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tACS generator as method for evaluating EEG electrodes: Initial validation using pig skin

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Abstract— **Electroencephalogram (EEG) systems commonly used in laboratory environments are rapidly moving towards real-world applications. In this paradigm shift, new methods for recording and classifying EEG signals are necessary. However, it is challenging to validate the efficacy of new electrodes or other data acquisition components when the target signal cannot be controlled. Here, we propose and validate a method for using an attenuated tACS unit to generate a ground-truth signal in conjunction with porcine skin to determine electrode effectiveness. We highlight the utility of this approach by objectively comparing the signal quality of a chirp signal though porcine skin as measured by four different EEG electrodes, three "dry" and a standard hydrogel. We believe a similar approach could be used with transdermal signals on live humans to observe effects of motion, electrode type, or environment on EEG in order to improve design for use in noisy, non-laboratory settings.**

I. Introduction

The electroencephalogram (EEG) is the premier tool for noninvasively measuring brain activity. For its application, an array of electrodes is applied to the scalp to record microvolt potential changes. These are processed by a range of algorithms for a variety of purposes from diagnostics to brain-computer interfaces [1]. However, the current gold standard for EEG electrodes involves a layer of conductive gel, either integral to the electrode or placed via an applicator, to facilitate contact with the scalp. This leads to a series of problems, some of which include: decreased patient comfort (gel in the hair is undesirable), long set up times, short usability windows, and costly disposable electrodes. A desirable solution to this is a so-called "dry" electrode, here defined as "an electrode that requires no additional gel, will not become unusable due to evaporative effects, and leaves no untoward residue." Several types of dry electrodes have been developed, such as conductive polymers [2], metal pins, shaped hydrogel, capacitive off-skin, and scalp-penetrating [3]. Each of these electrodes have advantages and disadvantages, however, they have one common problem: there exists no standard test for determining how well they interface with biological tissue in a real-use case [4]. Using human scalp and hair as a test scenario is difficult since we do not know the actual signals generating the scalp EEG potential. Due to this, we cannot mathematically determine how accurately we are measuring those sources, nor what effect each electrode type has on the EEG measurement when used.

The current best solution to this problem has been the development of EEG "phantoms" which mimic the conductive properties of bulk tissue [5]. Two previous examples are: dipoles seated in a conductive gel molded into a head [6] and carbon-doped materials with embedded signal electrodes [7]. While these phantoms are useful for studying the efficacy of electrodes, they do not mimic the properties of skin, the interference of the electrode and *stratum corneum,* nor the effects of hair. The differences between biological and phantom materials prevent accurate characterization of unusual artifacts, such as motion [5]. Thus, phantoms have the advantage of known sources, but the disadvantage of poor realism when attempting to determine electrode properties, can be expensive to create, and may be difficult to maintain. A better test is needed to evaluate EEG electrodes, especially as they perform in real-world scenarios.

Our proposed alternative approach is to inject a transdermal signal into a human scalp, establishing a ground-truth signal using the human as a pseudo-phantom, providing real tissue properties while ignoring subjects' unknown underlying EEG, in order to evaluate the effectiveness of electrodes. Transcranial electrical stimulators (tES), using DC or AC signals (tDCS, tACS respectively) have become more prevalent in the last decade, and have generally been accepted as a safe method for passing current through the scalp [8]. These units generate a specific signal with the intention of passing it through the brain for stimulation effects. However, we postulate that if this signal were attenuated to microvolt-level, it would not pass through the skull, rather remaining subcutaneous, and could be used as a ground-truth. An attenuated tACS-human head paradigm would be easy to obtain, relatively inexpensive, and easy to standardize across laboratories.

