Drawn-On-Skin Liquid Electrodes for Motion Artifact-Free Neurological Signal Recording

By

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Abstract

Signal quality is of the utmost importance when recording from the human body. In order to maximize the accuracy of biosignals recorded for monitoring and application, electrode systems are tailored to overcome the specific impediments the body's properties cause. In the case of electroencephalography (EEG), electrodes designed for recording are split into two groups: wet and dry, split by the usage or absence of a conductive electrolyte applied to the surface of the scalp. Both of these electrode types have issues that plague their usage. In order to subvert these issues, a conductive ink to apply customizable drawn-on-skin (DoS) electrodes is proposed. The DoS electrode system has demonstrated capabilities to record electrocardiogram (ECG) and electromyogram (EMG) successfully, without motion artifacts caused by movement induced deformation at the skin-electrode interface when compared to conventional electrode types. In applying the DoS electrode system, along with the implementation of protective insulation and encapsulation, a discrete, low visibility, motion artifact-free recording system can be created for high-quality electrophysiological signal acquisition. Herein, the characterization of DoS electrodes is described, and their implementation for motion artifact-free EEG and EOG recording, compared with conventional EEG electrodes, is demonstrated.

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I. Introduction

Electrical biosignals contain a plethora of physiological data that allows medical professionals insight into a person's health and well-being. The skin serves as the conduit through which information about the heart, the muscles, and the brain can be recorded and observed (1). When recording electrical signals from the human body, a solid and stable interface between the electrode and the skin is paramount (2, 3). In order to optimize the connection between the skin and the electrode, the ability for a recording electrode to account for the impedance of the skin-electrode interface and the changes in it caused by motion and time are important aspects that must be accounted for in novel devices. By accounting for these potential recording downfalls, a more superior recording system can be achieved.

Depending on where electrodes are placed, different signals can be recorded and measured. From electrodes placed on opposite sides of the heart, electrocardiograms (ECG) can be measured, giving insight into heart health and activity. From electrodes placed on muscles, electromyograms (EMG) can be measured, giving insight into muscle movements. From electrodes placed around the eye, electrooculograms (EOG) can be measured, allowing the eye to be tracked, and its movements observed. From electrodes placed on the around the head or on the scalp, electroencephalograms (EEG) can be recorded, representing the cumulative activity of groups of neurons, allowing brain activity to be monitored and measured.

For each type of signal, the quality of the electrode-skin interface used for recording is important for maximizing the quality of signal measured. Mathematically, the quality of a signal is quantified by its signal-to-noise ratio (SNR), and the conductive quality of the interface is measured by its impedance. By maximizing the former and minimizing the latter, the quality of the signal recorded, and thus the quality of the information about the patient being recorded, can be maximized. In order to maximize signal quality, impediments to recording must be accounted for. The primary obstacle to skin surface recordings is the outermost layer of the epidermis, the stratum corneum. This epidermal layer is made up of dehydrated skin cells with crosslinked keratin proteins, making it a poor conductor (4). Achieving quality signal recording from the skin surface in spite of the stratum corneum requires strategies to cope with it.

Another recording aspect to account for is user movement. Any application of an electrode to a human being will be subject to motion, ranging from obvious large voluntary motions to less obvious involuntary motions, like the motion the body performs during breathing or muscle contractions underneath the electrode. These movements can create disruptions in recordings, which can lead to issues in analysis of the subject's biosignals (5). Electrodes which do not account for wearer movement allow for two potential motion artifacts: longitudinal, where the electrode loses contact with the skin, and transverse, where the electrode slides on the skin (3). Both artifacts can cause recording disruptions, and therefore motion must be taken into account with electrode design.

When recording EEG specifically, the reduction of motion artifact has been attempted in order to maximize signal quality. While the stratum corneum is the primary issue of the skin-electrode interface, the conduction of EEG signal is also strongly influenced by the skull. EEG cannot record single neuron activity, but rather it represents the cumulative activity of over 100 million neurons (6). As a result of the skull's insulation, scalp EEG is only measured to an amplitude of about 100 μV, while from the surface of the brain directly it can be measured to an amplitude of over 1 mV (7). Because of this low amplitude, motion artifact interference is especially problematic, as the artifacts can have amplitudes many times larger than the desire biosignal (8). In order to minimize the effects of motion artifact, algorithms have been developed to remove the artifacts from recordings (9). However, an ideal recording system would record without motion artifacts in the raw recording, so that no biosignal information is lost in the filtering. Therefore, the development of motion artifact-free EEG electrodes would allow for superior recording.

The advantage of superior recording comes in the form of higher quality data for observation. Beyond recording for the sake of data collection, biosignals can be applied for usage. Specifically, EEG can be used for practical purposes by the system wearer. By identifying specific patterns in brain signal activity, tasks can be performed by computer systems, referred to as brain-computer interfacing (BCI). The most often used brain wave artifact used for BCI applications is the P300 (or P3) event, produced most prominently when the brain processes rare stimuli (10). There has been extensive research into using P300 for applications like communication systems, where the presentation of many stimuli leads to the production of a P300 wave when a desired, rare stimulus is presented, which programs use to form words for the subject, which can provide a means of communication for patients with issues otherwise communicating (11). With a recording system maximized for wearer comfort and signal quality, a superior BCI system can be designed for the wearers who need it.

In order to develop such a superior recording system, the current technology of EEG electrodes, the types as well as their flaws, will first be described. To ameliorate the issues with existing EEG electrodes, a novel type of electrode using a conductive liquid ink to fabricate drawn-on-skin (DoS) electrodes will be presented. The DoS electrode electrical properties will be characterized by demonstrating its recording capability with ECG and EMG usage. Once the DoS electrode's performance is established, the focus will shift to applications on the head for EOG and EEG recording, to demonstrate the full array of advantages presented by DoS electrodes over conventional and prefabricated electrodes, as well as to look to future uses of this novel type of electrode.

II. Existing Technology

i. Wet Electrodes

In the case of EEG recording, the usage of conductive gels or pastes as electrolytes to improve the interface between solid metal electrodes and the surface of the scalp is the gold standard used for measurement, referred to as wet electrodes (12). A diagram of dipole generation to recording via wet electrodes is shown in Figure 1a. When conductive electrolytes are used, recorded signal quality improves, because the addition of the electrolytes improves the overall conditions of recording. By hydrating the skin's surface and filling the skin's ducts, the skin's surface becomes more conductive (13). This hydration also serves to ameliorate the recording issues caused by the stratum corneum (4, 14, 15). The conductive electrolyte properties allow for the transmission of electrical signals from the skin to the metallic electrodes connected to the recording device. Using electrolytes in conjunction with metal electrodes allows for consistent recording in spite of the aforementioned issues with the skin surface. Electrodes used are made with metals such as gold, silver, and silver chloride, with example cup electrodes seen in Figure 1b (16). An alternate strategy previously documented utilized silver ink to print electrodes and circuit boards onto vinyl and polyester to create wet electrodes specifically designed for use with functional magnetic resonance imaging (fMRI) (17).

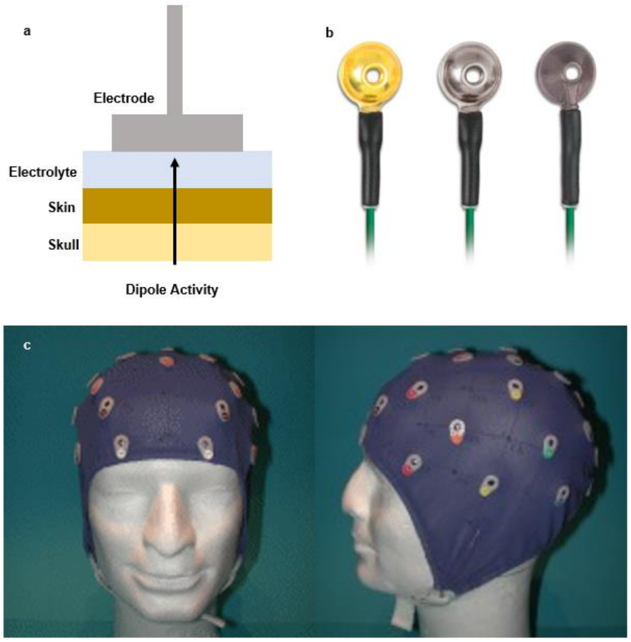

Figure 1 Explanation of wet EEG electrodes. a) Diagram of wet EEG recording. b) Example metal cup electrodes (Natus Grass® Reusable Stamped-Cup EEG Electrodes) c) Example 10-20 EEG cap (BioSemi 32 Channel Cap)

However, some of the properties of the EEG electrolyte also cause issues for the recorder that can limit their effectiveness. In order to achieve the best results from recordings, the electrode locations are sometimes abraded, which can lead to wearer injury (18). This is done to remove layers of the stratum corneum because it has been shown that thinning the stratum corneum lowers impedance (19). In addition, use of the conductive electrolytes for extended periods of time can lead to dehydration, which will lead to degradation in signal quality, and the expended gel may not be able to be adequately replaced for continued recording, which would then require a full reapplication of the entire electrode setup. In the case of excess or incorrect application of the electrolyte, the conductor for two different electrodes can run together, causing a short circuit, leading to recording issues (20, 21). Additionally, the electrolyte-electrode system has the potential to have issues with recordings from locations with hair due to reduced electrode-skin interface area (21). When functional, the electrolyte forms a firm bond with the skin surface, allowing for a solid metal electrode locked into position on an EEG cap to record signals from the scalp surface it would not otherwise be suitable to reach, where the EEG cap serves to hold the electrode in the proper place. An example EEG cap is shown in Figure 1c (22). However, EEG caps have a separate issue that inhibits their usefulness across recording scenarios. In the instances where EEG electrolyte was successfully applied for recording, they are extremely visible and obvious on the wearer's head, which makes it inconvenient in the best cases for long-term studies, due to discomfort caused to the wearer. This has led to alternative recording methods, such as recording from around the ear, in order to reduce visibility (23). Alternatively, collodion can be used as a glue to affix cup electrodes to the scalp with firm adhesion, but removing the electrodes can then become difficult and uncomfortable for wearers (24). There is then a need to develop novel strategies for scalp EEG recording in order to acknowledge and overcome these downfalls of traditional standard systems.

ii. Dry Electrodes

In order to subvert the issues with wet electrodes, alternative recording methods have been proposed and used. Instead of using conductive electrolytes to create an improved bond with the skin surface, these electrodes use specific structural designs in order to improve recording conditions. Electrodes that can record without the application of a conductive electrolyte exist, referred to as dry electrodes (25). One popular type of dry EEG electrode is spike electrodes, which use a number of long conductive projections to record. The smaller variant of spike electrodes is micro-electro-mechanical system (MEMS) electrodes, shown in Figure 2a, which use probes on the micrometer scale that pierce the stratum corneum in order to avoid the issues that the stratum corneum cause recordings (26, 27). The larger variant of spike electrode is shown in Figure 2b and uses spikes on the millimeter scale, which do not penetrate the skin surface (18, 28, 29). More limited quality improvement for EEG recording is achieved using a number of different methods such as utilizing naturally flexible substrates with stretchable electrodes, as shown in Figure 2e (30), and utilizing specific structural designs to flex non-intrinsically flexible materials, as demonstrated in Figure 2c (31).

