REVIEW ARTICLE

Strain Sensing Technologies: Recent Developments in Materials, Performance, and Applications

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Abstract

Strain sensors have become fundamental to contemporary sensing technology, driven by the growing demand for flexible, sensitive, and durable transducers across medical care, robotics, structural monitoring, and human-machine interfaces. Rapid advances in strain sensing performance—in terms of gauge factor, detection limit, working range, and operational stability—have enabled next-generation wearable and implantable sensing systems. This mini review examines sensing mechanism classifications, key performance parameters, sensor architectures, and application domains, with a unified focus on how sensing capability can be maximised for real-world deployment. Particular emphasis is placed on gauge factor enhancement, wide-range stretchability, ultrafast response time, and long-term sensing stability. The contributions of emerging material systems—including self-healing polymers, hybrid nanomaterials, graphene derivatives, carbon nanotube composites, and conductive hydrogels—are evaluated specifically in terms of these sensing performance metrics. Persistent challenges including biocompatibility, signal drift, packaging integration, and long-term operational stability are critically discussed alongside promising solutions. Finally, emerging directions toward multifunctional, self-healing, and scalable strain sensing platforms are outlined to guide future research and real-world deployment.

Keywords: strain sensors, sensing performance, gauge factor, flexible wearable sensing, human motion monitoring, biomedical sensing.

Submitted: 19 May 2025
Revised: 16 June 2025
Accepted: 18 June 2025
Published: 28 July 2025

Vol. 2, **No.** 3, 2025.
10.62762/TSCC.2025.665257

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1 Introduction

With advances in wearable and flexible sensing devices, next-generation strain sensors are of considerable significance for tracking health, human movement detection, robotics, and human-computer interaction. Sensors sense and convert the physical characteristics of their surroundings into measurable electrical signals. Flexible wearable strain sensors, in contrast to typical rigid sensors that are built on silicon substrates, offer

Citation
Bibi, M., Hassan, G., Asif, A., Shuja, A., Ahmad, H., Mushtaq, B., & Sajid, M. (2025). Strain Sensing Technologies: Recent Developments in Materials, Performance, and Applications. *ICCK Transactions on Sensing, Communication, and Control*, 2(3), 168–199.

improved flexibility, bio-compatibility, and mechanical durability. As a result, they are extremely well suited for use in biomedical applications [1]. Strain sensors have gained significant attention due to their ability to convert mechanical deformation into readable electrical signals, enabling a wide range of applications across various fields. Traditional strain sensors, typically based on metallic foils or semiconductor materials, offer high sensitivity and reliability but often suffer from limited flexibility, poor stretchability, and susceptibility to mechanical damage. In contrast, recent advancements have focused on developing next-generation strain sensors using novel materials and design strategies that offer enhanced mechanical compliance and multifunctionality [2]. The present investigation on strain sensors focuses on enhancing different parameters such as sensitivity, linearity, and stability, and incorporating self-healing characteristics and multi-functionality to ensure long lifespan and real-time health monitoring [3]. For example, strain sensors are extensively utilized in a variety of industrial solutions, including the following: continuous tracking of blood pressure [4, 6], implantable biomedical devices [5], and human-machine interfaces (HMI) [7, 8]. Long-term sensing reliability is a critical requirement for wearable strain sensors deployed in continuous health monitoring and human motion detection. Self-healing materials address this requirement by restoring sensing functionality after mechanical damage, thereby maintaining uninterrupted signal output throughout extended use [9, 10]. Energy autonomy represents another key challenge for continuous sensing: self-powered strain sensors that integrate piezoelectric or triboelectric energy harvesting eliminate the need for external power supplies, enabling truly untethered sensing operation [11]. Furthermore, conformal contact with curved and dynamic skin surfaces is essential for accurate strain acquisition; flexible and biocompatible substrates ensure intimate sensor-skin coupling, while advanced deposition techniques guarantee uniform sensing layer formation and reproducible signal output [12].

Achieving high sensitivity, broad sensing range, fast response, and long-term signal stability are the central challenges in strain sensor design. The choice of conductive constituents directly determines these sensing capabilities: carbon-based nanomaterials such as graphene and carbon nanotubes [13] provide exceptional sensitivity owing to their high aspect ratio and percolation network structure; metal nanoparticles [14] offer tunable conductivity and low detection limits; and two-dimensional transition metal carbides/nitrides (MXenes) [15] enhance signal stability through strong interfacial bonding with polymer matrices. Each material system presents a distinct sensing performance profile, and the selection must be guided by the requirements of the target sensing application. Notably, the integration of self-healing capability, hybrid conductive networks, and multifunctional transduction schemes has opened up new frontiers in sensing technology. These strategies not only improve sensitivity and sensing durability but also allow a sensor to recover its signal transduction capability after mechanical damage, significantly extending the usable sensing lifetime and measurement reliability. Such innovations mark a clear departure from conventional sensing devices and represent a paradigm shift toward intelligent and adaptive sensing platforms.

Achieving simultaneously high stretchability, biocompatibility, and stable sensing output remains a central challenge for wearable strain sensors [16]. Beyond performance, practical deployment imposes additional requirements: sensors must operate reliably over extended periods without signal drift, survive repeated mechanical deformation cycles without degradation, and remain safe for long-term skin contact. Biodegradable sensing platforms [17] address the specific need for transient monitoring—where the sensor must perform accurately during a defined clinical window and then safely dissolve—while recyclable sensing hydrogels [18] extend the operational life of biomedical sensing systems by enabling functional restoration rather than disposal. These developments directly expand the range of physiological sensing scenarios that flexible strain sensors can address.

Self-healing capability addresses one of the most persistent limitations of wearable strain sensors: the irreversible loss of sensing function caused by mechanical damage during use. By restoring conductive pathways after rupture, self-healing sensing layers maintain signal integrity across a far greater number of operational cycles than conventional systems [19]. Combined with self-powered transduction mechanisms, this enables strain sensing platforms that are both operationally autonomous and mechanically robust—critical attributes for continuous, unattended physiological monitoring [20].

This review systematically examines recent advances in strain sensing technology. The paper is organised around sensing performance: classification of sensing mechanisms, quantitative performance parameters (gauge factor, detection limit, linearity, response time, hysteresis, and durability), sensor type architectures, and application-driven performance requirements. Material systems are discussed specifically in terms of their contributions to these sensing metrics. Finally, unresolved sensing challenges and future directions toward higher-performance, more reliable strain sensing platforms are presented.

2 Classification of Sensors

In this section, we briefly survey the broader sensor landscape—including pressure, humidity, temperature, piezoresistive, and triboelectric sensors—to contextualise strain sensing within the wider field of flexible sensing. Material systems, fabrication approaches, and transduction mechanisms developed for these sensing modalities increasingly overlap with those of strain sensors, particularly in multifunctional and integrated wearable platforms. Strain sensors are then examined in detail in Section 2.6 and throughout the remainder of the article. These devices record data in real time, enabling automation, monitoring, and control across technological realms. Sensor technology drives innovation, improving efficiency and functionality in everyday technology and systems. These subsections analyse some fundamental sensors used in daily life, shown in Figure 1.

2.1 Pressure Sensor

The force per unit area on the surfaces of pressure sensors is measured and then translated into electrical signals for control or monitoring. The pressure sensor takes data from the real world and converts them into an electrical signal; the user may then make sense of the data on the screen [25]. The origins of pressure sensing can be traced to Evangelista Torricelli's mercury barometer (1643), which established the foundational principle of relating fluid column height to pressure. Electrical pressure sensors did not emerge until the early twentieth century, and it was not until the 1940s that strain gauge-based pressure sensors achieved practical measurement accuracy and adaptability. Pressure sensing performance—in terms of accuracy, detection range, and power efficiency—has improved substantially over subsequent decades, driven by advances in transduction mechanisms and miniaturisation technologies [26].

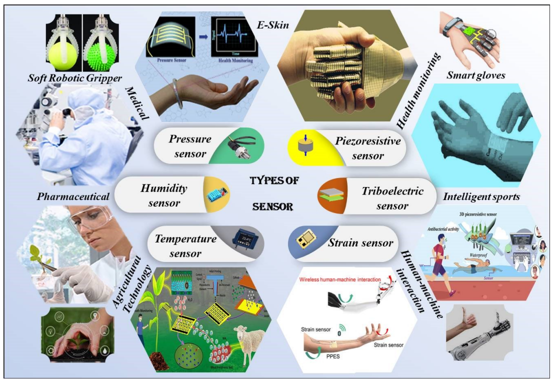

Figure 1. Robotic mechanism as robotic grippers. Reproduced with permission [20] Copyright © 2018 WILEY-VCH. Pharmaceutical industry. Wearable strain sensor integrated into clothing for human respiratory monitoring [21]. Flexible strain sensors for human arm/hand gesture capturing and recognition. Reproduced with permission [22] Copyright © 2020, IEEE. Robotic arm. 3D piezoresistive wearable sensor. Reproduced with permission [23] Copyright © 2023 Elsevier. Health monitoring sensor adhered on pulse. Smart gloves. E-skin. Medical application. Reproduced with permission [24] Copyright © 2015 Royal Society of Chemistry.

A number of devices employ liquid or pressure. Pressure-based IoT systems are easier to build using these sensors. Pressure deviations alert the system administrator. Factory and water and heating system maintenance benefit from these sensors' pressure change detection. Progressive improvements in signal conditioning, miniaturisation, and sensing element design have made pressure sensors more accurate, energy-efficient, and suitable for IoT integration. From simple instruments, pressure sensors are now essential parts of modern automotive, medical, environmental, and Internet of Things technology. Pressure sensors using MEMS technology are popular because they have advantages over electromechanical sensors. MEMS technology offers small size, low mass, low price, better functionality, extensive integration, low energy consumption, wider temperature range, and greater output signal [27, 28]. Table 1 compares pressure sensors based on different reported materials.

2.2 Humidity Sensor

Humidity refers to the amount of water vapor present in a gas. There are several ways to measure humidity, but the two most common are relative and absolute humidity. Quantifying the amount of water vapor in each volume uses absolute humidity. The ratio of the present water vapor pressure to the water vapor pressure needed for saturation at a

Table 1. Comparison of pressure sensors based on different materials.

