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Organ-specific bioelectronics for soft tissues

Xiaoyan Liu[1,#], Zhihui Zhang[2,#], Junwei Li[3], Zhixing Ge[1], Shaofei Shen[4], Chwee Teck Lim[1,5,6,*]

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Organ-specific bioelectronics, soft bioelectronic interfaces, stretchable electronics, implantable biosensors, soft tissue interfacing

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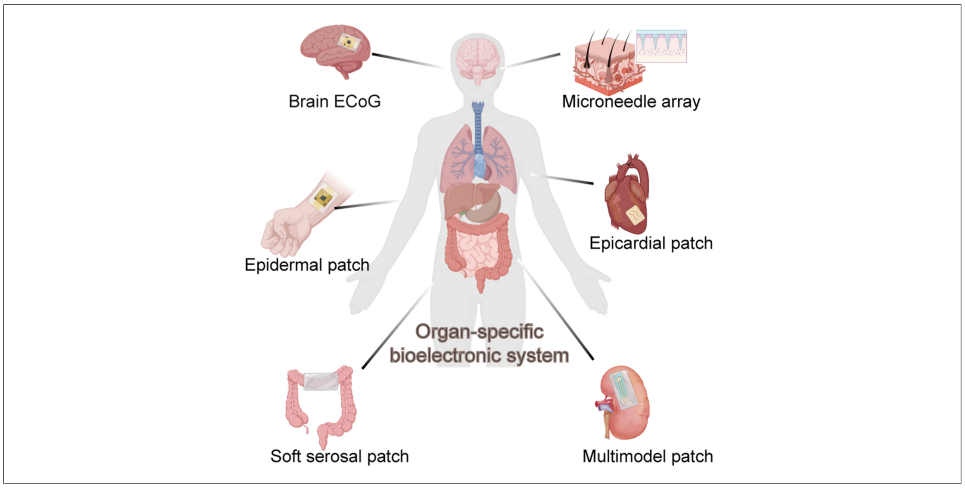

Abstract

Organ function relies on dynamic electrical and electrochemical signaling that governs processes ranging from cardiac conduction and neural activity to gastrointestinal (GI) regulation and endocrine communication. Bioelectronic devices have demonstrated clinical impact in applications such as cardiac pacing, cochlear implants, retinal prostheses, and continuous glucose monitoring. However, when deployed on soft, wet, and continuously moving organs, the long-term stability of the device–tissue interface becomes a key challenge due to mechanical mismatch, biofouling, and degradation in physiological environments. Increasing evidence suggests that universal device architectures are insufficient for reliable long-term operation across organs with distinct mechanical, biochemical, and immunological microenvironments. Organ-specific bioelectronics has therefore emerged as a design paradigm in which materials, device structures, and system architectures are co-optimized according to the deformation modes, chemical conditions, and biological responses of individual tissues. Recent advances include ultracompliant neural interfaces that minimize inflammatory responses, GI resident devices capable of operating under strong peristalsis and chemical exposure, stretchable epidermal

[1]Department of Biomedical Engineering, National University of Singapore, Singapore 117583, Singapore.
[2]School of Biomedical Engineering, Beihang University, Beijing 100191, China.
[3]School of Mechanical and Aerospace Engineering, Nanyang Technological University, Singapore 639798, Singapore.
[4]College of Life Science, Shanxi Agricultural University, Taiyuan 030000, Shanxi, China.
[5]Institute for Health Innovation and Technology, National University of Singapore, Singapore 117599, Singapore.
[6]Mechanobiology Institute, National University of Singapore, Singapore 117411, Singapore.
[#]These authors contributed equally.

***Correspondence to:** Prof. Chwee Teck Lim, Department of Biomedical Engineering, National University of Singapore, Singapore 117583, Singapore. E-mail: ctlim@nus.edu.sg

electronics that seamlessly integrate with skin mechanics, and epicardial or renal surface patches for monitoring visceral organs. This review summarizes recent developments in organ-specific bioelectronics from integrated perspectives of materials, device structures, and biological systems. Key material platforms, fabrication strategies, and representative applications are highlighted, followed by discussion of challenges in long-term biostability, scalable manufacturing, wireless power and data communication, and clinical translation, as well as future opportunities for organ-mimetic electronic interfaces enabling continuous monitoring and therapeutic modulation.

INTRODUCTION

Organ function maintenance and regulation rely on the dynamic variations of electrical signals and electrochemical processes[1-3]. From cardiac conduction to central and peripheral neural activity, and to the gut-brain axis and related endocrine regulation, long-term recording and quantitative analysis of these signals are of major importance for continuous monitoring and intervention[4-6]. Over the past decades, bioelectronic devices have demonstrated clinical value in multiple settings, such as cardiac pacing, cochlear and retinal prostheses, and continuous glucose monitoring[7-9]. These practices indicate that long-term coupling between electronic systems and physiological signals is achievable from an engineering standpoint and have further propelled the development of next-generation implantable and attachable platforms that interface more intimately with organ surfaces.

When devices are deployed onto soft, wet, and continuously moving organ surfaces, the key factor limiting performance and operational lifetime often shifts from the circuitry itself to the long-term stability of the device–tissue interface[10]. Rigid or semi-rigid structures, together with relatively thick encapsulation and interconnects, can mismatch soft tissues in modulus, curvature, and dynamic deformation, thereby leading to electrode delamination, interconnect fatigue failure, and concomitant fluctuations of interfacial impedance with degraded signal quality. Meanwhile, water and ion penetration from body fluids may cause degradation of insulating performance and electrochemical corrosion[11]. Interfacial processes such as protein adsorption and cell adhesion further accelerate biofouling and drive a progressive increase in impedance over time, ultimately manifesting as reduced signal-to-noise ratio (SNR), baseline drift, and time-dependent changes in stimulation threshold[12,13].

Furthermore, the microenvironment varies substantially across organs, implying that interfacial constraints are not uniform. Brain tissue is extremely soft and highly sensitive to immune responses, where even subtle micromotion can trigger tissue injury and glial activation[14]. The gastrointestinal (GI) tract experiences vigorous peristalsis and is continuously exposed to acidic and enzymatic fluids, imposing stringent requirements on encapsulation durability and adhesive stability. Skin interfaces undergo persistent stretching, shear, sweating, and repeated attach–detach cycles. Visceral organs such as the heart, liver, and kidney reside in fluid-rich environments and experience cyclic strain or perfusion-related mechanical fluctuations[15]. Under these conditions, a universal material and structural design is often unable to simultaneously satisfy conformal stability, reliable encapsulation, and low-impedance coupling, which has driven increasing interest in organ-specific bioelectronics strategies.

Organ-specific bioelectronics emphasizes defining design boundary conditions based on the deformation modes, chemical environment, and immune/biofouling behaviors of the target organ, and then co-optimizing the material system, structural form factor, and interfacial chemistry to achieve predictable long-term coupling stability[16]. Representative advances in recent years include ultra compliant neural interfaces that mitigate chronic inflammation and micromotion-induced damage, GI resident platforms capable of operating under weeks of peristalsis and chemical challenge, and stretchable epidermal patches that laminate onto skin with minimal mechanical loading[10,17] [Figure 1]. Despite differences in target

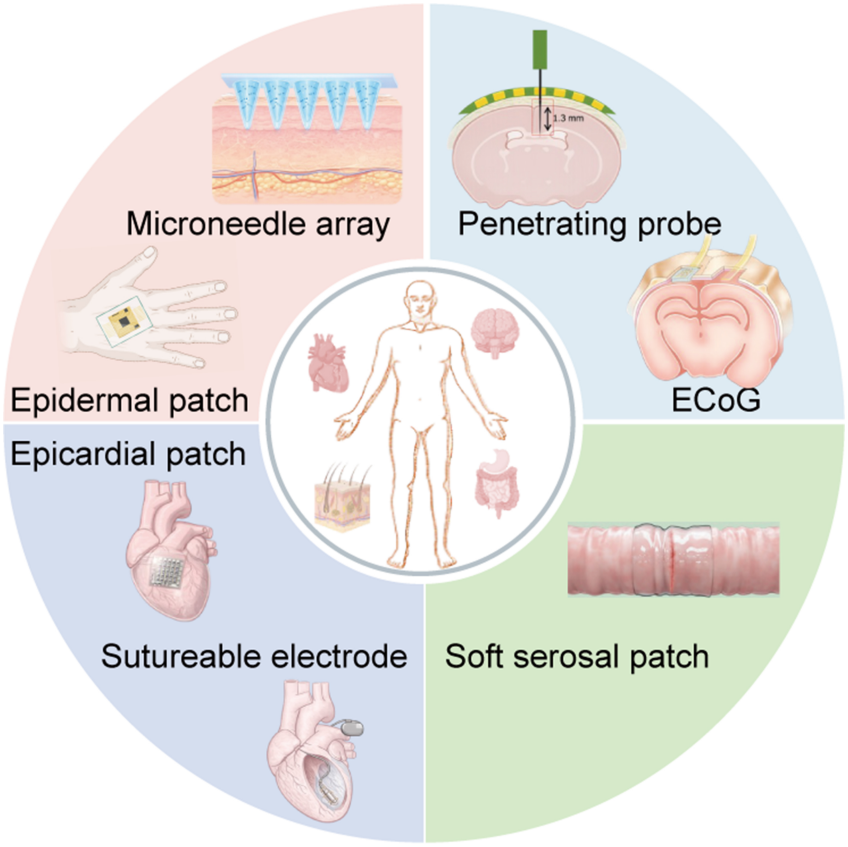

Figure 1. Representative organ-specific bioelectronic interfaces. Some schematic elements Created in BioRender. Liu, X. (2026) https://BioRender.com/w5rveah. ECoG: Electrocorticography.

applications, these systems exhibit a degree of commonality in materials and manufacturing, including ultrathin polymer films with multilayer encapsulation, elastomers combined with strain-relief geometric interconnects, hydrogel or mixed ionic–electronic conductor interfaces, as well as biodegradable supports and antifouling surface modifications[18].

In this review, we adopt a material–structure–biosystem co-design perspective to summarize structural/encapsulation substrates and conductive/interfacial materials for diverse organ interfaces and the associated key performance trade-offs, further survey relevant fabrication and integration routes and representative organ-targeted applications, and discuss open challenges in long-term stability evaluation, manufacturability, power and data links, and clinical translation[19].

MATERIALS FOR ORGAN-SPECIFIC BIOELECTRONICS

Structural and encapsulation substrates

Bioelectronic devices designed for soft-tissue organ interfaces must provide sufficient mechanical support and stress buffering to maintain stable electrical connections under bending and stretching, while also forming reliable chemical and moisture barriers that prevent water and ion penetration and protect internal circuits. In addition, minimizing inflammation and biofouling is essential for maintaining long-term biocompatibility during implantation or wearable use. A key design consideration is the mechanical matching between device substrates and target organs, since biological tissues span a wide range of Young's moduli, ranging from a few kilopascals in the brain to megapascal levels in the epidermis and tendons, whereas commonly used bioelectronic materials include soft hydrogels and elastomers as well as high-modulus polymer films [Figure 2][20-23]. In this section, we classify the relevant material systems into

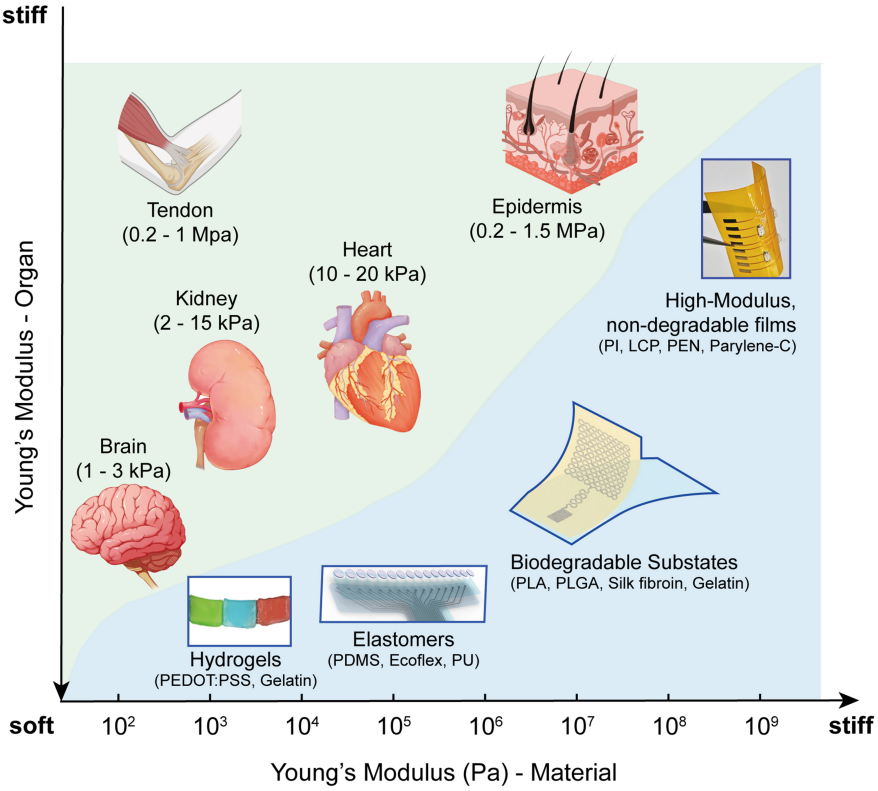

Figure 2. Mechanical matching between bioelectronic substrates and biological organs. Some elements were reproduced with permission, including hydrogels[24], Copyright © 2024, the American Association for the Advancement of Science; elastomers[21], Copyright © 2025, the American Association for the Advancement of Science; biodegradable substates[22], Copyright © 2025, WILEY-VCH Verlag GmbH & Co. KGaA, Weinheim; high-modulus, no-degradable films[25], Copyright © 2020, WILEY-VCH Verlag GmbH & Co. KGaA, Weinheim. Some schematic elements Created in BioRender. Liu, X. (2026) https://BioRender.com/tcvk8fu. PI: Polyimide; LCP: liquid crystal polymer; PEN: polyethylene naphthalate; PEDOT:PSS: poly(3,4-ethylenedioxythiophene):poly(styrenesulfonate); PDMS: polydimethylsiloxane; PU: polyurethane; PLA: polylactic acid; PLGA: poly(lactic-co-glycolic acid).

four categories: non-degradable substrate–encapsulation systems, elastomeric and stretchable platforms, biodegradable and bioresorbable structural layers, and antifouling interfacial coatings, and summarize their key characteristics and representative organ-specific applications.

Non-degradable substrate–encapsulation systems

At organ interfaces that require high-density electrode arrays and long-term stable recording and stimulation, the structural layer must both support precise micro- and nanopatterns and maintain its geometry and mechanical properties over years[26]. To meet these demands, a class of high-modulus, non-degradable polymer films has become the standard structural material set, including Parylene-C, polyimide (PI), polyethylene naphthalate (PEN), and liquid crystal polymers (LCPs) [Figure 3A][27-31]. These materials typically exhibit Young's moduli in the gigapascal range, far above those of soft tissues, but when the film thickness is reduced from tens of micrometres to a few micrometres or even sub-micrometre levels, their bending stiffness decreases substantially[32]. This thickness scaling allows them to conform to curved organ surfaces such as the cerebral cortex, retina, or peripheral nerves without pronounced wrinkling or structural instability, while still preserving in-plane dimensional stability and pattern fidelity.

These polymers are highly compatible with established microelectronic fabrication processes and can accommodate standard steps such as photolithography, metal interconnect patterning, and multilayer via formation. As a result, they are widely used as structural substrates for microelectrocorticography (μECoG)

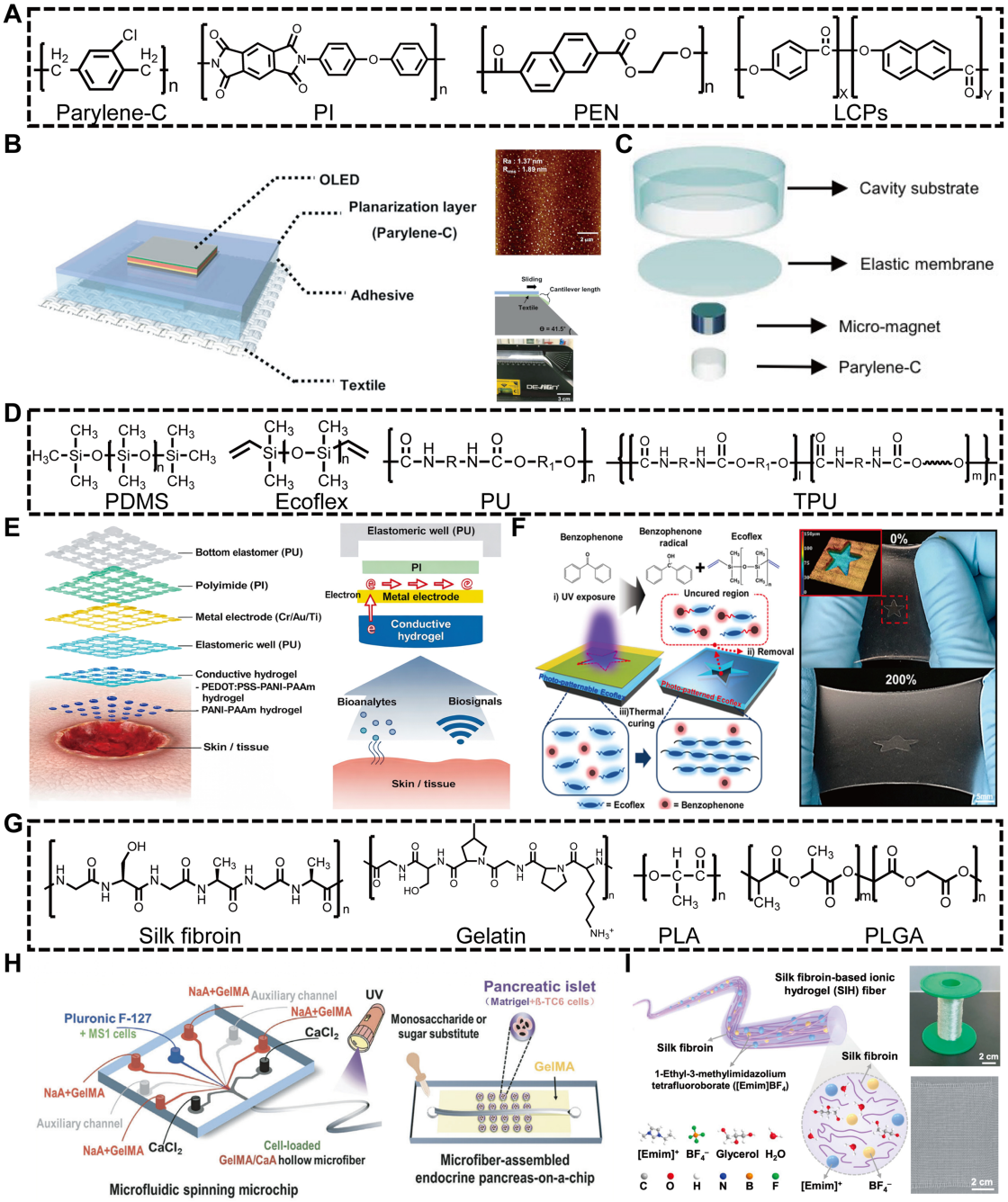

Figure 3. Representative material platforms for structural and encapsulation layers in organ-specific bioelectronics. (A) Representative non-degradable substrate and encapsulation materials; (B) Parylene-C-based planarization and device integration for flexible electronics[35]. Copyright © 2025, published by Springer Nature; (C) Parylene-C-based magnetic implant for wireless sensing[36]. Copyright © 2024, the American Association for the Advancement of Science; (D) Representative elastomeric and stretchable structural materials; (E) Stretchable bioelectronics with conductive hydrogels for low-impedance tissue interfaces[43]. Copyright © 2024, the American Association for the Advancement of Science; (F) Photo-patternable elastomeric substrates based on Ecoflex for stretchable bioelectronics[50]. Copyright © 2023, published by Springer Nature; (G) Representative biodegradable and bioresorbable structural materials. Created by the authors; (H) GelMA/CaA hollow microfiber assemblies as gelatin-derived biodegradable hydrogel microsystem[64]. Copyright © 2024, WILEY-VCH Verlag GmbH & Co. KGaA; (I) Silk fibroin-based ionic hydrogel fibers as biodegradable structural and conductive materials[56]. Copyright © 2024, published by Springer Nature. PI: Polyimide; PEN: polyethylene naphthalate; LCPs: liquid crystal polymers; OLED: organic light-emitting diode; PDMS: polydimethylsiloxane; PU: polyurethane; TPU: thermoplastic polyurethane; UV: ultraviolet; PLA: polylactic acid; PLGA: poly(lactic-co-glycolic acid).

arrays, retinal prostheses, and cuff electrodes targeting the auditory and peripheral nerves. Ultrathin parylene-C films can further serve as transferable and biocompatible carrier substrates because of their

excellent conformability, low-temperature processability, and water resistance[33,34]. For example, Cho *et al.* developed a textile-integrated OLED platform in which a thin parylene-C film was first formed on a guide glass and then transferred onto textile after thermal annealing, serving as a self-supporting planarization layer that replicated the smooth glass surface while preserving textile flexibility. This strategy enabled stable device fabrication and operation on rough, deformable textile substrates [Figure 3B][35]. At the same time, these materials provide good electrical insulation and moisture barrier performance, and can therefore serve as the primary encapsulation layer of the device. For example, parylene-C deposited by chemical vapor deposition can form dense, conformal, and nearly pinhole-free coatings with stable dielectric properties, making it one of the most widely used insulating and encapsulation materials for implantable neural electrodes and cardiac pacing leads [Figure 3C][36]. LCPs combine low water vapor transmission with the ability to be thermoformed and laminated, making them suitable for constructing hybrid rigid–flex encapsulation shells in intracranial or cochlear implants where higher mechanical robustness is required[37].

Overall, non-degradable high-modulus films represented by Parylene-C, PI, PEN and LCPs remain the predominant choice for organ interfaces that experience relatively small surface strain but demand high spatial resolution and long-term electrical stability, such as the cerebral cortex, retina and peripheral nerves[38-40]. Looking ahead, future efforts are likely to focus on retaining the mature processability and encapsulation reliability of these materials while introducing softer interlayers or biohybrid architectures to improve tissue integration. For example, Shi *et al.* developed a biohybrid multilayer interface combining PI, Au, SU-8, and living hydrogel layers to improve tissue-level compliance and interfacial integration with soft biological tissues[41].

Elastomeric and stretchable platforms

For highly deformable organs and tissues such as the skin, heart, and diaphragm, which experience substantial, multidirectional deformation during routine motion, an overly rigid structural layer can readily lead to electrode delamination, interconnect failure, or localized tissue compression. To address this, low-modulus elastomeric substrates based on polydimethylsiloxane (PDMS), Ecoflex, polyurethane elastomers (PU) and thermoplastic polyurethane (TPU) have become widely used at high-deformation organ interfaces [Figure 3D][42]. These materials combine flexible polymer backbones with relatively low crosslinking density, allowing their Young's modulus to be tuned into the 10^3-10^7 Pa range, while sustaining 50%-200% tensile strain without plastic damage. Shin *et al.* reported a stretchable multichannel sensor array, illustrating how mechanically compliant elastomeric substrates can be incorporated into bioelectronic systems to enable stable impedance and pH mapping under both static and dynamic conditions [Figure 3E][43]. This tissue-like mechanical compliance allows the device to conform and move with skin, cardiac, and diaphragmatic tissues during vigorous motion, respiration, and heartbeat, thereby reducing stiffness mismatch and improving interfacial stability.

