A Wireless, Multielectrode, User-generic Ear EEG Recording System

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***Abstract*—Recently it has been demonstrated that electroencephalography (EEG) can be recorded from the ear canal (in-ear EEG), opening the door to using discreet earpieces as wearable brain-computer interfaces (BCIs). We present, for the first time to our knowledge, a wireless neural recording platform for recording EEG from the ear canal with dry multielectrode, user-generic earpieces. A low-cost manufacturing process involving vacuum forming and spray coating was developed to improve ear canal contact in a range of users and combined with a 2.5 x 2.5 cm^2 wireless recording system. System performance was evaluated through electrode-skin interface (ESI) impedance characterization and measurement of common EEG signals simultaneously with wet scalp EEG, including eye blinks, alpha waves, and the auditory steady-state response (ASSR) across multiple users. The user-generic ear EEG recorded a mean alpha modulation of 2.17, outperforming the state-of-the-art.**

***Index Terms*—dry electrodes, ear EEG, electroencephalography (EEG), wireless neural recording**

I. Introduction

Electroencephalography (EEG) is a common clinical test that measures the brain's electrical activity from the surface of the scalp and is commonly used in clinical settings to diagnose epilepsy and sleep disorders. EEG can also record signals important for brain-computer interfaces (BCIs) such as eye blinks, auditory evoked potentials, alpha waves (neural signals correlated with relaxation) [1], and more. Clinical EEG systems use an array of wet electrodes that are individually placed on a subject's scalp that are bulky, uncomfortable, and result in a degraded signal-to-noise ratio (SNR) as the electrodes dry out [2]. To enable ambulatory recordings, wearable EEG devices have emerged using dry electrodes with recording electronics designed to accommodate dry electrode properties (e.g., higher impedance and interference susceptibility [2]). While easier to use than clinical systems, current wearable devices still require bulky headsets [3].

Recently, recording EEG from inside the ear (in-ear EEG) with dry electrodes has been investigated [4]–[6]. While in-ear EEG cannot provide the same spatial coverage of cortical activity or as high an SNR as clinical EEG devices, it can be significantly more discreet, comfortable, and well-suited for everyday use. To date, all instances of dry electrode in-ear EEG either depend on (1) custom-molded earpieces made per user [4], [7] or (2) exotic material-based earplugs that require wet electrode references. Neither style can be scaled with standard processes and materials to produce multichannel, dry neural recording wearables that fit multiple users. To make in-ear EEG widely usable, this paper presents a wireless, dry electrode in-ear EEG recording platform (user-generic ear EEG) that is fabricated using a scalable, low-cost manufacturing process. Multiple earpieces were fabricated and used to measure three classes of electrophysiological signals across multiple users. Measurements of electrode-skin interface (ESI) impedance, eye blinks, alpha modulation, and auditory steady-state response (ASSR) are presented and compared with wet scalp EEG recordings and state-of-the-art in-ear EEG. The user study was approved by UC Berkeley's Institutional Review Board (CPHS protocol ID: 2018-09-11395).

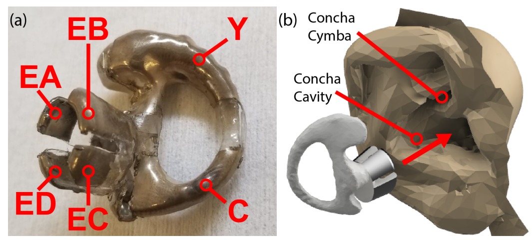

Fig. 1. (a) User-generic earpiece design. In-ear Ag electrodes (EA, EB, EC, & ED) are 60 mm^2. Out-ear Ag electrodes (Y & C) are 4 cm^2 and fit on the ear's concha cymba and concha cavity, respectively. (b) Earpiece fit render.

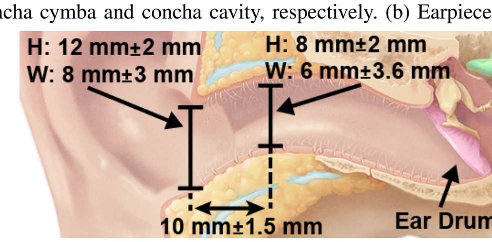

Fig. 2. Ear canal cross-section [8] with average measurements from [9].

