Rapid Fabrication of Low Impedance, 3D Dry Electrodes for Physiological Sensing

Ryan Kaveh*, Natalie Tetreault*, Karthik Gopalan, Julian Maravilla, Michael Lustig, Rikky Muller, Ana C. Arias

Abstract:

The performance of medical devices that record electrophysiological activity to diagnose epilepsy and cardiac arrhythmia depend on consistently low impedance interfaces between conductors and the skin. Clinically standard devices, wet electrodes, use hydrogels and skin abrasion to improve the interface and recorded signal quality. These electrodes and their required preparation are challenging to self-administer which reduces the frequency of disease monitoring and impedes in-home care. Wearable dry electrodes are more practical; however, they show higher impedance relative to wet electrodes and are costly to customize. In this work, a fabrication method for producing anatomically fit, dry electrodes that can be optimized for individuals and/or specific recording locations on the body is presented. Electroless gold plating is used in combination with 3D printing to enable anatomically fit, high-performance, 3D dry electrodes that do not require any skin preparation and are comfortable to wear. The performance of the example 3D dry electrodes is compared to clinically standard devices. The resulting electrodes exhibited an average electrode-skin impedance of 71.2 kΩ at 50Hz and DC offset of -20 mV, which is within the range achieved by clinical wet electrodes.
**Introduction:**

In clinical settings, non-invasive electrophysiological (ExG) recording devices use electrocardiography (ECG), electromyography (EMG), and electroencephalography (EEG) to enable expansive brain-machine interfaces [1], arrhythmia sensors [2], and advanced prostheses [3]. These devices require extensive skin preparation, use wet electrodes, record with high signal-to-noise ratios (SNR), and, with the help of a medical professional, offer restorative and life-changing treatments. Wet electrode hydrogels ensure consistent and low-impedance electrode-skin contact until the gel dries out. This drying leads to increased electrode-skin impedance (ESI), which reduces recorded single amplitude, increases susceptibility to power-line interference, ultimately reducing SNR [4]. The skin preparation required for wet electrodes (especially in EEG setups for sleep/epilepsy studies) also often leads to skin-irritation, hair loss [5], and discomfort from residual gel left in hair.

Some devices have attempted to expand these clinical techniques to everyday users by incorporating general purpose dry electrodes. These systems increase usability and patient comfort but generally result in higher ESI (>1 MΩs at <250 Hz) relative to wet-electrodes (10-100s kΩs at <250 Hz) [6]. Microneedles, pin electrodes, conductive composites, and conformal electrodes have been used to lower ESI and improve the mechanical stability of dry electrodes [6] [7] [8], but introduce new use or fabrication trade-offs. Microneedles, which pierce the top layer of skin, achieve lower ESIs and enable higher SNR recordings. However, prolonged use of these electrodes may result in lesion formation and risk of infection. Non-contact and conductive composites such as silicone carbon black and silver-glass silicone show the opposite trade-off, achieving greater comfort but suffers from higher ESIs relative to other dry electrodes [9] [6] [7]. Other electrode arrays have used flexible planar structures, machined metals, or metal printed devices to increase
electrode compliance, comfort, and options in sensing location [10] [11] [12] [13]. Printed flexible electrode arrays enable high density electrode placement, high-volume fabrication, and comfortable electrode compliance along a single axis, such as around arms [14]. Flexible printed arrays also require hydrogels to record through hairy surfaces. 3D printing and micromachining of insulating and conductive materials have been used in combination with vacuum deposition processes such as sputtering or evaporation [15] [16] to achieve low impedance scalp electrodes. Ultimately, existing techniques emphasize that no existing single dry-electrode design can target all biopotential signals. Furthermore, existing fabrication techniques for comfortable, anatomically customized electrodes are low-cost nor scalable or are limited to specific electrode shapes.

