



Review

Advanced ambulatory and continuous sensing technologies for cardiovascular connected healthcare

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ABSTRACT

Cardiovascular diseases remain the leading cause of mortality worldwide, highlighting the urgent need for accurate, continuous, and personalized monitoring strategies. In this review, we highlight hemodynamic decoding as a central framework to advance multimodal connected cardiovascular healthcare. We provide a comprehensive overview of seven major sensing technologies, including piezoelectric, triboelectric, magnetoelastic, optical, capacitive, piezoresistive and bioimpedance sensors, and emphasize that their monitoring capabilities and clinical applications are fundamentally based on hemodynamic signals such as arterial pulse, blood pressure, vascular stiffness, and multiparameter interactions. The underlying physical principles and representative case studies of each modality are summarized, together with a comparative analysis of their performance metrics and clinical relevance. Material selection strategies are systematically discussed, focusing on the balance between sensitivity, flexibility, biocompatibility, and long-term stability. Critical challenges are also addressed, including signal accuracy under dynamic physiological conditions, long-term durability, multimodal data integration, and compliance with privacy and regulatory standards. Finally, we outline future directions involving artificial intelligence, the Internet of Things, and implantable bioelectronic systems, which hold great promise for enabling next-generation cardiovascular monitoring technologies that are intelligent, scalable, and clinically transformative.

1. Introduction

The prevalence of cardiovascular diseases (CVDs) remains one of the most significant public health challenges globally. Despite

advancements in society and technology that have led to notable progress in certain treatments, the incidence and mortality rates of CVDs remain alarmingly high worldwide. Cardiovascular diseases remain the leading cause of death worldwide, accounting for approximately 19.2

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million deaths globally in 2023 [1]. CVDs include coronary artery disease [2], hypertension [3], heart failure [4] and rheumatic heart disease [5]. These diseases are characterized by structural and functional disorders of blood vessels, leading to insufficient blood supply to multiple organs [6]. In developing countries, limited resources and overburdened public health systems exacerbate the impact of cardiovascular diseases. Risk factors such as unhealthy diets [7], lack of physical activity [8], and poor lifestyle habits [9] are prevalent worldwide, further intensifying this issue.

Recent technological progress has highlighted the critical role of cardiovascular health monitoring in disease prevention, early diagnosis, and effective treatment. Timely and effective monitoring allows for early diagnosis, prompt treatment, and a reduction in the risk of severe complications, significantly lowering mortality rates [10,11]. Continuous health monitoring enables healthcare providers to assess treatment efficacy and adjust therapeutic strategies accordingly. Modern monitoring technologies, such as wearable devices and remote monitoring systems, have made daily cardiovascular disease monitoring possible in home environments [12]. These technologies not only continuously record key cardiac health indicators [13] like heart rate (HR), blood pressure (BP), and electrocardiograms (ECG) but also provide real-time feedback through advanced data analysis, helping patients and healthcare providers better understand disease progression.

Traditional monitoring methods such as routine ECG and BP measurements remain indispensable in clinical practice, yet their episodic and hospital-centered nature restricts long-term disease management [14,15]. In contrast, emerging sensor-based platforms are enabling continuous, non-invasive, and user-friendly cardiovascular monitoring

across daily life settings [16–18]. Technologies including piezoelectric sensors [19], triboelectric sensors [20], magnetoelastic sensors [21], optical technology [22], capacitive sensors [23] and piezoresistive sensors [24] have shown strong potential to capture physiological signals with high fidelity and wirelessly transmit data for real-time assessment (Fig. 1). When integrated with AI-based analytics, these systems further enhance predictive capability by analyzing heart rate variability, respiratory dynamics, and multimodal physiological features, facilitating the early detection of arrhythmias and hypertension [25–27]. Such innovations are gradually shifting cardiovascular care toward continuous, home-based monitoring and personalized prevention, opening new pathways for proactive disease management.

To advance this rapidly evolving field, the present review introduces hemodynamic decoding as a central framework for multimodal connected cardiovascular healthcare, positioning clinically relevant parameters such as blood pressure, arterial stiffness, and vascular impedance at the core of monitoring. While preprocessing, waveform analysis, feature extraction, and learning-based inference are widely used in conventional cardiovascular signal processing, the distinction of this framework lies in its target-driven and physiology-constrained organization, which aims to reconstruct intermediate hemodynamic observables such as pulse wave dynamics, pressure-related signatures, and flow-associated characteristics, rather than directly mapping signals to features or diagnostic outcomes. By shifting the focus toward physiologically interpretable variables, this approach enables improved cross-modality consistency and clinically meaningful inference, and represents an emerging conceptual direction integrating multimodal sensing, physics-informed reasoning, and data-driven inference.

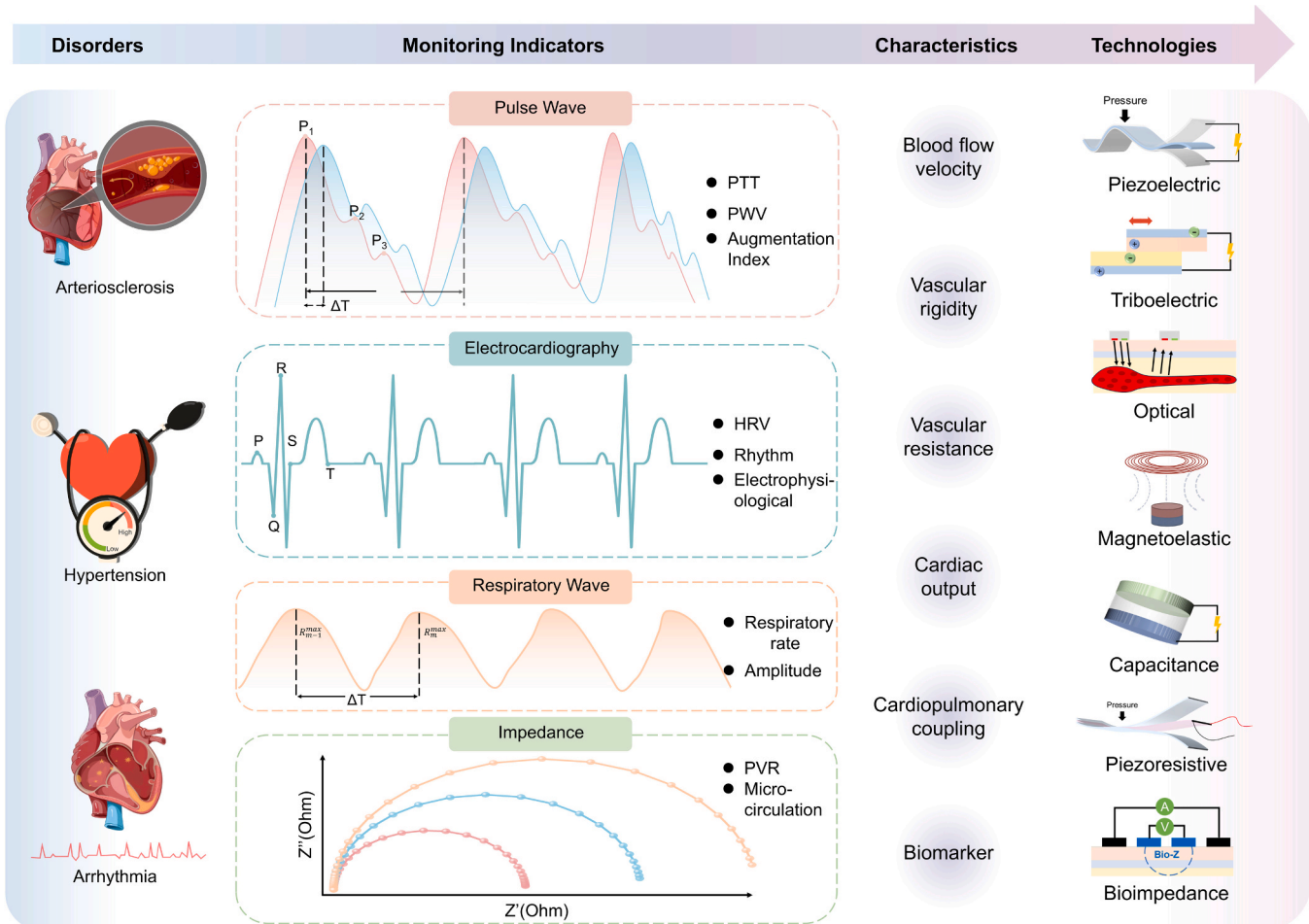


Fig. 1. Overview of key technologies and indicators driving cardiovascular connected healthcare through continuous monitoring.

2. Fundamentals of cardiovascular monitoring and sensing mechanisms

Cardiovascular health monitoring employs various technologies to track and assess the heart's function and vascular health, each with unique principles, design considerations, and engineering applications. This section explores the fundamental knowledge of different monitoring technologies, delving into the working principles of various sensors, comparing the performance of different materials and technologies.

2.1. From traditional to integrated ambulatory systems

The evolution of cardiovascular health monitoring technology is a significant area of innovation and practice in medical technology. From the basic tools of the past to today's highly sophisticated and intelligent systems, the advancement of these technologies reflects our deepening understanding of the prevention, diagnosis, and management of cardiovascular diseases.

Early attempts at cardiovascular monitoring began in the late 19th century when doctors could only assess heart function through direct auscultation. In 1902, one of the most notable advancements came from Dutch physiologist Willem Einthoven, who developed the string galvanometer [28]. This device could record the heart's electrical activity [29]. Einthoven's experiment demonstrated that the string galvanometer was easier to use and more sensitive than the capillary electrometer, providing a revolutionary tool for diagnosing cardiovascular diseases. The advent of the ECG not only improved the accuracy of heart disease diagnosis but also laid the groundwork for subsequent technological developments.

By 1961, Norman Jeff Holter had designed the Holter monitor for continuous ECG monitoring in hospitals [30] (Fig. 2a). Holter created a portable backpack recorder weighing 75 pounds, capable of continuously recording ECG for up to 24 h. This technology allowed doctors to monitor ECGs during the patient's daily activities, identifying

intermittent arrhythmias that were difficult to capture otherwise [31].

In the 1970s, Stanley's computerized monitoring of the long-term effects of cardiac assist devices was found to trigger problems such as myocardial decline and vasodilation, highlighting the critical role of computerized monitoring [32] (Fig. 2b). Real-time online algorithms for ECG waveform analysis began to be developed [43]. These algorithms provided information whenever the values exceeded predefined thresholds for a single-lead ECG signal. In 1974, Scheidt et al. utilized standard telephone lines and small devices capable of multi-lead ECG transmission and automatic switching to transmit 203 ECGs from 27 patients [44]. Additionally, the latter half of the 20th century saw the emergence of non-invasive BP monitoring technologies, such as oscillometric BP measurement. These advancements further simplified the diagnosis and management of hypertension, enabling patients to monitor their BP at home and improving hypertension management.

The 21st century has ushered in a new era of cardiovascular health monitoring driven by technological innovation. Since flexible, stretchable silicon was named one of the top ten technologies by MIT Technology Review in 2006 [33] (Fig. 2c), research into flexible sensors made from flexible materials has intensified. Most research on non-invasive monitoring of cardiovascular diseases has focused on a series of signals generated during the heart's activity [45], such as pulse waves [46], heart sounds [47], ECG [48] and seismocardiogram (SCG) [49] signals. In 2012, Zhenan Bao's team introduced an environmentally repeatable self-healing electronic composite material capable of sensing mechanical forces [34] (Fig. 2d). The ability to mimic the self-healing properties of natural skin in conductive and piezoresistive materials significantly expands the potential applications of electronic skin. In 2014, the Rogers team demonstrated how to create a soft, wearable system that adheres gently and non-invasively to the skin for physiological signal monitoring [35] (Fig. 2e). With the commercialization of smartwatches, a prospective study by Perez and others starting in 2017 highlighted the accuracy of the Apple Watch in reporting atrial fibrillation [50]. In 2018, Wang et al. described the design and operation of a skin-compliant ultrasound device capable of capturing BP waveforms at

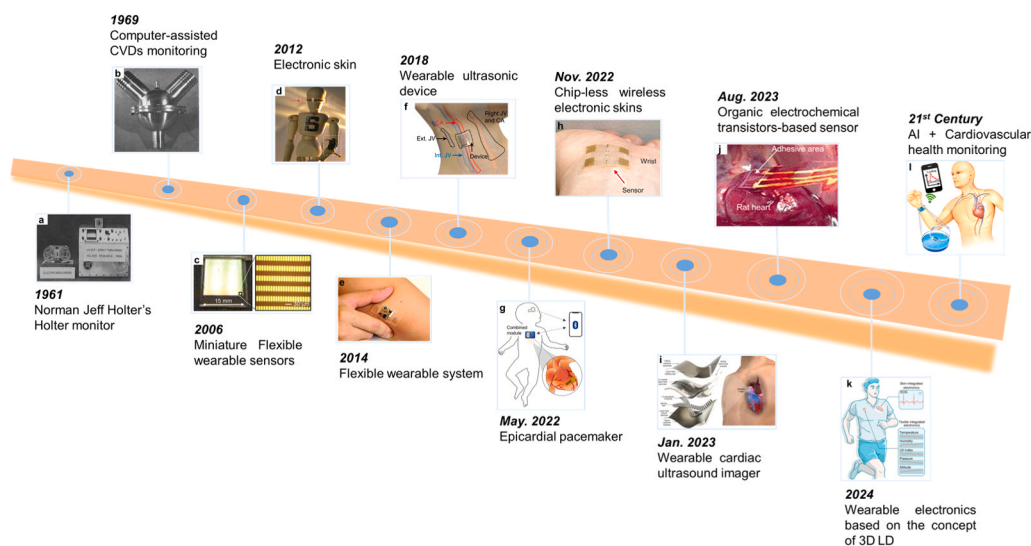


Fig. 2. Evolution of cardiovascular health monitoring technologies. (a) Holter monitor. Reproduced with permission. [30] Copyright 1961, The American Association for the Advancement of Science. (b) Artificial ventricular devices. Reproduced with permission. [32] Copyright 1969, Elsevier. (c) Stretchable silicon. Reproduced with permission. [33] Copyright 2006, The American Association for the Advancement of Science. (d) Epidermal electronics. Reproduced with permission. [34] Copyright 2012, Springer Nature. (e) Flexible wearable sensor. [35] Copyright 2014, The American Association for the Advancement of Science. (f) Wearable ultrasound devices. Reproduced with permission. [36] Copyright 2018, Springer Nature. (g) Biodegradable closed-loop wireless MEMS. Reproduced with permission. [37] Copyright 2022, The American Association for the Advancement of Science. (h) Chip-less wireless electronic skins. Reproduced with permission. [38] Copyright 2022, The American Association for the Advancement of Science. (i) Wearable cardiac ultrasound imager. Reproduced with permission. [39] Copyright 2023, Springer Nature. (j) Fully bioadhesive and stretchable OECT sensor. Reproduced with permission. [40] Copyright 2023, The American Association for the Advancement of Science. (k) Wearable electronics based on the concept of three-dimensional liquid diode. Reproduced with permission. [41] Copyright 2024, Springer Nature. (l) The concept of mobile ECG monitoring. Reproduced with permission. [42] Copyright 2019, American Chemical Society.

deeply buried arterial and venous sites [36] (Fig. 2f). In 2022, Rogers' team developed a biodegradable, closed-loop wireless micro-electromechanical system for HR monitoring and control [37] (Fig. 2g). Shortly afterward, Kim et al. demonstrated that single-crystal gallium nitride (GaN) films grown epitaxially on flexible substrates can be utilized in chip-free, flexible electronic skin [38], effectively addressing the traditional dependence of wireless electronic skin on rigid integrated circuit (IC) chips (Fig. 2h). Due to the large size of traditional non-invasive cardiac imaging methods, continuous measurement has been challenging [51,52]. In 2023, Hu et al. introduced a wearable ultrasound device combined with a deep learning (DL) model that can automatically extract characteristic information of cardiovascular diseases from continuous image recordings [39]. This device enables dynamic, wearable monitoring of cardiac performance across various environments, significantly enhancing accuracy (Fig. 2i). Li et al. reported a bioadhesive polymer semiconductor, and the further fabrication of transistor sensors enables it to produce high-quality and stable electrophysiological recordings on ex vivo rat hearts and muscles [40] (Fig. 2j). In 2024, Zhang et al. proposed a wearable electronic device based on a three-dimensional liquid diode configuration that provides stable signal output even in the presence of sweating by constructing spatially heterogeneous wettability [41] (Fig. 2k).

With the progress of internet and mobile communication technologies, remote cardiovascular health monitoring has become feasible. Modern remote monitoring systems can collect and transmit patient data in real-time to healthcare providers, allowing for complex cardiovascular function assessments in a home environment. For example, using

mobile ECG devices and cloud technology, patients' heart health can be remotely monitored and analyzed through intelligent algorithms [42] (Fig. 2l), facilitating early intervention and customized disease management plans.

These historical milestones and recent advancements illustrate the continuous innovation in cardiovascular health monitoring technologies. By integrating advanced materials and digital technologies, these systems offer promising solutions for enhancing patient care through precise and real-time health monitoring.

2.2. Fundamental principles of advanced sensors

Cardiovascular health monitoring technologies primarily rely on various sensors to capture the physiological signals of the heart and blood vessels. Each sensor technology operates on specific principles suited for detecting specific physiological changes (Fig. 3).

A range of physiological indicators has been employed to evaluate cardiovascular health by reflecting cardiac rhythm, vascular compliance, and hemodynamic function. HR and heart rate variability (HRV) are primary parameters typically extracted from ECG, photoplethysmography (PPG), or chest-wall micro-vibrations. HR quantifies the frequency of cardiac cycles, while HRV measures beat-to-beat fluctuations, offering insight into autonomic regulation. Clinically, reduced HRV is strongly associated with increased risk of arrhythmias, heart failure, and adverse cardiovascular outcomes [60–62]. BP is increasingly estimated without cuffs via time-domain features such as pulse transit time (PTT) and pulse arrival time (PAT), which measure the

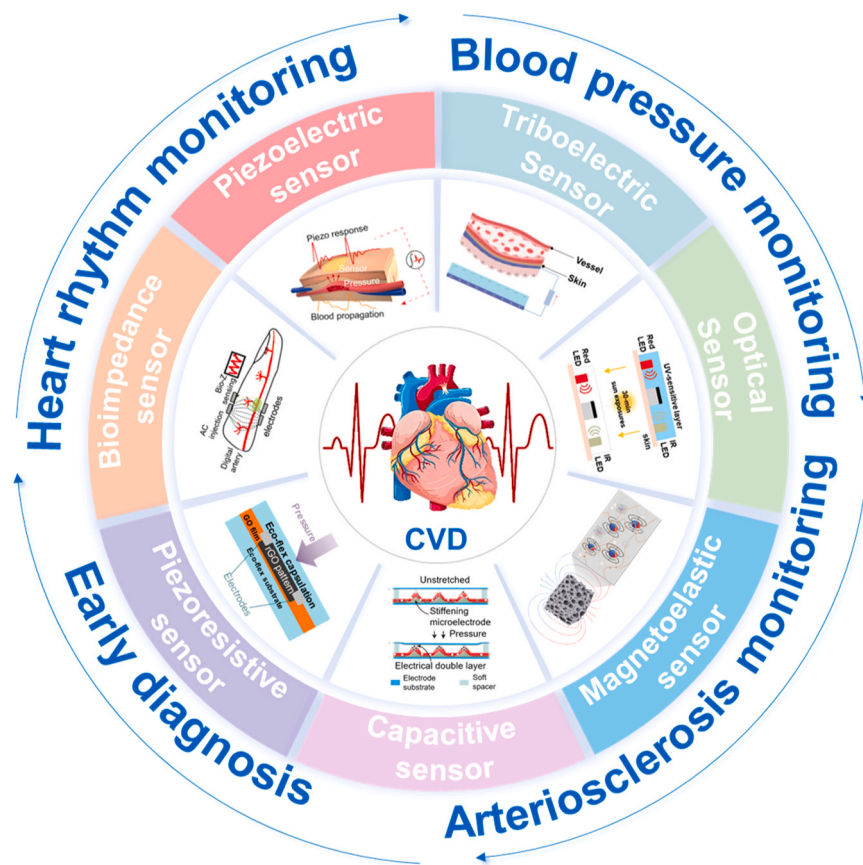


Fig. 3. The application of seven sensing monitoring mechanisms in CVDs. Clockwise from the top, the sequence is “Piezoelectric sensor.” Reproduced with permission. [53] Copyright 2023, Springer Nature. “Optical sensor.” Reproduced with permission. [54] Copyright 2016, The American Association for the Advancement of Science. “Magnetoelastic sensor.” Reproduced with permission [55]. Copyright 2021, Springer Nature. “Capacitive sensor.” Reproduced with permission. [56] Copyright 2021, The American Association for the Advancement of Science. “Piezoresistive sensor.” Reproduced with permission. [57] Copyright 2020, American Chemical Society. “Bioimpedance sensor.” Reproduced with permission. [58] Copyright 2023, Springer Nature. “Triboelectric sensor.” Reproduced with permission. [59] Copyright 2021, Wiley-VCH.

delay between proximal and distal signals. These parameters are closely linked to pulse wave velocity (PWV), a recognized marker of arterial stiffness. Clinically, shortened PTT and elevated PWV are early indicators of hypertension and vascular aging [63,64]. The augmentation index (AIx), derived from pulse waveform morphology, further quantifies wave reflections and arterial compliance [65,66]. Oxygen saturation (SpO₂) is obtained via PPG at sites like the fingertip or earlobe and represents the proportion of oxygen-bound hemoglobin. SpO₂ monitoring is critical in assessing respiratory and circulatory performance and in detecting hypoxemia across diverse clinical settings [67]. Bioimpedance analysis has emerged as a powerful modality for cardiovascular monitoring. It detects changes in electrical impedance due to blood volume shifts, vascular tone, and tissue composition, enabling indirect assessment of cardiac output and fluid status [68]. Bioimpedance-based approaches have been extended to detect circulating biomarkers, including C-reactive protein (CRP), D-dimer and endothelial microparticles, thereby supporting early diagnosis of inflammation and thrombosis.

2.2.1. Piezoelectricity

Piezoelectric sensors utilize the piezoelectric effect, where

mechanical stress applied to piezoelectric materials generates an electric charge. Materials such as quartz crystals or certain ceramics produce charges when subjected to pressure, vibration, or strain, enabling piezoelectric sensors to detect minute pressure changes with high sensitivity [69] (Fig. 4a).

In piezoelectric materials, the coupling between mechanical and electrical domains is governed by a set of constitutive equations that describe how mechanical stress and electric field simultaneously influence strain and electric displacement. Under the stress-electric field Eqs. (1) and (2), this relationship can be expressed as [70,71]:

$$S = s^E \cdot T + d \cdot E \quad (1)$$

$$D = d \cdot T + \epsilon^T \cdot E \quad (2)$$

where S is the strain tensor and T is the applied stress tensor. The matrix s^E represents the mechanical compliance at constant electric field, while d represents the piezoelectric coefficient associated with the converse and direct piezoelectric effects. E denotes the electric field vector, D is the resulting electric displacement vector, and ϵ^T stands for the permittivity measured under constant mechanical stress.

