling of various kinds of electrode systems is the program QUICKFIELD. Isotropic or anisotropic properties of the study area may be defined by components ε_x and ε_y. The analysed structure (Fig. 2) is drawn and analysed as a system - substrate - conductive strip (of the material) - cutaneous environment. Individual areas are differentiated by their size, relative permittivity (Fig. 1(a)) and the energized or de-energized area. By using this program, not only an overview of the layout of the field can be obtained, but also the values at various points in a two-dimensional structure. Based on that computer analysis we can optimize the electrode system. In our conditions, we worked with two different planar electrodes:

- symmetric electrodes: $100 \times 100 \times 100$ μm (Fig 2(a)),
- asymmetric electrodes: $100 \times 50 \times 30$ μm (Fig 2(b)),

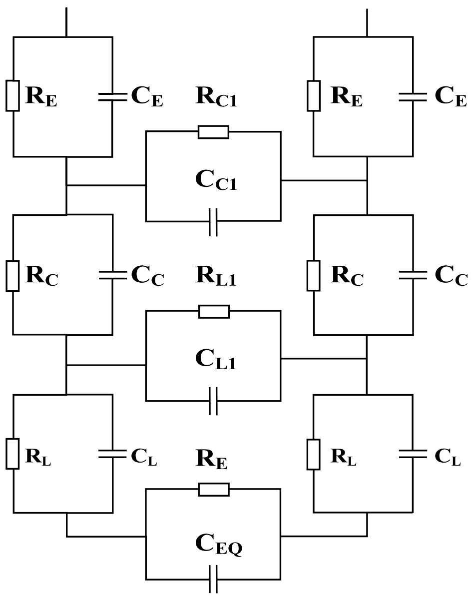
where 1$^{\text{st}}$ electrode width (w_1) $\times$ slit (d) $\times$ 2$^{\text{nd}}$ electrode width (w_2) [6].

Based on electrical model of interdigital arrays (IDAE) microelectrode/skin interface and simulations we have found that electric field distribution and depth of penetration into the outer skin layers depend mainly on the configuration and size of an electrode system.

This knowledge provides the possibility to examine different (separate) layers of epidermis by electrical impedance method. In short we can say that "electric field penetration depth" into human body is relatively close to "distance between the coplanar electrodes". It is perfectly valid for symmetric configuration, where the distance between electrodes is equal to electrodes width. The results of analysis also showed that in case of non-symmetric electrodes the electric field is more enclosed in outer layers of skin [10].

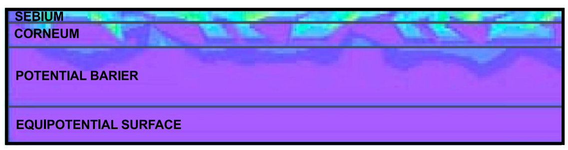


(a) Skin structure physical parameters and typical dimensions.

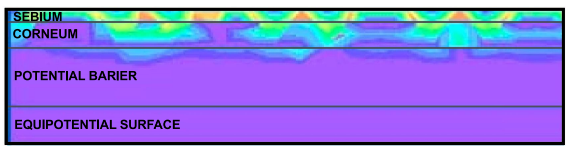


(b) Reciprocal electrical model.

Fig. 1: Model of human skin.