With these substrates, geometric layout further determines the effective operating range and failure mode of stretchable platforms under high strain. A widely adopted strategy is to confine rigid or semi-rigid functional units to local low-strain zones and connect them to the surrounding elastomer via deformable interconnects. Representative strategies also exploit interfacial engineering between conductive layers and elastomeric or fibrous substrates. For example, conductive coatings can be stabilized through molecular locking, pre-strain-induced wrinkling, and confined interpenetrating nanofiber networks, thereby maintaining electrical continuity and mechanical compliance under repeated deformation[44]. Such strain-engineered architectures enable graded dissipation of mechanical deformation around rigid functional components, thereby maintaining electrical continuity and signal stability during repeated stretching and release. At the material level, such interconnects are typically realized using metal thin films, metal meshes or embedded liquid-metal channels, all encapsulated within PDMS, Ecoflex or PU matrices[45-47]. By tailoring interconnect

geometry, including interconnect length, linewidth, and radius of curvature, externally applied strain can be redistributed into bending, twisting, and local rotation of the interconnects, thereby reducing strain concentration in functional electrode and chip regions. This strain-isolation strategy helps preserve interfacial contact, contact impedance, and signal amplitude during prolonged body motion or repeated organ contraction[48,49].

From the standpoint of encapsulation and interfacial stability, elastomeric platforms are usually realized as a multilayer system, in which an elastic substrate is combined with an elastic encapsulation layer and tailored surface modification. Kim *et al.* developed a photo-patternable Ecoflex encapsulation strategy that enables process-compatible formation of patterned elastic encapsulation layers with multi-windows, thereby improving strain dissipation, electrical stability, and selective multi-analyte sensing in intrinsically stretchable wearable bioelectronics [Figure 3F][50]. The internal conductors and electrodes are first fully embedded in PDMS, PU or TPU to provide electrical insulation and to reduce stress concentrations at geometric discontinuities[20,51,52]. For devices that are worn on the skin or exposed to sweat and body fluids, an additional thin fluoropolymer overlayer, such as polytetrafluoroethylene (PTFE), can be applied to the outer surface to reduce water and ion permeation while maintaining sufficient water vapor transmission[53-55]. For devices adhering to the epicardium or diaphragm, the thickness and modulus of the encapsulation layer must also be balanced against the frictional interaction with pericardium or pleura, so as to avoid excessive shear during cardiac or respiratory cycles.

Biodegradable and bioresorbable structural layers

In many short-term or single-use implant applications, the ideal interface material should gradually degrade and be resorbed *in vivo* once monitoring or therapy is complete, thereby avoiding secondary removal surgery and reducing long-term foreign body reactions. With this goal in mind, protein-based materials such as silk fibroin and gelatin, together with synthetic polyesters such as polylactic acid (PLA) and poly(lactic-co-glycolic acid) (PLGA), have become representative systems for biodegradable structural layers [Figure 3G][56]. Protein materials are degraded enzymatically through cleavage of amide bonds, whereas polyesters undergo hydrolysis of ester bonds to generate lactic acid, glycolic acid and other small molecules that subsequently enter normal metabolic pathways and are cleared[57]. By pre-embedding such cleavable chemical bonds in the polymer backbone, these materials can be programmed to gradually lose mechanical integrity and disappear within a defined time window under the combined action of water and enzymes, providing the chemical foundation for fully resorbable support and encapsulation layers[58].

Accordingly, the central design challenge for biodegradable structural layers is to coordinate mechanical support lifetime, electrical stability and degradation behaviour within an appropriate range[59]. For PLA and PLGA, effective support time can be extended from several weeks to several months by tuning crystallinity, molecular weight and the copolymer ratio of lactic to glycolic acid[60]. Higher crystallinity and molecular weight slow water ingress and backbone scission, making them suitable for medium-term cardiac or neural stimulators, whereas a higher glycolic acid content usually accelerates hydrolysis and is better suited to postoperative short-term monitoring patches or temporary electrodes[61,62]. Silk fibroin and gelatin can be adjusted in an analogous way by varying β-sheet content, crosslinking density or integration with inorganic fillers, allowing stiffness and degradation rate to be tuned while maintaining good cell compatibility[63]. Beyond conventional films and bulk scaffolds, biodegradable protein matrices can also be engineered into hollow microfiber assemblies and organ-mimetic microsystems. For example, Tian *et al.* developed a microfiber-assembled endocrine pancreas-on-a-chip based on microfluidically spun GelMA/CaA hollow fibers, in which the biodegradable hollow microfibers were used to mimic vascular lumens and support material transport, and were integrated with a 3D pancreatic islet culture layer for islet culture and functional evaluation [Figure 3H][64]. Silk fibroin-based polymers can also be engineered into mechanically robust and

ionically conductive fibers by incorporating ionic liquid components. Lu *et al.* reported a silk fibroin-based ionic hydrogel fiber composed of silk fibroin, ionic liquid, and glycerol, which exhibited high strength, large extensibility, and stable ionic conductivity, thereby enabling deformable conductive fibers for wearable bioelectronic and human–machine interface applications [Figure 3I][56]. In addition to structural substrates, biodegradable matrices can also be integrated with transient sensing interfaces and soft conductive layers. For example, agar- and hydrogel-based ionic interfaces combined with thin metal electrodes can provide temporary skin-compatible sensing platforms with minimized long-term foreign-body burden after use[65]. Another approach integrates ultrathin bioresorbable inorganic oxides as transient insulating or encapsulation layers and hydrolyzable conductive polymers as temporary charge-transport components, enabling the required electrical functions to be maintained over a prescribed time frame before gradual resorption[66,67].

At the level of organ interfaces, biodegradable and bioresorbable structural layers are particularly attractive in three types of scenarios. The first includes the heart and peripheral nerves in the postoperative or acute phase, where temporary electrophysiological interventions such as transient pacing, vagus nerve modulation or post-surgical rhythm monitoring are required. The second concerns wound beds, graft sites or remodelling soft tissues, where staged mechanical and electrophysiological monitoring or local electrical stimulation is needed during healing, but complete device disappearance is desired once tissue repair is complete[68]. The third involves patient groups for whom device retrieval is difficult or undesirable, such as neonates or very elderly individuals with multiple comorbidities. Recent work has begun to introduce dynamic crosslinks, inorganic fillers and multilayer gradient architectures into biodegradable frameworks, more tightly coupling mechanical support, electrical stability and controlled resorption[69]. These strategies aim to align the functional lifetime of the structural layer more precisely with organ healing timelines, while reducing issues such as swelling during degradation, fragment migration and local irritation, thereby opening broader materials and structural design space for single-use implants and resorbable physiological monitoring systems.

Interfacial coatings for antifouling control

In real physiological environments, once a device surface is exposed to plasma, cerebrospinal fluid or intestinal fluid, plasma proteins, inflammatory cells and microbes begin to adsorb and accumulate within a very short time[70]. This rapidly triggers inflammatory reactions, thrombosis and fibrotic encapsulation, which ultimately manifest as increased electrode impedance, reduced signal amplitude and, in severe cases, complete device failure[71,72]. Consequently, beyond the structural and encapsulation layers, a thin outermost interfacial coating that directly contacts body fluids and is engineered to resist protein adsorption, cell adhesion and even bacterial colonization has become a critical element for achieving long-term stable organ–device interfaces[73].

One of the earliest systematically explored strategies is the construction of a hydrated, hydrophilic layer. A representative approach is to graft or coat polyethylene glycol (PEG) or related hydrophilic polymers onto the device surface so that, in aqueous environments, a dense, nanometre-scale hydration layer is formed[74,75]. This hydrated shell introduces steric hindrance and a high energetic penalty for dehydration, thereby suppressing the residence of proteins and cells at the interface[76]. Because PEG is susceptible to oxidation and hydrolysis under long-term implantation conditions, recent work has increasingly shifted toward zwitterionic polymer systems, such as coatings bearing sulfobetaine or carboxybetaine side chains[77]. These materials carry both positive and negative charges on the same side chain, enabling the formation of more stable and tightly bound hydration shells. As a result, they exhibit more durable resistance to protein adsorption and cell adhesion in complex media such as plasma and cerebrospinal fluid, and have been applied to the surface modification of neural electrodes, vascular stents and catheters, where animal studies

Table 1. Conductive materials for bioelectronic interfaces

Material	Main conduction	Representative values	Biointerface relevance	Ref.
OMIECs	Electronic + ionic	Pristine PEDOT:PSS < 1 $S \cdot cm^{-1}$; hydrogels/composites up to 28-4,000 $S \cdot cm^{-1}$	Low-Z recording; OECTs	[83-85]
Conductive hydrogels	Ionic/mixed	Ion-dependent; tunable by salts, polymers, fillers	Wet coupling; soft electrodes	[49,86]
Liquid metals	Electronic	~10^6 $S \cdot m^{-1}$ level	Stretchable interconnects	[87,88]
2D materials	Electronic	Material-dependent; high in-plane transport	Thin electrodes; high surface area	[89,90]

OMIECs: Organic mixed ionic-electronic conductors; PEDOT:PSS: poly(3,4-ethylenedioxythiophene):poly(styrenesulfonate); OECTs: organic electrochemical transistors.

have shown reduced inflammation and attenuated glial scarring.

A second line of development draws inspiration from naturally "slippery" interfaces such as those of pitcher plants, through the creation of slippery liquid-infused porous surfaces (SLIPS)[78]. In these systems, a porous fluoropolymer or micro/nano-rough substrate serves as a scaffold in which a thin layer of medical lubricant is retained within pores or topographical features, forming a continuous, stable and self-healing liquid interface that offers virtually no effective anchoring sites for bacteria, proteins or cells. Multiple animal models have demonstrated that SLIPS coatings can markedly suppress bacterial biofilm formation and fibrotic encapsulation, making them attractive for high-contamination-risk devices such as urinary catheters, vascular implants and GI systems. Unlike hydrophilic coatings that rely primarily on chemical repulsion, SLIPS function mainly through physical dewetting and re-spreading of the infused lubricant, allowing damaged or abraded regions to recover their antifouling properties from the lubricant reservoir.

For brain, retinal and peripheral nerve electrodes, interfacial coatings must not only mitigate inflammation and glial scar formation but also preserve low impedance and ionic transport[79]. In intravascular or intestinal settings, additional requirements include antithrombotic and antibacterial performance and compatibility with fast-flowing fluids. Rationally matching these interfacial coatings with the non-degradable, elastic or biodegradable structural–encapsulation systems discussed above offers a route to constructing integrated interfaces that combine mechanical matching, chemical stability and biological tolerance across diverse organ environments, thereby providing a robust foundation for the long-term reliable operation of conductive layers and active electronic components[80].

Conductive layers

Conductive layers form the functional core of bioelectronic devices, enabling the conversion and transmission of electrical and ionic signals. Unlike the substrate layer, where a small number of materials can fulfill mechanical support roles, the conductive layer encompasses a diverse family of materials with fundamentally different conduction mechanisms and microstructures[81,82]. To capture this diversity and its relevance to device design, this section is organized by material families: organic mixed ionic-electronic conductors (OMIECs), conductive hydrogels, liquid metals, and two-dimensional nanomaterials.

In addition to differences in softness and processability, conductive biointerface materials differ substantially in their quantitative transport properties. OMIECs and conductive polymer hydrogels rely on coupled ionic and electronic transport, whereas liquid metals provide primarily electronic, metal-level conductivity and ionic hydrogels mainly conduct through mobile ions[83]. These differences determine interfacial impedance, charge-injection capability, sensing bandwidth, and long-term stability at wet tissue interfaces. Therefore, representative conductivity values and mixed-conduction characteristics of major conductive-layer materials are summarized in Table 1.

OMIECs

OMIECs are soft polymeric materials that transport both electronic and ionic charges while remaining mechanically compatible with biological tissues[91,92]. They are typically built from polymers with π-conjugated backbones combined with hydrophilic side chains or polyelectrolyte components, giving moduli in the MPa range, good bendability, and compatibility with solution processing[93,94]. Unlike purely electronic organic semiconductors, OMIECs undergo bulk doping in aqueous electrolytes so that charge is stored throughout the material volume[83]. This volumetric electrochemical coupling is particularly relevant at soft organ interfaces, where tissue impedance is dynamic and signal amplitudes are often small, requiring stable, low-impedance charge exchange rather than purely interfacial capacitive coupling[43].

Their mixed conduction can be viewed as the cooperation of electronic and ionic pathways within a single network[85]. Electrons or holes move along the π-conjugated backbone through locally ordered domains, with conductivity governed by backbone structure and doping level[95,96]. For representative poly(3,4-ethylenedioxythiophene):poly(styrenesulfonate) (PEDOT:PSS)-based OMIECs, electronic conductivity can range from below 1 $S \cdot cm^{-1}$ in pristine or structurally disordered films to 10^2-10^3 $S \cdot cm^{-1}$ after secondary doping, acid treatment, or structural optimization. Ions migrate within hydrophilic side chains or polyelectrolyte phases and compensate the charges created on the backbone during oxidation or reduction. Because this electron–ion compensation occurs throughout the bulk rather than at a confined interface, OMIECs exhibit high volumetric capacitance, typically on the order of tens to hundreds of F/cm^3 for PEDOT-based systems, enabling high organic electrochemical transistor (OECT) transconductance at low operating voltages as well as low-impedance, high-charge-injection electrode interfaces. Such characteristics are advantageous in organ-specific bioelectronic chips that must operate under strict electrochemical safety limits while maintaining efficient coupling to excitable tissues such as brain and myocardium. Recent work has shown that the electrochemical doping rate in OMIECs strongly depends on film microstructure, film morphology, and electrolyte environment, which in turn determine device response speed and operational stability[97,98]. These kinetic factors are particularly important in neural and cardiac electrophysiology, where millisecond-scale temporal resolution is required[83,99].

Recent studies further underscore the importance of electrochemical kinetics in OMIEC-based organ interfaces. Keene *et al.* showed that electrochemical doping in conjugated polymers can be hole-limited, highlighting that OMIEC response speed depends not only on ion transport but also on electronic charge propagation within the polymer phase[100]. For organ-specific bioelectronic chips, such behavior directly links material-level transport dynamics to system-level bandwidth and energy constraints. Beyond transport kinetics, recent material-design strategies have further expanded the utility of OMIECs in soft biointerfaces. Montazerian *et al.* showed that replacing conventional hydrophobic PSS dopants with hydrophilic biomacromolecular AlgS dopants in PEDOT:AlgS improves aqueous dispersibility, molecular degradability, and ionic integration with hydrogel matrices, thereby enabling injectable and 3D-printable OMIEC-based bioelectronics [Figure 4A][101]. Together, these findings shift the focus from purely material characterization toward mechanism-informed optimization of kinetics, stability, and energy efficiency in physiological environments.

Overall, OMIECs represent a commonly adopted conductive-layer strategy in organ-specific bioelectronic chips, enabling low-impedance electrochemical coupling at wet, soft, and dynamically active tissue interfaces under low operating voltages[102,103]. Their mixed ionic–electronic transport characteristics make them particularly suitable for high-density neural interfaces in the brain and retina, as well as conformal sensing and stimulation devices deployed on mechanically deformable tissues such as myocardium and intestinal mucosa[102].

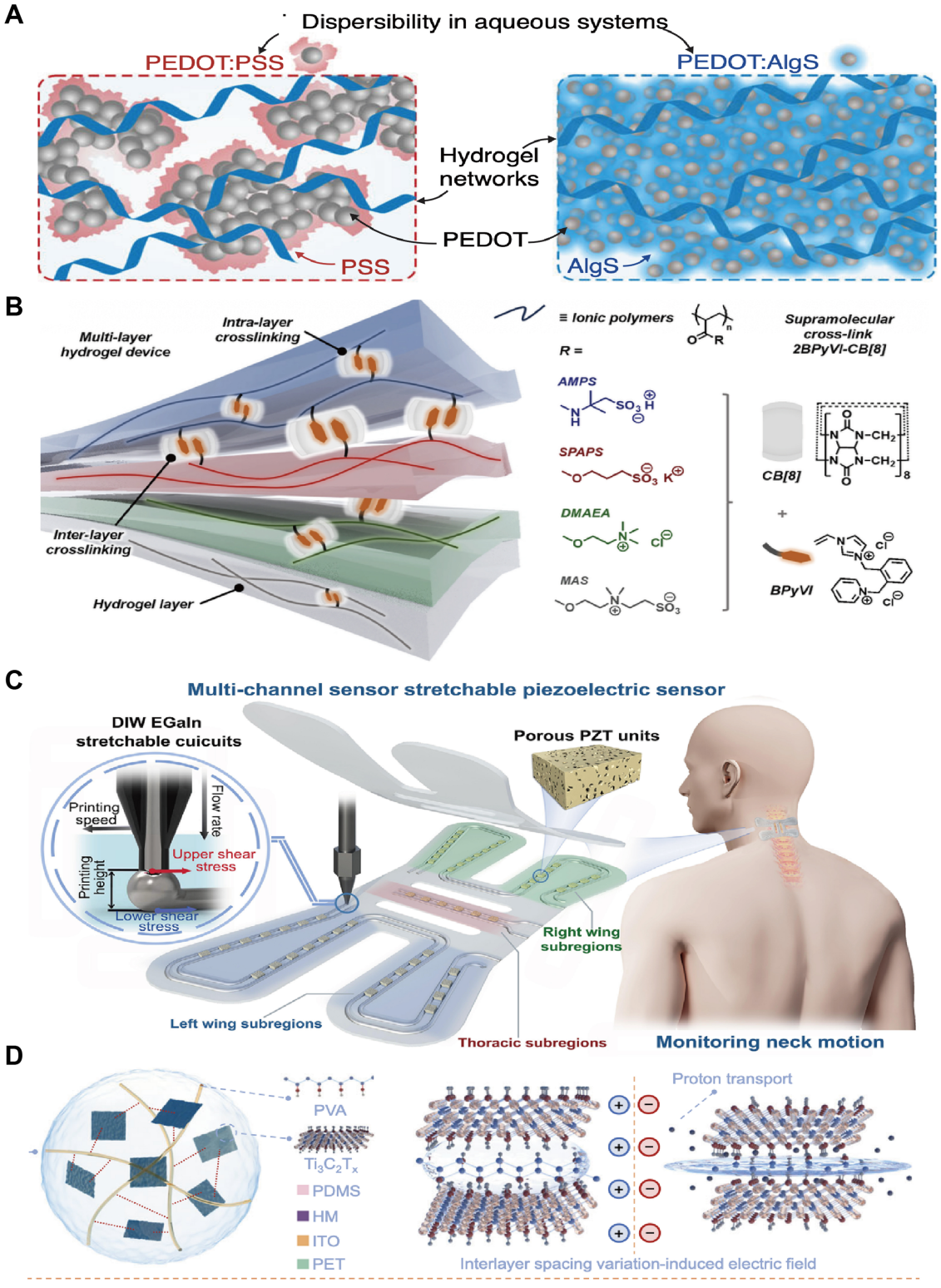

Figure 4. Representative material platforms for conductive and interfacial layers in organ-specific bioelectronics. (A) Hydrophilic-dopant-engineered PEDOT systems for degradable hydrogel bioelectronics[101]. Copyright © 2025, published by Springer Nature; (B) Supramolecular poly(ionic) networks enabling stretchable conductive hydrogels[24]. Copyright © 2024, the American Association for the Advancement of Science; (C) Topology-optimized stretchable piezoelectric sensors enabled by direct-ink-written liquid-metal circuits[128]. Copyright © 2026, WILEY-VCH Verlag GmbH & Co. KGaA; (D) Biomimetic microstructure-enabled piezoionic mechanoreceptors for ultrasensitive multimodal sensing and object recognition[150]. Copyright © 2025, published by Springer Nature. PEDOT: Poly(3,4-ethylenedioxythiophene); PSS: poly(styrenesulfonate); AMPS: 2-acrylamido-2-methyl-1-propanesulfonic acid; SPAPS: 3-sulfopropyl acrylate potassium salt; DMAEA: [2-(methacryloyloxy) ethyl] trimethylammonium chloride; MAS: [2-(methacryloyloxy)ethyl]dimethyl-(3-sulfopropyl)ammonium hydroxide; DIW: direct-ink-written; EGaIn: eutectic gallium–indium; PZT: porous lead zirconate titanate; PVA: polyvinyl alcohol; PDMS: polydimethylsiloxane; HM: hydrogel microneedles; ITO: indium tin oxide; PET: polyethylene terephthalate.

Conductive hydrogels

Conductive hydrogels are soft, water-rich polymer networks that are engineered to carry electrical signals while remaining mechanically similar to biological tissues. They are typically formed by three-dimensional crosslinking of natural or synthetic polymers such as polyacrylamide, alginate, or gelatin, which creates a stable, highly hydrated network[104]. By further introducing ionic or electronic conductive components into this matrix, conductive hydrogels can preserve softness and tissue affinity and at the same time support reliable electrical signal transmission[105-107].

Their electrical behavior is supported by two complementary mechanisms. First, the abundant water phase inside the hydrogel provides pathways for mobile ions, giving rise to ionic conduction[108,109]. Second, adding conductive elements such as metal nanowires, carbon nanotubes, graphene, or conducting polymers like PEDOT:PSS allows the formation of continuous electronic or mixed ion–electron pathways within the network[110-113]. Depending on the conductive filler, polymer composition, and percolation structure, conductive hydrogels can span several orders of magnitude in conductivity, from predominantly ion-conducting networks with conductivities typically in the 10^{-3}-10^{-1} S·cm^{-1} range to electronically conductive composite or conductive-polymer hydrogels with conductivities reaching 1-10^{2} S·cm^{-1} or higher[114]. Dynamic crosslinks based on hydrogen bonding, metal–ligand coordination, or catechol chemistry can break and reform during stretching, bending, or repeated loading. These reversible bonds help maintain the integrity of the conductive pathways and endow the material with self-healing and network reconfiguration[115-117]. Recent work has shown that supramolecular poly(ionic) networks can simultaneously achieve ionic conductivities up to 0.1 S·cm^{-1} and stretchabilities exceeding 1,500%, illustrating how dynamic crosslinking can reconcile efficient ionic transport with tissue-like deformability in multilayer conductive hydrogels [Figure 4B][24]. For example, metal-catechol coordination conductive hydrogels have been shown to recover both mechanical strength and electrical continuity after physical fracture, highlighting the role of dynamic bonding in sustaining cyclic stability[118]. Such dynamic bonding is particularly beneficial for organ interfaces subjected to cyclic deformation, such as myocardium and diaphragm.

Recent advances continue to expand the functional scope of conductive hydrogels toward environments and use-cases previously inaccessible to soft materials. One major direction focuses on improving environmental robustness, as exemplified by Zhang *et al.*, who introduced a "hydro-locking" strategy that stabilizes water within double-network hydrogels, maintaining softness and conductivity from -115 to 143 °C for extreme-condition sensing[119]. Although originally demonstrated for extreme-condition sensing, such hydration-stabilization strategies may also improve long-term stability of implanted organ interfaces. A complementary direction integrates energy-harvesting and therapeutic functions, demonstrated by Xin *et al.*, who developed a Fe^{2+}/Fe^{3+}-alginate thermogalvanic dressing capable of converting wound-site temperature gradients into therapeutic electrical stimulation while enabling real-time monitoring[120]. This multifunctional integration hints at future organ-resident patches capable of simultaneous sensing and localized therapy.

Overall, conductive hydrogels integrate softness, high water content, adhesion and mixed ionic-electronic transport, making them especially well-suited for electrochemical interfaces with the brain, heart and skeletal muscle, where intimate, low-modulus contact to wet, excitable tissue is essential[121]. As these materials evolve toward printable, adaptive and biodegradable architectures, they are expected to form a core platform for the next generation of implantable bioelectronics and soft robotic systems and to enable broader clinical and engineering applications[122].

Liquid metals and soft metal composites

Liquid metals and their soft composites form a class of conductors that combine metal-level electrical conductivity with fluid- or rubber-like mechanics[123]. In a narrow sense, liquid metals refer to metals or alloys that remain liquid near room temperature, most prominently gallium-based eutectic alloys such as

eutectic gallium–indium (EGaIn) and Galinstan[124,125]. These alloys exhibit conductivities on the order of 10^6 S·m^{-1} together with low vapour pressure, relatively low toxicity and good chemical stability, making them attractive as soft yet highly conductive layers in deformable bioelectronic systems.

