mEDA: Mobile DC-EDA Circuit Validation

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Abstract - Electrodermal activity (EDA) provides a direct indicator of sympathetic nervous system arousal through changes in skin conductance. However, wearable EDA sensing poses challenges such as inconsistent skin contact, electrode impedance variability, motion artifacts, and power constraints. To address these issues, this study presents mobile EDA (mEDA), a compact device driven by a stabilized direct-current source. A validation study was conducted on ten healthy adult participants in a time-synchronized protocol to collect data from BIOPAC and mEDA concurrently. mEDA recordings employed gel electrodes for P1–P5 and dry (textile) electrodes for P6–P10, while the BIOPAC MP160 system used gel electrodes for all participants. Participants underwent a 30-minute protocol of resting, deep breathing, and three cognitive tasks. The preprocessing pipeline consisted of low-pass filter and artifact (sharp peaks and flat line) removal. Cleaned signals were converted into frequency domain components for decomposition into low and high frequency components, skin conductance level (SCL), and skin conductance response (SCR) respectively. SCL and SCR were converted back to the time domain to analyze performance metrics between both devices. Pearson correlation, coherence, and Dynamic Time Warping (DTW) were computed on SCL, while zero-crossing peaks were counted for SCR analysis. With gel electrodes, the average Pearson correlation was 0.92 and the SCR peak count difference was 38. For textile electrodes, the correlation was 0.88 with a peak count difference of 119. Both configurations achieved coherence above 0.95 and DTW below 0.5 for most participants. These results demonstrate mEDA's reliable performance in capturing both tonic and phasic EDA across electrode configurations.

Keywords—EDA, SCL, SCR, Dynamic Time Warping (DTW), BIOPAC, dry electrode, tonic, phasic

I. Introduction

Human skin is the largest bodily organ, acting as a direct indicator of internal emotional, cognitive, and physiological processes. These internal states are closely linked to the Sympathetic Nervous System (SNS), which governs the body's automatic responses to arousal and stress. One of the most direct outcomes of SNS activation is emotional sweating: secretion of sweat due to psychological triggers alongside thermoregulation. Changes in the electrical properties of the skin, primarily caused by emotional sweating, are known as Electrodermal Activity (EDA) [1].

Electrically, EDA is explained using a parallel resistor model. In this framework, the thousands of individual sweat ducts are conceptualized as a set of resistors arranged in parallel. When sympathetic activation causes these ducts to fill with conductive sweat, the skin's overall impedance decreases. This is measured as a corresponding increase in skin conductance [2].

EDA is measured by applying either a direct (DC) or alternating (AC) current. The AC-based measurement method is more complex, involving phase shifts that require measuring a complex value (admittance) with both real (conductance) and imaginary (susceptance) parts [3]. Although AC can reveal detailed skin properties, the DC-based measurement method provides the conductance value that is sufficient for tracking sympathetic arousal [4]. The skin conductance signal, measured in micro siemens (μS), is differentiated into tonic and phasic components. The tonic component, or Skin Conductance Level (SCL), is the slow-changing baseline that reflects general arousal. In contrast, the phasic component consists of Skin Conductance Responses (SCRs), which are the sharp, transient peaks caused by stimuli or an internal emotional state. Each SCR has a characteristic sharp rise and slower decline [1]. EDA is a slow-moving signal that can be represented as a sum of the SCL and SCRs, typically with a dominant frequency below 1 Hz. SCL or tonic changes are lower frequency components typically falling in the range of below 0.05 Hz. SCR or phasic component is relatively a high-frequency component of EDA up to 2 Hz [5].

Shifting EDA measurements from laboratories to wearables presents several challenges: ensuring stable electrode–skin contact, controlling electrode impedance and induced currents, and mitigating motion artifacts caused by changes in pressure and electrode placement [6]. EDA can be measured with both standard gel-based and dry electrodes. For long-term recordings, dry electrodes are generally preferred to prevent sweat gland saturation by gel, although they are more susceptible to motion artifacts and fluctuating skin impedance [1].

