Feedback-based Electrode Rehydration for High Quality, Long Term, Noninvasive Biopotential Measurements and Current Delivery

Ashwati Krishnan[1], *Member, IEEE*, Harper Weigle,
Shawn Kelly, *Senior Member, IEEE*, and Pulkit Grover, *Senior Member, IEEE*

***Abstract*— Diagnoses and treatments of neurological disorders rely heavily on non-invasive measurement of bio-potentials, or non-invasive injection of currents into the body, by placing electrodes on the skin. A major impedance in using these techniques for continuous, long-term monitoring is the deterioration of electrode performance over time. Long term operation of electrodes requires that they should continue to have low electrode-skin impedance, even in presence of hair, for extended periods of time. There are currently no electrode systems that satisfy all these requirements. Our work proposes and demonstrates a proof-of-concept, feedback-based electrode impedance monitoring and control system. We use a sponge electrode, housed in a custom-designed casing that is connected to a low-speed peristaltic pump. The pump releases a small amount of saline solution when the electrode-skin impedance crosses a threshold. Our system succeeds in keeping the impedance below a target impedance for at least 20 hours on bench-top experiments. On human participant experiments, our results from 90 minutes of continuous measurement show that our saline electrodes outperform conventional clinical wet electrodes in the presence of hair.**

I. Introduction

The advent of microelectronics has increased our ability to measure and affect the electrical nature of the human body. Non-invasive electrical measurements such as electrocardiography (ECG, heart), electroencephalography (EEG, brain) and electromyography (EMG, muscle) etc. are some of the first and the most critical tools in diagnosing and tracking many disorders. Similarly, non-invasive electrical current delivery has been used to treat chronic pain, depression, and more recently, to arrest the growth of brain tumor cells [1], using "TTFields" that utilize low intensity non-invasive electrical current delivery for up to 18 hours/day.

Despite significant advancements in electrode design, e.g. the development of hydrogels [2], foam electrodes [3], and active (dry) electrodes, these advancements have resisted clinical adoption. To understand reasons for this resistance, let us focus on the widely used EEG systems: (i) most new devices on the market use dry electrodes, which result in noisier recordings [4], and clinical systems require high signal-to-noise (SNR) measurements; (ii) wet electrode systems are extremely cumbersome to use, yet they are the preferred choice due to their signal quality; (iii) wet electrode systems require a long setup time; and (iv) long term measurements are often needed (e.g. critically ill, comatose, or epilepsy patients), during which time wet electrodes require constant maintenance. For instance, because of unpredictability in the timing of seizures, seizure patients are monitored in Epilepsy Monitoring Units (EMUs) with simultaneous EEG and video recordings for long periods of time (4-7 days).

It is important to understand the value of technician time in these scenarios. For example, besides setting up and removing electrodes, EEG technicians perform critical tasks such as monitoring data quality, real-time EEG and video of every patient while they are in the hospital to observe electrical seizures as they happen. Because these measurements are critical to subsequent diagnoses and treatments, valuable time is taken up in ensuring that the EEG electrodes are at nominal impedance values (typically 5-10kΩ), by monitoring the impedance and refilling the gel as needed. While electrolyte gels perform well in clinical applications, they are sticky and are very inconvenient both to refill, and to wash off from patients' heads. Many practical problems make their work harder: in long term measurements, irregularities in signals can also be caused by pieces of dried gel or by too much gel under the electrodes, which directly affect the impedance. Another common issue, particularly in EEG measurements, is caused by thick hair, which prevents good electrode-skin contact in short and long term, resulting in higher impedance values. To mitigate these problems, in this paper we propose a solution that uses a sponge electrode with saline to automate the process of monitoring and controlling electrical impedance.

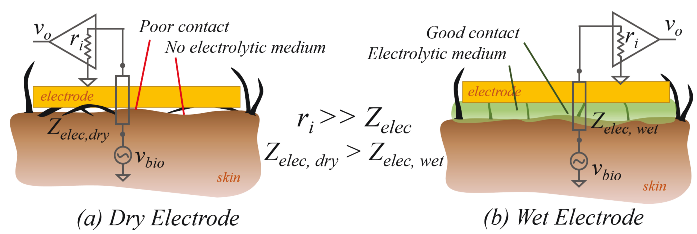

Fig. 1. Schematic description of the practical difficulties with dry and wet electrodes on the skin or scalp. Dry electrodes have higher impedance because they do not make good physical contact with the skin. Moreover, they do not have an ionic electrolytic medium that provides a smooth electrical connection with the skin, resulting in higher noise and the requirement for amplifiers with a high input impedance.

