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RESEARCH: REVIEW – HIGHLIGHTED PAPER

Multifunctional porous soft bioelectronics

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Soft bioelectronics, seamlessly interfacing with the human body to enable both recording and modulation of curvilinear biological tissues and organs, have significantly driven fields such as digital healthcare, human–machine interfaces, and robotics. Nonetheless, intractable challenges persist due to the onerous demand for imperceptible, burden-free, and user-centric comfortable bioelectronics. Porous soft bioelectronics is a new way to a library of imperceptible bioelectronic systems, that form natural interfaces with the human body. In this review, we provide an overview of the development and recent advances in multifunctional porous engineered soft bioelectronics, aiming to bridge the gap between living biotic and stiff abiotic systems. We first discuss strategies for fabricating porous, soft, and stretchable bioelectronic materials, emphasizing the concept of materials-level porous engineering for breathable and imperceptible bioelectronics. Next, we summarize wearable bioelectronics devices and multimodal systems with porous configurations designed for on-skin healthcare applications. Moving beneath the skin, we discuss implantable devices and systems enabled by porous bioelectronics with tissue-like compliance. Finally, existing challenges and translational gaps are also proposed to usher further research efforts towards realizing practical and clinical applications of porous bioelectronic systems; thus, revolutionizing conventional healthcare and medical practices and opening up unprecedented opportunities for long-term, imperceptible, non-invasive, and human-centric healthcare networks.

Introduction

Bioelectronics, as one of humanity's most aspirational endeavors, interfacing with the human body to monitor biophysical and biochemical signals, is a burgeoning technology poised to reshape the way of lives in the foreseeable future, such as wearable/implantable digital healthcare, human–machine interface, telemedicine, and soft robotics [1–3]. Biointerfaces between soft, curvilinear, living biological tissues and rigid electronic systems are the heart of this impending technology to capture, process, and visualize biological activity from the human body with high fidelity, posing a daunting challenge [4,5]. Notably, bioelectronic interfacing was not born at a single "eureka" moment, as decades of research in flexible bioelectronics have been engaged to standardize materials and processing techniques within the electronics community [6–8]. In 1985, General Electric (GE) produced the first sensitive skin, a flexible sheet integrated with infrared sensor arrays that allowed the robot's arm to avert potential obstacles [9]. However, the rigid and bulky configurations, mechanical mismatch between devices and skin, and impaired body home-

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ostasis are severe barriers to their widespread adoption and healthcare applications. In this context, to bridge the gap between dynamic biological systems and stiff electronics, soft bioelectronics have emerged, characterized by their flexibility, stretchability, and compliant form factors, enabling seamless integration with biological tissues (e.g., skin, heart, and brain) to create high-quality bio-interfaces [10,11]. Recent advances in i) molecular engineering for synthesizing intrinsic flexible/stretchable conductive materials (e.g., organic semiconductor, conjugated polymer, and hydrogel) and ii) structure-engineering (e.g., kirigami, wavy, serpentine, and island-bridge) for fabricating stretchable inorganic conductive materials have spurred the development of soft bioelectronics with tissue-like compliance [12,13]. However, the ongoing demand for imperceptibility, chronophysiological comfort, and user-friendliness-oriented customization pose immense and upgraded challenges in the materials design and manufacturing of next-generation bioelectronics.

Porous soft bioelectronics is the way to a library of user-friendly bio-integrated systems that facilitate the bidirectional communications between human organs and their surrounding microenvironments. Typically, sweat evaporation, heat dissipation, moisture transport, and biochemical diffusion could be synchronously facilitated by the porous biointerface, thus alleviating side effects such as inflammation, skin irritation, and immune responses even over extended temporal windows [14–17]. For over a decade, continuous materials-driven efforts have been made to synthesize and fabricate soft bioelectronics with porous-engineered configurations. Both cellular devices processed from bulk materials by top-down approaches (such as laser engraving, etching, and photolithography) and porous conductive materials synthesized by bottom-up approaches (such as electrospinning, 3D printing, and phase separation) have been recruited to fabricate porous soft bioelectronics (Fig. 1) [17–28]. In addition, multifunctional applications of soft porous bioelectronics, such as perspiration sensor patches, smart bandages, and neuron probes, are progressing hand-in-hand with innovations in materials development [29–32]. However, many of these approaches are still limited to a simple additive way of combining the porous conductor or substrate with the existing device. Future innovations and developments in porous bioelectronics necessitate the exploitation of the unique advantages and breakthrough properties of porous soft bioelectronic systems based on the innovative and fundamental understanding of biological tissue-electronic device interactions.

This review aims to provide rational guidelines for designing and fabricating multifunctional porous soft bioelectronics. We focus on materials synthesis, fabrication, and innovation aspects of the porous bioelectronic systems that enable imperceptible integration with soft biological systems for wearable and implantable applications. We start with discussing the design and fabrication strategies of porous conductive materials for creating porous soft bioelectronics. Then, we review state-of-the-art multifunctional wearable and implantable applications based on the porous soft bioelectronic individual devices and integrated multimodal systems, highlighting the impact of porous configuration on multiscale biointerface control from molecular to tissue-, organ-scale to human-level integration. Finally, we conclude with a perspective on the emerging challenges, bottlenecks, and opportunities for innovation in multifunctional porous soft bioelectronics.

Design and fabrication of porous soft bioelectronics

User-friendly and imperceptible porous bioelectronics are gaining intensive research attention due to the growing demand for long-term wearing comfort. Important considerations for developing porous soft bioelectronics are the material design with the ability to mimic the biological and biomechanical properties of human organs, such as softness, flexibility, and matter adoption/emission. Compared with conventional dense bulk materials, which are rigid and nonpermeable, porous soft electronic materials offer mechanical softness, thermophysiological comfort, and tailorable conductivity and even exhibit unconventional properties that could be deployed to tackle unmet biomedical challenging issues [33]. The following section will discuss the recent advances in materials design concepts and fabrication strategies for developing soft porous bioelectronics.

Electrospinning

Electrospinning is a large-scale, versatile, and cost-effective method for fabricating nanofibers with exceptional properties, such as high porosity, large surface area, softness, flexibility, and structural diversity [34–37]. Typically, electrospinning involves electrifying a liquid or melt droplet to generate fluid jets, which are then stretched and solidified into ultrafine fibers with diameters ranging from tens of nanometers to a few micrometers **(**Fig. 2**a)** [38–42]. Moreover, fibrous assemblies in various porous configurations, such as meshes, scaffolds, nonwoven mats, etc., could be used as gas-permeable, non-inflammatory, and conformal biointerfaces between the device and bio-objects of different shapes [43–45]. Despite these structural advantages, most conventional nanofibers are electrically insulating. The transformation of non-conductive nanofibers into conductive ones with promising electrical properties is a necessary step towards fiber-based porous soft electronics. To date, various strategies such as e-beam deposition, in-situ chemical synthesis/conversion, physic coating, and transfer pattering have been developed to enable the formation of metallic fiber while well-retaining the intrinsic high porosity and mechanical compliance of fiber networks [20,46–48]. As a demonstration of inflammation-free on-skin electronics, a substrate-free design using a nanomesh sensor was reported (Fig. 2**b**), in which an ultrathin Au layer was deposited and patterned on the surface of PVA nanofibers [21]. The subsequent dissolving of PVA sacrificing templates further enabled the direct conformal attachment of metal nanomeshes to the skin despite irregular structures. However, attach-type or wrap-type sensors, even in ultrathin configurations, can interfere with and degrade the human sensation of touching objects. To address the issue, an ultrathin nanomesh sensor with compliant nanoporous structures was developed (Fig. 2**c–d**) [19]. The nanomesh sensor consists of four nanofiber layers: a polyurethane (PU) nanomesh-embedded passivation layer, a top Au nanomesh electrode layer, a parylene-coated PU nanomesh intermediate layer with an air gap, and a

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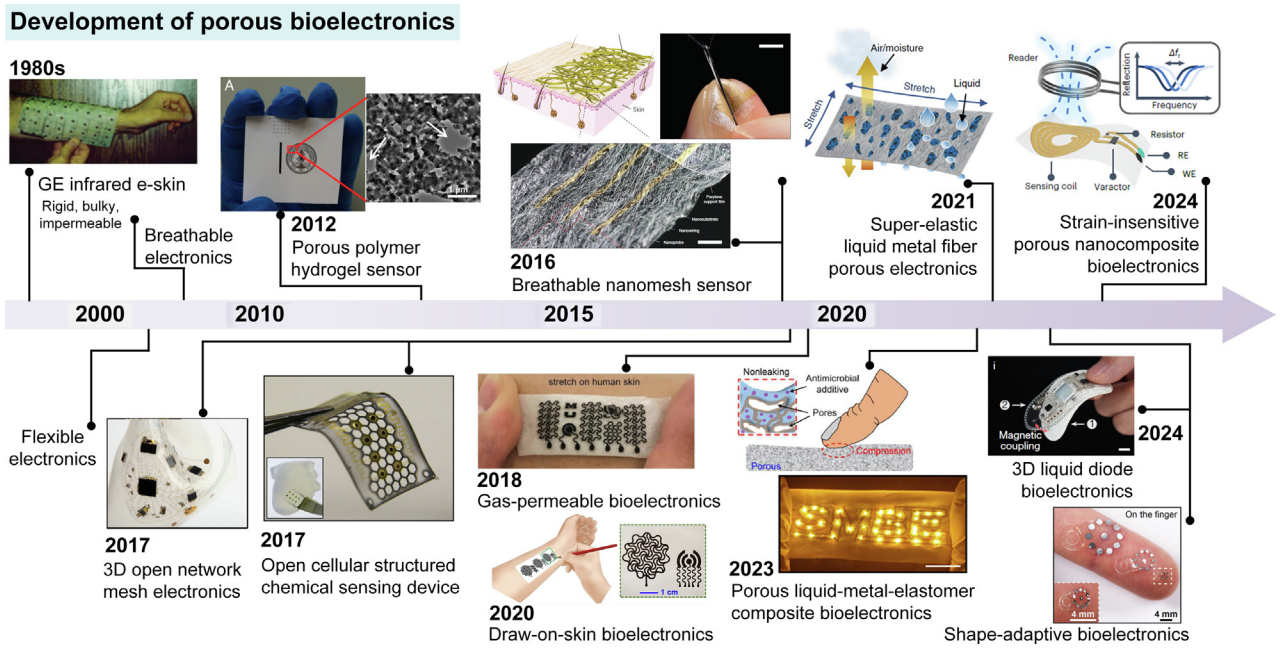

FIG. 1

A brief chronology of the development of porous soft bioelectronics. Reproduced with permission [17–24,26–28,33,140]. Copyright 2012, National Academy of Sciences. Copyright 2017, Wiley-VCH. Copyright 2016–2024, Nature Publishing Group. Copyright 2024, American Association for the Advancement of Science.

bottom Au nanomesh electrode layer. The capacitance changes between the top and bottom Au nanomesh layers enable the monitoring of pressure applied to on-skin sensors. Advances in electrospun nanomesh with multilayer configurations have also been made in on-tissue bioelectronic devices for monitoring dynamically pulsing cardiomyocytes **(**Fig. 2**e-f)**. The nanomesh devices, with only one or two nanofiber layers, equipped with Au nanoprobes, communication nanowires, and porous substrates, exhibited ultra-softness and could freely deform to minimize perturbations of cardiomyocyte motion [20].

