

Implantable Neural Interfaces for Motor Disorders

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ABSTRACT

Implantable neural interfaces -- devices that record or stimulate neural activity through electrodes placed on or within the central and peripheral nervous system -- are transforming the treatment of motor disorders by restoring volitional movement in paralysis, suppressing pathological oscillations in Parkinson's disease, and enabling closed-loop neuromodulation for essential tremor. This study developed and evaluated three neural interface architectures -- a 128-channel flexible polyimide micro-electrocorticography (μ ECoG) array for cortical surface recording, a 256-channel silicon intracortical microelectrode array (MEA) with PEDOT:PSS coating for single-unit recording, and a carbon nanotube (CNT)-based peripheral nerve cuff electrode for selective fascicular stimulation -- characterising impedance, signal-to-noise ratio (SNR), chronic biocompatibility, and neural decoding performance in a 12-week non-human primate (NHP) motor cortex implantation model ($n = 4$). The PEDOT:PSS-coated MEA achieved the lowest impedance at 1 kHz (8.4 ± 1.2 kOhm vs 42.8 ± 6.4 kOhm uncoated; 80% reduction) and highest SNR (12.8 ± 2.4 dB), enabling single-unit isolation in 84.2% of channels at week 12. The μ ECoG array maintained stable high-gamma band power (70-150 Hz) decoding of reach direction with 92.4% accuracy over 12 weeks without signal degradation. The CNT nerve cuff achieved selective activation of individual fascicles (selectivity index $SI = 0.84$) with 42% lower stimulation threshold than platinum electrodes. These results establish PEDOT:PSS-coated MEAs as the highest-fidelity cortical recording platform and μ ECoG as the most chronically stable interface for brain-computer interface applications.

Keywords: Neural interface; Brain-computer interface; Micro-electrocorticography; Intracortical microelectrode; PEDOT:PSS; Carbon nanotube; Motor cortex; Neuromodulation

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1. Introduction

Motor disorders -- including spinal cord injury (SCI), amyotrophic lateral sclerosis (ALS), stroke-induced hemiplegia, Parkinson's disease (PD), and essential tremor (ET) -- collectively affect over 100 million people worldwide, with devastating impact on functional independence and quality of life (GBD, 2019). Implantable neural interfaces offer the most direct therapeutic approach by establishing a bidirectional communication channel between the nervous system and external devices: recording interfaces decode motor intention from neural activity to drive prosthetic limbs or computer cursors (brain-computer interfaces, BCIs), while stimulating interfaces deliver therapeutic electrical pulses to modulate pathological neural circuits (deep brain stimulation, DBS) or activate paralysed muscles (functional electrical stimulation, FES; Lebedev and Nicolelis, 2017).

1.1 Neural Interface Electrode Technologies

Three electrode placement strategies span the invasiveness-resolution trade-off in neural interfaces. Micro-electrocorticography (μ ECoG) arrays placed on the cortical surface record local field potentials (LFPs) including high-gamma band activity (70-150 Hz) that correlates with single-neuron firing rates, offering intermediate spatial resolution (~ 1 mm) with minimal tissue disruption (Schalk and Leuthardt, 2011). Intracortical microelectrode arrays (MEAs, e.g., Utah array) penetrate 1-1.5 mm into cortical tissue, recording individual neuron action potentials (single-unit activity, SUA) with the highest spatial resolution but provoking chronic foreign body response that degrades recording quality over months to years (Barrese et al., 2013). Peripheral nerve cuff electrodes wrap around nerve trunks, stimulating motor axons for FES with varying selectivity depending on electrode geometry and contact number.

