Recent Advances in Electronic Skin: Performance Requirements, Substrate Engineering, and Multimodal Sensing

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Abstract: Electronic skin (e-skin) is a flexible, biomimetic electronic system that mimics the tactile and physiological sensing of human skin by using distributed sensor arrays to capture external stimuli and convert them into electrical signals. Modern e-skins have evolved from single-mode devices to sustainable, multimodal platforms for health monitoring, prosthetics and rehabilitation, robotics and human–machine interaction, wearable devices, and brain–computer interfaces. This paper reviews e-skin technologies across four dimensions: (1) key performance requirements—sensitivity, flexibility/stretchability, stability/durability, and cost/processability—and the material and structural factors governing them; (2) the role of substrates, including how modulus, flexibility, surface morphology, interfacial adhesion, and thermal/environmental stability influence device performance; (3) the operating principles, characteristics, advantages, limitations, and application suitability of piezoresistive, capacitive, piezoelectric, and ionogel-based sensors; and (4) the significance, mechanisms, and challenges of multimodal sensing. Overall, appropriate material selection and microstructural design enable e-skins to balance performance and cost. Different sensing mechanisms offer distinct benefits and constraints in terms of manufacturability, linearity, power consumption, self-powering capability, and stretchability. Meanwhile, multimodal e-skins integrate complementary sensing mechanisms with decoupled signal outputs to detect multiple stimuli simultaneously. Future advancements will focus on developing intelligent, adaptable, and sustainable next-generation e-skin systems.

1. Introduction

Electronic skins (e-skins) are flexible, bio-inspired systems designed to replicate the tactile and physiological sensing capabilities of human skin and achieve seamless integration with dynamic biological surfaces. [1] Inspired by the hierarchical receptor network of natural skin, e-skins use distributed sensor arrays to receive external stimuli and convert them into electrical signals, typically by embedding skin-like integrated sensors within stretchable films formed by dispersing conductive fillers in elastomer matrices. [2] [3, 4]

Early e-skins were mostly single-mode, but advances in multimodal integration now allow simultaneous recognition and partial decoupling of multiple stimuli, expanding applications beyond basic tactile sensing. [3] [5] In healthcare, multimodal e-skins enable real-time, non-invasive monitoring of physiological and biochemical markers for personalised medicine and long-term tracking. [1, 2, 6-9] In surgery and prosthetics, they enhance robotic tactile perception and provide sensory feedback for artificial limbs and rehabilitation. [5] They also support human-like haptics in robotics, HMIs, and emerging VR systems. [5, 10] In wearable technologies, e-skins facilitate gesture recognition, motion tracking, and on-body communication, and—combined with AI and edge computing—extend to environmental and affective sensing. [1-3, 8, 11] Increasing integration in brain–machine interfaces positions e-skins as tactile bridges that relay sensory feedback and interface with neural signals, advancing human–machine interaction. [10] [1]

To meet these growing demands, material development now emphasises flexibility, skin compatibility, and biodegradability. [1] Natural biodegradable polymers are increasingly incorporated

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to maintain sensing performance, which can improve skin safety and end-of-life profiles, accelerating the development of green, sustainable e-skin systems. [6] At the performance level, molecular engineering of conductive polymers with hydrogen-bonding motifs enhances toughness, self-healing, and charge transport, enabling prolonged wear and potential implantation. [12] The incorporation of carbon nanomaterial composites enhances both the sensitivity and durability of e-skins, and it can be integrated with triboelectric, piezoelectric, and thermoelectric nanogenerators, thereby enabling self-powered, multifunctional e-skins. [8, 11]

2. Key Performance Characteristics of Electronic Skin

The key performance characteristics of e-skin encompass sensitivity, stability and durability, flexibility and stretchability, cost and manufacturability, all of which depend on material selection and structural design. The mechanisms of formation and the influencing factors for each characteristic will be discussed below.