We propose to eventually test the effectiveness of using a live human as an EEG pseudo-phantom; however, we first need to establish a tACS-skin phantom protocol as a proof of concept. Particularly, we must ensure that signals injected into the skin at such a low level (microvolts) do indeed propagate in a predictable, non-distorted manner. The best tool for this characterization is electrical impedance spectroscopy (EIS); however, EIS machines are not generally approved for use on live humans due to safety concerns, and human-safe equipment would be single-purpose and cost prohibitive in most cases. Therefore, for these initial tests, we describe use of a pig-skin substitute for human skin with a tES providing a ground truth signal as an inexpensive, easy to set up, disposable test for quantifying the signal efficacy of multiple dry electrodes.

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II. Methods

A. Skin Preparation

Ice-glazed porcine skin was acquired from a local butcher shop within 24 hours of slaughter, where it had been prepared by scalding at 60 C water for 5-6 minutes and singed as necessary to remove hair. Unsinged pieces were selected and thawed at 4 C overnight, then cut to size to provide sufficient area for electrode placement without undue excess. Selected pieces measured 100 x 150 millimeters +/- 10%, with an average thickness of 4mm. To compensate for dehydration of the epidermal skin layers from glazing, the skin was rehydrated at 40 C, porcine body temperature [9], in 50 mM Calcium Chloride/20 mM HEPES solution due to its approximation of biological ionic availability as reported in [7]. The skin was heated for approximately one hour in a bath, then placed in a glass specimen dish where the surface was blotted dry with a paper towel.

B. Electrical Impedance Spectroscopy

Covidian Kendall foam/hydrogel electrodes were used in conjunction with a Solartron Modulab XM for EIS measurements. The electrode sites were prepared with 70% isopropyl alcohol/pumice prep pads before electrode attachment. A 5 mV rms signal sweep from 1Hz to 1kHz was performed using a four-electrode potentiostatic impedance measurement method (one source-sink pair, one measurement pair) to remove electrode junction effects from the EIS data. Measurements were taken both epidermally and transdermally to discern any surface or bulk frequency dependent attenuation or phase effects.

C. Dry Electrode Evaluation

Three common, commercially available dry electrodes were used as examples: Mindo Spring Loaded Sensors, g.tec g.SAHARA, and Cognionics Flex Sensors. These were compared against the standard Covidian hydrogel electrode. The electrodes can be binned into two conduction types: ionic and electronic. The Covidian hydrogel and Cognionics Ag/AgCl tipped dry electrodes are ionic conductors, while the g.tec and Mindo electrodes are gold-pin electronic conductors. A set of four of each electrode type were placed on each piece of skin as per figure 1. A g.tec g.USBamp with a g.SAHARA amplifier unit was used for recording. A custom Soterix Medical MXN-32 HD-tES voltage-controlled unit with arbitrary stimulation capacity was used for signal generation, with an accompanied Soterix Medical MultiChannel Attenuator to reduce the signal from stimulation to typical EEG scalp levels (attenuation factor of 20,000, reducing a +/- 2 V stimulation waveform to +/- 100μV). We generated a 0-50Hz bidirectional sweep chirp signal, created and played at 1000Hz for 10s. This signal was played to two of the output channels, with one channel connected directly to the g.tec amplifier for a ground truth comparator, and the other connected to the skin via a foam/hydrogel Covidian electrode.

Nine pieces of skin, prepared as above, were tested with each set of electrodes. The first trial set was done with gel electrodes. For each subsequent trial set, all electrodes except the stimulation source/ground and the reference/ground g.tec connected combinations were exchanged for the appropriate dry electrode. In all cases, an approximately 1kg glass plate was placed over the electrodes to press them to the skin. For each electrode set, three 20s trials (10s of signal with 5s of silence on either side) were performed.