These sensors play a crucial role in arterial pressure monitoring by

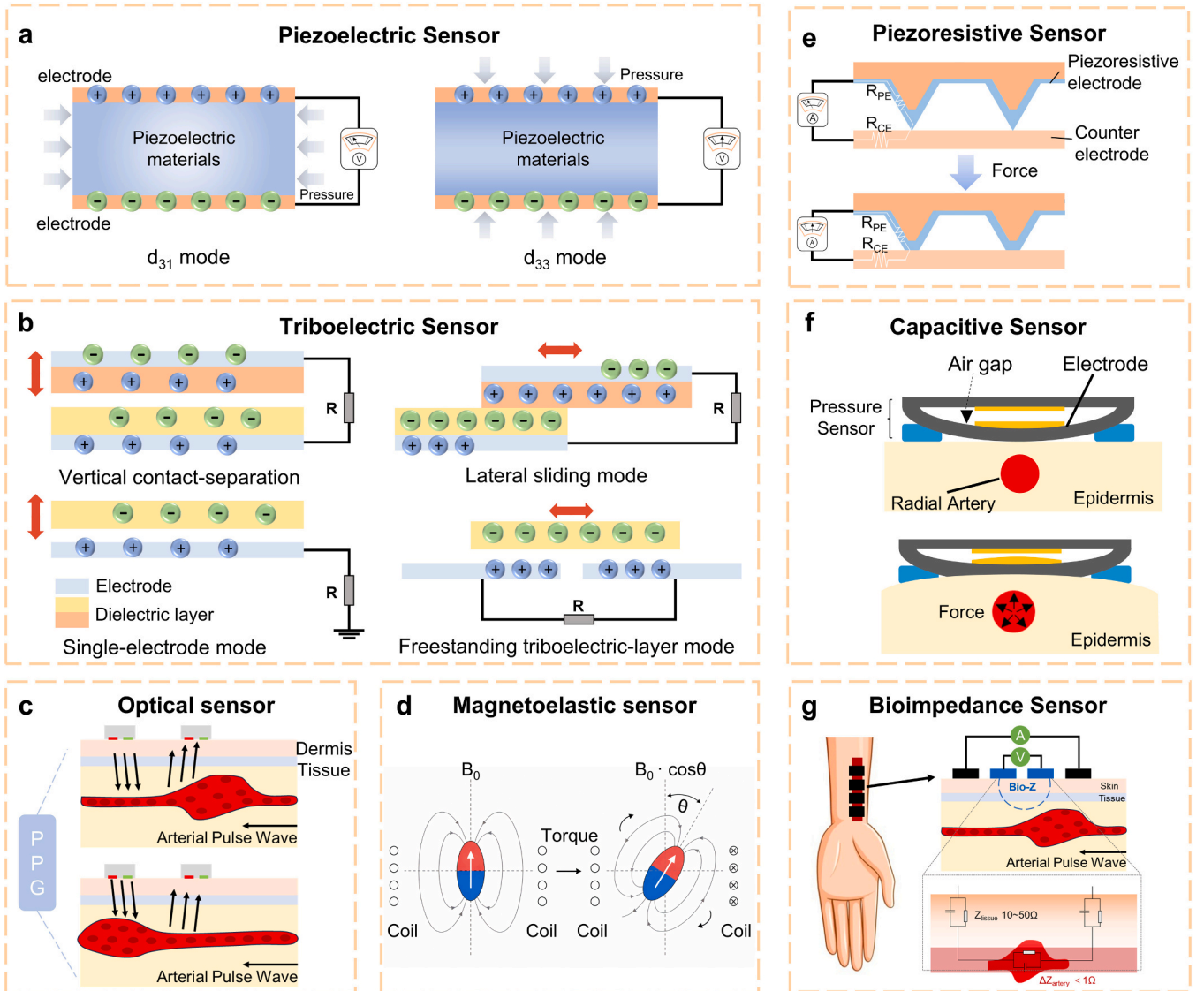


Fig. 4. Diagram of seven sensing monitoring mechanisms. (a) Piezoelectric effect. (b) Triboelectric effect. (c) Optical technology. (d) Magnetoelastic sensor. (e) Piezoresistive effect. (f) Capacitive sensing. (g) Bioimpedance technology.

assessing pressure changes in pulse waves to evaluate BP [19]. Additionally, in heart sound detection, piezoelectric sensors convert mechanical vibrations of heart sounds into electrical signals, providing phonocardiograms to aid in cardiac health assessment [72]. A distinctive strength of piezoelectric sensors in cardiovascular health monitoring lies in their inherent self-powered capability, which is achieved through the direct transduction of biomechanical deformation, such as arterial pulsation or myocardial motion, into electrical signals without the need for external power sources. In clinical or home-based cardiovascular monitoring, piezoelectric sensors often encounter challenges related to long-term stability and environmental exposure. Although these sensors demonstrate excellent sensitivity and fast response in controlled environments, their performance may degrade when exposed to moisture, temperature variations, or repeated mechanical deformation. For example, sweat and skin heat during prolonged wear can lead to signal drift, while mechanical fatigue may alter material polarization, reducing output amplitude [73–75]. These limitations constrain their sustained use in wearable or implantable formats, particularly in mobile and high-humidity conditions where passive encapsulation is insufficient to ensure performance consistency.

2.2.2. Triboelectricity

Triboelectric nanogenerators (TENGs) have emerged as a groundbreaking power generation technology. Under the leadership of Academician Zhonglin Wang [76], extensive research and experimental testing over the years have revealed that the theoretical basis for nanogenerators originates from the displacement current term in Maxwell's equations, established in the 19th century. In a triboelectric sensor, when two different materials come into contact and then separate, surface charges are generated and remain on the surfaces. This creates a voltage difference between the electrodes. To balance this voltage, free charges move through an external circuit, generating an electric current. The distance between the two layers changes due to mechanical movement, which causes both the capacitance C and the voltage V to vary over time. The resulting current I can be described by Eq. (3) [77]:

$$I = C \frac{\partial V}{\partial t} + V \frac{\partial C}{\partial t} \quad (3)$$

Experimental validation has shown that the instantaneous energy conversion efficiency of TENGs can reach up to 70%. The mechanism of TENGs is based on the triboelectric effect, which occurs when two materials with different triboelectric polarities come into contact, resulting in charge transfer and a potential difference [78]. When a load is connected between the electrodes of these materials or in a short-circuit state, charges are driven by the potential difference to flow back and forth between the electrodes, generating an electric current. Depending on the charge transfer mechanism, TENGs mainly operate in four modes: vertical contact-separation, lateral sliding, single-electrode, and independent layer modes (Fig. 4b). In contrast to the intrinsic dipole reorientation in piezoelectric materials, triboelectric sensing leverages interfacial charge interactions to achieve energy-autonomous signal acquisition from larger-scale or irregular mechanical inputs [79,80]. This self-powered capability is particularly well-suited for long-term monitoring needs without interfering with patients' daily activities [81–85]. However, in real-world cardiovascular applications, especially in wearable or epidermal formats, their long-term reliability is often compromised. Especially during motion, variations in contact pressure, displacement, and skin moisture often lead to signal instability [86,87]. Additionally, triboelectric outputs are inherently nonlinear and susceptible to environmental noise [88]. Reliable data extraction therefore requires real-time signal processing and advanced algorithms for artifact suppression and baseline correction. These technical challenges currently limit the widespread use of triboelectric sensors in long-term, ambulatory cardiovascular applications.

2.2.3. Optical

Optical sensing technologies offer powerful, noninvasive solutions for cardiovascular monitoring across hemodynamic and molecular levels. Optical sensors for pulse monitoring typically integrate a light source and a photodetector. Red and near-infrared LEDs are commonly used to ensure sufficient tissue penetration and sensitivity to blood absorption. Reflected or transmitted light is captured by photodiodes, and changes in intensity are translated into electrical signals [89,90]. In biological tissues, light attenuation is wavelength-dependent and can be described using the modified Beer-Lambert law [91]:

$$I(\lambda) = I_0(\lambda)e^{-\mu d} \quad (4)$$

In Eq. (4), $I(\lambda)$ is transmitted intensity, $I_0(\lambda)$ is laser incident intensity, d is slab thickness, and μ is the wavelength-dependent attenuation coefficient. This equation describes the exponential attenuation of light in tissue.

PPG, widely used in wearable devices, captures HR, HRV, and SpO₂ by detecting pulsatile changes in light absorption (Fig. 4c)[92–94]. Advances in flexible substrates, wireless integration, and AI-assisted processing have improved accuracy and comfort in real-world use. Optical fiber sensors, including fiber Bragg gratings and interferometers, further enable high-resolution monitoring of pulse waveforms, arterial stiffness, and central BP in both contact and contactless formats [95]. At the molecular scale, SERS leverages nanostructured substrates to detect biomarkers such as cTnI, CRP, and IL-6 with ultra-high sensitivity [96]. Recent developments in microfluidics, label-free detection, and multiplexing have brought SERS closer to clinical deployment. Together, these optical strategies support real-time, multimodal cardiovascular assessment from vital signs to disease-specific markers. However, their accuracy and robustness are strongly influenced by real-world conditions. PPG, for instance, can experience substantial error during physical activity due to motion artifacts and light scattering caused by skin deformation [97]. Studies have shown that HR estimation errors may reach 30 % during daily activity, with darker skin tones and ambient light fluctuations further impairing signal quality [98]. Optical fiber and SERS-based platforms demonstrate high analytical precision in laboratory settings, but their integration into wearable devices remains technically complex. Issues such as bending losses, optical fiber fragility, and the need for stable surface contact limit their reliability under ambulatory or long-term monitoring conditions [99].

2.2.4. Magnetoelastic

Magnetoelastic sensors rely on the coupling between mechanical deformation and magnetic properties [100], where applied stress or strain induces measurable changes in magnetization, magnetic permeability, or resonance characteristics. This behavior originates from magnetostriction, a fundamental property of ferromagnetic materials in which mechanical and magnetic domains are intrinsically coupled [101]. When subjected to external forces, the redistribution and rotation of magnetic domains lead to variations in magnetic response, enabling the transduction of biomechanical stimuli into remotely detectable signals without direct electrical contact (Fig. 4d).

For inductive readout of magnetoelastic transducers, the generated electrical signal can be described by Faraday's law [102,103]:

$$\varepsilon = -N \frac{d\Phi}{dt} \quad (5)$$

$$\Phi = BS \cos \theta \quad (6)$$

where ε is the induced electromotive force, N is the number of turns in the coil, Φ is the magnetic flux, B is the magnetic flux density, S is the effective coil area, and θ is the angle between the magnetic field and the normal direction of the coil. This relationship indicates that the electrical output can be enhanced by increasing the number of turns, the effective coil area, or the rate of change of magnetic flux. In resonant

magnetoelastic systems, these variations can be detected as shifts in resonance frequency or amplitude, providing a basis for wireless and passive sensing. The underlying magneto-mechanical coupling has been extensively studied in magnetostrictive alloys and soft magnetic composites, forming the foundation for remote sensing applications. Recent advances have extended magnetoelastic sensing to cardiovascular monitoring, where its contactless and battery-free nature enables the detection of pulse waves, pressure variations, and biomechanical motion via external magnetic interrogation, with particular advantages for implantable and deep-tissue applications [104–107]. However, clinical translation remains limited by susceptibility to magnetic interference, the need for external excitation and readout, and challenges in material optimization, miniaturization, and biocompatible integration. Ensuring stable performance under long-term mechanical and physiological conditions also requires improved control over material properties and encapsulation strategies. Continued progress in magnetostrictive materials, low-noise readout, and system integration is expected to further enhance the feasibility of magnetoelastic sensing for cardiovascular monitoring.

2.2.5. Capacitance

Capacitive sensors function by detecting variations in capacitance induced by changes in the distance or dielectric properties between two conductive plates [108] (Fig. 4f). These sensors are particularly sensitive to subtle mechanical deformations, such as chest expansion during respiration or slight alterations in cardiac geometry, making them highly suitable for continuous, non-invasive physiological monitoring. In respiratory rate tracking, capacitive sensors capture micro-scale thoracic displacements, while in cardiac monitoring, they can detect variations in the surrounding electric field caused by myocardial contraction and relaxation [109–111]. The capacitance C of such a parallel-plate system can be calculated using Eq. (7):

$$C = \frac{\epsilon_0 \epsilon_r A}{d} \quad (7)$$

Here, ϵ_0 is the vacuum permittivity, ϵ_r is the relative permittivity of the dielectric, A is the overlapping area of the electrodes, and d is the separation distance between them. When blood flows beneath the skin, it causes small mechanical displacements that alter the electrode spacing. These changes modify the capacitance value, allowing the sensor to convert mechanical signals into electrical outputs [112].

One of the most notable advantages of capacitive sensing lies in its compatibility with LC resonance-based wireless circuits, enabling battery-free and passive signal acquisition over distance. This feature supports fully integrated, maintenance-free cardiovascular monitoring systems, particularly for implantable or difficult-to-access anatomical sites. Additionally, capacitive sensors inherently exhibit low power consumption, making them ideal for continuous, long-term monitoring in wearable and remote health platforms. With non-contact detection, wireless operation and ultra-low power requirements, capacitive sensors are well positioned for next-generation cardiovascular monitoring in clinical and home care settings. In prolonged cardiovascular monitoring, particularly for implantable or skin-mounted capacitive systems, sensor performance can deteriorate due to mechanical fatigue and dielectric instability [113]. Fluctuations in ambient temperature, changes in skin hydration, and repeated mechanical loading may alter the dielectric properties or electrode separation, resulting in baseline drift and signal variability [114]. Without appropriate calibration or compensation mechanisms, these variations can reduce the accuracy of parameters such as respiratory rate, pulse wave timing, or arterial compliance.

2.2.6. Piezoresistance

Piezoresistive sensors operate by converting applied mechanical pressure into changes in electrical resistance. When pressure is applied to the sensing material, its internal conductive network or atomic structure is altered, resulting in a measurable resistance variation [115]

(Fig. 4e). The resistance can be described by Eq. (8) [116]:

$$R = \frac{\rho L}{A} \quad (8)$$

In this equation, R is the resistance, ρ is the resistivity of the sensing material, L is the effective conductive-path length, and A is the effective cross-sectional area. Under external pressure, changes in ρ , L , and A , together with variations in contact resistance, collectively contribute to the resistance response.

Piezoresistive sensors are primarily used in cardiovascular monitoring for BP and pulse wave measurements. In BP measurement, these sensors provide precise readings by detecting pressure changes in arteries [117–119]. For pulse wave analysis, sensors assess arterial stiffness by measuring PWV, a key indicator of cardiovascular health [120, 121]. The wide range of available materials, from traditional silicon structures to flexible carbon-based composites, enables piezoresistive sensors to be effectively integrated into both wearable and implantable devices while maintaining good mechanical compatibility. These features support their practical use in cardiovascular monitoring systems for clinical and home healthcare settings. Despite these advantages, the performance of piezoresistive sensors remains constrained by mechanical fatigue, temperature sensitivity, and signal drift in real-world applications [122,123]. In dynamic clinical settings such as ambulatory monitoring or postoperative recovery, repeated deformation of the sensing matrix can alter its microstructure, leading to resistance instability. Temperature fluctuations caused by skin contact, perspiration, or ambient changes further impact baseline consistency, reducing measurement accuracy over time [124]. These issues are particularly problematic in long-term monitoring scenarios, where stable and repeatable signals are essential for clinical decision-making. Although calibration algorithms and redundant sensor arrays can partially compensate for these effects, the reliable deployment of soft piezoresistive sensors in diverse patient populations remains technically challenging.

2.2.7. Bioimpedance

Bioimpedance sensors assess the electrical properties of biological tissues by introducing a low-amplitude alternating current and measuring the resulting voltage drop across the body. As electrical impedance varies with fluctuations in blood volume, fluid composition, and tissue structure, these sensors can infer a range of cardiovascular parameters, including blood flow, stroke volume, and related hemodynamic parameters [125,126] (Fig. 4g). The frequency-dependent impedance of biological tissues can be described using the Cole impedance model, as shown in Eq. (9) [127]:

$$Z(j\omega) = R_\infty + \frac{R_0 - R_\infty}{1 + (j\omega\tau)^\alpha} \quad (9)$$

Here, R_0 and R_∞ represent the impedance limits at zero and infinite frequency, respectively, τ is a time constant related to tissue relaxation behavior, and α reflects the dispersion of relaxation times within heterogeneous biological structures. This formulation allows for accurate modeling of bioimpedance responses across a broad frequency range, facilitating interpretation of dynamic physiological conditions.

This technology can assess blood volume and flow, particularly valuable for monitoring heart failure and other cardiovascular conditions [128–130]. Bioimpedance sensing captures integrated physiological signals from deep and superficial tissues, providing a more comprehensive view of cardiovascular dynamics than surface-based methods. Its capacity to extract multiple vital signs, such as HR, respiration, and fluid status, from a single sensor enhances diagnostic efficiency while minimizing system complexity. Bioimpedance technology supports deep-tissue and multi-parametric cardiovascular monitoring, yet its clinical implementation is constrained by variability in electrode-tissue interface conditions. In wearable or implantable settings, changes in skin impedance, electrode displacement, or tissue hydration can cause significant fluctuations in measured impedance values [131].

These effects are especially pronounced during motion, prolonged wear, or in individuals with compromised skin integrity. Furthermore, current clinical validation of bioimpedance-based systems is limited, with few large-scale trials demonstrating long-term accuracy under diverse physiological and environmental conditions [132]. As a result, despite its potential, bioimpedance sensing remains underutilized in routine cardiovascular diagnostics.

2.3. Material selection for advanced sensors

Selecting appropriate materials for cardiovascular health monitoring devices is essential, as it directly affects functionality, durability, safety, and patient comfort. The following discussion details the material choices for several key technologies in cardiovascular health monitoring (Fig. 5).

In cardiovascular health monitoring, the selection of piezoelectric materials must be tailored to specific application scenarios and device requirements, such as long-term implantation, skin-mounted wearability, or transient biodegradability. For durable applications requiring high and stable piezoelectric output, ceramic materials including lead zirconate titanate (PZT) [133] and barium titanate (BaTiO_3) [134] are commonly employed due to their strong electromechanical coupling and high piezoelectric coefficients, making them suitable for clinical ultrasound imaging and pressure sensing. However, their intrinsic rigidity and brittleness limit their integration into flexible or skin-conformal systems. For wearable or stretchable devices, piezoelectric polymers such as polyvinylidene fluoride (PVDF) [135] and its copolymers are often preferred, as they offer good mechanical flexibility, chemical stability, and processability, enabling integration into thin, soft platforms for long-term physiological monitoring. In implantable or biodegradable systems, materials such as, glycine-based crystals [136], and

peptide-derived films offer moderate piezoelectric performance while enabling safe in vivo degradation, thus eliminating the need for surgical removal. In these cases, material selection involves balancing piezoelectric activity, degradation behavior, mechanical compliance, and biocompatibility to meet application-specific demands.

In triboelectric sensors for cardiovascular health monitoring, material selection is primarily guided by the need to maximize charge generation through contact electrification, while also ensuring mechanical flexibility, biocompatibility, and integration compatibility. For wearable applications involving dynamic mechanical inputs such as pulse detection or motion-triggered monitoring, flexible polymeric materials with high surface charge affinity are commonly used. PVDF [137], polydimethylsiloxane (PDMS) [138], and thermoplastic polyurethane (TPU) [139] are frequently selected due to their excellent flexibility, triboelectric polarity, and ease of microstructuring, which enhances effective surface area and charge transfer. In epidermal or conformal sensor platforms, these polymers can be further engineered with surface textures or nanostructures to improve sensitivity under low-pressure or small-displacement conditions. For performance enhancement, nanomaterials such as ZnO nanowires, TiO_2 nanoparticles, and carbon-based additives are often incorporated to increase dielectric constant and interfacial charge mobility [81,140]. In implantable or bioresorbable systems, the use of degradable triboelectric layers such as gelatin, silk fibroin, or PLA is emerging, enabling temporary monitoring with reduced long-term bioburden [141,142]. The choice of triboelectric materials therefore depends not only on their charge affinity but also on their mechanical behavior, skin compatibility, fabrication feasibility, and suitability for the intended operating environment.

Optical cardiovascular sensors encompass multiple modalities, each with distinct material requirements based on their underlying transduction mechanisms. In PPG, which monitors pulsatile blood volume

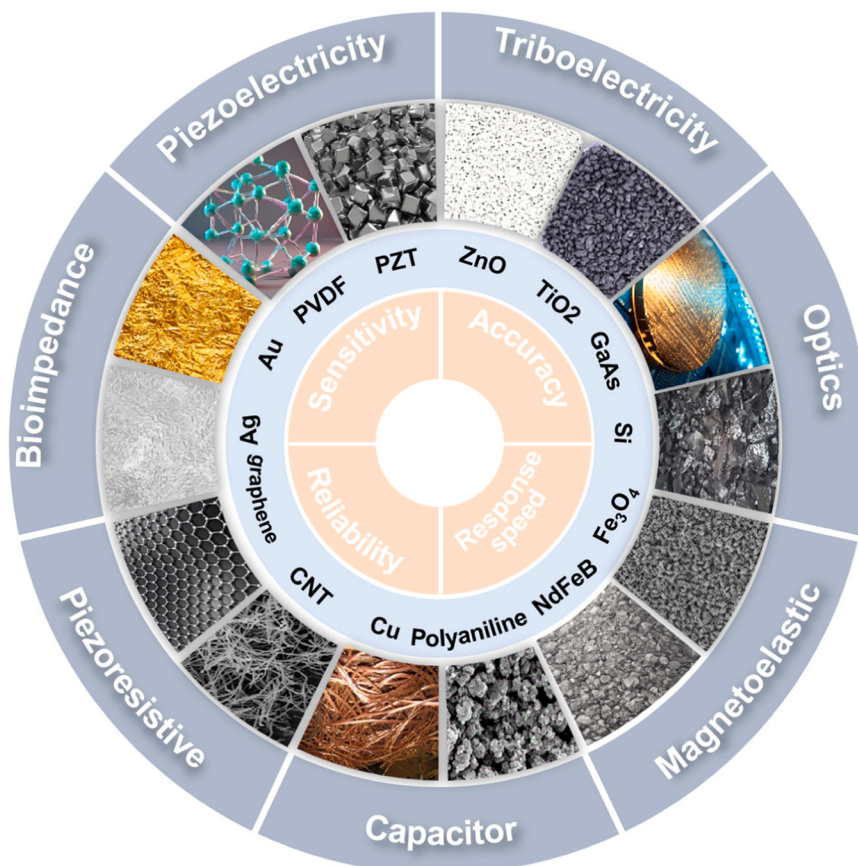


Fig. 5. Diagram of common materials used in seven major sensors and their biomedical application characteristics.

variations through light-tissue interactions, semiconductor materials such as gallium arsenide (GaAs) and silicon (Si) are frequently utilized due to their efficient optical emission, spectral compatibility with biological tissues, and stable photodetection performance [143,144]. GaAs enables near-infrared emission suitable for deep tissue penetration, while Si photodiodes offer reliable performance and compatibility with integrated electronics. For optical fiber sensing, which relies on changes in light transmission through mechanically responsive media, materials such as polymethyl methacrylate (PMMA), SiO₂ fibers, and stretchable hydrogels are selected for their low optical loss, mechanical flexibility, and stability under strain [145–147]. These systems are well suited for capturing vascular dynamics such as PWV and arterial stiffness. SERS sensors are designed for molecular-level detection, typically employ noble metals like Au or Ag in nanostructured configurations to amplify weak Raman signals through localized surface plasmon resonance [148, 149]. These plasmonic elements are often integrated with biocompatible and flexible substrates such as PDMS, enabling wearable or implantable operation. Across all optical modalities, materials are selected to optimize optical efficiency, biocompatibility, mechanical compliance, and integration with compact, stable cardiovascular monitoring platforms.

Magnetoelastic sensors rely on materials that couple mechanical deformation to changes in magnetic properties while remaining compatible with wireless readout. Soft magnetic fillers such as NdFeB particles or Fe-based alloys are typically dispersed in elastomers (e.g., PDMS or Ecoflex) to provide flexibility and conformability [102,150]. In wearable formats, conductive elements such as liquid metals or conductive yarns are sometimes introduced to support electrical readout. For implantable use, material choices also need to account for long-term stability and tissue compatibility, often requiring thin encapsulation layers such as parylene-C. In vibration-based systems, micro-magnets combined with compliant membranes enable environmental changes to be reflected in dynamic magnetic responses. Overall, material selection is a trade-off between magnetic response, mechanical compliance, and stability in physiological environments.

In capacitive sensors for cardiovascular monitoring, materials are selected to ensure stable capacitive coupling, mechanical compliance, and compatibility with either wired or wireless readout schemes. For electrodes, conductive metals such as Au, Ag, and Cu are widely used due to their high conductivity and long-term signal stability. In flexible and wearable systems, conductive polymers like Polyaniline (PANI) and PEDOT:PSS offer advantages in stretchability, processability, and integration with soft substrates [108,151–153]. Dielectric layers typically consist of elastomers such as PDMS or PU, which enhance interfacial deformability and improve sensitivity to pressure or displacement [154–156]. For wireless or battery-free LC-based sensing systems, material selection focuses on achieving stable electromagnetic coupling, mechanical adaptability, and biocompatibility. Conductive nanomaterials such as silver nanowires, carbon nanotubes, and graphene are often employed for their printability, flexibility, and compatibility with soft substrates. Biodegradable polymers including PLA and polycaprolactone (PCL) have also been explored for transient LC devices, enabling safe resorption post-monitoring. For the dielectric layer, soft materials like PDMS and ionic hydrogels offer both mechanical compliance and stable capacitive response. Ultimately, materials must be matched to the device's operational frequency, implantation duration, and structural requirements to ensure reliable wireless performance [157,158].

In piezoresistive sensing systems for cardiovascular monitoring, material selection is driven by the need for high sensitivity to mechanical deformation, long-term signal stability, and compatibility with both rigid and wearable platforms [159]. In clinical settings such as intracardiac pressure measurement or arterial catheterization, silicon remains the predominant material due to its well-characterized piezoresistive behavior, thermal stability, and compatibility with microelectromechanical systems (MEMS) fabrication. For wearable applications involving arterial pulse detection, blood pressure estimation,

or motion-coupled cardiovascular analysis, soft and stretchable composites are preferred. These often consist of conductive nanomaterials such as carbon nanotubes (CNTs), graphene, or MXenes embedded within elastomeric matrices like PDMS or TPU [160–163]. The sensitivity of such composites can be enhanced through structural engineering strategies that introduce microcracks, porous scaffolds, or surface corrugations to amplify resistance changes under strain [164, 165]. In long-term epidermal applications, where sensors must maintain conformal contact and mechanical resilience under repeated loading, material selection must balance gauge factor, hysteresis, fatigue resistance, and adaptability to scalable fabrication methods such as screen printing or inkjet deposition.