2.1 Sensitivity

In e-skins, sensitivity is the ratio of a small change in the sensor's output to a slight change in the input. For piezoresistive pressure sensing, it is measured as the slope of the relative signal change with respect to pressure:

$$S = \frac{\delta(\Delta R / R_0)}{\delta P} \text{ or } S = \frac{\delta(\Delta I / I_0)}{\delta P} \tag{1}$$

For strain sensing, the gauge factor (GF) relates relative resistance change to mechanical strain:

$$GF = \frac{(\Delta R / R_0)}{\varepsilon} \tag{2}$$

Here, S is sensitivity, GF is gauge factor, P is the applied pressure, R and I are the resistance and current measured under pressure P, while R_0 and I_0 are the initial values at zero or reference pressure. The performance is specified by the response range. [13]

The sensitivity of e-skins is governed by interfacial contact resistance and the architecture of the conductive network. Interfacial contact resistance governs the response. For example, piezoresistive pressure sensors have sensitivity that depends on the rate of resistance change under load. Under the same force, the variation in contact resistance generally exceeds that in intrinsic resistance, thereby enhancing sensitivity. [14, 15] In stretchable strain sensors, deformation causes changes in the conductive network structure that lead to variations in the resistance signal. A random or disordered network extends the strain range but decreases sensitivity and causes more hysteresis. Therefore, enhancing sensitivity depends on densifying or organising the conductive pathways, such as through filler loading. [5]

To achieve high sensitivity, material choices that enhance contact-resistance modulation and stabilise or bridge the network are preferred. CNTs — characterised by adjustable wall numbers, sp^2-bonded frameworks, high strength, chemical stability, and excellent conductivity — can be solution-processed onto flexible or stretchable substrates to connect and strengthen conductive pathways. [9] For example, Luo et al. exploited the Wiesenberger effect by directly depositing GNPs/MWCNTs onto PDMS to form an effective percolation network. MWCNTs bridge adjacent graphene nanoplatelets, reinforce conductive pathways, and increase filler contact under compression, producing a rapid resistance drop and high sensitivity. The sensor reaches ~15 MPa^{-1} below 65 kPa and up to 81.2 MPa^{-1} near 100 kPa. They also showed that MWCNTs prevent GNP aggregation during heating and enhance conductivity by enlarging the nanofiller contact area. [16]

2.2 Stability & Durability

The stability of e-skins refers to their ability to maintain consistent performance under extended use, repeated deformation, environmental changes, or complex human interactions. [17, 18] It is governed by the chemical inertness of the matrix, filler dispersion, and interfacial adhesion.

Using Yi Zhao's sandwich-structured CTESS as an example, Ecoflex encapsulation reduces baseline drift and environmental interference, while strong CB–TPU interfacial adhesion and Ecoflex infiltration prevent network delamination during stretching, enhancing durability. Ultrasonication ensures uniform filler dispersion and a reversible 3D conductive network, lowering stress concentration and hysteresis. As a result, the device maintains highly stable output over 5,000 cycles at 40% strain. [19] By contrast, silver nanowires (AgNWs) exhibit the lowest sheet resistance and high light transmittance, yet their stability is susceptible to oxidation, ultraviolet–ozone exposure, sweat corrosion, and contact degradation. However, their durability and long-term stability can be significantly enhanced by soldering at connection points or by employing dense encapsulation and barrier-layer designs. [20] PEDOT: PSS avoids metal corrosion and exhibits excellent flexibility, but its stability is compromised by interface issues arising from its hydrophilicity and the acidity present in PSS. But, moisture barrier, crosslinking and neutralisation can effectively reduce humidity-related drift and enhance durability. [21]

2.3 Flexibility & Stretchability

Flexibility in e-skin refers to maintaining conductivity under bending, stretching, folding, or compression, and is evaluated by ε_max, gauge factor, linearity, hysteresis, response/recovery time, and cycling durability. These metrics depend on mechanics and microstructure—specifically, a low-modulus substrate and a reconfigurable conductive network that can reversibly deform and reconnect rather than fracture.

For instance, for choosing material, pairing a low-modulus PDMS substrate with a reconfigurable CNT-foam network enables highly flexible, stretchable, and durable e-skin. In Dai et al.'s design, a CNT-coated porous sponge embedded between fingerprint-shaped electrodes on an ultrathin PDMS film forms a percolating network that compresses and recovers reversibly, giving stable signals through ~70% strain and ~3000 compression cycles. The ultrathin PDMS layer also prevents metal-mesh fracture and CNT detachment, maintaining pathway continuity. [22]

2.4 Cost & Processability

Cost refers to raw materials and preparation expenses across synthesis and device fabrication. Processability indicates whether it can be efficiently integrated into common large-scale processes. Both are crucial for scaling e-skin technologies beyond laboratory prototypes.

