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Monitoring and analysis of cardiovascular pulse waveforms using flexible capacitive and piezoresistive pressure sensors and machine learning perspective

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A B S T R A C T

The growing interest in flexible electronics for physiological monitoring, particularly using flexible pressure sensors for cardiovascular pulse waveforms monitoring, has potential applications in cuffless blood pressure measurement and early diagnosis of cardiovascular disease. High sensitivity, fast response time, good pressure resolution and a high signal-to-noise ratio are essential for effective pulse waveform detection. This review focuses on flexible capacitive and piezoresistive pressure sensors, which have seen significant enhancements due to their simple operation, superior performance, wide range of materials, and easy fabrication. The comparison of sensing methods for acquiring pulse waveforms from the wrist artery, device integration configurations, high-quality pulse waveforms collection, and performance analysis of capacitive and piezoresistive sensors are discussed. The review also covers the use of machine learning for analyzing pulse waveforms for cardiovascular disease diagnosis and cuff-less blood pressure monitoring. Lastly, it provides perspectives on current challenges and further advancements in the field.

1. Introduction

Cardiovascular disease (CVD) is a major global health concern, responsible for significant morbidity and mortality worldwide. Hypertension, characterized by elevated blood pressure, affects 47% of adults in the United States, 30% worldwide (CDC) and is a major risk factor for various CVDs, such as arteriosclerosis, diabetes, myocardial infarction, and coronary heart disease (Bingger et al., 2012; Meng et al., 2022). Early diagnosis and continuous monitoring of physiological signals are crucial for managing CVDs, necessitating the development of lightweight, user-friendly, and cost-effective wearable devices with a personalized biomarker monitoring system using portable energy storage device such as lithium ion batteries (Henriques et al., 2022). Pulse waveforms provide valuable information about an individual's cardiovascular health, including heart problems, blood pressure, and arterial stiffness (Meng et al., 2022). Continuous monitoring of pulse waveforms using wearable devices can serve as an excellent alternative to traditional cuff-based blood pressure monitoring systems, potentially reducing CVD risk factors and the related mortality rates.

A typical pulse waveform contains three peaks: systolic peak S, reflected peak R, and diastolic peak D (Fig. 1) (Cheng et al., 2020; He et al., 2019). Detecting subtle changes in these peaks is necessary to correctly characterize cardiovascular health, such as arterial stiffness (He et al., 2019; Qiu et al., 2019; Yang et al., 2019b). Traditional Chinese medicine (TCM) practitioners have long relied on pulse waveform analysis for diagnosing various health conditions. However, this approach is highly subjective and can vary between practitioners. Wearable photoplethysmogram (PPG) sensors, which use optical sensing to monitor real-time pulse waveforms, have emerged as a popular alternative. Despite their widespread use, PPG sensors face limitation in accurately detecting detailed characteristics of pulse waveforms due to data acquisition from capillaries rather than arterial sites, skin tonality, light intensity variation, and distance from the artery (Ibrahim and Jafari, 2022) (Lin et al., 2020) (Meng et al., 2022).

Recent advances in flexible pressure sensors have paved the way for improved cardiovascular monitoring through pulse waveform analysis (Kim et al., 2019b; Luo et al., 2016). These wearable sensors are lightweight, conformable to the skin, biocompatible, stable, low-cost, and

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low-power, making them suitable for a variety of applications, including the internet of things (IoT) (Cho et al., 2017; Jang et al., 2016), electronic skins (e-skins) (Kang et al., 2016; Kim et al., 2019a; Nie et al., 2017; Wang et al. 2013, 2017a, 2017b; Yang et al., 2018; Zhang et al., 2017a; Zhong et al., 2016), electronic textiles (e-textiles) (Cherenack et al., 2010; Chhetry et al., 2017; Eom et al., 2017; Lee et al., 2015), flexible touch displays (Fan et al., 2012; Lipomi et al., 2011; Luo et al., 2019), soft robotics (Kim et al., 2014), mobile healthcare aids (Boutry et al., 2015; Zang et al., 2015), and energy harvesting devices (Wang et al., 2015; Yang et al., 2013; Zhu et al., 2014). Several types of pressure sensors, such as capacitive pressure sensors, piezoresistive pressure sensors, piezoelectric pressure sensors, and triboelectric pressure sensors, have been successfully used for detecting arterial pulse waveform. Among these different pressure sensing mechanisms, capacitive and piezoresistive pressure sensors have simple working mechanisms, good pressure sensitivity, fast response time, and a compact circuit layout.

Machine learning, a branch of artificial intelligence, offers a promising avenue for quantitative pulse waveform analysis, which could lead to more accurate disease diagnosis and cuffless blood pressure monitoring (Meng et al., 2022). By extracting essential features from pulse waveforms, machine learning algorithms can provide valuable insights for diagnosing various health conditions, including those related to CVDs. Furthermore, machine learning has been utilized to estimate blood pressure by analyzing pulse waveforms features, such as pulse shape, pulse transit time, and pulse wave velocity and comparing them with standard cuff-based system. However, the complexity of blood circulation presents significant challenges to developing a highly stable and reliable cuffless blood pressure measurement system. Accurate acquisition of pulse waveforms using pressure sensors is crucial for overcoming these challenges.

The primary aim of this review paper is to provide a comprehensive comparison of different sensing mechanisms employed in wearable sensors for detecting pulse waveforms from the wrist artery, focusing on flexible capacitive and piezoresistive pressure sensors. We will discuss the essential sensor parameters for obtaining stable pulse waveforms from arterial sites and summarize the state-of-the-art research for capacitive and piezoresistive pressure sensors for acquiring pulse waveforms. Additionally, we will explore the potential applications of machine learning in analyzing pulse waveforms for disease diagnosis and cuffless blood pressure monitoring.

By examining the latest advancements in flexible pressure sensors and their applications in pulse waveform analysis, this review paper aims to contribute to the ongoing efforts to develop more accurate and reliable wearable devices for continuous monitoring of cardiovascular health. As the prevalence of CVDs continues to rise globally, the need for innovative solutions to improve early diagnosis and management of these conditions is more pressing than ever. The integration of advanced sensor technology and machine learning in wearable devices holds great promise for transforming the landscape of CVD prevention and treatment, ultimately reducing the burden of these diseases on individuals and healthcare systems worldwide.

2. Sensing mechanism for detecting pulse waveform

Wearable sensing devices detecting pulse waveforms often rely on changes in blood volume, impedance due to the blood flow, or mechanical deformation due to palpation (Fig. 2a). Most commercial wearable devices use optical sensing technologies, specifically photoplethysmogram (PPG) techniques, to detect pulse waveforms. The PPG sensor consists of one optical source and a photodetector, where the light source emits light through the skin, and the photodetector detects the reflected light carrying information proportional to blood volume (Fig. 2b–i) (Castaneda et al., 2018). Due to the volumetric flow rate of the blood going through the capillary or major arteries, the reflected light carries information proportional to the blood volume.

Bioimpedance sensors are another approach that relies on changes in electrical impedance due to the change in blood flow through arteries or capillaries (Ibrahim and Jafari, 2022). As blood has different resistance than skin and fat, the impedance changes as blood propagates through the body, which can be detected using bioimpedance electrodes (Fig. 2b–ii).