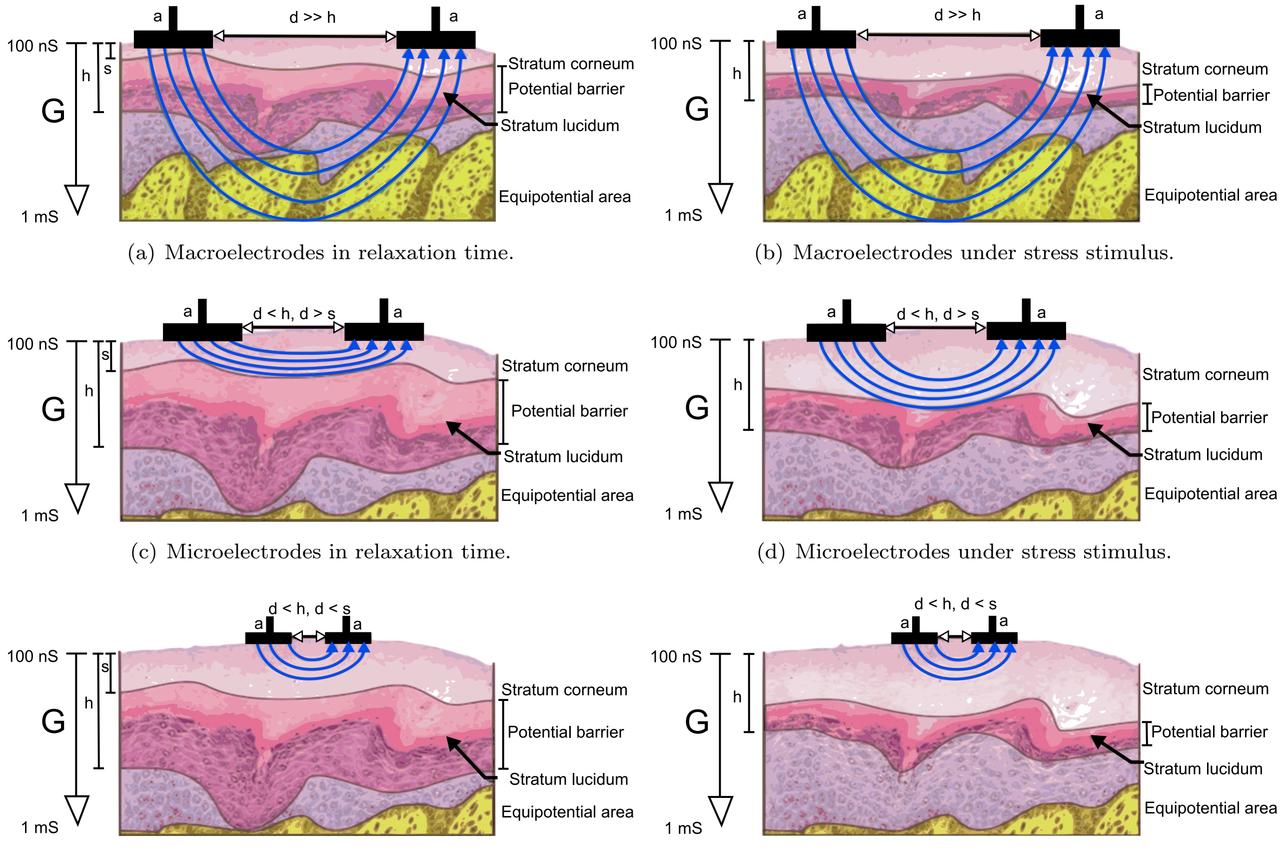


(a) Symmetric electrodes: $100 \times 100 \times 100$ μm.



(b) Asymmetric electrodes: $100 \times 50 \times 30$ μm.

Fig. 2: IDA microsensor/human skin interface simulations - distribution of electric field intensity [6].



(a) Macroelectrodes in relaxation time.

(b) Macroelectrodes under stress stimulus.

(c) Microelectrodes in relaxation time.

(d) Microelectrodes under stress stimulus.

(e) "Too small" microelectrodes in relaxation time.

(f) "Too small" microelectrodes under stress stimulus.

Fig. 3: The dominant vector intensity lines of the electric field in human skin.

If there are different electrodes applied on human skin, various space distributions of electrical field in the skin can occur.

In case of using macroelectrodes, when the distance between the coupled electrodes is greater than the thickness of electric active layers of skin h (stratum corneum (the outermost layer of the skin) with potential barrier) $d \gg h$, the vector intensity lines of the electric field are enclosed perpendicularly to the skin surface across the planar skin structures through dermis with high conductance (Fig. 3(a) and Fig. 3(b)).