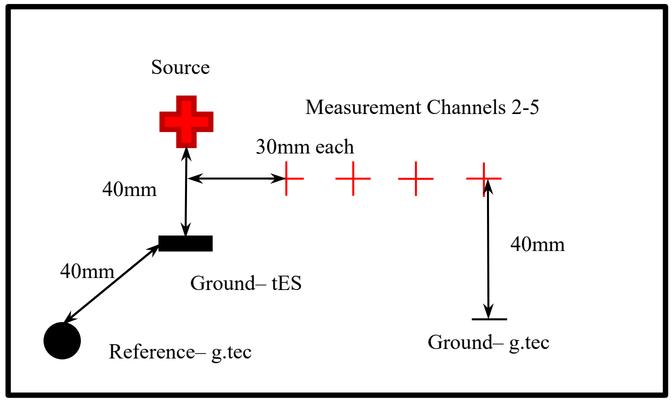

Figure 1. *Electrode setup for pig skin tACS testing. Channel 1 of the g.tec was a pass-through channel attached to a separate but identical source. All measurements are center to center. The source, ground, and reference electrodes remained foam/hydrogel, while all measurement electrodes were switched to the appropriate type during each trial.*

D. Analysis

Data were analyzed in MATLAB R2015b. First, the raw signal was filtered with a band-stop filter from 59-61 Hz to remove mains noise, though it is worth noting that the g.SAHARA has an onboard 0.1-40Hz bandpass filter. Then, each trial's power spectrum from 0-40Hz was computed via Welch's method. These spectra were averaged over the 3 trials. Each of these spectra were compared to the pass-through signal's spectrum to determine the attenuation across frequency from 0-40Hz. In parallel, the filtered signals were extracted by the start-stop markers, leaving only the 10s of stimulation in each waveform. Each recording electrode was compared to the ground-truth signal via magnitude squared coherence (MSC), where 1 represents perfect coherence and zero is none. We chose MSC as it allows a comparison that is sensitive to noise, but robust against positive or negative change in signal, as compared to squared-difference methods. Given the impedance data in figure 2, there is no reason to expect any non-linearity in the transfer function between the signals, thus the linearity assumptions of MSC are not violated. The MSC was binned across the common EEG delineations of *delta* (1-4Hz), *theta* (4-8Hz), *alpha* (8-13Hz), *beta* (13-20Hz), and *gamma* (20-40Hz) [10]. These bins were fit to beta distributions, as MSC is bound to [0,1].

III. Results

A. Electrical Impedance Spectroscopy

Figure $2_{a,b}$ shows a sample of EIS data. For EEG frequency ranges (1-100Hz), the EIS response is approximately flat. Though both changes in resistance and phase were significantly different from zero in the epidermal case, and phase alone was significant in the transdermal case (at P<.05 significance), the magnitude of such is small enough to be considered as having no effect, suggesting porcine skin to be a suitably stable surrogate without inherent signal distortion.

B. Dry Electrode Attenuation

Figure 3 shows averaged attenuation spectra from the same on-skin electrode type from all skin samples and positions (relative to pass-through), as the individual spectra did not differ significantly. It is clear that the signal is somewhat distorted in the dry electrodes. Phase shift was tested, but cases with nonzero maximum cross correlation phase shift were found to be close to zero and within statistical error. There is a significant difference in the cross-frequency noise floor for the wet vs Ag/AgCl vs gold electrode examples.

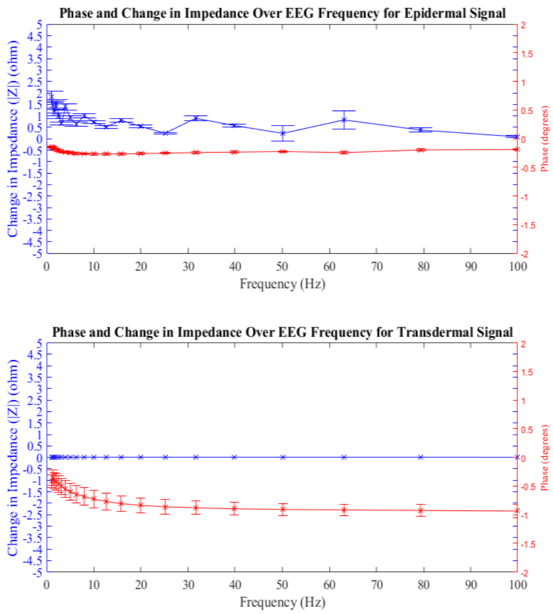

Figure 2. *a) Epidermal averaged EIS data for porcine skin with standard error. b) Transdermal averaged EIS data for porcine skin with standard error. Note the relatively small phase response and impedance change.*

