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Soft mechanical sensors for wearable and implantable applications

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Abstract

Wearable and implantable sensing of biomechanical signals such as pressure, strain, shear, and vibration can enable a multitude of human-integrated applications, including on-skin monitoring of vital signs, motion tracking, monitoring of internal organ condition, restoration of lost/impaired mechanoreception, among many others. The mechanical conformability of such sensors to the human skin and tissue is critical to enhancing their biocompatibility and sensing accuracy. As such, in the recent decade, significant efforts have been made in the development of soft mechanical sensors. To satisfy the requirements of different wearable and implantable applications, such sensors have been imparted with various additional properties to make them better suited for the varied contexts of human-integrated applications. In this review, focusing on the four major types of soft mechanical sensors for pressure, strain, shear, and vibration, we discussed the recent material and device design innovations for achieving several important properties, including flexibility and stretchability, bioresorbability and biodegradability, self-healing properties, breathability, transparency, wireless communication capabilities, and high-density integration. We then went on to discuss the current research state of the use of such novel soft mechanical sensors in wearable and implantable applications, based on which future research needs were further discussed.

This article is categorized under:

- Diagnostic Tools > Biosensing
- Diagnostic Tools > Diagnostic Nanodevices
- Implantable Materials and Surgical Technologies > Nanomaterials and Implants

KEYWORDS

biocompatible soft electronics, health monitoring, implantable sensors, pressure, shear, soft sensor, strain, vibration, wearable sensors

1 | INTRODUCTION

Mechanical sensing stands as an essential component for human-integrated electronics owing to the myriad of mechanical stimuli exerted on, by, and within the human body, such as pressure, strain, shear, and vibration applied on the skin and by internal organs. Measuring such mechanical stimuli allows for a variety of important applications including, but not limited to, high accuracy monitoring of vital signs such as pulse rate, blood pressure, and so forth (Boutry et al., 2015; Fan et al., 2020), motion sensing for diagnosis and rehabilitation of musculoskeletal disorders (Hong et al., 2021; Sun, Tasnim, et al., 2020), mechanotransduction for tactile sensing in electronic and prosthetic skins (Tee et al., 2015; Wang et al., 2023; Ying et al., 2012), and continuous monitoring of internal organs such as the brain (Kang et al., 2016; Shin et al., 2019), blood vessels (Boutry et al., 2019; Herbert et al., 2022), small intestines (Dagdeviren et al., 2017; Nan et al., 2022), and so forth (Arab Hassani et al., 2020; Boutry, Kaizawa, et al., 2018).

Overall, most mechanical sensors fall under four categories based on the type of mechanical stimuli they sense, which are pressure sensors, strain sensors/gauges, shear sensors, and vibration sensors (Figure 1). Common to all these sensor types, the operation principles mainly involve two steps: first, the physical deformation of the sensing structure by the mechanical stimulus to be detected, and then such geometric change resulting in the change of certain electrical properties (i.e., resistance, capacitance, voltage output) of the sensor. These two steps collectively decide the main performance metrics for such mechanical sensors, including sensitivity, sensing range, specificity for the targeted type of deformation over others, response speed, signal linearity, and hysteresis. Therefore, the efforts in the development of mechanical sensors have been focused on either (1) innovations in mechanical enhancement or amplification of the physical deformation triggered by the mechanical stimuli or (2) innovations in the material and/or device designs that effectively transfer geometric changes into electrical outputs.

Broadly speaking, the recent development of soft mechanical sensors largely evolved from long-term research efforts in mechanical sensors initially in rigid form factors. Mechanical sensors consisting of rigid silicon-based microelectronics made using conventional electronics manufacturing techniques have been used since the 1970s in a wide variety of application settings, such as in the manufacturing, aerospace, and medical equipment industries, among others (Bogue, 2007; Göpel et al., 1994). More recently, this technology has been used in commercially available sensors for

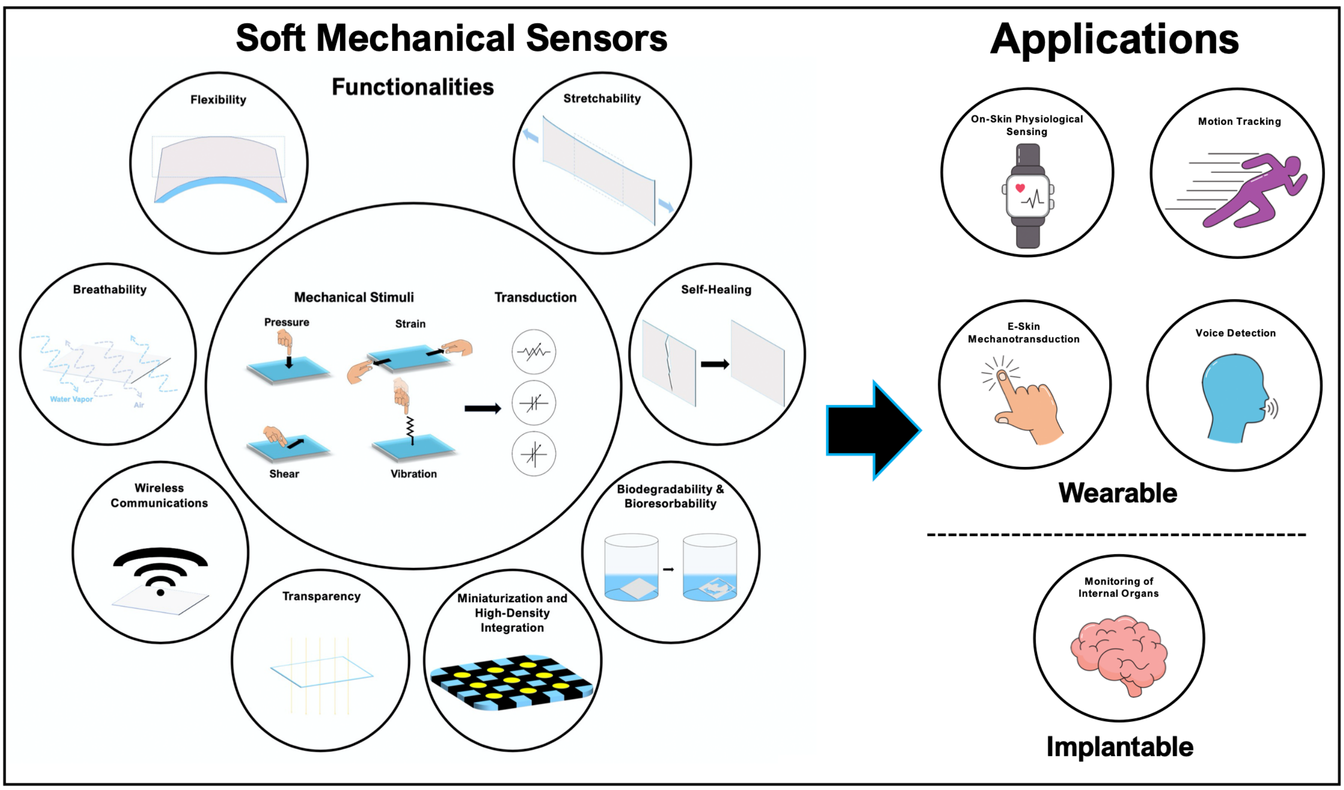

FIGURE 1 Various types of mechanical sensors and functionalities (left) and their human-integrated applications (right).

biomedical applications such as wearable and implantable hemodynamic monitoring systems like the Omron Healthcare HeartGuide™ (OMRON Healthcare, 2023) and Abbott's CardioMEMS™ HF System (Abbott, 2023). In the recent decade, there has been rapid development of mechanical sensors with greater deformability and softness than relatively more rigid MEMS-based devices in order to enable better mechanical integration of such devices with highly soft and dynamic human tissue so as to achieve better conformability, enable more effective transduction of mechanical signals, and improve long-term biocompatibility in biomedical mechanical sensing applications (Kim et al., 2012; Someya et al., 2016). Over the past couple of years, significant progress has been made in imparting flexibility and stretchability to different types of mechanical sensors while still achieving high sensitivity and other sensing characteristics. The main design strategies include both strain engineering for rigid devices such as rigid islands (Park, Jeong, et al., 2015; Yang et al., 2019), wrinkles/buckles (Chang et al., 2019; Wen et al., 2018), and kirigami cuts (Chen et al., 2021; Hong et al., 2021) as well as the use of intrinsically flexible and stretchable materials such as elastomer/nanofiller composites (Amjadi et al., 2014; Suzuki et al., 2016; Zhao, Li, et al., 2019), soft conductive polymers (Bhattacharjee et al., 2020; Choong et al., 2014), liquid metals (Cooper et al., 2017; Nan et al., 2022), hydrogels (Ge et al., 2018; Liu & Li, 2017), and ionogels (Xu et al., 2021; Zhang, He, et al., 2019).

Beyond flexibility and stretchability, the successful use of such mechanical sensors on the human body in different application scenarios also relies on other important features, including biodegradability/bioresorbability for transient implantable applications (Boutry, Kaizawa, et al., 2018), breathability for prolonged wearable uses with intimate skin contact (Lee et al., 2020), self-healability for enhanced robustness in mechanically harsh environments (Tee et al., 2012), transparency for inconspicuous device appearance (Lee, Reuveny, et al., 2016), wireless communication capabilities for improved convenience in wearable and implantable settings (Han et al., 2018), and high-density integration for high spatial resolution sensing (Wang, Xu, et al., 2018). Recent years have witnessed tremendous progress in incorporating these properties onto the four types (i.e., pressure, strain, shear, and vibration) of mechanical sensors through the use of new materials and device designs. In parallel, these newly created sensors have made several novel human-interfaced applications possible, both in the wearable and implantable spaces, which in turn not only shows the future merits of these new sensors but also provides first-hand feedback on sensor performance for the further development of such sensors.

This review will first give a brief overview of the four types of soft mechanical sensors and their working mechanisms. This will be followed by a discussion of commonly reported strategies for achieving the several major types of new design features that are important for human-interfaced applications, which include flexibility and stretchability, bioresorbability and biodegradability, self-healability, breathability, transparency, wireless communication capabilities, and high-density integration. This will further lead to a discussion about the current status of the wearable and implantable applications of soft mechanical sensors and future research needs in the field.

2 | MAJOR TYPES OF SOFT MECHANICAL SENSORS

In this section, we will review the general design considerations and strategies of mechanical sensors for the sensing of mechanical signals: pressure, strain, shear, and vibration.

2.1 | Soft pressure sensors

Pressure sensors transduce normal forces into electrical signals and have mainly been used for sensing pressure exerted by vital signs such as pulse and respiration, pressures reflecting organ/tissue conditions inside the human body, and pressures applied on robotic or prosthetic e-skins.

In general, soft pressure sensors are designed to convert deformations along the normal axis into changes in electrical performance as read-out signals. Such geometric changes are further facilitated in soft pressure sensors through the use of low-modulus silicones or hollow/porous structures (Choong et al., 2014; Chou et al., 2015; Mannsfeld et al., 2010). In particular, low-density microstructure designs, which are often combined with soft materials, help generate pronounced deformation or structural changes under pressure. Demonstrated examples of such microstructures include microdomes (Park, Lee, Hong, Ha, et al., 2014), micropillars (Park, Lee, Lim, Lee, et al., 2014), micropyramids (Choong et al., 2014; Mannsfeld et al., 2010), wrinkles (Chen et al., 2017; Gao et al., 2016), and micropores (Kwon et al., 2016), among others (Shi et al., 2018; Zhu et al., 2022) (Figure 2b).