Their functional behavior is governed by the interplay between metallic bonding and interfacial mechanics. Intrinsically, liquid metals are highly conductive fluids whose conductivity remains nearly unchanged under large mechanical deformation, allowing stable signal transmission under strain[126]. A nanometre-scale oxide skin and high surface tension stabilise discrete droplets and filaments and allow damaged traces to self-heal as the liquid flows and closes cracks. In soft metal composites, micron- or nanoscale droplets dispersed in PDMS, PU or hydrogels remain electrically insulated at rest; under stretching, compression or shear, the oxide shells rupture and neighbouring droplets fuse into percolating networks, producing stress-activated conductivity and inherent strain sensitivity. These features are particularly beneficial at organ interfaces subjected to continuous motion, such as skin, skeletal muscle and cardiovascular tissues, where conductive layers must tolerate repeated deformation without loss of conductivity[127]. For example, Zeng *et al.* showed that direct-ink-written EGaIn circuits can serve as highly stretchable electrodes in topology-optimized piezoelectric sensors for anisotropic motion monitoring, illustrating how liquid-metal interconnects enable precise patterning, stable conductivity under large deformation, and wearable bioelectronic integration [Figure 4C][128]. Beyond serving as passive fillers, gallium-based droplets can also initiate free-radical polymerisation and participate in crosslinking, as shown by Jaseem *et al.*, who obtained tough, injectable and self-healing conductive hydrogels suitable for injectable electrodes and soft encapsulation layers on skin or muscle[129].

Building on these platforms, recent work has shifted from simply achieving conductivity and stretchability to ensuring stable operation at complex biological interfaces. A notable example is a magnetically reshapable three-dimensional liquid-metal multi-electrode array, in which EGaIn-filled deformable microtubes are folded into three-dimensional shapes under magnetic guidance and gently inserted into brain organoids for multichannel, deep electrophysiological recording and stimulation, highlighting the plasticity and compliance of liquid metals at three-dimensional brain and organoid interfaces[130]. Such demonstrations highlight the capacity of liquid metals to conform to soft, three-dimensional brain and heart while maintaining metallic-level conductivity, a property difficult to achieve with conventional rigid electrodes[131,132].

However, liquid-metal conductors also present important fabrication and stability challenges. Their high surface tension and fluidity make high-resolution and reproducible patterning difficult, while the spontaneously formed gallium oxide skin can both stabilize traces and alter wettability, adhesion, and contact resistance[133]. Poor wetting on polymer substrates may cause line retraction or discontinuous patterns, and leakage from microchannels or composite matrices under repeated deformation can compromise device reliability[134]. These issues become more pronounced in large-area or high-throughput fabrication, where ink rheology, substrate adhesion, encapsulation, and droplet stabilization must be precisely controlled[135]. Recent patterning and stabilization strategies, including surface modification, microfluidic confinement, transfer printing, direct ink writing, laser-assisted processing, and elastomer/hydrogel encapsulation, are therefore important for translating liquid-metal conductors into reliable organ-specific bioelectronic interfaces[136].

Overall, liquid metals and their soft composites combine metal-like conductivity, rubber-like softness and reconfigurable interfaces, providing a conductive-layer strategy centered on strain-invariant conductivity and mechanical robustness in highly deformable biological systems[137]. Within organ-specific bioelectronic chips, they are particularly suitable for interfaces that experience large or repetitive deformation, including epidermal and muscular electromyography (EMG) systems, cardiovascular monitoring devices, and

emerging three-dimensional neural or organoid interfaces[138,139]. As encapsulation strategies, droplet stabilization methods, and biocompatible formulations continue to improve, liquid-metal-based conductors are expected to expand from wearable platforms toward more demanding implantable organ interfaces[140,141].

Two-dimensional nanomaterials

Two-dimensional conductors *are* atomically thin or few-layer materials *i*n which charge carriers are largely confined to in-plane transport[142,143]. Representative systems include graphene and its derivatives, transition metal dichalcogenides (TMDs) and MXenes[144]. Compared with bulk conductors, these sheets offer very high specific surface area, tunable band structures and excellent flexibility: monolayer graphene combines high carrier mobility with nanometre-scale thickness, while MXenes provide hydrophilicity and easy dispersion through surface –O/–OH/–F terminations[145]. These characteristics enable 2D materials to act as ultrathin conductive layers that enhance interfacial conductivity and device density without significantly increasing the bending stiffness of soft, organ-facing substrates.

Their interfacial behaviour is governed by sp^2-conjugated carbon networks or transition-metal d orbitals together with engineerable defects and surface groups[146,147]. Extended π or *d* bands support fast in-plane electronic transport, whereas edge sites, vacancies and terminations can be chemically tuned to adjust carrier density, Fermi level and charge-transfer kinetics. Reported conductivities vary widely depending on material type, flake quality, oxidation state, and film assembly; graphene-based and MXene films can reach conductivities from 10^3 to 10^5 $S \cdot m^{-1}$ or higher in optimized films, whereas oxidized or defect-rich derivatives show substantially lower values[19,148]. Unlike OMIECs, these materials do not usually exhibit bulk ionic–electronic mixed conduction, but their surface groups and electrolyte-accessible interfaces facilitate charge transfer and reduce electrode–tissue impedance[149]. For example, Ding *et al.* integrated MXene nanosheets into a flexible multilayer sensing architecture composed of hydrogel microneedles, PET/ITO films, and PDMS encapsulation, where the layered MXene morphology, stress-induced interlayer transport, and non-faradaic interfacial coupling enabled deformable multimodal bioelectronic sensing [Figure 4D][150]. When coated onto elastomers or hydrogels, 2D flakes form composite films that markedly reduce electrode–tissue impedance and maintain stable conduction under repeated bending and stretching, supporting long-term mechanical compatibility at dynamic tissue interfaces[104,151].

Beyond conductive 2D sheets, emerging two-dimensional polymers provide a distinct materials strategy for ultrathin encapsulation. For example, Ritt *et al.* showed that the molecularly impermeable 2D polyaramid 2DPA-1 can be processed into nanometre-thin, electrically insulating barrier films with exceptionally low gas permeability. A 60-nm 2DPA-1 coating markedly retarded MAPbI3 degradation by suppressing O_2 and water-vapour permeation, highlighting the promise of such ultrathin 2D polymer membranes for conformal encapsulation[152]. This combination of conductive 2D layers with atomically thin barrier films offers a strategy for simultaneously improving electrical performance and protecting organ-resident devices from biofluid infiltration and molecular diffusion[153].

Overall, two-dimensional conductors combine atomic-scale thickness, high conductivity and tunable interfacial chemistry, providing a conductive-layer strategy centred on high spatial resolution, minimal mechanical footprint and chemically programmable interfaces[154,155]. In the brain and retina they can function as transparent, low-impedance electrodes and protective layers for high-fidelity recording and stimulation, whereas on the heart, skeletal muscle and intestinal mucosa, 2D conductor-elastomer or -hydrogel coatings can enhance electrical coupling and stability while minimally perturbing tissue mechanics[156]. Within organ-specific bioelectronic chips, 2D materials are particularly advantageous when high-density electrode integration and minimal structural intrusion are required, such as in retinal prostheses, cortical microarrays and conformal visceral patches[157].

Table 2. Application-specific material requirements for bioelectronic modalities

Modality	Mechanical	Interfacial	Material	Benchmark	Devices	Ref.
Recording	Conformal	Low-Z	OMIECs; hydrogels	Low impedance; stable wet coupling	ECG/EEG/ECoG	[86,161]
Stimulation	Durable	High-CIC	PEDOT; Pt/Ir	High charge injection; electrochemical safety	Stimulator	[162-164]
Biosensing	Wet-operable	Selective	Enzymes; aptamers	Analyte permeability; antifouling	Sweat/ISF	[20,49,165,166]
Pressure	Stretchable	Low hysteresis	Elastomers; LMs	High deformability; cycling stability	E-skin	[134,167,168]
Temperature	Conformal	Thermally stable	Thin films	Stable thermal response	Patch	[169]
Delivery	Compliant	Controlled transport	Hydrogels; iontronics	Reservoir stability; biofluid compatibility	Pump/patch	[170,171]

OMIECs: Organic mixed ionic-electronic conductors; ECG: electrocardiography; EEG: electroencephalography; ECoG: electrocorticography; CIC: charge injection capacity; PEDOT: poly(3,4-ethylenedioxythiophene); ISF: interstitial fluid; LMs: liquid metals.

Beyond mechanical and electrical matching, biocompatibility is a central design consideration in organ-specific bioelectronics[158,159]. This issue extends beyond acute cytotoxicity to include skin compatibility, foreign-body and immune responses, inflammatory remodeling, and long-term safety during chronic use[160]. These concerns depend strongly on the target organ, implantation mode, exposure duration, and local biochemical environment. Therefore, material selection should be evaluated not only by conductivity and mechanics, but also by the biological responses expected in each tissue context.

In addition to organ-specific constraints, bioelectronic materials must also be selected according to the intended modality. Recording, stimulation, biosensing, pressure sensing, temperature sensing, and delivery each impose distinct requirements for contact, impedance, charge injection, selectivity, stability, and transport. These modality-specific considerations are summarized in Table 2.

FABRICATION STRATEGIES FOR ORGAN-SPECIFIC BIOELECTRONICS

Developing appropriate fabrication strategies for organ-specific electrode interfaces is critical, as the fabrication process directly determines interfacial properties such as biocompatibility, stability, and functional performance[172]. Because living systems are highly sensitive to their microenvironment, bioelectronic interfaces should be designed and constructed to closely mimic the native physiological milieu of the target tissue. To ensure long-term stable operation, electrodes must achieve robust integration and conformal contact with the organ surface[173]. To enable efficient communication between bioelectronic devices and biological tissues, the fabrication method should allow precise control over interfacial structural features, including size, geometry, and spatial distribution. The manufacturing strategies should support high-resolution patterning, large-area fabrication, and 3D curved-surface structuring, while maintaining conformal contact with soft tissues[174-176]. In this part, we discuss key design considerations for constructing tissue-electrode interfaces by using stretchable conductive materials and summarize recent advances in several representative fabrication techniques.

High-resolution patterning

Organ-targeted bioelectronic interfaces commonly require micrometer-scale resolution and high-density electrode arrays to match fine anatomical structures and to support localized stimulation and recording at high spatial resolution[177]. In particular, closely spaced microelectrode grids can serve as epidermal EMG arrays capable of resolving subtle body movements with an interelectrode spacing below 500 μm, enabling discrimination of fine motor units and complex activation patterns[178]. At the organ surface, similar spatial precision is essential for mapping heterogeneous electrophysiological activity, targeting small functional

units, and implementing multiplex sensing and stimulation. In this setting, high-resolution patterning is not only a question of achieving dense layouts, but also of defining sharp, well-controlled micro- and nanostructures on mechanically soft, curved, and often transient substrates that must operate stably in aqueous biological environments[179]. Meeting these requirements requires re-engineering conventional microelectronics processes, originally developed for rigid wafers, into hybrid fabrication workflows in which high-resolution patterning is decoupled from the final organ-matched mechanical support.

A common approach is to first pattern microscale structures on rigid donor wafers and then transfer them onto soft substrates such as elastomers, hydrogels, or bioresorbable polymers[179]. Standard complementary metal–oxide–semiconductor (CMOS)-compatible photolithography, e-beam lithography, and dry/wet etching are used to define metal interconnects, semiconductor membranes, and sensing elements at micrometer to sub-micrometer scales[180,181]. These ultrathin "islands" are then arranged into stretchable layouts and transferred onto compliant substrates by deterministic transfer printing or stamping[182]. Researchers have clarified the key materials and mechanics design principles needed to realize such devices, enabling stretchable circuits that retain the electrical performance of conventional silicon while matching the deformations of organs[183]. High-resolution transfer printing also allows light-emitting diodes (LEDs), OECTs, and micro-integrated circuits (micro-ICs), and various sensors to be integrated into conformal epicardial membranes, neural meshes, or epidermal patches without loss of lithographic precision.

Photolithographic microfabrication

Photolithography is a widely used microfabrication technique that provides excellent spatial resolution by using patterned light to transfer features onto thin-film substrates[184]. It is commonly applied to define metal electrodes and interconnects on polymeric substrates such as PI or SU-8, yielding flexible microelectrode arrays[185,186]. This strategy has been used to fabricate high-density arrays for neural probes, retinal implants, and other interfaces, with feature sizes on the order of single cells and finely controlled electrode spacing, allowing highly localized interaction with biological tissues[187]. A key limitation, however, is that conventional photolithography is largely restricted to planar, wafer-based processes; additional steps such as transfer printing or lamination are often required to integrate these micro-patterned structures onto organ-mimicking 3D surfaces.

Direct photopatterning offers a complementary route in which functional materials are structured *in situ* on a substrate without separate photoresist processing and etching [Figure 5A][188]. In this approach, light-sensitive formulations, such as photopolymerizable hydrogels containing conductive fillers or photo-crosslinkable conductive polymers, which are exposed through a mask or projected pattern to define the desired microstructures[189]. This method allows conductive pathways, electrode sites, or microelectrode arrays to be "written" directly into soft, biocompatible matrices with high precision. For organ-specific bioelectronics, direct photopatterning is particularly attractive because it can produce very soft, tissue-like electrode architectures that conform to brain, cardiac, or other organ surfaces while minimizing mechanical mismatch, thereby improving the stability and fidelity of the tissue-electrode interface.

Laser writing

Laser writing (or laser engraving) employs focused laser beams to locally ablate, modify, or convert materials, thereby defining conductive tracks or electrode patterns without the need for masks. It is an inherently maskless and programmable technique that lends itself well to rapid prototyping of complex layouts. A widely studied example is laser-induced graphene (LIG), in which irradiation of polymeric or biomass-derived films converts the surface into a porous, conductive graphene network that can serve as a flexible electrode[190]. This direct-write process can achieve microscale features and can be readily

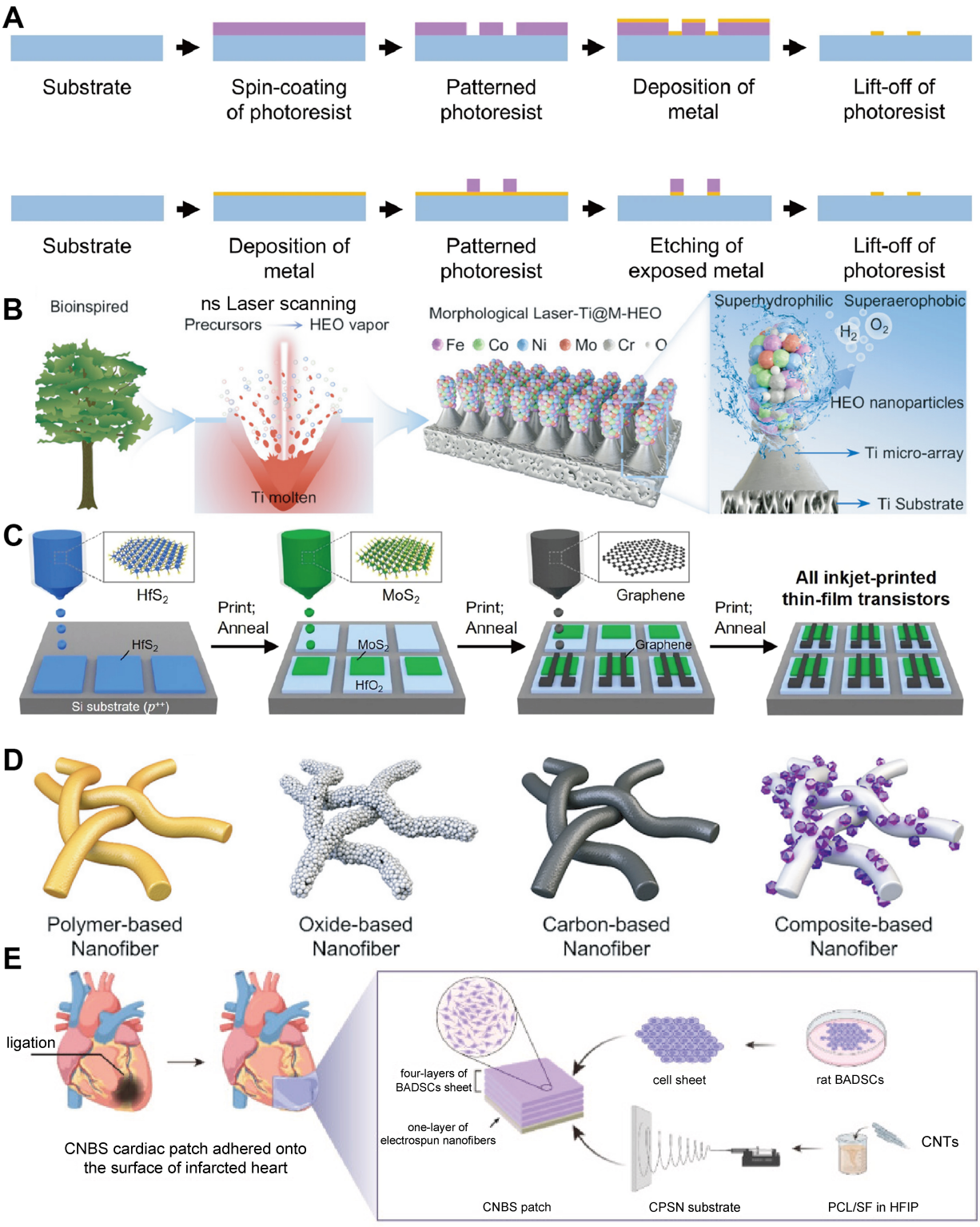

Figure 5. Fabrication strategies for organ-specific bioelectronics with high-resolution and large-area fabrication. (A) Schematic illustration of photolithographic microfabrication[188]. Copyright © 2025, published by Springer Nature; (B) LIG fabrication[191]. Copyright © 2025, WILEY-VCH Verlag GmbH & Co. KGaA; (C) Inkjet-based printing of bioelectronic devices[190]. Copyright © 2022, published by Springer Nature; (D) Electrospun nanofiber-based soft electronics[194]. Copyright © 2025, WILEY-VCH Verlag GmbH & Co. KGaA; (E) An electrospun conductive cardiac patch designed to conformally wrap around the surface of an infarcted heart[195]. Copyright © 2023, published by Elsevier. LIG: Laser-induced graphene; HEO: high-entropy oxide; BADSCs: brown adipose-derived stem cells; CNTs: carbon nanotubes; CNBS: CPSN-BADSCs sheets; CPSN: CNTs-containing electrospun polycaprolactone/silk fibroin nanofibers; PCL: polycaprolactone; SF: silk fibroin; HFIP: hexafluoroisopropanol.

reconfigured at the design stage, which is advantageous for organ-specific devices that require iterative optimization or patient-specific tailoring [Figure 5B][191]. Moreover, laser writing can be performed on thin films that are later laminated onto curved tissues, or directly on preformed substrates intended to conform to skin or organ surfaces[192]. In this way, it helps bridge the gap between high-resolution patterning and practical fabrication of conformal bioelectronic interfaces.

Inkjet-based printing

Inkjet printing is a digital, additive method where conductive or functional inks are deposited in tiny droplets to form patterns. Modern inkjet systems can produce microscale features, offering fairly high printing resolution for flexible circuits. This technique allows custom electrode layouts to be printed on demand, accommodating organ-specific geometries by simply adjusting the digital design. Because it is a non-contact process, inkjet printing can pattern delicate substrates without damage, making it suitable for creating fine electrode arrays on soft, tissue-like materials [Figure 5C][190]. Its capability to deposit multiple ink materials also enables integration of sensors or stimulators tailored to organ needs. For example, printing biocompatible conductive inks for cardiac or neural patches.

Large-area fabrication

Electrospinning

Electrospinning produces nonwoven mats of micro- or nanofibers by ejecting a polymer solution (often containing conductive nanomaterials) under a high-voltage field. The collected fiber mesh can cover large areas and serves as an inherently porous, flexible substrate or conductor network. Electrospinning is a low-cost, high-throughput technique that can be scaled up for mass production[193]. The resulting fibrous networks have high surface area and can be made from biocompatible polymers, making them well-suited for interfaces with tissues and organs [Figure 5D][194]. In organ-specific bioelectronics, electrospun scaffolds can function as soft electrodes or supporting matrices that conform over an organ's surface. For example, a cardiac patch with electrospun conductive fibers that wrap around the heart [Figure 5E][195]. Because the fiber diameter and mesh architecture can be tuned, this method allows optimization of mechanical properties (matching the compliance of an organ) while covering the target area uniformly with conductive pathways. Moreover, electrospun electronics often remain breathable and permeable, an advantage for long-term tissue integration.

Cut-and-pattern assembly

"Cut-and-paste" fabrication is a paper-crafting-inspired approach where electronic patterns are first cut out of thin conductive films or foils and then transferred onto target substrates[51,196]. In practice, this often uses a desktop vinyl cutter or laser cutter to outline circuits from metal-coated polymer sheets (such as gold on polymer). The patterned pieces (or the negative stencil) are then pasted onto surfaces like medical tape, skin patches, or organ models. This method bypasses the complexities of photolithography, achieving circuit feature sizes on the order of ~100 μm with simple equipment. It is highly suitable for large-area flexible electronics because it is not constrained by wafer size, allowing meter-long flexible circuits to be fabricated using roll-to-roll feedstocks[183,197]. In organ-specific device fabrication, cut-and-paste approaches have been used to rapidly prototype epidermal sensors and electronic tattoos that conform to the skin, as demonstrated in epidermal electronic systems (EES) and related wearable devices[198]. The same concept can also be extended to creating custom-shaped electronics for organs, such as cutting sensor meshes that match the geometry of heart or brain surfaces[199]. The main advantages of this approach are speed and low cost, although the achievable resolution is lower than that of photolithographic methods. In addition, this technique can be integrated with other large-area fabrication strategies, for instance by using a cut stencil to spray-coat or print conductive inks in subsequent steps, thereby combining pattern definition with large-area surface coverage.

Solution-based thin film coating

Chemical solution deposition involves coating a substrate with functional material from a liquid phase over a large area[200]. Techniques in this class include spin coating, dip coating, blade coating (also known as doctor blading), and similar solution-based film deposition methods. They enable uniform, thin films of conductive or active materials (polymers, nanoparticle inks, *etc.*) to be laid down over device areas far larger than typical wafers. For instance, one can spin-coat a conductive polymer across an entire flexible sheet, or dip-coat a 3D object to blanket it with a conformal conductive layer. These processes are simple, rapid, and cost-effective, often requiring only basic lab equipment. In organ-bioelectronics, solution deposition can be used to create continuous electrodes or sensing skins that cover whole organ surfaces (like a "electronic membrane" for an organ). While the coatings themselves are unpatterned, they can be combined with shadow masks or post-deposition patterning (laser ablation, selective etching) to define electrode regions. An example application is coating a balloon catheter or a neural probe with an even layer of conductive polymer to improve electrical interface over its entire surface area.

3D conformal fabrication

Fabricating bioelectronics for three-dimensional curved surfaces such as the epicardium of the heart or spherical organoids presents unique challenges[201]. The strategies discussed in this subsection aim to create devices that not only cover large areas but also conform closely to non-flat geometries[202]. Methods including direct three-dimensional printing, embedded printing, spray coating, and conformal molding enable the integration of electronic components with complex three-dimensional structures. These approaches ensure that sensors and electrodes maintain intimate contact with target tissues across curved surfaces and during dynamic movements, which is essential for achieving stable and reliable bioelectronic performance *in vivo*[203]. Accordingly, emphasis is placed on achieving mechanical conformability and three-dimensional architecture during the device fabrication process.