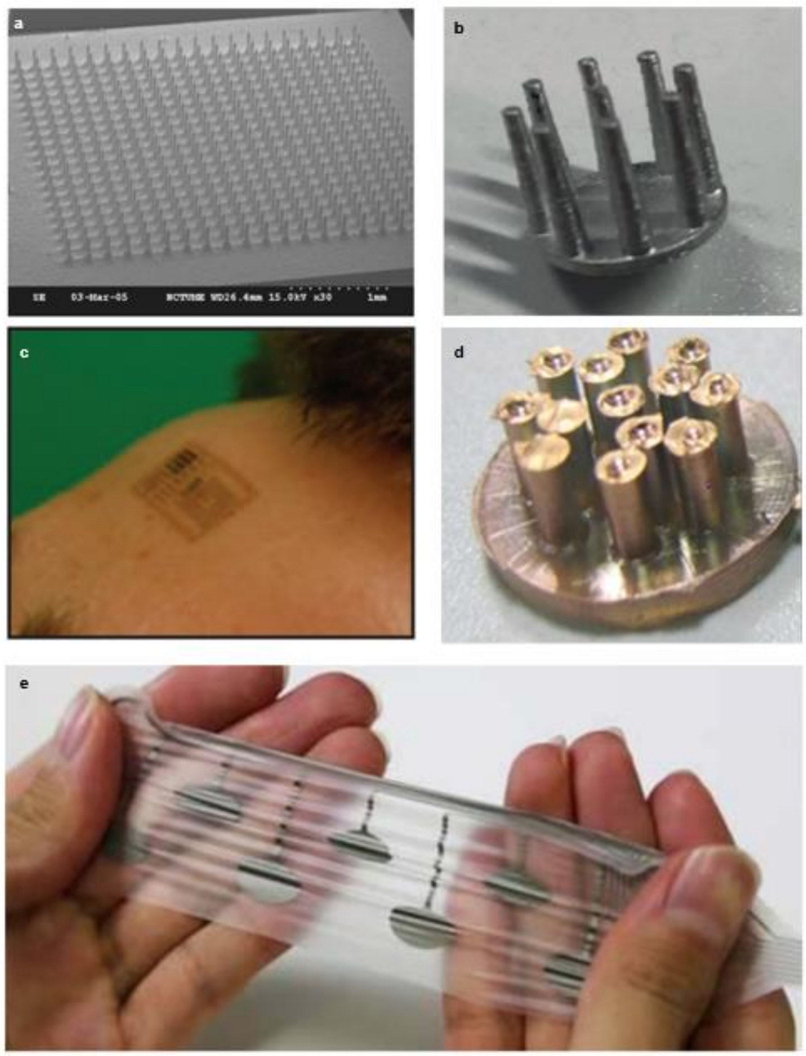

Figure 2 Examples of dry EEG electrodes. a) MEMS spike electrode. b) Millimeter scale pin electrode. c) Skin conformable electrode. d) Flexible spike electrode. e) Flexible sheet electrode.

The dry electrodes are not without their own issues. Electrode material selection becomes more important because the electrode comes into contact with the skin. Metals that cause allergic reactions or corrode in contact with sweat are therefore suboptimal (32). MEMS electrodes pins are thin, and they can break in the skin and even lead to infection, though this is rare (25). Among the dry electrodes, only the larger spikes allow for recording in the presence of hair, where the MEMS spikes cannot penetrate (18). The conformable electrodes are likewise impeded by hair and only work on clear skin. The longer, stiffer pins, while not in danger of breaking, still can cause discomfort in wearers, which has led to the development of flexible pin electrodes as an alternative, pictured in Figure 2d (33).

iii. Active Electrodes

EEG recording systems can also be augmented by the development of active electrodes, which include amplification at the site of recording in order to reduce sensitivity to noise (34). Especially useful for dry electrodes, active components allow for high impedance electrodes to reliably monitor activity (35). Active components allow dry electrodes to serve as more capable alternative approaches to conventional wet electrodes (36). However, active electrodes also have issues. Due to the additional electronics on the electrode, a significant amount of power is required and the size of the electrodes requires further development to minimize (37).

III. Drawn-on-Skin Electronics

i. DoS Ink Characterization

As a novel method to subvert the issues with both wet and dry EEG electrodes, the use of a conductive liquid mixture as ink to fabricate electrodes directly onto the skin is proposed as an alternative method. Under the same premise in which the conductive electrolytes work to improve the contact interface, the DoS electrodes would be able to flow freely to the scalp, to form a direct bond with the skin. Utilizing conductive ink, the liquid would serve as the electrode itself, rather than as a conductor to a separate metal electrode. The ink forms a direct and conformal bond with the skin, flows with the topography of the skin's surface, and fills in any gaps in the surface, creating an electrode that no longer needs an electrolyte to create a bond between the electrode and the skin. This allows for the removal of the electrolyte and the issues it can cause. Additionally, the electrodes being fabricated directly onto the skin means that the cap is no longer required in order to hold electrodes in place, and it can, therefore, be removed entirely as well. By removing the cap and conductive electrolyte, a discrete and more accommodating EEG recording system can be created, without sacrificing performance.

Beyond being more discrete and accommodating, the use of DoS electrodes directly applied onto the surface of the skin creates a more stable adhesion with the skin, leading to less interference as a result of movement, as movement causes deformations in the skin. To optimize the signals recorded from a living subject, a motion artifact-free recording system is required. The DoS electrode recording system has demonstrated the ability to record in spite of disruptions in the skin-electrode caused by movement, giving the potential for its implementation for more superior biosignal recording.

Because the issue with long-term EEG recordings does not exist in the same manner with the proposed electrode system, it would be possible to do long term recording studies or applications with the system. In order to prepare for such a potential usage, full system application would require the inclusion of a protective layer of insulation atop the ink electrodes. The conductive liquid mixture used for ink is water soluble, which would mean that any potential exposure to water, from a shower to the rain, would disrupt the integrity of the DoS electrodes, threatening the quality of long-term recording sessions outside of a controlled laboratory environment. In order to protect the electrodes, while still allowing the electrode wearer to go about, it is proposed here that a layer of insulation atop the liquid electrode be applied in order to preserve and protect it from the environment. By layering the insulation solely and directly over top of the electrode system on the skin, the DoS system achieves both protection and discretion.

The strategy proposed here to utilize the unique properties of conductive liquid ink for EEG recording would allow for more convenient and stable biosignal recording, with the ability to maintain consistent data acquisition over a long period of time, without being overly cumbersome or restrictive upon the electrode wearer. This improved system would hold advantages over the standardly used EEG cap and electrode with electrolyte procedures. Long term stability, combined with an improved overall interface, with less interference from motion artifact, in conjunction with a system that allows for more convenient and less discrete usage opens up the potential for easier long-term EEG studies on both the part of the experimenter and the wearer. These advantages and the properties giving rise to them will be demonstrated and detailed to follow.

The first aspect of the proposed system to be discussed is the chemical makeup of the conductive ink. The ink is a mixture of two component parts: a conductive polymer solution and a solid metallic conductor. The conductive polymer used is poly(3,4-ethylenedioxythiophene) polystyrene sulfonate, in the form of PH1000, a solution of the conductive polymer in water, to which the surfactant triton was added at a 1 wt% concentration. Mixing the PEDOT:PSS with triton was required to ensure that the silver flakes could fully dissolve in the conductive polymer mixture and to improve the ability for the PEDOT:PSS to stretch without fracture, thereby improving the overall suitability of the conductive polymer for skin surface application. The metallic conductor used is silver, in the form of silver flakes (10 µm size, 99.9% trace metals basis, 327077). Silver is the best inorganic conductor, as well as being more cost effective than other conductors like gold or carbon nanotubes, informing the decision to utilize it as the conductor (38). The PEDOT:PSS and the silver flakes are mixed in a 1:2 weight mixture ratio composition, mixed and stored simply at room temperature and under atmospheric pressure. For general electrode fabrication uses, ink was made in 1.5 g batches, with 0.5 g of silver flakes measured into a vial, and 1.0 g of PEDOT:PSS added and stirred to dissolve the silver completely. The PEDOT:PSS and triton mixture is stored at ~4C when not in use, and it was left to stir until it reached room temperature, in preparation for use. The silver flakes are stored in inert gas until needed for ink creation. Mixed silver-PEDOT:PSS ink is stirred prior to use, as the silver and PEDOT:PSS separates when left stationary, rendering the ink ill-suited for usage. The application of the ink electrodes is carried out by using ballpoint pens. The ink cartridges are drained of their original pen ink and cleaned with acetone, and the balls of the pen tips are removed. Using syringes with a 26-gauge needle point, the cartridges are loaded with the ink mixture. If left stationary, silver ink within the cartridge would become dried and unusable, so the application necessarily comes immediately after injection into the cartridge. Ink is applied by writing with the meniscus the ink forms at the tip opening, demonstrated in Figure 3a. Writing solely with the meniscus ensures that the metal of the pen tip does not cause scratches and discomfort to the wearer or disruptions in the continuity of the electrodes.

The specific parameters of the ink mixture were developed via testing. Mixture ratios of 1:1, 1:1.5, 1:2, and 1:2.5 silver to PEDOT:PSS were created and tested, and the 1:2 ratio mixture had the best properties at optimal circumstances and when stretched. Increasing the amount of PEDOT:PSS increased the ink's ability to survive stretches, as measured by the change in resistance caused by a 10% stretch returning to 0% stretch, divided by the resistance measured before any stretch. However, the highest relative quantity of PEDOT:PSS tested, the 1:2.5 ratio, was a much poorer conductor than all other tested ratios, as the concentration of conductive metallic flakes decreases. A graph depicting the properties of each tested ratio is shown in Figure 2e. Therefore, the 1:2 ratio mixture was selected as the optimized ink ratio for this proposal, as the most superior compromise between these two properties. To test the biocompatibility of the ink and ensure it is safe for skin application, it was applied to the bare back skin of shaved mice. After 48 hours of application on the animal, histological analyses were performed. There was no inflammation in any layer of the skin visible in the conductive ink treated skin samples when compared to the control samples of skin, with examples shown in Figure 3c and 3d, demonstrating the safety of skin application (39).

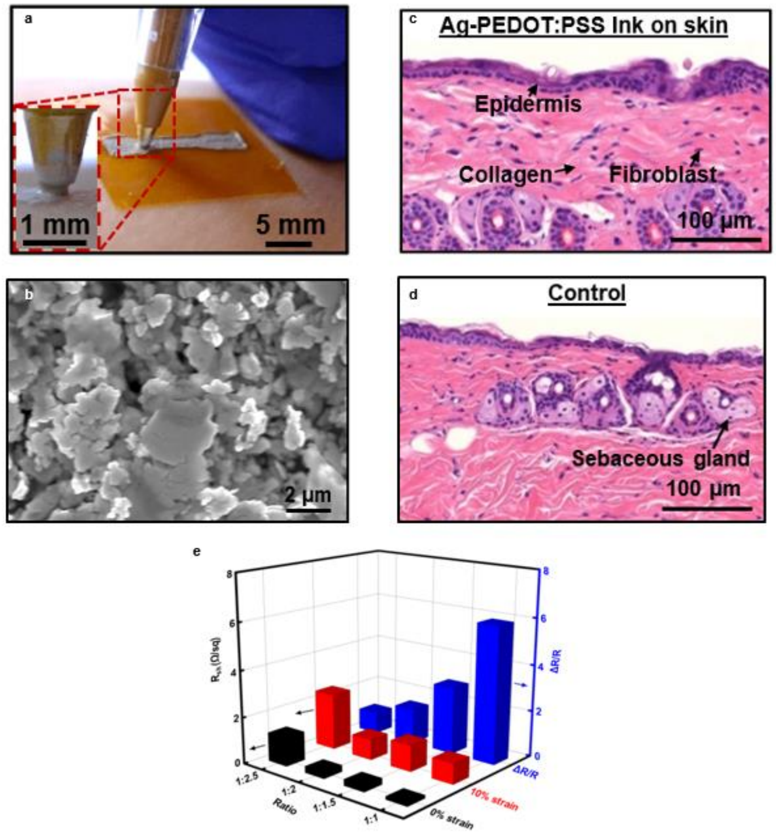

Figure 3 Conductive ink characterization. a) example of writing with ink on human skin. b) SEM image of ink c) Histological analysis of rat skin after 48 hours of ink application. d) Histology test control. e) Comparison graph of different ink ratio resistance, stretchability, and resistance change.

ii. Comparison between DoS and Traditional Electrodes

In order to validate the ability of the ink to function as electrodes for recording biosignals, electrocardiogram (ECG) data was recorded utilizing the DoS electrodes, gel adhesive electrodes (Meditrace 450 Foam Electrodes, Kendall), and gold mesh serpentine electrodes in order to compare the DoS electrodes to known electrode types. Because gold is not intrinsically stretchable or flexible, a serpentine arrangement was designed and fabricated in order to allow for a degree of stretch and flex in the electrode, so as to better serve as a comparable device for motion artifact related testing. Comparisons between the three types of electrodes demonstrated that the proposed ink electrodes function with equivalent or superior quality to the established electrodes. The same subject and recording locations were used across all three electrode types in order to minimize variation. The right wrist was used as the working electrode, and the left wrist was used as the reference electrode across all trials. The recording areas were cleaned with isopropyl alcohol before affixing electrodes in order to reduce the skin impedance levels and ensure optimal recording for all three electrode types. ECG data was recorded using a data acquisition system (RHD 2000, Intan Technologies), with the same amplifier used for each electrode type (RHD2216, Intan Technologies). The same snap cables were used for all trials. In order to adhere the snap cable wires to the DoS electrode, conductive carbon tape (ARclad® 8001-77, Adhesives Research) was adhered to the DoS electrode and the cable was attached to the tape, where the conductive tape served as wire and adhesive both. As can be observed below, the DoS electrode performs favorably when compared with the two other electrode types tested (39). The DoS electrodes were fabricated in 15.0 mm by 15.0 mm squares for recording, shown in Figure 4a, but size can be variable. Square electrodes with sides of lengths 7.5 mm, 5.0 mm, and 2.5 mm are also able to record signals, shown in Figures 3b, 3c, and 3d respectively. However, creating a stable interface between the skin and the recording device became increasingly difficult as the electrode surface area decreased, leading to a decrease in signal quality. The success of the size variance tests shows that the DoS electrodes can be customized to fit the circumstances a particular usage requires of them.

