Low-cost Carbon Fiber-based Conductive Silicone Sponge EEG Electrodes

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***Abstract*— We propose a novel carbon fiber-based conductive silicone sponge for low electrode-skin impedance EEG recordings. When this sponge is used with water or saline solution, no gel is required, lowering the setup time drastically compared to classical wet electrodes. Moreover, the wet conductive carbon fiber silicone sponges achieve an electrode-skin impedance as low as 2.5kΩ at 1kHz when wet, making them better than state of the art gel electrodes. Additionally, even as the sponge dries out, it continues to remain conductive and performs as a reliable dry electrode. We demonstrate through experiments that these conductive carbon fiber silicone sponge electrodes, wet or dry, are able to measure alpha wave activity. Our carbon fiber conductive sponge electrodes are low-cost and are highly suitable for designs of portable high density EEG measurement systems.**

I. Introduction

Scalp ElectroEncephaloGraphy (EEG) is a non-invasive method of measuring the brain's electrical activity used widely in epilepsy diagnosis, studying neurological disorders, neuroscientific studies, and brain-machine interfaces. EEG offers high temporal resolution, but classically, the low-electrode count of these systems has limited them to low spatial resolutions. There have been recent advancements in improving spatial resolution of EEG [1], [2] by increasing the number of sensors. High-Density EEG (HD-EEG) systems, using several hundreds of electrodes, have a lot of potential to become a low-cost imaging technology, but their development is not without challenges [3]. For instance, to correctly localize an epileptic seizure, there is a need to measure EEG signals before and at the onset of a seizure, requiring extended-time HD-EEG recordings. Wet electrodes provide high Signal-to-Noise Ratio (SNR), but are cumbersome to setup; dry electrodes have a poor SNR. In order to develop a high density EEG system that is robust, low-cost, and portable, we construct and evaluate a novel *conductive* carbon fiber, silicone sponge electrode that can be easily and frequently re-hydrated for long-term high-quality EEG observations. In Section II, we discuss the motivation for developing these electrodes, and in Section III, we describe our method of production and testing. Finally, in Section IV and V, we present results and discuss the applications of our conductive carbon fiber silicone sponge electrodes.

*The authors are with Carnegie Mellon University, Pittsburgh, PA 15213. Research was supported by the Chuck Noll Foundation for Brain Injury Research, a fellowship from the Center for Machine Learning and Health (CMLH), and the Carnegie Mellon Brain Hub grants.

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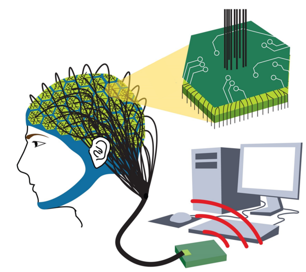

Fig. 1. High Density Electroencephalography (HD-EEG) systems are an emerging low-cost imaging modality with high temporal and spatial resolution [1].

II. Background

The medium of communication within the body is neuronal electrical signals. Because the dominant medium in the body is aqueous, electrical signals are realized through the movement of ions, as opposed to electrons. When an electrode is placed on the skin for measurement, there is a separation of charge that occurs at the electrode-skin interface. This is because unlike the body, electrical current in the electrode amplifier circuit is through the movement of electrons. Human skin consists of several layers, the outermost of which is the stratum corneum, which acts as a barrier to the flow of ions [4], thereby increasing the impedance of any electrode material that is placed to acquire signals from the body. To improve SNR, electrode-skin interface impedance needs to be lowered [5]. The skin is inherently a moist material, so EEG technicians obtain the most reliable signals from *wet* electrodes, which use an electrolyte gel between the electrode and the skin. Although the use of wet electrodes is widespread, they present several problems, especially for HD-EEG: (i) they require the use of special gels that dry out within just a few hours of use; (ii) they take a long time to set up, typically 30-45 minutes for 64 or 128 electrodes; (iii) the gel tends to spread and cause bridging between adjacent electrodes, thereby reducing spatial resolution of HD-EEG [2]. To address some of these issues, there has been significant progress in use of *hydrogels*. Hydrogels are materials that retain a large amount of water compared to the material's own volume [6] They have been incorporated increasingly in commercial disposable EEG electrodes, and are a very promising development for EEG [6]. However, hydrogels are unsuitable for long-term use because they lose their conductivity once they dry out. To avoid the use of electrolyte gels, advancements have been made in the design of dry electrodes and sponges. Portable consumer EEG devices often use dry electrodes that have conductive tips that