3D printing

3D printing (additive manufacturing) provides a direct route to build 3D device architectures that follow organ geometry. For organ-specific bioelectronics, this enables anatomically contoured implants, or electrode lattices whose overall shape and local electrode placement are defined in a single fabrication sequence. High-resolution methods such as two-photon polymerization (2PP) are particularly relevant because they produce sub-micrometer features and truly 3D structures on top of planar microelectronics. In a recent example, 2PP was used to print polymer microelectrode templates directly onto a CMOS array, followed by conformal metal coating, patterning, and passivation to yield 6,600 tissue-penetrating microelectrodes at 35 μm pitch, with independently tunable height and tip geometry[204]. These pillar electrodes penetrate the retina and place the recording sites within the retinal ganglion cell layer while avoiding the axon bundle layer, enabling high-density, large-area recordings with reduced axonal interference. This type of "direct-print" 3D electrode strategy shows that additive manufacturing can separate the design of tissue-interfacing microstructures from the underlying electronics [Figure 6A][205]. It enables organ-matched curvature, depth targeting, and heterogeneous electrode shapes within a single array. These capabilities are difficult to achieve with planar lithography alone. For instance, researchers have printed flexible electronics in a dome shape to fit a cardiac ventricle[201]. In this work, a scaled anatomical heart model is used as a 3D template for device construction. The electronic components are fabricated as serpentine metal meshes by conventional planar processing, but the overall 3D geometry and mechanical properties are set by a soft elastomeric membrane cast onto the 3D-printed heart model. After demolding, the electronics-integrated membrane can conformally wrap the entire epicardial surface, forming a stable bioelectronic interface with the heart. Beyond *ex situ* fabrication, recent advances have begun to extend additive manufacturing directly into living tissues. For example, Wei Gao and colleagues recently demonstrated *in vivo* ultrasound printing, where acoustic fields are used to localize and polymerize bioinks

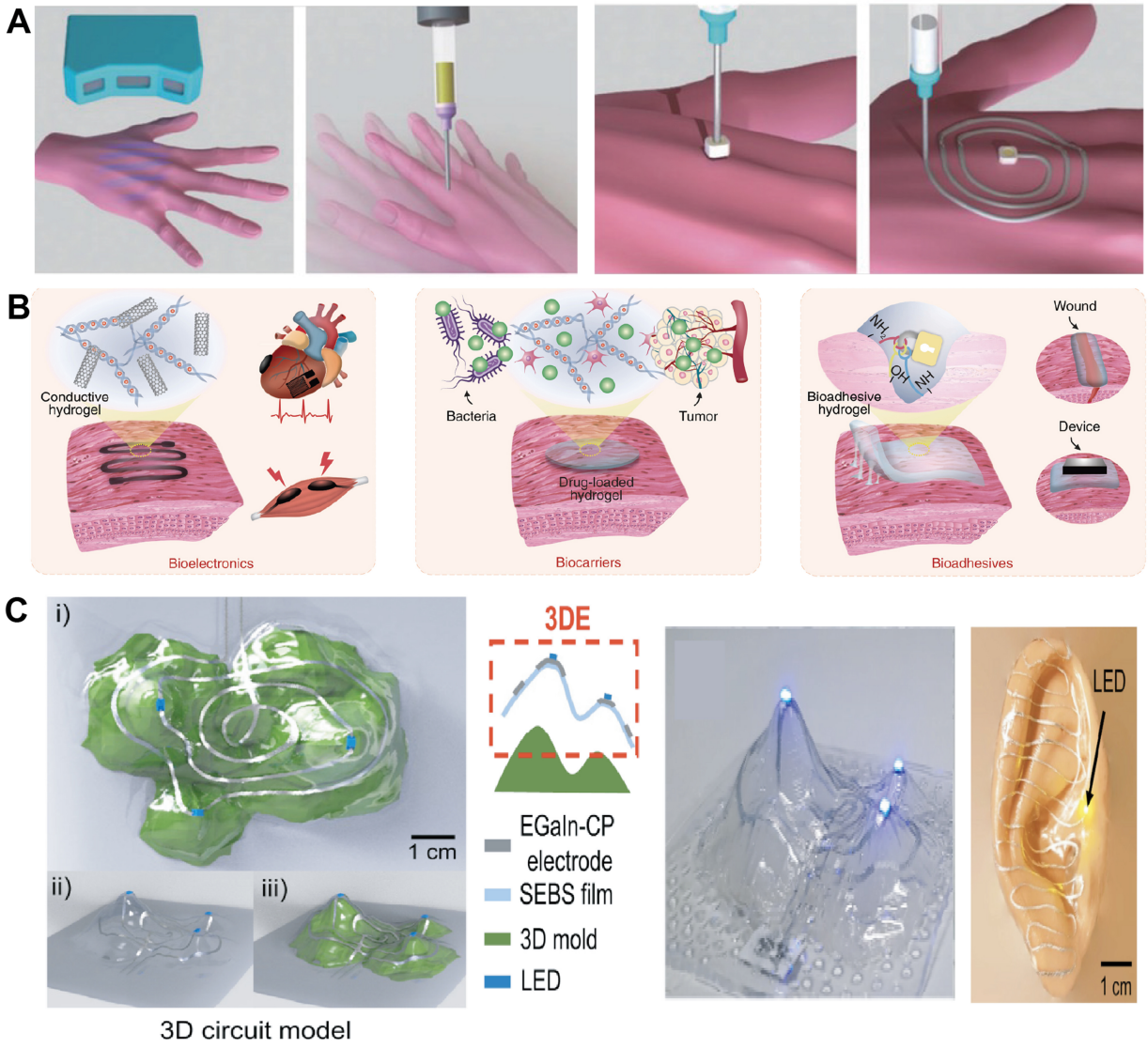

Figure 6. 3D conformal fabrication strategies for bioelectronics. (A) Adaptive 3D printing[205]. Copyright © 2018, WILEY-VCH Verlag GmbH & Co. KGaA; (B) Image-guided *in vivo* sound printing in deep tissues[206]. Copyright © 2025, the American Association for the Advancement of Science; (C) Conformal fabrication of 3D circuits on complex curvilinear surfaces[215]. Copyright © 2021, the American Association for the Advancement of Science. EGaIn-CP: eutectic gallium–indium with copper particles; SEBS: styrene–ethylene–butylene–styrene; LED: light-emitting diode.

within deep tissues, enabling the formation of functional structures directly inside the body[206]. This approach represents a paradigm shift from pre-fabricated implants toward minimally invasive, *in situ* fabrication of bioelectronic components, potentially allowing device formation in anatomically constrained or otherwise inaccessible regions [Figure 6B]. By choosing biocompatible, soft materials (e.g., stretchable inks or polymer composites), printed electronics can combine precise micro-patterning with bulk 3D shapes[207]. In this way, 3D printing directly contributes to organ-interfacing devices by co-defining both device geometry and microstructure, reducing the need to bend or mechanically deform flat electronics to fit complex organ topographies[208].

Spray coating

Spray coating (or spray printing) atomizes a solution or dispersion of functional materials into fine droplets and deposits them onto a target surface. Because the spray can wrap around edges and recessed regions, this

method is well suited for coating substrates with irregular or curved geometries. In soft bioelectronics, spray processes are commonly used to form thin, percolating films of conductive nanomaterials [e.g., silver nanowires (AgNWs), graphene, carbon nanotubes] or polymer blends over large areas in a single step, while allowing control over thickness and sheet resistance through the number of passes and ink concentration[209]. The technique is fast and can produce ultra-thin, uniform layers across the entire target, from convex domes to concave wells[210]. Xu *et al*. exemplified this strategy by spray printing AgNWs onto multiscale porous styrene–ethylene–butylene–styrene (SEBS) elastomer substrates to construct on-skin electronic devices with integrated passive-cooling capability[211]. The porous SEBS provided strong mechanical interlocking and π-π interactions with the AgNWs, allowing the sprayed networks to adhere firmly to the sponge-like surface while preserving high breathability and low thermal resistance. In this way, a single spray-coating step defined conductive traces and electrodes across complex, gas-permeable elastomer sheets, enabling multifunctional e-skin devices for electrophysiological sensing, temperature monitoring, and Joule heating without resorting to multi-step lithography. For organ-specific bioelectronics, spray coating can be applied to preformed 3D supports such as balloon catheters or molded elastomer shells[212]. A conformal conductive or sensing layer can be deposited on the entire surface or in selected regions through patterned masks, while the underlying porous or compliant substrate sets the mechanical properties and curvature[183,213]. Because spray processes are compatible with many polymers, hydrogels, and elastomers and are already mature in industrial settings[214]. They provide a practical route to scalable, conformal coatings for implantable or on-organ devices that must match complex anatomy yet remain thin, breathable, and mechanically compliant[203].

Conformal molding

Conformal molding refers to fabrication schemes in which electronic structures are formed or transferred while already constrained by a 3D target geometry. Instead of fabricating devices only on flat wafers and later forcing them to bend, the substrate or stamp is shaped to match the desired curvature, and micropatterns are created or printed in this state. For example, Choi *et al.* fabricated electrodes, interconnects, and LEDs on thin elastomeric films and then thermoformed the patterned circuits onto curved 3D molds, enabling conformal electronic systems on complex curvilinear surfaces [Figure 6C][215]. Alternative approaches use compliant stamps bearing micro- and nanopatterns (for example, PDMS or other elastomers); under heat and pressure, these stamps conform to spherical or highly curved surfaces and imprint conductive features directly onto them[184]. In all cases, the goal is to obtain intimate, continuous contact over the full 3D surface, which is essential for stable electrical coupling and reduction of local stress concentrations at the tissue-device interface.

Conformal additive stamp (CAS) printing provides a representative example of this concept applied with wafer-level precision. In CAS printing, ultrathin devices or "inks" are first fabricated on planar donor wafers using standard microfabrication, then retrieved by a pneumatically inflated elastomeric balloon and printed onto non-developable 3D substrates such as hemispherical shells and contact-lens molds. The balloon stamp deforms to match complex curvilinear surfaces while keeping the strain in brittle silicon elements below fracture limits, enabling high-yield transfer of photodetector arrays, antennas, hemispherical solar cells, and multifunctional smart contact lenses[216,217]. For organ-specific bioelectronics, similar conformal molding strategies allow sensor grids and electrode meshes to be fabricated in pre-curved forms that match the epicardium, ocular globe, or visceral organs[201,216,218]. Once released from the mold, these meshes "hug" the organ with minimal additional fixation and can be combined with stretchable designs to accommodate physiological motion while maintaining the imposed 3D shape.

Table 3. Organ- and application-specific requirements for bioelectronics

Organ	Modulus	Strain	Readout	Devices	Ref.
Brain/surface	0.1-10 kPa	Micromotion	EEG/LFP	ECoG/patch	[85,219]
Brain/penetrating	0.1-10 kPa	Micromotion	Spikes/LFP	Probe	[162,220]
Heart	8-15 kPa	20%-60%	ECG/EGM	Patch/mesh	[202,221-223]
Gut	0.1-3 MPa	50%-100%	EGG/pH	Patch/capsule	[170,224-226]
Skin/epidermis	0.4-20 MPa	20%-30%	ECG/EMG	Patch/tattoo	[86,167,227]

EEG: Electroencephalography; LFP: local field potential; ECoG: electrocorticography; ECG: electrocardiography; EGM: electrogram; EGG: electrogastrogram; EMG: epidermal and muscular electromyography.

APPLICATIONS OF ORGAN-SPECIFIC BIOELECTRONICS

The emerging landscape of bioelectronics is shifting from generic implantable or wearable devices toward organ-specific architectures engineered to match tissue stiffness, curvature, biochemical milieu, motility, and signal characteristics. Such precision interfacing is expected to improve long-term stability, reduce foreign-body responses, and enhance the fidelity of sensing or stimulation. Here, we highlight representative strategies for the brain, GI tract, skin, and visceral organs, including the heart, liver, kidney, and lung, with emphasis on how organ-level constraints guide material selection, device mechanics, and system-level design.

To clarify these design constraints, Table 3 summarizes representative mechanical environments, deformation modes, readouts, and device formats across major soft-tissue interfaces. Because interface requirements vary with application mode, device configuration, and tissue-contact depth, organ-specific bioelectronics should be viewed as a coupled design problem rather than a fixed organ-by-organ material selection.

Bioelectronics for the brain

The brain is among the most demanding application areas for organ-specific bioelectronics. Neural interfaces must accommodate extremely soft tissue that is mechanically fragile and highly immunoreactive, while detecting low-amplitude electrical signals with long-term stability. Traditional silicon or metaxl electrodes have a modulus order of magnitude higher than brain tissue, so even micromotion of the brain leads to micro-injury, glial scar formation, and progressive signal loss. Thus, organ-specific neural bioelectronics focus on minimizing mechanical and biochemical perturbation, essentially designing the interface as a tissue-like, long-lived "electronic scaffold" rather than a rigid probe. This means building electrodes and wires that more closely match the brain's softness, curvature, and microstructure to achieve *in vivo* "stealth". For example, the device becomes mechanically and structurally invisible to the host tissue over time.

To achieve this, mesh and injectable electronics employ sub-micrometer-thick, microporous architectures that can be delivered through fine needles and then unfold within neural tissue, reaching bending stiffness comparable to brain and allowing cellular infiltration with greatly reduced chronic inflammation and gliosis. Zhou *et al.* showed that syringe-injectable mesh electronics can integrate seamlessly with brain tissue while eliciting minimal chronic immune response, thereby enabling stable long-term neural interfacing[228]. In follow-up studies, researchers showed that such ultra-flexible mesh probes produce virtually no gliosis even months after implantation. Neurons and neurofilaments grow through the mesh structure by 12 weeks post-implantation, and the distribution of astrocytes and microglia becomes nearly uniform at the probe-tissue interface. This seamless tissue integration allowed stable neural recordings over three months with negligible signal loss. These findings highlight that making implants as soft and open as the brain is a

viable route to chronic biocompatibility and signal stability. Researchers are complementing this approach with ultrathin inorganic conductors and soft packaging: for example, thin gold nanomembranes in serpentine (fractal) layouts, liquid-metal interconnects, and conductive polymer hydrogels have been used to create electrodes and wires that deform readily with brain movement. Such designs drastically lower the effective stiffness of the device and reduce interfacial stress, further mitigating foreign-body response.

Another organ-specific innovation for brain interfaces is the use of organic electronic materials that can directly amplify neural signals on-site in the aqueous, ion-rich environment of the brain. Traditional metal electrodes pick up microvolt-level local field potentials that then must be amplified by distant circuitry, which limits SNR. OECTs, made from mixed ionic–electronic conductors like PEDOT:PSS, have been developed as flexible, biocompatible amplifying electrodes. OECTs operate at low voltages and use ions from the tissue to modulate their conductivity, effectively acting as local amplifiers at the tissue interface. Polymer OECT-based electrode arrays on ultrathin plastic films can record brain activity with much higher SNR than conventional electrodes, due to the transistors' built-in amplification[229]. *In vivo* tests of these transistor arrays on the cortical surface showed they could detect low-amplitude neural oscillations and even epileptiform spikes that were previously indiscernible with passive electrodes. Such OECT-based brain interfaces have been used for electrocorticography with excellent signal quality, and they represent a promising route to multiplexed neural sensors that record both electrical and neurochemical signals from the cortex[230].

The convergence of these organ-matched innovations is yielding a new generation of neural interfaces that can potentially operate stably for years. In the coming years, such networks of organ-specific bioelectronic devices could enable closed-loop neuromodulation therapies and high-bandwidth brain-computer interfaces that remain reliable over the patient's lifetime. However, translating these experimental systems into clinical practice will require overcoming several key challenges. Scalable manufacturing and deployment of micro-scale, polymeric devices must be achieved to cover large brain areas or thousands of channels. Soft packaging is an unsolved problem, which can protect electronics from corrosion by body fluids, typically requires rigid, glass-like encapsulants, so new strategies (e.g., thin-film coatings) are needed to robustly seal circuits while preserving elasticity[231]. Fully wireless operation is another challenge: integrating onboard power sources or developing safe wireless power and data links is crucial for untethered implants[232]. Despite these challenges, the progress in brain-specific bioelectronics demonstrates a clear path forward: by engineering devices to physically and functionally blend into the brain's own environment, we can improve the longevity and fidelity of neural interfaces, unlocking new capabilities in neuroscience research and clinical neuromodulation.

Beyond the mechanically matched mesh electrodes and OECT arrays discussed above, recent work has also introduced biologically assisted routes for delivering soft electronic devices into the brain. Sheng *et al.* reported an implantation approach during early brain development: ultra-soft mesh electronics were laminated onto the embryonic neural plate and subsequently carried inward as the plate folded and expanded into the three-dimensional brain[233]. Liang *et al.* developed a silk-enabled conformal intraventricular interface that can self-unfold and conformally adhere to periventricular brain structures, enabling minimally invasive and chronically stable neural recordings [Figure 7A][234]. These "developmentally integrated" implants enabled long-term recording of neural activity from single cells to populations, while markedly reducing gliosis and scar formation, indicating a relatively gentle and durable structural and functional integration with the host tissue during brain formation. Building on this concept, Yadav *et al.* used immune cells as carriers: subcellular wireless stimulators were covalently attached to circulating monocytes, which then trafficked to inflamed brain regions via endogenous immune surveillance pathways[235]. Following intravenous injection, the resulting cell-electronics hybrids provided targeted,

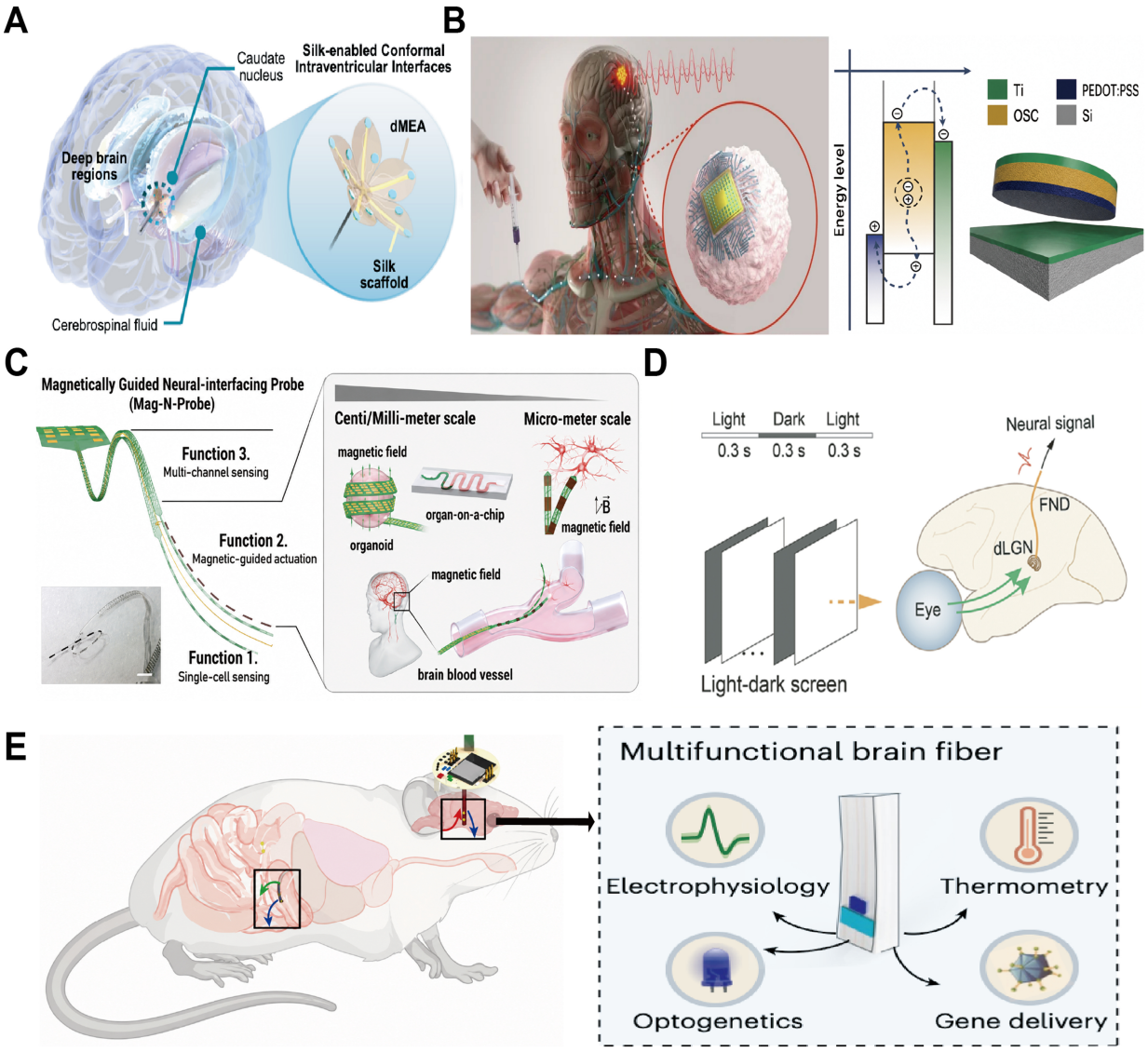

Figure 7. Bioelectronics for the Brain. (A) Silk-enabled conformal intraventricular interface for minimally invasive and chronically stable neural recordings[234]. Copyright © 2025, published by Springer Nature; (B) Cell–electronics hybrid enabling nonsurgical and targeted focal neuromodulation[235]. Copyright © 2025, published by Springer Nature; (C) Magnetically guided neural-interfacing probe for steerable implantation and multifunctional sensing[236]. Copyright © 2025, WILEY-VCH Verlag GmbH & Co. KGaA; (D) Soft-fiber neural interface for stable long-term recording under dynamic *in vivo* conditions[237]. Copyright © 2024, WILEY-VCH Verlag GmbH & Co. KGaA; (E) Multifunctional microelectronic fibers for multimodal brain interfacing and wireless neural modulation[238]. Copyright © 2023, published by Springer Nature. dMEA: Deformable microelectrode array; PEDOT:PSS: poly(3,4-ethylenedioxythiophene):poly(styrenesulfonate); OSC: organic semiconductor; FND: fiber neural device; dLGN: dorsal lateral geniculate nucleus.

wirelessly addressable focal neuromodulation without craniotomy or conventional leads [Figure 7B]. Taken together, these "development-driven" and "cell-driven" implantation modes illustrate an important direction in which soft neural interfaces are deployed by leveraging intrinsic biological processes, with the goal of reducing surgical trauma and improving long-term stability of the brain-device interface.

In parallel with these advances in implantation, the physical form of neural interfaces is evolving from locally fixed probes toward fiber-like systems whose position can be adjusted *in vivo*. Kim and colleagues embedded magnetic nanoparticles into ultra-flexible conductors to create magnetically responsive probes that can be steered by external fields, enabling centimetre-scale navigation in tissue and near single-cell positioning for

high-resolution electrophysiology [Figure 7C][236]. Tang *et al.* reported an axon-like soft-fiber bioelectronic device that enables reliable *in vivo* neural recording even under vigorous activities; the representative visual-stimulation experiment further demonstrated stable recording of light-evoked neural activity from the dLGN [Figure 7D][237]. Further extending this concept, Sahasrabudhe *et al.* developed multifunctional microelectronic fibers that enable wireless modulation of both gut and brain neural circuits, highlighting the potential of soft fiber-based bioelectronics for multimodal interfacing across distributed organ systems [Figure 7E][238]. Collectively, these guided and reconfigurable high-density soft fibres indicate that future soft brain interfaces are trending toward devices that not only match organ mechanics and materials, but also offer dynamic spatial adjustability to track and modulate neural circuits as they evolve over time.

Bioelectronics for the GI tract

The GI tract's continuous peristaltic motion and harsh chemistry (stomach acid, enzymes) presents a unique challenge for bioelectronic devices. Any implant or ingestible in the GI system must combine robust mechanical compliance with corrosion-resistant encapsulation to survive acidic gastric fluids. At the same time, these devices should not interfere with normal GI motility or cause obstruction. Researchers are therefore developing organ-specific GI bioelectronics that can function reliably in this extreme milieu while remaining biocompatible with the digestive system's motions and secretions.

One representative approach uses ingestible hydrogel systems that swell into soft, stomach-conforming spheres, enabling weeks-long gastric residence for continuous sensing (e.g., core temperature) while avoiding premature expulsion or mucosal irritation seen with rigid capsules; these devices can be actively deswelled on demand for safe retrieval[239]. More broadly, recent ingestible platforms combine miniaturized electronics, wireless links, and advanced materials to realize gastric- or intestinal-resident devices: star- or cage-like geometries, tough hydrogels, and elastomeric arms unfold post-swallow to resist pyloric passage yet disassemble on cue for clearance[240-242]. In parallel, micro-bio-electronic capsules integrate engineered bacteria with low-power electronics for *in situ* detection of GI bleeding and inflammatory biomarkers[243]. To protect such complex devices through the GI tract, researchers are exploring specialized encapsulation materials. These include biodegradable polymers like silk fibroin and PLGA, as well as pH-responsive enteric coatings and hydrogel coatings that remain intact in stomach acid but dissolve or become permeable in the intestines, thereby releasing drugs or exposing sensor surfaces only at the desired GI location[244]. A growing body of work from Traverso, Langer and colleagues has outlined design principles for these ingestible systems, emphasizing that devices must balance mechanical durability (to withstand gastric contractions), triggerable disassembly or transit, and careful biocompatibility with GI physiology to avoid irritation[245]. In essence, the materials and designs are tuned to match the GI environment. In parallel, researchers are innovating ways to power GI devices without batteries, harnessing the body's own chemical energy. Battery-free and biochemically powered capsules have been demonstrated that harvest energy from intestinal fluids using onboard biofuel cells. A recent ingestible sensor capsule used a glucose-fueled biofuel cell to generate power from sugars in the small intestine, allowing it to continuously monitor gut glucose levels and transmit data via a wireless (magnetic) telemetry link[244]. All in a pill-sized form factor with no internal battery. By exploiting biofuel or even mechanical energy from gut peristalsis, these designs eliminate the need for bulky batteries and toxic battery materials, making the capsules smaller and safer for long-term use. Such self-powered capsules have successfully performed real-time chemical sensing in animal models, proving the feasibility of sustained metabolite monitoring using only the energy available inside the GI tract.

