

High-gain and broadband in vivo amplifiers enabled by matched materials

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In-sensor biosignal amplification could enable superior signal quality, lower data transmission redundancy, and enhanced device integration. Organic electrochemical transistor (OECT)-based circuits offer an attractive candidate due to their low-voltage operation, high transduction efficiency, and seamless interfacing with biological tissues. However, the fundamental design principles governing high-performance OECT-based amplifiers, particularly the coupling between material design and circuit-level performance, remain insufficiently understood. Here, we present a matched n-p material system to elucidate the design principles and in-sensor amplification capability of complementary OECT circuits. Specifically, we develop a high-performance, low-operation-voltage n-type polymer, P(IgFTDPP-2FT), and integrate it with a complementary p-type DPP polymer. The resulting inverter-based amplifier can operate at low voltage while demonstrating a high gain exceeding 800 V V^{-1} , a bandwidth above 600 Hz at 0.6 V, and an ultralow power consumption down to 0.57 nW at 0.2 V. We further demonstrate in vivo amplification of mouse electrocorticographic (ECoG) signal with good biocompatibility and superior signal quality. This work establishes a general material–device–circuit framework and design guidelines for OECT-based amplifiers.

Advances in biomedical monitoring, especially for detecting faint physiological signals such as neural activity and cardiac rhythms^{1–5}, necessitate biosensing platforms that combine high-gain signal amplifiers with low power consumption and uncompromised biocompatibility^{6,7}. However, current state-of-the-art silicon-based electronics, including conventional complementary metal-oxide

semiconductor (CMOS) technologies, face inherent limitations such as high mechanical rigidity and poor interfacing ability with soft, dynamic biological tissues^{8–10}. In contrast, organic electrochemical transistors (OECTs) have emerged as a promising alternative, offering excellent biocompatibility, high transconductance (g_m), low operating voltage (<0.8 V), and the ability to operate in aqueous environments—

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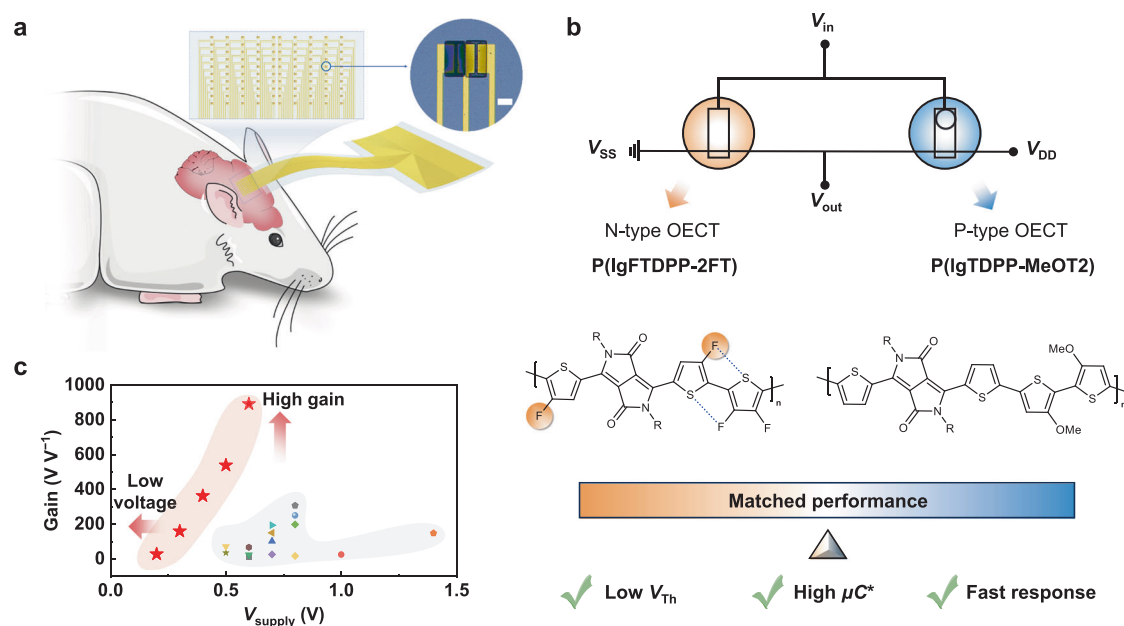


Fig. 1 | Schematic illustration of materials, devices, and applications for the in vivo signal amplifiers with ultralow power and high gain. **a** Schematic illustration of a flexible OECT inverter-based biosignal amplifier array for in vivo electrocorticographic (ECoG) signal recording, with an insert photo showing the detailed architecture of a single amplifier (scale bar: 50 μm). **b** Circuit diagram of

an OECT inverter-based amplifier enabled by *n*-type polymer P(lgFTDPP-2FT), achieving high μC^* , low V_{Th} , and fast response, matching with its *p*-type counterpart P(lgTDPP-MeOT2). **c** Comparison of the gain versus supply voltage of our amplifier with inverters from literature^{31,32,43–55}.

ideal for real-time, in vivo, high-sensitivity physiological signal monitoring^{7,11–13}. While individual OECT devices have demonstrated effectiveness for biosignal recording¹⁴, complementary OECT inverters are more promising for advanced and practical biosignal processing, providing high signal-to-noise ratios (SNRs) and potentially low power consumption¹⁵. Additionally, their voltage-output features facilitate seamless integration with backend data acquisition and digital processing systems, further supporting highly integrated bioelectronics¹⁶.

Achieving high-performance complementary amplifiers for reliable biosignal monitoring requires matched *p*- and *n*-type OECTs with balanced electrical and dynamic characteristics¹⁷. Current research efforts, however, predominantly focus on individual material optimization, overlooking integrated optimization. This has resulted in a significant performance disparity between *p*- and *n*-type materials^{18,19}. While state-of-the-art *p*-type OECTs have achieved exceptional performance with a figure of merit (μC^*) surpassing $500 \text{ F cm}^{-1} \text{ V}^{-1} \text{ s}^{-1}$, low threshold voltage (V_{Th}), and fast response time^{20–22}. *N*-type counterparts remain lagging behind²³. Although a few *n*-type OECT materials exhibited high μC^* over $150 \text{ F cm}^{-1} \text{ V}^{-1} \text{ s}^{-1}$, they always show high $V_{\text{Th}} > 0.5 \text{ V}$ ^{24,25}, which could result in non-ideal transistor behaviors and probably overestimated μC^* values^{26,27}. We believe single-parameter enhancement, such as μC^* , cannot reflect the practical application of OECT materials²⁸. Therefore, such imbalanced development and evaluation method severely constrains the performance of OECT-based amplifiers for biosignal recording applications in three key aspects: i) shifted transition voltage and reduced noise margins arising from current mismatch, increasing susceptibility to logic errors and potential failures; ii) increased operation voltage and power consumption due to higher *n*-type V_{Th} ; and iii) bandwidth limitations imposed by slower *n*-type response, leading to signal distortion at high frequency. Although device architecture engineering can partially alleviate these issues, these strategies often involve trade-offs in response frequency, power consumption, and fabrication complexity, which limit their practical deployment¹⁷. Ambipolar materials offer an alternative²⁹, but still suffer from high V_{Th} , unbalanced performance with non-real-to-real voltage output³⁰. Therefore, closing the

performance gap between *n*- and *p*-type OECT materials through molecular engineering is still challenging³¹. More importantly, the limits and design principles of complementary OECT-based amplifiers, particularly the interplay between material matching and device operation, have not been fully explored so far.

Here, we report a matched n-p material system that enables high-performance OECT-based amplifiers for in vivo biosignal amplification (Fig. 1a). We further establish general design guidelines for realizing high-performance OECT amplifiers, based on coordinated optimization of material properties, device operating regime, and circuit design. Specifically, a simple fluorination strategy was employed, yielding novel fluorinated thiophene-flanked diketopyrrolopyrrole (FTDPP) to the best of our knowledge and a high-performance *n*-type OECT polymer P(lgFTDPP-2FT). P(lgFTDPP-2FT) demonstrated more balanced performance metrics: a high μC^* of $186.9 \text{ F cm}^{-1} \text{ V}^{-1} \text{ s}^{-1}$, a low V_{Th} of 0.32 V, and fast response speeds of $\tau_{\text{on}}/\tau_{\text{off}} = 319/56 \mu\text{s}$. Its reduced V_{Th} facilitates operation within the moderate regime under low supply voltage. When pairing with a matched *p*-type material, P(lgTDPP-MeOT2), and optimizing the device configuration (Fig. 1b), we achieved a high gain of over 800 V V^{-1} at $V_{\text{DD}} = 0.6 \text{ V}$ (Fig. 1c) and an ultralow power consumption down to 0.57 nW at 0.2 V. These amplifiers can achieve a bandwidth of over 600 Hz, ranking among the fastest OECT-based amplifiers. We also demonstrate their application in implantable biosignal acquisition and potentially high-density integration, establishing a general material–device–circuit framework and design guidelines for OECT-based amplifiers, and offering a new approach for next-generation, high-fidelity biosignal recording.

