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Wearable Instrument for Skin Potential Response Analysis in AAL applications

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Abstract – **A novel instrument able to acquire Skin Potential Response (SPR) signals is proposed; SPR is a branch of Electrodermal Activity (EDA) and consists of reading nervous electric pulses that arise when the sympathetic nervous system activates sweat glands as a reaction of an external stressing stimulus. Scientific literature shows that EDA is a good methodology to detect workload, increased stress level and many neurological diseases or addictions. In this paper, we present the design and characterization of a wearable, battery operated, wireless device which can acquire SPR data and can be integrated in a wireless sensor network for the AAL. The developed control panel is responsible of receiving the SPR data, plotting them in real time and, on a secondary screen, providing to the patient visual and auditory stimuli; as a further stimuli generator, an adaptive game controlled by the SPR signal has been developed. The synchronous plot of stimuli markers and SPR signal on the same graph can provide a very useful information for the SPR analysis.**

I. INTRODUCTION

Since the early 1900s, Electro-Dermal Activity (EDA) received a considerable interest in the field of psychoanalysis as a useful methodology to quantify the stress level of a patient subject to specific stimuli during a psychoanalysis session. In the recent past, EDA methodology has been applied to healthy patients, like athletes to improve their performances, or gamblers, to prevent gambling addiction, and to patients with different diseases. In particular, studies on children with Attention Deficit Hyperactivity Disorder (ADHD) [1] or Autism (ASD) [2], and on patients with major depressive disorder [3], showed that EDA can be used as a reliable indicator of emotions.

In this paper, we present a low cost, wearable, small and accurate device to perform SPR analysis; the SPR acquired data are in real time sent, via Bluetooth protocol, to a laptop that plots the data and applies to patients visual, auditory and gaming stimuli on a different screen. In particular, the proposed adaptive gaming can be very useful in Ambient Assisted Living (AAL) applications e.g. for patients with ADHD or ASD.

The principle of EDA is related to the sympathetic nervous system activity, which cannot be controlled by the patient and can be correlated to emotional status; EDA methodology is applied by measuring electrical quantities, which are related to the sweat glands activity controlled by the sympathetic nervous system. EDA is divided into two classes: exosomatic and endosomatic [4].

Exosomatic methodology is mainly referred to as Skin Resistance Response (SRR), Galvanic Skin Response (GSR) or Skin Conductance Response (SCR).

Endosomatic methodology is mainly referred to as Skin Potential Response (SPR), Galvanic Skin Potential (GSP) or Skin Potential Level (SPL). The main strength of the endosomatic methodology is the response speed, since the measurand is the electric pulse that activates the sweat gland that is the cause of sweating.

SPR measurements are commonly performed with bulky and expensive acquisition systems, electrocardiographic or electromyographic instrumentation [5 – 8], since low cost and wearable solutions are not presently available to our knowledge. In wireless sensor networks for AAL applications, in addition to acceleration, position, pressure and temperature sensors, stress sensors can be useful [9], and the developed instrument lend itself to be integrated in the network.

The proposed wearable instrument operates amplifying the electric nervous pulses that activate sweat glands on the hand; in particular, three electrodes are needed: one sensing electrode on the palm (where sweat glands are activated), one sensing electrode on the back of the hand, in order to pick and amplify the differential voltage, and a third reference electrode posed on the wrist which is used as a reference electrode. The nervous SPR pulses are characterized by amplitudes in the order of few millivolts and have band-pass bandwidth of [0.1, 10] Hz [4 – 6]; the input impedance of the device must be by far greater than skin output impedance which is in the order of megaohm.

II. SPR DEVICE DESCRIPTION

The device block diagram is shown in Fig. 1. Three electrodes are used to acquire the SPR signal: a reference

electrode is posed on the wrist providing the reference voltage V_{REF}, a second electrode is posed on the palm of the hand and the third electrode is posed on the back of the hand.
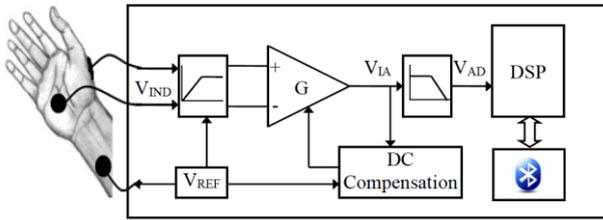

Fig. 1. Block diagram of the SPR wearable instrument.

