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Next-Generation Wearable and Stretchable Strain Sensors for Healthcare: Materials, Mechanisms, Architectures, and Applications

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Abstract: Wearable and stretchable strain sensors are central to smart wearable technology, enabling continuous, real-time, and noninvasive tracking of human motion and physiological signals for healthcare, sports, and human–machine interaction. Rapid progress in soft materials, structural design, and flexible electronics has led to devices with improved sensitivity, large strain range, durability, and skin conformability, yet practical translation into robust products remains limited. This review provides a comprehensive overview of recent advances in wearable and stretchable strain sensors with emphasis on healthcare applications. We first summarize key sensing mechanisms—including piezoresistive, capacitive, piezoelectric, triboelectric, and optical modes—and compare their operating principles and performance trade-offs. We then discuss material platforms such as elastomeric substrates, conductive polymers, nanomaterials, and hybrid composites, followed by critical design parameters (gauge factor, stretchability, hysteresis, response time, durability) that govern device performance. Structural and device engineering strategies, including microstructured surfaces, percolated networks, serpentine interconnects, porous scaffolds, and liquid-metal architectures, are highlighted as routes to achieving high performance under complex deformations. Representative applications in vital-sign monitoring, motion and gait analysis, rehabilitation, wound care, and smart textiles are reviewed, along with broader uses in soft robotics and structural health monitoring. Finally, we identify key challenges—long-term stability on skin, biocompatibility, power autonomy, secure data handling, standardized benchmarking, and scalable manufacturing—and outline future directions in self-powered systems, multimodal sensing, and AI-assisted analytics. This review aims to provide a consolidated framework to guide the design of next-generation, clinically relevant wearable strain-sensing platforms.

Keywords: wearable strain sensors, stretchable electronics, healthcare monitoring, smart textiles, human–machine interfaces

1. Introduction

In recent years, wearable electronics have transformed from experimental prototypes into highly functional devices capable of monitoring a variety of physiological and biomechanical signals

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in real time. Among these, wearable and stretchable strain sensors have emerged as a pivotal technology, enabling continuous and noninvasive tracking of human motion, health parameters, and human–machine interactions. Strain sensors measure deformation or displacement through changes in electrical, optical, or mechanical properties, providing critical insights into joint motion, respiration, pulse, gait patterns, and muscular activity [1, 2]. The integration of stretchable materials with electronic sensing platforms allows these devices to conform to complex body geometries, maintain functionality under repeated deformation, and deliver high-fidelity data for both clinical and consumer applications.

The evolution of wearable strain sensors is deeply intertwined with advances in materials science, nanotechnology, and soft electronics. Traditional rigid sensors, while accurate, are limited by their lack of flexibility, discomfort, and susceptibility to mechanical damage when applied to dynamic surfaces such as skin or soft tissue. To address these challenges, researchers have developed stretchable substrates such as polydimethylsiloxane (PDMS), polyurethane, and elastomeric composites, combined with conductive fillers including carbon nanotubes (CNTs), graphene, metallic nanowires, and liquid metals [3–5]. These hybrid material systems exhibit remarkable mechanical compliance and maintain electrical performance under strains exceeding 100%, enabling sensors to conform to the curvature of joints, muscles, and other deformable surfaces without compromising signal integrity.

Sensing mechanisms in wearable strain sensors have also evolved significantly, moving beyond conventional resistive or capacitive designs. Modern platforms incorporate piezoresistive, capacitive, piezoelectric, triboelectric, and optical modalities, each with specific advantages and trade-offs in terms of sensitivity, response time, stretchability, linearity, and energy consumption [5–8]. Piezoresistive sensors are widely adopted due to their simple fabrication and high sensitivity to strain; capacitive sensors offer superior linearity and low power operation; piezoelectric and triboelectric sensors enable self-powered sensing for dynamic movements; and optical-based sensors provide high spatial resolution and are less susceptible to electromagnetic interference (EMI) than all-electrical readouts in some setups [9–11]. The combination of these mechanisms in hybrid designs presents opportunities for multimodal sensing, enhancing the robustness and versatility of wearable devices.

Device architecture and structural innovations are central to the performance of next-generation wearable strain sensors. Hierarchical micro- and nano-patterning, porous networks, serpentine interconnects, and liquid-metal channels have been introduced to improve stretchability, durability, and sensitivity [4]. These design strategies allow devices to withstand repeated mechanical deformation while maintaining stable electrical or optical output, which is crucial for long-term monitoring in real-life applications. Additionally, the development of multifunctional sensors capable of simultaneously detecting strain, temperature, humidity, or biochemical signals represents a significant trend toward integrated wearable health platforms [1].

The applications of wearable strain sensors in healthcare have expanded rapidly. They have demonstrated efficacy in monitoring vital signs such as respiration rate and heart rate, tracking rehabilitation exercises, assessing musculoskeletal health, detecting falls, and even evaluating wound healing dynamics [2, 3]. The continuous and noninvasive nature of these sensors allows for remote patient monitoring and early diagnosis, reducing the need for hospital visits and enabling proactive healthcare interventions. Moreover, wearable strain sensors facilitate human–machine interfaces, contributing to the development of prosthetics, exoskeletons, and robotic assistive devices [1]. These applications illustrate the profound impact of stretchable sensors beyond conventional medical monitoring, bridging the gap between health, mobility, and human–computer interaction.

Despite these advances, several critical challenges remain. Long-term mechanical stability and performance under repeated deformation are not fully addressed, limiting the practical deployment of wearable strain sensors in daily life. The integration of power sources, wireless communication modules, and data processing units remains an ongoing engineering challenge, particularly in compact and skin-conformal devices [5]. Additionally, the translation of laboratory-scale prototypes into clinically validated systems is constrained by the lack of standardized testing protocols, regulatory approvals, and large-scale manufacturing methods. Biocompatibility, skin irritation, and environmental robustness (temperature, humidity, sweat) are also essential considerations that require further study [1–3].

Given these factors, a comprehensive synthesis of current developments is necessary to guide future research and commercialization efforts. Unlike material-specific reviews (e.g., graphene-focused surveys) or mechanism-specific reviews (e.g., triboelectric monitoring), this review emphasizes cross-mechanism integration and translation constraints that span all platforms. Specifically, we synthesize how materials, sensing mechanisms, and device structures interact to determine failure modes relevant to wearables (drift/hysteresis, humidity sensitivity, parasitics, packaging, and skin-interface effects), and we propose testable hypotheses and a consolidated framework to guide research beyond incremental demonstrations.

Lastly, wearable and stretchable strain sensors (Figure 1) represent a rapidly advancing field at the intersection of materials science [1, 12, 13], biomedical engineering [1, 13], and soft electronics [1]. Their ability to deliver continuous, noninvasive, and high-fidelity monitoring positions them as transformative tools for healthcare, rehabilitation, and human–machine interaction. By evaluating the current state of the art, identifying persistent challenges, and outlining emerging opportunities, this review provides a foundational reference for researchers aiming to design robust, multifunctional, and clinically relevant wearable strain-sensing systems. The following sections detail the materials, sensing mechanisms, device architectures, healthcare applications, challenges, research gaps, and future directions for these technologies [2, 14–17].

2. Classification of the Sensing Mechanism for Wearable and Stretchable Strain Sensors

A translation-oriented comparison of sensing mechanisms, including typical performance ranges and dominant failure modes, is summarized in Table 1. Typical ranges summarize common wearable implementations and depend strongly on geometry, strain rate, mounting, and test protocol; extreme maxima reported in single demonstrations are not treated as representative.

2.1. Piezoresistive sensors

Piezoresistive strain sensors [1, 18] transduce deformation into a change in electrical resistance and remain among the most widely used wearable strain-sensing modalities because they offer direct electrical readout, simple device integration, and a favorable sensitivity-to-complexity ratio for soft, skin-conformal systems. In healthcare wearables, they are commonly deployed for

Figure 1
Schematic overview of advanced wearable and stretchable strain sensors, highlighting the interplay between (top left) desirable performance parameters, (top right) key working mechanisms, (bottom left) representative conducting and functional materials, and (bottom right) major application domains