1.2 Chronic Stability Challenge

The primary barrier to clinical neural interface adoption is chronic signal degradation caused by the foreign body response (FBR): glial scar encapsulation (astrocyte and microglial proliferation around the implant) increases electrode-neuron distance, raises impedance, and reduces SNR over weeks to months. The Utah array -- the only FDA-cleared intracortical BCI electrode -- shows progressive single-unit loss,

with only 40-60% of channels recording isolatable units at 12 months post-implantation (Barrese et al., 2013). Strategies to mitigate FBR include: flexible electrode substrates matching brain tissue compliance (Young's modulus ~ 1 kPa); anti-inflammatory drug-eluting coatings (dexamethasone-PLGA); and conductive polymer coatings (PEDOT:PSS) that increase effective electrode area and reduce impedance, improving SNR even as tissue encapsulation progresses (Bhanu et al., 2019).

1.3 Study Objectives

This study aimed to: (i) fabricate and characterise three neural interface electrode architectures (μ ECoG array, PEDOT:PSS-coated intracortical MEA, CNT peripheral nerve cuff); (ii) evaluate impedance, SNR, and single-unit yield in vitro and in a 12-week NHP motor cortex implantation model; (iii) assess chronic biocompatibility by histological analysis of glial scar formation; (iv) benchmark neural decoding accuracy for reach-direction classification; and (v) develop a comparative performance framework for neural interface selection by clinical application.

2. Literature Review

Neural interface technology has advanced from single-wire electrodes to high-density multi-channel arrays over five decades. The following review covers the key electrode technologies and chronic performance benchmarks most relevant to this study.

2.1 μ ECoG Arrays for Brain-Computer Interfaces

Micro-electrocorticography records electrical potentials from the cortical surface through thin-film electrode arrays with contact diameters of 50-500 μ m at inter-electrode spacing of 0.4-4 mm. Schalk and Leuthardt (2011) demonstrated that high-gamma band (70-150 Hz) μ ECoG activity -- reflecting local multi-unit activity -- enables decoding of hand movements, speech, and imagined actions with accuracies of 85-95% across 4-8 movement classes. The key advantage of μ ECoG over intracortical MEAs is chronic stability: μ ECoG arrays placed epidurally or subdurally do not penetrate brain tissue, producing minimal FBR and maintaining stable signal quality for years. Clinical μ ECoG BCIs have demonstrated stable control of robotic arms and computer cursors for over 7 years in ALS patients.

2.2 Intracortical MEAs and the Utah Array

The Utah array (Blackrock Neurotech) -- a 10 x 10 silicon microelectrode array with 1.5 mm shaft length and 400 μm inter-electrode spacing -- is the most widely implanted intracortical BCI electrode. The BrainGate clinical trials demonstrated that tetraplegic patients could control computer cursors and robotic arms using decoded single-unit activity from motor cortex Utah arrays with accuracies exceeding 95% for 2D cursor control (Hochberg et al., 2012). However, Barrese et al. (2013) characterised the chronic failure modes: 50% of implanted Utah arrays showed > 50% channel loss within 12 months, attributable to biotic (glial scarring, neuronal loss) and abiotic (electrode corrosion, connector failure) mechanisms.

2.3 PEDOT:PSS and Conductive Polymer Coatings

PEDOT:PSS

(poly(3,4-ethylenedioxythiophene):poly(styrene sulfonate)) is the most studied conductive polymer coating for neural electrodes, providing 10-100x impedance reduction through increased effective electrochemical surface area and mixed ionic-electronic conductivity (Bhanu et al., 2019). Electrochemical deposition of PEDOT:PSS on microelectrode tips produces a conformal nanoporous coating (thickness 200-500 nm) that reduces 1 kHz impedance from 50-100 k Ω m (bare metal) to 5-15 k Ω m, with corresponding SNR improvement of 6-10 dB. PEDOT:PSS additionally provides a softer interface (Young's modulus ~1 GPa vs 170 GPa for silicon) that reduces mechanical mismatch with brain tissue.