This assessment shows up to 10 dB of difference in noise floor between g.tec and Cognionics electrodes in low frequency ranges. When the chirp is examined across all frequencies, a $1/f$ relationship is noted for the noise profile. This is oddly consistent with the idea of EEG power following $1/f$ and warrants further investigation. As the tACS device is experimental, we confirmed it was supplying consistent signal over several trials prior to this study, and in each trial we recorded the ground truth signal for comparison to remove any potential device variance from our calculations.

C. Dry Electrode Coherence

We found that, similar to attenuation, MSC was significantly higher for ionic conducting electrodes (Hydrogel and Cogionics) and exhibited a linear relationship with frequency. Figure 4 summarizes the mean and standard error of the MSC for each frequency bin.

Each MSC data set was compared to a beta distribution fit via a Kolmogorov-Smirnov (KS) goodness-of-fit test, and for all MSC < 0.9, the beta fit was not rejected at P<.05. At high values of MSC (>0.9), the beta distribution failed to fit. This is likely due to edge effects associated with MSC's ceiling, and the KS test's inaccuracies when the fit focuses near an edge. Z-scoring would allow for a fit in these cases, but such complexities are beyond the scope of this paper.

After initial fitting, each distribution was compared iteratively against all other distributions, and it was found that at P<.05 significance, every distribution was unique. However, it is clear that the ionic conductors tend to follow each other and the gold conductors do likewise. This is to be expected as per previous results. At high frequency (gamma-band), these differences shrink to the point of being nearly inseparable.

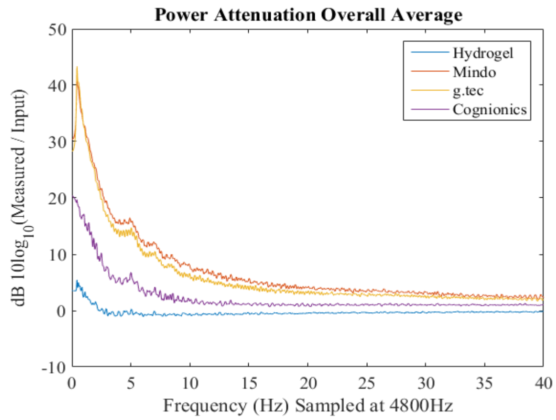

Figure 3. *Attenuation spectra of each electrode averaged across all trials. Note the relatively high noise in the low frequency bands, especially in the gold electrodes. This measurement is only possible through the presence of a ground-truth signal.*

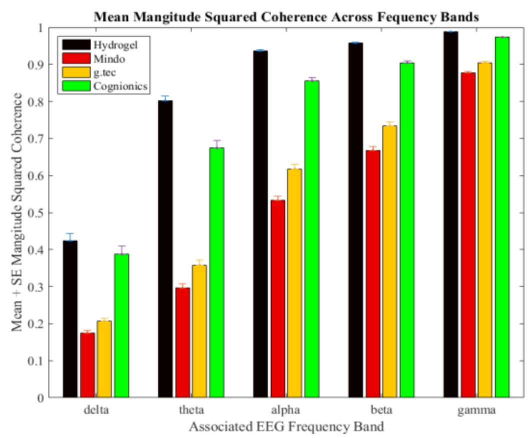

Figure 4. *Mean + SE of the fit beta distributions across frequency bins. As sampling was consistent per frequency unit, sample size shifts with the size of the bin, thus explaining the decrease in SE as bands expand. Note the direct relationship of frequency and MSC.*

IV. Discussion

It has been reported in the literature that skin tissue does not significantly low pass filter EEG frequencies [10], which is further supported with the EIS data presented here. Given this response, we can expect that any observed signal distortion is likely due to the properties of the electrodes rather than the tissue on which the recordings are taken. We checked both transdermal and surface conduction, and found a very small statistically significant phase shift, but only a fraction of a degree. Likewise, the attenuation profile was statistically significant, but the overall change was a small percentage of the base resistance (<0.1%), which should not cause any high-magnitude effects, and certainly not on the scale we observe in later experiments.