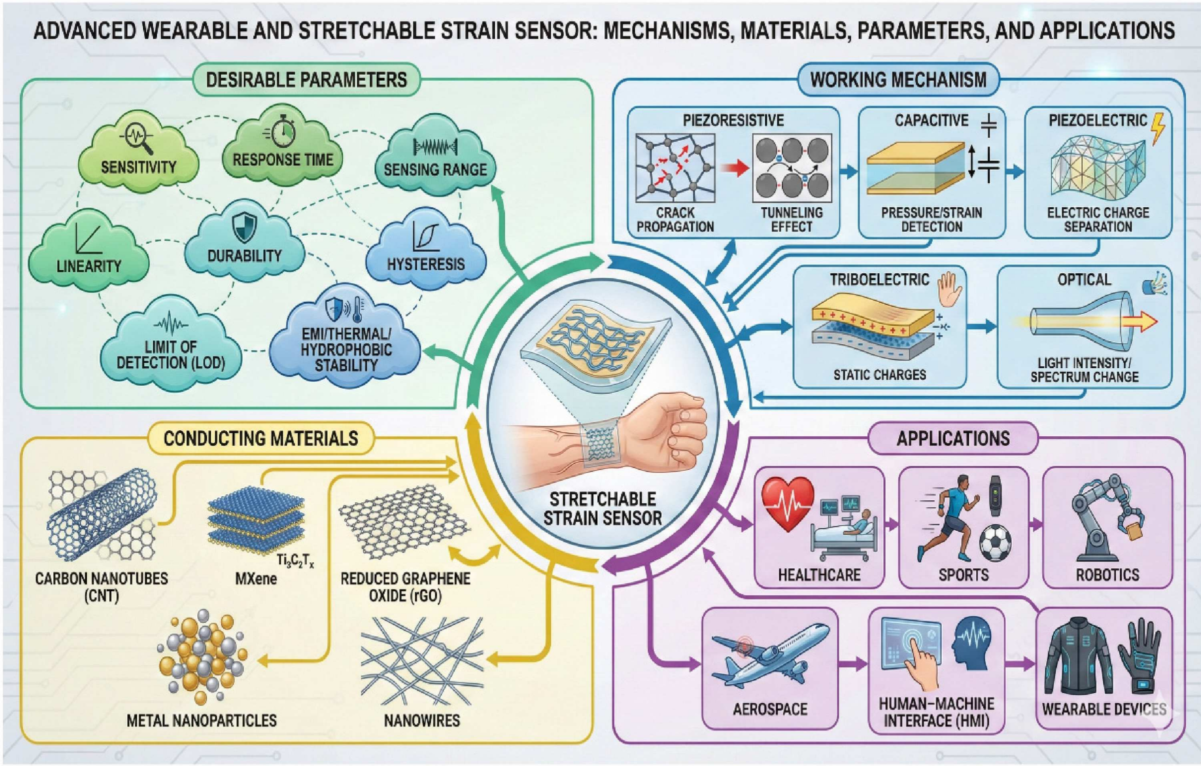


Table 1
Comparison of sensing mechanisms

Mechanism	Transduction principle	Typical sensitivity/benchmark	Practical strain range (typical)	Key strengths	Dominant limitations (real wear)	Translation notes (why/when it wins)	Representative examples (from your refs)
Piezoresistive	Strain changes resistance via geometry + intrinsic resistivity (percolation/tunneling/crack evolution)	GF ~ 1–10^2 (common composites); 10^3–10^5 (microcrack/engineered networks)	0–50% (common); up to 100–300%+ with elastomeric networks	Very high sensitivity, simple readout, easy integration with compact electronics	Hysteresis, baseline drift, temperature/humidity sensitivity, aging of conductive network	Often dominates early clinical prototypes because electronics are simple; requires drift/hysteresis management (encapsulation + calibration + stable mounting)	[1, 7, 12, 15, 18, 25]

(*Continued*)

Table 1
(*Continued*)

Mechanism	Transduction principle	Typical sensitivity/benchmark	Practical strain range (typical)	Key strengths	Dominant limitations (real wear)	Translation notes (why/when it wins)	Representative examples (from your refs)
Capacitive	Strain changes capacitance via ΔA and/or Δd (and dielectric changes in engineered structures)	Effective GF typically < 1–2 (geometry-limited); higher values reported in specialized ionic/hydrogel microstructures	0–50% (common); can be extended with stretchable electrodes/dielectrics	High linearity, low power, good repeatability, less temperature-dependent than resistive	Susceptible to parasitic capacitance, EMI/pickup, lead motion artifacts; requires shielding/grounding	Preferred for long-duration monitoring where stability matters; best with careful packaging + circuit design	[11, 19]
Piezoelectric	Strain/stress induces charge separation → voltage/current	Best for dynamic strain; sensitivity often reported as voltage/charge per force/strain (varies strongly by material/geometry)	Dynamic motions; not ideal for static holds	Self-powered sensing, strong transient response, good for impacts/vibrations	Weak/zero static response, brittleness (ceramics), packaging constraints	Good for gait/impact and event detection; often paired with a baseline-capable sensor for continuous tracking	[27–29, 34–36]
Triboelectric (TENG)	Contact electrification + electrostatic induction → voltage/current	Strong outputs for dynamic motion; highly design-dependent (surface microstructure, contact area, frequency)	Dynamic motions; output depends on contact/separation	Self-powered, lightweight, broad material choice, high output for motion events	Humidity sensitivity, wear/charge decay, motion-dependent calibration, variability with contact mechanics	Excellent for self-powered event detection; needs humidity mitigation (coatings/barriers) and calibration if used quantitatively	[37, 40, 42, 43, 47, 48]
Optical strain sensing (FBG/waveguides/polymer fibers)	Strain modulates wavelength/intensity/phase in fiber/waveguide	Reported as $\Delta\lambda/\varepsilon$, intensity change per ε, or $dB \cdot \varepsilon^{-1}$ (platform-specific)	Wide, platform-dependent; can support distributed sensing	Immune to electrical pickup at the sensing site; high resolution; potential multiplexing	Bulkier optoelectronics, alignment/packaging, bending loss, motion artifacts (for some wear setups)	Attractive when EMI or electrical isolation is critical; translation depends on packaging + cost reduction	[50–54, 57]

(*Continued*)

Table 1
(*Continued*)

Mechanism	Transduction principle	Typical sensitivity/benchmark	Practical strain range (typical)	Key strengths	Dominant limitations (real wear)	Translation notes (why/when it wins)	Representative examples (from your refs)
Hybrid: tribo–piezo	Dual self-powered channels: triboelectric + piezoelectric	Two complementary dynamic signals (broadband events + vibration signatures)	Dynamic motions	Higher information content; improves robustness for event classification	Still limited for static strain; humidity affects tribo channel	Useful for HMI and activity recognition where events dominate; fusion improves reliability	[42, 47, 48]
Hybrid: tribo–capacitive	Self-powered tribo channel + stable capacitive baseline	Capacitive gives stable low-freq baseline; tribo adds dynamic transients	Mixed (quasi-static + dynamic)	Stability + self-powered transient sensing in one system	Added integration complexity; parasitics + humidity must be managed	Strong pathway to practical wearables: stable baseline + self-powered features	[8]
Hybrid: piezoresistive–capacitive	Dual electrical channels: resistive sensitivity + capacitive stability	Resistive high GF + capacitive stable baseline	Mixed	Improves robustness; mitigates single-mode failure (drift vs parasitics)	More wiring/circuit complexity; needs fusion/calibration	Useful for long-duration monitoring where drift or parasitics alone would degrade performance	[1, 11]
Hybrid: optical–electrical	Optical strain channel + electrical strain channel	Redundant sensing for reliability	Mixed	Robustness to specific artifacts; redundancy	Higher integration complexity and cost	Beneficial when placement/EMI constraints vary; redundancy improves reliability	[50–54, 57]

continuous monitoring of respiration-related expansion, pulse-linked skin motion, and joint kinematics using thin, flexible substrates that minimize motion restriction [19].

At a fundamental level, resistance modulation under strain reflects two coupled contributions: (i) geometry-driven changes in the current path (e.g., elongation and cross-sectional contraction) and (ii) strain-dependent transport changes within the sensing layer, which are especially important in soft composites. In polymer–filler systems, mechanical deformation alters the spacing and contact between conductive domains, changing the effective conduction pathways and the tunneling barriers that govern electron transport. Importantly, in many stretchable composites, the transport contribution can dominate the signal—enhancing sensitivity—but it is also the primary origin of nonideal behaviors such as hysteresis and drift [20, 21].

For wearable deployment, the key limitation is that the conductive microstructure evolves with time and cycling. Under repeated deformation, viscoelastic relaxation and filler rearrangement can cause the loading and unloading curves to diverge (hysteresis) and can shift the baseline even at nominally constant strain (drift). This also introduces strain-rate dependence, where the output differs for slow versus rapid motions—an important consideration for dynamic monitoring scenarios (e.g., gait vs resting respiration) that require reliable calibration [22].

Translation-ready piezoresistive designs therefore treat hysteresis/drift as system-level constraints, not merely material properties. Stabilization strategies include microstructure engineering (to promote repeatable deformation modes), composite architectures that reduce irreversible network rearrangement, and encapsulation/skin-interface control to limit humidity ingress and mounting-induced artifacts [23]. In addition, multimodal designs can exploit complementary channels: for example, piezoresistive–triboelectric hybrids can use the resistive signal as a quasi-static baseline, while the triboelectric output captures dynamic transients; however, this hybridization must explicitly manage piezoresistive hysteresis/drift alongside triboelectric humidity sensitivity, typically via packaging (barrier layers) and signal fusion/normalization routines [24–26].

2.2. Capacitive sensors

Capacitive strain sensors transduce deformation into a change in capacitance and are often favored in wearable monitoring because they can provide a stable baseline with low power consumption and high repeatability over long-duration use. A typical device consists of two compliant electrodes separated by a deformable dielectric, and the measured capacitance can be approximated by $C = \varepsilon A/d$, where ε is the effective permittivity, A is the electrode overlap area, and d is the separation distance [1, 23]. In practice, stretching usually reduces the effective overlap and/or increases separation, yielding a decrease in C, while compression produces the opposite trend. Compared with piezoresistive composites, capacitive sensors are often less prone to irreversible microstructural rearrangement, which helps explain their reputation for lower drift in continuous monitoring settings.

However, the primary challenges for wearable capacitive sensors are typically system level rather than purely material level. Because capacitance values are small, the readout can be influenced by parasitic capacitance, lead motion, shielding quality, and environmental coupling, which can appear as “noise” or baseline wandering during movement. Additionally, nonplanar deformation (bending, wrinkling, shear) can change geometry in ways that deviate from the ideal parallel-plate assumption, requiring careful mechanical design and calibration. Moisture and sweat can also modify dielectric properties or interfacial conditions, affecting repeatability in real-world wear.

Performance improvements therefore often combine materials and architecture: soft dielectrics (including microstructured or porous layers) amplify deformation-induced capacitance change; stretchable electrodes (e.g., liquid metals, carbon composites, conductive polymers) maintain conductivity under strain; and differential layouts or shielding strategies suppress parasitics [1, 23]. In clinical translation, capacitive sensors are especially attractive for low-frequency physiological signals such as respiration, posture, and slow joint motion, where stable baseline behavior is prioritized over extreme sensitivity. For applications that require both stability and high sensitivity, capacitive sensors are often paired with a resistive or triboelectric channel in hybrid configurations.