Regarding wearable and epidermal bioelectronics, the air/moisture/liquid permeability of the electrospun nanofiber substrates enables long-term user comfort, while system-level nanofiber electronic devices featuring high stretchability with strain-resilience is unambiguously essential for wearables and biomedical implants. Integrating the conductive additives (such as silver (Ag) nanoflakes, Ag nanowires (AgNWs), and gold (Au) nanowires) with elastomer fiber substrates using methods such as blending, spray coating, and growth-loading is a straightforward strategy to construct macroscopically stretchable conductors [49,50]. However, issues such as the cracking and segregation of rigid conductive fillers lead to irreversible deterioration or failure during long-term deformation. Recently, gallium-based liquid metals with liquid-like fluidity and metallic electrical conductivity have emerged as a promising alternative due to their ability to deform while preserving their electrical conductivity [51]. A super-elastic liquid–metal fiber mat conductor was reported by directly coating the liquid metal onto an electrospun elastomeric fibrous mat **(**Fig. 2**g)**. Notably, during a pre-stretch activation process at a strain of 1800 %, the liquid metal embedded in fibrous frameworks self-organized into a laterally mesh-like and vertically buckled structure [22]. Such a unique, organized liquid metal mesh could transform from circular into ribbon-like configurations when stretched, and return to circular configurations when the external load is removed, synergistically enabling the stretchability, electrical stability, and permeability of liquid–metal/elastomer fibrous composites (Fig. 2**h**).

Pattering electronic circuits onto the electrospun fibrous substrates is another effective strategy for fabricating fibrous bioelectronics with well-preserved breathability. However, it is still a critical challenge to pattern microelectrodes on the electrospun substrate with high resolution and pattern density due to the intrinsic large surface roughness and high porosity of nanofiber mats. To address this issue, recently, a wafer-scale patterning strategy was developed to transfer liquid metal microelectrode patterns onto elastomer fibrous substrates (Fig. 2**i**) [52]. The poly(styrene-block-butadiene-block-styrene) SBS fiber mat was directly electrospun onto the Ag microelectrode pattern, which was photolithographed on a silica (SiO_2) wafer modified with water-soluble dextran. The fibrous porous substrates facilitate water penetration and, thus, rapid and isotropic dissolution of the dextran sacrificing layer, enabling large-area pattern transfer. The resulting porous bioelectronic device demonstrated high resolution (down to 2 μ m), high patterning density (>75,000 electrodes cm^{-2}), and superior conductivity at stretch strains up to 1000 % (Fig. 2**j-k**). While electrospinning is widely employed for fabricating nanofiber porous materials as substrates for soft, porous bioelectronics, producing intrinsically conductive nanofibers with superior conductivity remains a significant challenge [53]. Although incorporating a substantial amount of conductive fillers can effectively enhance the conductivity of electrospun fibers, this approach inevitably comprises the electrospinnability of the material and reduces the flexibility of resulting membranes, posing a daunting challenge [50].

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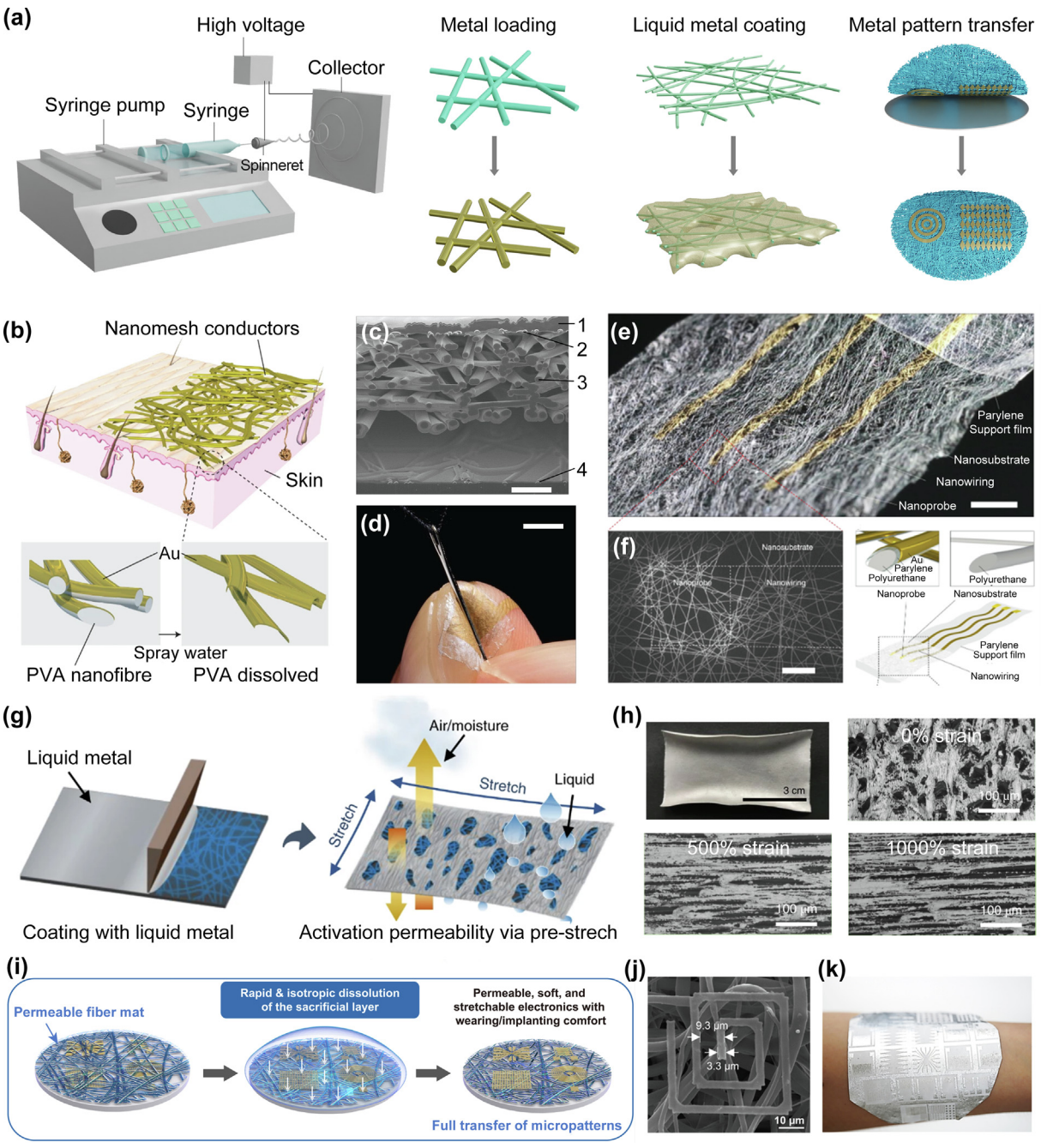

FIG. 2

Electrospun fibrous porous bioelectronics. (a) Typical configuration of electrospinning set-up and strategies for fabricating porous bioelectronics. (b) Inflammation-free and stretchable nanomesh conductors. Reproduced with permission [21]. Copyright 2017, Nature Publishing Group. (c–d) Imperceptible nanomesh pressure sensors. Reproduced with permission [19]. Copyright 2020, American Association for the Advancement of Science. (e–f) On-tissue nanomesh bioelectronic devices consist of nanoprobes, nanowires, and nanosubstrates. Reproduced with permission [20]. Copyright 2019, Nature Publishing Group. (g–h) Super-elastic liquid–metal/SBS fibrous mats. Reproduced with permission [22]. Copyright 2021, Nature Publishing Group. (i–k) Wafer-scale patterning of electrodes on porous fibrous substrates with high resolution and pattern density. Reproduced with permission [52]. Copyright 2023, American Association for the Advancement of Science.

3D printing

3D printing, a subset of additive manufacturing, has garnered significant interest in bioelectronics due to its capability to pattern devices and systems with high resolution and flexibility through direct material deposition [2,54]. To date, advanced 3D printing techniques, including nozzle-based 3D printing (such as direct ink writing, inkjet printing, and fused deposition modeling) and light-based methods (such as selective laser sintering and stereolithography), have been developed [55,56]. Consequently, 3D printing explosively promotes its potential as a multiscale "living" platform, fostering the integration of multimaterials, printing microscale structures, and, ultimately, the manufacture of macroscale devices for soft bioelectronics systems.

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Despite significant progress and promising potential, 3D-printed multifunctional porous bioelectronics, particularly for wearables and implantables, are still in the early stages. In light of the pressing demand for human-specific wearable devices with wear comfort and patient-specific biomedical implants that minimize inflammation, materials design and structure modulation are of paramount significance [57]. For instance, the generation of porous architectures during 3D printing facilitates the permeability of soft bioelectronics, and the adoption of novel nanoscale and functional materials would endow the resulting devices with unique electrical and mechanical properties [58].