Flexible pressure sensors, on the other hand, rely on the artery's pressure on the skin, which induces tiny deformation that can be detected (Fig. 2b–iii). Pressure sensors can be categorized into capacitive, piezoresistive, piezoelectric, and triboelectric sensors based on their sensing mechanisms (Cui et al., 2022). Capacitive pressure sensors work based on the parallel plate mode of a capacitor, where two parallel electrodes are separated by using an insulating material (Fig. 2c–i). The

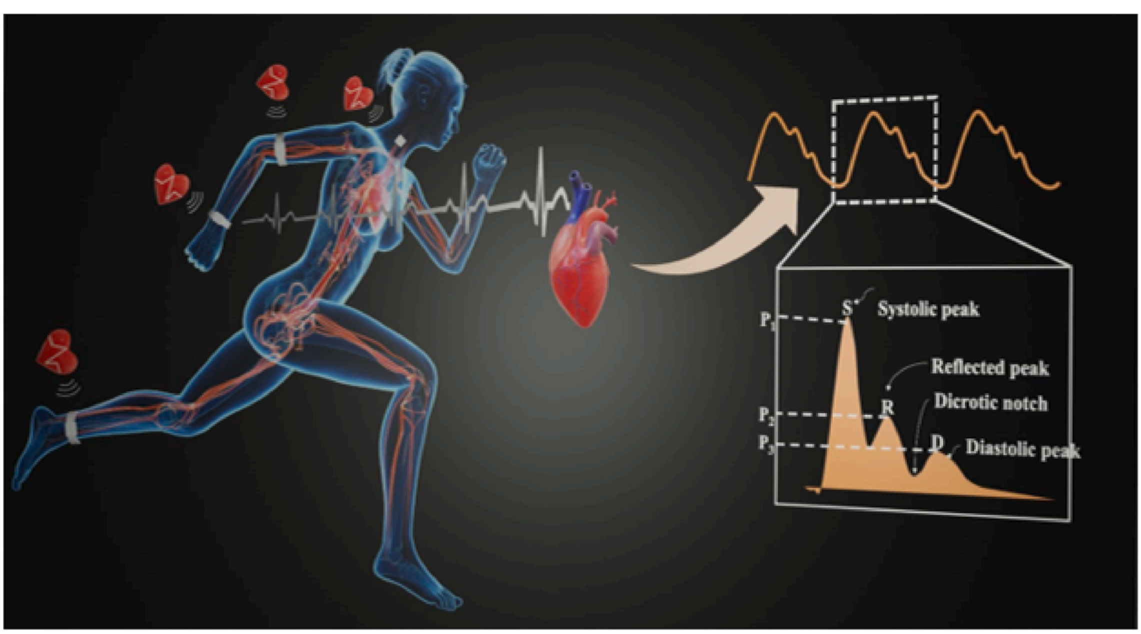

Fig. 1. Pulse waveform obtained from the wrist artery. A healthy pulse waveform contains a systolic or S peak corresponding to systolic pressure P_1, reflected peak R from the periphery corresponding to the reflected pressure P_2, and diastolic peak D corresponding to the diastolic pressure P_3.

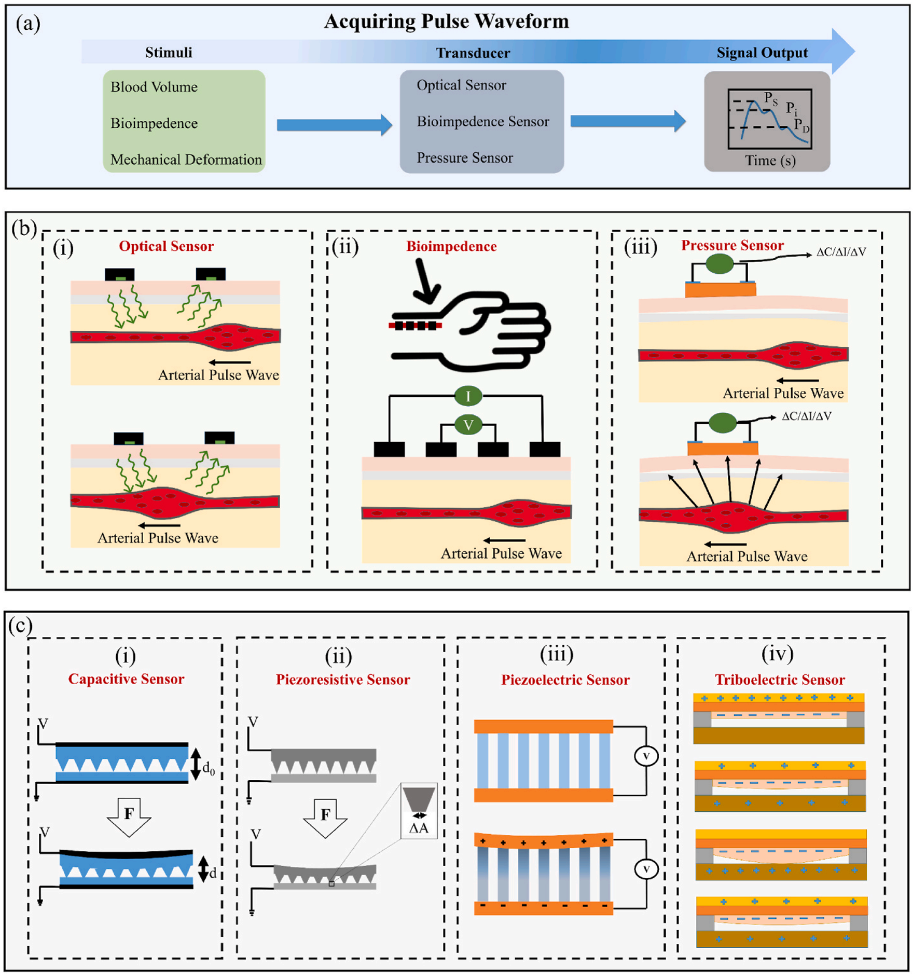

Fig. 2. Existing techniques for detecting pulse waveforms. Sensing technologies are based on different techniques for detecting pulse waveforms. (a) Optical and bioimpedance sensing depends on the volume of blood flow to acquire pulse waveforms. Flexible mechanical sensors rely on skin deformation due to palpation to generate pulse waveforms data. (b) Figures i-iii show sensing techniques for the optical sensor (i), bioimpedance sensor (ii), and mechanical sensor (iii). (c) Figures (i-iv) show details of different flexible mechanical sensors' sensing mechanisms.

distance between the electrodes plays a key role for capacitive sensors, and research focuses on this design rule to obtain high sensitivity (He et al., 2018; Lee et al., 2016a; Miller and Bao, 2015; Shao et al., 2017). Piezoresistive sensors operate based on the change in resistance between two layers of sensing materials (Fig. 2c–ii) (Zhang et al., 2016). The resistance is dependent on the resistivity of the materials and the substrate area.

Piezoelectric sensors use mechanical strain-induced polarization occurring in certain solid materials. Due to the external force, dipole moments in the anisotropic crystalline material generate macroscopic potential in the crystal (Fig. 2c–iii) (Li et al., 2018b). However, their response is dynamic and incapable of detecting static pressure stimuli. Triboelectric sensors are dynamic pressure sensing technologies that utilize mechanical motion to induce charges on two opposing substrates (Fig. 2c–iv). When two materials with different electronegativity come into physical contact and separate, the two substrates induce a potential that can run electricity through an external circuit (Pu et al., 2021).

Each of these sensing mechanisms has its advantages and limitations, which are considered during the design and development of wearable devices for pulse waveform monitoring. Researchers continue to explore new materials, fabrication processes, and sensor configurations to improve sensitivity, response time, and overall performance of these devices. The ultimate goal is to develop wearable sensors that are reliable, accurate, and user-friendly for continuous monitoring of cardiovascular health.