Results

Polymer design and synthesis

Thienyl diketopyrrolopyrrole (TDPP) was employed as the building block due to its extended π -conjugated backbone and strong inter-molecular interactions. While our previously developed *n*-type polymer, P(lgTDPP-2FT), exhibited high performance³², it suffers from both suboptimal performance and high operational voltages. To address these issues, we adopted a simple yet effective strategy^{33,34} by

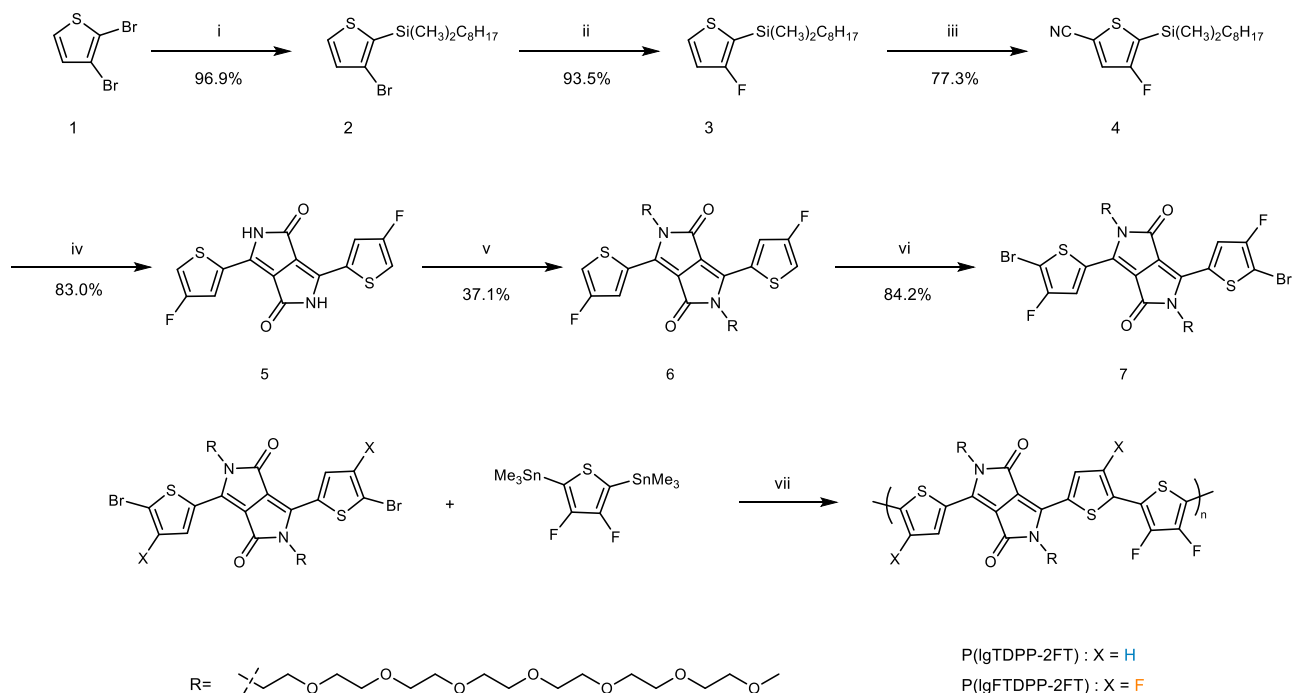


Fig. 2 | Synthetic route to the *n*-type OECT polymers used in this study. (i) EtMgCl, $\text{C}_8\text{H}_{17}(\text{Me})_2\text{SiCl}$, THF, room temperature (R.T.). (ii) *tert*-BuLi, *N*-fluorobenzenesulfonimide (NFSI), diethyl ether, -84°C . (iii) *n*-BuLi, phenyl cyanate,

diethyl ether, -78°C to 0°C . (iv) Na, *tert*-amyl alcohol, FeCl_3 , diisopropyl succinate, 65°C . (v) RBr, K_2CO_3 , KI, DMF, 120°C . (vi) *N*-bromosuccinimide (NBS), dichloromethane (DCM), 0°C to R.T. (vii) $\text{Pd}(\text{PPh}_3)_4$, CuI, CB/DMF, 130°C .

fluorination of the thiophene moiety on the TDPP core, yielding FTDPP. Subsequent introduction of ethylene glycol (EG) side chains and Stille polymerization with 2FT produced the polymer P(lgFTDPP-2FT). The small atomic radius of F facilitates noncovalent F $\cdots$ H and F $\cdots$ S interactions, which can enhance backbone planarity and further improve charge carrier mobility³⁵.

Unlike the conventional synthetic route³⁶, we employed a Grignard reaction in the first step to avoid the use of *n*-butyllithium, thereby greatly reducing side reactions, improving overall yield, and simplifying purification (Fig. 2). Given the known challenges of electrophilic fluorination^{37,38}, we optimized the second step by minimizing THF volume to suppress undesired hydrogen exchange between *tert*-butyllithium and the thiophene α -position, increasing the yield from 50–90%. Both polymers were obtained via Pd-catalyzed Stille polymerization reactions using CuI as the co-catalyst, followed by purification through Soxhlet extraction and finally collected by dichloromethane or chloroform (CF). Their molecular weights were evaluated by gel permeation chromatography (GPC) using CF as the eluent. Both polymers show very similar molecular weights of $M_w/M_n = 32.5/16.3$ kDa for P(lgTDPP-2FT) and $M_w/M_n = 33.9/16.1$ kDa for P(lgFTDPP-2FT) (Supplementary Fig. 1). Both polymers exhibit good thermal stability with high decomposition temperatures (Supplementary Figs. 2, 3).

To elucidate the structure-property relationships, we conducted systematic characterizations of the materials³⁰. Ultraviolet-Visible-Near-Infrared (UV-Vis-NIR) absorption spectroscopy revealed that P(lgFTDPP-2FT) exhibits minimal red shift upon transitioning from solution to solid state, in contrast to the much stronger red-shift absorption observed for P(lgTDPP-2FT) (Fig. 3a). This disparity suggests enhanced backbone planarity and better conformational stability of P(lgFTDPP-2FT). The higher O–O–O–1 vibrational absorption ratio of P(lgFTDPP-2FT) further corroborates its more rigid backbone. These results were further supported by relaxed potential energy surface (PES) scans of the dihedral angles (Fig. 3b, c, Supplementary Figs. 4–6). Both polymers show similar rotational barriers in neutral and charged states. However, P(lgFTDPP-2FT) shows a single dominant stable

conformation with a steeper change in energy as the dihedral angles, which can be attributed to the F-induced noncovalent interactions (Supplementary Fig. 7). Such features could reduce the conformational disorder in the solid state³⁹.

Electrochemical measurements revealed that fluorination lowered the energy levels, particularly the LUMO. The electron affinity (EA) was substantially deepened from -3.83 eV for P(lgTDPP-2FT) to -4.14 eV for P(lgFTDPP-2FT), while the ionization potential (IP) also shifted from -5.19 eV to -5.42 eV (Fig. 3d, Supplementary Table 1). Consequently, the electrochemical band gap was slightly narrowed from 1.36 eV to 1.28 eV. These results indicate that the introduction of fluorine atoms into TDPP can lower the LUMO energy level, which is consistent with the results of DFT calculations. Continuous CV measurements of P(lgFTDPP-2FT) in 0.1 M NaCl aqueous solution demonstrated its better electrochemical stability upon cycling (Supplementary Fig. 8). Consistent with the CV results, spectroelectrochemistry revealed that P(lgFTDPP-2FT) undergoes absorption changes from neutral to polaron/bipolaron states at -0.3 V, earlier than P(lgTDPP-2FT) (-0.6 V) (Fig. 3e, Supplementary Fig. 9). Grazing incidence wide-angle X-ray scattering (GIWAXS) and atomic force microscope (AFM) were employed to explore the molecular packing and morphology. In pristine films, both polymers adopt mixed face-on and edge-on packing, with P(lgFTDPP-2FT) exhibiting a stronger edge-on preference (Fig. 3f–i, Supplementary Fig. 10 and Supplementary Table 2). The edge-on arrangement facilitates efficient ion penetration, thereby contributing to a fast electrochemical response. Compared to P(lgTDPP-2FT), P(lgFTDPP-2FT) displayed more pronounced (300) and (010) peaks, indicative of enhanced crystallinity and structural order. Upon exposure or reduction, P(lgFTDPP-2FT) consistently retained its edge-on dominant orientation, whereas P(lgTDPP-2FT) transitioned from a mixed orientation toward a more edge-on arrangement. These results highlight the better morphological stability of P(lgFTDPP-2FT), which likely stems from its enhanced conformational stability and further supports the realization of high charge-carrier mobility, as discussed later. AFM images reveal that both polymers exhibited porous morphologies (Supplementary

