PC-based instrumentation for electrodermal activity measurement

Christian Tronstad[1], Sverre Grimnes[1,2], Ørjan G. Martinsen[2,1], Vegard Amundsen[2] and Slawomir Wojniusz[3,4]

[1]Department of Clinical and Biomedical Engineering, Oslo University Hospital, Rikshospitalet, Norway.
[2]Department of Physics, University of Oslo, Rikshospitalet, Norway.
[3]Department of Neuropsychiatry and Psychosomatic Medicine, Oslo University Hospital, Rikshospitalet, Norway.
[4]Faculty of Health Sciences, Oslo University College, Norway

E-mail: christian.tronstad@gmail.com

Abstract. A PC-based EDA measuring system is presented. The system is composed of a laptop with a PCMCIA DAQ-card running LabVIEW® software, a small front-end with a dual op-amp IC and a few passive components, and three skin-surface electrodes. The electrode system gives a monopolar measurement below the measuring electrode regardless of the electrode sizes, unless very small. Usage of the system is demonstrated by measurements from a mental stress experiment on 17 volunteers. There was a significant correlation ($R=0.51$, $p<0.001$) between the self-assessed stress-level and the EDA response frequency. The system allows easy on-site customization in software of measuring parameters, signal-quality monitoring and non-linearity detection in real time. We believe that the most suited use for the system is for stationary experimental purposes where this flexibility is desired. The system is easy to reproduce by engineers interested in doing EDA research.

1. Introduction

The activity from the sympathetic nervous system (SNS) regulates the secretory part of the sweat glands, which in turn changes the electrical properties of the skin due to the filling of electrolyte-containing sweat in the ducts. These changes are mostly in the stratum corneum (SC) layer of the skin, and by measuring the SC conductance we can obtain a link to the SNS activity. This phenomenon is usually called electrodermal activity (EDA). The best sites for measuring EDA are known to be the palmar and plantar dermatomes, which probably differ from the innervation of sweat glands on the rest of the body [1]. There is a wide range of uses for EDA recordings, with applications within psychophysiology, ambulatory monitoring, intensive care and biofeedback among others.

EDR instrumentation generally requires very few electronic components and can be built into small devices with low current consumption [2,3]. However, for stationary experimental purposes where a laptop is the monitor of choice, the instrumentation is easily realized by a DAQ card and a few off-the-shelf components. This paper presents one such solution. EDA is known as an indicator of stress [1], and measurements from usage of the system in a mental stress experimental setting are presented.

2. Materials and methods

2.1. Instrumentation

The measuring system is composed of three main parts, illustrated in figure 1:

- Three skin surface gel electrodes.
- Front-end electronics consisting of one dual op-amp IC, 5 resistors and two capacitors.
- A laptop with a National Instruments 6062E DAQ-card and LabVIEW® software for signal processing, display and data storage.

A low amplitude sine voltage generated by the D/A of the DAQ is applied to the skin through the E and R electrodes. Remote sensing at the current-less R electrode causes the right op-amp to drive a current through E which keeps the potential under R equal to the excitation sine, thus eliminating the SC impedance below E [4,5].