The quality of the recording is measured by SNR. For comparison across the different electrodes used, the SNR was comparable across all three types. The DoS, gel, and mesh electrodes each registered SNR values above 45. When DoS size varied, electrode SNR remained high as the electrode shrunk, though once the electrodes decreased in size past 5.0 mm, the SNR decreased below 45 dB. The small tested electrodes of size 2.5 mm and 1.5 mm both still recorded SNR values above 40 dB (39). SNR values were calculated using the frequency domain method. Because features and morphology of the QRS complex can be recorded with a 50 Hz bandpass, data recorded in the 0-50 Hz range was determined to contain relevant ECG data, and all higher frequencies were considered noise (40).

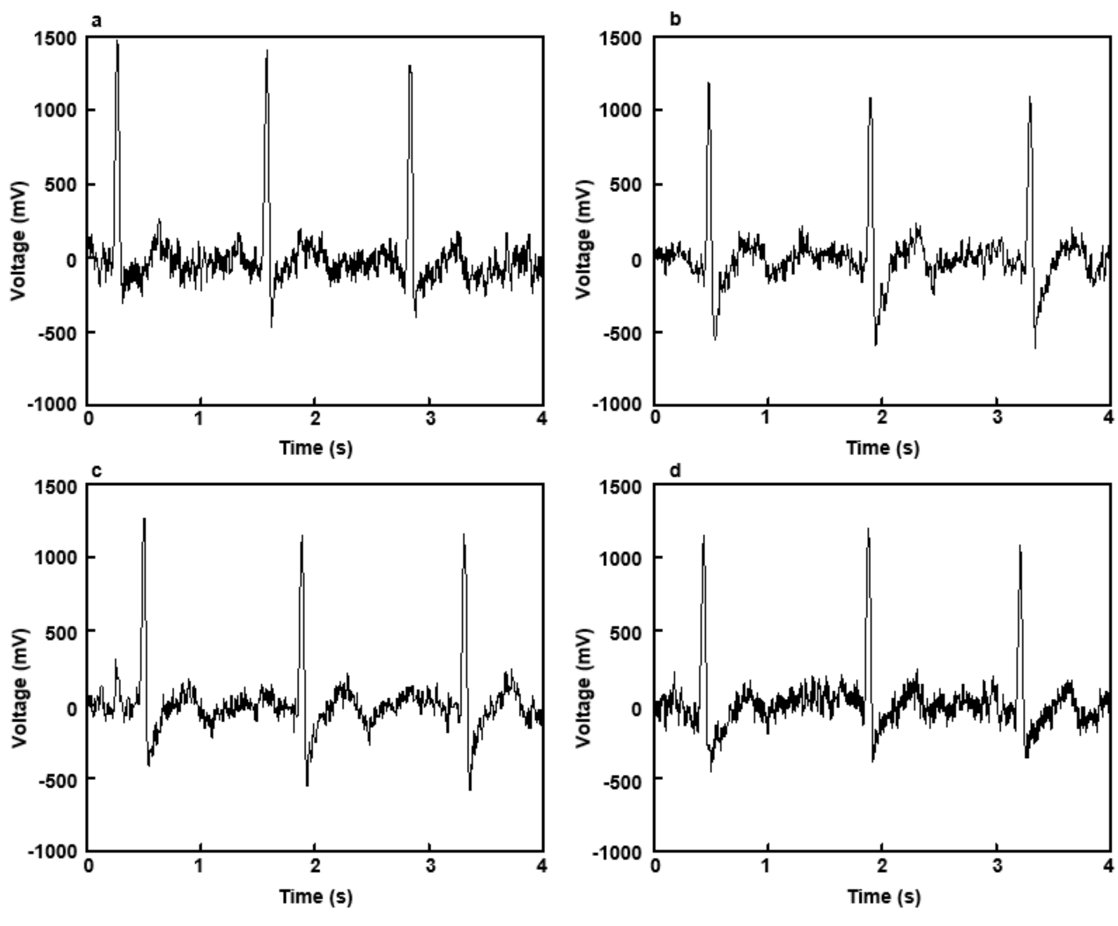

Figure 4 DoS electrode size evaluation. a) 15 mm square electrode test. b) 7.5 mm square electrode test. c) 5 mm square electrode test. d) 2.5 mm square electrode test.

a. Durability

With the potential for implementation with long-term studies, it, therefore, must then be demonstrated that the DoS electrodes can survive wear for extended lengths of time. In order to test electrode durability, the three electrode types were worn on the subject's wrist over the course of a day, recording ECG at three time points: hour 0, hour 4, and hour 7. Hour 0 represents the time immediately after the application of the electrode on the subject. Each of the electrodes was exposed to the environment with no added special protection. The DoS electrode survived the hours of wear favorably, demonstrating little degradation in the quality of recording between hours 0 and 7. The SNR values recorded for each electrode type exceeded 40 dB, even at hour 7. Of all the electrodes, the DoS electrodes maintained the highest signal quality. Additionally, a test for the ability to the electrodes to withstand sweat from underneath was performed, as another factor of durability. Each of the electrode types was applied to the subject, and the subject was exposed to high environmental temperatures, in excess of 35°C, in order to induce sweating. While an increase in signal noise in DoS electrodes can be observed, the ECG waveform and R-peak amplitude remain constant across both dry and sweat-damaged electrodes, both important features of the signal recording. The gold serpentine electrodes showed a more substantial increase in noise when compared to the DoS electrodes. The gel adhesive electrodes showed no increase in noise but showed a noticeable decrease in R-peak amplitude across all peaks after sweat damage. This response can be explained by a drop in conductivity due to dilution. While sweat is an electrolyte, its conductivity is less than that of the gel electrolyte (41). When the subject sweats underneath the electrode, the overall conductivity between the skin and the electrode decreases. Both the sweat and time tests are shown in Figure 5. SNR values calculated for sweat affected electrodes were similar to the values calculated prior to sweat exposure. Both the DoS and gold mesh electrodes endured slight SNR decreases of ~1 dB, while the gel adhesive electrodes experienced a larger quality decrease of ~3 dB (39). SNR values were calculated with the same parameters as previously described.

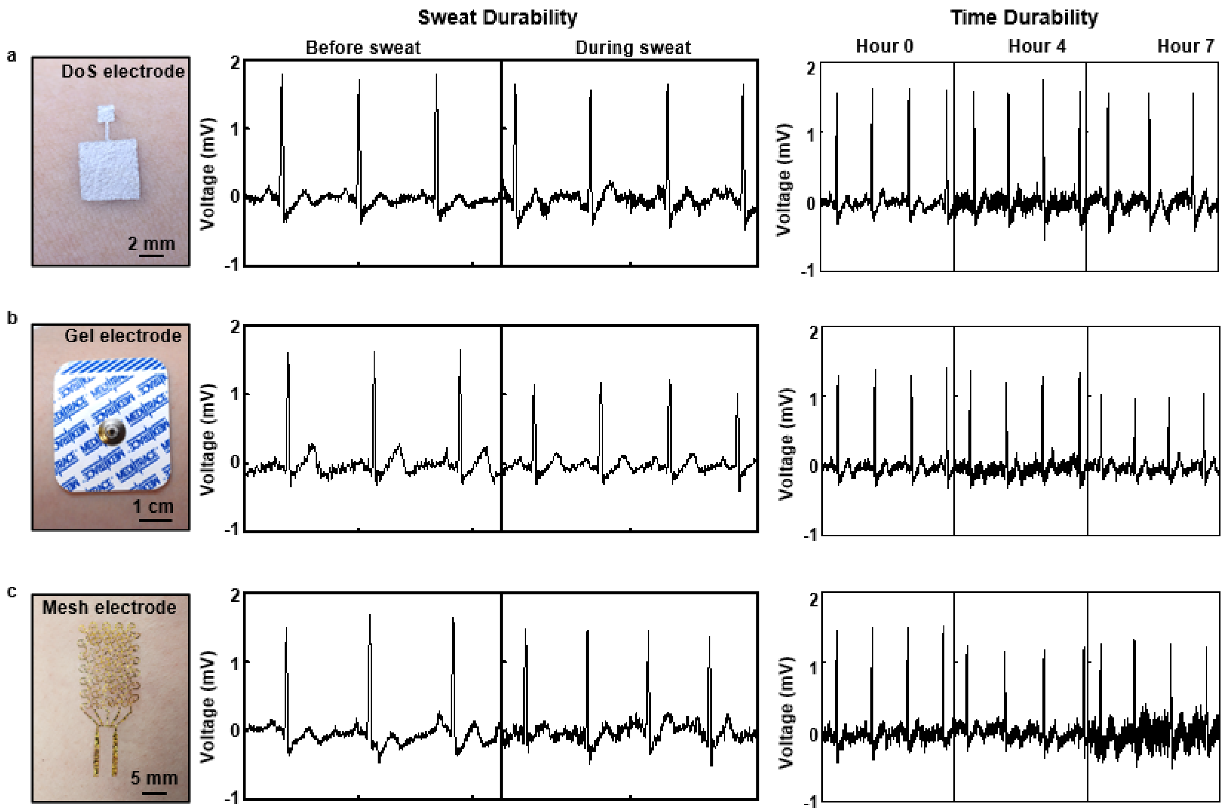

Figure 5 Electrode performances after time and sweat exposure. a) DoS electrode sweat and time tests. b) Gel adhesive sweat and time tests. c) Gold mesh electrode sweat and time tests.

In order to demonstrate the quality of adhesion and resiliency of the DoS electrodes, the physical durability of the electrodes was tested. To test adhesion, a piece of tape was layer on top of the exposed electrode and removed. With firm adhesion, the electrode will survive the tape removal. The gel, without its adhesive package, and the mesh electrode did not survive the tape, while the DoS electrode was not disrupted by the tape. The resiliency of the DoS electrodes was tested by rubbing a finger vigorously over the electrode. In this test too, the DoS electrodes survived, while the mesh electrodes were severely damaged by the physical disruption.

iii. Insulation and Encapsulation

a. Environmental Encapsulation

Because the ink electrodes are made from a dried aqueous solution, when exposed to water, the electrodes break down and will dissolve. Within the laboratory setting, the electrodes can be protected from potential damages, however outside, environmental dangers exist. Where the conductive electrolyte has the potential to dry out after extended periods of wear, the DoS ink maintains its properties after time. Because of this, there is the potential to utilize the DoS electrodes for long-term EEG studies, but environmental factors pose a more prominent danger to the electrodes, and therefore the recording. In order to prevent this weakness from harming the integrity of the recording, protection of the exposed liquid ink electrode is required, so the addition of a layer of insulation is proposed. Using an existing compound as insulation to ensure its safety for human use, liquid bandage (NewSkin) was chosen as the insulator. Liquid bandage is a waterproof adhesive. It dries clear within minutes, so as to not be over incumbent on the wearer and allowing the recorder to monitor the electrode. Given these qualities, liquid bandage was chosen as a suitable insulator for this purpose. With a conductive stainless steel wire laid against the DoS electrode, the liquid bandage is applied and allowed to dry, adhering the wire to the DoS electrode pad and creating a protective layer atop both.