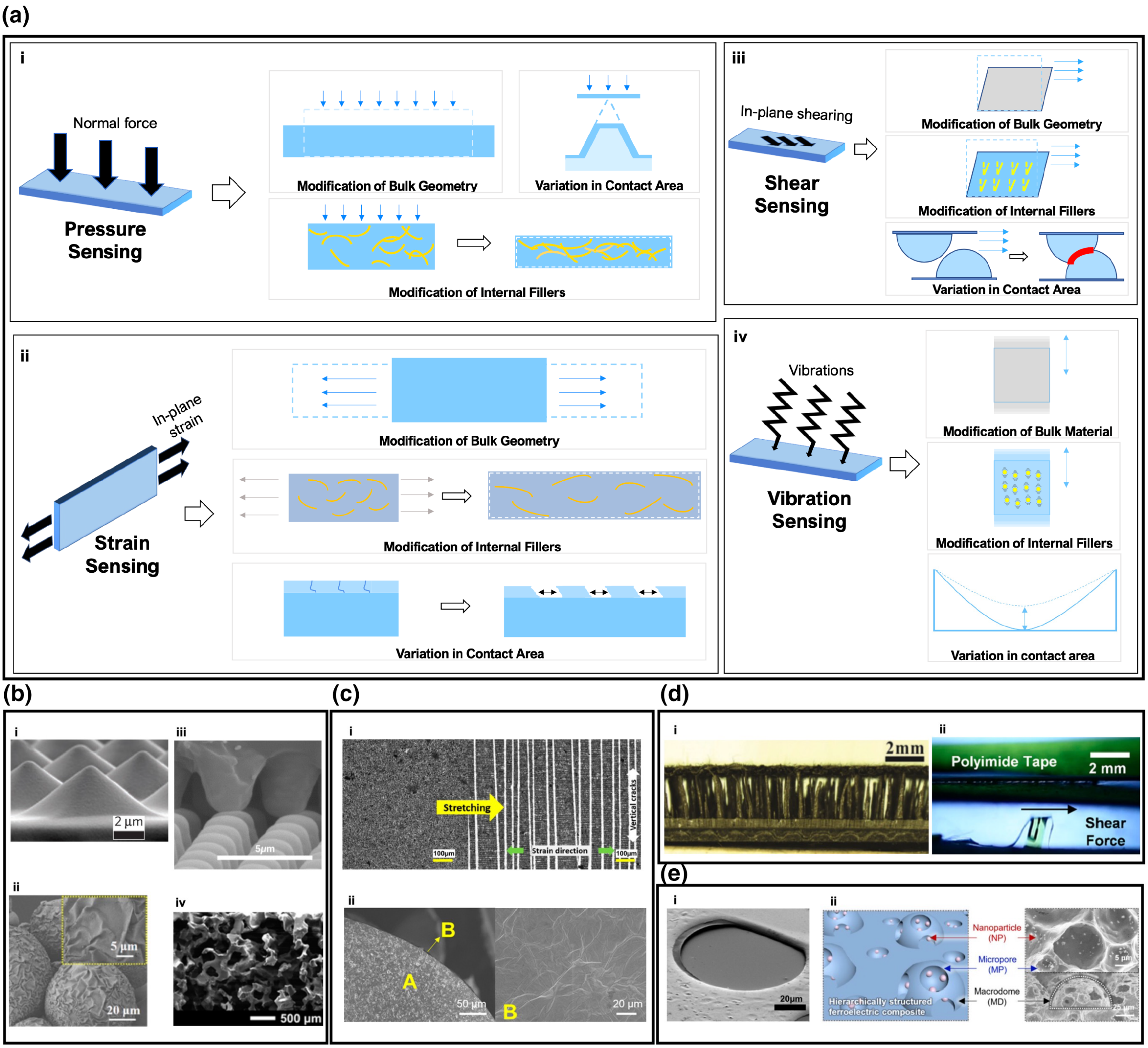

FIGURE 2 (a) Basic operation principles and device design features of the four types of mechanical pressure sensors. (i) Pressure sensors. (ii) Strain sensors. (iii) Shear sensors. (iv) Vibration sensors (b) Examples of microstructures used in pressure sensors (i) Micropyramids (ii) Hierarchically structured wrinkled micropillars (iii) Microdomes (iv) Micropores. Reprinted with permission from Mannsfeld et al., 2010; Gao et al., 2016; Park, Lee, Hong, Ha, et al., 2014; Kwon et al., 2016. Copyright 2010 Macmillan Publishers, 2016 American Chemical Society, 2014 Springer Science + Business Media New York, 2016 American Chemical Society. (c) Examples of microstructures used in strain sensors (i) Microcracks (ii) Mesh structure. Reprinted with permission from Amjadi et al., 2016; Ma et al., 2019. Copyright 2016 American Chemical Society, 2019 American Chemical Society. (d) Examples of microstructures used in shear sensors (i) Spine-shaped microstructures (ii) Vertically buckled membranes. Reprinted with permission from Ji et al., 2020; Won et al., 2019. Copyright 2020 Royal Society of Chemistry, 2019 American Chemical Society (e) Examples of microstructures used in vibration sensors (i) Micromembranes (ii) Hierarchically structured porous macrodomes. Reprinted with permission from Lee et al., 2019; Park et al., 2022. Copyright 2019 Springer Nature, 2022 American Association for the Advancement of Science.

Further transduction of such normal deformations into changes in electrical parameters mainly happens through three mechanisms (Figure 2a). First, simply through geometric changes in the overall structure, such as the reduction of the thickness of the active layer, normal pressure can cause either an increase in the electrostatic capacitance over a dielectric layer (Boutry et al., 2015; Lipomi et al., 2011; Mannsfeld et al., 2010) or a decrease in the bulk resistance of a conductor due to the piezoresistive effect (Han et al., 2018; Kang et al., 2016; Shin et al., 2019) or changes in cross-sectional area/length (Darabi et al., 2017; Gao et al., 2017). Besides these, other geometric mechanisms also include the generation of a piezoelectric potential from a piezoelectric layer (Dagdeviren et al., 2014; Park et al., 2016; Persano et al., 2013) and change in effective inductance of an inductive layer (Jang, Shin, et al., 2016; Nie et al., 2019). Second, normal compression of an active layer can lead to alterations in blended nanofillers that produce electrical signals. For instance, compression of an active layer may lead to an improvement of the percolation of blended conducting nanofillers, resulting in a decrease in the layer's resistivity (Chen et al., 2019; Hwang et al., 2011; Lee, Reuveny, et al., 2016), or compression of blended piezoelectric fillers, resulting in the generation of voltage (Liu, Zheng, et al., 2020; Yang et al., 2020). Third, an increase in the contact area between two layers in physical contact can lead to a decrease in the contact resistance between two conductors (Choong et al., 2014; Chou et al., 2015; Tee et al., 2015; Wang et al., 2023), an increase in the capacitance at an electric double layer (Nie et al., 2015; Su et al., 2021), or the generation of voltage from triboelectrically charged surfaces (Lee, Yoon, et al., 2016; Pu et al., 2017). For embodying the above mechanisms, the electric device types that have so far been used for pressure sensors mainly include resistors (Choong et al., 2014; Darabi et al., 2017; Shin et al., 2019; Wang et al., 2023), both electrostatic (Boutry et al., 2019; Kwon et al., 2016; Lipomi et al., 2011; Mannsfeld et al., 2010) and electrolyte type (Liu, Liu, et al., 2020; Nie et al., 2015; Su et al., 2021) capacitors, inductors (Jang, Shin, et al., 2016; Nie et al., 2019), transistors (Schwartz et al., 2013; Wang, Xu, et al., 2018; Zang et al., 2015), piezoelectric elements (Dagdeviren et al., 2014; Park et al., 2016; Yang et al., 2020), and triboelectric nanogenerators (Lee, Yoon, et al., 2016; Lin et al., 2013; Pu et al., 2017).

2.2 | Soft strain sensors

Strain sensors transduce tensile deformations into electrical signals. They have primarily been used for motion tracking, recording vital signs (e.g., pulse), and monitoring the operations of organs/tissues. Strain sensors show changes in their electrical parameters in response to in-plane deformations. For specific applications, strain sensors need to have sufficient stretchability for the strain range to be detected. As such, stretchable materials and structures are of particular value in this context—strategies for achieving stretchability will be reviewed in greater detail in Section 3.1.

There are three typical mechanisms for the transduction of in-plane strains into the change of a soft strain sensor's electrical parameters (Figure 2a). First, stretching-induced decreases in thickness due to Poisson's ratio and increases in length can result in changes in resistance (Liu & Li, 2017; Son et al., 2014; Xu, Wang, et al., 2019), capacitance (Atalay et al., 2017; Cai et al., 2013; Nur et al., 2018), piezoelectric potential (Gullapalli et al., 2010; Sun et al., 2018), or inductance (Fassler & Majidi, 2013; Tavassolian et al., 2020), similar to transduction mechanisms used in pressure sensors. Second, in nanofiller-based stretchable layers, strain can lead to changes in the percolated connectivity of conductive fillers (Amjadi et al., 2014; Liu et al., 2016). Third, the change of contact area resulting from strain-induced separation of individual sensing elements can increase contact resistance between conductive sensing elements (Amjadi et al., 2016; Li et al., 2014) or generate voltage signals through the triboelectric effect (Ning et al., 2022). Examples of such contact-variable sensing elements include microcracks (Amjadi et al., 2016), serpentine structures (Araromi et al., 2020; Liu et al., 2021), and mesh structures (Ma et al., 2019; Miyamoto et al., 2017) (Figure 2c).

2.3 | Soft shear sensors

Shear sensors measure forces that are tangential to a surface. In the context of the human body, they have been used to detect frictional forces on the skin (Wang, Jones, et al., 2020). Shear sensors are designed to be deformable along the direction parallel to the sensor cross-section. Relative lateral displacement of the top surface of the sensor to the bottom surface requires a low effective shear modulus, that is, low lateral stability. This has been achieved through the use of easily deformable layers made of materials such as silicones (Viry et al., 2014; Wang et al., 2021) and high aspect ratio microstructure patterns such as spine-shaped structures (Ji et al., 2020), vertical films (Li, Akiyama, et al., 2020), vertically interlocked microdomes (Park, Lee, Hong, Lee, et al., 2014), or vertically buckled membranes (Won et al., 2019) (Figure 2d).

Transduction of shear deformations into changes in electrical parameters can be achieved via the aforementioned resistive (Pang et al., 2012; Park, Lee, Hong, Lee, et al., 2014; Won et al., 2019), capacitive (Boutry, Negre, et al., 2018; Ji et al., 2020; Viry et al., 2014), piezoelectric (Liu, Li, et al., 2020), triboelectric (Wang et al., 2021), and inductive (Wang, Jones, et al., 2018) mechanisms in many ways (Figure 2a). Dimensional changes or displacement of the sensor due to shear-induced lateral displacement can lead to variations in capacitance owing to changed dielectric layer thickness (Boutry, Negre, et al., 2018; Ji et al., 2020), variations in resistance owing to changes in the contact area between conductive microstructures (Pang et al., 2012; Park, Lee, Hong, Lee, et al., 2014), generation of voltage upon variation of contact area between triboelectric layers (Wang et al., 2021), or the generation of piezoelectric voltages upon the deformation of piezoelectric layers (Li, Akiyama, et al., 2020). In sensors consisting of multiple sensing elements, variations in the readings of individual sensing units caused by the spatial heterogeneity in deformation induced by shearing can be analyzed to yield information about the direction of the shear force (Boutry, Negre, et al., 2018; Viry et al., 2014; Wang et al., 2021; Won et al., 2019).

2.4 | Soft vibration sensors

Vibration sensors transduce high frequency, and often low amplitude, mechanical deformations. Common vibrations that exist in the context of bioelectronics include the human voice, which has a frequency range of around 85–255 Hz (Fitch & Holbrook, 1970), and texture sensing in tactile sensing applications, which requires a frequency range that goes up to 200–300 Hz (Park et al., 2018).

Soft vibration sensors commonly employ elastomeric membranes/substrates. To achieve sufficiently high resonance frequency, design strategies include decreasing the thickness of the membrane/matrix and introducing an air cavity or pores throughout the membrane/matrix owing to the extremely high compressibility of air. Under vibration, sensing happens through high-frequency deformations of the elastomer (Figure 2a). By embedding active nanofillers and/or introducing mechanically sensitive structures (e.g., microcracks, microdiaphragms) (Figure 2e), vibrations can be further transduced into electrical readout through aforementioned resistive (Gong et al., 2020; Tong et al., 2022), capacitive (Lee et al., 2019), piezoelectric (Lee et al., 2014; Park et al., 2022; Yan et al., 2022), or triboelectric (Kang, Cho, et al., 2018; Kang, Son, et al., 2018) effects.

For detecting more complicated waveforms consisting of various frequencies, such as in cochlear implant applications, integration of data from multiple sensing units with different resonant frequencies is required. These sensors consist of several sensing units that differ in the area, thickness, and/or porosity of their elastomeric membrane/matrix to realize a broader frequency sensing range (Jang, Lee, et al., 2016; Lee et al., 2014; Park et al., 2022).

3 | IMPARTMENT OF HUMAN-COMPATIBLE PROPERTIES

Achieving the properties needed for human-interfaced applications while simultaneously obtaining high sensing performance has been a fundamental goal of the development of the soft mechanical sensors reviewed in this paper. In this section, we review the research efforts for several types of such human-compatible properties.

3.1 | From flexible to stretchable

Imparting devices with flexibility and stretchability is crucial to achieving highly conformable attachment to dynamic, curvilinear skin/tissue surfaces, accommodating skin/tissue deformations, and minimizing discomfort and/or tissue damage.

Moving away from traditional rigid sensors, flexibility, that is bendability, is relatively easy to achieve through the use of plastics or elastomers as substrates and thin films as functional layers such as electrodes (Boutry et al., 2015; Lee, Reuveny, et al., 2016). As such, some flexible sensors are commercially available (Novel Electronics Inc., 2023) or undergoing clinical trials (Xu, Jayaraman, & Rogers, 2019).