Currently, various polymers, inorganic materials, 2D materials, hydrogels, and multi-material composites have been reported for the 3D printing fabrication of porous soft bioelectronics [59]. Poly (3,4-ethylene dioxythiophene): polystyrene sulfonate (PEDOT:PSS) is a representative conductive polymer with mixed ionic and electronic conductivity, holding great potential as a promising candidate for bioelectronics [60]. Given the low viscosity of commercial PEDOT: PSS dispersion, condensing the pristine dispersion to enhance the solid content is a straightforward method to synthesize printable PEDOT:PSS inks with high viscosity. A method of lyophilization in liquid nitrogen and redispersion in water/DMSO mixtures was reported to transform the dilute PEDOT:PSS solution into thixotropic, past-like, and printable inks with high conductivity, owing to the generation of interweaved nanofibril networks [61]. The resultant printability enables the fabrication of PEDOT:PSS into high resolution (over 30 μm), high aspect ratio, overhanging, and micromesh porous structures (Fig. 3**a**).

In addition, next-generation personalized healthcare and early diagnostics necessitate constructing patient-specific anatomical models. It is still challenging to fabricate sophisticated 3D objects with high-resolution anatomical features based on 3D printing. To this end, Cho et al. reported a 3D powder bed printing method by using sugar and elastomer as stuff materials to fabricate a 3D microcellular network-type, interconnected microporous, and conductive elastomer scaffold (Fig. 3**b**) [62]. Sugar grains were employed as a template for forming interconnected porous frameworks, which acted as a scaffold to accommodate soft silicone elastomer and conductive single-walled carbon nanotubes. The subject-specific shape of the as-fabricated sensors fostered faster tracking performances in the real-time monitoring of rapidly changing body signals.

Recently, two-dimensional materials (such as graphene, $Ti_3C_2T_X$ (MXene), and molybdenum disulfide (MoS_2)) have been particularly highlighted due to their extraordinary carrier transport properties [63]. Assembling 2D materials into macroscopic 3D porous bulks with customized patterns, geometry, and conductivity using 3D printing has gained considerable interest. Intuitively, modulating the gelation of nanosheet inks is a straightforward strategy. For example, the one-step oxidation and gelation treatment enable graphene powders to be processed into graphene oxide-based aqueous inks for additive-free 3D printing. However, fabricating 3D-printed 2D materials with tailored macroscopic porous structures and mitigated restack of nanosheets remains challenging. Notably, a recently reported 3D freeze-printing (3DFP) method allows the fabrication of genuinely 3D MXenes aerogels with tailored micro- and macroporous structures (Fig. 3**c**) [64]. Unlike conventional extrusion-based 3D printing, 3DFP does not require the shear-thinning property of inks. Instead, water or ice are deployed as supporting materials to form 3D architectures with overhang features.

Conductive hydrogels have garnered increasing attention for bioelectronic interfaces due to the integrated advantages: i) mixed ionic and electronic conduction that allow seamless integration with living tissues and ii) structures mimicking the extracellular matrix, which facilitate cell proliferation and differentiation [65]. Recently, additive manufacturing of hydrogel-based conductive inks has opened a plethora of new avenues to fabricate porous bioelectronic interfaces. However, the 3D printing of hydrogel with simultaneous excellent electrical conductivity and mechanical robustness has faced lingering challenges. To address the trade-off between electrical and mechanical properties, a bi-continuous ink design strategy was developed using hydrophilic PU as the mechanical phase and PEDOT:PSS as the electrical phase (Fig. 3**d**) [66]. Notably, the simultaneous mild phase separation of PEDOT:PSS and PU enables the formation of homogeneous 3D printable inks. Subsequent solvent evaporation after printing results in a bi-continuous conductive PEDOT:PSS/PU hydrogel with high electrical conductivity (>10 S cm^{-1}) and stretchability (>300 % strain).

Ultimately, 3D printing can be leveraged to formulate multi-materials into a single geometry with structural robustness and multifunctionality. Intuitively, the inks with sensing material composites enable high-performance multimodal physicochemical sensing capabilities. Recently, a universal semisolid extrusion-based 3D printing strategy was reported using composite inks to fabricate epifluidic electronic skin with multifunctional sensing capabilities (Fig. 3**e**) [67]. Given the high surface area and electrochemical activity desired for electrochemical working electrodes, styrene–butadiene–styrene/carbon nanotubes/polyethylene glycol (SBS/CNTs/PEG) composites were employed as inks; thus, 3D porous structures could be generated by the phase elimination of PEG within a printed composite.

Despite the advantages of 3D printing as a mask-free additive manufacturing strategy that offers a promising avenue for advancing customizable soft bioelectronic components with porous structures, integrating 3D-printed porous bioelectronic sensors with porous soft substrates holds great promise for producing lightweight, burden-free wearable bioelectronics in practical wearable applications and clinical settings [68]. However, to date, producing high-quality bioelectronic components with 3D printing on porous substrates remains challenging, largely because of the inherent surface roughness.

Laser engraving

The laser-induced fabrication technique, also known as laser engraving, has emerged as a promising method for the rapid, mask-free, and customizable fabrication of porous bioelectronics with broad material compatibility spanning from 2D materials, metal, metal oxide to metal carbide [69–71]. Over the past decade, synthesis paradigms for porous electronic materials and devices have been explosively developed by regulating laser parameters (e.g., power, speed, and focus) and exploiting light-matter interactions to facilitate localized thermal transforma-

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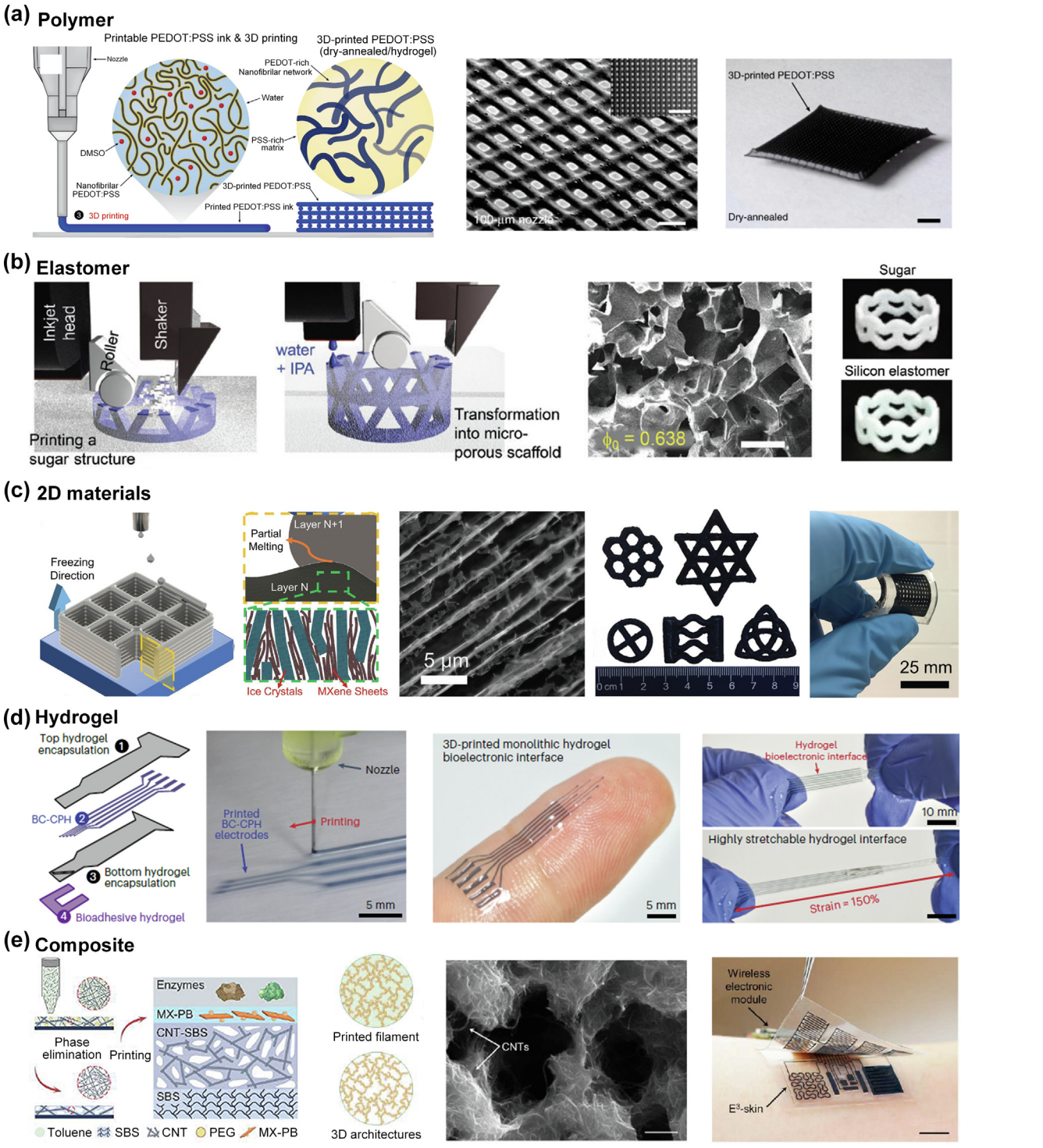

FIG. 3

3D-printed porous bioelectronics. (a) 3D printing of porous PEDOT:PSS overhanging meshes. Reproduced with permission [61]. Copyright 2020, Nature Publishing Group. (b) Fabrication of porous elastomer scaffold via 3D powder bed printing. Reproduced with permission [62]. Copyright 2020, Wiley-VCH. (c) 3D freeze printing of MXene aerogels with truly macroscopic porous structure. Reproduced with permission [64]. Copyright 2022, Wiley-VCH. (d) 3D printing of PEDOT:PSS/elastomer hydrogel bioelectronic interface. Reproduced with permission [66]. Copyright 2023, Nature Publishing Group. (e) Semisolid extrusion-based 3D printing of porous conductive composites for epifluidic electronic skin. Reproduced with permission [67]. Copyright 2023, American Association for the Advancement of Science.

tions [72,73]. Graphene, the most extensively investigated 2D material, has drawn attention within the bioelectronics community due to its intrinsic nanoporous structures, outstanding electrical conductivity, and striking electrochemical activity [74]. Nonetheless, the tedious fabrication process, such as chemical vapor deposition, exfoliation, and flash joule heating,

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are insurmountable impediments to the programmable and miniaturized patterning and wide adoption of graphene in bioelectronic fields [75]. In 2014, the fabrication of laser-induced porous graphene (LIG) on polyimide films using a commercial CO_2 infrared laser system at ambient conditions was reported [76]. LIG has been deemed a significant milestone, eliminating the need for precursors and simplifying the fabrication process. Typically, LIG exhibits hierarchical porous structures and high specific surface area due to the instantaneous liberation of gas molecules during the laser scribing process, thus showing great promise as biophysical or biochemical sensing electrodes for bioelectronic devices [77].