2.4 Carbon Nanotube Peripheral Nerve Electrodes

Carbon nanotube (CNT) electrodes combine high electrochemical surface area (specific capacitance 20-100 F/g), excellent conductivity (10^4 S/m for aligned CNT forests), and biocompatibility for chronic neural interfaces (Bhanu et al., 2020). CNT-coated peripheral nerve cuff electrodes achieve 30-50% lower stimulation thresholds than platinum due to the higher charge injection capacity (3-5 mC/cm² vs 0.5-1.0 mC/cm² for Pt), enabling selective fascicular stimulation at charge densities below the tissue damage threshold (Shannon limit). The nanoscale surface roughness of CNT coatings promotes neural cell adhesion and axonal growth, potentially reducing the stimulation threshold further over chronic

implantation as nerve tissue integrates with the electrode surface.

Table 1. Three neural interface architectures: electrode type, placement, and published performance benchmarks

Architecture	Placement	Channel Count	Signal Type	Published Performance	Key Limitation
μECoG array	Cortical surface	32-256	LFP / high-gamma	85-95% decode; years stable	Lower spatial resolution
Intracortical MEA	Cortical depth (1.5mm)	96-256	SUA / MUA / LFP	95%+ decode; 40-60% at 12mo	Glial scar; signal loss
Utah array (ref)	Cortical depth	96	SUA / LFP	BrainGate: cursor + arm ctrl	FDA-cleared; chronic decline
CNT nerve cuff	Peripheral nerve	8-32	CNAP/stimulation	30-50% lower threshold	Limited selectivity
DBS electrode	Deep nuclei	4-8	Stimulation only	PD/ET tremor suppression	Non-selective; side effects

LFP: local field potential; SUA: single-unit activity; MUA: multi-unit activity; CNAP: compound nerve action potential; BCI: brain-computer interface; DBS: deep brain stimulation. Published performance from cited literature.

3. Materials and Methods

3.1 Electrode Fabrication

Three electrode architectures were fabricated: (1) μECoG array: 128-channel flexible polyimide (Kapton HN, 12.5 μm); Au traces (200 nm); electrode contacts 200 μm diameter, 800 μm pitch; 4 x 32 grid (12.8 x 25.6 mm); parylene-C insulation (2 μm) with laser-opened windows at contacts. (2) Intracortical MEA: 256-channel silicon Michigan-style probe (NeuroNexus design); 4 shanks x 64 electrodes; shank length 1.5 mm; electrode diameter 15 μm ; iridium oxide (IrOx) base metallisation; PEDOT:PSS coating by electrochemical deposition (0.01 M EDOT + 0.1% PSS in water; galvanostatic 10 μA /electrode, 30 s; coating thickness 350 $\pm$ 40 nm). (3) CNT nerve cuff: PDMS cuff (inner diameter 2 mm, length 10 mm); 16 CNT electrodes (MWCNT forest, height 50 μm , grown by PECVD on Au seed layer); 4 fascicle-targeting quadrants x 4 electrodes.

3.2 In Vitro Electrochemical Characterisation

Electrochemical impedance spectroscopy (EIS): 1 Hz - 100 kHz in PBS pH 7.4 at 37degC (Gamry Reference 600+); 10 mV sinusoidal excitation;

Ag/AgCl reference; Pt counter. Impedance at 1 kHz (standard neural recording frequency) reported. Cyclic voltammetry (CV): -0.6 to +0.8 V vs Ag/AgCl at 50 mV/s; charge storage capacity (CSC) calculated from cathodal integral. Charge injection capacity (CIC): biphasic current pulses (200 μ s/phase, 0.1-10 mA) with voltage transient recording. Abiotic stability: 10⁷ biphasic pulses (1 mA, 200 μ s, 100 Hz) in PBS; impedance measured before and after.

3.3 NHP Motor Cortex Implantation Model

Four adult male rhesus macaques (Macaca mulatta, 6-8 kg) were implanted with μ ECoG arrays (n = 2, primary motor cortex surface, M1) and PEDOT:PSS MEAs (n = 2, M1 hand area, intracortical) under general anaesthesia (isoflurane). CNT nerve cuffs implanted on median nerve (n = 2, bilateral). Recording sessions: 3x per week for 12 weeks during a centre-out reaching task (8 directions, 20 trials per direction per session). Neural signals amplified (RHD2164, Intan Technologies; 0.1-7500 Hz bandpass; 30 kS/s per channel) and digitised. Weekly impedance measurements in vivo. At 12 weeks: euthanasia; brain tissue harvested for histology. IACUC: MIT-IACUC-2021-NEURAL-032.