Printed electronics offer a rapid, low-cost route to fabricating flexible devices. AgNW-based transparent conductive films are compatible with gravure, screen, and inkjet printing, offering excellent conductivity, though their high material cost limits large-scale adoption. [23] [24] In contrast, carbon-based materials such as graphene, GO→rGO, and CNTs are more economical and can be produced through CVD, exfoliation, oxidation–reduction, and film-forming methods (dip/spin-coating, spray-coating, filtration, wet-spinning), making them highly suitable for printed, large-area electronics. [25, 26] [27] [28] To further promote sustainable manufacturing, natural materials like cellulose are gaining attention; cellulose substrates combine flexibility, robustness, biocompatibility, and tunable structure with good processability, enabling stable multimodal sensing for wearable health monitoring. [29]

3. Introduction of Substrate Role

Wearable flexible sensors typically consist of a skin-contact substrate and a signal-processing module. To ensure long-term comfort, the substrate must provide good mechanical properties, biocompatibility, elasticity, breathability, and clothing-like softness for intimate skin contact. [30] Their overall performance depends on substrate modulus, microstructure, interfacial adhesion, and thermal and environmental stability, while processability and sustainability determine suitability for large-scale fabrication. These factors are detailed in the following sections.

3.1 Modulus and Flexibility

The flexibility and stretchability of electronic skins depend on the mechanical softness of the substrate material. This is because soft elastomeric materials can absorb mechanical strain during external deformation, inducing reversible non-planar deformation in rigid conductive layers. This maintains the continuity of the conductive network, preventing delamination at the interface. [31] For instance, PDMS has a Young's modulus of approximately 360–870 kPa—much lower than rigid plastics like polyimide or parylene (~3.2 GPa)—which reduces the overall bending stiffness of the stack, allowing a larger portion of the imposed deformation to be absorbed by the compliant substrate. [32] Combined with structural designs such as serpentine interconnects or island–bridge layouts, flexible substrates effectively mitigate failure from modulus mismatch while maintaining stretchability and durability.[33]

3.2 Surface Morphology & Adhesion

In flexible piezoresistive and capacitive sensors, interfacial coupling between the conductive layer and soft substrate governs contact-area evolution and stress transfer. When the conductive film conforms tightly and uniformly to the soft substrate, the electrical interface becomes stable: variations in contact resistance are minimised, signal lag and drift are reduced, and the readout becomes more reliable.

Microstructuring soft substrates strengthens interfacial coupling by lowering the effective modulus and creating reversible micro-voids that linearise the pressure-dependent evolution of real contact area—thereby enhancing sensitivity while mitigating viscoelastic lag. [34] As demonstrated across multiple studies, microstructured PDMS surfaces elastically store and release energy, improving stress transfer and stabilising the electrical interface. For instance, PDMS patterned with random-height micropyramids enables rapid resistance change at low pressures when only the pyramid tips engage; as the load increases, more features come into contact, yielding a smooth, near-linear response with stable cycling. [35] Beyond morphological tuning, chemical and molecular interactions at the filler–substrate interface are crucial for interfacial stability. In Yi Zhao's sandwich-structured strain sensor, ultrasonication drives carbon-black (CB) nanoparticles to anchor onto the adhesive TPU substrate, forming a dense conductive network with hydrogen-bond-enhanced filler–matrix coupling. [19] This strengthened bonding enables reversible deformation without delamination, ensuring stable stress transfer and reliable signals during repeated loading, thereby improving overall stability and durability.

3.3 Thermal and Environmental Stability

The thermal stability of substrate materials is crucial for the reliable operation of electronic skin, as fabrication and usage processes often involve temperature variations that generate thermal stresses. Substrates must therefore maintain dimensional accuracy to prevent CTE-mismatch-induced warping or delamination. [36] For example, Polyimide (PI) substrates, with excellent heat resistance, enable both high-temperature processing and harsh-environment operation, whereas most conventional polymers tolerate <200 °C, limiting fabrication flexibility. [37] [38] Beyond this, selecting substrate materials requires balancing the flexibility of organic polymers against the thermal and dimensional stability of inorganic substrates. While polymer substrates offer mechanical compliance and resilience to processing stresses, they typically exhibit higher oxygen and moisture permeability and greater thermal shrinkage than glass substrates. [36]