In bioimpedance-based cardiovascular systems, material selection is guided by the diverse sensing modalities spanning from molecular biomarker detection to deep-tissue hemodynamic monitoring. For label-free electrochemical impedance platforms targeting soluble markers such as CRP, D-dimer, and platelet-derived microparticles, electrode materials must support high surface area, electron transfer efficiency, and stable electrochemical behavior. Nanostructured Au, graphene oxide (GO), and CNT-based composites are frequently employed to enhance sensitivity and enable multiplexed detection within miniaturized sensor arrays [166]. In wearable platforms, especially those requiring long-term epidermal contact, soft, low-impedance electrodes such as PEDOT-coated porous PDMS, MXene-integrated elastomers, and carbon-infused hydrogels are preferred to improve conformal adhesion, reduce interfacial impedance, and minimize motion artifacts [167]. These materials support the acquisition of high-fidelity signals for parameters such as HR and BP, even under ambulatory conditions. For implantable applications, where in situ vascular assessment is needed, stable and biocompatible conductors such as Au and Pt are integrated with microstructured substrates or bioresorbable polymers to ensure reliable signal acquisition in complex biological environments [168]. With the increasing integration of AI-assisted modeling and wireless data transmission, materials must also meet additional requirements including thermal stability, signal reproducibility, and mechanical durability across repeated usage cycles. Overall, material choices in bioimpedance sensing reflect a balance among electrical performance, interface stability, biocompatibility, and suitability for either epidermal or implantable deployment.

The selection of materials for cardiovascular health monitoring sensors plays a crucial role in determining their performance, durability, and biocompatibility. Each sensing technology requires tailored materials to optimize efficiency and reliability, from piezoelectric and triboelectric energy conversion to optical, capacitive, and bioimpedance-based detection. The integration of nanotechnology, flexible electronics, and biodegradable materials continues to drive advancements in sensor performance, improving patient comfort and enabling long-term, real-time cardiovascular health monitoring [169, 170].

3. Transformative advances in sensing technologies for cardiovascular monitoring

Cardiovascular health monitoring technologies have found extensive applications in both clinical and home health management, offering numerous benefits ranging from early disease detection to long-term health management. This section explores the development of various cardiovascular health monitoring technologies, their applications in specific medical practices, and their potential to optimize treatment and improve quality of life.

3.1. Piezoelectric sensing for arterial pressure and pulse wave detection

Piezoelectric nanogenerators (PENGs) represent an emerging nanotechnology capable of converting ubiquitous mechanical stimuli into electrical energy, thereby addressing energy supply challenges, which

represents one of their key advantages [171–177]. In addition to high sensitivity to mechanical stimuli, piezoelectric sensing inherently enables self-powered signal generation by directly converting biomechanical deformation into electrical output. This capability reduces reliance on external power sources and supports continuous cardiovascular monitoring, particularly in wearable and implantable scenarios where long-term operation is required. Self-polarized natural piezoelectric materials have garnered significant attention due to their higher sensitivity, non-toxicity, biocompatibility, and flexibility [178–180]. Ghosh et al. [181] proposed a wearable biomimetic piezoelectric

pressure sensor capable of accurately detecting real-time human physiological signals without any external power source. This sensor exhibited excellent sensitivity ($27 \text{ mV} \cdot \text{N}^{-1}$) and a rapid response time (4.9 ms), making it valuable for diagnosing arteriosclerosis. In addition to monitoring related indicators, the sensor can be directly implanted to contact blood vessels for pressure monitoring, providing a visual display of arterial health (Fig. 6a). Dae et al. [182] showcased a self-powered piezoelectric pulse sensor capable of measuring radial/carotid pulse signals in superficial arteries in vivo (Fig. 6b). The ultra-thin piezoelectric sensor conforms to the complex textures of rugged skin, thereby

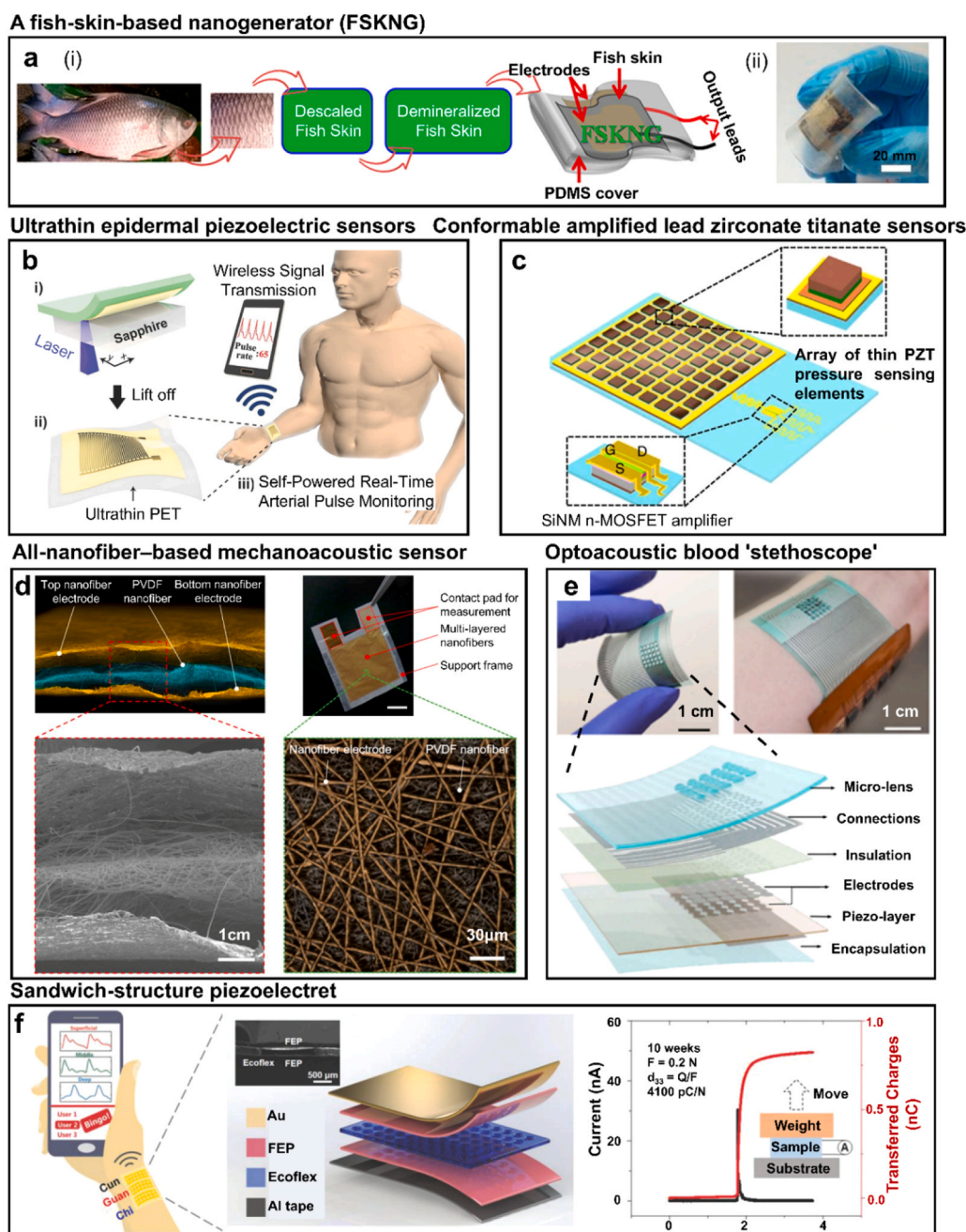


Fig. 6. Piezoelectric devices enable real-time hemodynamic decoding. (a) Illustration of the preparation process for fish skin nanogenerators from raw fish skin. Reproduced with permission [181]. Copyright 2017, American Chemical Society. (b) Fabrication and application of an ultrathin self-powered piezoelectric sensor for real-time arterial pulse monitoring, based on transferred PZT films on PET substrates. Reproduced with permission [182]. Copyright 2017, WILEY-VCH. (c) Array of square piezoelectric film transducers based on PZT (green: 400 nm) film. Reproduced with permission [183]. Copyright 2014, Springer Nature. (d) Cross-sectional, optical, and 3D microscopic images of the all-nanofiber mechanoacoustic sensing system. Reproduced with permission. [184] Copyright 2020, PNAS. (e) Schematic layout and an image showing the bending capability of the OBS. Reproduced with permission [185]. Copyright 2023, Springer Nature. (f) A piezoelectric electret sensor system designed for pulse wave detection along with its equivalent piezoelectric coefficient. Reproduced with permission. [186] Copyright 2018, Wiley-VCH.

responding to subtle pulse variations on the epidermal surface. Continuous monitoring of arterial pulses is crucial for early detection of cardiovascular diseases and assessing individual health status.

3.1.1. Early detection of cardiac rhythm anomalies via piezoelectric sensing

In cardiac rhythm monitoring, piezoelectric sensors can precisely capture the minute mechanical changes produced by each heartbeat, aiding in the evaluation of cardiac pumping efficiency and rhythm stability. Their high sensitivity and rapid response are particularly suited for detecting arrhythmias and other cardiac dysfunctions. Dagdeviren et al. [183] introduced an ultra-thin, compact device employing inorganic materials, a heterogeneous design, and theoretical modeling (Fig. 6c). A 400 nm high-quality PZT film served as the active element in a capacitor-type structure, which was connected to the gate of metal oxide semiconductor field effect transistors based on silicon nanomembranes. This design amplified the piezoelectric voltage response of PZT and converted it to a current output through capacitive coupling, thereby enabling the precise measurement of arterial pulse pressure waves. Despite the impressive piezoelectric coefficients (d_{33}) of ceramics or single crystals reaching 1500 or 4100 pC/N, their inherent brittleness and rigidity limit conformal contact with the skin and impair pressure measurement accuracy [187,188]. In contrast, polymers do not suffer from these limitations. Nayeem et al. [184] designed a fully nanofiber mechanical sensor with PVDF piezoelectric nanofiber sheets sandwiched between nanofiber electrode sheets or continuous film sheets, significantly increasing peak output voltage and maintaining high mechanical robustness and stability during bending, enabling continuous 10-hour SCG measurement (Fig. 6d). Jin et al. [185] utilized PVDF as an alternating piezoelectric sensor array to "listen" to acoustic waves representing blood characteristics, advancing over other opto-acoustic effect-based sensors for continuous arrhythmia monitoring (Fig. 6e). Chu et al. [186] utilized a flexible piezoelectric pulse wave sensor system based on a fluorinated ethylene propylene (FEP)/Ecoflex/FEP sandwich structure, achieving a d_{33} coefficient up to 4100 pC/N (Fig. 6f). Pulse sensors placed at the Cun, Guan, and Chi positions successfully diagnosed arrhythmias by detecting blood arterial waves induced by cardiac ejection, corresponding to blood volume, pressure, and flow patterns.

3.1.2. Continuous monitoring of hemodynamics and arterial compliance

While primarily used for monitoring cardiac rhythm and activity, piezoelectric sensors have shown potential in measuring BP changes related to heart pumping function. These sensors can be integrated into BP monitoring bands to estimate BP by capturing the chest wall's subtle movements during cardiac cycles [189,190]. This method allows for continuous BP monitoring without the need for traditional cuffs. Current research focuses on enhancing the sensitivity of piezoelectric sensors and reducing their susceptibility to external motion interference to improve measurement accuracy. Tian et al. [19] reported a wearable sensor based on heterogeneous layered piezoelectric composite ($d_{33}=41.67$ pC/N) (Fig. 7a,b), measuring local PWV and validating cardiovascular parameters through clinical Doppler ultrasound. Using a regression algorithm, a comprehensive index was proposed for evaluating cardiovascular function throughout the day. Li et al. [191] prepared a fabric-based ultrasound sensor for continuous, non-invasive BP monitoring. This was achieved by combining "S"-shaped stretchable island-bridge electrodes with a fabric substrate and embedding PZT-5H piezoelectric material through a hot pressing transfer process (Fig. 7c). The results demonstrated that the 4×5 array ultrasound sensor could accurately locate deep arterial vessels, allowing for continuous BP monitoring in both resting and post-exercise states (Fig. 7d). Due to low signal quality and the lack of a functional mapping relationship between sensor signals and BP values, the accuracy of continuous BP monitoring largely depends on the sensor type and the user's physiological parameters, presenting significant challenges [192,193]. Min et al. [194] reported a wearable BP sensor utilizing an ultra-thin piezoelectric sensor

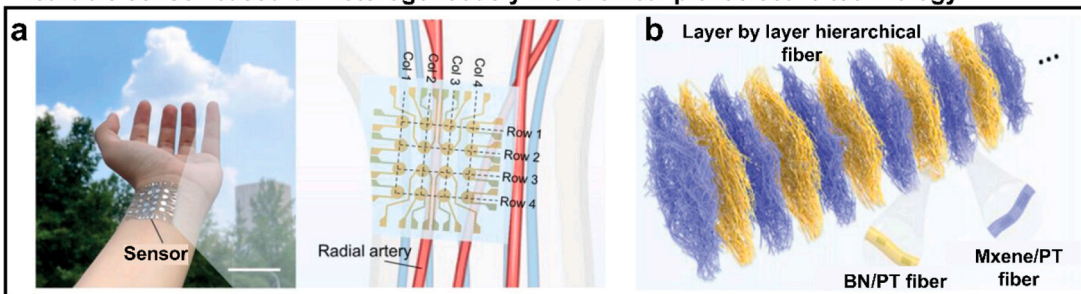
correlated with accurate BP transfer functions. By incorporating a PDMS passivation layer and a medical-grade adhesive layer, the wearable BP sensor achieved high sensitivity (0.062 kPa^{-1}), a rapid response time (23 ms), and excellent mechanical stability, with no significant degradation after 50,000 compression tests (Fig. 7e). Formal clinical validation was conducted on 35 participants, comparing the device's performance to an FDA-approved sphygmomanometer (OMRON HEM-7121). Although the WPBPS itself has not yet received FDA approval, the trial was approved by the Korean Ministry of Food and Drug Safety (MFDS) and an institutional IRB, marking an important step toward regulatory readiness. Cheng et al. [195] developed an implantable, self-powered, and visualized BP monitoring system based on piezoelectric thin films. During in vitro testing, the system demonstrated excellent linearity ($R^2>0.99$) and high sensitivity (173 mV mmHg^{-1}), maintaining outstanding stability over more than 50,000 operational cycles (Fig. 7f). In vivo experiments conducted on adult Yorkshire pigs showed linearity ($R^2 = 0.971$) and sensitivity of 14.32 mV/mmHg , with the device's maximum instantaneous power output reaching 40 nW. Notably, this system can provide real-time hypertension alerts without the need for an internal power supply.

For hemodynamic and arterial compliance assessment, implantable electronic devices have been demonstrated to offer significant advantages in continuous and real-time monitoring [198–200]. Li et al. [201] introduced a unique molecular structure design strategy, creating core-shell PVDF/hydroxylamine hydrochloride organic piezoelectric nanofibers with well-controlled, self-oriented nanocrystals on uniaxially oriented β -phase fiber spaces. When implanted in pigs, these sensors exhibited ultra-high detection sensitivity and accuracy, capable of capturing micro-pressure changes on the outer wall of the cardiovascular system. The measured output voltage displayed rhythmic piezoelectric signals, providing diagnostic evidence for early assessment and diagnosis of vascular diseases, particularly for monitoring deep vein thrombosis recurrence post-surgery. Prosthetic vascular grafts are designed according to the patient's arterial structure and can be wrapped around diseased vessels. However, current technology has yet to adopt a single blood flow sensing technology for real-time monitoring of prosthetic grafts, leading to a significant risk of failure [202,203]. Natta et al. [196] introduced an ultra-thin, lightweight, and flexible smart micro-system based on piezoelectric aluminum nitride (AlN) (Fig. 7g). This system is integrated on the outer surface of vascular grafts for continuous blood flow monitoring. The AlN piezoelectric thin film, wrapped around the outer surface of the graft, generates an electrical response due to the deformation of the vascular wall. The sensor can detect frequency changes in flow rate, maintaining signal amplitude stability without significant fluctuations. Gil et al. [197] developed a force sensor made from P(VDF-TrFE) as the piezoelectric sensing layer (Fig. 7h). This sensor, integrated into the tips of medical catheters and implantable devices, has medical applications in monitoring bodily pressures. When the implant was pressed against the skin on a volunteer's wrist to detect the pulse signal directly above the radial artery, it was able to clearly distinguish important temporal features within the arterial pulse.

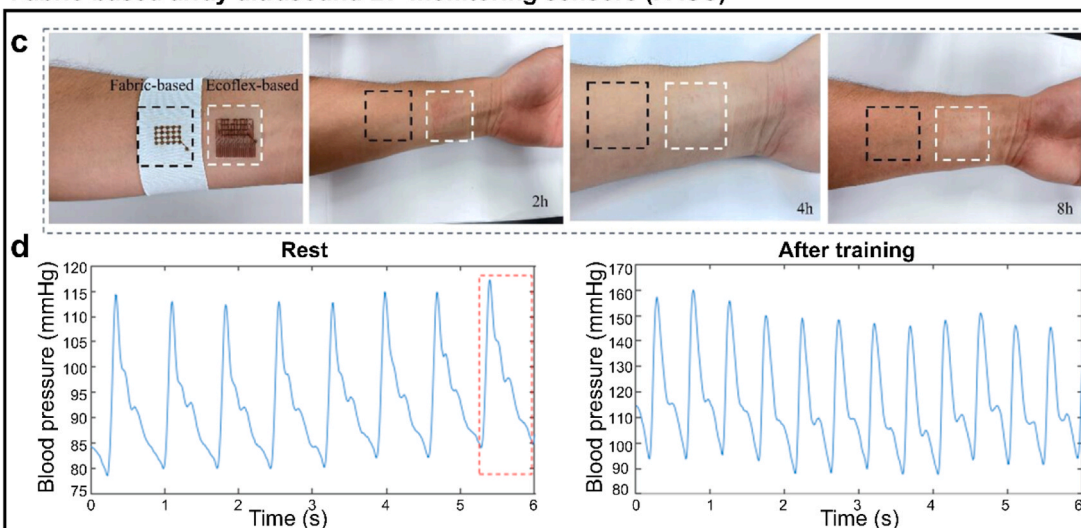
3.2. Triboelectric platforms for self-powered cardiovascular monitoring

With the continuous intersection of medical and engineering disciplines, flexible wearable electronic monitoring devices have seen rapid advancements in recent years [204,205]. However, their heavy reliance on traditional battery power limits their lifespan and consequently restricts the implementation of real-time or continuous biomedical monitoring [206,207]. Early diagnosis of heart diseases is crucial for effective prevention and treatment. Triboelectric sensors, with their high sensitivity and multiparametric detection capabilities, can monitor various cardiac function indicators [208,209], such as HRV and pulse wave characteristics. Triboelectric platforms are particularly attractive for cardiovascular monitoring due to their intrinsic capability for energy harvesting and self-powered operation. By converting low-frequency

Wearable sensor based on heterogeneously hierarchical piezoelectric technology

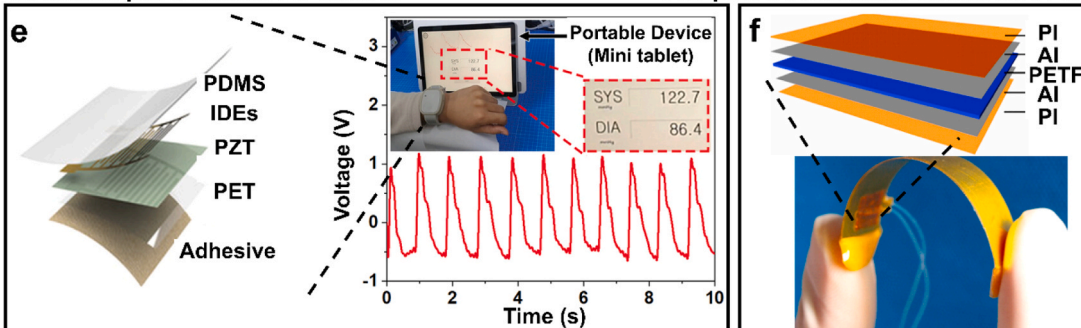


Fabric-based array ultrasound BP monitoring sensors (FAUS)



Wearable piezoelectric BP sensor

Self-powered monitor based on PETF



Flexible smart patch based on piezoelectric AlN

Small miniaturized sensor

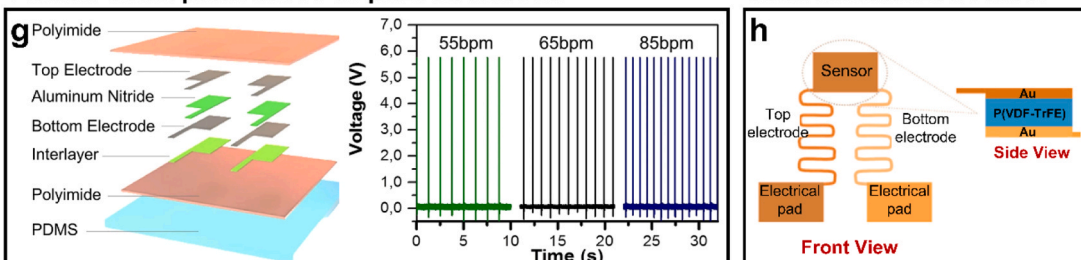


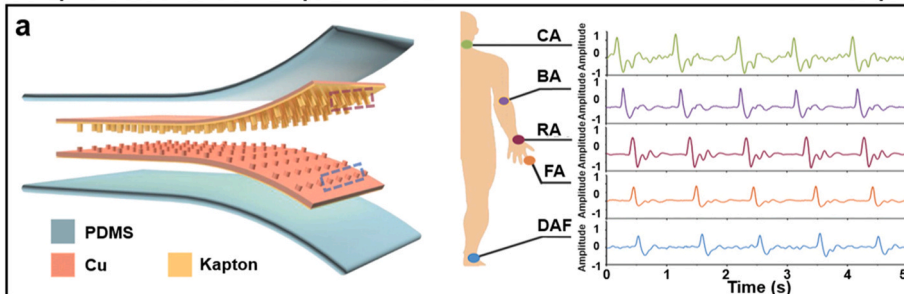
Fig. 7. Implantable and wearable piezoelectric devices enable precise hemodynamic tracking. (a-b) Composition and optical image of the thin-film piezoelectric array sensor, accompanied by a schematic illustration depicting its use in radial artery testing (Scale bar: 5 cm). Reproduced with permission [19]. Copyright 2024, Wiley-VCH. (c-d) Breathability experiment of FAUS and monitoring of radial artery BP waveform at rest and post-exercise. Reproduced with permission [191]. Copyright 2023, Wiley-VCH. (e) Schematic diagram of pulse waveform monitored by a wearable piezoelectric BP sensor transmitted to a portable device via a wireless communication circuit. Reproduced with permission [194]. Copyright 2023, Wiley-VCH. (f) Self-powered and implantable sensor based on piezoelectric film and its in vivo power waveform for BP monitoring. Reproduced with permission [195]. Copyright 2016, Elsevier. (g) Definition of different layers in the sensor's heterogeneous structure and the output voltage generated under varying flow rate conditions. Reproduced with permission [196]. Copyright 2019, Springer Nature. (h) Schematic of a miniaturized piezoelectric force sensor with vertical and planar electrode configurations. Reproduced with permission [197]. Copyright 2020, American Chemical Society.

biomechanical motions into electrical signals, these systems enable continuous sensing without the need for external power supplies, making them well suited for long-term wearable and implantable applications. Ou Yang et al. [210] introduced a flexible, self-powered, ultra-sensitive pulse sensor (SUPS) based on active triboelectric sensors. This sensor exhibits excellent output performance, high peak signal-to-noise ratio, ultra-sensitivity, and low cost (Fig. 8a). SUPS can provide accurate, wireless, real-time physiological information monitoring for cardiovascular diseases based on pulse signal analysis. Through feature index analysis, including HRV index and pulse profile analysis from patients with atrial fibrillation (AF), indicative diagnostics for cardiovascular diseases can be provided. Ma et al. [211] introduced a self-powered, multifunctional implanted triboelectric active sensor

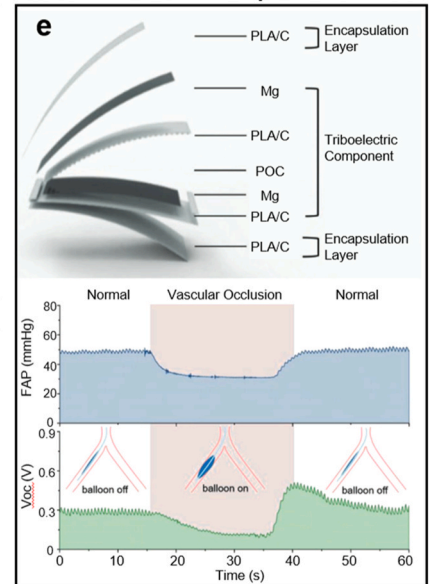
(iTEAS) (Fig. 8b). Once properly encapsulated and installed on the heart's surface, the integration of the nanostructured triboelectric layer with flexible encapsulation materials enables iTEAS to detect subtle changes in cardiac motion. The monitoring capability remains excellent even 72 h after chest closure, accurately detecting heart rate with 99% accuracy. Atrial fibrillation, premature ventricular contractions, and other arrhythmias can be detected in real-time.