To further achieve stretchability for operating under large in-plane strains, special designs are needed either on the device or material level. The device-level strategies mainly comprise the use of strain engineering designs to achieve stretchability with conventional, non-stretchable materials. The most commonly used stretchable structures include buckles (Arab Hassani et al., 2020; Chang et al., 2019), serpentine patterns (Araromi et al., 2020), kirigami patterns (Chen et al., 2021; Hong et al., 2021), and helices (Gao et al., 2020; Ning et al., 2022) (Figure 3a). Strain engineering has been used to impart stretchability to rigid functional materials. Hong et al. reported a stretchable piezoelectric strain sensor able to reach strains up to 100% owing to the honeycomb kirigami pattern of rigid ceramic PZT embedded in a

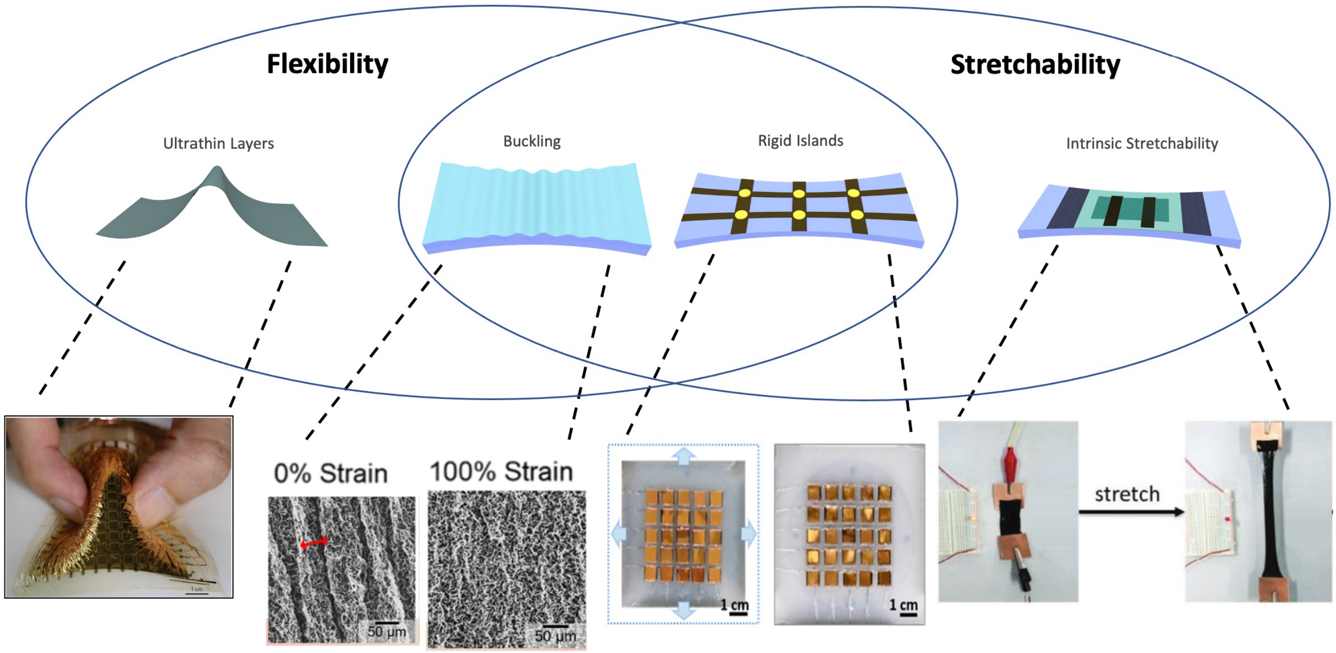

FIGURE 3 Strategies for achieving flexibility and stretchability. Common strategies for achieving flexibility and stretchability. Reprinted with permission from Lee, Reuveny, et al., 2016; Chang et al., 2019; Park, Jeong, et al., 2015; Zhang, He, et al., 2019. Copyright 2016 Macmillan Publishers Limited, 2019 American Chemical Society, 2015 American Chemical Society, 2019 WILEY-VCH Verlag GmbH & Co. KGaA, Weinheim.

stretchable PDMS matrix (Hong et al., 2021). The honeycomb pattern was achieved by mixing a PZT suspension into a textile template, which was later removed by sintering to leave behind the patterned PZT. A wide variety of other strain-engineering strategies have been explored, such as the application of pre-strain to induce wrinkling or cracking (Arab Hassani et al., 2020; Chang et al., 2019; Chen et al., 2021), winding of functional materials around substrate strands to create helices (Gao et al., 2020; Ning et al., 2022), laser-patterning to create serpentine structures (Araromi et al., 2020), among others.

Another major strategy for achieving stretchable sensors is the use of intrinsically stretchable materials to serve as stretchable functional materials. Commonly used stretchable electronic materials include elastomer/conductive filler composites (Amjadi et al., 2014; Suzuki et al., 2016), hydrogels (Liu & Li, 2017; Zhao et al., 2021), ionogels (Xu et al., 2021; Zhang, He, et al., 2019), conducting polymers and their composites (Bhattacharjee et al., 2020; Choong et al., 2014), and liquid metals (Cooper et al., 2017; Kang, Cho, et al., 2018; Kang, Son, et al., 2018). Recent advances in intrinsically stretchable electronic materials have been well summarized in several recent review papers (Kim et al., 2020; Matsuhisa et al., 2019), so we will not go into details here. Intrinsically stretchable sensor components are attractive owing to their high mechanical robustness and facile fabrication as compared to strain-engineered rigid electronics (Wang, Oh, et al., 2018).

3.1.1 | Decoupling of different deformations

A significant challenge facing flexible and stretchable pressure and strain sensors is the coupling of undesired, operation-induced mechanical deformations (i.e., bending, stretching) with the mechanical stimulus of sensing interest. While a generally applicable strategy for multi-sensor systems (e.g., arrays) is a device-level rigid-island design, which can limit such stretching/bending deformations on each sensor pixel (Park, Jeong, et al., 2015; Yang et al., 2019), such a strategy is always at the cost of the density, sensitivity, softness, and deformation of the sensing elements. Therefore, endowing sensors with intrinsic insensitivity to bending/stretching deformations is highly desirable.

For resistive pressure sensors, strain-induced changes in the performance of soft pressure sensors are mainly through alterations in conductive paths within conductive sensor components. As such, several reported strain-insensitive pressure sensors have relied on strain-dispersing structures throughout the bulk of the active layer that minimizes strain-induced changes in conductive pathways within the material. Examples of such strain-dispersing elements include nanofibers (Lee, Reuveny, et al., 2016), micropores (Kim et al., 2019), or wrinkles (Chang et al., 2019). For instance, Kim et al. reported a resistive pressure sensor consisting of a porous CNT-coated PDMS sponge that showed bending insensitivity at bending radii of up to 7.5 mm owing to the ability of the ~300–500 μm micropores throughout the bulk of the sensor to compress in response to bending-induced strains, ensuring that the local strain on individual micropores is minimized. As such, bending does not result in appreciable changes in the contact area, and so the sensor is sensitive to normal forces as opposed to bending-induced strains (Kim et al., 2019).

For the general micropatterned soft pressure sensor design, strategies have also been proposed for achieving intrinsically strain-insensitive pressure sensing without sacrificing stretchability, such as introducing mechanical micro-hierarchical designs. Specifically, our group used a combination of an electrical double layer (EDL)–based capacitive sensing mechanism and hierarchical microstructure to achieve strain-insensitive pressure sensing up to strains of 50% in addition to high-pressure sensitivity of up to 4.5 kPa^{-1} (Figure 4a) (Su et al., 2021). The sensor's strain-unperturbed sensing capability arises from the locally stiffened electrodes underneath each micropyramid. This design ensures that

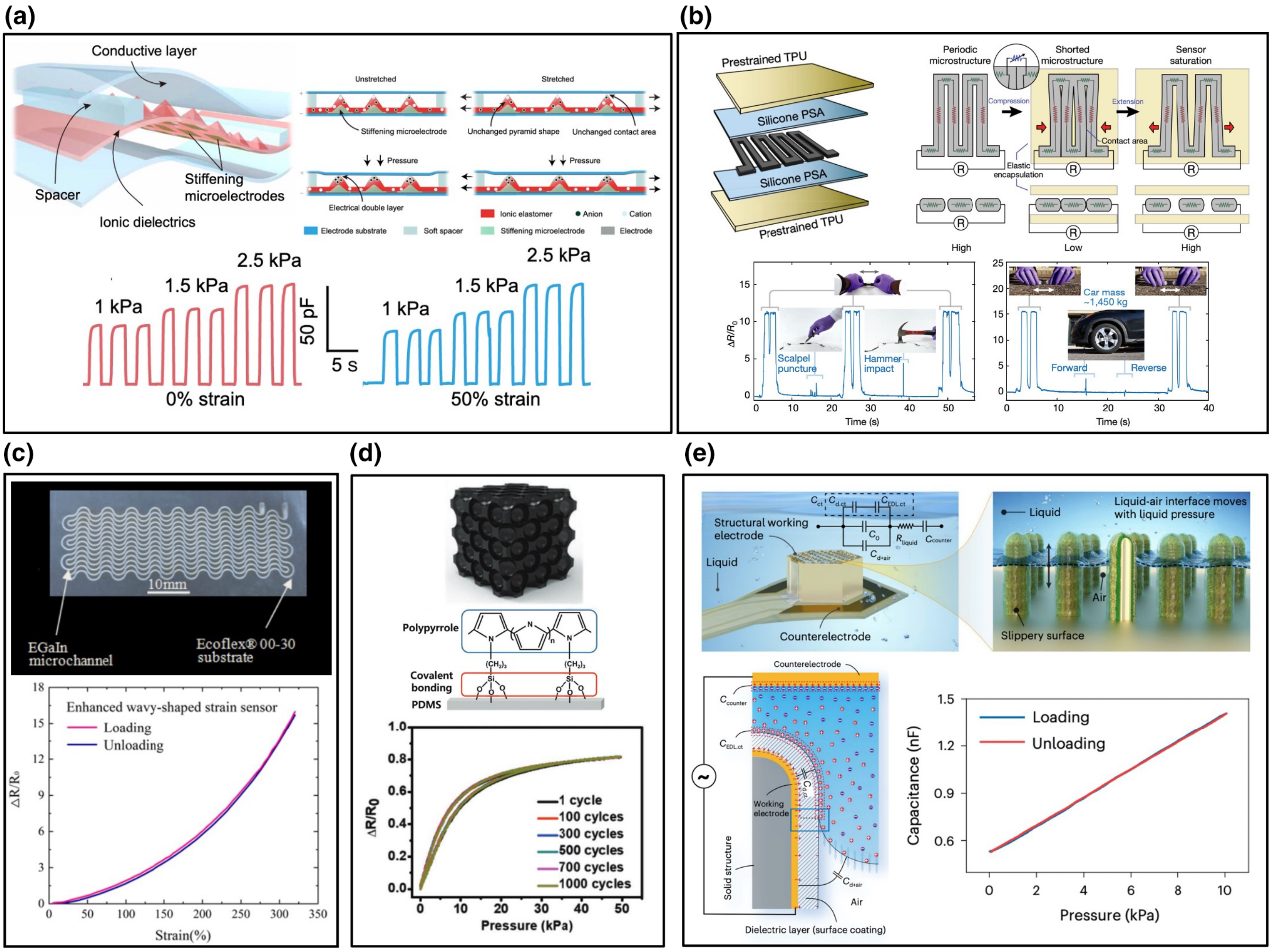

FIGURE 4 Strategies for minimizing hysteresis and decoupling of different deformations and inputs. (a) Strain-insensitive stretchable pressure sensor. Reprinted with permission from Su et al., 2021. Copyright 2021 American Association for the Advancement of Science. (b) Pressure, bending, and twisting-insensitive strain sensor using sensing elements arranged in serpentine meander/interdigitated pattern. Reprinted with permission from Araromi et al., 2020. Copyright 2020 Springer Nature Limited. (c) Strain sensor showing low hysteresis enabled by liquid metal resistive sensing element and wavy channel design. Reprinted with permission from Chen, Zhang, et al., 2020. Copyright 2020 American Chemical Society. (d) Pressure sensor showing low hysteresis enabled by covalent grafting of the resistive layer to the elastomeric substrate in order to minimize slippage between layers. Reprinted with permission from Oh et al., 2019. Copyright WILEY-VCH Verlag GmbH & Co. KGaA, Weinheim. (e) Highly linear, low hysteresis pressure using frictionless mobile air–water interface as a pressure-sensing mechanism. Reprinted with permission from Cheng et al., 2023. Copyright 2023 Springer Nature Limited.

the micropryamids do not deform under stretching, resulting in an unchanged contact area between the top and bottom electrodes and, thus, unchanged ionic capacitance under strain. It should be noted that while the relative stiffness of the bottom electrode micropyramids enabled strain-unperturbed sensing, it did not significantly impair sensor stretchability or modulus.

With regard to strain sensors, the decoupling of normal pressures has primarily been achieved through the use of sensing mechanisms that rely on the separation of sensor components solely along the axial plane of the sensor. For instance, numerous resistive and capacitive pressure-insensitive strain sensors have employed interdigitated electrodes that generate a strain-induced decrease in capacitance (Atalay et al., 2017; Kim, Kim, & Park, 2017; Zhang et al., 2022) or an increase in contact resistance (Araromi et al., 2020; Liu et al., 2021). Using such an approach, Aaromi et al. reported a resistive strain sensor consisting of a laser-patterned interdigitated carbon fiber polymer composite electrode that could withstand a normal force of ∼1400 kN without showing any significant electrical response (Figure 4b) (Araromi et al., 2020). Such interdigitated electrodes have also been made using conductors such as liquid metal (Zhang et al., 2022), silver nanowires (Kim, Kim, & Park, 2017), and conductive fibers (Atalay, 2018; Liu et al., 2021), among other materials.