II. System Overview

A. User-Generic, Dry Electrode Earpiece

A user-generic electrode scheme, physical earpiece design, and scalable, low-cost manufacturing process are critical to building a versatile neural recording platform. Electrode locations and earpiece design are informed by physiological recordings and anatomical measurements. Initially, a custom-molded earpiece based on a high-resolution scan of a human ear was used to characterize the manufacturing process and test electrode placements similar to those in [4]. In these initial designs, ten 12 mm^2 electrodes were spaced evenly across a customized earpiece. ESI impedance measurements and correlations in physiological measurements informed a final design with four 60 mm^2 electrodes placed in the ear canal and two 4 cm^2 electrodes on the ear's concha cymba and

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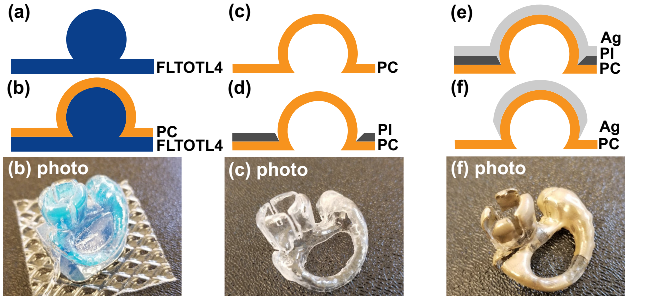

Fig. 3. Thermoforming manufacturing process cross section renders and photographs: (a) 3D-printed master mold; (b) 0.762 mm thick polycarbonate (PC) is thermoformed around master mold; (c) liberated PC from mold; (d) masked PC with laser cut polyimide (PI); (e) masked piece is spray coated with Ag; (f) mask is removed to reveal Ag electrodes on hollow PC earpiece.

concha cavity (Fig. 1). These larger electrodes were optimized for low ESI impedance (relative to the first earpiece and other dry electrode devices [4]) across multiple subjects and centered along physical features near the ear canal aperture [9]. Having two electrodes outside the ear canal provides flexibility when choosing sense/reference schemes for different subjects and their ear morphologies. The final physical earpiece design was generated using ear morphology measurements (Fig. 2) from a large human population [9] and scaling the features highlighted in Fig. 2 up and down by a standard deviation to make small, medium, and large earpieces.

An earpiece manufacturing process was developed based on commercial thermoforming practices. First, a master mold of the physical earpiece was 3D-printed using a temperature-resistant polymer resin (FLTOTL4, Formlabs) (Fig. 3(a)). This produces a durable master mold capable of periodically withstanding high temperatures (important for thermoforming multiple instances of the earpiece). A thin polycarbonate sheet is then heated to 148°C and thermoformed around the master mold using a heated vacuum form chamber (450DT, Formech) (Fig. 3(b)). When liberated from the master mold, the polycarbonate is a hollow representation of the earpiece (providing space for wires and components inside the earpiece) (Fig. 3(c)). With a flared design, the polycarbonate earpiece exhibits a spring constant of 176 N/m at each in-ear electrode (EA, EB, EC, & ED). This embedded spring allows each electrode to apply approximately 9.3 kPa (at 50% strain) on the ear canal. This pressure is on the order of polyurethane foam (at 50% strain) [10], reducing contact impedance and increasing earpiece/electrode stability.

After thermoforming, the hollow earpiece is masked with a laser cut polyimide sheet (Fig. 3(d)) and spray-coated with Ag (Fig. 3(e)). Finally, the polyimide mask is removed to reveal the earpiece with Ag electrodes (Fig. 3(f)). This process can be reliably reproduced and extended to different molds, substrates, and electrode shapes/materials. Additionally, after spray coating a Ag base layer, the electrodes can be electroplated with different materials to optimize for different electrode properties (e.g., Pt to increase long-term stability). This study uses Ag electrodes due to their low impedance properties and prevalence in biomedical recording applications.

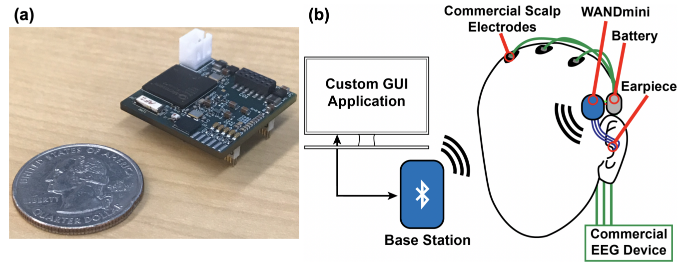

Fig. 4. (a) WANDmini wireless neural recording module. (b) Rendering of the experimental set up. Headworn WANDmini records EEG from the earpiece. All recorded data is transmitted to the base station and visualized in real-time in a GUI, which also provides cues to users.