Recent commercial, dry electrode ExG [17] [18] [19] systems employ user-generic electrodes and require hairless recording locations on the wrist, forehead or limbs to improve SNR reliability for very specific target applications across large populations. An exploratory medical device that is accepted by a patient for continuous electrophysiological recordings must be anatomically customized for a sensing location that does not interfere with daily activities. It is also preferable to record through hair without the need for shaving, contacting gels or skin preparation without compromising SNR. In addition to low (ESI) in frequency bands of interest (<500 Hz for ExG), near zero electrode DC offset (EDO) is essential for signal quality. An illustration of such ideal electrodes that do not require skin preparation, are reusable, and are fabricated with scalable methods is given in Figure 1. The fabrication process presented in this work combines additive manufacturing and an electroless plating processes to enable the rapid prototyping of any anatomically customized, low ESI, reusable dry electrodes without the use of vacuum. Furthermore, the resulting metallized surfaces allow traditional soldering which is needed for integration with conventional data processing electronics and require no skin preparation.
To establish the range of devices that can be fabricated with this process, two dry sensor topographies were designed based off pin-electrodes and smooth surface electrodes that have been used in sleep, epilepsy, and fitness studies. Pin electrodes with bulbed tips were fabricated for increased patient comfort (Figure 1a) during EEG recordings through hair. Versatile circular electrodes were integrated with flexible 3D printed wristband structures for use with different arm and wrist sizes for ECG and EMG measurements (Figure 1b). When compared to clinically standard, wet electrodes of the same area, they express comparable electrode-skin impedances. The results demonstrate a scalable electrode fabrication process that can rapidly produce anatomically optimized, low impedance, 3D dry electrodes for repeatable physiological recordings over long periods of time (Figure 1c). Unlike existing fabrication techniques for similarly customizable electrodes, the presented process does not require vacuum and can be cost-effective to both quickly iterate on prototypes or produce wearable electrodes at scale.
Figure 1: Additive manufacturing and electroless plating processes can rapidly fabricate dry electrodes customized for different wearables and partial limbs. This single process can produce designs ranging from dense (a) pin electrodes for EEG recordings through hair to (b) curved/conformal electrodes myoelectric prosthetics at scale. (c) The resulting gold-finished dry electrodes can also be reused over the course of months without any skin preparation.
Results:

Fabrication Process:

The electrode structures were printed using a stereolithography 3D printer (Formlabs Form 2 Printer) with a standard, clear methacrylate photopolymer (Figure 2a). Stereolithographic (SLA) printers create 3D structures by precisely laser-curing photosensitive polymer resins in a layer-by-layer fashion. This method of 3D printing results in much finer resolutions and wider choices of materials than fused deposition modeling (FDM) printing, which melts and extrudes plastic filaments. After printing, the samples were post-processed with a 20-minute IPA bath to rinse away uncured resin, followed by an approximately hour-long UV curing process to fully cure the surface.

The structures were subsequently sandblasted with 100 grit white fused aluminum oxide blasting media (Industrial Supply, Twin Falls, ID) to increase their surface area [19] (Figure 2b). Sandblasting promotes better film adhesion and lower skin-electrode impedance. The samples were sonicated in a bath of DI water with Alconox cleaning solution for approximately 10 minutes before rinsing with DI water. The surface energy of the printed structures is modified in a bath of 1% benzalkonium chloride (Sigma Aldrich 12060-100G) surfactant solution for 10 minutes. These steps ensure the plating surface is clean and promote adhesion of the catalyst.

The metal plating process is a result of subsequent plating and cleaning steps. Prior to each plating step, the samples were rinsed and dried thoroughly. First, the 3D printed electrodes were submerged in a beaker of a palladium-tin catalyst for 10 minutes, followed by the copper plating solution for a minimum of 4 hours, which provides a thick base layer of metal for gold plating as shown in Figure 2c. After copper plating, electrodes are soaked in the surfactant and catalyst solutions and then moved to a gold plating solution for approximately 15 minutes (Sigma Aldrich
901670-250ML) (Figure 2d). The second gold plating step is achieved by placing the samples in the same surfactant, catalyst, and gold plating solutions to form the top layer of the dry electrodes (Figure 2e). After all plating steps, tinned copper wires were directly soldered to the electrode surface to facilitate electrode integration with recording systems. Detailed preparation instructions and processing steps for catalyst and copper plating solutions can be found below in the methods section.