For shear and vibration sensors, the designs for decoupling bending and stretching influence have been limited so far. Although some of the design strategies for pressure and strain sensors can also be applicable, the unique operation mechanisms of these two types of sensors should also be considered when designing new decoupling strategies for them.

3.1.2 | Minimizing hysteresis

While intrinsically stretchable materials can indeed possess desirable mechanical characteristics, their viscoelastic nature can result in hysteresis effects in mechanical sensors (Hwang et al., 2020). To overcome such effects, a few strategies have been implemented. For example, intrinsically low hysteresis functional components such as ionic liquid (Choi et al., 2017) or liquid metal conductors (Chen, Zhang, et al., 2020) have been used to lower hysteresis in intrinsically stretchable strain sensors. In particular, Chen et al. used microchannels EGaIn in an Ecoflex substrate to fabricate a resistive strain sensor that showed a low hysteresis of ∼5% owing to the low viscosity of the liquid metal EGaIn, which was enclosed in a wavy channel shape to reduce energy dissipation in the encapsulating elastomer layer (Figure 4c) (Chen, Zhang, et al., 2020). Besides the use of materials that intrinsically display low hysteresis, the aforementioned microstructure designs (e.g., micropyramids, microdomes, etc.) have commonly been used to reduce hysteresis in pressure (Choong et al., 2014; Su et al., 2021), shear (Boutry, Negre, et al., 2018; Park, Lee, Hong, Lee, et al., 2014), and vibration (Kang, Cho, et al., 2018; Kang, Son, et al., 2018; Tong et al., 2022) sensors due to the reduction of the viscoelastic nature of elastomers, owing to the presence of air gaps within the elastomer matrix (Ruth et al., 2020). In addition to microstructure designs, other morphologies, such as microcracks (Gu et al., 2020; Yao et al., 2020), have also been used to further reduce hysteresis.

Because mechanical sensors consisting of elastomers typically operate on the basis of the deformation of a relatively low-modulus elastomeric substrate or matrix, hysteresis has also been partly attributed to slippage between conductive components and the elastomeric substrate or matrix on the application of a mechanical stimulus due to poor adhesion between the two. As such, improving adhesion between such components through various strategies including, but not limited to, chemical grafting (Oh et al., 2019), surface energy modification (Liang, Sun, et al., 2023), and microphase separation (Shen et al., 2022) have been reported to reduce hysteresis in pressure and strain sensors. For instance, Oh et al. reported a resistive pressure sensor wherein the conductive polypyrrole layer was covalently bonded to the underlying PDMS porous substrate, preventing the relative sliding of layers (Figure 4d) (Oh et al., 2019). The sensor showed a low degree of hysteresis of 2% owing to the strong bonding between layers.

In addition to the use of intrinsically low hysteresis materials and strategies to mitigate hysteresis caused by the viscoelasticity of elastomeric materials, alternative sensing mechanisms have also been explored to tackle the issue of hysteresis. For instance, taking advantage of movable liquid–gas contact lines present atop a highly hydrophobic surface due to entrapment of air underneath overlying water, Cheng et al. demonstrated a capacitive pressure sensor showing extraordinarily low hysteresis owing to the friction-free motion of the contact line at the water-trapped air interface, which on moving due to the application of pressure would change the interfacial capacitance of the electric double-layer present between the water and electrodes (Figure 4e) (Cheng et al., 2023). In all, while significant progress has been made in tackling the issue of hysteresis in soft mechanical sensors, further research into near-complete elimination of hysteresis under the sorts of harsh mechanical conditions wearable and implantable sensors are exposed to in real use cases is crucial in order to enable highly repeatable long-term sensing performance.

3.2 | Bioresorbability and biodegradability

Implantable sensing for short-term physiological, post-surgery monitoring represents a critical application of soft mechanical sensors. As such, the ability of sensors to degrade under physiological conditions into non-toxic substances that can easily be absorbed by the body—that is, bioresorbability—is an incredibly useful functionality.

Bioresorbability has been achieved through the use of materials that can be hydrolyzed into non-toxic, soluble, and bloodstream-absorbable substances. Popular polymer options mainly include synthetic polyesters such as poly-lactic acid (PLA) (Boutry et al., 2019; Boutry, Kaizawa, et al., 2018; Curry et al., 2018; Ouyang et al., 2021), poly(1,-8-octanediol-co-citrate) (POC) (Ouyang et al., 2021), polyhydroxybutyrate/polyhydroxyvalerate (PHB/PHV) (Boutry et al., 2019), poly(octamethylene maleate (anhydride) citrate) (POMaC) (Boutry et al., 2019), poly(lactic-co-glycolic acid) (PLGA) (Kang et al., 2016; Lu et al., 2020), polyglycerol sebacate (PGS) (Boutry et al., 2019; Boutry, Kaizawa, et al., 2018), and polycaprolactone (PCL) (Palmroth et al., 2019; Palmroth et al., 2020). These insulating polymers are commonly utilized to fabricate non-conductive sensor components such as substrate, dielectric, and encapsulation layers. Naturally derived polymers such as silk fibroin and cellulose, which can undergo enzymatic degradation (Jiang et al., 2018), have also been used to fabricate such non-conductive sensor components. Conductive and semiconductive resorbable sensor components are commonly fabricated from metals and metalloids that can be hydrolyzed into soluble, non-toxic hydroxides, such as magnesium (Boutry et al., 2019; Boutry, Kaizawa, et al., 2018; Curry et al., 2018; Ouyang et al., 2021), molybdenum (Curry et al., 2018; Kang et al., 2016), zinc (Lu et al., 2020), silicon (Kang et al., 2016; Shin et al., 2019), germanium (Zhao, Xue, et al., 2022), and so forth. In 2019, using a selection of these biodegradable materials, including Mg as electrodes, PGS as a micropyramidal dielectric layer, and PLLA and POMAC as the encapsulation, Boutry et al. reported an implantable, bioresorbable capacitive sensor for monitoring arterial pulse post-vessel anastomosis (Figure 5a) (Boutry et al., 2019). The sensor was capable of wireless communication enabled by inductive coupling (which will be discussed in further detail in Section 3.6). The sensor showed complete dissolution 3 months post-implantation and did not trigger a significant immune response around the implantation site. Overall, bioresorbable sensors open up the possibility for various transient, implantable applications. Moving toward actual clinical

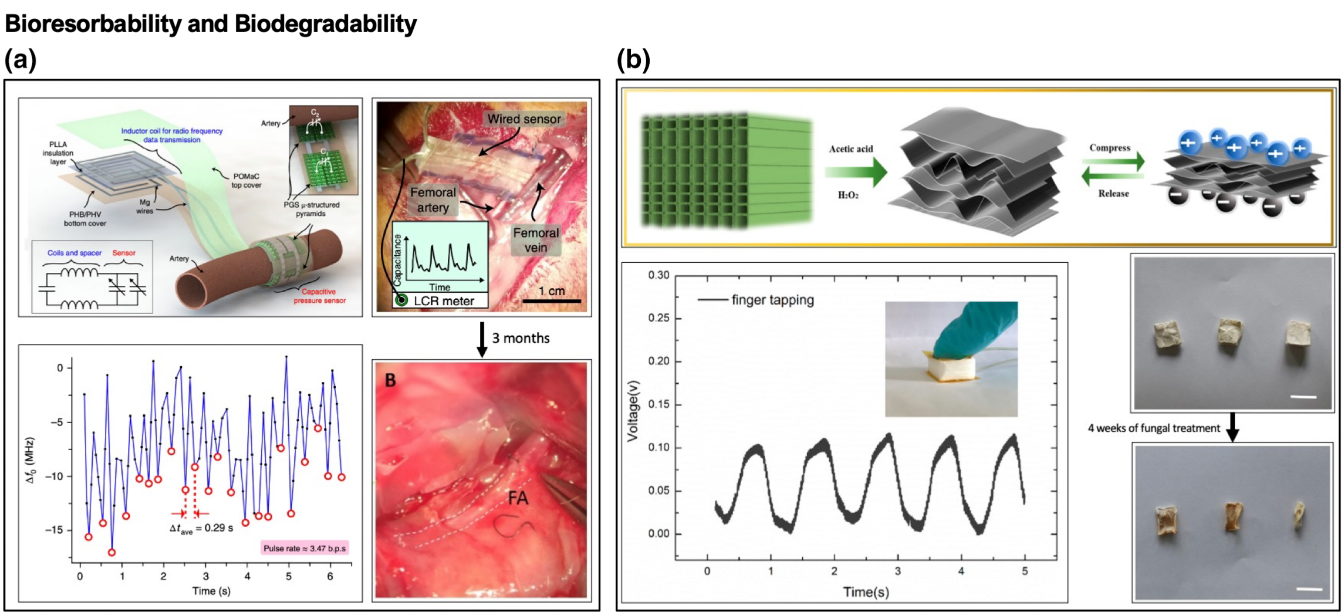

FIGURE 5 Bioresorbable and biodegradable mechanical sensors. (a) Bioresorbable capacitive sensor for arterial pulse monitoring. Reprinted with permission from Boutry et al., 2019. Copyright 2019 Springer Nature Limited. (b) Biodegradable piezoelectric pressure sensor made of delignified wood. Reprinted with permission from Sun, Guo, et al., 2020. Copyright 2020 American Chemical Society.

application, more rigorous assessments of the long-term device toxicity of bioresorbable sensors in animals and the human body are still needed.

Beyond degradation under physiological conditions, biodegradability, in general, is also an important feature for the alleviation of the burden of plastic pollution associated with soft sensors, thereby reducing their environmental impact. An increasingly diverse range of materials have been employed in the production of biodegradable sensors through either physical compositing or chemical modification strategies that can be degraded. For instance, biodegradable conductive components in piezoresistive sensors have been fabricated through the impregnation of a biodegradable substrate such as cotton (Wei et al., 2016) and paper (Gao et al., 2019; Guo et al., 2019) with conductive nanofillers. In other works, biodegradable piezoelectric layers have been fabricated through the modification of naturally derived materials like wood (Sun, Guo, et al., 2020) and glycine (Hosseini et al., 2020). For instance, Sun et al. used acetic acid/hydrogen peroxide to delignify balsa wood into a piezoelectric wood sponge capable of undergoing fungal degradation (Figure 5b) (Sun, Guo, et al., 2020).

3.3 | Self-healability

Imparting sensors with self-healing properties can make them better withstand unexpected physical damages and preserve device functions over extended periods of time. Self-healing properties have largely been achieved through the use of materials designed to be capable of reversible dynamic bonding such as hydrogen bonding (Chakraborty et al., 2019; Chen, Koh, et al., 2020; Tee et al., 2012), π-π stacking (Xu, Wang, et al., 2019), electrostatic interactions (Cao et al., 2018; Cao et al., 2019; Wang, Zhang, et al., 2018), metal–ligand coordination (Li et al., 2016; Liu et al., 2018; Zhang et al., 2018), and dynamic covalent bonding (e.g., Diels-Alder Zhao, Jiang, & Huang, 2019), imine bonds (Zhang et al., 2020), disulfide linkages (Khatib et al., 2020, etc.). An extended overview of these materials has been well covered in several review papers published recently (Kang et al., 2019; Khatib et al., 2021).

For fabricating mechanical sensors, self-healing elastomer matrices can be directly used to replace the insulating components as substrate, dielectric, and triboelectric layers (Kang, Cho, et al., 2018; Kang, Son, et al., 2018; Xu et al., 2017; Ying et al., 2020; Zhang et al., 2018). In particular, Kang et al. used a dielectric layer made of PDMS modified with 4,4′ methylenebis(phenyl urea) and isophorone bisurea to fabricate a self-healing capacitive strain sensor wherein the urea-based hydrogen bonds provided the PDMS self-healing properties (Figure 6a) (Kang, Cho, et al., 2018; Kang, Son, et al., 2018). When cut into half, the sensor could self-heal to nearly reinstate pre-cut performance.

For the fabrication of self-healing conductive components in mechanical sensors, the most commonly reported strategy is the dispersion of conductive fillers into a self-healing elastomeric matrix to form conducting percolating pathways (Guo et al., 2020; Liu et al., 2018; Tee et al., 2012; Zhang, He, et al., 2019). In addition to conductive fillers, conductive polymers (Chakraborty et al., 2019) and electrolytes (Chen, Koh, et al., 2020) have also been used to facilitate conductivity, especially in self-healing hydrogel matrices. Besides conductivity, other forms of functionality, such as piezoelectric properties, have been imparted to self-healing matrices through the dispersion of piezoelectric fillers into self-healing matrices (Yang et al., 2021).

