Measurement 75 (2015) 111–121

Contents lists available at ScienceDirect

Measurement

journal homepage: www.elsevier.com/locate/measurement

Design and characterization of a real-time, wearable, endosomatic electrodermal system

A. Affanni [a,*], G. Chiorboli [b]

[a] University of Udine, Dept. of Electrical, Management and Mechanical Engineering, Via delle Scienze 208, 33100 Udine, Italy
[b] University of Parma, Dept. of Information Engineering, Parco Area delle Scienze 181/a, 43100 Parma, Italy

ARTICLE INFO

Article history:
Received 19 February 2015
Received in revised form 10 July 2015
Accepted 21 July 2015
Available online 10 August 2015

Keywords:
Psychophysiology
Electrodermal activity
Skin potentials
Wearable instruments
Measurement uncertainty
Traceability to SI

ABSTRACT

This paper presents the design and characterization of a compact wearable system for long-term assessment of skin potential response, with the aim of monitoring mental stress in a variety of applications. Literature reports that the expected skin potential has peak-to-peak amplitudes of few millivolts in the frequency band [0.1, 10] Hz. The designed system is characterized by a slightly wider bandwidth of [0.08, 40] Hz, and it is based on a 12-bit ADC working with a sampling rate of 200 Sa/s, which can be increased up to 3.5 kSa/s. Data can be continuously acquired for up to 40 h with a battery of 3.7 V/1800 mAh. A Graphical User Interface was also developed for the host computer in .NET framework. The system, to our knowledge the first example of wearable endosomatic electrodermal activity sensor, joins to several skin conductance wearable measuring systems recently proposed in literature, and opens up opportunities for future comparisons of endosomatic and exosomatic responses in real life.

The device is thoroughly characterized in accordance with the state-of-the-art of the metrological research in the field.

1. Introduction

The measurement of emotional response through analysis of the Electro-Dermal Activity (EDA) has been employed since the nineteenth century [1,2] to quantify attention and stress of patients subject to specific stimuli, or to detect deception for forensic purposes. In the past decade EDA-based techniques have been applied in psychophysiology research to improve athlete performances [3], to prevent gambling addiction [4–6], to monitor patient with attention deficit hyperactivity, autism, and depressive disorders [7–9], or to detect mental stress in combatants [10], in traders [11], in vehicle drivers, in pilots, and in human and robot interaction [12–14].

EDA analysis has particular attraction in several applications since it is non-invasive, cost-effective, and relatively sensitive when properly implemented. EDA is correlated to the sympathetic nervous system activity which can not be controlled by the patient and, in particular, it depends on the sweat glands behavior and on the change in pore size. As soon as the autonomic nervous system activates these glands which secrete sweat, the resistance between two electrodes dramatically drops. Since, at the same time, it is also possible to measure skin potential between these electrodes, EDA measurements can be classified in *exosomatic*, if a current, on which the resistance or conductance measurement is based, is introduced from outside, or *endosomatic* if the electric pulses generated by the sympathetic nervous system are the measured signals. In the first case one can measure the Skin Conductance Level (SCL) or Response (SCR, also known as Skin Resistance Response – SRR, or Galvanic Skin Response – GSR), and in the second

* Corresponding author.
E-mail addresses: antonio.affanni@uniud.it (A. Affanni), giovanni.chiorboli@unipr.it (G. Chiorboli).

http://dx.doi.org/10.1016/j.measurement.2015.07.047
0263-2241/

case Skin Potential Level (SPL) or Response (SPR). The EDA is usually measured by two electrodes placed on the palm of the hand (or of the feet) where the density of sweat glands is the highest (>420 cm^{-2}), with a third electrode in the forearm for inactive voltage reference which is necessary in endosomatic measurements [15].