2.3. Piezoelectric sensors

Piezoelectric sensors convert mechanical stress into an electrical response through charge separation in non-centrosymmetric materials, including Polyvinylidene Fluoride (PVDF) and its copolymers, ceramic-based systems (e.g., Lead Zirconate Titanate (PZT)), ZnO nanostructures, and emerging hybrid formulations [24–31]. Their key advantage in wearables is that they naturally encode dynamic deformation—rapid motion, vibration, impacts—into measurable electrical signals, enabling event-driven monitoring and, in some cases, self-powered operation. This is why piezoelectric wearables frequently appear in gait/impact sensing, tremor or vibration detection, and dynamic biomechanical signatures associated with cardiovascular activity (e.g., mechanical components of pulse or heart motion) [31–36].

The same physics also imposes an important limitation: piezoelectric output is strongest when strain is time-varying, and it typically diminishes when deformation becomes static or very slow. As a result, piezoelectric sensing alone is not ideal for applications where the clinical target is a stable baseline strain (e.g., sustained posture holds or slow breathing trends), unless the system focuses on extracting transient features rather than continuous strain magnitude. Practical wearables also face packaging challenges: ceramic piezoelectrics can be brittle, and even flexible polymers require structural engineering (thin films, serpentine layouts, neutral-plane placement, or encapsulation) to survive repeated bending and stretching.

Translation-ready piezoelectric wearables therefore often adopt one of two strategies: (i) treat piezoelectric sensing as a dynamic-event channel for gait, gesture, or acoustic/vibration signatures or (ii) combine piezoelectric elements with resistive/capacitive sensors to capture both dynamic transients and quasi-static strain in a single platform [33, 35]. This hybrid logic is especially useful in rehabilitation and locomotion monitoring, where both continuous posture information and event-level movement features matter.

2.4. Triboelectric sensors

Triboelectric sensors generate electrical signals from the coupled effects of contact electrification and electrostatic induction when two materials repeatedly contact and separate [37–48]. In wearable systems, they are attractive because they can deliver high signal amplitudes for motion events and can support self-powered sensing, converting everyday human motion into electrical output that can be used for detection (and sometimes energy harvesting). Triboelectric wearables have been demonstrated for finger/joint motion, posture changes, respiration-related movement, and tactile interaction—applications where intermittent motion produces distinct signal patterns [37–48].

The translation bottleneck for triboelectric sensing is that the output is strongly influenced by surface condition and environment, especially humidity. Moisture can accelerate charge dissipation, reducing signal stability and making calibration user- and environment-dependent. Triboelectric signals are also inherently tied to contact/separation dynamics (motion mode, frequency, force, surface roughness), which complicates attempts to interpret the output as a unique “strain value” under arbitrary real-world use. These characteristics make triboelectric sensors exceptionally good for event detection and pattern recognition, but less straightforward for continuous, absolute strain measurement without careful controls.

Accordingly, robust triboelectric wearable designs increasingly rely on strategies that explicitly address variability: engineered microstructures to increase repeatable contact area, barrier layers or hydrophobic coatings to reduce humidity effects, and signal normalization routines to handle user-to-user differences. Importantly, triboelectric sensors are frequently deployed in multimodal hybrids: a resistive or capacitive channel provides a stable baseline for quasi-static strain, while the triboelectric channel captures motion-driven transients and can contribute to self-powered operation. In such hybrids, the combined design must manage piezoresistive hysteresis/drift alongside triboelectric humidity sensitivity, which motivates both packaging solutions (environmental barriers) and signal-fusion approaches [49].

2.5. Optical sensor

2.5.1. Optical physiological sensing (PPG)

Optical physiological sensing in wearables—most commonly photoplethysmography (PPG)—infers cardiovascular variables by measuring light absorption changes associated with pulsatile blood-volume variation in tissue. Wearable implementations typically use LEDs and photodiodes; green wavelengths are commonly used for heart-rate monitoring due to strong hemoglobin absorption, while red/near-infrared bands enable estimation of oxygenation by comparing absorption differences between oxygenated and deoxygenated hemoglobin [50–52]. Because PPG measures hemodynamics rather than deformation, it should be treated as a biophotonics modality, not a strain-sensing mechanism. In practice, the dominant deployment challenges include motion artifacts, skin-contact variability, and user-to-user physiological differences, which have motivated algorithmic approaches for artifact rejection and robust feature extraction [53–55].

2.5.2. Optical strain sensing (FBGs, polymer fibers, elastomeric waveguides)

Optical strain sensors measure deformation through strain-induced modulation of an optical path—such as wavelength shifts in fiber Bragg gratings (FBGs), intensity changes due to bending/strain loss in polymer fibers, or guided-mode modulation in stretchable elastomeric waveguides. These approaches can support high-resolution deformation tracking and, in some implementations, reduced susceptibility to certain electrical interference mechanisms compared with purely electrical sensors. However, optical wearables often face integration challenges (packaging, alignment/coupling stability, bend/temperature cross-sensitivity, and cost), which can limit manufacturability relative to all-electrical solutions [56, 57].

2.6. Translation-oriented comparison: why some mechanisms dominate clinically

In clinical translation, the supposed ideal mechanism is often the one that minimizes system complexity while maintaining stable, interpretable signals under real-world conditions (skin-contact variation, sweat/humidity, motion artifacts). For this reason, piezoresistive and capacitive sensors frequently dominate wearable clinical prototypes and early products: they provide a direct electrical readout compatible with compact electronics and wireless modules, and capacitive platforms in particular are valued for low power consumption and stability in long-term monitoring. The main translation bottlenecks are hysteresis and baseline drift (especially in polymer composites), plus environmental dependence; accordingly, practical designs increasingly rely on microstructural engineering, encapsulation, and calibration routines to maintain repeatability over extended cycling.

By contrast, piezoelectric and triboelectric sensors are attractive for self-powered operation and dynamic-event detection, but translation is often limited by their stronger dependence on dynamic strain rate/frequency and by environmental variability (e.g., humidity sensitivity in triboelectric outputs), which complicates calibration for continuous, low-frequency physiological monitoring. Optical approaches offer compelling advantages such as high spatial resolution and potential less susceptible to EMI than all-electrical readouts in some setups, but they typically introduce additional packaging/alignment and optoelectronic integration complexity, which can raise cost and slow manufacturability relative to all-electrical readouts.

2.7. Cross-mechanism trade-offs and design rules

While the previous subsections describe individual transduction mechanisms, real-world wearable performance is usually dictated by trade-offs and non-idealities (hysteresis, drift, humidity sensitivity, parasitics, motion artifacts, and packaging). Below, we summarize cross-mechanism contradictions and provide practical design rules that can guide mechanism and material selection.

2.7.1. Cross-mechanism contradictions that drive design choices

Sensitivity vs stability: Piezoresistive sensors can achieve high sensitivity but often exhibit hysteresis and baseline drift due to viscoelastic relaxation and evolving conductive networks. Capacitive sensors are often more stable for continuous monitoring but can be more susceptible to parasitic capacitance, shielding demands, and motion-induced geometric artifacts.

Static/low-frequency vs dynamic-only response: Piezoelectric and triboelectric platforms are well suited for dynamic deformation and event detection but are inherently less appropriate for static or quasi-static strain tracking (e.g., long-hold posture, slow respiration baselines) unless combined with another modality.

Environmental robustness vs self-powering: Triboelectric signals can be strong and self-powered but are often sensitive to humidity and surface condition, complicating calibration. This motivates hybrid designs where an electrical strain channel (resistive/capacitive) provides a stable baseline while triboelectric output captures dynamic transients.

Signal integrity vs integration complexity: Optical strain sensing can reduce susceptibility to certain electrical interference and enable high-resolution mapping but may introduce optoelectronic integration, alignment, and packaging complexity that can slow scaling compared with fully electrical readouts.

2.7.2. Design rules for translation-oriented wearable strain sensors

1) **Choose the mechanism by clinical signal bandwidth first.**

   a. For continuous low-frequency monitoring (respiration belts, posture, long-duration joint flexion): prioritize capacitive or drift-mitigated piezoresistive designs.

   b. For high-dynamic-event detection (gait impacts, fast gestures): piezoelectric/triboelectric designs can be advantageous, especially if self-powered operation is desirable.

2) **Treat hysteresis and drift as system-level problems, not only material problems**. Reduce hysteresis/drift via a *stack* of measures: microstructure engineering (e.g., crack-based or serpentine structures), stable encapsulation, controlled skin-contact mechanics, and calibration protocols (including periodic baseline correction).
3) **Humidity/sweat robustness must be designed, not assumed.** Any mechanism relying on surface charge or ionic transport should be evaluated under humidity/sweat exposure, and mitigation should be explicit (encapsulation, hydrophobic coatings, breathable barrier layers, or redundant sensing channels).
4) **Hybrid sensors should be justified by complementary failure modes.** Combine mechanisms only when the added modality addresses a specific weakness:

a. resistive + capacitive → sensitivity + stability

b. triboelectric + resistive/capacitive →' self-powered transients + stable baseline

c. optical strain + electrical strain → redundancy under EMI/motion/placement constraints

5) **Benchmarking must reflect use conditions.** Report strain-rate dependence, long-term cycling, mounting method, sweat/humidity exposure, and calibration repeatability. Without these, comparisons across mechanisms are not meaningful for healthcare translation.
6) **Commercialization readiness depends on manufacturability and repeatability.** Prefer material/architecture choices compatible with scalable processes (printing/lamination/molding) and specify what controls lot-to-lot variability (QC metrics, calibration spread, yield).