To validate the ability of the liquid ink electrodes to withstand dissolving under the liquid bandage, ECG from the same location on the same subject as previous experiments was recorded with insulated liquid ink electrodes. As can be observed, the addition of the encapsulation layer did not disrupt the recording. A single layer of the liquid bandage will not damage the ink electrode, though layering additional layers causes issues. The method of liquid bandage removal is layering an additional layer of liquid bandage over the top of the hardened protective layer, causing it to dissolve, allowing it to be removed. This dissolution can get underneath the ink layer, and cause it to break up, which would cause issues with the electrode's recording capabilities. As seen in Figure 6, a single layer of encapsulation does not interfere with the ability for the DoS system to record, allowing for completely protected electrodes to be fabricated directly onto the skin.

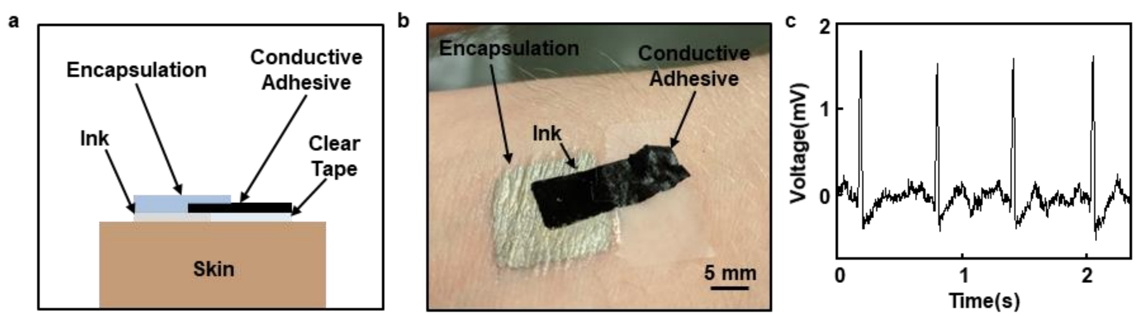

Figure 6 Demonstration of DoS electrode encapsulation. a) diagram of electrode encapsulation setup. b) Photo of application on human skin. c) Example ECG recorded during the experiment.

b. Skin Insulation

The setup hereto described, while functional, would be as cumbersome and unwieldy as any conventional system. Just as a traditional EEG system uses a cap connected to a recording device with wires from each electrode, the current proposed DoS system is connected to a recording device with wires from each electrode, and this connection meant that the system is not ideal for user wear or movement, as wire induced artifacts can still affect the recording (42). Therefore, it is proposed that in addition to using the ink to draw electrodes, the ink could also be used to draw wires on the subject's skin, in order to arrange pads for more convenient connection of the electrode to the recording device, allowing for connections between recording device and electrode away from the actual site of recording. Drawing with the ink on bare skin would lead to biosignal contamination along the wire, which would render moot any recording at the electrode itself. Therefore, a layer of insulation between the bare skin and the DoS wire connection would be required. Because liquid bandage dissolves its own hardened form, it would be ill-suited to serving as both insulation from the skin and insulation from the environment. As a result, a second material is thus proposed to serve as the insulation from the skin. Medical adhesive (Pros-Aide Adhesive) is used, since it is already used in medical situations, meaning its safety with human applications is not in question. It was confirmed to serve as an insulator by applying a layer to a subject's skin and testing the resistance in the connection between nearby exposed skin and the area covered by medical adhesive. The resistance was measured and found to be undetermined, because no current flowed between the two regions, proving that the adhesive would function as adequate electrical insulation for the proposed usage. Cross-sectional microscope images of the full proposed encapsulation system fabricated on PDMS samples were taken. The surface height of the full encapsulation arrangement: skin insulation, DoS electrode, and encapsulation, was determined to be under 150 µm, with the medical adhesive skin base insulation layer being the thickest layer. The conductive ink has been previously tested with a profilometer for thickness. Two layers would be the most required for usage, making up the electrode pad and interconnection to a larger circuit, which would yield a thickness of under 3 µm. The low thickness of the full array allows it to be discrete, maximizing convenience on the part of the wearer.

With the insurance that the insulator functions, it must then be shown that the liquid ink can serve as a functional wire. Sheet resistance was used as the measurement of conductivity in this regard. An insufficiently conductive trace will not act as a suitable wire for this purpose, so it must be demonstrated the liquid ink wires did not have overly high sheet resistances. It must also be shown that a DoS electrode could be connected via the liquid ink to a separate pad insulated by the medical adhesive, which then was connected to the signal recording device. The cascade of uses of the conductive ink requires all the aforementioned properties to be exhibited successfully. Previously, biosignal recording with encapsulated electrodes has been reported, without losing functionality (cite DoS paper). Application of the full encapsulation setup demonstrated sheet resistance properties consistent with non-encapsulated ink. Applied to a subject's arm, a mock DoS electrode was drawn directly on the skin, and a layer of medical adhesive was applied to the skin, atop which a layer of conductive ink was drawn. To each end of the setup, a 22-gauge wire was affixed, and the entire setup was encapsulated in liquid bandage. An increase in sheet resistance when compared to the uncovered ink was observed, though the recorded encapsulated sheet resistance of 2.9 Ω/sq remains a low value.

iv. Motion Artifact Experiments

A major advantage of the DoS electrode system is the reduced interference caused by motion artifacts caused by the wearer's movement. In order to demonstrate the DoS electrodes' ability to record consistent and noise-free data despite deformation of the skin-electrode interface, a 10 mm diameter circular vibrating motor (HUELE micro-motor) was taped to the subject's arm equidistant between two electrodes on the right arm, 10 cm from each electrode, as shown in Figure 7c. Electromyogram (EMG) data was recorded and used as the biosignal to be monitored. At the wrist was the reference electrode, on the upper arm was the working electrode. The vibrating motor was connected to a function generator (DG4062, RIGOL), with parameters set to 1 kHz, 5 V_{PP}, 1.5 V DC offset, with a ramp waveform. The vibrations of the motor on the arm are meant to replicate and model the deformations in the skin caused by real-world scenario movement of the arm in order to model actual data recording during the subject's movement (43). The subject kept their arm stationary. In order to maintain a constant baseline EMG signal, the subject avoided any voluntary movements for the duration of the experiment. Each of the three electrode types being compared was tested with this motion modeling, with the DoS electrodes demonstrating superior performance over both other types of electrodes. The raw EMG signal is shown in Figure 7a, and time-frequency maps of the EMG recordings are shown in Figure 7b. During durations of vibration, indicated in pink, increases in noise can be observed in the gel and mesh electrodes. The other way that the ability for electrodes to continue to function under less than ideal circumstances was tested was by locally deforming the devices on a subject's skin. The electrodes were positioned in the ECG recording form, one electrode on each wrist. The skin around the working electrode on the right wrist was stretched, compressed, and released in regular intervals to replicate skin deformation. In each trial, the skin was stretched, then released and then compressed, then released. Each movement lasted an estimated 1 second, and the skin was stretched at a rate of approximately 2 mm/s, as determined by video analysis. The results are shown in Figure

8. There are visible artifacts produced on the gel and gold mesh electrode recordings by the deformations, while the DoS electrodes showed minimal visible artifacts caused by the deformations. Artifacts directly caused by the motion are indicated with red arrows. Local deformation was found to cause a significant change in electrode SNR. For the gel adhesive electrodes, during deformation SNR was calculated to be only 20 dB, less than half of its calculated SNR under ideal condition. For the gold mesh electrodes, local deformation reduced SNR to 10 dB, a quarter of its ideal value. In contrast, the DoS electrodes maintained an SNR value over 40 dB, demonstrating an ability to consistently record at a high-quality level while the skin-electrode interface is being deformed. In regards to both vibration and stretching, used to model motion and deformation respectively, the DoS electrodes proved to be the most stable electrodes and the best recording platform under unideal conditions.

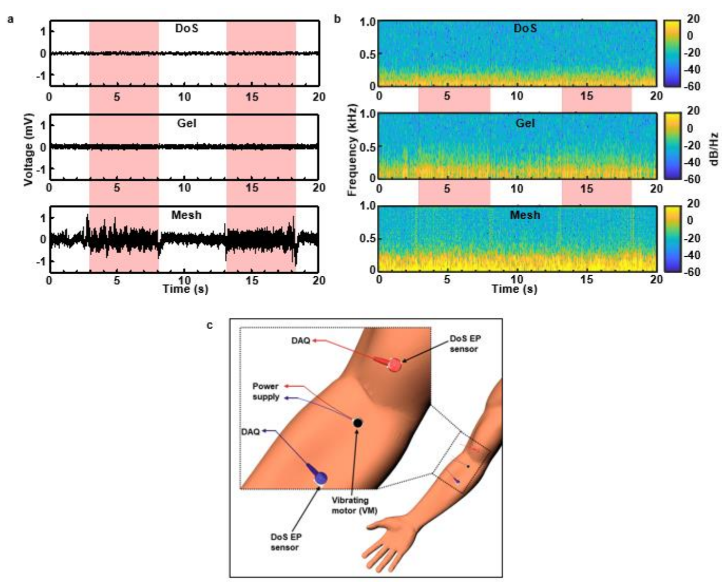

Figure 7 Electrode performance during skin deformation due to vibration. a) EMG recording during vibration. b) Time-frequency maps of the EMG data in (a). c) Experimental setup.

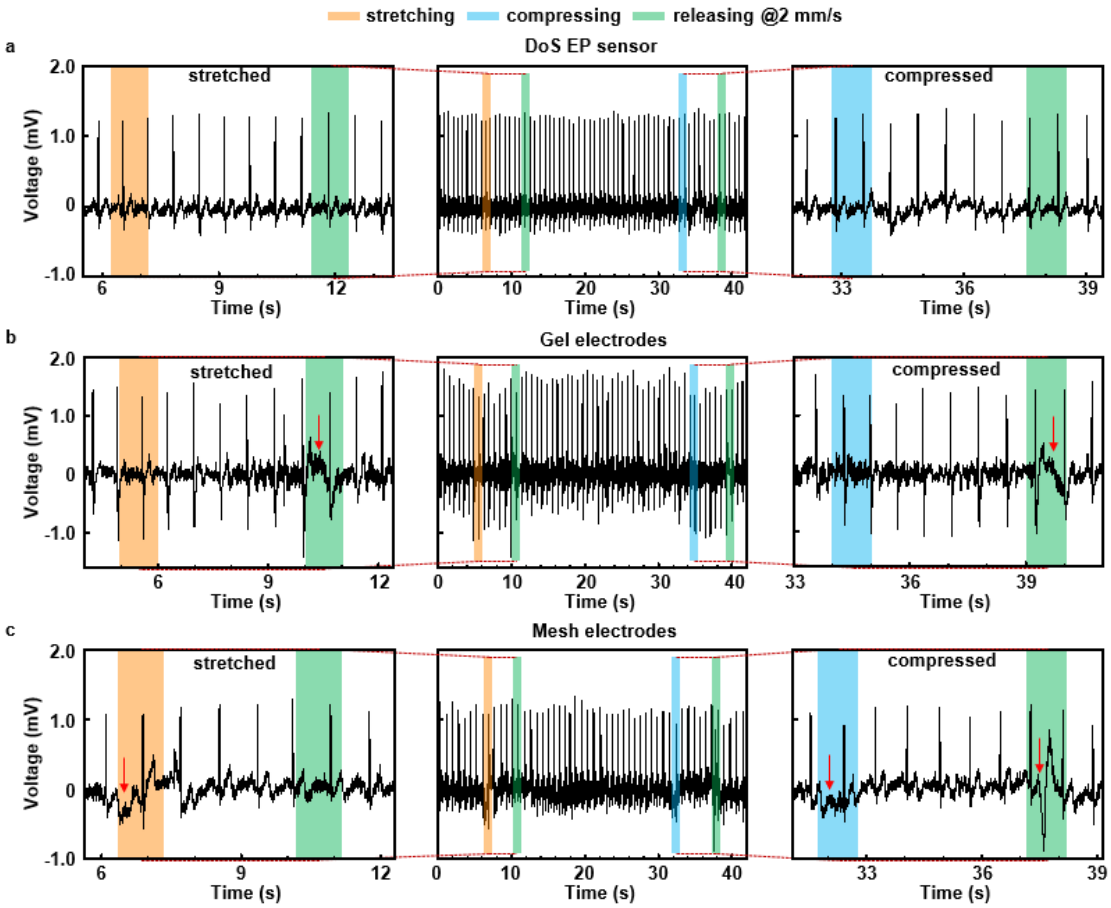

Figure 8 Electrode performance during local skin deformation. a) DoS electrode test. b) Gel adhesive electrode test. c) Gold mesh electrode test.

