

Scalable Fabrication of 3D-Protruded Neural Microelectrodes Using Flexible PCB Technology for High-Fidelity Neural Interfaces

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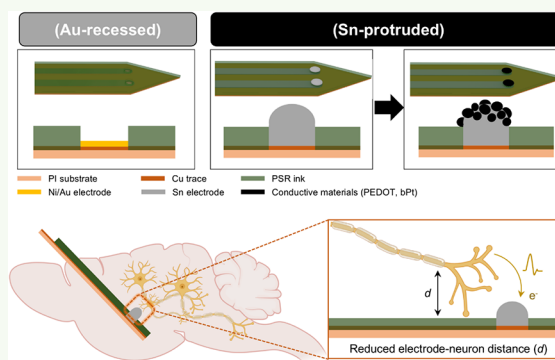
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ABSTRACT: Flexible neural electrodes hold great promise for monitoring electrophysiological activity in the brain, but achieving both fabrication scalability and high signal fidelity remains a significant challenge. Conventional high-performance neural interfaces typically require complex and costly microfabrication processes, while simpler approaches often fail to optimize the electrode–tissue interface. Here, we introduce a scalable, cost-effective platform for fabricating high-performance three-dimensional (3D) neural electrodes using an industrial flexible printed circuit board (fPCB) process. We systematically compare two electrode architectures: (i) recessed microelectrodes with gold (Au) electroless plating, serving as a benchmark, and (ii) 3D protruded microelectrodes formed by tin (Sn) electroplating. To isolate the role of geometry, both electrode types were further functionalized with high-surface-area platinum black (Pt black) and poly(3,4-ethylenedioxythiophene) (PEDOT) coatings. Electrochemical analysis revealed that the Sn-protruded electrodes exhibited lower impedance at 1 kHz and a higher charge storage capacity (CSC) than Au-recessed electrodes. In vivo recordings from the mouse hippocampus further demonstrated that Sn-protruded electrodes achieved a significantly higher signal-to-noise ratio (SNR) and stronger spike amplitudes compared to both bare and coated Au-recessed electrodes. These findings establish 3D protruded electrode geometry as a key determinant of recording fidelity, primarily by reducing the electrode–neuron distance. More broadly, our results demonstrate that the fPCB-based approach provides a rapid, accessible, and scalable route to high-performance neural interfaces, highlighting its potential for widespread adoption in neural engineering.

KEYWORDS: flexible neural electrode, 3D electrode, flexible PCB, electrophysiology, electrodeposition



1. INTRODUCTION

The brain is an extraordinarily complex network of billions of interconnected neurons that generate and transmit bioelectrical signals to form functional circuits. These electrochemical dynamics underpin higher cognitive functions, including memory, learning, and consciousness, as well as motor control.^{1–6} Accordingly, the ability to record neural activity with high fidelity in real time is essential for advancing our understanding of brain function and for elucidating the mechanisms of neurological and psychiatric disorders such as epilepsy, Parkinson's disease, and depression. Implantable neural electrodes provide the critical interface for detecting neuronal activity, and the quality of these recordings directly shapes applications ranging from fundamental neuroscience research to clinical neurotechnologies and next-generation brain–computer interfaces (BCIs).^{7–10}

For decades, silicon-based microelectrode arrays, such as Michigan-style probes, have defined the gold standard in high-resolution neurophysiology.^{11–17} Manufactured through mature semiconductor processes, these devices offer exceptional channel density, reliable signal quality, and proven long-term

functionality. However, their advantages come at a significant cost. Fabrication relies on centralized cleanroom facilities and multistep processes—including photolithography, thin-film deposition, and reactive-ion etching—that are time-intensive and costly, often exceeding \$1,000 per probe with production lead times of weeks. More critically, rigid silicon (Young's modulus ~ 170 GPa) is profoundly mismatched with the mechanical softness of brain tissue (~ 1 – 10 kPa).^{18,19} This mismatch induces micromotion, chronic inflammation, and foreign-body encapsulation, which degrade the electrode–tissue interface and ultimately diminish recording quality over time.

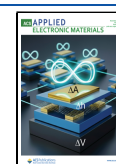
To overcome these limitations, flexible neural probes based on soft polymers such as polyimide (PI), parylene, and SU-8

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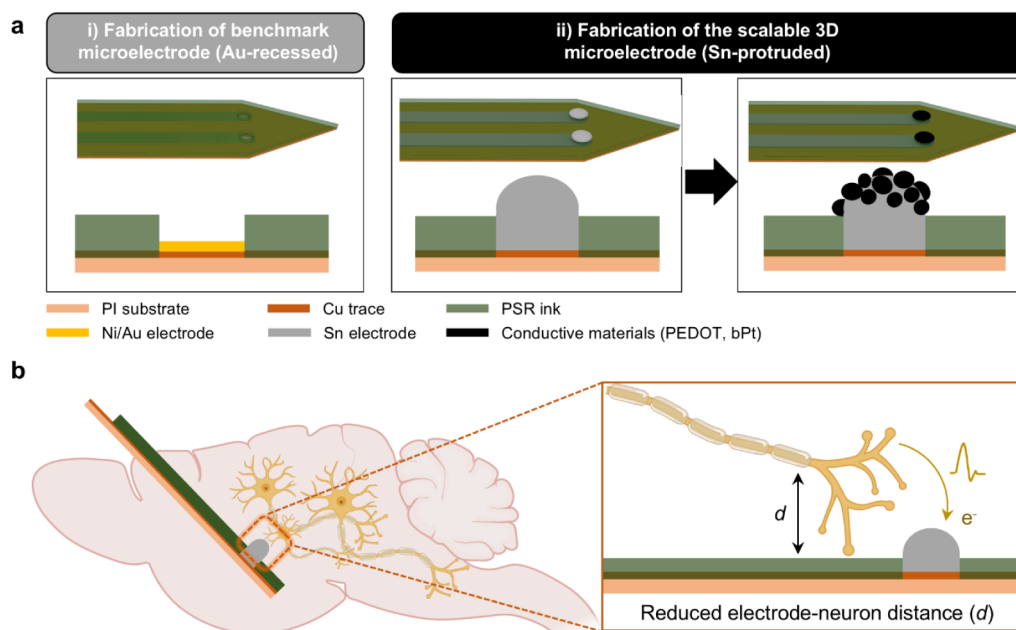