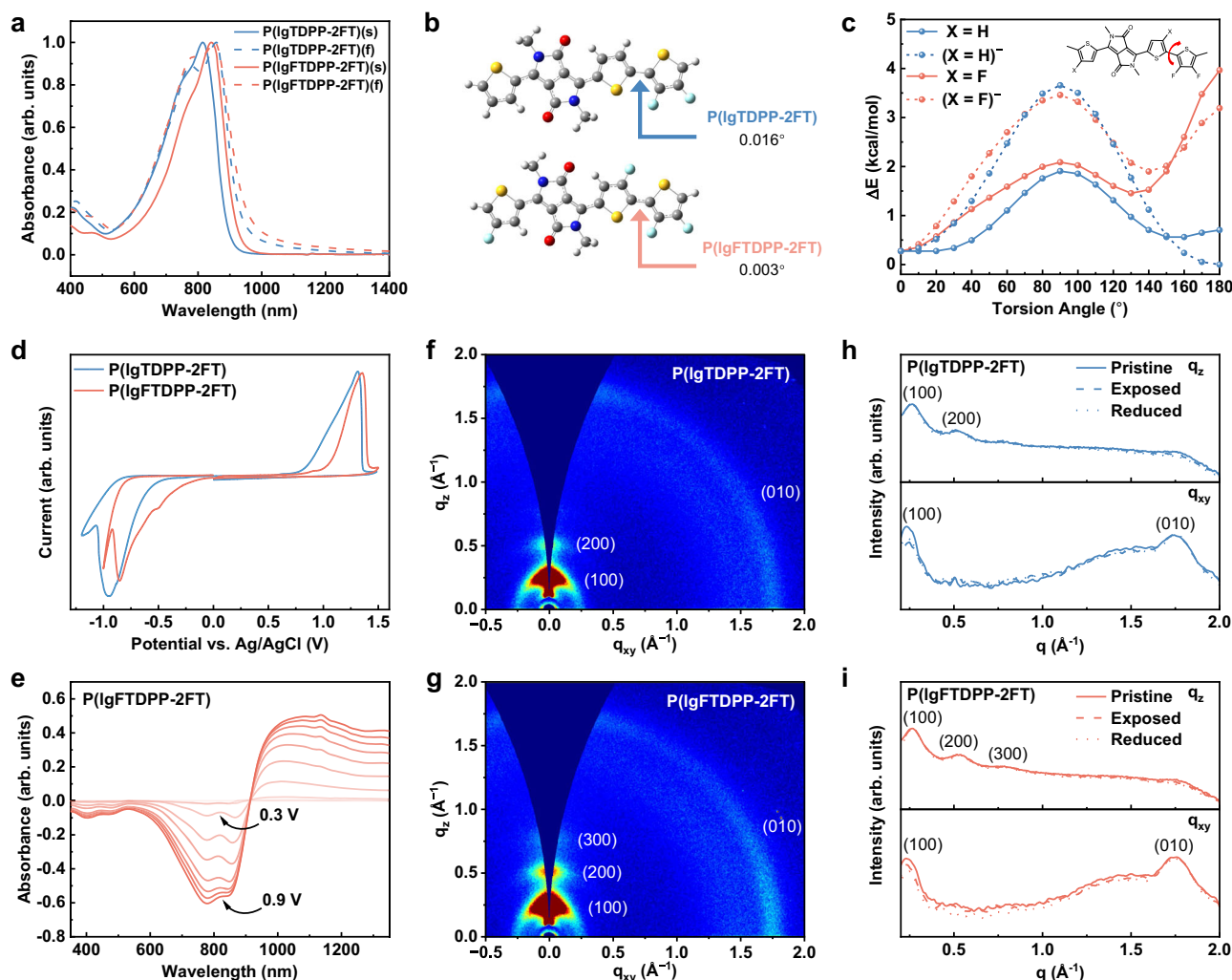


Fig. 3 | Optoelectronic properties and microstructures of the polymers. **a** Normalized UV-Vis-NIR absorption spectra of P(lgTDPP-2FT) and P(lgFTDPP-2FT) in chloroform (CF) solution and as thin films. **b** DFT-optimized backbone structures, bond lengths, and dihedral angles of the monomers. The atomic coordinates are provided in the Supplementary Dataset 1. **c** Comparison of the relaxed PES scans of the dihedral angles of the monomers in the neutral and negatively charged states. **d** Cyclic voltammetry of both polymers, normalized for comparison. **e** Electrochemical absorption spectra of P(lgFTDPP-2FT) on ITO glass in 0.1 M NaCl

aqueous solution. **f, g** 2D-GIWAXS patterns of both polymer films.

h, i Corresponding line cuts of the 2D-GIWAXS patterns. Cuts along the q_{xy} direction represent scattering in the in-plane direction, while the q_z is from the out-of-plane direction. “Pristine” stands for dry films without any treatment. “Exposed” stands for the films immersed in 0.1 M NaCl for 10 min and blow-dried. “Reduced” stands for the films biased at 0.9 V in 0.1 M NaCl for 10 min and blow-dried. Source data are provided as a Source Data file.

Fig. 11). The vertically oriented pores facilitate rapid ion transport, while the in-plane interconnected network forms efficient electronic pathways.

OECT device fabrication and characterization

To evaluate the electrical performance of the polymers, OECT devices were fabricated. All the OECT devices were fabricated using photolithography and parylene patterning method. The semiconducting layers were deposited by spin-coating (see Supporting Information for more details). The following equation (Eq. (1)), based on the Bernards’ model⁴⁰, was used to evaluate the performance of both polymers:

$$g_m = W/L \cdot d \mu \cdot C^* |V_{Th} - V_{GS}| \quad (1)$$

where g_m is the transconductance in the saturation regime; W , L , and d are the channel width, length, and film thickness, respectively; μ denotes the charge carrier mobility; C^* denotes the capacitance of the

channel; V_{Th} is the threshold voltage; and V_{GS} is the voltage between the gate and source electrode.

Both polymers showed typical n -type OECT behaviors. P(lgFTDPP-2FT) exhibited a μC^* value of $186.9 \text{ F cm}^{-1} \text{ V}^{-1} \text{ s}^{-1}$, much higher than that of P(lgTDPP-2FT) ($72.4 \text{ F cm}^{-1} \text{ V}^{-1} \text{ s}^{-1}$) (Fig. 4a, b, Supplementary Fig. 12a, b and Supplementary Table 3). Moreover, P(lgFTDPP-2FT) showed a much lower V_{Th} of 0.32 V than that of P(lgTDPP-2FT) (0.67 V), which is essential for low-voltage operation and matching the performance of p -type polymers. The bulk capacitance (C^*) was measured by electrochemical impedance spectroscopy (EIS) (Supplementary Fig. 13). The maximum C^* values extracted were averaged to be 156.3 F cm^{-3} for P(lgTDPP-2FT) and 107.6 F cm^{-3} for P(lgFTDPP-2FT). Based on the μC^* and C^* values, the electron mobility (μ) of P(lgFTDPP-2FT) was calculated to be $1.74 \text{ cm}^2 \text{ V}^{-1} \text{ s}^{-1}$, much higher than P(lgTDPP-2FT) ($0.46 \text{ cm}^2 \text{ V}^{-1} \text{ s}^{-1}$) (Table 1). The high μ of P(lgFTDPP-2FT) can be attributed to its uniquely stable and dominant backbone conformation. We also tested the transient characteristics of the OECT devices to evaluate their response speed (Fig. 4c, Supplementary Fig. 12c). The