Figure 2

Figure 2: Fabrication process. (a) Electrode design is 3D printed with an SLA printer. (b) Samples are sandblasted and then cleaned. (c) Electrodes are metalized with copper via exposure to surfactant, catalyst, and copper plating solutions in sequence. (d) The first layer of gold is formed by submersion in surfactant, catalyst, and gold plating solution. (e) The previous step is repeated once more to improve the device longevity and performance.
Material characterization:

**Sheet Resistance:**

Sheet resistance was characterized with 4-point impedance measurements immediately after plating and once a week for nine weeks. Samples of single metal plating layers were compared against double layers, using copper as the control film. As prepared, copper plated control samples, single layer gold samples, and double layer gold samples exhibited an average sheet resistance and standard deviation of $19.2 \pm 1.7 \text{ m}\Omega/\square$, $17.6 \pm 1.1 \text{ m}\Omega/\square$, and $16.5 \pm 0.7 \text{ m}\Omega/\square$, respectively. The second plated layer of gold was shown to lower the sample’s average sheet resistance as well as its standard deviation (Figure 3a) and was adopted for the longevity study. It was observed that the deposition of a single layer of gold was not adequate to prevent copper oxidation after a week. After nine weeks, an oxide layer was observed in the copper samples. In addition, a 20% increase in sheet resistance was measured. This contrasts with the 2-layer gold samples which showed no meaningful change in sheet resistance or oxide formation. (Figure 3b). The sheet resistance of both the copper and gold are displayed in reference to their respective sheet resistances on day 0.

*Film Integrity/Acid Dip Tests:*

Kapton tape was applied around entire electrode surfaces and then removed. No visible gold or copper was removed with the tape. Samples also underwent nitric acid baths (Figure 3c). Nitric acid, a typical copper etchant, will readily dissolve copper but is not suitable for etching gold [20]. No noticeable differences were observed after dipping the samples with two gold layers. Control samples made of copper were quickly and completely etched down to the bare substrate.
**Surface Roughness:**

To assess the surface roughness, light microscopy photographs were taken and stylus profilometry was performed on flat, slide-like samples of the same 3D printed photopolymer (Figure 3d & Table 1). The samples were printed, masked with tape, and subjected to the same plating process described above to form selectively patterned films for comparison. A Dektak stylus profilometer was used to collect measurements on these samples at different points in the process and the surface roughness average, $R_a$, was calculated. Due to each plating layer being $\leq 1$ um thick, the surface roughness was virtually unchanged after the initial sandblasting.

<table>
<thead>
<tr>
<th></th>
<th>Bare print</th>
<th>Sandblasted Surface</th>
<th>Cu</th>
<th>1 layer Au</th>
<th>2 layer Au</th>
</tr>
</thead>
<tbody>
<tr>
<td>Avg Roughness, $R_a$ (µm)</td>
<td>3.34 ± 0.90</td>
<td>3.34 ± 0.67</td>
<td>4.32 ± 1.10</td>
<td>3.33 ± 1.19</td>
<td>3.50 ± 0.82</td>
</tr>
</tbody>
</table>

*Table 1: Table listing the sample surface roughness at different points along the plating procedure*
Figure 3: Electrical and physical characterizations of Au plated surfaces. (a) Box plot of absolute sheet resistance immediately after plating. (b) Mean (solid line) sheet resistance and standard deviation (shaded region) over time normalized to day 0. Photos of the copper and copper + two layers of gold samples across the 9-week longevity study. (c) Before and after photographs of acid dip test. The copper sample was etched bare, while the double layered gold sample showed no effect. (d) Light microscopy images of plated surfaces showcasing the roughness resulting from sandblasting.
**Electrode Skin Impedance and Electrode DC Offset:**

Impedance spectroscopy and EDO measurements with and without electrolyte gel were compared to those of gold cup electrodes (surface area of roughly 70mm²). All ESI measurements were performed between two equivalent electrodes, one on the back of the head through the hair, and the other on the ipsilateral mastoid. The presented spectra in Figure 4 are half of the measurement and represent the impedance of one electrode. To emulate real-world scenarios for the dry electrodes, no skin preparation was performed before each trial and measurement sessions were repeated over the course of several weeks. Gold cup electrode measurements used typical skin cleaning and preparation prior to each trial. All measurements were performed with an LCR meter (E4980 A, Keysight) and results were fit to an equivalent circuit model using a constant phase element (CPE) (spectra shown in Figure 4a, circuit models shown in Figure 4b and c).

