

REVIEW

Fiber-type soft bioelectronics for wearable and implantable sensing and therapy

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Abstract

Fiber-type soft bioelectronics are revolutionizing wearable and implantable healthcare technologies by addressing critical clinical challenges, particularly minimizing the mismatch in mechanical stiffness between bioelectronics and biological tissues. These devices can seamlessly integrate with dynamic in vivo environments. Their inherent mechanical flexibility and structural adaptability enable applications in both confined sensitive regions and expansive highly mobile areas of the body. Beyond adaptability, fiber-type soft bioelectronics offer multifunctionality, enabling real-time biological signal acquisition, targeted drug delivery, and localized electrical stimulation. Moreover, fabric-based designs offer excellent conformability, making them suitable for long-term monitoring of physical, electrochemical, and electrophysiological signals. This article presents a comprehensive review on fiber-type soft bioelectronics technologies, with a focus on their wearable and implantable applications in healthcare. First, the fundamental requirements for these devices are outlined, describing the foundation for their design and functional integration. Technological advancements that fulfill those requirements are described based on actual examples. The review also examines the materials used for the fibers, highlighting their mechanical, electrical, and biocompatible properties. Next, strategies for fiber fabrication are discussed, including methods for transforming fibers into fabrics. Finally, recent breakthroughs in the applications of fiber- and fabric-type soft bioelectronics in health monitoring and therapeutic interventions are explored.

KEYWORDS

fiber-type bioelectronics, implantable bioelectronics, sensing and therapy, soft materials, wearable bioelectronics

Haneul Kim and Hak Yong Kim contributed equally; Dae-Hyeong Kim and Hye Jin Kim contributed equally.

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1 | INTRODUCTION

Recent advancements in biomaterials have addressed the challenges related to using conventional rigid materials for bioelectronics, by focusing on developing soft and stretchable materials with mechanical properties similar to those of biological tissues (e.g., skin, muscles, and other internal organs).^[1–3] The use of traditional rigid materials in bioelectronics often induces unwanted immune responses and exhibits poor long-term stability within the body due to the mismatch in mechanical stiffness between these materials and biological tissues. In contrast, soft and stretchable materials enhance mechanical compatibility with biological systems, reducing tissue damage, preventing immune responses, and enabling stable, long-term functionality.^[4–6] These unique attributes have garnered significant interest in medical device development, driving innovations in biosensing and therapeutic device technologies.^[7–9]

Bioelectronics with soft and stretchable form factors can operate stably in dynamic *in vivo* environments subjected to repetitive movements, achieving seamless integration with biological tissues due to their exceptional mechanical compliance.^[10,11] Examples include nano/micro particle-based bioelectronics,^[12,13] fiber-type bioelectronics,^[14,15] and patch-type or wearable bioelectronics,^[16,17] each tailored for specific biomedical applications. Particle-based bioelectronics are typically solubilized in water, allowing facile integration into complex biological environments after injection and enabling localized drug delivery,^[18,19] and thermal and/or electrical stimulation (ES).^[20,21] Fiber-type soft bioelectronics are optimized for narrow and intricate environments, such as blood vessels and intestinal tracts, and are particularly effective in accessing delicate regions like the brain and heart.^[22–24] Patch-type or wearable bioelectronics are designed to be applied to larger surface areas, providing capabilities for large-area mapping, and patterned electrical modulation.^[25,26] These versatile devices enable a wide range of biosensing and therapeutic interventions.^[27–29]

Among various types of soft bioelectronics, fiber-type soft bioelectronics stand out for their thin structure, low stiffness, and multifunctionality, enabling minimally invasive insertion into soft, complex, and delicate organs such as the brain,^[30,31] heart,^[32] intestinal tracts,^[14,33] blood vessels (Figure 1). These devices are invaluable for studying complex environments that encompass physical, electrochemical, and electrophysiological processes, such as neural networks in the brain and the functional dynamics of other internal organ systems.^[34–36] Additionally, they can be woven into wearable fabric configurations while retaining their original advantages

including adaptability, breathability, and permeability, etc.^[37,38] These properties enable conformal integration with dynamic, curved body surfaces, allowing the monitoring of diverse biological signals, such as heart rate and muscle activity, over extended periods, making them suitable for real-time health management.^[39,40] Fabric-type soft bioelectronics also support localized ES therapies^[41,42] and human-machine interfacing functions.^[43]

This review explores the potential of fiber-type soft bioelectronics as next-generation technologies for wearable and implantable healthcare applications. The essential requirements for these fiber devices to achieve seamless integration with soft biological systems are first described. Then, the materials utilized for fibers are introduced with an emphasis on their roles in achieving the desired mechanical and electrical properties for soft bioelectronics. Furthermore, fabrication strategies and design approaches for weaving fibers into yarn- and fabric-type soft bioelectronics are discussed, highlighting the potential to expand their applicability across diverse healthcare domains. The state-of-the-art applications of fiber- and fabric-type soft bioelectronics are also introduced, focusing on their biosensing and therapeutic functionalities in implantable and wearable applications. Finally, this review concludes with an assessment of future opportunities and challenges associated with the development and implementation of these technologies.

2 | DESIGN STRATEGIES FOR RELIABLE PHYSIOLOGICAL SIGNAL ACQUISITION

Fiber-type soft bioelectronics facilitate precise monitoring of physiological states by capturing a wide range of physical, electrochemical, and electrophysiological signals, as well as biomolecular concentrations. These devices combine flexibility, biocompatibility, and adaptability, positioning them as indispensable tools for real-time diagnostics and therapeutic interventions. This section presents a detailed overview of critical electrophysiological signals and biomolecules, emphasizing their importance in both clinical and research contexts.

2.1 | Electrophysiological signals

Electrophysiological signals, representing the electrical activities of excitable tissues, serve as essential biomarkers for evaluating the functional state of various tissues and organs.^[44,45] These signals are widely used in clinical diagnostics and therapeutic interventions, including applications in neurology, cardiology, and gastroenterology.

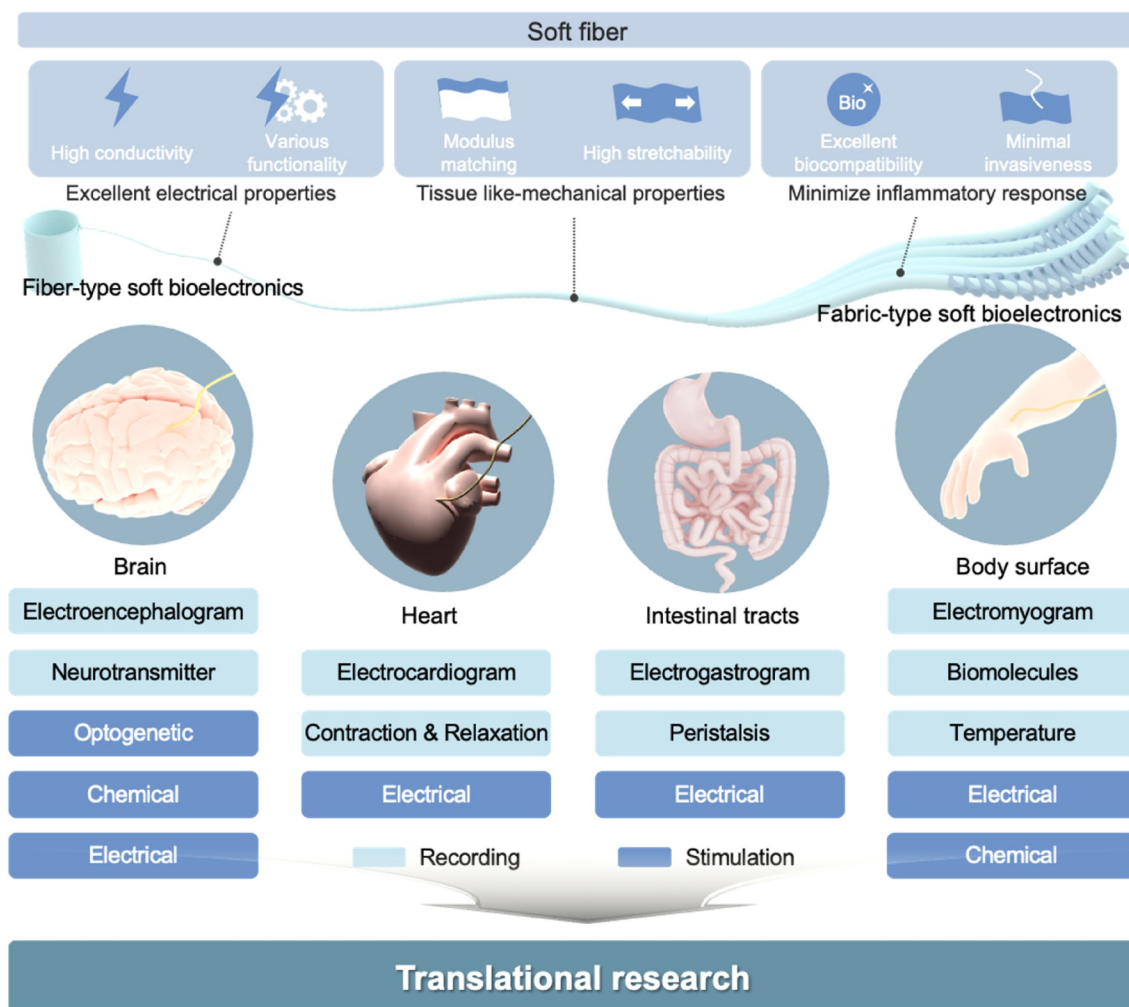


FIGURE 1 Schematic illustration of the evolution of fiber-type soft bioelectronics for translational research in health monitoring and therapeutic interventions.

Representative electrophysiological signals include electrogastrogram (EGG), electroencephalogram (EEG), electrocorticogram (ECoG), electrocardiogram (ECG), and electromyogram (EMG) (Figure 2a).^[46] Each signal is characterized by distinct amplitude ranges and frequency bands, depending on its anatomical origin and physiological function.

EGG reflects the rhythmic electrical activity of the gastrointestinal tract, typically in the range of 50–500 μ V and at low frequencies (0.015 and 0.15 Hz), corresponding to the slow wave contractions regulated by interstitial cells of *Cajal*.^[47,48] EEG and ECoG are widely used to monitor brain activity.^[49–51] EEG captures cortical signals non-invasively via scalp electrodes, with amplitudes ranging from 0.5 to 100 μ V and frequencies between 0.5 and 100 Hz.^[52] In contrast, ECoG involves direct recording from the cortical surface, offering higher signal amplitudes (100 μ V and several mV) and frequency resolution (>100 Hz),^[53,54] suitable for applications requiring precise brain mapping, such as surgical

planning for epilepsy or brain tumor cases.^[55,56] ECG signals typically range in intensity from 0.05 to 5 mV, with a frequency spectrum spanning 0.05–150 Hz.^[57,58] The detailed waveform analysis, including P waves, QRS complexes, and T waves, aids in diagnosing arrhythmias and myocardial ischemia.^[59] Similarly, EMG measures the electrical activity of muscles, offering critical insights into muscle function and serving as a valuable tool for diagnosing muscle disorders and neurological conditions.^[60,61] EMG signals are generated by the depolarization and repolarization of muscle fibers during contractions and relaxations. These signals typically exhibit intensities ranging from 0.01 to 5 mV, and a frequency spectrum spanning between 20 and 500 Hz.

To enable real-time and stable acquisition of these signals, fiber-based electrodes have been developed using conductive materials and stretchable designs. Their ability to conform to dynamic body surfaces makes them suitable for wearable and implantable applications. Fiber-integrated ECG textiles, soft ECoG grids, and knitted

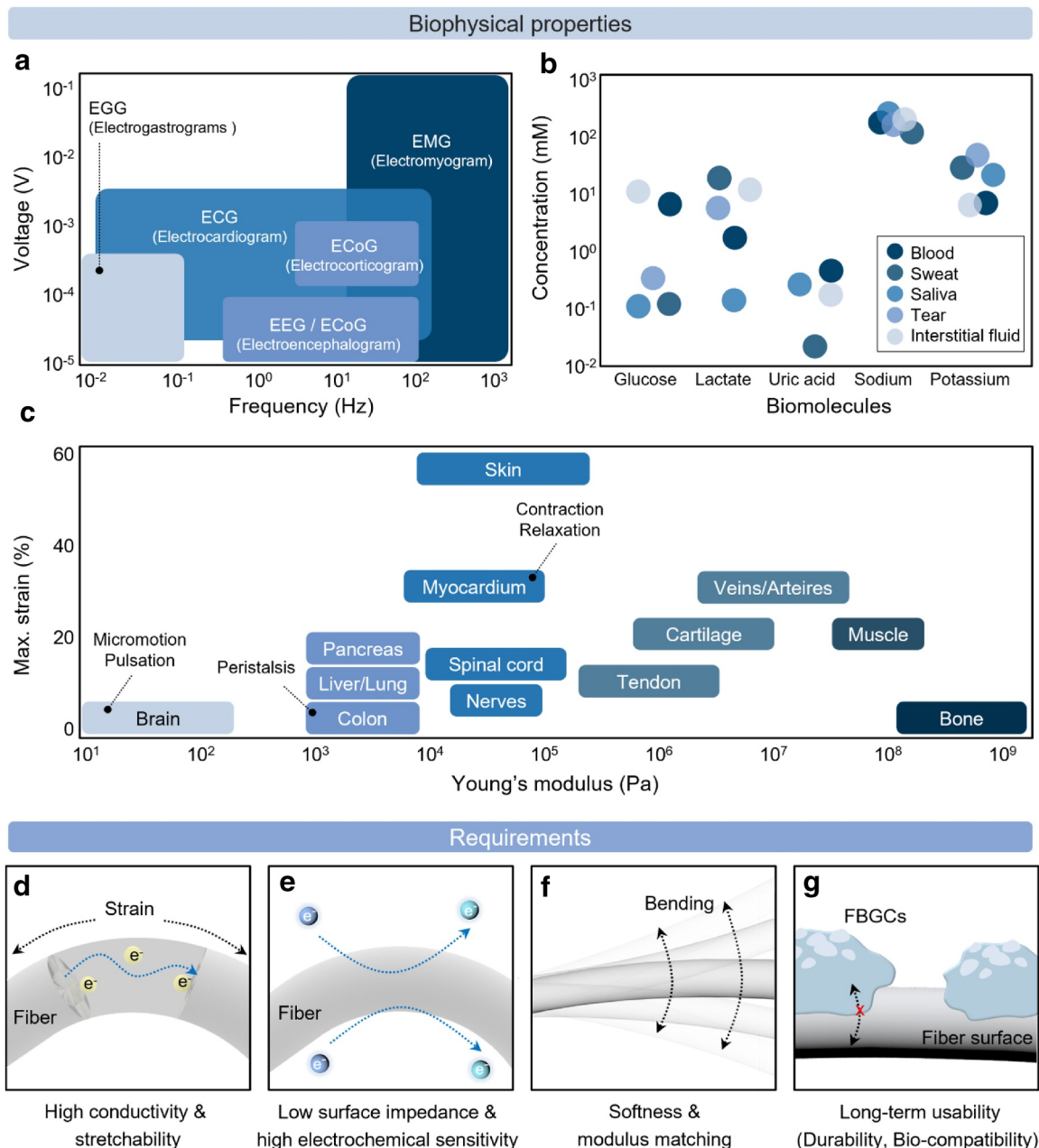


FIGURE 2 Biophysical properties and requirements of fiber-type soft bioelectronics. (a–d) Biophysical properties: (a) Various physioelectrical signals generated in the human body that indicate physiological states. (b) Diverse biomolecules present in various biofluids used for monitoring body conditions. (c) Mechanical properties of different organs. (d–g) Requirements of fiber-type soft bioelectronics: (d) High conductivity and stretchability. (e) Low surface impedance and high electrochemical sensitivity. (f) Softness and modulus matching for biocompatibility. (g) Long-term usability for sustained operation.

EMG electrodes have demonstrated effective performance in long-term biosignal monitoring.

2.2 | Biomolecules

Biomolecules present in biofluids, such as blood, sweat, saliva, tears, urine, and interstitial fluid, are essential indicators for assessing the physiological state of the

human body.^[62,63] These biomolecules, including glucose, lactic acid, uric acid, electrolytes, pH, cortisol, and hormones, play critical roles in maintaining homeostasis and regulating physiological processes. The physiological reference ranges for metabolites and electrolytes vary across biofluids (Figure 2b). In blood, glucose levels typically range from 3.9 to 5.6 mM, lactate from 0.5 to 1.5 mM, sodium from 135 to 150 mM, and potassium from 5 to 6 mM, reflecting its role in energy

metabolism and osmotic balance. Sweat contains lower glucose concentrations (between 0.06 and 0.11 mM) but significantly higher lactate levels (between 16 and 30 mM), highlighting its function in thermoregulation. Sodium and potassium levels in sweat are highly variable, ranging from 10 to 100 and 24 mM, respectively, due to their role in electrolyte excretion. In saliva, glucose levels range from 0.03 to 0.1 mM, while sodium concentrations vary widely between 8.7 and 217.3 mM, reflecting the influence of salivary gland activity. Tears maintain glucose concentrations between 0.06 and 0.3 mM and sodium levels between 120 and 170 mM, both crucial for ocular health. In interstitial fluid, glucose levels range between 3.2 and 9.2 mM, and sodium concentrations between 120 and 154 mM, reflecting its role in maintaining cellular environments.

2.3 | Design strategies guided by physiological signal characteristics

Based on the physiological signal types and characteristics discussed above, this section outlines the design considerations that guide the development of fiber-type soft bioelectronics. These strategies emphasize how mechanical and biological compatibility should be tailored to meet the demands of real-time signal acquisition and long-term use.

2.3.1 | Mechanical properties of tissue and moving organs

Young's modulus, a fundamental mechanical property, quantifies tissue stiffness and reflects its ability to resist deformation under applied stress. Soft tissues, such as the brain, exhibit exceptionally low Young's modulus values, typically between 100 and 500 Pa, indicating their high flexibility and softness. In contrast, hard tissues such as cortical bone have Young's modulus values exceeding 10 GPa, reflecting their significant stiffness (Figure 2c).^[64, 65] This wide range underscores the necessity of designing fiber-type soft bioelectronics with tailored mechanical properties to ensure compatibility with various tissues. The low stiffness of brain tissue requires bioelectronic materials with comparable softness to minimize mechanical damage or inflammation. Skin, muscles, and nerves exhibit progressively higher stiffness, with Young's modulus values of 10–100 kPa for skin, around 100 kPa for muscles, and approximately 1 MPa for nerves. Tendons and cortical bone, ranging between 1 and 2 GPa and exceeding 10 GPa, respectively, demand robust materials for effective integration. Aligning device stiffness

with tissue properties is crucial for ensuring long-term functionality and precise signal detection.

Beyond stiffness, tissues experience varying strain due to physiological movements, which bioelectronic devices must accommodate for seamless integration and reliable performance.^[29,66] Skin undergoes substantial deformation, stretching up to 60% during movement. In contrast, cardiac tissues experience around 30% strain due to their continuous contraction and relaxation. Muscles typically endure strain levels of approximately 20% during activity, while peripheral nerves and brain tissue are subjected to relatively minimal strain, at about 10% and less than 1%, respectively.

2.3.2 | Biocompatibility and long-term stability

In addition to mechanical compatibility, fiber-based soft bioelectronics must exhibit high biocompatibility to minimize immune responses and inflammation. The materials used should be non-toxic, chemically stable, and capable of maintaining their functionality over extended periods in vivo. Furthermore, they must withstand various physical and chemical stresses commonly encountered in physiological environments, such as mechanical deformation caused by tissue movement, chemical composition changes in biofluids, and oxidative stress. To enhance biocompatibility, the device surface can be coated with bio-friendly materials such as polymer-based coatings (e.g., polyethylene glycol, collagen, gelatin, poly(lactic acid), or poly(lactic-co-glycolic acid) or protein-based coatings (e.g., fibrinogen, albumin, fibronectin) that minimize immune responses and inflammation.^[67–69] These coatings help reduce adverse biological reactions and ensure long-term stability within the body. In addition, the device surface can be directly coated with anti-inflammatory or immunosuppressive agents to suppress inflammation and tissue rejection. Such drug coatings are particularly effective in mitigating acute immune responses during the initial stages of implantation. By stabilizing interactions between the device and surrounding tissues, these coatings contribute to improved biocompatibility and enhanced long-term stability.