Ref.	Materials	Sensing Mechanism	Detection Rate	Response	Sensitivity
[28]	Multi-layer textile materials	Pressure	0-120 kPa	N/A	0.0012 kPa^{-1}
[29]	PVDF/AgNWs/TiO_2	Capacitive	0–1500 kPa	166.9 ms	1.448 kPa^{-1}
[30]	MWCNT/PDMS	Capacitive	0.2 Pa	N/A	17.8 kPa^{-1}
[31]	AFICPS	Capacitive	5 Pa	11 ms	0.49 kPa^{-1}
[32]	Gr/PEDOT:PSS	Piezoresistive	N/A	42 ms	17.65 kPa^{-1}
[33]	RGO/PDMS	Piezoresistive	3.2 kPa	N/A	164.5 kPa^{-1}
[34]	PAAMPSA/PEDOT:PSS	Piezoresistive	N/A	19 ms	44.31 V kPa^{-1}
[35]	PEDOT:PSS fabric	Triboelectric	10 kPa	N/A	0.228 V/N

given temperature is known as relative humidity or RH. The temperature directly impacts the relative humidity. A hygrometer employs a humidity-sensitive material that changes physically or electrically with moisture. Water vapor is a big part of air and its concentration varies, thus it is vital to predict the exact amount for varied needs. Hygrometry is the art and science of humidity measurement. Nicolas Cryfts invented the hygrometer in 1450 to measure humidity. It monitored humidity using wool. Over a century later, this hygrometer was modified to measure moisture with a sponge instead of wool. After that, acetate, paper, hair, and nylon replaced sponge. As humidity sensors became increasingly popular, it was crucial to develop a technique to build them that were not expensive or harmful while offering accurate readings with low hysteresis, high sensitivity, and low power consumption. In response to industrial demand, capacitive, resistive, semi-conductive, optical, surface acoustic wave, and other humidity sensors were created . Capacitive sensors offer power consumption, range, sensitivity, and cost. However, they have difficulty connecting with external circuitry and have non-linear responses. Resistive humidity sensors offer superior linearity and simpler signal conditioning interfaces, though their sensitivity is temperature-dependent, requiring compensation for accurate humidity quantification. Low-cost, flexible resistive humidity sensors based on screen-printed graphene/polypyrrole/carbon black nanocomposite inks on paper substrates have also been reported, demonstrating stable response across a wide relative humidity range [42]. Metallic nanowire composites such as sandwich-structured Ag/Fe_3O_4 nanowire films have further demonstrated high sensitivity and multi-modal dynamic monitoring capability in flexible resistive humidity sensors. Electrospun nanofiber-based membranes, such as LiCl/cellulose nanofiber composites, have demonstrated wide-range and highly sensitive humidity detection through one-step fabrication [43]. Two-dimensional MXene/WS_2 heterostructure-based flexible humidity sensors have further demonstrated

Table 2. Comparison of humidity sensors based on different materials.

Ref. #	Sensing Materials	RH Range (%)	Response/Recovery Time	Sensitivity
[40]	Graphene/methyl-red composite	35–100	0.251 s / 0.35 s	96.36%
[41]	DMAc/LiCl	5–98	99 s / 110 s	up to 4191%
[42]	Graphene/PPy/CB nanocomposite	23–92.7	5 s / 7 s	∼12.2 Ω $\%RH^{-1}$
[43]	LiCl/Cellulose nanofiber	11–98	<30 s / <60 s	$\sim10^4$ % $decade^{-1}$ (impedance)
[44]	$Ti_3C_2T_x$ MXene/WS_2	11–97	4.65 s / 1.33 s	1707%
[47]	MWCNT/PVA	58–99	∼300 s / N/A	509%

ultrafast response and recovery times alongside high sensitivity across a wide relative humidity range [44]. Table 2 illustrates different nanomaterials-based humidity sensors and their responses. The material systems and fabrication strategies developed for humidity sensing—particularly conductive polymer composites, MXene heterostructures, and nanofiber substrates—are increasingly being co-integrated with strain sensing layers in multifunctional wearable platforms, motivating their inclusion in this review.

2.3 Temperature Sensor

A temperature sensor is a device that detects and monitors the thermal energy in its environment, transforming it into an electrical signal or digital output to measure temperature. Skin temperature sensing improves illness diagnosis, accident prevention, and environmental data collection. Normal body temperature is necessary for health, but variations might signify serious disease. Most people maintain a core temperature of 37–37.5 degrees Celsius. Temperature sensors measure skin temperature and changes in real time [45, 46]. Skin temperature can usually be monitored with thermometers. The biggest challenge is placing hard sensor devices against the skin to monitor conditions in tight spaces like garments or prosthetics. This approach is unsuitable for paediatric patients, as axillary thermometry is poorly tolerated in young children [47]. Skin-to-skin contact boosts temperature measurements. However, stiff temperature detectors cannot handle uneven surfaces. Flexible and wearable temperature sensor research is hampered by this issue. Wearable temperature sensors should be soft, flexible, biocompatible, lightweight, durable, and skin-friendly. Comfort matters, especially for long-term use [48].

Flexible temperature sensors include thermocouple, resistance, chromic, and thermistor. Seebeck effect powers many self-powered temperature sensors by converting temperature into electricity. These sensors have energy and temperature monitoring capabilities [49–54]. Resistance temperature detectors exploit the predictable variation of electrical resistance with temperature to achieve high measurement accuracy and stability, making them suitable for precision clinical monitoring. Thermocouples utilise the Seebeck effect to generate a voltage proportional to a temperature difference, enabling self-powered temperature sensing across wide temperature ranges without an external excitation source—a significant advantage for remote or implantable sensing applications [49–54]. Temperature sensors enhance daily comfort, safety, and efficiency. They control house heating and cooling using thermostats for comfort. Weather station temperature sensors give accurate weather data for forecasting and climate monitoring. Diagnostic medical thermometers detect body temperature correctly using sensors. Sensors regulate food storage and transit temperatures to preserve fragile items. Environmental control, health monitoring, and industrial process optimization are improved by temperature sensors [55, 56]. Table 3 illustrates the reported temperature sensor that are flexible and have enormous features.

2.4 Piezoresistive Sensor

A piezoresistive sensor is a sensor that alters its electrical resistance when exposed to mechanical stress or pressure, appropriate for measuring and identifying pressure fluctuations in electronic devices and applications in industry. The electron tunneling paths are affected by variations in the spacing between conductive fillers or particles under strain. Greater gaps decrease the potential of tunneling when the material extends, increasing resistance. Because of its extreme sensitivity to nanoscale separations, this tunneling technique allows for accurate strain monitoring [62, 63]. In the late 1990s, piezoresistive sensors used rigid metal ions or metallic substrates integrated with silicon circuits, which caused mechanical mismatches with human tissue, poor bio interface, high noise-to-signal ratios, inflammation, and metal toxicity. In the early 21st century, piezoresistive sensors were integrated with human skin to create tactile-perception-focused "electronic skin" (E-skin). Since 2020, piezoresistive sensors have improved in flexibility, biocompatibility, lightweight design, high sensitivity, stability, low cost, and reduced power consumption. These improvements have expanded their sensing applications into IoT, flexible touch displays, soft robotics, electronic textiles, and mobile healthcare, making soft piezoresistive sensors promising for future wearable sensing and long-term health monitoring [64, 65]. A broad range of sensing scenarios has been addressed through piezoresistive platforms, with conductive network design—spanning one-dimensional nanowires, two-dimensional graphene and MXene, and hybrid architectures—tuned to deliver specific combinations of sensitivity, working range, and mechanical compliance required by each application [66–69]. Using multiple graphene nanoplatelet layers on a polyurethane (PU) substrate, an alternative electronic skin sensor displayed a GF of 67 at 0.2% strain and 320 at 2% strain, capable of detecting both minute deformations and larger stimuli [70].

2.5 Triboelectric Sensor

Triboelectric sensor utilizes both electrostatic induction and the triboelectric phenomenon to transform mechanical input into electrical impulses. Triboelectric phenomena have been observed within nano-generators with the purpose of generating self-powered sensors. For wearable and portable electronics that demand battery-free operation, this feature is extremely beneficial [71]. Mechanical disturbances can affect time and place, which affect power generation. A human finger gains a positive charge (+) when it touches the sensor surface. The sensor's insulating surface holds negative charges for a long period since they are hard to induce. Raising the finger off the surface transfers positive charge to the electrodes [72]. The generated current enters and leaves the electrodes as the finger approaches or leaves the gadget. With a reaction time of 68 milliseconds, the sensor can monitor extremely low pressure. A nanotexture-enhanced nano-generator array used triboelectricity for self-powering sensing. Even in wet conditions, the sensor provides reliable voltages. When no load is attached, the sensor produces an open-circuit voltage of 1.613 V, as reported for the nanotexture-enhanced nanogenerator in [73–75]. The device can detect both the lateral force resulting from friction and the perpendicular force exerted on it. The voltage frequencies produced in response to various external stimuli, such as sliding and pressure, exhibited distinct differences [76]. Triboelectric-piezoresistive sensor development is intriguing. The structures' porosity makes the embedded sensor more sensitive [77]. Table 4,

Table 3. Comparison of flexible temperature sensors based on different materials.

Ref. #	Sensing Materials	Features	Nature
[57]	Non-ionic polymer, Polyelectrolyte layers	ultrahigh sensitivity, thermal index surpassing 10,000 K^{-1}, (10 mK min to exceeds 1.5 mK.)	Flexible
[58]	NiO/CNTF	broad working range(-15 to $60°C$) high sensitivity (max. $-20.2\%°C^{-1}$ and B value of 3332 K)	Flexible
[59]	(PAAm-SA) hydrogel, Parylene C, UV-cured resin	fast response ($2.02s$ of $40°C$), Sensing range (22–$100°C$)	Flexible
[60]	(Ag-MWCNT)/ (PEDOT: PSS)	high sensitivity of-0.45% $°C^{-1}$ and linearity value of 0.998 in the temp. range of 25–80$°C$	Flexible

Table 4. Comparison of triboelectric sensors based on different materials.

Ref.	Materials	Stability Cycle	Response Time	Sensitivity
[78]	PVA/PU/3D structure	3000	7.9 milliseconds	6.875 $VkPa^{-1}$
[79]	Ag NWs/TPU/PA	stable	75 milliseconds	0.048 kPa^{-1}
[80]	PU film/Nylon/Mica	excellent	0.25 milliseconds	0.082 kPa^{-1}
[82]	PVDF-BTO	stable	26 milliseconds	2.620.11 $VkPa^{-1}$

illustrates the reported triboelectric sensor with different features.

2.6 Strain Sensors

Movement causes both tiny and major deformations. Flexible strain sensors are needed to detect and monitor the movement of the intended parts of the human body, whether they are caused by vocal cord vibrations or joint activity. Unlike traditional silicon-based strain sensors, the strain sensors must be made of a stretchy and flexible material due to the flexibility requirements. Wearable strain sensors require their component materials to be sensitive, flexible, mechanically dependable, possess hysteresis, and provide an output signal in a linear form. The strain sensor measures and quantifies material deformation or strain. The classification of wearable flexible strain sensors is according to the working mechanism used for sensing. Resistive and capacitive strain sensors collect mechanical deformation and convert it into an electrical signal. Resistive strain sensors measure strain using electrical resistance, while capacitive sensors use capacitance. Due to their transformational feature, strain sensors are essential for monitoring and understanding material structural behavior. Figure 2 illustrates typical strain sensor architecture.