The incorporation of micro/nanoscale structures into triboelectric sensors has been shown to significantly enhance signal output and sensing performance [215–220], enabling non-invasive and continuous BP monitoring. By analyzing the amplitude and velocity of pulse waves at arterial sites, these sensors offer a comfortable and wearable alternative to conventional cuff-based devices for daily use [221]. Typically,

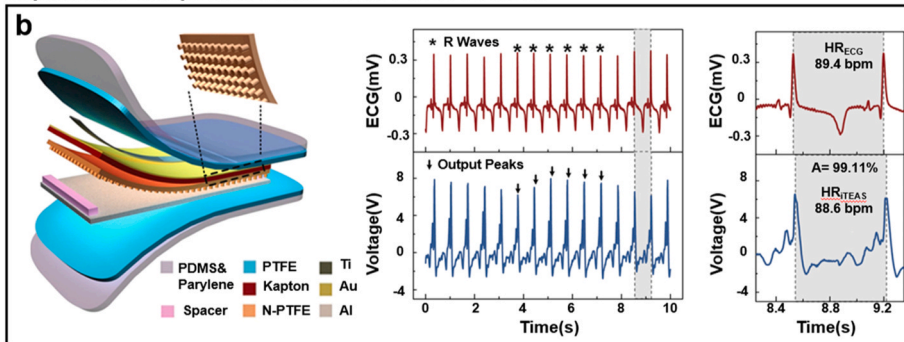
Self-powered ultrasensitive pulse sensor



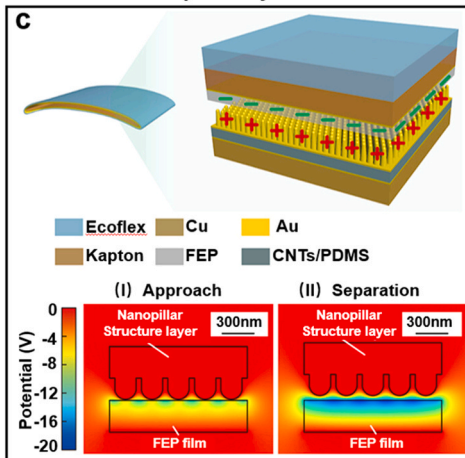
Implantable bioresorbable self-powered sensor



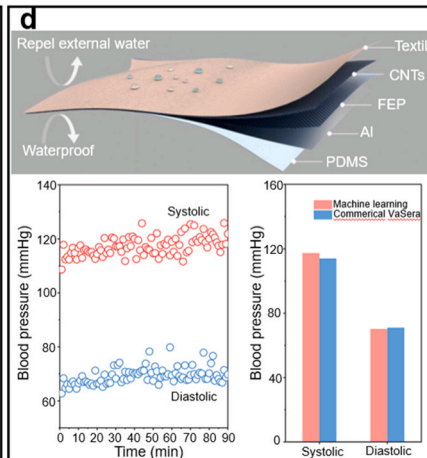
Implantable self-powered triboelectric active sensor



A triboelectric pulse sensor based on biomimetic nanopillar layer



ML-assisted textile triboelectric pulse sensor



Textile-based sensor

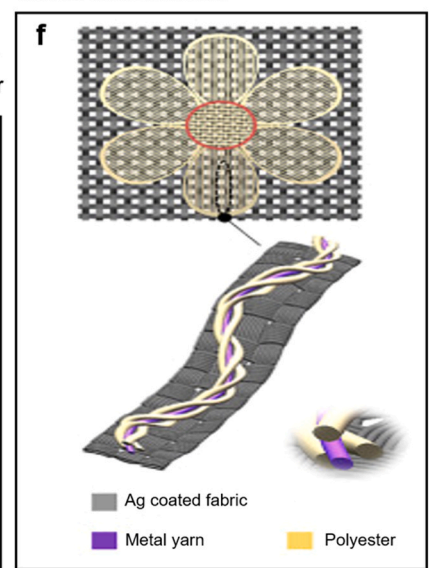


Fig. 8. Triboelectric systems enable battery-free monitoring and early diagnosis of CVDs. (a) The signal output of SUPS pressed on various artery position. Reproduced with permission. [210] Copyright 2017, Wiley-VCH. (b) Schematic diagram of the hierarchical structure of iTEAS and the R-wave voltage output of ECG. Reproduced with permission. [211] Copyright 2016, American Chemical Society. (c) Photographs of the potential difference simulated in the triboelectric layer during contact and separation, and the dual pulse wave signals detected by the wearable system. Reproduced with permission. [212] Copyright 2023, American Chemical Society. (d) The concept of a cardiovascular monitoring system based on self-powered textile triboelectric sensors. Reproduced with permission. [59] Copyright 2021, Wiley-VCH. (e) Simulated abnormal vascular occlusion events are detected by the sensor. Reproduced with permission. [213] Copyright 2021, Wiley-VCH. (f) Schematic illustration of the textile-based sensor as fabricated. Reproduced with permission [214]. Copyright 2020, Elsevier.

triboelectric pressure sensors can be viewed as an electrical model of several capacitors connected in series [222]. However, balancing sensitivity with detection range remains a pressing issue [223]. Yang et al. [224] reported the first self-powered bionic membrane sensor based on the triboelectric effect. This sensor is wearable, lightweight, cost-effective, and capable of multimodal operation, laying a solid foundation for the application of triboelectric sensors in health monitoring. Lei et al. [225] reported a highly sensitive flexible triboelectric patch with zero power consumption. Additionally, thanks to the porous microstructure of the Polyurethane (PU) sponge frame, the proposed flexible triboelectric patch achieved a sensitivity increase of five times, reaching 5.93 kPa^{-1} , in the pressure range of 0–5 kPa compared to devices with solid triboelectric layers, allowing for precise monitoring of faint pulse wave signals. Yao et al. [212] developed a wearable triboelectric pulse sensor based on a biomimetic nanopillar layer (Fig. 8c). By replicating the soft lithography of cicada wings, flexible conductive nanopillars were fabricated as the triboelectric layer, significantly improving the sensor's output performance and enabling the detection of weak signal characteristics in pulse waveforms. The sensor is integrated with personalized ML to provide accurate and continuous BP monitoring. By combining the sensor with personalized partial least squares regression (PLSR) ML, unknown BP can be derived based on individual pulse characteristics with reasonable accuracy, circumventing the variability issues encountered with generic PLSR ML or formulaic algorithms. Fang et al. [59] developed a ML-assisted textile triboelectric sensor capable of operating in sweaty conditions, which converts subtle skin deformations caused by arterial pulsations into electrical signals (Fig. 8d). The layered carbon nanotube triboelectric layer used in this sensor enhances its sensitivity, achieving a signal-to-noise ratio of 23.3 dB, a response time of 40 ms, and a sensitivity of up to $0.21 \mu\text{A kPa}^{-1}$. With the assistance of ML algorithms, the textile triboelectric sensor can continuously and accurately measure SBP and DBP, validated against cuff-based commercial BP measurements in hospitals. Minimally invasive monitoring of endocardial pressure (EP) changes is of great clinical significance for heart failure patients with impaired cardiac function [226]. Liu et al. [227] proposed a miniaturized, flexible, ultrasensitive, self-powered endocardial pressure sensor based on triboelectric principles for real-time EP monitoring. In vitro testing shows the device oscillates more than 100 million times in humid environments with mechanical and electrical stability. Arrhythmias such as ventricular fibrillation and premature ventricular contractions can also be detected according to EP changes. Gao et al. [228] developed a triboelectric cardiovascular monitoring system incorporating a gradient micro/nanostructured hydrogel interface to enhance pressure sensitivity and signal fidelity. The device was evaluated in 214 measurements against an FDA-approved sphygmomanometer (OMRON), showing mean errors of $0.61 \pm 4.72 \text{ mmHg}$ (SBP) and $1.9 \pm 3.83 \text{ mmHg}$ (DBP). Although not yet FDA-approved, the system underwent IRB-approved human testing, indicating strong potential for clinical translation. These advances demonstrate the potential to safely sense pressure conditions, diagnose, and monitor cardiovascular diseases, showing promise for implantable healthcare monitoring.

Arteriosclerosis is a critical risk factor for numerous cardiovascular diseases. Triboelectric sensors have emerged as effective tools for assessing arterial stiffness by detecting changes in pulse wave velocity and waveform. Increased arterial stiffness typically results in faster pulse wave transmission, and triboelectric sensors can record these variations in real time, providing a valuable means for early arteriosclerosis detection. Jiang et al. [229] proposed a flexible, waterproof, and breathable self-powered textile pulse sensor based on a triboelectric nanogenerator. Upon integrating signal acquisition modules and a Bluetooth chip, the device can transmit pulse waveforms to smart terminals in real-time, enabling wireless health monitoring. By analyzing pulse waveform features, including stiffness index (SI) and AIx, the wireless health monitoring system facilitates early diagnosis of arterial stiffness. OuYang et al. [213] introduced an implantable, bioresorbable,

self-powered sensor based on the triboelectric effect, with a sensitivity of up to $11 \text{ mV}\cdot\text{mmHg}^{-1}$, showcasing excellent linearity ($R^2=99.3\%$), antibacterial properties, and degradability (Fig. 8e). A vascular occlusion model was constructed through implantation into a balloon, and the sensor accurately identified simulated vascular blockage events, indicating substantial application potential. Meng et al. [214] developed a textile-based sensor with a sensitivity of 3.88 V/kPa , which enhances wearing comfort and can accurately capture pulse wave signals from elderly and frail individuals. This sensor is suitable for long-term continuous physiological signal monitoring (Fig. 8f).

3.3. Optical technologies for hemodynamic and oxygenation sensing

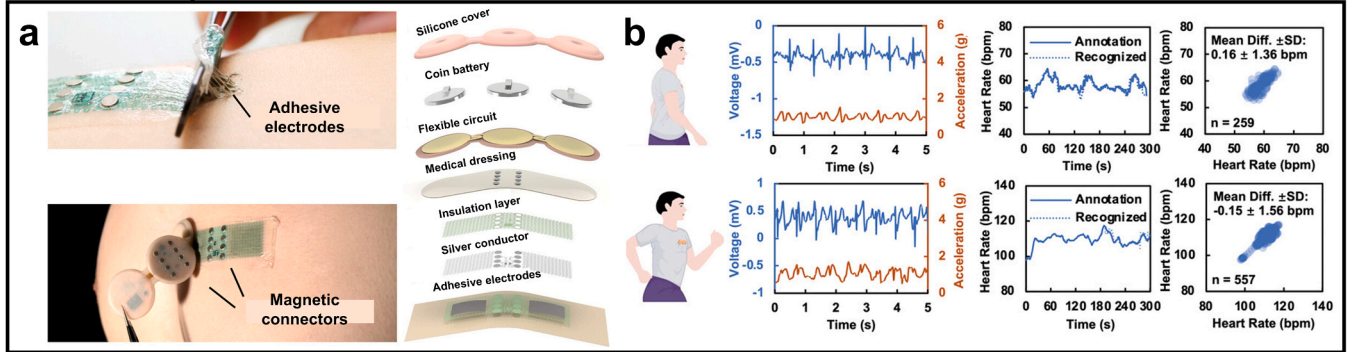
Optical sensing technologies for cardiovascular monitoring can be categorized into three complementary subtypes: PPG, optical fiber-based sensing, and surface-enhanced Raman spectroscopy (SERS), which differ in sensing principles, physiological scales, and functional roles. PPG captures pulsatile blood volume variations and enables continuous monitoring of heart rate, oxygen saturation, and pulse waveform morphology. Optical fiber sensors provide mechanically sensitive and spatially flexible platforms for detecting vascular dynamics such as pulse wave propagation and arterial stiffness. In contrast, SERS operates at the molecular level and enables highly sensitive detection of cardiovascular biomarkers. Together, these approaches cover multiple levels of cardiovascular information, including hemodynamic signals, biomechanical responses, and biochemical markers, supporting a complementary and integrated sensing strategy for cardiovascular assessment.

3.3.1. PPG

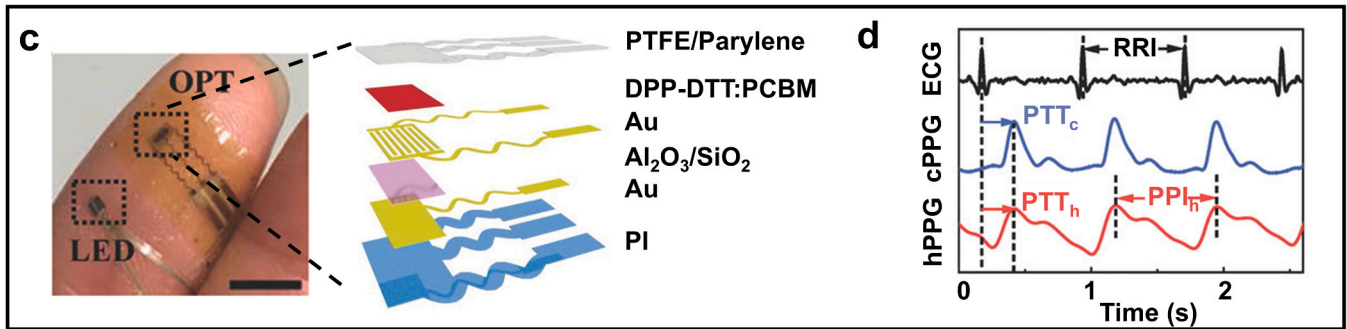
PPG is a widely used optical sensing technique that measures blood volume changes in the microvascular bed by detecting variations in light absorption. Light emitted onto the skin is modulated by pulsatile blood flow, producing characteristic waveforms that reflect cardiac activity and vascular dynamics. By employing multiple wavelengths, typically red and infrared, PPG can also estimate SpO_2 , making it valuable for assessing both circulatory and respiratory function. Because it is noninvasive, power-efficient, and easy to integrate into wearable devices, PPG has become a core technology in smartwatches and fitness trackers for continuous cardiovascular and oxygen monitoring in daily life.

Traditional multimodal sensing often relies on combining single-function systems distributed across different body locations. This approach is frequently hampered by signal fluctuations [230,231] and motion artifacts [232,233] caused by unreliable skin contact, which significantly affect effective signal acquisition. To overcome these challenges, Wu et al. [234] used a skin patch with self-adhesive conductive electrodes and a low-latency self-compensation algorithm, employing a unique synchronous correlation peak extraction algorithm to simultaneously detect dynamic multi-parameter signals, including ECG, PPG, and temperature on the chest, reducing HR inference errors compared to traditional methods (Fig. 9a, b). However, most examples of these emerging devices require batteries and/or hardwired connections to operate [235,236]. To address this limitation, Kim et al. [54] introduced a thin, stretchable wireless optical system using NFC technology to monitor heart rate by measuring backscattered light from an infrared LED with a silicon (PIN) photodetector, allowing intimate skin contact for various health information acquisition. Ma et al. [237] demonstrated a wearable ear-mounted PPG system capable of robust HR and SpO_2 detection under low power conditions and wireless transmission, emphasizing the importance of form factor and motion robustness in real-world scenarios. Jarchi et al. [238] evaluated commercial wrist-worn devices (e.g., Wavelet Health, Huawei Watch, Creative Medical PC-68B) and confirmed that mature PPG-based systems can reliably extract both HRV and SpO_2 in clinical settings. Their findings demonstrated good agreement with bedside monitors, highlighting

Flexible skin patches based on PPG



Multimodal and wireless skin-interfaced system



PPG-based hemorrhagic risk prediction

Soft wearable system based on fiber optic technology

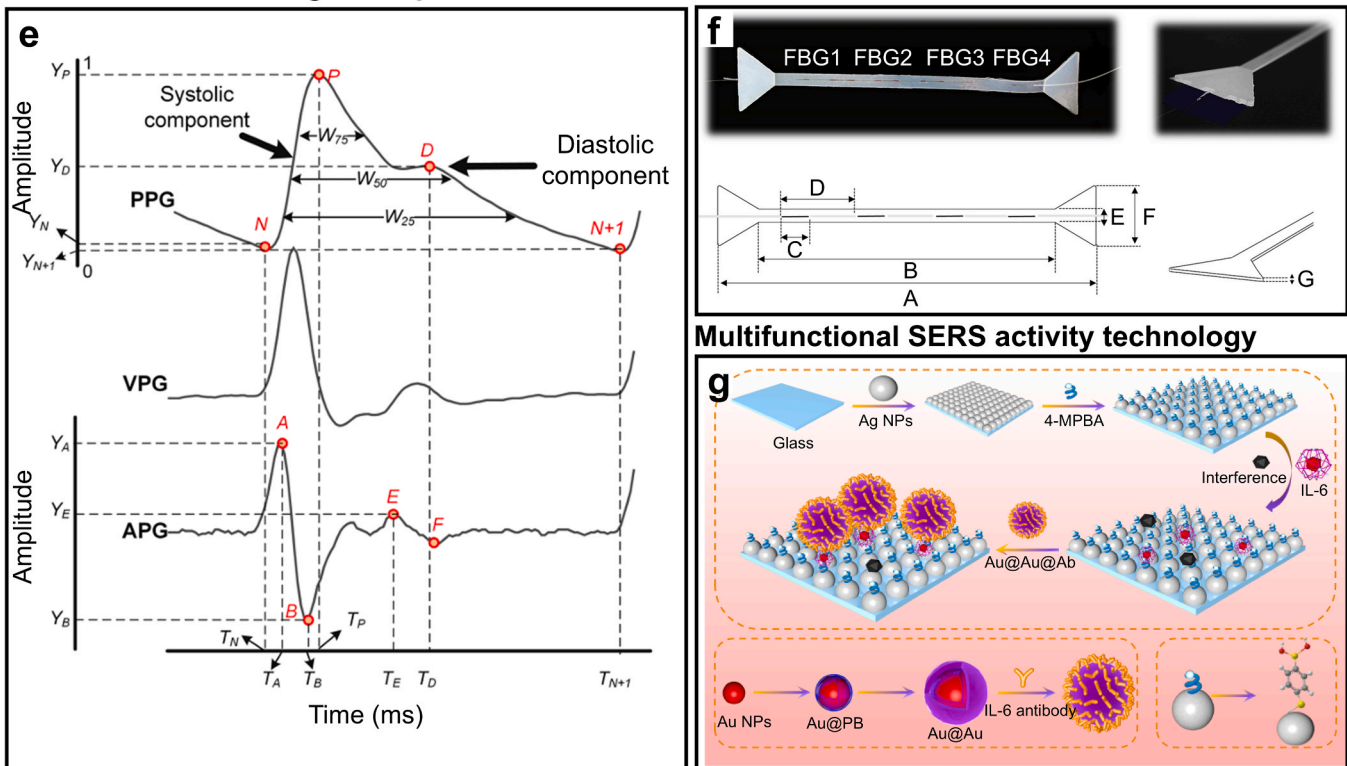


Fig. 9. Advanced PPG sensors integrate multimodal signals for real-time cardiovascular monitoring. (a) Image of the skin patch incorporating various sensing components. (b) Synchronized ECG and acceleration signals recorded during walking and running states. Reproduced with permission. [234] Copyright 2023, Wiley-VCH. (c) Schematic representation of the flexible OPT device structure. (d) Typical output signals captured by ECG, cPPG, and hPPG sensors, along with an illustration depicting pulse transit time. Reproduced with permission. [240] Copyright 2017, Wiley-VCH. (e) PPG, the first derivative of PPG (VPG), the second derivative of PPG (APG) signal and their features. The abbreviations T and Y denote the time (ms) and amplitude of the corresponding points, respectively. Reproduced with permission. [241] Copyright 2022, Springer Nature. (f) The soft skin like wearable system with the four FBGs encapsulated into: pictures and geometrical features. Reproduced with permission [242]. Copyright 2021, Springer Nature. (g) The scheme of IL-6 detection procedure based on SERS. Reproduced with permission [243]. Copyright 2021, Elsevier.

the feasibility of deploying consumer-grade PPG technology for multi-modal physiological monitoring in real-world healthcare scenarios. Jiang et al. [239] developed a hybrid photoacoustic and piezoelectric ultrasound implant, which was evaluated *ex vivo* under porcine tissue. Stable wireless energy transfer and signal transmission were demonstrated within FDA-recommended safety limits for diagnostic ultrasound, including mechanical and thermal indices. Although the device has not received FDA approval, it represents a preclinical-stage sensing platform designed in accordance with regulatory safety guidelines.

Despite the existence of various continuous BP measurement methods, they often face issues of poor generalizability across different individuals and wearing conditions [244,245]. To address this, Wang et al. [246] proposed a novel BP measurement method and developed a dual-sided smartwatch device comprising dual PPG sensors and a custom interface sensor. The device utilizes PPG sensors placed on the wrist and the back of the hand to detect cardiac output and pulse waveform characteristics. ML algorithms were employed to integrate the detected multi-channel signals, enabling real-time estimation of continuous BP. This dual PPG sensing method effectively mitigates individual differences in PPG signals caused by varying wear conditions and individual variations. Li et al. [247] reported a high-performance skin-like photonic system integrated with ultra-thin flexible circuits. By introducing a theoretical model based on a virtual working principle, this system predicts accurate BP and suppresses motion artifacts. The system employs optical differentiation in the frequency domain for stable optical measurements within the skin-like device. Encased in a 50 μm thick biocompatible film, the thin and flexible device adheres to the skin through van der Waals interactions, allowing precise, continuous, non-invasive BP monitoring. To address the rigidity and power consumption issues of PPG sensors, Xu et al. [240] developed an epidermal flexible near-infrared (NIR) PPG sensor integrating low-power, high-sensitivity organic phototransistors and efficient inorganic light-emitting diodes (Fig. 9c, d). This epidermal/flexible PPG sensor, combined with ECG monitoring, reliably tracked pulse pressure changes in various body postures with lower power consumption than commercial PPG sensors. He et al. [241] combined PPG and ML to develop a predictive bleeding risk model (Fig. 9e). The model achieved an AUC of 0.762, offering a dynamic and continuous assessment of bleeding risk, enabling real-time treatment adjustments for improved patient safety.

3.3.2. Optical fiber sensor

Flexible optical fiber sensing has emerged as a compelling strategy for non-invasive and continuous cardiovascular monitoring, owing to its intrinsic advantages in biocompatibility, electromagnetic immunity, and high mechanical sensitivity. By embedding optical fibers into soft wearable substrates, these sensors can detect subtle physiological signals, such as pulse waves, blood pressure variations, and respiratory motion, through real-time modulation of light transmission. Recent progress in fiber structuring (e.g., twin-core fibers, FBGs, interferometers), waveguide integration, and advanced signal decoding algorithms has significantly improved their performance in terms of sensitivity, adaptability, and long-term stability [248].

Tan et al. [249] introduced a mattress-based interferometric system using twin-core optical fibers for non-contact, simultaneous monitoring of heart rate and respiration, highlighting its potential for unobtrusive home health tracking. Building on this, Lyu et al. [250] developed a non-invasive fiber-optic ballistocardiography (BCG) monitoring system embedded in a smart cushion, capable of capturing cardiac-induced micro-vibrations using an interferometric 3×3 coupler. The system demonstrated high accuracy in heart rate and contractility assessment, showing strong agreement with ECG-derived metrics across various physiological states. This approach offers a practical and contact-free solution for continuous cardiovascular monitoring in daily settings. Pang et al. [251] developed a wrist-worn sensor based on a single-mode-multimode-singlemode fiber structure. Combined with a fluidic capsule and ML algorithms, the system achieved Association for the

Advancement of Medical Instrumentation (AAMI) grade A accuracy, demonstrating strong potential for wearable integration.

Advancing multipoint and multiparameter sensing, Presti et al. [242] embedded multiple FBGs into a soft chest patch for synchronized SCG monitoring (Fig. 9f). The system enhanced HR estimation accuracy and maintained signal quality in high-interference environments such as MRI. Wang et al. [146] designed a miniature interferometric sensor with a reflective aluminum diaphragm tip, enabling ΔL_x measurement when affixed to the radial artery. Results showed good alignment with clinical standards. Gowda et al. [252] further developed an FBG-based pulse pressure sensor with a 3D-printed adapter that eliminated the need for precise alignment, achieving < 3 mmHg error in radial artery measurements across human subjects-demonstrating strong adaptability for real-world applications.

As a result, a wide range of system configurations has been developed and validated in both laboratory and clinical settings, including mattress-based sensors, seat-integrated devices, and wrist- or chest-worn wearables. These efforts collectively demonstrate the increasing feasibility of optical fiber technologies for next-generation cardiovascular health monitoring.

3.3.3. SERS

SERS has emerged as a powerful analytical technique for non-invasive, molecular-level monitoring in cardiovascular health. By leveraging signal amplification from nanostructured metallic substrates, SERS enables ultrasensitive detection of cardiovascular biomarkers, including lipids, reactive oxygen species (ROS), and circulating micro-RNAs, at physiologically relevant concentrations. Unlike conventional optical methods, SERS offers both high chemical specificity and low detection limits, making it particularly suited for early-stage diagnosis of cardiovascular diseases. Recent advances in substrate design, microfluidic integration, and portable Raman platforms have further enhanced its applicability in point-of-care and wearable diagnostic systems, paving the way for real-time, label-free cardiovascular assessment [253].

SERS has emerged as a powerful tool for ultrasensitive cardiovascular biomarker detection. In one early effort, Zhou et al. [243] introduced a dual-model calibration strategy that integrates experimental and theoretical data to improve IL-6 detection accuracy in complex serum environments, resulting in picomolar sensitivity and enhanced reproducibility (Fig. 9g). Expanding on platform integration, Wang et al. [254] developed a microcavity-based SERS chip that enables rapid, high-specificity detection of cardiac troponin I (cTnI), offering a compact format suitable for clinical and wearable use. Similarly, Muhammad et al. [255] demonstrated a label-free biosensor using aptamer-functionalized gold nanoparticle arrays for IL-6 detection, removing the need for antibodies or tags and simplifying point-of-care deployment.