3. Key performance metrics of pressure sensors

Although the gold standard for acquiring pulse waveforms is invasive catheter measurement using a high-fidelity pressure transducer, its invasive nature makes it impractical for routine clinical use. A common non-invasive method is using optical sensing-based PPG, wearable pressure sensors (capacitive, piezoresistive, piezoelectric, and triboelectric), or bioimpedance sensors. Regardless of the sensing method used, accurate signal collection is crucial for practical applications. To ensure that the measured pulse waveforms are not distorted, several parameters should be assessed, such as waveform shape and timing, amplitude, and consistency.

Detecting high-quality and stable pulse waveforms from the wrist artery depends on various factors, such as the sensor footprint area, appropriate location, and sensitivity (Choong et al., 2014). The radial artery, which is the most common location for detecting pulse, lies under approximately 5 mm of skin with a diameter less than 2.5 mm. Consequently, the sensor footprint area should be small to transmit the pressure efficiently from the radial artery to the sensor. Choosing the right location and decisive point for detecting the arterial pulse waveform is crucial.

Pulse waveforms from a healthy individual contain three peaks: systolic peak, reflected peak, and the diastolic peak (Suen et al., 2018). Resolving these peaks requires high sensitivity and resolution from the pressure sensor. For instance, a piezoresistive pressure sensor with a micro pyramid structure was able to detect wrist pulse effectively, where a pulse pressure of 170Pa (1.3 mmHg) was detected on the skin (Choong et al., 2014).

The pressure sensor should have a broad and linear pressure sensing range with adequate sensitivity to monitor pulse waveforms effectively. A linear sensing range ensures undistorted pulse waveforms without additional signal processing (Bae et al., 2016; Luo et al. 2016, 2017). When a flexible sensor is worn, external pressure is applied to the sensor to keep it in place. If this pressure surpasses the highest sensitivity range, the sensor may struggle to detect pulse waveforms efficiently. A broad pressure sensing range with adequate sensitivity ensures that the sensor maintains enough sensitivity after the external pressure is applied, helping detect pulse waveforms easily.

Fast response time is also essential for resolving individual peaks in the time domain, particularly as a person's pulse may increase during physical activity (Luo et al., 2016). An effective pressure sensor should be capable of adjusting to these changes in BPM and providing accurate readings.

A high signal-to-noise ratio (SNR) is critical for high-performance sensors, as noisy signals may require extensive processing. The noise may come from bending, twisting, body movement, or external factors such as temperature, and electromagnetic interference. Techniques to increase SNR for pressure sensors include using polymeric nanopillar adhesion layers inspired by insects and lizards (Fig. 3a) (Pang et al., 2015) or applying external pressure (He et al., 2019). These techniques have been shown to improve the SNR without compromising sensitivity.

Studies have shown that steady pulse waveforms from the wrist artery can be obtained without ultrahigh sensitivity; however, excessive pressure may distort the waveforms and produce erroneous data. According to a report, the preferable window for external pressure is up to 10 kPa without distorting the pulse waveform signal with a pressure sensitivity of less than 1 kPa^{-1} (Fig. 3d). The SNR increases under similar pressure sensitivity as the external pressure increases.

In summary, obtaining accurate pulse waveforms from the wrist artery involves balancing factors such as sensor footprint area, sensitivity, pressure sensing range, response time, and signal-to-noise ratio. Designing sensors with these considerations in mind will lead to more reliable and accurate pulse detection devices.

4. Recent progress on pressure sensors for pulse waveforms detection

4.1. Capacitive pressure sensor

4.1.1. Materials and fabrication

4.1.1.1. Substrate and dielectric materials. To develop a flexible capacitive pressure sensor, selecting appropriate substrate and dielectric materials is crucial. The polymer materials for flexible sensors should possess bendability, stretchability, and low Young's modulus, similar to the skin, to ensure comfortable usage. Dielectric materials should have good permittivity, low modulus of elasticity, low viscoelasticity, and low leakage current for capacitive pressure sensors to achieve an optimal pressure response.

Polydimethylsiloxane (PDMS) is an excellent option for flexible pressure sensors due to its biocompatibility with human skin and living cells, flexibility, stretchability, and high dielectric constant, making it suitable for electrode and substrate fabrication (Chowdhury et al., 2020; Li et al., 2019c; Mahata et al., 2020; Mannsfeld et al., 2010; Yang et al., 2019a; Zeng et al., 2019). Various flexible dielectric materials have been reported, including PDMS, polyvinylidene fluoride (PVDF) (Shuai et al., 2017), polyvinylidene fluoride trifluoro ethylene [P(VDF-TrFE] (Guo et al., 2019a), Parylene C (Ha et al., 2012; Li et al., 2020; Pritchard et al.,

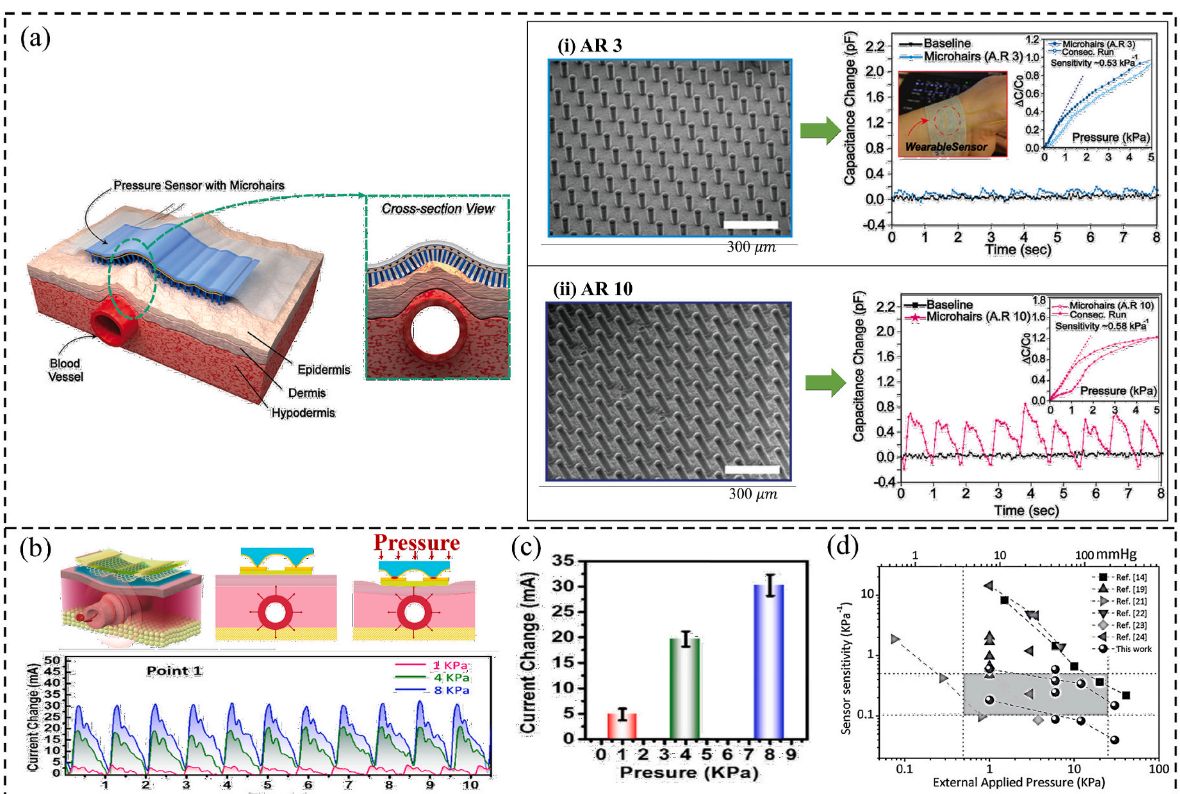

Fig. 3. Increasing conformal contact is crucial for a higher SNR. (a) Gecko-inspired nano hair substrate increases contact between the skin and the sensor (Pang et al., 2015). The adhesion further improves resulting higher SNR as the aspect ratio (AR) increases from 3 (a–i) to 10 (a-ii). (b–d) Applying external pressure also increases the SNR. (b) The SNR dramatically improves as the external pressure increases (He et al., 2019). (c) The current signal increases as the external pressure increases from the wrist artery (He et al., 2019). (d) With an external pressure window of up to 10 kPa, a low-sensitive sensor can capture pulse waveform with a good SNR (Luo et al., 2016).