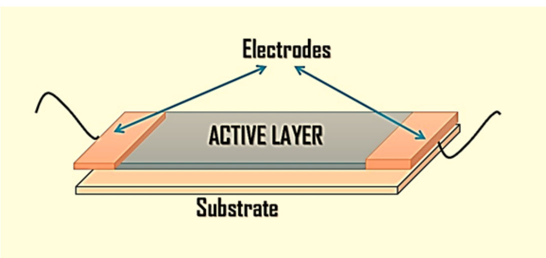

Figure 2. Typical strain sensor architecture.

Strain sensors serve as critical transducers across civil engineering, aerospace, biomechanics, and biomedical fields, converting mechanical deformation into measurable electrical signals to enable quantitative monitoring of structural and physiological states. In biomedical contexts, they quantify tissue deformations to advance prosthetics, rehabilitation protocols, and musculoskeletal diagnostics [84]. In structural applications, they provide continuous integrity monitoring of safety-critical components. The sensing performance of a strain sensor is characterized by a core set of parameters: sensitivity (gauge factor), stretchability (working strain range), response time, detection limit, linearity, hysteresis, and long-term stability. These parameters collectively determine the sensor's fitness for a given sensing task, and their optimization—rather than material selection per se—is the central objective of strain sensor engineering [85].

2.6.1 *Flexibility*

The flexible strain sensor has received considerable attention in recent years due to its immense potential for use in e-skin, human-computer interface, human health monitoring, movement detection, and soft robotics, as advancements in science and technology proceed. Flexible strain sensors (FSSs), owing to their inherent flexibility, lightweight nature, and compatibility with living organisms, are very important. Conventional devices composed of metals or semiconductors have a limited ability to detect slight deformations, typically below 5% strain, and are inflexible, causing discomfort when worn [86]. Due to its high performance, flexible strain sensors are employed in structural health monitoring, display technology, and pressure sensing. Flexible strain sensors can detect small muscle, skin, and pulse movements from movement, breathing, vocalization, and heartbeat in electrical equipment or on garments and skin. Additionally, they may detect considerable joint bending-induced body stresses. Also, robots can use them to govern joint mobility [87]. The ability to detect strains below 1% while sustaining mechanical deformations exceeding 50% defines the core sensing requirement for flexible wearable applications [88–90]. Achieving this requires sensing networks that deform reversibly with the substrate without loss of electrical continuity. One-dimensional carbon nanotubes and two-dimensional graphene have proven particularly effective in this role: their high aspect ratio and large surface area promote percolation network formation at low loading, enabling simultaneous high sensitivity and mechanical compliance [91, 92]. The key design criterion is not the intrinsic stiffness of the constituent material, but rather the architecture of the conductive network and its ability to sustain sensing performance across the full deformation range required by the target application. Organic transistor-based strain sensor platforms have been integrated into wearable formats for monitoring wrist flexion and extension, demonstrating the viability of low-voltage flexible electronics for continuous biomechanical sensing in rehabilitation contexts [36]. Huang et al. [93] reported a PU-based flexible strain sensor that can detect small movements when mounted on different parts of the body, such as human joints. Zhang et al. [94] recently reported a flexible strain sensor that employs an MXene/BC double network-based hydrogel for human motion detection, achieving remarkable results with stability surpassing 2000 cycles.

2.6.2 Sensitivity

The sensitivity of a strain sensor is quantified by the gauge factor (GF), defined as the ratio of the relative resistance change ($\Delta R/R_0$) to the applied strain (ϵ). A high GF is essential for detecting subtle physiological signals such as pulse waveforms and micro-expressions, while a sufficiently wide sensing range ($\epsilon > 50\%$) is required to capture large-amplitude joint movements. These two demands are inherently in tension: microcrack-based sensing mechanisms typically yield ultrahigh GF but limited working range, whereas percolation network-based mechanisms offer broader range at moderate sensitivity [95]. Identifying the appropriate sensitivity–range trade-off is therefore the primary design criterion, dictated by the specific sensing scenario rather than by material preference alone. Strain sensors should also have a higher gauge factor (GF) and a lower limit of detection of 1% strain. This lets it identify and respond to sensitive stresses like pulse, respiration, phonation, and microexpressions. Even with little strain, reducing conduction routes is important for high sensitivity. However, conductivity must be maintained under extreme strain to have a wide sensing range [96]. By converting an Au film into a brittle conductive layer, Li et al. [97] achieved a 100% maximum tensile limit and high sensitivity. Nevertheless, the properties of "wide stretchability" and "high sensitivity and low limit of detection" are in a trade-off relationship due to the specific demands of the sensing materials [98]. Silk fibroin-based conductive hydrogels designed by Tao [99] and colleagues are recyclable, freeze-resistant, and dry-resistant. Strain sensors are sensitive to macro and micro human motions due to their gauge factor of 2.18. Strain sensors and integrated supercapacitors excel in harsh situations. Cai et al. [84] developed a $Ti_3C_2T_x$ MXene/CNT composite strain sensor by intentionally introducing fractures in the MXene layer and bridging them with CNTs. The sensor achieved a GF of 35 in the 0–40% strain range and 772.6 in the 40–70% strain range, with an overall working range of approximately 80%. The strain sensors, which utilize metal nanoparticles (e.g., Pd nanoparticles molded into polyimide substrates), exhibit excellent sensitivity (GF = 390) and a low limit of detection (0.06%), but they lack stretchability ($\epsilon = 0.2\%$) due to their rigidity. A strain sensor constructed from ternary nanocomposites of zero-dimensional materials exhibits a remarkably high gauge factor of 2393 at a strain of 62% [100].

Wang et al. [101] developed a star-nose-inspired multi-mode flexible sensor composed of parallel polyimide films mimicking the nasal rays of the star-nosed mole. The sensor achieved a high gauge factor of up to 4000 under 55% strain. Although the work highlights high sensitivity, the exact limit of detection is not clearly stated, and further improvements may be required for detecting ultralow strains. Flexible strain sensors with a high gauge factor (GF), long-term durability, extensive stretchability, and a low limit of detection are in demand. Recent research by Huang et al. [102] generated strain-detecting polymer nanoball-adorned graphene sensors. The sensors exhibit GFs of 51 in the 0–5% strain range and 87 in the 5–8% strain range within their sensing working window, while the serpentine-structured porous fiber substrate sustains mechanical integrity beyond 100% strain without fracture [102].

2.6.3 Self-Healing

The sensing function of a strain sensor depends on the integrity of its conductive network: mechanical damage that severs conductive pathways produces permanent signal loss, terminating the sensor's operational life. Self-healing strain sensors address this fundamental reliability limitation by autonomously restoring electrical conductivity after damage, thereby recovering sensing capability without external intervention [103, 104]. This restoration of sensing function—rather than merely structural recovery—is the critical metric: a self-healing sensor must return to its pre-damage gauge factor, detection limit, and response time within a clinically or operationally acceptable timeframe [105, 106]. Self-healing capability has been demonstrated to extend sensor operational life across repeated damage-recovery cycles while maintaining consistent signal output, a prerequisite for long-term continuous monitoring applications [107, 108]. Ion-conductive hydrogel platforms that combine stretchability, transparency, and autonomous self-healing represent a particularly promising material class for multifunctional sensing applications where mechanical robustness and optical clarity are simultaneously required [123]. From a sensing standpoint, self-healing sensors are categorised by how their sensing function is restored: autonomic self-healing sensors recover their conductive sensing pathways without any external intervention, whereas non-autonomic sensors require an external stimulus such as light, heat, or humidity to re-establish reliable signal transduction [109].

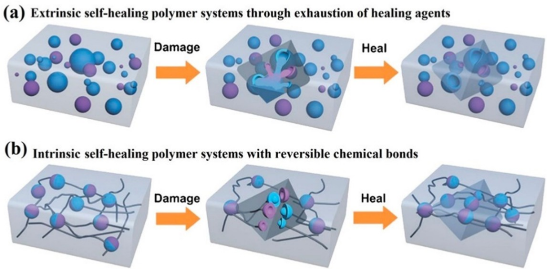

Figure 3. Schematic illustration of (a)Intrinsic self-healing (b) Extrinsic self-healing polymer systems. Reproduced with Permission. [110] Copyright © 2017 WILEY-VCH.

Self-healing mechanisms in strain sensors are classified as extrinsic or intrinsic, as shown in Figure 3, and each has distinct implications for sensing recovery performance. Extrinsic systems release healing agents from embedded capsules or channels upon damage [111, 112]: sensing restoration is rapid at the point of damage but is limited to a finite number of healing events, making these systems suited to sensors operating in predictable damage scenarios. Intrinsic systems rely on reversible dynamic bonds that reform spontaneously [113]: sensing recovery can occur repeatedly at any damage site, and the recovered gauge factor and linearity are typically closer to the original values. Because intrinsic healing does not deplete a finite reservoir, it sustains sensing performance across a greater number of damage-recovery cycles [114], which is the decisive advantage for sensors intended for continuous long-term monitoring where damage events are unpredictable. Table 5 shows a comparison of different materials-based strain sensor healing efficiencies.

Table 5. Comparison of the self-healing efficiencies of strain sensors based on different materials.

Ref. #	Materials	Self-healing efficiency
[115]	MWCNTs-UPy/IU-PAM	95%
[116]	PAAm/NaCl	99%
[117]	Fe_3O_4@PAA/PAA	>95%
[118]	Col/$AlCl_3$/ DCMC	100%
[119]	CNFs/PU	97.7%
[120]	(P(AM-co-AA)) /Al^{3+} network	97%
[121]	(PVA)/(TP)/EG/AgNPs/ MWCNTs	96.07%
[122]	glutaraldehyde (GA)- (PVA)/ (CNC/ (PEDOT: PSS)	>98.3 %

2.6.4 *Adhesiveness*

Reliable strain sensing on the human body depends not only on the intrinsic performance of the sensing element but also on the quality of the sensor–skin interface. Stable mechanical coupling between the sensor and the monitored surface determines how faithfully surface deformation is transferred into the sensing layer, making interfacial adhesion a direct determinant of sensing accuracy.