In light of the demand for porous soft bioelectronics, elastomers are considerable candidates as supporting substrates for fabricating LIG-based flexible devices. However, direct LIG synthesis on tissue-like elastomer membranes remains a persistent challenge, owing to the lower carbon contents and melting points. Alternatively, transferring LIG from pristine precursor substrates onto target elastomers emerges as a promising approach. Our group reported a transfer-fabrication strategy to yield highly permeable porous skin-interfaced electronics (Fig. 4**a**) [26]. LIG derived from the polyimide (PI) film with nanoporous structures was first laminated onto partially cured sugar-templated elastomer substrates and, thus, the subsequent solidification and sugar dissolution facilitated the transfer of LIG onto porous elastomer sponge substrates. Such a strategy could synergistically enhance the electrode sensitivity, expedite perspiration evaporation, and prevent sweat aggregation to improve the long-term feasibility of epidermal bioelectronics. Additionally, the direct writing of LIG on flexible porous substrates is a more straightforward strategy for achieving porous soft bioelectronic devices. For example, a laser scribing process to write LIG on Kevlar textile in the air was reported (Fig. 4**a**) [78]. The laser-induced photothermal effect facilitated the cleavage of C═O and C—N bonds in Kevlar and the reorganization of residual carbon atoms into graphene. The laser-assisted fabrication of LIG/textile composites may enable the facile manufacturing of wearable textile electronics. Notably, the ultrafast kinetics induced by laser irradiation predominantly yield intrinsic amorphous structures in LIG, leading to diminished performance, particularly increased resistance, in electronic devices. To address this challenge, a pioneering flash healing strategy was recently developed to rectify topological defects of LIG within milliseconds, thereby improving its crystallinity, conductivity, and flexibility [79]. The resulting LIG-based strain sensor exhibited a gauge factor of 129.3 under a strain of 10 %, demonstrating significant potential for human–machine interfaces.

Furthermore, laser-assisted sintering or printing enables the custom-built fabrication of metal-based integrated circuits and sensors. To date, laser-sintering of solution printable metal nanoparticles (e.g., Au and Ag) has demonstrated high conductivity [72,80,81]. However, the formation of surface oxide layers during laser sintering processes inevitably impedes the enhancement of conductivity. To address this issue, an evaporation–condensation-mediated laser sintering method was reported to fabricate a robust conductor (Fig. 4**b**) [82]. The zinc (Zn) nanoparticle suspension was coated on a glass slide to fabricate homogeneous Zn films, which were then pressed onto a sodium carboxymethylcellulose (Na-CMC) substrate. Upon laser sintering, thermal evaporation in vacuum chambers induces the direct printing of Zn patterns on the Na-CMC substrate. Interestingly, the surface Zinc oxide (ZnO) layer hindered the coalescence of Zn nanoparticles while dominating the evaporation of the Zn core, thereby contributing to the formation of Zn nanoparticles with a hollow porous structure. Arbitrary and grid Zn patterns can also be readily fabricated via such a versatile laser sintering strategy.

Moreover, other transition metals, such as copper and nickel, and their oxide forms with thermal instability have also been demonstrated for laser sintering, reduction, or oxidation to realize the manufacturing of micro/nanoelectronics [83–85]. Recently, our group reported the direct conversion of molybdenum chloride precursors to molybdenum dioxide on various substrates, including polyimide, glass, and hair, while a porous SEBS substrate was employed to fabricate Janus on-skin porous electronics (Fig. 4**c**) [86]. The interconnected porous structures endow the epidermal electronic devices with high breathability. Furthermore, a direct-write laser patterning method for fabricating conductive molybdenum carbide/graphene composites on porous paper substrates has been developed for paper electronics (Fig. 4**d**) [87]. The paper substrates were coated with gelatin-mediated inks containing Mo^{5+}, and subsequent laser scanning directly converted the fibrous paper substrates to molybdenum carbide-graphene composites with multiscale porous structures. The as-prepared porous composites exhibited an electrical resistance of 30 Ω per square, showing great potential as porous electrodes for electrochemical sensors and 3D foldable electronics. Notably, refractory metals, such as molybdenum (Mo) and tungsten, are notoriously challenging to employ for direct laser-assisted additive manufacturing because of the high melting temperature [88,89]. To address the challenge, a versatile laser manufacturing method was developed, in which tar, a byproduct of petroleum processing, was used as both a light absorber and binder to effectively sinter Mo powders into patterned Mo/MoC_x sensors with a sheet resistance below 1.6 Ω per square [90]. Although recent progress of laser-assisted fabrication has been made for soft porous bioelectronic materials, it remains challenging to enable laser engraving as a universal strategy to directly pattern various highly conductive materials on tissue-like soft porous substrates.

Phase separation

Over recent decades, the phase separation method has garnered increased interest due to its adaptivity and versatility for fabricating porous, soft bioelectronics. Specifically, the success story of phase separation stems from the concept of introducing adjustable and interconnected pores into a dense polymer matrix in a simple, template-free, and controlled manner [91]. Typically, the initial homogeneous solution consists of a polymer/solvent/nonsolvent path across the binodal and spinodal upon solvent evaporation, further entering the unstable region where liquid–liquid phase separation occurs, generating polymer-rich and nonsolvent-rich (pore) phases [92]. Given that the constellation of properties of phase-separated porous polymers, such as permeability, liquid resistance, thermal insulation, and moisture transport, can be realized through pore structure regulation,

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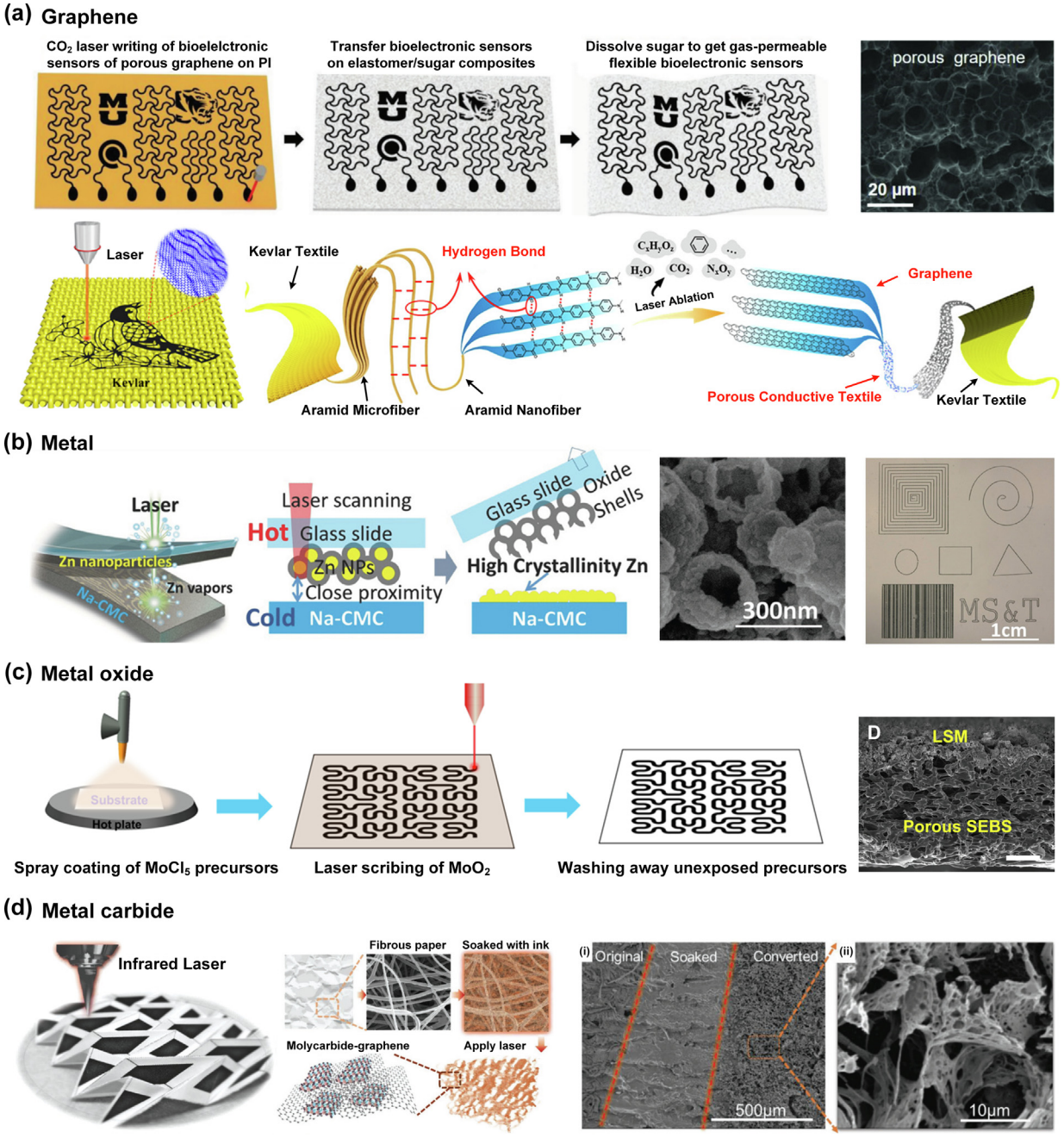

FIG. 4

Laser-engraving-based fabrication of porous soft bioelectronics. (a)Schematic illustration of the fabrication of LIG-based bioelectronic sensors on porous elastomer sponges and Kevlar textiles. Reproduced with permission [26,78]. Copyright 2018, Wiley-VCH. Copyright 2020 American Chemical Society. (b) Evaporation-condensation mediated laser sintering and patterning of porous Zn nanoparticles for bioelectronics. Reproduced with permission [82]. Copyright 2017, Wiley-VCH. (c) Laser scribing of MoO_2 on porous elastomer substrates for Janus epidermal electronics. Reproduced with permission [86]. Copyright 2022, American Association for the Advancement of Science. (d) Direct laser writing of conductive molybdenum carbide-graphene composites on porous paper substrates. Reproduced with permission [87]. Copyright 2018, Wiley-VCH.