3. Materials for Wearable and Stretchable Strain Sensors

The performance of wearable and stretchable strain sensors is heavily dependent on the choice of materials, which influence sensitivity, stretchability, durability, biocompatibility, and manufacturability. Researchers have developed a diverse range of materials, broadly categorized into substrates, conductive fillers, and composite/hybrid materials (see also Figure 2).

3.1. Substrates

Substrates form the foundation layer on which sensing elements, interconnects, and encapsulation layers are built. In wearable healthcare devices, substrates must be lightweight, flexible, biocompatible, and capable of closely matching the mechanical properties and curvature of human skin to maintain stable contact and reliable signal acquisition during daily activities [58–60]. Common examples include PDMS, Ecoflex, polyimide, polyurethane, textiles (woven/knitted fabrics), paper, and hydrogels, as summarized in Table 2 [61–64]. These materials allow the device to bend, stretch, twist, or compress without cracking or losing sensing performance. Their softness also improves user comfort and enables long-term contact with the body. Some substrates, such as hydrogels, offer additional benefits like moisture retention, which enhances adhesion to the skin and facilitates biosignal acquisition. Overall, substrates serve as the mechanical backbone of the sensor, preserving structural integrity while supporting flexible electronics [65–68].

Figure 2
Hierarchical materials system for wearable and stretchable strain sensors: substrates, conductive fillers, and composite architectures.

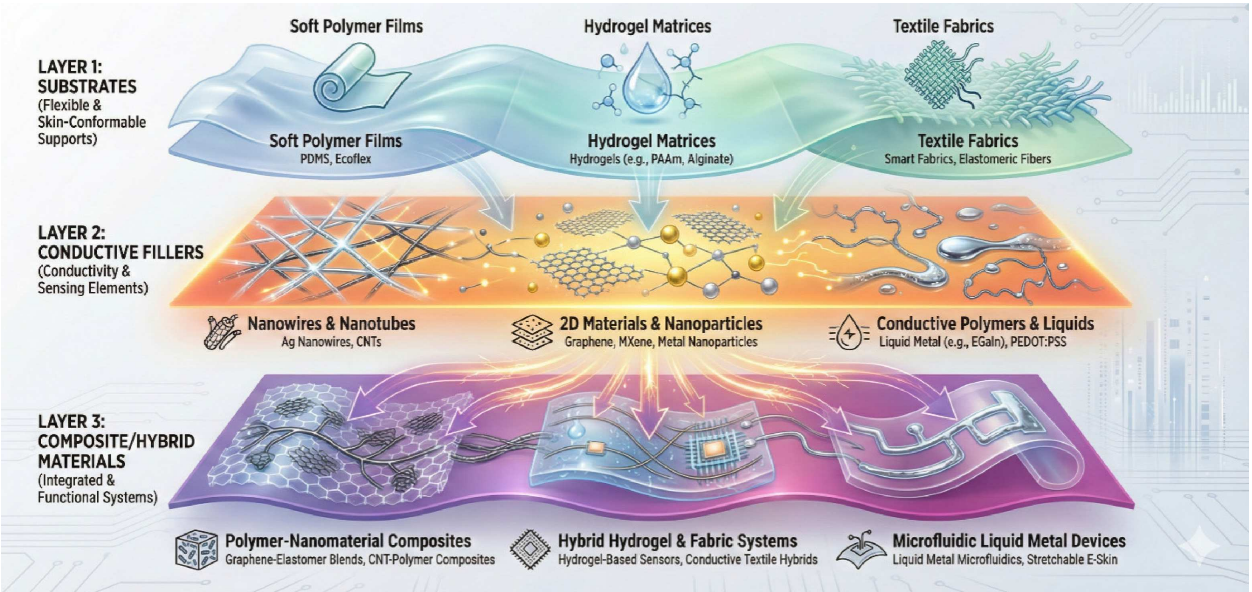


Table 2
Substrate materials for stretchable strain sensors

Material	Properties	Advantages	Limitations
PDMS (polydimethylsiloxane) [1, 15]	Elastomeric, flexible, biocompatible	High stretchability; skin-friendly; easy to mold	Hydrophobic; limited adhesion to some fillers
Polyurethane (PU) [23]	Soft, flexible, elastic	Durable; good mechanical recovery	Can degrade under UV exposure; moderate thermal stability
Ecoflex [1, 24]	Highly stretchable, low modulus	Ultra-flexible; excellent for conformal sensors	Relatively low tensile strength
Thermoplastic elastomers (TPE) [1, 45]	Flexible, heat-processable	Recyclable; compatible with various fabrication methods	Lower chemical resistance than PDMS

3.2. Conductive fillers

Conductive fillers are the functional components responsible for electrical conductivity. They form the conductive pathways that detect mechanical deformation (such as strain or pressure) or physiological signals (such as temperature, sweat ions, or electrophysiological activity) [69–71].

Common conductive fillers include metal nanoparticles/-nanowires (Ag, Au, Cu), carbon materials (graphene, CNTs, carbon black) [72–76], and liquid metals (eutectic gallium–indium). Each filler offers distinct advantages: metals provide high conductivity, carbon materials offer excellent flexibility and chemical stability, while liquid metals maintain conductivity even under extreme deformation [77–80]. Conductive fillers are typically embedded into soft substrates or patterned onto films through printing, coating, or laser writing. Their ability to maintain percolation networks under mechanical stress makes them essential for high-performance stretchable electronics [81, 82].

3.3. Composite/hybrid materials

Composite or hybrid materials combine substrates and conductive fillers into a single multifunctional matrix, integrating mechanical flexibility with stable electrical behavior. Examples include CNT–PDMS composites, graphene–elastomer blends, silver nanowire–polyurethane composites, and conductive polymer hydrogels. By tuning filler concentration, dispersion, or alignment, researchers can precisely control sensitivity, durability, stretchability, and response time [83–87]. Hybrid materials often outperform their individual components because they leverage the strengths of both: the softness of elastomers and the conductivity of nanomaterials. These materials also enable tailoring of sensor behavior for specific applications, such as ultrahigh sensitivity for pulse monitoring or large strain tolerance for motion tracking [88–90]. As a result, composite materials are widely used in next-generation wearable sensors due to their durability, customizable properties, and compatibility with scalable fabrication processes like 3D printing and screen printing [87]. Table 3 summarizes recent trends in hybrid materials, while also the translation drivers and bottlenecks by sensing mechanism are shown in Table 4.

3.4. Material–mechanism synergies and dominant failure modes

Material selection in wearable strain sensors should be evaluated in terms of synergy: how a substrate, conductive

Table 3
Hybrid material systems for wearable strain sensors

Hybrid system	Functionality	Advantages	Challenges
CNT–PDMS [23, 91]	Piezoresistive sensing	High sensitivity; stretchable	Uniform dispersion; potential aggregation
Graphene–PDMS [1, 43]	Capacitive/piezoresistive	Flexible; durable	Large-area fabrication; cost
AgNW–PU [1, 45]	Transparent, stretchable conductors	High conductivity; printable	Oxidation; strain-induced cracking
Liquid metal–elastomer [34, 56]	Self-healing, stretchable	Ultra-stretchable; multifunctional	Encapsulation; stability
PEDOT:PSS–elastic polymer [30, 31]	Biocompatible strain sensors	Solution-processable; flexible	Humidity sensitivity; limited conductivity

Table 4
Translation drivers and bottlenecks by sensing mechanism

Mechanism	Why it translates well clinically	Main barriers in practice	Where it wins most
Piezoresistive	Simple fabrication + direct electrical read-out; high sensitivity; easy integration into patches/textiles	Hysteresis, drift, temperature/humidity dependence	Respiration/pulse/motion where low-cost integration is key
Capacitive	Low power, stable output; good for continuous monitoring	Parasitics/shielding, moisture/adhesion effects; readout circuitry complexity	Long-term monitoring (breathing/posture), array sensing
Piezoelectric	Self-powered potential for dynamic motion	Weak for static/slow strain; integration of high-impedance readout; packaging	Dynamic motion events, intermittent monitoring
Triboelectric	Self-powered, high output for motion signals	Output variability; environmental sensitivity; calibration challenges	Motion/gesture, event detection, low-resource use-cases
Optical	High resolution; EMI advantages	Optoelectronics, alignment/packaging complexity; cost	EMI-sensitive settings; high-resolution deformation mapping

phase, and microstructure jointly determine (i) sensitivity and dynamic range, (ii) stability under repeated deformation, and (iii) biocompatibility and manufacturability. For example, elastomer–nanofiller composites (e.g., CNT/graphene–PDMS or AgNW–PU) offer high compliance and simple electrical readout, but their dominant non-idealities are often hysteresis and baseline drift caused by viscoelastic relaxation and evolving conductive networks. These non-idealities can be mitigated by stabilizing the conductive architecture (segregated networks, controlled filler alignment), adopting microstructures that enforce repeatable deformation modes, and using encapsulation strategies that reduce sweat/humidity ingress and mounting variability.

Liquid-metal conductors provide exceptional stretchability and self-healing electrical continuity, yet their key barriers are leakage risk, interface stability, and packaging complexity for long-term skin contact. Hybridizing liquid metal with biocompatible barrier layers or mechanically robust top encapsulation can improve safety and reliability. Emerging 2D materials (e.g., MXenes) and conductive hydrogels offer a complementary pathway: they can deliver soft, tissue-like interfaces and high compliance and can be integrated as coatings or composite phases to improve skin compatibility and signal stability. In this review, we therefore evaluate material systems not only by peak conductivity or stretchability but also by their dominant failure modes and their compatibility with scalable manufacturing and clinical validation.

4. Design Parameters for Stretchable Strain Sensors

The performance of stretchable strain sensors is critically influenced by a set of interrelated design parameters, including stretchability, sensitivity, linearity, hysteresis, response and recovery time, overshooting, and dynamic durability. Understanding and optimizing these parameters are essential for developing high-performance sensors suitable for wearable electronics, soft robotics, and biomedical monitoring [1] (see also Figure 3).