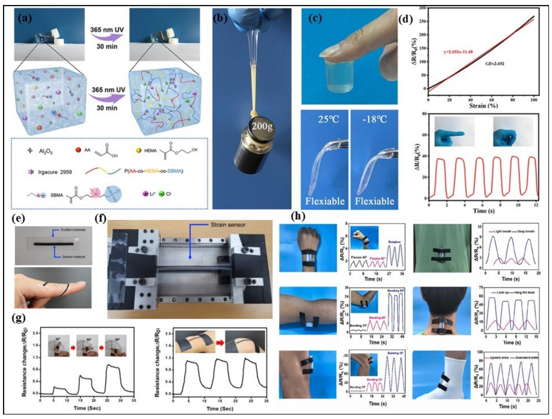

Figure 4. (a) Fabrication mechanism of PAHS gels (b) Under various external forces-mechanical properties (c) Adhering to substrates and at different temperatures (d) PAHS gel sensor strain sensitivity. Reproduced with permission [127] Copyright © 2023 Elsevier (e) Stretchable strain sensor stretched by hand rolling. (f) A home-made tensile tester stretched a strain sensor of 300% strain. (g) Response to bending motion detection. Reproduced with permission [134] Copyright © 2023 the Korean Institute of Metals and Materials. (h) Strain sensor for movement detection. Reproduced with permission [135] Copyright © 2025 The Authors. Published by Elsevier B.V

As technology grows such as the Internet of Things (IoT), people expect better living conditions, multiple sensors that monitor both health and environmental conditions are now gradually making their way into the everyday lives of people globally such as e-skin, speech recognition, health monitoring, human-computer interactions, and soft robotics with flexibility, adhesiveness, and biocompatibility, used to monitor minute changes brought on by external factors [125, 126]. In terms of all attributes, Strain sensor adhesiveness directly influences the sensor's ability to be firmly attached to the surface undergoing evaluation. Achieving high adhesion is essential for reliable strain-sensing data collection [125]. Skin-attachable hydrogels driven by supramolecular interactions, such as nucleotide-crosslinked networks, have demonstrated stable conformal contact with skin surfaces during dynamic deformation, enabling accurate strain acquisition for gesture recognition and human-machine interface applications [124]. Maintaining a stable connection with the substrate and minimizing the possibility of slippage or detachment is essential for a well-designed strain sensor with high adhesiveness, especially in dynamic or harsh situations. Across a broad spectrum of structural monitoring techniques, and industrial and biomedical applications, this is critical for getting accurate strain data and making sure the sensor lasts a long time. A strain sensor's whole performance and usefulness are greatly affected by the adhesiveness factor, which in turn affects the sensor's practical utility in real-world circumstances [126].

2.6.5 *Stretchability*

The working strain range of a sensor determines which physiological or structural sensing tasks it can perform: pulse detection requires sensing at strains below 5%, while full joint angle monitoring demands ranges exceeding 50%, and some soft robotic applications require strain sensing beyond 500%. Matching the sensor's stretchability to the target sensing range—without sacrificing sensitivity or signal stability at either extreme—is therefore the primary stretchability design requirement. A single-step in situ free-radical polymerization technique based PAHS gels-based strain sensor (Figure 4 (a)) shows more than 85% transparency at 550 nm, 800% stretchability, and 2.25 S/m ionic conductivity as shown in Figure 4 (a, b, d). Furthermore, these gels maintained their flexibility and conductivity even at a temperature as low as -18 ℃ (Figure 4 (c)), superior strain sensitivity (GF = 2.69), and application of using PAHS gel sensors to detect human action, the finger bends and straightens repeatedly, indicating high repeatability [127]. The achievable sensing range spans several orders of magnitude depending on sensor architecture. Resistive sensors based on aligned CNT networks sustain stable sensing up to 280% strain [128], while capacitive architectures using percolating CNT-elastomer networks extend this to approximately 300% with lower hysteresis [129]—the choice between these two types should be dictated by the hysteresis tolerance of the sensing application. For applications requiring detection across an even wider deformation range, textile and fabric-based sensing platforms achieve strains from 140% to 500% [83, 130–133], enabling whole-body motion sensing scenarios that rigid or thin-film sensors cannot address. Among these, core-spun elastic thread-based sensors with integrated electrode and sensing cell have demonstrated gauge factors comparable to metallic strain gauges [83], graphene-dyed wool fabric sensors have demonstrated moisture resilience while maintaining stable strain sensing performance [131], and fragmented conductive cotton fabric sensors have achieved high sensitivity under large deformations [130]. A stretchable strain sensor designed by using multi-walled carbon nanotubes (MWCNTs)-Ecoflex composite paste on an Ecoflex substrate using a straightforward stencil printing method (Figure 4(e,f,g)) exhibits maximum stretchability of 1,000% (Figure 4(f,g)) and a sensitivity range that extends from 3 to 12,287 [134] . Long et al. [135] reported a flexible strain sensor with a stretchability feature mounted on different parts of the body to detect movement, shown in Figure 4(h). Printable and wearable graphene-based strain sensors have further demonstrated high sensitivity for human motion monitoring through scalable fabrication processes [39]. Table 6 illustrates different materials-based strain sensor comparison in terms of stretchability, gauge factor and response time.

3 Parameters of Strain Sensor

3.1 Gauge Factor

The gauge factor is defined as the ratio of the percentage change in the sensor's response to the strain that is being applied. Strain sensors reported for use in a variety of applications exhibit a stretchability of several hundred percent in terms of linear strain and a sensitivity (gauge factor) ranging from <10 to ≈1000 [141–143]. For resistive and capacitive sensors, GFs are calculated as $\Delta R/R_0$ divided by ϵ (resistive) and $\Delta C/C_0$ divided by ϵ (capacitive), respectively. Conventional metal-foil strain gauges exhibit GFs in the range of 2–5 [85, 143], whereas semiconducting

Table 6. Comparison of different material-based strain sensor gauge factor, stretchability, and response time.

Ref.	Materials	Stretchability	Gauge factor	Response time
[136]	TPU/PDA/MWCNTs/GO	760 %	41.57	59 milliseconds
[137]	CNTs/AgNWs hybrid network	100%	79	42 milliseconds
[138]	P(HEA-co-AA) DES-IL gels	1783%	1.01	143 milliseconds
[139]	CNC)/ (A-MWCNT) conductive hydrogel	431.6 %	4.32	100 milliseconds
[140]	POE/carbon nanostructures (CNS)	952.5%	5230*	200 ms

* Peak GF measured near the fracture limit; GF in the linear working range is substantially lower.

and nanocomposite strain sensors can achieve GFs of 100 or more [144–146].

The gauge factor landscape across sensing architectures reveals clear performance–range trade-offs that must guide sensor selection for specific sensing tasks. Percolation-network resistive sensors (CNT–elastomer composites) yield GFs of 1–3.37 [147, 148]: sufficient for large-amplitude joint motion detection but insufficient for subtle physiological signals such as pulse or respiration. Microcrack-based resistive sensors achieve GFs exceeding 1000 [149, 150], enabling detection of micro-scale skin deformations associated with arterial pulse and facial muscle activity, but their working strain range is correspondingly narrow. Ideal parallel-plate capacitive sensors are theoretically constrained to GF ≤ 1 [151, 152], making them unsuitable where high sensitivity is required; structured dielectric architectures incorporating microstructured or porous dielectric layers can exceed this limit through geometric amplification, though at the cost of fabrication complexity. Their inherently low hysteresis nonetheless makes capacitive sensors preferable for applications demanding repeatable, drift-free signal output. The selection of sensing architecture should therefore begin with the sensitivity and range requirements of the target physiological or structural sensing task, not with material availability. Figure 5 presents a graphical comparison of gauge factors across different material-based strain sensors [153–159].

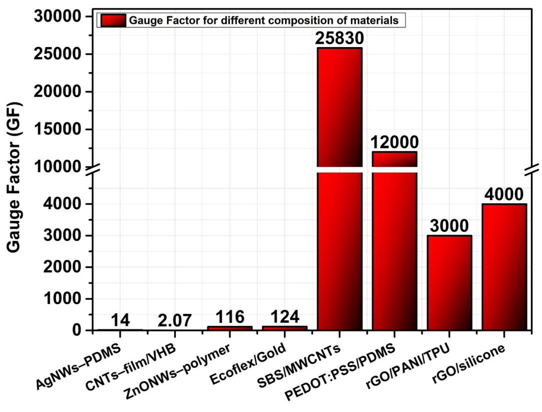

Figure 5. Graphical bar chart comparison of gauge factor of strain sensor fabricated by utilizing different materials.

3.2 Linearity

Linearity, in simple terms, represents a correlation where the resistance changes or variation is directly proportional to applied strain. A linear regression analysis is used to generate the coefficient of determination (R_2), which is then used to quantify the linearity of a flexible strain sensor. This is accomplished by determining the R2 value. The R_2 value of a linear strain sensor is higher, which indicates that the output signal follows a straight line with just a small amount of variance. When the value of R_2 is low, on the other hand, a strain sensor is said to be nonlinear [160]. The linearity parameter is crucial for stretchable strain sensors as it must be able to accommodate extremely large strains. The nonlinearity of sensors adds complexity and difficulty to the calibration process. The presence of nonlinearity is a significant disadvantage commonly found in strain sensors of resistive type. Typically, strain sensors exhibit a non-linear response to applied strain when the thin film microstructures transform from a "homogeneous morphology" to a "nonhomogeneous morphology" during stretching [161, 162]. Non-linearity in resistive strain sensors arises from structural transitions within the conductive network during stretching: as the network shifts from a homogeneous to an inhomogeneous percolation state, the resistance–strain relationship deviates from linearity [163, 164]. For sensing applications requiring signal interpretation without complex calibration—such as real-time joint angle estimation or continuous blood pressure monitoring—this non-linearity directly limits measurement accuracy and increases signal processing burden. Hybrid network architectures that maintain network homogeneity across the full strain range can substantially improve linearity while recovering sensitivity. Capacitive sensors provide inherently superior linearity across large strain ranges [165], making them the preferred architecture where measurement accuracy takes precedence over sensitivity.

Table 7. Comparison of strain sensor based on different materials and polymers composite highlighting response time.

Ref #	Materials	Response Time
[171]	methylpropiophenone/PVA	660 ms
[172]	Poly(vinyl alcohol)/polyacrylamide (PVA/PAM)	368 ms
[173]	TPU@CNT	75 ms
[175]	ECPCs/CNT/thermoplastic polyurethane	180 ms
[176]	zinc oxide	300 ms
[177]	CNT/silver nanowire (AgNW)	600 ms

Table 8. Comparison of strain sensor materials highlighting dynamic durability and stability cycles.

Ref #	Materials	Dynamic Durability and Stability
[180]	MWCNTs)/(GR)/(SR)/Fe3O4	9000 cycles
[181]	graphene/TPU	1000 cycles at 50% strain
[182]	CNTs-CB/PDMS composites	2500 cycles at 200% strain (GF retention not reported in source)
[145]	AgNW/elastomer nanocomposite	1000 cycles at 70% strain
[184]	PET/ PUE	1000 cycles at 20

3.3 Response Time

Response time defines the temporal resolution of a strain sensing system and directly determines which dynamic physiological signals can be accurately captured. Human arterial pulse waveforms require sensing bandwidths corresponding to response times below 100 ms; vocal cord vibrations and micro-expressions demand even faster response. A systematic delay in reaching stable output is an intrinsic characteristic of all stretchable strain sensors, arising from the viscoelastic relaxation behaviour of their polymer components [166, 167]; the critical question is whether this delay falls within the temporal tolerance of the target sensing application. The 90% response time criterion is the standard benchmark for quantifying this delay in stretchable sensors [168]. Resistive sensors with viscoelastic polymer matrices typically exhibit response times in the range of 300–660 ms [169], which is adequate for slow joint movements but insufficient for pulse or phonation monitoring. Capacitive sensors achieve faster responses of 50–100 ms [170], extending the range of detectable physiological signals. For sensing applications with strict temporal requirements, response time must be explicitly matched to the frequency content of the target signal, as shown in Table 7.