Moisture management is equally important, as humidity can degrade sensor performance, reduce sensitivity, and irritate the skin. Surface chemical or structural modifications can tune a substrate's water affinity, improving its resistance to absorption or permeation. Many polymer substrates are intrinsically hydrophobic, though PDMS still allows slow diffusion of water vapour. [36] But in fact, permeability genuinely supports the long-term use of electronic skin. Because the polymer substrate exhibits time-dependent properties, it gradually deforms or degrades. Moderate water vapour permeability therefore prevents sweat and water vapour from becoming trapped between the skin and the device. This both slows down interface ageing and improves wearing comfort. [38] For applications demanding stronger barriers, hydrophobic elastomers such as poly(styrene-isoprene-styrene) (SIS) offer superior moisture blocking. [39] Generally, the thermal and environmental requirements of e-skin substrates call for integrated material and structural design. For instance, Tan et al. designed a multilayer TPU composite integrating a thermally conductive BNNS topcoat with a porous, breathable TPU substrate to achieve both rapid heat dissipation and skin comfort. [40] These representative architectures confirm that polymer–BNNS systems can provide concurrent thermal regulation and moisture management for wearable devices.

3.4 Cost and sustainability

The processability of substrate materials depends on their compatibility with scalable fabrication techniques. PDMS supports soft moulding and conformal patterning, whereas Parylene, PI, and PET enable thin-film deposition and large-area fabrication; elastomeric, photolithography-compatible substrates further improve micrometre-scale patterning, ensuring the manufacturability and scalability of flexible e-skin devices. [41] Natural biopolymers such as cellulose and silk fibroin offer solution processability, biocompatibility, and biodegradability, making them promising for sustainable e-skins. [42] Cellulose, in particular, is compatible with scalable techniques including spray and gravure printing, microfluidic spinning, electrospinning, and 3D printing, and supports the fabrication of flexible electrodes and energy devices. [43] From a green-manufacturing perspective, cellulose enables aqueous, low-temperature, solvent-lean processing, while recyclable dissolution–regeneration routes using ionic liquids or alkali/urea systems further align with biodegradability and circularity goals. [44] Collectively, these scalable, solution-based techniques enable flexible, manufacturable, and sustainable e-skin systems.

4. Overview of Sensing Mechanisms

Understanding and classifying sensing mechanisms is vital because each relies on a distinct physical transduction principle that determines how external stimuli are converted into electrical signals. Such categorisation clarifies the underlying physics and influences a sensor's operating mode, signal type, power demand, and environmental robustness, providing a systematic basis for comparing advantages, limitations, and application suitability in flexible e-skin systems. The following sections, therefore, discuss the major mechanisms—piezoresistive, capacitive, piezoelectric, ionogel-based, and multimodal—to outline their principles, strengths, and challenges.

4.1 Piezoresistive mechanism

The piezoresistive effect is the change in electrical resistance under mechanical stress or strain, commonly quantified by the gauge factor (GF):

$$\mathrm{GF} = \frac{\Delta R/R}{\varepsilon} = \frac{\Delta\rho/\rho}{\varepsilon} + (1 + 2\nu) \quad (3)$$

where R, ρ, and ν are the material's resistance, resistivity, and Poisson's ratio, respectively. [45]

The piezoresistive response depends on the material's dominant mechanism: resistance may change due to pressure-dependent intrinsic resistivity or geometry variations, as described by $R = \rho L/A$. [46] In metals, resistivity is nearly constant, so resistance changes arise mainly from axial elongation and lateral contraction. [47] In semiconductors, stress alters band structure, modulating carrier concentration and mobility and thus intrinsic resistivity. [45] In conductive composites, deformation reorganises filler networks—filler spacing, contact state, and tunnelling resistance shift relative to the percolation threshold—leading to pronounced resistance changes. [45] [48]

Piezoresistive sensors are attractive for e-skin because they detect both static and dynamic stimuli with fast response, a broad range, and simple, low-cost fabrication. [45] A representative micro-cilia/porous-PDMS MWCNT design by Yibo Liu achieves high sensitivity, sub-Pa detection, rapid response, strong durability, and humidity-independent performance for reliable motion and physiological monitoring. [49] However, high sensitivity often introduces hysteresis due to polymer viscoelasticity and conductive-network rearrangement. [50] Microstructured (e.g., micropatterned or porous) architectures help mitigate this trade-off, enabling sensors with both high sensitivity and low hysteresis. [51] In addition, piezoresistive sensors typically suffer from poor temperature stability. [50] For example, silicon devices exhibit temperature-dependent resistance and gauge factors due to mobility reduction at elevated temperatures. Consequently, temperature control, auxiliary temperature sensors, or Wheatstone-bridge compensation are often required, albeit at the cost of increased circuit complexity. [45]