4.1. Stretchability

Stretchability, defined as the maximum strain a sensor can withstand while maintaining structural integrity and reliable sensing, is a key design criterion. It is mathematically expressed as:

$$Stretchability\ (\%) = \frac{l - l_0}{l_0} \times 100\% \tag{1}$$

where l_0 is the initial sensor length and l is the extended length under applied strain. Stretchability depends on multiple factors, including the choice of substrate, fabrication process, and the aspect ratio of micro- or nanomaterials used as sensing elements. For example, composites of graphene, poly (acrylic acid), and amorphous calcium carbonate leverage double crosslinking to achieve high elasticity, enabling stretchability up to 500%. The use of two- or three-dimensional nanomaterials, such as CNTs or silver nanowires, enhances stretchability compared to one-dimensional structures due to the formation of more robust percolation networks under strain [15, 18].

Natural fiber-based sensors have emerged as highly stretchable alternatives, with carbonized silk and cotton fabrics showing strain tolerances of up to 500% and 140%, respectively, when embedded in flexible polymer matrices like Ecoflex. Such innovations are crucial for wearable applications, where large deformations are common, and sensor failure must be avoided.

4.2. Sensitivity

Sensitivity reflects how effectively a sensor converts mechanical strain into a measurable electrical signal and is quantified by the gauge factor (GF):

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

Figure 3
Performance metrics of wearable and stretchable strain sensor

for resistive-type sensors, where ΔR is the change in resistance, R_0 is the baseline resistance, and ε is the applied strain. Resistive sensors often achieve ultrahigh sensitivity through microstructural engineering. For example, fragmented CNT–PDMS composites exhibit GF up to 107 at 50% strain, while open-mesh graphene–PDMS sensors achieve GF values exceeding 88,000 at high strains due to sliding and disconnection mechanisms within the nanostructure [13].

Capacitive-type sensors, on the other hand, have GF defined as:

$$GF = \frac{\Delta C/C_0}{\varepsilon} = \frac{C - C_0}{C_0 \varepsilon} \tag{3}$$

where C_0 and C represent the capacitance before and after strain, respectively. Although most capacitive sensors have GF < 1 due to geometric limitations, hybrid designs using ionic hydrogels and conductive nanofilms can reach values up to 165 under ultrahigh strain [92]. Optical strain sensors, which detect mechanical deformation through changes in light propagation, offer sensitivities quantified as output power loss per strain unit, with reported values ranging from 3.6 to 18.75 $\mathrm{dB{\cdot}\varepsilon^{-1}}$.

4.3. Linearity

Linearity determines how directly the sensor output correlates with the applied strain. High linearity ensures accurate calibration and reliable measurements across the sensor's working range. It is quantified by the coefficient of determination (R^2):

$$R^2 = 1 - \frac{\sum_i (y_i - \hat{y}_i)^2}{\sum_i (y_i - \bar{y})^2} \tag{4}$$

where y_i are the measured output values, $\hat{y}_i$ are predicted values from linear regression, and $\bar{y}$ is the mean of the measured values. A R^2 value approaching 1 indicates a strong linear correlation. Nonlinearity arises from asymmetric microcrack propagation, separation of conductive networks, or tunneling effects at high strain. Innovations in sensor design, such as PU-PVDF composites, have achieved linearity >0.98, demonstrating stable performance under repeated stretching cycles [1, 15].

4.4. Hysteresis

Hysteresis describes the difference in sensor output between loading and unloading cycles. It is influenced by the viscoelastic behavior of the polymer substrate and interactions at the nanofiller–polymer interface:

$$DH\,(\%) = \frac{A_S - A_R}{A_S} \times 100\% \tag{5}$$

where A_S and A_R are the areas under the stretching and releasing curves, respectively. Significant hysteresis can compromise the accuracy of dynamic measurements. Minimizing hysteresis involves optimizing polymer–filler interactions and ensuring uniform nanofiller dispersion. Advanced sensors, such as VACNT–PDMS composites, achieve negligible hysteresis while retaining high sensitivity [18–20].

4.5. Response and recovery time

The response (t_{resp}) and recovery (t_{rec}) times quantify how quickly a sensor reacts to strain and returns to baseline after unloading:

$$t_{resp} = t_{signal\ rise\ to\ 90\%\ of\ max}, \quad t_{rec} = t_{signal\ decay\ to\ 10\%\ of\ max} \tag{6}$$

Short response times are essential for real-time monitoring in wearable electronics. Polymer-based sensors often exhibit slower response due to viscoelastic effects, but hybrid structures combining conducting polymers with carbon nanostructures or hydrogels have reduced reaction times to as low as 22 ms [13–18].

4.6. Overshooting

Overshooting occurs when the sensor output temporarily exceeds its steady-state value due to rapid stress relaxation of the polymer matrix. This phenomenon is influenced by viscoelasticity, strain rate, and sensitivity and can indirectly indicate nonlinearity. Multi-linear zone calibration strategies are commonly employed to mitigate overshooting and maintain accurate sensing.

4.7. Dynamic durability

Dynamic durability assesses the ability of a sensor to maintain performance under repeated stretching–releasing cycles. The relative resistance change over N cycles is used to evaluate stability:

$$Dynamic\ Stability = \frac{\Delta R}{R_0}\Big|_{N\ cycles} \tag{7}$$

Sensor degradation can result from polymer fatigue, filler fracture, or tunneling effects. Optimized filler dispersion, substrate selection, and mechanical compatibility can significantly improve cycle life. Some 3D-printed sensors withstand over 5000 cycles with minimal performance loss, highlighting their suitability for long-term wearable applications [21].

4.8. Design rules that connect parameters to real-world performance

The design parameters in Section 4 should not be optimized independently, because wearable deployment is typically limited by a small set of coupled constraints: stability on skin, repeatability over cycling, and robustness to sweat/humidity. The following design rules summarize the dominant couplings observed across sensor classes:

1) **High GF is not sufficient:** For healthcare, stability (low drift/hysteresis) and calibration repeatability often dominate utility over peak sensitivity.
2) **Hysteresis and drift are system level:** They depend on material viscoelasticity, filler network evolution, encapsulation, and skin-interface mechanics; they must be managed via structure + packaging + calibration, not only by materials.
3) **Dynamic vs quasi-static signals require different mechanisms:** Piezo/tribo excel in transient events, while resistive/capacitive platforms typically provide more interpretable continuous strain baselines.
4) **Reporting must match use conditions:** Strain-rate dependence, sweat/humidity exposure, mounting method, and long-term cycling should be reported to enable fair comparison and translation.

5. Structural and Device Design Innovations

The performance of wearable and stretchable strain sensors is strongly influenced by structural design, which complements material selection and sensing mechanisms. Advanced architectures improve stretchability, sensitivity, linearity, and durability, enabling sensors to function reliably on complex, moving surfaces such as the skin, joints, and muscles [1, 15].

5.1. Hierarchical and microstructured designs

Hierarchical micro- and nanostructures are widely adopted to enhance sensor performance. Patterns such as pyramids, micro-domes, and porous networks concentrate stress at targeted locations, improving sensitivity while maintaining linear response over large strains. For instance, pyramidal microstructures localize deformation at the tips of the conductive regions, increasing the GF without introducing excessive hysteresis. Dome-shaped microstructures distribute strain more uniformly across the sensing surface, improving durability and repeatability under repeated deformation cycles [22].

Porous networks are another key innovation, particularly for composite sensors where conductive fillers are embedded in elastomers. These networks allow the material to stretch while maintaining continuous conductive pathways, thereby preserving sensor output even under high strain. Nanowire networks further enhance performance by forming percolation pathways that respond sensitively to minute deformations [24]. Collectively, these hierarchical designs provide a balance between sensitivity, linearity, and mechanical resilience.

5.2. Serpentine and meander interconnects

While flexible substrates provide stretchability, conductive elements such as metallic nanowires or thin films often remain relatively rigid. Serpentine and meander interconnects overcome this limitation by introducing geometric flexibility. These interconnects unfold under strain, distributing stress along their length and maintaining electrical connectivity. Serpentine patterns allow materials to stretch over 100% without failure, while meander patterns reduce local stress concentration, enhancing long-term durability [12].

These interconnect designs are particularly useful for multi-electrode sensor arrays, where maintaining uniform signal transmission across large, deformable areas is critical. By combining serpentine interconnects with stretchable substrates, devices achieve high conformability, making them suitable for wearable applications such as joint motion tracking, respiration monitoring, and human–machine interfaces.

5.3. Liquid-metal channels

Embedding liquid metals like EGaIn or Galinstan into elastomeric channels has emerged as a transformative approach for ultra-stretchable sensors. These channels are inherently self-healing and maintain electrical conductivity even under extreme deformation (>200% strain). The liquid metal flows within the channel during stretching, preventing rupture and signal loss. However, practical implementation requires careful encapsulation to prevent leakage, and fabrication can become complex at microscale dimensions. Despite these challenges, liquid-metal designs are particularly valuable for high-deformation sites such as elbows, knees, and other joints, where traditional solid conductors would fail [34].

5.4. Structure–material co-design for scalable, wearable deployment

Structural innovations (microstructures, serpentine interconnects, and liquid-metal channels) are most effective when paired with material systems that support repeatable deformation modes and robust encapsulation. For example, microstructured elastomers can enhance sensitivity by strain localization, but they can also amplify hysteresis if the sensing layer undergoes irreversible microcrack evolution; thus, microstructures should be paired with conductive architectures that recover elastically over the target strain range. Serpentine interconnects reduce local stress and improve durability for arrayed devices, but they place constraints on electrode materials and packaging to prevent delamination under repeated motion. Liquid-metal channels achieve ultra-stretchability, but translation depends on leakage-resistant channel design and manufacturable sealing/encapsulation.