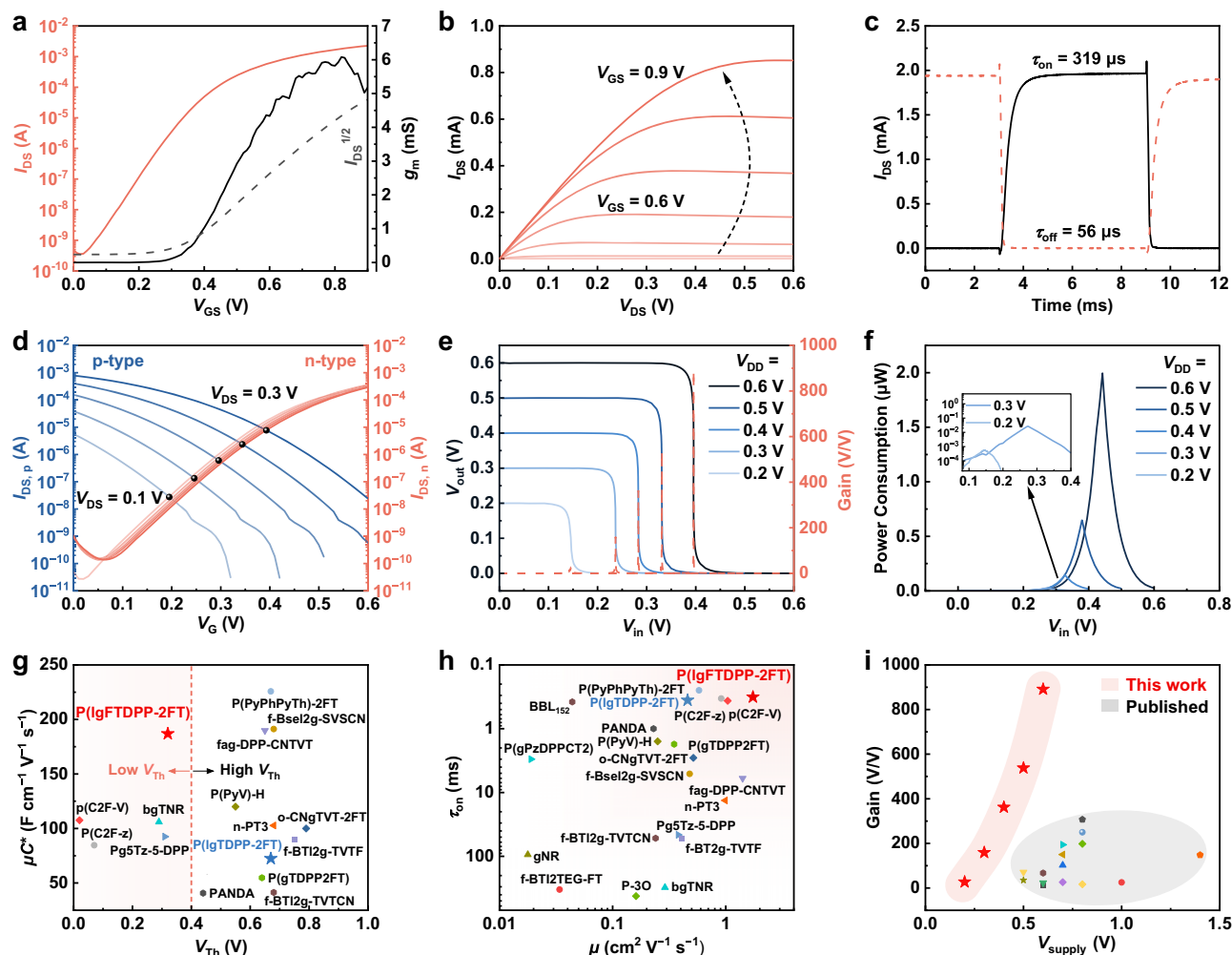


Fig. 4 | OECT device characterization. **a** Transfer, **b** output characteristics, and **c** transient on/off curves of P(lgFTDPP-2FT). $V_{GS} = 0–0.9$ V and $V_{DS} = 0.6$ V. **d** Transfer characteristics of *p*-type P(lgTDP-2FT) and *n*-type P(lgFTDPP-2FT) biased at different V_{DS} according to the inverter's working potential. For the *n*-type OECT, $V_{SS} = 0$ V and $V_{DD} = 0.1–0.3$ V; For the *p*-type OECT, $V_{SS} = 0.1–0.3$ V and $V_{DD} = 0.2–0.6$ V. **e** Voltage transfer and gain characteristics of the complementary inverter based on P(lgTDP-2FT) and P(lgFTDPP-2FT). **f** Power consumption of

the inverter based on P(lgTDP-2FT) and P(lgFTDPP-2FT) under various operation voltages. Inset: a magnified view of the low-voltage region. ($W/L = 100$ $\mu\text{m}/10$ μm). Comparison of the **g** μC^* vs. V_{Th} and **h** τ_{on} vs. μ of P(lgFTDPP-2FT) with other state-of-the-art *n*-type OECT materials (Supplementary Table 4)^{33–25,31,32,45,46,52–75}. **i** Comparison of the gain and supply voltage of the state-of-the-art inverters (Supplementary Table 5)^{31,32,43–55}. Source data are provided as a Source Data file.

Table 1 | OECT device performance of the polymers used in this study

Polymer	Type	$g_{m, \text{norm}}^a$ [S cm ⁻¹]	V_{Th}^b [V]	μC^*^c [F cm ⁻¹ V ⁻¹ s ⁻¹]	μ^d [cm ² V ⁻¹ s ⁻¹]	C^* [F cm ⁻³]	τ_{on}/τ_{off} [μs]	d [nm]
P(lgTDP-2FT)	<i>n</i>	22.0 (21.4 ± 2.2)	0.67	72.4 (69.1 ± 3.4)	0.46 (0.44 ± 0.02)	156.3 ± 3.9	358/87	83.1
P(lgFTDPP-2FT)	<i>n</i>	66.7 (61.3 ± 6.4)	0.32	186.9 (169.2 ± 19.1)	1.74 (1.57 ± 0.22)	107.6 ± 9.4	319/56	86.4
P(lgTDP-2MeOT2)	<i>p</i>	115.6 ^e	−0.22	174 ^e	2.15 ^e	80.8 ^e	63/11	51.2

^aTransconductance was normalized by the channel geometry. The W/L of all the devices is 100 $\mu\text{m}/10$ μm . All the OECT devices were operated in a 0.1 M NaCl aqueous solution.

^b V_{Th} was determined by extrapolating the corresponding $I_{DS}^{1/2}$ vs. V_{GS} plot;

^c μC^* was calculated according to Eq. (1);

^d μ was calculated from μC^* and the measured C^* . Five devices were analyzed for μC^* and $g_{m, \text{norm}}$ ($n = 5$). Three devices were analyzed for C^* ($n = 3$). The data outside the brackets are maximal value, and the inside ones indicate mean ± SD. ^eThe parameters were extracted from Ref. 22.

devices of P(lgTDP-2FT) and P(lgFTDPP-2FT) demonstrate rapid response times within the sub-millisecond regime, with τ_{on}/τ_{off} of 358/87 μs and 319/56 μs , respectively (Table 1). The faster response of P(lgFTDPP-2FT) is probably due to its lower C^* combined with higher μ ⁴¹. The combination of low V_{Th} , high μC^* , and high μ coupled with a rapid response of P(lgFTDPP-2FT) is particularly remarkable among state-of-the-art *n*-type OECT materials (Fig. 4g, h), demonstrating an

excellent balance between efficient charge transport and low-power operation.

