Measurement and Analysis of System Parameter Effects on Noise in EEG Systems

Meghna Roy Chowdhury and Shreyas Sen
School of Electrical and Computer Engineering, Purdue University, West Lafayette, IN, USA
{mroycho,shreyas}@purdue.edu

***Abstract*—Electroencephalography (EEG) is a widely used method for monitoring brain activity. Traditionally, wet electrodes have been the preferred choice due to their superior signal-to-noise ratio (SNR). However, as the demand for wearable and long-term EEG systems grows, researchers are increasingly exploring dry electrodes, which offer greater practicality despite their lower SNR. In this paper, we analyze how system parameters affect the noise characteristics that contribute to the lower SNR in dry EEG systems. Through experimental measurements and theoretical insights, we examine key factors influencing signal quality, particularly contact impedance and uncorrelated pickup. Specifically, we demonstrate that increasing pressure and electrode contact area significantly reduces noise levels, by approximately 50 dB and 30 dB, respectively, by lowering contact impedance. Furthermore, we observe that while dry electrodes maintain stable noise levels over time, wet electrodes experience a significant noise increase of approximately 38 dB after just two hours. Additionally, we highlight the presence of flicker noise in dry EEG systems, a phenomenon previously overlooked in this context. Our findings provide critical insights into the noise behavior of dry electrodes, paving the way for more reliable and practical wearable EEG systems. By addressing key challenges in signal quality, we contribute to advancing long-term neurological monitoring technologies for real-world applications**

***Index Terms*—EEG, Noise Analysis, Electrodes, Flicker Noise**

I. Introduction

Electroencephalography (EEG) is a key tool for measuring brain activity in neurotechnology and wearable brain-computer interface (BCI) devices. However, these sub-microvolt signals are often drowned out by noise as high as 10–60 mV, which poses a major challenge in biomedical recordings [1]. As BCIs continue to evolve, there is a growing need for robust EEG subsystems. A non-invasive EEG system (see Fig.1(a)) typically places electrodes on the scalp to detect neural activity. This activity involves ionic currents, carried by Na^+, K^+, Ca^{++}, and Cl^-which are then converted into electronic signals at the scalp surface [2]. Traditionally, wired systems with wet electrodes have been favored for their low impedance and relatively better SNR. However, their need for conductive gels limits their practicality in long-term or wearable applications. Recent technological advancements drive the shift toward dry electrodes [3], [4]. These eliminate the need for gels, offering ease of use, especially for long-term wearable applications. However, with this convenience comes a trade-off of lower SNR due to increased impedance at the electrode-skin interface.

Researchers have explored software-based post-processing techniques to enhance EEG SNR [5]–[7], and various studies have investigated the design of analog front-end circuits for EEG improvement [8], [9]. *However, there is a gap in identifying the noise source in an EEG system.* Current literature predominantly addresses wet electrodes, with limited exploration of the distinct noise characteristics in dry electrodes. The contribution of impedance components — skin impedance, contact impedance, and electrode impedance — to the noise profile remains insufficiently studied.