*The authors are with Carnegie Mellon University, Pittsburgh, PA,USA.
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[1] Corresponding author email: `ashwatik@andrew.cmu.edu`

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In Section II, we provide some background explaining the scientific and engineering basis of electrode-skin contact through the example of wet versus dry electrodes. In Sections III and IV, we describe our design consisting of electrodes with sponges in our tailor-made casing, and use of saline solution to rehydrate the electrode using an impedance measurement-based feedback system. Finally, we demonstrate with bench-top as well as human participant experiments that, with repeated feedback-based rehydration, our mechanism succeeds in keeping the electrode-skin impedance within a required impedance range, resulting in a system for long-term, high quality measurements. In particular, we show that while the impedance of standard gold-cup electrodes increases over time, our electrodes maintain the impedance below a set point.

II. Background

Electrical signals in the body are realized through the movement of ions, unlike electrons in metals. The electrode-skin interface presents an interesting problem in the measurement of bio-signals with devices such as EEG/EMG/ECG. When an electrode is placed on the skin, the outermost layer of skin acts as a barrier to the flow of ions [5], creating a high impedance at the electrode-skin interface. By adding electrolyte gel, or saline solution, there is an increase in ionic conduction between the electrode and the skin and a significant reduction in impedance. Additionally, in the presence of hair, using an electrolyte gel also increases the electrical contact of the electrode with the skin (Fig. 1). One can also use a dedicated high input impedance amplifier for each electrode to improve the SNR of the signal [6]. The use of sponge or foam electrodes [7], [8] has been gaining traction because of increased user comfort, and increased electrical contact due to their flexibility in conforming around hair on the scalp. Although all these approaches can help, they do not address the need for quick-to-install electrodes that maintain low impedance for long-term measurements.

III. Proposed Idea

The main issue with long-term bio-potential measurements or current delivery systems is that the electrodes dry up after a prolonged use. As the electrodes dry up, the impedance increases because the ionic conduction between the electrode and skin decreases. Because electrolyte gels can be sticky and difficult to work with, we use sponge electrodes with saline solution to achieve the same mechanism of conduction. The advantage of sponge electrodes is that they are comfortable to wear, and conform around hair thereby making good contact with the scalp. One limitation (not addressed, e.g., in [7]) is the difficulty of performing long term measurements using sponge electrodes, which we present in this work. The key driver of this work is the fact that the interface between the sponge electrode and the skin needs to stay damp – to enable ionic conduction between the electrode and the skin. By monitoring the impedance of the interface, and subsequently actuating a drip system for a short time when the impedance exceeds a threshold, we can ensure a nominal impedance. In this way, we can achieve an effective, long term, bio-potential measurement and/or current delivery.

IV. Methods

To describe and validate a long-term low-impedance, low-setup time electrode-skin interface, we present our experimental methods and results in three main segments: (i) description of the measurement circuit and feedback system; (ii) reporting electrode impedance across time on a bench-top setup for long term monitoring of impedance; and, finally, (iii) comparing, across time, electrode-skin impedance on a human participant with different types of commercially available electrodes.

A. System Design

The system consists of an impedance measurement circuit with and a multiplexer for switching between electrodes. A LabVIEW virtual instrument interfaces the instrumentation with software control that actuates the feedback system, which is a peristaltic pump connected to a power supply. The pump connects a tube to a sponge electrode, placed either on a bench-top system or on the scalp.

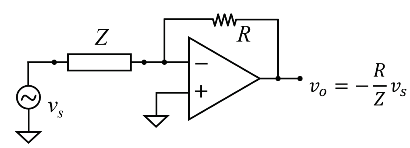

Fig. 2. Op-amp V-to-I converter circuit for impedance measurement.

We used an op-amp (LT1632) based voltage-to-current converter circuit for impedance measurements (Fig. 2). The input was a sine waveform with a peak-to-peak voltage between 10-30mV, at a frequency of 1kHz. While there are applications that operate at a lower frequencies (such as EEG), we used 1kHz because most commercial electrode values are reported at a frequency of 1kHz. We varied that feedback resistor, R, between 10kΩ and 100kΩ, based on the gain that was required. A higher impedance (Z) required a higher input voltage or a higher gain. To achieve a slow drip on activation with no leakage, we used a 12V, 35rpm peristaltic pump (SimplyPumps.com, Westmoreland City, PA). To control the amount of fluid, we drove the motor at 6V (80mA), reducing the output when required, resulting in a two-state proportional feedback control system. To avoid jitter and frequent on/off cycles of the pump, we used high and low set points for impedances. We activated the pump based on the state machine shown in Fig. 3 through the LabVIEW virtual instrument. We report only magnitude values in this work for simplicity, because that is the standard metric according to which electrodes are frequently compared.