978-1-5386-3646-6/18/$31.00

are directly pushed against the skin, but these offer signals with lower SNR than wet electrodes because of their high impedance [7]. The main idea behind the use of sponges is to use a simple mechanism to "wet" the electrode, by soaking it in an easily available conductive electrolyte, such as a saline solution [8]. The sponge approach is attractive because it is low cost and can be quickly applied. However, the saline solution dries out quickly, and consequently, the dry sponges are non-conducting. All the above mentioned issues become unmanageable for high electrode count HD-EEG systems, and they make long term, ambulatory EEG measurement systems almost impossible.

Our Approach

When wet electrodes dry out over prolonged use, the electrode-skin impedance can increase to unacceptably high values. A key aspect of our approach is to ensure a low electrode-skin interface impedance, regardless of the wetness of the interface. To that end, we provide a novel silicone-based foam/sponge that is embedded with conductive carbon fibers. When the conductive sponge is infused with saline, it provides an aqueous conductive medium between the electrode and the skin. Furthermore, due to the presence of conductive carbon fibers, the sponge conducts even when it *is dry*. Carbon fibers are strands of $\approx 5\mu$m diameter, and are mainly carbon atoms bonded together in microscopic crystals. The crystalline arrangement accounts for their high tensile strength. Because carbon fibers comprise mostly carbon (or graphite), they are also good conductors of electricity and are inert to chemical reactions such as corrosion. Silicones are inert, synthetic polymers that have repeating units of siloxanes (Si-O). Silicones are biocompatible, non-corrosive, thermally stable and have been used in the medical field for implants and bandages. These properties make silicone and carbon fibers appealing for their use in portable HD-EEG systems. Our conductive carbon-fiber silicone sponge is designed to function as a reliable wet electrode and a convenient dry electrode.

III. Methods

A. Preparation of Conductive Sponge

We used two-part silicone foam obtained from Smooth-On Inc. (*Soma Foama 15*, Macungie, PA, USA). The carbon fiber (CF) was procured as a woven fabric from ACP Composites (Livermore, CA, USA). The CF was manually cut from strands extracted from the woven fibers to achieve an average fiber length of $2-5$ mm. We mixed Part A of the two-part silicone foam with CF thoroughly at 25°C in the ratios presented in Table I. Silicone thinning fluid, sourced from Eager Plastics (Chicago, IL, USA), was added to allow for better flow for molding. After thorough mixing, Part B of the silicone foam was add to the Part A-CF blend, stirred and immediately poured into molds to cure. The time taken for the mixture to become a solid foam (cure time) is 1 hr at room temperature. Table I shows variations in preparations in different samples.

TABLE I
CONDUCTIVE CARBON SILICONE SPONGE SAMPLES

#	Silicone (g)			Carbon Fiber (g)
	Part A	Part B	Thinning Fluid	
I	3.11	1.5	0.7	0.2
II	3.07	1.5	0.62	0.25
III	4	2	0.2	0.6
IV	4.06	2.1	0.7	0.81

Notes on Foam Preparation: Foams can be open-cell or closed-cell. Open-cell foams have many interconnected pores, which retain fluid to create an aqueous electrode environment that is required for low electrode-skin impedance. However, most silicone foams are closed-cell foams.