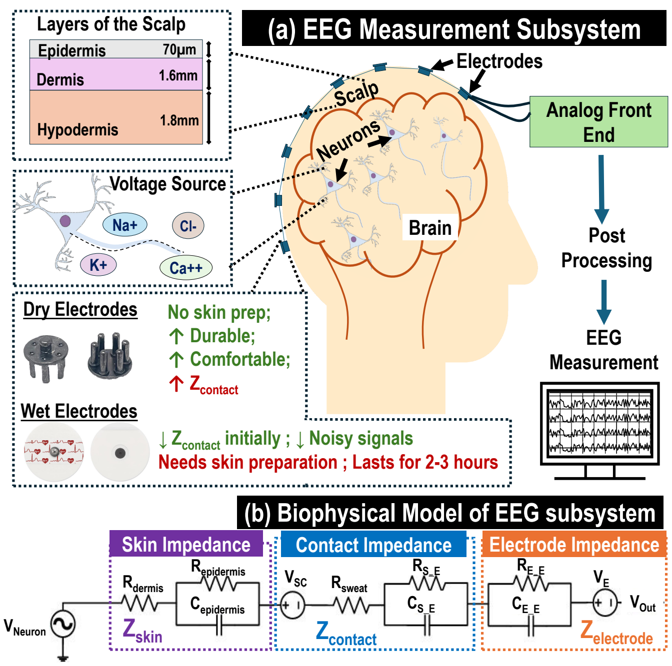


Fig. 1: (a) EEG measurement subsystem (b) Biophysical model of the electrode-skin interface.

This study aims to fill this gap by analyzing the effects of system parameters on the noise behavior in dry electrodes in a typical EEG system. The main contributions of this work are as follows:

- **Impact of Contact Impedance**: We show that increasing pressure and contact area reduces noise by up to 50 dB, highlighting the dominant role of contact impedance.
- **Uncorrelated Pickup of Noise plays a role**: We show through equations and experiments that the uncorrelated nature of noise impacts the overall noise level in the EEG system.
- **Medium Conductivity's Role**: Noise is more correlated

in conductive environments like salt water, leading to lower overall noise.

II. Understanding the EEG Subsystem

To understand the noise sources in EEG subsystems, we first analyze a biophysical circuit model (Fig.1(b)). This EEG sensing system can be viewed as an in-body to out-of-body communication modality, where the neurons serve as the transmitter, the electrode as the receiver, and the channel consists of layers with varying conductivity. Similar to the galvanic mode in Human Body Communication (HBC) [10], where the transmitter is placed inside the body, the neural signal is transmitted as ionic current until it reaches the scalp, where it transitions to electron current. This transition point, which begins at the scalp (as depicted in Fig. 1), is crucial in the biophysical model. In this model, three key impedance components, Z_{skin}, Z_{contact}, and $Z_{\text{electrode}}$ are considered. Z_{skin} is influenced by the properties of the skin, Z_{contact} depends on the pressure and surface area between the skin and the electrode, and $Z_{\text{electrode}}$ reflects the intrinsic properties of the electrode. Each component introduces resistance that affects the overall noise profile.

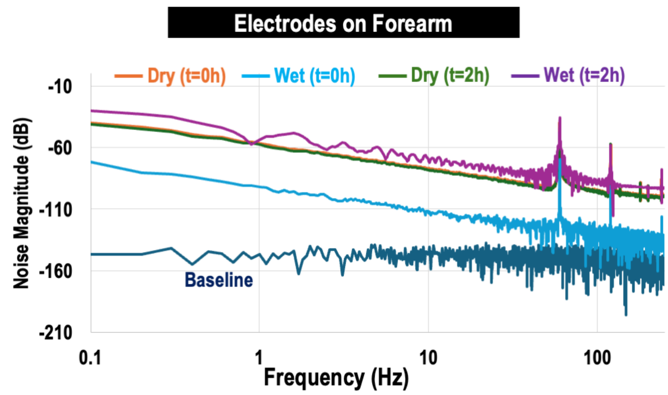
By isolating these impedance components through controlled experiments, the individual contributions to the noise in dry electrodes can be better understood. This knowledge can then be used to mitigate noise in real-world EEG applications. In this paper, the focus is on the noise associated with electrodes, particularly dry electrodes, with an emphasis on analyzing Z_{contact} and $Z_{\text{electrode}}$ through hypothesis-driven experiments.

III. Experimental Analysis of Noise

To systematically investigate the sources of noise in dry electrodes, we perform a series of experiments that isolate different varying system parameters (Z_{contact}, and $Z_{\text{electrode}}$). By varying pressure, contact area, and environmental conditions, we aim to identify the contribution of contact and electrode property in inducing noise in an EEG subsystem.