phase-separated polymer porous materials have been explicitly deemed a promising candidate for constructing comfortable, breathable, and allergy-free biointerfaces. Moreover, numerous conductive fillers with different configurations, spanning from 0D powders, 1D nanowires, and 2D nanosheets to liquid metal, can be readily embedded into porous phase-separated substrates, thus integrating the complementary advantages of fillers' superior conductivity with the substrates' breathability and mechanical conformability for soft bioelectronic applications. (Fig. 5**a**). Notably, although 0D powders have been explored in bioelectronics, they typically exhibit a higher percolation threshold compared to 1D nanowires and 2D sheets, which poses challenges in achieving optimal electrical conductivity without necessitating excessive filler loading [93]. In contrast, 1D conductive nanowires, with their high aspect ratio, offer enhanced overall conductivity by improving electrical conduction through the

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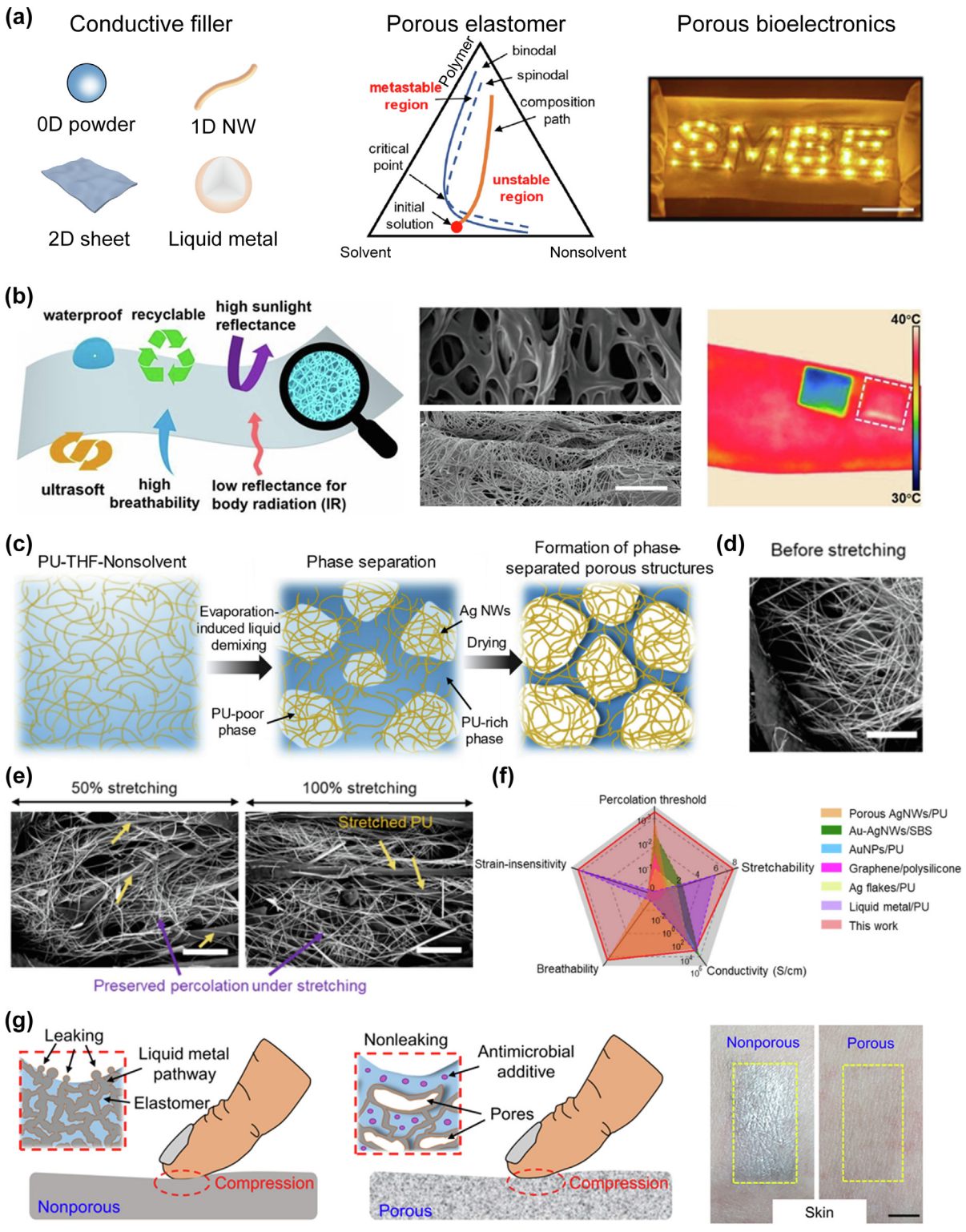

FIG. 5

Phase-separation-based fabrication of porous soft bioelectronics. (a) schematic illustration of the fabrication of porous soft bioelectronics by integrating conductive fillers with phase-separated porous elastomer. Reproduced with permission [23,33]. Copyright 2023, American Association for the Advancement of Science. Copyright 2024, Nature Publishing Group. (b) Multiscale porous engineered soft SEBS substrates with passive cooling, waterproof, and breathable properties. Reproduced with permission [25]. Copyright 2020, National Academy of Sciences. (c–f) One-step phase-separation-based synthesis of strain-insensitive porous PU/AgNWs composites with ultralow percolation thresholds. Reproduced with permission [33]. Copyright 2024, Nature Publishing Group. (g) Phase-separated porous EGaIn/PU composites with LM-leakage-resistant properties. Reproduced with permission [23]. Copyright 2023, American Association for the Advancement of Science.

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materials themselves while minimizing resistance at the junctions between individual nanowires. 2D conductive sheets can also form a percolated conductive network with area-based contacts, thereby reducing contact resistance. Our group designed the SEBS/chloroform/isopropyl solution system to enable the phase separation synthesis of multifunctional SEBS substrates with cross-scale porous architectures. The rapid evaporation of chloroform with a lower boiling point induced the spontaneous phase separation of isopropyl alcohol (IPA) from SEBS to generate micro/nanoscale droplets, which were further evaporated to form multiscale pores with sizes ranging from 0.2 to 7 μm, enabling sunlight backscattering to minimize heat absorption while retaining heat dissipation (Fig. 5**b**) [25]. Covered by the porous SEBS film, the skin temperature is maintained at approximately 6 °C lower than exposed skin under a solar intensity of 840 $W{\cdot}m^{-2}$, demonstrating superior passive-cooling capabilities. Moreover, the hierarchical and interconnected pores endow the SEBS substrate with the desired properties of waterproofing, breathability, and conformability. Various multifunctional and stretchable on-skin electronic devices, including temperature, pressure, and electrophysiological sensors, can be readily fabricated by spray-printing serpentine-like layouts of AgNWs on the pre-stretchable porous substrates. However, challenging issues remain a persistent concern, such as inevitable electrical–mechanical coupling and deteriorated electrical conductivity under long-term cyclic deformation. Significantly, electromechanical stability and strain-insensitive properties are essential for next-generation soft bioelectronics and soft robotics. To this end, strain-induced rearrangement of rigid conductive fillers inside the porous elastomer matrix represents a promising strategy. Our group reported a one-step in-situ phase separation strategy to trigger the adaptive self-organization of conductive fillers within porous polymer substrates (Fig. 5**c**) [33]. Notably, liquid–liquid phase separation occurs upon the evaporation of the volatile solvent (THF) and nonsolvent (ethanol), promoting the formation of PU-rich and PU-poor phases. During the entire phase separation process, Ag NWs with amphiphilic ligands remain in the PU-poor phase due to their immiscibility with the PU solution. Driven by the Pickering effect to minimize system free energy, microscale pores further guide the self-assembly of Ag NWs to form strain-adaptive percolation networks (Fig. 5**d–e**). The energy-dissipative porous PU microarchitectures and adaptable Ag NWs conductive pathways synergistically induce effective and stable electron transport even under extraordinary macroscopic strains of up to 600 % (Fig. 5**f**).

Compared to rigid conductive fillers such as Au NWs, Ag NWs, Ag nanoflakes, graphene, and MXene nanosheets, liquid metal, with liquid-like fluidity, striking deformation capabilities, and metallic electric conductivity, is of particular interest for stretchable electronic devices [51,94]. Embedding liquid metal into an elastomer substrate is a straightforward approach to fabricating soft electronics. For example, a stretchable conductive composite was reported by directly dispersing liquid metal particles into a polydimethylsiloxane film [95]. During an external imprinting process, the liquid metal droplets ruptured and coalesced to in-situ generate conductive pathways with superior electrical conductivity (1.37×10^3 S cm^{-1}). Nonetheless, poor adhesion between liquid metal and substrate, undesired leakage upon deformations, and the impermeability of the membranes remain serious concerns for long-term soft bioelectronics. Our group reported the phase separation-based synthesis of EGaIn-embedded porous soft conductors with high leakage resistance (Fig. 5**g**) [23]. Notably, during the rapid phase separation, EGaIn microparticles self-organized on pore surfaces and further generated conductive trajectories *via* mechanical sintering. Interestingly, the porous structures can dampen external stress to minimize deformation-induced eutectic gallium-indium (EGaIn) leakage. Indiscernible smearing occurred after wearing the porous composites for 3 days, enabling the bioelectronic device to interact with human skin in a long-term, imperceptible, and comfortable manner. Although the phase separation method can be employed to fabricate various porous bioelectronic materials with excellent conductivity and even strain-resilience, supplementary pattering processes are still required to precisely structure these soft conductors into functional bioelectronic devices.

Multifunctional applications of porous soft bioelectronics

System-level bioelectronic devices, constructed from soft porous electronic materials, form artificial biointerfaces with the human body, serving as hubs for collecting physical and chemical biomarker information and processing signals. Advances in the molecular design and porous engineering of soft electronic materials have boosted the creation of pioneering, imperceptible, breathable, and multifunctional biointerfaces between electronics and the human body. This section comprehensively reviews representative porous wearable and implantable sensors and multiplexed devices.

