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Research Article

The Stentrode™ System by Synchron: Architectural Design and Clinical Translation of a Minimally Invasive Brain–Computer Interface

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Abstract

The Stentrode™ system, developed by Synchron, introduces a minimally invasive endovascular architecture for brain–computer interfacing by enabling chronic electrocorticographic recording from within the superior sagittal sinus adjacent to the motor cortex. This review analyzes the system's engineering framework, emphasizing the integration of mechanical design, materials science, signal conditioning, and chronic biostability. The device consists of a self-expanding nitinol stent embedded with thin-film platinum–iridium electrodes, delivered transvenously via the internal jugular vein and deployed without penetrating cortical tissue. Neural signals are amplified and digitized through a subcutaneous telemetry unit designed for low-noise, high-fidelity acquisition, particularly in the high-gamma band. Preclinical and clinical data confirm endothelialization of the implant, preserving venous patency and ensuring long-term signal stability. While the system sacrifices the spatial resolution of intracortical arrays, it offers a lower-risk, outpatient-compatible alternative that redefines the trade space between signal fidelity, procedural complexity, and translational scalability. By shifting the neurointerface paradigm from neurosurgical to endovascular access, the Stentrode™ establishes a new class of BCIs optimized for chronic deployment with minimal disruption to neural architecture.

Keywords: neurosurgical; endovascular; brain

1.Introduction

Brain–computer interfaces (BCIs) are rapidly evolving from experimental tools into viable clinical systems for restoring lost motor and communicative function. Historically, BCI platforms have prioritized signal fidelity by employing intracortical or subdural electrode arrays. However, these approaches often necessitate open-brain surgery and carry risks related to tissue injury, inflammation, and long-term device stability. In response, alternative architectures that seek to reduce procedural burden while preserving meaningful neural signal acquisition are being explored.

The Stentrode™ system represents one such architectural innovation: a fully implantable, endovascular neural interface capable of recording brain activity without penetrating cortical tissue. By integrating a flexible electrode array onto a self-expanding stent structure, the device can be deployed through the venous system and positioned near eloquent cortical areas, enabling neural recording through the vessel wall. This paradigm shift in neurointerface design offers the potential for broader clinical adoption by leveraging well-established endovascular techniques and minimizing neurosurgical complexity.

This manuscript provides a detailed engineering analysis of the Stentrode™ system, emphasizing its mechanical configuration, materials selection, signal processing pipeline, and long-term compatibility with the neurovascular environment. In doing so, it aims to elucidate the design considerations that distinguish this platform from more invasive BCI systems and clarify the trade-offs in achieving chronic functionality within a minimally invasive framework.

2.0 Methods

This review synthesizes scientific and technical sources published between 2019 and 2025, including peer-reviewed articles, clinical trial reports, conference proceedings, and regulatory filings. Targeted searches

were conducted via PubMed, IEEE Xplore, bioRxiv, and Google Scholar using terms such as "Stentrode," "endovascular brain–computer interface," "venous neural implant," and "Synchron BCI system." Medical Subject Headings (MeSH) were applied where appropriate to refine results.

The methodology was structured to capture the full engineering stack and translational evolution of the Stentrode™ system, with emphasis on thin-film neural electronics, mechanical compliance, wireless telemetry, and long-term endothelial integration. Priority was given to sources detailing device architecture, deployment techniques, signal acquisition pathways, and chronic biocompatibility.

Where peer-reviewed data were limited due to proprietary constraints, supplementary information was drawn from company white papers, FDA documentation, and investigator-authored briefings. All technical claims were cross-validated across multiple sources, and references with persistent identifiers—such as DOIs or stable institutional URLs—were prioritized to ensure transparency and reproducibility.

3.0 Device Architecture

The Stentrode™ system, developed by Synchron, represents a paradigm shift in brain–computer interface (BCI) design by leveraging endovascular access to the cerebral cortex. Unlike traditional intracortical or subdural systems that require craniotomy, the Stentrode is deployed via the venous system, specifically the superior sagittal sinus (SSS), to record neural activity from the motor cortex [1]. This approach minimizes surgical morbidity while maintaining sufficient signal fidelity for motor decoding applications.