Figure 1. Scalable fabrication of 3D-protruded neural microelectrodes using an fPCB process. (a) Schematic of two types of flexible neural microelectrodes. Left: Benchmark Au-recessed electrodes were formed by patterning Ni/Au on Cu traces and encapsulating the surface with a thick PSR ink layer, resulting in recessed recording sites. Right: Scalable 3D-protruded electrodes were fabricated by selectively electroplating Sn onto exposed Cu traces, followed by surface modification with conductive materials such as Pt black or PEDOT:PSS. (b) Conceptual diagram illustrating the reduced electrode-to-neuron distance achieved by the 3D-protruded geometry.

have been developed.^{20–25} These materials better match the compliance of brain tissue, thereby reducing inflammatory responses and supporting more stable long-term integration.^{26,27} In parallel, researchers have explored engineering electrode geometries in three dimensions—including porous,^{28–30} recessed,^{31–34} and protruded structures^{32,35,36}—to improve recording fidelity. Such designs are hypothesized to increase effective surface area, reduce impedance, and improve neuronal coupling.³² However, the field still lacks a clear consensus on optimal electrode design. A key challenge is that many prior studies conflate geometric effects with surface modifications, making it difficult to isolate the true role of electrode architecture. Furthermore, even flexible probes frequently rely on laboratory-scale cleanroom fabrication, which remains costly, labor-intensive, and poorly scalable for widespread adoption.

In this work, we introduce a neural probe platform that uniquely integrates scalable manufacturing with systematic electrode structural optimization. Our probes are fabricated using an industrial fPCB process—a mature, high-throughput technology capable of producing multilayered flexible devices rapidly and at low cost. This industrial pathway addresses key limitations of traditional microfabrication by enabling high-volume production and straightforward customization of electrode layouts, channel counts, and geometries. Building on this platform, we engineered electrode performance through a dual optimization strategy. First, we created three-dimensional protruded electrodes via selective tin (Sn) electroplating, introducing a geometry that physically extends into tissue to reduce electrode–neuron distance. Second, we enhanced the electrode surface with high-surface-area platinum black (Pt black) or the conductive polymer poly (3,4-ethylenedioxythiophene):poly(styrenesulfonate) (PEDOT:PSS), thereby reducing impedance and increasing charge transfer capacity. Together, these geometric and surface

modifications provide synergistic improvements in neural recording fidelity (Figure 1a).

To rigorously validate this approach, we directly compared Sn-protruded electrodes—both bare and surface-enhanced—with conventional Au-recessed electrodes fabricated through electroless plating and equivalently coated with Pt black to control for surface effects. Our results demonstrate that protruded geometry is the dominant factor in enhancing in vivo recording performance, with surface-enhanced Sn-protruded electrodes achieving substantially higher signal-to-noise ratios than recessed counterparts (Figure 1b). In summary, this work establishes a cost-effective and customizable fPCB-based neural probe platform while providing clear experimental evidence that 3D-protruded electrode geometry plays a decisive role in improving electrode–tissue coupling and neural recording fidelity. By bridging scalable industrial fabrication with rational electrode design, our approach paves the way toward next-generation neural interfaces that are simultaneously high-performance, affordable, and broadly accessible.

2. MATERIALS AND METHODS

2.1. Materials. Poly(3,4-ethylenedioxythiophene) (PEDOT), poly(sodium 4-styrenesulfonate) (PSS; $M_w \sim 70,000$), lead(II) acetate trihydrate (AR, 99.0%), and hexachloroplatinic acid solution (H_2PtCl_6) were obtained from Sigma-Aldrich (USA). All other reagents were purchased from Sigma-Aldrich unless otherwise specified.

2.2. Fabrication of Flexible PCB (fPCB)-Based Neural Probes. fPCB-based probes were fabricated to realize three-dimensional (3D) neural electrodes. The probe layout was optimized based on our prior designs and conformed to standard fPCB manufacturing specifications.³⁴ Each probe featured a line width of 125 μm , line spacing of 100 μm , and

electrode diameter of 30 μm , with shanks spaced 3 mm apart to permit simultaneous recordings from multiple brain regions.

Fabrication was carried out through a commercial fPCB foundry using a multistep process. First, a 12 μm copper (Cu) layer was laminated onto a 13 μm PI substrate with a heat-activated adhesive. The Cu layer was patterned by chemical etching to define the circuit traces. For the insulation layer, photoimageable solder resist (PSR) ink was used instead of the more commonly employed PI film. Although PI can be fabricated as thin as $\sim 12 \mu\text{m}$ in standard FPCB processes, it requires an additional adhesive layer ($\sim 20 \mu\text{m}$) to bond it to the substrate, resulting in a total thickness of approximately 30 μm . In contrast, PSR ink does not require an adhesive layer, thereby simplifying the fabrication process and reducing the overall stack height. Additionally, PSR enables patterning of the electrode openings via standard photolithography. This allowed precise exposure of electrode sites as small as $\sim 30 \mu\text{m}$. By comparison, CO_2 laser ablation—commonly used to open PI layers—offers relatively poor resolution ($>100 \mu\text{m}$), which limits the minimum electrode size and can increase misalignment risk. Therefore, within the limitations of the general fPCB fabrication process, 20 μm PSR ink was then applied to insulate the circuitry, leaving only the electrode and connector regions exposed.

Photolithographic exposure and development were used to open electrode windows in the PSR layer. For recessed electrodes, these exposed pads were metalized by electroless nickel/gold (Ni/Au) plating. For 3D-protruded electrodes, selective tin (Sn) electroplating was employed to build vertically elevated structures extending above the PSR surface.

Electroless Ni/Au plating, the conventional method for fPCB circuits, yields relatively rough surfaces with limited deposition thickness (2.5–5 μm). As a result, electrodes fabricated this way remain recessed within the PSR layer, which increases the electrode–neuron distance and may compromise recording fidelity. In contrast, Sn electroplating enables the electrochemical deposition of substantially thicker coatings (up to $\sim 30 \mu\text{m}$), producing protruded electrodes that reduce electrode–neuron separation and thereby enhance signal detection.

2.3. Surface Modification of 3D Electrodes via Electrochemical Coating. To further enhance the electrochemical performance of the 3D-structured neural electrodes, the exposed electrode surfaces were coated with platinum black (Pt black) or poly(3,4-ethylenedioxythiophene):polystyrene-sulfonate (PEDOT:PSS). Both coatings are known to substantially increase effective surface area, reduce impedance, and thereby improve neural signal fidelity.^{37–40}

For Pt black electroplating, the plating solution was prepared by dissolving 3% (w/v) hexachloroplatinic acid hydrate (H_2PtCl_6), 0.025 N hydrochloric acid (HCl), and 0.025% (w/v) lead acetate in deionized (DI) water. Electroplating was performed in a standard three-electrode configuration: the Au pad served as the working electrode (WE), a platinum (Pt) wire as the counter electrode (CE), and an Ag/AgCl wire as the reference electrode (RE). A constant current of $-4 \mu\text{A}$ (vs CE) was applied for 20 s using a potentiostat (NanoZ, Switzerland).