A. Setup

Our experimental setup includes electrodes and an analog front end. Electrodes for EEG measurements are broadly categorized as dry or wet. Wet electrodes have a saline-gel or water-based gel for better contact. In our study, we utilized Bio ProTech's surface wet electrodes, featuring an Ag/AgCl sensing element and hydro-gel for adhesion, with a diameter of 55mm [11]. The area of the wet electrodes was adjusted to align with the overall diameter of the dry electrodes. OpenBCI's snap electrodes, characterized by a comb-like structure and an Ag-AgCl coating, were employed for dry electrode measurements [12]. The analog front end utilized in the experiments was ADS1299 from Texas Instruments [13]. The experiments were conducted using off-the-shelf commercial components.



	Electrode	Noise Magnitude at 10Hz
1.	Dry at t=0h	-77 dB
2.	Wet at t=0h	-112 dB
3.	Dry at t=2h	-78 dB
4.	Wet at t=2h	-74 dB
5.	Baseline	-150 dB

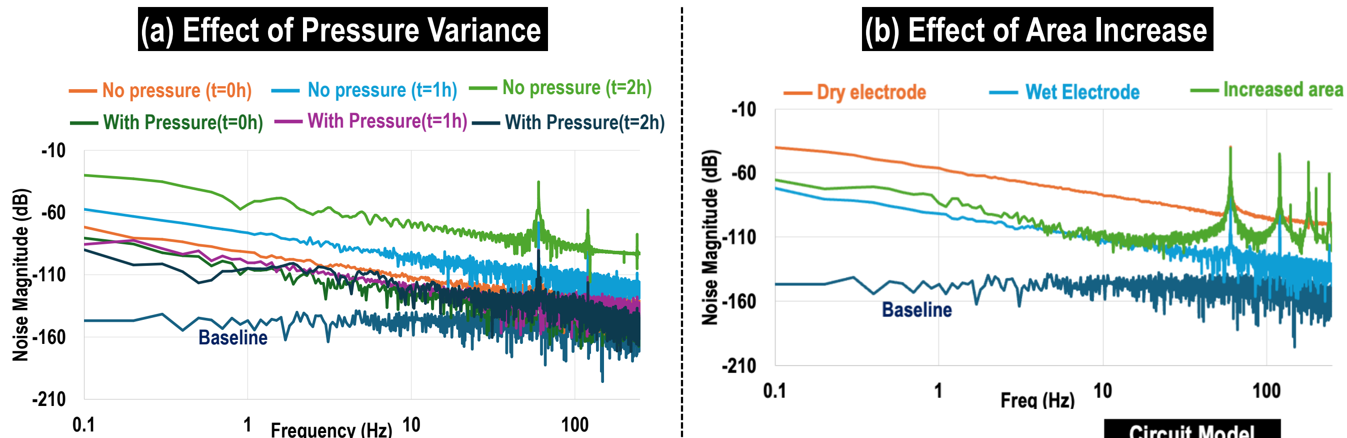
Fig. 2: Comparison of noise magnitude over time for wet and dry electrodes placed on the forearm, showing the stability of dry electrodes and the increasing noise in wet electrodes as the gel dries up.

B. Noise Profiling Methodology

We characterize noise by analyzing signal pickup in wet and dry EEG electrodes. To measure noise magnitude, we place the electrodes on the forearm and perform a frequency sweep from DC to 250 Hz, covering the primary EEG frequency range. The forearm serves as an ideal test site due to the absence of intrinsic EEG signals, ensuring that any recorded activity is just noise [14]. Building on this, we investigate how electrode contact influences noise in an EEG system. Prior studies suggest that wet electrodes exhibit lower contact impedance than dry electrodes due to the presence of conductive gels, but this advantage may vary under different conditions. To examine this variability, we compare the noise performance of wet and dry electrodes across various experimental conditions. Furthermore, we assess whether noise arises solely from contact impedance or if the electrode's intrinsic properties also contribute. To isolate these effects, we analyze key impedance components (Z_{contact} and $Z_{\text{electrode}}$) in the circuit model (Fig. 1(b)). By placing the electrodes in different mediums, we systematically evaluate their individual contributions to overall noise levels.