Accordingly, we emphasize structure–material co-design: the best-performing wearable sensors are rarely optimized by materials alone or structures alone; rather, they are enabled by combinations that minimize dominant failure modes (drift, humidity sensitivity, delamination, leakage) while remaining compatible with scalable fabrication routes.

5.5. Research gaps and testable hypotheses

Despite rapid progress, several gaps limit translation from proof-of-concept devices to clinically reliable systems. Based on the synthesis in Sections 2–5, we propose the following testable hypotheses that can guide next-generation research:

Drift vs architecture: For polymer–nanofiller piezoresistive sensors, stabilized conductive architectures (segregated or aligned networks) reduce long-term drift more effectively than increasing filler loading alone, when tested under identical cycling and humidity conditions.

Packaging as primary control: For triboelectric wearables, humidity-induced signal variability is more strongly reduced by barrier-layer encapsulation and surface protection than by tribo-material selection alone, under standardized humidity exposure tests.

Hybrid redundancy improves calibration: Multimodal hybrids (e.g., resistive + capacitive or tribo + resistive) improve calibration robustness across users by enabling self-consistency checks and artifact rejection compared with single-modality sensors.

Structure–material co-design: Microstructured designs improve sensitivity without sacrificing repeatability only when paired with sensing layers that exhibit near-elastic recovery (low viscoelastic relaxation) across the target strain-rate range.

Scalability penalty: The primary degradation during scale-up is not peak sensitivity but lot-to-lot variability (calibration spread and drift), which can be minimized through process control and in-line QC metrics.

6. Applications in Healthcare: Case-Study View, Validation, and Translation Constraints

Wearable and stretchable strain sensors have found widespread application in healthcare monitoring, offering continuous, noninvasive, and high-fidelity measurement of physiological signals. By integrating flexible substrates, advanced sensing mechanisms, and innovative structural designs, these sensors provide real-time insights into human health, improving diagnostics, rehabilitation, and personalized care [1, 16]. For healthcare applications, sensor performance must be interpreted in the context of (i) the strain regime (quasi-static vs dynamic), (ii) wear conditions (motion artifacts, sweat/humidity, adhesive creep), and (iii) validation against clinical references (e.g., respiratory belts/spirometry proxies, goniometers/motion capture, standardized task outcomes). Accordingly, the subsections below highlight not only representative use-cases but also the dominant failure modes (drift, hysteresis, parasitics, humidity sensitivity) and the mitigation strategies used in practice (encapsulation, stable mounting, calibration, and signal processing including denoising).

6.1. Vital signs monitoring

One of the most common applications of wearable strain sensors is the measurement of vital signs such as respiration rate, heart rate, and pulse. Piezoresistive and capacitive sensors embedded in chest straps or adhesive patches can detect minute skin deformations caused by breathing or heartbeat. Hierarchical microstructures amplify strain at conductive regions, improving the signal-to-noise ratio and enabling accurate detection even under subtle physiological movements [44].

Piezoelectric and triboelectric sensors can operate in self-powered modes, capturing dynamic changes in heartbeat or respiratory motion without the need for external power sources, which is particularly useful for remote monitoring or wearable patches in low-resource settings [15].

Clinical validation and robustness: Translation-ready studies typically validate respiration and pulse-related metrics against reference measurements such as respiratory belts/spirometry proxies and ECG/PPG-derived rate (for rate-only claims), reporting error metrics (e.g., Mean Absolute Error (MAE)/Root Mean Square Error (RMSE)), robustness under motion (sitting/walking), and multi-hour stability. In practice, performance is often limited by baseline drift (viscoelastic relaxation in composites), mounting variability, and sweat/humidity exposure; mitigation commonly combines encapsulation, stable skin-interface design, and algorithmic baseline correction.

6.2. Motion and gait analysis

Wearable strain sensors are also applied to musculoskeletal monitoring, including joint motion, gait analysis, and posture tracking. Sensors placed on elbows, knees, or along muscles can detect bending, stretching, and contraction by measuring strain-induced changes in resistance, capacitance, or generated voltage. Serpentine interconnects and liquid-metal channels allow these sensors to maintain performance under large deformations, providing real-time data for rehabilitation, sports medicine, and injury prevention [1–3].

Clinical/rehab validation: For gait and rehabilitation, credible validation compares sensor-derived features (joint angle, step timing, gait symmetry) against goniometers, Inertial Measurement Unit (IMUs), or motion capture, and reports repeatability across sessions and sensitivity to placement shifts. Dominant artifacts include re-wearing calibration shifts and strain–pressure cross-sensitivity; multimodal fusion (e.g., strain + IMU, or resistive + capacitive) is frequently used to improve robustness.

6.3. Human–machine interfaces

Next-generation sensors enable human–machine interactions by translating physiological signals into digital commands. For example, gesture recognition systems use triboelectric or piezoresistive sensors integrated into gloves to detect finger movements, while prosthetic devices utilize strain sensors to interpret muscle contractions for control [18–20]. These applications demonstrate the potential of wearable sensors to enhance mobility and accessibility for patients with disabilities.

Performance reporting: Beyond proof-of-concept demonstrations, Human–Machine Interface (HMI) systems should report task metrics such as classification accuracy, latency, stability under repeated donning/doffing, and generalization across users. A key practical challenge is "domain shift" (user-to-user and session variability), motivating normalization, calibration routines, and lightweight adaptive models for robust control.

6.4. Wound healing and skin monitoring

Emerging applications include monitoring wound healing and skin health. Stretchable sensors can track swelling, deformation, or pressure changes around wounds, providing clinicians with continuous feedback without invasive procedures. Capacitive and optical sensors are particularly suited for these applications due to their high sensitivity, ensuring accurate measurements even in complex clinical environments [16].

Regulatory/clinical context: Wound- and skin-adjacent wear requires multi-day skin compatibility and stable adhesion; therefore, studies should report irritation/comfort considerations and trend reliability under moisture exposure. If optical methods are discussed, it is important to distinguish optical strain sensing from optical physiological sensing (PPG), which can exhibit demographic performance bias; equity-aware validation requires diverse cohorts and artifact-aware processing when PPG-type readouts are used.

6.5. Clinical validation, regulatory translation, and equity considerations

Clinical translation of wearable strain-sensing systems requires evidence that performance persists beyond controlled benchtop tests; a translation-oriented validation checklist for such studies is provided in Table 5. Accordingly, validation should be framed around agreement with reference standards (gold standard or widely accepted comparators) and robustness under realistic wear [2]. For vital signs and motion monitoring, studies should report (i) comparison to a suitable reference (e.g., respiratory belts/spirometry proxies for respiration, goniometers/IMUs/motion capture for joint kinematics), (ii) cohort size and subject variability, (iii) sensitivity to placement and mounting conditions (adhesive type, contact pressure, garment tension), and (iv) multi-hour to multi-day stability, including baseline drift, recalibration frequency, and repeatability across re-wearing. In addition, reporting should explicitly include wear-condition stressors (sweat/humidity, temperature fluctuation, and motion artifacts) because these dominate long-term accuracy in real environments. From a regulatory standpoint, devices intended for medical monitoring may fall under risk-based pathways (often including Class II devices cleared through a 510(k) process depending on intended use and claims), so translation-oriented studies should document safety considerations (skin compatibility for prolonged contact, electrical/thermal safety where applicable)

Table 5
Translation-oriented validation checklist for wearable strain-sensing studies

Item	What to report	Why it matters clinically
Reference standard	Comparator method (belt/spirometry proxy; IMU/goniometer/motion capture; clinical labels)	Establishes credibility and meaningful accuracy
Cohort design	N subjects, demographics, inclusion/exclusion criteria	Generalizability beyond a single user
Mounting sensitivity	Adhesive/garment type, contact pressure, placement repeatability	Major source of error in wearables
Long-wear stability	Drift over hours/days, recalibration frequency, re-wear repeatability	Determines real-world utility
Wear stressors	Sweat/humidity, temperature, motion artifacts; testing conditions	Controls the “lab vs real life” gap
Safety/comfort	Skin irritation, breathability, materials exposure, thermal/electrical safety	Required for prolonged wear and clinical adoption
Regulatory alignment	Intended use (wellness vs clinical), high-level pathway awareness	Avoids over-claiming; guides study design
Equity/bias reporting	If PPG/optical vitals: subgroup metrics; if strain-on-skin: adhesion/skin compliance variability	Prevents biased performance and improves trust

and define the intended use case (wellness tracking vs clinical decision support). Finally, equity considerations should be addressed when optical physiological sensing (e.g., PPG) is discussed, since optical absorption/scattering and motion artifacts can produce demographic performance disparities; therefore, validation should include diverse cohorts and subgroup performance reporting. For purely mechanical strain sensing on skin, analogous equity-related variability can arise through differences in skin compliance, sweating, and adhesion across users, motivating standardized mounting protocols and reporting of calibration spread across participants [92].

7. Broader Applications of Wearable and Stretchable Strain Sensors

Wearable and stretchable strain sensors are no longer limited to healthcare applications. Their combination of flexibility, stretchability, and high sensitivity has enabled a wide range of uses in various fields. These sensors can provide continuous, real-time monitoring in dynamic and irregular environments, making them suitable for applications beyond physiological tracking [1, 17]. Beyond health monitoring, the same design requirements discussed in Sections 2–5—large strain compliance, fatigue resistance, environmental robustness, and scalable integration—are equally decisive in robotics, infrastructure, and smart textiles. We therefore summarize broader applications using the same translation lens (dominant signal regime, deployment constraints, and integration considerations), rather than presenting unrelated examples.