For PEDOT:PSS deposition, the electrolyte consisted of 1% (w/w) PSS in DI water premixed with EDOT monomer. Using the same three-electrode setup as for Pt black, the conductive polymer was electrochemically deposited onto the

electrode surface, forming a conformal coating that improved both conductivity and biocompatibility.

Following electrodeposition, all electrodes were rinsed thoroughly with DI water and dried under ambient conditions to ensure surface cleanliness and stability before subsequent testing.

2.4. Characterization of Customizable fPCB Electrodes. The surface morphology and elemental composition of the electrodes were characterized by field-emission scanning electron microscopy (FE-SEM; SU-8230, Hitachi), energy-dispersive X-ray spectroscopy (EDS; X-max 80, Oxford Instruments), and atomic force microscopy (AFM; NX20, Park Systems).

Electrochemical performance was evaluated using electrochemical impedance spectroscopy (EIS) and cyclic voltammetry (CV) with a potentiostat system (PalmSens4, PalmSens). Both techniques employed a three-electrode configuration, with the neural electrode as the working electrode (WE, a Pt wire as the counter electrode (CE), and an Ag/AgCl wire as the reference electrode (RE). EIS was conducted in 1 $\times$ phosphate-buffered saline (PBS) over a frequency range of 10^5 to 10 Hz with a 100 mV sinusoidal excitation. CV measurements were performed in PBS within a voltage window of -0.6 V to $+0.6 \text{ V}$ (vs Ag/AgCl) at scan rates of 50 mV/s to quantify charge storage capacity and assess redox behavior.

To investigate the role of electrode geometry in shaping the electric field distribution, finite element analysis (FEA) was performed using Sim4Life (ZMT Zurich MedTech AG). The simulation domain was modeled as a 3D volume with a minimum voxel resolution of 0.1 μm . Brain tissue conductivity was assigned as 0.04 S/m, based on the IT'IS Foundation database. Dirichlet boundary conditions (70 mV) were applied to the electrode surface, while Neumann boundary conditions were imposed on the outer boundaries of the domain.

Two electrode configurations were simulated: a recessed electrode and a 3D-protruded electrode. Both models assumed a probe shank thickness of 35 μm and an insulation layer thickness of 20 μm . The protruding Sn electrode was modeled with a height of 10 μm , and the gap between the electrode tip and a neuron-mimicking axon (modeled as a sphere, 2 μm diameter) was set to 10 μm .⁴¹ These simulations enabled direct comparison of the electric field concentration in the vicinity of the neuron-like structure.

2.5. In Vitro Biocompatibility Assessment of fPCB Electrodes. The biocompatibility of fPCB-based neural electrodes was evaluated using an MTT assay (MTT, Sigma-Aldrich, USA), following the ISO 10993 guidelines.⁴² The human hepatocellular carcinoma cell line HepG2 (HB-8065, ATCC, USA) was used for cytotoxicity testing. Sterilized Parafilm (PM-996, Bemis, USA) and standard culture medium (DMEM-high glucose, HyClone, USA) served as negative controls, while Triton X-100 (TR1020-500-00, Biosesang, Korea) and cisplatin (CS2210000, Merck, Germany) were used as positive controls.

The culture medium was supplemented with 10% (v/v) fetal bovine serum (FBS) and 1% (v/v) penicillin/streptomycin (P/S). HepG2 cells were seeded into 96-well plates and incubated for 24 h at 37 $^\circ\text{C}$ in a humidified incubator with 5% CO_2 to allow attachment.

The extraction medium was prepared by immersing sterilized fPCB electrodes in culture medium containing 1% FBS and 1% P/S. Samples were incubated for 48 h at 37 $^\circ\text{C}$ in