C. Observation & Insights

1) Dried-up wet electrodes are worse than dry electrodes: Fig. 2 shows the noise magnitude of the electrodes placed on the forearm over 2 hours. Initially (t=0), the difference between wet and dry electrodes is ≈35 dB, with the wet electrodes represented by the blue line and the dry electrodes by the orange line. After 2 hours, wet electrodes show a change in noise level, with the difference between wet at t=0 and wet at t=2h being about 38 dB (blue and purple lines, respectively).



Time (h)	Magnitude at 10Hz without pressure (dB)	Magnitude at 10Hz with pressure (dB)
0	-113	-124
1	-92	-120
2	-71	-121
Baseline	-150	-150

Time (h)	Magnitude at 10Hz (dB)
Dry	-77
Dry (↑ Area)	-107
Wet	-113
Baseline	-150

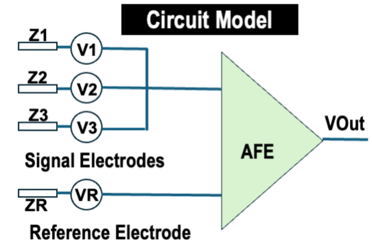


Fig. 3: Impact of pressure and contact area on noise magnitude for wet and dry electrodes, demonstrating that increasing contact pressure and area reduces noise, with contact impedance being the dominant factor.

In contrast, dry electrodes exhibit no significant change in noise over the same period, with the noise levels at t=0 and t=2h remaining similar (orange and green lines, respectively). After 2 hours, the difference in noise between dry and wet electrodes is reduced to about 4 dB (comparing the green and violet lines). This finding suggests that dry electrodes are more stable for long-duration EEG recordings, as they do not experience significant noise variation over time, unlike wet electrodes. Additionally, this highlights the influence of contact impedance on the noise performance of wet electrodes as the electrolyte gel dries up with time.

Insight 1 (a): *Initially, wet electrodes have ≈35dB less noise as compared to dry electrodes suggesting that $Z_{contact}$ and/or $Z_{electrode}$ of wet electrodes $\leq Z_{contact}$ and/or $Z_{electrode}$ of dry electrodes. However, with time, the electrolyte dries up in wet electrodes, which makes $Z_{contact}$ bad, and eventually, it becomes worse than dry electrodes.*

Insight 1 (b) : *In the case of dry electrodes, the noise stays the same at t=0 and t=2h, as $Z_{contact}$ and $Z_{electrode}$ of dry does not change with time, suggesting that dry electrodes are more reliable for longer duration EEG measurements.*

2) Increase Electrode Contact for a better SNR: Fig. 3(a) shows noise magnitude with varying pressure applied to wet and dry electrodes using a wristband. Initially (t=0), there is an 11 dB difference between the no-pressure and pressure-applied cases (orange vs. dark green lines). This difference grows to 28 dB at t=1 hour (sky blue vs. purple lines) and 40 dB at t=2 hours (light green vs. navy blue lines), showing the increasing impact of pressure on $Z_{contact}$. Noise variation among the pressure-applied conditions is minimal, confirming $Z_{contact}$ as the main noise contributor, consistent with capacitive HBC findings [10]. A similar trend is observed with dry electrodes, though discomfort increases with pressure.

Increasing electrode area is another way to improve skin contact [14]. This was proved by connecting three dry electrodes in parallel to represent one large signal electrode(Fig. 3(b)), which improved the signal by about 30 dB. This effect was modeled using the circuit shown in Fig. 3(b). Here, *Z1, Z2, & Z3* represent signal electrodes, each with area A, and corresponding noise sources V1, V2, and V3. With equal impedance and uncorrelated noise, the combined output voltage is inversely proportional to $\sqrt{A}$. However, no reduction occurs if the noise sources are fully correlated, as seen in Eqs. (3), (4). Consider voltage values V_1, V_2, and V_3:

$$V_{total} = \frac{1}{3}(V_1 + V_2 + V_3) \tag{1}$$

The effective RMS noise voltage for the combined signal, V_{total}, over N measurements is:

$$\begin{aligned} V_{\text{total,RMS}} &= \sqrt{\frac{1}{N}\sum_{i=1}^{N}\left(\frac{(V_1+V_2+V_3)}{3}\right)^2} \\ &= \frac{1}{3}\sqrt{\frac{1}{N}\sum_{i=1}^{N}(V_1^2+V_2^2+V_3^2+2V_1V_2+2V_1V_3+2V_2V_3)} \end{aligned} \tag{2}$$

Assuming *identical but uncorrelated* noise sources:

$$V_{\text{total,RMS}} = \frac{1}{3}\sqrt{(3V_{\text{RMS}}^2)} = \frac{V_{\text{RMS}}}{\sqrt{3}} \tag{3}$$

For fully correlated noise sources, the terms

$2V_1V_2, 2V_1V_3, 2V_2V_3$ will not cancel out, hence:

$$V_{\text{total,eff}} = V_{\text{RMS}} \tag{4}$$

Insight 2 (a): *Applying pressure, and increasing contact area significantly reduces $Z_{contact}$. It is also seen that $Z_{contact}$ makes more contribution in noise than $Z_{electrode}$.*

Insight 2 (b): *The noise pickup from electrodes is uncorrelated, which explains why an increased contact area leads to a reduction in noise magnitude.*

3) Noise correlation is influenced by the conductivity of the external medium: Fig.4 shows the setup and results of the next set of experiments conducted to examine the effect of noise when electrodes are placed in different mediums. After analyzing the forearm case, we placed electrodes in two different mediums of different conductivity: salt water and air. It is seen that the noise magnitude in salt water is ≈8dB less than in air. This observation can be attributed to the properties of salt-water having a higher conductivity and relative permittivity than air (Eqs.(5) (a), (b) and (c)), leading to more of a difference in thermal noise ($V_{Thermal,SaltWater} < V_{Thermal,Air}$).

$$\sigma_{SaltWater} > \sigma_{Air} \tag{5a}$$

$$\epsilon_{SaltWater} > \epsilon_{Air} \tag{5b}$$

$$V_{Thermal} = \sqrt{4KTR} = \sqrt{4KT\rho\frac{L}{A}} = \sqrt{4KT\frac{1}{\sigma}\frac{L}{A}} \tag{5c}$$

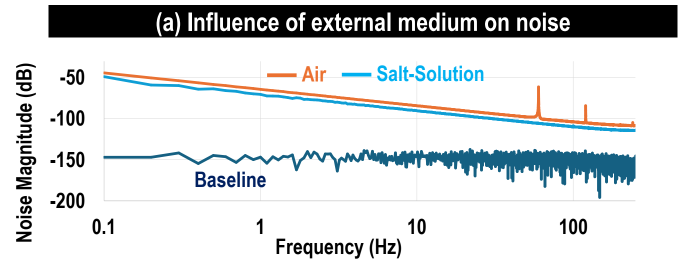
Secondly, thermal noise in a conductive medium is spatially correlated over small distances due to uniform charge movement. This means that the noise picked up from each electrode in salt water is more correlated than in air. Due to the differential pick-up mode, the noise gets canceled out; hence, we see a lower noise magnitude in salt water. This phenomenon is also observed in underwater communication [15]. Due to faster propagation, higher density, and lower absorption in water, noise is likely to be more spatially correlated over short distances than air. The correlation coefficient between two sensors in a noise field can be approximated by Eq. (6). Hence, for lower frequency, the correlation is higher in saline water than in air for the same sensor spacing. A similar phenomenon is observed in HBC channel measurements across various mediums [16], [17].

$$\rho = \sin\frac{kd}{kd}; kd = \frac{2\pi}{\lambda} \tag{6}$$

Insight 3*: The conductivity of the external medium plays a role in noise magnitude. In a more conductive medium, the noise pickup from each electrode is more correlated, leading to its cancellation in the differential modality. On the other hand, the noise pickup from each electrode in a lesser conductive medium is uncorrelated, leading to a higher noise magnitude.*