We designed and 3D-printed (Form Labs, Somerville, MA, USA) an electrode casing to house a cellulose sponge electrode with a silver/silver chloride pellet, and a tube fitting to wet the electrode based on the measured impedance, shown in Fig. 4.

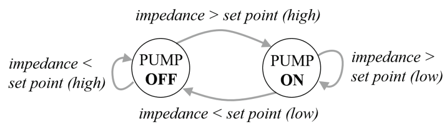

Fig. 3. State machine depicting the state of the pump based on the impedance measured on the sponge electrode.

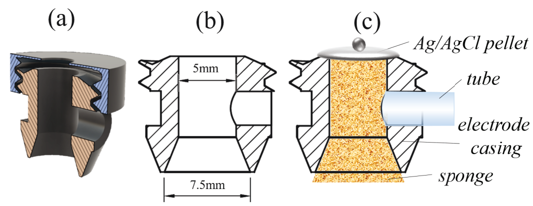

Fig. 4. Feedback-based sponge electrode casing: (a) 3D CAD model showing a screw and a base with an outlet for a tube, (b) cross-sectional view of the casing and (c) schematic of the electrode casing with an Ag/AgCl pellet and a sponge placed inside.

B. Long Term Measurement Setup

Because the measurement of long term impedance over days can be a tedious process for a human participant, we built bench-top setup (Fig. 5) for the sponge electrode to monitor the performance of the feedback over a longer time under standard temperature and pressure conditions. The sampling time for the impedance was between 4-12 seconds, based on the number of electrodes measured by the multiplexer.

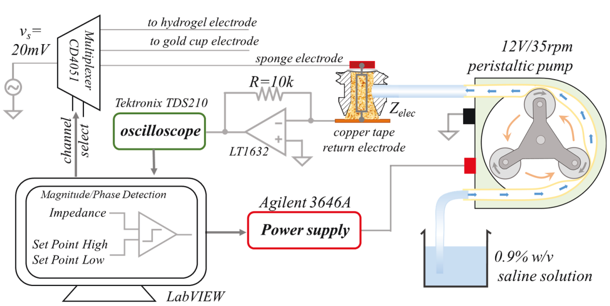

Fig. 5. Full system setup for the long-term measurement using feedback in sponge electrodes. A low-voltage, low rotation peristaltic pump was used to gently discharge saline solution based on the impedance set point fixed in the LabVIEW environment.

We used copper tape as the return electrodes for the bench-top measurement, and assumed any impedance changes that occurred due to its interaction with the saline to be negligible when compared to the impedance of the sponge electrode.

C. Human EEG Measurements

All experiments were conducted in accordance with Internal Review Board (IRB) protocols approved by Carnegie Mellon University (Pittsburgh, PA, USA). We measured the impedance of two commercial electrodes - disposable hydrogel electrode (Covidien Kendall, Minneapolis, USA), and a gold-cup electrode (Natus Neurology, Pleasanton, CA, USA). We placed the electrodes on areas on the scalp that contained some hair, in order to test our electrodes in real conditions (Fig. 6).

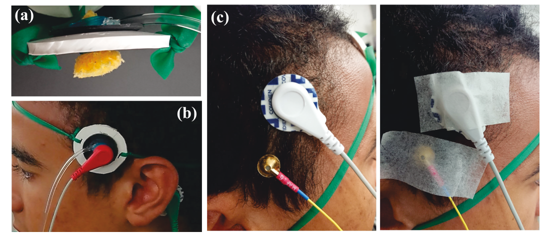

Fig. 6. Placement of electrodes on areas with hair: (a) Sponge electrode placed in a piece of silicone with elastic that is wrapped around the head (b) placement of the tube and connector on the sponge electrode, located between the F7 and T3 EEG locations [9] (c) *(top)* a hydrogel sticker and *(bottom)* a gold cup electrode with Ten20 conductive paste located between the F8 and T4 locations.

V. Results

The initial long term tests show the response of a cellulose sponge electrode observed for more than 20 hours on a continuously running bench top setup. The results presented was for an experiment we conducted with the setup shown in Fig. 5. As can be seen from Fig. 7(a), the time scale of evaporation from a cellulose sponge electrode to a set point of 15kΩ is on the order of 60 minutes.