Endosomatic recordings are likely more influenced by bias potential artifacts and by skin stretching than exosomatic ones, which can be obtained and interpreted more easily, since in the endosomatic recording the internal voltage source is less controllable, potentially leading to less accurate measurements. Even if some Authors point out that endosomatic responses are slightly faster than exosomatic ones [16], recent works demonstrate that endosomatic measurements are more sensitive to skin hydration and to temperature differences of the measuring sites, and that the response time depends on hydration of the skin and the filling of the sweat ducts [17]. Moreover, [15] observes that a pretreatment of sites used for exosomatic recording is unnecessary, while the site used for the inactive electrode in endosomatic recording must be pretreated, and the stratum corneum has to be removed to diminish the potential difference between the site and the body core underneath the electrode. This invasiveness creates possibilities for skin infections, which are most problematic for long term measurements. Disposable electrodes that do not require skin preparation, can simplify EDA measurements, against the potential risk of reduced SPR amplitudes if the instrument input impedance is not high enough; in the case of the proposed system, the input impedance is by far higher than skin impedance thus improving the SPR signal reading also without skin pretreatment. By contrast, endosomatic recording is more "physiological", since it is free from any external current applied to the system, an undoubted advantage in long-term recordings. Moreover, electrode polarization is prevented since no external current is applied.

In [15], the author observes that systematic comparison studies between endosomatic and exosomatic EDA recording are rare. In the past, the majority of EDA studies have been performed with exosomatic methods, probably because of the problems arising from the interpretation of the skin potential which can be mono, bi, or triphasic, but endosomatic recording might be a valuable tool for basic research, since the signal analysis may give rise to interpretations with respect to their different psychological meaning [15]. Finally, neurologists predominantly use endosomatic SPR recordings, which they call "sympathetic skin response" (SSR) [18].

Nowadays, technology evolution has made possible the use of wireless sensor for biomedical applications at low cost with high effectiveness, as observed in several recent works, where a lot of information is available about the design issues and challenges [19–23].

In the past decade, many EDA measurement systems, which were previously performed only in laboratory, with bulky and expensive electrocardiographs (ECG) and electromyographs (EMG) [18,24,25], were implemented with wireless technology in wearable devices [10,11,14,26–28], allowing research no more limited to measurement performed over a short period of time in laboratory. Notice that all the cited systems perform exosomatic measurements; to our knowledge, low-cost, endosomatic SPR wearable systems are not presently available neither as commercial or research product, even if, in theory, any wearable ECG or EMG system might be used. However, general purpose, multichannel ECG or EMG systems are not optimized for recording SPR, evoked by internal or external arousal stimuli, so that dedicated systems seems to be convenient. Nevertheless, useful design suggestions can be found in [16,21,29–32], where wearable ECG or EMG systems were described.

In this work we describe 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 host laptop or tablet that plots the data and applies to patients visual, auditory and gaming stimuli on a different screen. In Section 2 the system, originally proposed in [33], is described with more detail on hardware configuration and firmware for stimulation and data analysis.

Notice that EDA measurements, such as practically all measurements involving human perception and interpretation [34], are affected by a lot of variables: subject variability, which produce different responses to the same stimuli, environmental parameters such as temperature, which influences hand sweating, surface electrode placement, and electronic instrument accuracy. As recently observed, EDA measurement systems "are rarely metrologically evaluated" in accordance with [35]. Refs. [36,37] present a general procedure for the evaluation of EDA measurement systems based on exosomatic SCR, which can be adapted with slight modifications to endosomatic SPR. Therefore, Section 3 evaluates the proposed prototype with a procedure similar to [36]. The main difference consists in the fact that the precision resistance decade, used to simulate skin resistance in the exosomatic measurement systems, has to be substituted by a variable voltage reference to simulate skin potential. With respect to skin-conductance phasic amplitudes of about 2 μS, which are typical of exosomatic setup, in endosomatic measurements SPR pulses are characterized by amplitudes in the order of few millivolts, up to 20 mV, with spectral content in the frequency band [0.1, 10] Hz [15,24,38] ([0.05, 35] Hz in [39]), so that the dynamic evaluation of the system, in part performed by comparison with reference general purpose instruments like EMGs or acquisition boards, has to be carried out with attention to limit the reference instrument bandwidth.

2. System description

The typical architecture of wearable EDA systems for patients' remote monitoring consists of three main building blocks: (A) sensing and analog circuit front-end for signal conditioning of physiological data; (B) sampling and digitizing hardware; and (C) data logging and communication hardware and software to relay the acquired data to a remote device. Finally, data analysis tools are added to extract relevant information.

The design of the proposed SPR sensor is illustrated below, accordingly to this typical architecture. Fig. 1 shows