The electrodes are Ag/AgCl electrodes used in electrocardiography and electromyography. Electrodes and their positioning on the hand is well known in literature. Muscle activity and body movements are the most important physiological source of artifacts. However, they produce burst of high frequency, with the power spectrum concentrated beyond 40-80 Hz [10].

As shown in Fig. 1 the differential voltage V_{IND} between the palm and the back electrodes is properly filtered and amplified; a Digital Signal Processor (DSP) acquires the amplified analog voltage, implements a digital notch filter to remove residual power line disturbs and sends via Universal Asynchronous Receiver Transmitter (UART) the acquired data to the Bluetooth module which converts UART packets into Bluetooth protocol. The device is battery operated and it needs a single Li-Ion cell as a power supply: a 3.7 V 1800 mAh lithium polymer battery has been chosen. Since the device current consumption is 45 mA during transmission, 40 hours of continuous data acquisition can be recorded, by far greater than usual SPR sessions (in the order of one hour).

A power supply section of the circuit (not shown in Fig 1) converts the battery voltage into +3.3 V through a buck DC-DC converter; a voltage V_{REF} = 1.65 V for the reference electrode and for the instrument is provided by a linear voltage reference.

The two signals acquired from the palm and back electrodes are high-pass filtered by a couple of passive, RC, first order filters, with the aim of removing the common mode DC voltage that may be present on the skin. In order to minimize, in the pass-band, the load uncertainty at a level less than 1%, the input impedance of the instrument has been set to 100 MΩ, since the skin impedance is of the order of 1 MΩ. The output of the filters are connected to the instrumentation amplifier which amplifies the difference between the palm and back electrodes; the gain G is set to 80, since the expected largest SPR pulse is in the range of ±20 mV and must be converted into 3.3 Vpp.

The selected instrumentation amplifier voltage offset (V_{OS}), bias (I_B) and offset (I_{OS}) currents can strongly affect the output signal V_{IA}, since the filters DC output impedance is 100 MΩ. The selected instrumentation amplifier (MCP 6N11 from Microchip) is characterized by typical values I_B = 20 pA, I_{OS} = 1 pA and V_{OS} = 0.5 mV, CMRR = 115 dB. Naming V_{OS}^* the undesired voltage at the amplifier input due to V_{OS}, I_B and I_{OS}, the worst case V_{OS}^* is approximately 0.7 mV, corresponding to a worst-case output DC error $\delta V_{IA} = G \cdot V_{OS}^* \cong 60$ mV. This value is not acceptable especially with a small supply range of 3.3 V. To overcome this problem, the "DC compensation" block in Fig.1 has been added. This block consists of an inverting integrator, which integrates the quantity V_{IA}-V_{REF} in order to shift the DC component of the instrumentation amplifier. Defining τ_{INT} the integrator time constant and τ_F the high-pass filter time constant, the output of instrumentation amplifier V_{IA} is, in the Laplace domain,

$$V_{IA}(s) = V_{REF} + \frac{\tau_{INT}s}{1 + \tau_{INT}s} \cdot GV_{OS}^* + $$
$$+ \frac{\tau_{INT}\tau_F s^2}{(1 + \tau_{INT}s)(1 + \tau_F s)} \cdot GV_{IND}(s). \tag{1}$$

The above equation shows that the DC quantity V_{OS}^* does not affect the output because it is multiplied by a high pass transfer function.

The output of instrumentation amplifier is connected to a low-pass, anti-alias filter. This filter is a third order low-pass filter having three coincident poles each at frequency $(2\pi\tau_{AA})^{-1}$ = 72 Hz, being τ_{AA} the time constant of each pole. Thus, the expected overall frequency behavior is a band pass system with pass band gain of 38 dB, a lower cutoff frequency of 0.08 Hz with a slope of 40 dB/dec and an upper cutoff frequency of 40 Hz with a slope of -60 dB/dec.

The conditioned signal V_{AD} is then sent to an analog input of the DSP. The chosen DSP (DSPIC 30F3013 from Microchip) has an on board 12 bit A/D converter and operates at 8 Mega Instructions per Second; the sample rate has been set to 200 Sa/s.

Fig. 2. SPR wearable instrument.