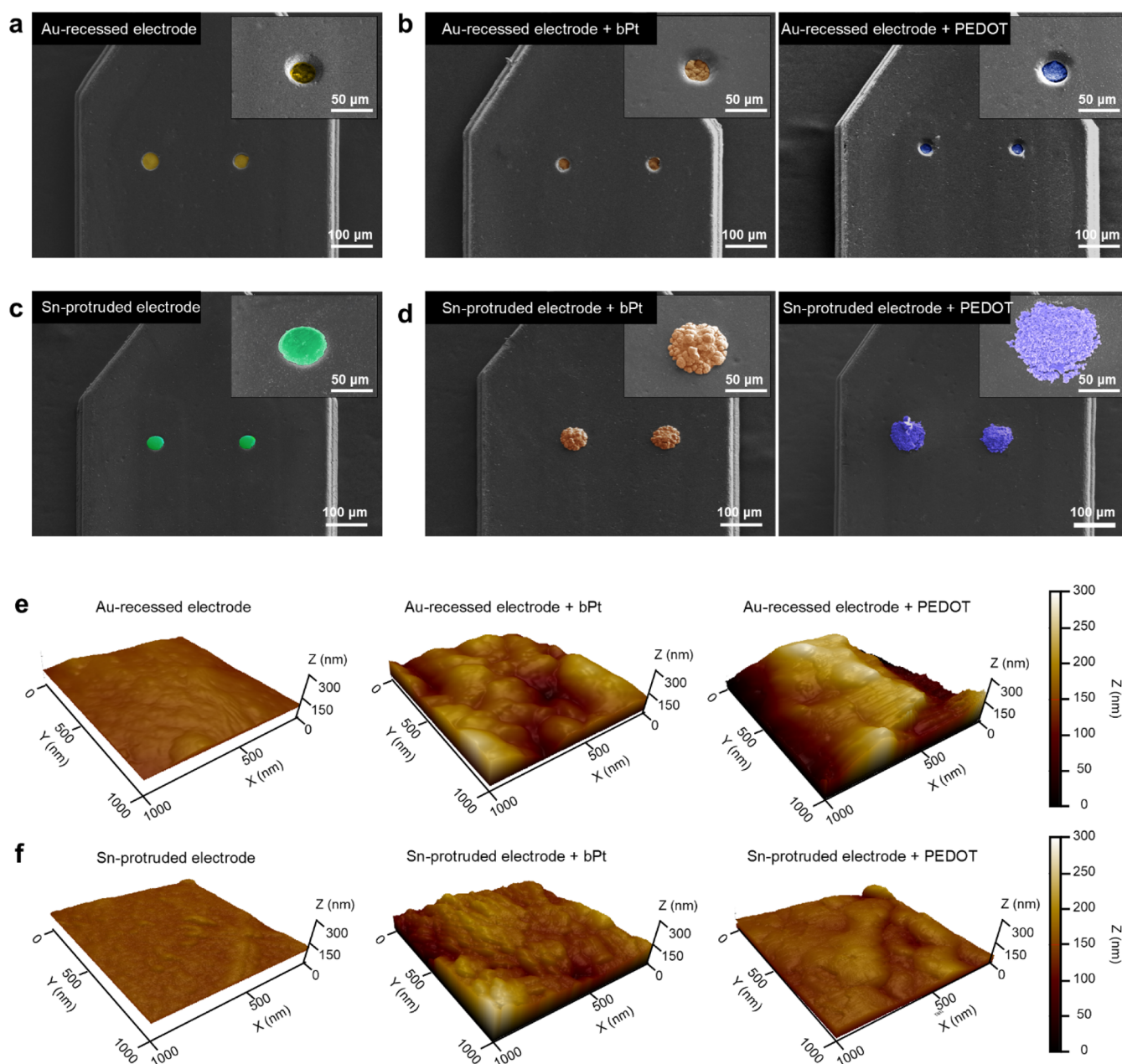


Figure 2. SEM and AFM images of electrode configurations fabricated on the fPCB platform. (a) Au-recessed electrodes (benchmark) showing planar recessed geometry with Ni/Au metallization. (b) Surface-modified Au-recessed electrodes coated with black Pt (bPt, left) and PEDOT:PSS (right), exhibiting porous metallic and polymeric morphologies, respectively. (c) Sn-protruded electrodes formed by selective Sn electroplating, producing dome-shaped protrusions above the PSR insulation. (d) Surface-modified Sn-protruded electrodes with bPt (left) and PEDOT:PSS (right), displaying significantly enlarged and textured surfaces due to the combined effects of 3D geometry and surface coatings. Insets: magnified SEM views of individual electrodes. Scale bars: 100 μm (main), 50 μm (insets). (e,f) AFM topography images of Au-recessed (e) and Sn-protruded (f) electrodes before and after bPt or PEDOT coating, showing increased surface roughness and height variation following surface modification.

a shaking incubator to promote elution of potential leachates. The resulting extract was serially diluted to concentrations of 12.5%, 25%, 50%, 75%, and 100% (v/v).

Following removal of the culture medium, 100 μL of each diluted extract was added to the cell-seeded wells and incubated for 72 h under identical conditions. The assay medium was then replaced with MTT solution (1 mg/mL in culture medium), and cells were incubated for an additional 2 h at 37 $^{\circ}\text{C}$. After incubation, the MTT solution was removed, and formazan crystals were solubilized in dimethyl sulfoxide

(DMSO). Absorbance was measured at 570 nm using a microplate reader (Synergy, BioTek, USA).

Cell viability was calculated by normalizing absorbance values to the negative control (defined as 100%), while the positive control was defined as 0%. This provided a quantitative assessment of any cytotoxic effects induced by the fPCB electrodes.

2.6. In Vivo Neural Recording Test. All animal experiments were approved by the Institutional Animal Care and Use Committee (IACUC) of the Korea Brain Research Institute (KBRI, Daegu, South Korea) and conducted in

accordance with institutional ethical guidelines. Adult male mice (10 weeks old, >25 g) were used for electrophysiological recordings.

Mice were anesthetized via intraperitoneal injection of 0.5% urethane (400 mg/kg body weight) and secured in an automated stereotaxic frame (71000, RWD Life Science, USA). The reference wire from the fPCB interface was connected to the cerebellum via a skull-mounted stainless-steel screw.

A craniotomy was performed at stereotaxic coordinates targeting the hippocampal CA1 region (ML: ± 1.5 mm; AP: -1.75 mm; DV: -1.75 mm), and the dura mater was carefully removed. Each probe shank was slowly inserted into the CA1 region using a motorized micromanipulator.

Neural activity was recorded with the Intan RHS system (RHS Stim/Recording Controller and 16-channel headstage, Intan Technologies, USA). Spiking signals were monitored for at least 5 min per insertion. After each recording session, electrodes were carefully retracted to minimize tissue damage.

Signals were digitized using RHX acquisition software (sampling rate: 20 kS/s/channel; band-pass filter: 0.3–7.5 kHz; notch filter: 60 Hz; 16-bit ADC). Postprocessing and spike sorting were performed using a custom MATLAB algorithm, as described previously.¹⁹

2.7. Statistical Analysis. Independent-sample statistical analyses were performed using GraphPad Prism (GraphPad, USA). The statistical test applied to each data set is specified in the corresponding figure legend.

2.8. Data Availability. All data supporting the findings of this study are included in the main text and [Supporting Information](#) files or are available from the corresponding author upon reasonable request.

3. RESULTS AND DISCUSSION

3.1. Preparation of Customizable 3D Neural Microelectrodes. To systematically evaluate how electrode geometry influences neural signal acquisition, we developed two distinct 3D microelectrode architectures using our optimized fPCB fabrication process.^{34,43} Notably, the use of PSR ink instead of the conventional PI film as the upper insulation enabled the patterning of fine electrode openings as small as ~ 30 μm within a commercial fPCB workflow. This platform enables the rapid and cost-effective production of both recessed (baseline) and protruded electrode designs ([Table S1](#)).

The recessed microelectrode was fabricated as a standard benchmark architecture, fully compatible with conventional, mass-producible fPCB manufacturing. Using the electroless nickel immersion gold (ENIG) process, a ~ 0.1 μm -thick gold layer was chemically deposited onto the electrode site ([Figures S1a and S2](#)). While this method is widely adopted and reliable, it inherently restricts the formation of high-aspect-ratio features, resulting in a low-profile recessed geometry. Despite these limitations, recessed electrodes serve as a practical control for evaluating the benefits of protruded designs.

In contrast, the protruded microelectrode was realized by electroplating Sn onto the exposed Cu traces within the same fPCB process. This produced a hemispherical bump that extended above the surrounding PSR insulation, thereby creating a true 3D electrode ([Figures S1b and S3](#)). This architecture was deliberately designed based on the hypothesis that protrusion improves physical coupling with neural tissue

and reduces the electrode–neuron gap, thereby enhancing recording sensitivity and efficiency.

To ensure that differences in signal quality were attributable to geometry rather than material properties, both recessed and protruded electrodes were further modified with sequential electrochemical coatings of Pt black and the conductive polymer PEDOT:PSS. These coatings improved electrochemical performance—lowering impedance and increasing charge transfer capacity—while keeping the surface chemistry consistent across designs.