The architecture of the system comprises three primary components:

- A self-expanding nitinol stent scaffold serves as the mechanical backbone and delivery vehicle.
- A thin-film electrode array embedded along the stent's luminal surface for neural signal acquisition.
- A subcutaneous telemetry unit that digitizes, powers, and wirelessly transmits neural data to external processing hardware [2].

This section focuses on the engineering principles, materials science, and neurovascular interface dynamics underpinning Stentrode's functionality.

3.1 Stentrode Scaffold and Electrode Design

The mechanical core of the system is the nitinol stent scaffold. Nitinol, a near-equiatomic nickel–titanium alloy, is selected for its superelastic and shape memory properties, enabling it to undergo significant compressive deformation during catheter-based deployment and recover its original geometry intravascularly. A reversible martensitic phase transformation governs this thermomechanical behavior, and at normothermic physiological temperatures, nitinol resides in its austenitic phase. Its ability to maintain stable radial force within highly compliant venous environments ensures that the stent conforms closely to the wall of the superior sagittal sinus without exerting excessive pressure that might cause venous stenosis or flow disruption [3].

The stent's final deployed dimensions—approximately 40 mm in length and 8 mm in diameter—have been optimized to align anatomically with the posterior SSS overlying the primary motor cortex. Laser-cut from drawn nitinol tubing and subjected to electropolishing, the scaffold features a repeating open-cell geometry that provides both flexibility and circumferential anchoring. Finite element modeling simulations have confirmed that radial and hoop stresses remain within safe margins under pulsatile hemodynamic loading, even during Valsalva maneuvers or other transient intracranial pressure elevations [4].

Sixteen platinum-iridium electrodes are embedded along the stent's luminal surface, each coated with iridium oxide to enhance charge injection capacity and reduce electrode polarization. The selection of platinum–iridium (typically 90:10 or 80:20) is predicated on its high corrosion resistance, inert electrochemical behavior, and historical biocompatibility in neural stimulation applications. Iridium oxide, applied via sputtering or reactive deposition, further increases the effective surface area of the electrode through nanoroughness and promotes capacitive coupling for stable signal acquisition across a broad frequency spectrum [5].

These electrodes are lithographically patterned onto a polyimide film substrate using standard MEMS thin-film processes. The layout includes gold or platinum metal traces insulated by biocompatible dielectrics, such as parylene-C, to prevent electrical crosstalk and signal leakage. The flexible electrode array is subsequently wrapped around the interior curvature of the stent and adhesively bonded, ensuring that electrode contact sites are circumferentially distributed and maintained in close apposition to the venous endothelium following deployment [6].

Following implantation, the device undergoes a natural endothelialization process wherein the stent struts and electrode surfaces become enveloped by migrating endothelial cells, typically within four weeks. Preclinical studies in ovine models—commonly used due to their comparable cortical hemodynamics—demonstrated that endothelialization stabilizes the electrode–vessel interface without inducing thrombus formation or intimal hyperplasia. Histological analyses confirmed the preservation of endothelial integrity, absence of lymphocytic infiltration, and sustained vessel patency up to 190 days post-implantation. These findings are critical for long-term device stability, as they confirm that the neurovascular interface remains biologically inert and mechanically robust under chronic conditions [7].

To mitigate early-stage thromboembolic risk, Synchron employs a dual antiplatelet therapy regimen consisting of aspirin and clopidogrel for the first 90 days, followed by aspirin monotherapy [8]. Computational hemodynamic modeling supports this clinical protocol, showing that shear stress along the stent–vessel interface remains within thresholds that discourage platelet aggregation, particularly around electrode junctions. This is especially important given the device's proximity to cortical draining veins, where stagnant flow or turbulent eddies could otherwise pose an embolic risk [6].

Altogether, the Stentrode scaffold and electrode array design represents a balance between neuroanatomical access, mechanical compliance, electrochemical performance, and chronic biostability. The convergence of thin-film MEMS fabrication with endovascular deployment strategy establishes a viable paradigm for minimally invasive neurosensing and offers a compelling alternative to traditional intracortical or subdural BCI platforms.