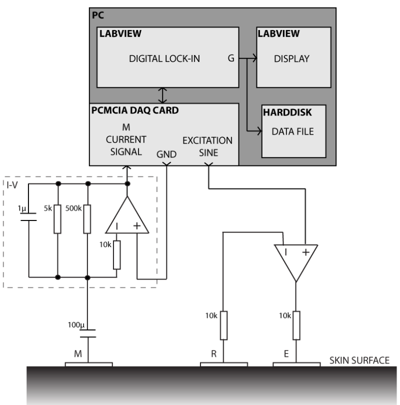

Figure 1. PC-based EDA instrumentation

Figure 2. Photo of the system in use

Because the impedance of the wet tissue is negligible compared to that of the SC, the only current-opposing element that is left is the impedance of the SC below the M electrode. Thus, the measurement is monopolar regardless of the effective electrode areas (EEA) (unless very small [4]), and confined to the SC of the EEA below M. The major contributor to changes in this impedance is the filling and reabsorption of sweat in the sweat ducts. Measuring with AC eliminates the endogenous potentials generated by the skin or electrodes. The I-V dashed box on the left constitute a current to voltage converter with a *Fc=200 Hz* RC lowpass filter. The 100µ capacitor blocks DC which may saturate the op-amp and its impedance needs to be very small compared to the SC below M, thus the large capacitance. The 10k resistors are for safety in order to reduce the current through the skin should a breakdown in the op-amp occur. The digitized current signal is demodulated by the phase-corrected excitation sine in software, giving the AC conductance G as the real part of the current divided by the excitation voltage. The voltage input and output of the DAQ is synchronized with the same sample clock. The op-amp power is supplied by the DAQ-card. Figure 2 shows a photo of the total system.

2.2. Software

Apart from the front-end with the op-amp IC as the only active component, all the measurement tasks are handled digitally in software. The digital lock-in technique is similar to that presented in [2]. For research purposes, the aim is to preserve as much as possible of the EDR signal information, because the EDR parameters may vary depending on the feature extraction methods when the sampling frequency Fs is decreased below a certain threshold. Very little signal energy is lost when Fs is as low as $1Hz$, but depending on the signal feature extraction method, a higher Fs may be required. The requirement of using a low excitation frequency Fe for the EDA measurement [6] decides the compensation between Fs and the signal quality by means of the number of periods N for one measurement. The software will also need a certain processing time Tp between each measurement for signal processing, storage etc, giving the following limit for the maximum Fs:

$$Fs = \left(\frac{N}{Fe} + Tp\right)^{-1} \tag{1}$$

2.3. Experimental protocol

17 volunteers (11M, 6F) who gave informed consent were subjected to three epochs of stress exposure, each lasting 10 min: Relaxation with classical music, quietly reading a text while provoked by irregular bursts of white noise and anticipation of public speaking while still receiving the white noise. The M electrode was placed on the hypothenar area of the palm, while the R and E electrodes were placed on the underarm, but the placement of these two is not critical. The skin AC conductance was sampled at $Fs=8Hz$.

2.4. Data analysis

As a measure of EDA response frequency, the number of EDA peaks (EDA TC) within a central five minute window within all the epochs was computationally determined by a top counting algorithm (A.1) and compared with a self-assessed stress score (1-10) of the volunteers using the Pearson product moment correlation. Comparison between the epochs was done using one way repeated measures ANOVA and the Holm-Sidak t-test. Normality was assumed based on the Shapiro-Wilk test.

3. Results

The SNR of the conductance measurement was 74 dB with a phase error of 0.02°, tested in a lab with a high degree of electromagnetic interference. The linear dynamic range of the system with the configuration as in fig.1 was 0 to approx 200µS. This also depends on the type of op-amp used.

There was a significant correlation between the EDA TC and the subjective stress-score ($R=0.51$, $p<0.001$). As shown in fig.3, both the EDA TC and the subjective score was significantly higher ($p<0.05$) during epoch 2 and 3 compared to epoch 1. Fig.4 shows an example measurement from epoch 2.

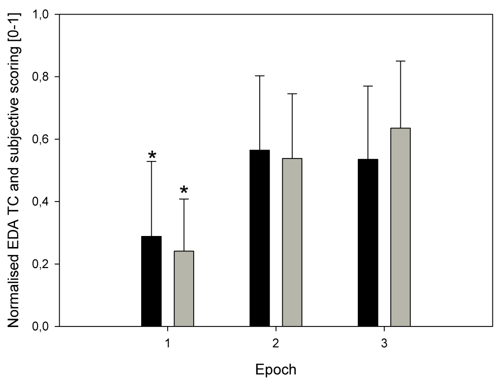

Figure 3. Mean and standard deviation of the EDA TC (black) and the subjective scoring (grey) for the three epochs. *p<0.05.