As self-healing needs to happen between each respective layer in a multi-layer sensor device, an open challenge for introducing device-level self-healing properties to multi-layer sensor designs is achieving accurate alignment and physical contact of the same layer at a damage front. This requires the development of new materials and/or device design concepts for enabling controlled and guided alignment for all the layers in a sensor device before or during the self-healing processes. To this end, Cooper et al. introduced a strategy of using identical dynamic bonds but immiscible properties between different polymers to achieve autonomous alignment (Figure 6b) (Cooper et al., 2023). While the identical dynamic bonds allow for bonding between the modified polypropylene glycol (PPG) and PDMS layers, the immiscibility of the two polymers ensures preferential bonding of each respective separated polymer layer to itself and largely restricts the bonding between the separated layers of the respective polymers to the interfaces of the different layers. Using this strategy, the authors demonstrated the accurate autonomous realignment of a misaligned and cut film of 11 alternating layers of modified PPG and PDMS of varying thicknesses. The authors further implemented this strategy to fabricate an autonomously realigning self-healing capacitive pressure sensor consisting of electrodes and a dielectric layer made of immiscible polymer layers loaded with conductive and dielectric nanoparticles, respectively.

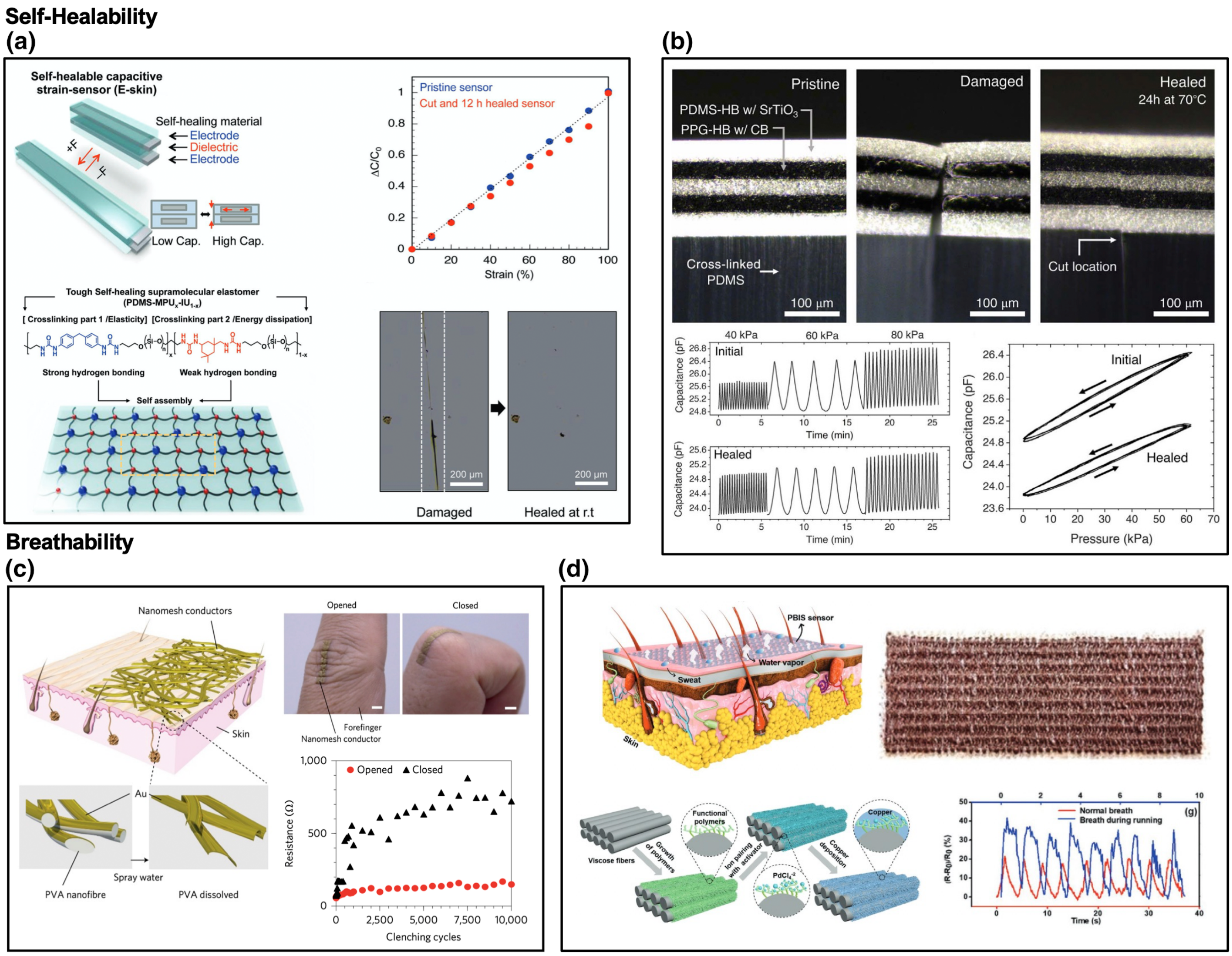

FIGURE 6 Self-healable and breathable sensors. (a) Self-healable elastomeric strain sensor made of liquid metal enclosed by a self-healing elastomer. Reprinted with permission from Kang, Son, et al., 2018, Kang, Cho, et al., 2018. Copyright 2018 WILEY-VCH Verlag GmbH & Co. KGaA, Weinheim. (b) Self-healable pressure sensor capable of autonomous alignment of separate layers at damage front. Reprinted with permission from Cooper et al., 2023. Copyright 2023 American Association for the Advancement of Science. (c) Breathable strain sensor made of gold nanowire mesh. Reprinted with permission from Miyamoto et al., 2017. Copyright 2017 Macmillan Publishers Limited. (d) Fabric-based strain sensor consisting of copper-coated viscose yarn embroidered with Lycra fibers. Reprinted with permission from Liu et al., 2021. Copyright 2021 Wiley-VCH GmbH.

3.4 | Breathability

Ensuring that devices are breathable is crucial to minimizing discomfort and skin irritation due to wearable sensors (Miyamoto et al., 2017). In particular, breathability refers to the permeability of the sensor to air and water vapor so that perspiration can freely evaporate. Breathability in sensors has been achieved by the use of sensor components with porous morphologies such as fibrous or spongey structures.

Fibrous morphologies have been achieved at both the nanoscale and macroscale. Common strategies for achieving a nanoscale fibrous structure include the electrospinning of a nanomesh of a desired material (Yang et al., 2018; Yu et al., 2021) or deposition of a desired material through a dissolvable template prepared by electrospinning (Lee et al., 2020; Miyamoto et al., 2017). Conductive nanofibrous layers have been achieved by the electrospinning of functional materials (Lee et al., 2020) or the conductive functionalization of a nanofibrous substrate (Chao et al., 2021; Qiao et al., 2022). Materials of widely varying properties have been used to fabricate fibrous and breathable substrate layers (Qiao et al., 2022; Yang et al., 2018; Yu et al., 2021), dielectric layers (Fu et al., 2020; Lee et al., 2020; Yu et al., 2021), piezoresistive layers (Chao et al., 2021; Miyamoto et al., 2017; Wang, Chao, et al., 2020), triboelectric layers

(Peng et al., 2020) and piezoelectric layers (Kim, Wu, et al., 2018). As for the use of such materials in developing mechanical sensors, Miyamoto et al. reported a highly breathable pressure sensor based on a nanomesh design (Figure 6c) (Miyamoto et al., 2017). By depositing gold onto electrospun PVA nanofibers which were subsequently dissolved to leave behind a gold nanomesh, a highly gas-permeable on-skin conductive mesh was obtained that was capable of piezoresistive pressure sensing through pressure-induced variations in conductive pathways. The nanomesh sensor was assessed by participants to induce less feelings of discomfort than a 1 mm thick silicone film and 1 μm thick parylene film, which the authors attribute to the high gas permeability of the nanomesh design. The use of nanofibrous sensor components to minimize the irritation associated with an on-skin sensor has been extended to more complicated device structures. For instance, Lee et al. reported a capacitive pressure sensor consisting of electrospun nanofibrous electrodes and a dielectric layer laminated onto each other that displayed on-skin tactile imperceptibility (Lee et al., 2020). This indicates the general potential for nanofibrous materials to enable high comfort in a wide variety of sensor designs.

In addition to nanofibrous morphologies, functionalized spongey polyurethane scaffolds have also been used to fabricate conductive layers in breathable piezoresistive pressure sensors as well, indicating the wide array of porous morphologies available to achieve breathability in mechanical sensors (He et al., 2020; Wang, Chao, et al., 2020).

At a macroscale, fibrous morphologies have commonly been achieved through the use of textile substrates such as nylon and cotton. Such textiles have been made conductive to serve as electrodes or piezoresistive components by coating textile fibers with conductive materials using methods such as polymer-assisted metal deposition (Liu et al., 2021), dip-coating (Liu, Li, et al., 2020) or wrapping textile fibers around a conductive core fiber (Lou et al., 2020; Zhang, Wang, et al., 2019). So far, textiles have been used to create resistive, capacitive, and triboelectric sensors for pressure (Fan et al., 2020; Liu, Li, et al., 2020; Zhang, Wang, et al., 2019) and strain (Chen et al., 2017; Ning et al., 2022). Liu et al. reported a textile-based resistive pressure sensor using copper-coated viscose yarn embroidered with Lycra yarn (Figure 6d) (Liu et al., 2021). The group was able to make viscose yarn conductive using a polymer-assisted metal deposition process. The yarn-based sensor was seamlessly integrated with clothing—a unique advantage of textiles for human-interfaced applications—and was able to accurately monitor motions associated with breathing and joint movement.

3.5 | Transparency

Certain applications require transparent sensors, such as smart contact lenses (Kim et al., 2021; Kim, Kim, Lee, et al., 2017) with intraocular pressure sensing capabilities or inconspicuous wearable sensors (Pu et al., 2017; Ren et al., 2019) for measuring motion or mechanical physiological signals like heart rate or respiration rate. Transparency in active and substrate layers of mechanical sensors has been achieved through the use of intrinsically transparent materials as well as highly porous mesh structures that owe their transparency to the presence of voids.

Some commonly used materials in soft mechanical sensors such as plastics (Nie et al., 2015) and elastomers (Ho et al., 2017; Pu et al., 2017) as well as their composites readily show high optical transmittance and have been used as transparent substrate (Ho et al., 2017; Pu et al., 2017), dielectric (Kim, Kim, Lee, et al., 2017), triboelectric (Pu et al., 2017), and piezoelectric (Kim, Jang, et al., 2018) layers. ITO (Nie et al., 2015) and PEDOT:PSS (Choong et al., 2014; Wen et al., 2018) have been used as typical materials for transparent electrodes. In addition, hydrogel and hydrogel composites have also been reported to serve as transparent ion-conducting electrodes (Ge et al., 2018; Jing et al., 2019; Pu et al., 2017) and iontronic (Nie et al., 2015) layers. For example, Pu et al. reported a transparent triboelectric pressure sensor for tactile sensing using elastomeric substrate and triboelectric layers made of either PDMS or VHB and a hydrogel transparent electrode consisting of PAAm-LiCl (Figure 7a) (Pu et al., 2017). With the substrate and triboelectric layers made of either PDMS or VHB, the device showed high transparency of up to 96.2% transmittance for visible light and stretchability of up to 1160% uniaxial strain.

In addition to intrinsically transparent conductive materials, another commonly reported strategy for achieving transparent and conductive layers involves the use of highly porous conductive nanomaterial networks. Examples include metal nanomeshes (Ho et al., 2017; Kim et al., 2021; Li, Zhao, et al., 2020), hybrid metal/carbon nanomaterial nanomeshes (Kim, Kim, Lee, et al., 2017; Lee et al., 2013), and electrospun nanofibers embedded with conductive carbon nanoparticles (Lee, Reuveny, et al., 2016; Wang et al., 2017). Using such a material design, Ho et al. fabricated a transparent and stretchable resistive strain sensor consisting of a conductive mesh network of silver nanowires bridged by gold nanowires with optical transparency of up to 66.7% and stretchability of up to 70% (Ho et al., 2017). In