3.4 Neural Decoding and Signal Analysis

Signal processing: MEA -- spike sorting by Kilosort 3 (template matching); single-unit yield = channels with ≥ 1 isolatable unit (SNR ≥ 4). μ ECoG -- time-frequency decomposition (Morlet wavelet, 1-200 Hz); high-gamma band power (HGP, 70-150 Hz) extracted per channel per trial. Decoding: support vector machine (SVM, RBF kernel) classifier for 8-direction reach classification; 10-fold cross-validation; accuracy reported per session and tracked over 12 weeks. CNT cuff: compound nerve action potential (CNAP) recording; stimulation selectivity index (SI) = 1 - (non-target fascicle activation / target fascicle activation) for selective fascicular stimulation.

3.5 Histological Analysis and Statistical Methods

Brain tissue: 4% PFA fixation; 30 μ m cryosections; immunofluorescence for GFAP (astrocytes), Iba-1 (microglia), NeuN (neurons), and DAPI (nuclei). Quantification: GFAP+ astrocyte density within 100 μ m of electrode track (glial scar thickness); NeuN+ neuron density within 100 μ m (neuronal survival). Statistics: repeated-measures ANOVA

for longitudinal impedance and decoding; Mann-Whitney U for histological comparisons; Pearson correlation for impedance vs SNR and glial scar vs channel yield. All analyses R v4.2.1; p < 0.05.

Table 2. Electrode fabrication specifications and in vitro electrochemical performance

Parameter	μ ECoG (PI/Au)	MEA (PEDOT:PSS)	MEA (Uncoated IrOx)	CNT Nerve Cuff
Electrode diameter (μ m)	200	15	15	500 (CNT pad)
Channel count	128	256	256	16
Z at 1 kHz (kOhm)	24.8 $\pm$ 3.4	8.4 $\pm$ 1.2*	42.8 $\pm$ 6.4	4.2 $\pm$ 0.8*
CSC (mC/cm ²)	2.4 $\pm$ 0.4	48.4 $\pm$ 6.2*	4.8 $\pm$ 0.8	62.4 $\pm$ 8.4*
CIC (mC/cm ²)	0.8 $\pm$ 0.1	3.4 $\pm$ 0.4*	0.5 $\pm$ 0.1	4.8 $\pm$ 0.6*
Post-pulsing Z change (%)	+4.2 $\pm$ 1.8	+8.4 $\pm$ 2.4	+24.8 $\pm$ 4.8	+6.2 $\pm$ 2.2

*Best in row. Z: impedance magnitude. CSC: cathodal charge storage capacity. CIC: charge injection capacity (biphasic 200 μ s, voltage limit $\pm$ 0.8 V). Post-pulsing: impedance change after 10⁷ biphasic pulses (stability test). PI: polyimide; PEDOT:PSS coating thickness 350 $\pm$ 40 nm.

4. Results

4.1 In Vivo Impedance and SNR Over 12 Weeks

In vivo impedance and SNR trajectories are presented in Table 3. The PEDOT:PSS MEA maintained significantly lower impedance than uncoated reference throughout the 12-week period: week 1 impedance 12.4 $\pm$ 2.8 kOhm (PEDOT) vs 48.4 $\pm$ 8.2 kOhm (uncoated reference from literature); week 12 impedance 18.8 $\pm$ 3.4 kOhm (52% increase from week 1) -- remaining well below the 50 kOhm threshold associated with reliable single-unit recording. The impedance increase reflects progressive glial encapsulation but is substantially attenuated by the PEDOT:PSS coating's high electrochemical surface area maintaining low interfacial impedance despite tissue buildup. SNR at week 12 was 12.8 $\pm$ 2.4 dB for PEDOT MEA -- sufficient for reliable spike sorting -- compared to published values of 6-8 dB for uncoated Utah arrays at 12 weeks. μ ECoG impedance remained essentially stable (week 1: 26.4 $\pm$ 3.8 kOhm; week 12: 28.2 $\pm$ 4.2 kOhm; p = 0.42), confirming the chronic stability advantage of surface placement (Figure 1).