3.2 Lead Routing and Telemetry Unit Design

A critical component of the Stentrode™ system's functionality lies in the design and integration of its lead routing and subcutaneous telemetry unit, which together form the conduit between the intravascular electrode array and external signal processing infrastructure. This subsystem must satisfy a complex set of engineering constraints: it must preserve signal fidelity, withstand chronic mechanical stress, ensure biocompatibility, and support wireless data transmission, all within the anatomical and physiological limitations of the human body.

The electrode array is connected to a flexible, insulated lead that traverses the venous system via the internal jugular vein. This lead is tunneled subcutaneously and terminates in a titanium-encased implantable receiver–transmitter unit (IRTU) housed in a subclavicular pocket. The lead itself is constructed from helically wound, biocompatible conductors embedded in a silicone or polyurethane sheath. It accommodates repetitive neck and shoulder motion without inducing mechanical fatigue or insulation breach. Strain-relief loops are incorporated at the venous exit site and the IRTU interface to mitigate tensile loading and reduce the risk of lead fracture or migration over time [1][8].

The IRTU performs several essential functions. First, it digitizes the analog neural signals received from the electrode array using low-noise amplification and analog-to-digital conversion (ADC) circuitry. The ADC operates at a sampling rate sufficient to capture high-gamma ECoG activity (typically ≥1 kHz per channel), with resolution optimized to preserve signal-to-noise ratio while minimizing power consumption. Second, the IRTU handles wireless telemetry, transmitting digitized neural data to an external telemetry unit (ETU) via a secure, low-power radiofrequency (RF) link. The system employs Bluetooth Low Energy (BLE) protocols with custom firmware to ensure robust packet delivery, forward error correction, and low-latency communication suitable for real-time decoding applications[9].

Power is delivered to the IRTU via inductive coupling from the ETU, which is worn externally over the subclavicular region. The ETU contains a primary coil that transmits energy at a resonant frequency (typically 6.78 MHz), which is received by a secondary coil embedded in the IRTU. This wireless power transfer eliminates the need for percutaneous connectors, thereby reducing infection risk and improving patient comfort. The IRTU includes onboard energy storage capacitors to buffer transient power fluctuations and ensure continuous operation during brief misalignments of the external coil [10].

Thermal management is a nontrivial design consideration. The IRTU's power budget must remain below thresholds that could induce local tissue heating, typically constrained to <2°C above baseline to comply with ISO 14708-1 standards for implantable devices. Finite element thermal modeling has been employed to simulate heat dissipation under worst-case load conditions, confirming that the device remains within safe operating limits during continuous telemetry and signal acquisition [6].

The entire telemetry subsystem is hermetically sealed using laser-welded titanium enclosures with ceramic feedthroughs for electrical isolation. This packaging strategy ensures long-term biostability and resistance to fluid ingress, with accelerated aging studies predicting operational lifetimes exceeding five years. The IRTU's form factor—approximately 50 × 30 × 8 mm—has been optimized for subcutaneous implantation without impinging on adjacent anatomical structures such as the clavicle, brachial plexus, or subclavian vessels [1][8].

Together, the lead and telemetry unit form a robust, fully implantable interface that bridges the intravascular neural recording array with external computational infrastructure. Their design reflects a careful balance between mechanical resilience, electrical performance, and biological integration, enabling the Stentrode™ system to extend the user's neural intent seamlessly.

3.3 Endovascular Deployment Methodology

The deployment of the Stentrode™ system leverages established neurointerventional techniques to access the cerebral venous system and position the electrode array adjacent to the motor cortex. This approach is modeled on procedures routinely performed in interventional neuroradiology, such as venous sinus stenting for idiopathic intracranial hypertension or embolization of dural arteriovenous fistulas. The procedure is typically performed under general anesthesia in a hybrid angiography suite equipped with biplanar fluoroscopy and digital subtraction angiography (DSA) capabilities.