These innovations accelerate gut-specific bioelectronics that are now capable of comprehensive, real-time monitoring and intervention. Swallowable devices have been built to map a wide range of GI parameters: they can measure local pH, temperature, pressure, gas composition, and even sense biomarkers of infection or inflammation in different sections of the gut. Despite the rapid progress, significant challenges remain

before GI bioelectronics can be broadly deployed. A key challenge is ensuring that devices can maintain secure adhesion or positioning at a target site under the constantly moving, contracting conditions of the gut. Even for devices designed to stick to the mucosal wall, the GI tract's continual peristalsis and mucus turnover can make long-term attachment difficult. Recent work on bioadhesive hydrogel interfaces is tackling this by improving electrode-tissue contact in the stomach, but robust retention over weeks is still an open problem[225]. Another challenge is engineering encapsulation and coatings that can withstand months of corrosive attack by gastric acid and enzymes. By surmounting these challenges, GI bioelectronics will be poised to revolutionize how we monitor and treat diseases of the digestive tract, offering a future of smart pills that can diagnose, report, and respond from within our own bodies.

The first is chronically resident platforms for large-area physiological mapping. Kong *et al.* used 3D-printed gastric-resident frameworks with deployable arms to keep electronics in the stomach for weeks, addressing the long-standing problem that conventional capsules are cleared within hours and cannot follow slow disease dynamics [Figure 8A][242]. Building on this concept, Boys *et al.* developed implantable bioelectronic platforms for *in vivo* gut electrophysiology, demonstrating stable recording of GI electrical activity in mice, rats, and pigs, thereby highlighting the translational potential of chronic bioelectronic interfaces for the digestive tract [Figure 8B][246]. Gopalakrishnan *et al.* reported a wireless smart capsule that integrates pH and oxidation–reduction potential sensing to profile inflammatory status throughout the GI tract, providing a promising strategy for *in situ* monitoring of reactive oxygen species (ROS)-related gut inflammation and inflammatory bowel disease [Figure 8C][247]. Taken together, these studies show that resident geometries, soft ribbons and tissue-matched implants can convert the GI tract from a transient passage into a site for stable, organ-scale physiological monitoring.

The second emerging direction is self-powered, multimodal biochemical profiling. De la Paz *et al.* demonstrated an ingestible capsule in which a glucose-fuelled biofuel cell harvests energy from luminal sugars in the small intestine to power onboard electronics, allowing continuous wireless monitoring of intestinal glucose without a conventional battery [Figure 8D][244]. Building on earlier bacteria–electronics capsules for detecting GI bleeding, Inda-Webb *et al.* recently reported a sub-1.4 cm^3 device that integrates genetically engineered probiotic biosensors with a custom photodetector and readout chip, allowing *in situ* detection of highly labile inflammatory mediators such as nitric oxide and hydrogen sulfide directly inside the gut[248]. By combining biochemical energy harvesting with high-density sensing, these devices directly tackle two bottlenecks of earlier capsules - short energy budget and single-analyte readout - and point toward GI systems that can profile metabolites, inflammatory mediators and therapeutic drugs over long time scales. Together with advances in adhesive interfaces and corrosion-resistant encapsulation, these two classes of technologies outline a realistic path toward future GI bioelectronics that not only survive the harsh digestive environment, but also provide continuous, organ-specific information to guide the management of motility disorders, inflammatory bowel disease and metabolic syndromes.

Bioelectronics for the epidermis

As a large, accessible and regenerating organ, skin offers a convenient interface for non-invasive bioelectronics but imposes requirements distinct from internal tissues. The epidermis is continuously renewed and subjected to multiaxial stretching, shear, perspiration and environmental exposure; contact is often intermittent and user controlled. Effective skin-mounted devices must therefore be ultrathin, compliant, breathable and hypoallergenic, forming intimate yet reversible contact without occlusion or irritation, while maintaining stable impedance during daily activities[81]. As highlighted in Bao's recent review on skin-inspired bioelectronic materials and systems, effective epidermal devices must therefore combine ultralow bending stiffness, conformal yet gentle adhesion, gas and moisture permeability, and stable electrical performance under daily wear, while using materials that are safe for long-term contact with human skin.

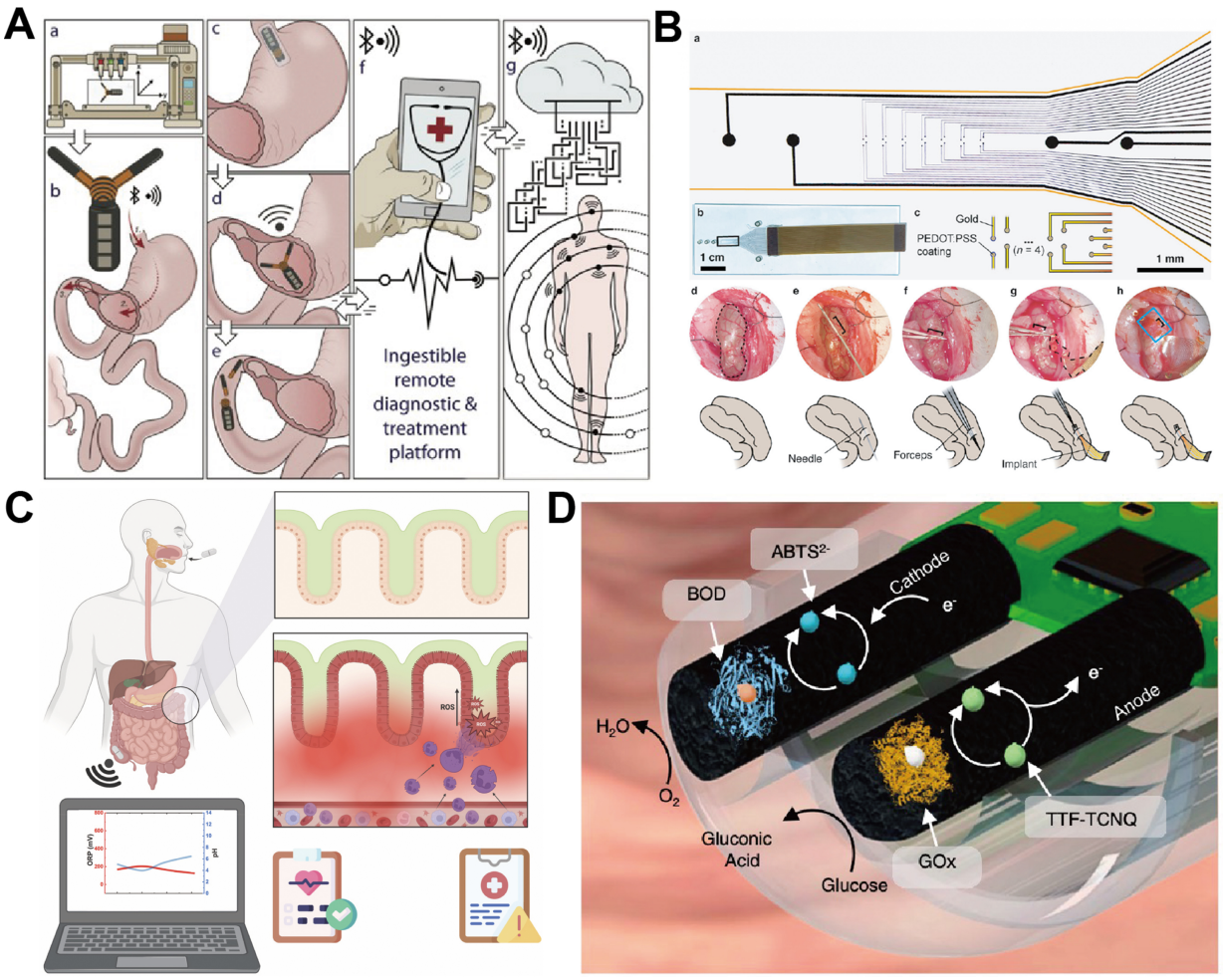

Figure 8. Bioelectronics for the GI tract. (A) Gastric-resident electronics for prolonged physiological monitoring[242]. Copyright © 2019, WILEY-VCH Verlag GmbH & Co. KGaA; (B) Implantable bioelectronics for gut electrophysiology[246]. Copyright © 2025, published by Springer Nature; (C) Smart capsule for inflammation profiling throughout the GI tract[247]. Copyright © 2023 The Authors, published by Elsevier; (D) Self-powered ingestible wireless biosensing system for *in situ* metabolite monitoring[244]. Copyright © 2022, published by Springer Nature. PEDOT:PSS: Poly(3,4-ethylenedioxythiophene):poly(styrenesulfonate); BOD: bilirubin oxidase; ABTS: 2,2′-azino-bis(3-ethylbenzothiazoline-6-sulfonic acid; GOx: glucose oxidase; TTF-TCNQ: tetrathiafulvalene-7,7,8,8-tetracyanoquinodiamethane.

In epidermal bioelectronics, skin bioimpedance can serve as a physiological readout, while skin–electrode interfacial impedance provides a practical indicator of device–skin coupling quality. Elevated interfacial impedance can reduce signal fidelity and increase susceptibility to motion artifacts during long-term wearable monitoring[249]. Therefore, soft, conformal, and hydrated contact layers are widely used to enhance skin coupling and lower contact impedance[250]. Beyond electrophysiological recording, impedance-based measurements can also provide information on skin hydration, tissue condition, and barrier-state changes, making them useful for monitoring both epidermal physiology and device–skin interface stability[251].

Pioneering EES by Kim *et al.* matched thickness, modulus, and bending stiffness of ultrathin serpentine metal circuits laminated on elastomer substrates to that of human skin, creating "mechanically invisible" devices that adhere via van der Waals forces alone[51]. Subsequent developments include tattoo-like electronics that can be transferred as temporary decals, large-area "electronic skin" matrices using organic semiconductors and nanomaterials for distributed pressure and temperature sensing, and conductive hydrogel or ionic patches that provide low-impedance, soft contacts for biopotential recording and

neuromodulation[252]. Porous elastomers, microperforated films, nanofiber meshes, and Janus structures have been introduced to enhance breathability and moisture management, while new supramolecular and bioadhesive chemistries enable strong yet painless removal, tailored for sensitive skin and long-term wear.

Rogers and Kim's group first demonstrated EES in which ultrathin serpentine metal interconnects on elastomer substrates were engineered to match the thickness, modulus and bending stiffness of the stratum corneum, creating "mechanically invisible" patches that adhere via van der Waals forces alone and deliver high-quality electrocardiography (ECG), EMG and electroencephalography (EEG) recordings. Building on this mechanical-matching concept, Du *et al.* recently developed a self-compliant ionic nanomesh: a highly porous network of elastic polymer nanofibres infiltrated with ionic conductors that laminates on skin with almost no normal stress [Figure 9A][253]. The nanomesh architecture is intrinsically gas-permeable and mechanically unconstraining, enabling stable electrophysiological recordings during sweating and motion while greatly reducing occlusion-induced irritation over multi-day wear. Together, these studies show how matching not only modulus but also normal stress and breathability is key to organ-specific epidermal interfaces.

To improve robustness under everyday mechanical damage, Son and colleagues realized reconfigurable assemblies of self-healing stretchable transistors and circuits[254]. Their platform uses self-healing polymer dielectrics and elastomeric conductors to form transistor "islands" connected by self-healable interconnects; after cutting and rejoining, both mechanical integrity and circuit functions such as inverters and ring oscillators recover with minimal performance loss. In parallel, Lee *et al.* reported a rapidly self-healing electronic skin in which a disulfide-bonded TPU matrix and printed conductive networks allow the device to restore over 80% of its mechanical strength and conductivity within seconds at room temperature after being fully severed [Figure 9B][255]. Integrated with low-power wireless electronics and on-patch machine-learning algorithms, this e-skin can continuously monitor motion and classify levels of muscular fatigue in real time. These examples illustrate how molecularly engineered self-healing polymers can extend device lifetime and support intelligent, long-term epidermal monitoring.

Bao's group has advanced skin-like active biosensors based on stretchable organic electronics to address signal drift and long-term stability. In their recent work on drift-free biosensors with stretchable diode-connected organic field-effect transistors (OFETs), they constructed diode-connected OTFT pixels on pre-strained elastomer substrates and integrated them with microfluidic sweat collectors and a soft wireless[256]. The diode configuration suppresses threshold-voltage drift and baseline fluctuations, enabling accurate, drift-free monitoring of sweat analytes such as glucose and cortisol during motion and temperature changes. Complementary efforts from the same group and others have shown that such stretchable OFET arrays can be reconfigured and combined with self-healing conductors, pointing toward adaptive epidermal circuits that maintain calibrated biochemical sensing over extended periods.

Skin-specific bioelectronics now underpin continuous ECG/EMG/EEG monitoring, motion and posture tracking, sweat-based biochemical sensing (electrolytes, metabolites and hormones) and transcutaneous stimulation for pain management, rehabilitation and cosmetic therapies. Beyond conventional biopotential sensing, impedance-based bioelectronic approaches have been explored for neuromuscular perception and human-machine interaction, including soft contact sensing *via* impedance imaging and muscle-driven joint torque estimation through voltage–torque mapping[257,258]. Recent work further demonstrates integrated impedance-based sensing platforms for enhanced neuromuscular perception in human–machine interfaces[259]. The field is converging toward multimodal patches that combine mechanical, electrophysiological and biochemical sensing with on-board processing and low-power wireless links, enabling "zero-burden" long-term monitoring in daily life. Looking ahead, integrating these mechanically

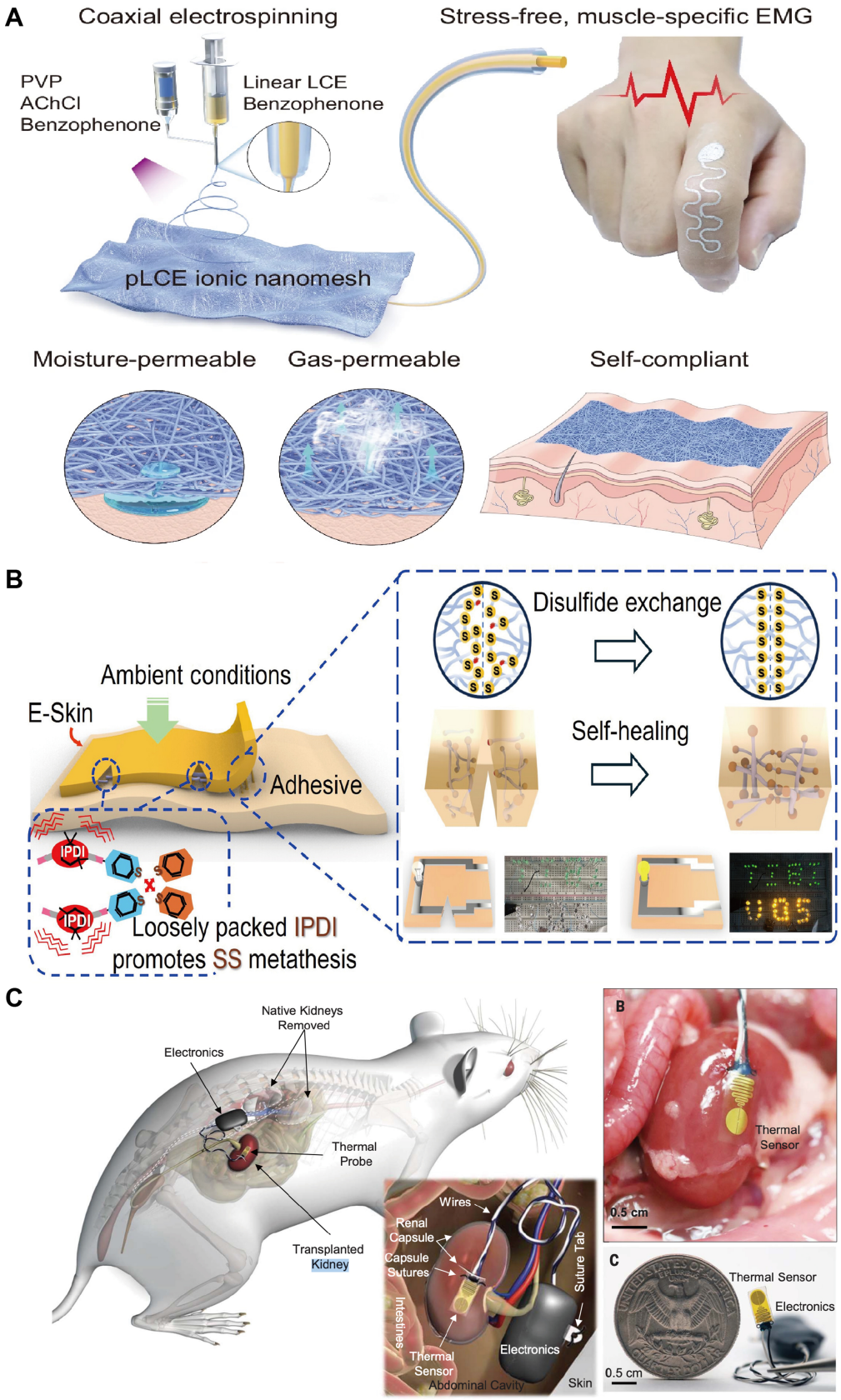

Figure 9. Bioelectronics for the epidermis and other visceral organs. (A) A self-compliant ionic nanomesh for skin electronics[253]. Copyright © 2025, published by Springer Nature; (B) A selfhealing electronic skin for movement evaluation[255]. Copyright © 2025, the American Association for the Advancement of Science; (C) A bioelectronic patch for monitoring organ transplant rejection[261]. Copyright © 2023, the American Association for the Advancement of Science. PVP: Polyvinylpyrrolidone; LCE: liquid crystal elastomers; pLCE: permeable LCE; EMG: epidermal and muscular electromyography; IPDI: isophorone diisocyanate; SS: disulfide bonds.

invisible, gas-permeable, self-healing and drift-free device concepts with sustainable substrates, secure data architectures and haptic or AR/VR interfaces will further position epidermal bioelectronics as organ-specific platforms at the intersection of healthcare, human–machine interaction and soft robotics.

Bioelectronics for other visceral organs

Among visceral organs, the heart represents one of the most prominent examples where organ-specific bioelectronics have achieved substantial progress. The heart's constant rhythmic beating (~1 Hz) and complex three-dimensional curvature create a challenging environment for conventional rigid implants such as metallic leads and canister pacemakers, which often introduce severe mechanical mismatch and require invasive fixation. To address these limitations, Xu *et al.* developed three-dimensional multifunctional integumentary membranes, thin silicone epicardial membranes conformally wrapped around the entire heart and integrated with stretchable arrays of electrodes, temperature and strain sensors, and stimulators[201]. These devices provide high-density mapping of electrical activity and mechanical deformation across the entire epicardial surface while accommodating large stroke-volume changes without delamination. Earlier electronic web architectures further demonstrated that serpentine metal interconnects embedded in elastomers can conform to highly curved cardiac geometries and tolerate large biaxial strains during beating[202]. Such systems enable simultaneous electrical mapping and pacing over large ventricular areas while maintaining stable contact with the epicardium.

Building on these epicardial platforms, soft electronics have also been integrated with interventional catheters. Recent catheter-mounted electronic arrays unfold at the catheter tip to form soft, conformal sensor patches that provide real-time multimodal feedback during procedures such as radiofrequency ablation and irreversible electroporation[260]. These systems enable co-registered measurements of temperature, contact pressure, and intracardiac electrograms at the tissue–catheter interface, illustrating how cardiac bioelectronics are evolving from point-like rigid electrodes toward organ-matched, high-coverage sensing and stimulation systems.

Beyond the heart, similar organ-specific strategies are being extended to other visceral organs. The kidney, for example, does not experience large rhythmic contractions but resides deep within the body and is subject to slow perfusion variations and motion associated with respiration. Recent studies have demonstrated that soft implantable bioelectronic interfaces can be laminated onto transplanted kidneys to continuously monitor local biophysical signatures associated with early graft rejection[261,262]. Such systems can potentially provide early warning signals of transplant rejection before conventional serum biomarkers rise. For instance, Madhvapathy *et al.* developed an ultrathin stretchable bioelectronic patch that is inserted beneath the renal capsule and wirelessly monitors local temperature and thermal conductivity as indicators of inflammatory activity in rat kidney transplants[261]. Continuous measurements revealed characteristic temperature variations associated with early rejection, enabling detection weeks before traditional biomarkers such as serum creatinine and BUN changed, while producing minimal foreign-body response at the organ surface [Figure 9C].

In addition to device architecture, interfacial surface properties play a critical role in determining the host response to visceral implants. Studies on soft-tissue implants have shown that surface topography can strongly influence immune cell recruitment and fibrotic encapsulation. Micro- and macro-textured surfaces, for example, can elicit distinct immune responses compared with smooth interfaces, suggesting that micron-scale topographical design may provide an additional strategy for mitigating fibrosis around long-term visceral bioelectronics[263].

Overall, bioelectronics for visceral organs highlight the importance of secure yet minimally invasive interfaces with wet, highly deformable tissues. Strategies under active development include bioadhesive hydrogel coatings that anchor devices to organ surfaces without sutures, wirelessly powered architectures that eliminate transcutaneous leads, and biodegradable electronics that function temporarily before safely dissolving. The convergence of cardiac epicardial electronics, kidney-specific monitoring systems, and therapeutic nanofluidic patches suggests that organ-tailored electronic platforms capable of sensing and intervention across internal organs are becoming increasingly feasible, opening new possibilities for proactive management of conditions such as arrhythmias, organ failure, and transplant rejection.

CONCLUSION AND OUTLOOK

Organ-specific bioelectronics is redefining soft bioelectronic interfacing from a largely generic flexible-device strategy toward a tissue-adapted design paradigm[264]. As highlighted throughout this review, recent advances in conductive and interfacial materials, structural and encapsulation platforms, and organ-matched fabrication strategies have greatly expanded the capability of bioelectronic systems to establish stable, conformal, and functionally integrated interfaces with soft tissues[49]. Across neural, GI, epidermal, and visceral-organ applications, these studies collectively show that high-performance biointerfacing depends not simply on making electronics softer, but on co-optimizing materials, structures, and device functions according to the distinct mechanical, biochemical, and immunological environments of individual organs[265].

Despite this rapid progress, the field still faces a fundamental knowledge gap in understanding what truly governs long-term device–tissue integration in different organs[161]. Soft and stretchable mechanics alone do not guarantee stable chronic performance. Instead, electrical reliability and biological acceptance are jointly shaped by interfacial hydration, ion transport, encapsulation integrity, local immune activity, fibrotic remodeling, and organ-specific modes of deformation[49,161]. These factors vary substantially across tissues such as the brain, skin, gut, heart, and kidney, making it difficult to define universal material-selection or interface-design rules[85]. As a result, the long-term interpretation of device performance, failure, and biological response remains incomplete, and more predictive frameworks that connect material properties with organ-specific *in vivo* behavior are still needed.

Beyond this biological complexity, major engineering challenges remain in translating promising material systems into robust organ-resident devices. In addition to passive monitoring, organ-specific bioelectronics are expected to evolve toward closed-loop systems that integrate real-time sensing, signal processing, and adaptive therapeutic modulation. Electrophysiological, chemical, mechanical, or thermal signals from the target organ could be used to guide feedback-controlled interventions, including electrical stimulation, drug or ion delivery, thermal regulation, and neuromodulation. Such closed-loop systems are particularly important for dynamic organs such as the brain, heart, gut, and skin, where pathological states evolve over time and require organ-adapted therapeutic responses.