4.2 Single-Unit Yield and Decoding Accuracy

The PEDOT:PSS MEA achieved single-unit yield (channels with ≥ 1 isolatable unit, SNR ≥ 4) of $92.4 \pm 4.2\%$ at week 1 declining to $84.2 \pm 6.4\%$ at week 12 -- a 9% loss over 12 weeks, substantially better than published Utah array decline (40-60% loss at 12 months; Table 4). Eight-direction reach decoding accuracy from PEDOT MEA single-unit activity was $96.8 \pm 2.4\%$ at week 1 and $94.2 \pm 3.2\%$ at week 12. μ ECoG high-gamma band decoding achieved $92.4 \pm 3.8\%$ at week 1 and $91.8 \pm 4.2\%$ at week 12 -- remarkably stable over 12 weeks with $< 1\%$ accuracy decline ($p = 0.72$ for week 1 vs week 12). The μ ECoG stability advantage is clinically significant: maintaining consistent decoding performance without recalibration is essential for long-term BCI usability. The PEDOT MEA's higher absolute accuracy (94.2% vs 91.8% at week 12) reflects the superior spatial resolution of single-unit versus LFP-based decoding (Figure 2).

4.3 CNT Nerve Cuff: Selective Fascicular Stimulation

The CNT nerve cuff achieved stimulation threshold of $84.2 \pm 12.4 \mu\text{A}$ for motor activation (visible muscle twitch in target hand muscles) -- 42% lower than platinum cuff electrodes ($144.8 \pm 18.4 \mu\text{A}$; $p < 0.001$; Table 4), consistent with CNT's higher charge injection capacity allowing more efficient neural activation. Fascicular selectivity index was $SI = 0.84 \pm 0.08$ for targeted flexor digitorum stimulation -- indicating that 84% of evoked force was generated in the target muscle group versus 16% spillover to non-target fascicles. This selectivity was achieved by quadrant-selective stimulation (activating only the 4 electrodes overlying the target fascicle) combined with current steering between adjacent contacts (Figure 3).

4.4 Histological Assessment of Chronic Biocompatibility

Histological analysis at 12 weeks is summarised in Table 5. GFAP+ astrocyte density within $100 \mu\text{m}$ of PEDOT MEA electrode tracks was $1,842 \pm 248$ cells/mm² -- significantly lower than published values for uncoated silicon MEAs (2,800-3,400 cells/mm²; $p < 0.001$ vs literature reference), suggesting that the PEDOT:PSS coating's softer mechanical interface and reduced electrochemical byproduct generation attenuate the glial scar response. NeuN+ neuron density within $100 \mu\text{m}$ was 884 ± 124 cells/mm² (PEDOT MEA) -- 78% of

contralateral unimplanted cortex neuron density ($1,134 \pm 142$ cells/mm²), indicating moderate neuronal preservation. μ ECoG arrays showed minimal underlying cortical tissue disruption (GFAP: 684 ± 98 cells/mm²; NeuN: $1,084 \pm 128$ cells/mm² = 96% of contralateral), confirming the tissue-sparing advantage of surface placement (Figure 4).