- *Soma Foama 15* is a closed-cell silicone foam that expands to 4 times its volume through the release of gas bubbles, creating a lot of pores. We opened up the interior pores by applying pressure to the cured foam.
- The carbon fiber needs to be about 2-5mm long and mixed until the Part A-CF blend appears homogeneous with a shiny grey texture. This is because conduction in the silicone occurs through interconnected fibers that separate while mixing. We observed that milled carbon fiber was not as effective in increasing the conductivity of the silicone foam. Once the sample has cured, about 1mm of all surfaces need to be cut or filed in order to expose these fibers to metal contacts.
- Chopped carbon fibers of length ≈6mm are commercially available. However, we observed that this length makes the silicone-CF mixture difficult to pour into molds because it behaves like a flat sheet, rather than a pourable mixture. The pot life (the time elapsed before the mixture starts to cure) of *Soma Foama 15* is 30 seconds. Thus, it needs to be poured immediately after mixing in Part B, and this was observed to be done more reliably with shorter carbon fibers.
- The CF changes the mechanical properties of the resulting foam. If too much CF is added, the resulting mixture is too heavy to expand into a foam with many pores.

B. Material Properties

To study material characteristics relevant to EEG recordings, we tested the conductivity of the CF sponge, and the extent of water retention in a few samples described in Table I.

1) Conductivity: The conductivity of bulk materials is obtained by measuring the resistance of a sample of known geometry by forcing a current through one pair of leads and measuring the voltage through another pair. 3D printed rectangular molds were used to study the conductivity of the CF sponge using (1). The conductivity was measured using a Keithley 2400 source-meter (Tektronix, Inc., Beaverton, OR, USA), and was measured when the CF sponge was dry as well as after absorbing 0.9% w/v saline solution, which has a conductivity of 14.7 milli-Siemens per centimeter. Fig. 2 shows the dimensions of the mold and the circuit configuration we used to perform these tests.

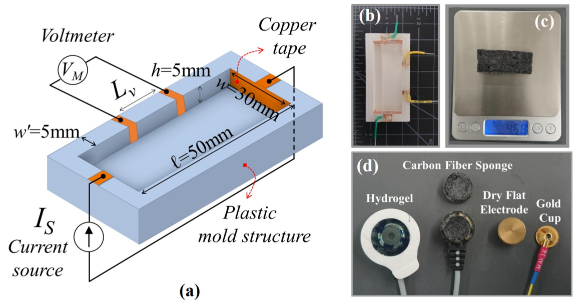

Fig. 2. (a), (b): Rectangular mold design for measuring bulk conductivity: The design was created with AutoDesk Inventor (TM) and printed on a MakerGear M2 (TM) 3D-printer. Copper tape was used to connect to the material and L_v=20mm (c) Image of carbon fiber silicone sponge sample from rectangular mold (d) Different electrodes used for comparison.

The conductivity, σ, of the bulk material is given by

$$\sigma = \frac{I_s}{V_M} \times \frac{L_v}{w \times h}, \tag{1}$$

where the variable notations are provided in Fig. 2, and the results are shown in Fig. 3.

2) Water Retention: The samples shown in Table I were squeezed in de-ionized water, dabbed on a clean paper towel to remove the excess drip and placed in a standard temperature and pressure environment. The samples were weighed repeatedly over 10 hours to observe the extent of evaporation over time.

C. Human Scalp Measurements

To evaluate the efficacy of our conductive carbon fiber silicone sponge electrodes for EEG, we performed impedance measurements and EEG recordings on a human participant. All experiments were conducted in accordance with the Institutional Review Board protocols approved by Carnegie Mellon University. We performed electrode-skin impedance measurements using the Intan Recording Controller (Los Angeles, CA, USA). We used a sampling rate of 20 kilo-samples/sec, bandpass filter settings of 0.1Hz to 7.5kHz and a notch filter setting at 60Hz. We compare our conductive sponge electrodes in wet and dry conditions to a Covidien Kendall (Minneapolis, MN, USA) disposable hydrogel electrode, a BrainVision (Morrisville, NC, USA) flat, metal passive dry electrode and a gold-cup electrode (Natus Neurology, Pleasanton, CA, USA) (Fig. 2d). The diameter of all electrodes were between 8-10mm and the thickness of our conductive carbon fiber sponge electrodes was 2-4mm. For these experiments, one electrode of each of the 4 types was placed close together on the left and right sides of the forehead. The 4 types of dry electrodes were compared simultaneously, followed by the wet electrodes on the same forehead.