Long-term stability can be further reinforced by designing materials and structures capable of enduring key stress factors in vivo. For instance, chemically stable materials such as polydimethylsiloxane (PDMS), parylene-C, or fluorinated polymers can resist chemical corrosion caused by ions and enzymes in biofluids, as well as changes in pH.^[70,71] Additionally, addressing oxidative stress through antioxidant treatments or

incorporating protective barrier layers can significantly improve the durability and functionality of the device over time.^[72]

3 | REQUIREMENTS OF FIBER-TYPE SOFT BIOELECTRONICS

To effectively detect and respond to a wide range of physiological signals and biomolecules introduced in the previous section, fiber-type soft bioelectronics must be designed with specific material and structural properties that ensure stable performance in biological environments. These devices must accommodate the mechanical characteristics of target tissues—such as Young's modulus and strain levels—while maintaining functional reliability under dynamic conditions. This mechanical compatibility enables seamless integration with soft and deformable tissues, supporting accurate biosignal acquisition and responsive therapeutic modulation. Accordingly, the key requirements include high conductivity, electrochemical sensitivity, stretchability, multifunctionality, softness, and biocompatibility.

3.1 | High conductivity and stretchability

Conductivity is a critical parameter for accurate recording and stable transmission of electrophysiological signals, addressing challenges such as signal distortion, energy dissipation, tissue damage, and signal delays associated with insufficient conductivity (Figure 2d). Electrophysiological signals within the body are typically of very low intensity, often in the μV range, necessitating precise recording without signal loss. Moreover, low conductivity results in energy loss, reducing the efficiency of ES delivery and necessitating higher voltage or current to stimulate biological tissues.^[73] This has two major adverse effects: first, excessive energy consumption shortens the battery life of bioelectronics, limiting their long-term sustainability. Second, elevated voltage or current levels can damage tissues, potentially causing thermal injury, inflammation, or functional disruption in sensitive areas, such as neural or cardiac tissues. Additionally, low conductivity can introduce signal transmission delays, hindering real-time monitoring and rapid feedback. These delays can distort data and compromise the timely delivery of therapeutic interventions, reducing the overall effectiveness of the bioelectronic system.

Stretchability is required for bioelectronics to preserve their electrical functionality under physical deformation, which is essential for reliable and efficient performance

in dynamic biological environments.^[10] The human body constantly undergoes physical deformations from movements such as heartbeats, lung expansion and contraction, and skin motion. A lack of stretchability can lead to disruptions in electrical connectivity, causing signal loss, distortion, or inconsistent delivery of ES.

3.2 | Low surface impedance and high electrochemical sensitivity

Electrochemical sensitivity refers to the device's ability to detect minute electrical changes induced by specific substances, ions, or chemical reactions (Figure 2e). These subtle variations, generated by physiological signals and chemical markers within the body, require high sensitivity for precise detection. In fiber-type soft bioelectronics designed for prolonged contact with skin or internal tissues, capturing these minor electrical variations is essential for reliable physiological and chemical signal acquisition.

Achieving high electrochemical sensitivity depends on two factors: low surface impedance and high charge storage capacity (CSC). Low surface impedance minimizes signal loss and distortion, improves energy efficiency, and ensures accurate signal transmission.^[74] High impedance, in contrast, can attenuate the signal and reduce energy transfer efficiency, hindering signal reliability and therapeutic outcomes. High CSC further enhances the functionality of fiber-type soft bioelectronics by increasing their capacity to store and transfer electrical charges. This capability is particularly advantageous in applications such as neural rehabilitation and cardiac stimulation, where strong and precise ES is required.

3.3 | Modulus matching and softness

Modulus matching is essential for ensuring mechanical compatibility between fiber-type soft bioelectronics and biological tissues, effectively minimizing the risk of tissue damage, irritation, and inflammation caused by mechanical mismatches (Figure 2f).^[75,76] This is achieved by tailoring the softness of fibers to align with the mechanical properties of the target tissues, primarily defined by Young's modulus. Modulus matching is particularly critical for implantable and wearable devices, as their requirements differ significantly based on their interaction with the body.^[77]

For implantable devices, modulus matching is vital for long-term tissue integration and stability. Stiff devices can exert excessive pressure on surrounding tissues, leading to irritation, inflammation, fibrosis, and compromised

functionality. By contrast, fibers with Young's modulus similar to that of target tissues—such as skin, muscles, or nerves—naturally interact with the surrounding environment, minimizing mechanical strain and ensuring durability and reliability over extended periods. In wearable devices, modulus matching enhances comfort and functionality by conforming to the natural movements of the skin. Materials with mechanical properties similar to the skin allow the device to adapt seamlessly to body dynamics, reducing discomfort and irritation caused by prolonged wear. This improves both the fitness and user experience, making wearable devices more practical for continuous monitoring and other applications.^[78]

3.4 | Long-term usability

Long-term usability is a critical consideration for fiber-type soft bioelectronics, as these devices are often designed for prolonged interaction with biological tissues (Figure 2g). Whether implanted or worn externally, these devices must maintain their functionality, structural integrity, and biocompatibility over extended periods to ensure their effectiveness in biomedical applications. Sustained performance is vital for achieving reliable monitoring, therapeutic intervention, and seamless integration with dynamic biological environments.

Mechanical durability is a key factor influencing the long-term usability of fiber-type soft bioelectronics. These devices must endure repetitive deformations caused by physiological movements such as breathing, muscle contractions, and daily activities. Materials with high stretchability and flexibility are essential to prevent mechanical fatigue, cracking, or delamination, thereby ensuring stable performance over an extended period. Biocompatibility is another critical factor determining the long-term usability of fiber-type soft bioelectronics. Materials that interact with biological tissues over an extended period must avoid triggering adverse reactions, such as inflammation, immune responses, or fibrosis. Failure to maintain biocompatibility can compromise the functional effectiveness of the device and lead to discomfort or complications for the user. Electrochemical stability also plays a crucial role in ensuring the long-term performance of fiber-type soft bioelectronics. Devices that record physiological signals or provide tissue stimulation depend on stable electrochemical properties to maintain accuracy and efficiency.^[79] Over time, a degradation of electrodes or a decline in charge transfer capabilities can significantly compromise device functionality. To address this, various strategies have been implemented to enhance electrode longevity in bio-electronic systems. Biocompatible encapsulation layers,

such as Parylene-C or PDMS, are widely employed to protect electrodes from ionic penetration and moisture-induced degradation. Self-healing materials have also been introduced to restore microcracks and mechanical discontinuities caused by repetitive strain, thereby maintaining electrical continuity. Furthermore, surface modification approaches using graphene or conductive polymer coatings have shown promise in suppressing electrochemical corrosion and preserving signal fidelity. These methods collectively contribute to improved operational stability and extended device lifespan under physiological conditions.

4 | KEY STRATEGIES FOR HIGH-PERFORMANCE FIBER-TYPE SOFT BIOELECTRONICS

Fiber-type soft bioelectronics are being developed as high-performance, multifunctional wearable, and implantable bioelectronics based on novel materials and structure designs, multi-channel and multi-modality implementation, and fiber-type fabric fabrication techniques. Early soft fibers were primarily fabricated using core-type structures, limiting stretchability and functionality (Figure 3a).^[30] To address these limitations, buckled sheath-core structures have been introduced, enhancing stretchability and sensitivity. For example, sensors utilizing this structure achieved a workable strain range exceeding 1135%, a response time of ~16 ms, and gauge factors of 21.3 and 34.22 in specific strain ranges (Figure 3b).^[80] These advancements enable reliable monitoring of both subtle and large muscle motions, as well as applications in wearable and implantable devices. Recent research demonstrated the integration of triboelectric micro-flexure-sensitive fibers that achieved ultra-low curvature detection at 0.1 mm^{-1} and flexure factors exceeding 21.8% within a bending range of 10° (Figure 3c).^[81] These developments showcase tailored structural designs that enhance fiber performance for specific applications. Additionally, surface engineering approaches, such as creating gold-based networks on fibers, have demonstrated alternative methods to improve fiber functionality beyond buckling, exemplified by their application in organic electrochemical transistors (OECTs) with transconductance reaching 135 mS—one to two orders of magnitude higher than previous state-of-the-art devices (Figure 3d).^[82]

Material innovations have played a major role in enhancing the performance of soft fibers. Liquid metals (LMs) have been employed to maximize stretchability and ensure strain-insensitive performance, maintaining electrical conductivity ($\sim 2.5 \times 10^6 \text{ S m}^{-1}$) under mechanical

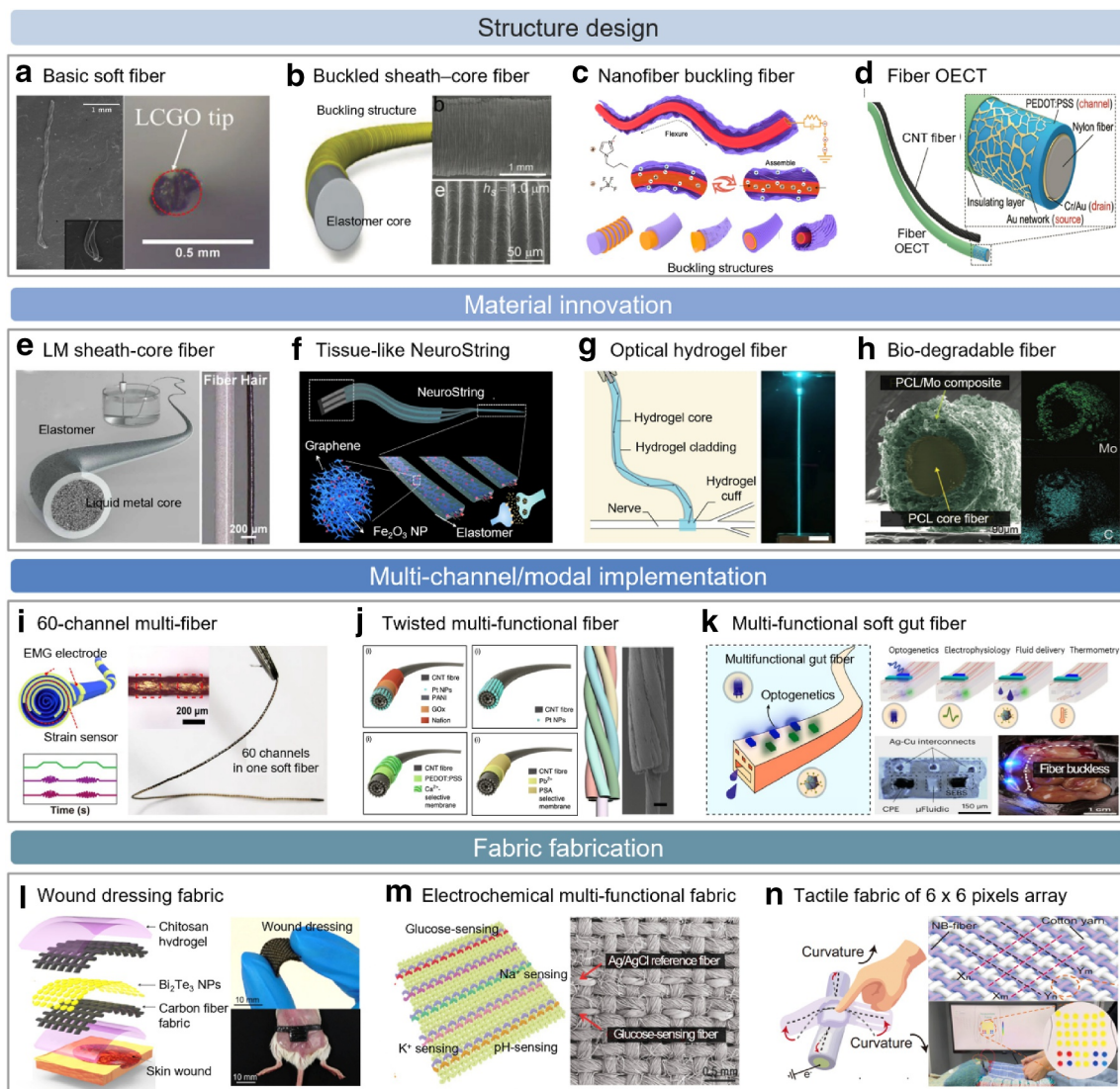


FIGURE 3 Technological advancements in fiber-type soft bioelectronics: (a–d) Surface design to optimize functional interfaces; reproduced (Adapted) with permission.^[30] Copyright 2015, Wiley-VCH. Reproduced (Adapted) with permission.^[80] Copyright 2018, Wiley-VCH. Reproduced (Adapted) with permission.^[81] Copyright 2024, Springer Nature. Reproduced (Adapted) with permission.^[82] Copyright 2023, Springer Nature and Tsinghua University Press. (e–h) Material innovations to improve performance and durability; reproduced (Adapted) with permission.^[83] Copyright 2023, Springer Nature and Tsinghua University Press. Reproduced (Adapted) with permission.^[14] Copyright 2022, Springer Nature Limited. Reproduced (Adapted) with permission.^[84] Copyright 2023, Springer Nature. Reproduced (Adapted) with permission.^[85] Copyright 2023, Wiley-VCH. (i–k) Multi-channel/modal implementation for comprehensive signal monitoring and therapeutic interventions; Reproduced (Adapted) under terms of the CC-BY license.^[86] Copyright 2023. Reproduced (Adapted) with permission.^[87] Copyright 2020, Springer Nature. Reproduced (Adapted) with permission.^[33] Copyright 2024, Wiley-VCH. And (l–n) fabric fabrication to enable scalability and integration into wearable applications. Reproduced (Adapted) with permission.^[88] Copyright 2023, Wiley-VCH. Reproduced (Adapted) with permission.^[89] Copyright 2018, Wiley-VCH. Reproduced (Adapted) with permission.^[81] Copyright 2024, Springer Nature.

stress (Figure 3e).^[83] Elastomers combined with conductive or electrochemically sensitive materials have been used in fibers to achieve superior conductivity, electrochemical sensitivity, and softness, meeting the demands of wearable bioelectronics (Figure 3f).^[14] The use of hydrogels, particularly fatigue-resistant variants, has further

strengthened their capabilities, producing optical fibers with attenuation coefficients as low as 1.07 dB cm⁻¹ at 473 nm and stretchability up to 200% over 30,000 cycles (Figure 3g).^[84] Biodegradable materials, like Mo-embedded polycaprolactone fibers, provide eco-friendly options with high conductivity (~43.5 Ω cm⁻¹) and

robustness over 4000 bending cycles, ideal for transient bioelectronics (Figure 3h).^[85]

Advanced fabrication techniques have enabled the development of multifunctional fibers with integrated features. For example, fibers fabricated by rolling patterned elastomers can incorporate 60 channels, enabling effective recording of physiological signals (Figure 3i).^[86] Also, twisting multifunctional fibers made from carbon nanotubes allows for the simultaneous detection of electrochemical signal variations induced by different biomolecules, providing versatile platforms for biochemical analysis (Figure 3j).^[87] These fibers achieve a bending stiffness compatible with biological tissues and facilitate real-time *in vivo* monitoring with minimal mechanical mismatch. Furthermore, microelectronic fibers incorporating optoelectronic components, such as micro-light-emitting diodes (μ LEDs), electrodes, and microfluidic channels, enable wireless control of neural circuits and physiological monitoring. These fibers are particularly suited for deep-brain and gastrointestinal applications, providing illumination intensities of up to 70 mW mm⁻² and maintaining stable functionality over long-term implantation (Figure 3k).^[33]

Fiber-type soft bioelectronics can also be seamlessly integrated into fabrics using techniques such as weaving, knitting, or embroidering. These fiber-based fabrics can serve as wound dressing layers, facilitating biomedical applications such as sensing diverse electrochemical signals or monitoring localized signal changes at specific points. For example, wound dressings that incorporate triboelectric nanogenerators (TENG) and thermocatalytic materials can provide localized ES and reactive oxygen species generation, enabling accelerated wound healing and infection prevention (Figure 3l).^[88] Also, fibers embedded into textiles facilitate real-time monitoring of physiological parameters such as glucose, sodium ions (Na⁺), potassium ions (K⁺), and pH levels (Figure 3m).^[89] These electrochemical fabrics can maintain high sensitivity and stability under repeated deformations, making them ideal for wearable health monitoring. Crossing fibers within fabric structures allows for the precise detection of signal variations, further enhancing their utility in medical and wearable technologies (Figure 3n).^[81] These fabrics are highly durable, maintaining their performance over more than 10,000 cycles, and offer precise monitoring of subtle physiological deformations such as muscle activity and pulse detection. Through these diverse strategies, fiber-type soft bioelectronics are emerging as an innovative platform for bioelectronics, paving the way for advanced wearable and implantable devices with a wide range of functionalities and applications.

5 | MATERIAL STRATEGIES FOR FIBER-TYPE SOFT BIOELECTRONICS

Fiber-type soft bioelectronics can be fabricated either by thinning traditionally rigid materials to achieve flexibility and softness (Figure 4a) or by utilizing intrinsically soft and stretchable materials (Figure 4b). The thinning-down approach is relatively straightforward and utilizes widely used materials and manufacturing processes. Fibers produced using this method exhibit high initial strength and stability, maintaining their electrical and mechanical properties well. However, these fibers have often limited stretchability and low mechanical compliance, making them unsuitable for dynamic or complex environments. In particular, they may not be appropriate for applications requiring close contact with soft and delicate surfaces, such as wearable and implantable devices.

In contrast, fabricating soft fibers using intrinsically soft materials ensures high mechanical compliance and flexibility. These fibers can function stably even in dynamic motions or complex biological environments, maintaining intimate contact with sensitive surfaces. Specifically, materials with excellent stretchability and biocompatibility provide superior user convenience and long-term stability in wearable and implantable devices.

One key strategy for creating intrinsically soft fibers is to combine soft matrix materials with functional nanofillers to form nanocomposites.^[90,91] Nanocomposites offer a balanced combination of mechanical softness and functional performance, providing high electrical conductivity and stretchability to meet requirements for various applications.^[92] Within the nanocomposites, nanofillers interact with the matrix to form continuous percolation networks.^[93,94] These networks enhance the electrical, thermal, and mechanical performance of the soft fibers while maintaining their inherent flexibility and softness.