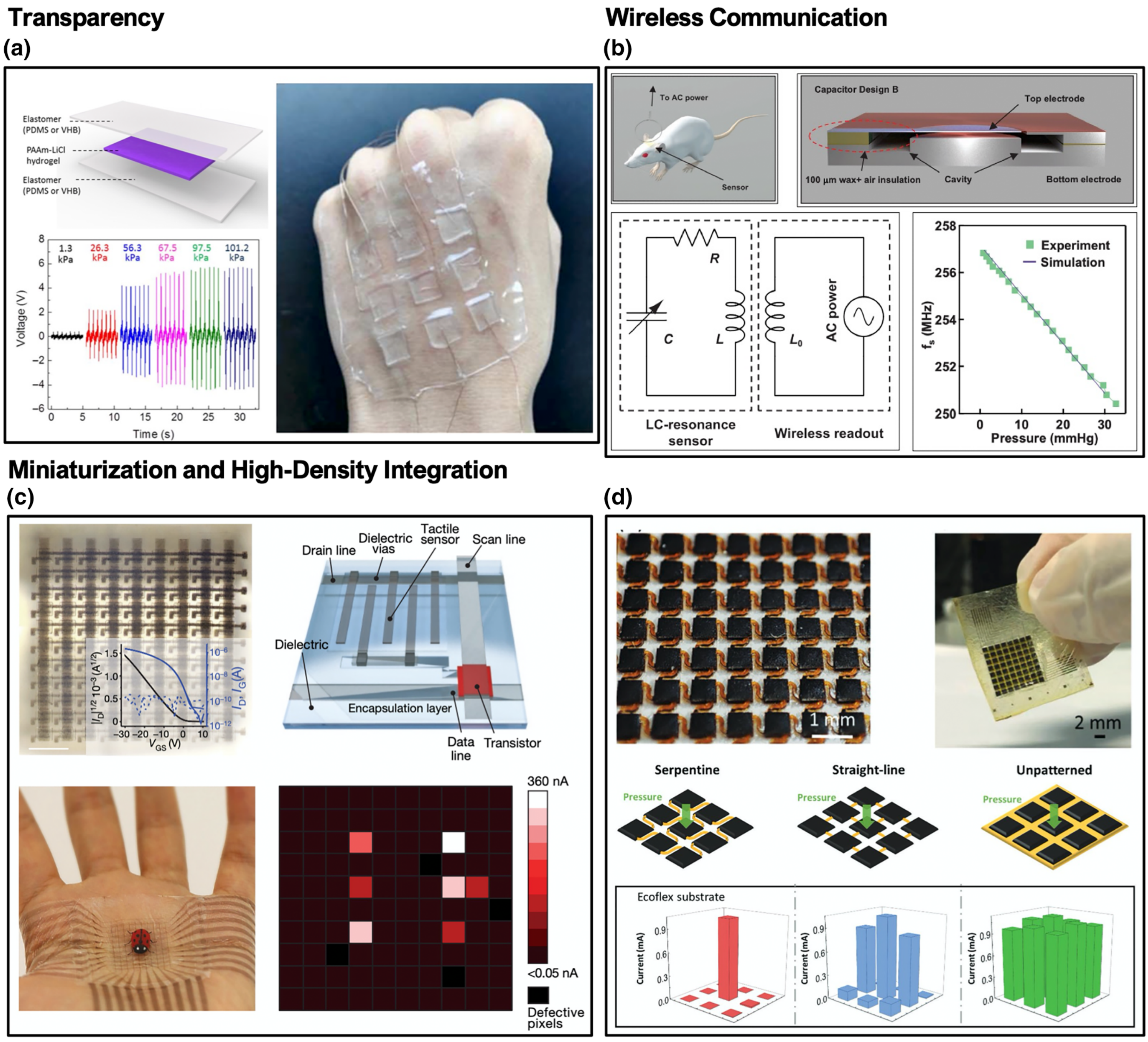

FIGURE 7 Examples of transparent, wireless communication-enabled, and high-density array integrated sensors (a) Transparent triboelectric pressure sensor using a hydrogel triboelectric layer. Reprinted with permission from Pu et al., 2017. Copyright 2017 American Association for the Advancement of Science. (b) RFID-based capacitive pressure sensor for wireless sensing of internal pressures. Reprinted with permission from Lu et al., 2020. Copyright 2020 Wiley-VCH GmbH. (c) High-density transistor array integrated with tactile sensors. Reprinted with permission from Wang, Xu, et al., 2018. Copyright 2018 Macmillan Publishers Limited. (d) Laser-manufactured high-density pressure sensor array with crosstalk minimization through serpentine interconnects. Reprinted with permission from Li et al., 2022. Copyright 2022 Wiley-VCH GmbH.

particular, the gold-nanowire bridging helped to achieve high conductivity while also maintaining low mesh density and, thus, high transparency.

3.6 | Wireless communications

The ability of mechanical sensors to communicate wirelessly is an important functionality for more convenient and continuous data collection from wearable and implantable mechanical sensors. Reported wireless communications methods include Bluetooth, near-field communications, and radio frequency identification.

One strategy for enabling wireless communications in soft mechanical sensors is the integration of rigid communication modules such as Bluetooth (Abramson et al., 2022; Kwak et al., 2020; Liu, Liu, et al., 2020; Park et al., 2020; Zhu et al., 2019) and NFC (Han et al., 2018; Kim et al., 2021). While such modules offer some benefits, such as being highly standardized and robust, they pose certain challenges for human-integrated applications. Both modules are rigid, limiting modulus compatibility between sensor and tissue. Moreover, Bluetooth modules, in particular, are further limited by their requirement for an active power source.

A promising strategy that overcomes both the issues of rigidity and the requirement for an active power source is radio frequency identification (RFID) (Boutry et al., 2019; Chen et al., 2014; Deng et al., 2018; Lu et al., 2020). Relying on inductive coupling between an inductor coil attached to the sensor and an external AC-voltage transponder, RFID enables wireless, passive communication. Typical sensor designs include a capacitive mechanical sensor connected to an inductor coil, creating an RLC circuit. Variations in the capacitance of the sensor due to a mechanical stimulus such as pressure or strain result in variations in the resonant frequency of the RLC circuit that may be detected by an external transponder from which information about the mechanical stimulus on the sensor may be extrapolated. Reported inductor coils typically consist of a thin film of metals such as Cu (Chen et al., 2014), Mg (Boutry et al., 2019; Lu et al., 2020), liquid metal (Niu et al., 2019), and so forth, patterned on a flexible or stretchable substrate. The simplicity of this inductor coil design has allowed for several additional features to be imparted to the inductor coil, such as stretchability (Niu et al., 2019) and bioresorbability (Boutry et al., 2019; Boutry, Kaizawa, et al., 2018). An example of a wireless sensor having bioresorbable properties was demonstrated by Lu et al. The group used RFID-based technology to fabricate capacitive pressure sensors for the wireless monitoring of internal physiological pressures (Figure 7b) (Lu et al., 2020). Consisting of a Zn top electrode, Mg spacer, bottom electrode, and inductive coil, as well as PLGA substrate layers, the sensor showed a sensitivity as high as 200 kHz $mmHg^{-1}$ and maintained stable operations in vivo up to 4 days before undergoing significant hydrolyzation into bioresorbable materials.

3.7 | Miniaturization and high-density integration

The capability of mechanical sensors for spatially-resolved mapping of mechanical stimuli, in addition to sensing their magnitude, is important for many applications such as wearable touch panels, prostheses, and so forth. Such a feature is especially important in the context of pressure sensing for e-skins, given the high spatial resolution of the human skin, which can require sensor pixel sizes as low as 1–2 mm (Dargahi & Najarian, 2004). In general, the fabrication of such arrays mainly requires the patterning of different components (e.g., conductors, dielectrics, and sometimes semiconductors) in the sensor design. Typical patterning techniques include inkjet printing (Baek et al., 2022; Lee et al., 2022; Shimura et al., 2023; Wang, Xu, et al., 2018), photolithography (Boutry, Negre, et al., 2018; Shimura et al., 2023; Wang, Xu, et al., 2018; Zhao, Tang, et al., 2022), and laser ablation (Li et al., 2022).

The most basic array designs can be made just by patterning the top and bottom electrodes into crossbar structures, giving rise to passive arrays (Pan et al., 2014; Woo et al., 2014). However, such passive matrix designs have the limitation of electrical crosstalk between adjacent pixels due to shunt current. To minimize the electrical crosstalk, active matrices are created by integrating a sensor array with an array of transistors to control the on/off of each pixel during the operation (Baek et al., 2022; Wang, Xu, et al., 2018; Zhao, Tang, et al., 2022). Wang et al. reported a high-density stretchable transistor array with up to 347 transistors per square centimeter capable of high spatial resolution pressure sensing, among other functions. The authors demonstrated a 10×10 transistor array integrated with tactile sensors with a resolution of one sensor per 2 mm (Figure 7c) (Wang, Xu, et al., 2018). However, as the design of the sensor can only provide on/off information about the touch of conductive objects, more complicated sensor designs with quantitative measurements of mechanical stimuli such as pressures need to be further integrated.

Mechanical crosstalk between miniaturized sensing units is a major challenge facing high-density sensing arrays and can lead to diminished signal mapping accuracy. This is especially prominent in flexible and stretchable sensing arrays, given the delocalization of mechanical deformation owing to the low modulus of commonly used soft materials. As such, effort has been made to minimize such crosstalk while retaining small pixel size through strategies that localize strain near individual sensing units. Miniaturized interconnects (Li et al., 2022; Zheng et al., 2023) and spacers (Pyo et al., 2017; Zhang et al., 2023) fabricated through the aforementioned high-resolution optical micropatterning techniques are a promising strategy for strain-localization in high-density arrays. By modeling interconnects as cantilevers, Li et al. found that laser-patterned serpentine structure interconnects between individual sensing units significantly reduced mechanical cross-talk between individual pressure sensing units by approximately two orders of magnitude compared to an unpatterned array (Figure 7d) (Li et al., 2022). Laser ablation of a polyimide substrate at varying power levels was used to convert polyimide into resistive laser-induced graphene and to pattern the substrate between sensing units into the desired interconnect shape. In this way, an array of resistive sensing pixels connected by serpentine interconnects was obtained.

4 | WEARABLE AND IMPLANTABLE APPLICATIONS

4.1 | Wearable applications

There have been a number of reported wearable applications of soft mechanical sensors ranging from prosthetic skins to implantable devices. The high conformability of soft mechanical sensors allows for intimate contact with soft biotissues and minimized sensor artifacts. Overall, demonstrated wearable applications for biomedical applications mainly include the following categories: mechanoreception for tactile restoration, physiological signal sensing for continuous health monitoring, motion detection for tracking of musculoskeletal function, and acoustic sensing for wearable microphones.

4.1.1 | Mechanoreception for tactile restoration

Soft mechanical sensors have been developed as artificial mechanoreceptors for "electronic skins" (e-skins) for the restoration of impaired tactile sense in amputees and patients with spinal cord or peripheral nerve injuries through neuroprosthetic technologies. This type of application requires sensors to not only mimic the four types of cutaneous mechanoreceptors (i.e., two slowly adaptive and two rapidly adaptive) but also to display mechanical and functional properties of skin, such as stretchability, self-healing properties, and the ability to interface with nervous tissue. In recent decades, substantial progress has been made in achieving these goals. Using an interlocked microdome structure of PVDF/rGO, which is both conductive and piezoelectric, Park et al. fabricated a flexible sensor capable of resistively detecting static tactile stimuli as well as piezoelectrically detecting dynamic stimuli such as surface textures and slipping between skin and objects, thereby achieving skin-like mechanotransduction (Park, Kim, et al., 2015). Beyond the replication of the skin's mechanotransduction abilities, significant progress has been made in imparting such sensors with properties and functionalities of the skin, such as stretchability and the ability to interface with nervous tissue. Using carbon nanotube-based stretchable piezoresistive sensors combined with stretchable circuits, Wang et al. were able to achieve pressure sensing capable of interfacing with the nervous system in a frequency-modulated manner, resembling action potentials involved in biological mechanotransduction, thereby achieving skin-like neural communication (Figure 8a) (Wang et al., 2023). Upon the application of pressure, the resistance of the sensor, which is attached to the ring oscillators, decreases the load resistance of the ring oscillators, resulting in increased frequency of the ring oscillator output. The output from the ring oscillators is then converted into action-potential-like voltage spikes by a stretchable edge detector circuit. For demonstrating a biological perception-actuation loop, the sensor system was interfaced with the sciatic nerve of a rat to induce greater leg contraction on the application of greater pressure.

Besides applications in tactile restoration for human patients, skin-like mechanotransduction enabled by mechanical sensors also has important uses in robotics, especially bioinspired soft robotics, for imparting human-like tactile sensation, and thereby physical intelligence, to such machines. Overall, the designs and features of such sensors are subject to the same requirements as those required for use on the human body, except for not having the need to interface with biological nervous systems. Moving forward, such skin-mimetic functions and properties need to be extended to other types of mechanosensation. Moreover, improved sensing density and resolution must be realized.