A digital notch filter is implemented in order to reduce the EMI power line noise, which, otherwise, can affect the low-level signal. The filter is designed as a second order

Butterworth filter with the notch frequency centered at 50 Hz. The filter has been implemented with 32-bit double precision in order to achieve the highest frequency accuracy. The processed data are finally sent via UART protocol to the Bluetooth module (BTC2-DATA from Adeunis RF). The data transfer Baud Rate has been set at 19.2 kbps in order to allow the data flow without crowding the channel.

The SPR wearable instrument has overall dimensions of $52{\times}56$ mm^2 as shown in Fig. 2.

The data sent by the SPR device can be plotted, saved and post processed by a developed control panel installed on a laptop. The software has been developed in .NET environment. The control panel is responsible to communicate with the SPR device, to plot real time the SPR signal, to insert graphical markers when stimuli are applied to the patient, to save the SPR data, the markers and markers comments. A first developed utility allows providing to the patient, in another window on a different screen, a slide show or a movie, which automatically projects different images or videos and sinchronously puts markers on the SPR graph. This utility can be used to evoke to the patient different emotions and record the SPR reactions.
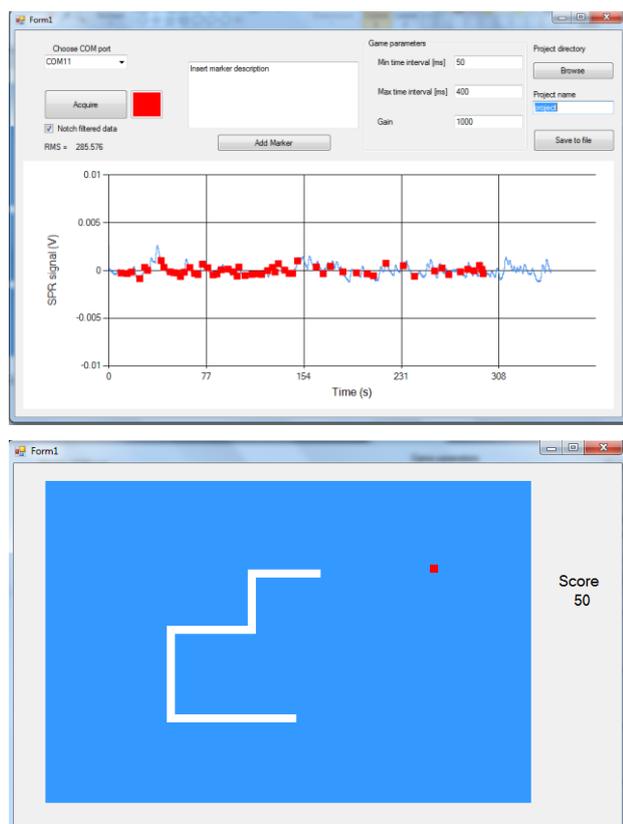

Fig. 3. Top: control panel of the SPR acquisition software. Bottom: the "snake" game window

A second developed utility, running on a different screen, is the "snake" game (Fig. 3). The "snake" can operate in two different modes: in open loop, at constant speed, as a stimulus to the patient while the instrument records the SPR signal, or in closed loop, with the snake speed modulated by the SPR signal used as a biofeedback. The higher the stress of the patient, the higher will be the SPR signal and the lower will be the snake speed; every time that the snake eats an apple, a marker is put on the SPR graph. The game ends when the snake touches the borders of the window.

Simple software modification are finally possible, for the integration in a multisensor system for AAL as such described in [11].

In order to reduce the computational weight of the real time plot, it has been decided to acquire 10 data long packets, so the graph is refreshed every 50 ms instead of 5 ms. At every event containing a 10 data long packet, a time stamp with the laptop system clock is created. It has been chosen to use the system clock time stamp instead of the number of samples multiplied by the sample rate because the uncertainty on sample rate propagates on the uncertainty on the instants when the data are received and plotted; for example, in one hour acquisition with 0.1 % uncertainty on the sample rate, there will be a time base drift of almost 4 s: this time misalignment is not acceptable and justifies the choice of recording the system clock time stamps. The performances of the software have been tested for 90 minutes in terms of CPU and RAM usage, providing, on a low performance laptop (Pentium T2310 with 2 GB RAM) a CPU usage of 35% and a memory occupation of 30 MB; during the test also the uncertainty on the time base of the plot has been verified to be in the order of milliseconds.