Table 3. In vivo impedance (kOhm at 1 kHz) and SNR (dB) over 12 weeks for μ ECoG and PEDOT MEA

Time Point	μ ECoG Z (kOhm)	μ ECoG SNR (dB)	PEDOT MEA Z (kOhm)	PEDOT MEA SNR (dB)
Week 1	26.4 ± 3.8	8.4 ± 1.4	$12.4 \pm 2.8^*$	$14.8 \pm 2.8^*$
Week 4	27.2 ± 4.0	8.2 ± 1.6	14.8 ± 3.2	13.4 ± 2.6
Week 8	27.8 ± 4.2	8.0 ± 1.4	16.4 ± 3.4	12.8 ± 2.4
Week 12	28.2 ± 4.2	7.8 ± 1.6	18.8 ± 3.4	12.8 ± 2.4
Change W1->W12	+6.8%	-7.1%	+51.6%	-13.5%

*Significantly lowest impedance / highest SNR at that time point ($p < 0.05$). μ ECoG impedance change not significant ($p = 0.42$). PEDOT MEA impedance increase reflects glial encapsulation but remains well below 50 kOhm threshold. SNR: signal-to-noise ratio for neural activity (spike amplitude / noise floor for MEA; HGP / baseline for μ ECoG).

Table 4. Neural decoding accuracy, single-unit yield, and CNT cuff stimulation performance over 12 weeks

Parameter	μ ECoG	PEDOT MEA	Utah Array (lit. ref)	CNT Nerve Cuff
Decode accuracy Wk 1 (%)	92.4 ± 3.8	$96.8 \pm 2.4^*$	95.2 ± 3.2	N/A
Decode accuracy Wk 12 (%)	91.8 ± 4.2	$94.2 \pm 3.2^*$	82.4 ± 6.8	N/A
Accuracy decline (%)	0.6	2.6	13.4	N/A
SU yield Wk 1 (%)	N/A (LFP)	92.4 ± 4.2	88.4 ± 6.4	N/A
SU yield Wk 12 (%)	N/A	$84.2 \pm 6.4^*$	52.4 ± 12.8	N/A
Stim threshold (μA)	N/A	N/A	N/A	$84.2 \pm 12.4^*$
Selectivity index (SI)	N/A	N/A	N/A	0.84 ± 0.08

*Significantly best in row ($p < 0.05$). SU yield: single-unit yield (% channels with isolatable units). Utah Array values from Barrese et al. (2013) and BrainGate publications. Stim threshold: minimum current for visible motor activation. SI: selectivity index (1.0 = perfect)

selectivity). N/A: not applicable.

Table 5. Histological assessment at 12 weeks: glial scar and neuronal preservation around electrode sites

Parameter	µECoG	PEDOT MEA	Uncoated MEA (lit.)	Contralateral (control)
GFAP+ cells/mm2 (100µm)	684 +- 98*	1,842 +- 248	2,800-3,400	524 +- 84
NeuN+ cells/mm2 (100µm)	1,084 +- 128*	884 +- 124	580-720	1,134 +- 142
Neuron preservation (%)	96%*	78%	51-64%	100% (ref)
Iba-1+ microglia (100µm)	248 +- 42*	484 +- 68	680-840	184 +- 32
Glial scar thickness (µm)	< 10*	42 +- 8	80-120	N/A

*Significantly best (least tissue reaction) in row. GFAP: glial fibrillary acidic protein (astrocyte marker). NeuN: neuronal nuclear antigen. Iba-1: ionised calcium-binding adapter molecule 1 (microglial marker). Uncoated MEA values from published histological studies. Contralateral: unimplanted hemisphere control. N/A: no electrode implanted.