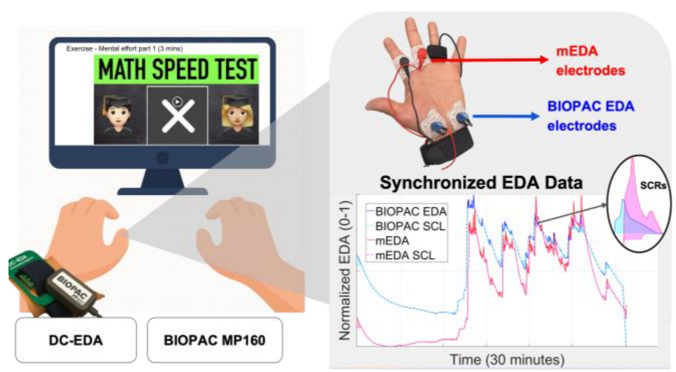

Figure 1: Concept image of validating mEDA

In this paper, we present the DC sourced mEDA circuit adaptive for a wide range of skin electrode impedances. While prior work demonstrated the feasibility of a DC-based EDA circuit [7], their validation was limited to morphological inspection without quantitative benchmarking against research-grade systems. This leaves a gap in confirming whether such circuits can reliably capture both tonic and phasic components of EDA compared to gold-standard instrumentation. We propose to enhance and validate the DC-based EDA circuit to make it more acceptable across different stimulations for both gel and dry electrodes. The key contributions include:

- **Circuit adaptation:** Optimized DC-EDA bioinstrumentation using an advanced amplifier (AD860x) for improved gain, resolution, and power efficiency.
- **System validation**: Benchmarked mEDA against a gold standard EDA device (BIOPAC MP160) in ten healthy adult participants, confirming comparable signal quality with both gel and textile electrodes.
- **Protocol and feasibility analysis**: Applied a stress–relaxation protocol and used standardized metrics (Pearson correlation, Dynamic Time Warping (DTW), coherence, ΔZC) to quantify agreement in SCL and SCR.

II. Materials and Methods

A. mEDA Bioinstrumentation System Development

The EDA acquisition bioinstrumentation comprises a three-stage operational amplifier [7]. The first stage provides a virtual ground for single-supply operation (AD8604), followed by a voltage-controlled linear current source and a low-pass filter. By choosing R_{ref} = 825 kΩ, excitation current is held at 2 µA, for high sensitivity across a wide range of skin conductance. The conditioned voltage (V_{eda}) is digitized via a 24-bit ADC ADS1219, interfaced to a Raspberry Pi Pico (Fig. 2).

$$G_{skin} = \frac{V_{DD}}{(V_{DD} - 2V_{EDA})R_{ref}} \times 10^6 \quad \text{……..(1)}$$

Skin conductance is computed from skin resistance using equation (1) where, G_{skin} is skin conductance (µS), V_{eda} is divider output voltage, V_{DD} is supply voltage (3.3V) and R_{ref} is the reference resistor (825 kΩ). A CSV file containing data sampled at 100 Hz was saved to the device's SD card every five minutes.

B. Human Study Setup and Procedure

Ten healthy adults were recruited to validate our custom EDA circuit against the gold-standard system (BIOPAC MP160) with the Bionomadix PPGED-R amplifier [8]. This study received approval from the University of Rhode Island Institutional Review Board (IRB No# 2125810-6).

The PPGED-R is a DC-based EDA module that connects to MP160. To ensure time-synchronized concurrent data capture, an external digital trigger is used from the mEDA system. Four EDA electrodes (two for mEDA and two for BIOPAC) were placed on standard EDA collection sites on the left hand [9]. To avoid crosstalk, BIOPAC electrodes were placed on the wrist, and mEDA electrodes were placed on the index and middle fingers as shown in Fig. 1. BIOPAC EDA was acquired at 1 KHz, whereas custom EDA was recorded at 100Hz and stored in local SD card.

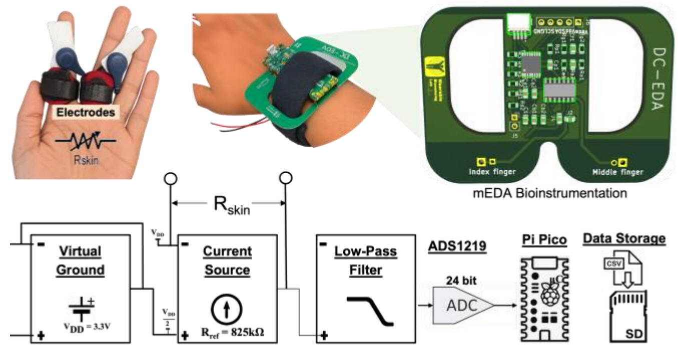

Figure 2: The Bioinstrumentation Architecture of the mEDA System.