3.4 Dynamic Durability and Stability

The term "dynamic durability" refers to the capacity of strain sensors to survive multiple cycles of stretching and releasing over a prolonged period, while simultaneously sustaining their electrical functioning. Stretchable wearable strain sensors need to have a high level of endurance since strain sensors are designed to detect stresses that are extremely vast, complicated, and continuously changing [178]. Dynamic durability directly governs whether a sensor can sustain accurate sensing across the lifetime of the intended application: a sensor deployed for continuous joint motion monitoring over weeks must maintain its gauge factor, detection limit, and signal-to-noise ratio across tens of thousands of stretch-release cycles. Signal drift, baseline shift, and sensitivity degradation under cyclic loading are the critical sensing failure modes, and they occur well before mechanical fracture becomes visible [179]. Quantifying durability purely in terms of cycle count is therefore insufficient; the stability of sensing output—measured as the variation in normalised resistance response across the full test duration—is the more meaningful performance indicator for practical deployment. Table 8 shows different cycles and strain percentages of the strain sensor device reported.

3.5 Hysteresis

The term "hysteresis" is used to describe the connection between the current state of a strain sensor, which is the output signal of cycle n, and its previous record, which is the output signal of cycles 1 to $n - 1$. Hysteresis is an essential key consideration component of strain sensors, particularly in wearable industries [185]. A significant amount of hysteresis behavior is responsible for the irreversible or inconsistent sensing performance of sensors when they are subjected to dynamic stressors. Strain sensor hysteresis produces signal drift and inconsistent readings during loading and unloading cycles, which can create severe measurement inaccuracies in long-term monitoring applications. Resistive sensors have higher hysteresis due to viscoelastic effects and material fatigue, making calibration difficult and limiting accuracy. Capacitive sensors have lower hysteresis and perform better in real-world applications like wearable health monitoring and robotics. Compared to resistive strain sensors, capacitive sensors perform better by showing less lag and delivering more accurate results [186]. Horizontal parallel inter-digital strain sensors have benefits over vertical sandwich resistive and capacitive strain sensors. It improves substrate transparency and reduces normal force connections. By using a

Table 9. Comparison of different material composite based strain sensor materials highlighting Hysteresis and strain percentage.

Ref.	Materials	Hysteresis	Strain
[188]	Ethylene glycol/sodium chloride	0.15%	250%
[189]	CB/PDMS foam	1.2%	70%
[190]	PEDOT: PSS nanofibers/PVA	<1.5%	300%
[191]	MWCNTs/ Ecoflex nanocomposites	1.63%	1300%
[192]	Polyacrylamide (PAAM) hydrogel	<3%	0–200%
[146]	P3HT-NFs/ PDMS	<12%	up to 100%

low-pressure coupling effect under normal load, 3D electrodes in the substrate increase their endurance. Interdigital 3D electrodes are made from vertically aligned carbon nanotubes in the strain sensor. It is translucent, stretchy, and has 0.35% ultralow hysteresis. It also has great transparency, a 60 ms reaction time, strong long-term stability, and a pressure-insensitive performance of less than 0.8% [187]. Table 9 shows the hysteresis and strain parameters of different fabricated strain sensors.

4 Types of Strain Sensor

4.1 Capacitive strain sensor

Capacitive strain sensors transduce mechanical deformation into a measurable change in capacitance, governed by the geometry of a parallel-plate structure: two flexible electrodes separated by a deformable dielectric layer [193]. The sensing principle offers inherent advantages for certain applications—low hysteresis, high linearity, and freedom from resistive heating—but imposes specific performance constraints. The theoretical maximum gauge factor for an ideal parallel-plate capacitive sensor is unity, limiting sensitivity relative to resistive architectures [197, 198]. Electrode conductivity must be maintained across the full working strain range: rigid metallic films fracture under stretching, severing the sensing circuit, so continuous sensing requires electrode architectures that accommodate deformation without conductivity loss [199, 200]. The critical design challenge for capacitive strain sensors is therefore not dielectric material selection per se, but achieving stable electrode conductivity and dielectric uniformity across the intended sensing range—since any spatial inhomogeneity in the dielectric layer produces localised capacitance variations that manifest as measurement error [194–196]. Recently, certain research has incorporated additional ingredients and fillers to enhance the chemical, mechanical, and electrical characteristics of stretchable composite electrodes. Porous elastomeric architectures—such as PDMS/AgNW sponge composites—have been shown to improve electrode compressibility and contact uniformity, which is directly relevant to capacitive sensor dielectric and electrode design [183]. Atalay et al. [201] demonstrated a highly stretchable capacitive strain sensor in which sputtered metal electrodes patterned by direct-write laser rastering retain their conductivity up to 250% strain, yielding a linear and repeatable sensing output up to 85% strain.

To increase capacitive strain sensor sensitivity, researchers construct new geometries or adjust material mechanical properties beyond the parallel plate capacitor. Kim et al. [202] using AgNW electrodes on PDMS, designed an interdigitated capacitive type strain sensor(ICSS) as shown in Figure 6(a, b, c) [193]. Figure 6 (d) demonstrates that the interdigitated capacitive strain sensor (ICSS) has an electric conductivity (σ) (at 30% ε) of approximately $\tilde{4}2.2$ S/cm, indicating no major cracks in the electrodes and allowing for effective capacitor performance up to 30% ε and sensing performance as applicable application shown in Figure 6(e). Huang et al. [203] proposed homogeneous, water-retaining hydrogel electrodes for capacitive strain sensors.SPGLH sensor in detecting human motion (Figure 6 (f)) and in human-machine interface (Figure 6 (g)) as an application in real-time. Xue et al. [204] reported a capacitive-type strain sensor for strain movement(Figure 6 (h)). A popular way to make stretchable capacitive strain sensors is by placing a thin insulating film, called a dielectric layer, between two flexible electrodes-kind of like a sandwich. When a voltage is applied, the dielectric layer prevents the opposing charges on each electrode from passing through. This creates a capacitor that starts with an initial capacitance of C_o, which can be described as follows:

$$C_o = \epsilon_o \epsilon_r \frac{A}{d} \quad (1)$$

where ϵ_o is the permittivity of free space, ϵ_r is the relative permittivity (dielectric constant) of the dielectric material, A is the electrode overlap area, and d is the thickness of the dielectric layer.

4.2 Resistive Strain Sensor

At the beginning of the 1940s, the first strain sensors of the resistive type were developed. These strain sensors make use of metallic foils. It has been demonstrated

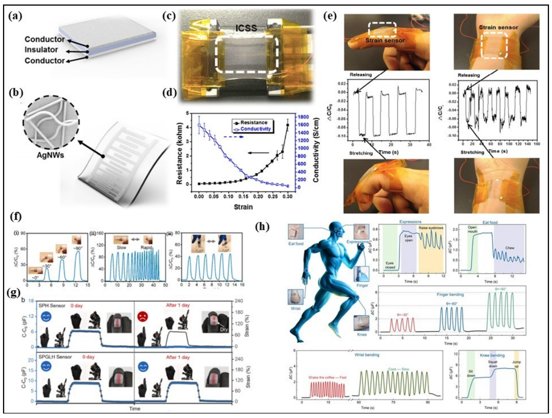

Figure 6. (a) Traditional parallel-plate capacitive strain sensor. (b,c) Schematic and photograph of an interdigitated capacitive strain sensor (ICSS). (d) Electrical conductivity of the ICSS electrodes at 30% strain, confirming crack-free electrode integrity. (e) ICSS sensing performance during finger and wrist muscle cyclic motions. Reproduced with permission [202]. Copyright © 2017, American Chemical Society. (f) SPGLH-based sensor detecting human motions: (i) finger angle, (ii) response to slow–fast bending, and (iii) tracking under saline. (g) SPH and SPGLH sensor response signals for robot control. Reproduced with permission [203]. Copyright © 2023, Elsevier. (h) Capacitive-type strain sensor application. Reproduced with permission [204]. Copyright © 2024, Springer.

that these sensors are capable of detecting strain of up to 5%, particularly for very slight deformities in composites or nanocomposites and rigid materials or bodies [205, 206]. Resistive strain sensors consist of sensing elements, which are conductive networks, and a flexible polymer or fiber substrate that can be stretched. Their operation is based on the concept that the electrical resistivity of a material changes when subjected to the strain that is applied [207].

The shift from rigid to flexible resistive strain sensors was driven by a fundamental sensing requirement: the need to maintain intimate, mechanically stable contact with soft, curved, and dynamically deforming surfaces such as human skin and soft tissue. A sensor that cannot conform to the monitored surface loses strain coupling efficiency, introducing systematic measurement error that cannot be corrected by calibration [208]. Flexible resistive sensors resolve this by distributing the sensing element across a compliant substrate, ensuring that the full applied strain is transmitted to the active sensing network. The quality of this strain transmission, quantified as the ratio of the strain in the sensing layer to the strain in the monitored surface, is the primary performance criterion governing substrate and sensing element integration design.

Using conductive Au clusters and Eco flex elastomeric films, Migliorini et al. [155] built a high-strain resistive sensor. This sensor is created using supersonic cluster beam deposition. A thorough electromechanical characterization of the sensor indicates steady functioning up to 4500 working cycles, 500% strain, and 124-gauge factor as shown in Figure 7 (a). The sensor can sense mechanical stimuli like stretching and compression as mounted on the finger and elbow (Figure 7(b)). Shang et al. [200] proposed a straightforward and efficient method for producing a superhydrophobic flexible resistive strain sensor. This method involved spraying carbon black (CB) nanoparticles in a thermoplastic elastomer (SEBS) solution onto a PDMS flexible substrate (Figure 7(c)). The behavior of strain sensing is depicted in Figure 7(d) under cyclic tensile testing with loads ranging from 10% to 70% (the maximum strain range), the normalized resistance $\Delta R/R0$ increased significantly and showed a consistent signal across stretching-releasing strain cycles. Applications for human motion detection as finger straight or bending and resistance variation in strain sensors, are shown in Figure 7(e).