To enable multiplexed analysis, Cheng et al. [256] designed a gold-patterned SERS chip for simultaneous detection of cTnI and CK-MB, achieving femtomolar sensitivity with high throughput. Chon et al. [257] further advanced this approach with a magnetic bead-based competitive immunoassay that accurately quantified the same two markers within minutes, showing excellent agreement with standard clinical methods. For more comprehensive diagnostics, Su et al. [258] engineered a bimetallic nanostar array that enhanced near-field plasmon coupling and enabled simultaneous detection of NT-proBNP, cTnI, and creatinine at attomolar levels in unprocessed serum. Hu et al. [259] developed a core-shell Au@Ag nanotag system integrated with magnetic enrichment for dual detection of cTnI and H-FABP, achieving rapid readout and low detection limits ideal for emergency settings. Together, these studies highlight SERS as a highly adaptable sensing strategy for cardiovascular health monitoring-offering exceptional sensitivity, multiplexing capability, and increasing clinical relevance.

3.4. Magnetoelastic platforms for wireless cardiovascular monitoring

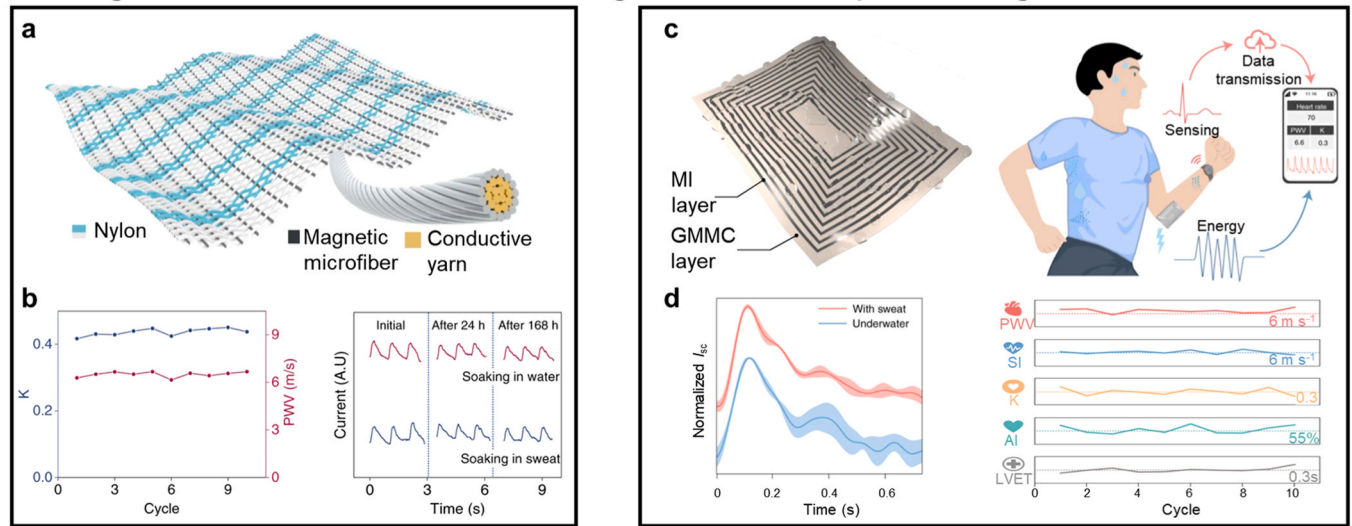
Magnetoelastic platforms have recently emerged as a promising approach for wireless cardiovascular monitoring, particularly in scenarios where passive operation and minimal physical coupling are required. Unlike conventional electrical sensing modalities, magnetoelastic systems transduce biomechanical stimuli into variations in magnetic properties, which can be interrogated remotely through external magnetic fields. This wireless and battery-free operation eliminates the need for direct electrical connections, making these platforms especially suitable for long-term monitoring in wearable and implantable settings.

Recent advances in magnetic-field-mediated sensing have enabled a

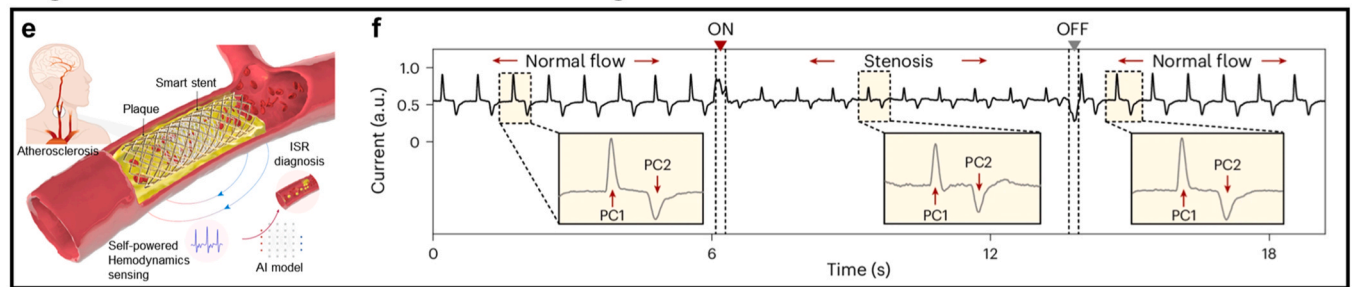
progressive transition from wearable to fully implantable platforms. For instance, Zhao et al. [260] developed soft magnetoelastic fibers that can be woven into textiles (Fig. 10a), where mechanical deformation modulates magnetic responses, enabling stable pulse-wave monitoring with minimal performance degradation over repeated cycles and under challenging conditions such as water or sweat exposure (Fig. 10b). Building on this textile-based architecture, the group further engineered a stretchable self-powered magnetoelastic sensor (Fig. 10c), in which magnetically induced electrical outputs enable continuous acquisition of physiological signals and cardiovascular indices, while maintaining reliable signal fidelity under dynamic deformation and humid or underwater conditions [55] (Fig. 10d).

Moving toward implantable applications, Chen et al. [261]

Soft magnetoelastic fibers for wearable sensing Stretchable self-powered magnetoelastic sensor



Magnetoelastic smart stents for restenosis diagnosis



Chipless magnetic implants for wireless multimodal sensing

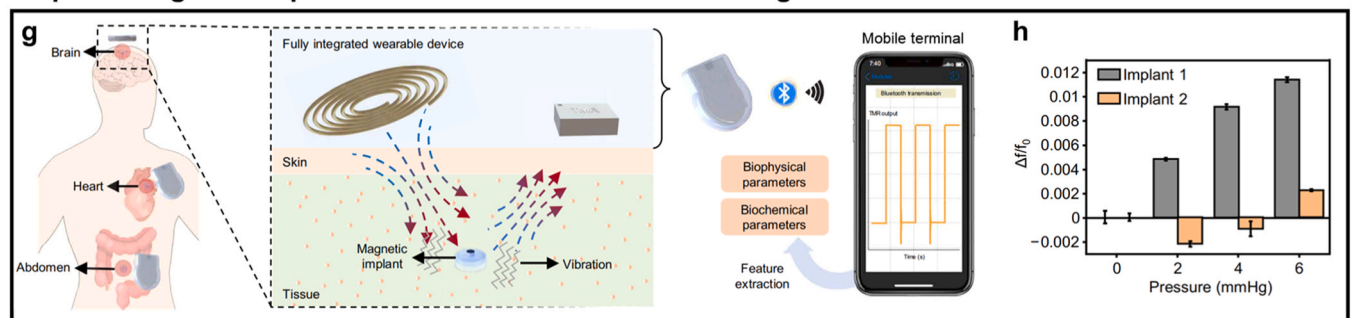


Fig. 10. Magnetoelastic and magnetic platforms for wireless physiological sensing from wearable to implantable systems. (a) Soft magnetoelastic fibers integrated into a textile architecture. (b) Stable pulse-wave monitoring under cyclic loading and in water/sweat environments. Reproduced with permission [260]. Copyright 2021, Springer Nature. (c) Stretchable self-powered magnetoelastic sensor with layered structure. (d) Robust signal output and cardiovascular parameter extraction under varying conditions. Reproduced with permission [55]. Copyright 2021, Springer Nature. (e) Magnetoelastic smart stent for in situ hemodynamic monitoring and restenosis diagnosis. (f) Distinct waveform features under normal and stenotic flow conditions. Reproduced with permission [261]. Copyright 2026, Springer Nature. (g) Chipless magnetic implant system for wireless multimodal sensing. (h) Reproducible pressure-dependent response across different implants. Reproduced with permission [262]. Copyright 2024, American Association for the Advancement of Science.

introduced a magnetoelastic smart stent (Fig. 10e) that transduces flow-induced structural deformation into wireless magnetic signals, enabling real-time identification of hemodynamic abnormalities such as restenosis, with distinct waveform features observed between normal and stenotic conditions (Fig. 10f). Extending this paradigm, Wan et al. [262] demonstrated millimeter-scale chipless magnetic implants integrated with a wearable interrogation system (Fig. 10g), where externally driven magnetic vibration encodes local mechanical and biochemical information, enabling quantitative wireless sensing of pressure as well as broader multimodal physicochemical parameters,

with reproducible signal responses across different implant units (Fig. 10h).

Together, these studies delineate a coherent evolution from textile-based and stretchable devices to minimally invasive and fully implantable systems, accompanied by a transition from motion-derived hemodynamic sensing to more integrated physicochemical monitoring. This progression highlights magnetic sensing as a promising pathway for cardiovascular applications requiring wireless operation, deep-tissue accessibility, and long-term stability.

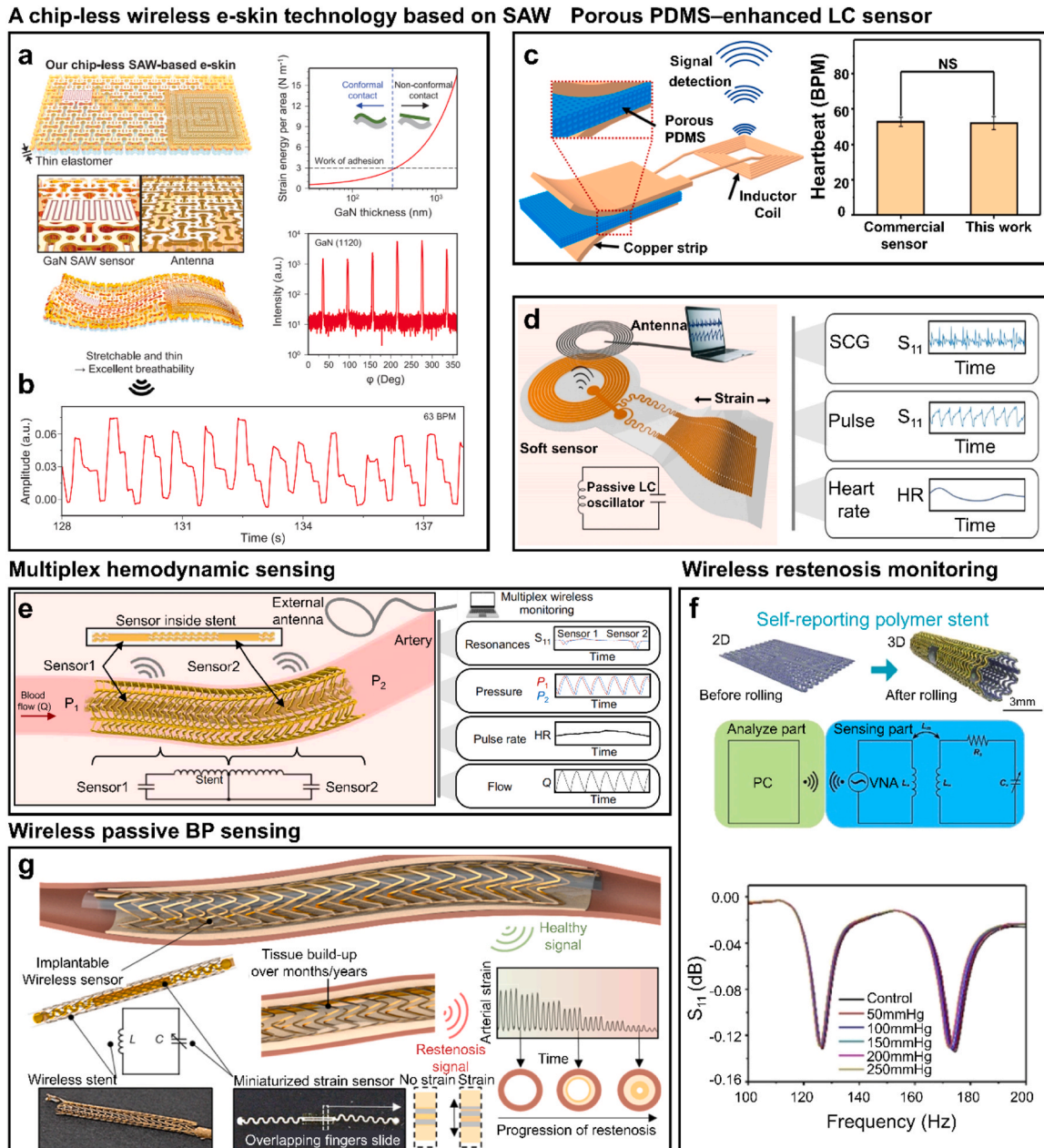


Fig. 11. Wireless LC sensors enable multimodal monitoring of cardiac function. (a-b) Chipless wireless electronic skin made of GaN-based SAW and real-time continuous wireless pulse measurements. Reproduced with permission [38]. Copyright 2022, American Association for the Advancement of Science. (c) Schematic illustration of porous PDMS based LC sensor for wireless monitoring. Reproduced with permission [268]. Copyright 2024, Elsevier. (d) Schematic of the inductive coupling measurement method. The soft wearable sensor can measure SCG, pulse, and HR. Reproduced with permission [269]. Copyright 2023, Woon-Hong Yeo. (e) Illustration of the wireless design and sensing scheme to simultaneously monitor pressure, HR, and flow. Reproduced with permission [272]. Copyright 2022, American Association for the Advancement of Science. (f) Device concept and working mechanism of the proposed self-reporting stent. Finite element method (FEM) simulation analysis of a planar inductor coil with a two-dimensional design and its self-assembled 3D structure with a 3 mm radius was used to determine the electrical parameters of the proposed device. Reproduced with permission [273]. Copyright 2023, American Chemical Society. (g) Overview of a soft implantable bioelectronic system for wireless restenosis monitoring. Reproduced with permission [274]. Copyright 2023, Elsevier.

3.5. Capacitive sensors for contactless and comfortable cardiovascular monitoring

3.5.1. LC-resonant wireless capacitive sensors for battery-free cardiovascular sensing

For broader application of electronic skin, the key technology is the ability to wirelessly transmit data [263]. Traditional wireless electronic skin requires rigid IC chips, leading to issues such as reduced sensitivity, significant heat generation [264], and short communication distances [265]. LC-resonant capacitive sensors have gained increasing attention in battery-free cardiovascular monitoring due to their ability to wirelessly transmit physiological signals through resonance frequency shifts. By eliminating the need for onboard power supplies, these systems enable long-term, continuous monitoring with minimal user burden. Depending on the application context, LC-based sensors can be implemented in either wearable formats for external signal acquisition or implantable designs for deep-tissue monitoring.

Bao Zhenan's group developed a battery-free, skin-conformal LC sensor integrating a stretchable antenna and pressure-sensitive capacitor for wireless pulse monitoring. The device enables stable resonance tracking during motion, supporting non-invasive, continuous cardiovascular sensing [266]. The team further developed a fully printed, skin-interfaced LC microsystem with a multilayer serpentine architecture that integrates pressure sensing, wireless transmission, and soft packaging into a single platform [267]. The system employs aerosol jet printing to fabricate microstructure capacitive sensors and inductive antennas directly onto elastomeric substrates, ensuring high sensitivity to arterial pulse pressure while maintaining mechanical robustness under dynamic skin deformation. Kim et al. [38] introduced a chipless wireless electronic skin based on a surface acoustic wave (SAW) sensor made of ultra-thin single-crystal piezoelectric GaN film, demonstrating a week-long pulse monitoring capability, offering new designs for wireless cardiac rhythm monitoring devices (Fig. 11a, b). Thorali et al. [268] developed a highly flexible LC-based wireless capacitive pressure sensor using a PDMS dielectric layer, synthesized via a novel solvent-free steaming method (Fig. 11c). By tuning porosity up to 75%, the sensor exhibited significantly enhanced compressibility and dielectric tunability, achieving up to $11 \times$ sensitivity improvement over non-porous PDMS and enabling detection sensitivities of up to 0.046 kPa^{-1} . Integrated into an LC resonant circuit, the sensor enabled battery-free wireless monitoring of vital physiological signals such as radial pulse, skin temperature, swallowing, and facial muscle activity. Guess et al. [269] developed a fully wireless, battery-free, and chipless soft sensor based on LC resonance for SCG monitoring (Fig. 11d). Fabricated via single-layer laser micromachining of copper on polyimide and encapsulated in stretchable elastomer, the ultrathin sensor ($<1 \text{ mm}$) adheres conformally to the chest and captures high-fidelity cardiac vibrations through inductive coupling. Compared with commercial accelerometers and ECG systems, the sensor reliably identified SCG fiducial points, demonstrating strong potential for unobtrusive, ambulatory cardiovascular health monitoring. Lyu et al. [270] proposed a chipless, battery-free LC wireless sensing system inspired by repeater design to enhance coupling between a lightweight reader and flexible capacitive sensors. The system enables multi-site, contactless monitoring of cardiovascular and respiratory signals through clothing, demonstrating strong potential for wearable body sensor networks. Hsu et al. [271] developed a pocket-sized, fully noncontact LC-based eddy current sensor for simultaneous cardiac and respiratory monitoring. Operating through clothing without skin contact, the device accurately tracks heart and breath signals in real time, offering a compact solution for unobtrusive cardiopulmonary monitoring.

Recent progress in implantable LC-based sensors has moved beyond standalone pressure detection toward multifunctional, structurally adaptive platforms capable of long-term, wireless cardiovascular monitoring. Bao Zhenan's team developed a fully biodegradable LC-based sensor integrating fringe-field capacitive sensing with a bilayer

inductor design for wireless, battery-free arterial pulse monitoring [275]. In vivo studies in rats confirmed high sensitivity, biocompatibility, and stable monitoring over 12 weeks, highlighting its potential for long-term postoperative cardiovascular surveillance. Herbert et al. [272] laid the foundation with a fully integrated LC vascular electronic system, embedding a printed capacitive sensor into an inductive stent for simultaneous tracking of pressure, flow, and pulse rate (Fig. 11e). Their system demonstrated high spatial integration, catheter compatibility, and extended wireless readout in vivo, establishing a benchmark for multimodal, clinically translatable implants. Ouyunbaatar et al. [276] introduced a self-rollable SU-8 polymer stent incorporating a planar-fabricated LC sensor. This structure offered process scalability and stable operation across biological media, achieving 21 kHz/mmHg sensitivity under wireless interrogation. To address implantability and fabrication constraints, they later developed a dual-sensor self-reporting stent that combines pressure and flow sensing via an expanded LC inductor coil [273]. This design significantly enhanced sensitivity (32 kHz/mmHg) and enabled early detection of in-stent restenosis (Fig. 11f). Focusing on mechanical compatibility and vascular integration, Yeh et al. [277] explored a thin, parylene-C-based LC sensor designed for seamless stent integration. While more foundational, this work emphasized linearity (up to 250 mmHg), phase-dip readout, and material biocompatibility-providing essential design principles for future clinical systems. Rigo et al. [274] presented a microneedle-printed, miniaturized LC sensor embedded along the inner stent wall (Fig. 11g). The optimized finger-like capacitive geometry allowed for high-resolution arterial strain sensing, showing improvement in wireless sensitivity for restenosis monitoring. In contrast to intravascular integration, Hu et al. [278] proposed an extra-arterial LC sensor mounted outside the vessel wall, minimizing direct trauma while maintaining high sensitivity (8.53 kHz/mmHg). Their system demonstrated accurate in vivo pressure tracking in rats, highlighting an alternative route for chronic BP monitoring with reduced procedural complexity.

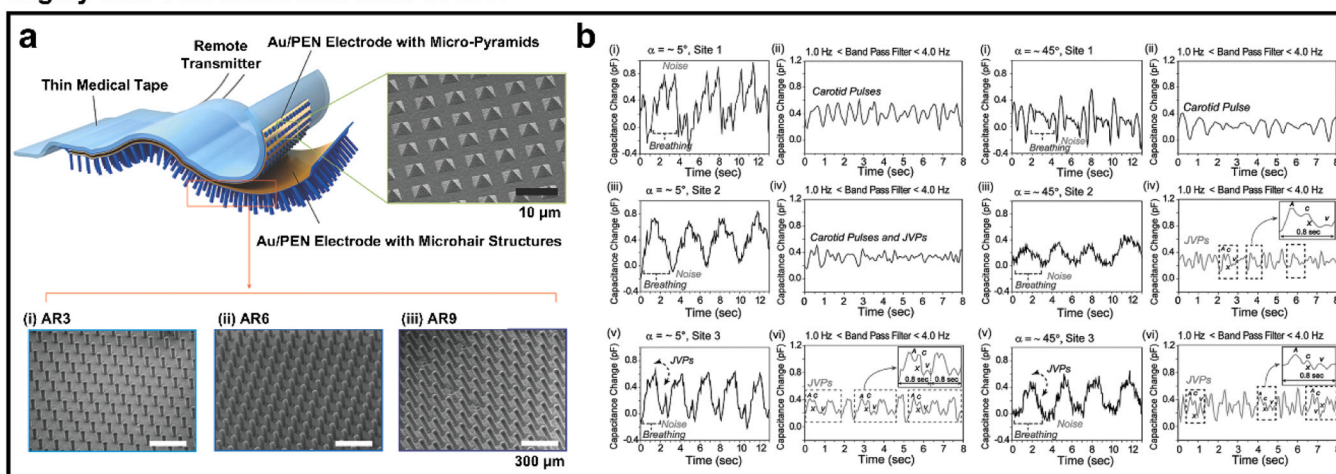
Advances in LC-resonant capacitive sensors have enabled battery-free, wireless cardiovascular monitoring in both wearable and implantable formats. Wearable systems emphasize flexibility, breathability, and noncontact operation for unobtrusive multi-site sensing, while implantable devices focus on structural integration, multimodal monitoring, and long-term stability. Together, they highlight the versatility of LC platforms in supporting seamless, continuous tracking of vital cardiovascular signals across diverse clinical scenarios.

3.5.2. Contactless and conformal capacitive sensing for comfortable cardiovascular monitoring

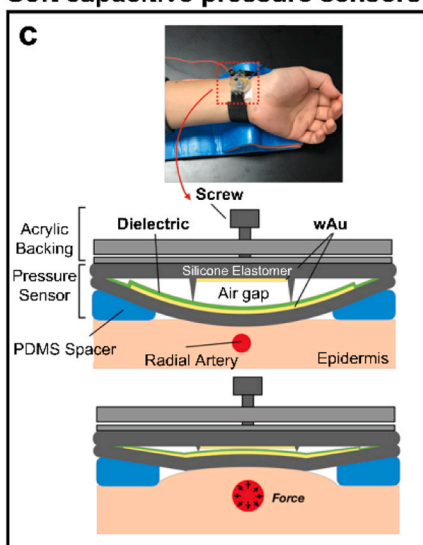
Capacitive sensors offer a promising route for unobtrusive cardiovascular monitoring due to their simple structure, high sensitivity, and compatibility with soft, skin-mounted systems. Recent efforts have focused on improving conformability, comfort, and mechanical coupling on irregular or moving skin surfaces through advanced materials and microstructural designs. This section summarizes representative strategies addressing these challenges for real-time, skin-integrated sensing.