Amplifier design and characterization

The inverter-based amplifier operates near the transition voltage (V_M), defined as the input voltage (V_{in}) where $V_{out} = V_{DD}/2$. One approach to estimate V_M is to take the intersection point where the I_{DS} of the *p*- and

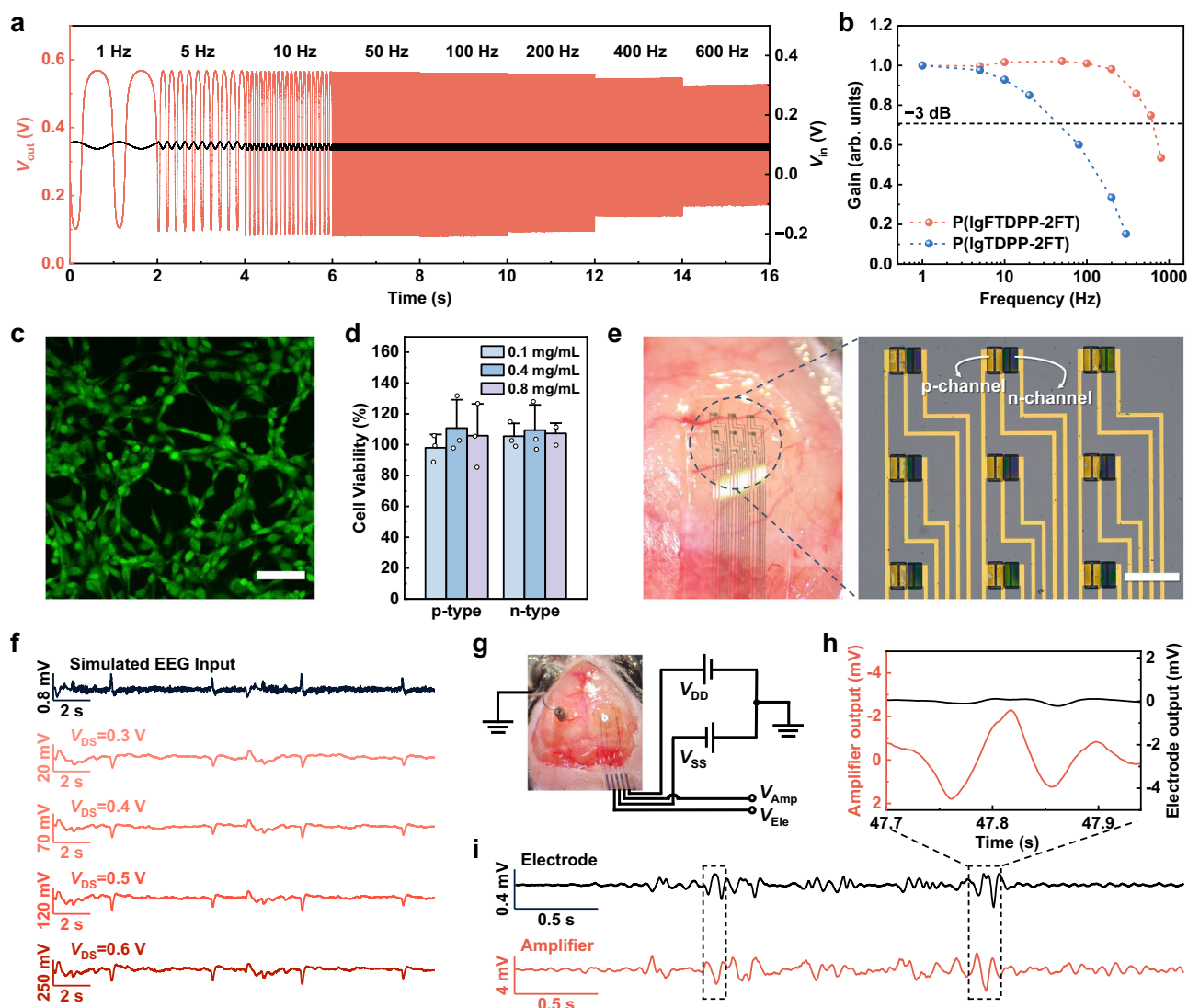


Fig. 5 | Characterization and applications of our biosignal amplifiers. **a** Dynamic response of the P(IgFTDPP-2FT)-based amplifier using 10 mV amplitude sinusoidal signals at different frequencies ($W/L = 30 \mu\text{m}/3 \mu\text{m}$). **b** Bandwidth comparisons of P(IgFTDPP-2FT)- and P(IgTDPP-2FT)-based inverters. **c** Live/Dead staining of mouse fibroblasts (NIH-3T3) cultured on P(IgFTDPP-2FT) film (scale bar: 100 μm). The experiment was repeated three times with similar results. **d** Cell viability of NIH-3T3 cells cultured with different polymer extractions, evaluated by MTT assay. *p*-type

stands for P(IgTDPP-MeOT2) and *n*-type stands for P(IgFTDPP-2FT). Data are presented as mean $\pm$ SD ($n = 3$ independent biological experiments). **e** Photograph of the flexible amplifier arrays. Scale bar: 250 μm . **f** Amplification of simulated EEG with P(IgFTDPP-2FT)-based amplifier. **g** Photograph of the in vivo ECoG recording and schematic of the electrical wiring. **h**, **i** Representative recordings of local field potentials (LFPs) obtained from an Ag/AgCl electrode and from our flexible amplifier. Source data are provided as a Source Data file.

n-type OECTs are equal¹⁷. In OECT-based inverters, a fundamental trade-off exists between gain and bandwidth, both of which are critical for on-site signal amplification. Our analysis indicates that operation closer to the subthreshold regime generally favors high intrinsic gain and low power consumption, but leads to slower response and narrower bandwidth. In contrast, operation in higher current regime improves switching speed and bandwidth, but sacrifices gain and increases power consumption (Supplementary Fig. 14 and Discussion). Therefore, realizing a high-performance inverter-based amplifier should not only employ matched materials, but should also consider the operating regime of each OECT to achieve a balanced trade-off among gain, bandwidth, and power consumption. In this context, OECTs operating in the moderate regime, which optimally balance high transconductance efficiency with sufficient charging current, are favorable for high-performance inverter operation (Supplementary Fig. 14a, d).

With this framework, the transfer characteristics, including V_{Th} and μC^* values of both *n*-type and *p*-type materials, largely influence

the transition point and the overall quality of the inverter (Supplementary Fig. 14d–f). Once the two materials are mismatched, the inverter requires a higher operating voltage with a pronounced shift in the V_{M} and working regime, or even exhibits complete switching failure (Supplementary Fig. 14e, f and Discussion). P(IgFTDPP-2FT) demonstrates a higher μC^* , higher μ , and lower C^* , which are advantageous for achieving high transduction efficiency and fast transient response. In addition, its lower V_{Th} facilitates operation within the moderate regime under low supply voltage. Therefore, we constructed complementary inverters by combining the P(IgFTDPP-2FT) with a suitably matched *p*-type material P(IgTDPP-MeOT2)²² based on the considerations of its V_{Th} (−0.22 V), μC^* ($174 \text{ F cm}^{-1} \text{ V}^{-1} \text{ s}^{-1}$), and response time ($\tau_{\text{on}}/\tau_{\text{off}} = 63/11 \mu\text{s}$) (Supplementary Fig. 15). Under inverter-relevant biasing conditions with V_{DD} ranging from 0.2 to 0.6 V, P(IgFTDPP-2FT) exhibited higher current and closer to $V_{\text{DD}}/2$ at the intersection (corresponding to the V_{M} in inverter) of their transfer characteristics compared with P(IgTDPP-2FT) (Fig. 4d, Supplementary Fig. 16).

Based on this material optimization, we further fabricated inverters with the same device geometry for *n*- and *p*-type OECTs, where polymers were processed by spin-coating combined with parylene-assisted patterning. Benefiting from the low V_{Th} and matched performance of both polymers, the inverters can operate at supply voltages (V_{DD}) from 0.2–0.6 V, achieving a maximum gain as high as 891 V V^{-1} at 0.6 V and a gain of 27 V V^{-1} even at 0.2 V (Fig. 4e). Such performance obviously outperforms all reported OECT-based complementary inverters due to combined effects of the materials' low SS , optimized device geometry, and the moderate operating regime enabled by the current-matching strategy (Fig. 4i, Supplementary Fig. 17). The inverter also exhibited low power consumption due to the low supply voltage, with a dynamic power consumption of only 0.57 nW at $V_{DD} = 0.2\text{ V}$, and $2\text{ }\mu\text{W}$ at 0.6 V (Fig. 4f, Supplementary Table 6). In addition, the inverter demonstrates good stability under continuous dynamic amplification and repeated measurements of static voltage transfer characteristics (VTCs) (Supplementary Fig. 18).

In inverter optimization, the noise margin (NM) depends on the transition point and gain, and mainly reflects the deviation of an inverter from the ideal switching point ($V_{DD}/2$), which is strongly related to the degree of material matching. Specifically, the high- and low-level noise margins are extracted from the VTCs:

$$NM_L = V_{IL} \quad (2)$$

$$NM_H = V_{DD} - V_{IH} \quad (3)$$

where V_{IL} and V_{IH} are defined as the low and high input voltages at the points on the VTC satisfying $\partial V_{out}/\partial V_{in} = -1$. Then, NM is defined as the minimum of the two noise margins, normalized by half of the supply voltage V_{DD} . Benefiting from better-matched V_{Th} between the two materials, only minor device-level tuning is required to achieve a nearly ideal VTC. Through adjustments to the aspect ratio of the channels, V_M was observed to approach the ideal value ($V_{DD}/2$) (Supplementary Fig. 19b, Supplementary Table 7). A noise margin (NM) of 92% was obtained at $V_{DD} = 0.6\text{ V}$. With decreasing supply voltage, minor voltage shifts result in proportionally larger changes in NM ; however, an NM of 78% was still preserved at 0.2 V. In contrast, the high V_{Th} of P(IgTDPP-2FT) limits the ability to tune the operating regime through device geometry (Supplementary Fig. 19a). With small-channel planar inverter devices for future single-cell recording, we further examined their amplification capability under different input frequencies. As illustrated in Fig. 5a, small sinusoidal inputs within the transition region were reliably amplified, and the devices exhibited a remarkable cut-off frequency of 622 Hz, higher than the 44 Hz obtained with inverters based on P(IgTDPP-2FT) (Fig. 5b, Supplementary Tables 8, 9). This improvement originates not only from the optimization of μ and response speed at the single material level, but also from the shift in the operating regime enabled by the material matching (Supplementary Fig. 20). These results suggest that matched materials contribute to outstanding inverter performance and exhibit significant potential for low-power, high-gain, and broadband biosignal amplification.