IV. EEG/EOG Implementation

i. Impedance Measurements

The conductivity of a skin-electrode is quantified by its impedance, and by minimizing impedance, optimal signal can be measured and recorded. Skin impedance tests were conducted by placing two electrodes of the same type to be tested on the right arm of a subject 15.0 mm apart and running an impedance analysis test through an impedance analyzer (Multi/Autolab M204, Metrohm) system process. The experiment setup on the skin is shown in Figure 9d. Frequencies beginning at 10^5 to 1 Hz were tested for skin impedance. It has been previously identified that hair poses a potential problem for wet electrodes, as it can reduce the contact area of the skin-electrode interface (21). Alternatively, because the DoS electrodes are applied directly to the skin, and the liquid ink can flow under and around hair, maintain a constant area of contact for the skin-electrode interface. In order to test these properties, a skin-electrode impedance test was conducted on a living subject. There were two different recording circumstances to be tested with regards to impedance: impedance over clear skin and impedance over hair-covered skin. By testing both situations, a difference in the quality of the skin-electrode interface between the two circumstances can be observed if the presence of hair causes issues. The results for both tests are shown in Figures 9a, 9b, and 9c. The standard EEG electrodes used for comparison purposes were silver cup electrodes, filled to the brim of the cup with EEG gel electrolyte. Dry silver spike electrodes were also used as a comparison, though their relatively much higher impedance led to the cessation of their usage as they made a considerably inferior interface, as shown in Figure 9c. One of the potential advantages intended for the DoS system is the ability to record around hair, as previously mentioned, so consistent impedance performance between clear and hairy skin would demonstrate that advantage, as can be observed in Figure 9a. Gel cup electrodes demonstrated a significant increase in skin impedance when hair was introduced as an interferent, shown in Figure 9b, demonstrating the difficulty it can have in maintaining a firm interface when circumstances are not perfectly ideal and demonstrating the advantage of electrodes drawn directly onto the skin when bodily features become obstacles. The gel and spike EEG electrodes used were housed in a spring apparatus, which keeps the electrode pinned in place when affixed properly to the wearer's head. The force exerted by the spring would have an effect on the recording properties of the electrode. Force applied to an electrode has the property of decreasing resistive properties and stabilizing resistive values during recordings, though it does increase contact impedance (44). The DoS electrodes had no force applied during recordings.

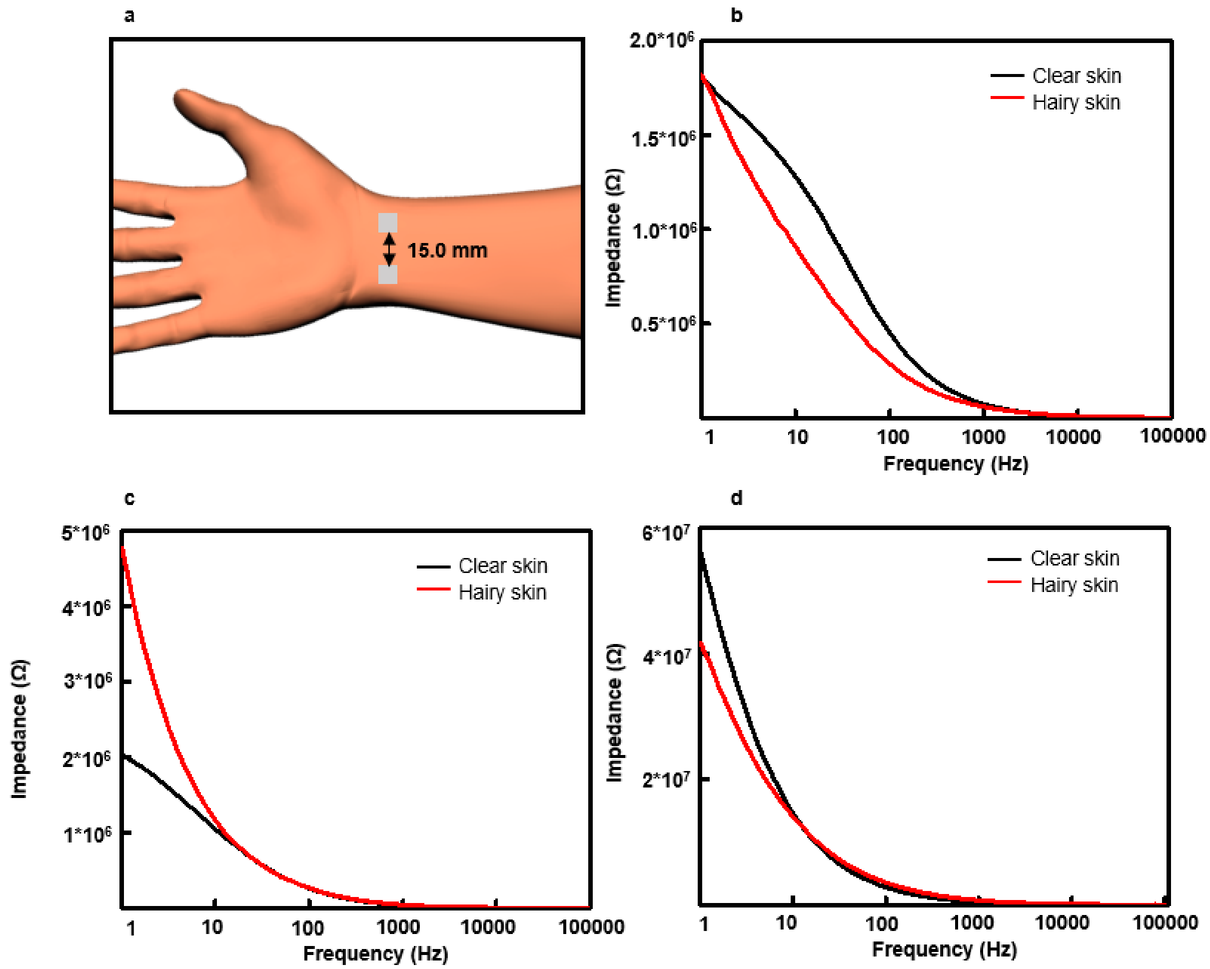

Figure 9 EEG electrode impedance measurements. a) DoS electrode test. b) gel cup electrode test. c) dry spike electrode test. d) experimental setup.

ii. EOG Recording

Using these favorable recording properties demonstrated during ECG and EMG recording, the next step was to modify the recording procedure to record from the head. The signal recorded from the head was EOG, the electrical signature caused by the polarity of the eye. The cornea is positively charged relative to the negatively charged retina, which allows for the detection of the corneoretinal potential(45). By placing electrodes above and below an eye, and to the far left and far right of both eyes, both horizontal and vertical eye movements can be recorded and observed. Blinks and eye movements can create noise in

EEG electrodes recording from the frontal region of the head. To set up a full EEG system, EOG measurement must also be recorded in order to filter out noise caused by these motions. The ideal full DoS system will, therefore, include EOG electrodes alongside EEG recording, so EOG recording must be successfully demonstrated. The gel adhesive electrodes previously used for comparison were also used to monitor EOG signals in order to validate the results acquired from the liquid electrode arrangement. The subject was instructed to move their eyes to the furthest extent they could on command. As shown in Figure 10, at time 1s, the subject was instructed to look up, and at time 3s, they were instructed to look down. Results recorded with the gel electrodes confirmed the measurements recorded with the liquid ink electrodes with recordings of like shape and amplitude.

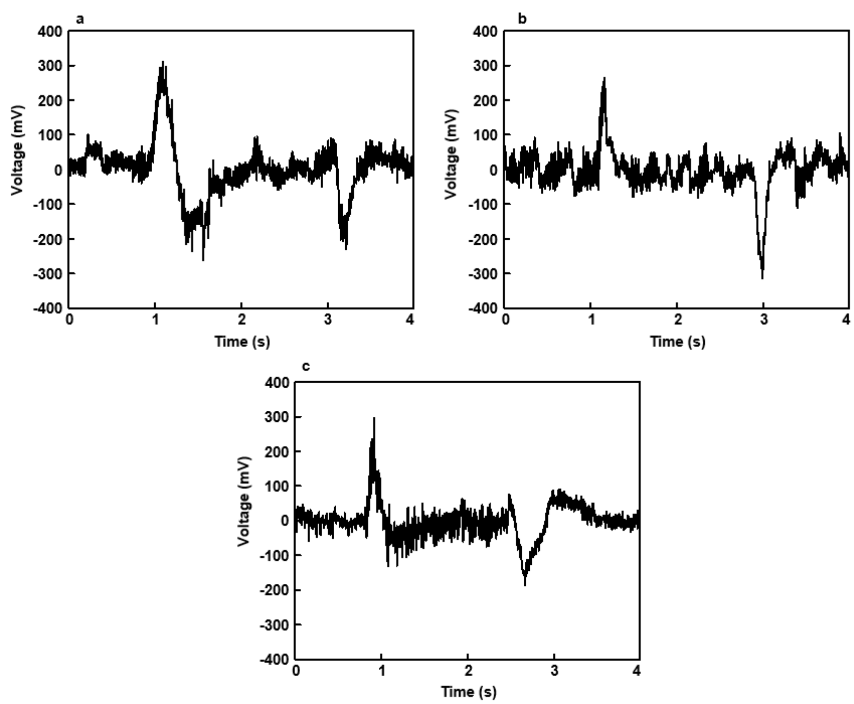

Figure 10 Demonstration of EOG recording. a) DoS EOG recording. b) gel adhesive EOG recording. c) encapsulated EOG recording.

In order to validate the encapsulation on the head, DoS EOG was repeated with encapsulated electrodes. Ink electrodes were drawn above and below the subject's right eye, and steel wires were affixed to the electrode with liquid bandage. For this test, vertical eye movement was again monitored. The insulated DoS electrodes measured comparable EOG signal, recorded using the same parameters as the unencapsulated electrode test, shown in Figure 10c.

iii. EEG Recording

With the demonstration of the feasibility of the premise, it was application that was the next step. To validate the DoS electrodes were capable of recording EEG at all, electrodes were placed at the O1 and O2 locations, as defined by the 10-20 electrode placement system, chosen because of the ease at which signals can be changed and observed. The relationship between relative alpha frequency band power and the opening and closing of the eyes has been extensively documented, and this relationship was monitored with the DoS and gel cup electrodes. Alpha band power desynchronizes and decreases when the eyes are open, and conversely synchronizes and increases when the eyes are closed (46, 47). This relationship can be clearly observed during trials and as such was chosen to validate the EEG recordings. As validated by the gel cup electrode recordings, such a result was acquired. Gel cup electrodes were held in place with a hard plastic headset (OpenBCI Ultracortex), to ensure firm affixation with the proper electrode locations. Utilizing the DoS electrodes, thin bare stainless steel wires (0.0762 mm diameter bare, 0.14 mm diameter insulated) were used to connect the electrodes to the recording system. All tests used the same recording system (Cyton 8-channel Biosensing board), recording at 250 Hz, connected wirelessly to OpenBCI. Recordings from both types of electrodes showed clear changes in frequency power at the marker, indicating the moment at which the subject closed their eyes, shown below. Data from O1 location electrodes is shown in Figures 11a and 11b. At the dashed line, the subject was instructed to close their eyes, leading to the observable increase in alpha band power. The similarity in recorded data demonstrates the ability for the DoS electrode to record EEG signal at a comparable level to gel cup electrodes. Figure 11c shows the power frequency graphs of the two recordings. As can be observed in the power frequency graph, the DoS electrodes demonstrated a more significant amount of alpha power, as observed in the large power spike at ~12 Hz relative to the compared gel electrodes.