In case of use of microelectrode pairs the lines of electric field are enclosed in parallel direction relative to laminar skin structures of epidermis. This is because the distance between the electrodes is less than the thickness of electric active layers of skin: $d < h$ and higher then thickness of stratum corneum: $d > s$ (d - electrode distance, h - thickness of electric active layers of skin, s – thickness of stratum corneum). From inner layers of skin, the electric field intensity lines are embossed to the surface (to the area with a lower conductivity) by the influence of the potential barrier which is generated by electrical double-layer around stratum lucidum (Fig. 3(c)).

Under a stress stimulus the potential barrier narrows down and the electric field can reach inner layers of human skin with higher conductivity, and therefore the total conductivity increases (Fig. 3(d)). Such configuration is ideal for the analysis of electrophysiological processes in human skin under stress.

In case of small sized microelectrodes the vector intensity lines of the electric field are enclosed in top layers of stratum corneum and the flow of electric lines is independent from thickness of potential barrier (Fig. 3(e) and Fig. 3(f)). Such electrodes are more ideal for surface analysis in cosmetics [10].

In our designed sensor system, we used IDA microelectrodes with utilized 200 µm × 200 µm × 200 µm (1^{st} electrode width × slit × 2^{nd} electrode width) dimensions. The obtained pre-experimental results proved that the optimal amplitude of driving signal should be selected from 1.5 V to 3 V. The driving signal frequency is not critical; however, an optimal value of several kHz has been proved.

3. Sensor Systems

3.1. Testing System

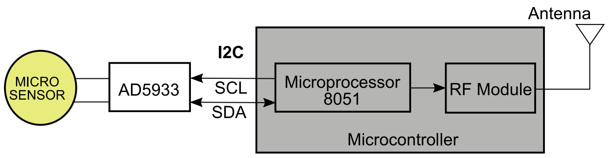
Several methods, applicable to continuous measurement of the human skin impedance, were considered, tested and analysed. As a result of this analysis we have chosen in the 1st setup the auto-balancing bridge method. It has high accuracy, short measurement time, high repeating rate, frequency and amplitude signal definition, possibility to measure both real and imaginary impedance components, controllability by a microprocessor, digital processing, etc. [11].

The proposed complex measurement system, offering these features, is composed of two main parts (Fig. 4). The core of the proposed portable monitoring system (Fig. 4(a)) is the integrated circuit AD5933 [12] that provides measurement of the human skin impedance sensed by the developed microsensor. The measurement process is controlled by the microprocessor nRF24E1 via I2C interface.

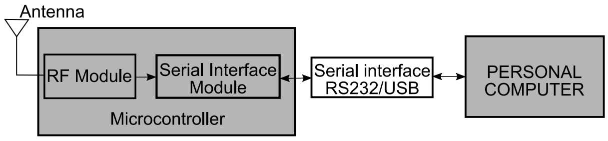
Using the RF wireless communication interface, the microprocessor sends the measured data to the receiver part (Fig. 4(b)) on the PC side. Consequently, the personal computer executes data storage and data post-processing. Additionally, the microcontroller also provides an initial configuration of integrated circuit AD5933 (setting the frequency and amplitude of the driving signal, measurement time slots, power management, etc.).

PCB has been realized on double layer FR4 board by SMT technology with minimum strip width of 0.2 mm and minimum clearance width of 0.2 mm. The total size of the measurement unit is 50 × 60 mm and 11 × 17 mm for the receiving module (Fig. 4(c)).