Wearable bioelectronics

In wearable biosensing applications across various practice scenarios, such as running, jogging, swimming, and sleeping, the interactions between the device and skin, such as sweat evaporation and air/moisture permeation, are essential for ensuring long-term wearable comfort, particularly in minimizing skin irritation [14]. Soft porous bioelectronics can be seamlessly laminated on human skin, enabling non-invasive, long-term signal acquisition for point-of-care monitoring, thus dynamically mapping physiological states without causing adverse effects [3]. In this section, we summarize the recent progress in various porous bioelectronic sensors and systems for monitoring biophysical, biochemical, and electrophysiological signals.

Biophysical sensors

Harnessing biophysical signals from the human body, such as temperature, pressure, strain, and electrophysiological signals, is essential for monitoring dynamic changes in an individual's health status [12,96]. In this regard, wearable soft biophysical sensors with porous configurations are of great interest for patient-centric digital healthcare, owing to the capability of building a comfortable on-skin diagnostic platform with high sensitivity, wide detection range, and imperceptibility.

Temperature is a physiological biosignal with temporal and spatial variance. Notably, deviations of a few degrees from the

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normal body temperature provide effective cues associated with skin disease, infection, inflammation, and metabolic disorders [97]. Porous LIG-based sensors have been extensively developed for thermal recording. The electron–phonon scattering and thermal velocity of electrons in LIG fluctuate in response to thermal changes, leading to dynamic changes in electrical conductivity [98]. To enhance sensitivity and lower the detection limit, LIG sensors with porous fiber-like structures were proposed, yielding accurate on-body temperature sensing with high sensitivity (-0.06 % $°C^{-1}$) and a low detection limit of 0.051 °C (Fig. 6**a**) [99].

Human skin can perceive external pressure stimuli for tactile sensation; achieving skin-like sensitivity and softness is highly desired for wearable electronics [100]. In response to the need for electronic skin, wearable pressure sensors, one branch of artificial tactile sensors, have been extensively developed, including capacitive, piezoresistive, piezoelectric, and ionotropic pressure sensors [101]. For instance, by sandwiching a highly porous conductive nanocomposite and an ultrathin insulating layer between two parallel electrodes, the device exhibited a hybrid capacitive and piezoresistive response with high sensitivity over a wide pressure range [102]. However, disturbance from laminated sensors to the natural feeling of the skin is a critical challenge. To minimize sensory interference, a fibrous nanomesh device with compliant and ultrathin nanoporous structures has been proposed (Fig. 6**b**) [19]. The repeated capacitance changes of nanomesh devices for lifted objects (e.g., cotton balls, cubic sponges, and bottles) vary due to their different shapes and weights. Notably, such imperceptible sensing will pave the way for continuous monitoring of human motions in natural states.

Moreover, incorporating porous structures into a conductive elastic matrix enables the fabrication of wearable strain sensors, which can conformally interface with curvilinear human body to transform mechanical signals (e.g., stretching, bending, and twisting) into high-sensitivity electrical signals. Notably, with regard to sophisticated human–machine interface and dynamic complex body motions, the next-generation wearable strain sensor necessitates multi-directional strain sensing performance [103]. Recently, an anisotropic and cross-plied strain sensor was developed to address the issue by orthogonally stacking two layers of aligned, anisotropic carbon nanofibrous membranes (Fig. 6**c**) [104]. Due to morphological changes and, thus, disparate electrical resistance changes in different loading directions, the nanofiber sensor device can precisely discriminate multiple degrees of freedom neck motion.

Biochemical sensors

Biochemical markers, such as pH, ions, uric acid, lactate, and glucose, secreted from body fluids (e.g., sweat, tears, saliva, and interstitial fluids), are significant indicators of human health [105]. Real-time monitoring and analysis of these biomarkers are crucial for advancing personalized medicine, therapeutic interventions, and overall healthcare, as they facilitate dynamic and timely adjustments in treatments based on immediate physiological feedback. Currently, wearable soft bioelectronic devices offer a non-invasive and continuous tool for real-time recording of disease-related biochemical species without painful blood collection [106]. Porous-engineered biochemical sensors, with the integrated advantages of high specific surface area and rich surface defects, facilitate electron transfer kinetics for improved biosensing performance [107]. For example, by employing porous carbon or nanoporous gold with a polyaniline (PANI) coating layer as a sensing electrode, the device exhibited an enhanced pH sensitivity (Fig. 6**d**) [108]. To monitor ionic electrolytes, ion-selective porous membranes are usually deposited on electrodes for potentiometric sensing of different ion species. However, the water layer formed between ionophores and 2D flat electrodes inevitably leads to degraded sensitivity and potential drift during long-term operation [109]. Recently, a 3D gradient porous design of graphene electrodes was developed to minimize water accumulation, reduce Na^+ diffusion, and enhance the electroactive surface area, thus yielding a high sensitivity of 65.1 mV $decade^{-1}$ (Fig. 6**e**) [110].

Enzymatic-based biosensors, in which specific enzymes are functionalized on the working electrode to catalyze reactions with target biomarkers, show great potential in wearable and implantable bioelectronics for the monitoring of metabolites [111]. For example, integrating porous bioelectronic materials such as LIG and silver nanowires with oxidases (e.g., glucose oxidase, uricase, and lactate oxidase) functionalization, the fabricated device could be used for real-time monitoring of glucose, uric acid, and lactate levels in biofluids [99]. In addition, it is highly desirable to simplify fluid collection and reduce sampling time to develop a wearable enzymatic biosensor for accurate monitoring of biomarkers. Microfluidic biofluid uptake and sampling, based on the fluidic channel design of substrates and devices, is a promising approach. For example, a porous and hydrophilic sweat-uptake layer was developed to be integrated with the glucose sensor, acting as a pump to rapidly absorb and deliver sweat to the sensor interface (Fig. 6**f**) [108].

Additionally, miniaturization is also an essential factor that should be considered for long-term wearables. However, for wearable electrochemical immunosensors, the degraded signal-to-noise ratio (SNR) of miniaturized devices limits their adoption in reliable molecular diagnostics [112,113]. Recently, a strategy of anchoring aptamers on the nanoporous electrode with optimized pore size and probe density was developed, demonstrating accelerated electron transfer due to deteriorated charge screening and, thus, increased signal gain with lower variation (Fig. 6**g**) [114]. The sensor limit of detection of the nanoporous electrodes was significantly reduced by almost four times compared to pristine nonporous devices.

Electrophysiological sensors

Biopotential signals, including electrocardiography (ECG), electroencephalogram (EEG), electromyography (EMG), and electrocorticography (ECoG), are useful indicators for early medical diagnosis of disorders of the heart, brain, muscle cells, etc. [115,116]. Wearable electrophysiological sensors laminated on the human body enable in-situ non-invasive signaling. Importantly, two prerequisites should be considered: i) ultrathin and conformal design of devices and ii) porous configurations with permeability, which enable the formation of the biotic-abiotic interface that decreases the contact impedance, acquires reliable biopotential signals, and enables long-term imperceptible sensing. For example, a laser-carbonaceous fibrous patch derived

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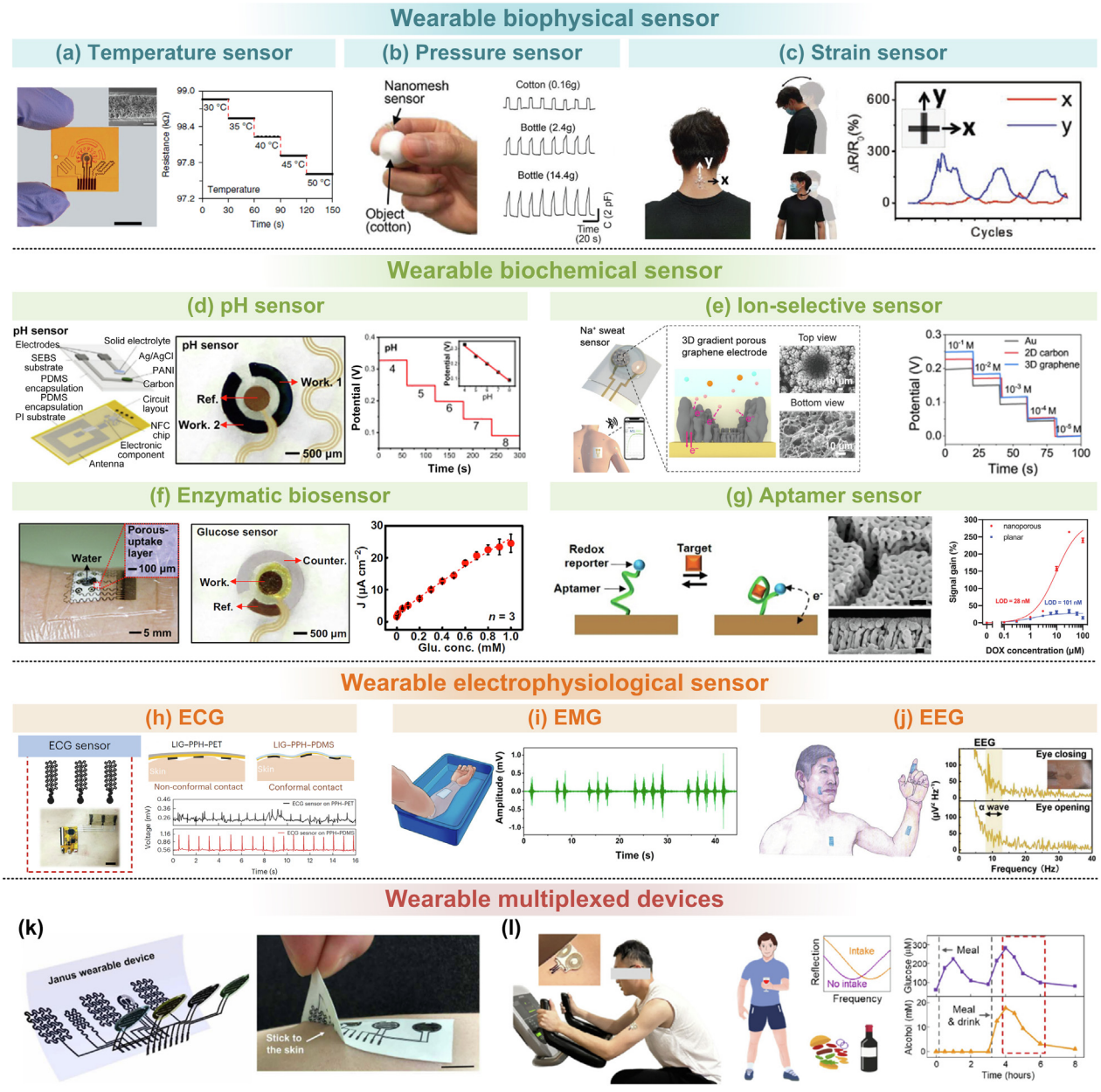