SNR calculations for this data were made with two separate methodologies. The first calculations were made by taking the signal power of a latency of interest divided by the baseline signal activity (48). The measurements recorded during the eyes open phases were defined as noise, and the activity in the alpha region during the eyes closed phase was defined to be the signal of interest. The bounds used for the alpha region were defined as 8 – 14 Hz (49). This value measures the ability of the electrodes to record an induced response relative to the baseline EEG signal. Because this value has also been referred to as the evoked to spontaneous ratio, a second SNR value was calculated by using the mean of trials recorded as the signal and using the average of deviations from the mean signal across trials as the noise (50). In both methods of SNR calculation, the DoS electrodes yielded a higher value than the gel cup electrodes. For the first calculation, the DoS SNR was measured to be 12.20 dB, and the gel cup SNR was measured to be 9.46 dB, demonstrating a superior ability for the DoS electrodes to record a significant event relative to a baseline recording. For the second calculation, the DoS SNR was calculated to be

11.73 dB, and the gel cup was calculated to be 8.95 dB, showing that the DoS electrodes recorded less relative deviation from the mean signal than the gel cup electrodes. These results show the DoS electrodes' ability to record EEG signal at a high-quality, even when compared to conventional gel cup electrodes.

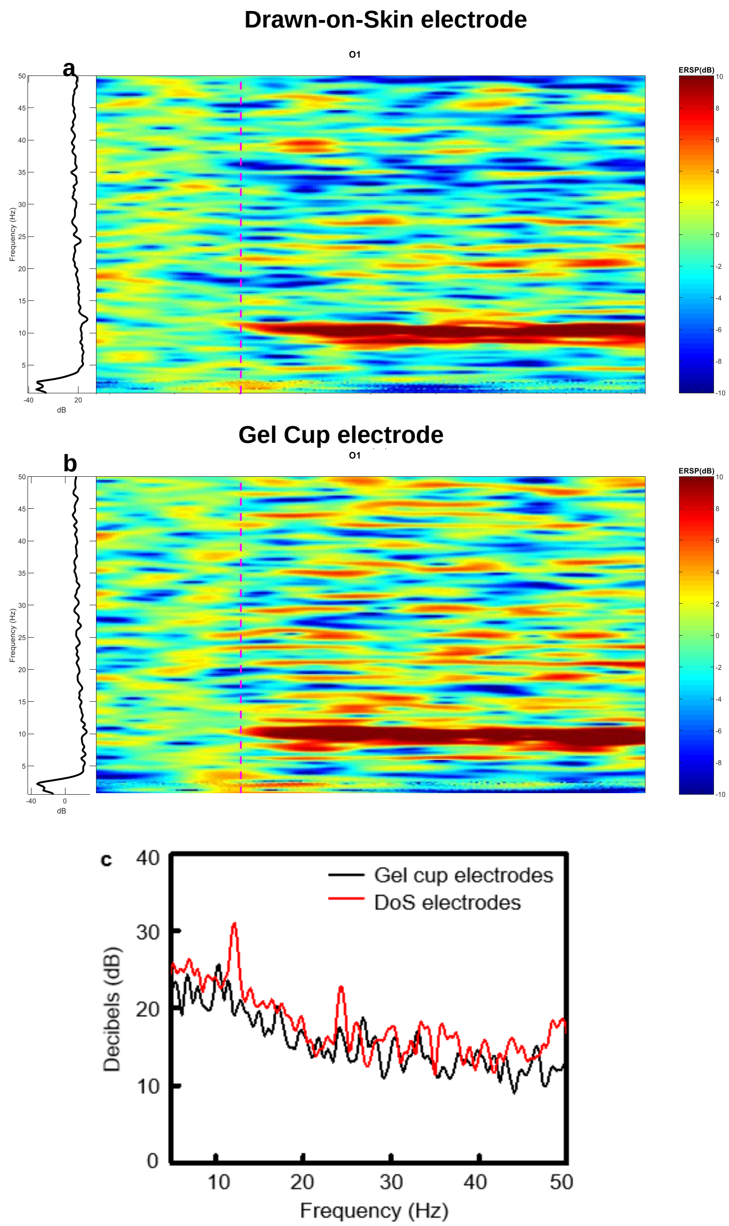

Figure 11 Occipital EEG recording comparison. a) DoS O1 recording. b) Gel cup O1 recording. c) Frequency power comparison.

iv. Simultaneous EEG-EOG Recording

With both EOG and EEG recording capability taken into account, recordings of frontal EEG signals were taken with the DoS recording system alongside horizontal and vertical EOG in order to filter out eye movements. Eye movement can cause recording artifacts as the shifting corneoretinal potential contaminates the EEG signal. EOG signal is recorded to use in signal processing to remove these artifacts. In order to simplify recording applications, the same electrode can be applied to both EEG and EOG situations. As can be observed, the unfiltered EEG recordings from the Fp2 location, per the 10-20 system, were contaminated with eye movement artifacts. Using the EOG signal recordings for noise removal proved successful at removing the eye movement artifacts, allowing the frontal EEG recording location to be used for recording with the DoS system. Frontal EEG was recorded during a visual test in which the subject was presented with a simple arithmetic problem with a provided solution. If the solution was correct, the subject would press the right arrow key, or they would press the left arrow key if the solution was incorrect. The simple mental calculations should cause increased activity in frontal regions of the brain, which can be observed in Figures 12a and 12b at the dotted line, indicating the moment of response to the stimulus, presented at the solid line. The dotted line also coincidentally marks the moment of a subject blink, and the effect of it and the second eye movement artifact, seen at the 3 second mark, on the recording can be observed in the raw data in Figure 12a. Use of the EOG recording as part of the filtering process allowed for the observation of increased activity in the delta range (4-8 Hz) during the task. This activity is most clearly visible around the 1 second mark, where immediately after the presentation of the stimulus, a burst of low frequency activity was recorded. An increase in delta range activity during mental tasks has been previously documented and recorded from the frontal region (51). The increased prominence of gamma range (30+ Hz) activity at the onset of the stimulus has also been documented during cognitive activity (52). Figure 12c shows the electrode arrangement for the experiment.

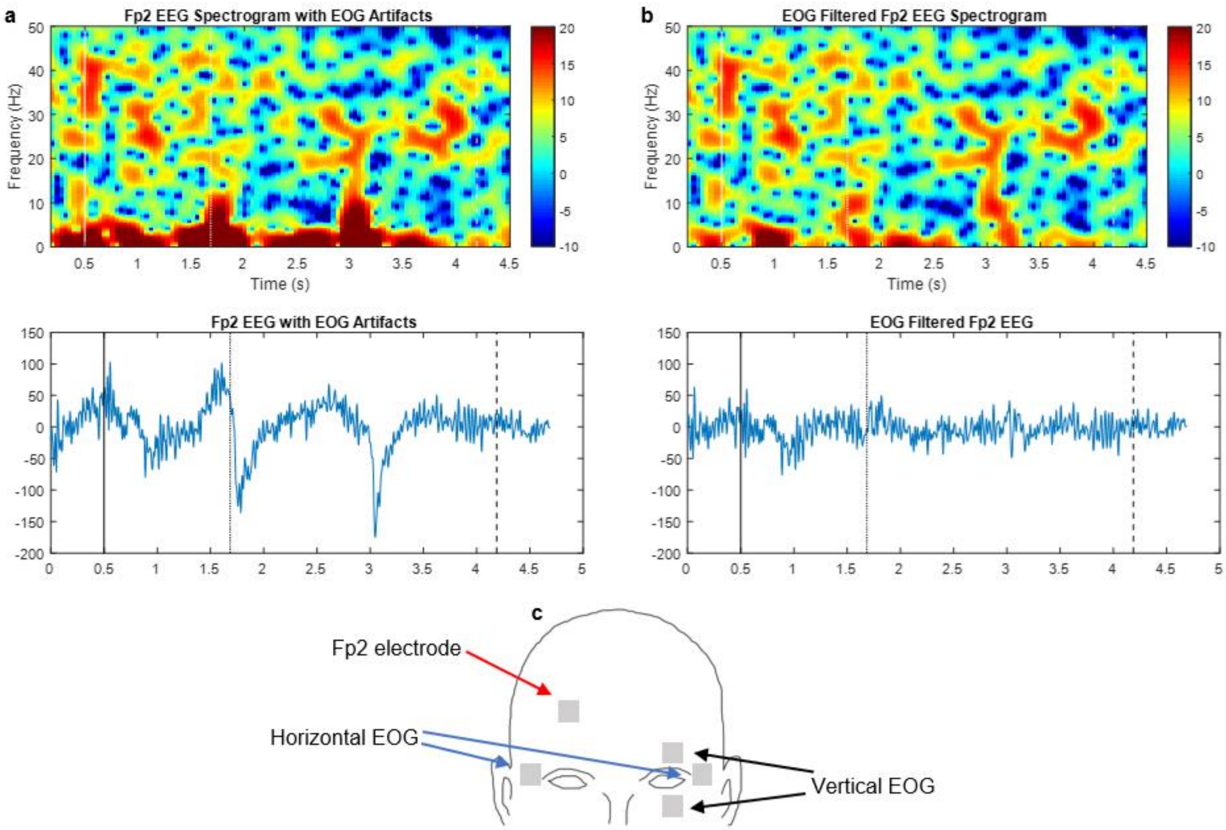

Figure 12 EOG filtered DoS EEG recording. a) Fp2 recording before EOG filtering. b) Fp2 recording after EOG filtering. c) experimental setup.

VI. Discussion

A novel method of recording neurological signals from humans has been developed in order to create a recording system without the effects of skin deformation induced motion artifacts. By using a conductive ink applied directly to the surface of the skin, biosignals can be recorded at high-quality. The DoS electrodes are resilient, as they can withstand time, sweat, and physical disruption, and with the addition of encapsulation and insulation, they can withstand water as well. This electrode type can be applied anywhere on the body, and electrodes can be customized by the experimenter to vary the size and to add circuit components on the skin surface. Applying the electrodes to the head allows for EEG recording, and the properties of the electrodes give them advantages in use over conventional electrodes.

As has been demonstrated, the novel DoS recording system has successfully demonstrated purposeful use with each of its component parts in accordance with its necessary electrical properties. The initial comparison with gel adhesive and gold mesh electrodes was necessary to form a recognizable baseline to establish the functionality of the liquid ink as a conductive electrode. Due to ease of application on the wrists and the ease at which the ECG waveform can be identified and observed, it was used for comparison purposes.

The explanation of the success of the chemical mixture as a conductor can be attributed to the electrical properties of the metal ions and conducting polymer. Silver is a very capable electrical conductor, which makes it particularly useful as an electrode material. Nanoparticles and flakes have been shown to self-reorganize upon stretching (53, 54), which makes them suitable for usage in applications where stretching and flexing would be necessary, such as in the case in applications on human skin. PEDOT:PSS has the highest conductivity reported among solution-processed polymers (54), however the PEDOT:PSS by itself would be too prone to fracturing for these purposes, as well as unable to dissolve completely the silver flakes added for conductivity. To remedy both of these issues, triton X-100, a surfactant, was added to the polymer. It has been shown that adding triton to PEDOT:PSS increases its stretchability without decreasing its electrical properties (55).