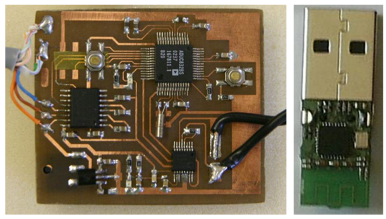
In order to verify the developed and realized system a comparison of our microelectrodes-based EDA method to the commercial macroelectrode approach [13], usually used in the laboratory medical or psychological experiments, was carried out. The 200 µm × 200 µm × 200 µm IDAE electrode microelectrode was placed on middle-finger of non-dominant hand, where macroelectrodes where connected between index and ring finger, so we could have expected that specific conductivities were similar. The comparison was performed by using the standard „Distraction" psychotests [14], where the signals were measured simultaneously. It shows that the responses given by both approaches were similar. Under stress stimulus the conductivity is increasing and in relax decreasing. However, the microelectrode signals are observed to be more stable with a shorter response time (Fig. 5).



(a) Portable measurement block.



(b) Receiver module.



(c) PCB design of measurement and receiver module.

Fig. 4: Realized EDA monitoring system.

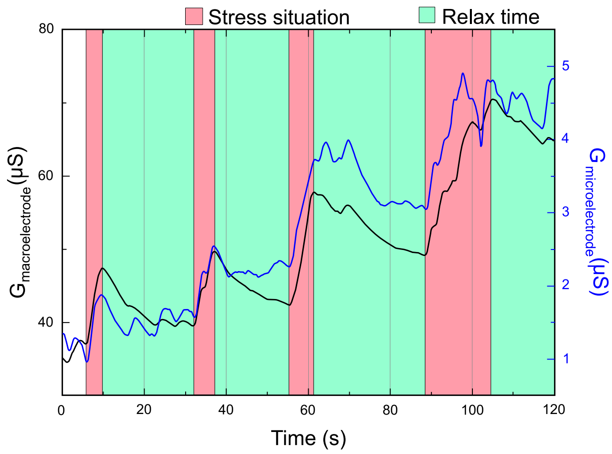


Fig. 5: Comparison of absolute conductivity values during a psychological experiment: standard macroelectrode versus IDA microelectrode.

3.2. EDA Ring

After detailed testing of the 1st design we decided to construct a practical sensor system in the form of small ring (Fig. 6) that could be useful in practical life. In this system, the driving generator provides a sinus signal wave with amplitude of 1.6–3 V and frequency 1 kHz to gold-plated 200 × 200 × 200 µm IDAE structure, where the equivalent current is measured and skin conductivity is calculated. The sampling rate can be set in range from 0.33 to 33 SPS (Samples Per Second). Total dimensions of EDA-ring are 20 × 20 × 5 mm and it is intended to be placed on the ring-finger of non-dominant hand.

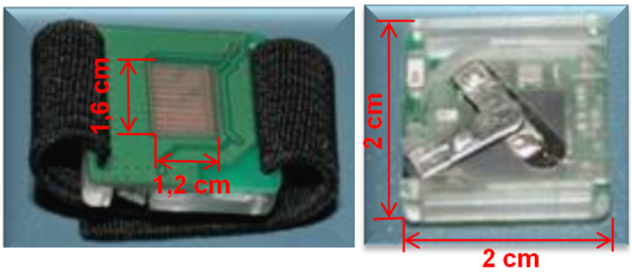


Fig. 6: EDA-ring: design.

Tab. 1: EDA-ring: technical parameters.

Technical parameters of sensor system	
Measuring signal Amplitude	1.6–3 V
Measuring signal Frequency	1 kHz (sinus)
ADC resolution	10 bit
Sample rate	0.33–33 SPS
Connectivity	ISM band 2.4 GHz, standard IEEE802.15.4
Transmission range	10 m at interior, 30 m at exterior
Supply voltage	1x CR 2032 (Lithium) 3 V, 12 - 24h stamina
Temperature range	0–70 °C
Weight	3 g
Dimensions	20 × 20 × 5 mm
Electrodes	IDAE 200 / 200 µm, Au galvanic plated on Cu

With this sensor system, a complex physiological research on group of 28 volunteers (mean age=23.5 years; SD=1.41; 13 males; 15 females) at Department of Psychology of Comenius University was done. The influence of body position and mental (Dual N-back [14]) or psychical activity (squatting) was evaluated. The experiment took about 30 minutes for each person.