The dry, gold-plated, pin electrodes (contact surface area of ~50mm²) exhibited an average impedance of 71.2 kΩ and phase of -20° at 50 Hz. When a small amount of electrolyte gel was added to the 3D printed electrodes, the 50 Hz impedance dropped slightly to 40 kΩ while the phase increased to -15°. This is consistent with expected electrode behavior [21]. Comparatively, clinical gold-cup electrodes performed very similarly to the gold-plated electrodes with hydrogel with a 50 Hz impedance of 39 kΩ and phase of -15°. Since the dry pin electrodes and wet gold cup electrodes have different contact areas, an overall resistivity should be used as a comparison heuristic to compare their performance. The dry electrode exhibits an electrode resistivity of 3.56 Ωm² while the wet electrode has a resistivity of 2.73 Ωm². While the wet electrode still has a lower resistivity overall, the dry electrode performance is significantly more area efficient than existing wearable dry electrodes [9]. Furthermore, the dry electrode’s gold finish was stable over time. No noticeable surface degradation was observed despite daily
reuse and cleaning over the course of two months nor did the measured impedance values increase throughout the observation period.

![Figure 4](image)

*Figure 4: Electrode-skin interface (ESI) characterization of the gold-plated pin electrodes vs clinical gold cup electrodes. (a) Average (n = 5) magnitude and phase of gold-plated pin electrodes on scalp with and without electrolyte gel alongside gold cup electrodes. An electrode model with a constant phase element (CPE) model, Z\text{elec}, was fitted to the dry gold-plated pin electrodes. (b) 3D-printed, gold-plated, dry scalp electrode and clinical gold cup electrode. (c) CPE model with fitted parameters for spread resistance R_s, charge transfer resistance R_{ct}, and CPE double layer.*

Electrode DC offset (EDO) measurements were taken between each dry sensing electrode and an equivalently sized dry reference electrode with the WANDmini [10] wireless recording device (see Methods section) with an input range up to 400 mV and an input impedance of 10 MΩ. No skin cleaning or abrasion was performed. The mean EDO and standard deviation were -20 mV ± 10 mV. The minimum and maximum EDO values are −32 mV and 23 mV, respectively. It is important to note that the measured EDO is not equivalent to the electrode's open-circuit potential due to the finite input impedance of the recording front end.
**ExG Measurements:**

ECG, EMG, and EEG signals were recorded using the WANDmini system (see Methods for more details). Three subjects (one male, two female) were recruited to perform proof-of-concept ExG measurements. To mimic realistic day-to-day scenarios, no skin preparation was performed prior to dry electrode donning in all experiments. When wet electrodes were used for comparison measurements, a trained technician performed skin preparation and hydrogel application. The proof-of-concept study was approved by UC Berkeley’s Institutional Review Board (CPHS protocol ID: 2018-09-11395).

**Dry Electrode Comfort:**

Subjects donned the curved, dry electrode armband (Figure 5e), dry pin-electrode headband (Figure 5h), and clinically standard, wet, gold cup electrodes (Figure 4b) for three 2 hour sessions. After the third session, subjects asked to rate the comfort of each electrode form factor on a scale of 1-5 (1: unbearably uncomfortable and 5: extremely comfortable). User scores are in Table 2. Wet electrodes had an average rating of 2.3 with subjects citing skin abrasion and residual gel as defining uncomfortable factors. Both dry electrodes were rated 4 on average. The dry electrodes had no skin preparation and left no lingering skin irritation, soreness, or redness. Subjects commented that the curved dry electrodes felt like wearing jewelry and could be worn long term. Wet electrode sites exhibited minor irritation and lingering redness.