On-site biological recording and amplification

The biocompatibility of P(IgFTDPP-2FT) and P(IgTDPP-MeOT2) was evaluated through cell viability assays. Live/dead staining revealed that mouse fibroblasts (NIH-3T3) cultured on polymer films exhibited complete viability (green fluorescence), comparable to that of the control group. (Fig. 5c, Supplementary Fig. 21). This was further confirmed by the MTT assays, where cells treated with polymer extracts for 24 h maintained near 100% viability (Fig. 5d), demonstrating excellent biocompatibility and minimal cytotoxicity of the polymers. In vivo histology from 1–4 weeks post-implantation revealed minimal neuronal loss near the device, with astrocyte density and connective

tissue thickness remaining stable (Supplementary Fig. 22), supporting its potential for long-term neural interfacing applications.

For potential flexible integration, we further fabricated flexible inverter amplifiers through a two-step parylene-assisted patterning process (Supplementary Fig. 23). We fabricated a 3×3 flexible inverter array with all devices operating properly and exhibiting gains exceeding 500 V V^{-1} (Fig. 5e, Supplementary Fig. 24), demonstrating the good uniformity of our polymers and their great potential for scalable integration in more complex bioelectronic platforms.

To assess the applicability of our devices for biosignal amplification, we first conducted in vitro tests using simulated electroencephalographic (EEG) signals. With the V_M biased close to 0 V, the simulated EEG waveform applied as V_{in} was amplified into a large V_{out} . Robust, distortion-free amplification was observed across supply voltages of 0.3–0.6 V, achieving a peak gain of 413 V V^{-1} (Fig. 5f). In contrast, the P(IgTDPP-2FT)-based inverter, despite having a similar static gain, exhibited a lower dynamic gain (95 V V^{-1}) due to a much lower bandwidth (Supplementary Fig. 25). Furthermore, we employed a single flexible inverter device for in vivo acquisition of electrocorticographic (ECoG) signals. A pellet Ag/AgCl electrode was externally placed on one side of the mouse dura mater to serve as the ground, while a flexible electrode together with the inverter device was positioned on the opposite side for raw signal recording (Fig. 5g). We recorded local field potentials (LFPs) from mice at a 2 kHz sampling rate and filtered them to retain activity in the 4–45 Hz range for better comparison. Due to anesthesia-induced suppression of neural activity, the recorded signals predominantly reflected θ -band (4–8 Hz) and α -band (8–12 Hz) low-frequency oscillations. Both the waveforms and spectrograms showed strong overlap between the electrode and inverter recordings, although minor differences arose due to the spatial separation ($\sim 1\text{ mm}$). (Fig. 5h, i, Supplementary Fig. 26a). Compared to the original signals, the inverter achieved an amplification factor of up to 16 V V^{-1} . After the same filtering process, the SNR of the inverter recordings was up to 33.6 dB, higher than 27.2 dB of the electrode recordings, largely due to the suppression of downstream noise provided by in situ amplification. In addition, another longer recording revealed that the inverter exhibits a stronger recording capability for low-amplitude neural activities, while the electrode signal becomes indistinguishable from noise (Supplementary Fig. 26b). Although the amplification factor in vivo was lower than that observed in vitro, probably due to noise interference, the complex biofluid environment, and mechanical disturbances⁴², in-sensor amplification improves the preservation of more information from weak neural signals. We believe that through further optimization in materials' in vivo stability, biocompatibility, and circuit-level design, higher performance can be achieved.

Discussion

In summary, we developed a matched n-p material and device system to elucidate the design principles of complementary biosignal amplifiers. For material design, we employed a simple fluorination strategy and synthesized P(IgFTDPP-2FT), which combined low threshold voltage ($V_{Th} = 0.32\text{ V}$), high electron mobility ($\mu \approx 1.74\text{ cm}^2\text{ V}^{-1}\text{ s}^{-1}$), and outstanding performance metrics ($\mu C^* \approx 186.9\text{ F cm}^{-1}\text{ V}^{-1}\text{ s}^{-1}$), making it one of the best-performing *n*-type OECT materials to date. When paired with a matched *p*-type polymer P(IgTDPP-MeOT2), the resulting inverters achieved a gain value exceeding 800 V V^{-1} and a broad bandwidth over 600 Hz at $V_{DD} = 0.6\text{ V}$. Additionally, the system can operate at low voltages down to 0.2 V with a power consumption as low as 0.57 nW. Our polymer also exhibited good biocompatibility and could be readily processed into flexible amplifier arrays for biosignal recording. Importantly, this work highlights the role of the device operating regime, where operation in a moderate regime enables a balanced trade-off among high gain and broad bandwidth. This work not only provides some new guidelines for OECT-based amplifiers but

also provides a materials-to-device design strategy for balancing electrical performance for practical needs.

Method

All animal procedures and experiments performed in this study were approved and supported by the Institutional Animal Care and Use Committee (IACUC) of Peking University (Approval No. MSE-LeiT-1) and Tsinghua University (Approval No. THU-LARC-2025-031), Beijing, China. We have complied with all relevant ethical regulations for animal use.

Statistics & reproducibility

No statistical method was used to predetermine sample size. Sample sizes were chosen based on previous studies in the field and are consistent with commonly used practices for device characterization. For in vivo histology test, sample sizes were chosen based on established practices while minimizing animal use in accordance with ethical guidelines. Data were excluded only when clear experimental artefacts were identified (e.g., device failure or measurement errors), according to pre-established criteria. For all statistical analyzes, data were obtained from independently fabricated devices unless otherwise specified. The experiments were not randomized. The investigators were not blinded to allocation during experiments and outcome assessment because each animal served as its own control.

Materials

All chemical reagents were purchased and used as received unless otherwise indicated. All air and water-sensitive reactions were performed under a nitrogen atmosphere. Ultradry solvents were obtained from the J&K reagent company. 2-methyl-2-butanol was dried via CaH_2 and distilled under argon.

Chemical structure and optoelectronic property characterization

^1H NMR and ^{13}C NMR spectra were recorded on Bruker ARX-400 (400 MHz) or Bruker ARX-600 (600 MHz). All chemical shifts were reported in parts per million (ppm). ^1H NMR chemical shifts were referenced to CDCl_3 (7.26 ppm) or DMSO (2.5 ppm), and ^{13}C NMR chemical shifts were referenced to CDCl_3 (77.00 ppm) or DMSO (39.5 ppm). High-resolution mass spectra were acquired on a Bruker Solarix XR Fourier Transform Ion Cyclotron Resonance Mass Spectrometer (FTICR-MS) in positive ion mode. Thermal gravity analyzes (TGA) were carried out on a TA Instrument Q600 SDT analyzer, and differential scanning calorimetry (DSC) analyzes were performed on a TA Instrument Q2000 analyzer. Absorption spectra were recorded on a UV-vis spectrometer (PerkinElmer Lambda 750). Cyclic voltammograms (CV) were measured through an electrochemical workstation SP-200 (Bio-Logic Science Instruments). A standard three-electrode setup was established by employing glassy carbon as the working electrode (WE), a block of platinum mesh as the counter electrode (CE), and an Ag/AgCl electrode as the reference electrode (RE), further calibrated against ferrocene (Fc/Fc^+). The measurements were carried out in acetonitrile with 0.1 M tetrabutylammonium hexafluorophosphate and 0.1 M NaCl solution as the supporting electrolyte with a scan rate of 50 mV/s.