Participants completed a 30-minute protocol (Fig. 3A) consisting of an initial seated baseline, a paced deep-breathing relaxation period, three back-to-back cognitive challenges, and a final seated cool-down:

- In the first mental task of M1 ("Speed Math"), they answered 25 multiple-choice arithmetic questions spanning four difficulty levels within 3 minutes.
- Next M2, there was a two-minute "Reverse Alphabet" task, they recited the letters from Z to A, restarting from Z whenever an error occurred.
- The third M3 task required participants to follow a video prompt and solve 40 arithmetic problems in 4 minutes, with question difficulty increasing over time.

Mental tasks M1 and M3 were based on math arithmetic operation questions from pre-recorded video. All the tasks were presented through continuous slide show. A brief resting period separated each task where we have detailed them about instructions for upcoming tasks, and the session concluded with seated rest to allow physiological measures to return toward baseline. Participants were asked to close their eyes during each five-minute rest to ensure a consistent decline in skin conductance level toward baseline, while calming music played in background. The first two minutes were treated as a stabilization phase for the slow-varying EDA signal, and only data collected after this settling time were used for analysis.

For first five participants (P1-P5), mEDA data were collected using standard gel electrodes. For the rest of the participants (P6-P10), mEDA data were recorded using silver-knit textile electrodes developed in our lab in a previous study [10]. BIOPAC data were recorded using gel electrodes as standard comparison for all the participants. The morphology of the waveforms recorded by both devices were very similar throughout the protocol. Signals acquired from mEDA gel-based electrodes exhibited amplitudes that closely matched with BIOPAC gel electrodes, whereas recordings from silver-knit textile dry electrodes were approximately 50% lower in amplitude.

C. Comparative Signal Analysis:

1) Preprocessing:

All recordings were resampled to 100 Hz via cubic interpolation and truncated to 30 minutes to ensure matching durations for BIOPAC and mEDA.