Despite rapid technological advancements, achieving ultra-conformal contact on irregular, non-flat, and rough skin surfaces remains a challenge for real-time diagnostics. Inspired by the micro-hair structures found in nature [279,280], Pang et al. [281] developed a wearable bandage-like capacitive pressure sensor with bio-inspired microhair structures of various aspect ratios for ultra-compliance on non-flat surfaces, significantly improving signal-to-noise ratio (Fig. 12a). The portable sensor with a wireless transmitter measured weak signals from deep jugular venous pulse, associated with arrhythmia diagnosis and pathology (Fig. 12b). Kim et al. [282] demonstrated an innovative approach to developing soft capacitive pressure sensors, showcasing exceptional electromechanical properties. Utilizing soft, highly wrinkled thin-film electrodes, these sensors can effectively couple with the human body to quantify local and subtle

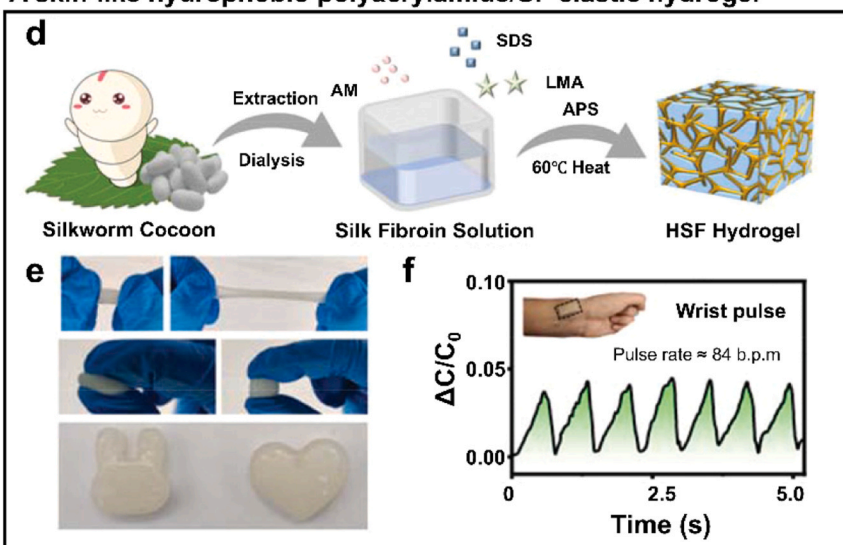
Highly skin-conformal microhair sensor



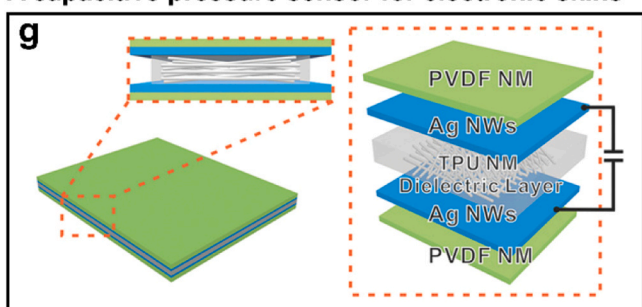
Soft capacitive pressure sensors



A skin-like hydrophobic-polyacrylamide/SF elastic hydrogel



A capacitive pressure sensor for electronic skins



An ionic capacitive pressure sensor

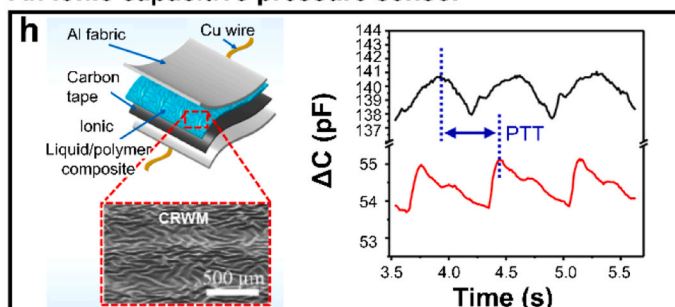


Fig. 12. Capacitive sensors enable conformal, contactless, and sensitive cardiovascular tracking. (a) Cross-sectional view of the micro-hair structure sensor. (b) Capacitive responses to various physical activities (i, iii, and v) and carotid pulse waveforms measured from a healthy individual and a cardiac patient. Reproduced with permission. [281] Copyright 2015, Wiley-VCH. (c) Soft capacitive pressure sensor placed on the wrist, detecting deformation caused by blood flow through the radial artery. Reproduced with permission. [282] Copyright 2019, Wiley-VCH. (d) Schematic illustration of the preparation of HSF hydrogel. (e) Optical images of the HSF elastomers. (f) Practical applications of HSF-SA pressure sensors in human motion monitoring. Reproduced with permission [283]. Copyright 2022, Wiley-VCH. (g) NM-based skin-like pressure sensors. Reproduced with permission. [284] Copyright 2018, Wiley-VCH. (h) Ionic capacitive pressure sensor used for measuring radial artery pulse and comparison of measured BP with that obtained from a cuff BP monitor. Reproduced with permission. [285] Copyright 2023, American Chemical Society.

pressures, enabling real-time monitoring of rapid and significant BP fluctuations. This capability is crucial for preventing acute cardiovascular events (Fig. 12c). Li et al. [283] synthesized a capacitive sensor based on hydrophobic polyacrylamide/silk fibroin by doping polyacrylamide hydrogel with silk fibroin (SF), resulting in a dual-network

skin elastomer (Fig. 12d, e). Compared to polyacrylamide hydrogel, HSF exhibited significantly improved stretchability, resilience, and tear resistance. Additionally, the device achieved a sensitivity of 0.4 $\text{kPa}\cdot\text{N}^{-1}$ and a low-pressure detection limit of 12 Pa, making it suitable for monitoring human movements, such as bending and releasing of

limbs (fingers, wrists, elbows), swallowing, vocalization, and wrist pulses (Fig. 12f). Su et al. [56] reported an expandable ionic capacitive pressure sensor based on a double electric layer (EDL) interface capacitance structure with pyramid-shaped microstructures. The overall sensor consists mainly of an EDL formed at the interface between the pyramid apex and the top electrode. The capacitance remained nearly constant under planar stretching conditions. By introducing the pyramid-shaped microstructures into the EDL, the sensor exhibited 98% strain insensitivity, up to 50% strain capability, and a low-pressure detection limit of 0.2 Pa. The fabricated soft robotic finger demonstrated the ability to monitor mechanical physiological signals from the skin.

In addition to contact challenges, the breathability between the device and the skin is also crucial [286]. Paper, composed of randomly interconnected cellulose fibers, has a natural porous structure and is widely used as a breathable substrate [287,288]. However, the rough surface of paper often requires mineral fillers that reduce its flexibility and increase its density [289]. Yang et al. [284] reported a breathable and screen-printed capacitive pressure sensor for electronic skin, composed of PVDF/AgNWs electrodes and a TPU dielectric layer, offering excellent sensitivity, fast response time (<26 ms), ultra-low detection limit (1.6 Pa), and breathability, for monitoring physiological signals like arrhythmias (Fig. 12g). Zhou et al. [290] proposed a flexible liquid metal-based capacitive microfluidic pressure sensor. The core principle of this sensor involves measuring pressure through capacitance changes induced by the movement of the interface between two immiscible liquids within a microchannel. This innovative method enables direct, real-time pressure measurement on a micro-scale. The detection channel and liquid metal electrodes can be fabricated in tandem with the working microchannel at any target location, greatly facilitating miniaturization and integration. In a complementary development, Cho et al. [285] introduced a novel and cost-effective manufacturing process for ionic liquid (IL)/polymer composite materials featuring microstructures. By employing IL/polymer composites as the dielectric layer, they developed an ionic capacitive pressure sensor characterized by ultra-high linear sensitivity and an extended operational range (Fig. 12h). This sensor effectively monitors pulse at the wrist and neck, and estimates BP based on pulse data, underscoring its feasibility as a healthcare monitoring device.

3.6. Piezoresistive sensors for high-resolution cardiovascular signal detection

Piezoresistive sensors have emerged as a promising solution for cardiovascular monitoring due to their high sensitivity, simple readout, and compatibility with flexible platforms. Current research has advanced along two major paths: microstructural engineering to enhance signal resolution and range, and the development of soft, biocompatible materials such as hydrogels to improve mechanical compliance and wearability.

To date, the performance of devices, including both surface and active layers, has been continually studied and optimized through structural redesign and raw material selection [291,292]. Researchers have attempted to use three-dimensional compressible structures with rough surfaces as active layers [293], achieving a broad measurement range but compromising adhesion to the substrate. Li et al. [294] reported an ultra-thin, high-performance piezoresistive sensor with layered nanonetwork piezoresistive materials and nanonetwork electrodes, offering potential applications in wearable electronics for arrhythmia monitoring. Previous research predominantly focused on increasing sensor sensitivity, with less attention given to the operational range and the linear relationship between sensor output and applied pressure [295,296]. Bae et al. [297] proposed a high-performance piezoresistive pressure sensor device (Fig. 13a). This device, composed of a layered structure array forming a pressure-sensitive electronic skin, utilizes single-layer graphene as the electrode material, allowing it to

operate at low voltages. The device exhibits a linear relationship between applied pressure and output, capturing radial artery pressure waves composed of three components: the incident wave from the heartbeat, the reflected wave from the hand, and the reflected wave from the lower body (Fig. 13b). Such data can be used to determine the resistance index (RI) and SI, which are related to arterial stiffness. Park et al. [298] developed a flexible piezoresistive pressure sensor based on wrinkled carbon nanotube thin films, utilizing a thermally induced shrinkage process to create the wrinkled structures (Fig. 13c). This sensor significantly enhances pressure sensitivity and exhibits rapid response capabilities, making it suitable for atherosclerosis monitoring. By tracking arterial pulse waveforms, the sensor provides precise blood flow pulsation data. Moreover, Yang et al. [299] designed a hierarchical microstructure biomimetic flexible piezoresistive sensor. This sensor comprises a layered polyaniline/polyvinylidene fluoride nanofiber film sandwiched between interlocking electrodes with microdome structures. It has a pressure detection range of 58.4–960 Pa and exhibits excellent cyclic stability over 50,000 cycles. Additionally, this conformal skin-adhered sensor successfully demonstrated the monitoring of physiological signals and motion states in humans.

Array microstructures are widely employed in flexible pressure sensors to enhance performance, with various forms developed, including micropillar arrays [303], microcone arrays [304], micro-needle arrays [305], and microsphere arrays [306]. Shen et al. [307] demonstrated a capacitive-piezoresistive hybrid flexible pressure sensor based on a conductive micropillar array. The piezoresistive mode is mainly used for higher precision and sampling rates, significantly simplifying IC design for wearable applications. This sensor can detect finer digital arterial pulse signals, illustrating its high sensitivity. The improved, simple, and highly adjustable manufacturing process, along with the flexible hybrid sensing strategy, will benefit the unified design, mass production, quantifiable optimization, and functional diversity of wearable/implantable bioelectronics, playing a critical role in arteriosclerosis monitoring. It is well-known that many piezoresistive pressure sensors exhibit high sensitivity to temperature fluctuations [308]. To address this challenge, Park et al. [300] employed an active compensation strategy by integrating four piezoresistive sensors into a full Wheatstone bridge configuration. This was achieved using a multilayer 3D microsystem design (Fig. 13d). Integrating this 3D pressure sensor into a soft, flexible electronic platform produces a system capable of real-time wireless measurements from the skin's surface. Placing the device above the radial and carotid arteries yields high-quality waveforms related to pulsatile blood flow, which quantitatively correlate with blood pressure (Fig. 13e).

Among various hydrogel-based sensors, piezoresistive hydrogels have attracted attention for their mechanical compliance, biocompatibility, and direct signal transduction capabilities in cardiovascular monitoring [309]. Yao et al. [301] designed an ultrasensitive and breathable hydrogel fiber-based strain sensor by introducing a localized crack into a dual-network hydrogel structure (Fig. 13f). The engineered crack served as a strain-amplifying site, significantly enhancing gauge factor (up to 3930) while maintaining mechanical robustness and rapid response (78 ms). When applied to the wrist, the sensor captured high-fidelity pulse waveforms, including P, T and D waves, and enabled extraction of hemodynamic parameters such as AIx and the time delay of the reflected pressure wave (ΔT_{DVP}), demonstrating potential for real-time cardiovascular assessment. Wang et al. [302] developed a high-performance hydrogel sensing system by integrating a porous polyacrylamide (PAM) pressure sensor with a conductive tannic acid-modified hydrogel electrode (Fig. 13g). The system combined excellent conformability, skin adhesion, and mechanical stretchability with a low detection limit (16 Pa) and high pressure sensitivity (0.95 kPa^{-1}). Beyond its use in gesture recognition and robotic rehabilitation, the sensor reliably detected subtle cardiovascular signals such as wrist pulse and eye blinks, suggesting its viability for multimodal, noninvasive health monitoring. Further advancing the field, Cohen-Gerassi et al.

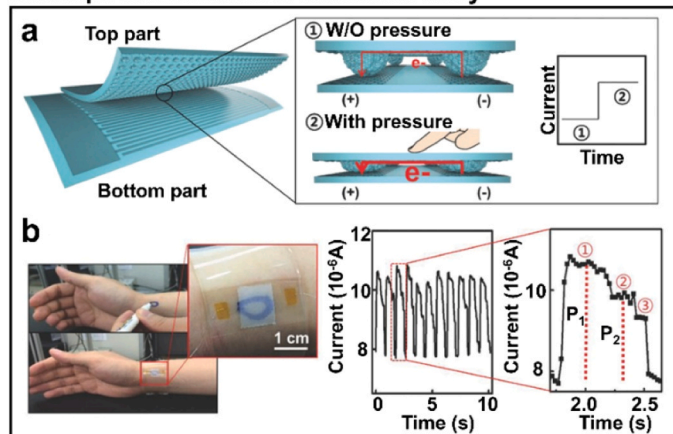
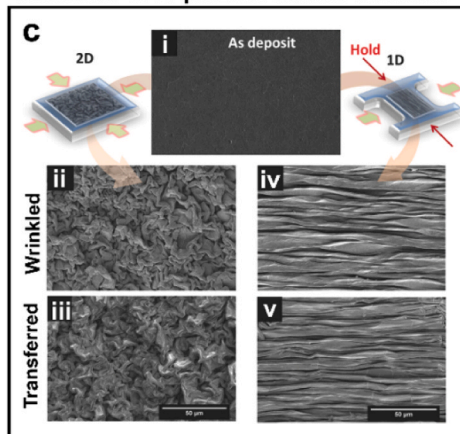
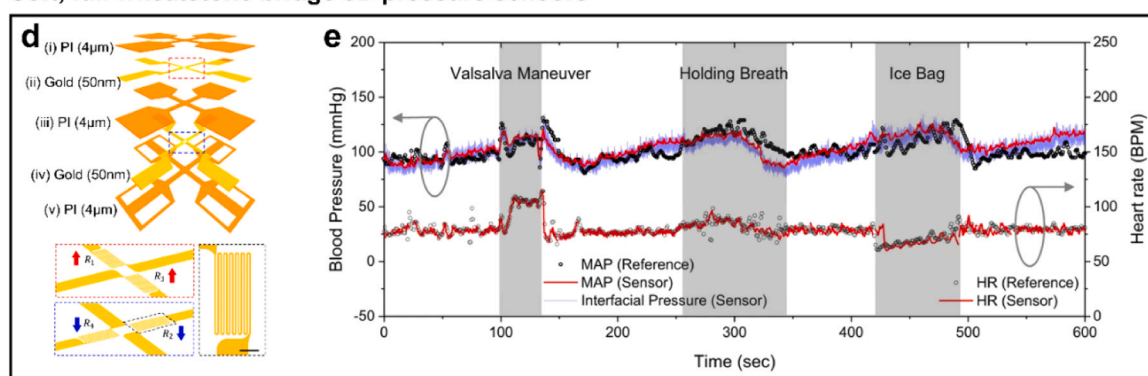
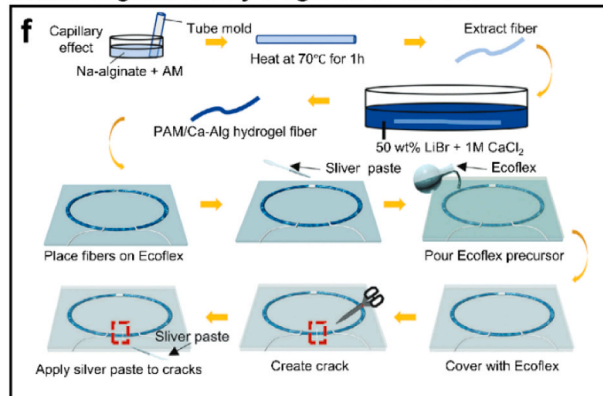
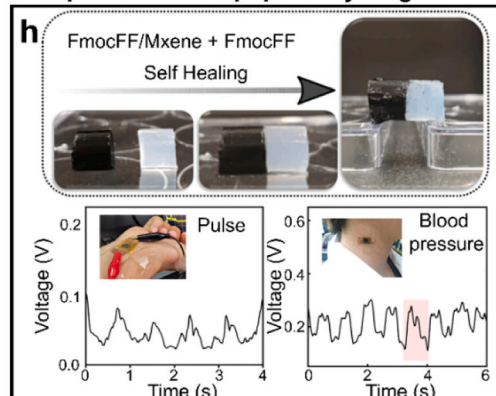
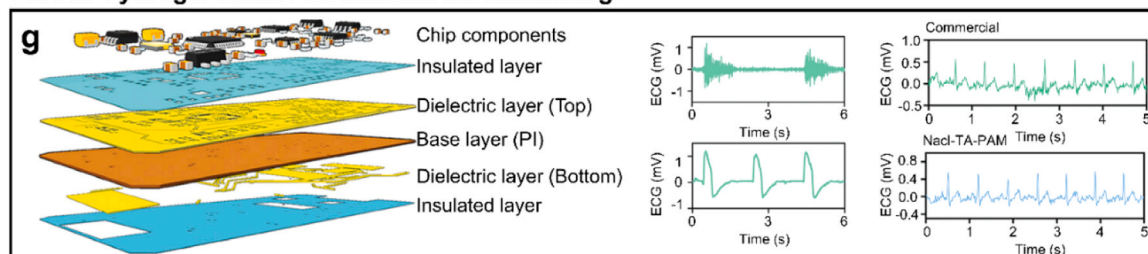
Bioinspired hierarchical structural array**Wrinkled CNT piezoresistive sensors****Soft, full wheatstone bridge 3D pressure sensors****Crack-engineered hydrogel fiber sensor****Composite MXene/peptide hydrogels****Porous hydrogel sensor for multimodal monitoring**

Fig. 13. Engineering strategies improve the precision and robustness of piezoresistive cardiovascular sensors. (a-b) Pressure sensor composed of a multistage graphene/PDMS array and the measurement data of radial artery physical force. Reproduced with permission. [297] Copyright 2016, Wiley-VCH. (c) SEM images illustrating wrinkled topography. Reproduced with permission. [298] Copyright 2018, Wiley-VCH. (d-e) Continuous measurement of blood pressure and heart rate using 3D piezoresistive sensors. Reproduced with permission. [300] Copyright 2024, Springer Nature. (f) Schematic illustrating the manufacturing process of the CLC sensor. Reproduced with permission [301]. Copyright 2024, Wiley-VCH. (g) Schematic of the multilayer structure of the self-designed flexible printed circuit board (FPCB). Reproduced with permission [302]. Copyright 2023, Wiley-VCH. (h) Demonstration of the self-healing property. Two separate blocks of Fmoc-FF and MXene/Fmoc-FF hydrogels healed after being kept in contact for 5 min. Reproduced with permission [119]. Copyright 2024, Wiley-VCH.

[119] designed a composite MXene/peptide hydrogel piezoresistive sensor based on Fmoc-FF peptide hydrogel, leveraging recent advancements in materials science and functional nanomaterials (Fig. 13h). To enhance peptide-MXene interactions, two peptide derivatives were studied, examining the effects of adding positively and negatively charged amino acids, lysine and aspartic acid, on self-assembly, self-healing, mechanical, and electrical properties. This sensor holds potential for various biomedical applications, such as wearable skin patches and integrated artificial skin capable of measuring pulse and blood pressure. Huang et al. [310] prepared a conductive aerogel with a 3D network structure using freeze-drying, exhibiting excellent mechanical and electrical properties (flexibility, elasticity, compressibility, and conductivity), constructing multifunctional flexible/wearable devices, including piezoresistive sensors for cardiac rhythm monitoring.

While most piezoresistive hydrogel sensors have been developed for skin-mounted or wearable applications, recent studies have extended their use to implantable systems for deeper, long-term physiological monitoring. Wu et al. [311] developed a compressible and biocompatible piezoresistive hydrogel sensor based on HEC/SPI/PANI, which maintained stable performance over 4 weeks of *in vivo* implantation. The sensor enabled real-time monitoring of physiological signals such as pulse and respiration, and successfully tracked sciatic nerve recovery in rats, demonstrating its potential for implantable cardiovascular and neuromuscular monitoring. Zhang et al. [312] developed a highly stretchable and biocompatible ionic-conductive hydrogel reinforced with nanoclay sheets and engineered with a surface wrinkled microstructure. This implantable sensor exhibited ultrafast response (50 ms), long-term durability (>5000 cycles), and excellent protein resistance, minimizing foreign body response. Notably, it enabled real-time *in vivo* detection of myocardial infarction (MI) in rats and even localized infarcted regions on the cardiac surface, offering a promising strategy for precise cardiac disease diagnosis. Together, advances in structural design and hydrogel-based materials have greatly broadened the application range of piezoresistive sensors, enabling their use from wearable pulse monitors to implantable devices capable of *in vivo* diagnosis. These developments lay the foundation for next-generation cardiovascular interfaces that are soft, precise, and adaptable to clinical needs.

3.7. Bioimpedance sensors for multifunctional and deep-tissue cardiovascular monitoring

Recent developments in bioimpedance technology have demonstrated its potential as a noninvasive approach for comprehensive cardiovascular monitoring. By measuring tissue-dependent electrical impedance, this technique allows access not only to deep-tissue physiological signals but also to circulating biomarkers relevant to cardiovascular inflammation, thrombosis, and vascular remodeling. Recent studies have shown its utility in detecting CRP, D-dimer, and microparticles using label-free electrochemical interfaces, as well as its integration into implantable systems for *in situ* vascular assessment. In parallel, advances in electrode design and signal processing have enabled wearable bioimpedance platforms to support continuous monitoring of BP, HR, and multi-parameter signals. These systems increasingly incorporate ML algorithms and physics-informed modeling to enhance signal interpretation, reduce calibration demands, and support early-stage cardiovascular risk assessment.

3.7.1. Noninvasive deep-tissue sensing with bioimpedance interfaces

Recent advances have extended bioimpedance sensing beyond macroscopic applications toward the noninvasive detection of molecular biomarkers. Studies have demonstrated its feasibility in monitoring inflammatory and pathological states, underscoring its potential for broader clinical diagnostics [313]. This molecular-level sensing capability lays the foundation for tracking cardiovascular-related biomarkers, including inflammation and thrombosis markers, via label-free

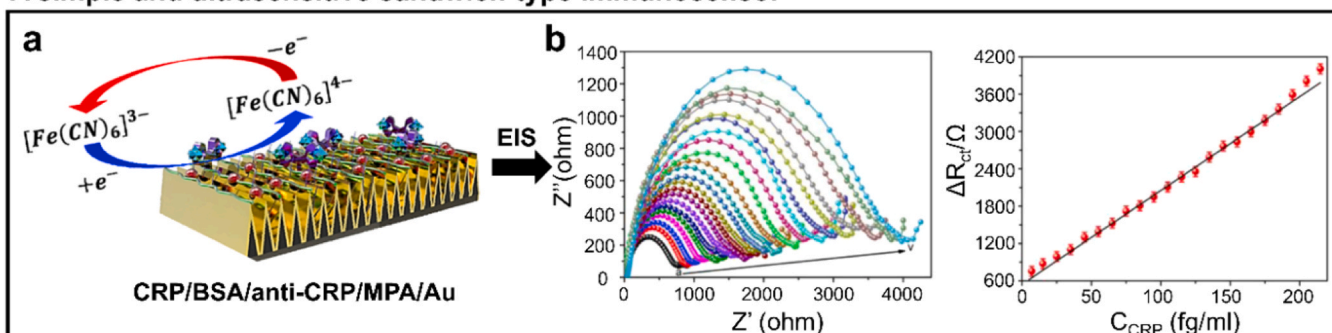
electrochemical impedance strategies.

Among established markers, CRP remains a central indicator of cardiovascular inflammation [314]. In a large-scale, multicenter clinical study, Tsigos et al. [315] explored the association between dual-frequency bioimpedance analysis and chronic stress–inflammation biomarkers in over 99,000 subjects. This early integration of impedance variables with circulating CRP established a noninvasive framework for identifying patients under chronic inflammatory load, laying foundational evidence for subsequent biomarker-oriented cardiovascular applications. Building on this framework, Vilian et al. [316] developed a high-selectivity, label-free, and non-invasive ultra-sensitive electrochemical immunosensor platform based on BSA/anti-CRP/MPA/Au substrates, demonstrating consistent impedance response across a concentration range of 7–215 fg/ML with good linearity, ideal for clinically relevant CRP levels (Fig. 14a, b). Beyond single-analyte detection, multiplexed sensing has emerged to address the multifactorial nature of thrombotic events, Wang et al. [317] proposed a nanostructured electrochemical immunosensing platform capable of simultaneously profiling multiple thrombosis-related biomarkers, including CRP, calprotectin, sP-selectin, and D-dimer. By integrating laser-engraved carbon nanotube electrodes with electrodeposited gold nanoparticles, they constructed a multi-channel impedance sensor array with high sensitivity and specificity. Notably, an unsupervised clustering algorithm was applied to the impedance signatures to stratify patient thrombotic risk, achieving an AUC of 0.95 across a clinical cohort of COVID-19 patients. This approach highlights the potential of combining multiplexed bioimpedance sensing with ML to enable rapid, noninvasive, and personalized cardiovascular risk assessment.