Beyond enabling systematic comparison of electrode geometries, our fPCB-based fabrication approach provides several practical advantages: it is inherently scalable, cost-efficient, and compatible with industrial PCB manufacturing pipelines. Batch fabrication ensures high reproducibility, while integrated connector pads on the flexible substrate allow direct interfacing with standard FPC connectors ([Figure S4](#)). This plug-and-play configuration eliminates the need for labor-intensive wire bonding, streamlining device deployment in experimental settings ([Figure S5](#)).

Collectively, these innovations establish a customizable and accessible 3D neural microelectrode platform that combines industrial scalability with enhanced recording performance.

3.2. Morphological Characterization of 3D Neural Microelectrodes. The distinct geometries and surface modifications of the fabricated neural microelectrodes were systematically examined using scanning electron microscopy (SEM).

For the recessed design, the 30 μm -diameter recording site was flush with the surrounding PSR insulation, yielding a completely flat surface profile ([Figure 2a](#)). Even after sequential functionalization with Pt black and PEDOT, the recessed gold electrode retained its sunken geometry with no protrusion above the insulation ([Figure 2b](#)). These findings confirm that surface modification alone cannot overcome the intrinsic geometric limitation of recessed electrodes, namely, the relatively large electrode–tissue gap that constrains recording fidelity.

In contrast, the Sn-based protruded electrode exhibited a well-defined three-dimensional hemispherical profile that extended prominently above the PSR insulation layer. This elevated geometry provided a stable scaffold for surface engineering. Electrochemical deposition of Pt black generated a porous, sponge-like morphology that substantially increased the effective surface area ([Figure 2c](#)). Subsequent PEDOT coating formed a uniform polymeric film over the porous substrate, further enhancing electrochemical performance and tissue integration ([Figure 2d](#)). Unlike in the recessed design, these modifications were applied to an already elevated 3D surface, thereby amplifying both the geometric advantage and the electrochemical interface.

To quantify the changes in surface topography, atomic force microscopy (AFM) was employed ([Figure 2e,f](#)). The root-mean-square (R_q) roughness of the bare Au surface was measured to be 7.61 nm, which increased to 33.56 and 65.39 nm after Pt black and PEDOT deposition, respectively ([Figure S6a](#)). A similar trend was observed for the Sn-based electrodes, with the R_q value increasing from 5.75 nm (bare) to 26.4 nm (Pt black) and 10.1 nm (PEDOT) ([Figure S6b](#)). These results demonstrate that the surface modifications create a rougher topography. This increased surface roughness leads to a larger effective surface area, which is consistent with the lower impedance values observed in the EIS data.

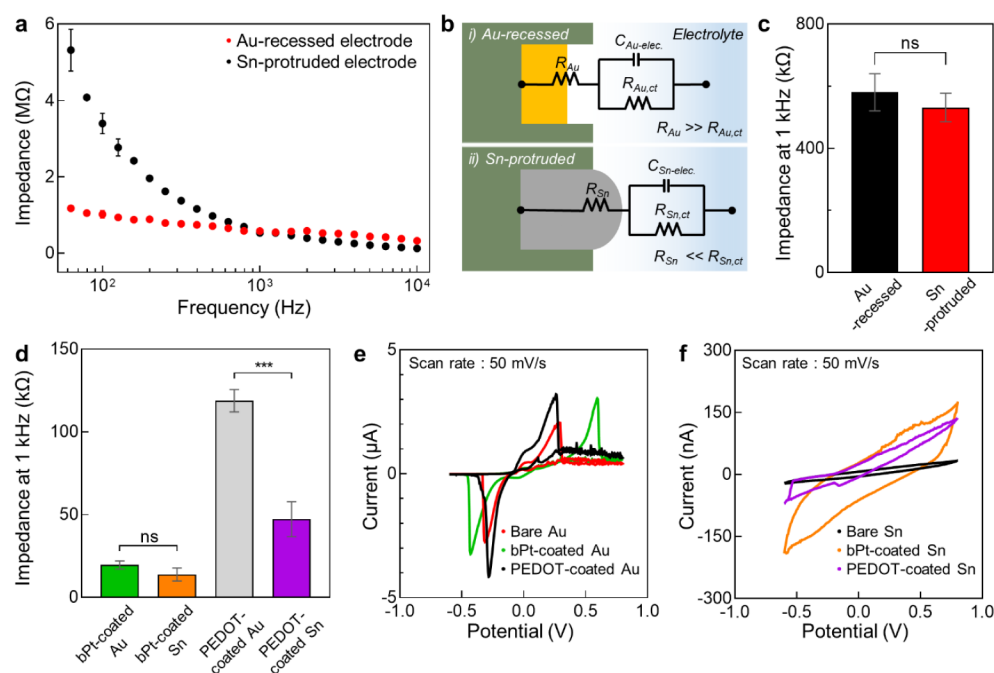


Figure 3. Electrochemical performance of recessed and protruded microelectrodes with and without surface modification. (a) Electrochemical impedance spectra in phosphate-buffered saline ($n = 3$). (b) Equivalent circuit models; the protruded geometry increases effective double-layer capacitance and reduces access resistance, enhancing charge transfer. (c) Impedance at 1 kHz for recessed and protruded electrodes ($n = 3$). (d) Impedance at 1 kHz for electrodes coated with Pt black (bPt) or PEDOT:PSS ($n = 3$). (e,f) Cyclic voltammetry (CV) curves of bare and surface-modified Au and Sn electrodes. Each bar in (c) and (d) shows the mean $\pm$ standard deviation. *** $p < 0.001$. ns $p > 0.05$, not significant. Statistical analysis was performed using a two-tailed unpaired Student's t test. (c) $t(4) = 1.133$, $p = 0.3207$; (d) $t(4) = 2.166$, $p = 0.1093$ (bPt-coated sample), $t(4) = 9.881$, $p = 0.0006$ (PEDOT:PSS coated sample).

Taken together, these morphological analyses demonstrate that the protruded electrode design uniquely integrates geometric protrusion with surface-level functionalization, creating a significantly more favorable electrode–tissue interface. This dual advantage directly underpins the superior recording performance observed in vivo.

3.3. Electrochemical Characterization of 3D Neural Microelectrodes. The electrochemical performance of the fabricated electrodes was systematically evaluated using EIS in 1× PBS. The bare Au-recessed and Sn-protruded electrodes exhibited distinct frequency-dependent impedance behaviors. At higher frequencies, the Au-recessed electrode showed greater impedance, whereas the Sn-protruded electrode displayed lower impedance, attributable to its reduced surface resistance. At lower frequencies, however, the impedance of the Sn-protruded electrode surpassed that of the Au-recessed electrode, resulting in a frequency-dependent impedance crossover (Figure 3a).