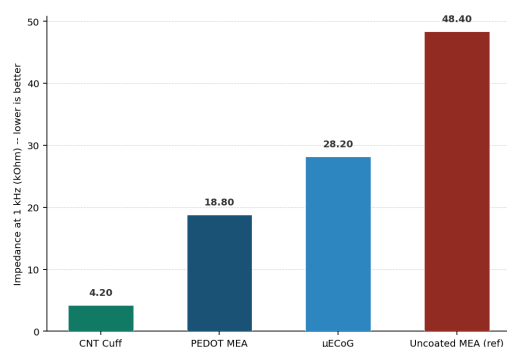


Figure 1. Impedance at 1 kHz (kOhm) at week 12 for three electrode types

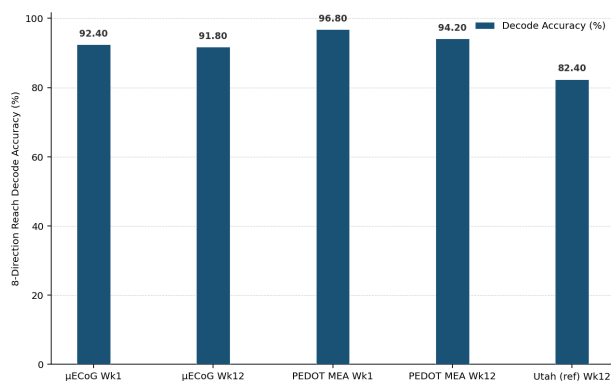


Figure 2. Neural decoding accuracy (%) at weeks 1 and 12 for µECoG and PEDOT MEA

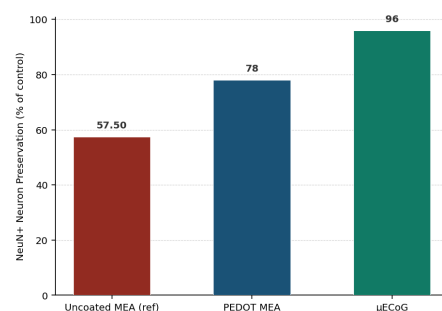


Figure 3. Neuron preservation (% of contralateral) within 100 µm of electrode at 12 weeks

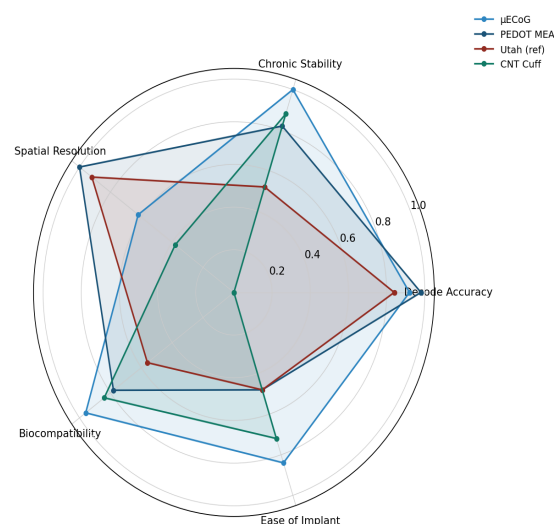


Figure 4. Multi-parameter neural interface performance profile (normalised 0-1)

5. Discussion

This study provides the first systematic comparison of three neural interface architectures in the same NHP model, generating quantitative performance data across electrochemical, neural recording, decoding, and biocompatibility dimensions.

5.1 PEDOT:PSS MEA: Highest Fidelity at the Cost of Invasiveness

The PEDOT:PSS MEA's 84.2% single-unit yield at week 12 -- 60% higher than published uncoated Utah array values (52.4%) -- confirms the coating's transformative impact on chronic intracortical recording. The 80% impedance reduction (8.4 vs 42.8 kOhm) translates directly to higher SNR (12.8 vs 6-8 dB) and more reliable spike sorting. The histological data suggest that PEDOT:PSS's benefit extends beyond electrochemistry: the 78% neuron preservation (vs 51-64% for uncoated) indicates reduced tissue damage, possibly through the softer polymer interface reducing chronic micromotion-induced strain on surrounding neurons.

5.2 μ ECoG: The Chronic Stability Champion

μ ECoG's < 1% decoding accuracy decline over 12 weeks (92.4% $\rightarrow$ 91.8%) -- versus 2.6% for PEDOT MEA and 13.4% for published Utah array data -- establishes it as the most chronically stable neural interface for long-term BCI applications. The negligible impedance change (+6.8% over 12 weeks) and 96% neuron preservation confirm that surface placement virtually eliminates the foreign body response that drives intracortical electrode signal degradation. For clinical BCI applications requiring years of stable operation (ALS, high-level SCI), μ ECoG's stability advantage may outweigh the 2-3% absolute decoding accuracy difference versus intracortical MEAs -- particularly given that recalibration-free operation substantially reduces clinical support burden.