D. Electrode-Skin Impedance

While electrode impedance values are typically reported at 1kHz, many relevant EEG signals are at a much lower frequency (5-40Hz). Therefore, we recorded electrode-skin impedance values at 20Hz, 200Hz and 1kHz and 3kHz (see Fig. 5). We did not abrade skin for the electrodes under evaluation, however, we placed a gold-plated cup electrode with Ten20 conductive paste over abraded skin on the right mastoid bone as a reference to ensure an unbiased comparison. To verify the low impedance of the reference, we use an identical cup electrode configuration over the left mastoid.

E. EEG Measurements

Alpha waves are a highly stereotypical form of EEG activity that can be measured when the participant is in a relaxed state, or when their eyes are closed. We measured 3 minutes of EEG signals from a participant under two conditions: with their eyes open, and closed. We performed frequency analysis of the acquired data using the MATLAB-based EEGLAB toolbox.

IV. Results

Using the 4-point measurement technique for bulk materials, in Fig. 3, we show how the conductivity of our carbon fiber (CF) sponge varies with the amount of CF in the sponge. The conductivity of the sample increases with CF content and in the presence of saline. The *change* in conductivity due to the addition of saline *decreases* with increase in CF, because higher concentration of CF implies fewer pores in the material to hold in the saline solution. The sponge structure ensures the presence of an aqueous ionic solution for a low electrode-skin impedance. To study the extent of liquid retention, we measured the rate of evaporation of de-ionized water in a few samples over several hours, and the results are shown in Fig. 4. Using the curve fitting toolbox in MATLAB, we found that the weight of the sample undergoing evaporation decreased in a logarithmic manner. The data are shown in Fig. 4 along with the generalized model equation.

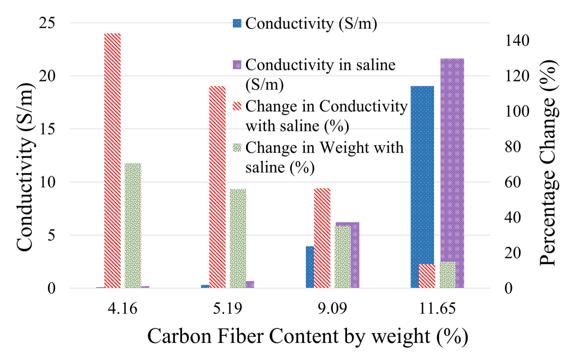

Fig. 3. Conductivity data for carbon fiber silicone sponge samples: The conductivity was measured for dry samples as well as samples saturated with 0.9%(w/v) saline solution. As the carbon fiber percentage increases, the dry conductivity increases.

The magnitude of the electrode-skin impedance is shown in Fig. 5. The reference electrode impedance was between 0.3 - 0.5kΩ. The wet conductive sponge electrode achieved an impedance of around 2kΩ at 1kHz, which was lower than the wet gold cup electrode with Signa electrode gel and the disposable hydrogel electrode. The dry CF-sponge electrode's impedance was comparable to that of standard dry electrodes.

To demonstrate the efficacy of the conductive carbon fiber electrode material as an electrode to detect muscular activity,

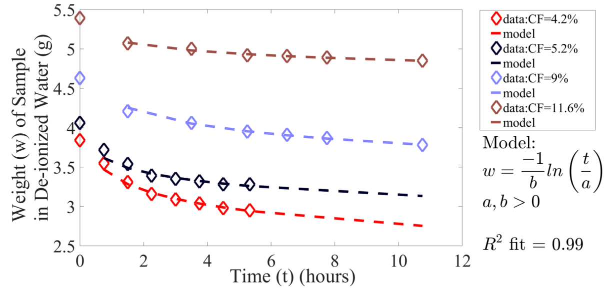

Fig. 4. Rate of evaporation of de-ionized water in conductive carbon fiber samples (at STP): There is a logarithmic decay of the weight of the sample containing deionized water and the time taken for the water to evaporate.