FIG. 6

Multifunctional porous soft wearable bioelectronics. (a) A LIG-based porous perspiration sensing device for real-time recording of temperature. Reproduced with permission [99]. Copyright 2020, Nature Publishing Group. (b) Nanomesh pressure sensor device for measurements of grasp force. Reproduced with permission [19]. Copyright 2020, Nature Publishing Group. (c) Multi-directional strain sensor for monitoring multi-degrees-of-freedom neck. Reproduced with permission [104]. Copyright 2019, Wiley-VCH. (d) Nanoporous carbon- and gold-based pH sensor for the in situ recording of sweat pH value. Reproduced with permission [108]. Copyright 2017, American Association for the Advancement of Science. (e) 3D Gradient porous graphene sensor for monitoring ion concentration of sweat. Reproduced with permission [110]. Copyright 2022, American Chemical Society. (f) Enzymatic-based biosensor device with integrated porous microfluidic channel and electrodes for glucose monitoring. Reproduced with permission [108]. Copyright 2017, American Association for the Advancement of Science. (g) Nanoporous Au-based electrochemical aptamer sensor. Reproduced with permission [114]. Copyright 2021, Wiley-VCH. (h) ECG sensor based on stretchable porous graphene-hydrogel interface. Reproduced with permission [118]. Copyright 2023, Nature Publishing Group. (i) Porous elastomer/Ag NWs-based EMG sensors. Reproduced with permission [25]. Copyright 2020, National Academy of Sciences. (j) LIG-based porous device for EEG signal recording. Reproduced with permission [120]. Copyright 2018, Wiley-VCH. Copyright 2023, Wiley-VCH. (k) Multimodal Janus on-skin bioelectronic systems for electrochemical and electrophysiological sensing. Reproduced with permission [86]. Copyright 2022, American Association for the Advancement of Science. (l) Multiplexed wireless biochemical sweat sensing system. Reproduced with permission [33]. Copyright 2024, Nature Publishing Group.

from MXene/fluoropolymer was developed as an electronic textile for long-term ECG monitoring with a SNR of 37.63 dB [117]. Nonetheless, reliable ECG monitoring necessitates high-quality biointerfaces that can ultra-conformally contact skin even over a larger strain. Most recently, a cryogenic transfer strategy was developed to allow the transfer of LIG onto ultra-thin polyvinyl alcohol-phytic acid-honey (PPH) hydrogel films [118]. Interestingly, the low-temperature atmosphere boosted

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the interfacial bonding between the porous graphene layer and the crystallized water inside PPH hydrogels. The LIG/hydrogel nanocomposites exhibited an intrinsic stretchability of 110 %, owing to the electrically consistent deflected cracks generated as hydrogel could dissipate external energy and provide out-of-plane conductive pathways. High-quality ECG signals were observed from the LIG/PPH device without fluctuation due to the ultrathin, deformable, and conformal interface (Fig. 6**h**).

Wearable EMG sensors facilitate the development of rehabilitation and prosthetic motor control, as the precise EMG signals acquired can interact with external linked devices to control robotic systems in a programmed manner [119]. Recently, the phase-separated porous SEBS/Ag NWs composite on-skin device with waterproofness was developed and mounted on the extensor pollicis longus and extensor pollicis brevis muscles [25]. The porous elastomer device with tissue-like compliance promotes the high-fidelity extraction of EMG signals with a SNR of 20.7 dB, which enables the control of a soft robot hand with a short response time of 8 s (Fig. 6**i**). Similar to wearable EMG sensors, porous elastomer sponges with open-mesh LIG patterns were adopted to record EEG signals [26]. The gas-permeable on-skin sensors with low contact impedance (17 kΩ at 100 Hz) were mounted onto the forehead to acquire the alpha rhythm with the frequency centered at ≈10 Hz (Fig. 6**j**) [120].

Multimodal systems

Multimodal recording of electrophysiological signals, biochemical data, and biophysical activities on the skin can be helpful for comprehensively monitoring health status and early diagnosis [121]. Recently, a multimodal on-skin Janus wearable bioelectronics system was reported based on patterned porous SEBS/MoO_2 composites [86]. Interestingly, the Janus device layout enables the wearable system to synergistically capture multiple biological data (ECG, EEG, EMG, temperature, caffeine, uric acid, and glucose) from human skin and extract signals from human breath (alcohol and NH_3) and surrounding environments (UV and humidity) (Fig. 6**k**). In addition to multiplexed sensing devices, for fully standalone wearable bioelectronic systems, system-level battery-free design should be leveraged to circumvent the need for external power sources. Addressing the issue, a wireless near field communication (NFC) powering solution was proposed based on a stretchable spiral coil made from porous PU/AgNWs nanocomposites with strain-insensitive conductivity (Fig. 6**l**) [33]. Such a battery-free system innovatively facilitated imperceptible and real-time perspiration sensing during everyday activities.

Implantable bioelectronics

Compared with wearable bioelectronics laminated on the skin, implantable bioelectronics are mounted within the body, making intimate contact with internal curvilinear tissues and organs, thus enhancing the fidelity and sensitivity of weak endogenous electrophysiological and biochemical signals from the brain, heart, intestines, or sciatic nerves [122,123]. Notably, implantable bioelectronics necessitates rigorous requirements for the design of materials and devices, including biocompatibility, tissue-like compliance, adhesion, and durability [124–126]. Recently, porous soft bioelectronics have shown great promise as smart implantable devices, interfacing with tender tissues with negligible mechanical mismatch and facilitated biofluid-device interactions. In this section, we review porous soft bioelectronics for implantable electrophysiological recording, biochemical sensing, electrostimulation, and wireless energy powering.

Electrophysiological and chemical recording

Tissue activities, such as hormone release, contraction and relaxation of muscles, and neuron communication, are all induced by the generation and propagation of biopotential signals (Fig. 7**a**) [127]. Hence, recording electrophysiological signals *in vivo* has gained great interest for precisely monitoring bioactivities, diagnosing disorders, and gaining a fundamental understanding of electrophysiological mechanisms [128]. Electrical signals from the cortical surface of the brain, usually known as ECoG signals, are much weaker than ECG and EMG signals. Thus, it is crucial to fabricate devices with low mechanical moduli, superior electrical conductivity, and softness for ECoG recording. Of note, patterned porous stretchable liquid–metal microelectrodes were recently developed as implantable bioelectronics covering the major cortical subdomains of the rat (Fig. 7**b**) [52]. Benefiting from the high channel density (100 electrodes cm^{-2}) and mechanical compliance with the cortical surface of brain tissue, somatosensory evoked potentials in the cortex were successfully recorded under electrical stimulation.

Implantable ECG sensors have also been widely investigated recently. When adapting to heart tissues with complex geometries, the insusceptible electrical conductivity is an essential factor for the long-term recording of cardiac activity. Recently, a stretchable LIG-hydrogel interface with an over fivefold enhancement in intrinsic stretchability was developed, enabling the creation of a soft cardiac patch to seamlessly contact rat hearts for in vivo ECG signal detection (Fig. 7**c**) [118]. Notably, the normal sinus rhythm of a healthy rat and the agonal signals in a rat caused by arrhythmia could be diagnosed.

Real-time in vivo monitoring of the biochemical dynamics of biofluids enables the early warning of the onset of various diseases, such as infections, mental disorders, and cardiovascular diseases [129,130]. To date, significant progress has been made in implantable biosensing electronics. For example, functionalized helical fiber bundles consisting of carbon nanotube-based electrochemical sensors enable in-vivo spatial monitoring of hydrogen peroxide (H_2O_2) in tumors in mice and glucose in the venous blood of cats (Fig. 7**d**) [131]. However, a significant challenge is that when the devices are implanted and penetrate the tissue for signal acquisition, some tender tissues (e.g., amnion) rupture irreversibly and are difficult to regenerate. To address the issue, recently, an interface-stabilized fiber sensor consisting of a restorative gel sheath and multi-ply CNT fiber electrodes has been proposed and seamlessly integrated into the amnion *via* tissue adhesion [132]. Numerous hierarchical gaps between CNT fibers yield high degrees of freedom, thereby enhancing the flexibility of fiber sensing devices. Moreover, the released collagen from the gel sheath can, in situ, promote cell aggregation to prevent amnionic rupture. As demonstrated, implanted fiber sensors with minimized invasiveness showed