This section explores the diverse material strategies employed in the fabrication of intrinsically soft fibers. Research on various materials used for soft fiber production is summarized in Table 1. Matrix materials can be broadly categorized into three types: natural materials, elastomers, and gels, each exhibiting unique properties such as stretchability which means strain limits like breaking point limited to describing matrices, softness, biocompatibility, biodegradability, and filler compatibility means the adaptability of matrices on fillers. Meanwhile, fillers can be classified into five main types: metallic nanomaterials, carbon-based nanomaterials, LMs, functional polymers, and metal oxide nanomaterials. The size, shape, and concentration of fillers influence the formation of percolation networks, which

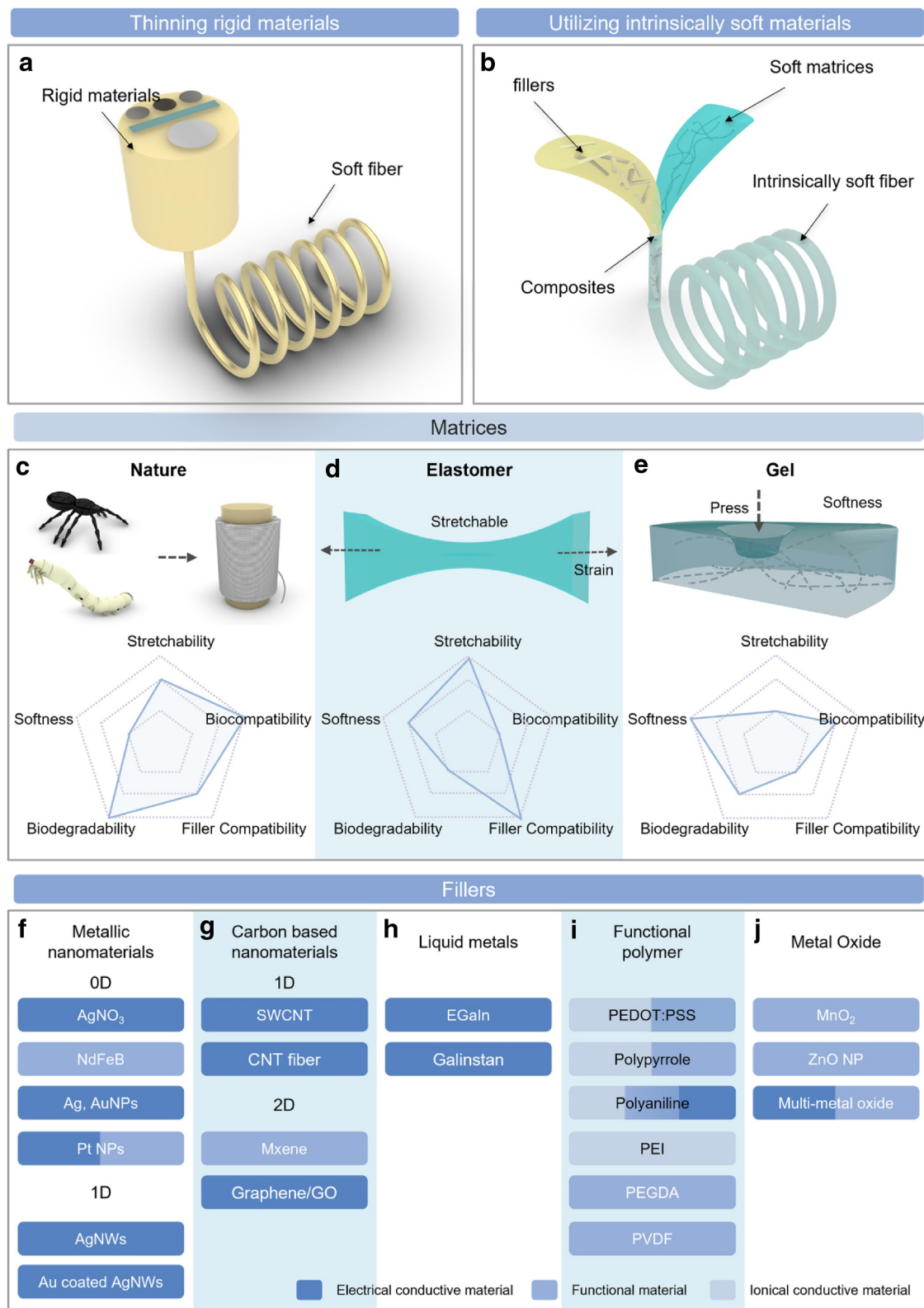


FIGURE 4 Representative methods for fabricating fiber-type soft bioelectronics (a) thinning rigid materials and (b) utilizing intrinsically soft materials. (c–j) Material strategies for fiber-type intrinsically soft bioelectronics. Various matrices, including (c) natural materials. (d) Elastomers. (e) Gels; Various fillers, such as (f) metallic nanomaterials. (g) Carbon-based nanomaterials. (h) Liquid metals. (i) Functional polymers. (j) Metal oxides.

TABLE 1 Overview of fiber-type soft bioelectronics and their performance metrics.

Matrix	Filler	Mechanical property	Performance	Application	Ref.
Wool	MXene ($\text{Ti}_3\text{C}_2\text{T}_x$)	–	Pressure sensitivity: $\sim 80.79 \text{ kPa}^{-1}$	Wearable (wrist, sitting)	[95]
Wool (keratin)	CNT	–	Pressure sensitivity: $\sim 12.64 \text{ kPa}^{-1}$	Wearable (wrist, pulse)	[96]
Nylon, vinyl silicon resin	BaTiO_3 , $\text{ZnS}:\text{Cu}^{2+}$, Ag flakes,	Breakage strength: $\sim 56.4 \text{ MPa}$	Electric field intensity: $\sim 3 \times 10 \text{ V mm}^{-1}$	Wearable (skin touch to fabric)	[97]
Nylon, ecoflex	NdFeB , Iron oxide, AgNO_3	Young's modulus: $\sim 630 \text{ kPa}$	Magneto-mechanical coupling factor: $1.05 \times 10^{-7} \text{ T Pa}^{-1}$	Wearable (wrist)	[98]
SEBS, PVA	EGaIn	Young's modulus: $\sim 1.6 \text{ MPa}$	Peak power density: $\sim 10.2 \mu\text{W m}^{-2}$	Wearable (sports)	[99]
SEBS	EGaIn	Stretchability: $\sim 650\%$ Tensile strength: $\sim 10 \text{ MPa}$	Harvesting efficiency: $\sim 160 \text{ V m}^{-1}$ and $360 \mu\text{W m}^{-1}$	Wearable (forearm elbow, knee)	[100]
Geniomer	GaInStan	Young's modulus: $3.4\text{--}3.8 \text{ MPa}$	Harvesting efficiency: $\sim 490 \text{ V m}^{-1}$ and 175 nC	Wearable (waistband)	[101]
PET	PVDF	Young's modulus: $\sim$ hundreds of MPa	Pressure sensitivity: $\sim 83 \text{ mV N}^{-1}$	Wearable (shoe)	[99]
Glycine	PLLA	–	Pressure sensitivity: $\sim 0.44 \text{ V Pa}^{-1}$	Implantable (rat colon)	[102]
Spandex, PDMS	CNT, AgNWs	–	–	–	[103]
PANI	CNT, CSA, Silicone oil	Young's modulus: $\sim 2.8 \text{ GPa}$	Maximum power factor: $\sim 176 \mu\text{W m}^{-1} \text{ K}^{-2}$	Wearable (wrist band)	[104]
PVDF	BaTiO_3	–	$V_{\text{OC}}: \sim 36.05 \text{ V}/I_{\text{SC}}: \sim 3.126 \mu\text{A}$	–	[105]
PVDF-HFP	ZnO	–	Pressure sensitivity: $\sim 1.9 \text{ V kPa}^{-1}$	Wearable (mask)	[106]
PAN, PDA	ZnO, Mxene	Young's modulus: $\sim 80.14 \text{ MPa}$	Pressure sensitivity: $\sim 28.56 \text{ V N}^{-1}$	Wearable (body)	[107]
TPU, EVA	Ionic liquid (BMimBF ₄)	Breakage strength: $\sim 20 \text{ MPa}$	Flexure sensitivity: $\sim 0.1 \text{ mm}^{-1}$	Wearable (forearm)	[81]
WPU, PEDOT:PSS	Ionic liquid (Emim-TCM), Ag flakes	Young's modulus: $\sim 22 \text{ GPa}$	Power factor: $\sim 26.1 \mu\text{W m}^{-1} \text{ K}^{-2}$	Wearable (wrist)	[108]
PC, PMMA	Carbon fiber, Tungsten	Yong's modulus: $\sim 2.83 \text{ GPa}$	Dopamine sensing: $\sim 172.56 \text{ nA } \mu\text{M}^{-1}$ Attenuation: $0.885\text{--}0.899 \text{ dB cm}^{-1}$	Implantable (rat brain)	[109]
PVA	PEI, SWCNTs	Breakage strength: $\sim 56.4 \text{ MPa}$	Seebeck coefficient P-type: $39.5 \mu\text{V K}^{-1}$ N-type: $48 \mu\text{V K}^{-1}$	Robotic arm	[110]
Alginate	PEGDA	Young's modulus: $\sim 0.5 \text{ kPa}$	Anti-freezing ($< 80^\circ\text{C}$)	Wearable (hand)	[111]
Silk (cocoon)	Ionic liquid (EmimBF ₄)	Break strength: $\sim 55 \text{ MPa}$	Conductivity: $\sim 0.45 \text{ S m}^{-1}$	Robot hands	[112]
Silk (spider), PAA, PVA	ZrOCl_2	Young's modulus: $3.74 \pm 0.16\text{--}118.53 \pm 5.49 \text{ MPa}$	Conductivity: $\sim 0.0014 \text{ mS cm}^{-1}$	–	[113]
Alginate, PEGDA-co-AAM	–	Young's modulus: $0.61\text{--}2.58 \text{ MPa}$	Attenuation $0.46\text{--}0.59 \text{ dB cm}^{-1}$ (450 nm) $0.20\text{--}0.63 \text{ dB cm}^{-1}$ (515 nm)	Implantable (rat brain)	[114]
PVA	–	Young's modulus: $\sim 1.6 \text{ MPa}$	Attenuation $\sim 1.07 \text{ dB cm}^{-1}$ (473 nm) $\sim 0.53 \text{ dB cm}^{-1}$ (589 nm)	Implantable (rat sciatic nerve)	[84]

Abbreviations: AAM, Acrylamide; AgNWs, Silver nanowires; CSA, Camphorsulfonic acid; EVA, Ethylene-vinyl-acetate; PAN, Poly(acrylonitrile); PANI, Poly(aniline); PC, Polycarbonate; PDA, Polydopamine; PDMS, Polydimethylsiloxane; PEDOT:PSS, Poly(3,4-ethylenedioxythiophene) polystyrene sulfonate; PEGDA, Poly(ethylene glycol) diacrylate; PEI, Polyetherimide; PET, Poly(ethylene terephthalate); PMMA, Poly(methyl methacrylate); Poly-L-lactic acid; PVA, Poly(vinyl alcohol); PVDF, Poly(vinylidene fluoride); PVDF-HFP, Poly(vinylidene fluoride-co-hexafluoropropylene); SEBS, Styrene-ethylene-butylene-styrene; SWCNT, Single-wall carbon nanotube; TPU, Thermoplastic polyurethane; WPU, Waterborne polyurethane.

significantly determine the overall physical and functional properties of soft fibers.

5.1 | Various matrices for fiber-type intrinsically soft bioelectronics

5.1.1 | Natural materials

Natural materials, such as those derived from sheep, silkworms, and spiders, provide excellent biocompatibility and biodegradability (Figure 4c).^[115] These materials exhibit high resistance to deformation and possess antimicrobial and water-resistant properties.^[95] However, their high Young's modulus (~5 GPa) often leads to mechanical mismatches with tissues, limiting their use in implantable devices. To address this limitation, recent advancements have focused on extracting protein from natural materials and reprocessing and polymerizing them to produce softer and more flexible fibers. For example, silk fibroin extracted from spiders and silkworms, as well as keratin protein derived from sheep's wool, have been combined with other polymers to produce fibers with improved properties.^[104,112,113] These fibers have undergone molecular adjustment, reducing their Young's modulus to the MPa range (e.g., ~130 MPa), making them mechanically compatible with tissues and suitable for implantable applications.^[116]

5.1.2 | Elastomers

Elastomers, with their low Young's modulus (in the MPa range) and high stretchability, can be easily processed into any desired shape and exhibit high compatibility with various fillers, enabling soft fibers with various functionalities (Figure 4d). Elastomers are categorized on the basis of their crosslinking mechanisms into physically crosslinked elastomers and chemically crosslinked elastomers.

Physically crosslinked elastomers belong to the category of thermoplastic elastomers, where crosslinking occurs through reversible physical interaction, such as phase separation, entanglements, van der Waals forces, or hydrogen bonding. This reversible interaction allows thermoplastic elastomers to reprocess upon heating and maintain low hysteresis under repeated strain cycles. Examples include thermoplastic polyurethane (TPU),^[117] styrene-ethylene-butylene-styrene (SEBS),^[99] ethylene vinyl acetate (EVA),^[81] and Geniomer.^[101] TPU's block copolymer structure provides both flexibility and durability, making it ideal for wearable sensors, cardiac patches, and other dynamic applications. SEBS, with its

excellent mechanical stability and resistance to environmental stress, is widely used for stretchable electronics and biosensor encapsulation.

In contrast, chemically crosslinked elastomers, categorized as thermoset elastomers, form permanent three-dimensional networks through covalent bonds between polymer chains. This structure provides high mechanical strength and durability but makes them unsuitable for reprocessing. Chemically crosslinked elastomers, such as PDMS,^[102] and Ecoflex,^[98] are extensively used in applications like soft robotics, neural interfaces, and microfluidic devices. PDMS offers excellent biocompatibility and chemical stability, with tunable mechanical properties controlled by adjusting the curing agent ratio. Ecoflex, with its high stretchability and softness, is ideal for dynamic implantable devices and soft robotics. Elastomers also exhibit exceptional dielectric properties compared to other matrix materials, making them suitable for various advanced applications. These include multi-layer fiber packaging, triboelectric fiber fabrication, and piezoelectric fiber fabrication.

5.1.3 | Gels

Gels are colloidal solutions composed of a polymer network that retains liquid, providing a unique combination of solid-like structural stability and semi-solid properties (Figure 4e). The physical properties of gels can be finely tuned by adjusting their water content, with Young's modulus typically ranging from tens of Pa and single digits of MPa.^[118] This low modulus ensures mechanical compatibility with biological tissues, minimizing tissue damage and enhancing biocompatibility, making gels particularly suitable for implantable devices.

The polymers forming gels contain hydrophilic functional groups, including ester, hydroxyl, and amide groups, which enable liquid retention and maintain structural stability. Ester groups facilitate biodegradability through hydrolysis, hydroxyl groups enhance structural strength via strong hydrogen bonding, and amide groups maximize water retention through hygroscopic interactions. These functional groups collectively ensure that gels retain their mechanical properties and structural integrity in biological environments.^[119–121]

Natural polymers, such as chitosan and alginate, are commonly employed as gel materials due to their hydrophilic nature.^[111,122,123] Chitosan contains amine and hydroxyl groups, which offer antimicrobial properties and form crosslinked structures in mildly acidic environments to improve structural stability. Alginate contains carboxyl groups that enable ionic interactions with multivalent ions such as calcium ions (Ca^{2+}), which can

regulate the viscoelasticity and mechanical strength of gels.^[124]

Unlike elastomers, gels possess exceptional softness and diffusion properties due to their water content.^[125] The polymeric water network within gels not only contributes to their softness, allowing gel-based fiber to maintain mechanical compatibility with tissue,^[114,116] but also enables controlled drug release in delivery systems and promotes ion diffusion in implantable bioelectronics, enhancing signal transmission efficiency.^[126,127] Furthermore, modulating the water composition in gels can significantly expand their functionalities. For example, incorporating additives such as ionic liquids^[82,112,128] or glycerol^[108,113] significantly enhances ionic conductivity and structural stability. Ionic liquids improve electrical performance by increasing ion mobility, while glycerol prevents dehydration and maintains mechanical flexibility, improving the gel's physical stability. Despite these advantages, water content in gel inhibits the interaction between polymers that maintain morphology and make gels consist of low polymer density. Thus, gels have the most powerful softness but lower stretchability than other materials.

5.2 | Various fillers for fiber-type intrinsically soft bioelectronics

To ensure fiber-type intrinsically soft bioelectronics maintain their intrinsic softness while achieving high performance, the development of an effective percolation network between the matrix and fillers is crucial. The percolation network depends on several factors, including the geometry, spatial distribution, and interfacial interactions of the fillers within the matrix.^[129] Fillers with high aspect ratios are particularly advantageous, as they promote the formation of interconnected pathways, reduce the percolation threshold, and significantly enhance conductivity. The performance of the percolation network depends not only on the quantity but also on the type and properties of fillers. The physical, electrical, and chemical characteristics of fillers play a critical role in determining the overall functionality of the fiber. Properly selected fillers can impart desirable properties including electrical and thermal conductivities as well as mechanical reinforcement. However, inappropriate choices can hinder uniform network formation or compromise the fiber's flexibility and stretchability. Thus, selecting fillers that align with the design objectives and application requirements of the fiber is crucial. In the following sections, we will explore these fillers in detail and discuss their specific roles in enhancing the performance of fiber-type intrinsically soft bioelectronics. The

main filler materials used to fabricate soft fiber bioelectronics are listed in Table 2.

5.2.1 | Metallic nanomaterials

Metallic nanomaterials have been widely used to enhance the conductivity of intrinsically soft fibers, offering the versatility of synthesis in various dimensions (Figure 4f). Each dimensional class provides distinct properties that significantly impact the performance and applications of soft fibers.

Zero-dimensional (0D) nanomaterials include precursors and nanoparticles (NPs). Precursors such as silver nitrate (AgNO_3) and gold chloride (HAuCl_4) are dissolved in a solution with matrix materials or applied in the deposition process for fiber fabrication and surface treatment.^[98,151,152] This approach imparts unique properties to the fiber, eliminating the need for additional filler deposition and ensuring excellent dispersion and uniformity. However, it sometimes makes fiber vulnerable to deformation. NPs, such as silver nanoparticles (AgNPs)^[153] and gold nanoparticles (AuNPs),^[154] are commonly used to enhance electrical conductivity, while platinum nanoparticles (PtNPs)^[87,133] enhance surface impedance and act as catalysts for hydrogen oxidation and oxygen reduction. These materials take advantage of their high surface area to enhance fiber performance and provide functionality.

One-dimensional (1D) nanomaterials are primarily synthesized as nanowires (NWs), characterized by their high aspect ratio. This elongated structure increases the contact area between fillers, facilitating the formation of an efficient percolation network. Consequently, 1D nanomaterials can achieve high conductivity at lower filler concentrations, while preserving the flexibility and softness of the matrix. For this reason, they are widely employed in strain sensors that endure repeated deformation.^[103,155] Common examples include silver NWs (AgNWs)^[156–158] and Au-coated AgNWs.^[159,160] While AgNWs can be synthesized with high aspect ratios and yields, gold NWs (AuNWs) are biocompatible and chemically stable on oxidation but are challenging to produce at scales and yields comparable to those of AgNWs. To overcome these limitations, Au is often coated onto AgNWs, combining the high yield of AgNWs with the stability of gold.^[161]

Two-dimensional (2D) nanomaterials include metallic thin films^[86] and metallic flakes.^[162–164] These materials offer a large surface area and layered structures that provide excellent electrical conductivity and mechanical properties. Metallic thin films, for example, can be deposited onto matrix surfaces, such as

TABLE 2 Filler properties for fabricating fiber.