The ability for the silver-PEDOT:PSS ink electrodes to stretch and flex is necessary for human applications because a living surface is not stagnant and will move and bend and stretch. By the addition of triton in the polymer mixture, which enhances the stretchability of the ink, the resulting DoS devices are able to record without interference from motion artifact as the wearer moves. The vibrations caused by the motor mimicked natural movement in a measurable and controllable method. The explanation for the usage of gel adhesive electrodes in hospital settings can be clearly observed in the EMG recording. The gold mesh is designed in a serpentine pattern, which allows the metallic electrode to flex and stretch (56). Despite this tailored design, the mesh clearly demonstrated a weakness to motion artifact when compared to the other electrode types. The most probable explanation for this disparity between electrode types would be the lack of firm adhesion on the part of the gold mesh electrode. While the mesh electrode does adhere to the skin, there is no dedicated adhesive ensuring a consistent bond between electrode and skin, allowing for two potential motion artifacts: longitudinal, where the electrode loses contact with the skin, and transverse, where the electrode slides on the skin (3). In the case of the vibrating motor experiment, it is likely that the lack of consistent adhesion to the skin allowed the electrode to become detached from the skin in some areas as the skin vibrated. In the case of deformation, the lack of adhesion led to the electrode sliding. The gel adhesive electrodes use a firm adhesive which acts similar to a sticker, ensuring a stable electrode connection, to prevent longitudinal motion artifacts. However, the gel underneath the adhesive pad was still likely able to slide with skin deformation (57). As demonstrated by the local deformation experiment, the connection created by the foam adhesive was not able to prevent all motion artifacts. As demonstrated by the direct adhesion test, the gel by itself does not create a firm adhesive bond with the skin, allowing the gel by itself to potentially shift. The DoS electrode itself was adhered firmly and directly onto the skin. The same conductive adhesive tape that was used to connect the DoS electrode to the data acquisition system was used with the gold mesh electrode. Because of the success of the DoS electrode, the conductive adhesive was therefore not the problem. Given this, the lack of consistent adhesion between electrode and skin surface is likely the cause for the artifact issues with the gold mesh electrode. With the necessity for the proposed DoS electrode system to minimize the impact of motion artifacts, the isolation of major motion artifacts to the electrode type without firm adhesion would reinforce the findings that the proposed system would successfully minimize noise resulting from movement.

The minimization of motion artifacts in DoS electrode recordings allows for more stable data acquisition, which would be of benefit with EEG recordings. Changes in the interface geometry between the skin and the electrode caused by user motion likewise cause issues in recordings with EEG signals (8). Just as fluctuations in the skin, from vibrations and deformation, caused issues with the gel adhesive and gold mesh electrodes, movement can cause similar issues with conventional EEG electrodes, both dry and wet. The inclusion of a conductive electrolyte causes EEG electrodes to respond differently at the skin interface, leading to different electrical properties between dry and wet electrodes (44). With the DoS electrodes' firm adhesion directly to the skin, they remain unchanged with motion, giving an advantage over the traditional electrodes.

Another problem faced by conventional EEG electrodes is that gel used by wet electrodes is the potential for drying out during long term usage. Since the DoS electrodes are not wet, there is no liquid to risk drying. As shown by the time durability test, the DoS electrodes demonstrate consistent recording over the course over several hours, where the gel electrodes demonstrate a significant decrease in R-peak amplitude. With the low voltage of scalp EEG, a decrease in recording amplitude can lead to missing brain activity in recordings. Because the DoS electrodes allow for consistent recording over the period of hours, this allows for the potential for long-term recordings, and it would allow for the ability to take recordings outside of the laboratory environment.

The potential for long-term applications means that the DoS electrodes must be capable of withstanding potential environmental hazards. Adhesion tests performed with the different electrodes showed the ability of the DoS electrodes to survive the tape pull, where the gold mesh and exposed gel both did not bond with the skin firmly enough to survive. Using physical force to damage electrodes demonstrated the DoS electrodes' resiliency, with negligible damage observed after the electrode was vigorously rubbed. There is a weakness of the DoS electrodes, though. To remove the DoS electrodes, the electrode area is wiped with water, which dissolves the ink and easily removes it. The specific goal for the insulation of the DoS electrodes was to preserve the electrodes from damage due to water. Therefore, any sort of long-term application would need to be guarded against water, else the electrodes would be dissolved during use if the wearer showers or gets caught in the rain or otherwise gets covered in water.

In order to protect the DoS electrodes, the liquid bandage insulation was utilized. To test the efficacy of liquid bandage as insulation, two separate aspects of the integration were tested: the ability for the liquid bandage to protect the electrode beneath and the conductivity of electrodes beneath. If the electrode still breaks down when water is applied even when insulated, then the liquid bandage will not suffice for the purpose of this proposed system. Similarly, if the conductivity of the electrodes becomes compromised as a result of the insulation, then the insulation will be equally ill-suited for usage. Experimental results validated both aspects of the insulation as sufficient for the purpose of this proposal. The durability of the electrode was substantially increased through the liquid bandage application and served dual purposes in practice: protection and adhesion. It was for the purpose of protection that the insulation was proposed at all, and in this regard, liquid bandage demonstrated excellent properties. The adhesive property of the liquid bandage was not an initial concern to its usage, but the firm connection between electrode and wire connecting the electrode to the recording device formed by the adhesion ensured recording survived motion that would be natural in usage on a human subject.

With the advantages of DoS EEG electrodes demonstrated to each work, there are certain specific ways that the proposed electrodes can be useful. The use of silver-PEDOT:PSS ink to make complex circuits featuring functional devices has been previously established, so with it already shown that the ink can be used to make DoS EEG electrodes, then these electrodes can be featured as a part of a larger complex. Each of the trials that have been mentioned here utilizes wired connections from the electrode pad to the data acquisition system. Proposed then is the possibility of in the future, integrating complex device components to the skin surface to remove the wires from the head to the system. It has been previously mentioned that wire movement can cause motion artifacts (42). Therefore, to create the most superior system, free wires must be eliminated entirely. By using the established ability of the conductive ink to act as a wire, the electrode pads recording biosignals can then be connected to contact pads on a flexible patch on the skin, which can then process data directly on the skin. Such a patch can include such devices as amplifiers, alternate to digital signal converters, and wireless transmission antennas. Antennas have been integrated into wearable devices to transmit data from electrodes wirelessly (58, 59). Integrating wearable sensors with wearable antenna creates a fully wearable sensor to record and transmit data. By integrating wireless antenna with near-field communication (NFC) or radiofrequency identification (RFID) technology, battery-free wireless communication can be achieved from the surface of the skin (60, 61). It has been demonstrated that antenna can maintain performance during stretching, allowing for communication from the subject’s skin, even as they move (62). This would allow the transmission of EEG data from the electrodes directly to a smartphone. The most likely location for this hub would be the base of the skull on the neck, a location that would not be overly visible on the subject, while allowing for the traces from the electrode pads to converge together. This would allow the system to record and transmit data entirely from the surface of the skin, without being overly incumbent on the patient while the system is being worn.

The development of such a system would be the ideal implementation of all the advantages expressed by the DoS electrode system. It would allow for high-quality recording, free from artifacts caused by skin deformation, but also free from artifacts caused by the movement of parts of the circuit. The whole circuit would be low-visibility and barely perceptible, so it would be more comfortable for the wearer to use for extended periods of time. For those periods of use, the electrodes and their traces would be capable of withstanding the trials of wear, and they would be encapsulated from the environment and insulated from signal contamination. Those traces would lead to a soft electronic device, affixed to the skin, complete with wireless transmission capabilities. Once the full system is applied to the subject, they would be able to record high-quality that they could record directly from their phone for observation or for application.

VII. Conclusion

With the advantages hereto described, using an ink mixture of silver flakes and PEDOT:PSS should be further explored for expanded future usage in recording and practically using EEG signals measured from the scalp's surface. The ability of the DoS electrode system to record consistent and high-quality biosignal data in spite of skin-electrode interface deformation and the ability to integrate insulation from the skin's surface and from the dangers of the environment to protect from multiple types of unwanted noise interference combine to form a system that is capable of creating complex and customizable devices directly onto the surface of the subject's skin that can withstand the potential rigors of everyday wear. By creating minimally obtrusive and minimally visible device systems on the subject directly, then systems can be set up and experiments designed for long term usage in a manner more convenient for the subject.

References

1. Liu Y, Pharr M, Salvatore GA. "Lab-on-Skin: A Review of Flexible and Stretchable Electronics for Wearable Health Monitoring." *ACS Nano*. 2017;11(10):9614-35. doi: 10.1021/acsnano.7b04898.

2. Wang Y, Qiu Y, Ameri SK, Jang H, Dai Z, Huang Y, Lu N. "Low-cost, µm-thick, tape-free electronic tattoo sensors with minimized motion and sweat artifacts." *npj Flexible Electronics*. 2018;2(1):6. doi: 10.1038/s41528-017-0019-4.

3. Li X, Hui H, Sun Y, editors. Investigation of motion artifacts for biopotential measurement in wearable devices. 2016 IEEE 13th International Conference on Wearable and Implantable Body Sensor Networks (BSN); 2016 14-17 June 2016.

4. McAdams ET, Jossinet J, Lackermeier A, Risacher F. "Factors affecting electrode-gel-skin interface impedance in electrical impedance tomography." *Medical and Biological Engineering and Computing*. 1996;34(6):397-408. doi: 10.1007/BF02523842.

5. Webster JG. "Reducing Motion Artifacts and Interference in Biopotential Recording." *IEEE Transactions on Biomedical Engineering*. 1984;BME-31(12):823-6. doi: 10.1109/TBME.1984.325244.

6. Nunez PS, R. Electrical Fields of the Brain Second Edition ed: Oxford University Press; 2006.

7. Malmivuo J, Plonsey R. Bioelectromagnetism. 13. Electroencephalography. 1995. p. 247-64.

8. Mihajlović V, Patki S, Grundlehner B, editors. The impact of head movements on EEG and contact impedance: An adaptive filtering solution for motion artifact reduction.

2014 36th Annual International Conference of the IEEE Engineering in Medicine and Biology Society; 2014 26-30 Aug. 2014.

9. Gwin JT, Gramann K, Makeig S, Ferris DP. "Removal of movement artifact from high-density EEG recorded during walking and running." *J Neurophysiol*. 2010;103(6):3526-34. Epub 04/21. doi: 10.1152/jn.00105.2010. PubMed PMID: 20410364.

10. Verleger R, Śmigasiewicz K. "Do Rare Stimuli Evoke Large P3s by Being Unexpected? A Comparison of Oddball Effects Between Standard-Oddball and Prediction-Oddball Tasks." *Adv Cogn Psychol*. 2016;12(2):88-104. doi: 10.5709/acp-0189-9. PubMed PMID: 27512527.

11. Käthner I, Kübler A, Halder S. "Rapid P300 brain-computer interface communication with a head-mounted display." *Frontiers in neuroscience*. 2015;9:207-. doi: 10.3389/fnins.2015.00207. PubMed PMID: 26097447.

12. Hinrichs H, Scholz M, Baum AK, Kam JWY, Knight RT, Heinze H-J. "Comparison between a wireless dry electrode EEG system with a conventional wired wet electrode EEG system for clinical applications." *Scientific Reports*. 2020;10(1):5218. doi: 10.1038/s41598-020-62154-0.

13. Kleffner-Canucci K, Luu P, Naleway J, Tucker DM. "A novel hydrogel electrolyte extender for rapid application of EEG sensors and extended recordings." *Journal of Neuroscience Methods*. 2012;206(1):83-7. doi: https://doi.org/10.1016/j.jneumeth.2011.11.021.

14. Alba NA, Sclabassi RJ, Sun M, Cui XT. "Novel Hydrogel-Based Preparation-Free EEG Electrode." *IEEE Transactions on Neural Systems and Rehabilitation Engineering*. 2010;18(4):415-23. doi: 10.1109/TNSRE.2010.2048579.

15. Kalia YN, Guy RH. "The Electrical Characteristics of Human Skin in Vivo." *Pharmaceutical Research*. 1995;12(11):1605-13. doi: 10.1023/A:1016228730522.

16. Natus. Grass® Reusable Stamped-Cup EEG Electrodes [June 28, 2020]. Available from: https://natus.com/.

17. Vasios CE, Angelone LM, Purdon PL, Ahveninen J, Belliveau JW, Bonmassar G. "EEG/(f)MRI measurements at 7 Tesla using a new EEG cap (“InkCap”)." *NeuroImage*. 2006;33(4):1082-92. doi: https://doi.org/10.1016/j.neuroimage.2006.07.038.

18. Salvo P, Raedt R, Carrette E, Schaubroeck D, Vanfleteren J, Cardon L. "A 3D printed dry electrode for ECG/EEG recording." *Sensors and Actuators A: Physical*. 2012;174:96-102. doi: https://doi.org/10.1016/j.sna.2011.12.017.