2008), Ecoflex (Atalay et al., 2018; Baek et al., 2017; Ding et al., 2019), Polyurethane (PU) (Hwang et al., 2019; Liu et al., 2018; Qiu et al., 2019; Yang et al., 2018), Al_2O_3 (Huang et al., 2020), polypropylene (Cheng et al., 2018), colorless polyimide (CPI) (Wan et al., 2018a), Polymethyl methacrylate (PMMA) (Joo et al., 2017), graphene oxide (Wan et al., 2017), and many others.

4.1.1.2. Electrode materials. A variety of materials are available for electrode layer fabrication, including intrinsically conductive polymer, conductive polymers synthesized with conductive fillers, and thin metallic film deposited on the polymers. Electrodes should exhibit good electrical conductivity, mechanical stability against strain, and maintain conductivity under cyclic loading. Intrinsically conductive polymers, or organic semiconductors, possess a delocalized p-electron system with an intrinsic wide-bandgap, which influences their electron affinity (Hatchett and Josowicz, 2008). Polyaniline (PANI), Poly pyrrole (PPy), and polythiophene (PTH) are examples of such polymers (Hatchett and Josowicz, 2008; Zhou et al., 2021). Another approach to create electrode materials is by adding conductive fillers into non-conductive polymer matrices. Carbon-based 0D, 1D, and 2D nanofillers like carbon black, carbon nanotubes, graphene, and metal, and metal oxide nanowires such as AgNWs, ZnNWs, etc. are gaining popularity as conductive fillers (Li et al., 2019c; Liu et al., 2018; Wang et al., 2014).

4.1.1.3. Fabrication

4.1.1.3.1. Fabrication of porous sensing material. There are several fabrication techniques for creating porous polymer substrate. One cost-effective method is the sacrificial template method, which involves incorporating solvent-dissolvable porogens such as salt, sugar, polystyrene beads into the polymer substrate to create the pores. A straightforward approach to this method is mixing the sacrificial templates with a pre-cured polymer solution. After curing the polymer, the template is removed by dissolving it in an appropriate solvent. Dissolving the templates is relatively simple, as materials like PDMS are permeable to water. The pore size in the substrate depends on the size of the porogens used. Researchers have also reported alternative techniques for creating pores, such as using a water-in-oil emulsion process, baking-inspired foaming method, and modifying commercially available sponges with active materials. Utilizing readily available sponges and functionalizing them with suitable chemicals provides another approach for achieving porous substrates.

4.1.1.3.2. Fabrication of micropattered sensing material. Among various fabrication techniques, photolithography is a highly effective method for creating micropatterned sensing layer. Micropatterned molds can be achieved using photolithography by applying a mask on a silicon wafer. The micropatterned sensing layer can then be produced by casting the polymer material into the mold via spin coating. After curing the polymer, it can be peeled off to reveal the desired microstructures on the surface. Due to the silicon wafer's crystal structure, pyramid shaped micropatterns are common on flexible substrates. Micro dome features have also been reported using photolithography.

In addition to photolithography, additive manufacturing methods such as 3D printing have been employed to create various micropatterned sensing materials. Researchers have also utilized natural templates, such as lotus leaf (Li et al., 2016a; Wan et al., 2018a), calathea zebrine leaf (Qiu et al., 2018), and rose petal (Mahata et al., 2020; Wan et al., 2018b) to fabricate micropatterned substrates for capacitive sensors. Sandpaper and anodized aluminum have been reported as mold materials for crafting micropatterned sensing materials as well (Bai et al., 2020; Guo et al., 2019a). Beyond these methods, numerous innovative techniques have been developed for fabricating micropatterned sensing layers. The creation of wrinkled microstructure using UV/plasma-treated polymer as the template has gained popularity as a non-photolithography-based fabrication approach for sensor realization. Plasma treatment forms a SiOx layer, which generates a wavy pattern when the stretch is reduced due to the mismatch between the polymer and SiOx layer (Baek et al., 2017; Liu et al., 2018; Ma et al., 2018; Shuai et al., 2017; Zeng et al., 2019).

4.1.2. Performance

4.1.2.1. Evaluation of pressure sensitivity

4.1.2.1.1. Porous dielectric layer. Incorporating porosity within the dielectric layer decreases the modulus of elasticity and enhances the pressure sensitivity (Fig. 4a) (Kwon et al., 2016). The effective dielectric constant is a result of the combined influence of the sensor's air and polymer dielectric properties (Yoon et al., 2017a). When pressure is applied, permittivity increases as air is compressed, leaving behind a solid dielectric layer with a higher dielectric permittivity than air. The pore size and spatial density significantly impact capacitive sensors (Fig. 4b) (Kang et al., 2016; Mu et al., 2018; Pruvost et al., 2019; Yoon et al., 2017a). Increasing pore size augments the overall porosity of the dielectric layer, reducing its stiffness and consequently improving sensitivity. However, excessive porosity can undermine sensor performance by introducing high hysteresis, low modulus, and mechanical instability (Xiong et al., 2020).

Capacitive pressure sensors with a porous dielectric layer provide an extensive working range due to the thick dielectric material (Xiong et al., 2020). Nonetheless, pressure sensitivity is typically low (Chen et al., 2016; Han et al., 2020; Jeon et al., 2020; Kwon et al., 2016; Pruvost et al., 2019; Wang et al., 2019b; Yang et al., 2019c). Furthermore, such foam-type capacitive pressure sensors exhibit low baseline capacitance, making them susceptible to noise (Mannsfeld et al., 2010). Noise sources include parasitic and electromagnetic interferences originating from the body (Li et al., 2017; Zhu et al., 2018). To enhance sensor sensitivity, an effective strategy is to increase the material's dielectric constant by doping it with high-dielectric-constant materials (Fig. 4c) (Chhetry et al., 2020; Choi et al., 2020; Cui et al., 2016; Jang et al., 2016; Liu et al., 2018; Mu et al., 2018). Various dopants, such as AgNWs, Ag nanoparticles, carbon black, $CaCu_3Ti_4O_{12}$, $BaTiO_3$, MXene, and others, elevate the dielectric permittivity of the capacitive pressure sensor (Choi et al., 2020; Sharma et al., 2020). The permittivity increases due to enhanced polarization at the interface, thus forming local micro-capacitors (Pruvost et al., 2019). However, as the filler concentration rises up to a certain level, the capacitive behavior declines (Cui et al., 2016). Additionally, incorporating fillers raises the polymer's modulus of elasticity, thereby reducing the sensitivity derived from the softness of the dielectric material (Wei et al., 2019). As a result, determining the optimal concentration is crucial for maximizing sensitivity.