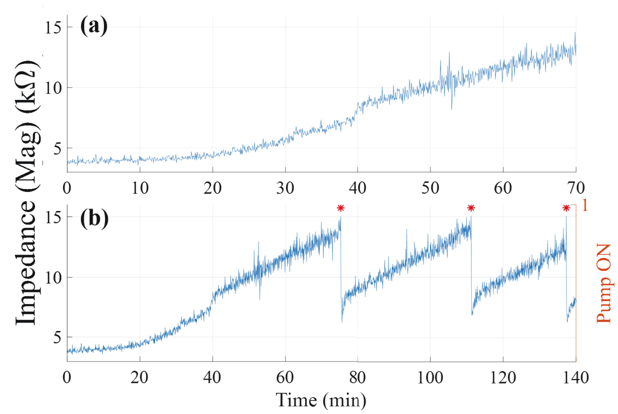

Fig. 7. (a) Sponge electrode on bench-top without feedback: The impedance increases by about 100% in 15 minutes on a bench-top setup due to drying (b) Feedback turned on with an arbitrary set point of 15kΩ, with a drive voltage for the pump = 6V, for 2 seconds. The * indicates when the pump is ON.

Zooming out of the data collected from the same setup, Fig. 7(b) shows the impedance and the state of the pump (ON/OFF; a red star indicates the ON state). There are two observations that we would like to highlight: (i) for a short, small volume discharge of the pump, the electrode sees a dramatic reduction in the impedance and (ii) the pump needs to turn on only for 1-2 seconds every hour or so (for the given set point). We surmise that the reduction in impedance is due to increased ionic conduction due to the saline solution as well as increased physical contact or surface area.

Fig. 8 shows the performance for more than 20 hours in two bench-top experiments. In Fig. 8(a), we used a set point (high) of 15kΩ in order to see the effect of one-time re-hydration on the impedance. Because the set point was so high, feedback was required approximately every hour. In Fig. 8(b), we introduced hysteresis into the system, and changed the set point limits to 2.2kΩ to 2.5kΩ. A lower range resulted in more frequent saline pumps.

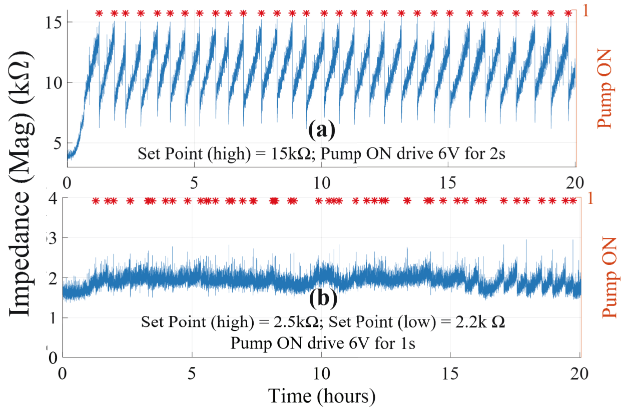

Fig. 8. Long term impedance monitoring with (a) open ended feedback with a set point of 15kΩ and (b) feedback with hysteresis with set points of 2.2kΩ and 2.5kΩ

Two human participants had electrodes placed in areas with similar amounts of hair for about 90 minutes. One participant had a gold and hydrogel electrode, and the other had two sponge electrodes, with and without feedback. We present the data in Fig. 9 and found that the response with hair was eye-opening: the Covidien hydrogel sticker is usually known for its robust performance on bare skin. However, as shown by the large impedance in Fig. 9, the adhesive on the sticker does not allow the electrode to be used for areas with hair at all, even if the hair is not very dense. While the gold cup electrode with the Ten20 conductive paste shows a nominal value initially, over tens of minutes the electrode tends to shift in position or dry and lose contact, similar to the sponge electrode without feedback. Our feedback system was set between 5kΩ and 9kΩ, and it consistently maintained the impedance between those values due to the slow discharge of saline through the pump, based on the set points.