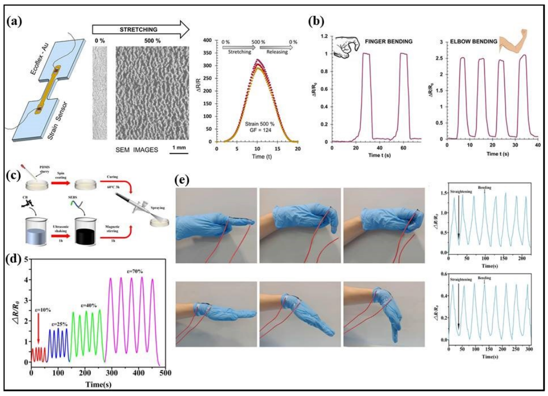

Figure 7. (a) Ecoflux-Au based resistive strain sensor and (b) RSS mounted on finger and elbow: stretching vs. resistance. Reproduced with permission [155]. Copyright © 2023, American Chemical Society. (c) Schematic representation of flexible strain sensor, (d) normalised resistance variation over five stretching–releasing cycles (10%–70% strain), and (e) output signal for various deformation degrees. Reproduced with permission [209]. Copyright © 2016, American Chemical Society.

5 Material Systems for Strain Sensing

5.1 Conductive Nanomaterials

The sensing performance of resistive strain sensors is fundamentally governed by the architecture and connectivity of their conductive networks. Carbon-based nanomaterials have emerged as the dominant platform for constructing such networks, owing to a combination of properties that are directly relevant to sensing: high intrinsic electrical conductivity enables low baseline resistance and high signal-to-noise ratio; exceptional mechanical flexibility allows the conductive network to deform reversibly with the substrate without permanent damage; and the high aspect ratio of one-dimensional structures (carbon nanotubes) or the large surface area of two-dimensional structures (graphene) [210] facilitates the formation of percolation networks at low filler content, maximising sensitivity while preserving stretchability [211]. The choice among carbon black, carbon nanotubes, and graphene derivatives is therefore driven by the sensing performance requirements of the target application. Carbon black, while lower in intrinsic conductivity than carbon nanotubes or graphene, provides a cost-effective percolation network that is particularly suited to sensors requiring moderate sensitivity across a wide strain range, where its isotropic network morphology sustains conductivity without directional sensing bias.

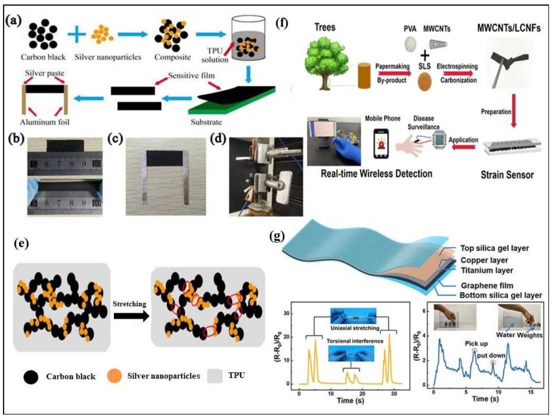

Figure 8. (a) Fabrication procedure of the CB/AgNPs composite based strain sensor (b) Stretchable characteristic of a strain sensor (c) strain sensor sample with electrodes (d) Sensor in testing machine (e) During stretching, the morphology of a CB/AgNPs-based strain sensor changes. Reproduced with permission [212] Copyright © 2018, Open Access. (f) MWCNTs based strain sensor fabrication and application. Reproduced with permission [213] Copyright © 2023, American Chemical Society (g) Strain-sensitive layer, a Cu/G micro crack composite film/self-adhesive silica gel-waterproofing and self-adhesive layer. Reproduced with permission [214] Copyright © 2023, American Chemical Society.

Its incorporation into elastomeric sensing composites extends the conductive percolation network across large deformations, sustaining measurable resistance change at strains where higher-aspect-ratio networks would fracture [148]. Zhang et al. [212] demonstrated a carbon black/silver nanoparticle-based flexible strain sensor, as depicted in Figure 8 (a). The strain sensors that have been constructed have a high degree of stretchability and demonstrate good sensitivity, with a gauge factor of 21.12 at a tensile strain of 100%. Additionally, they show great performance in static conditions at a strain of 50% for 600 seconds, as well as dynamic stability as in Figure 8 (b, c, d) with 100 cycles and morphology changes of a strain sensor based on CB/AgNPs during stretching as shown in Figure 8 (e). Carbon nanotubes have gained substantial attention due to their quick fabrication, affordability, and chemical-based durability. Wang et al. [213] developed a strain sensor utilizing a combination of multi-walled carbon nanotubes and lignin-based carbon nanofibers as shown in Figure 8 (f). The fabricated sensor exhibits a GF of 137.3 in the 0–160% strain range and 2746.4 in the 160–240% range, with the sharp sensitivity transition attributed to a microcrack-induced percolation change in the conductive network at large deformations. Bioinspired composite architectures, such as multiwalled carbon nanotube/polyvinyl butyral/nylon cloth strain sensors, have further enabled wireless strain sensing with stable electrical performance across repeated deformation cycles [81]. The broad applicability of such carbon nanomaterial-based flexible sensors across wearable electronics, robotics, and structural health monitoring has been comprehensively reviewed in the literature [38]. Zhu et al. [214] researched a strain sensor that is flexible and utilizes composite films made of copper and graphene. The sensor has a broad response range extending up to 110% strain, with particularly high sensitivity demonstrated in the 92%–110% strain regime. Additionally, it possesses excellent endurance in detection capabilities. Additionally, it can identify ultrasmall strains, namely strains as low as 0.2%. The sensor's self-adhesive silica gel clings directly to the skin, enhancing the monitoring of complicated strain status during underwater activities. The versatile strain sensor has the potential to become an advanced underwater sensor in the future as shown in Figure 8 (g). Metal nanoparticles, such as silver or MXene, improve strain-sensing conductivity and stability. These nanoparticles can be integrated into flexible and elastic materials to create strain sensors that can adapt to distinct surfaces and endure mechanical deformation. Metal nanoparticles such as silver and gold, beyond their role as conductive fillers in strain-sensing composites, have also been explored as plasmonic transducers for environmental sensing, illustrating the broader versatility of bio-fabricated nanomaterials across sensing modalities [216]. Within strain sensing specifically, the integration of nanotechnology enhances sensor performance and enables applications in wearable devices, robotics, and structural health monitoring where accurate mechanical strain detection is essential [215].

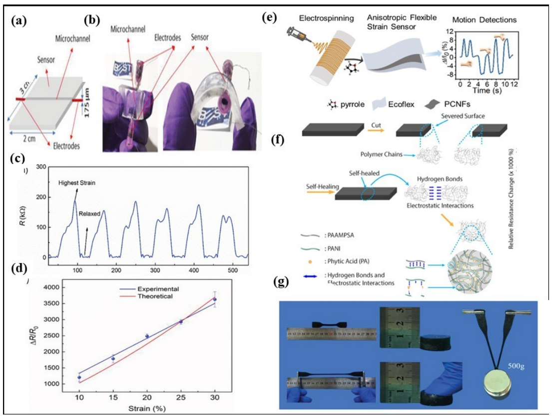

Figure 9. (a) Schematics illustration of the fabricated strain sensor (b) Flexible sensor optical image (c) Sensor's response to a 10% strain d) Sensor response (experimental and theoretical) for various strain values. Reproduced with Open Access [157], 2020.Wiley VCH (e) Aligned Cellulose and Polypropylene Composite Nanofibers through Electrospinning and ITS application. Reproduced with permission [218] Copyright © 2023, American Chemical Society (f) Polymer Complex-Conductive, self-healing and ultra-stretchable Strain Sensor.Reproduced with permission [108] Copyright © 2019, American Chemical Society. (g) Strain sensor polymer hydrogel mechanical characteristics with compression, stretch and accept 500 g's weight. Reproduced with permission. [220] Copyright © 2024 Wiley-VCH GmbH

5.1.1 *Conductive Polymer*

Conductive polymers offer a distinctive set of sensing-relevant properties that complement carbon nanomaterial-based systems. Their tunable electrical conductivity—adjustable through doping level and processing conditions—allows the sensor's baseline resistance and sensitivity to be tailored for specific sensing ranges [157, 219]. The integration of conductive elastomeric composites with microstructured geometries further amplifies sensitivity by concentrating deformation at specific contact points, as demonstrated by micropyramid-array pressure sensors [217]. Among available options, PEDOT:PSS has demonstrated the most consistent sensing performance, combining high signal stability under repeated strain cycling, processability into thin sensing films, and compatibility with flexible substrates. These attributes make it particularly suitable as an active sensing layer in applications requiring reliable detection of small, repetitive strains such as pulse monitoring and respiration sensing. Bhattacharjee et al. [157] fabricated a PEDOT: PSS micro channel-based highly sensitive stretchable strain sensor as shown in Figure 9 (a,b). The sensor has a significantly low average level of hysteresis (<9%). The sensor displays roughly an order ($\Delta R/R\_0 \approx 1200$) increase in the resistance (R) with 10% applied strain (Figure 9 (c)). The response of the sensor (both experimental and theoretical) for various strain values is shown in Figure 9 (d). Polypyrrole (PPy) is extensively utilized in sensors and electronic devices due to its simple production process and exceptional electrical conductivity. Li et al. [218] demonstrated strain sensors through electrospinning on highly aligned cellulose/Polypyrrole composite. The strain sensor has exceptional anisotropic sensing capabilities with a sensitivity of 0.73 and 0.01 for tensile applied perpendicular and parallel to the nanofiber's alignment. Strong hydrogen bonding between cellulose nanofibers and polypyrrole makes the sensor durable (2000 cycles). Flexible strain sensors can also detect human joint movements, indicating prospective motion detection applications as in Figure 9 (e). Polyaniline (PANI) offers tunable conductivity and strong electrochemical activity, making it particularly effective in sensors requiring sensitivity to both mechanical strain and chemical stimuli simultaneously [61]. The primary sensing limitation of conductive polymers is their relatively modest mechanical stretchability compared to elastomeric composites [219]. This constrains their working strain range and can lead to sensing saturation or irreversible conductivity loss at large deformations. Hybridisation with elastomeric matrices or incorporation into composite architectures addresses this limitation by decoupling the electrical sensing function from the mechanical deformation burden, allowing the polymer to operate within its optimal conductivity range while the substrate accommodates the full applied strain. Lu et al. [108] demonstrated an ultrastretchable conductive polymer complex (composed of poly(2-acrylamido-2-methyl-1-propanesulfonic acid), polyaniline, and phytic acid) as a strain sensor with repeatable autonomous self-healing ability, as shown in Figure 9(f). Its ultrahigh stretchability (1935%), repeatable autonomous self-healing (repeating healing efficiency >98%), quadratic strain response ($R^2 > 0.9998$), and linear flexion bending response ($R^2 > 0.9994$) exceed those of conventional wearable strain sensors. Wang et al. [220] reported a strain sensor based on polymer hydrogel with effective weight abilities (Figure 9(g)).