Biocompatibility is another critical consideration for organ-specific bioelectronics, because material safety is strongly dependent on both the intrinsic chemistry of the material and the target tissue environment[159,266]. For epidermal systems, skin compatibility involves not only low cytotoxicity, but also breathability, irritation-free adhesion, sweat tolerance, and avoidance of allergic or inflammatory reactions during long-term wear[267]. For implantable interfaces, the major concerns shift toward chronic immune activation, fibrotic encapsulation, degradation products, ion or nanoparticle release, and long-term stability in wet and enzymatically active environments[268]. Different material systems therefore require different safety

considerations: conductive polymers and OMIECs require evaluation of dopants, additives, and electrochemical by-products; hydrogels require control over residual monomers, crosslinkers, swelling, and degradation behavior; liquid metals and metal composites require stable encapsulation to prevent leakage or metal ion exposure; and 2D materials require careful assessment of flake size, oxidation state, persistence, and inflammatory potential[269,270]. Soft conductors, conductive hydrogels, OMIECs, liquid-metal composites, and ultrathin encapsulation layers offer attractive combinations of compliance and electrical performance, but their long-term behavior under wet, ion-rich, enzymatic, and mechanically dynamic conditions is still not fully understood[271-273]. Repeated bending, stretching, swelling, delamination, biofluid infiltration, and material fatigue can progressively degrade adhesion, impedance stability, and signal fidelity[274]. At the same time, many high-performance organ-conformal devices still depend on low-throughput or highly customized fabrication workflows, which complicates reproducibility, scaling, and integration with clinically relevant device formats[275]. Thus, improving long-term stability while preserving manufacturability remains a central engineering challenge for the field.

At the system level, organ-specific bioelectronics is expected to evolve from passive sensing interfaces toward multifunctional and closed-loop platforms. Future systems will likely integrate multimodal sensing, localized stimulation or delivery, low-power signal conditioning, wireless communication, and adaptive data analysis within unified soft architectures[276]. In this context, materials innovation will continue to play a central role, including self-healing conductors and encapsulants, biodegradable and transient structural materials, adhesive and repair-supportive interfaces, and biohybrid or living components that improve chronic tissue integration[69,277,278]. In parallel, advances in wireless power transfer, miniaturized electronics, and AI-assisted interpretation of organ-level signals may enable more autonomous bioelectronic systems capable of not only monitoring disease progression, but also actively modulating tissue function in real time[49].

Ultimately, the transition of organ-specific bioelectronics from compelling laboratory demonstrations to widespread biomedical use will depend on scalable manufacturing, rigorous long-term validation, and clear demonstration of clinical utility [Figure 10][75]. This includes reproducible large-area and high-resolution fabrication, sterilization-compatible packaging, reliable implantation or resorption strategies, and systematic evaluation in chronic animal models and human-relevant settings[279,280]. Just as importantly, future progress will require closer integration among materials scientists, device engineers, biologists, and clinicians so that organ-specific platforms are designed not only for mechanical conformity and electrical performance, but also for compatibility with real surgical workflows and therapeutic needs. With such advances, organ-specific bioelectronics is well positioned to become a foundational technology for precision monitoring, localized intervention, and next-generation bioelectronic medicine.

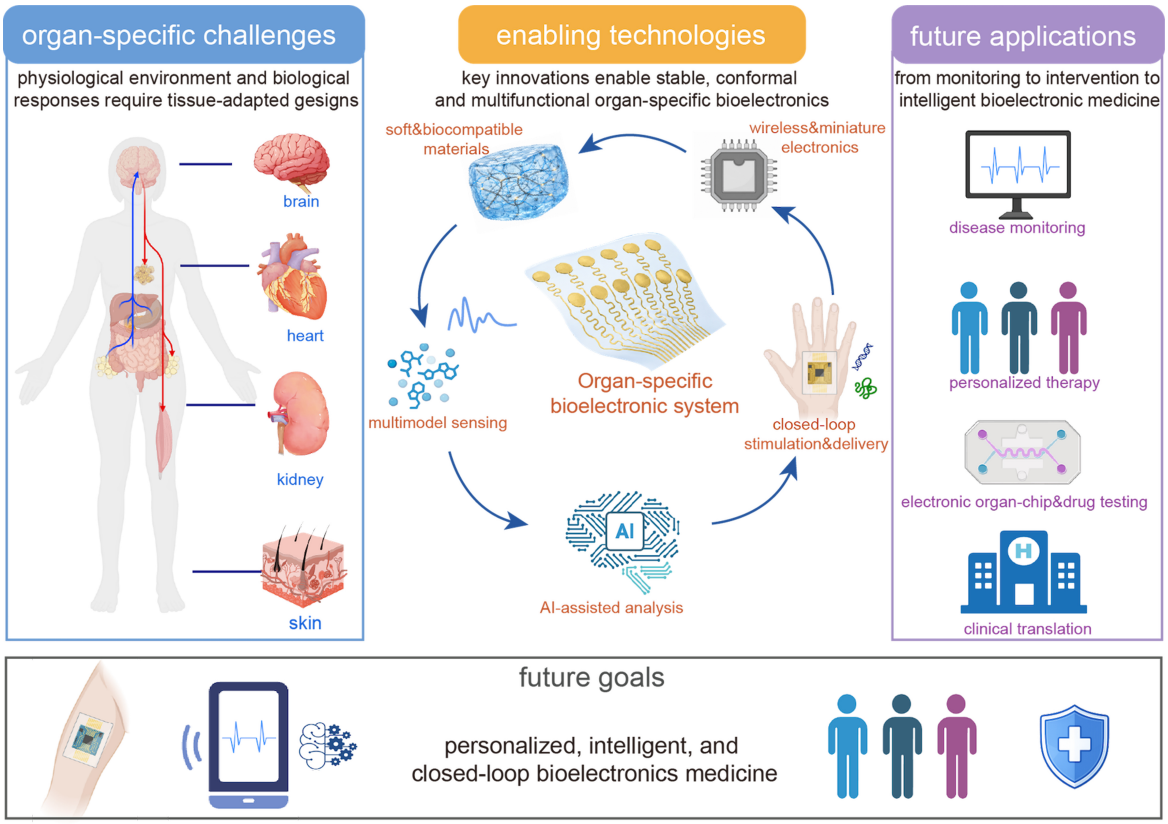

Figure 10. Future perspectives for organ-specific bioelectronics. Created by the authors using Adobe Illustrator. Some schematic elements created in BioRender. Liu, X. (2026) https://BioRender.com/k8j8vrc.

DECLARATIONS

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Authors' contributions

Designed the original draft: Liu, X.; Zhang, Z.
Wrote the original draft: Liu, X.; Zhang, Z.
Reviewed and revised the manuscript: Liu, X.; Zhang, Z.; Lim, C. T.
Provided comments and suggestions on the revision of the manuscript: Li, J.; Ge, Z.; Shen, S.

Availability of data and materials

Not applicable.

AI and AI-assisted tools statement

During the preparation of this manuscript, the AI tool ChatGPT by OpenAI based on GPT-4o (released May 13, 2024) was used solely for language editing. The tool did not influence the study design, data collection, analysis, interpretation, or the scientific content of the work. All authors take full responsibility for the accuracy, integrity, and final content of the manuscript.

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Conflicts of interest

Lim, C. T. is the Advisory Editor of the *Soft Science* journal. He had no involvement in the review or editorial process of this manuscript, including but not limited to reviewer selection, evaluation, or the final decision, while the other authors have declared that they have no conflicts of interest.

Ethical approval and consent to participate

Not applicable.

Consent for publication

Not applicable.

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REFERENCES

1. Chen, W.; Zhao, H.; Li, Y. Mitochondrial dynamics in health and disease: mechanisms and potential targets. *Signal. Transduct. Target. Ther.* **2023**, *8*, 333. DOI PubMed PMC
2. Li, T.; Shi, C.; Jin, F.; et al. Cell activity modulation and its specific function maintenance by bioinspired electromechanical nanogenerator. *Sci. Adv.* **2021**, *7*, eabh2350. DOI PubMed PMC
3. Shi, J.; Li, P.; Kim, S.; Tian, B. Implantable bioelectronic devices for photoelectrochemical and electrochemical modulation of cells and tissues. *Nat. Rev. Bioeng.* **2025**, *3*, 485-504. DOI PubMed PMC
4. Park, D. S.; Fishman, G. I. The cardiac conduction system: development, function and therapeutic targets. *Nat. Rev. Cardiol.* **2026**, *23*, 303-23. DOI PubMed
5. Espinoza, D.; Seibold, F.; Stanley, S. Central and peripheral neural circuits regulating glucose homeostasis. *NPJ. Biomed. Innov.* **2025**, *2*, 34. DOI PubMed PMC
6. Doenyas, C.; Clarke, G.; Cserjési, R. Gut-brain axis and neuropsychiatric health: recent advances. *Sci. Rep.* **2025**, *15*, 3415. DOI PubMed PMC
7. Zhou, W.; Jiang, Y.; Xu, Q.; et al. Soft and stretchable organic bioelectronics for continuous intraoperative neurophysiological monitoring during microsurgery. *Nat. Biomed. Eng.* **2023**, *7*, 1270-81. DOI PubMed
8. Xin, Y.; Sun, B.; Kong, Y.; et al. Advances in integrated power supplies for self-powered bioelectronic devices. *Nanoscale* **2025**, *17*, 2423-37. DOI PubMed
9. Nair, V.; Dalrymple, A. N.; Yu, Z.; et al. Miniature battery-free bioelectronics. *Science* **2023**, *382*, eabn4732. DOI PubMed
10. Lv, S.; Xu, Z.; Mo, F.; et al. Long-term stability strategies of deep brain flexible neural interface. *npj. Flex. Electron.* **2025**, *9*, 410. DOI
11. Lim, K.; Seo, H.; Chung, W. G.; et al. Material and structural considerations for high-performance electrodes for wearable skin devices. *Commun. Mater.* **2024**, *5*, 490. DOI
12. Weyer, H.; Roth, T. A.; Frey, E. Protein pattern morphology and dynamics emerging from effective interfacial tension. *Nat. Phys.* **2026**, *22*, 94-102. DOI
13. Bijonowski, B. M.; Park, J.; Bergert, M.; et al. Intercellular adhesion boots collective cell migration through elevated membrane tension. *Nat. Commun.* **2025**, *16*, 1588. DOI PubMed PMC
14. Osetrova, M.; Tkachev, A.; Mair, W.; et al. Lipidome atlas of the adult human brain. *Nat. Commun.* **2024**, *15*, 4455. DOI PubMed PMC
15. Chen, Y.; Jiang, B.; Hao, H.; et al. Atomic-level regulation of cobalt single-atom nanozymes: engineering high-efficiency catalase mimics. *Angew. Chem. Int. Ed. Engl.* **2023**, *62*, e202301879. DOI PubMed
16. Cho, I. H.; Jang, D. Y.; Kim, H. J. Adaptive conductors for organ-specific soft bioelectronic interfaces. *Biomed. Eng. Lett.* **2026**, *16*, 283-306. DOI PubMed PMC
17. Ghosh, A.; Li, L.; Xu, L.; et al. Gastrointestinal-resident, shape-changing microdevices extend drug release in vivo. *Sci. Adv.* **2020**, *6*, eabb4133. DOI PubMed PMC
18. Chaves, A.; Azadani, J. G.; Alsalman, H.; et al. Bandgap engineering of two-dimensional semiconductor materials. *npj. 2D. Mater. Appl.* **2020**, *4*, 162. DOI
19. Thompson, B. C.; Murray, E.; Wallace, G. G. Graphite oxide to graphene. Biomaterials to bionics. *Adv. Mater.* **2015**, *27*, 7563-82. DOI PubMed
20. Gao, W.; Emaminejad, S.; Nyein, H. Y. Y.; et al. Fully integrated wearable sensor arrays for multiplexed in situ perspiration analysis. *Nature* **2016**, *529*, 509-14. DOI PubMed PMC

21. Li, J.; Wu, K.; Xiao, J.; et al. Flexible multichannel muscle impedance sensors for collaborative human-machine interfaces. *Sci. Adv.* **2025**, *11*, eadv3359. DOI PubMed PMC
22. Mirbakht, S. S.; Golparvar, A.; Umar, M.; Kuzubasoglu, B. A.; Irani, F. S.; Yapici, M. K. Highly self-adhesive and biodegradable silk bioelectronics for all-in-one imperceptible long-term electrophysiological biosignals monitoring. *Adv. Sci.* **2025**, *12*, e2405988. DOI PubMed PMC
23. Liu, J.; Li, Z.; Sun, M.; et al. Flexible bioelectronic systems with large-scale temperature sensor arrays for monitoring and treatments of localized wound inflammation. *Proc. Natl. Acad. Sci. U. S. A.* **2024**, *121*, e2412423121. DOI PubMed PMC
24. O'Neill, S. J. K.; Huang, Z.; Chen, X.; et al. Highly stretchable dynamic hydrogels for soft multilayer electronics. *Sci. Adv.* **2024**, *10*, eadn5142. DOI PubMed PMC
25. Kim, H.; Kim, Y. S.; Mahmood, M.; et al. Fully integrated, stretchable, wireless skin-conformal bioelectronics for continuous stress monitoring in daily life. *Adv. Sci.* **2020**, *7*, 2000810. DOI PubMed PMC
26. Cho, Y. U.; Lim, S. L.; Hong, J.; Yu, K. J. Transparent neural implantable devices: a comprehensive review of challenges and progress. *npj. Flex. Electron.* **2022**, *6*, 178. DOI
27. Wu, S. D.; Hsu, S. H.; Ketelsen, B.; et al. Fabrication of eco-friendly wearable strain sensor arrays via facile contact printing for healthcare applications. *Small. Methods.* **2023**, *7*, e2300170. DOI PubMed
28. Ma, R.; Kwon, S.; Zheng, Q.; et al. Carbon-nanotube/silver networks in nitrile butadiene rubber for highly conductive flexible adhesives. *Adv. Mater.* **2012**, *24*, 3344-9. DOI PubMed
29. Yi, C.; Li, W.; Shi, S.; et al. High-temperature-resistant and colorless polyimide: preparations, properties, and applications. *Solar. Energy.* **2020**, *195*, 340-54. DOI
30. Wang, S.; Chinnasamy, T.; Lifson, M. A.; Inci, F.; Demirci, U. Flexible substrate-based devices for point-of-care diagnostics. *Trends. Biotechnol.* **2016**, *34*, 909-21. DOI PubMed PMC
31. Rihani, R.; Tasnim, N.; Javed, M.; et al. Liquid crystalline polymers: opportunities to shape neural interfaces. *Neuromodulation* **2022**, *25*, 1259-67. DOI PubMed
32. Zhang, T.; Chai, Y.; Wang, S.; et al. Recent study advances in flexible sensors based on polyimides. *Sensors* **2023**, *23*, 9743. DOI PubMed PMC
33. Zhou, E.; Wang, X.; Liang, J.; et al. Chronically stable, high-resolution micro-electrocorticographic brain-computer interfaces for real-time motor decoding. *Adv. Sci.* **2025**, *12*, e06663. DOI PubMed PMC
34. Wang, S.; Jiang, C.; Yu, Y.; et al. Tellurium nanowire retinal nanoprosthesis improves vision in models of blindness. *Science* **2025**, *388*, eadu2987. DOI PubMed
35. Cho, H.; Kim, M. J.; Chang, J.; et al. Advanced textile-based OLEDs utilizing parylene-C planarization for enhanced flexibility and stability in true wearing displays. *npj. Flex. Electron.* **2025**, *9*, 413. DOI
36. Wan, J.; Nie, Z.; Xu, J.; et al. Millimeter-scale magnetic implants paired with a fully integrated wearable device for wireless biophysical and biochemical sensing. *Sci. Adv.* **2024**, *10*, eadm9314. DOI PubMed PMC
37. Chong, H.; Majerus, S. J.; Bogie, K. M.; Zorman, C. A. Non-hermetic packaging of biomedical microsystems from a materials perspective: a review. *Med. Devices. Sens.* **2020**, *3*, e10082. DOI
38. Zeng, Q.; Zhao, S.; Yang, H.; Zhang, Y.; Wu, T. Micro/nano technologies for high-density retinal implant. *Micromachines* **2019**, *10*, 419. DOI PubMed PMC
39. Márton, G.; Orbán, G.; Kiss, M.; Fiáth, R.; Pongrácz, A.; Ulbert, I. A multimodal, SU-8 - platinum - polyimide microelectrode array for chronic in vivo neurophysiology. *PLoS. One.* **2015**, *10*, e0145307. DOI PubMed PMC
40. Richner, T. J.; Thongpang, S.; Brodnick, S. K.; et al. Optogenetic micro-electrocorticography for modulating and localizing cerebral cortex activity. *J. Neural. Eng.* **2014**, *11*, 016010. DOI PubMed PMC
41. Shi, J.; Kim, S.; Li, P.; et al. Active biointegrated living electronics for managing inflammation. *Science* **2024**, *384*, 1023-30. DOI PubMed
42. Wu, Y.; Liu, C.; Lapiere, M.; et al. Thermoplastic elastomers for wireless, skin-interfaced electronic, and microfluidic devices. *Adv. Mater. Technol.* **2023**, *8*, 2300732. DOI
43. Shin, Y.; Lee, H. S.; Hong, Y. J.; et al. Low-impedance tissue-device interface using homogeneously conductive hydrogels chemically bonded to stretchable bioelectronics. *Sci. Adv.* **2024**, *10*, eadi7724. DOI PubMed PMC
44. Liang, C.; Zhu, M.; Chen, Y.; et al. Multiscale interfacial confined locking from nano to macro enables strain insensitivity in epidermal electronic devices. *Adv. Mater.* **2026**, *38*, e06843. DOI PubMed
45. Lee, H.; Song, S.; Yea, J.; et al. Vialess heterogeneous skin patch for multimodal monitoring and stimulation. *Nat. Commun.* **2025**, *16*, 650. DOI PubMed PMC
46. He, J.; Huang, J.; Li, R.; et al. Hysteresis-free and dynamically resilient strain sensor enabled by interfacial coordination. *Sci. Adv.* **2026**, *12*, eaea2450. DOI PubMed PMC

47. Wang, Y.; Wang, Z.; Zhang, Y.; et al. A 2.7-μm-thick robust, permeable, and antifreezing hydrogel electrode for long-term ambulatory health monitoring. *Sci. Adv.* **2025**, *11*, eadt2286. DOI PubMed PMC
48. Su, Y.; Ping, X.; Yu, K. J.; et al. In-plane deformation mechanics for highly stretchable electronics. *Adv. Mater.* **2017**, *29*, 1604989. DOI PubMed
49. Jang, J.; Jin, S.; Yoon, S.; Shin, M.; Son, D. Materials strategy and device fabrication for stable closed-loop bioelectronics. *npj. Biosensing.* **2025**, *2*, 45. DOI
50. Kim, S.; Kang, J.; Lee, I.; et al. An intrinsically stretchable multi-biochemical sensor for sweat analysis using photo-patternable ecoflex. *npj. Flex. Electron.* **2023**, *7*, 268. DOI
51. Kim, D. H.; Lu, N.; Ma, R.; et al. Epidermal electronics. *Science* **2011**, *333*, 838-43. DOI PubMed
52. Qi, D.; Zhang, K.; Tian, G.; Jiang, B.; Huang, Y. Stretchable electronics based on PDMS substrates. *Adv. Mater.* **2021**, *33*, e2003155. DOI PubMed
53. Gao, F.; Liu, C.; Zhang, L.; et al. Wearable and flexible electrochemical sensors for sweat analysis: a review. *Microsyst. Nanoeng.* **2023**, *9*, 1. DOI PubMed PMC
54. Yuan, X.; Li, C.; Yin, X.; et al. Epidermal wearable biosensors for monitoring biomarkers of chronic disease in sweat. *Biosensors* **2023**, *13*, 313. DOI PubMed PMC
55. Erdem, A.; Eksin, E.; Senturk, H.; Yildiz, E.; Maral, M. Recent developments in wearable biosensors for healthcare and biomedical applications. *TrAC. Trends. Anal. Chem.* **2024**, *171*, 117510. DOI
56. Lu, H.; Zhang, Y.; Zhu, M.; et al. Intelligent perceptual textiles based on ionic-conductive and strong silk fibers. *Nat. Commun.* **2024**, *15*, 3289. DOI PubMed PMC
57. Shire, E.; Coimbra, A. A. B.; Barba Ostria, C.; Rios-Solis, L.; López Barreiro, D. Molecular design of protein-based materials - state of the art, opportunities and challenges at the interface between materials engineering and synthetic biology. *Mol. Syst. Des. Eng.* **2024**, *9*, 1187-209. DOI
58. Hwang, S. W.; Tao, H.; Kim, D. H.; et al. A physically transient form of silicon electronics. *Science* **2012**, *337*, 1640-4. DOI PubMed PMC
59. Middleton, J. C.; Tipton, A. J. Synthetic biodegradable polymers as orthopedic devices. *Biomaterials* **2000**, *21*, 2335-46. DOI PubMed
60. Dutta, R.; Chowdhury, S.; Kar, K.; Mazumder, K. Silk fibroin–based biomaterial scaffold in tissue engineering: present persuasive perspective. *Regen. Eng. Transl. Med.* **2025**, *11*, 531-52. DOI
61. Choi, Y. S.; Yin, R. T.; Pfenniger, A.; et al. Fully implantable and bioresorbable cardiac pacemakers without leads or batteries. *Nat. Biotechnol.* **2021**, *39*, 1228-38. DOI PubMed PMC
62. Shuai, Y.; Zheng, M.; Kundu, S. C.; Mao, C.; Yang, M. Bioengineered silk protein-based 3D in vitro models for tissue engineering and drug development: from silk matrix properties to biomedical applications. *Adv. Healthc. Mater.* **2024**, *13*, e2401458. DOI PubMed
63. Wen, D. L.; Sun, D. H.; Huang, P.; et al. Recent progress in silk fibroin-based flexible electronics. *Microsyst. Nanoeng.* **2021**, *7*, 35. DOI PubMed PMC
64. Tian, L.; Shi, J.; Li, W.; Zhang, Y.; Gao, X. Hollow microfiber assembly-based endocrine pancreas-on-a-chip for sugar substitute evaluation. *Adv. Healthc. Mater.* **2024**, *13*, e2302104. DOI PubMed
65. Zou, Y.; Chen, Z.; Jin, B.; et al. A closed-loop bioelectronic patch for intelligent blood pressure management. *Sci. Adv.* **2025**, *11*, eadx6438. DOI PubMed PMC
66. Hwang, S.; Kim, D.; Tao, H.; et al. Materials and fabrication processes for transient and bioresorbable high-performance electronics. *Adv. Funct. Mater.* **2013**, *23*, 4087-93. DOI
67. Jia, X.; Ma, X.; Zhao, L.; et al. A biocompatible and fully erodible conducting polymer enables implanted rechargeable Zn batteries. *Chem. Sci.* **2023**, *14*, 2123-30. DOI PubMed PMC
68. Baumgartner, M.; Hartmann, F.; Drack, M.; et al. Resilient yet entirely degradable gelatin-based biogels for soft robots and electronics. *Nat. Mater.* **2020**, *19*, 1102-9. DOI PubMed
69. Wu, H.; Lyu, H.; Jiang, H.; et al. Bioinspired supramolecular fibrillization enables stretchable and biodegradable piezoelectric bioelectronics. *Sci. Adv.* **2025**, *11*, eadu6759. DOI PubMed PMC
70. Lee, S. H.; Yoo, S.; Kim, S. H.; Kim, Y. M.; Han, S. I.; Lee, H. Nature-inspired surface modification strategies for implantable devices. *Mater. Today. Bio.* **2025**, *31*, 101615. DOI PubMed PMC
71. Kohli, S.; Shahzad, K.; Jouppila, A.; Holthöfer, H.; Isermann, B.; Lassila, R. Thrombosis and inflammation - a dynamic interplay and the role of glycosaminoglycans and activated protein C. *Front. Cardiovasc. Med.* **2022**, *9*, 866751. DOI PubMed PMC
72. Kong, P.; Cui, Z. Y.; Huang, X. F.; Zhang, D. D.; Guo, R. J.; Han, M. Inflammation and atherosclerosis: signaling pathways and therapeutic intervention. *Signal. Transduct. Target. Ther.* **2022**, *7*, 131. DOI PubMed PMC
73. Xu, J.; Duan, C.; Wan, X.; et al. A soft magnetoelastic sensor to decode levels of fatigue. *Nat. Electron.* **2025**, *8*, 709-20. DOI PubMed PMC