III.   EXPERIMENTAL RESULTS AND DEVICE CHARACTERIZATION

In this Section, the performances of the entire SPR system are described. The overall performances have been characterized in terms of linearity, gain accuracy and bandwidth. In order to characterize the linearity and the gain accuracy of the instrument, the following measurement procedure has been set up.

The output of an Agilent 33220A waveform generator has been connected to a resistive attenuator (attenuation factor A=0.00813) whose output impedance has been set to 1 MΩ to simulate skin impedance; the generator has been set to provide a 10 Hz sine wave with an amplitude V_G variable from 100 mVpp to 5 Vpp with 50 mV step size. The output of the attenuator has thus provided an amplitude $V_{IND} = A{\cdot}V_G$ variable from 800 μVpp to 40 mVpp with 400 μVpp step size and a total amount of N = 99 steps.

The quantized voltage provided by the A/D converter $V_Q = V_{AD}-V_{REF}$ has been acquired for a duration of 10 periods of the input sine wave and, for each ith step, the measurement has been repeated 10 times.

At the same time, with an Agilent 34401A multimeter,

the RMS value of V_G has been acquired 10 times for each ith step.

The gain G has been estimated by least squares linear fitting as

$$G = \frac{\sum_{i=1}^{N}[(AV_{Gi} - A\overline{V_G}) \cdot (V_{Qi} - \overline{V_Q})]}{\sum_{i=1}^{N}(AV_{Gi} - A\overline{V_G})^2}. \qquad (2)$$

Let consider that the uncertainty contributions are $u(V_{Qi})$, $u(V_{Gi})$ and $u(A)$ obtained from the A/D datasheet and from the instrument manual. From (2), since

$$u^2(G) = \sum_{i=1}^{N}\left(\frac{\partial G}{\partial V_{Qi}}u(V_{Qi})\right)^2 \\ + \sum_{i=1}^{N}\left(\frac{\partial G}{\partial V_{Gi}}u(V_{Gi})\right)^2 + \sum_{i=1}^{N}\left(\frac{\partial G}{\partial A}u(A)\right)^2, \qquad (3)$$

it has been pointed out that $G = 80.05 \pm 0.04$. Fig. 4 shows the linearity of the SPR device; the linearity error is calculated as the difference between data and the least square linear regression, and is less than 0.5%. Error bars represent the uncertainty of each point.

The SPR device has been characterized also in terms of frequency response. Similarly to the linearity characterization, the output of an Agilent 33220A waveform generator has been connected to a resistive attenuator whose output impedance has been set to 1 MΩ to simulate skin behavior; the generator has been set to provide a sine wave of amplitude 5 Vpp, corresponding to 40 mVpp at the output of the attenuator, with frequencies from 0.01 to 1000 Hz logarithmically spaced with 10 points/decade. The voltage V_Q has been acquired for each frequency step on a duration of ten periods of the sine wave and its RMS value is calculated. The frequency response is then evaluated in terms of $V_Q(j\omega)/V_{IND}(j\omega)$. The overall bandwidth results in the range [0.08, 40] Hz and the digital notch filter allows a power line frequency rejection of 35 dB. The gain flatness in the pass-band is better than 0.5 dB. As previously stated, this frequency response minimizes the artifacts due to the muscular activity. The chosen sample rate is, as previously said, 200 Sa/s; since the sample frequency is generated with an internal counter of the DSP, the uncertainty of the sample rate depends both on the counter resolution and on the DSP clock stability, resulting 200.0 ± 0.6 Sa/s.
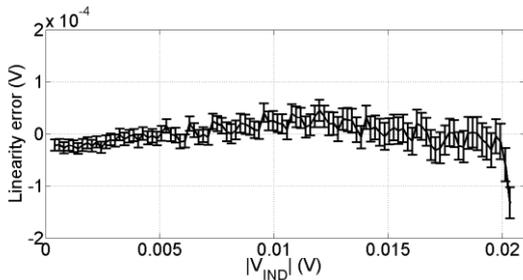

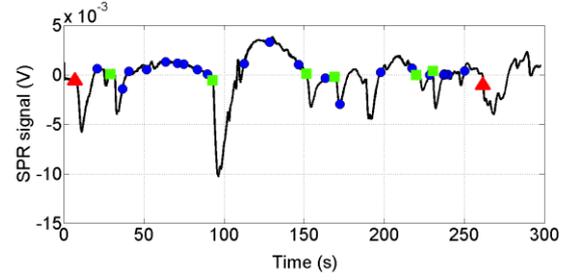Fig. 4. Linearity error of the SPR wearable instrument.