Vascular access is obtained via percutaneous puncture of the right internal jugular vein using a modified Seldinger technique. A 9 French introducer sheath is advanced under ultrasound guidance, and systemic anticoagulation is initiated with intravenous heparin to maintain an activated clotting time (ACT) of 250–300 seconds. A guide catheter is then navigated superiorly through the jugular bulb into the transverse sinus and ultimately into the superior sagittal sinus (SSS), which overlies the precentral gyrus of the dominant hemisphere [1][8].

Once the target segment of the SSS is reached, a microcatheter–microwire system is used to deliver the Stentrode™ device. The stent-electrode array is preloaded into a delivery catheter and constrained in a compressed configuration. Under continuous fluoroscopic visualization, the device retracts the outer sheath, allowing the nitinol scaffold to self-expand and anchor against the venous endothelium. The deployment is performed with submillimeter precision, guided by preoperative MRI-venography fusion imaging and intraoperative roadmap overlays to ensure alignment with the cortical motor strip [9].

Device positioning is confirmed using contrast-enhanced DSA, which delineates the stent's radiopaque markers and verifies patency of the SSS. Intraoperative cone-beam CT may assess three-dimensional conformation and proximity to cortical landmarks. Once deployed, the electrode lead is tunneled subcutaneously through the neck and chest wall to the subclavicular pocket, which is connected to the implantable receiver–transmitter unit (IRTU). The venous access site is closed using a figure-of-eight suture or vascular closure device, and hemostasis is confirmed prior to extubation[11].

Intraoperative monitoring includes impedance testing of each electrode channel to verify electrical continuity and detect potential short circuits or open leads. Baseline neural signals may be recorded intraoperatively to confirm functional contact with the cortical surface. The entire procedure typically lasts 90–120 minutes, with the endovascular phase requiring less than 30 minutes of fluoroscopy time. Compared to traditional craniotomy-based BCI implantation, this approach significantly reduces operative time, blood loss, and postoperative recovery duration [5].

Post-deployment imaging at 3 and 12 months has demonstrated submillimeter positional stability of the device, with no evidence of migration, thrombosis, or venous occlusion[12]. The minimally invasive nature of the procedure, reproducibility, and compatibility with standard neurointerventional workflows underscore its potential for widespread clinical adoption.

4.0 Signal Conditioning and Architectural Constraints

The fidelity of neural signal acquisition in the Stentrode™ system is governed by electrode design and placement and by the architecture of its analog front-end (AFE), digitization pipeline, and packaging constraints imposed by its fully implantable form factor. These components must operate within stringent power, thermal, and spatial budgets while preserving the spectral and temporal integrity of electrocorticographic (ECoG) signals recorded from within the superior sagittal sinus. This section thoroughly examines the signal conditioning architecture, emphasizing the trade-offs inherent in implantable neural electronics.

4.1 Analog Front-End Design

The AFE is the first stage in the signal chain and is responsible for amplifying low-amplitude neural signals—typically in the range of 10–100 µV peak-to-peak—while rejecting common-mode noise and preserving bandwidth. Each electrode channel is connected to a low-noise amplifier (LNA) with an input-referred noise floor below 2.5 µV_rms, optimized for the 1–300 Hz frequency band that encompasses motor-related µ (8–13 Hz), β (13–30 Hz), and low-γ (30–100 Hz) oscillations [9][13].

The LNA employs a capacitively coupled architecture with chopper stabilization to achieve high input impedance (>1 GΩ) and minimize signal attenuation across the electrode–tissue interface. This technique modulates the input signal to a higher frequency band where 1/f noise and DC offset drift are less pronounced, then demodulates it post-amplification. The result significantly reduces low-frequency noise and thermal drift, which is critical for stable long-term decoding performance [14].

The AFE also incorporates a programmable gain amplifier (PGA) stage, allowing dynamic channel gain adjustment to accommodate inter-subject variability in signal amplitude and electrode impedance. This adaptability is crucial in endovascular systems, where signal transduction occurs across the venous endothelium and may be influenced by vascular remodeling or fibrin sheath formation over time [7].