D. Observation of Flicker Noise Across All Experiments

Our experiments show that both wet and dry electrodes exhibit flicker noise at lower frequencies that overlap with key EEG signal bands. This 1/f noise is common in electronic systems and arises from random charge trapping at interfaces or junctions. Studies have observed this phenomenon in vacuum tubes, BJTs, and MOSFETs, where flicker noise results from charge fluctuations caused by surface impurities or trapping at the oxide-semiconductor interface [18], [19].



Magnitude at 10Hz in Air (dB)	Magnitude at 10Hz in Salt solution (dB)
-84	-92

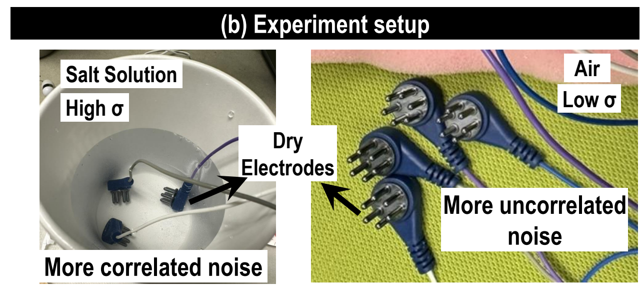


Fig. 4: The influence of the external medium on noise. (a) Noise magnitude is higher in air due to uncorrelated noise sources, while a conductive medium like salt water reduces noise through partial cancellation. (b) Experiment setup with electrodes placed in air (low σ) and a salt solution (high σ).

In EEG systems, flicker noise likely originates from similar mechanisms, particularly Faradaic processes at the electrode-skin interface [2], [20]. Variations in skin impedance cause current fluctuations, resembling charge trapping effects in electronic components. This similarity suggests that flicker noise in EEG systems follow similar underlying mechanisms as in electronic systems. Although reducing flicker noise is beyond the scope of this paper, our findings confirm its presence and its role in degrading EEG signals. Addressing this issue will be crucial for improving EEG reliability.

IV. Conclusion and Future Work

This study provides a comprehensive analysis of system parameter effects of noise in dry electrodes, supported by experiments. We identify $Z_{contact}$ and uncorrelated noise pickup as significant contributors to high SNR in EEG. We demonstrate that optimizing contact impedance through pressure and contact area adjustments can achieve a ≈50dB and ≈30dB reduction in noise, respectively. Furthermore, our analysis of spatial correlation of noise and observation of flicker noise contribute valuable insights for designing more reliable dry electrode EEG systems. These findings have significant implications for developing long-term wearable EEG devices, particularly in scenarios where wet electrodes are impractical. Future research will aim to integrate these contributions into a long-term wearable EEG Cap, enhancing signal quality across diverse operating environments.

References

[1] C. Gondran, E. Siebert, S. Yacoub, and E. Novakov, “Noise of surface bio-potential electrodes based on nasicon ceramic and ag- agcl,” *Medical and Biological Engineering and Computing*, vol. 34, pp. 460–466, 1996.

[2] M. R. Baidillah, R. Riyanto, P. Busono, S. Karim, R. Febryarto, A. Astasari, D. Sangaji, and W. P. Taruno, “Electrical impedance spectroscopy for skin layer assessment: A scoping review of electrode design, measurement methods, and post-processing techniques,” *Measurement*, p. 114111, 2024.

[3] H. Hinrichs, M. Scholz, A. K. Baum, J. W. Kam, R. T. Knight, and H.-J. Heinze, “Comparison between a wireless dry electrode eeg system with a conventional wired wet electrode eeg system for clinical applications,” *Scientific reports*, vol. 10, no. 1, p. 5218, 2020.

[4] Y. Fu, J. Zhao, Y. Dong, and X. Wang, “Dry electrodes for human bioelectrical signal monitoring,” *Sensors*, vol. 20, no. 13, p. 3651, 2020.