4.1.2.1.2. Microstructured dielectric layer/electrodes. Alternatively, micro structuring the electrodes or the dielectric layer in capacitive sensors effectively enhances the sensing performance (Huang et al., 2020). Periodic microstructures enable controlled deformation of the distance between the electrodes due to reduced stiffness on the substrate surface, resulting in improved sensor sensitivity (Bai et al., 2020; Guo et al., 2019a; Kwon et al., 2016; Li et al., 2020; Luo et al., 2019; Xiong et al., 2020; Yang et al. 2019a, 2019b). Various studies have investigated different microstructures in capacitive pressure sensors, such as the micro pyramid (Cheng et al., 2018; Huang et al., 2020), micro dome (Mahata et al., 2020), microcylinder (Yang et al., 2019a), micro/-nanopillars (Guo et al., 2019a), tilted micropillars (Luo et al., 2019), and micro tower (Wan et al., 2018a). Regular microstructures like micro pyramids efficiently distribute stress and saturate when flattened under increasing loads. By controlling the spatial density of the microstructures, substrate stiffness can be managed. For instance, substrates with dense microstructures exhibit low-pressure sensitivity due to increased surface stiffness (Cheng et al., 2018; Li et al., 2020). Cho et al. presented a bimodal sensor with temperature and pressure sensing capabilities, where fewer microstructured pyramids led to better sensitivity (Fig. 4d) (Bae et al., 2018). The sensitivity for micropatterned capacitive sensors

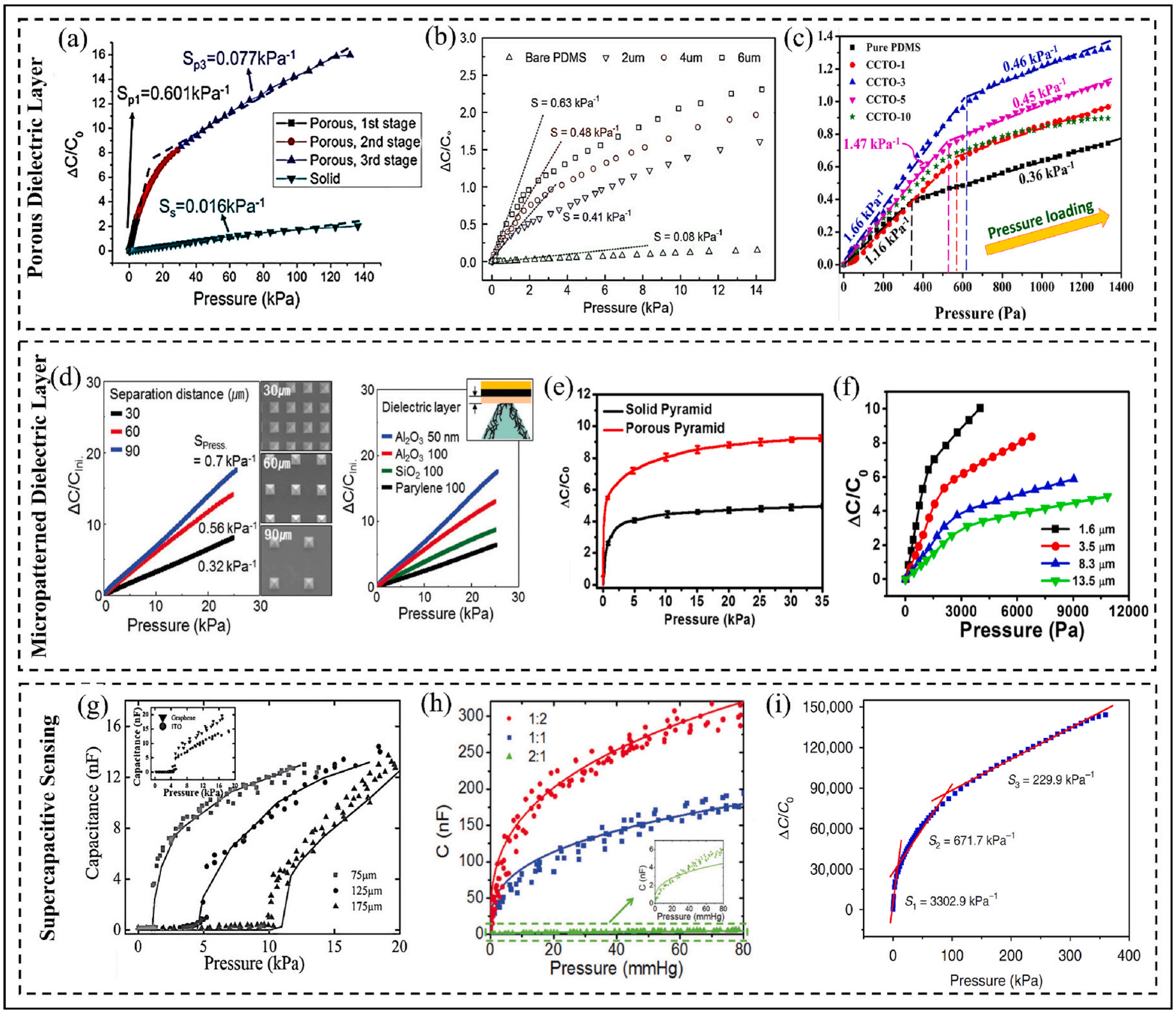

Fig. 4. The sensing performance of capacitive pressure sensor. (a) Capacitive sensor with porous dielectric layer shows sensitivity enhancement than planar dielectric layer (Kwon et al., 2016). (b) The sensitivity improvement through tuning the porosity (Kang et al., 2016). (c) Pressure sensitivity enhancement by improving the dielectric permittivity (Mu et al., 2018). (d) The effect of spatial density of micropyramid and the thickness of the dielectric layer on sensor performance (Bae et al., 2018). (e) The pressure sensitivity of capacitive pressure sensor with porous micro pyramid dielectric layer (Yang et al., 2019b). (f) Effects of the size of the surface microstructure on the sensing performance (Shuai et al., 2017). (g) Supercapacitive sensor with planar dielectric layer sandwiched between ITO/PET electrodes with nF-kPa^{-1} pressure sensitivity (Nie et al., 2015). (h) Supercapacitive pressure sensing with ionic nano fabric dielectric layer (Li et al., 2017). (i) Supercapacitive pressure sensor with graded intrafillable architecture using sandpaper mold (Bai et al., 2020).

drops at high pressure due to microstructures flattening. Therefore, the sensitivity shows two linear regions, one with high sensitivity and with low sensitivity. Capacitive sensors with micropyramid structures typically exhibit high sensitivity at the low-pressure ranges with a low linear sensing range. However, introducing hierarchical microstructures can improve linearity (Bae et al., 2018). Adjusting the dielectric layer's thickness is a favorable approach for enhancing capacitive sensor sensitivity (Fig. 4d) (Han et al., 2020; Huang et al., 2020; Li et al., 2020; Xiong et al., 2020). Sensors with thin device structures can achieve higher baseline capacitance and an elevated SNR (Mannsfeld et al., 2010; Shuai et al., 2017). Duan et al. reported a conductive micro-structured air-gap capacitive sensor with a thin Al_2O_3 dielectric layer, achieving a peak sensitivity of 770 kPa^{-1} (Huang et al., 2020).

Since creating a porous dielectric layer effectively reduces the polymer's elastic modulus, further microstructuring the polymer surface should increase capacitive pressure sensor sensitivity. A recent study suggests a porous micro pyramid structured capacitive sensor could achieve a peak sensitivity of 44 kPa^{-1} under 100Pa with an ultralow detection limit of 0.14Pa (Yang et al., 2019b). Due to the simultaneous presence of pores and surface microstructures, the sensor's sensitivity surpasses that of a sensor with a solid micro pyramid structure (Fig. 4e). The size of micrographic features also plays a crucial role in sensitivity, as depicted in Fig. 4f (Shuai et al., 2017). In addition to photolithography, researchers have utilized readily available templates such as tree leaves, sandpaper mold, and anodized alumina to create micropatterned sensing layer. However, one drawback of using these templates is the irregular order of microstructures, which can lead to reproducibility issues in fabrication (Ma et al., 2018).