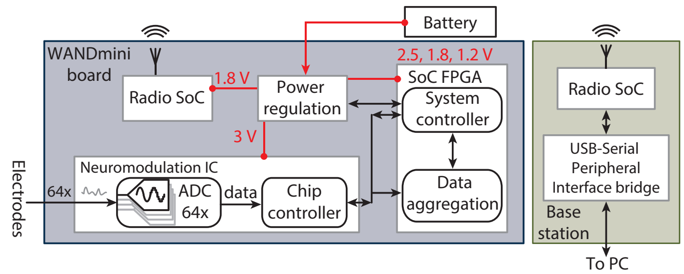

Fig. 5. System architecture of WANDmini wireless neural recording module as used in the user-generic ear EEG platform.

TABLE I
WANDMINI WIRELESS NEURAL RECORDING MODULE SPECIFICATIONS

Maximum Recording Channels	64
Recording Channels Used	5
Input Range	100 mVpp
Noise Floor	70 nV/$\sqrt{\text{Hz}}$
ADC Resolution	15 bits
ADC Sample Rate	1 kS/s
Wireless Data Rate	2 Mbps
PCB Dimensions	25.4 mm $\times$ 25.4 mm
Weight (w/o battery)	3.8 g
Power	46 mW

B. Wireless Neural Recording Module - WANDmini

EEG signals are acquired using a compact neural recording module (WANDmini) that wirelessly transmits data to a base station connected to a laptop (Fig. 4 and 5). The device is based on a previous design for a wireless, artifact-free neuromodulation device (WAND) [11] with a reduced form factor of 2.5 $\times$ 2.5 cm^2. The recording, digitization, and serialization are performed by a custom neuromodulation IC [12] (NMIC, Cortera Neurotechnologies, Inc.) with 64 digitizing frontends integrated into a compact footprint and serving as a platform for recording applications with higher electrode counts. WANDmini specifications are listed in Table I. The wide input range can accommodate large DC offsets generated by dry electrode half-cell potentials and provides robustness to interference. The flat noise spectrum down to $<$1 Hz ensures minimal electronic noise in the band of interest. The NMIC also has stimulation capabilities, which are not used in this study. System control and data aggregation are performed by an on-board SoC FPGA with a 166 MHz ARM Cortex-M3 processor (SmartFusion2 M2S025, Microsemi), enabling reprogrammable on-board computations. Digitized EEG data are transmitted to the base station by a 2.4 GHz Bluetooth Low Energy (BLE) radio (nRF51822, Nordic Semiconductor).

TABLE II
PARAMETRIC FIT PARAMETERS FOR ELECTRODE MODEL IN FIG. 6(B)

Mean Values	$R_s (\Omega)$	$R_{ct} (\Omega)$	CPE_{dl_Q}	CPE_{dl_n}
CPE Model	678	793 k	0.793 x10^{-9}	0.760

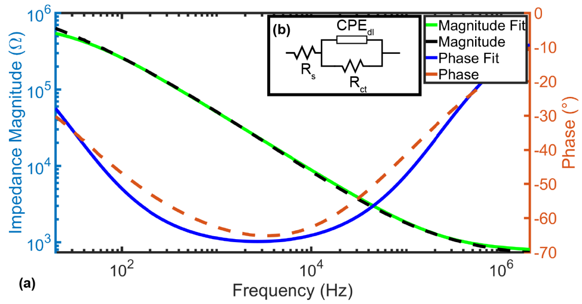

Fig. 6. (a) Average impedance spectrum of dry, in-ear Ag electrodes across all users (n = 44). (b) Electrode circuit model: a spread resistance, R_s, in series with a double layer constant phase element (CPE), CPE_{dl}, and charge transfer resistance, R_{ct}.

The radio protocol also allows uplink from the base station to send commands such as starting and stopping acquisition and configuration commands to the NMIC.

When streaming 64 channels of data, the device consumes 46 mW, of which <1 mW is dissipated by the NMIC, 10 mW are dissipated by the radio, and the remaining power is consumed by the SoC FPGA and power management circuits. Powered by a 3.7 V, 550 mAh LiPo battery, the device can operate for ~44 hours. The user-generic ear EEG system's firmware is configurable to adjust channel count, input range, and error correcting codes for the BLE transmission. Data streamed to the laptop's base station is visualized in real-time with a Python GUI that also configures the WANDmini and cues the user.