This crossover behavior reflects the influence of electrode geometry on electrochemical properties. At high frequencies, surface resistance (R) dominates; the Sn-protruded electrode exhibits lower resistance due to improved electrolyte access enabled by its 3D structure. At low frequencies, the double-layer capacitance (C) becomes the controlling factor, and the larger effective surface area of Pt black- or PEDOT-coated Sn-protruded electrodes supports higher capacitance relative to recessed electrodes (Figure 3b).

At 1 kHz, which is a standard benchmark for neural electrode performance, the Sn-protruded electrode demonstrated lower impedance (531.2 ± 45.9 kΩ) compared to the Au-recessed electrode (580.5 ± 59.6 kΩ), underscoring the

intrinsic electrical advantage of the protruded geometry (Figure 3c).

Following surface modification, these differences became more pronounced. After Pt black deposition, the Sn-protruded electrode exhibited significantly lower impedance (13.8 ± 3.9 kΩ) than the Au-recessed counterpart (19.6 ± 2.4 kΩ) at 1 kHz. A similar trend was observed with PEDOT coating, where the Sn-protruded electrode showed lower impedance (47.3 ± 10.5 kΩ) compared to the coated Au-recessed electrode (118.8 ± 6.77 kΩ) (Figure 3d).

These results establish that electrode geometry, specifically protrusion, has a greater influence on impedance performance than surface chemistry alone, particularly within the frequency range relevant to neural recording.

Cyclic voltammetry (CV) further validated these findings. The bare Au-recessed electrode displayed characteristic redox peaks, which increased markedly after Pt black and PEDOT modification, indicating enhanced charge storage capacity (CSC) from increased surface area (Figure 3e). By contrast, the uncoated Sn-protruded electrode showed minimal faradaic currents in the nA range, despite its low impedance. X-ray photoelectron spectroscopy (XPS) analysis confirmed this effect originated from the formation of a native SnO_2 layer, highlighting the necessity of surface modification for effective charge transfer (Figure S7).

After Pt black and PEDOT deposition, the Sn-protruded electrode exhibited substantially enlarged CV areas, confirming synergistic CSC enhancement through the combination of geometry and surface modification (Figure 3f).

Mechanical durability was evaluated by repeatedly inserting Pt black-coated Sn-protruded electrodes into 0.9% agarose gel, a brain tissue surrogate (Figure S8a). Impedance was

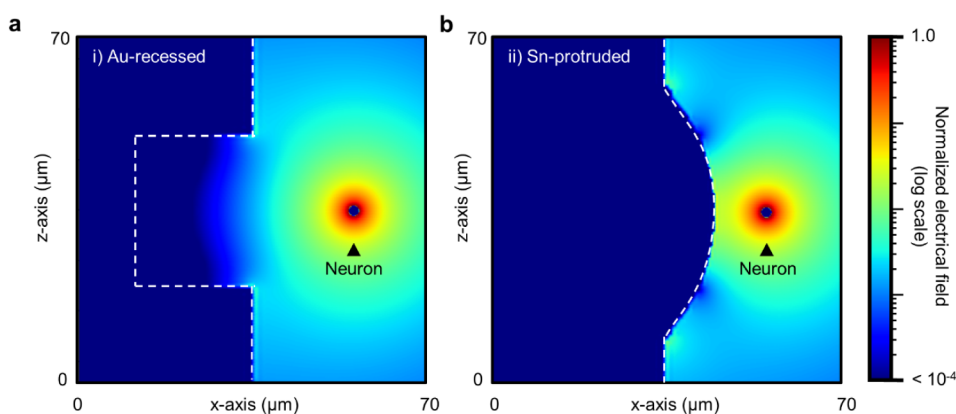


Figure 4. Finite element modeling of extracellular potential fields around recessed and protruded electrodes. (a) Simulated potential distribution around an Au-recessed electrode. (b) Simulated potential distribution around a Sn-protruded electrode. Color maps represent normalized extracellular potentials generated by a point-source axon near each electrode.

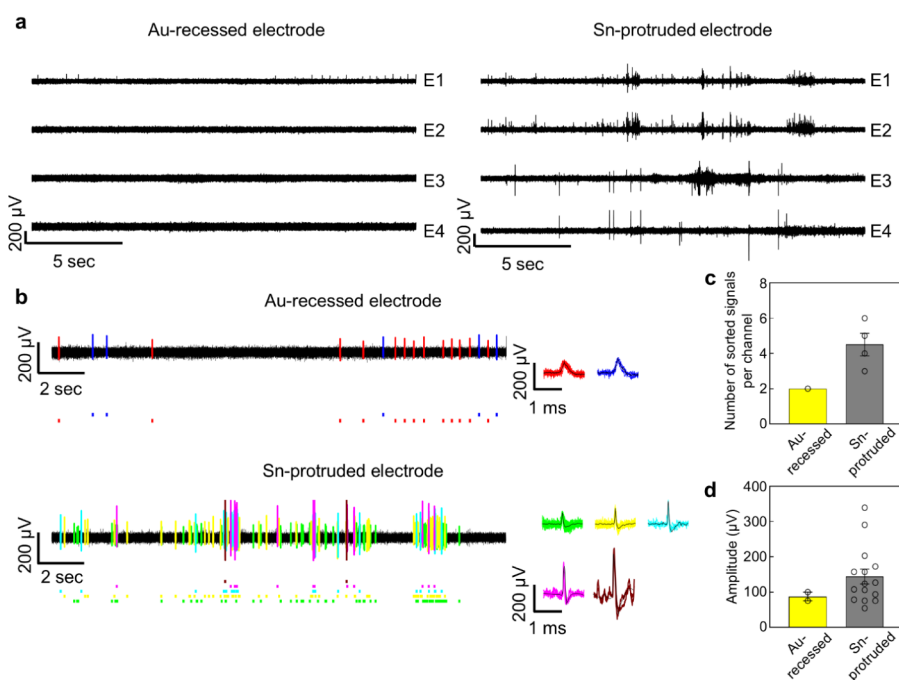


Figure 5. Neural signal acquisition using Au-recessed and Sn-protruded electrodes in vivo. (a) Representative recordings from four channels (E1–E4) in the CA1 region. (b) Spike sorting results with raster plots and color-coded waveforms. (c) Number of sorted units per channel. $n = 1$ (Au), $n = 4$ (Sn). (d) Comparison of spike amplitudes. $n = 2$ (Au), $n = 15$ (Sn). Each bar in (c) and (d) shows the mean $\pm$ standard deviation; dots represent individual channels.

measured after every 10 insertions, up to 50 cycles. No significant changes in impedance or morphological degradation were observed, confirming both the mechanical robustness and long-term electrical stability of the modified electrodes (Figure S8b).