4.1.2.1.3. Supercapacitive sensing. In recent years, supercapacitive or ionotropic sensing has led to the successful realization of highly sensitive sensors using an ionic gel as the dielectric layer (Bai et al., 2020; Chhetry et al., 2019; Cho et al., 2017; Li et al., 2017; Yoon et al., 2017b). The sensing mechanism allows for the accumulation of ultra-high areal capacitance due to the formation of electric double layers, resulting in exceptional pressure sensitivity in the nano-Farad range (Nie et al., 2015). This remarkable sensitivity can be achieved without relying on sophisticated micropattern fabrication techniques.

Nie et al. reported a capacitive pressure sensor using a super-capacitive sensing mechanism (Fig. 4g) (Nie et al., 2015). The sensor comprises two planar electrodes sandwiching an iontronic film and an airgap structure. This simple planar structure can achieve ultrahigh pressure sensitivity of 3.1 nF kPa^{-1}, which surpasses its solid-state counterparts, where pressure sensitivity falls within the pF-kPa^{-1} range. Li et al. presented a supercapacitive iontronic nano fabric pressure sensor, achieving a high-pressure sensitivity of 114 nF kPa^{-1}, significantly higher than many capacitive pressure sensors reported so far (Fig. 4h) (Li et al., 2017). The pressure sensor's sensing mechanism utilizes the deformation of the ionic nanofiber layer.

Supercapacitive sensors can achieve an extensive sensing range with excellent linearity due to the formation of an electric double layer. Guo et al. demonstrated a supercapacitive sensor using an ionotropic mechanism with graded intrafillable architecture, achieving a sensitivity window of 220-3300 kPa^{-1} (Fig. 4i) (Bai et al., 2020). The significant sensitivity resulted from the combined effect of graded architecture and electric double layers (EDL). The graded architecture was fabricated using a low-cost sandpaper template. This sensor displayed three regions of sensitivity: 3302.9 kPa^{-1}, 671.7 kPa^{-1}, and 229.9 kPa^{-1} at <10 kPa,10–100 kPa, and >100 kPa, respectively, making it one of the most successful capacitive sensors in terms of broad working range and sensitivity. They reported an ultra-high resolution of only 18Pa over the entire pressure range. However, the temperature and moisture dependence are inherent drawbacks of using an ionic gel as the dielectric layer (Nie et al., 2015; Yang et al., 2019b).

4.1.2.2. Response time, pressure resolution. Response time is a critical parameter for flexible pressure sensors used in physiological monitoring. These flexible sensors employ various polymer materials that exhibit viscoelasticity, resulting in a slow mechanical response. Viscoelastic behavior is particularly noticeable in capacitive pressure sensors, as capacitive sensing relies on the geometric changes and polymer materials always exhibit viscoelasticity (Ruth and Bao, 2020). Typically, a capacitive pressure sensor with a solid polymer dielectric layer has a long recovery time after pressure application due to polymer's viscoelasticity. The introduction of pores reduces the viscoelastic property of the polymer material, thereby decreasing the response time (Kang et al., 2016). Fig. 5a and b illustrate the response times of two capacitive pressure sensors with porous dielectric layers (Sharma et al., 2020; Choi et al., 2020). A sensor featuring electro-spun nanofiber has a rise and fall time of 0.15s, while a sensor with a stable porous structure obtained from the sacrificial template method exhibits a similar response time of 0.1s. However, capacitive pressure sensors with porous dielectric layers can still present significant hysteresis (Kang et al., 2016). This may result from the bulk polymer layer that significantly contributes to the dielectric layer's viscoelastic behavior.

Ultrafast response time can be achieved by creating surface microstructure. When pressure is applied, only the surface microstructures are compressed compared to the porous dielectric layer, significantly improving response time. Fig. 5c displays a quick rise time of 21ms and a fall time of 22ms for a micro pyramid structured supercapacitive pressure sensor (Yang et al., 2019b).

High-resolution pressure detection is another essential factor for identifying intricate features of pulse waveforms. Fig. 5d–f shows the pressure responses of various capacitive sensors with electro-spun nanofiber dielectric layer (Fig. 5d), a porous micropyramid based dielectric layer (Fig. 5e), and micro pyramid structured supercapacitive dielectric layer (Fig. 5f). The figures demonstrate that the relative change in capacitance varies significantly among different sensors. For instance, a relative capacitance changes of less than 0.75 could be achieved under 1.5 kPa pressure for the sensor with an electro spun nanofiber dielectric layer (Fig. 5d). In contrast, a capacitive sensor with

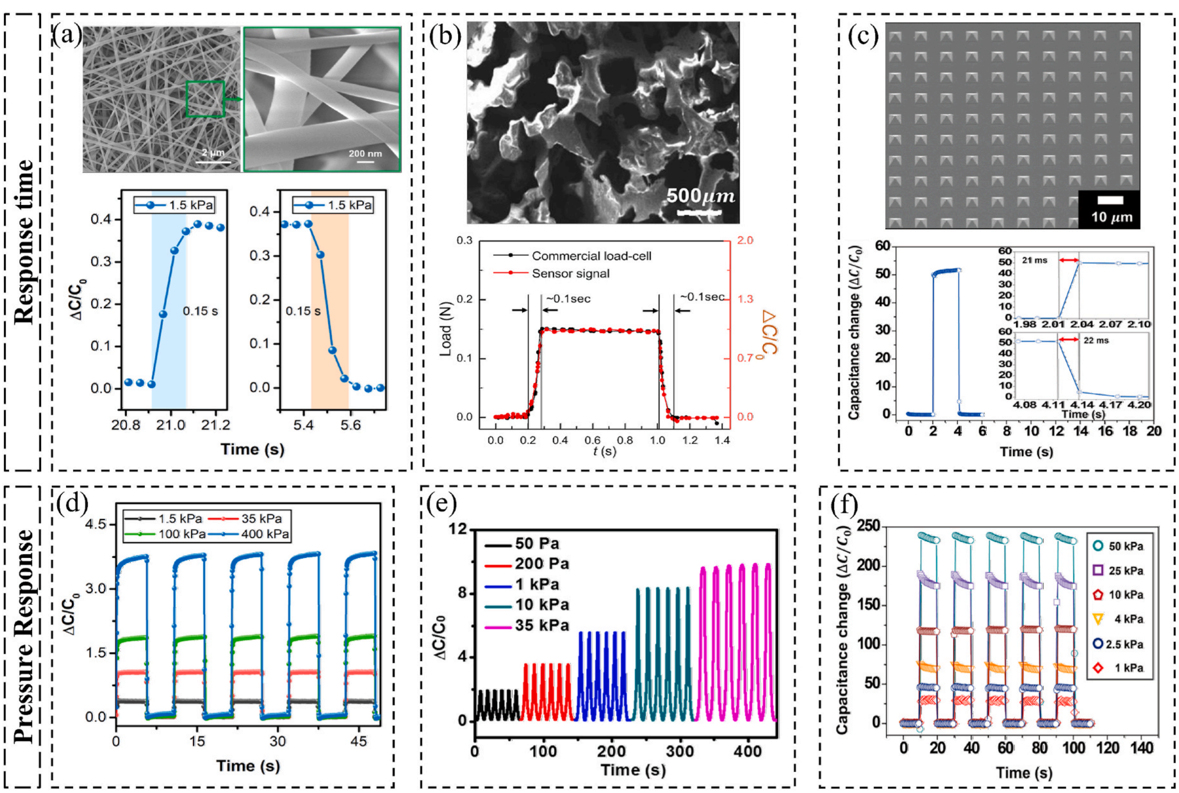

Fig. 5. Pressure response of capacitive sensor with several types of dielectric layer/electrode. (a) A capacitive pressure sensor with an electro spun nanofiber-based dielectric layer shows a total response time of 0.3s (Sharma et al., 2020). (b) The response time of a capacitive pressure sensor with a porous dielectric layer obtained from the sacrificial template method shows a response time of 0.2s (Choi et al., 2020). (c) A capacitive pressure sensor with micropatterned ionic dielectric layer shows a fast response time of 43ms (Cho et al., 2017). (d) Capacitive pressure sensor with electro spun nanofiber-based dielectric layer shows moderate pressure resolution under different pressure (Sharma et al., 2020). (e) Improved pressure resolution with porous micropyramid dielectric layer-based capacitive sensor (Yang et al., 2019b). (f) The pressure resolution of a micropatterned supercapacitive pressure sensor shows a very high-pressure resolution compared to the other sensors (Cho et al., 2017).