74. Jang, T. M.; Han, W. B.; Han, S.; et al. Stretchable and biodegradable self-healing conductors for multifunctional electronics. *Sci. Adv.* **2024**, *10*, eadp9818. DOI PubMed PMC
75. Zhang, Y.; Hu, X.; Yan, Z.; et al. Ultra-soft organic combined film with piezoelectricity induced by liquid-liquid interface polar engineering. *Nat. Commun.* **2025**, *16*, 6410. DOI PubMed PMC
76. Shao, Y.; Yan, J.; Zhi, Y.; et al. A universal packaging substrate for mechanically stable assembly of stretchable electronics. *Nat. Commun.* **2024**, *15*, 6106. DOI PubMed PMC
77. GhavamiNejad, A.; Flynn, C. D.; Geraili, A.; et al. Continuous insulin monitoring using an antibody-protecting zwitterionic microneedle patch. *Nat. Biomed. Eng.* **2026**, *10*, 445-57. DOI PubMed PMC
78. Wong, T. S.; Kang, S. H.; Tang, S. K.; et al. Bioinspired self-repairing slippery surfaces with pressure-stable omniphobicity. *Nature* **2011**, *477*, 443-7. DOI PubMed
79. Li, L.; Jiang, C. Electrodeposited coatings for neural electrodes: a review. *Biosens. Bioelectron.* **2025**, *282*, 117492. DOI PubMed
80. Talasaz, A. H.; Sadeghipour, P.; Ortega-Paz, L.; et al. Optimizing antithrombotic therapy in patients with coexisting cardiovascular and gastrointestinal disease. *Nat. Rev. Cardiol.* **2024**, *21*, 574-92. DOI PubMed PMC
81. Zhao, C.; Park, J.; Root, S. E.; Bao, Z. Skin-inspired soft bioelectronic materials, devices and systems. *Nat. Rev. Bioeng.* **2024**, *2*, 671-90. DOI
82. Kim, H.; Won, Y.; Song, H. W.; Kwon, Y.; Jun, M.; Oh, J. H. Organic mixed ionic-electronic conductors for bioelectronic sensors: materials and operation mechanisms. *Adv. Sci.* **2024**, *11*, e2306191. DOI PubMed PMC
83. Paulsen, B. D.; Tybrandt, K.; Stavrinidou, E.; Rivnay, J. Organic mixed ionic-electronic conductors. *Nat. Mater.* **2020**, *19*, 13-26. DOI PubMed
84. Rashid, R. B.; Ji, X.; Rivnay, J. Organic electrochemical transistors in bioelectronic circuits. *Biosens. Bioelectron.* **2021**, *190*, 113461. DOI PubMed
85. Li, J.; Mo, D.; Hu, J.; et al. PEDOT:PSS-based bioelectronics for brain monitoring and modulation. *Microsyst. Nanoeng.* **2025**, *11*, 87. DOI PubMed PMC
86. Tang, H.; Li, Y.; Liao, S.; Liu, H.; Qiao, Y.; Zhou, J. Multifunctional conductive hydrogel interface for bioelectronic recording and stimulation. *Adv. Healthc. Mater.* **2024**, *13*, e2400562. DOI PubMed
87. Park, Y. G.; Lee, G. Y.; Jang, J.; Yun, S. M.; Kim, E.; Park, J. U. Liquid metal-based soft electronics for wearable healthcare. *Adv. Healthc. Mater.* **2021**, *10*, e2002280. DOI PubMed
88. Dickey, M. D. Stretchable and soft electronics using liquid metals. *Adv. Mater.* **2017**, *29*, 1606425. DOI PubMed
89. Luo, Z. D.; Yang, M. M.; Liu, Y.; Alexe, M. Emerging opportunities for 2D semiconductor/ferroelectric transistor-structure devices. *Adv. Mater.* **2021**, *33*, e2005620. DOI PubMed
90. Driscoll, N.; Erickson, B.; Murphy, B. B.; et al. MXene-infused bioelectronic interfaces for multiscale electrophysiology and stimulation. *Sci. Transl. Med.* **2021**, *13*, eabf8629. DOI PubMed PMC
91. Jiang, H.; Taranekar, P.; Reynolds, J. R.; Schanze, K. S. Conjugated polyelectrolytes: synthesis, photophysics, and applications. *Angew. Chem. Int. Ed. Engl.* **2009**, *48*, 4300-16. DOI PubMed
92. Wu, R.; Matta, M.; Paulsen, B. D.; Rivnay, J. Operando characterization of organic mixed ionic/electronic conducting materials. *Chem. Rev.* **2022**, *122*, 4493-551. DOI PubMed
93. Maier, J. Nanoionics: ion transport and electrochemical storage in confined systems. *Nat. Mater.* **2005**, *4*, 805-15. DOI PubMed
94. Lin, Y.; Fang, S.; Su, D.; Brinkman, K. S.; Chen, F. Enhancing grain boundary ionic conductivity in mixed ionic-electronic conductors. *Nat. Commun.* **2015**, *6*, 6824. DOI PubMed PMC
95. Noriega, R.; Rivnay, J.; Vandewal, K.; et al. A general relationship between disorder, aggregation and charge transport in conjugated polymers. *Nat. Mater.* **2013**, *12*, 1038-44. DOI PubMed
96. Lee, J.; Han, A. R.; Yu, H.; Shin, T. J.; Yang, C.; Oh, J. H. Boosting the ambipolar performance of solution-processable polymer semiconductors via hybrid side-chain engineering. *J. Am. Chem. Soc.* **2013**, *135*, 9540-7. DOI PubMed
97. Rebetez, G.; Bardagot, O.; Affolter, J.; Réhault, J.; Banerji, N. What drives the kinetics and doping level in the electrochemical reactions of PEDOT:PSS? *Adv. Funct. Mater.* **2022**, *32*, 2105821. DOI
98. Ohayon, D.; Druet, V.; Inal, S. A guide for the characterization of organic electrochemical transistors and channel materials. *Chem. Soc. Rev.* **2023**, *52*, 1001-23. DOI PubMed
99. Wu, R.; Ji, X.; Ma, Q.; Paulsen, B. D.; Tropp, J.; Rivnay, J. Direct quantification of ion composition and mobility in organic mixed ionic-electronic conductors. *Sci. Adv.* **2024**, *10*, eadn8628. DOI PubMed PMC
100. Keene, S. T.; Laulainen, J. E. M.; Pandya, R.; et al. Hole-limited electrochemical doping in conjugated polymers. *Nat. Mater.* **2023**, *22*, 1121-7. DOI PubMed PMC
101. Montazerian, H.; Davoodi, E.; Wang, C.; et al. Boosting hydrogel conductivity via water-dispersible conducting polymers for injectable bioelectronics. *Nat. Commun.* **2025**, *16*, 3755. DOI PubMed PMC

102. Choi, Y.; Oh, S.; Qian, C.; Park, J. H.; Cho, J. H. Vertical organic synapse expandable to 3D crossbar array. *Nat. Commun.* **2020**, *11*, 4595. DOI PubMed PMC
103. Zeglio, E.; Inganäs, O. Active materials for organic electrochemical transistors. *Adv. Mater.* **2018**, *30*, e1800941. DOI PubMed
104. Ohm, Y.; Pan, C.; Ford, M. J.; Huang, X.; Liao, J.; Majidi, C. An electrically conductive silver–polyacrylamide–alginate hydrogel composite for soft electronics. *Nat. Electron.* **2021**, *4*, 185-92. DOI
105. Zhang, Y.; Tan, Y.; Lao, J.; Gao, H.; Yu, J. Hydrogels for flexible electronics. *ACS. Nano.* **2023**, *17*, 9681-93. DOI PubMed
106. Liu, Y.; He, K.; Chen, G.; Leow, W. R.; Chen, X. Nature-inspired structural materials for flexible electronic devices. *Chem. Rev.* **2017**, *117*, 12893-941. DOI PubMed
107. Yang, S.; Cheng, J.; Shang, J.; et al. Stretchable surface electromyography electrode array patch for tendon location and muscle injury prevention. *Nat. Commun.* **2023**, *14*, 6494. DOI PubMed PMC
108. Wang, Z.; Wei, H.; Huang, Y.; Wei, Y.; Chen, J. Naturally sourced hydrogels: emerging fundamental materials for next-generation healthcare sensing. *Chem. Soc. Rev.* **2023**, *52*, 2992-3034. DOI PubMed
109. Shao, Y.; Shigenobu, K.; Watanabe, M.; Zhang, C. Role of viscosity in deviations from the Nernst-Einstein relation. *J. Phys. Chem. B.* **2020**, *124*, 4774-80. DOI PubMed PMC
110. Xu, C.; Yang, Y.; Gao, W. Skin-interfaced sensors in digital medicine: from materials to applications. *Matter* **2020**, *2*, 1414-45. DOI PubMed PMC
111. Yuk, H.; Lu, B.; Zhao, X. Hydrogel bioelectronics. *Chem. Soc. Rev.* **2019**, *48*, 1642-67. DOI PubMed
112. Zhang, Y. Z.; Lee, K. H.; Anjum, D. H.; et al. MXenes stretch hydrogel sensor performance to new limits. *Sci. Adv.* **2018**, *4*, eaat0098. DOI PubMed PMC
113. Nezakati, T.; Seifalian, A.; Tan, A.; Seifalian, A. M. Conductive polymers: opportunities and challenges in biomedical applications. *Chem. Rev.* **2018**, *118*, 6766-843. DOI PubMed
114. Li, G.; Huang, K.; Deng, J.; et al. Highly conducting and stretchable double-network hydrogel for soft bioelectronics. *Adv. Mater.* **2022**, *34*, e2200261. DOI PubMed
115. Xie, Y.; Martini, N.; Hassler, C.; et al. In vivo monitoring of glial scar proliferation on chronically implanted neural electrodes by fiber optical coherence tomography. *Front. Neuroeng.* **2014**, *7*, 34. DOI PubMed PMC
116. Amirthalingam, S.; Rajendran, A. K.; Moon, Y. G.; Hwang, N. S. Stimuli-responsive dynamic hydrogels: design, properties and tissue engineering applications. *Mater. Horiz.* **2023**, *10*, 3325-50. DOI PubMed
117. Xue, B.; Bashir, Z.; Guo, Y.; et al. Strong, tough, rapid-recovery, and fatigue-resistant hydrogels made of picot peptide fibres. *Nat. Commun.* **2023**, *14*, 38280. DOI PubMed PMC
118. Li, H.; Zhang, H.; Peng, Y.; Liu, X.; Du, J.; Liao, J. Rapid synthesis of functions-integrated hydrogel as a self-powered wound dressing for real-time drug release and health monitoring. *Adv. Healthc. Mater.* **2024**, *13*, e2401704. DOI PubMed
119. Zhang, X.; Li, D.; Yang, X.; et al. Hydro-locking in hydrogel for extreme temperature tolerance. *Science* **2025**, *387*, 967-73. DOI PubMed
120. Xin, J.; Gao, L.; Zhang, W.; et al. A thermogalvanic cell dressing for smart wound monitoring and accelerated healing. *Nat. Biomed. Eng.* **2026**, *10*, 80-93. DOI PubMed
121. Zhou, T.; Yuk, H.; Hu, F.; et al. 3D printable high-performance conducting polymer hydrogel for all-hydrogel bioelectronic interfaces. *Nat. Mater.* **2023**, *22*, 895-902. DOI PubMed
122. Yuk, H.; Zhang, T.; Lin, S.; Parada, G. A.; Zhao, X. Tough bonding of hydrogels to diverse non-porous surfaces. *Nat. Mater.* **2016**, *15*, 190-6. DOI PubMed PMC
123. Xu, H.; Lu, J.; Xi, Y.; Wang, X.; Liu, J. Liquid metal biomaterials: translational medicines, challenges and perspectives. *Natl. Sci. Rev.* **2024**, *11*, nwad302. DOI PubMed PMC
124. Hu, L.; Wang, H.; Wang, X.; Liu, X.; Guo, J.; Liu, J. Magnetic liquid metals manipulated in the three-dimensional free space. *ACS. Appl. Mater. Interfaces.* **2019**, *11*, 8685-92. DOI PubMed
125. Truong, V. K.; Hayles, A.; Bright, R.; et al. Gallium liquid metal: nanotoolbox for antimicrobial applications. *ACS. Nano.* **2023**, *17*, 14406-23. DOI PubMed
126. Tang, L.; Cheng, S.; Zhang, L.; et al. Printable metal-polymer conductors for highly stretchable bio-devices. *iScience* **2018**, *4*, 302-11. DOI PubMed PMC
127. Jiang, C.; Li, W.; Wu, Q.; et al. Shape-adaptive electronics based on liquid metal circuits printed on thermoplastic films. *Nat. Electron.* **2026**, *9*, 45-58. DOI
128. Zeng, H.; Xu, Q.; Zhang, J.; et al. Topology-optimized stretchable piezoelectric sensors with tailored liquid-metal circuits for anisotropic stress-adaptive motion monitoring. *Adv. Mater.* **2026**, *38*, e18168. DOI PubMed PMC
129. Jaseem, S. A.; Rahmani, P.; Sakorikar, T.; et al. Liquid metals as initiators of free-radical polymerization of hydrogels: a perspective. *Adv. Funct. Mater.* **2026**, *36*, e14024. DOI

130. Kim, E.; Jeong, E.; Hong, Y. M.; et al. Magnetically reshapable 3D multi-electrode arrays of liquid metals for electrophysiological analysis of brain organoids. *Nat. Commun.* **2025**, *16*, 2011. DOI PubMed PMC
131. Dong, R.; Liu, X.; Cheng, S.; et al. Highly stretchable metal-polymer conductor electrode array for electrophysiology. *Adv. Healthc. Mater.* **2021**, *10*, e2000641. DOI PubMed
132. Dong, R.; Wang, L.; Hang, C.; et al. Printed stretchable liquid metal electrode arrays for in vivo neural recording. *Small* **2021**, *17*, e2006612. DOI PubMed
133. Kim, J. H.; Kim, S.; Dickey, M. D.; So, J. H.; Koo, H. J. Interface of gallium-based liquid metals: oxide skin, wetting, and applications. *Nanoscale. Horiz.* **2024**, *9*, 1099-119. DOI PubMed
134. Deng, Y.; Bu, F.; Wang, Y.; Chee, P. S.; Liu, X.; Guan, C. Stretchable liquid metal based biomedical devices. *npj. Flex. Electron.* **2024**, *8*, 298. DOI
135. Lin, Z.; Qiu, X.; Cai, Z.; et al. High internal phase emulsions gel ink for direct-ink-writing 3D printing of liquid metal. *Nat. Commun.* **2024**, *15*, 4806. DOI PubMed PMC
136. Singh, M.; Bhuyan, P.; Jeong, S.; Park, S. Directly printable, non-smearable and stretchable conductive ink enabled by liquid metal microparticles interstitially engineered in highly entangled elastomeric matrix. *Adv. Funct. Mater.* **2025**, *35*, 2412178. DOI
137. Hang, C.; Ding, L.; Cheng, S.; et al. A soft and absorbable temporary epicardial pacing wire. *Adv. Mater.* **2021**, *33*, e2101447. DOI PubMed
138. Peng, Y.; Song, J.; Zhang, Y.; et al. Permeable, wet-adhesive, and EMI-resistant liquid metal electronic skin for high-fidelity electrophysiological monitoring in sweaty and electromagnetic environments. *Adv. Mater.* **2025**, *37*, e08041. DOI PubMed
139. Liu, X.; Dong, R.; Hang, C.; Lim, C. T.; Jiang, X. Brain extracellular matrix-based electronic brain biochip. *ACS. Nano.* **2026**, *20*, 9482-94. DOI PubMed
140. Boateng, D.; Li, X.; Zhu, Y.; et al. Recent advances in flexible hydrogel sensors: enhancing data processing and machine learning for intelligent perception. *Biosens. Bioelectron.* **2024**, *261*, 116499. DOI PubMed
141. Hajalilou, A. Liquid metal-polymer hydrogel composites for sustainable electronics: a review. *Molecules* **2025**, *30*, 905. DOI PubMed PMC
142. Geim, A. K.; Novoselov, K. S. The rise of graphene. *Nat. Mater.* **2007**, *6*, 183-91. DOI PubMed
143. Molle, A.; Goldberger, J.; Houssa, M.; Xu, Y.; Zhang, S. C.; Akinwande, D. Buckled two-dimensional Xene sheets. *Nat. Mater.* **2017**, *16*, 163-9. DOI PubMed
144. Novoselov, K. S.; Geim, A. K.; Morozov, S. V.; et al. Electric field effect in atomically thin carbon films. *Science* **2004**, *306*, 666-9. DOI PubMed
145. Tao, W.; Kong, N.; Ji, X.; et al. Emerging two-dimensional monoelemental materials (Xenes) for biomedical applications. *Chem. Soc. Rev.* **2019**, *48*, 2891-912. DOI PubMed
146. Qi, B.; Shen, R.; Ke, Z.; et al. 2D sp^2 carbon-conjugated covalent organic frameworks: photocatalytic platforms for solar energy conversion. *Rare. Met.* **2025**, *44*, 9543-87. DOI
147. You, X.; Bao, R.; Zhang, L.; et al. A combined experimental and computational study of the Cu/C (sp^2) interface. *Carbon. Trends.* **2021**, *4*, 100046. DOI
148. Li, S.; Ma, L.; Zhou, M.; et al. New opportunities for emerging 2D materials in bioelectronics and biosensors. *Curr. Opin. Biomed. Eng.* **2020**, *13*, 32-41. DOI
149. Zhang, Q.; Wei, Q.; Huang, K.; et al. Defects boost graphitization for highly conductive graphene films. *Natl. Sci. Rev.* **2023**, *10*, nwad147. DOI PubMed PMC
150. Ding, M.; Xie, P.; Wang, J.; et al. Biomimetic microstructure design for ultrasensitive piezoionic mechanoreceptors in multimodal object recognition. *Nat. Commun.* **2025**, *16*, 8129. DOI PubMed PMC
151. Dong, H.; Qiao, Y.; Peng, S.; et al. 2D material/epoxy composite coatings, a perspective from the regulation of 2D materials. *Prog. Org. Coat.* **2023**, *183*, 107817. DOI
152. Ritt, C. L.; Quien, M.; Wei, Z.; et al. A molecularly impermeable polymer from two-dimensional polyaramids. *Nature* **2025**, *647*, 383-9. DOI PubMed PMC
153. Dastgeer, G.; Zulfiqar, M. W.; Nisar, S.; et al. Emerging role of 2D materials in photovoltaics: efficiency enhancement and future perspectives. *Nanomicro. Lett.* **2025**, *18*, 32. DOI PubMed PMC
154. Yin, R.; Xu, Z.; Mei, M.; et al. Soft transparent graphene contact lens electrodes for conformal full-cornea recording of electroretinogram. *Nat. Commun.* **2018**, *9*, 2334. DOI PubMed PMC
155. Lim, J.; Lee, S.; Kim, J.; et al. Hybrid graphene electrode for the diagnosis and treatment of epilepsy in free-moving animal models. *NPG. Asia. Mater.* **2023**, *15*, 464. DOI

156. Alex, M.; Khan, K. R. B.; Al-Othman, A.; Al-Sayah, M. H.; Al Nashash, H. MXene-based flexible electrodes for electrophysiological monitoring. *Sensors* **2024**, *24*, 3260. DOI PubMed PMC
157. Duvan, F. T.; Cunquero, M.; Masvidal-Codina, E.; et al. Graphene-based microelectrodes with bidirectional functionality for next-generation retinal electronic interfaces. *Nanoscale. Horiz.* **2024**, *9*, 1948-61. DOI PubMed
158. Li, Y.; Li, N.; De Oliveira, N.; Wang, S. Implantable bioelectronics toward long-term stability and sustainability. *Matter* **2021**, *4*, 1125-41. DOI
159. Robinson, K. J.; Voelcker, N. H.; Thissen, H. Clinical challenges and opportunities related to the biological responses experienced by indwelling and implantable bioelectronic medical devices. *Acta. Biomater.* **2025**, *193*, 49-64. DOI PubMed
160. Carnicer-Lombarte, A.; Chen, S. T.; Malliaras, G. G.; Barone, D. G. Foreign body reaction to implanted biomaterials and its impact in nerve neuroprosthetics. *Front. Bioeng. Biotechnol.* **2021**, *9*, 622524. DOI PubMed PMC
161. Fallegger, F.; Schiavone, G.; Lacour, S. P. Conformable hybrid systems for implantable bioelectronic interfaces. *Adv. Mater.* **2020**, *32*, e1903904. DOI PubMed
162. Green, R.; Abidian, M. R. Conducting polymers for neural prosthetic and neural interface applications. *Adv. Mater.* **2015**, *27*, 7620-37. DOI PubMed PMC
163. Berggren, M.; Głowacki, E. D.; Simon, D. T.; Stavrinidou, E.; Tybrandt, K. In vivo organic bioelectronics for neuromodulation. *Chem. Rev.* **2022**, *122*, 4826-46. DOI PubMed PMC
164. Uguz, I.; Shepard, K. L. Spatially controlled, bipolar, cortical stimulation with high-capacitance, mechanically flexible subdural surface microelectrode arrays. *Sci. Adv.* **2022**, *8*, eabq6354. DOI PubMed PMC
165. Koklu, A.; Ohayon, D.; Wustoni, S.; Druet, V.; Saleh, A.; Inal, S. Organic bioelectronic devices for metabolite sensing. *Chem. Rev.* **2022**, *122*, 4581-635. DOI PubMed
166. Sarac, B.; Yücer, S.; Ciftci, F. MOF-based bioelectronic supercapacitors. *Small* **2025**, *21*, e2412846. DOI PubMed PMC
167. Liu, Y.; Pharr, M.; Salvatore, G. A. Lab-on-skin: a review of flexible and stretchable electronics for wearable health monitoring. *ACS. Nano.* **2017**, *11*, 9614-35. DOI PubMed
168. Linh, V. T. N.; Han, S.; Koh, E.; Kim, S.; Jung, H. S.; Koo, J. Advances in wearable electronics for monitoring human organs: bridging external and internal health assessments. *Biomaterials* **2025**, *314*, 122865. DOI PubMed
169. Wang, Y.; Feng, X.; Chen, X. Autonomous bioelectronic devices based on silk fibroin. *Adv. Mater.* **2025**, *37*, e2500073. DOI PubMed
170. Mariello, M.; Eş, I.; Proctor, C. M. Soft and flexible bioelectronic micro-systems for electronically controlled drug delivery. *Adv. Healthc. Mater.* **2024**, *13*, e2302969. DOI PubMed
171. Wang, S.; Chen, X.; Zhao, C.; et al. An organic electrochemical transistor for multi-modal sensing, memory and processing. *Nat. Electron.* **2023**, *6*, 281-91. DOI
172. Lee, C.; Hu, S.; Christy, J.; Chou, F.; Ramli, T. C.; Chen, H. Biointerface coatings with structural and biochemical properties modifications of biomaterials. *Adv. Mater. Interfaces.* **2023**, *10*, 2202286. DOI
173. Zhao, Y.; Zhang, S.; Yu, T.; et al. Ultra-conformal skin electrodes with synergistically enhanced conductivity for long-time and low-motion artifact epidermal electrophysiology. *Nat. Commun.* **2021**, *12*, 4880. DOI PubMed PMC
174. Moonen, P. F.; Yakimets, I.; Huskens, J. Fabrication of transistors on flexible substrates: from mass-printing to high-resolution alternative lithography strategies. *Adv. Mater.* **2012**, *24*, 5526-41. DOI PubMed
175. Yin, X.; Yang, J.; Wang, H. Vertical phase separation structure for high-performance organic thin-film transistors: mechanism, optimization strategy, and large-area fabrication toward flexible and stretchable electronics. *Adv. Funct. Mater.* **2022**, *32*, 2202071. DOI
176. Rich, S. I.; Jiang, Z.; Fukuda, K.; Someya, T. Well-rounded devices: the fabrication of electronics on curved surfaces - a review. *Mater. Horiz.* **2021**, *8*, 1926-58. DOI PubMed
177. Kim, J. J.; Bae, M.; Cho, D. W. Multi-organ microphysiological systems targeting specific organs for recapitulating disease phenotypes via organ crosstalk. *Small. Sci.* **2024**, *4*, 2400314. DOI PubMed PMC
178. Jiang, Y.; Zhang, Z.; Wang, Y. X.; et al. Topological supramolecular network enabled high-conductivity, stretchable organic bioelectronics. *Science* **2022**, *375*, 1411-7. DOI PubMed
179. Zhang, Y.; Lee, G.; Li, S.; Hu, Z.; Zhao, K.; Rogers, J. A. Advances in bioresorbable materials and electronics. *Chem. Rev.* **2023**, *123*, 11722-73. DOI PubMed
180. Wu, E.; Tsarev, S.; Proniakova, D.; et al. A CMOS-compatible fabrication approach for high-performance perovskite photodetector arrays. *Adv. Opt. Mater.* **2025**, *13*, 2402979. DOI
181. Zhu, C.; Ekinci, H.; Pan, A.; Cui, B.; Zhu, X. Electron beam lithography on nonplanar and irregular surfaces. *Microsyst. Nanoeng.* **2024**, *10*, 52. DOI PubMed PMC
182. Kim, D. W.; Kong, M.; Jeong, U. Interface design for stretchable electronic devices. *Adv. Sci.* **2021**, *8*, 2004170. DOI PubMed PMC
183. Rogers, J. A.; Someya, T.; Huang, Y. Materials and mechanics for stretchable electronics. *Science* **2010**, *327*, 1603-7. DOI PubMed