Filler types	Performances	Ref.
Silver (Ag)	Electrical conductivity: $66.7 \times 10^6 \text{ S m}^{-1}$	[130]
Gold (Au)	Electrical conductivity: $49.0 \times 10^6 \text{ S m}^{-1}$	[131]
Neodymium iron boron (NdFeB)	Remanent magnetization: 1.0–1.4 T	[131, 132]
Platinum (Pt)	Electrical conductivity: $10.2 \times 10^6 \text{ S m}^{-1}$ Catalyze wide range of reactions especially H_2O_2 oxidase, lower impedance	[87, 131, 133]
SWCNT	Electrical conductivity: $10^2\text{--}10^6 \text{ S m}^{-1}$	[134]
CNT fiber	Electrical conductivity: $1.5\text{--}1.7 \times 10^4 \text{ S m}^{-1}$	[135]
Mxene	Electrical conductivity: $11.8 \times 10^5 \text{ S m}^{-1}$	[136]
Graphene	Electrical conductivity: $1.46 \pm 0.82 \times 10^6 \text{ S m}^{-1}$	[137]
Graphene oxide (GO)	Functionality: Enhanced reactivity for surface modification	[138]
Eutectic gallium-Indium	Electrical conductivity: $3.4 \times 10^4 \text{ S m}^{-1}$	[139]
Galinstan	Electrical conductivity: $3.46 \times 10^6 \text{ S m}^{-1}$	[140]
PEDOT:PSS	Ionic conductivity: $10^2\text{--}10^4 \text{ S m}^{-1}$ Functionality: Use as p-doping material	[141]
Ppy	Conductivity: $10^4\text{--}7.5 \times 10^5 \text{ S m}^{-1}$	[142]
PANI	Conductivity: $3 \times 10^3\text{--}2 \times 10^4 \text{ S m}^{-1}$	[143]
PEI	Functionality: Use as n-doping material	[110, 143]
PEGDA	Functionality: High optical transparency	[144, 145]
PVDF	Functionality: Piezoelectric	[146]
MnO_2	Functionality: Increase transducing surface area	[147]
ZnO NPs	Functionality: Piezoelectric	[148, 149]
Multi-metal oxide	Functionality: Multifunctionality based on metal materials	[150]

Abbreviations: CNT, Carbon nanotube; PANI, Polyaniline; PEDOT:PSS, Poly(3,4-ethylenedioxythiophene) polystyrene sulfonate; PEGDA, Poly(ethylene glycol) diacrylate; Ppy, Polypyrrole; PVDF, Poly(vinylidene fluoride); SWCNT, Single-walled carbon nanotube; ZnO NPs, Zinc oxide nanoparticle.

elastomers, to significantly enhance electrical conductivity and introduce additional functionalities. Gold-thin films deposited on elastomer surfaces form conductive pathways, promoting percolation networks while maintaining the material's softness and biocompatibility. This approach is particularly advantageous for wearable and implantable devices, which require both flexibility and functionality.

5.2.2 | Carbon-type nanomaterials

Carbon-type nanomaterials are widely utilized in soft fibers due to their high mechanical strength, excellent electrical properties, and compatibility with soft matrices (Figure 4g).^[165] These materials exhibit outstanding electrochemical sensitivity and electrical and mechanical stability, and provide significant design flexibility and functionalization potential.^[166] Carbon-type nanomaterials can be categorized by their dimensionality into

1D and 2D materials, each contributing unique functionalities to soft fibers.

Single-walled carbon nanotubes (SWCNTs) and carbon nanotubes (CNT) fibers are representative of 1D carbon nanomaterials.^[167] SWCNTs, with diameters as small as 0.4–2 nm, exhibit exceptional electrical conductivity and mechanical properties. Their small size allows them to mix easily with matrix materials, enhancing both flexibility and conductivity without requiring extensive processing.^[104,110] The CNT fibers, fabricated using SWCNTs as building blocks through processes such as floating catalyst chemical vapor deposition (FCCVD) or chemical vapor deposition (CVD),^[82,168] offer high mechanical stability with Young's modulus values in the GPa range. These fibers perform reliably under dynamic conditions, making them ideal electrode materials for electrochemical sensing applications in intrinsically soft fibers.

Graphene, graphene oxide (GO), and MXene are key examples of 2D carbon-type nanomaterials. Graphene

provides excellent electrical conductivity, mechanical strength, and biocompatibility, making it particularly suitable for wearable and implantable fibers.^[14,117]

GO, the oxidized form of graphene excels at depositing nanoparticles and enzymes onto fiber surfaces, enhancing surface functionality.^[169] MXene, composed of single atomic layers of transition metal carbides or nitrides, offers outstanding electrical and catalytic properties.^[95,107] Titanium-based MXene enhances electrical conductivity, while molybdenum-based MXene contributes electrochemical functionalities, such as hydrogen evolution reaction catalysis. These materials can be incorporated into matrix materials or deposited onto fiber surfaces to enhance conductivity and introduce specialized functionalities.

5.2.3 | Liquid metals

LMs are materials that remain in a liquid state at or near room temperature, characterized by their high electrical conductivity, low viscosity, and excellent deformability (Figure 4h).^[170] These properties make LMs highly suitable for integration into flexible and soft fibers. Prominent examples include eutectic gallium-indium (EGaIn), an alloy of gallium and indium with a melting point of 15.5°C, and Galinstan. Both materials exhibit high conductivity and deformability, as well as non-toxicity and biocompatibility, making them ideal candidates for biomedical applications. In intrinsically soft fiber electronics, LMs play a critical role due to their strain-insensitive nature, maintaining conductivity even under significant deformation.^[171] This unique property allows LMs to support the development of soft fiber electronics capable of withstanding large strains. Furthermore, their fluidity facilitates seamless integration with fiber matrices, enabling the creation of multifunctional fibers by combining LMs with nanoparticles or enzymes.^[172] Despite their remarkable properties, LMs face challenges such as the formation of an oxide layer on their surface, which reduces conductivity and hinders adhesion to other materials. Furthermore, due to their fluidity, careful design and encapsulation strategies are required to prevent leakage into surrounding tissues.^[99,100]

A study by Lee et al. demonstrated the fabrication of fibers with semi-solid LM particles for use in implantable devices.^[83] This study proposed a solution-based process using blade coating to deposit LM particles on the surface of fiber that is resistant to deformation and retains its particle shape, thereby preventing leakage.

5.2.4 | Functional polymers

Functional polymers, containing diverse functional groups, serve critical roles in intrinsically soft fibers by enhancing performance while preserving the soft physical properties of the matrix (Figure 4i). These polymers are utilized either to support the performance of fillers that struggle to form percolation networks within the matrix or to impart specific functionalities intrinsic to the polymers themselves.

Poly(3,4-ethylenedioxythiophene) polystyrene sulfonate (PEDOT:PSS) is a widely employed conductive polymer known for its excellent electrical conductivity. PEDOT's performance is enhanced through P-doping,^[173,174] although the insulating nature of PSS can limit its overall conductivity. Techniques such as acid doping and solvent annealing have been developed to address this limitation by modifying conductive pathways and reducing the insulating effect of PSS. Through these optimizations, PEDOT:PSS enhances the conductivity of soft fibers while synergistically integrating with the matrix to maximize performance.

Polypyrrole (Ppy) exhibits conductivity when oxidized, owing to electron delocalization within its conjugated pyrrole ring backbone. While undoped Ppy is insulating, doping with chemical oxidants enhances conductivity without compromising the fiber's inherent softness.^[117,175] Similarly, polyaniline (PANI) provides tunable conductivity influenced by its oxidation state and pH. This property makes PANI particularly suitable for pH-sensitive applications and biosensors, where ionic conductivity is crucial for dynamic environments.^[104,176]

Polyethyleneimine (PEI), known for its high ionic conductivity, is commonly used for N-doping.^[104] By promoting ion movement within fibers, PEI significantly enhances conductivity while maintaining the matrix's flexibility and softness.

Additional functional polymers, such as poly(ethylene glycol) diacrylate (PEGDA) and polyvinylidene fluoride (PVDF), contribute unique physical and electrical properties to soft fibers. PEGDA, recognized for its high transparency in hydrogel form and exceptional mechanical properties, is ideal for optical applications.^[112, 115] In contrast, PVDF, with its high fluorine content, exhibits strong electron-attracting properties due to the electronegativity of fluorine. This characteristic makes PVDF highly effective for activating triboelectric and piezoelectric properties, broadening its application potential in advanced fiber technologies.^[105,174,177–180]

5.2.5 | Metal oxides

Metal oxides, through oxidation reactions, modify the physical and chemical properties of metals, introducing new characteristics that are not present in their original state (Figure 4j). These attributes make metal oxides valuable fillers for intrinsically soft fibers. By synthesizing metal oxides in various dimensions, their properties can be tailored to meet specific application needs. For instance, metal oxides can exhibit enhanced electrical conductivity, piezoelectric effects, and catalytic properties, making them highly advantageous in soft fiber fabrication.

Manganese dioxide (MnO_2), frequently employed as an electrical catalyst, is synthesized in NWs form to facilitate high electron storage. This property makes MnO_2 an effective filler, enhancing electrical conductivity and improving the energy storage efficiency of intrinsically soft fibers.^[181] In contrast, zinc oxide (ZnO) exhibits piezoelectric characteristics through oxidation, enabling fibers to convert mechanical energy into electrical signals. This property positions ZnO as an ideal filler for energy harvesting applications.^[106,107,174] Additionally, multi-metal oxides offer multifunctionality by combining the properties of their constituent metals. When mixed and oxidized, they retain individual metal functions while enhancing electrical conductivity. Recent studies have demonstrated the potential of multi-metal oxides to produce multifunctional fibers, further broadening the scope of intrinsically soft fiber applications.^[182,183]

6 | FABRICATION STRATEGIES FOR FIBER-TYPE SOFT BIOELECTRONICS

So far, we have explored various materials, from matrix materials to filler materials, used in the fabrication of fiber-type soft bioelectronics. There are several strategies for creating soft fibers using these materials, and in this section we will categorize and present these strategies under three main groups: spinning, coating, and others.

6.1 | Spinning

The spinning methods widely used for fabricating fiber-type soft bioelectronics are broadly classified into two main categories: melt-spinning and solution-spinning. Melt-spinning involves melting thermoplastic polymers to form continuous fibers by extrusion while solution-spinning uses polymer solutions to fabricate fibers. Melt-spinning can be further subdivided into extrusion spinning and direct spinning, while solution spinning

includes techniques such as dry-spinning, silk-spinning, blow-spinning, electrospinning, wet spinning, and alternating extrusion spinning.

6.1.1 | Melt-spinning

Melt-spinning is a process in which thermoplastic polymers are heated above their melting point to achieve a molten state and then extruded through a nozzle to form continuous filaments. The extruded molten material is solidified through rapid cooling, typically using cold air. This method is straightforward as it enables polymer processing without requiring additional solvents or chemical treatments. Furthermore, the characteristics of the fibers can be tailored by adjusting process parameters such as extrusion temperature, cooling rate, and draw ratio.^[184,185] Unlike other spinning techniques such as electrospinning or wet spinning, melt spinning allows for control of materials in their molten state, which makes it less sensitive to environmental factors. This flexibility enables the independent embedding of diverse functionalities through multiple channels, facilitating the simple fabrication of multi-functional fibers.

The melt-spinning method enables the integration of complex architectures—such as multi-core configurations or embedded microfluidic channels—within a single fiber by incorporating diverse functional materials. In this approach, thermoplastic polymers such as polycarbonate or conductive polymers such as polyethylene were used as core materials, while thermoplastic elastomers served as cladding materials. The preform, structured with these materials, was drawn through heating in a furnace to produce multi-functional fibers (Figure 5a). The fabricated fibers, with a diameter of 10 μm , can withstand strains of up to 500%. Additionally, these fibers integrate conductivity and optical properties, enabling multi-functional sensing capabilities for pressure, strain, and temperature.^[186] Also, other studies have introduced a melt-spinning method of fabricating high-performance piezoelectric fibers by incorporating functional nanomaterial such as barium-titanium oxide (BT) and reduced graphene oxide (rGO) into polyvinylidene fluoride (PVDF) matrices. These fibers, produced via twin-screw extrusion at 180–220°C with a 3 mm spinneret, exhibited a β -phase content of up to 52%, a diameter of ~170 μm , a tensile strength of 420 MPa, and an elongation at break of 141%. Various structural configurations, including coiled and knitted designs, were optimized for enhanced electromechanical performance. Coiled fibers withstood 10% strain, generating 1.3 V, while knitted structures achieved 4 V with a sensitivity of 10V N^{-1} , making them ideal for real-time physiological

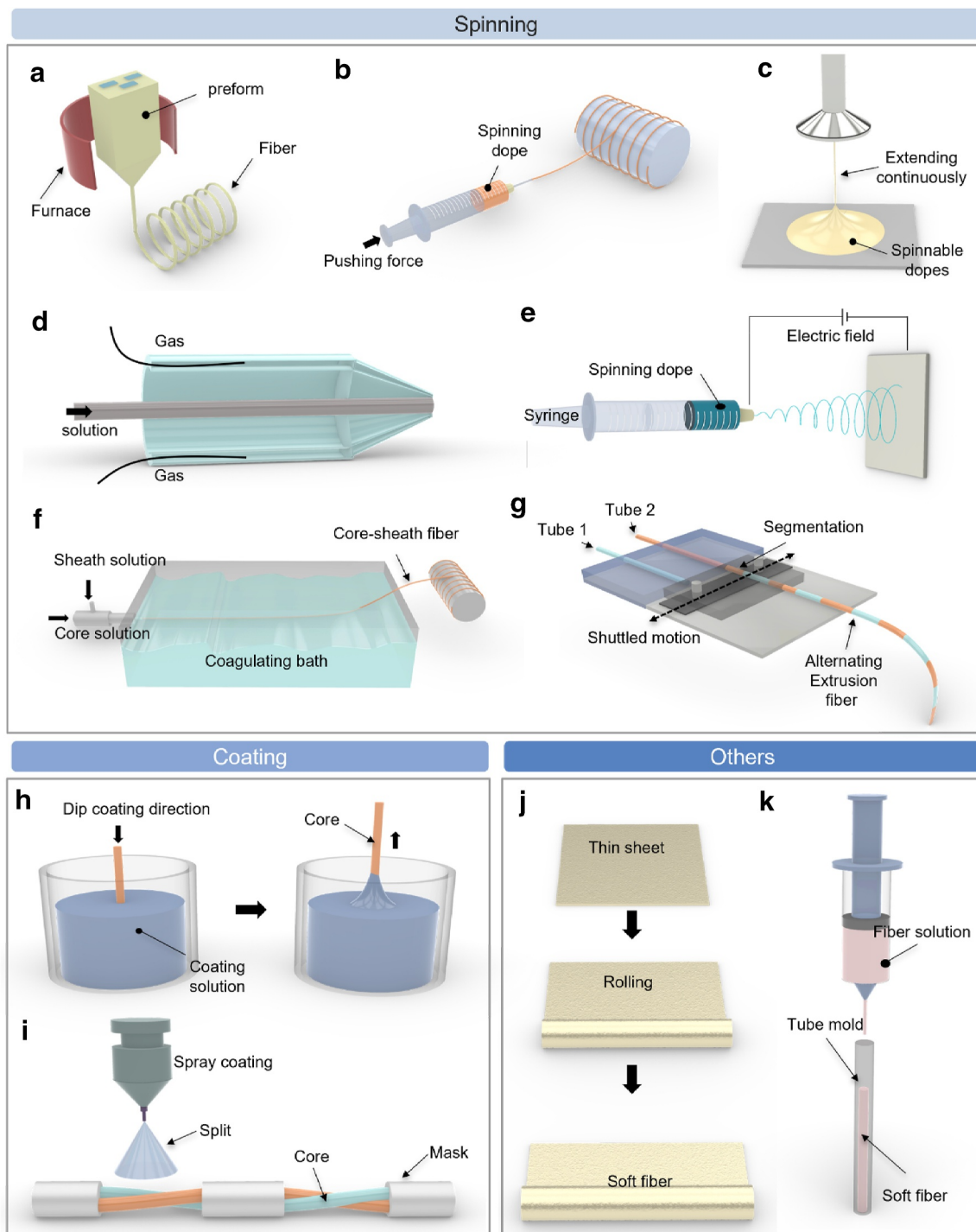


FIGURE 5 Various fabrication strategies for fiber-type soft bioelectronics. Techniques for creating and enhancing fiber-type soft bioelectronics can be categorized as follows: (a–g) Spinning techniques, including (a) melt spinning, (b) dry spinning, (c) silk-spinning, (d) blow spinning, (e) electrospinning, (f) wet spinning, and (g) alternating extrusion spinning; (h, i) coating techniques, such as (h) dip coating and (i) electrospray coating; (j, k) other techniques, including (j) rolling, and (k) molding.

monitoring. These melt-spun fibers hold promise for self-powered biomedical devices, such as wearable sensors, implantable biosensors, and energy-harvesting systems, offering a scalable and biocompatible solution for next-generation healthcare technologies.^[179]

6.1.2 | Dry-spinning

Dry-spinning is one of the methods in the solution-spinning category and is regarded as a simpler and more efficient process compared to other spinning

techniques. While most spinning methods require specially designed equipment, multi-step processes, and pre-or post-spinning treatments, dry-spinning allows for fiber fabrication with fewer additional chemical treatments (Figure 5b). In this method, a polymer solution is extruded through a syringe, and the solvent evaporates in ambient air at room temperature, forming a continuous fiber of solidified polymer. The dry spinning process also relies on phase transition to solidify fibers. Unlike other spinning methods, which often control the fiber diameter through stretching, dry spinning controls both fiber diameter and mechanical properties by adjusting the rotation speed of the bobbin (collector) during spinning, while simultaneously stretching the fiber.

A study introduced an energy-efficient dry spinning technique utilizing a polyacrylonitrile/dimethylformamide (DMF) solution mixed with AgNO_3 salt—referred to as PANSion—for fiber fabrication. This method employs non-solvent vapor-induced phase separation (NVIPS) to trigger a phase transition from liquid to solid, enabling the production of fibers with high stretchability (200–600%) and electrical functionality ($0.5\text{--}2\text{ S m}^{-1}$) without requiring additional processing. The resulting fibers had diameters of approximately $50\text{ }\mu\text{m}$. This study highlights a novel dry spinning method that significantly reduces energy consumption compared to conventional techniques, providing valuable insights into energy-saving spinning technology.^[151]

6.1.3 | Silk-spinning

Unlike dry-spinning, recent methods inspired by the silk-spinning process of spiders have been developed that utilize spontaneous phase separation to produce functional soft fibers. The silk-spinning process mimics the way spiders or silkworms secrete silk proteins to form silk fibers. By inducing phase separation of silk proteins from highly concentrated protein solutions in air at ambient temperature, this method enables the production of solid free-standing fibers with both strength and flexibility (Figure 5c). In contrast to dry-spinning, which requires high pressure and thermal heating, silk-spinning operates under atmospheric pressure and ambient temperature.^[187,188] This method facilitates the alignment and aggregation of protein molecules in the spinning solution, allowing for more energy-efficient fiber formation with greater process simplicity.

A recent study introduced a phase-separation-enabled ambient spinning approach inspired by the supramolecular assembly and phase transition of spider silk proteins. In this work, a dope solution—referred to as PANSion—was formulated by dissolving polyacrylonitrile and silver

ions in DMF, facilitating fiber formation under ambient conditions. The dope solution was exposed to air, triggering water-induced NVIPS, which transformed the dope into solid free-standing fibers. During this process, silver ions were reduced to AgNPs, which impart electrical conductivity to the fibers. This study eliminates the need for the high pressure and thermal heating often required in traditional spinning methods. The resulting fibers exhibit exceptional stretchability (more than 500%), mechanical strength (more than 6 MPa), and conductivity (approximately 1.82 S m^{-1}). The fibers, with diameters ranging from 50 to $200\text{ }\mu\text{m}$, demonstrate potential applications in sensing gloves and smart face masks, showcasing their versatility.^[152]

6.1.4 | Blow-spinning

Blow-spinning is a technique that utilizes a high-velocity gas stream, typically compressed air, to spin polymer solutions or melt into nanofibers. The high-pressure air accelerates and fragments the polymer solution near the nozzle, stretching it into fine threads, while the solvent rapidly evaporates, forming fibers (Figure 5d). Blow-spinning allows for higher production rates, providing an efficient fiber-spinning process.^[189] In comparison to dry-spinning, blow-spinning does not require high temperatures or high pressures; instead, it utilizes airflow to stretch polymer solutions or melts to create fibers, enabling faster production with reduced dependence on external energy sources.