7.1. Human–machine interaction and soft robotics

Stretchable strain sensors are increasingly used in human–machine interaction systems and soft robotics. When integrated into robotic skins, artificial muscles, or flexible grippers, the sensors provide proprioceptive feedback that allows robots to detect bending, stretching, and tactile forces accurately. Gesture recognition gloves with strain sensors can translate finger and wrist movements into digital commands, enabling natural control of robotic arms, drones, and virtual reality systems. Their mechanical compliance ensures reliable performance under repeated deformation, which is critical for soft robots designed to interact safely with humans or delicate objects. These sensors are also used in assistive devices for people with disabilities, where precise motion tracking improves accessibility and control [1, 8].

7.2. Sports performance and ergonomics

In sports and physical training, wearable strain sensors provide detailed biomechanical insights that were previously obtainable only with large motion-capture systems. Sensors mounted on joints or muscles can measure range of motion, joint angles, strain patterns, and muscle activity, allowing coaches and athletes to optimize performance while reducing the risk of injury. Beyond sports, these sensors are applied in occupational ergonomics to monitor posture and repetitive movements in workplaces. By detecting improper lifting techniques or sustained awkward positions, they help prevent musculoskeletal disorders and improve workplace safety. Wearable sensors provide continuous feedback in real-world conditions, with performance often benchmarked against reference motion-capture or IMU systems in field-like settings, unlike traditional laboratory-based systems [90].

7.3. Structural health monitoring

Stretchable strain sensors are also applied in structural health monitoring of civil, aerospace, and mechanical infrastructures. Their flexibility allows them to conform to curved or irregular surfaces, which is useful for monitoring stress, strain, vibration, and fatigue in bridges, pipelines, aircraft wings, composite materials, and wind turbine blades. Embedded sensors can detect micro-cracks or early deformation, providing real-time alerts that prevent catastrophic failure. Compared to conventional rigid sensors, stretchable sensors are lightweight, low power, and easier to integrate into structural components [93].

Embedded sensors can detect micro-cracks, early deformation, and fatigue evolution, enabling real-time alerts and predictive maintenance. In recent work, stretchable and porous conductive architectures (including MXene-based foams and composites) have been explored to improve conformal contact, sensitivity at low strain, and durability for long-term monitoring on rough structural surfaces, complementing established elastomer–nanofiller and fiber-based strain-sensing approaches [93, 94]. Compared to conventional rigid gauges, stretchable sensors can offer lightweight integration and wider-area coverage; however, translation to infrastructure requires demonstrating long-duration stability under temperature cycling, humidity, UV exposure, and adhesive aging, in addition to strain sensitivity [95].

7.4. Consumer electronics and smart textiles

Wearable strain sensors are increasingly incorporated into smart textiles and interactive consumer electronics. In clothing, footwear, and accessories, they can measure body motion, detect posture changes, and monitor pressure distribution. Smart textiles enable gesture control, activity recognition, personalized fitness tracking, and adaptive clothing that responds to the wearer's movements. In consumer electronics, flexible sensors facilitate touch-sensitive surfaces for curved displays, foldable devices, and flexible keyboards, enhancing user interactivity and durability [18]. Many of these nonmedical deployments also align with scalable manufacturing routes (e.g., printing/lamination and roll-to-roll compatible processes), which can reduce cost and improve reproducibility for large-area smart textiles and flexible surfaces [20].

7.5. Environmental and industrial monitoring

Stretchable strain sensors are finding applications in environmental and industrial monitoring. In environmental systems, they can detect soil displacement, structural vibrations, water flow pressure, or early signs of landslides, which aid in disaster prevention and management. In industrial settings, sensors monitor machinery vibration, equipment wear, and structural fatigue, providing predictive maintenance data that reduces downtime and improves safety. Their ability to conform to irregular surfaces and operate under mechanical stress makes them suitable for environments where conventional rigid sensors would fail [12, 56].

8. Challenges of Wearable and Stretchable Strain Sensors

Wearable and stretchable strain sensors offer significant potential for healthcare, robotics, and other emerging fields, yet they face a few technical and practical challenges that have limited their widespread deployment. One of the most pressing issues is maintaining mechanical durability over extended periods of use. Because these sensors are designed to undergo repeated stretching, bending, or compression, the soft substrates and composite materials used in their construction are often prone to fatigue. Over time, this can lead to the formation of cracks, delamination between layers, or even loss of conductivity in the sensing elements, which in turn affects the reliability and accuracy of the sensor output. In addition, continuous mechanical stress can induce drift in the measured signals, making long-term monitoring less dependable unless careful material design and structural reinforcement are employed [23].

Another major challenge involves achieving a balance between sensitivity and stretchability. Sensors that are engineered to tolerate large deformations often exhibit a reduction in sensitivity because the conductive networks embedded within the substrate may become disrupted as they are stretched. Conversely, sensors designed to maximize sensitivity can fail when subjected to large strains or repeated stress cycles, which limits their operational range [95–97]. To overcome this trade-off, researchers are exploring strategies such as optimizing the dispersion and alignment of conductive fillers, developing hierarchical or microstructured architectures, and integrating hybrid materials that combine elasticity with high electrical performance. Achieving this balance is crucial for wearable sensors, particularly for applications such as precise motion tracking, physiological monitoring, or soft robotics, where both large-range deformation and high signal fidelity are required [97–100].

Signal stability is another critical concern, as wearable sensors are often used in dynamic environments where the device is subject to motion artifacts, variations in temperature, humidity, and other environmental factors. These external influences can introduce noise, baseline drift, or fluctuations in the measured signals, reducing both the reproducibility and reliability of the data. For instance, small changes in skin contact, sweat accumulation, or body movements can significantly affect the output of piezoresistive or capacitive sensors. Addressing these issues requires not only the development of robust sensing materials but also the implementation of sophisticated signal-processing algorithms, calibration routines, and encapsulation techniques to mitigate the impact of external disturbances and ensure accurate readings during real-world use.

Biocompatibility and skin integration present additional challenges, particularly for wearable devices intended for prolonged use. Sensors must be comfortable, nonirritating, and safe for direct contact with human skin, yet also mechanically robust and resistant to deformation. Chemically safe materials may be too stiff or bulky, compromising comfort and flexibility, while soft or adhesive materials may degrade over time, lose adhesion, or allow sweat and moisture to affect performance. Designing wearable sensors that strike the right balance between user comfort, mechanical resilience, and biocompatibility is essential for ensuring long-term usability, especially for medical monitoring or continuous health tracking applications [16].

Power consumption and data management also limit the effectiveness of wearable strain sensors in practical applications. While some sensing mechanisms, such as piezoelectric and triboelectric devices, can generate power through mechanical deformation, many sensors still rely on external power sources and supporting electronics for signal acquisition and wireless transmission [22]. Integrating energy-efficient electronics, compact wireless modules, and on-device data processing without increasing the size, weight, or rigidity of the sensor remains a substantial challenge. Efficient power management is critical to enable continuous, long-term monitoring without frequent battery replacement or recharging, particularly for applications in healthcare and sports where uninterrupted operation is necessary [45–47].

Finally, scalability and manufacturing remain major hurdles for bringing wearable strain sensors to the market. Fabrication of high-performance sensors often requires complex material synthesis, precise micro- or nanoscale structuring, and careful assembly to ensure consistent performance. Producing these devices in large quantities while maintaining uniform quality, reliability, and cost-effectiveness is a persistent challenge. Advances in scalable

Table 6
Lab-scale vs scalable fabrication routes for wearable strain sensors

Category	Lab-scale prototyping (typical)	Scalable manufacturing (typical)	What usually breaks during scale-up
Patterning of conductors/composites	Drop-casting, manual coating, stencil deposition	Screen/gravure/inkjet printing, roll-to-roll coating/lamination	Uniformity, batch-to-batch variability, percolation-network consistency
Microstructures	Hand-molded textures, lab lithography	Continuous embossing, scalable molding, replicated stamps	Feature fidelity vs throughput; defect rate
Device assembly	Manual wiring/encapsulation	Automated lamination, pick-and-place, in-line curing	Yield, alignment tolerance, packaging integrity
Encapsulation/skin interface	Lab silicone encapsulation; tape adhesives	Medical-grade encapsulation + controlled adhesive stacks	Sweat ingress, delamination, irritation risk
QC/calibration	Small-sample characterization	Statistical process control + calibration protocols	Drift/hysteresis variation across lots

Table 7
Practical barriers to clinical validation and regulatory readiness

Domain	What must be demonstrated	Typical failure mode in wearables
Measurement validity	Agreement vs reference (gold standard), across users	Motion artifacts, placement sensitivity
Repeatability and drift	Stable baseline and calibration over days/weeks	Viscoelastic relaxation, hysteresis, sweat effects
Biocompatibility and comfort	Safe prolonged skin contact, minimal irritation	Adhesive degradation, skin irritation
Environmental robustness	Performance under humidity/sweat/temperature	Signal shifts, delamination
Manufacturing reproducibility	Consistent performance across lots	Nonuniform materials/assembly
Data integrity	Reliable logging/wireless transfer, QC flags	Dropouts, untracked calibration changes

Table 8
Mechanisms–materials–performance tendencies–application mapping

Mechanism	Common material systems	Typical performance tendencies	Best-fit application domains
Piezoresistive	Elastomer + CNT/-graphene/AgNW; hybrid composites	High sensitivity; drift/hysteresis risks; simple readout	Respiration, pulse waveforms, joint motion
Capacitive	Dielectric elastomers + patterned electrodes	Stable/low power; sensitive to noise/parasitics	Posture/motion tracking; arrays; muscle activity
Piezoelectric	PVDF, ceramic/polymer composites	Strong for dynamic strain; limited static response	Gait, dynamic motion; event detection
Triboelectric	Layered polymer interfaces, microstructured surfaces	Self-powered dynamic signals; environment-dependent outputs	Motion-triggered sensing; HMI gestures
Optical strain sensing	Flexible waveguides, polymer optical fibers/FBGs	EMI advantages; packaging/cost complexity; high resolution	Joint angle, deformation mapping, respiration belts
Optical physiological sensing (PPG)	LED + photodiode; skin optics	Measures blood-volume/absorption changes, not strain	Heart rate, SpO_2, perfusion trends