4.1.2 | Physiological signal sensing for continuous health monitoring

On-body measurement of physiological signals such as heart rate, pulse waveforms, and respiration rate makes up another important application of mechanical sensors by enabling high comfort and continuous health monitoring in various use cases, from fitness tracking to disease detection. Much progress has been made in fabricating sensors sensitive enough to detect the subtle physiological signals of the aforementioned types while achieving highly comfortable

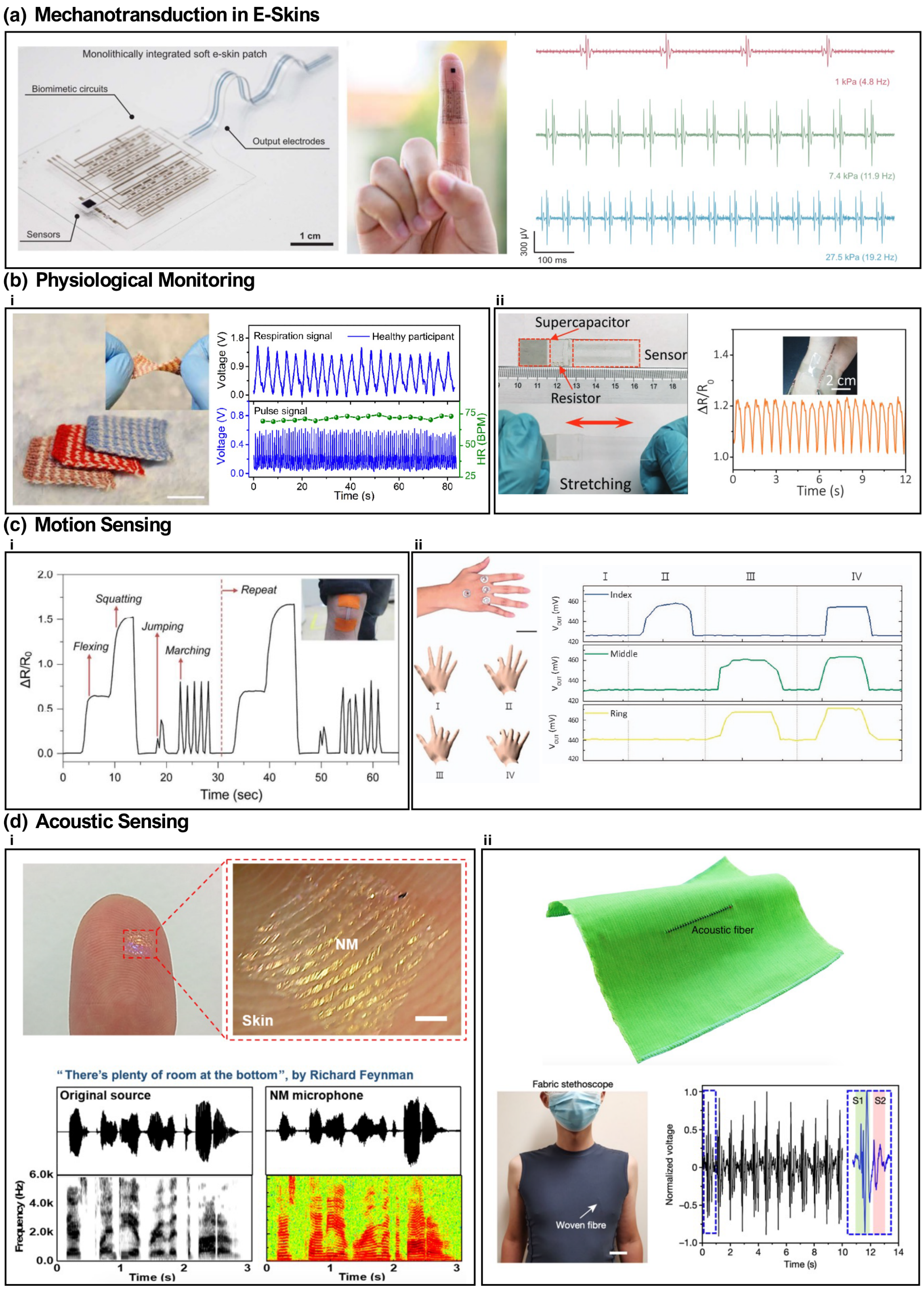

FIGURE 8 Legend on next page.

and even imperceptible skin attachment. Fan et al. reported a textile-based triboelectric pressure sensor for wearable monitoring of arterial pulse and respiratory signals (Figure 8b (i)) (Fan et al., 2020). Made of triboelectric layers of yarn with a stainless-steel core and nylon, the sensor showed a high sensitivity of 7.84 mV Pa^{-1} and a fast response time of

20 ms, allowing for high-accuracy detection of arterial pulse waves and respiratory signals while also being highly breathable and washable. The group further demonstrated the sensor's potential for high comfort by integrating it seamlessly with a shirt. Beyond integration with clothing, progress has also been made in more seamlessly integrating physiological sensors directly with the body. Using a mesh network of MXenes bridged by silver nanowires embedded in a PDMS matrix, Liang et al. reported a stretchable and transparent strain sensor with a high gauge factor of up to 220 under 1% strain and a fast response time of 50 ms (Figure 8b (ii)) (Liang, Sheng, et al., 2023). Such performance allowed for the accurate monitoring of highly subtle physiological signals such as wrist pulse. At the same time, its transparency and stretchability enabled visual and on-skin imperceptibility.

For this application direction, future developments of soft mechanical sensors need to achieve more precise control of sensitivity and sensing range to allow for better measurements of different types of physiological signals. Toward realizing continuous health monitoring, innovations are also needed at the system level for integrating soft mechanical sensors with other electrical components for data processing and even wireless communication.

4.1.3 | Motion detection for tracking of musculoskeletal function

Wearable mechanical sensors, especially strain sensors, have also been demonstrated for use in motion sensing, with possible applications in the diagnosis of musculoskeletal disorders, rehabilitation, monitoring of athletic activity, and human–machine interfaces. Strain sensors used for such applications are often designed with high stretchability in order to operate under large strain ranges associated with muscular movements. Using an ethylene glycol/sodium chloride mixture enclosed by Ecoflex, Choi et al. reported a highly stretchable ionic liquid-based resistive strain sensor with a sensing range of up to 300% strain, allowing for the measurement of knee flexion associated with a wide variety of motions such as squatting, jumping, and marching (Figure 8c (i)) (Choi et al., 2017). Beyond sensing performance, progress has been made in making such sensors better suited for wearable motion-sensing applications with minimal restrictions to motions and activities in daily life. To this end, efforts have been made to integrate wireless operations. Jeong et al. reported a skin-attachable, stretchable strain sensor for motion monitoring capable of wireless communication (Figure 8c (ii)) (Jeong et al., 2017). Using a liquid metal (GaInSn) based resistive strain sensor interfaced with an NFC chip, wireless sensing up to 30 cm was realized for motions ranging from swallowing to finger and wrist flexion.

For this research application, further development into improving the communication distance of such wireless data while retaining high on-skin conformability and comfort could make motion sensing even more convenient, thereby better enabling the collection of highly accurate biomechanical data unimpaired by relatively rigid communication modules or wires. Strategies involving more sophisticated system-level integration such as networked communication between passive, stretchable on-skin sensors and relatively rigid textile-attached readout circuits could potentially be used to achieve high on-skin sensor conformability and long-distance communication simultaneously (Niu et al., 2019).

4.1.4 | Acoustic sensing for wearable microphones

Acoustic sensing realized by soft mechanical sensors has important uses in wearable hearing aids, artificial larynxes, and human–machine interfaces. Mostly, this type of sensing function is served by vibrational sensors. In particular, as acoustic vibrations typically have low amplitude and high frequency, vibrational sensors need to have characteristic

FIGURE 8 Examples of wearable applications. (a) Stretchable pressure sensor connected to a stretchable circuit capable of biomimetic amplitude-decoupled frequency modulated output in an e-skin system. Reprinted with permission from Wang et al., 2023. Copyright 2023 American Association for the Advancement of Science. (b) (i) High-comfort physiological sensing enabled by textile-based (Fan et al., 2020) and (ii) stretchable, transparent (Liang, Sheng, et al., 2023) sensors. Copyright 2020 American Association for the Advancement of Science and 2023 Wiley-VCH GmbH. (c) (i) Motion sensing enabled by highly stretchable and (ii) wireless sensors. Reprinted with permission from Choi et al., 2017 and Jeong et al., 2017. Copyright 2017 American Chemical Society and 2017 Springer Nature. (d) (i) Highly soft, transparent on-skin microphone (Kang, Cho, et al., 2018; Kang, Son, et al., 2018) and (ii) acoustic fabric (Yan et al., 2022). Reprinted with permission from Kang, Son, et al., 2018, Kang, Cho, et al., 2018 and Yan et al., 2022. Copyright 2018 American Association for the Advancement of Science and 2022 Macmillan Publishers Limited.

resonant frequencies close to the acoustic frequency under measurement. Since the resonant frequency is inversely proportional to the Young's modulus of the structure, it can be a particular challenge for soft vibration sensors to achieve high enough vibrational frequencies.

In recent works, progress has been made in achieving wearable acoustic sensors that seamlessly integrate with the human body. Kang et al. reported a highly conformable and transparent microphone using a triboelectric sensor consisting of a freestanding conductive membrane made of an orthogonal silver nanowire array for highly comfortable on-skin usage capable of facilitating accurate voice recognition (Figure 8d (i)) (Kang, Cho, et al., 2018; Kang, Son, et al., 2018). The sensor could accurately record sound waveforms and was used to recognize voice commands with an average reliability of 98.6%, closely matching the average reliability of 99.1% achieved by a commercial microphone while also maintaining conformable and imperceptible skin contact.

In addition to improving the comfort of sensors placed directly on the skin, progress has been made in integrating vibration sensors with clothing. Yan et al. reported a piezoelectric fiber consisting of an elastomer/piezoelectric nanoparticle composite that showed recording performance comparable to a commercial microphone, with a sensitivity of 19.6 mV at 94 dB and 1 kHz while also retaining a fibrous form factor that can easily be woven to fabric and machine washed (Figure 8d (ii)) (Yan et al., 2022). The authors demonstrated the use of such acoustic fibers in detecting speech waveforms as well as in auscultating heart sounds, thereby showing its potential in a wide range of human-interfaced applications.

For this application direction, further areas for development include sensitivity improvement for better-detecting low-volume sound, broader frequency bandwidth (especially on stretchable substrates), and the incorporation of wireless capabilities to enable seamless voice-controlled human–machine interfacing applications (e.g., electrolarynx).

Overall, while much progress has been made in achieving the fabrication of soft sensors highly sensitive to a variety of mechanical stimuli, further research is needed in the implementation of soft mechanical sensors in practical wearable applications. System-level integration of soft wearable mechanical sensors with other important components such as soft power sources, communication modules, or actuators is crucial. Furthermore, enabling wireless communication in sensors involved in continuous monitoring is essential for maximizing comfort, and ultimately, the practicality of soft sensing modalities. Beyond this, imparting all components of such systems with adequate encapsulation and properties such as self-healability and hydrophobicity is essential to maintaining long-term performance under the harsh conditions they are exposed to on the skin.

4.2 | Implantable applications

Implanted mechanical sensors can potentially provide invaluable information for tissue/organ monitoring, postoperative monitoring, and restoration of lost tissue/organ function. Owing to the need for intimate contact between such sensors and the internal organs of the body, biocompatibility is a key consideration. Furthermore, additional features, such as wireless communication capabilities and bioresorbable properties, are required for fully implanted sensing function and surgery-free sensor removal at the end of sensor life, respectively. To date, significant progress has been made in the use of soft mechanical sensors in three major application directions, as reviewed below.

4.2.1 | Implanted tissue/organ monitoring

Several implantable sensors have been reported for the monitoring of conditions of various tissues/organs, including the gastrointestinal tract (Dagdeviren et al., 2017), connective tissue (Boutry, Kaizawa, et al., 2018; Lee et al., 2021), blood vessels (Boutry et al., 2019; Herbert et al., 2022), heart (Dual et al., 2020), thoracic cavity (Li et al., 2023), and brain (Kang et al., 2016; Shin et al., 2019). Dagdeviren et al. reported a flexible piezoelectric sensor based on PZT nanoribbons for diagnosing and treating gastrointestinal motility disorders (Dagdeviren et al., 2017). The sensor was implanted into the stomach of a pig model and could clearly differentiate between pre and post-ingestion states of the stomach.