A recent study demonstrated the use of a precursor solution composed of metal oxides, metal salts, and polyvinyl butyral dissolved in ethanol, which was ejected through an airbrush nozzle at 150 kPa, where the airflow facilitated its solidification into inorganic–polymer hybrid fibers. The resulting fiber exhibited diameters ranging from 70 to 300 nm. When integrated with elastic substrates such as SEBS, the thin-film transistors made from these fibers demonstrated durability through over 1000 bending cycles without performance degradation and withstood strains of up to 50%. The indium-gallium-zinc oxide fiber-type devices fabricated through this process showed potential for multi-functional sensing applications, including strain, pressure, temperature, gas, and ultraviolet (UV) detection.^[183]

6.1.5 | Electrospinning

Electrospinning is a unique fiber fabrication strategy that differs from other spinning strategies by utilizing an electric field rather than high pressure or thermal

heating. Unlike blow-spinning or dry-spinning, electrospinning does not require significant external pressure or heat, making it particularly advantageous for producing nanoscale fibers with ultrafine diameters. In electrospinning, a high voltage is applied between the nozzle of a syringe containing a polymer solution and a collector, generating an electric field. This electric field overcomes the surface tension of the polymer solution, forming a sharp Taylor cone at the nozzle tip. The polymer solution is ejected from the cone, and as it travels toward the collector the solvent evaporates or the molten polymer solidifies, resulting in the formation of fine fibers (Figure 5e).^[190] Electrospinning enables the production of ultra-fine fibers ranging from nanometers to micrometers in diameter. These fibers typically exhibit a random arrangement, characterized by high surface area and porosity. These characteristics make electrospun fibers ideal for various applications, including filters, scaffolds, and membranes. One notable application involves the creation of fiber membranes containing LM particles.

Researchers employed electrospinning to fabricate fiber membranes embedded with LM particles by injecting a mixture of TPU solution and LM particles through a syringe pump. Through a selective stamping method that applies pressure to specific areas of the membrane, semi-embedded LM particles rupture and release, forming high-resolution conductive pathways with minimum line widths of 50 μm . As a result, the LM nanomembrane fabricated through the electrospinning process and stamping technique exhibits high resolution and stability (retaining stability for over 30,000 cycles at 100% strain). This membrane demonstrates potential applications as a flexible sensor for collecting bioelectrical signals such as ECG and EMG.^[191]

6.1.6 | Wet-spinning

Wet-spinning is a fiber fabrication method that utilizes pressure and a coagulation bath to create fibers from polymer solutions, differing from other spinning techniques that rely on high temperatures or electric fields (Figure 5f). Unlike previously described spinning techniques such as melt-spinning, electrospinning, or dry-spinning, wet-spinning involves extruding a polymer solution through a pressurized nozzle into a coagulation bath, where the solvent is removed via diffusion or chemical reaction with the coagulation medium.^[192] This process forms solid fibers, which are then washed, dried, and further treated to achieve the desired mechanical or chemical properties.^[193,194]

An innovative technique based on light-triggered dynamic wet spinning was proposed for the continuous synthesis of core-sheath hydrogel optical fibers. In this technique, PEGDA-co-acrylamide (PEGDA-co-AAm)-type hydrogel was used as the core material, while Calcium alginate hydrogel served as the sheath material. Both materials were extruded under pressure, and the core and sheath materials underwent simultaneous crosslinking: the core material crosslinks via ionic exchange reactions in a CaCl_2 coagulation bath, and the sheath is crosslinked by UV photopolymerization. This results in continuous fiber formation. The hydrogel fibers produced, exhibiting diameters ranging from 70 to 1172 μm , exhibit excellent biocompatibility and a tissue-like Young's modulus (<2.60 MPa). These fibers show significant potential for applications such as deep-tissue cancer therapy and brain optogenetic stimulation, highlighting their versatility in deep-tissue therapeutic applications.^[114]

6.1.7 | Alternating extrusion spinning

Next, we introduce alternative extrusion, a modified processing technique developed to improve material properties, and precision, or enable unique structural designs compared to conventional extrusion. Conventional extrusion typically involves melting or dissolving a homogeneous material and continuously shaping it into fibers through a mold or die, with additional post-processing required to create functionally integrated structures. In contrast, alternative extrusion uses pressure to sequentially extrude multiple solutions into a core tube, creating continuous segments. By systematically or periodically adjusting parameters such as material flow or processing conditions during extrusion, this technique is highly effective for fabricating fibers with customized microstructures or complex shapes (Figure 5g).

An experimental approach demonstrated that alternative extrusion could be used to produce continuous p/n-type thermoelectric (TE) fibers, which were then woven into fabrics to investigate their multifunctional potential. The TE fibers were fabricated using two materials: SWCNTs and polyvinyl alcohol (PVA). By utilizing a programmed and computer-controlled extrusion-segment assembly line, the two materials were alternately extruded to produce continuous p/n-type TE fibers. The resulting TE fibers, with a diameter of 300 μm , exhibited high durability and flexibility, maintaining the TE performance even after more than 1000 bending cycles. Additionally, the fibers demonstrated high tensile strength (over 20 MPa), making them highly

suitable for use in flexible weaving applications. This work highlights the broad potential of multifunctional Fabric electronics fabricated through alternative extrusion.^[110]

6.2 | Coating

In the previous sections, we explored various strategies for fabricating soft and stretchable fibers using different spinning methods. In this section, we focus on coating methods, which primarily aim to impart diverse functionalities to fibers with elastic or soft cores. These coating processes enhance the electrical, mechanical, and chemical properties of the fiber, enabling the creation of fibers with specific capabilities, such as sensing, chemical reactivity, or improved durability. Coating methods can be categorized into dip-coating and electrospray coating. Dip-coating involves immersing the fiber into a solution, ensuring an even coating, which is a simple yet effective method. On the other hand, electrospray coating uses high voltage to spray fine particles onto the fiber, allowing precise and controlled coating. These techniques enable the fibers to acquire various functionalities tailored to specific applications.

6.2.1 | Dip-coating

Dip-coating is a process where a soft or stretchable core is immersed in a coating solution and then slowly withdrawn, allowing a uniform coating to form on the fiber's surface, driven by the solution's viscosity and surface tension (Figure 5h). The thickness of the coating can be precisely controlled by adjusting parameters like solution concentration, viscosity, and withdrawal speed. This technique enables the incorporation of various electrical and chemical properties into the fiber by using different core materials and coating solutions.

A research group introduced a solution-based dip-coating method to coat fibers with Bi-layer LM particles (BiLMP), a dual-layer structure that enhances durability, stretchability, and conductivity. They used a customized blade coating machine to dip-coat polyurethane (PU) fibers with polymer-attached LMPs and CNT-attached LMPs. This approach significantly improved the mechanical durability of the fibers, enabling them to withstand over 10,000 cycles at 100% strain, and enhanced conductivity ($\sim 2.5 \times 10^6 \text{ S m}^{-1}$). The resulting BiLMP fibers, with a diameter of 200 μm , showed significant potential for minimally invasive 1D bioelectronics

applications, such as brain activity monitoring electrodes, optogenetic neural probes, and peripheral nerve stimulators.^[83]

6.2.2 | Electrospray coating

Electrospray coating is a technique used for rapidly applying polymer solutions or NPs-type materials over large areas. It utilizes an electric field to spray coating materials, such as polymers, metal particles, or ceramics, through a nozzle onto the surface of a soft and stretchable core, forming a thin layer on the surface. As the solution reaches the surface, the solvent evaporates, solidifying the coating layer. Similar to electrospinning, the electric field forms a Taylor cone at the nozzle, causing the liquid to break into nm to μm size fine particles that create a uniform coating layer (Figure 5i). Coating thickness and uniformity can be controlled by adjusting parameters like solution concentration, viscosity, nozzle voltage, and spray distance.^[195] The method also allows selective coating of specific areas, which is a significant advantage.

A recent study used electrospray coating to selectively apply oleyamine doping on CNT fibers after p-hybridization with a PEDOT:PSS solution. This approach produced alternating p-n type patterns on the CNT fibers, which were then used to develop high-performance TE devices. The study demonstrated the potential for wearable TE generators with excellent power density (70 mW m^{-2}) and stretchability ($\sim 80\%$), showcasing the versatility of the electrospray method in functionalizing fiber for advanced applications.^[174]

6.3 | Others

6.3.1 | Rolling

In addition to traditional spinning processes and coating techniques, there are various methods for fiber fabrication using composite materials and advanced techniques. One such method is the rolling technique, which involves wrapping thin films of polymer or composite material around a needle with a specific diameter to form cylindrical fibers (Figure 5j). The rolling technique converts planar devices made of multiple functional thin films into compact cylindrical fiber structures, enhancing their implantability in the human body. This method is particularly advantageous for offering tubular structural support for nerve tissue regeneration,

making it suitable for implantable neural therapeutic applications.

Employing a rolling strategy, investigators developed a self-electrified biodegradable conduit aimed at advancing neural regeneration therapies. They assembled a multilayer structure comprising electrospun polycaprolactone (PCL) (thickness $\sim 350\ \mu\text{m}$), copolymer of poly(L-lactic acid) and poly(trimethylene carbonate) (PLLA-PTMC) (thickness, $\sim 300\ \mu\text{m}$), and magnesium (Mg; thickness, $3.5\ \mu\text{m}$) and iron-manganese alloy (FeMn; thickness, $1.5\ \mu\text{m}$) (Mg-FeMn) thin-film electrodes, which were then rolled around a needle with a diameter of $1.5\ \text{mm}$ to form a 3D cylindrical fiber structure. Subsequently, a thin layer of PCL (thickness: $\sim 20\ \mu\text{m}$) was electrospun over the rolled structure to enhance the stability and alignment of the cylindrical fiber. The resulting fiber, with a diameter of $1.5\ \text{mm}$, retained its functionality after over 1000 bending cycles at an angle of 180° . It demonstrated sufficient electrical conductivity ($3.87 \times 10^5\ \text{S m}^{-1}$) to promote neural regeneration. Furthermore, the 3D cylindrical fiber structure, with its structural stability and biodegradability, is advantageous for use as an implantable device. The continuous ES provided by the Mg-FeMn electrodes offers a therapeutic approach to damaged nerves, promoting regeneration and functional recovery. This innovative approach is highly regarded for its potential in neural regenerative medicine.^[196]

6.3.2 | Molding

Molding techniques offer a simpler and more efficient alternative to traditional fiber fabrication strategies, enabling the production of high-quality fibers with precise shape control and compatibility across a wide range of materials. Injection molding, for example, involves melting a plastic or molten material, injecting it into a mold, and then cooling it to solidify the material (Figure 5k). This method achieves precision up to $\pm 0.01\ \text{mm}$, allowing the creation of complex 3D shapes. It is highly versatile and can be applied to plastics, metals, ceramics, and both thermoplastic and thermosetting materials.

Researchers applied this method to produce elastomer core fibers by injecting a silicone rubber precursor into polyethylene tubes and curing them at room temperature for 4 h. The $2\ \text{mm}$ -diameter fibers maintained excellent structural integrity under 250% strain and survived more than 3500 bending cycles. Additionally, multifunctional fibers were fabricated by wrapping materials such as copper (Cu) wire, polyamide fiber, or ultra-high

molecular weight polyethylene fiber, which can be combined with other materials to develop stretchable functional materials or devices, opening new possibilities for research and practical applications.^[197]

7 | DESIGN STRATEGIES FOR TRANSFORMING SINGLE FIBER INTO YARN AND FABRIC

Fibers produced using the previously discussed strategies can be extended into yarns and fabrics through various design strategies (Figure 6). These strategies not only enhance the mechanical properties of the fibers but also enable multifunctionality by integrating individual fibers with varied capabilities. Furthermore, they allow for the detection of physiological signals over large areas. This section presents methods for transforming fibers into yarns and fabrics.

7.1 | Strategies for functional yarns

7.1.1 | Twisting

Yarns fabricated by integrating single fibers are typically produced through twisting and coating methods. Twisting transforms single fibers into yarn-like structures through the use of a rotating disk. In this method, fibers pass through a high-speed rotating disk, and the rotational force naturally twists the fibers into a uniform yarn structure (Figure 6a). This technique enables the creation of multi-functional fibers with a uniform twist, enabling properties such as conductivity, electrochemical sensitivity, piezoelectricity, and thermoelectricity without clumping. Additionally, the twisted structure enhances friction and bonding forces between fibers, enhancing tensile strength while preserving structural stability.

A research group demonstrated the twisting of multifunctional fibers into helical structures to enhance mechanical stability and functionality. The twisted structure increased the contact area, boosting electrical conductivity and energy storage capacity. By twisting highly flexible PVA fibers with PEDOT:PSS or CNT, they fabricated a temperature thread sensor (diameter: $\sim 200\ \mu\text{m}$) that maintained stable conductivity even under 300% strain. They also developed a thread battery (diameter: $\sim 500\ \mu\text{m}$) by twisting lithium cobalt oxide (LiCoO_2) with conductive carbon materials and an electroluminescent thread device (diameter: $\sim 400\ \mu\text{m}$) by twisting PVA, copper-doped Zinc Sulfide (ZnS:Cu), and AgNWs fibers. These applications highlight the potential

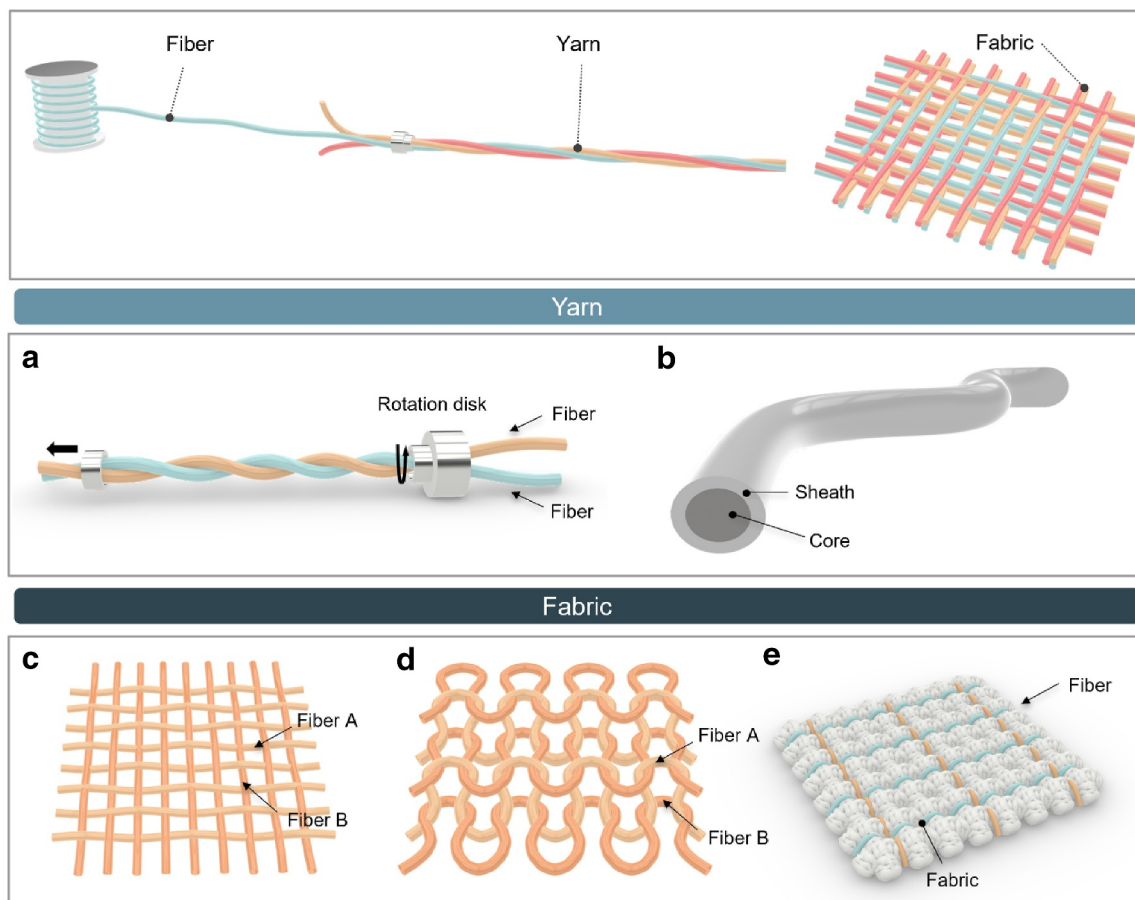


FIGURE 6 Design strategies for transforming a single fiber into yarn and fabric. (a) Twisting method. (b) Coating strategy. (c) Weaving method. (d) Knitting method. (e) Embroidering method.

of multifunctional fibers as flexible and wearable information-processing tools, paving the way for wearable fabrics with improved flexibility and usability.^[198]

7.1.2 | Coating

Coating techniques can impart multifunctionality to fibers by enhancing the surface properties of the fiber. The coating method discussed in Section 6.2 primarily pertains to single fiber-type device fabrication, where the process is optimized for achieving uniform coating on individual fibers to enhance their functional properties. On the other hand, the coating method described in this section applies to multi-fiber or fabric-based structures, where scalability and process efficiency are key considerations. This method allows for the addition of different functionalities, such as conductivity, biocompatibility, or chemical resistance, to the fiber core. By utilizing a core-sheath structure in the dip-coating process, a conductive material can be applied to the core, while the outer

sheath can provide protective or insulating properties. This design provides core stability and improves the overall performance of the fiber. The core-sheath configuration in the conductive fibers prevents deformation, ensuring reliable performance even under mechanical stress (Figure 6b).

A recent approach demonstrated the fabrication of LM-core, core-sheath microfibers via coaxial wet spinning, offering both high stretchability and stability for integration into self-powered wearable motion sensors. The process involved a dual-structured coaxial nozzle, where a LMs-containing PVDF-hexafluoropropylene-tetrafluoroethylene (PVDF-HFP-TFE) solution was injected through the inner nozzle to form the core, while PVDF-HFP-HFE/PEGDA solution was injected through the outer nozzle to form the sheath. The fibers were solidified in a coagulation bath. The resulting LM core-sheath microfibers (diameter: $\sim 270 \mu\text{m}$) exhibited strain-insensitivity, high conductivity ($\sim 4.35 \times 10^4 \text{ S m}^{-1}$), and exceptional stretchability ($\sim 1170\%$). These fibers demonstrated significant potential for applications in smart fabrics and

wearable sensors, underscoring their versatility and exceptional performance.^[199]

7.2 | Strategies for functional fabric

Fabrics made by weaving single fibers or yarns into a 2D structure provide a larger surface area, improving the sensitivity of electronics and supporting the integration of complex functional systems like smart fabrics and wearable devices. Additionally, these fabrics act as stable fiber matrix structures, facilitating the extension of stretchable interconnections. This section explores various techniques for fabricating fabrics by interconnecting single fibers or yarns into fabric forms using methods such as weaving, knitting, and embroidering.

7.2.1 | Weaving and knitting

Weaving and knitting are two fundamental techniques for fabricating fabrics from fibers, each offering distinct characteristics and advantages. While both methods involve interconnecting fibers to form fabric, the processes and resulting materials differ significantly. A research group developed various stretchable fabrics using weaving and knitting techniques with Cu wires.^[200] Weaving involves intersecting warp and weft threads at right angles, resulting in a structure that emphasizes stability and strength. This method forms a rigid structure, providing high durability, stability, and mechanical strength. However, woven fabrics typically have limited stretchability, with a maximum of approximately 67%, and are prone to structural damage under large mechanical deformations. For instance, in the study by Yun et al., a Jacquard machine was used to rotate the weaves by 45°, forming a diamond-shaped space to enhance stretchability (Figure 6c). Despite their limited stretch, woven fabrics are highly durable and commonly used in applications where structural integrity and stability are critical.

In contrast, knitting connects fibers in loop formations, enabling excellent stretchability at the intersections of the loop (Figure 6d). While knitting is more complex and results in a lower mechanical strength compared to weaving, it offers greater flexibility and stretchability. Knitted fabrics can typically stretch up to 60% and maintain their performance over 500 cycles without significant resistance loss. This process is especially useful in applications that require high flexibility and stretchability. Furthermore, these fabrics can be produced in extended lengths and areas, demonstrating their considerable potential for commercial applications.