4.2 Analog-to-Digital Conversion and Sampling Constraints

Following amplification, neural signals are digitized using a low-power analog-to-digital converter (ADC) integrated within the implantable receiver–transmitter unit (IRTU). The ADC operates at a sampling rate of 1 kHz per channel with 12- to 16-bit resolution, balancing the need for temporal precision with constraints on data throughput and implant power consumption [5].

The choice of sampling rate reflects a compromise between capturing high-gamma activity—often associated with motor intention—and minimizing aliasing and quantization noise. Anti-aliasing filters are implemented prior to digitization using low-pass finite impulse response (FIR) filters with a cutoff near 400 Hz, ensuring compliance with the Nyquist criterion while preserving signal bandwidth [13][14].

The digitized signals undergo on-chip compression to reduce data volume and transmission latency using delta encoding or predictive coding algorithms. These methods exploit the temporal correlation of neural signals to encode only the difference between successive samples, thereby reducing the bit rate without significant loss of information. This is essential for maintaining real-time performance over the Bluetooth Low Energy (BLE) telemetry link, which has a practical upper limit of ~1 Mbps in implantable configurations [15].

4.3 Electromagnetic and Motion Artifact Mitigation

The endovascular location of the Stentrode™ confers several advantages in artifact suppression. The superior sagittal sinus is shielded from scalp muscle artifacts and environmental electromagnetic interference (EMI), which commonly degrade signal quality in surface EEG systems. However, the system remains susceptible to motion-induced artifacts, particularly cardiac pulsatility and respiratory excursions [16].

To mitigate these effects, the AFE incorporates differential amplification with a high common-mode rejection ratio (CMRR > 100 dB), and the signal processing pipeline includes adaptive filtering algorithms that subtract out periodic artifacts synchronized to physiological rhythms. Additionally, the mechanical compliance of the stent and lead reduces micromotion at the electrode–endothelium interface, further stabilizing the signal baseline [16].

5.4 Thermal and Power Constraints

To prevent tissue heating, all signal conditioning components must operate within a strict thermal envelope. The total power budget for the implant is constrained to <10 mW, with the AFE and ADC accounting for the majority of this consumption. Thermal modeling studies have demonstrated that, under continuous operation, the implant surface temperature remains <1.5°C above baseline, well within ISO 14708-1 safety thresholds [6].

The system employs duty-cycling strategies and low-leakage CMOS design techniques to achieve this. Power gating deactivates unused channels or reduces sampling rates during idle periods, extending battery life and reducing thermal load. These architectural decisions are critical for enabling chronic implantation without active cooling or frequent recharging [6].

5.0 Biostability and Long-Term Structural Compatibility

The chronic viability of the Stentrode™ system hinges on its ability to maintain structural integrity and electrochemical performance within the neurovascular environment over extended periods. Unlike intracortical or subdural interfaces, which often contend with gliosis and fibrotic encapsulation, the endovascular approach leverages the natural healing dynamics of the venous endothelium to stabilize the implant. This section synthesizes preclinical and clinical findings that characterize the device architecture's long-term mechanical, biological, and electrochemical compatibility [5].

5.1 Endothelialization and Vascular Integration

Following deployment in the superior sagittal sinus, the Stentrode™ undergoes rapid endothelialization, a process by which endothelial cells migrate over the stent struts and electrode surfaces to form a continuous, non-thrombogenic lining. In ovine models, complete endothelial coverage has been observed within 28 days, confirmed via scanning electron microscopy and histological staining for CD31 and von Willebrand factor [7]. This biological encapsulation is essential for anchoring the device, minimizing micromotion, and reducing the risk of thrombus formation or neointimal hyperplasia [17].

Computational fluid dynamics (CFD) simulations have demonstrated that the stent’s open-cell nitinol geometry maintains wall shear stress above the critical threshold (~0.4 Pa) required to support endothelial quiescence and alignment [6]. These simulations also show that the strut spacing and curvature avoid regions of recirculating flow, which promote platelet aggregation and fibrin deposition [4].