To evaluate the flexibility of the probe, we measured electrode impedance after bending (Figure S9a). The impedance remained stable across 0° , 45° , and 60° bends, measuring $16.25 \pm 4.11 \text{ k}\Omega$, $17.00 \pm 7.44 \text{ k}\Omega$, and $18.75 \pm 2.22 \text{ k}\Omega$, respectively (Figure S9b). In a durability test with 50 bending cycles ($0^\circ \leftrightarrow 60^\circ$), impedance changed from $17.75 \pm 3.78 \text{ k}\Omega$ to $21.5 \pm 1.00 \text{ k}\Omega$, showing no statistically significant degradation (Figure S10). These results confirm that the probe maintains its electrical performance under mechanical deformation, validating its flexibility.

Finally, finite element analysis (FEA) was used to simulate the electrical potential fields generated by a representative neural source. Compared with recessed electrodes, the protruded design extended the conductive surface into regions of higher potential, thereby improving coupling with local extracellular fields (Figure 4). These simulations support the experimental data by demonstrating that protruded geometries facilitate stronger neural signal acquisition, consistent with the higher spike amplitudes and signal-to-noise ratios (SNRs) observed in vivo.

3.4. Biocompatibility Test of fPCB-Based Neural Electrodes. To evaluate potential cytotoxicity, a biocompatibility assessment was conducted using a standard MTT assay in accordance with ISO 10993 guidelines.⁴² Extracts from sterilized fPCB-based electrodes were applied to HepG2 cell cultures at varying concentrations (12.5%, 25%, 50%, 75%, and

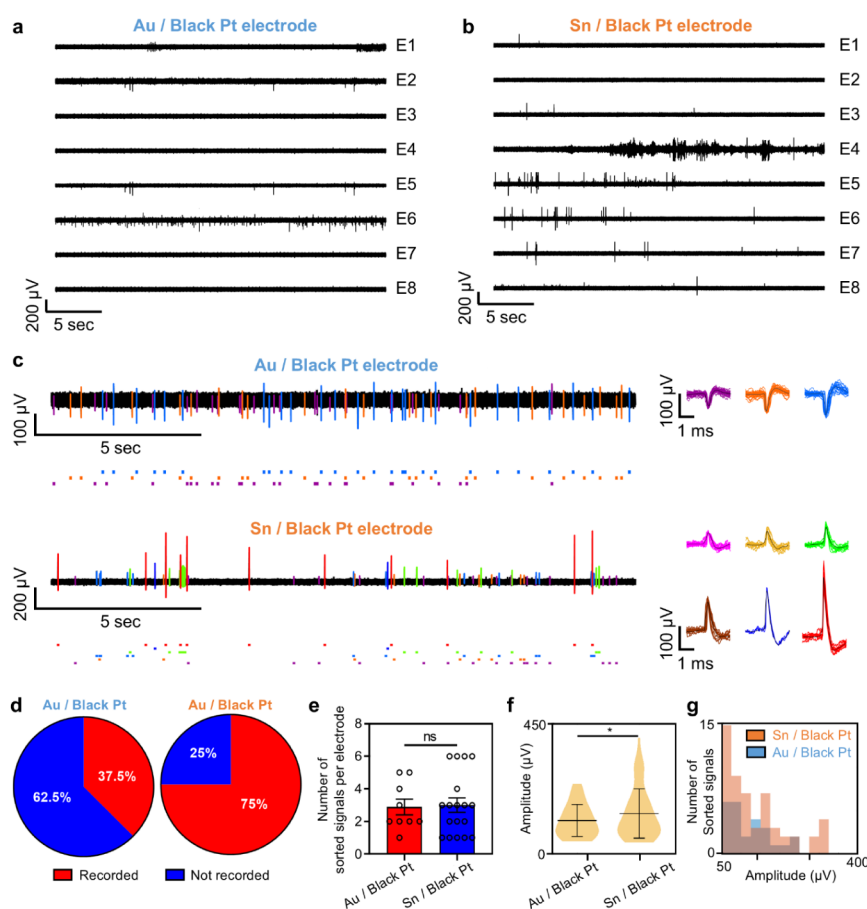


Figure 6. In vivo recording performance of Au-recessed and Sn-protruded electrodes coated with black Pt. (a,b) Representative raw neural signals from (a) Au-recessed and (b) Sn-protruded electrodes coated with black Pt. (c) Spike sorting results with raster plots and color-coded waveforms. (d) Pie charts showing the proportion of electrodes with successful signal detection (red) versus nonrecording electrodes (blue). (e) Number of sorted spikes per electrode. $t(25) = 0.1552$, $p = 0.8779$. $n = 9$ (Au/Black Pt), $n = 18$ (Sn/Black Pt). (f) Comparison of spike amplitudes. Although the means were not significantly different (unpaired t -test, $t(78) = 1.306$, $p = 0.1954$), the variance between the two groups was significantly different (F -test, $F(54, 24) = 2.399$, $p = 0.0214$). $n = 25$ (Au/Black Pt), $n = 55$ (Sn/Black Pt). (g) Histogram of spike amplitude distribution. $n = 25$ (Au/Black Pt), $n = 55$ (Sn/Black Pt). Each bar in (e) and (f) shows the mean $\pm$ standard deviation. ns, not significant. $*p < 0.05$. Statistical analysis was performed using a two-tailed unpaired Student's t test.

100% v/v). Culture medium served as the negative control, while Triton X-100 was used as the positive control (Figure S11a).

Cell viability remained above 80% across all extract concentrations, with no statistically significant differences compared to the negative control group ($p > 0.05$). In contrast, cells treated with Triton X-100 exhibited nearly complete death, confirming the sensitivity and validity of the assay (Figure S11b).

Importantly, even at the highest extract concentration (100%), the fPCB-based electrodes maintained cell viability of $100\% \pm 3.2\%$, indicating negligible cytotoxicity. These results confirm the excellent biocompatibility of the fabricated electrodes, supporting their suitability for long-term implantation and in vivo neural interfacing.

3.5. In Vivo Neural Recording from Bare 3D Neural Microelectrodes. To assess functional performance before surface modification, acute in vivo recordings were performed in the hippocampal CA1 region of anesthetized mice using bare (uncoated) Au-recessed and Sn-protruded electrodes. This experiment isolated the effect of electrode geometry on neural signal acquisition (Figure Sa,b).