For ease of implantation, one strategy is to integrate such sensors with commonly used medical implants that have well-developed surgical procedures. In this regard, Herbert et al. reported the integration of miniaturized, flexible pressure sensors with a wireless and batteryless stent for hemodynamic monitoring (Figure 9a) (Herbert et al., 2022). Through implantation into the right iliac artery of a rabbit using a balloon catheter, this device realized accurate

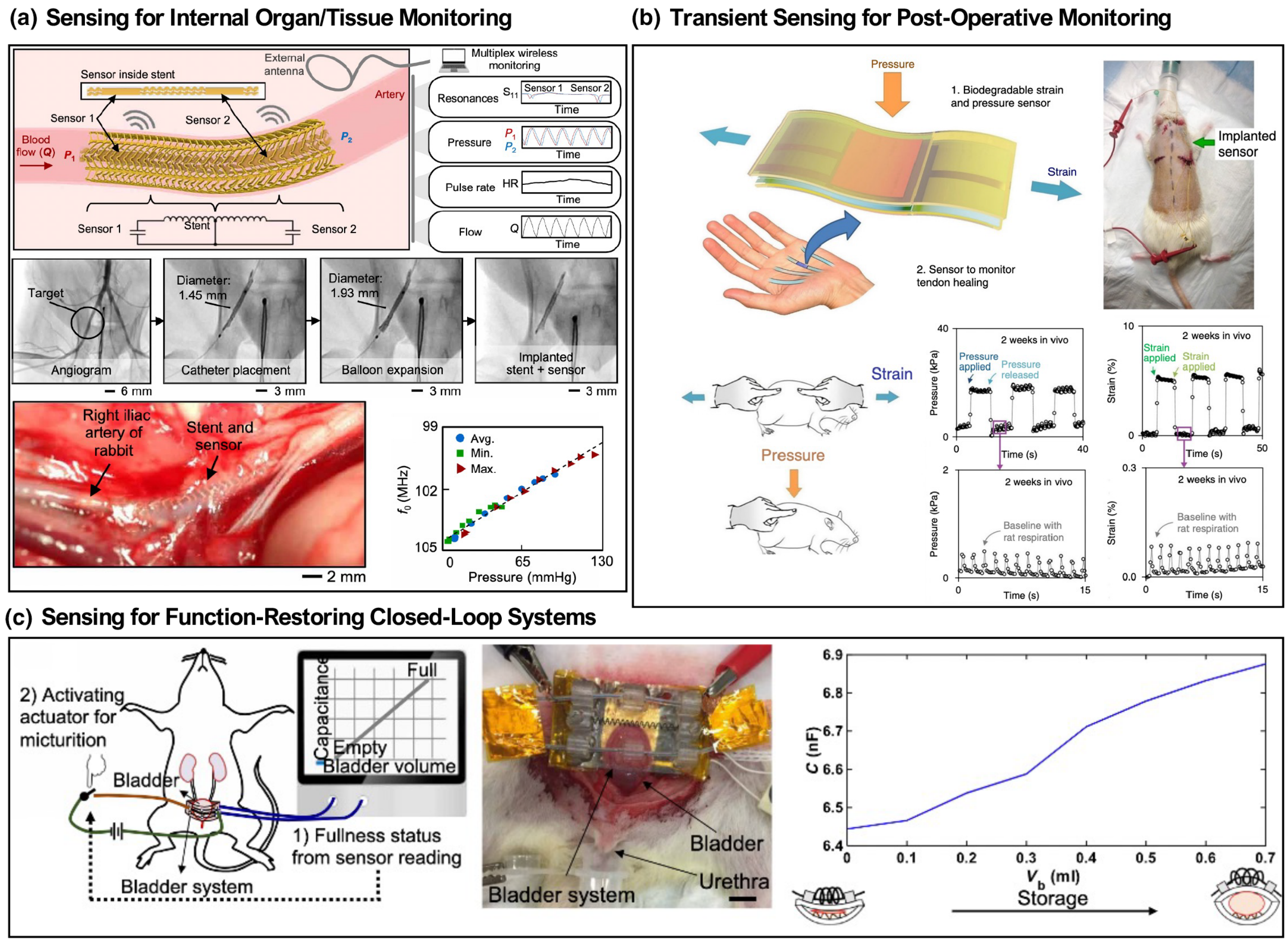

FIGURE 9 Examples of implantable applications. (a) Pressure sensor/smart stent for wireless hemodynamic monitoring. Reprinted with permission from Herbert et al., 2022. Copyright 2022 American Association for the Advancement of Science. (b) Bioresorbable multiplex pressure/strain sensor for transient monitoring of tendon healing. Reprinted with permission from Boutry, Kaizawa, et al., 2018. Copyright 2018 Macmillan Publishers Limited. (c) Strain sensor for measuring bladder volume in a sensing-actuation system for restoration of bladder function. Reprinted with permission from Arab Hassani et al., 2020. Copyright 2020 American Association for the Advancement of Science.

sensing of various hemodynamic parameters, including hemodynamic pressure, pulse rate, and flow. The stent simultaneously serves as an inductor coil antenna for RF-based wireless communication, which, coupled with its long-term performance stability, gives it the potential to chronically monitor the state of cardiovascular diseases.

Overall, the main need for future development of soft mechanical sensors for implantable uses is to suppress long-term immunogenic responses so as to minimize inflammation and other side reactions. In addition, new materials and device designs for hermetic encapsulation are also highly critical for the stability of device functions in physiological environments.

4.2.2 | Postoperative monitoring

Another promising application of soft implantable sensors is post-operative monitoring. Biodegradable and/or bioresorbable properties are critical to such an application so that such sensors may eventually dissolve into the body in a relatively short-term frame without the need for a second surgery to remove the sensor. To this end, Boutry et al. reported a stretchable and bioresorbable strain and pressure sensor for monitoring tendon healing for improved rehabilitation following tendon repair surgery (Figure 9b) (Boutry, Kaizawa, et al., 2018). The sensor, composed entirely of bioresorbable polymers and metal thin films, was shown to dissolve in PBS solution at 37°C over a period of 8 weeks as evidenced by its progressively diminished fatigue life with time submerged. Such bioresorbable properties were subsequently realized for wireless pressure sensors for arterial pulse monitoring, as discussed in Section 3.2 (Boutry et al., 2019). The sensor's wireless communication capabilities make it ideal for implantable post-operative monitoring following surgeries such as vessel anastomoses. Intracranial pressure is another important physiological parameter to monitor in healing processes following neurological pathologies such as traumatic brain injury or hydrocephalus. Shin et al. reported bioresorbable pressure sensors made of ultrathin hydrolyzable inorganic materials with a measurement accuracy comparable to that of a clinical ICP monitor (Shin et al., 2019).

While current biocompatibility data of such implantable sensors is promising, further investigation into the long-term health effects of the dissolved products of these sensors is crucial to achieving clinical translation. Also, to have more precise control of biodegradation processes, stimulation-triggered degradation properties will be the next level of design to gradually move toward, for which new material and device innovations will be needed.

4.2.3 | Restoration of lost tissue/organ function

Due to neurological diseases, surgical removal of under-skin tissue/organ, and physical injuries, mechanosensation on certain parts of the body could be lost. Implantation of soft mechanical sensors under the skin could help to restore mechanical sensation through neuroprosthetic technology. In general, this is the same type of application as reviewed in Section 4.1.1 but with more intimate integration with the body, thereby possessing higher requirements on biocompatibility and hermetic encapsulation. One such example is a subcutaneously implantable triboelectric sensor implanted into the hindfoot of a mouse that was connected to the mouse's transected distal tibial nerve (Shlomy et al., 2021) via a cuff electrode. The authors reported that the mice with the implanted sensor had a significantly lower pressure threshold for reacting to a force applied on their paws as compared to those mice with transected distal tibial nerves but no implanted sensor. Sensor encapsulation consisted of two layers of waterproof VHB tape sealed at the edges with biocompatible fibrin glue. The sensor improved rat hind paw sensitivity up to 19 days post-implantation, showing the encapsulation layer's ability to at least partially serve as an effective biofluid barrier and preserve sensor function over an extended period of time.

Another function of mechanical sensors in such applications is to monitor the physical state (e.g., volume) of an organ and form closed-loop systems with stimulation devices to bypass a dysfunctional nervous system. For instance, for patients with loss of bladder sensation due to neurogenic underactive bladder (UAB), Arab Hassani et al. reported the use of a capacitive strain sensor on the bladder for monitoring bladder volume in a closed-loop system that could trigger urination (Figure 9c) (Arab Hassani et al., 2020). Consisting of wrinkled gold electrodes on a prestrained elastomer substrate, the stretchable sensor was able to deform together with the volume expansion of the bladder and could linearly correlate bladder volume with the capacitance readout with a sensitivity of 0.7 μF/liter. The sensor was further integrated with a shape memory alloy-based actuator capable of squeezing the bladder to achieve urination, possibly providing a future alternative to urinary catheters in cases of UAB. A 600 nm thick layer of parylene was used to encapsulate the sensor from biofluids.

Overall, there has been tremendous progress in the field of soft implantable mechanical sensors with the achievement of sensors displaying non-toxicity, biodegradability, wireless communication capabilities, as well as integration into closed-loop systems. Such sensors offer the great clinical promise by allowing for continuous data collection of previously inaccessible internal organ conditions in a biocompatible way. Further research is required to ascertain the long-term biocompatibility with suppressed foreign-body responses by combining materials science with immunology. Moreover, research into the development of highly passivating yet mechanically compliant encapsulation is required to ensure robust sensor performance in the harsh biofluidic environment of the body over extended periods of time. Detailed studies into the hermeticity of such soft encapsulation layers, along with encapsulated sensor performance over time under implanted conditions, must be done to enable chronic usage of function-restoring implantable sensors. Furthermore, new device designs and surgical methods are also needed to achieve minimally invasive implantation.

5 | SUMMARY AND OUTLOOK

In this review, we have discussed the major types of soft mechanical sensors for wearable and implantable applications as well as some commonly reported strategies for achieving functionalities specific to these various applications. Besides the continued improvement of key sensing metrics (e.g., sensitivity, response time, repeatability), significant progress has been made in incorporating skin/tissue conformable and compatible properties for long-term stable and minimally-invasive operations. As discussed in this review, these properties include, but are not limited to, flexibility and stretchability, biodegradability and bioresorbability, self-healability, breathability, transparency, wireless communication capabilities, and high-density integration. The achievement of these properties comes from innovations in both material and device designs. These new properties have greatly expanded the uses of soft mechanical sensors in wearable and implantable applications, as we reviewed in this paper.

Toward realizing clinical and commercial applications of such sensors, we envisage that major advancements need to be made in several aspects to overcome the limitations associated with them. Briefly, some major avenues for further research in the field of soft mechanical sensors include:

1. Improvements in sensor performance: While there has been significant progress in improving the sensitivity and sensing range associated with soft mechanical sensors, further optimization of sensor performance is required for robust and reliable sensing in real-use cases. In particular, the prevalence of elastomers in the fabrication of soft mechanical sensors necessitates further development in reducing the hysteresis and response time associated with their viscoelasticity. While progress has been made in the reduction of hysteresis and response time associated with soft mechanical sensors (as discussed in Section 3.1.2), more general implementation of hysteresis-minimizing strategies across a wider range of sensors consisting of different materials would pave the way forward for universally repeatable sensing performance among soft mechanical sensors. Furthermore, improving the durability of sensor components to ensure stable device performance over several cycles of mechanical deformation and harsh mechanical conditions in real use cases is crucial.
2. High comfort and biocompatible system-level integration of sensors to power sources, wireless transmission devices, circuits, and actuators: In order to fully realize the potential for soft mechanical sensors to enable high comfort and biocompatibility in wearable and implantable applications, seamless integration of other circuit components is vital to sensor function and application to the human body. Such circuit components include, but are not limited to, power sources, wireless transmission components, data-processing circuits, and actuators. Recently, important progress has been made in this regard through the use of intrinsically stretchable circuit components (Wang et al., 2023) or hybrid integration of stretchable and rigid sensor components in a highly ergonomic fashion (Niu et al., 2019).
3. Further investigation into the long-term biocompatibility of implantable sensors and immunomodulatory strategies to reduce foreign-body response: In order to ensure the safety of soft mechanical sensors in implantable applications, establishing the long-term biocompatibility of sensor materials is required so as to minimize risks of toxicity or inflammation associated with the implanted sensor. In addition to establishing the biocompatibility of sensor materials, further investigation into strategies to modulate the immune responses elicited by implanted sensors, such as surface modification or sustained immunomodulatory drug release, is another promising area of research that can increase the long-term safety associated with implanted sensor use (Li, Yuan, et al., 2020; Su, Kong, et al., 2022; Vishwakarma et al., 2016).
4. Hermetic encapsulation and long-term sensor stability: Given the highly biofluidic environments wearable and implanted sensors are subjected to, breakthroughs need to be made in the development of soft hermetic encapsulation that can grant sensors resistance to water penetration and the resulting electrical leakage and moisture-induced degradation of sensor components. In addition to high conformability and low vapor permeability, such encapsulation must also be biocompatible and have minimal interference on the encapsulated sensor's sensing capabilities and mechanical properties. Recent progress has been made in the employment of different soft materials, such as liquid metals (Shen et al., 2023) and lubricant-infused polymers (Lemaire et al., 2023), in the fabrication of mechanically compliant encapsulation layers, providing potential research directions for the development of highly robust encapsulation of soft mechanical sensors in wearable and implantable settings.
5. Large-scale manufacturing with low device-to-device variation: In order to realize commercial and clinical viability, further advancements in the use of scalable manufacturing techniques to fabricate soft mechanical sensors are necessary. While techniques such as inkjet printing and photolithography have already been used to fabricate sensors in a highly scalable way (as discussed in Section 3.7), further exploration of large-scale manufacturing techniques for a broader range of materials that can enable the different aforementioned human-compatible properties is required. In addition to advances in high-volume manufacturing processes, standardized device performance is a critical factor for commercial and clinical applications. As such, a more thorough understanding of the causes of

device-to-device variations in mechanical sensors manufactured in laboratory settings is required to achieve high device-to-device uniformity.

In all, soft mechanical sensors represent an emerging class of bioelectronic devices that have the potential to unlock high-accuracy, high-comfort continuous monitoring of mechanical biodata, allowing for promising biomedical applications in electronic skins, physiological and postoperative monitoring, motion tracking, and restoration of organ function, and more.

AUTHOR CONTRIBUTIONS

Rithvik Papani: Conceptualization (equal); writing – original draft (lead); writing – review and editing (equal). **Yang Li:** Conceptualization (supporting); writing – original draft (supporting); writing – review and editing (supporting). **Sihong Wang:** Conceptualization (equal); funding acquisition (lead); project administration (lead); writing – review and editing (equal).

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CONFLICT OF INTEREST STATEMENT

The authors declare no conflicts of interest.

DATA AVAILABILITY STATEMENT

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

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