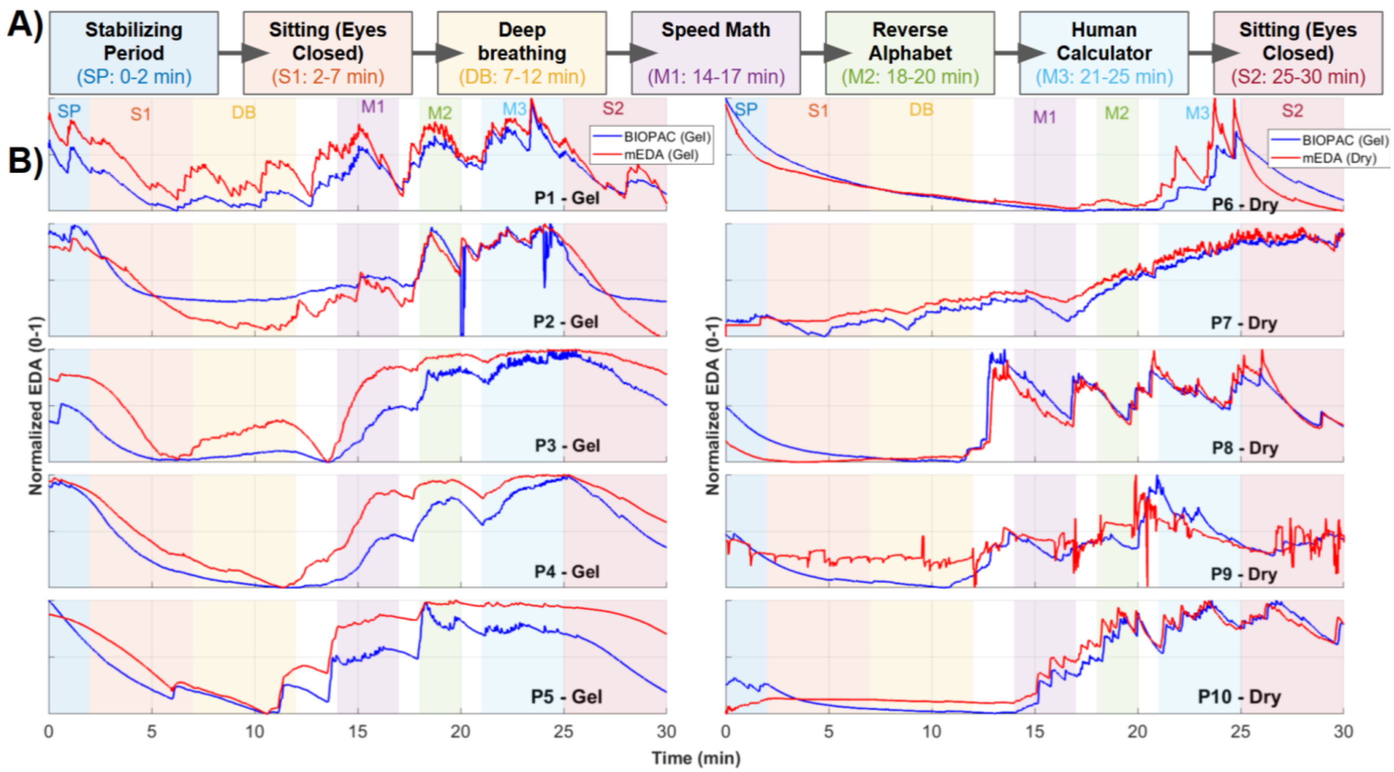

Figure 3: A) Study Protocol with relaxing and mental task activities. B) Normalized EDA signals after preprocessing during different activities.

A 1.5 Hz low-pass filter was used to remove high-frequency noise [7], and artifacts were excluded by zeroing and interpolating any 10-second window with abrupt amplitude changes exceeding 20 % or a standard deviation below 10^{-6} [11]. Finally, each participant's mEDA and BIOPAC data was normalized [0 1] (Fig. 3B) before calculating metrics for validation of mEDA data [12] (Fig. 4).

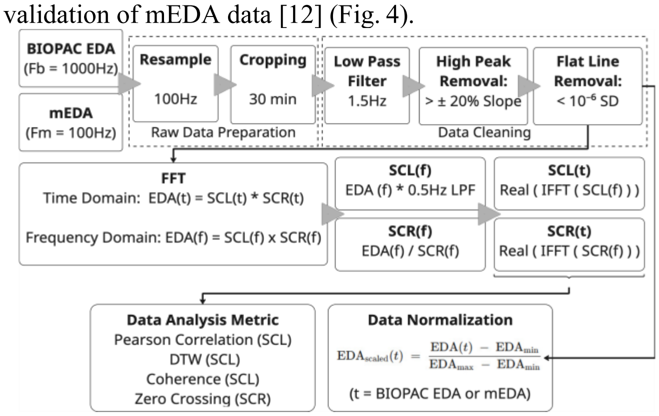

Figure 4: EDA Signal Processing Pipeline.

2) Analysis Metrics:

Preprocessed signals were overlaid to assess both BIOPAC and mEDA waveform morphology as shown in Fig. 3B. EDA signal was decomposed into SCL and SCR components before normalization using an FFT-based cutoff at 0.5 Hz [7] [13]. SCL similarity was quantified with the Pearson cross-correlation coefficient, ranging from -1 (inverse) to +1 (perfect) linear agreement. Temporal alignment was evaluated using DTW, with normalized scores classified as excellent (0.0-0.5), good (0.5-1.5), weak (1.5-3.0), or poor (>3.0). Frequency-domain coupling was measured by coherence, yielding values between 0 (no shared activity) and 1 (perfect coupling) at each frequency [12]. SCR peaks were detected via zero-crossing analysis against a moving-average baseline (1s sample window, 1s sample shift), and agreement between devices was expressed as the absolute zero-crossing difference $\Delta ZC = |ZC_{BIOPAC} - ZC_{mEDA}|$ [6]. All computations were performed in MATLAB.