We have shown differences in electrode performance using porcine skin, which has been widely used as a close substitute for human skin in other applications [11]. Its electrical properties have been previously characterized and match properties for human skin reasonably well [12]. It is easily prepared, and can be obtained from a butcher at low cost. Meanwhile, the use of a known signal with known degradation properties in tissue helps confirm what has been previously reported concerning dry electrodes, namely their high noise level and signal degradation [13], [14]. The gold electrodes showed higher noise at low frequency, possibly due to their electrical conductance charge coupling vs the ionic conducting Ag/AgCl and hydrogel electrodes. Using a tissue biologically similar to human skin allows us to better predict how the electrodes will behave in a real-world situation. Whereas current phantoms tend to be made from hard plastics or homogenous gels, porcine skin could serve as a low-cost improved outer layer for phantom testing. Driving the testing with an on-skin tACS unit further simplifies this idea, allowing for a quick mock-up with no embedded electronics to rapidly ascertain the conductive efficiency of a new electrode or confirm a current system is indeed functioning correctly. Given the apparent shifts in the MSC of the signal, electrode noise profiles might be able to be calculated with a known signal, then subtracted out in dry EEG system algorithms to compensate for the reduced signal fidelity.

The $1/f$ pattern of noise power in the EEG electrodes observed warrants further investigation. Given this distribution is usually attributed to underlying signal, comparing a scalp-only known signal to a person's EEG would allow for an advantage even a synthetic phantom would be unable to provide [10]. This was an unexpected finding, but further justifies pursuing "human-as-a-phantom" experiments.

Given the fits were rejected at the higher coherence, it is possible that the associated distribution of the noise profile of an electrode may shift across frequency. Once again, this highlights the need for high-quality realistic phantoms for testing EEG electrodes and systems, as minor differences in spectral effects could have a large effect in signal processing. As the signal is quite small with EEG, any minor change in signal-to-noise ratio could extract important features or eliminate features generated by the measurement system.

We have demonstrated the utility of a tACS-skin testbed, and thus, plan to move forward into human testing. Full-power tACS shows no significant danger [15]; attenuated tACS should pose even less. We have shown that the tACS signal will transmit across the surface of the skin as expected, and as a ground-truth signal, is an effective measure of electrode properties. Notably, one challenge to overcome with live humans will be the presence of their own natural electrical signals, which will add noise to the recorded data. However, we believe this can be overcome using trial epoching and averaging, given the ability to tightly control the correspondence between input and output signals with the proposed paradigm.

V. Conclusion and Future Work

We have demonstrated a method for determining fidelity of different types of EEG electrodes using a porcine skin substitute and an easily controllable, safe, EEG-level skin-input signal. Even in these preliminary experiments, we were able to identify factors of interest for future work that, without a known input signal, could be attributed to any number of causes, such as tissue or native signal effects. Further refinement of this model will allow for proper classification of these effects and potentially better design or processing methods to enhance signal fidelity.

As we move forward, we would like to compare EEG to the phantom signal to fully categorize the effects of each electrode in a more realistic setting, and to further validate this model. If a sample of unscalded skin with hair intact were obtained, some basic hair penetration testing could be completed as well. We intend to validate further by comparing measured data to established computational models. Eventually, we will use a live human limb or head in a similar method to further increase utility of this model, and allow for collection of motion artifact in real time. The porcine model is a first step towards our human-as-a-phantom concept.