19. Yamamoto T, Yamamoto Y. "Electrical properties of the epidermal stratum corneum." *Medical and biological engineering*. 1976;14(2):151-8. doi: 10.1007/BF02478741.

20. Searle A, Kirkup L. "A direct comparison of wet, dry and insulating bioelectric recording electrodes." *Physiological Measurement*. 2000;21(2):271-83. doi: 10.1088/0967-3334/21/2/307.

21. Lin C, Liao L, Liu Y, Wang I, Lin B, Chang J. "Novel Dry Polymer Foam Electrodes for Long-Term EEG Measurement." *IEEE Transactions on Biomedical Engineering*. 2011;58(5):1200-7. doi: 10.1109/TBME.2010.2102353.

22. Biosemi. Headcaps [June 6, 2020]. Available from: https://www.biosemi.com/.

23. Debener S, Emkes R, De Vos M, Bleichner M. "Unobtrusive ambulatory EEG using a smartphone and flexible printed electrodes around the ear." *Scientific reports*. 2015;5:16743-. doi: 10.1038/srep16743. PubMed PMID: 26572314.

24. Casson A, Abdulaal M, Dulabh M, Kohli S, Krachunov S, Trimble E. Electroencephalogram. 2018. p. 45-81.

25. Lopez-Gordo MA, Sanchez-Morillo D, Pelayo Valle F. "Dry EEG electrodes." *Sensors (Basel, Switzerland)*. 2014;14(7):12847-70. doi: 10.3390/s140712847. PubMed PMID: 25046013.

26. Jin-Chern C, Li-Wei K, Chin-Teng L, Chao-Ting H, Tzyy-Ping J, Sheng-Fu L, Jong-Liang J, editors. Using novel MEMS EEG sensors in detecting drowsiness application. 2006 IEEE Biomedical Circuits and Systems Conference; 2006 29 Nov.-1 Dec. 2006.

27. Ng WC, Seet HL, Lee KS, Ning N, Tai WX, Sutedja M, Fuh JYH, Li XP. "Micro-spike EEG electrode and the vacuum-casting technology for mass production." *Journal of Materials Processing Technology*. 2009;209(9):4434-8. doi: https://doi.org/10.1016/j.jmatprotec.2008.10.051.

28. Krachunov S, Casson AJ. "3D Printed Dry EEG Electrodes." *Sensors*. 2016;16(10). doi: 10.3390/s16101635.

29. Fiedler P, Mühle R, Griebel S, Pedrosa P, Fonseca C, Vaz F, Zanow F, Haueisen J. "Contact Pressure and Flexibility of Multipin Dry EEG Electrodes." *IEEE Transactions on Neural Systems and Rehabilitation Engineering*. 2018;26(4):750-7. doi: 10.1109/TNSRE.2018.2811752.

30. Sekitani T, Yoshimoto S, Araki T, Uemura T. "12-2: Invited Paper: A Sheet-type Wireless electroencephalogram (EEG) Sensor System using Flexible and Stretchable Electronics." *SID Symposium Digest of Technical Papers*. 2017;48(1):143-6. doi: 10.1002/sdtp.11594.

31. Kim DH, Ghaffari R, Lu N, Rogers JA. "Flexible and stretchable electronics for biointegrated devices." *Annual review of biomedical engineering*. 2012;14:113-28. Epub 2012/04/25. doi: 10.1146/annurev-bioeng-071811-150018. PubMed PMID: 22524391.

32. Fonseca C, Cunha JPS, Martins RE, Ferreira VM, Sa JPMd, Barbosa MA, Silva AMd. "A Novel Dry Active Electrode for EEG Recording." *IEEE Transactions on Biomedical Engineering*. 2007;54(1):162-5. doi: 10.1109/TBME.2006.884649.

33. Wang L, Liu J, Yang B, Yang C. "PDMS-Based Low Cost Flexible Dry Electrode for Long-Term EEG Measurement." *IEEE Sensors Journal*. 2012;12(9):2898-904. doi: 10.1109/JSEN.2012.2204339.

34. Kappenman ES, Luck SJ. "The effects of electrode impedance on data quality and statistical significance in ERP recordings." *Psychophysiology*. 2010;47(5):888-904. Epub 03/29. doi: 10.1111/j.1469-8986.2010.01009.x. PubMed PMID: 20374541.

35. Mathewson KE, Harrison TJL, Kizuk SAD. "High and dry? Comparing active dry EEG electrodes to active and passive wet electrodes." *Psychophysiology*. 2017;54(1):74-82. doi: 10.1111/psyp.12536.

36. Xiong Z, Qiang L, Kilsgaard S, Moradi F, Kappel SL, Kidmose P, editors. A wearable ear-EEG recording system based on dry-contact active electrodes. 2016 IEEE Symposium on VLSI Circuits (VLSI-Circuits); 2016 15-17 June 2016.

37. Xu J, Yazicioglu RF, Grundlehner B, Harpe P, Makinwa KAA, Hoof CV. "A 160 μW 8-Channel Active Electrode System for EEG Monitoring." *IEEE Transactions on Biomedical Circuits and Systems*. 2011;5(6):555-67. doi: 10.1109/TBCAS.2011.2170985.

38. Houghton T, Vanjaria J, Yu H, editors. Conductive and Stretchable Silver-Polymer Blend for Electronic Applications. 2016 IEEE 66th Electronic Components and Technology Conference (ECTC); 2016 31 May-3 June 2016.

39. Ershad F, Thukral A, Yue J, Comeaux P, Lu Y, Shim H, Sim K, Kim N-I, Rao Z, Guevara R, Contreras L, Pan F, Zhang Y, Guan Y, Yang P, Wang X, Wang P, Wu X, Yu C (In Press). "Ultra-Conformal Drawn-on-Skin Electronics for Multifunctional Motion Artifact-Free Sensing and Point-of-Care Treatment." *Nature Communications*. 2020. doi: doi.org/10.1038/s41467-020-17619-1.

40. Jurak P, Halamek J, Leinveber P, Vondra V, Soukup L, Vesely P, Sumbera J, Zeman K, Martinakova L, Jurakova T, Novak M. Ultra-high-frequency ECG measurement2013. 783-6 p.

41. Tronstad C, Johnsen GK, Grimnes S, Martinsen ØG. "A study on electrode gels for skin conductance measurements." *Physiological Measurement*. 2010;31(10):1395-410. doi: 10.1088/0967-3334/31/10/008.

42. Symeonidou E-R, Nordin AD, Hairston WD, Ferris DP. "Effects of Cable Sway, Electrode Surface Area, and Electrode Mass on Electroencephalography Signal Quality during Motion." *Sensors (Basel, Switzerland)*. 2018;18(4):1073. doi: 10.3390/s18041073. PubMed PMID: 29614020.

43. Nawrocki RA, Jin H, Lee S, Yokota T, Sekino M, Someya T. "Self-Adhesive and Ultra-Conformable, Sub-300 nm Dry Thin-Film Electrodes for Surface Monitoring of Biopotentials." *Advanced Functional Materials*. 2018;28(36):1803279. doi: 10.1002/adfm.201803279.

44. Mihajlović V, Grundlehner B, editors. The effect of force and electrode material on electrode-to-skin impedance. 2012 IEEE Biomedical Circuits and Systems Conference (BioCAS); 2012 28-30 Nov. 2012.

45. Jiang RA-m, Somaya; Bouridane, Ahmed; Crookes, Danny; Beghdadi, Azeddine Biometric Security and Privacy: Opportunities & Challenges in The Big Data Era. Celebi ME, editor. Baton Rouge, LA: Springer; 2017.

46. Barry RJ, De Blasio FM. "EEG differences between eyes-closed and eyes-open resting remain in healthy ageing." *Biological Psychology*. 2017;129:293-304. doi: https://doi.org/10.1016/j.biopsycho.2017.09.010.

47. Barry RJ, Clarke AR, Johnstone SJ, Magee CA, Rushby JA. "EEG differences between eyes-closed and eyes-open resting conditions." *Clinical Neurophysiology*. 2007;118(12):2765-73. doi: https://doi.org/10.1016/j.clinph.2007.07.028.

48. Viola FC, editor. Towards artifact-free auditory evoked potentials in cochlear implant users2011.

49. Abo-Zahhad M, Ahmed S, Seha SN. "A New EEG Acquisition Protocol for Biometric Identification Using Eye Blinking Signals." *International Journal of Intelligent Systems and Applications (IJISA)*. 2015;07:48-54. doi: 10.5815/ijisa.2015.06.05.

50. Schultz S. "Signal-to-noise ratio in neuroscience." *Scholarpedia*. 2007;2:2046. doi: 10.4249/scholarpedia.2046.

51. Fernández T, Harmony T, Rodríguez M, Bernal J, Silva J, Reyes A, Marosi E. "EEG activation patterns during the performance of tasks involving different components of mental calculation." *Electroencephalography and Clinical Neurophysiology*. 1995;94(3):175-82. doi: https://doi.org/10.1016/0013-4694(94)00262-J.

52. Tallon-Baudry C, Bertrand O, Peronnet F, Pernier J. "Induced γ-Band Activity during the Delay of a Visual Short-Term Memory Task in Humans." *The Journal of Neuroscience*. 1998;18(11):4244-54. doi: 10.1523/jneurosci.18-11-04244.1998.

53. Kim Y, Zhu J, Yeom B, Di Prima M, Su X, Kim J-G, Yoo SJ, Uher C, Kotov NA. "Stretchable nanoparticle conductors with self-organized conductive pathways." *Nature*. 2013;500(7460):59-63. doi: 10.1038/nature12401.

54. Wang Y, Zhu C, Pfattner R, Yan H, Jin L, Chen S, Molina-Lopez F, Lissel F, Liu J, Rabiah NI, Chen Z, Chung JW, Linder C, Toney MF, Murmann B, Bao Z. "A highly stretchable, transparent, and conductive polymer." *Science Advances*. 2017;3(3):e1602076. doi: 10.1126/sciadv.1602076.

55. Oh JY, Kim S, Baik H-K, Jeong U. "Conducting Polymer Dough for Deformable Electronics." *Advanced Materials*. 2016;28(22):4455-61. doi: 10.1002/adma.201502947.

56. Kim D-H, Xiao J, Song J, Huang Y, Rogers JA. "Stretchable, Curvilinear Electronics Based on Inorganic Materials." *Advanced Materials*. 2010;22(19):2108-24. doi: 10.1002/adma.200902927.

57. Ödman S, Åke Öberg P. "Movement-induced potentials in surface electrodes." *Medical and Biological Engineering and Computing*. 1982;20(2):159. doi: 10.1007/BF02441351.

58. Monti G, Corchia L, Tarricone L. "UHF Wearable Rectenna on Textile Materials." *IEEE Transactions on Antennas and Propagation*. 2013;61(7):3869-73. doi: 10.1109/TAP.2013.2254693.

59. Paracha KN, Rahim SKA, Soh PJ, Khalily M. "Wearable Antennas: A Review of Materials, Structures, and Innovative Features for Autonomous Communication and Sensing." *IEEE Access*. 2019;7:56694-712. doi: 10.1109/ACCESS.2019.2909146.

60. Niu S, Matsuhisa N, Beker L, Li J, Wang S, Wang J, Jiang Y, Yan X, Yun Y, Burnett W, Poon ASY, Tok JBH, Chen X, Bao Z. "A wireless body area sensor network based on stretchable passive tags." *Nature Electronics*. 2019;2(8):361-8. doi: 10.1038/s41928-019-0286-2.

61. Kim J, Salvatore GA, Araki H, Chiarelli AM, Xie Z, Banks A, Sheng X, Liu Y, Lee JW, Jang K-I, Heo SY, Cho K, Luo H, Zimmerman B, Kim J, Yan L, Feng X, Xu S, Fabiani M, Gratton G, Huang Y, Paik U, Rogers JA. "Battery-free, stretchable optoelectronic systems for wireless optical characterization of the skin." *Science Advances*. 2016;2(8):e1600418. doi: 10.1126/sciadv.1600418.

62. Jeong SH, Hagman A, Hjort K, Jobs M, Sundqvist J, Wu Z. "Liquid alloy printing of microfluidic stretchable electronics." *Lab on a Chip*. 2012;12(22):4657-64. doi: 10.1039/C2LC40628D.