In Fig. 7, there are shown exemplary EDA results from 3 persons. Each person provides individual signal dependent on his own characteristic features. The signals have similar tendencies – in relaxation, the conductivity is decreasing and under mental or physical burden (stress stimulus) it is increasing.

3.3. Smart Clothes

Last sensor system is designed as multifunctional holter and in this case, it was built-in into specially designed smart clothes. Holter (Fig. 8) is based on analog font-end TI ADS1292R and microcontroller ATxmega 128A3 (Fig. 8(a)). The ADS1292R is two-channel, 24-bit, delta-sigma analog-to-digital converter with a built-in programmable gain amplifier, internal reference and an on-board oscillator [15]. The system has been extended by the gyroscope L3GD20, accelerometer with magnetometer LSM303D [16] and barometer with temperature sensor BMP180 [17]. The latest design was reworked and improved, new firmware, controlling software and data transfer system were reprogrammed, acting now as USB flash device. The ADS1292R incorporates all features commonly required in portable, low-power medical electrocardiogram with sports and fitness applications. Power consumption of one channel is only 335 µW. Used version of analog-to-digital converter ADS1292R also includes a fully integrated impedance measurement function (Fig. 8(b)), where 32/64 kHz modulating square wave signal is driving the human body impedance trough known impedances Z_k. After demodulation and low pass (2–4 Hz) filtering we can obtain the EDA impedance Z_b (Fig. 3) [11]. Also in this version, compared to previous two systems, the impedance is measured at different frequency and so the relative sensitivity may be different.

Tab. 2: Smart clothes EDA monitoring system: technical parameters.

Technical parameters of sensor system	
Number of Channels	2
Programmable Gain	1, 2, 3, 4, 6, 8 or 12
Input-Referred Noise	8 mVPP(150-Hz BW, G = 6)
ADC resolution	24-bit, no data missing
Sample rate	125–8 kSPS
MCU	16 bit AVR, 32 MHz
SRAM	8 kbytes
Built-In	Impedance circuit (32, 64 kHz)
Acceleration sensor	3D ±2g/±4g/±8g/±16g (16 bit) 3–1600 SPS
Gyroscope	250, 500, 2000 dps (16 bit) 95–750 SPS
Magnetometer	2, 4 ,8 ,12 G (16 bit) 3–100 SPS
Barometer	300–1100 hPa (16 bit) 2–50 SPS
Data storage	Integrated 4 GB SD card
Output data format	CSV, EDF+
Power supply	1x Li-Pol 120 mAh
Connectivity	Micro USB
Optional	Bluetooth - 10 m range
Temperature range	0–70 °C
Electrodes	Disposable Ag/AgCl
Dimensions	37 × 25 × 15 mm
Weight	20 g
Next features: RGB LED and acoustic signalization, trigger button, real-time clock.	

In Fig. 9, there is our implementation of designed holter in smart shirt. Whole system, including electronics, fabric electrodes and T-shirt technology was prepared at our department. In comparison with the first two sensor systems, this design uses conductive fabric electrodes (MedTex P180, E130, Zel etc.). Therefore, the quality of the resulting signal is slightly lower.

The system was compared with standard macroelectrodes measurement systems [13]. Macroelectrodes were connected between index and ring finger and fabric electrodes of smart T-shirt measured signal in the area of brachioradialis muscle.

The comparison was performed by using the standard „Distraction" psychotests [14], where the signals were measured simultaneously. Results (Fig. 10) shows, that due to using different sensing method, the measured absolute values and sensitivities are different, but the responses, curve shapes, given by both approaches were similar. Under stress stimulus the conductivity is rapidly increasing and in relax it is rapidly decreasing. Also, the smart clothes offer lower sensitivity, mainly due to lower response of examined body areas to neural activity, but the suitability of smart clothes for long-term monitoring is incomparable with macroelectrodes.