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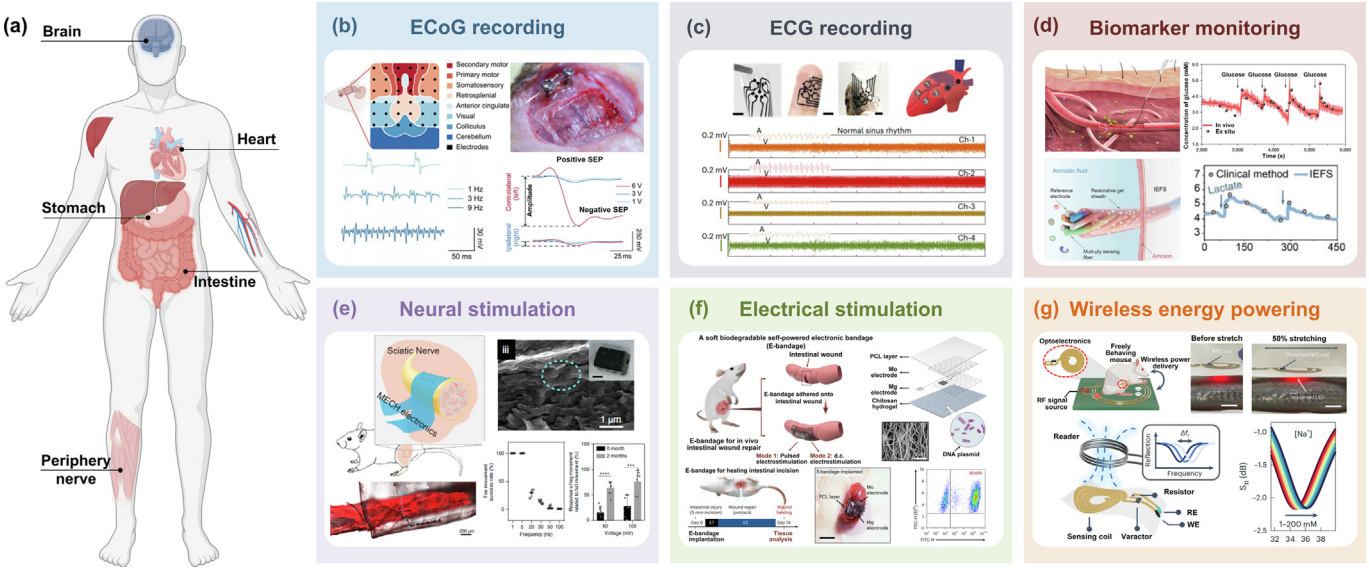

FIG. 7

Multifunctional porous soft implantable bioelectronics. (a) Anatomical characteristics of the human body. (b) Permeable and soft liquid metal/SBS microelectrode for implantable ECoG recording. Reproduced with permission [52]. Copyright 2023, American Association for the Advancement of Science. (c) Fingertip-size LIG/hydrogel electrode array for in vivo detection of ECG signals. Reproduced with permission [118]. Copyright 2023, Nature Publishing Group. (d) Porous helical CNT fiber bundles for implantable biomarker monitoring. Reproduced with permission [131]. Copyright 2024, Wiley-VCH. Copyright 2020, Nature Publishing Group. (e) Hydrogel-based elastic micro-electrodes for in vivo neural stimulation. Reproduced with permission [136]. Copyright 2019, Nature Publishing Group. (f) Soft electronic bandage for in vivo electrical stimulation and accelerated intestinal wound healing. Reproduced with permission [137]. Copyright 2024, Nature Publishing Group. (g) Wireless porous bioelectronics based on phase-separated porous PU/AgNWs composites for wireless energy powering and biochemical sensing. Reproduced with permission [33]. Copyright 2024, Nature Publishing Group.

high sensitivity to biomarkers such as lactate from amniotic fluids.

Electrostimulation

Implantable bioelectronic devices can also be deployed to deliver electrical stimulation as feedback-based electronic therapy [133]. Notably, even a slight mechanical or structural mismatch at the device/tissue interface can trigger an immune response and lead to severe tissue rejection, significantly increasing the risk of tissue damage [134]. To date, most approaches have focused on device-level structural design to yield ultrathin soft electronics or porous engineered mesh electronics with lower bending stiffness or enhanced stretchability, significantly narrowing the mechanical mismatch between implantable stimulators and tissues to reduce adverse immune responses while accommodating body motion [135]. Alternatively, developing intrinsically stretchable and soft porous electronic materials with tissue-analogous Young's modulus and permeability is a fundamental strategy for creating intimate biointerfaces with high spatial resolution and nutrient permeability, thereby mitigating biological mismatch. For example, conductive hydrogel-based elastic microelectrodes with Young's modulus values in the kilopascal range were proposed (Fig. 7**e**) [136]. Electrode arrays of such ultrathin hydrogel 'electronics' enable the generation of a high-quality interface upon being wrapped on sciatic nerves. Thus, a low-voltage electrical stimulation at 50 mV can be effectively delivered on a peripheral nerve to induce leg response. Most recently, another example of a self-powered electronic bandage made from soft and bioresorbable materials was reported (Fig. 7**f**) [137]. The biodegradable magnesium and Mo microelectrode pair were patterned on polycaprolactone nanofibrous films to produce dual electrical therapy, including a direct current electrical stimulation and pulsed electrostimulation-based electrotransfection, promoting healing factor exocytosis to accelerate intestinal wound healing.

Multiplexed systems

The next-generation digital healthcare applications trigger the ongoing demand for the miniaturization of implantable devices and integration of single sensing or stimulation devices into smart multiplexed systems. Although several implantable multiplexed sensing platforms have been demonstrated as a proof of concept, the challenging issues still remain as a standalone implantable system requires power sources for data extraction and communication [138]. Therefore, wireless networks for data transmission and power delivery need to be addressed to circumvent the use of batteries and chips. Recently, our group reported a strain-resilient wireless powering implantable system, in which phase-separated porous PU/AgNWs were used as strain-insensitive conductors to fabricate stretchable spiral coils, which were further implemented in a radiofrequency wireless power transfer system (Fig. 7**g**) [33]. As proof of concept, the transmitter coil could power an implanted light-emitting diode (LED) in mice and fish, even when operated at a 50 % stretch strain. Such an NFC-based wireless power delivery and data transmission system could enable the development of fully battery-free implantable bioelectronic systems for wireless healthcare.

Conclusion and perspective

Recent achievements have demonstrated the promising potential of multifunctional porous soft bioelectronics in paving a new way for a library of imperceptible and user-comfortable

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biointerfacing systems for long-term wearable and implantable healthcare applications, including breathable on-skin perspiration biosensing, non-invasive implantable biosensing, electrical stimulation therapy, and miniaturized wireless power delivery. Advances in the development of porous soft configurations of materials, devices, and systems facilitate the bidirectional communications between bio-tissues and surroundings and the extraction of high-quality bio-signals, making porous soft bioelectronics highly promising as a reliable and burden-alleviated biointerface for next-generation human-centric bioelectronic systems. As presented in this review, we discussed the recent advances in porous soft bioelectronics, focusing on the materials design, fabrication strategies, and porous engineering to develop user-comfortable bioelectronic interfaces. Furthermore, the pioneering progress in multifunctional wearable and implantable applications of soft porous bioelectronics is summarized.

Despite encouraging progress to date, many existing challenges lie ahead in light of the ongoing demands of next-generation whole-body soft bioelectronics to be translated into real-world clinical applications. For wearables, (1) the long-term durability of the electrochemical biosensing performance of porous, soft bioelectronics remains insufficient due to rupture of porous structures and interfaces, device degradation, and severe biofouling. Interface stabilization strategies should be involved based on molecular design and device optimization; (2) Wearable porous bioelectronics for the real-time monitoring of biomarkers such as DNAs/RNAs and proteins from saliva, tears, and exhaled breath is still limited. (3) In regard to wearable items, current wearable porous bioelectronics are still difficult for regular washing as porous configurations and system integrity will be damaged. For implantables, (1) the trade-off between performance attenuation and permeability since porous bioelectronics are more susceptible to inside-body scenarios and usually face challenges of rapid oxidation and degradation rates. The molecular design of the encapsulation layer with simultaneous permeability and extended performance window should be involved; (2) fibrous tissue layers adhering to the implantable porous bioelectronics may induce electronic failure, and a porous interface with robust anti-bio-adhesion behaviors and immunomodulation capabilities should be implemented in the implantable systems. Additionally, following challenges are critical for both wearable and implantable bioelectronics in real-world applications. First, adhesiveness between the porous interface and tissues is still a crucial issue, especially in improving the adhesiveness without sacrificing the intrinsic breathability. Second, most studies aiming to explore optimized designs of porous materials, devices, and systems are still at the trial-and-error stage, which is time-consuming. Future research should focus more on constructing a library of porous material designs, scalable device fabrication, and system integration with machine learning and artificial intelligence [139]. Machine learning can be employed to extract information from scientific literature and provide protocol guidance for the synthesis of porous bioelectronic materials, with a focus on optimizing key properties such as conductivity, stretchability, breathability, cooling capability, and biocompatibility, specifically tailored for wearable and implantable healthcare devices. AI-driven advancements can facilitate the development of enhanced porous soft bioelectronics by enabling efficient material selection, optimization of synthesis pathways, and the potential for rapid large-scale production. Third, the commercial barriers facing porous soft bioelectronics should be addressed to enable practical applications in personalized healthcare. Specifically, future research should prioritize developing high-throughput manufacturing methods, improving cost-efficiency, and enhancing continuous assembly and integration processes. Fourth, for translational wearable and implantable applications, standalone porous soft bioelectronic systems require wireless communication and data transmission and thus, enabling real-time feedback for users. Battery-free, wireless porous soft bioelectronics are ideal for daily use outside hospital settings. Although radiofrequency identification systems powered by transmitters that utilize electromagnetic waves can deliver sufficient energy to support wireless communication, the maximum power output is constrained by the potential risk of tissue damage from excessive exposure to irradiated waves. Additionally, when monitoring multiple biomarkers simultaneously, issues like crosstalk and signal jamming continue to pose significant challenges. Fifth, future research should prioritize clinical trials involving porous soft bioelectronics, particularly in addressing urgent medical challenges. For instance, porous soft bioelectronics, configured as wearable dressings or implantable patches, could be used for clinical monitoring of metabolite concentrations, enhancing neural prostheses, and enabling personalized drug delivery, among other applications, ultimately advancing next-generation healthcare.

In summary, the field of multifunctional porous soft bioelectronics provides unprecedented opportunities to drive progress toward a new paradigm of human-centric, burden-free, and imperceptible bioelectronic systems, forming a whole-body network interface to improve human healthcare, enabling other unforeseen applications, and even reshape the way of life in the near future.

CRediT authorship contribution statement

Feng Zhang: Writing – review & editing, Writing – original draft, Visualization, Conceptualization. **Yadong Xu:** Writing – review & editing, Writing – original draft, Conceptualization. **Ganggang Zhao:** Visualization, Validation. **Zehua Chen:** Visualization, Validation. **Can Li:** Visualization, Validation. **Zheng Yan:** Writing – review & editing, Writing – original draft, Supervision, Project administration, Funding acquisition, Conceptualization.

Data availability

No data was used for the research described in the article.

Declaration of competing interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

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