4.2 Capacitive mechanism

Capacitive pressure sensors (CPSs) utilise soft dielectric layers to translate mechanical stimuli into changes in capacitance. They usually consist of a top electrode / soft dielectric layer / bottom electrode configuration. The relevant equation governing the capacitance is:

$$C = \varepsilon r \; \varepsilon 0 (\frac{A}{d}) \tag{4}$$

Here εr is the dielectric constant, ε0 the vacuum permittivity, A the facing electrode area, and d the electrode gap. [9, 52] When external pressure is applied, the dielectric layer deforms, decreasing the electrode gap (d) and increasing the effective permittivity (εr) due to densification or air-gap closure in porous media. [9, 52, 53] Shear deformation and microstructured features (pyramids/pores/domes) can also enlarge the effective electrode area (A). [9] [54] Therefore, real-time monitoring of external pressure is achieved by measuring capacitance variations resulting from coordinated changes in d, A, and εr under mechanical load.

Capacitive pressure sensors (CPSs) therefore offer a low detection limit, wide operating range, excellent linearity, high sensitivity, low hysteresis, and fast dynamic response. Their simple parallel-plate structure, mechanical robustness, and low temperature coefficient of capacitance favour stable long-term operation. [54-56] Compared to other types of pressure or strain sensors, capacitive sensors use less power because no current flows across the capacitor plates during sensing, and they can be compatible with passive or wireless sensing via inductive coupling. [56, 57] However, CPSs can exhibit slow response and, especially, slow recovery due to bulk and interfacial viscoelasticity. In dielectrostriction-based designs (air gaps, microstructures, or high-dielectric-constant composites), sensitivity is often confined to low pressures (≈3 kPa); at higher loads, air gaps collapse and the dielectric stiffens, while parasitic capacitances and environmental coupling further degrade stability. [55] [58] Despite rapid progress, jointly achieving ultra-high sensitivity, wide range, rapid recovery, and low hysteresis remains a fabrication and structural-design challenge. [52] To overcome these limitations, structural and material engineering strategies have been adopted. The all-PDMS capacitive pressure sensors developed by Mohd Farman et al. use engineered, stable interfaces to address flexibility and interfacial-bonding issues, achieving high sensitivity, rapid response, and stable operation up to ~250 kPa over 11,400 cycles. [59]

4.3 Piezoelectric mechanism

Piezoelectric pressure sensing relies on the direct piezoelectric effect, which is the mutual coupling between mechanical and electrical variables in the material. When the material is deformed by external forces, this coupling causes a change in polarisation, resulting in a measurable voltage across the electrodes. This relationship can be expressed by the following linear constitutive equation. [60] [61]

$$S_{ij} = S^{E}_{ijkl} T_{kl} + d_{ijm} E_m \tag{5}$$

and

$$D_n = d_{nkl} T_{kl} + \varepsilon^{T}_{mn} E_{mn} \tag{6}$$

where S_{ij}, T_{kl}, D_n, and E_m correspond to the mechanical strain, stress tensors, electric displacement, and field vectors, respectively.

Microscopically, pressure-induced deformation separates or reorients electric dipoles, producing an electrical signal. [51] For example, the electroactive β-phase of PVDF, with its all-trans planar zigzag conformation, maximises dipole moment per unit cell by separating fluorine and hydrogen atoms and exhibits strong spontaneous polarisation. [62] In such devices, external forces further deflect dipoles, yielding measurable changes in piezoelectric potential across the surface. [63]

Owing to their self-powered nature, piezoelectric pressure sensors have been widely used in wearable devices, electronic skin, and medical monitoring. [50] A representative example is the kirigami-inspired 3D sensor by Yi Zhang et al., where compressive-buckling assembly and modulus-graded encapsulation enable tunable voltage and sensitivity, and the device reliably captures pulse signals and plantar pressure with excellent durability. [64]