An integrated BCI system ideally also performs extensive signal processing. WANDmini employs a Cortex M3 microprocessor that consumes the majority of the power but also provides resources for local interference mitigation, feature extraction, and classification. Further optimization of the wireless neural recording module's form factor for an ear-worn device would enable a fully discreet and comfortable non-invasive BCI system.

III. EXPERIMENTAL RESULTS

To verify system performance, four types of measurements were performed on three subjects: ESI impedance characterization, eye blink tests, alpha modulation measurements, and ASSR. All electrophysiological measurements were simultaneously recorded using wet Au cup electrodes on the scalp. It is expected that scalp measurements will have higher SNR than measurements performed in the ear since (1) wet, scalp electrodes have a lower noise floor than dry electrodes and (2) are spaced farther apart [4]. Wet electrodes were placed at C3, C4, Cz, M1, M2, and the forehead as ground (i.e., 10-20 system). Recordings used a commercial EEG recorder (MPR ST+, Embletta) referenced against M1 and M2. The user-generic ear EEG was referenced and grounded separately with the reference on the concha cymba (Fig. 1) and ground on the ipsilateral mastoid. Subject 1 used the large earpiece while subjects 2 and 3 used the medium-sized earpiece.

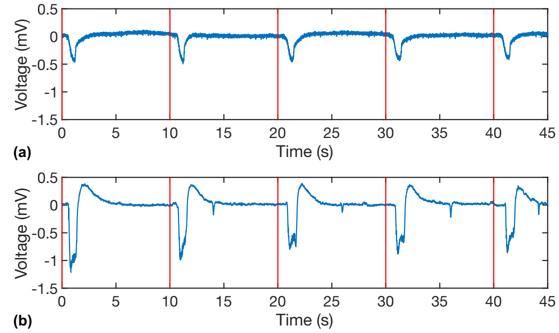

Fig. 7. Recorded eye blinks using (a) user-generic ear EEG and (b) scalp EEG. Visual cues are marked by red lines.

A. Impedance Measurements

Impedance measurements were performed between each dry electrode on the user-generic earpiece and a wet electrode placed on the subject's ipsilateral mastoid. Wet electrodes have an order of magnitude lower impedance than dry electrodes [2], thus impedance measurements are dominated by the single dry ESI. Measurements were performed with an LCR meter (E4980A, Keysight) and results were fitted to an equivalent circuit model (Fig. 6) comprising resistors and a constant phase element (CPE) (Table II). This constant phase element can be approximated with a 5.27 nF capacitor. At 50 Hz, the interface has an average impedance of 392 kΩ and phase of -39°. As expected, impedance measurements are not normally distributed due to user-to-user and time-to-time variation [4]. Impedance measurements were used to grade electrode-skin contact. Acceptable contact was determined by a 20 Hz ESI impedance <1 MΩ; 80% of all measurements meet this criteria.

B. Eye Blinks

To characterize the use of eye blinks as a BCI modality, subjects were asked to blink when prompted by a visual cue. Five visual cues were presented 10 s apart. The recordings were bandpass filtered from 0.05 to 50 Hz. Fig. 7 shows eye blink amplitudes are 0.4 mV and 0.8 mV for user-generic ear EEG and scalp EEG, respectively. Baseline signals are 15 μVrms and 11 μVrms, respectively.

C. Alpha Modulation

Alpha rhythm is a neural signal from 8 to 12 Hz that reflects a person's state of attention or relaxation. It is a useful signal for applications such as drowsiness detection and neurofeedback and is known to increase in power when a person's eyes are closed. Subjects were asked to switch between two conditions every 30 s over the course of 2 minutes: (1) eyes open and focused, and (2) eyes closed and relaxed. The spectrogram in Fig. 8(a) shows a clear example of alpha modulation recorded with user-generic ear EEG. Fig. 8(b) shows the grand average of mean alpha (8–12 Hz) power. Mean alpha modulation is defined as the ratio of mean alpha power from eyes closed to eyes open. The mean alpha modulation (standard deviation) was 2.17 (± 0.69) (V^2/V^2) for all user-generic ear EEG subjects and 5.54 (± 1.84) (V^2/V^2) for wet scalp EEG. Though modulation is greater in scalp EEG, as expected [4], the signal maintains sufficient SNR for user-generic ear EEG detection.