7.2.2 | Embroidering

Embroidering is a manufacturing technique in which fibers or yarns are placed in specific patterns or structures on a fabric and secured in place. It involves stitching fibers or yarns onto a base fabric with needles and threads. Intricate patterns can be automatically created with computer-controlled embroidering machines, allowing for the fabrication of diverse shapes and intricate designs (Figure 6e). Researchers designed a core-sheath yarn-type electrochemical fabric sensor (CSSY-EFS) that enables efficient sweat capture and stable sensing, utilizing CNT-based fibers for the conductive core and cotton fibers for the hydrophilic sheath. The hydrophobic CSSY was embroidered onto a super-hydrophobic fabric, allowing selective sweat capture and detection in specific areas. This enabled the development of a real-time sweat monitoring system with reliable signal transmission. The study demonstrated that the embroidered sensor maintained high stability under various dynamic conditions, confirming its practical applicability as a wearable electrochemical sensor.^[201]

8 | APPLICATIONS

The diverse material and design strategies discussed earlier have enabled significant advancements in fiber-type wearable and implantable bioelectronics, providing novel solutions for body monitoring and therapeutic interventions. Wearable bioelectronics leverage fiber materials to create fabric-like structures that conform to the skin, enabling efficient and non-invasive measurement of biological signals as well as the delivery of therapies. In contrast, implantable bioelectronics utilize fibers in their standalone form, enabling minimally invasive integration into the body, and thereby mitigating inflammation and immune responses. This section explores recent advancements in fiber-type fabrics and standalone fibers, with a focus on their applications in monitoring biological signals and delivering targeted therapies. Tables 3 and 4 summarize the characteristics of the upcoming fiber-type bioelectronic applications research, divided into wearable and implantable applications.

8.1 | Advances in fabric-type soft wearable bioelectronics for health monitoring

Recent advancements in wearable technologies have been driven by the integration of soft and stretchable fibers and fabrics, enabling precise physiological

TABLE 3 Various types of fiber-type soft wearable bioelectronics.

Matrix	Filler	Fabrication	Performance	Ref.
TPU, EVA	AAm, AA	Electrospinning, Molding	Stretchability: ~100% Voltage response of 0.9 V at a curvature radius of 6 mm ⁻¹	[81]
CS/PEDOT, PCL	PVDF, PP	Electrospinning	Stretchability: ~100%	[202]
CPs (CP-1, CP-2, CP-3)	CP-1 fiber, TDAE, BMI-TSFI	Wet spinning	Conductivity: 76–173 S cm ⁻¹ Tensile strength >200 MPa	[203]
SBS	EGaln, oLM	Electrospinning	Electrical stability up to 1500% strain Air permeability: 177 mm s ⁻¹ Stable resistance over 1000 stretch-release cycles	[191]
Cotton fiber	CNT fiber, PANI, Pt, GOx, PEDOT:PSS	Embroidering	Sweat capture efficiency: 75–90% Minimum sweat volume required for sensing: 0.5 μ L Bending cyclic stability: 5000 bending	[201]
HA/Gel, PDMS, PI	Dopamine	Electrospinning	Tensile strength: 21.9–26.1 kPa Wound healing acceleration: 35% versus 25% (control)	[204]
PA6	DTAC	Electrospinning	Filtration efficiency: 99.95% Antibacterial efficiency: 99.9%	[205]
CFF	Bi ₂ Te ₃ NPs, H ₂ O ₂	Electrospinning, Dip coating	TENG output voltage: 25V, TENG output current: 1 μ A Bacterial inhibition: 87.3%	[88]
TPU fiber	CNTs, AgNPs, PDA	Wet spinning	Stretchability: 1468% Strain sensitivity: 2.3×10^6	[42]

Abbreviations: AA, Acrylic acid; AAm, Acrylamide; AgNPs, Silver nanoparticles; Bi₂Te₃ NPs, Bismuth Telluride nanoplates; BMI-TSFI, 1-butyl-3-methylimidazolium bis(trifluoromethylsulfonyl)imide; CFF, Chitosan-coated carbon fiber fabric; CNT, Carbon nanotube; CP-1, P(PzDPP-2FT); CP-2, P(gTDPP-2FT); CP-3, P(IgDPP-MeOT2); CPs, Conjugated polymers; CS/PEDOT, Chitosan/poly(3,4-ethylenedioxythiophene); DTAC, Dodecyltrimethylammonium chloride; EGaln, Eutectic gallium-indium; EVA, Ethylene-vinyl acetate; GOx, Glucose oxidase; HA/Gel, Hyaluronic acid/gel; oLM, Oxidized liquid metal; PANI, Polyaniline; PA6, Polyamide 6; PCL, Polycaprolactone; PDA, Polydopamine; PEDOT:PSS, Poly(3,4-ethylene dioxothiophene); polystyrene sulfonate; PI, Polyimide; PP, Polypropylene; Pt, Platinum; PVDF, Polyvinylidene fluoride; SBS, Poly(styrene-block-butadiene-block-styrene); TDAE, Tetrakis(dimethylamino)ethylene.

monitoring, therapeutic interventions, and real-time health management. These innovations have expanded the scope of bioelectronic devices and unlocked groundbreaking applications in health monitoring and therapeutic applications.

To enhance real-time sensitivity and adaptability, a recent study developed a triboelectric-effect-type microflexure-sensitive fiber inspired by the lamellar cell structure of the Meissner corpuscle in the skin. This design utilized TPU nanofibers twisted around hollow EVA fibers, forming high-specific-surface tribo-interfaces with enhanced mechanical responsiveness through thermal treatment (Figure 7a).^[81] To achieve this, TPU nanofibers were twisted and wrapped around pre-stretched hollow EVA fibers, enhancing the surface area and forming high-specific surface tribo-interfaces. The fiber structure's spacing was adjusted through thermal treatment to maximize mechanical responsiveness. Subsequently, the ionogel was injected into the fiber core

to improve its electrical signal responsiveness during flexure, enabling efficient mechanical deformation sensing via the triboelectric effect (Figure 7b). With high stretchability up to 100% and stable performance over 10,000 deformation cycles, this fiber was integrated into a cotton-blend fabric to create a breathable moisture-permeable biomechanical fabric (breathability: ~1380 mm s⁻¹, permeability: ~0.055 g cm⁻² h⁻¹) (Figure 7c). The Fabric effectively measured pulse signals and monitored muscle movements across various activities, such as walking and running. When worn on a patient's wrist, it delivers real-time feedback on muscle activity and force changes, highlighting its potential in personalized rehabilitation training (Figure 7d).

Extending this concept, a research group developed a triboelectric/piezoelectric hybrid nanogenerator (TP-hNG) system to address the limitations of conventional invasive electrical stimulation (iES) therapy. By generating bioinspired electrical signals synchronized with

TABLE 4 Various types of fiber-type soft implantable bioelectronics.

Matrix	Filler	Fabrication	Performance	Ref.
PU, Ecoflex	AuNPs	Dip-coating	~15.1 pF (under 10–20% strain)	[154]
SBS	Au-AgNWs, Pt black, EGaIn, IrO ₂ NPs	Molding	~10 Ω (under 150% strain)	[32]
PDMS, CNT fiber	CNT fiber, Ag/AgCl	CVD process, dip-coating	Accuracy on HVA ~97.8% Detection sensitivity: ~410 pA nM ⁻¹	[206]
CNT fiber	PtNP, GOx, PEDOT:PSS et al.	CVD process, dip-coating	Multifunctionality Spatial resolution: μ m scale arrangement Sensitivity: 0.84 μ A μ M ⁻¹ /5.6 nA μ M ⁻¹ /4.0 mV mM ⁻¹ (H ₂ O ₂ /glucose/Ca ²⁺)	[87]
SEBS	LIG, EGaIn, PEDOT:PSS, PANI	Rolling, molding, Photolithography	Detection limit: ~5.6 nM/~7.2 nM/~3.5 nM (dopamine/norepinephrine/epinephrine)	[14]
pAAM, PEGDA, alginate		Wet spinning	Optical attenuation: ~0.63 dB cm ⁻¹ at 515 nm	[114]
PVA	rGO	Molding	Optical attenuation: ~0.53 dB cm ⁻¹ at 589 nm Fatigue threshold: ~625 J m ⁻²	[84]
CNT fibers	CNT fibers, PANI, GOx, Pt, CNT, Nafion	CVD process, dip-coating	Power density: ~22.6 mW cm ⁻²	[207]
Polycarbonate, SEBS	Carbon-doped polyethylene	Thermal drawing	Multifunctionality	[33]
SEBS	pMMA, PANI, EGaIn, LIG, Nafion et al.	Rolling, Photolithography	Spatial resolution: ~40 μ m Detection limit: ~24 nM (serotonin)	[176]

Abbreviations: AgNW, Silver nanowire; AuNP, Silver nanoparticle; CNT, Carbon nanotube; CVD, Chemical vapor deposition; EGaIn, Eutectic Gallium-Indium; GOx, Glucose oxidase; HVA, Homovanillic acid; IrO₂ NP, Iridium dioxide nanoparticle; LIG, Laser induced graphene; pAAM, PolyAcrylamide; PANI, Polyaniline; PDMS, Polydimethylsiloxane; PEDOT:PSS, Poly(2,3 = dihydrothieno-1,4-dioxin)-poly(styrenesulfonate); PEGDA, Poly(ethylene glycol) diacrylate; pMMA, Polymethyl methacrylate; Pt black, Platinum black; PtNP, Platinum nanoparticle; PU, Polyurethane; PVA, Polyvinyl acid; rGO, Reduced graphene.

respiratory movements, the TP-hNG dynamically adjusted ES, greatly improving neural regeneration and functional recovery.^[202] The system employed PVDF and PP nanofibers with a soft PDMS spacer to form triboelectric layers, mimicking the elasticity of human skin up to 7 MPa (Figure 7e,f). Capable of generating electrical signals up to 3.44 V kPa⁻¹ during chest and diaphragm movements (Figure 7g), the device maintained signal stability over 36,000 cycles. Experiments in rats demonstrated the ability of TP-hNG to synchronize with the vagus and phrenic nerve signals, autonomously adjusting bio-iES signals to physiological activity levels, thereby promoting motor neuron stimulation and neural recovery (Figure 7h). This highlights its clinical potential as a self-adaptive stable-output therapeutic device.

In addition to these wearable systems, a recent study introduced flow-enhanced crystallization (FLEX) technology to fabricate semiconductor polymer fibers with enhanced mechanical and electrical performance. By utilizing low-concentration solutions, FLEX-aligned polymer chains form highly crystalline structures, resulting in fibers with up to 10 times higher tensile strength over 200 MPa and toughness over 80 MJ m⁻³

compared to conventional fibers (Figure 7i).^[203] The fibers were wet-spun in conjugated polymer coagulation baths and crystallized using the FLEX method resulting in enhanced crystalline structures. This resulted in previously unattainable electrical conductivity (~1.82 S m⁻¹), Young's modulus exceeding 100 GPa, tensile strength of approximately 200 MPa, and strain-enhanced electronic performance. These fibers, integrated into wearable fabrics, enabled self-powered devices, such as organic thermoelectric fibers (OTES) and OECT fibers, to harvest body heat or sense lactate and ECG signals (Figure 7j,k). The FLEX method demonstrated scalability and durability, unlocking new possibilities for advanced applications in wearable electronics.

A recent study further advanced the field of wearable bioelectronics by developing LM nanofiber membranes (LMNMs) designed for bioelectrical signal monitoring. These membranes combined electrospun TPU fibers with semi-embedded LM particles using a pressure-stamping technique, resulting in conductive pathways with a minimum line width of 50 μ m (Figure 7l).^[191] The LMNMs exhibited exceptional stretchability up to 400% and maintained resistance stability over 30,000 strain

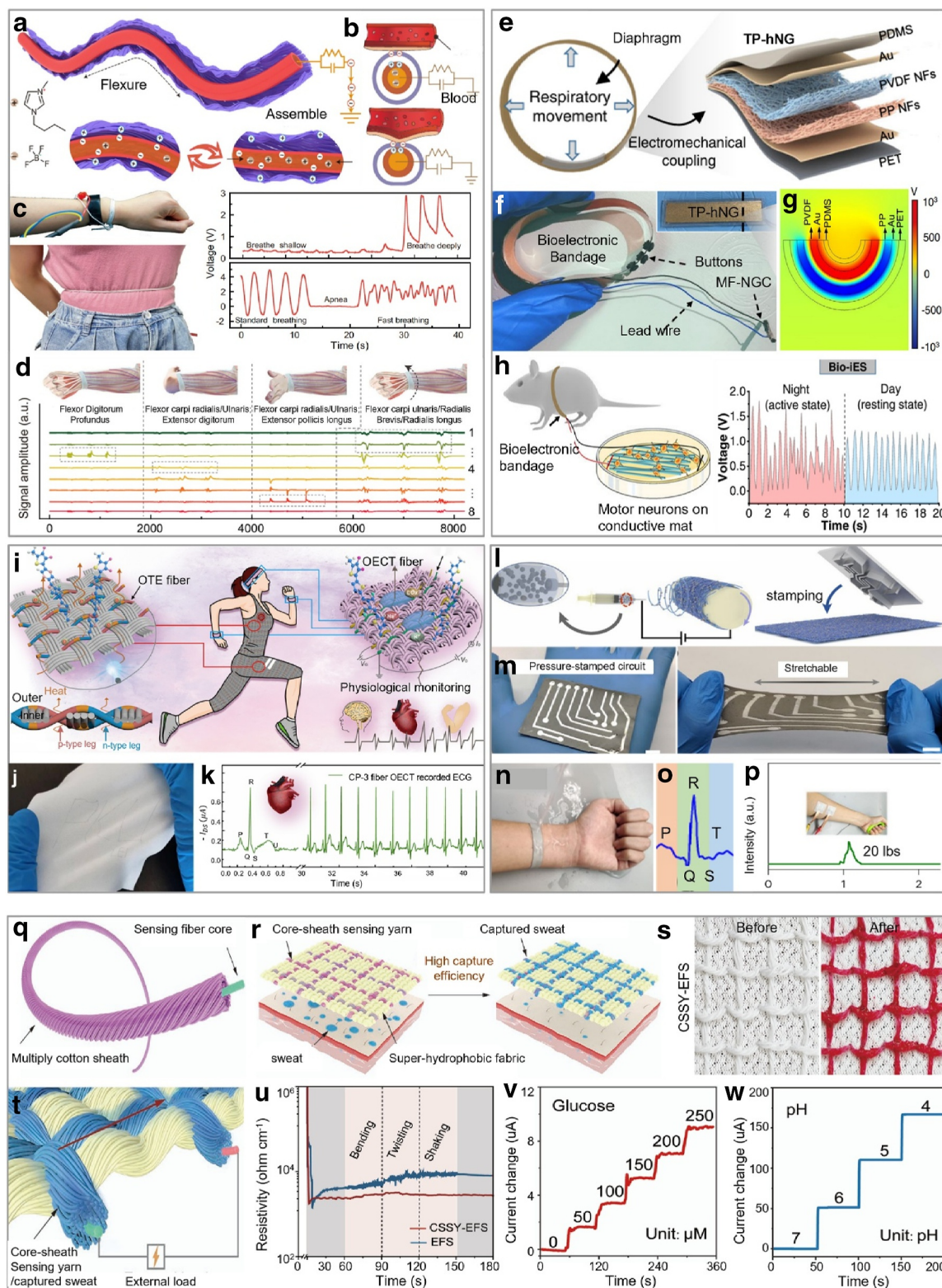


FIGURE 7 Legend on next page.

cycles (Figure 7m). Additionally, the fibers exhibited high moisture permeability ($\sim 2491 \text{ g m}^{-2} \text{ d}^{-1}$), ensuring excellent breathability and skin compatibility, while maintaining high electrical conductivity ($\sim 2.13 \times 10^6 \text{ S m}^{-1}$). Integrated into ECG and EMG sensors, the LMNMs demonstrated high signal accuracy even in artificial sweat environments, surpassing commercial electrodes in impedance and signal stability (Figure 7n,o). Their ability to map EMG signals, distinguish grip strength variations, highlighting their potential in advanced bioelectrical monitoring applications (Figure 7p).

As part of efforts toward molecular-level biosensing, the CSSY-EFS was introduced to simultaneously absorb sweat and measure glucose and pH concentrations.^[201] This system utilized CNT-based sensor fibers as the core, encased in multi-layer cotton sheaths to facilitate sweat absorption, and incorporated TiO_2 -coated substrates for superhydrophobic properties (Figure 7q). The yarn achieved a sweat capture efficiency of 75–90% exclusively at the hydrophilic sensing fibers due to the contrast in hydrophilic and hydrophobic properties between the yarn and substrate (Figure 7r,s). The super-hydrophobic cotton fabric substrate, coated with TiO_2 nanopowder, prevented sweat absorption, while the hydrophilic sensing yarn efficiently captured sweat, maintaining stable circuit connectivity with as little as $0.5 \mu\text{L}$ of sweat (Figure 7t). The functional sensing fiber served as the working electrode, while multi-ply cotton-wrapped CNT/Ag/AgCl fibers functioned as the reference electrode, providing stable and accurate sensing signals during dynamic body movements, such as bending, twisting, and shaking (Figure 7u). The electrochemical sensing performance of CSSY-EFS was evaluated with various functional fibers. For glucose-sensing fibers, sensitivity was enhanced by

electrodepositing polyaniline and PtNPs and coating the fiber surface with glucose oxidase. This allowed glucose molecules in the analyte to react, generating a current response linearly proportional to glucose concentration (Figure 7v). Similarly, the pH-sensing yarn demonstrated excellent linear correlation within a pH range from 4 to 7 (Figure 7w).

8.2 | Advances in fabric-type soft wearable bioelectronics for therapeutic applications

Recent advancements in material science have enabled the integration of soft and stretchable fibers or fabrics into wearable devices, offering superior comfort and seamless adaptability to the dynamic movements of the human body. These innovations have been widely applied across various fields, particularly in the development of highly flexible and conductive fibers capable of monitoring physiological signals, delivering therapeutic interventions, and administering drugs. Among these applications, advances in photomedicine technology have garnered significant attention for its role in enabling safe and non-invasive wound healing therapies.