5.2 Mechanical Strain and Lead Fatigue

The Stentrode™ must endure the mechanical demands of the cerebral venous system, which is subject to pulsatile flow, respiratory-induced pressure fluctuations, and cervical motion. The nitinol scaffold exhibits superelastic behavior, allowing it to accommodate cyclic deformation without plastic fatigue. Mechanical testing has shown that the stent withstands over 10^8 loading cycles without fracture, delamination, or loss of radial force [4].

The electrode lead, which traverses the jugular vein and subcutaneous tissues to reach the subclavicular implant site, is engineered with helically wound conductors and strain-relief loops to mitigate tensile stress. Accelerated aging and mechanical cycling studies have demonstrated that

the lead maintains electrical continuity and insulation integrity after simulated multi-year use [5]. In the COMMAND trial, no lead fractures, telemetry failures, or migration events were reported over 12-month follow-up intervals [8].

5.3 Histological and Immunological Response

Histological analysis of explanted devices from ovine models has revealed minimal perivascular inflammation or foreign body response. Hematoxylin and eosin (H&E) staining showed intact venous architecture and preserved endothelial monolayers, while immunohistochemistry for CD68 and CD45 indicated negligible macrophage or lymphocyte infiltration [7]. These findings suggest the device elicits a low-grade immune response, consistent with other nitinol-based endovascular implants.

Clinical data from the SWITCH and COMMAND trials corroborate these findings. Across all participants had no device-related serious adverse events (SAEs) such as venous thrombosis, hemorrhage, or infection. Imaging follow-up using MR venography and contrast-enhanced CT confirmed stable device positioning and vessel patency up to 12 months post-implantation [8].

5.4 Electrochemical Stability and Signal Integrity

The electrochemical performance of the Stentrode™ electrodes remains stable over chronic implantation periods. In vivo impedance spectroscopy has shown that electrode impedance stabilizes within the first two weeks post-implantation and remains within the functional range (10–50 kΩ at 1 kHz) thereafter [16]. This plateau corresponds to the completion of endothelialization and establishing a stable electrode–tissue interface.

Signal-to-noise ratios (SNRs) for motor-related ECoG signals have remained consistent over multi-month recordings, with no significant channel dropout or baseline drift. These metrics are critical for maintaining decoder performance in assistive communication applications and suggest that the device's structural and electrochemical interfaces are robust to chronic implantation [14].

6.0 Comparative Discussion: Architectural Trade-offs

The architectural design of the Stentrode™ system reflects a deliberate prioritization of surgical accessibility and chronic biostability over spatial resolution and channel density. This trade-off distinguishes it from intracortical and subdural brain–computer interface (BCI) platforms, which offer higher signal granularity but at the cost of increased procedural complexity and long-term biological risk. Understanding these trade-offs is essential for contextualizing the Stentrode's engineering philosophy and translational trajectory.

6.1 Invasiveness vs. Signal Resolution

Intracortical systems such as the Utah array or Neuralink's N1 platform achieve single-unit resolution by penetrating cortical layers II–V with microneedle electrodes. These architectures enable high-bandwidth decoding of motor intention and speech-related activity but require craniotomy and cortical penetration, which introduce risks of hemorrhage, gliosis, and long-term signal degradation due to reactive tissue encapsulation [18]. In contrast, the Stentrode™ records electrocorticographic (ECoG) signals from within the superior sagittal sinus, avoiding direct cortical contact. While this limits spatial resolution to the mesoscale (millimeter-level), it preserves the blood–brain barrier and reduces the risk of neuroinflammatory sequelae [7].

Preclinical comparisons have shown that the Stentrode™ achieves signal-to-noise ratios and bandwidths comparable to subdural ECoG arrays, particularly in the high-gamma band (70–150 Hz), critical for motor decoding [14]. However, it lacks the laminar specificity and spike resolution of penetrating arrays, making it less suitable for applications requiring fine-grained neural discrimination, such as speech synthesis or high-dimensional prosthetic control [19].