a porous micropyramid dielectric layer can achieve a relative change in capacitance of 6 under 1 kPa pressure (Fig. 5e). As a result, the latter sensor can resolve the pressure better than the former. However, Fig. 5f shows that under 1 kPa pressure, the micropatterned supercapacitive pressure sensor can achieve a relative change in capacitance of 25, surpassing both electrospun nanofiber-based sensor and MWCNTs functionalized eco flex sensor. Therefore, micropatterned supercapacitive sensors have superior pressure sensing capabilities and better resolution than the nanofiber-based sensors and porous micropyramid dielectric layer-based sensor. Table 1 summarizes recent progress in the field of capacitive pressure sensors.

4.2. Piezoresistive pressure sensor

4.2.1. Materials and fabrication

4.2.1.1. Piezoresistive materials. In recent years, piezoresistive pressure sensors based on micro/nanostructured materials have been researched. Among these materials, conductive polymers hold immense potential for developing piezoresistive pressure sensors due to their physical durability, excellent conductivity, and biocompatibility. However, they also face challenges such as unstable mechanical responses and low sensitivity (Lou et al., 2016). The incorporation of conductive nanofillers can enhance the properties of these inherent conductive polymers, ultimately improving pressure sensor performance (Lou et al., 2016).

Conductive polymers are created by functionalizing various biocompatible polymers like PDMS (Ai et al., 2018; Cheng et al., 2020; Li et al., 2016b; Wang et al., 2016), polyurethane (Chang et al., 2018; Lee et al., 2016b; Nie et al., 2017; Wei et al., 2017a), sponge (Zhang et al., 2016), polyimide (Huang et al., 2018; Jeong et al., 2020), polyvinyl difluoride (PVDF) (Costa et al., 2017; Kong et al., 2020), ecoflex (Lee et al., 2019a), polylactic acid (Guo et al., 2019b), poly(3, 4-ethylenedioxythiphene)-poly(styrenesulfonate) PEDOT: PSS (Eom et al., 2017; Jang et al., 2019; Wang et al., 2019a), polyglycerol sebacate (Yan et al., 2017), and others using 0D, 1D, and 2D conductive nanofillers (Boland et al., 2016; Lou et al., 2016; Zhang et al., 2018). CNTs, graphene, metal nanowires, ZnO nanowires, carbon black, and other materials significantly affect the charge transport properties of conductive polymers (Wei et al., 2017b).

Apart from conductive polymers, ultralight cellular materials can serve as sensing materials, offering a combination of softness and elasticity (Qiu et al., 2016). Various ultralight cellular materials have been studied, including metallic and ceramic lattices, carbon nanotubes aerogels, graphene/graphite foams, and carbonaceous nanofibrous aerogels (Qiu et al., 2016; Si et al., 2016). Recently, conductive textile materials have gained popularity as they can be used for energy storage, strain sensors, and health care devices, making them a promising material for resistive pressure sensor fabrication (Liu et al., 2017; Ren et al., 2017). Textile-based pressure sensors employ different nanofillers, such as CNTs (carbon nanotubes), graphene, graphene/polymer nanocomposites, carbonized silk fabrics, electrospun nanofibers, and regular cloth coated with carbonaceous materials (Liu et al., 2017).

4.2.1.2. Electrode materials. Based on the packaging, piezoresistive sensors can employ various types of electrodes, including two planar electrodes sandwiching a sensing layer (Li et al., 2019d), pair of interdigitated electrodes on a flexible substrate carrying the sensing material (Li et al., 2019d), microstructured electrode fabrication (Jeong et al., 2020), among others. A thin metal film deposited on a flexible substrate, such as PDMS serves as one of the simplest and most effective electrodes for piezoresistive pressure sensors (Lee et al., 2016c). Different metallic thin films, like gold and platinum films, have been utilized in creating electrodes for the piezoresistive pressure sensors (Chun et al., 2016; Lee et al., 2016c; Luo et al., 2016).

Conductive polymers are suitable for creating stretchable electrodes for resistive sensors (Choong et al., 2014). Conductive polymers can be functionalized with conductive nanofillers beyond the percolation threshold to produce highly conductive electrodes. Various 0D, 1D, and 2D nanomaterials can be easily employed as filler materials. As mentioned earlier, conductive textiles have gained popularity for developing pressure sensors. Conductive fabrics such as Al-fabric with different composites like woven Au–Ni, Ag knitted, and electro-spun

Table 1
Recent advancements in capacitive pressure sensors with different approaches.

Type	Dielectric Layer	Material Structure	Sensitivity (kPa^{-1})	Response Time (ms)	Working Range (kPa)	Ref.
Parallel Plate	Ecoflex, MWCNT	Porous Dielectric Layer	6.42 (0–2 kPa)	Not Available	0–10	(Choi et al., 2020)
	PDMS, CB	Porous Dielectric Layer	35 (0–0.2 kPa)	Not Available	0–11	(Pruvost et al., 2019)
	PDMS, CCTO	Porous Dielectric Layer	1.66 (0–640 Pa)	33	0–1.4	(Mu et al., 2018)
	MXene, PVDF, PEDOT: PSS	Nanofibrous Dielectric Layer	0.51 (0–1 kPa)	300	0–400	(Sharma et al., 2020)
	PDMS	Micropatterned electrodes with porous dielectric layer	2.51 (0–12 kPa)	201	0–30	(Zhang et al., 2021)
	PDMS	Micropyramid dielectric layer with porosity	44.5 (0–0.1 kPa)	50	0–35	(Yang et al., 2019b)
	PDMS	Micropatterned Dielectric Layer from Rose Petal Template	.055 (0.5–10 kPa)	350	0–12	(Mahata et al., 2020)
	PVDF	Micropatterned Electrode with Thin Dielectric Layer	30.2 (0–130 Pa)	75	>9	(Xiong et al., 2020)
	Parylene C	Micropatterned Electrode with Thin Dielectric Layer	70.6 (0–50 Pa)	Not Available		(Li et al., 2020)
	PVDF-TrFE	Interlocked Nanocone Dielectric Layer	6.583 (0–100 Pa)	84	0–0.9	(Niu et al., 2020)
Supercapacitive	MXene/ionic liquid in PVA	Ionic Nanofibrous Membrane Dielectric Layer	5.5 (0–30 kPa)	163.2	0–250	(Sharma et al., 2021)
	PVDF, EMIM-TFSI	Micropatterned Supercapacitive Dielectric Layer	41 (0–400 Pa)	43	0–50	(Cho et al., 2017)
	PVA, H_3PO_4	Supercapacitive Hair-like Microstructure	2296 (0–4 kPa)	85	100	(Pan et al., 2022)
	PVDF, EMIM-TFSI	Micropatterened Supercapacitive Dielectric Layer	131.5 (<2 kPa)	114	0–30	(Chhetry et al., 2019)
	MXene, EMIM-TFSI, PVDF	Nonwoven fabric	31.40 (0–10 kPa)	90	0–80	(Sun et al., 2022)
	$BMIM-PF_6$	Iontronic fabric	6.5 (0–10 kPa)	60	0–175	(Lin et al., 2020)

poly(styrene-*block* -butadiene- *block* -styrene) nanofiber mat decorated with Ag nanoparticles, can be excellent choices for electrodes formation (Choong et al., 2014). These textile-based electrodes have rough surfaces, which contribute to higher sensitivity.