4. Conclusion

Linear equations modelling and simplified power losses calculation have been used to predict approximate optimal DG size. Each location of DG at load buses has optimal size of DG for minimum power losses. It is noted that optimal size for optimal location is not necessary to be the same as optimal location for optimal size. The proposed method has acceptable accuracy with less time and memory consumption where it is crucial factor in real-time management of power grids. The loss reduction by properly placed and appropriate size of DG is one of the more significant findings to emerge from this study. With these benefits, control and assessment of large scales power grid will become easily predictable as more intermittent power sources, such as wind and solar, come online. Different IEEE test bus systems have been tested and results are validated with exact calculations.

Acknowledgment

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References

[1] BOQUETE, L., J. M. R. ASCARIZ, J. CANTOS, R. BAREA, J. M. MIGUEL, S. ORTEGA and N. PEIXOTO. A portable wireless biometric multi-channel system. *Measurement*. 2012, vol. 45, iss. 6, pp. 1587–1598. ISSN 0263-2241. DOI: 10.1016/j.measurement.2012.02.018.

[2] CARMO, J. P. and J. H. CORREIA. RF CMOS transceiver at 2.4 GHz in wearables for measuring the cardio-respiratory function. *Measurement*. 2011, vol. 44, iss. 1, pp. 65–73. ISSN 0263-2241. DOI: 10.1016/j.measurement.2010.09.027.

[3] BOUCSEIN, W., D. C. FOWLES, S. GRIMNES, G. BEN-SKAKHAR, W. T. ROTH, M. E. DAWSON and D. L. FILION. Committee Report - Publication recommendations for electrodermal measurements. *Psychophysiology*. 2012, vol. 49, iss. 8, pp. 1017–1034. ISSN 1469-8986. DOI: 10.1111/j.1469-8986.2012.01384.x.

[4] HANDLER, M., R. NELSON, D. KRAPOHL and C. R. HONTS. An EDA Primer for Polygraph Examiners. *Polygraph*. 2010, vol. 39, iss. 2, pp. 68–108. ISSN 1533-9793.

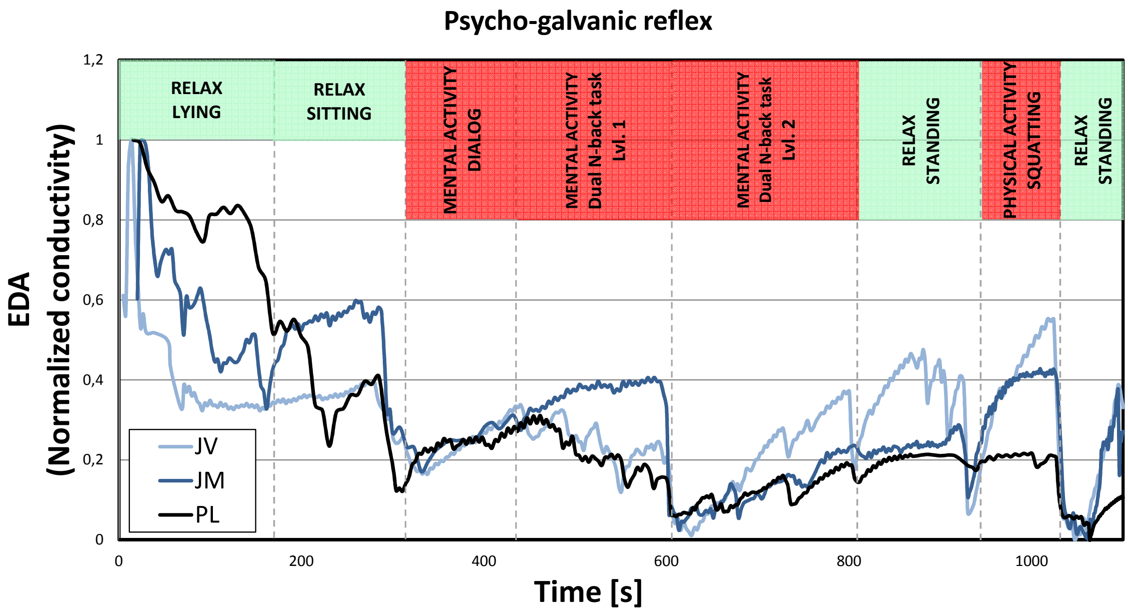
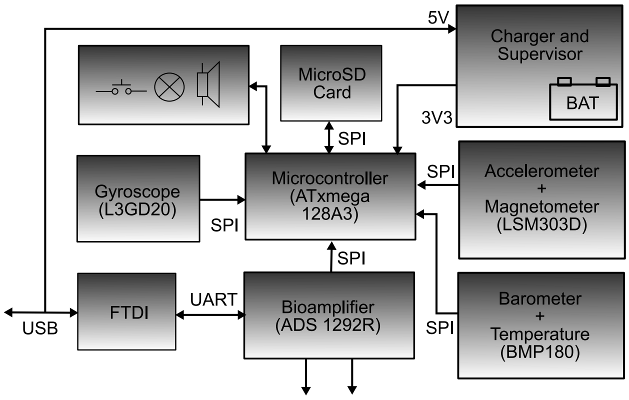
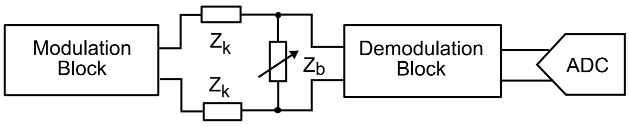