6.2 Scalability and Channel Density

The Stentrode™ currently supports 16 recording channels, constrained by the physical dimensions of the venous sinus and the need to maintain hemodynamic patency. In contrast, subdural platforms such as Precision Neuroscience's Layer 7 array offer up to 4,096 channels across an 8 cm^2 footprint, and intracortical systems like Paradromics' Connexus platform scale to 6,400+ channels via modular silicon shanks [20]. These high-density systems enable population-level decoding and spatiotemporal mapping of cortical ensembles but require extensive cranial access and complex implantation workflows.

Efforts to scale the Stentrode™ architecture must contend with vascular anatomy, electrode miniaturization, and telemetry bandwidth. While future iterations may incorporate multiplexed leads or endovascular branching arrays, the current design is optimized for low-channel, high-safety applications such as binary click selection and directional cursor control [8].

6.3 Manufacturability and Deployment Complexity

From a manufacturing standpoint, the Stentrode™ benefits from its compatibility with established stent fabrication workflows and MEMS-based thin-film electrode processing. Laser-cut nitinol scaffolds and polyimide-based electrode arrays allow batch fabrication and quality control within existing medical device supply chains [9]. In contrast, intracortical arrays often require manual assembly, micromachining, and cleanroom-intensive processes that limit scalability and increase cost.

Deployment of the Stentrode™ is performed via standard neurointerventional techniques, requiring only fluoroscopic guidance and jugular vein access. This contrasts sharply with the neurosurgical infrastructure required for craniotomy-based systems, which may involve stereotactic navigation, robotic insertion platforms, and intraoperative electrocorticography. The procedural simplicity of the Stentrode™ reduces operative time, anesthesia exposure, and hospital resource utilization, making it more amenable to outpatient workflows and broader clinical adoption [1].

6.4 Upgradeability and System Modularity

The modularity of the Stentrode™ system is currently limited by its monolithic design and fixed telemetry interface. In contrast, subdural and intracortical platforms are increasingly exploring modular architectures that allow for staged implantation, hardware upgrades, and multi-array configurations. However, the Stentrode's endovascular route offers a unique opportunity for future upgradeability via catheter-based interventions, potentially enabling in situ replacement or augmentation without reoperation [21].

7.0 Conclusion

The Stentrode™ system introduces a novel architectural solution to the chronic brain–computer interfacing challenge by combining thin-film neural electronics with an endovascular delivery platform. Its design reflects a deliberate rebalancing of priorities, trading the micrometer-scale resolution of penetrating arrays for a minimally invasive, biologically stable interface that integrates seamlessly into existing

neurointerventional workflows. The system achieves long-term neural recording without breaching the cortical surface or compromising vascular patency through mechanical compliance, biocompatible materials, and low-power signal processing.

This review has outlined the technical principles underpinning Stentrode's functionality, from stent-based electrode deployment and endothelial integration to real-time telemetry and thermal regulation. Notably, the system's architecture demonstrates that high-fidelity neural data can be acquired within the venous system using scalable manufacturing techniques and outpatient-compatible procedures. While current limitations in channel density and spatial resolution constrain the complexity of decoded outputs, early clinical data validate the platform's feasibility for motor restoration and digital communication applications.

Looking forward, the endovascular route provides a unique foundation for future enhancements, including multiplexed electrode arrays, bidirectional stimulation capabilities, and in situ upgradability. These possibilities position the Stentrode™ not merely as an alternative to traditional BCIs, but as the prototype of a new class of neural interfaces engineered for safety, scalability, and long-term integration within the dynamic physiological environment of the brain.

Declarations

All journal policies and submission guidelines were carefully reviewed to ensure full compliance, and the manuscript has not been previously published or submitted elsewhere. The author declares no conflicts of interest. No human, animal, or plant subjects were involved in this literature review, so ethics approval, participant consent, and studies involving plants are not applicable. Additionally, no personal details, images, or videos of individuals are included, which makes publication consent unnecessary. The research did not receive external funding, and no data or supplementary materials are associated with the manuscript. Grammarly AI was used solely to refine grammar, syntax, and paragraph structure. It did not generate ideas or content, thereby preserving the originality of the work.

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