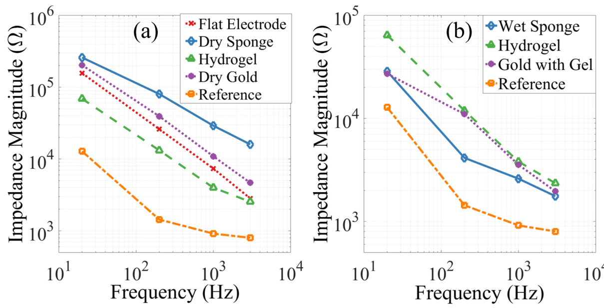

Fig. 5. Electrode-skin impedance measured on the forehead: (a) Dry electrode; (b) Wet electrode; our conductive sponge was dipped in 0.9% (w/v) saline solution, and showed an impedance of around 2.5kΩ at 1kHz. The impedance of the reference was on the order of 0.5kΩ.

we show a time series plot in Fig. 6, depicting different rates of blinking. Alpha wave measurements manifest when people close their eyes and are typically within 8 - 12 Hz. Fig. 7 shows the frequency spectrum peaking in the presence of alpha waves when eyes are closed and absent when eyes are open. While it has been well established that wet electrodes are a reliable means detecting alpha waves, we found that our dry conductive sponge electrodes are as effective as wet electrodes in measuring alpha wave activity in the brain.

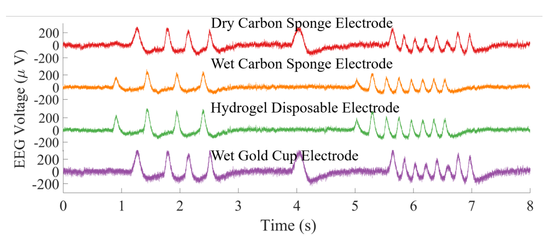

Fig. 6. Transient plots showing eye blinks illustrate that the dry conductive sponge electrodes are effective in detecting electrical activity associated with muscle movement, and are on par with standard electrodes.

V. Conclusion

We have developed a novel carbon fiber-based conductive silicone sponge for use in biomedical applications such as EEG. We demonstrated through bench-top experiments that as the percentage of carbon fiber increases, the conductivity increases. On the other hand, the amount of solution the material can hold decreases, because there are fewer pores in the material. While we observed a lower electrode-skin impedance with a dry conductive sponge with high carbon fiber content (9-11%), increasing fiber content reduces the amount of time the electrode can be used as a wet electrode. The impedance of two 9mm diameter circular carbon

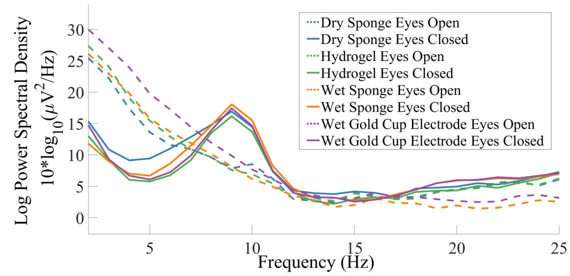

Fig. 7. Frequency response of EEG signals acquired for eyes-open versus eyes-closed. Our electrodes show prominent alpha wave peaks around 8-11Hz as expected for normal EEG activity when eyes are closed.

fiber sponges soaked in 0.9%(w/v) saline solution was an average of 2.5kΩ, which is better than a gold electrode with electrolyte gel. Our EEG experiments show that our conductive sponge electrodes (dry and wet) can reliably measure alpha waves on the forehead. The conductive carbon-fiber sponge electrodes are a low cost, fast-installation solution for high-quality EEG measurements. They are non-magnetic, so they can be used in conjunction with Magnetic Resonance Imaging machines. Because there is no electrode gel involved, the delivery of saline solution is a convenient way to achieve excellent wet electrodes with a short setup time. The purpose of using a conductive sponge is to maintain a low electrode-skin impedance even as the electrode dries out. We foresee the use of these carbon conductive sponge electrodes in portable ambulatory and low-cost high density EEG measurement systems.

Acknowledgments

We thank Deepshika Acharya and Uday Jagadisan for their help with the implementation and testing of the conductive sponges.

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