[5] J. Rodrigues, M. Weiß, J. Hewig, and J. J. Allen, “Epos: Eeg processing open-source scripts,” *Frontiers in neuroscience*, vol. 15, p. 660449, 2021.

[6] C. Q. Lai, H. Ibrahim, M. Z. Abdullah, J. M. Abdullah, S. A. Suandi, and A. Azman, “Artifacts and noise removal for electroencephalogram (eeg): A literature review,” in *2018 IEEE Symposium on Computer Applications & Industrial Electronics (ISCAIE)*, pp. 326–332, IEEE, 2018.

[7] N. Alharbi, “A novel approach for noise removal and distinction of eeg recordings,” *Biomedical signal processing and control*, vol. 39, pp. 23–33, 2018.

[8] P.-W. Chen, C.-W. Huang, and C.-Y. Wu, “An 1.97 μ, w/ch 65nm-cmos 8-channel analog front-end acquisition circuit with fast-settling hybrid dc servo loop for eeg monitoring,” in *2018 IEEE International Symposium on Circuits and Systems (ISCAS)*, pp. 1–5, IEEE, 2018.

[9] D. Yates, E. Lopez-Morillo, R. G. Carvajal, J. Ramirez-Angulo, and E. Rodriguez-Villegas, “A low-voltage low-power front-end for wearable eeg systems,” in *2007 29th Annual International Conference of the IEEE Engineering in Medicine and Biology Society*, pp. 5282–5285, IEEE, 2007.

[10] S. Maity, M. He, M. Nath, D. Das, B. Chatterjee, and S. Sen, “Bio-physical modeling, characterization, and optimization of electro-quasistatic human body communication,” *IEEE Transactions on Biomedical Engineering*, vol. 66, no. 6, pp. 1791–1802, 2018.

[11] Protech. https://www.protechsite.com/_subpage/eng/product/list.php?viewMode=view&ca_id=010201&sel_search=&txt_search=&page=1&idx=26. Accessed: 2024-10-10.

[12] “EEG Snap Electrodes — shop.openbci.com.” https://shop.openbci.com/products/eeg-snap-electrodes. [Accessed 10-10-2024].

[13] “Ads1299.” https://www.ti.com/tool/ADS1299EEGFE-PDK. [Accessed 10-10-2024].

[14] E. Huigen, A. Peper, and C. Grimbergen, “Investigation into the origin of the noise of surface electrodes,” *Medical and biological engineering and computing*, vol. 40, pp. 332–338, 2002.

[15] I. F. Akyildiz, D. Pompili, and T. Melodia, “Underwater acoustic sensor networks: research challenges,” *Ad hoc networks*, vol. 3, no. 3, pp. 257–279, 2005.

[16] S. Sarkar, M. Nath, A. Datta, D. Yang, S. Maity, and S. Sen, “Material property based analysis of electro-quasistatic human-structure interactions,” in *2023 IEEE Biomedical Circuits and Systems Conference (BioCAS)*, pp. 1–5, IEEE, 2023.

[17] S. Sarkar, M. R. Chowdhury, Q. Huang, and S. Sen, “Material property based analysis of human body communication in body resonance regime,” in *2024 IEEE MTT-S International Microwave Biomedical Conference (IMBioC)*, pp. 69–71, IEEE, 2024.

[18] A. Van der Ziel, “Flicker noise in electronic devices,” in *Advances in electronics and electron physics*, vol. 49, pp. 225–297, Elsevier, 1979.

[19] R. F. Voss, “1/f (flicker) noise: A brief review,” in *33rd Annual Symposium on Frequency Control*, pp. 40–46, IEEE, 1979.

[20] K. Polachan, B. Chatterjee, S. Weigand, and S. Sen, “Human body–electrode interfaces for wide-frequency sensing and communication: A review,” *Nanomaterials*, vol. 11, no. 8, p. 2152, 2021.