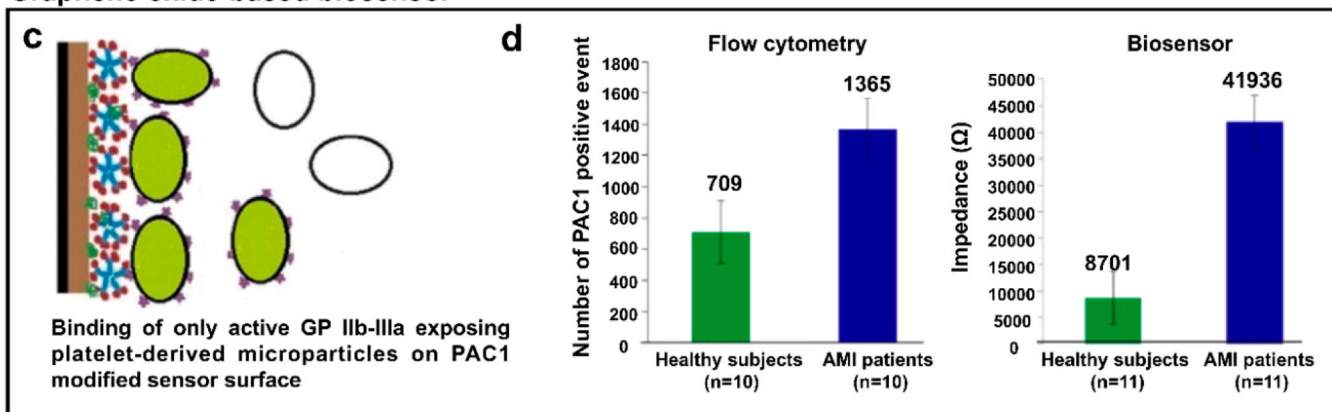
At the cellular level, circulating microparticles have gained traction as dynamic markers of vascular injury. Kailashiya et al. [318] developed methods to detect circulating platelet-derived microparticles (PMP) in individual blood samples, with impedance never exceeding 15000Ω in 80% of healthy individuals, while most acute myocardial infarction (AMI) patients showed higher values, indicating more circulating PMP as an AMI biomarker [321,322]. GO-manufactured biosensors successfully monitored more PMP in AMI patients' blood, demonstrating high specificity and sensitivity for early screening of "high-risk" individuals (Fig. 14c, d). Zheng et al. [323] conducted a clinical study showing that specific Microparticles (MP), such as those expressing ICAM-1 and E-selectin, are inversely associated with physical activity levels in obese adults. These findings support the use of EMPs as noninvasive biomarkers for early vascular inflammation and cardiovascular risk assessment. In addition to circulating biomarkers, tissue-embedded impedance sensors have been explored for *in situ* vascular pathology assessment, Hoare et al. [319] developed an implantable micro-sensor and telemetry system capable of distinguishing different cell types associated with arterial stenosis, including differentiating between blood and clots (Fig. 14e). Through impedance analysis of human carotid endarterectomy specimens, it was found that lipid cores exhibited similar impedance magnitude and characteristics across the spectrum. For moderate disease, impedance in the 10–100 KHz range was slightly higher compared to other conditions. Calcified caps showed increased impedance compared to lipid cores, with more significant differences in magnitude at lower frequencies as the disease severity increased. Bussoo et al. [320] developed a cell sensing and apoptosis system based on interdigitated electrodes (IDE) (Fig. 14f). The results demonstrated that the device is scalable, and miniaturizing the IDE improved detection sensitivity. Continuous impedance measurements at 10 kHz were used to monitor the apoptosis of vascular smooth muscle cells. When combined with vascular devices, this system effectively monitors arterial diseases.

Collectively, these advances illustrate the expanding role of bioimpedance technology in cardiovascular diagnostics, ranging from the detection of circulating biomarkers such as CRP, D-dimer, and microparticles to the real-time assessment of vascular pathology through implantable systems. By combining molecular sensitivity, cellular

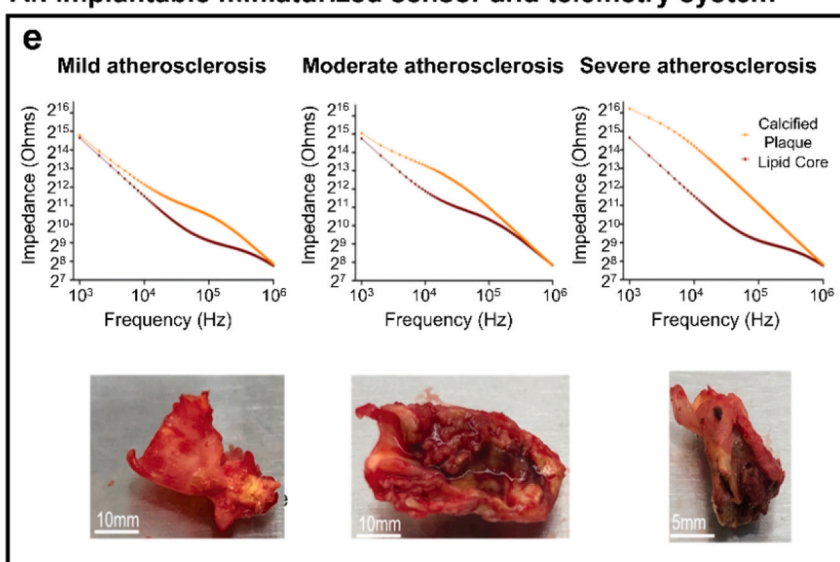
A simple and ultrasensitive sandwich-type immunosensor



Graphene oxide-based biosensor



An implantable miniaturized sensor and telemetry system



A cell sensing and apoptosis system

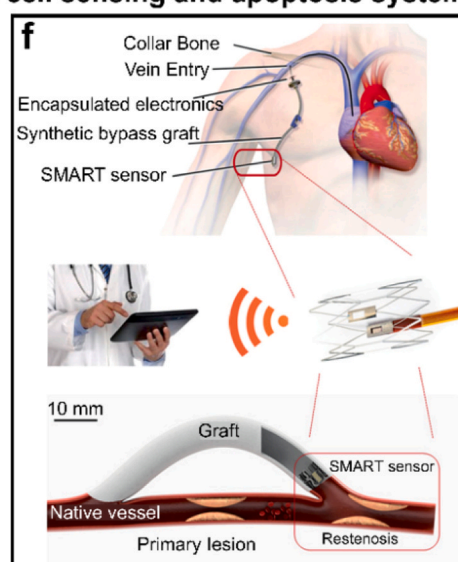


Fig. 14. Bioimpedance sensing advances noninvasive monitoring of cardiovascular biomarkers. (a-b) EIS response and calibration curve of CRP immunosensor for electrochemical immunoassay. Reproduced with permission. [316] Copyright 2019, Elsevier. (c-d) Detection of PMP in healthy subjects and AMI patients using flow cytometry and biosensors. Reproduced with permission [318]. Copyright 2015, Elsevier. (e) Concept of a smart stent with multiple impedance sensors, and impedance analysis of lipid core versus calcified cap with corresponding plaques. Reproduced with permission. [319] Copyright 2022, Wiley-VCH. (f) Sensors integrated into implantable medical devices. Reproduced with permission. [320] Copyright 2020, Wiley-VCH.

specificity, and deep-tissue accessibility, bioimpedance sensing provides a unified and noninvasive framework for early risk evaluation, disease monitoring, and long-term cardiovascular management.

3.7.2. Intelligent bioimpedance platforms for wearable, multiparametric cardiovascular monitoring

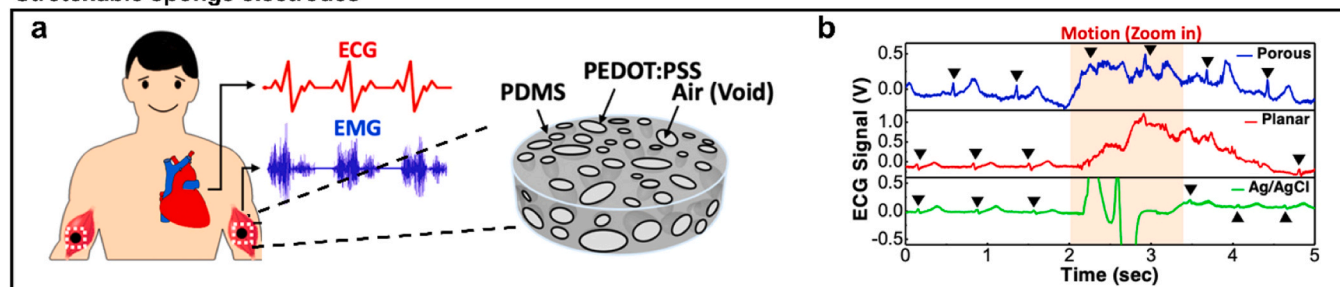
Achieving reliable and high-quality bioimpedance signals begins with improving the electrode-skin interface. While traditional rigid

electrodes often suffer from poor conformity and high contact impedance, recent studies have turned to soft, epidermal materials designed to reduce interfacial impedance and motion artifacts during body movement [324–326]. Although microneedle-based designs offer better coupling, their invasive nature raises biosafety concerns. Addressing these challenges, Lo et al. [327] introduced elastic sponge electrodes fabricated from porous PDMS templated by sacrificial sugar and coated with PEDOT. This architecture significantly enhanced the effective

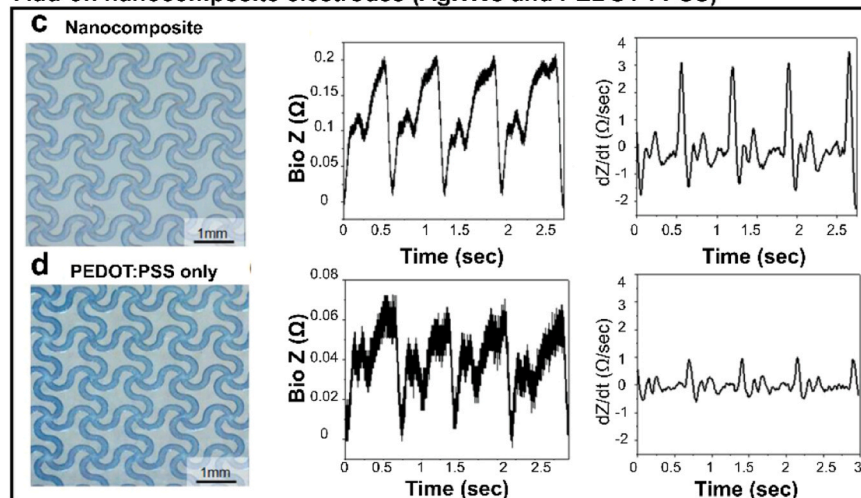
contact area, reducing skin-electrode impedance by up to 6.7-fold and boosting SNR in ECG and electromyograms (EMG) acquisition—crucial for arrhythmia diagnosis (Fig. 15 a,b). In electrophysiological recordings, electrode design is crucial for obtaining high-quality signals and minimizing motion artifacts. Soft dry electrodes (such as nanowires, graphene, and conductive polymers) are more suitable but are prone to mechanical damage [328–330]. Kong et al. [331] addressed the challenge of connecting soft electrodes with rigid data acquisition circuits by employing an additional soft electronic interface for bioimpedance monitoring of cardiac signals, including heart rate and cuffless blood pressure. The bioimpedance collected using additional nanocomposite electrodes displayed a SNR of 37.0 dB, significantly higher than the 25.9 dB obtained with standard Ag/AgCl gel electrodes, enabling high-precision continuous blood pressure monitoring (Fig. 15 c,d).

To translate bioimpedance signals into clinically meaningful parameters, recent studies have increasingly incorporated artificial intelligence frameworks to enhance signal interpretation, personalize cardiovascular modeling, and reduce calibration demands [334]. Sel et al. [58] designed a ring-shaped bioimpedance sensor, investigating the optimal design parameters for electrode position and size (Fig. 15e). By leveraging the deep tissue sensing capabilities of bioimpedance, the sensor achieves high sensitivity to arterial volume changes without differentiating between different skin tones. ML algorithms were employed to capture unique features from bioimpedance signals and build a BP estimation model, demonstrating the framework's potential to enhance traditional diagnostic tools used in medical practice (Fig. 15f). To further reduce data requirements, they introduced a physics-informed neural network (PINN) capable of reconstructing

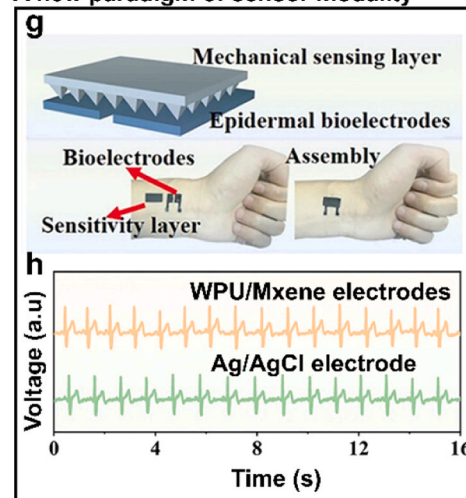
Stretchable sponge electrodes



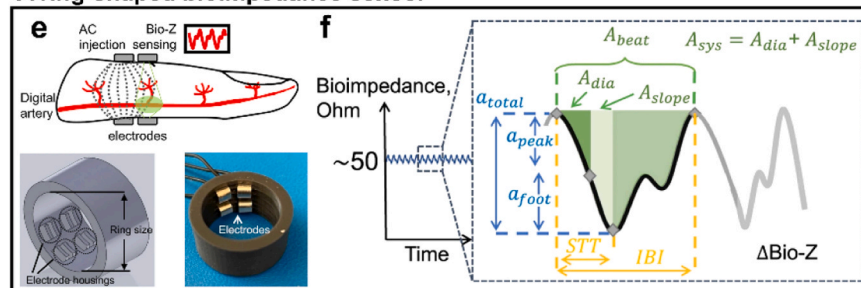
Add-on nanocomposite electrodes (AgNWs and PEDOT : PSS)



A new paradigm of sensor modality



A ring-shaped bioimpedance sensor



MXene hydrogel-based epidermic sensors

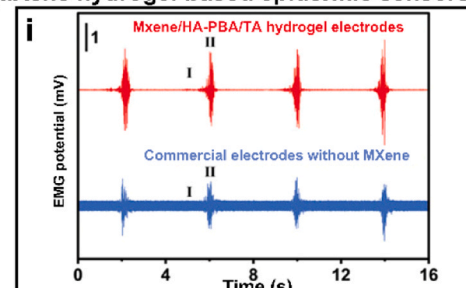


Fig. 15. Bioimpedance technologies drive intelligent and multifunctional cardiovascular health platforms. (a–b) Schematic illustration of ECG and EMG recording using PEDOT/PDMS sponge electrodes, displaying ECG peaks and motion artifacts. Reproduced with permission. [327] Copyright 2022, American Chemical Society. (c–d) Comparison of bioimpedance measurements using nanocomposite and PEDOT electrodes. Reproduced with permission. [331] Copyright 2023, Wiley-VCH. (e–f) Cross-sectional bioimpedance sensing using four electrodes, and the definition of corresponding bioimpedance features used in blood pressure correlation analysis. Reproduced with permission [58]. Copyright 2023, Springer Nature. (g) Schematic and digital images of the sensors. (h) ECG signals derived from printed WPU/MXene-30% epidermal bioelectrodes and Ag/AgCl gel electrodes. Reproduced with permission [332]. Copyright 2023, Wiley-VCH. (i) EMG signals measured by the MXene/HA-PBA/TA hydrogel electrode and commercial hydrogel electrode without MXene ((I) initial, (II) after clenching right fist) respectively. Reproduced with permission [333]. Copyright 2024, Wiley-VCH.

cardiovascular dynamics using minimal ground-truth input [335]. This model extracts complex cardiovascular information using minimal ground-truth data. When tested on the same dataset with state-of-the-art time-series models, the PINN retained high correlation and low error, reducing the amount of ground-truth training data needed by an average of 15 times. This advancement is pivotal for the future development of artificial intelligence algorithms. Furthermore, Ibrahim et al. [336] proposed a cuffless BP monitoring method based on a novel bio-impedance sensor array. This sensor array is embedded in a compact, flexible wristband that provides robust blood pulsation sensing and BP estimation without requiring calibration of the sensing position. Wang et al. [337] introduced the research and development of a single-channel wrist-worn impedance plethysmography (IPG) system for wearable continuous blood pressure measurement. A two-step calibration BP mathematical model based on IPG was derived for single-session physiological signal acquisition. High-resolution measurements were achieved by optimizing the injection current frequency and electrode arrangement, targeting the significant impedance differences between the SBP and DBP feature points in the IPG waveform.

As bioimpedance systems evolve, multifunctionality and compactness have become key considerations. One notable strategy is the co-integration of mechanical and electrical sensing within a unified platform [338,339]. Balancing the multifunctionality and simplicity of wearable sensors presents a challenge. Song et al. [332] proposed a new paradigm for mechanical and bioelectrical sensing modalities based on a single material system and a simple solution processing method. The multifunctional sensor consists of a pair of highly conductive ultrathin electrodes (WPU/MXene-1) and an elastic microstructured mechanical sensing layer (WPU/MXene-2) (Fig. 15g). This configuration provides high-pressure sensitivity and low skin-electrode interface impedance, enabling simultaneous monitoring of physiological pressures (such as arterial pulse signals) and epidermal bioelectrical signals, including ECG and EMG (Fig. 15h). Moreover, Wang et al. [333] developed sensors based on adhesive MXene/HA-PBA/TA hydrogels that conformally attach to the chest and wrist for real-time monitoring of ECG and radial artery pulses, displaying normal arterial waveforms (Fig. 15i). The two characteristic peaks of the radial artery pulse are distinctly identifiable, which is crucial for diagnosing arterial stiffness and vascular aging. To enhance long-term wearability and convenience, Sel et al. [340] introduced ImpediBands, compact distributed intelligent bioimpedance (Bio-Z) patches, where communication between patches is established through the body without the need for wires, creating common potential points between sensors. The team also proposed a blind source separation technique based on second-order blind identification for signal reconstruction [341], extracting cardiopulmonary activity from Bio-Z signals for early diagnosis of cardiac diseases. Furthermore, they explored non-invasive bioimpedance patterns across different body areas. Results demonstrated that bioimpedance effectively monitors cardiac and respiratory activities in the limbs and upper body. Therefore, this mode shows strong potential for adoption in various wearable devices (such as smartwatches, smart bands, and smart patches) to meet the needs of early heart disease diagnosis.

Collectively, these advances illustrate how bioimpedance sensing is evolving into a multifunctional cardiovascular monitoring modality. From molecular-level detection of inflammatory and thrombotic biomarkers to AI-assisted wearable systems for continuous hemodynamic tracking, the field is moving toward integrated, minimally invasive solutions that span molecular diagnostics, physiological monitoring, and personalized analytics. This convergence of capabilities positions bioimpedance sensing as a practical and scalable interface for future cardiovascular healthcare delivery.

Moreover, recent regulatory reviews have highlighted the importance of intended use, safety, data integrity, and cybersecurity, which are key criteria defined by agencies like the FDA and supported by international standards such as ISO 14971 and ISO 14155 [342]. Considering these regulatory aspects early in device development is

essential to accelerate clinical adoption and ensure patient safety.

Regulatory approval for emerging cardiovascular sensing technologies remains constrained by several structural challenges. A major limitation arises from the absence of standardized evaluation protocols for novel sensing modalities, which complicates cross-platform validation and comparison. In parallel, AI-integrated systems face regulatory uncertainty regarding algorithm transparency, reproducibility, and adaptive learning capabilities. The evolving guidance on software as a medical device (SaMD), including the FDA's AI/ML Action Plan, highlights the need for frameworks that can accommodate both static and continuously learning models. Moreover, robust clinical validation often relies on large-scale, demographically diverse datasets; however, access to such data is frequently limited by privacy regulations and institutional silos. Compounding these issues, the deployment of connected and wearable devices introduces data governance complexities, particularly under regulatory frameworks such as HIPAA and GDPR. Existing approval pathways, including the FDA's 510(k) and De Novo mechanisms, are not always compatible with the rapid iteration cycles characteristic of digital health technologies. Addressing these interconnected barriers will require coordinated efforts across industry, healthcare systems, and regulatory agencies to modernize approval processes while ensuring safety, transparency, and clinical relevance.

3.8. Comparative evaluation

Evaluating and comparing performance metrics of various monitoring devices is critical in developing cardiovascular health monitoring technologies, as these metrics directly impact the practicality and reliability of monitoring results. Materials must not only possess excellent physical and chemical properties but also meet the high standards for biomedical applications. The main performance metrics (sensitivity, accuracy, stability, and response speed) are discussed in detail and compared across different monitoring technologies (Table 1).

Sensitivity is determined by applying controlled physical stimuli, such as calibrated pressure or mechanical strain, using devices like linear actuators or pressure controllers, and measuring the resulting change in electrical output (voltage, current, or charge) per unit input [345]. Accuracy is assessed by comparing sensor outputs to those obtained from clinical-grade reference instruments, such as sphygmomanometers or Holter monitors, under standardized physiological conditions, with performance quantified by the mean absolute error and standard deviation across multiple measurements [346]. Stability is evaluated through repeated measurements over time, often spanning hundreds to thousands of cycles, or through continuous operation under dynamic conditions, to monitor signal drift, baseline shifts, or output degradation. Response speed is measured by applying a step-function stimulus, such as a rapid pressure pulse or mechanical tapping, and recording the time required for the signal to reach a defined proportion (commonly 90%) of its peak value, as captured by high-speed data acquisition systems [347].

Although ECG and Holter monitoring remain the clinical gold standards for cardiovascular assessment due to their diagnostic reliability and regulatory maturity, emerging sensing mechanisms offer complementary advantages in specific contexts [348] (Table 2). Piezoelectric and triboelectric sensors offer a distinct advantage in their ability to operate without external power sources, a capability rooted in the intrinsic properties of their active materials. This energy-autonomous behavior makes them particularly well suited for long-term, continuous monitoring in wearable and implantable cardiovascular systems, where dynamic signals such as pulse waves and cardiac vibrations must be captured with high temporal fidelity and minimal energy burden [349]. Optical sensors are especially well suited for noninvasive, multimodal, home-based cardiovascular monitoring. Their unique ability to detect oxygen saturation, which is inaccessible to other sensing modalities, along with compatibility with biochemical and hemodynamic signal integration, makes them ideal for comprehensive

Table 1

Flexible sensors for monitoring CVD-related parameters: materials, working principles, and performance.

Mechanism	Materials	Structure	Sensitivity	Response time	Stability
Piezoelectric	FEP/Ecoflex/FEP	Sandwich-structure piezoelectret [186]	4100 pC N ⁻¹ 32.6 nA kPa ⁻¹	18.6 ms	Number of cycles > 3600 > 30 days
	BN/PT & MXene/PT	Heterogeneously hierarchical piezoelectric composite [19]	41.67 pC N ⁻¹ 39.3 mV kPa ⁻¹	30.1 ms	Number of cycles > 1000 > 50,000
	PVDF	All-nanofiber-based mechanoacoustic sensor [184]	10050.6 mV Pa ⁻¹	-	Number of cycles > 1000 > 50,000
	PVDF	Piezoelectric film [195]	173 mV·mmHg ⁻¹	51 ms	Number of cycles > 50,000
	PZT	Piezoelectric film [194]	0.062 kPa ⁻¹	23 ms	Number of cycles > 50,000
	PZT	E-skin [183]	1.36 μA Pa ⁻¹	~0.1ms	Number of cycles > 1000 > 75,000
	Peptide	E-skin [181]	27 mV·N ⁻¹	4.9 ms	Number of cycles > 75,000
	AlN	Flexible smart patch [196]	0.012 V·Pa ⁻¹ ·m ⁻²	0.84 μs	-
	P(VDF-TrFE)	Small miniaturized force sensor [197]	0.63 μV·Pa ⁻¹	50 ms	Number of cycles > 400,000
	Glycine/PVA PLLA/glycine	Localized stress concentration [343] Piezoelectric film [344]	1.13 V·kPa ⁻¹ 13.2 mV·kPa ⁻¹	- 10 ms	- Number of cycles > 4000
Triboelectric	PTFE & Au	Self-powered endocardial pressure sensor [227]	1.2 mV·mmHg ⁻¹	-	Number of cycles > 100 million
	FEP & CNTs	Textile triboelectric pulse sensor [59]	0.21 μA·kPa ⁻¹	40 ms	Number of cycles > 10,000
	CNTs & PDMS	A triboelectric pulse sensor based on biomimetic nanopillar layer [212]	0.14 nA·kPa ⁻¹	7.99 ms	Number of cycles > 4500
	PCL	Self-powered textile pulse sensor [229]	37 V·kPa ⁻¹	-	Number of cycles > 50,000
	poly(lactic acid)- (chitosan 4%)	Implantable bioresorbable self-powered sensor [213]	11 mV·mmHg ⁻¹	0.5 ms	Number of cycles > 450,000
	Polyester & Nylon yarn & Ag	Textile-based sensor [214]	3.88 V·kPa ⁻¹	-	Number of cycles > 80,000
Optical	OPT & NIR	Multimodal and wireless skin-interfaced system [240]	3.5 × 10 ⁵ A·W ⁻¹	-	Number of cycles > 100
	LED & PD GaAs & Al ₂ O ₃	Flexible Skin Patches [234] Skin-like systems [247]	- ± 5 mmHg	- -	> 72 h Number of cycles > 300
Magnetoelastic	NdFeB/SiO ₂	Textile magnetoelastic generator [260]	Low detection limit: 0.05 kPa	-	Number of cycles > 20,000
	NdFeB/SiO ₂	Ultrastretchable thin-film [21]	4.16 × 10 ⁻⁸ T·Pa ⁻¹	3 ms	Number of cycles > 12,000
	NdFeB SiO ₂	Soft magnetoelastic generator [55] Magnetoelastic mesh-integrated stent [261]	7.19 × 10 ⁻⁸ T·Pa ⁻¹ -	- Real-time response	- -
Capacitive	NdFeB/PDMS	Millimeter-scale magnetic implant [262]	-	-	> 3 months
	PEN & Cr/Au & PVA	Highly Skin-Conformal Microhairy Sensor [281]	0.55–0.58 kPa ⁻¹	< 0.1s	Number of cycles > 3000
	PVDF & Ag & polyurethane GaN	A Breathable and Screen-Printed Pressure Sensor [284] E-skin [38]	4.2 kPa ⁻¹ 10.06 A W ⁻¹	< 26 ms -	- Number of cycles > 3000
	SS/Au & PI POMaC & PLLA & PHB/ PHV	Wireless soft sensors [272] Wireless soft sensors [275]	0.013 kPa ⁻¹ -	- -	> 3 months > 12 weeks