<table>
<thead>
<tr>
<th>Electrode</th>
<th>User 1</th>
<th>User 2</th>
<th>User 3</th>
<th>Average</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pin Dry Electrode</td>
<td>4</td>
<td>4</td>
<td>4</td>
<td>4</td>
</tr>
<tr>
<td>Curved Dry Electrode</td>
<td>5</td>
<td>4</td>
<td>4</td>
<td>4.3</td>
</tr>
<tr>
<td>Wet Electrode</td>
<td>3</td>
<td>2</td>
<td>2</td>
<td>2.3</td>
</tr>
</tbody>
</table>

*Table 2: User-specific scoring of the dry electrode headband, dry electrode arm band, and wet electrodes.*
**Electrocardiography (ECG):**

Cross body ECG measurements were performed using 3-electrode armbands with embedded curved electrodes on each arm. Differential measurements were taken across both arm bands to record a cross-body ECG (Figure 5b). One of the dry electrodes was selected as a ground to reduce interference in the recordings. A clear ECG rhythm (PQRST complex) was recorded and is labeled in Figure 5a.

**Electromyography (EMG):**

Bicep EMG was recorded on a single subject wearing two bands on a single arm, one across the bicep and another by the elbow. The elbow electrodes were used as both a system ground and as references for the bicep electrodes. Sensing electrodes were placed along the bicep. Users were cued to flex their biceps every 5 seconds, resulting in clear increase in broad spectrum activity (0-200 Hz) both in frequency and time domain (Figure 5c, d).

**Electroencephalography (EEG):**

A Single user’s alpha attenuation response was measured with both wet and dry electrode-based recording setups. Alpha rhythms are a spontaneous neural signal centered between 8-12 Hz that reflect a person’s state of attention and can be modulated by opening and closing their eyes. Originating from the occipital lobe, alpha waves are large amplitude signals that are commonly used to benchmark EEG systems. First, the user would wear a 3D printed headband with dry pin-style electrodes. The headband was oriented to roughly correspond to T3 and T4 recording sides (according to the traditional 10-20 map). Afterwards, the dry electrodes were swapped out for
clinically standard gold cup electrodes (Figure 4b) and the experiment was repeated. To reduce interference and make the comparison as consistent as possible between the two measurements, a single wet ground electrode was placed on the subject’s left mastoid and maintained for both wet and dry electrode measurements. In both setups, the subject was tasked to close/open their eyes every 30 seconds to modulate alpha band activity. When the subject closed their eyes, alpha band power increased while ocular related artifacts decreased (Figure 5f). The results showed comparable performance between the wet and dry electrodes. Both wet and dry electrode setups recorded alpha modulation ratios of 4.89 and 4.91, respectively (Figure 5g).
Figure 5: Sample physiological measurements taken with gold-plated, dry electrodes. (a) Time domain measurement of single lead, cross-body ECG measurement with labeled PQRST complexes. (b) ECG measurement set up with 3D-printed bands equipped with dry electrodes on each arm. (c) Spectrogram and (d) time-domain of EMG measurements taken with two electrode bands (e) placed along a single subject’s arm. The subject was cued to flex their bicep every 5 seconds. Characteristic broad spectrum muscle activity (~0 - 200 Hz) was observed when the user was cued to flex. (f) Sample Time-frequency spectrogram of a single dry electrode alpha attenuation response. (g) Averaged (n=3) alpha (8-12 Hz) band power of alpha modulation recorded across a subject’s scalp for both dry pin and wet gold cup electrode measurements. Alpha power increased by a factor of 5x in the eyes-closed state for both dry and wet measurements. (h) Alpha attenuation response measurement setup. Two dry electrodes were placed on either side of the head.
Discussion:

This multi-layer electroless plating fabrication technique enables the rapid prototyping of customized electrodes for all applications. Unlike techniques such as sputtering, evaporation, or spin-coating, the presented electroless plating process will evenly metalize any electrode shape with a gold finish, enabling optimization for any recording location without vacuum or constraints on surface feature density, overhangs, and assembly. The resulting electrode surfaces are robust enough to be directly soldered to, which allows integration with the WANDmini and other systems that use conventional sensor data processing and transmission electronics. Furthermore, this procedure has the added benefit of being performed at relatively low temperatures with standard laboratory equipment, thus preventing carbon scoring or mid-process thin film thermal expansion that can lead to film flaking and peeling (a common issue with electroplating thin films). This increased ease-of-manufacture and ease-of-use also allows for easier iteration to find comfortable dry electrode designs for specific users and or recording locations (e.g. inside the ear). Since the electroless gold plating process is self-limiting and plates only 0.25µm before ceasing, introducing a two-layer stack method increases the thickness. This limits grain-boundary diffusion [22] and extends electrode lifetime, as the electrodes show robust stability over time even after exposure to human skin and ambient air. Tape and acid dip tests further confirm robust gold film integrity and adhesion to the different layers as well as the substrate.

Direct comparisons between the 3D printed, gold-plated dry electrodes and clinical controls are shown by the impedance spectra and EDO measurements with and without electrolyte gel. The ESI, which is crucial to electrode performance, is inversely proportional to the effective electrode surface area [21]. The ESI is further lowered in this process by creating a rough surface via sandblasting, which increases the surface area without increasing the overall size of electrodes.
This rough surface is maintained throughout the plating steps as evidenced by the lack of change in surface roughness over each step. The dry, gold-plated electrodes were specifically designed to record through head hair and (without any skin preparation) achieved similar ESI performance to the scalp electrode controls when accounting for the difference in surface area. They also maintain EDOs within the input range of state-of-the-art recording front ends [23] [24].

The ECG, EMG, and EEG experiments comprehensively demonstrate the benefits of this optimized fabrication process in sample use cases. In ECG, clear waveforms were shown, demonstrating potential to detect heart rate, heart rate variability, and cardiac arrhythmias. The presented EMG exhibited expected broadband muscle activity. With EEG, the alpha attenuation response recorded with the dry electrodes exhibited the same alpha modulation as the wet electrode case. All three signal paradigms were recorded without any skin cleaning, abrasion, or hydrogel.

Finally, as these electrodes were repeatedly used and tested over the course of 4 months, there was no degradation in appearance or ESI, nor any adverse skin reaction on users. Ultimately, the presented rapid, adaptable, and low-complexity fabrication process results in re-usable, long lasting, and anatomically fit dry electrodes that can enable new neural wearables and devices for day-to-day brain-computer interfaces.
Methods:

Plating Solutions:

Both the catalyst and copper plating solutions were made in house. The catalyst solution was prepared between 60-70°C and was stirred for approximately one hour after all components were added. This solution is prepared in full and lasts several weeks before the salts precipitate and is no longer usable.

The two main components for an electroless plating solution are a metallic salt and a reducing agent, in this case, copper(II) sulfate and formaldehyde respectively. At a sufficiently high pH (the solution is adjusted to 12.8 by adding NaOH), formaldehyde reacts with hydroxide ions in solution to reduce copper ions in the salt:

\[
\text{Cu}^{2+} + 2\text{HCHO} + 4\text{OH}^- \rightarrow \text{Cu} + \text{H}_2 + 2\text{H}_2\text{O} + 2\text{HCO}_2^-
\]

In the solution we add EDTA, which acts as a complexing agent (as copper salts are insoluble at pH > 4), and ferrocyanide, which acts to stabilize the solution over time (the solution is stored without the addition of formaldehyde, which is then added to the appropriate quantity being used for plating). Layers are typically built at a rate of about <1μm/h at ambient temperature [25], so leaving the samples for several hours or overnight in a covered plating solution provides sufficient coverage. A lightly bubbling nitrogen line was left in the solution to provide light agitation, promote even coverage by displacing the hydrogen gas product, and to limit oxidization of the copper during the plating process.