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References

[1] K. McDowell *et al.*, "Real-World Neuroimaging Technologies," *IEEE Access*, vol. 1, pp. 131–149, 2013.

[2] G. A. Slipher, W. D. Hairston, J. C. Bradford, E. D. Bain, and R. A. Mrozek, "Carbon nanofiber-filled conductive silicone elastomers as soft, dry bioelectronic interfaces," *PLOS ONE*, vol. 13, no. 2, p. e0189415, Feb. 2018.

[3] M. Lopez-Gordo, D. Sanchez-Morillo, and F. Valle, "Dry EEG Electrodes," *Sensors*, vol. 14, no. 7, pp. 12847–12870, Jul. 2014.

[4] A. S. Oliveira, B. R. Schlink, W. D. Hairston, P. König, and D. P. Ferris, "Proposing Metrics for Benchmarking Novel EEG Technologies Towards Real-World Measurements," *Front. Hum. Neurosci.*, vol. 10, May 2016.

[5] B. J. Lance et al., "2012 Year-End Report on Neurotechnologies for In-Vehicle Applications", US Army Research Laboratory, Aberdeen Proving Ground, MD, Tech. Report. ARL-SR-267, June 2013

[6] W. D. Hairston, G. A. Slipher, and A. B. Yu, "Ballistic gelatin as a putative substrate for EEG phantom devices," arXiv:1609.07691 [physics.med-ph], Aug. 2016

[7] T. J. Collier, D. B. Kynor, J. Bieszczad, W. E. Audette, E. J. Kobylarz, and S. G. Diamond, "Creation of a Human Head Phantom for Testing of Electroencephalography Equipment and Techniques," *IEEE Trans. Biomed. Eng.*, vol. 59, no. 9, pp. 2628–2634, Sep. 2012.

[8] S. Zaghi, M. Acar, B. Hultgren, P. S. Boggio, and F. Fregni, "Noninvasive Brain Stimulation with Low-Intensity Electrical Currents: Putative Mechanisms of Action for Direct and Alternating Current Stimulation," *The Neuroscientist*, vol. 16, no. 3, pp. 285–307, Jun. 2010.

[9] L. E. Mount and J. G. Rowell, "Body size, body temperature and age in relation to the metabolic rate of the pig in the first five weeks after birth," *J. Physiol.*, vol. 154, no. 2, pp. 408–416, Dec. 1960.

[10] P. L. Nunez, E. P. of B. E. P. L. Nunez, R. Srinivasan, and A. P. of C. S. R. Srinivasan, *Electric Fields of the Brain: The Neurophysics of EEG*. Oxford University Press, 2006.

[11] A. Summerfield, F. Meurens, and M. E. Ricklin, "The immunology of the porcine skin and its value as a model for human skin," *Mol. Immunol.*, vol. 66, no. 1, pp. 14–21, Jul. 2015.

[12] C. Gabriel, A. Peyman, and E. H. Grant, "Electrical conductivity of tissue at frequencies below 1 MHz," *Phys. Med. Biol.*, vol. 54, no. 16, pp. 4863–4878, Aug. 2009.

[13] Y. M. Chi, T.-P. Jung, and G. Cauwenberghs, "Dry-Contact and Noncontact Biopotential Electrodes: Methodological Review," *IEEE Rev. Biomed. Eng.*, vol. 3, pp. 106–119, 2010.

[14] A. J. Ries, J. Touryan, J. Vettel, K. McDowell, and W. D. Hairston, "A Comparison of Electroencephalography Signals Acquired from Conventional and Mobile Systems," *J. Neurosci. Neuroengineering*, vol. 3, no. 1, pp. 10–20, Feb. 2014.

[15] H. Matsumoto and Y. Ugawa, "Adverse events of tDCS and tACS: A review," *Clin. Neurophysiol. Pract.*, vol. 2, pp. 19–25, 2017.