(continued on next page)

Table 1 (continued)

Mechanism	Materials	Structure	Sensitivity	Response time	Stability
Piezoresistive	PS & Au	Soft capacitive pressure sensors [282]	0.148 kPa^{-1} ; $-0.054 \pm 2.09 \text{ mmHg}$	< 10 ms	Number of cycles > 5000
	Liquid metal	Liquid metal capacitive pressure sensor [290]	$\pm 7.5 \text{ mmHg}$	< 0.2 ms	-
	Liquid metal	An ionic capacitive pressure sensor [285]	56.91 kPa^{-1}	60 ms	Number of cycles > 10,000
	GO/CNT & HA	Hybrid aerogel [310]	-	33 ms	Number of cycles > 20,000
	Ag NWs & GR & PANFs	E-skin [294]	134 kPa^{-1}	< 20 ms	Number of cycles > 8000
	Fmoc-FF peptide	A novel composite hydrogel [119]	38.5 kPa^{-1}	0.64 ms	Number of cycles > 20,000
	rGO & PDMS	Flexible resistive pressure sensor [117]	0.27 mmHg^{-1}	-	> 1 h
	Graphene & PDMS	Bioinspired Hierarchical Structural Array [297]	$-0.9 \pm 4.7 \text{ mmHg}$ 8.5 kPa^{-1}	40 ms	Number of cycles > 10,000
	PAM/Ca-Alg	Hydrogel [301]	-	78 ms	Number of cycles > 5000
	NaCl-TA-PAM	Hydrogel [302]	0.95 kPa^{-1}	20 ms	Number of cycles > 1000
Bioimpedance	HEC/SPI/PANI	Hydrogel [311]	-	0.14 s	Number of cycles > 100
	PAA-HEMA/PAM/SPE	Hydrogel [312]	-	50 ms	Number of cycles > 5000
	PEDOT:PSS	Soft sponge electrode [327]	-	-	< 1 h
	CRP/BSA/anti-CRP/MPA/Au	CRP sensor [316]	2.25 fg mL^{-1}	-	> 30 days
	GO & PAC1 & PMP	PMPs sensor [318]	-	-	-
	Ag	A ring-shaped bioimpedance sensor [58]	$\pm 5 \text{ mmHg}$	-	> 2 h
	AgNWs & PEDOT:PSS	Soft electronic devices [331]	$4.5 \Omega \text{ s}^{-1}$ $-2.0 \pm 6.3 \text{ mmHg}$	-	Number of cycles > 10
Capacitive & Piezoresistive	Ag	A novel Bio-Z sensor array [336]	$0.2 \pm 6.5 \text{ mmHg}$	-	> 90 min
	Ag/AgCl	A single-channel wrist-worn impedance plethysmography system [337]	$2.01 \pm 1.40 \text{ mmHg}$	-	-
	PDMS	Conductive micropillar arrays [307]	-	< 100 ms	Number of cycles > 6000
Piezoelectric & Optical	ITO & PVDF	Optoacoustic blood 'stethoscope' [185]	-	0.5 ns	Number of cycles > 1000
Triboelectric & Piezoresistive	GO & CNT	Hybrid aerogel [310]	$98.4 \text{ V (90\% strains)}$	-	Number of cycles > 20,000

physiological assessment. These advantages are largely attributed to the diverse optical transduction strategies available, enabling the detection of both volumetric blood flow and molecular signatures [350,351]. Despite these strengths, optical signals are inherently susceptible to motion artifacts, skin pigmentation, and ambient light interference, requiring advanced compensation algorithms to ensure accuracy in dynamic conditions. Magnetoelastic sensing offers a complementary, contactless approach by converting mechanical stimuli into magnetic responses that can be read wirelessly. This enables passive operation and reduces the need for direct electrical interfaces, although performance can be affected by magnetic interference and typically requires dedicated signal processing [104,352]. Capacitive sensors offer strong mechanical compliance and long-term stability, and their operating principle allows for diverse circuit-level implementations. Electrode designs such as interdigital electrodes enhance sensitivity to subtle mechanical deformations, while integration with LC resonant circuits enables wireless signal transmission without the need for active electronics [353]. These features make capacitive sensors highly adaptable

for low-power, continuous monitoring of slow physiological events such as respiration or arterial wall movement. Piezoresistive sensors are highly valued for their structural versatility and sensitivity to subtle mechanical deformation, making them well suited for capturing arterial wall expansion, pulse pressure variations, or motion-induced strain. Their performance can be significantly enhanced through microstructural engineering, such as incorporating hierarchical cracks, porous architectures, or wrinkled conductive films, which amplify resistance changes under strain [354,355]. These design strategies enable high gauge factors and allow seamless integration with soft, skin-conformal substrates for wearable applications. Additionally, their fabrication process is compatible with scalable and low-cost techniques, supporting large-area deployment [356]. Bioimpedance sensors enable the characterization of tissue composition and fluid distribution by measuring frequency-dependent conductivity. This modality is particularly advantageous for assessing deep-tissue parameters such as blood volume, hydration status, and vascular tone. Its robustness against motion and skin color variation enhances applicability in ambulatory environments,

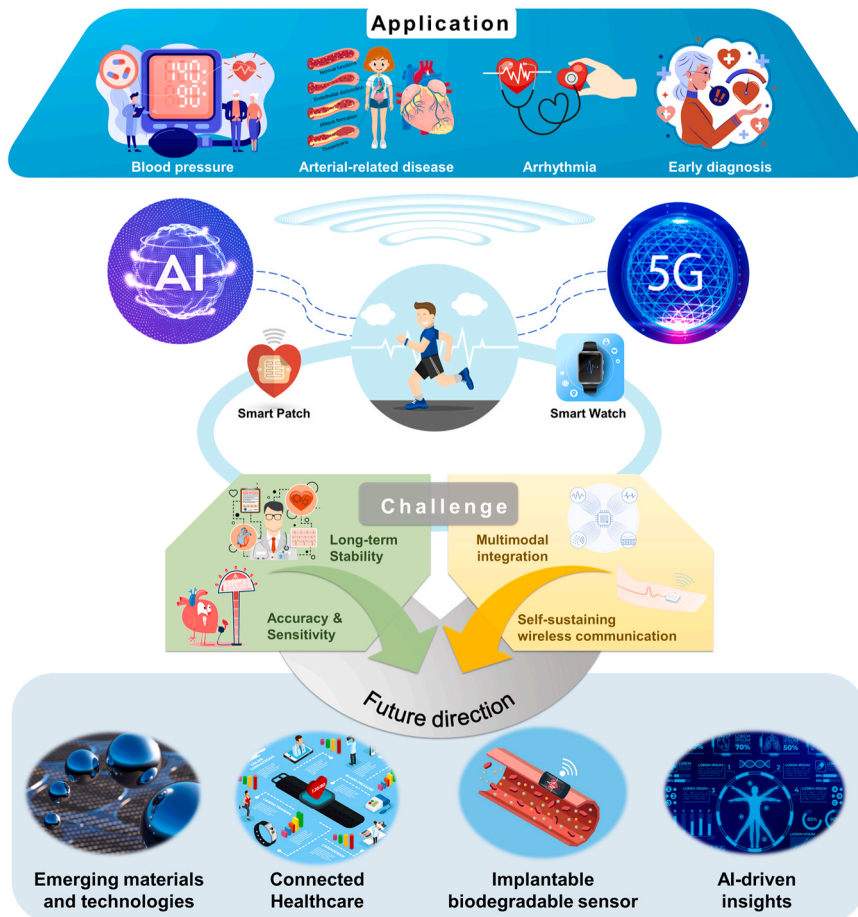
Table 2

Comparison of key properties, materials, advantages, and limitations of various mechanisms.

Sensing mechanism	Key properties	Representative materials	Power consumption	Limitations
Piezoelectric	High piezoelectric coefficient, mechanical flexibility	PZT, PVDF, quartz, BaTiO ₃	Self-powered	■ Sophisticated processing
Triboelectric	Charge affinity, surface roughness, flexibility	PTFE, FEP, PVDF, PI	Self-powered	■ Sensitivity to interference
Optical	Optical bandgap, transparency, emission efficiency	Si, GaAs, InGaN	>10 mW [359]	■ Poor linearity
				■ Low stability
				■ Individual differences
				■ Easily influenced by the environment
Magnetoelastic	Magnetomechanical coupling, wireless transduction, passive operation	NdFeB, Fe-based alloys, magnetoelastic composites	Battery-free	■ Signal interference
				■ Limited signal amplitude
				■ Sensitivity to external magnetic noise
				■ Relatively complex signal decoding
Capacitive	Dielectric constant, stability, conformability	Au, Ag, PU, Polyaniline	mW-level [360]	■ Limited detection range
				■ Environmental sensitivity
				■ Complex signal processing
Piezoresistive	Gauge factor, mechanical durability	CNTs, Graphene, Carbon black	3 nW [361]	■ Physical contact
				■ Poor stability
Bioimpedance	Electrical conductivity, biostability	Ag, Au, PEDOT:PSS	80 μ W [362]	■ Individual differences
				■ Complex data interpretation
Traditional ECG	Electrical conductivity, Temporal resolution	Ag/AgCl	16.4–373.2 mW [363]	■ Lack of portability
Holter	Portability, Continuity	Ag/AgCl	> 100 mW [364]	■ Operator dependence
				■ Limited resolution
				■ Patient discomfort

although multi-electrode configurations and contact stability remain technical challenges [357,358]. Rather than replacing conventional methods, these emerging technologies are expected to supplement traditional tools through multimodal integration, enhanced

spatial-temporal resolution, and expanded sensing dimensions across varied clinical and resource-constrained scenarios.

**Fig. 16.** Challenges and outlook in cardiovascular health monitoring technologies.

4. Challenges and outlook

As cardiovascular monitoring technologies become increasingly precise and integrated, significant challenges remain in translating multimodal sensing outputs into reliable and clinically meaningful cardiovascular information. Despite advances across sensing modalities such as piezoelectric, optical, and piezoresistive systems, stable signal acquisition and interpretation under complex physiological conditions remain difficult to achieve. These bottlenecks can be broadly categorized into four key aspects: reliable signal acquisition under dynamic physiological conditions, long-term device stability and biocompatibility, challenges in multimodal data integration and interpretation, and the lack of standardized frameworks for clinical validation. These challenges arise from limitations in materials, device architecture, signal fidelity, and long-term operational stability, which directly affect the consistency and interpretability of hemodynamic-related signals. Emerging strategies, including biodegradable materials and battery-free wireless systems, offer new opportunities but also introduce additional complexities in system integration and performance control. Addressing these challenges requires coordinated advances in materials design, system engineering, and clinically grounded validation to enable robust and scalable cardiovascular monitoring platforms (Fig. 16).

4.1. Technical and clinical challenges

4.1.1. Challenges in monitoring accuracy and sensitivity

Achieving precise and high-fidelity cardiovascular monitoring remains technically demanding due to limitations in material properties, sensor design, and integration complexity. Functional materials must meet diverse and sometimes conflicting requirements, including mechanical compliance, electrical responsiveness, and environmental durability. While ceramics like PZT offer high piezoelectric output, their brittleness hinders use in deformable systems [365–368]. Polymers such as PVDF provide flexibility but compromise performance. Optical materials like silicon operate efficiently within specific wavelengths yet underperform in deeper tissue penetration [369], and organic alternatives are prone to photodegradation [370]. Conventional strain gauge metals lack sufficient sensitivity for subtle physiological signals [371], and materials with high dielectric constants, although enhancing capacitance, often introduce losses that destabilize signal output [372]. Electrodes such as gold or platinum may induce bioincompatibility or corrosion over time, complicating long-term deployment.

Sensor architecture further governs sensitivity, stability, and adaptability in real-world scenarios. In piezoelectric systems, stress concentration and directional strain alignment are essential for consistent output, but multiaxial deformations *in vivo* complicate these optimizations [373]. Optical sensors demand precise light path engineering to mitigate attenuation and ambient interference, particularly under motion or variable skin conditions [374]. For resistive and capacitive sensors, spatial electrode configuration and field uniformity significantly affect signal resolution, while triboelectric systems depend on a delicate balance of charge generation and dissipation, where gains in output are often nonlinear and saturation-prone [375]. Bioimpedance performance is highly dependent on interface conformity, making mechanical design and surface engineering pivotal to signal integrity and biocompatibility.

Fabrication processes impose additional constraints. Variations in purity, morphology, and processing parameters across materials like PZT, Si, and polymer composites introduce inconsistencies that propagate through device performance. Micro- and nano-structural features, which are often critical for transduction efficiency, require highly precise techniques such as photolithography, electron beam patterning, or vapor deposition [376–382]. Multilayer systems, increasingly adopted to enable multimodal sensing, present challenges in interfacial bonding; inadequate adhesion or mismatched mechanical properties at these junctions can result in delamination, signal loss, or premature failure.

In practice, different sensing modalities exhibit distinct sensitivities

to motion artifacts and physiological noise. Mechanical sensing approaches such as piezoelectric and triboelectric sensors are generally more robust to ambient electromagnetic interference but are susceptible to motion-induced disturbances, while optical and bioimpedance techniques are more sensitive to tissue motion and contact instability, requiring advanced signal processing or multimodal compensation strategies to ensure reliable measurements.

4.1.2. Biocompatible and biodegradable materials for long-term and transient monitoring

The development of cardiovascular sensors that are both biocompatible and biodegradable presents a complex balance between physiological safety, device longevity, and controlled degradation. In recent years, significant progress has been made in engineering cardiovascular sensors using materials that offer both biocompatibility and biodegradability. Biocompatibility extends beyond avoiding acute toxicity, requiring long-term immunotolerance, minimal fibrotic encapsulation, and stable tissue integration [383,384]. To address these needs, strategies such as surface functionalization with zwitterionic or ECM-mimetic coatings, incorporation of anti-inflammatory agents, and the use of hydrogels with tissue-matched mechanical properties have been explored. Materials like gelatin methacrylate, poly(L-lactide-co-ε-caprolactone), and conductive polysaccharide-based hydrogels have demonstrated improved interface compatibility in both subcutaneous and vascular environments [385–388]. Natural biomaterials, including amino acids, peptides, proteins (collagen, silk fibroin, and elastin), polysaccharides (cellulose and chitosan), and certain viral protein assemblies, have attracted increasing attention due to their intrinsic biocompatibility, biological adaptability, and environmental sustainability [170,389,390]. Derived from natural sources, these materials generally exhibit excellent biocompatibility without the need for additional surface modification.

Devices such as transient pacemakers and pressure monitors have demonstrated safe degradation profiles *in vivo*, functioning effectively for several weeks before complete resorption [391]. However, precise control over degradation timing remains a major challenge, as physiological variability can alter dissolution rates and by-product profiles. Rapid degradation risks early device failure, while prolonged persistence may cause fibrosis or foreign-body reactions. Beyond material stability, ensuring reliable electrical and mechanical performance throughout the functional lifespan is critical. As biodegradable sensors degrade, changes in conductivity, flexibility, or structural integrity may compromise signal fidelity [392–394]. For implantable systems, additional challenges arise from the need for stable encapsulation to protect device functionality, reliable power availability under constrained energy conditions, and long-term biocompatibility with controlled immune response, all of which are essential for maintaining consistent *in vivo* performance. Strategies to compensate for such signal drift through circuit design or real-time calibration are still under development. Integrating non-degradable components like chips or antennas with resorbable substrates introduces further complexity, often requiring hybrid packaging or full-system biodegradability. Overall, biocompatible and biodegradable materials represent a promising pathway for next-generation implantable cardiovascular sensors, but require coordinated progress in materials engineering, device architecture, and biological validation to reach clinical viability.

4.1.3. Self-sustaining wireless communication based on SAW and radio-frequency (RF) techniques

Battery-free wireless systems are central to the future of long-term cardiovascular monitoring, yet achieving reliable communication without external power remains a significant technical challenge. Energy harvesting technologies such as triboelectric, piezoelectric, and thermoelectric mechanisms have been investigated to convert biomechanical motion or body heat into electrical energy [349]. These self-powered approaches provide a promising pathway toward

continuous and maintenance-free cardiovascular monitoring by reducing dependence on conventional batteries and enabling energy-autonomous operation. However, the harvested energy is often limited and variable under dynamic physiological conditions, which poses challenges for sustaining continuous sensing and wireless transmission.

SAW and RF telemetry offer feasible solutions for wireless data communication. SAW devices can function passively by reflecting external signals, while RF systems support active real-time transmission [395–397]. Despite these advantages, miniaturizing antennas and integrating them into flexible or implantable platforms is a major hurdle. Biological tissues absorb and scatter high-frequency signals, often resulting in substantial transmission loss and degraded communication quality. This signal attenuation limits the depth and reliability of data transfer, especially in subcutaneous or intravascular applications.

Several emerging strategies aim to address these limitations. Intra-body communication and near-field inductive coupling reduce propagation loss by transmitting at lower frequencies or using the body as a conductive channel. Simultaneously, efforts in circuit miniaturization and ultra-low-power design have improved the efficiency of data transmission under constrained energy conditions. Combining these advances with adaptive energy management systems could allow sensors to operate autonomously and transmit clinical-grade cardiovascular data over extended periods. Progress in this field depends on integrating advances in materials, electronics, and wireless engineering to support reliable and scalable communication without reliance on conventional batteries.

4.1.4. Enhancing cardiovascular monitoring through multimodal sensing integration

Multimodal sensing, which integrates multiple complementary mechanisms such as piezoelectric, optical, capacitive, or bioimpedance signals, has emerged as a promising approach to improve data accuracy, reliability, and contextual interpretation in cardiovascular monitoring. By capturing physiological signals through different transduction principles, multimodal systems can enhance signal fidelity, enable cross-validation, and compensate for modality-specific artifacts. The multimodal integration of PPG and bioimpedance has been shown to improve the accuracy and robustness of HR monitoring, particularly under challenging conditions such as motion, variations in skin tone, and changes in sensor placement. By leveraging the complementary strengths of optical and electrical signals, this approach enhances pulse detection stability, enables more precise estimation of pulse transit time and blood pressure, and offers advantages in power efficiency and skin conformity [398]. It has been shown that a multimodal system integrating electrochemical lactate sensing and ECG monitoring in the same flexible platform can achieve simultaneous acquisition of metabolic and electrophysiological signals during the exercise state, which effectively enhances the ability to comprehensively assess the cardiovascular health status [399]. In addition, the flexible photoacoustic ‘blood stethoscope’, which integrates optical and piezoelectric acoustic mechanisms, can realise non-invasive, real-time, three-dimensional imaging monitoring of a variety of cardiovascular parameters, and provide a more in-depth and wide-ranging assessment of cardiovascular health, providing a potential direction for future multiparameter continuous monitoring [185]. Complementing this, a wearable platform combining piezoelectric ultrasound and electrochemical sensors enables real-time monitoring of blood pressure, heart rate, and metabolic markers such as glucose and lactate. By integrating physical and chemical sensing in a single patch, this approach expands the scope of cardiovascular assessment and supports comprehensive, multimodal health monitoring [400].

While multimodal integration offers significant advantages, it also introduces substantial complexity in system design, signal synchronization, and data interpretation. Variations in sampling rates, signal formats, and noise profiles require advanced computational strategies

such as sensor-specific preprocessing, temporal alignment, and ML-based fusion. To enable clinical validation and scalable deployment, standardization is essential. Frameworks like HL7 FHIR and ISO/IEEE 11073 support interoperable data encoding, while modular hardware architectures and defined communication protocols improve device compatibility. In parallel, regulatory guidelines from bodies such as the FDA provide pathways for benchmarking multimodal systems against clinical endpoints, promoting their integration into real-world healthcare settings. In addition to technical frameworks, achieving real-world application requires the establishment of consensus-based benchmarking protocols, cross-disciplinary standard committees, and early collaboration between developers and regulators. Pilot studies conducted under standardized clinical workflows can serve as a bridge between prototype testing and scalable deployment.

4.2. Conclusion and outlook

Cardiovascular health monitoring has progressed from conventional ECG-centered techniques to integrated platforms capable of continuously capturing diverse physiological parameters. Within this transition, hemodynamic decoding has emerged as a unifying framework, since signals such as arterial pulse, blood pressure, and vascular stiffness directly reflect the dynamic state of the cardiovascular system. Advanced sensing modalities, including piezoelectric, triboelectric, magnetoelastic, optical, capacitive, piezoresistive and bioimpedance devices, provide unique advantages such as high sensitivity, self-powered operation, and non-invasive measurement, thereby extending cardiovascular assessment beyond clinical settings and into personalized, preventive care. Although significant progress has been made, challenges remain in achieving robust accuracy, long-term reliability, and biocompatibility, particularly for implantable and continuous monitoring applications. Physiological variability, motion artifacts, and environmental interferences can impair signal fidelity, while device stability and user comfort remain critical for real-world adoption. Addressing these limitations will require synergistic progress in material innovation, device engineering, and algorithmic robustness.

The next generation of cardiovascular monitoring will be defined by the integration of AI-enabled hemodynamic decoding and IoT-based connected healthcare infrastructures. AI-based models are increasingly used to integrate multimodal hemodynamic signals and extract physiologically relevant features for clinical inference, supporting applications such as arrhythmia detection, pulse waveform analysis, and BP estimation. Current multimodal fusion strategies are predominantly based on data-driven ML approaches, while physics-informed models are increasingly explored to improve interpretability and constrain model behavior under varying physiological conditions. Future systems will prioritize interpretability and personalization, adjusting to individual baselines and behavioral contexts to enhance diagnostic accuracy and decision support. At the same time, IoT-enabled architectures will enable seamless communication between wearable or implantable sensors and cloud or edge computing networks, ensuring real-time feedback, longitudinal monitoring, and decentralized healthcare delivery.

From a translational perspective, the clinical integration of hemodynamics-based multimodal monitoring technologies will depend on alignment with regulatory frameworks, algorithm accountability, and data protection standards. Building clinician-facing platforms with validated thresholds and explainable outputs will be essential to foster trust and ensure responsible clinical adoption. With continuous innovation at the intersection of sensing materials, bioelectronics, and intelligent data analytics, hemodynamic decoding is poised to redefine multimodal connected cardiovascular healthcare, offering transformative opportunities for precision medicine and long-term disease prevention.

Abbreviations

Association for the Advancement of Medical Instrumentation, AAMI; Atrial fibrillation, AF; Artificial intelligence, AI; Augmentation index, AIx; Aluminum nitride, AlN; Acute myocardial infarction, AMI; Blood pressure, BP; C-reactive protein, CRP; Cardiac troponin I, cTnI; Cardiovascular diseases, CVDs; Electrocardiograms, ECG; Double electric layer, EDL; Electromyograms, EMG; Endocardial pressure, EP; Finite element method, FEM; Fluorinated ethylene propylene, FEP; Gallium arsenide, GaAs; Gallium nitride, GaN; Graphene oxide, GO; Heart rate, HR; Heart rate variability, HRV; Integrated circuit, IC; Interdigitated electrodes, IDE; Ionic liquid, IL; Internet of Things, IoT; Impedance plethysmography, IPG; Implanted triboelectric active sensor, iTEAS; Micro-electromechanical systems, MEMS; Myocardial infarction, MI; Machine learning, ML; Near-infrared, NIR; Poly (vinylidene fluoride-ran-trifluoroethylene), P(VDF-TrFE); Polyacrylamide, PAM; Polyaniline, PANI; Pulse arrival time, PAT; Polycaprolactone, PCL; Polydimethylsiloxane, PDMS; Piezoelectric nanogenerators, PENGs; Physics-informed neural network, PINN; Polylactic acid, PLA; Poly-L-lactic acid, PLLA; Partial least squares regression, PLSR; Polymethyl methacrylate, PMMA; Platelet-derived microparticles, PMP; Photoplethysmography, PPG; Pulse transit time, PTT; Polyurethane, PU; Polyvinylidene fluoride, PVDF; Pulse wave velocity, PWV; Lead zirconate titanate, PZT; Resistance index, RI; Reactive oxygen species, ROS; Surface acoustic wave, SAW; Seismocardiogram, SCG; Surface-enhanced Raman spectroscopy, SERS; Silk fibroin, SF; Silicon, Si; Stiffness index, SI; Oxygen saturation, SpO₂; Triboelectric nanogenerators, TENGs; Titanium dioxide, TiO₂; Thermoplastic polyurethane, TPU; Zinc oxide, ZnO.

CRediT authorship contribution statement

Tong Li: Writing – review & editing, Writing – original draft. **Wanghuai Xu:** Writing – review & editing, Writing – original draft. **Jun Li:** Writing – review & editing, Writing – original draft, Investigation, Conceptualization. **Yun Ke:** Writing – review & editing, Writing – original draft, Visualization, Investigation, Conceptualization. **Zhong Lin Wang:** Writing – review & editing, Supervision, Conceptualization. **Zhou Li:** Writing – review & editing, Supervision, Conceptualization. **Fan Yang:** Writing – review & editing, Supervision, Funding acquisition, Conceptualization. **Dalong Ni:** Writing – review & editing, Investigation. **Xiaofeng Ye:** Writing – review & editing, Supervision, Conceptualization.

Declaration of Competing Interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

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Data availability

No data was used for the research described in the article.

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