184. del Campo, A.; Arzt, E. Fabrication approaches for generating complex micro- and nanopatterns on polymeric surfaces. *Chem. Rev.* **2008**, *108*, 911-45. DOI PubMed
185. Ji, B.; Guo, Z.; Wang, M.; et al. Flexible polyimide-based hybrid opto-electric neural interface with 16 channels of micro-LEDs and electrodes. *Microsyst. Nanoeng.* **2018**, *4*, 27. DOI PubMed PMC
186. Altuna, A.; Menendez de la Prida, L.; Bellistri, E.; et al. SU-8 based microprobes with integrated planar electrodes for enhanced neural depth recording. *Biosens. Bioelectron.* **2012**, *37*, 1-5. DOI PubMed
187. Xu, C.; Lemon, W.; Liu, C. Design and fabrication of a high-density metal microelectrode array for neural recording. *Sens. Actuators. A. Phys.* **2002**, *96*, 78-85. DOI
188. Ma, Y.; Wang, S.; Wu, Z. S. Photolithographic microfabrication of microbatteries for on-chip energy storage. *Nanomicro. Lett.* **2025**, *17*, 105. DOI PubMed PMC
189. Sifringer, L.; De Windt, L.; Bernhard, S.; et al. Photopatterning of conductive hydrogels which exhibit tissue-like properties. *J. Mater. Chem. B.* **2024**, *12*, 10272-84. DOI
190. Song, O.; Rhee, D.; Kim, J.; et al. All inkjet-printed electronics based on electrochemically exfoliated two-dimensional metal, semiconductor, and dielectric. *npj. 2D. Mater. Appl.* **2022**, *6*, 337. DOI
191. Lin, C.; Wang, Q.; Liu, H.; et al. Direct laser writing of bioinspired high-entropy oxide nanoarrays for practical water electrolysis. *Adv. Energy. Mater.* **2025**, *15*, e03929. DOI
192. You, R.; Liu, Y. Q.; Hao, Y. L.; Han, D. D.; Zhang, Y. L.; You, Z. Laser fabrication of graphene-based flexible electronics. *Adv. Mater.* **2020**, *32*, e1901981. DOI PubMed
193. Wang, Y.; Yokota, T.; Someya, T. Electrospun nanofiber-based soft electronics. *NPG. Asia. Mater.* **2021**, *13*, 267. DOI
194. Cho, Y.; Beak, J. W.; Sagong, M.; Ahn, S.; Nam, J. S.; Kim, I. D. Electrospinning and nanofiber technology: fundamentals, innovations, and applications. *Adv. Mater.* **2025**, *37*, e2500162. DOI PubMed PMC
195. Wei, X.; Wang, L.; Duan, C.; et al. Cardiac patches made of brown adipose-derived stem cell sheets and conductive electrospun nanofibers restore infarcted heart for ischemic myocardial infarction. *Bioact. Mater.* **2023**, *27*, 271-87. DOI PubMed PMC
196. Xu, R.; He, P.; Lan, G.; et al. Facile fabrication of multilayer stretchable electronics via a two-mode mechanical cutting process. *ACS. Nano.* **2022**, *16*, 1533-46. DOI PubMed
197. Wei, R.; Li, H.; Chen, Z.; Hua, Q.; Shen, G.; Jiang, K. Revolutionizing wearable technology: advanced fabrication techniques for body-conformable electronics. *npj. Flex. Electron.* **2024**, *8*, 370. DOI
198. Yang, S.; Chen, Y. C.; Nicolini, L.; et al. "Cut-and-Paste" manufacture of multiparametric epidermal sensor systems. *Adv. Mater.* **2015**, *27*, 6423-30. DOI PubMed
199. Brooks, A. K.; Chakravarty, S.; Ali, M.; Yadavalli, V. K. Kirigami-inspired biodesign for applications in healthcare. *Adv. Mater.* **2022**, *34*, e2109550. DOI PubMed
200. Zou, G. F.; Zhao, J.; Luo, H. M.; McCleskey, T. M.; Burrell, A. K.; Jia, Q. X. Polymer-assisted-deposition: a chemical solution route for a wide range of materials. *Chem. Soc. Rev.* **2013**, *42*, 439-49. DOI PubMed
201. Xu, L.; Gutbrod, S. R.; Bonifas, A. P.; et al. 3D multifunctional integumentary membranes for spatiotemporal cardiac measurements and stimulation across the entire epicardium. *Nat. Commun.* **2014**, *5*, 3329. DOI PubMed PMC
202. Kim, D. H.; Ghaffari, R.; Lu, N.; et al. Electronic sensor and actuator webs for large-area complex geometry cardiac mapping and therapy. *Proc. Natl. Acad. Sci. U. S. A.* **2012**, *109*, 19910-5. DOI PubMed PMC
203. Song, E.; Li, J.; Won, S. M.; Bai, W.; Rogers, J. A. Materials for flexible bioelectronic systems as chronic neural interfaces. *Nat. Mater.* **2020**, *19*, 590-603. DOI PubMed
204. Wang, P.; Wu, E. G.; Uluşan, H.; et al. Direct-print 3D electrodes for large-scale, high-density, and customizable neural interfaces. *Adv. Sci.* **2025**, *12*, e2408602. DOI PubMed PMC
205. Zhu, Z.; Guo, S. Z.; Hirdler, T.; et al. 3D printed functional and biological materials on moving freeform surfaces. *Adv. Mater.* **2018**, *30*, e1707495. DOI PubMed PMC
206. Davoodi, E.; Li, J.; Ma, X.; et al. Imaging-guided deep tissue in vivo sound printing. *Science* **2025**, *388*, 616-23. DOI PubMed PMC
207. Valentine, A. D.; Busbee, T. A.; Boley, J. W.; et al. Hybrid 3D printing of soft electronics. *Adv. Mater.* **2017**, *29*, 1703817. DOI
208. Li, H.; Liu, H.; Sun, M.; Huang, Y.; Xu, L. 3D interfacing between soft electronic tools and complex biological tissues. *Adv. Mater.* **2021**, *33*, e2004425. DOI PubMed
209. Say, M. G.; Brooke, R.; Edberg, J.; et al. Spray-coated paper supercapacitors. *npj. Flex. Electron.* **2020**, *4*, 79. DOI
210. Govind, R. K.; Mondal, I.; Baishya, K.; et al. Large-area fabrication of high performing, flexible, transparent conducting electrodes using screen printing and spray coating techniques. *Adv. Mater. Technol.* **2022**, *7*, 2101120. DOI
211. Xu, Y.; Sun, B.; Ling, Y.; et al. Multiscale porous elastomer substrates for multifunctional on-skin electronics with passive-cooling capabilities. *Proc. Natl. Acad. Sci. U. S. A.* **2020**, *117*, 205-13. DOI PubMed PMC

212. Jeong, J. W.; Shin, G.; Park, S. I.; Yu, K. J.; Xu, L.; Rogers, J. A. Soft materials in neuroengineering for hard problems in neuroscience. *Neuron* **2015**, *86*, 175-86. DOI PubMed
213. Dong, Z.; He, Q.; Shen, D.; et al. Microfabrication of functional polyimide films and microstructures for flexible MEMS applications. *Microsyst. Nanoeng.* **2023**, *9*, 31. DOI PubMed PMC
214. Corzo, D.; Tostado-Blázquez, G.; Baran, D. Flexible electronics: status, challenges and opportunities. *Front. Electron.* **2020**, *1*, 594003. DOI
215. Choi, J.; Han, C.; Cho, S.; et al. Customizable, conformal, and stretchable 3D electronics via predistorted pattern generation and thermoforming. *Sci. Adv.* **2021**, *7*, eabj0694. DOI PubMed PMC
216. Sim, K.; Chen, S.; Li, Z.; et al. Three-dimensional curvy electronics created using conformal additive stamp printing. *Nat. Electron.* **2019**, *2*, 471-9. DOI
217. Rao, Z.; Lu, Y.; Li, Z.; et al. Curvy, shape-adaptive imagers based on printed optoelectronic pixels with a kirigami design. *Nat. Electron.* **2021**, *4*, 513-21. DOI
218. Bo, R.; Xu, S.; Yang, Y.; Zhang, Y. Mechanically-guided 3D assembly for architected flexible electronics. *Chem. Rev.* **2023**, *123*, 11137-89. DOI PubMed PMC
219. Minev, I. R.; Musienko, P.; Hirsch, A.; et al. Biomaterials. Electronic dura mater for long-term multimodal neural interfaces. *Science* **2015**, *347*, 159-63. DOI PubMed
220. Ganji, M.; Kaestner, E.; Hermiz, J.; et al. Development and translation of PEDOT:PSS microelectrodes for intraoperative monitoring. *Adv. Funct. Mater.* **2018**, *28*, 1700232. DOI
221. Cox-Pridmore, D. M.; Castro, F. A.; Silva, S. R. P.; Camelliti, P.; Zhao, Y. Emerging bioelectronic strategies for cardiovascular tissue engineering and implantation. *Small* **2022**, *18*, e2105281. DOI PubMed
222. Yu, C.; Shi, M.; He, S.; et al. Chronological adhesive cardiac patch for synchronous mechanophysiological monitoring and electrocoupling therapy. *Nat. Commun.* **2023**, *14*, 6226. DOI PubMed PMC
223. Sim, K.; Ershad, F.; Zhang, Y.; et al. An epicardial bioelectronic patch made from soft rubbery materials and capable of spatiotemporal mapping of electrophysiological activity. *Nat. Electron.* **2020**, *3*, 775-84. DOI
224. You, S. S.; Gierlach, A.; Schmidt, P.; et al. An ingestible device for gastric electrophysiology. *Nat. Electron.* **2024**, *7*, 497-508. DOI
225. Nan, K.; Wong, K.; Li, D.; et al. An ingestible, battery-free, tissue-adhering robotic interface for non-invasive and chronic electrostimulation of the gut. *Nat. Commun.* **2024**, *15*, 6749. DOI PubMed PMC
226. Srinivasan, S.; Antonini, M. J.; Alshareef, A.; et al. Gastrointestinal neuroprosthesis for motility and metabolic neuromodulation. *Nat. Commun.* **2025**, *16*, 7374. DOI PubMed PMC
227. Wei, B.; Wang, Z.; Guo, H.; et al. Ultraflexible tattoo electrodes for epidermal and in vivo electrophysiological recording. *Cell. Rep. Phys. Sci.* **2023**, *4*, 101335. DOI
228. Zhou, T.; Hong, G.; Fu, T. M.; et al. Syringe-injectable mesh electronics integrate seamlessly with minimal chronic immune response in the brain. *Proc. Natl. Acad. Sci. U. S. A.* **2017**, *114*, 5894-9. DOI PubMed PMC
229. Khodagholy, D.; Doublet, T.; Quilichini, P.; et al. In vivo recordings of brain activity using organic transistors. *Nat. Commun.* **2013**, *4*, 1575. DOI PubMed PMC
230. Khodagholy, D.; Rivnay, J.; Sessolo, M.; et al. High transconductance organic electrochemical transistors. *Nat. Commun.* **2013**, *4*, 2133. DOI PubMed PMC
231. Shen, Q.; Jiang, M.; Wang, R.; et al. Liquid metal-based soft, hermetic, and wireless-communicable seals for stretchable systems. *Science* **2023**, *379*, 488-93. DOI PubMed
232. Kang, S. K.; Murphy, R. K.; Hwang, S. W.; et al. Bioresorbable silicon electronic sensors for the brain. *Nature* **2016**, *530*, 71-6. DOI PubMed
233. Sheng, H.; Liu, R.; Li, Q.; et al. Brain implantation of soft bioelectronics via embryonic development. *Nature* **2025**, *642*, 954-64. DOI PubMed PMC
234. Liang, J.; Wang, X.; Chen, Z.; et al. Silk-enabled conformal intraventricular interfaces for minimally invasive neural recordings. *Nat. Commun.* **2025**, *16*, 9366. DOI PubMed PMC
235. Yadav, S.; Lee, R. X.; Kajale, S. N.; et al. A nonsurgical brain implant enabled through a cell-electronics hybrid for focal neuromodulation. *Nat. Biotechnol.* **2025**. DOI PubMed
236. Kim, J. Y.; Kim, H.; Kim, M. H.; et al. Magnetically guided flexible bioelectronic probe for single-cell recordings in multi-scale biosystems. *Adv. Mater.* **2026**, *38*, e11700. DOI PubMed
237. Tang, C.; Han, Z.; Liu, Z.; et al. A soft-fiber bioelectronic device with axon-like architecture enables reliable neural recording in vivo under vigorous activities. *Adv. Mater.* **2024**, *36*, e2407874. DOI PubMed
238. Sahasrabudhe, A.; Rupprecht, L. E.; Orguc, S.; et al. Multifunctional microelectronic fibers enable wireless modulation of gut and brain neural circuits. *Nat. Biotechnol.* **2024**, *42*, 892-904. DOI PubMed PMC

239. Mau, M. M.; Sarker, S.; Terry, B. S. Ingestible devices for long-term gastrointestinal residency: a review. *Prog. Biomed. Eng.* **2021**, *3*, 042001. DOI
240. Liu, J.; Pang, Y.; Zhang, S.; et al. Triggerable tough hydrogels for gastric resident dosage forms. *Nat. Commun.* **2017**, *8*, 124. DOI PubMed PMC
241. Verma, M.; Vishwanath, K.; Eweje, F.; et al. A gastric resident drug delivery system for prolonged gram-level dosing of tuberculosis treatment. *Sci. Transl. Med.* **2019**, *11*, eaau6267. DOI PubMed PMC
242. Kong, Y. L.; Zou, X.; McCandler, C. A.; et al. 3D-printed gastric resident electronics. *Adv. Mater. Technol.* **2019**, *4*, 1800490. DOI PubMed PMC
243. Mimee, M.; Nadeau, P.; Hayward, A.; et al. An ingestible bacterial-electronic system to monitor gastrointestinal health. *Science* **2018**, *360*, 915-8. DOI PubMed PMC
244. De la Paz, E.; Maganti, N. H.; Trifonov, A.; et al. A self-powered ingestible wireless biosensing system for real-time in situ monitoring of gastrointestinal tract metabolites. *Nat. Commun.* **2022**, *13*, 7405. DOI PubMed PMC
245. Traverso, G.; Langer, R. Perspective: Special delivery for the gut. *Nature* **2015**, *519*, S19. DOI PubMed
246. Boys, A. J.; Güemes, A.; Ma, L.; et al. Implantable bioelectronics for gut electrophysiology. *Nat. Commun.* **2025**, *16*, 10240. DOI PubMed PMC
247. Gopalakrishnan, S.; Thomas, R.; Sedaghat, S.; et al. Smart capsule for monitoring inflammation profile throughout the gastrointestinal tract. *Biosens. Bioelectron. X.* **2023**, *14*, 100380. DOI PubMed PMC
248. Inda-Webb, M. E.; Jimenez, M.; Liu, Q.; et al. Sub-1.4 cm^3 capsule for detecting labile inflammatory biomarkers in situ. *Nature* **2023**, *620*, 386-92. DOI PubMed
249. Yang, L.; Gan, L.; Zhang, Z.; et al. Insight into the contact impedance between the electrode and the skin surface for electrophysical recordings. *ACS. Omega.* **2022**, *7*, 13906-12. DOI PubMed PMC
250. Xue, H.; Wang, D.; Jin, M.; et al. Hydrogel electrodes with conductive and substrate-adhesive layers for noninvasive long-term EEG acquisition. *Microsyst. Nanoeng.* **2023**, *9*, 79. DOI PubMed PMC
251. Lu, F.; Wang, C.; Zhao, R.; et al. Review of stratum corneum impedance measurement in non-invasive penetration application. *Biosensors* **2018**, *8*, 31. DOI PubMed PMC
252. Someya, T.; Sekitani, T.; Iba, S.; Kato, Y.; Kawaguchi, H.; Sakurai, T. A large-area, flexible pressure sensor matrix with organic field-effect transistors for artificial skin applications. *Proc. Natl. Acad. Sci. U. S. A.* **2004**, *101*, 9966-70. DOI PubMed PMC
253. Du, Q.; Liu, L.; Sun, S.; Wu, P. Self-compliant ionic nanomesh for gas-permeable and stress-free on-skin electronics. *Nat. Commun.* **2025**, *16*, 11510. DOI PubMed PMC
254. Jang, J.; Choo, H.; Lee, S.; et al. Reconfigurable assembly of self-healing stretchable transistors and circuits for integrated systems. *Nat. Electron.* **2025**, *8*, 474-84. DOI
255. Lee, Y.; Tian, X.; Park, J.; et al. Rapidly self-healing electronic skin for machine learning-assisted physiological and movement evaluation. *Sci. Adv.* **2025**, *11*, eads1301. DOI PubMed PMC
256. Zhao, C.; Park, J.; Maulà, D.; et al. Skin-like drift-free biosensors with stretchable diode-connected organic field-effect transistors. *Nat. Electron.* **2025**, *8*, 981-93. DOI
257. Li, J.; Lee, K. A novel method for soft contact sensing based on electrical impedance sensitivity images. *IEEE. Sens. J.* **2022**, *22*, 9296-305. DOI
258. Li, J.; Lee, K. Muscle-driven joint-torque estimation based on voltage-torque mapping of electrical impedance sensing. *IEEE. Sens. J.* **2023**, *23*, 13966-77. DOI
259. Li, J.; Niu, B.; Wu, K.; Xiao, J.; Liu, X.; Wang, Y. Bioelectronic sensors for neuromuscular perception in human-machine interfaces. *Adv. Robot. Res.* **2025**, *1*, e70074. DOI
260. Han, M.; Chen, L.; Aras, K.; et al. Catheter-integrated soft multilayer electronic arrays for multiplexed sensing and actuation during cardiac surgery. *Nat. Biomed. Eng.* **2020**, *4*, 997-1009. DOI PubMed PMC
261. Madhvapathy, S. R.; Wang, J. J.; Wang, H.; et al. Implantable bioelectronic systems for early detection of kidney transplant rejection. *Science* **2023**, *381*, 1105-12. DOI PubMed
262. Wang, M. Taking kidney temperatures to detect rejection. *Nat. Rev. Nephrol.* **2023**, *19*, 753. DOI PubMed
263. Doloff, J. C.; Veiseh, O.; de Mezerville, R.; et al. The surface topography of silicone breast implants mediates the foreign body response in mice, rabbits and humans. *Nat. Biomed. Eng.* **2021**, *5*, 1115-30. DOI PubMed
264. Kim, H. J.; Koo, J. H.; Lee, S.; Hyeon, T.; Kim, D. Materials design and integration strategies for soft bioelectronics in digital healthcare. *Nat. Rev. Mater.* **2025**, *10*, 654-73. DOI
265. Song, E.; Chiang, C. H.; Li, R.; et al. Flexible electronic/optoelectronic microsystems with scalable designs for chronic biointegration. *Proc. Natl. Acad. Sci. U. S. A.* **2019**, *116*, 15398-406. DOI PubMed PMC

266. Kalashnikov, N.; Barralet, J.; Vorstenbosch, J. Implantable medical devices, biomaterials, and the foreign body response: a surgical perspective. *J. Biomed. Mater. Res. A.* **2025**, *113*, e37983. DOI PubMed
267. Guo, S.; Huang, H. Design strategies for skin-interfaced sensors. *Sens. Actuators. A. Phys.* **2024**, *376*, 115671. DOI
268. Mariello, M. reliability and stability of bioelectronic medicine: a critical and pedagogical perspective. *Bioelectron. Med.* **2025**, *11*, 16. DOI PubMed PMC
269. Zhou, Y.; Burgoyne, Morris. G. H.; Nair, M. Current and emerging strategies for biocompatible materials for implantable electronics. *Cell. Rep. Phys. Sci.* **2024**, *5*, 101852. DOI
270. Lee, M. Y.; Lee, E. S.; Ko, N. Y.; et al. Emerging roles of hydrogels, organogels, and their hybrids in soft bioelectronics and bioplatforms. *npj. Biosensing.* **2025**, *2*, 55. DOI
271. Zhang, J.; Cheng, Z.; Li, P.; Tian, B. Materials and device strategies to enhance spatiotemporal resolution in bioelectronics. *Nat. Rev. Mater.* **2025**, *10*, 425-48. DOI PubMed PMC
272. Lee, H. S.; Jeong, E.; Choi, H.; et al. Facile and robust integration of functional hydrogels into micropillar-structured elastomer platforms for stable cardiac bioelectronics. *Sci. Adv.* **2026**, *12*, eaeb9059. DOI PubMed PMC
273. Cao, J.; Li, X.; Liu, Y.; Zhu, G.; Li, R. W. Liquid metal-based electronics for on-skin healthcare. *Biosensors* **2023**, *13*, 84. DOI PubMed PMC
274. Moon, H.; Aymon, B. F. G.; Deng, J.; et al. Adhesive nonfibrotic bioelectronic interfaces on diverse peripheral nerves for long-term functional neuromodulation. *Sci. Adv.* **2025**, *11*, eadz3668. DOI PubMed PMC
275. Zhao, Y.; Landau, S.; Okhovatian, S.; et al. Integrating organoids and organ-on-a-chip devices. *Nat. Rev. Bioeng.* **2024**, *2*, 588-608. DOI
276. Wang, Q.; Dong, X.; Jiang, D.; et al. Bioelectronic Interfaces And Sensors For Neural Organoids. *Microsyst. Nanoeng.* **2025**, *11*, 172. DOI PubMed PMC
277. Liu, C.; Kelley, S. O.; Wang, Z. Self-healing materials for bioelectronic devices. *Adv. Mater.* **2024**, *36*, e2401219. DOI PubMed
278. Indana, D.; Agarwal, P.; Bhutani, N.; Chaudhuri, O. Viscoelasticity and adhesion signaling in biomaterials control human pluripotent stem cell morphogenesis in 3D culture. *Adv. Mater.* **2021**, *33*, e2101966. DOI PubMed
279. Hua, W.; Gaharwar, A. K. 3D biofabricated in vitro models as new approach methodologies for animal alternatives. *NPJ. Biomed. Innov.* **2026**, *3*, 20. DOI PubMed PMC
280. Moroni, L.; Burdick, J. A.; Highley, C.; et al. Biofabrication strategies for 3D in vitro models and regenerative medicine. *Nat. Rev. Mater.* **2018**, *3*, 21-37. DOI PubMed PMC