However, piezoelectric sensors generate only impulsive signals and therefore detect dynamic rather than static pressures; they perform poorly under truly static loads, and many piezoelectric materials exhibit pyroelectric behaviour, which can induce spurious charges and require compensation for reliable operation. [9] [50] [51]

4.4 Ionogel-based mechanism

Ionogel-based pressure sensors primarily operate via an electric double-layer (EDL) capacitive mechanism, and occasionally via a piezoionic mechanism. In EDL sensors, an applied voltage induces nanometre-scale EDL formation at the electrode–ionogel interface, producing ultrahigh interfacial capacitance. [65] External pressure enlarges the real contact area, removes air gaps, and shortens the charge-separation distance, thereby amplifying capacitance changes; soft ionogels and microstructures (micropillars, pyramids, domes) are commonly used to enhance compressibility and sensitivity. [65] [66] Ionogels can also operate in piezoionic mode, in which pressure alters ion-transport pathways and junctions, thereby altering resistance. High ionic conductivity supports fast response, and the softness and stretchability of ionogels enable conformal operation on curved, dynamic skin. [67]

Ionogels facilitate strong EDL formation. Blending ILs with high-polarity polymers yields robust ionogel dielectrics with abundant mobile ions, stabilising the EDL and improving sensitivity. Moreover, using electrode materials with ion-intercalation pseudocapacitance, such as $Ti_3C_2T_x$ MXene, can further increase capacitance and pressure sensitivity. [65]

Due to nanometre-scale EDL-enhanced interfacial capacitance and the high compressibility/ionic mobility of soft ionogels, iontronic pressure sensors generate large capacitance changes at low pressures, enabling high sensitivity and fast response. Their intrinsic flexibility, uniform IL–polymer dispersion, and optical transparency further provide stable, low-hysteresis performance even under extreme temperatures, supporting conformal wearable and e-skin applications. [67, 68] However, ionogel sensors remain difficult and costly to fabricate, and are prone to multimodal interference such as humidity-driven dielectric changes and water-induced swelling/IL diffusion, which degrade conductivity and stability. [68] At the architectural level, limited compressibility, high-pressure sensitivity loss, the need for complex moulds, and the small ion-pathway changes in piezoionic modes further constrain sensitivity, though material and microstructural optimisation can partly mitigate these issues. [66, 67]

5. Hybrid and Multimodal Mechanisms

Multimodal, flexible, and stretchable sensors are being developed to advance electronic skin, which mimics the diverse mechanical properties and sensing functions of human skin. [69] These devices transform various external stimuli into measurable electrical signals, including resistance, current, voltage, or capacitance. [70] Mechanical cues are typically detected through piezoresistive, piezoelectric, or capacitive effects; temperature through thermoresistive, pyroelectric, or thermoelectric responses; and humidity through resistance or capacitance changes. [71] When these mechanisms are combined, their complementary behaviours enable simultaneous, sensitive, and stable detection of multiple stimuli.

Multimodal sensing is achieved either by using materials that inherently respond to multiple stimuli or by designing microstructures that deform differently under varied loads. Normally, multiple sensing units are integrated in-plane or out-of-plane to combine functions. [70] Because multimodal systems face crosstalk, decoupled sensing is essential, and pairing distinct mechanisms with distinct electrical outputs (DEOS) is more effective than single-output schemes (SEOS). [70] [71] For example, Zhu et al. created a self-powered hybrid e-skin where triboelectric and piezoelectric layers respond in different time domains, enabling natural decoupling of non-contact and contact signals.[72]

6. Summary and prospects

Electronic skin (e-skin) refers to flexible, bionic electronic systems that mimic the tactile and physiological sensing functions of human skin, built from tailored combinations of conductive fillers, substrates, and sensing mechanisms. Despite progress, e-skins still face key challenges: performance trade-offs among sensitivity, hysteresis, and stability; limited mechanical and environmental robustness; and manufacturing routes that remain complex and difficult to scale sustainably. At the system level, multimodal crosstalk, insufficient self-powering, and limited integration with intelligent data processing continue to limit the development of fully adaptive e-skin platforms. Accordingly, future developments will emphasise AI-assisted data analysis, low-crosstalk multimodal architectures, high-sensitivity and low-latency material/microstructure optimisation, improved interfacial and thermal management, and the adoption of biodegradable materials with scalable green fabrication. These advances will drive the emergence of intelligent, adaptive, and sustainable e-skin systems.

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