III. Results

1) Linearity and Stability of Circuit Performance:

Linearity and stability were calibrated using known resistance with our device. Precision resistors of 100-680kΩ were used over time to test our circuit in the benchtop version before employing it on human participants. Each resistance showed <±0.3 mV (ADC values) drift, confirming constant conductance over the targeted range. This shows that employing R_{ref} of 825 KΩ provides consistent current to measure shift in conductance fluctuating within 20µS range for dry and gel electrodes.

2) Morphological Comparison:

Fig. 3B presents normalized EDA signals from both mEDA and BIOPAC systems. The left subplot (P1–P5) compares gel electrodes, while the right subplot (P6–P10) compares mEDA dry electrodes against BIOPAC gel electrodes. Across all participants, the signal trends between devices were well-aligned, indicating consistent tracking of EDA changes. Although dry electrodes showed lower amplitude, their signal shape closely matched BIOPAC, likely due to differences in skin-electrode impedance affecting current flow. Participant P9 exhibited noticeable motion artifacts, likely from hand movement during recording. Overall, a downward trend was observed during the breathing task, followed by increased EDA during cognitive challenges (14–25 min), and a decline during the final rest period. Most participants showed an immediate EDA rise during task instructions, with three distinct peaks corresponding to each mental task. However, deviations from protocol were evident in P1 and P9, affecting signal consistency.

3) SCL and SCR Quantitative Performance Metrics: Table 1 provides quantitative metrics comparing tonic SCL and phasic SCR agreement between BIOPAC and mEDA. Gel electrodes (P1-P5) achieved a high average Pearson correlation 0.92, coherence greater than 0.95, and DTW below 0.5 for all except P3 (DTW - 0.62), reflecting consistent baseline tracking; their |ΔZC| values were low (5-78), indicating nearly identical SCR counts. P3's slightly elevated DTW corresponds to a brief timing offset around the M2 to M3 transition in Fig. 3B.

Table 1: SCL and SCR comparison metrics.

PID	SCL			SCR
Gel: P1-P5 Dry: P6-P10	Pearson Correlation (Closer to 1 is better)	DTW (Lower is better)	Coherence (Closer to 1 is better)	ΔZC (Lower is better)
P1-Gel	0.88	0.24	0.95	78
P2-Gel	0.92	0.40	0.85	5
P3-Gel	0.94	0.62	0.82	42
P4-Gel	0.95	0.34	0.96	7
P5-Gel	0.89	0.30	0.97	59
P6-Dry	0.84	0.17	0.98	46
P7-Dry	0.98	0.15	0.96	35
P8-Dry	0.93	0.38	0.97	98
P9-Dry	0.66	0.49	0.95	288
P10-Dry	0.97	0.27	0.98	126

On the other hand, dry electrodes (P6 - P10) also maintained strong average correlation of 0.88, coherence higher than 0.95, and DTW below 0.5 for all but P9 (DTW= 0.49), demonstrating that even with higher impedance, dry contacts reliably follow slow EDA trends. However, ΔZC values were more variable (35-288), driven by motion artifacts (notably P9) and impedance fluctuations, leading to occasional SCR over- or under-counting. Despite gel electrodes outperforming dry in uniformity, particularly in consistently low DTW and ΔZC metrics, dry electrodes still delivered robust SCL and SCR agreement.

IV. Conclusion and Future Works

In this study, a DC-based EDA circuit was modified through an integration of a new ADC (ADS1219), and skin conductivity was calculated via a Raspberry Pi Pico to achieve enhanced resolution. A custom bioinstrumentation system (mEDA) was developed and characterized using both known resistor values and human skin. Validation against a research-grade standard was conducted for gel and dry electrodes across ten healthy adult participants subjected to relaxation and stress stimuli. Time-synchronized EDA signals underwent detailed signal processing and were compared using established metrics. The mEDA system demonstrated performance comparable to the BIOPAC device for both electrode types. Interestingly, dry electrodes often showed stronger tonic alignment with BIOPAC than gel electrodes, likely due to electrode placement and the potential for gel to interfere with sweat gland activity. This highlights the promise of advancing dry electrodes further if circuit adaptation and optimized form factors are implemented. Future work will involve further circuit characterization, evaluation of alternative current sources in relation to skin-electrode impedance for multiple dry electrodes, and extensive long-duration human studies incorporating additional biosignals for event-based skin conductivity assessment.

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