Fig. 5. SPR signal (solid line) during a "snake" game: red triangular markers represent the game start/end, blue circular markers represent the easy apples eaten, green square markers represent the difficult apples eaten

As an example of application of the developed device, Fig. 5 shows the SPR data acquired from a healthy subject, female, 37 years old, while playing at the snake game on the second screen. Solid line represents the SPR signal, red triangular markers represent the start and the end of the game, blue circular markers represent the "easy" apples (the ones in the middle of the screen) and the green square markers represent the "difficult" apples (the one on the borders of the screen). In order to understand the SPR graph, let consider that the positive wave of the skin potential response is an indicator of the defense reflex, while the negative wave is an indicator of the orienting reflex [12]. As it can be seen, after the start there is a negative peak of the SPR signal because the snake starts running inducing arousal to the player; similarly, the negative peaks appear also after each difficult apple, when the stress is higher, since the player must move quickly away from the borders. Finally, the peak appears strongly also when the game is over thus frustrating the player. In correspondence of these peaks, the SPR signal reduced the snake speed in order to enhance the relaxation of the player.

IV. CONCLUSIONS

We developed a low cost, wearable, wireless instrument for acquiring SPR data. The description of the system composed by an analog section, a firmware section and a software section was presented. The instrument can be very useful to detect or study many forms of addiction or neurologic disease; since commercial wearable solutions for endosomatic EDA methodology are not available, the proposed instrument can be a good aid for patients in AAL applications.

REFERENCES

[1] Rosenthal, R. H., Allen, T. W., "An examination of attention, arousal, and learning dysfunctions of hyperkinetic children", Psychological Bulletin, Vol 85(4), Jul 1978, pp. 689-715.
[2] Welch K. C., "Physiological Signals of Autistic Children Can be Useful", IEEE Instrumentation and

measurement magazine, February 2012, pp. 28-32.

[3] Boettger M. K., Greiner W, Rachow T., Brühl C., Bär K. J., "Sympathetic skin response following painful electrical stimulation is increased in major depression", Pain, vol. 149 No. 1, 2010, pp. 130-134.

[4] Boucsein, W. "Electrodermal activity", second edition, 2012, Springer.

[5] Kucera P., Goldenberg Z., Kurca E., "Sympathetic skin response: review of the method and its clinical use", Bratisl Lek Listy, 2004, vol. 105 No. 3, pp. 108–116.

[6] Ohme R., Reykowska D., Wiener D., Choromanska A., "Analysis of Neurophysiological Reactions to Advertising Stimuli by Means of EEG and Galvanic Skin Response Measures", Journal of Neuroscience, Psychology and Economics, vol. 2 No. 1, 2009, pp. 12-31.

[7] Jabbari A., Johnsen B., Grimnes S., Martinsen O.G., "Simultaneous measurement of skin potential and conductance in electrodermal response monitoring", Journal of Physics: conference series, vol. 224 No. 1, 2010, doi:10.1088/1742-6596/224/1/012091.

[8] Boucsein W., Fowles C., Grimnes S., Ben-Shakhar G., Roth W.T., Dawson M.E., Filion D.L., "Publication recommendations for electrodermal measurements", Psychophysiology, vol. 49, 2012, pp. 1017-1034.

[9] McAdams E. T. et al., "Biomedical Sensors for Ambient Assisted Living" in S.C. Mukhopadhyay and A. Lay-Ekuakille, "Advances in Biomedical Sensing, Measurements, Instrumentation and Systems", Springer, 2010, pp. 240-262.

[10] A. Van Boxtel, "Optimal signal bandwidth for the recording of surface EMG activity of facial, jaw, oral, and neck muscles", Psychophysiology, Vol. 38, 2001, pp. 22-34.

[11] V. Bianchi, F. Grossi, I. De Munari, P. Ciampolini (2012) "Multi Sensor Assistant: A Multisensor Wearable Device for Ambient Assisted Living", Journal of Medical Imaging and Health Informatics, 2012, vol. 2 No 1, pp.70- 75.

[12] D. C. Raskin, H. Kotses, J. Bever, "Autonomic indicators of orienting and defensive reflexes", Journal of Experimental Psychology, Vol 80 (3, Pt.1), Jun 1969, pp. 423-433.