Fig. 7: Complex physiological monitoring: (3 probands).



(a) Block diagram of holter module.



(b) Schematic of EDA monitoring circuit.

Fig. 8: Smart clothes EDA monitoring system.

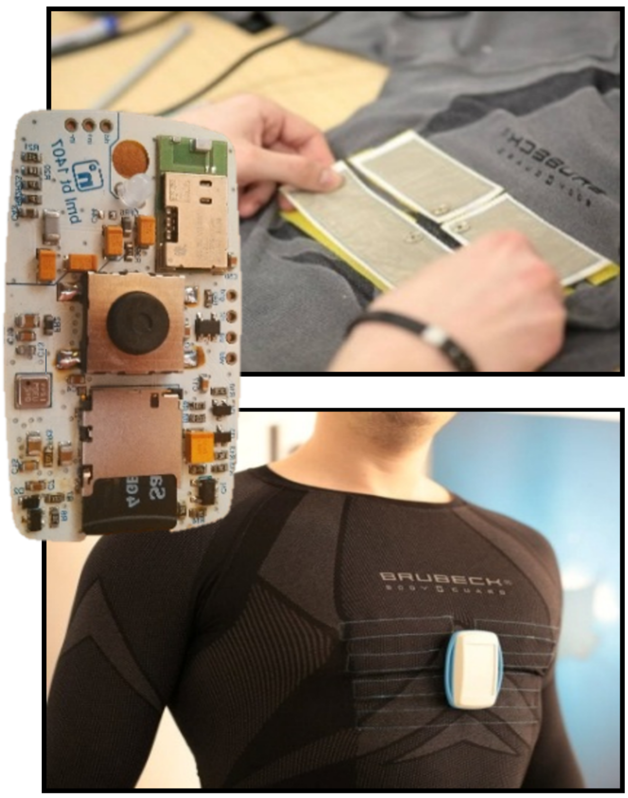


Fig. 9: Realization ofsmart clothes EDA monitoring system.

[5] WEIS, M., T. DANILLA, L. MATAY, P. HRKUT and J. KAKOS. Noninvasive Biomedical Sensors on the Biology - Interface of Human Skin. In: *7th International Conference on Measurement in Clinical Medicine*. Stara Lesna, 1995, pp. 89–91.