Representative traces revealed that only one of the four Au-recessed electrodes detected discernible neural activity, whereas all four Sn-protruded electrodes captured robust spike signals (Figure 5c). The protruded architecture appeared to facilitate more intimate contact with surrounding neurons, thereby minimizing the electrode–neuron gap and enhancing electrical coupling.

Spike detection and sorting further highlighted these differences. As shown in Figure 5b, Sn-protruded electrodes recorded higher spike amplitudes and more frequent events compared to recessed electrodes, which exhibited sparse or negligible spiking (Figure 5b). Sorted waveforms revealed multiple distinct clusters from Sn-protruded channels, demonstrating their ability to resolve single-unit activity with improved SNR (Figure 5d). Notably, several Sn channels isolated multiple well-defined neuronal units, underscoring the superior fidelity of the 3D design.

In summary, bare Sn-protruded electrodes consistently outperformed recessed electrodes, capturing more abundant and reliable neural activity even without surface coatings. These findings establish 3D geometry as a decisive factor in enabling high-quality acute recordings.

3.6. In Vivo Neural Recording from Surface-Modified 3D Microelectrodes. To further enhance recording performance, in vivo experiments were conducted with surface-modified electrodes. Recordings from the hippocampal CA1 region compared Au/black Pt and Sn/black Pt electrodes, thereby combining geometric and surface-level optimization (Figure 6a,b, Figure S12).

The Sn/black Pt electrodes exhibited significantly stronger spiking activity across multiple channels, while Au/black Pt electrodes showed sparse or negligible spike signals (Figure 6c). These results suggest that protruded geometry provides a substantial advantage in neuronal interfacing, which is further amplified by the high-surface-area Pt black coating.

Spike sorting confirmed the superior performance of Sn/black Pt electrodes. A greater diversity of spike waveforms was observed, with 75% of Sn electrodes successfully recording sorted units, compared to only 37.5% of Au electrodes (Figure 6d,e). Violin plots of spike amplitudes demonstrated significantly larger signals from Sn/black Pt electrodes, consistent with higher SNR (Figure 6f). Histogram analyses further supported this observation by showing a broader distribution of large-amplitude spikes (Figure 6g).

Collectively, these results demonstrate the synergy between protruded geometry and conductive surface modification. Sn/black Pt electrodes consistently achieved higher spike detection probability, stronger amplitudes, and improved fidelity, establishing them as a mechanically and electrically optimized platform for acute neural recordings. These findings highlight the promise of this approach for advancing high-performance brain–machine interface technologies.

4. CONCLUSION

In this study, we introduced a scalable and customizable neural interface platform based on fPCB technology. This industrially compatible fabrication method enables rapid and cost-effective production of high-density, multilayered neural probes, overcoming key limitations of conventional silicon-based and cleanroom-fabricated flexible devices. Leveraging this platform, we systematically investigated the impact of electrode geometry and surface modification on neural signal acquisition.

We fabricated and evaluated both recessed gold and protruded tin microelectrodes, with and without high-surface-area coatings such as platinum black and PEDOT:PSS. Mechanical durability tests and finite element simulations confirmed the robustness and functional advantages of the 3D protruded architecture. In vivo electrophysiological recordings in the mouse hippocampus demonstrated that protruded electrodes consistently outperformed recessed counterparts, achieving higher spike amplitudes, signal-to-noise ratios (SNR), and unit yield, even under identical surface treatment conditions. These results provide compelling evidence that electrode geometry is a primary determinant of recording fidelity, independent of surface engineering alone.

However, the fPCB electrodes used for neural signal measurement in this study—despite their advantages in mass production and low cost through geometric and surface modifications—have a major limitation in terms of low electrode density. This limitation has been partially addressed in our recent work, where the number of electrodes per shank was increased by enhancing the resolution of lithographic and wet etching processes.⁴⁴ Yet, it is important to note that such improvements inevitably introduce trade-offs in fabrication yield, which should be carefully balanced in future scalable

designs. Also, in this study, a 20 μm thick PSR layer was chosen as the upper insulation to balance process reliability and electrode recessed depth, enabling Sn electroplating in a well-confined area. While a thinner PSR insulation layer (e.g., $\sim 10\ \mu\text{m}$) could further improve the recessed geometry and electroplating efficiency, such modifications are contingent upon the ink formulation and the process capabilities of the FPCB foundry. Future work may explore optimization of solvent composition or resin properties to reduce the insulation thickness while maintaining pattern fidelity and yield.

By decoupling structural and surface effects, this study provides clear experimental evidence for the synergistic benefits of combining 3D protrusion with surface modification. Overall, our work establishes a practical, scalable, and high-performance platform for neural interfaces, offering a versatile approach optimized for high-fidelity recordings. We anticipate that this platform will enable broader adoption of customizable, high-performance neural electrodes across basic neuroscience, neuroengineering, and clinical applications.

■ ASSOCIATED CONTENT

Supporting Information

The Supporting Information is available free of charge at <https://pubs.acs.org/doi/10.1021/acsaelm.5c02055>.

Structural differences between Au-recessed electrodes and Sn-protruded electrodes; fabrication process of the flexible PCB-based neural probe with the recessed structure by Ni/Au electroless plating; fabrication process of the flexible PCB-based neural probe with a 3D protruded structure by Sn electroplating; mass production capability of fPCB for cost-effective batch fabrication; schematic of the scalable fPCB-based microelectrode platform; root mean square (Rq) roughness values of AFM analysis data; XPS spectra of the surface of the Sn-protruded electrode; insertion test of the fPCB-based neural microelectrode into tissue-mimic hydrogel to confirm mechanical durability; evaluation of electrode impedance under probe bending; impedance stability of the fPCB probe under repeated bending; MTT assay to assess the biocompatibility of the Sn-protruded microelectrode; in vivo neural signal recording performance of black Pt-coated Au-recessed electrodes and black Pt-coated Sn-protruded electrodes; comparison of electrode performance between Au-recessed electrodes and Sn-protrude probe (PDF)

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Author Contributions

*M.K. and G.K. contributed equally to this work. M.K., G.K., and H.S. performed most of the experiments, analyzed the data, prepared the figures, and wrote the manuscript. S.L. was involved in electrode modifications. H.L., S.P., and J.L. were involved in preparing the in vivo experiment. N.C. and H.S. discussed the results, provided comments, and wrote the manuscript. All of the authors reviewed the manuscript.

Notes

The authors declare no competing financial interest.

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