Lastly, the gold layer was applied by heating the gold plating solution to about 90°C and submerging the samples for about 15 minutes. This process is self-limiting, as gold layers adhere to the copper only and not upon itself. Recipes for each solution can be found in Table 3.
<table>
<thead>
<tr>
<th>Solution</th>
<th>Components</th>
<th>Purpose</th>
</tr>
</thead>
<tbody>
<tr>
<td>Catalyst</td>
<td>• 1 L deionized water &lt;br&gt;• 60 mL HCl &lt;br&gt;• 0.25g PdCl₂ 12g SnCl₂, after PdCl₂ completely dissolves</td>
<td>Provides very thin initial palladium layer for copper adhesion</td>
</tr>
<tr>
<td>Electroless Copper</td>
<td>• 1000 mL deionized water &lt;br&gt;• 18g CuSO₄·5H₂O &lt;br&gt;• 48g EDTA &lt;br&gt;• 57.2mg K₄Fe(CN)₆·3H₂O &lt;br&gt;• 1 mL HCl &lt;br&gt;• NaOH as needed to adjust pH to 12.8 &lt;br&gt;• Formaldehyde, when ready for use in 22.5:1 ratio of plating solution:formaldehyde</td>
<td>Plates a thick layer of highly conductive material</td>
</tr>
<tr>
<td>Electroless Gold</td>
<td>Bright electroless gold plating solution (Sigma-Aldrich Part Number: 901670)</td>
<td>Prevents copper oxidation and improves biocompatibility</td>
</tr>
</tbody>
</table>

*Table 3: Plating process solution components and purpose. All materials were purchased from Sigma-Aldrich.*
**Electrode Model Fitting:**

While many electrode models exist within the literature, CPEs most accurately fit electrode behavior due to their ability to model the imperfect double layer formed at the electrode-skin interface [21]. CPEs are modeled by:

\[
Z_{CPE} = \frac{1}{(j\omega)^n Q}
\]

*Equation 1*

where \(0 < n < 1\). \(Q\) is a measure of the magnitude of \(Z_{CPE}\) while \(n\) fits the bilayer phase offset.

The CPE based electrode model’s impedance, \(Z_{Elec}\), can be described by

\[
Z_{Elec} = R_s + \frac{R_{CT}}{1 + (j\omega)^n Q R_{CT}}
\]

*Equation 2*

where \(R_s\) is spread resistance and \(R_{ct}\) is the charge-transfer resistance.

**Experimental ExG Recording System:**

In this work, ExG signals are acquired using a miniature, wireless, artifact-free neuromodulation device (WANDmini) [10], a low-profile, custom neural recording system that streams recorded data over Bluetooth Low Energy (BLE) to a base station connected to a laptop. WANDmini is derived from a previous design for a wireless, artifact-free neuromodulation device (WAND) [14], reduced to a form factor of 2.5 \(\times\) 2.5 cm\(^2\), and embedded with custom firmware (Figure 6). Recording and digitization are performed by a custom neuromodulation IC [26] (NMIC, Cortera Neurotechnologies, Inc.) integrated with 64 digitizing frontends, thereby expandable to recording applications with higher electrode counts. NMIC and WANDmini specifications are listed in Table 4.
The NMIC was selected for its low power and high dynamic range, supporting a 100–400 mV input range with a flat input referred noise voltage spectrum of 70 nV/√Hz. The analog-to-digital converters (ADCs) have a resolution of 15 bits and sample at 1 kSps, providing sufficient resolution and bandwidth for EMG, ECG, and EEG signals. The wide linear input range can accommodate the large electrode DC offsets and provides robustness to interference.

![Figure 6: WANDmini neural recording module beside a scale quarter](image)

Table 4: Relevant NMIC and WANDmini specifications to presented ExG measurements

<table>
<thead>
<tr>
<th>Specification</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Max Recording Channels</td>
<td>64</td>
</tr>
<tr>
<td>Input Range</td>
<td>100 – 400 mV</td>
</tr>
<tr>
<td>Input referred noise voltage spectrum</td>
<td>70 nV/√Hz</td>
</tr>
<tr>
<td>Input Impedance</td>
<td>40 Ω</td>
</tr>
<tr>
<td>ADC Resolution</td>
<td>15 bits</td>
</tr>
<tr>
<td>ADC Sample Rate</td>
<td>1 kS/s</td>
</tr>
<tr>
<td>Wireless Data Rate</td>
<td>2 Mbps</td>
</tr>
<tr>
<td>Board Dimensions</td>
<td>25.4 mm x 25.4 mm</td>
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Acknowledgements:

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