[6] IVANIC, R., I. NOVOTNY, V. REHACEK, V. TVAROZEK and M. WEIS. Thin film non-symmetric microelectrode array for impedance monitoring of human skin. *Thin Solid Films*. 2003, vol. 433, iss. 1–2, pp. 332–336. ISSN 0040-6090. DOI: 10.1016/S0040-6090(03)00389-4.

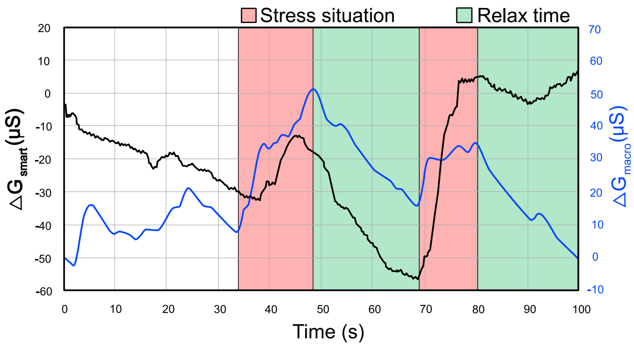


Fig. 10: Newton Raphson and linear method Comparison for 30 Bus.

[7] BALIN, A. K. and A. M. KLIGMAN. *Aging and the Skin*. New York: Raven Press, 1989. ISBN 978-0890047781.

[8] SHEPPARD, Jr., N. F., R. C. TUCKER and C. WU. Electrical conductivity measurements using microfabricated interdigitated electrodes. *Analytical chemistry*. 1993, vol. 65, iss. 9, pp. 1199–1202. ISSN 0003-2700. DOI: 10.1021/ac00057a016.

[9] WILHELM, K. P. Possible Pitfalls in Hydration Measurements. *Skin Bioengineering: Techniques and Applications in Dermatology and Cosmetology*. Basel: Karger, 1998. ISBN 3-8055-6519-4.

[10] VAVRINSKY, E., V. STOPJAKOVA, I. BREZINA, L. MAJER, P. SOLARIKOVA and V. TVAROZEK. Electro Optical Monitoring and Analysis of Human Cognitive Processes. In: *Semiconductor Technologies*. Rijeka: InTech, 2010, pp. 465–490. ISBN 978-953-307-080-3. DOI: 10.5772/8558.

[11] MAJER, L., V. STOPJAKOVA and E. VAVRINSKY. Wireless Measurement System for Non-Invasive Biomedical Monitoring of Psycho-Physiological Processes. *Journal of Electrical Engineering*. 2009, vol. 60, iss. 2, pp. 57–68. ISSN 1335-3632.

[12] 1 MSPS, 12 Bit Impedance Converter Network Analyzer. *Analog Devices* [online]. 2008. Available at: `http://www.analog.com`.

[13] BOUCSEIN, W. *Electrodermal Activity*. New York: Springer, 2015. ISBN 978-1-461-41126-0.

[14] OWEN, A. M., K. M. MCMILLAN, A. R. LAIRD and E. BULLMORE. N-Back Working Memory

Paradigm: A Meta-Analysis of Normative Functional Neuroimaging Studies. *Human Brain Mapping*. 2005, vol. 25, iss. 1, pp. 46–59. ISSN 1097-0193. DOI: 10.1002/hbm.20131.

[15] *Texas Instruments*. [online]. 2012. Available at: http://www.ti.com.

[16] *STMicroelectronics*. [online]. 2010. Available at: http://www.st.com.

[17] *Bosch Sensortec*. [online]. 2014. Available at: http://www.bosch-sensortec.com.

About Authors

Erik VAVRINSKY received his M.Sc. and Ph.D. degree from the Faculty of Electrical Engineering and Information Technology, Slovak University of Technology in years 2002 and 2005. He has carried out postgraduate studies in area of thin-film microsensors and electrochemical biosensors. In present, he participates at the University education process as a lecturer. His current research interests include design, characterization and testing of sensor systems and bio-monitoring devices. Dominant part of his multidisciplinary research is focused on human physiology.

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