Size exclusion chromatography measurement

Polymer number-average molecular weight (M_n) and polydispersity index ($\text{PDI} = M_w/M_n$) were measured by size exclusion chromatography (SEC). SEC analyzes were performed in chloroform using a Waters 1515 system equipped with a PLMIXED 7.5 $\times$ 50 mm guard column, two PLMIXED-C 7.5 $\times$ 300 columns, and a differential refractive index detector at 35 $^\circ\text{C}$ with a flow rate of 1 mL min^{-1} . The instrument was calibrated with 10 PS standards, and chromatograms were processed with Waters Breeze software.

DFT calculations

Geometry optimization, molecular energy level calculation, and relaxed potential energy surface scan were performed at the (U) B3LYP/6-311 g(d,p) or wB97XD/6-311 g(d,p) level using Gaussian 16¹ and Gaussian View 6². Long side chains were replaced with methyl groups to simplify the calculation. Energy level calculations for both polymers, each containing three fragments, were conducted at the B3LYP/6-311 G(d, p) level. Relaxed potential energy surface (PES) scans of the polymers in neutral and negatively charged states were performed using the wB97XD/6-311 G(d, p) level.

AFM and GIWAXS characterization

Atomic force microscopy (AFM) measurements were performed with a Cypher atomic force microscope (Asylum Research, Oxford Instruments). Two-dimensional grazing incidence wide-angle X-ray scattering (2D-GIWAXS) measurements were conducted on a Xenocs-SAXS/WAXS system with an X-ray wavelength of 1.5418 Å and 0.2 $^\circ$ as an incidence angle. Pilatus 300 K was used as a 2D detector. Data processing was performed in Igor Pro software with Nika and WAXTools packages. Coherence length (L_c) was calculated from the breadth (Δq) of a diffraction peak: $L_c = 0.89 \times 2\pi/\Delta q$. Para crystalline disorder was calculated from the center position (q_0) and breadth (Δq) of a diffraction peak: $g = (\Delta q/(2\pi q_0))^{1/2}$.

Spectroelectrochemistry

Spectroelectrochemistry was performed using an ITO-coated glass slide, which was spin-coated with the polymer solution (5 mg mL^{-1} chloroform solution) at a rotating speed of 1000 rpm. These polymer-coated ITO slides were employed as the working electrode (WE) and immersed in a cuvette filled with a 0.1 M aqueous NaCl solution. A Pt wire was used as the counter electrode (CE), and an Ag/AgCl pellet served as the reference electrode (RE). A UV-vis spectrometer (PerkinElmer Lambda 750) was used, with the beam path passing through the electrolyte-filled cuvette and the polymer-coated ITO samples. A background spectrum was recorded with the cuvette/electrolyte/ITO-coated glass setup before a potential was applied to the cell. The potential was applied to the WE for 5 s before recording the spectra and was maintained for a certain amount of time until the completion of spectrum scanning.

Electrochemical impedance spectra

Electrochemical impedance spectroscopy (EIS) was performed on polymer-coated electrodes using an electrochemical workstation (BioLogic Science Instruments, SP-200). The polymer films were patterned into squares of defined areas on gold (Au) electrodes through the photolithography technique. These polymer-coated Au electrodes with silica substrate were fully covered with 0.1 M NaCl solution, followed by the employment of a Pt wire and an Ag/AgCl pellet. The capacitances of polymers were determined using a potential-EIS method. A DC offset voltage corresponding to the maximum doping level of each polymer was applied, along with a sinusoidal AC voltage of 10 mV (RMS) at frequencies ranging from 1–100 kHz. The as-obtained Bode or Nyquist plots were fitted to the Randle's circuit, $R_s(R_p||C)$, using the software EC-Lab view. The thickness of the film was determined in the dry state using a DEKTA profilometer (Bruker). Each material underwent more than three independent measurements (sample size $n \geq 3$), with the mean and standard deviation (SD) calculated to summarize the results.

Fabrication of OECTs and inverters

The fabrication of OECTs and inverters includes the deposition and patterning of the metallic electrode, the parylene layer, and the polymer in the channel. In detail, silica substrates were cleaned by ultrasonication in acetone, DI water, and isopropyl alcohol, followed by nitrogen blow-drying and a brief oxygen plasma treatment. Metal pad

interconnects and source/drain contacts were patterned and deposited (5 nm Cr and 35 nm Au) using a lift-off process. About 1 μm of parylene-C was deposited as a dielectric layer using a PDS 2010 Lab-coater-2, with a 3-(trimethoxysilyl) propyl methacrylate (A-174 Silane) adhesion promoter. A 2.5% aqueous solution of industrial cleaner (Micro-90) was subsequently spin-coated as an anti-adhesive layer for a second sacrificial 1 μm parylene-C film. The channel areas were subsequently patterned with AZ 10XT photoresist and AZ-400 K developer. The exposed channel areas were etched by reaction ion etching (RIE) with O_2 plasma using an LCCP-6A reactive ion etcher (Leuven Instruments). Remove the photoresist and perform a brief oxygen plasma clean. The *n*-type polymer was dissolved in CF or DCM at a concentration of 5 mg mL^{-1} or 8 mg mL^{-1} , and the *p*-type polymer was dissolved in HFIP at a concentration of 3 mg mL^{-1} , followed by spin-coating at 800 rpm (*n*-type) and 1000 rpm (*p*-type). After a peeling-off process of the second sacrificial parylene layer, the OECTs or inverters were ready for measurement.

OECTs and inverters characterization

The device characterization was performed on a probe station using an Fs-Pro semiconductor parameter analyzer (PDA). An Ag/AgCl pellet (Warner Instruments) was employed as the gate electrode and immersed in a 0.1 M NaCl solution, which covered the polymer film in the channels. The thickness of the films was determined in a dry state after testing with a DEKTAK profilometer (Bruker). The bandwidth was measured using the PDA for V_{DD} bias and measurement. A function generator (Keithley 3390) was employed to provide the DC V_{in} bias and superimpose a small-amplitude sinusoidal signal.

Simulated EEG amplification

The measurement was performed using rigid devices with $W/L = 100 \mu\text{m}/10 \mu\text{m}$, the same as those used in inverter characterization. An Ag/AgCl pellet was employed as the gate electrode and was immersed in a 0.1 M NaCl electrolyte that covered the polymer film in the channel. First, the voltage transfer curve of the inverter was measured at a given $V_{\text{DD,C}}$ to determine the switching voltage V_{M} . Then, the supply voltages V_{DD} and V_{SS} were adjusted so that the switching point was located near 0 V. Specifically, V_{DD} was set to $V_{\text{DD,C}} - V_{\text{M}}$, while V_{SS} was set to $-V_{\text{M}}$. The simulated EEG voltage signal was applied to the gate as the input signal, and the corresponding V_{out} was recorded. Notably, V_{SS} was applied using a signal generator (Keithley 3390), while the remaining signals were applied and recorded using an Fs-Pro semiconductor parameter analyzer (PDA). Finally, the output signals were digitally filtered using a 0.1 Hz high-pass filter and a 50 Hz notch filter.

Cytotoxicity

For live/dead cell fluorescence imaging, polymers were spin-coated onto glass substrates, baked at 100 °C, washed with DI water, and blow-dried. After a brief oxygen plasma treatment, the samples were placed into 12-well plates. Mouse fibroblasts (NIH-3T3) (CellCook, CC9002) were seeded at 1×10^5 cells per well and cultured in high-glucose DMEM (CM2016, CellCook) supplemented with 10% fetal bovine serum (FBS, BL201A, Biosharp), 100 U/mL penicillin, and 100 U/mL streptomycin (SV30010, HyClone) at 37 °C in a humidified atmosphere containing 5% CO_2 . After 24 h, a live/dead staining assay (E-CK-A354, Elabscience) was performed to evaluate the biocompatibility of the substrates. Briefly, cells were washed with PBS, and a working solution was prepared by adding 1 μM Calcein-AM and 7.5 μM propidium iodide (PI) to DMEM. The working solution was added to the cultures and incubated at 37 °C for 30 min. Live cells (green fluorescence) and dead cells (red fluorescence) were then observed using an ECLIPSE LV100ND POL/DS microscope (Nikon) equipped with a C-HGFI (Nikon) light source and a 490 nm excitation filter.

For the MTT assay, polymer solutions were drop-cast onto OTS-modified silica wafers, vacuum-treated for 12 h after peeling, and sequentially washed with DI water, PBS, and DMEM. The samples were then extracted in complete culture medium at 37 °C for 24 h at a concentration of 0.8 mg mL^{-1} . The extracts were filtered through a 0.22 μm filter and diluted to the desired concentrations. NIH-3T3 cells were seeded into 96-well plates (2×10^4 cells per well) and incubated for 24 h. After washing with PBS, the cells were treated with extraction media of different concentrations for another 24 h. The extract was then removed, and MTT solution (0.5 mg/mL in DMEM, MA0198, MeilunBio) was added. Following incubation for 4 h in the dark, the solution was carefully removed, and 100 μL of DMSO was added to each well to dissolve the formazan crystals. The absorbance was measured at 570 nm using a microplate reader (Multiskan FC, Thermo Scientific).