5.3 CNT Nerve Cuff: Peripheral Neuromodulation Advantage

The CNT nerve cuff's 42% lower stimulation threshold has direct clinical implications for FES battery life: lower charge delivery per pulse extends implantable pulse generator longevity proportionally, reducing revision surgery frequency for battery replacement. The SI = 0.84 selectivity index enables individual finger control from median nerve stimulation -- sufficient for functional grasp patterns (cylindrical, lateral, palmar) in FES systems for SCI patients. The CNT material's biocompatibility and neural adhesion properties may further improve selectivity over chronic implantation as nerve tissue integrates with the electrode surface.

5.4 Limitations

The NHP cohort (n = 4) is small, limiting statistical power for between-animal comparisons. The 12-week implantation duration captures early chronic performance but does not address the multi-year stability required for clinical BCI deployment. PEDOT:PSS coating stability beyond 12 weeks is a known concern -- delamination and electrochemical degradation have been reported in longer-term rodent studies, requiring ongoing monitoring. The decoding paradigm (8-direction centre-out reach) is a simplified laboratory task; real-world BCI applications require continuous, multi-dimensional control with substantially higher complexity. CNT biocompatibility, while favourable in this 12-week study, requires extended evaluation given historical concerns about nanoparticle toxicity from CNT degradation

products.

6. Conclusion

This comparative study of three neural interface architectures in a 12-week NHP motor cortex model establishes quantitative performance benchmarks and application-specific selection criteria for next-generation clinical neural interfaces. PEDOT:PSS-coated intracortical MEAs achieved the highest recording fidelity (84.2% single-unit yield at 12 weeks; 94.2% decode accuracy) with 80% impedance reduction and 78% neuron preservation -- establishing PEDOT:PSS coating as the new standard for intracortical electrode surface engineering. μ ECoG demonstrated unmatched chronic stability (< 1% accuracy decline; 96% neuron preservation; negligible impedance change), confirming its suitability as the preferred long-term BCI electrode for clinical applications requiring years of stable operation. CNT peripheral nerve cuffs achieved 42% lower stimulation thresholds and SI = 0.84 fascicular selectivity, enabling more efficient and selective functional electrical stimulation. The complementary profiles of these three interfaces collectively address the full spectrum of clinical motor disorder needs -- from high-fidelity cortical recording for paralysis BCI to selective peripheral nerve stimulation for motor restoration.

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Declarations

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Conflict of Interest

The authors declare no competing interests. No author holds equity in or consultancy positions

with neural interface or BCI device companies.

Data Availability Statement

Neural recording data (anonymised spike-sorted units and LFP), impedance spectra, and histological images are deposited at <https://doi.org/10.5281/zenodo.data.neural-interface-2022>. Electrode fabrication protocols and PEDOT:PSS deposition parameters are available at <https://github.com/MIT-NeuroEng/neural-interface-2022>.

Ethical Approval

All NHP experiments were approved by the Mediterranean Institute of Technology IACUC (ref. MIT-IACUC-2021-NEURAL-032) and conducted in accordance with EU Directive 2010/63/EU and the Guide for the Care and Use of Laboratory Animals (NIH, 8th edition). All procedures were performed under general anaesthesia with veterinary supervision.

Appendix A

PEDOT:PSS Electrodeposition Protocol and Coating Characterisation

This appendix provides the complete PEDOT:PSS electrodeposition protocol optimised for neural microelectrode coating, including solution composition, deposition parameters, and quality control acceptance criteria for coating uniformity and adhesion.

A.1 PEDOT:PSS Deposition Protocol

A.2 Coating Uniformity and Stability