An integrated system of electrospun nanofiber wound dressings and light-emitting diodes (LEDs) was introduced as a novel strategy to accelerate early wound healing.^[204] Their study introduced hyaluronic acid (HA)/Gel-based electrospun nano fabric membranes designed to manage severe wounds with high exudate levels. Gelatin, derived from porcine skin, was dissolved in 1,1,1,3,3,3-hexafluoro-2-propanol and processed into gelatin nanofibers through electrospinning. The fibers

FIGURE 7 Advances in fabric-type wearable bioelectronics for health monitoring. (a, b) The triboelectric mechanism of electron-ion conduction (a) in nanofiber buckling (NB)-fiber. (b) For pulse detection is presented. (c) Photograph of the sensor for pulse and respiration monitoring (left), and respiratory waveforms in different states (right). (d) The strength of electromyography (EMG) signals during different gestures. Reproduced (Adapted) with permission.^[181] Copyright 2024, Springer Nature. (e) Schematic illustration of the working principle for converting respiratory movement into real-time bio-invasive electrical stimulation (Bio-iES) signals. (f) Photograph of the fabricated bioelectronic bandage. (g) FEA analysis of potential generated in a triboelectric/piezoelectric hybrid nanogenerator (TP-hNG). (h) Schematic illustration of Bio-iES stimulating motor neurons in vitro and waveform of Bio-iES obtained from SD rats. Reproduced (Adapted) with permission.^[202] Copyright 2022, Springer Nature. (i) Schematic illustration of semiconducting polymer fibers. (j) Photograph of fabricated woven fabric. (k) Real-time recorded electrocardiography (ECG) signals from CP-3 fiber organic electrochemical transistors (OECTs). Reproduced (Adapted) with permission.^[203] Copyright 2024, AAAS. (l) Fabricating process of liquid metal nanomembrane (LMNM) through electrospinning and pressure-stamping. (m) Photographs showing pressure-stamped circuit on LMNM, unstretched and stretched. (n) LMNM electrode when the subject was in a wet condition. (o) ECG collected with LMNM. (p) EMG signals were generated by holding grip rings and collected by LMNM. Reproduced (Adapted) with permission.^[191] Copyright 2024, Springer Nature. (q) Core-sheath sensing yarn structure of a multi-ply cotton sheath and sensing fiber core. (r) Schematic illustration of the sweat capture process using a core-sheath sensing yarn-type electrochemical fabric sensor (CSSY-EFS). (s) Photographs of a $4 \times 4 \text{ cm}$ CSSY-EFS before and after the capture of artificial sweat. (t) Sweat sensing mechanism of CSSY-EFS. (u) The resistance changes of CSSY-EFS and EFS under different formations after capturing sweat. (v) Chronoamperometric response plot of the CSSY-EFS to glucose. (w) Open-circuit potential response plots of the CSSY-EFS to pH. Reproduced (Adapted) with permission.^[201] Copyright 2022, Wiley-VCH.

were then cross-linked using a solution of ethanol containing HA and 1-ethyl-3-(3-dimethyl aminopropyl) carbodiimide/N-hydroxysuccinimide/dopamine, to improve biocompatibility, tissue adhesion, and structural stability (Figure 8a). These HA/Gel-based membranes effectively mimicked the extracellular matrix to support tissue regeneration and demonstrated excellent wound healing performance (Figure 8b). In excisional wound models in

rats, the membranes markedly reduced the wound area and accelerated collagen synthesis, achieving complete wound closure within 2 weeks. Notably, dopamine-functionalized D-HA/Gel membranes outperformed untreated controls, reducing wound area by approximately 35% compared to ~25% in the control group after 7 days (Figure 8c,d). Furthermore, the integration of a flexible multi-LED array under near-infrared (NIR) irradiation

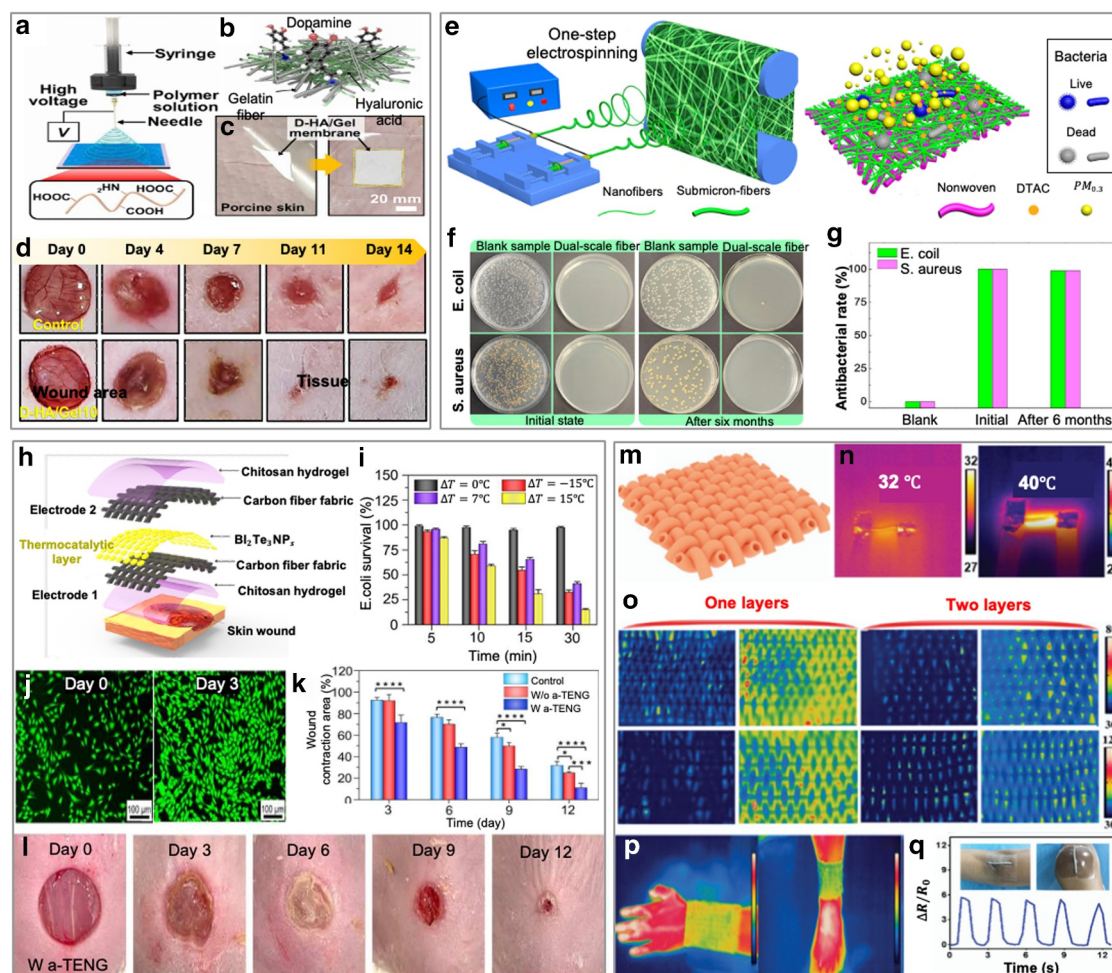


FIGURE 8 Advances in fabric-type soft wearable bioelectronics for therapeutic applications. (a) Preparation process of dopamine-modified HA (D-HA)/Gel nanofibrous mesh by electrospinning. (b) Structure of synthesized D-HA/Gel nanofibrous mesh. (c) Photographs of the D-HA/Gel membrane on the injured rat skin. (d) Sequential photographs of each wound site after 14 days. Reproduced (Adapted) with permission.^[204] Copyright 2022, AAAS. (e) Fabrication process and structural characteristics of dual-scale fibrous filter. (f) Photographs of quantitative antibacterial activity images. (g) Antibacterial rate of blank sample and dual-scale fiber filter. Reproduced (Adapted) with permission.^[205] Copyright 2024, Springer Nature. (h) Illustration of the wound dressing with Bi₂Te₃ nanoplates (NPs) the thermocatalytic layer, chitosan hydrogel, and carbon fiber fabrics (CFFs) layer. (i) Antibacterial efficiency of Bi₂Te₃ NPs for *Escherichia coli*. (j) Fluorescence images of fibroblasts after electrical stimulation (ES) from an arch-shaped TENG (a-TENG) recorded on 0 and 3 days. (k) Wound contraction area of wounds taken on 0, 3, 6, 9, and 12 days for the control, w/o a-TENG, and w/a-TENG groups. (l) Digital photographs of wounds taken on 0, 3, 6, 9, and 12 days for the w/a-TENG group. Reproduced (Adapted) with permission.^[88] Copyright 2023, AAAS. (m) Illustration of thermoplastic polyurethane (TPU) fabric-type soft bioelectronics for thermal therapeutic applications. (n) Thermal infrared image of AgNPs/polydopamine/CNT-TPU/TPU conductive composite fibers (APTTF) at 25°C and under a voltage of 2 V for 18 s. (o) Double-layer of TPU fabric heated at 80–120°C for 0 min (images in the left column) and 5 min (images in the right column), respectively. (p) Thermal infrared image of fabrics on the wrist, ankle, and knee at 25°C. (q) Application for recording human elbow bending motion. Reproduced (Adapted) with permission.^[42] Copyright 2020, RSC Publishing.

enhanced tensile strength ($\sim 9.44 \pm 0.74$ N), tissue bonding, and structural recovery, while minimizing scar tissue formation. These results demonstrated the potential of nanofibrous LED-integrated membranes as advanced wound healing platforms.

Beyond wound healing, nanofiber membranes have gained attention for antibacterial applications due to their favorable properties such as biocompatibility, biodegradability, and mechanical stability. A recent study introduced an ultrathin, dual-scale fibrous network that offers superior antibacterial efficacy, thermal regulation, and moisture management.^[205] Using a dual-needle electrospinning technique, polyamide 6 (PA6) fibers were incorporated with dodecyltrimethylammonium chloride (DTAC), an antibacterial ionic surfactant, to enhance hydrophilicity and antibacterial performance. Under a high-voltage electric field, DTAC-induced ionic-dipole interactions created jet splitting, forming a dual-scale fibrous network (Figure 8e). This network, with a thickness of ~ 1.49 μm and weight of ~ 0.57 g m^{-2} , reduced raw material usage by over 97% compared to commercial N95 masks, while achieving 99.95% PM0.3 filtration efficiency and superior water vapor transmission rates (~ 7541 $\text{g m}^{-2} \text{ day}^{-1}$). DTAC within the fibers disrupted bacterial cell membranes, ensuring sustained antibacterial effects. Experimental results showed that *Escherichia coli* and *Staphylococcus aureus* were completely eradicated on the PA6/DTAC-2 filter, which maintained over 99% antibacterial efficiency after 6 months of storage, establishing it as a sustainable alternative to conventional mask filters (Figure 8f,g).

Building on multifunctional design strategies, a team of researchers developed a self-powered wound dressing that integrates antibacterial functionality with self-generated ES to accelerate the wound-healing process.^[88] This wearable platform utilized bismuth telluride (Bi_2Te_3) nanoflakes, carbon fiber fabrics (CFFs), and chitosan hydrogel offering tailored treatment for various biological responses. Bi_2Te_3 nanoflakes, applied to the bottom CFF layer functioned as thermocatalysts to generate hydrogen peroxide (H_2O_2) under temperature gradients, disrupting bacterial cell membranes and reducing survival rates of *E. coli* and *S. aureus* to 12.7% and 18.7%, respectively (Figure 8h). This process accelerated wound healing while showing enhanced bacterial inhibition with increased temperature gradients (Figure 8i). The TENG system, which utilized Bi_2Te_3 -coated CFFs as electrodes, generated an output voltage of 25 V at 1 Hz, promoting fibroblast proliferation by 2.1 times and migration speed by 3.2 times. In mice wound models, the combined system achieved approximately 89% wound contraction within 12 days, achieving a 58%

improvement compared to controls (Figure 8j-l). This innovative dressing highlights the potential of combining TE materials with ES for non-invasive wound healing.

Thermal therapy, another significant application of soft and stretchable fibers, has been explored for tissue regeneration, inflammation reduction, and muscle rehabilitation. To advance thermal therapy applications, a novel class of TPU fibers with hollow-porous architecture—modeled after human skin and animal fur—was introduced, demonstrating superior insulation and multifunctional performance.^[42] Wet spinning techniques were used to fabricate conductive composite fibers (simplified as APTTF) by incorporating AgNPs/polydopamine/CNT fillers (Figure 8m). These fibers demonstrated exceptional stretchability ($\sim 1468\%$) and a thermal conductivity that is 3.8 times lower than that of solid TPU, rendering them ideal for wearable fabrics. When integrated into woven fabrics, these fibers enabled efficient electrical heating at voltages as low as 2 V, achieving thermal equilibrium within 18 s (Figure 8n). Additionally, the thermal insulation performance was evaluated by heating the woven fabric at 80°C and 120°C for 5 min. Thermal results showed that one-layer and two-layer fabrics retained insulation temperatures of approximately 40.6°C and 55.2°C under heating conditions (Figure 8o). Additionally, these fibers demonstrated excellent strain and pressure sensing capabilities, enabling precise motion detection, particularly at human joints, where capacitance changes correspond to different weights and positions (Figure 8p).

8.3 | Advances in fiber-type soft implantable bioelectronics for physiological signal monitoring

Implantable devices are essential for capturing signals from body organs that are challenging to monitor using wearable devices, including localized electrical signals and changes in metabolite concentrations. To achieve effective implantation, soft bioelectronic devices require enhanced biocompatibility and a wide range of elastic modulus to match the diverse properties of tissues, ranging from soft brain tissue (~ 0.5 kPa) to rigid bone (~ 10 GPa) (Figure 2c). Recent advancements in implantable fibers have enabled real-time monitoring of strain, bioelectrical signals, and biomolecules, highlighting their suitability for long-term biomedical applications.

One study developed a strain-sensing fiber integrated with electromagnetic induction for wireless signal transmission for monitoring mechanical strain during tissue recovery. The fiber, fabricated by depositing

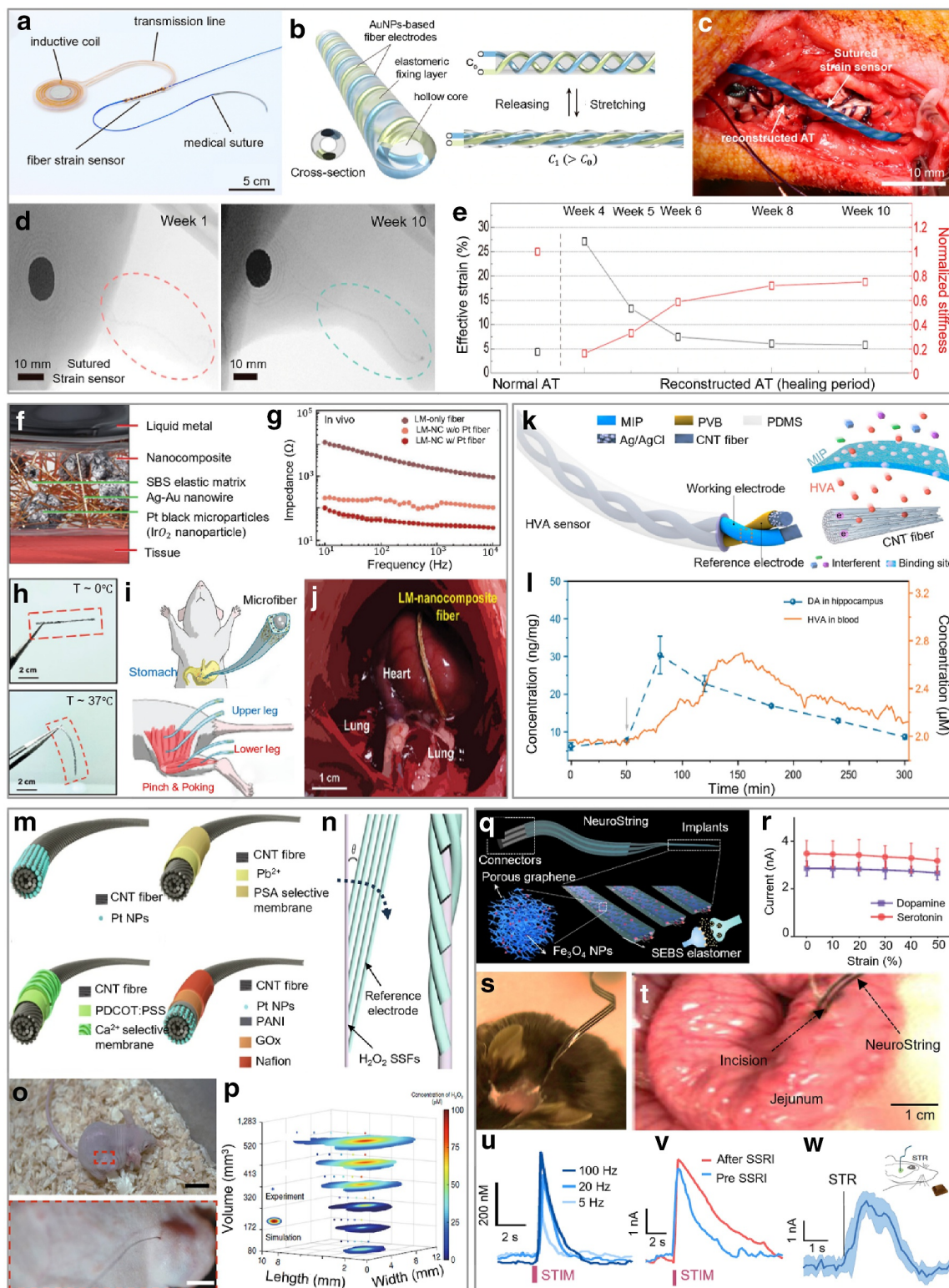


FIGURE 9 Legend on next page.

AuNPs onto PU, which was arranged in a double-helix structure around a hollow core, was designed to reduce deformation under tensile strain (Figure 9a).^[154] This configuration allowed capacitance changes to be measured accurately without significant fiber distortion (Figure 9b). The fiber exhibited high sensitivity (~ 15.1 pF in a 10–20% strain range) and excellent durability, enduring more than 10,000 stretching cycles at 10% strain. Additionally, saline surface treatment reduced tissue friction by $\sim 71.6\%$, minimizing damage. When applied to an injured porcine Achilles tendon, the fiber monitored strain changes during joint movement and remained fixed for 10 weeks, reliably transmitting wireless data. It tracked the tendon's stiffness recovery to $\sim 75\%$ of its normal levels during healing, demonstrating its utility for long-term rehabilitation monitoring (Figure 9c–e).

Strain monitoring is critical, but bioelectrical signals such as EMG, ECG, and EGG are equally important for understanding organ functionality. To measure these electrical signals, a research group introduced a frozen-injection fiber with enhanced impedance. This hollow fiber incorporated Au-coated AgNWs for stability, SBS for flexibility, and Pt black for low impedance (Figure 9f).^[32] The fiber maintained low impedance even under 150% tensile strain (Figure 9g). To facilitate minimally invasive implantation, the fiber was frozen via spray cooling, temporarily raising its modulus to 145 MPa for insertion through an 18-gauge needle (Figure 9h). The fiber successfully recorded electrical signals from the stomach, heart, and thigh muscles, demonstrating its versatility. For example, in gastric applications, it recorded rhythm changes in response to dietary stimuli and ES. In cardiac monitoring, it monitored heart rate changes induced by norepinephrine and diltiazem injections, showcasing its ability to provide accurate bioelectrical measurements in various organ systems (Figure 9i,j).

Beyond mechanical and electrical signal monitoring, real-time detection of metabolites offers essential insights into disease progression and therapeutic responses. To enable this, a fiber-type sensor was developed using molecularly imprinted polymer (MIP) techniques for detecting homovanillic acid (HVA), a dopamine metabolite and biomarker for CNS diseases. The fiber was fabricated by polymerizing an MIP layer onto CNT fibers, twisting them with a reference electrode, and coating them with PDMS for biocompatibility (Figure 9k).^[206] The sensor exhibited high specificity of up to 12.6-folds compared to other metabolites, and accuracy of up to 97.8% for HVA, even in complex environments containing analogs. Implanted into the hepatic portal vein of rats, the fiber successfully monitored changes in HVA concentration over 4 weeks and correlated them with the time delay in the increase in blood HVA levels caused by levodopa-induced dopamine metabolism (Figure 9l). The results demonstrated its potential for real-time monitoring of dopamine-related disorders.