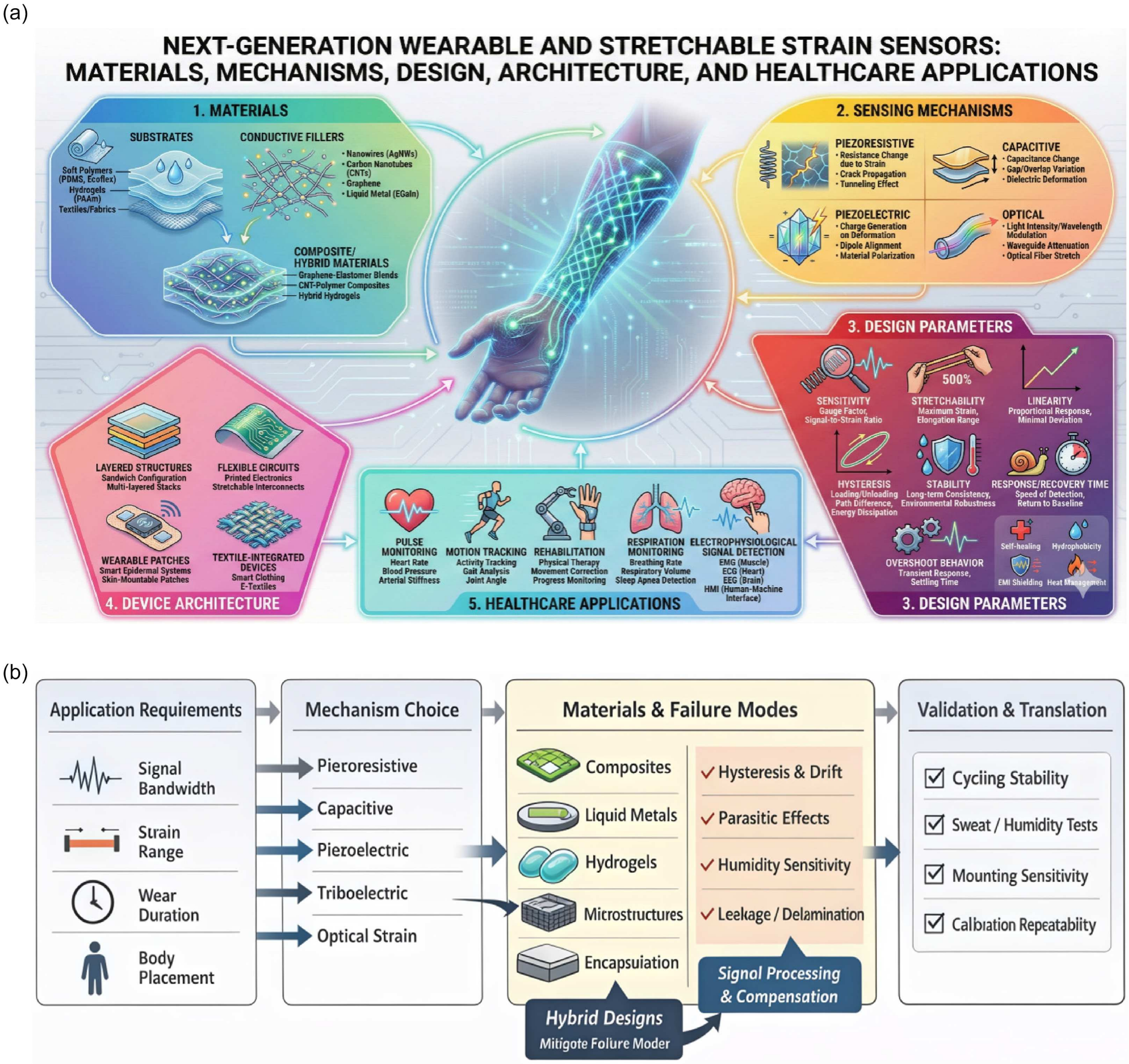
manufacturing methods, such as printing technologies, roll-to-roll processing, and automated assembly, are being explored, but achieving the combination of performance, affordability, and mass production capability continues to be an area of active research.

8.1. Scalability and clinical translation considerations

Although many wearable strain sensors demonstrate excellent benchtop performance, translation to real-world healthcare devices is frequently limited by (i) manufacturing scalability and reproducibility, (ii) long-term stability on skin under sweat/humidity and repeated deformation, and (iii) validation requirements (repeatability, calibration stability, and safety/biocompatibility).

To address this gap, we summarize fabrication routes from lab prototyping to scalable manufacturing and outline practical validation milestones needed before clinical deployment (see also Tables 6, 7, and 8).

Figure 4
(a) Integrated roadmap of next-generation wearable and stretchable strain sensors for healthcare. (b) Design and translation workflow for wearable and stretchable strain sensors



9. Conclusion

Next-generation wearable and stretchable strain sensors represent a powerful convergence of materials science, device engineering, and healthcare. Through the development of soft, conformable substrates, advanced conductive fillers, and microstructured hybrid composites, today's strain sensors achieve a level of flexibility and sensitivity previously unattainable. These sensors leverage a variety of transduction mechanisms, including piezoresistive, capacitive, piezoelectric, triboelectric, and optical principles, which have demonstrated their potential in continuous physiological monitoring, human-motion tracking, and wearable interfaces, as illustrated in Figure 4.

However, despite remarkable advances, significant challenges remain. Durability under repeated deformation, trade-offs between sensitivity and stretchability, signal stability in real-world conditions, biocompatibility, power constraints, and scalable manufacturing are all critical hurdles that must be overcome. Addressing these issues will require not only fundamental research but also system-level innovation and cross-disciplinary collaboration.

The future of wearable strain sensors is bright. With ongoing efforts in multifunctional material development, low-power and self-powered designs, machine learning (ML)–enabled data processing, and manufacturing scale-up, these devices are poised to transform how we monitor health, interact with machines, and integrate electronics into our daily lives. By bridging the gap between proof-of-concept lab devices and clinically validated systems, the next wave of wearable strain sensors has the potential to revolutionize personalized healthcare and enable truly intelligent, skin-like devices.

Recommendations

To accelerate clinical and commercial translation of wearable and stretchable strain sensors, future work should prioritize standardized validation, failure-mode–driven design, and scalable manufacturing, rather than incremental gains in peak sensitivity. First, we recommend adopting application-matched benchmarking protocols that report strain-rate dependence, long-term cycling stability, humidity/sweat exposure, and mounting sensitivity, because these factors dominate real-world performance yet remain inconsistently reported across the literature. Second, materials and structures should be selected using a failure-mode lens: for piezoresistive composites, the central challenge is hysteresis and drift from viscoelastic relaxation and evolving conductive pathways, motivating stabilized architectures (segregated networks, elastic recovery designs), robust encapsulation, and routine calibration; for triboelectric platforms, humidity-driven variability motivates barrier coatings and multimodal redundancy; for capacitive systems, parasitics and motion-induced wiring artifacts motivate shielding and geometry-aware calibration; and for liquid-metal architectures, leakage risk and long-term skin safety motivate channel design and biocompatible encapsulation. Third, algorithmic processing should be framed as a mitigation tool, not a buzzword: ML-based denoising and drift compensation, motion-artifact detection, and domain adaptation across users should be systematically evaluated against simple baselines (filters, normalization) and reported with transparent metrics and ablation studies. Fourth, scalability must be addressed explicitly by demonstrating manufacturing-compatible routes (screen/inkjet/-gravure printing, roll-to-roll coating/lamination, embossing of microstructures) and reporting lot-to-lot variability and yield, since calibration spread often becomes the limiting factor during scale-up. Fifth, healthcare deployment requires attention to regulatory readiness and equity: studies should include diverse testing cohorts, report performance across skin types and body morphologies where relevant, and document skin compatibility for prolonged wear; data handling should follow ethical AI principles (privacy, security, bias assessment), especially as strain-sensor outputs are integrated into telehealth workflows. Sixth, sustainability should be treated as a design constraint by considering e-waste, material recoverability, and environmentally responsible choices for nanomaterials and encapsulants. Finally, beyond current paradigms, emerging directions such as quantum-enhanced sensing concepts may offer long-term opportunities for ultra-precise readouts, but near-term impact will likely come from integrating robust materials, packaging, scalable fabrication, and validated analytics into complete wearable systems. Overall, the most valuable next-generation contributions will be those that convert laboratory prototypes into repeatable, manufacturable, and clinically validated devices.

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Ethical Statement

This study does not contain any studies with human or animal subjects performed by any of the authors.

Conflicts of Interest

The authors declare that they have no conflicts of interest to this work.

Data Availability Statement

Data sharing is not applicable to this article as no new data were created or analyzed in this study.

Author Contribution Statement

Boluwatife Oluwasegun: Formal analysis, Investigation, Writing – original draft, Visualization. **Oluwaseun Oni-Adimabua:** Conceptualization, Writing – original draft. **Kyrian Odo:** Conceptualization, Methodology. **Opeyemi Akanbi:** Conceptualization, Writing – review & editing, Supervision, Project administration. **Aimanose Eigbedion:** Methodology, Formal analysis. **Joseph Igbama:** Validation, Investigation. **Joseph Alieme:** Validation, Investigation. **Folakemi Ijagbemi:** Validation, Resources. **Ayotunde Igbekele:** Resources, Visualization. **Michael Adelere:** Supervision, Project administration. **Hakeem Oyeshola:** Formal analysis, Writing – review & editing.

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