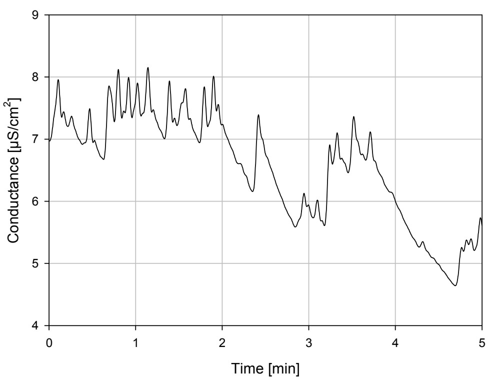

Figure 4. Example measurement from epoch 2.

Discussion

Expectedly, epoch 2 and 3 led to a higher self-assessed stress level than epoch 1. Although the protocol failed to produce a significant difference in mean stress level between the epoch 2 and 3, large intraindividual differences between these epochs were frequently observed. Thus, the correlation between the EDA parameter and the self-assessed score may be a better suited measure of the link between EDA and mental stress than the ANOVA. However, the variation in intersubjective understanding of the scoring will impact the correlation. Although the first 2.5 minutes of the epochs were excluded in the analysis to allow habituation to the noise, part of the EDA responses in epochs two and three may be directly caused by the acoustic stimulus, and whether this represents stress or not can be debated.

With this setup, measuring parameters such as the excitation voltage and frequency, phase correction, filtering and calibration is easily modified by a few mouse-clicks. In addition, the system enables automatic detection of artefacts or non-linearity from the raw input current signal, which can be important for this type of electrode system [4]. The simplicity of a laptop with only a cable, a front-end box which can be made very small and electrodes as the total measuring system makes it convenient for use in an experimental setting where a laptop is the monitor of preference. This system would be easy to reproduce by biomedical engineers interested in doing EDA research and it would work just as well with most other types of DAQ-cards, e.g. the USB-DAQ which has become more popular lately.

For electrical safety, it is important to run the laptop on battery while measuring on live subjects in order to obtain a body floating connection with the equipment.

Appendix

With FFT analysis revealing negligible energy above 1 Hz, the window length was set to $1Hz \cdot Fs = 8$ points and the EDA TC was calculated as:

$$TC = \sum_{i=Fs/2}^{N-Fs/2} \left(pos\left(G[i] - G\left[i - \frac{Fs}{2}\right] - \Delta G\right)\right.$$

$$\left. \cdot \prod_{j=0}^{\frac{Fs}{2}} \left(pos(G\,[i-j] - G[i-j-1]) \cdot pos(G[i+j] - G[i+j+1])\right)\right) \qquad (A.1)$$

$$Where\ pos(x) = \begin{cases} 1, & x \geq 0 \\ 0, & x < 0 \end{cases}, \Delta G = \frac{0.01\mu S}{cm^2}$$

References

[1] Boucsein W. 1992. *Electrodermal Activity*. NewYork: Plenum Press. 460 pages, ISBN-10: 0306442140, ISBN-13: 978-0306442148.

[2] Tronstad C., Gjein G.E., Grimnes S., Martinsen Ø.G., Krogstad A-L., Fosse E. 2008 Electrical measurement of sweat activity. *Physiol.Meas*. **29** 407-415

[3] Tronstad C., Martinsen Ø.G., Grimnes S. 2008 Embedded instrumentation for skin admittance measurement. *Engineering in Medicine and Biology Society, 2009. EMBS 2008. 30th Annural International Conference of the IEEE, Vancouver*. 2373-2376

[4] Grimnes S., Martinsen Ø.G., Tronstad C. 2009. *Noise properties of the 3-electrode skin admittance measuring circuit. 4th European Conference of the IFMBE, Antwerp. Springer Berling Heidelberd, IFMBE Proceedings* **22** 720-722.

[5] Grimnes S. 1983 Impedance measurement of individual skin surface electrodes. *Med.Biol.Eng.Comput*. **21** 750-755

[6] Martinsen Ø.G., Grimnes S., Haug E. 1999 Measuring depth depends on frequency in electrical skin impedance measurements. *Skin Res. Technol.* **5** 179-81.