VI. Conclusion & Future Work

Many sensing and stimulation/current delivery applications require long term, continuous operation of electrodes placed on the scalp, while ensuring low electrode-skin impedance. This work proposes a mechanism to ensure that electrode impedance stays low in the long term, while requiring a low setup time. Our mechanism works on conventional sponge-based electrode-skin interfaces which monitors electrode-skin impedance, and uses a simple feedback mechanism to rehydrate and lower the impedance when high impedance is measured. A typical electrolyte gel is too viscous for thin tubes and can be very difficult to clean after it has dried up. The advantage of using saline solution is that there is a smooth flow through a low-speed pump, that moistens the sponge as required. The actuation time is short, roughly 1 - 3s, and is needed infrequently, ensuring that it can

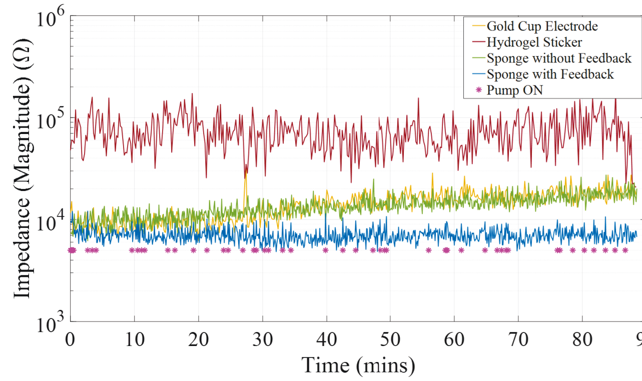

Fig. 9. Comparison of different electrode types on human participants on scalp area with hair. The hydrogel sticker does not adhere to scalp with hair, showing high impedance. The gradual increase in impedance of the gold cup electrode and the sponge electrode without feedback after 20 minutes, shows effects of drying as well as contact on scalp areas with hair. In contrast, the sponge with feedback shows steady maintenance of impedance for 90 mins due to discharge of saline from the pump when required.

be interleaved with EEG measurement without a significant loss of information. While the system presented in this work is not ready for clinical adoption, it is our belief that for low densities of electrodes commonly used in clinical EEGs, our process is scalable when designed with efficient microfluidics networks. Our work suggests that long term, high SNR bioelectrical signal measurements and current delivery can be performed reliably on scalps with hair.

References

[1] E. J. Mun, H. M. Babiker, U. Weinberg, E. D. Kirson, and D. D. Von Hoff, "Tumor-treating fields: A fourth modality in cancer treatment," *American Association for Cancer Research*, vol. 24, no. 2, pp. 266–275, Aug. 2018.

[2] N. A. Alba, R. J. Sclabassi, M. Sun, and T. C. Xinyan, "Novel Hydrogel-Based Preparation-Free EEG Electrode," *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 18, no. 4, pp. 415–423, Aug. 2010.

[3] C.-T. Lin, L.-D. Liao, Y.-H. Liu, I.-J. Wang, B.-S. Lin, and J.-Y. Chang, "Novel Dry Polymer Foam Electrodes for Long-Term EEG Measurement," *IEEE Transactions on Biomedical Engineering*, vol. 58, no. 5, pp. 1200–1207, May 2011.

[4] K. E. Mathewson, T. J. L. Harrison, and S. A. D. Kizuk, "High and dry? Comparing active dry EEG electrodes to active and passive wet electrodes," *Psychophysiology*, vol. 54, pp. 74–82, 2017.

[5] E. McAdams, J. Jossinet, A. Lackermeier, and F. Risacher, "Factors affecting electrode-gel-skin interface impedance in electrical impedance tomography," *Medical and Biological Engineering & Computing*, vol. 34, pp. 397–408, Nov. 1996.

[6] T. C. Ferree, P. Luu, G. S. Russell, and D. M. Tucker, "Scalp electrode impedance, infection risk and eeg data quality," *Clinical Neurophysiology*, vol. 112, no. 3, pp. 536–544, Mar. 2001.

[7] A. Krishnan, K. Ritesh, A. Etienne, A. Robinson, S. K. Kelly, M. Behrmann, M. Tarr, and P. Grover, "Challenges and Opportunities in Instrumentation and Use of High-Density EEG for Underserved Regions," *EAI Second International Conference on Innovations and Interdisciplinary Solutions for Underserved Areas*, 2018.

[8] A. Yeung, H. Garudadri, C. v. Toen, P. Mercier, O. Balkan, S. Makeig, and N. Virji-Babul, "Comparison of foam0based and spring-loaded dry eeg electrodes with wet electrodes," *37th Annual Conference of the IEEE Engineering in Medicine anc Biology Society*, pp. 7131–7134, Nov. 2015.

[9] V. Jurcak, T. Daisuke, and I. Dan, "10/20, 10/10, and 10/5 systems revisited: Their validity as relative head-surface-based positioning systems," *NeuroImage*, vol. 34, no. 4, pp. 1600–1611, 2007.