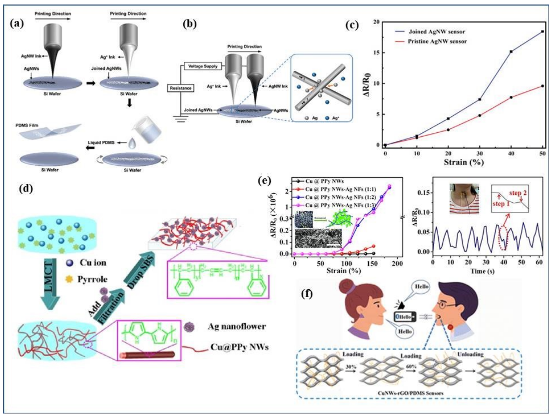

Figure 10. (a) High-resolution electro hydrodynamic (EHD) printing AgNW/PDMS based strain sensor. b) Schematic illustration of electrohydrodynamic (EHD) printing (c) Change in relative resistance with deformation-strain in sensor. Reproduced with permission [222] Copyright © 2023, Wiley VCH.(d) Fabricating strain sensor (based on Cu@PPy NW–threaded Ag NFs).(e) Plots of resistance vs. tensile strain of the composite-based strain sensor and Throat swallowing. Reproduced with permission [223] Copyright © 2021. Elsevier.(f) CuNW-rGO/PDMS strain sensor for speech recognition, lip reading. Reproduced with permission [224] Copyright © 2023, Royal Society of Chemistry.

5.1.2 *Metal based nanomaterials*

Metallic nanomaterials contribute a specific and complementary set of sensing-relevant properties to strain sensor design. Their intrinsically high electrical conductivity—significantly exceeding that of carbon-based networks—enables low-resistance sensing elements with high signal-to-noise ratios even at small deformations [221]. Among available geometries, one-dimensional nanowires and nanofibers have proven most effective for strain sensing: their high aspect ratio promotes long-range conductive pathway formation at low areal density, allowing the network to remain conductive under large strain while generating a strong resistance signal in response to deformation. This geometry-driven sensing advantage—rather than bulk conductivity alone—explains the prevalence of silver nanowires and copper nanowires as preferred metallic sensing constituents. Among these geometries, nanowires and nanofibers deliver the most consistent strain sensing performance across large deformation ranges, and have consequently become the most widely deployed metallic sensing constituents in wearable applications. Using electrohydrodynamic printing, Hu et al. [222] fabricate (high-resolution) self-reduced soldered-based silver nanowire(AgNWs) patterns for flexible and wearable strain sensors as shown in Figure 10 (a,b) and resistance variation with applied strain shown in Figure 10 (c).

Copper nanowires deliver sensing-relevant performance comparable to silver nanowires at lower cost: their combination of high conductivity and mechanical flexibility enables conductive networks that sustain low-resistance signal pathways under repeated large-strain deformation cycles. Zhao et al. [223] Construct wearable strain sensors by applying a layer of Polypyrrole onto copper nanowires (Cu@PPyNW) that are connected through silver nanoflowers (Figure 10 (d)). Electrical resistance increases slowly from 0 to 80% strain and then quickly as stretching strain increases, showing conductive network structural differences under different strains and can be stretched up to 150% before insulation and sensor test on throat swelling as shown in Figure 10 (e). Cheng et al. [224] developed a CuNW-rGO/PDMS piezoresistive strain sensor that is both extremely sensitive and stretchable. This sensor was designed specifically for lip-reading extraction and speech recognition Figure 10 (f).

5.1.3 *Substrate Materials*

The substrate of a strain sensor is not a passive structural element: its mechanical compliance, surface adhesion, and long-term dimensional stability directly determine the accuracy and consistency of strain transmission from the monitored surface to the sensing layer. A substrate that is too stiff relative to the monitored surface will partially absorb the applied strain, reducing the effective gauge factor; one that creeps or relaxes under sustained load will introduce signal baseline drift. Polydimethylsiloxane (PDMS) has become the dominant substrate choice for wearable strain sensing because its elastic modulus closely matches that of soft tissue, ensuring efficient strain coupling at the skin interface with minimal signal attenuation, while its surface energy supports conformal contact across irregular anatomical geometries [225]. Hassan et al. [226] fabricated an ink-jet printed strain sensor that is stretchable using a composite of Graphene and Zinc Oxide(ZnO) on (micro-random) ridged Polydimethylsiloxane (PDMS) substrate. The schematic diagram of the inkjet material printer and DMP-3000 insets is shown in Figure 11 (a). In Figure 11 (b), all strain sensor fabrication stages are illustrated as having a composite film thickness of 500 nm (Figure 11 (c)).

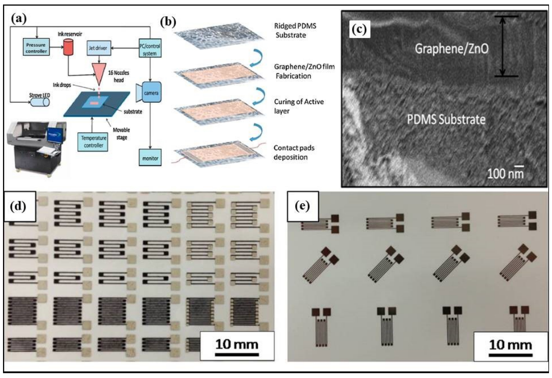

Figure 11. (a) Inkjet material printer layout with commercialised DMP-3000 inset. (b) Fabrication process for PDMS substrate-based strain sensor. (c) Cross-sectional SEM image of a strain sensor. Reproduced with permission [226] Copyright © 2018, Elsevier (d) Images of screen-printed and (e) Inkjet-printed strain sensors on PET substrates. Reproduced with permission [227] Copyright © 2017, Elsevier.

Substrate selection should be matched to the mechanical environment of the sensing application. For structural health monitoring where substrates experience only small, quasi-static strains, rigid silicon and polyimide (Kapton) substrates provide the dimensional stability needed for high-precision, drift-free measurements [227, 231]. For dynamic, large-deformation sensing on curved or moving surfaces, compliant elastomeric substrates are essential to maintain conformal contact and efficient strain transfer [228–230]. Piezoelectric substrates such as lead zirconate titanate convert strain directly into voltage without an external power supply, making them suitable for self-powered sensing in structural monitoring and industrial applications where access for battery replacement is impractical [232, 233]. The governing criterion in each case is how well the substrate transmits the applied strain to the active sensing layer while maintaining mechanical and electrical stability across the operational lifetime of the sensor.

6 Application of Strain Sensors

6.1 Biomedical Application

Continuous physiological sensing is increasingly recognised as a key enabler of remote and personalised health monitoring: by supplying clinicians with uninterrupted, quantitative measurement data, wearable sensing systems can improve diagnosis and early treatment and extend monitoring capability well beyond clinical settings [234]. Wearable devices such as strain sensors, are gaining popularity because they are easy to use, can diagnose issues early, and can monitor them for a long time [235]. Stretchable strain sensors naturally collect environmental and physiological data from the body. Data include oxygen saturation, blood pressure, heartbeat, breathing rate, motions, body temperature, and more. Strain sensors can communicate data to a device or access point via wireless or internet, updating a medical facility. Emergency service centers receive alerts for quick response. With correct treatment, doctors can remotely monitor and make medical decisions [236]. Researchers have developed strain sensors that are highly sensitive, can be mounted on the skin, and are wearable. These sensors are designed to detect small movements of the skin caused by activities such as speaking, making facial expressions, swelling of tissues, healing of wounds, breathing, and pulse [237]. Skin-like wearable strain sensors based on silver nanowire capacitive networks have demonstrated accurate finger kinematic tracking for continuous and unobtrusive quantification of hand function, with direct application in stroke rehabilitation monitoring [174].

Vital signs including blood flow, respiration rate and pulse rate can be monitored over time for personalized monitoring of health and early disease diagnosis. The patellar reflex has been evaluated with a stretchable capacitive sensor. Strain sensors are a promising solution for real-time monitoring of a variety of human motion signals. A strain sensor can detect finger touching, wrist bending, radial artery pulse, and ulnar artery pulse as shown in Figure 12(a-d) demonstrated by Shi et al. [238]. Figure 12(e) illustrates an e-skin-based ACG-strain sensor for heart cyclic volume or pressure fluctuations [239]. The metal NP-elastomer composite sensor was placed on the fifth intercostal space apex and measured in real time. Figure 12(f, g) compares volunteer ACG signals during rest and exercise. Interestingly, the sensor recorded respiration and ACG. Pre- and post-exercise respiration rates increased from 12 to 26 breaths min^{-1}, respectively. As shown in the figures, deep breathing increased the sensor's signal amplitude during exercise, allowing it to handle greater strains. Zhang et al. [240] reported strain sensors with applications in biomedical such as in soft tissues (Figure 12(h). You et al. [241] reported a sensor with application in limb actuation, hand gesture interpretation, human heart rate monitoring, etc (Figure 12(I)).

For in vivo sensing applications, signal drift and long-term biocompatibility remain critical challenges for biomedical strain sensors. For instance, despite their superior conductivity, metal nanoparticles (NPs) are still being studied for possible safety and long-term stability in medical applications [242]. Integration of wireless technologies is advancing the functionality of wearable strain sensors, enabling real-time, remote monitoring. For example, Radio Frequency Identification (RFID) technology allows battery-free, wireless data transmission, making it ideal for continuous health monitoring [243].

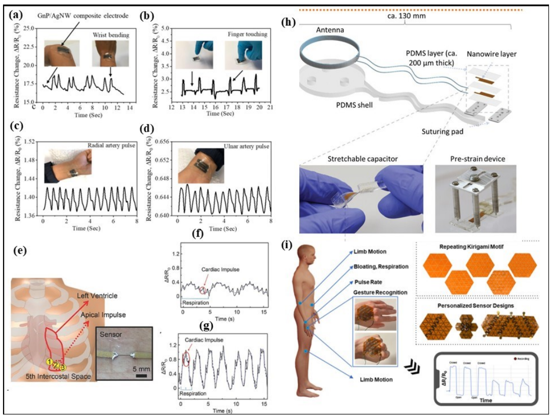

Figure 12. Real-time motion monitoring (a) wrist bending (b) finger contact, (c) radial artery pulse (d) ulnar artery pulse (2019) [238] (e) Schematic illustration of metal NPs–PDMS composite based ACG strain sensor. (f) ACG sensor response of the volunteer in the rest state. (g) ACG sensor response after exercise. Reproduced with permission [239] Copyright © 2016, Wiley-VCH. (h) In vivo monitoring of musculoskeletal soft tissue strain using wireless sensor technologies. Reproduced with permission [240] Copyright © 2023 The Authors. Advanced Materials Technologies published by Wiley-VCH GmbH (i) Examples of strain sensor mounted on different parts of body (2025) [241].