In vivo experiments

Mature female C57BL/6 mice (8 weeks old) were purchased from the Peking University Laboratory Animal Center and the Laboratory Animal Resources Center of Tsinghua University and housed in the animal facility. The mice were maintained under specific-pathogen-free conditions with a 12/12-h light/dark cycle, 22–26 °C, 30–70% relative humidity, and access to sterile pellet food and water ad libitum. Sex was not considered as a variable in the experimental design, as no sex-dependent effects were expected. At the end of the experiments, animals were euthanized by gradual-fill CO_2 inhalation in accordance with institutional animal care and use guidelines.

In vivo histology test

For device preparation, P(IgFTDPP-2FT) solution (5 mg/mL in chloroform) was spin-coated onto a 5 μm -thick parylene-C substrate, which was subsequently patterned into circular disks. Prior to implantation, the devices underwent vacuum treatment followed by prolonged immersion in sterile saline to remove any residual organic solvents.

During surgery, mice were anesthetized with 2–3% isoflurane and maintained with 1–1.5% isoflurane during the surgery on a heating pad. A craniotomy (circular, diameter ~4 mm) was performed to expose the cortical surface. Following the craniotomy, the prepared device was carefully placed onto the exposed cortical surface, ensuring full contact with the dura mater. A sterile glass coverslip was then positioned over the craniotomy window to stabilize the electrode against the tissue. Subsequently, the skull surface was lightly etched and dried, and the dental cement was applied to cover the glass coverslip, electrode leads, and exposed skull, embedding the connector base. The cement was allowed to cure completely. After surgery, meloxicam (5 mg/mL) was administered via subcutaneous injection and again at 24 h post-operation for a total of two days. Mice were allowed to recover before any subsequent experiments.

Immunofluorescence staining and analysis

After anesthesia (described above) was introduced, mice were perfused transcardially with 1X PBS (Servicebio) followed by 4% paraformaldehyde (Biosharp). The animal was decapitated and the brain was removed from the skull and set in 4% formaldehyde for 24 h as post-fixation. The incrementally increasing sucrose solutions (10–30%) were used to soak the brains overnight for cryoprotection. Sagittal frozen sections (30 μm) were cut on a cryostat (RWD FS800A) and collected in PBS.

Sections were washed $3 \times 5 \text{ min}$ in cold PBS and subjected to antigen retrieval in 0.01 M sodium citrate buffer (95 °C, 15 min). After cooling and washing, sections were blocked and permeabilized in PBS containing 0.3% Triton X-100 and 5% normal goat serum (Servicebio) for 1 h at room temperature. Next the slices were first incubated in PBST (1X PBS with 0.3% Triton X-100) with primary antibodies overnight at 4 °C for astrocytes (anti-GFAP mouse mAb, GB12090, 1:1000,

Servicebio) and neurons (anti-NeuN rabbit pAb, GB15138, Servicebio). The slices were then washed with PBS and incubated with secondary antibodies for 1 h at room temperature (Alexa Fluor 488-conjugated Goat Anti-Rabbit IgG, GB25303, 1:1000, Servicebio; Cy3 conjugated Goat Anti-mouse IgG, GB21301, 1:1000, Servicebio) followed by DAPI (1:1000, Sigma-Aldrich) for 20 min. After staining, the slices were mounted with a glass coverslip and Fluoromount-G (0100-01, SouthernBiotech) before imaging.

Image acquisition and analysis

Stained sections were imaged on a Nikon A1Rsi HD25 super resolution laser scanning confocal microscope using a 10 X objective. Large field tile scans (with Z stacks) were acquired with excitation lasers at 405 nm (DAPI), 488 nm (Alexa Fluor 488), and 561 nm (Cy3).

For quantification, two regions of interest (ROIs) were defined in each section: (1) the device contact region (tissue directly underlying the implant) and (2) the contralateral control region (homotopic area in the opposite hemisphere). Using NIS Elements Viewer (Nikon), NeuN positive neurons and GFAP positive astrocytes were automatically identified and counted within a 500 μm -wide band centered on the cortical surface in each ROI. The thickness of connective tissue (from pia to white matter) was also measured in both ROIs. Statistical comparisons of cell densities and cortical thickness between device and control regions were performed.

Fabrication of flexible inverters

Detailed procedures are provided in the Supplementary Fig. 23. First, an 8 μm -thick parylene substrate was deposited on a silicon wafer. The prior fabrication procedures of interconnects patterning and deposition of 2 parylene layers were similar to those of rigid inverter devices. After exposing one side of the channel by lithography and RIE, the *p*-type polymer and fluoroacrylate (Novec 2704, 3 M) were spin-coated at 1000 rpm and 1500 rpm, respectively. The sacrificial parylene layer was then peeled off, leaving the *p*-type material confined to the left channel region. Subsequently, a 2.5% Micro 90 aqueous solution was spin-coated, and 1 μm parylene was deposited. The *n*-type channel and electrodes were then exposed using the same approach. A small amount of 2.5% Micro 90 aqueous solution was drop-cast onto the electrode region and dried. The device was spin-coated with the *n*-type polymer at 800 rpm. After peeling off the sacrificial parylene layer, the *n*-type material was retained only on the *n*-channel. The device was soaked in Novec 7200 (3 M) to remove fluoroacrylate. Ag/AgCl paste was drop-cast on the electrode areas and cured at 80 °C. The devices were finally cut and directly delaminated from the wafer. To achieve electrical connection with measuring instruments, an Anisotropic Conductive Film (ACF) bonding machine (GZC-COF280, Hailunda Technology, Shenzhen, China) was used to connect flexible devices and flexible printing circuits (FPC) cable.

In vivo ECoG recording

During surgery, mice were anesthetized with oxygen-vaporized isoflurane (3% for induction, 1.5–2% for maintenance, 0.5 L/min) and secured in a stereotaxic frame. Body temperature was maintained at 37 °C throughout the procedure. Craniotomies (~10 mm²) were performed to expose the cortical surface without removing the dura mater. An Ag/AgCl pellet was inserted into the cortex through the craniotomy to a depth of approximately 0.5 mm to serve as the ground electrode, and the device was fixed to another craniotomy site. Before in vivo testing, the static test was conducted using an Fs-Pro semiconductor parameter analyzer (PDA). The voltage transfer curve of the inverter was measured at $V_{\text{DD}} = 0.6 \text{ V}$ to determine the switching voltage V_{M} . Then, the supply voltages V_{DD} and V_{SS} were adjusted so that the switching point was located near 0 V. Specifically, V_{DD} was set to $0.6 - V_{\text{M}}$ by PDA, while V_{SS} was set to $-V_{\text{M}}$ by a signal generator (Keithley 3390). The output signals from the inverter and the electrode

were recorded with PDA. Signals from the inverter and the electrode were acquired simultaneously, with a spatial separation of ~1 mm. The acquired signals were bandpass filtered at 4–45 Hz and notch filtered at 50 Hz and its harmonics. To facilitate comparison, the time axis was slightly adjusted based on the peak of activity. Spectrograms were generated using continuous wavelet transform (CWT) via the *letswave* toolbox in MATLAB.

Reporting summary

Further information on research design is available in the Nature Portfolio Reporting Summary linked to this article.

Data availability

The datasets generated in this study have all been deposited in Figshare and are publicly available under <https://doi.org/10.6084/m9.figshare.31819981>. Source data are provided with this paper.

Code availability

No central code was developed for this work.

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Author contributions

Q.L., J.-P.C., Z.R., X.L., and G.-Y.G. synthesized the polymers. Y.Z., Q.L., X.P., Z.X., P.L., and J.L. performed polymer characterizations. C.W. carried out the DFT calculations. Y.Z., S.G., and J.X. fabricated the devices and performed electrical characterizations. X.W., Y.Z., and B.Y. conducted the biocompatibility tests. Y.Z., X.W., Y.C., W.L., W.S., and S.F. performed the in vivo recording experiments. Y.Z., Q.L., X.D., T.L., and Z.Z. wrote the manuscript. All authors discussed the results and contributed to the final version of the manuscript.

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Competing interests

The authors declare no competing interests.

Additional information

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