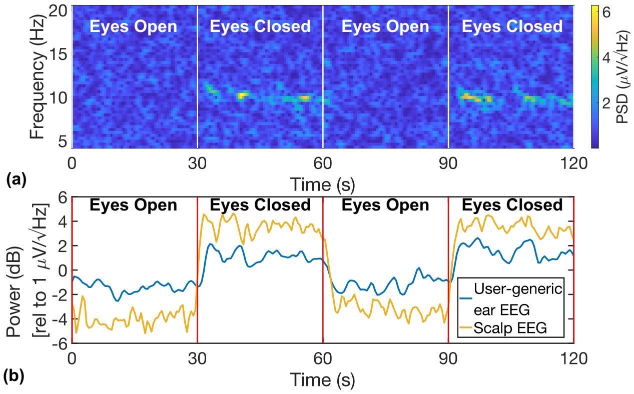

Fig. 8. (a) Time-frequency spectrogram of alpha modulation recorded using the user-generic ear EEG on subject 2. Alpha (8–12 Hz) power increases by a factor of 3.4 during the eyes-closed state. (b) Grand average of mean alpha power for 3 subjects and 44 total trials recorded with user-generic ear EEG and scalp EEG.

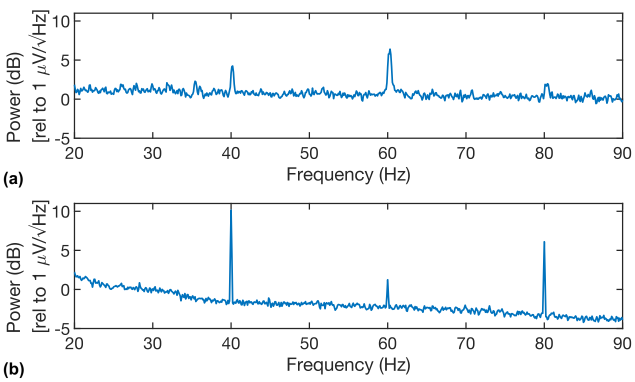

Fig. 9. Auditory evoked potentials (ASSR for 40 Hz click stimulus). (a) Grand average PSD of ASSR for 3 subjects and 28 total trials for user-generic ear EEG showing a mean SNR of 2.5 dB; (b) scalp EEG showing a mean SNR of 11.7 dB.

D. Auditory Steady-State Response (ASSR)

We confirmed the ability to record evoked potentials by measuring ASSR. In an ASSR experiment, the subject listens to a low-frequency auditory stimulus, which elicits a neural response at that frequency. ASSR is useful for applications such as hearing threshold estimation and binary choice via left-vs-right focus. Subjects listened to 40 Hz clicks sampled at 10 kHz played through desktop speaker for 100 s. Fig. 9 shows the grand average of the power spectral density showing clear neural responses at 40 Hz. The mean SNR was 2.5 dB for user-generic ear EEG and 11.7 dB for scalp EEG. Mean SNR was calculated as the ratio of the power at 40 Hz to the mean power from 35 to 45 Hz (excluding 40 Hz) [4]. The SNR is lower for user-generic ear EEG due to higher noise floor, as previously discussed, and shorter electrode distance, which reduces signal amplitude [4].

IV. Summary

A discrete ear EEG recording system with a user-generic earpiece and wireless neural recording module is presented. A low-cost, scalable manufacturing process for building generic earpieces is developed along with a small form factor, wireless neural recording module. Eye blinks, alpha modulation, and ASSR recordings are demonstrated in multiple trials across 3 subjects. The mean recorded alpha modulation was used to compare the user-generic ear EEG platform's sensitivity with a state-of-the-art dry electrode in-ear EEG system [4]. Though user-to-user variability may play a role, the user-generic ear EEG platform recorded a mean alpha modulation of 2.17 while [4] recorded a mean modulation of 1.2 (Table III). To the best of the authors' knowledge, this is the only work that supports multichannel wireless user-generic ear EEG recording across multiple users with the same earpiece design (Table III).

TABLE III
Table Comparing Recent In-ear EEG Systems

	[5]	[7]	[4]	This Work
Electrode Material	CNT/ PDMS	Dry	Dry IrO2	**Dry Ag**
Dry Electrodes	1	3	6	6
Reference	Wet	Dry	Dry	**Dry**
Electrode Area	–	–	9.6 mm^2	60 mm^2**
Impedance @ 50 Hz	50 kΩ*	–	1.1 MΩ*	392 kΩ
Wireless	–	100 kbps	–	**2 Mbps**
Earpiece Style	Generic	Custom	Custom	**Generic**
Scalable	No	No	No	**Yes**
Mean Alpha Modulation	–	–	1.2	**2.17**

*estimated **sense electrode area only

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