6.2 Sport and Fitness Monitoring

Stretchable, wearable strain sensors can be mounted on various parts of the body to quantify athletic performance and support fitness monitoring. In sports sensing scenarios, strain sensors complement conventional acceleration, heart rate, and blood oxygen sensing modalities by directly capturing skin and muscle deformation, providing motion information that inertial sensing alone cannot resolve. The acquired sensing data can be transmitted to a mobile device for analysis, enabling users to evaluate movement quality and plan training on the basis of quantitative monitoring data [91, 244]. Figure 13 (a) illustrates

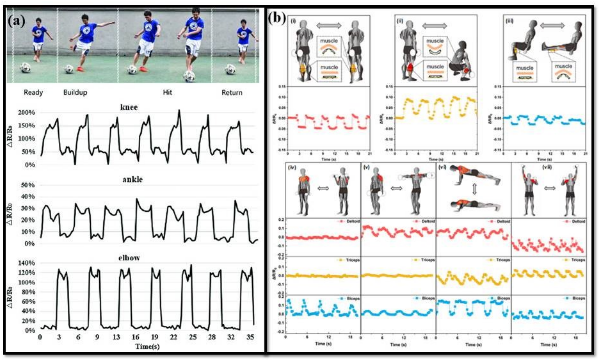

Figure 13. (a) Soccer game pictures during the shooting event, as well as the responses of strain sensors mounted to the ankle, elbow and knee. Reproduced with permission [245] Copyright © 2018, Elsevier. (b) Muscle contractions monitoring (during extensions of standing heel lifts, legs and squats) (i-iii) and during standing, the state of biceps, triceps and deltoid (iv-vii) side lateral raise, dumbbell curls, push-ups and shoulder press. Reproduced with permission [246] Copyright © 2021, Springer Nature.

the way wearable graphene-fiber strain sensors assess soccer players' shooting gestures [245]. Players had strain sensors on their knees, ankles, and elbows to monitor their sensory reaction. The strain sensors monitored the whole soccer shot process, providing detailed data for gesture analysis, shown in the figure. Figure 13(b) presents the impressive performance metrics of the fiber-junction bending sensor proposed by Li et al. [246]. The calf muscle epidermis sensor can measure and direct epidermis stresses in standing heel lifts, squats, and leg extensions (Figure 13 b (i–iii)). Furthermore, multichannel monitoring of contraction in various muscles is being analysed (Figure 13 b (iv-vii)). Three sensors were placed near the bicep, tricep, and deltoid muscles, and muscular contractions induced by standing dumbbell curls, side lateral raises, push-ups, and shoulder presses were detected simultaneously. In another instance, flexible strain sensors have been attached to the knee joint to observe and monitor varied knee motions, such as walking, running, and jumping from a hunched pose [247]. Stretchable and wearable strain sensors offer improvements in the on-going monitoring of health and fitness, rehabilitation that is compatible with human physiology, and the assessment of athletes' performance in sports [248, 249].

6.3 Human Machine Interface and Soft Robotics

Recent technologies aim to construct intelligence-based, sophisticated interactions between humans and machines. Wearable sensory systems with stretchability, flexibility, and adhesive qualities play a vital role. Smart robots can be controlled by strain sensor signals. Many smart gloves use adhesive, stretchable, and wearable strain sensors [250–252]. Wu et al. [253] reported a strain sensor based on liquid metal for human-machine interface-based applications with an ultralow detection limit, having a wide strain sensing range (105%), low detection limit of 0.05%, little hysteresis, high repeatability, and fast reaction time (58ms). This sensor successfully monitored voice intensity differentiation, blink motion, wrist pulse and heartbeat. In addition, the sensor identified numerous human gestures in smart gloves and knee caps. Figure 14(a,b) demonstrates its capacity to monitor human health and activities, as well as contact between human and machine interfaces and virtual reality applications.

Low-latency strain sensors are critical for delivering real-time control and feedback in human-machine interactions, particularly in robotic applications. Sensors with response rates under 100 milliseconds are critical for ensuring precise, seamless, and timely interaction between humans and machines. This is especially important in applications like implants, surgical robots, and collaborative robotics, where synchronized motion and quick response are required. As a result, optimizing sensor reaction time is an important design issue when building effective human-machine interfaces [254].

Wu et al. [255] developed a dual-mode piezoresistive sensor combining strain and pressure sensing capabilities for human-machine interface applications, with the sensor attached to a robotic hand (Figure 14(c)), demonstrating a pressure sensitivity of $0.088\ \mathrm{kPa}^{-1}$, a sustainable strain of 70%, and a gauge factor of 37. Truby et al. [20] work to create soft somatosensitive actuators (SSAs) based on embedded 3D printing with numerous conducting characteristics. A soft robotic gripper (Figure 14(d)) with embedded inflation, curvature, and contact sensors, including fine and deep touch sensors, gives proprioceptive and haptic feedback. Screen-printed curvature sensors on elastomeric substrates have also been demonstrated as a scalable approach for integrating strain sensing directly into soft pneumatic actuators, enabling detection of blocked motion states and supporting closed-loop control of soft robotic systems [37]. Yang et al. [256] address ultra-robust strain sensors for soft robot sensing and autonomy (Figure 14(e)). In a particular research study, to determine the bending angle of the finger joints, strain sensors were carefully attached to the fingers of the gloves. Strain sensors monitored and responded to finger joint bending, varying electrical output. Straightening the finger relaxed the strain sensor, reducing strain and restoring the electrical signal. Soft-bodied robots could use stretchable and flexible strain sensors to send input to feedback controllers and sensory artificial skins to perceive and interact with their surroundings [257, 258]. Additionally, the smart glove wirelessly communicated sensor data to the gripper robot. The robot's actions were controlled by finger movements [259]. When it comes to performing surgical operations or executing intricate and hazardous duties that are beyond the physical capabilities of the human body, the utilization of smart glove systems to remotely supervise slave robots might be advantageous. Smart glove systems with wearable strain sensors have various advantages over conventional systems (cyber gloves) [260].

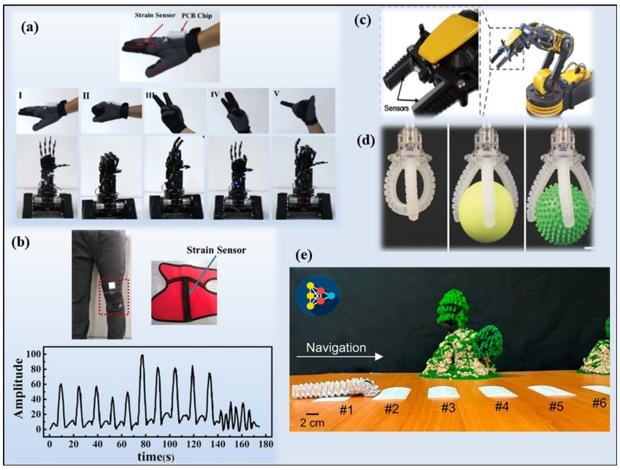

Figure 14. (a) Robotic hand operation and (b) monitoring of human knee motion amplitude in real time. Reproduced with permission [253]. Copyright © 2021, Wiley-VCH. (c) Strain sensor connected to a robotic hand gripper surface. Reproduced with permission [255]. Copyright © 2018, Elsevier. (d) Soft robotic gripper. Reproduced with permission [20]. Copyright © 2018, Wiley-VCH. (e) Robot navigation across a synthetic landscape with different heights. Reproduced with permission [256]. Copyright © 2024, Springer Nature.

7 Challenges

Despite significant progress, several fundamental sensing challenges remain unresolved. Multi-axial strain sensing represents a critical unmet need: current sensor architectures accurately detect uniaxial strain but fail to decouple simultaneous multi-directional deformations—a capability essential for robotic electronic skin and full-body motion capture [84, 256, 263]. Signal cross-sensitivity between strain and pressure further complicates strain quantification in dynamic environments where both stimuli coexist, requiring advances in transduction mechanisms, sensor array architecture, and signal processing strategies.

Packaging and integration remain important obstacles, particularly at the interface between soft sensors and stiff electrical components. Techniques such as serpentine interconnects [261] and liquid-metal electrodes [262] can accommodate mechanical mismatch during repetitive deformation, preserving stable signal transmission and improving system durability.

Signal conditioning, readout, and power modules remain predominantly rigid, limiting the mechanical compliance of the complete signal-acquisition chain [261, 262]. Although recent demonstrations of flexible energy-storage and transmission modules integrated with elastic strain sensors indicate promising directions, a fully flexible platform combining sensing, power, and data readout with consistent long-term fidelity has not yet been realised [263]. Finally, stable skin attachment is itself a sensing requirement: partial detachment corrupts strain transfer and yields erroneous readings, motivating micro-hair adhesive structures and skin-like compliant sensor architectures.

8 Outlook

Future advances in strain sensing will be driven by the demand for sensors that simultaneously achieve ultrahigh sensitivity, broad working range, fast response, and long-term stability—performance combinations that current single-material systems cannot fully deliver. Hybrid nanomaterial networks are being engineered specifically to extend the percolation-based sensitivity–range trade-off beyond its present limits, while miniaturisation and IoT integration will further expand the deployment contexts for next-generation sensing platforms.

In healthcare, the integration of strain sensors into medical devices enables precise and continuous physiological monitoring, supporting personalised treatment, rehabilitation, and chronic disease management. In industrial and structural applications, embedding distributed strain sensing allows real-time quantification of mechanical state, enabling systems to respond dynamically to changing loading conditions. Across both domains, advances in sensing materials and transduction mechanisms must be accompanied by progress in signal analytics: machine learning methods, in particular, show strong promise for decoupling complex multi-axis strain signals and improving interpretation accuracy in dynamic environments [263].

9 Conclusion

This review has examined strain sensing technology from the perspective of sensing performance: how sensitivity, stretchability, response time, linearity, hysteresis, and long-term stability can be simultaneously optimised to meet the demands of real-world monitoring applications. Advances in nanocomposite, conductive polymer, and hybrid material systems have been the primary enablers of these sensing improvements, expanding the achievable performance envelope beyond the limits of conventional metallic foil gauges. Fitness monitoring and healthcare can benefit from these sensors' precise and continuous data for injury prevention and personalized therapy. However, contemporary technologies still struggle with long-term stability, wireless system integration, performance evaluation standardization, and large-scale, cost-effective production. Addressing these constraints is crucial for future advancement. Therefore, future research should build multifunctional and self-powered sensors, improve energy efficiency, integrate sophisticated data analytics and AI, and stimulate multidisciplinary collaboration to hasten real-world deployment. Strain sensors will transform next-generation smart systems, healthcare, wearable electronics, and engineering.

Data Availability Statement

Not applicable.

Funding

This work was supported by the National Research Program for Universities (NRPU) funded by the Higher Education Commission, Pakistan, under Grant 20-15054/NRPU/R&D/HEC/2021.

Conflicts of Interest

The authors declare no conflicts of interest.

Ethical Approval and Consent to Participate

Not applicable.

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