To meet the need for multifunctionality, working electrodes using CNT fibers with specific surface treatments tailored for sensing different substances were designed. These electrodes were then arranged around a reference fiber to create a yarn structure.^[87] The reference electrode was prepared by depositing Ag/AgCl onto CNT fibers via cyclic voltammetry, using CNT fibers produced via the CVD process. For H_2O_2 sensing, fibers were prepared by depositing PtNPs on CNT fibers through the electrochemical double potential method. Similarly, glucose-sensing fibers were created by first depositing a PANI layer through chronoamperometry, followed by PtNPs deposition, as used in the H_2O_2 fibers. Then, a chitosan/SWCNTs/glucose oxidase (GOx) solution was applied via dip coating, followed by a Nafion coating, which was subsequently cross-linked. For cation sensing, fibers were coated with a PEDOT:PSS layer and

FIGURE 9 Advances in fiber-type soft implantable bioelectronics for physiological signal monitoring. (a) Schematic illustration of helical strain sensing fiber and inductive coil for wireless sensing. (b) Helical structure of strain sensing fiber. (c) Photograph of the fiber implanted on Achilles tendon of a porcine model. (d) C-Arm images displaying the stable position of the fiber for 10 weeks. (e) Change of mechanical stiffness and effective strain during reconstruction. Reproduced (Adapted) with permission.^[154] Copyright 2024, ACS Publication. (f) Schematic illustration of freezable low-impedance fiber consisting of nanocomposite and liquid metals. (g) Impedance changes by tensile strain and frequency (in vivo). (h) Photographs of fiber according to temperature (0°C , 37°C). (i, j) Implantable gastric, muscular, and cardiac applications of fiber in mice model. Reproduced (Adapted) with permission.^[32] Copyright 2024, Wiley-VCH. (k) Schematic illustration of homovanillic acid (HVA) sensing fiber and selective sensing of molecularly imprinted polymer. (l) Time difference of levodopa and HVA concentration by the fiber. Reproduced (Adapted) with permission.^[206] Copyright 2024, ACS Publication. (m) Schematic illustration of various single fibers for sensing H_2O_2 , prostate-specific antigen, cations, and glucose. (n) Fabrication process by twisting multiple sensing fibers for spatial sensing of H_2O_2 . (o) Application on tumor cells in mice model. (p) H_2O_2 concentration according to tumor cell size. Reproduced (Adapted) with permission.^[87] Copyright 2020, Springer Nature. (q) Schematic illustration of “NeuroString.” (r) Stable of monoamine oxidation current peak according to strain. (s, t) Application on brain of mice and gut of porcine model. (u) Dopamine concentration after optic stimulation. (v) Serotonin concentration after selective serotonin reuptake inhibitor drug treatment in mice model and (w) feeding chocolate. Reproduced (Adapted) with permission.^[14] Copyright 2022, Springer Nature.

dip-coated in a cationic cocktail before drying. Prostate-specific antigen (PSA) sensing fibers were fabricated by dip-coating CNT fibers in a solution containing chitosan-Pb₂[Fe(CN)₆]-poly(diallyldimethylammonium chloride)-graphene oxide (Figure 9m). These fibers were spun around a reference fiber and coated with PDMS to form multifunctional yarns. In an *in vivo* experiment, some H₂O₂ sensing fibers were arranged around the reference electrode to construct a yarn, which was then implanted into solid tumors in mice. This setup enabled real-time monitoring of the spatial distribution of H₂O₂ in tumors of varying sizes (80–1283 mm³) (Figure 9n,o). The researchers found that as tumor size increased, the distribution of H₂O₂ expanded as concentrations increased from 16 to 98 μM at the tumor center (Figure 9p).

Finally, to overcome the mechanical mismatch between CNT fibers and soft tissues, NeuroString, which consists of a laser-induced graphene (LIG)-metal nanoparticle-based fiber capable of measuring monoamines by utilizing its catalytic properties to facilitate monoamine oxidation, was developed.^[14] The researchers utilized metalloporphyrin to synthesize iron-containing metal oxide NPs, which act as catalysts for monoamine oxidation. These NPs were deposited onto a LIG layer, which was subsequently coated with a soft SEBS layer. The fiber was then laser-cut and dip-coated with pullulan to facilitate its implantation in biological tissues. This fabrication approach maintained the monoamine oxidation functionality while ensuring tissue compatibility (Figure 9q). The resulting fiber exhibited exceptional physical and electrochemical sensing capabilities, such as a breaking strain exceeding 1700% and minimal oxidation voltage degradation up to 10% under 50% strain (Figure 9r). It demonstrated low detection limits for the monoamine family, including dopamine (~5.6 nM), norepinephrine (~7.2 nM), serotonin (~3.5 nM), and epinephrine (~6.6 nM) (Figure 9s,t). To validate its functionality, the fiber was assessed *in vivo* in both rats and pigs. First, the fiber was implanted in the brain of AAV9-DAT-Cre transgenic mice to measure dopamine level changes upon optical stimulation of dopamine neurons and during a Pavlovian reward learning task. This experiment successfully measured dopamine increases in response to optical stimulation and reward, confirming stable signal acquisition (Figure 9u). Next, using SERT-Cre mice, the researchers measured serotonin level changes following optical stimulation of the DRN and observed increased serotonin expression after SSRI drug treatment (Figure 9v). Further experiments in rat and pig models focused on serotonin monitoring under various conditions, including drug treatment, inflammatory intestinal disease, and intestinal motility. Notably, in rats, the fiber enabled simultaneous

measurement of brain catecholamine concentrations and intestinal serotonin levels after chocolate feeding. The results showed a delayed increase in the gut's serotonin concentration (30–60 min post-feeding) after a rise in brain catecholamine levels, illustrating the temporal dynamics of brain-gut neurotransmitter interactions (Figure 9w).

8.4 | Advances in fiber-type soft implantable bioelectronics for therapeutic applications

Soft fiber electronics demonstrate significant potential in measuring various signals, from electrical and electrochemical signals to optical signals. Expanding upon these applications, implantable devices that utilize electrical and optical signals for therapy are gaining attention, notably for delivering precise stimuli and treatments to target tissues.

Researchers engineered implantable optical fibers using biocompatible hydrogels to enable both optical signal transmission and photothermal therapeutic applications. The fibers were fabricated by wet spinning a composite of alginate, PEGDA, and polyacrylamide (pAAM), with core and cladding layers formed through controlled polymerization processes. The alginate formed a cladding layer upon polymerization with calcium, while the PEGDA-pAAM solution was polymerized under UV light to form the core (Figure 10a).^[114] The resulting fibers exhibited a low Young's modulus of up to 2.60 MPa and low optical attenuation of up to 0.63 dB cm⁻¹, making them suitable for brain and deep tissue applications. Their efficacy was validated through two *in vivo* experiments. First, in a deep tissue photothermal therapy model, the fibers were implanted near cancer cells in mice and irradiated with NIR light for 15 min. The localized temperature increased to 48°C within 4 min, reducing tumor size while sparing surrounding tissues (Figure 10b,c). Second, in a PV-ChR2 mice model, the fibers were inserted into the M1 brain region to transmit optical signals. Light stimulation reduced the mice's movement, and the fibers showed lower expression of post-traumatic Glial fibrillary acidic protein and Immunoglobulin G compared to commercial silica optical fibers, confirming superior biocompatibility (Figure 10d). This study highlighted the potential of hydrogel-based optical fibers for safe and effective photothermal therapy and optical signal transmission.

To address the challenges of fatigue deformation in optical fibers caused by repetitive movements, optimized hydrogel optical fibers by adjusting the nanocrystalline structure of PVA hydrogel for brain-to-sciatic nerve

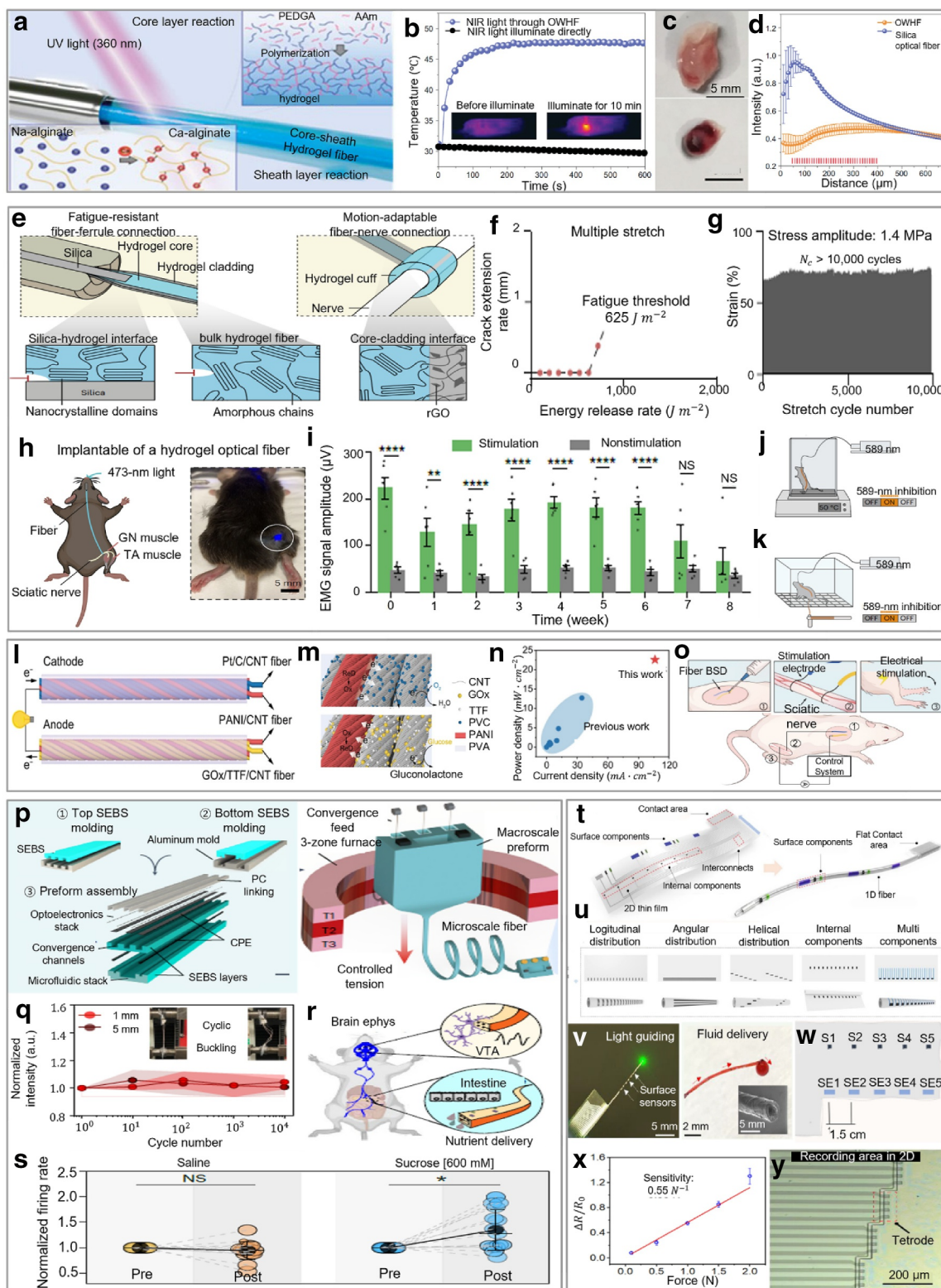


FIGURE 10 Legend on next page.

connections were introduced. They achieved fibers with enhanced fatigue resistance and optical performance. The cladding region, rich in nanocrystalline structures, offered strength and high refractive index, while the core region ensured flexibility by reducing nanocrystalline content. Reduced GO was added to enhance mechanical stability, and a cuff structure was added for sciatic nerve anchoring (Figure 10e).^[84] The optimized fibers exhibited low optical attenuation ($\sim 0.53 \text{ dB cm}^{-1}$ at 589 nm), high fatigue threshold of $\sim 625 \text{ J m}^{-2}$, and retained deformation capabilities at 70% strain over 10,000 cycles (Figure 10f,g). Two experiments validated their performance. First, optical signals transmitted from the brain to the sciatic nerve in a Thy1:ChR2-EYFP mouse model induced muscle contraction, maintaining functionality over 6 weeks (Figure 10h,i). Second, in a TRPV1:NpHR-EYFP mouse model, optical stimulation delayed pain responses, demonstrating effective signal transmission for therapeutic applications (Figure 10j,k). These results confirmed the fiber's durability and potential for long-term phototherapy in highly deformable body regions.

A bio-supercapacitor fiber utilizing glucose oxidation for ES was developed as an energy-harvesting solution for implantable medical devices. CNT fibers functionalized with PANI served as supercapacitor electrodes. The anode was designed with GOx to catalyze glucose oxidation. At the same time, the cathode used Pt/CNT/Nafion to enhance electron transfer, forming a yarn structure insulated with a PVA coating (Figure 10l,m).^[207] This fiber delivered a power density of $\sim 22.6 \text{ mW cm}^{-2}$ (Figure 10n), exhibited stability over ~ 8460 pulse discharges, and maintained performance during 2000 friction and bending cycles. In vivo experiments demonstrated its efficacy;

fibers implanted subcutaneously in rats generated energy from glucose oxidation and applied synchronized ES to the sciatic nerve, inducing muscle contractions without external power sources (Figure 10o). This study emphasized the potential of glucose-powered implantable fibers for continuous ES during long recovery processes.

Advancing multifunctional design strategies, one study developed thermally drawn fibers capable of simultaneously transmitting optical and electrical signals, as well as delivering drugs, between the brain and gut. Polycarbonate-based preforms containing carbon-doped polyethylene and SEBS were melt-spinning to create fibers with microfluidic channels for drug and nutrient delivery alongside optical and electrical pathways (Figure 10p).^[33] The fibers exhibited stability under 10,000 cyclic buckling tests and sustained drug delivery even at 90° bending (Figure 10q). In a Cck:ChR2 rat model, these fibers enabled light stimulation and nutrient delivery to the gut. Light stimulation using μLEDs from the brain to the gut activated anorexia-inducing cells in the intestine. In terms of nutrients, sucrose delivery from the brain to the gut increased dopamine-related neural activity in the VTA region, highlighting the gut's role in influencing brain signaling (Figure 10r,s). This study demonstrated the potential of multifunctional fibers in exploring gut-brain connections.

Finally, highly soft and multifunctional fibers capable of single-cell resolution using a rolling technique and soft lithography are developed. SEBS substrates were patterned to include electrodes for optical signal transmission (Poly(methyl methacrylate) (PMMA)), fluidic channels (Polypropylene glycol (PPG)-PU Urea (PUU)),

FIGURE 10 Advances in fiber-type soft implantable bioelectronics for therapeutic applications. (a) Schematic illustration of ultra-violet radiation polymerization of alginate, poly(ethylene glycol) diacrylate-polymethyl methacrylate hydrogel fiber by wet spinning process. (b) Fast temperature increases for photothermal therapy. (c) Cancer cell size decreases by photothermal therapy. (d) Immunoglobulin immunostaining of fiber compared to conventional silica optical fiber. Reproduced (Adapted) with permission.^[114] Copyright 2021, Oxford University Press. (e) Schematic illustration of polyvinyl acid (PVA) hydrogel fiber manipulated nanocrystalline structure. (f) Crack extension rate according to the energy release rate of PVA hydrogel fiber during cyclic peeling test. (g) Maximum strain during cyclic tensile test. (h) Implantation of PVA hydrogel fiber in mice model. (i) Intensity of EMG signals during implantation of fiber in mice model with and without optogenetic stimulation. (j, k) Illustration of the hot-plate test, of von Frey test in pain latency of TRPV1:NpHR-EYFP transgenic mice model. Reproduced (Adapted) with permission.^[84] Copyright 2023, Springer Nature. (l) Schematic illustration of bio supercapacitor fiber. (m) Principle of bio supercapacitor fiber. (n) Power and current density of bio supercapacitor fiber. (o) Schematic illustration of implantation of bio supercapacitor fiber. Reproduced (Adapted) with permission.^[207] Copyright 2023, Wiley-VCH. (p) Schematic illustration of soft gut fiber by thermal drawing process. (q) Light intensity during a cyclic test of soft gut fiber. (r) Schematic illustration of brain ventral tegmental area electrophysiology during nutrition delivery on the duodenal region. (s) Dopamine firing rate according to intraduodenal sucrose treatment compared with saline. Reproduced (Adapted) with permission.^[33] Copyright 2023, Springer Nature. (t) Schematic illustration of “spiral NeuroString.” (u) 2D styrene-ethylene/butylene styrene (SEBS) film with various patterning designs and fiber structures after the rolling process. (v) Photographs of light guiding and fluid delivery of “spiral NeuroString” with hollow structure. (w) Illustration of patterned “spiral NeuroString” for application on the gut of porcine model. (x) Pressure sensitivity of “spiral NeuroString” on the gut of porcine model. (y) Microscopy of patterned recording area for the brain of mice model. Reproduced (Adapted) under terms of the CC-BY license.^[176] Copyright 2023, the authors.

pH sensing (PANI), conductivity (EGaIn), pressure sensing (LIG), serotonin sensing (Nafion), and ES (PEDOT:PSS) (Figure 10t).^[176] The resulting fibers, with electrode densities of up to 150 sensors mm⁻², exhibited a low Young's modulus of up to ~5 MPa and maintained functionality under biological strain. Soft multifunctional fibers fabricated using the rolling technique have the potential to create various functional morphologies depending on the pattern and rolling direction (Figure 10u). In porcine intestinal models, the fibers monitored serotonin levels, induced intestinal movement via ES, and recorded pressure changes, even following SSRI administration (Figure 10v,w). The resulting fiber has a low detection limit of 24 nM for serotonin and high-pressure sensitivity (Figure 10x). The fiber was used for measuring serotonin inside the porcine small intestine, inducing small intestinal movement through ES, and measuring pressure. This demonstrated an increase in the concentration of serotonin, which inhibited the reuptake of serotonin after SSRI drug delivery and measured changes in small intestinal movement in response to ES. In rat brain models, fibers with 40 μ m spatial resolution recorded neuronal responses to treadmill activity at the single-cell level (Figure 10y). This approach demonstrated the versatility of multifunctional fibers for simultaneous measurement and stimulation across biological systems.

9 | CONCLUSION AND PERSPECTIVE

Fiber-based soft bioelectronics offer a mechanically compliant and biologically integrated platform for long-term operation in dynamic physiological environments. Their inherent softness, stretchability, and dimensional scalability enable precise interfaces with curved and moving tissues, providing a basis for continuous sensing, localized stimulation, and on-demand therapeutic intervention.^[208] Integration of multiple functionalities into a single fiber, and further assembly into textile-scale systems, has extended their applicability across wearable,^[209] implantable, and regenerative healthcare technologies.

Realizing the full potential of these systems requires addressing several interrelated challenges. Material platforms must satisfy simultaneous demands for electrical conductivity, mechanical durability, and biocompatibility over extended timeframes. While conventional conductive polymers and nanomaterials provide partial solutions, their performance under chronic exposure remains limited. Current efforts focus on developing bioresorbable conductors,^[85] ion-conductive hydrogels,^[210] and composite systems that maintain functionality in biological environments. Mechanical fixation and stable adhesion to tissue are critical for implantable systems. Existing

adhesive strategies show limited performance under wet and dynamic conditions. Bioinspired interfacial structures, stimulus-responsive adhesion, and micro-structured anchoring approaches are being explored to improve long-term interfacing without inducing damage. System-level integration remains a bottleneck. Power modules, signal processors, and communication circuits must be incorporated without compromising the mechanical and biological compatibility of the fiber platform. This includes the development of stretchable energy harvesters, soft supercapacitors, and wireless systems optimized for fiber geometries and textile architectures. Deployment and delivery methods are essential for clinical translation. Despite their softness, fiber devices require precise and minimally invasive insertion techniques. Temporary stiffening strategies, bioresorbable insertion guides,^[211] and catheter-assisted placement^[212] are under investigation to ensure positional accuracy and safety during implantation. Long-term functionality must be maintained under physiological and environmental stress. Repeated mechanical loading, exposure to biofluids, and external factors such as washing or temperature fluctuations can degrade performance. Encapsulation strategies, fatigue-resistant materials, and self-healing structures are being developed to extend the operational lifetime.

Fiber-type soft bioelectronics represent a modular and scalable framework for multimodal biointerfaces. Continued progress in material development, device engineering, and system integration will enable their transition from laboratory prototypes to clinically relevant tools. With interdisciplinary efforts, these technologies are expected to support the next generation of bioelectronic systems for health monitoring, disease intervention, and tissue modulation.

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CONFLICT OF INTEREST STATEMENT

The authors declare no conflicts of interest.

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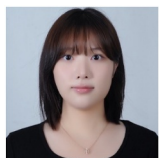
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