4.2.1.3. Fabrication

4.2.1.3.1. Porous sensing layer. Various techniques can be employed to create a porous piezoresistive sensing layer. The sacrificial template method is an effective approach for generating a porous sensing layer, where materials can be functionalized with different 0D, 1D, and 2D conductive nanomaterials to achieve piezoresistive effects. Moreover, commercial foam materials, such as polyimide foam and melamine foam, can be functionalized by dip coating in conductive nanofiller dispersions. Another method involves pyrolyzing polymeric foam to create a porous conductive sensing layer, which achieves piezoresistive effects. Electrospinning serves as a controlled method for producing a porous sensing layer with adjustable thickness, where the thickness can be regulated by altering the deposition time. Alternative fabrication methods for generating a porous sensing layer include freeze-drying, metal oxide frameworks, baking-inspired foaming, among others.

4.2.1.3.2. Micropatterened sensing layer. Fabrication of micropatterned sensing layers for piezoresistive sensors follows a similar approach to that of capacitive pressure sensor. However, for piezoresistive pressure sensors, polymer materials are functionalized with different conductive nanofillers to achieve piezo resistivity (Wang et al., 2016).

Contrary to the capacitive pressure sensors, the performance of piezoresistive sensors may depend on the design of the electrodes. Numerous studies have investigated plane, interdigital, and micropatterned electrodes for realizing high-performance piezoresistive pressure sensors (Mannsfeld et al., 2010). Various techniques can be employed to achieve different interdigital patterns and microstructured electrodes, such as standard photolithography techniques. In addition to photolithography, several other methods for realizing interdigitated electrodes include using a shadow mask for depositing patterned metallic electrodes, employing a laser writer, using a stencil to screen print the conductive ink on the flexible substrate, and utilizing a laser printer to print conductive ink on the substrate.

4.2.2. Performance

4.2.2.1. Evaluation of pressure sensitivity

4.2.2.1.1. Porous sensing layer. Porous polymers functionalized with various 0D, 1D, and 2D conductive nanofillers have been demonstrated to be beneficial for developing low-cost piezoresistive sensors with moderate sensitivity (Ge et al., 2018; Li et al., 2016b). Nanofillers such as carbon nanotubes (CNTs), carbon black (CB), graphene, metal nanowires, etc. are added to the spongy backbone of the polymer. The sensing mechanism for porous piezoresistive sensors can rely on microcrack formation or increasing contact points between the conductive backbones of the sensing material (Wu et al., 2016). Conductive paths increase when a force is applied, allowing more current to flow under a fixed voltage (Ding et al., 2018). Although adding conductive filler to the polymer matrix increases the stiffness of the polymer, controlling the pre-polymer mixing ratio can help manage the stiffness (Luo et al., 2017).

Achieving high sensitivity requires tuning the resistivity by optimizing the concentration of conductive fillers to produce high signal output at low pressure. Ideally, these conductive nanofillers are mixed with the polymer matrix according to the percolation threshold, as shown in the following formula:

$$\sigma = \sigma_0 (m - m_c)^t \quad (1)$$

where m represents the volume fraction of the conductive filler, m_c is the percolation threshold, and t is the critical exponent determined by the dimension of the conductive network. The percolation threshold is a critical concentration of filler material at which a conduction path is formed, turning insulating material into conductive material. If the polymer material achieves excessive conductivity, the sensor output will saturate at a low-pressure limit. The interaction between the porous/spongy sensing material and the electrodes can further improve the sensing performance of the piezoresistive pressure sensor.

Piezoresistive sensing performance can be achieved by sandwiching the porous sensing layer between two flat electrodes or placing the sensing material on interdigital electrodes. The latter shows better performance according to the literature; however, research comparing both electrode types is necessary to better understanding performance. In addition to placing the sensor on top of the interdigital electrodes, studies show that having some insulating spacer between the electrodes and the sensing material can further increase pressure sensitivity. For example, Gao et al. reported an MXene functionalized melamine sponge with high sensitivity of 147 kPa^{-1} below 5.37 kPa and 442 kPa^{-1} within the 5.37–18.56 kPa pressure range *(Yue et al. 2018)*. This high sensitivity arises from non-conductive PVA nanowire spacers that act as a barrier between the conductive MXene sheets (Fig. 6a). The sensor responds linearly to applied pressure with two straight slopes. Similar studies have also been conducted by other groups, showing comparable performance. A drawback of using porous material for piezoresistive sensor fabrication is inferior recoverability, poor mechanical properties, and reduced mechanical durability *(Luo et al. 2017)*.

4.2.2.1.2. Microstructured sensing layer. Utilizing micro/nanostructures on sensing materials can lead to significant variations in resistance for piezoresistive sensors (Kim et al., 2018). By incorporating these microstructures, air gaps are created that reduce the contact area between the opposing substrates and lower the polymer's stiffness at contact points. This allows for easier deformation under pressure, resulting in improved sensitivity, faster response time, and enhanced reproducibility (Kim et al., 2018; Lee et al., 2016c; Li et al., 2018a). Researchers have conducted numerous studies on different microstructures and their effects on piezoresistive sensors, such as micro pyramids (Chang et al., 2018; Jeong et al., 2020; Park et al., 2018), nanorods (Suen et al., 2018), micro dome (Bae et al., 2016; Park et al., 2018), semi-spheroids (Zhong et al., 2016), microcylinder (Peng et al., 2018), and micropillars (Park et al., 2018). Micro pyramid and micro dome structures are particularly popular among the research community for their advantages in the fabrication process and their ability to achieve good sensitivity among various microstructures.

Previous research has demonstrated that the spatial density and size of microstructure play a crucial role in sensitivity. Sensors with a high spatial density of microstructures tend to have lower sensitivity due to increased surface stiffness, and vice versa (Bae et al., 2016). However, the viscoelastic property of the sensor increases with a decreased spatial density of microstructures. It is essential to determine the optimal spatial density of microstructures to achieve the best performance from the sensor. A comparison of studies on the effects of different microstructures reveals that sensors with micro dome surface features can achieve the most significant change in the area between the electrodes and the sensing material (Jafarizadeh et al., 2022; Park et al., 2018). This is due to localized stress concentration, which results in the highest-pressure sensitivity.

Several approaches can be used to package piezoresistive pressure sensors with micropatterned structures, such as micropatterned sensing layer-flat electrode layouts, micropatterned sensing layer interlocked structures, and micropatterned sensing layer-interdigital electrodes. The presence of roughness on the electrodes can further increase pressure sensitivity (Chen et al., 2019; Park et al., 2018). Yang et al. demonstrated this concept with their piezoresistive pressure sensor, achieving a pressure sensitivity of 259.32 kPa^{-1} (Chen et al., 2019).

One limitation of piezoresistive pressure sensors with micropatterned sensing layers is the saturation of the pressure sensitivity in low-pressure regions due to the smaller size of the micropatterns (Wang

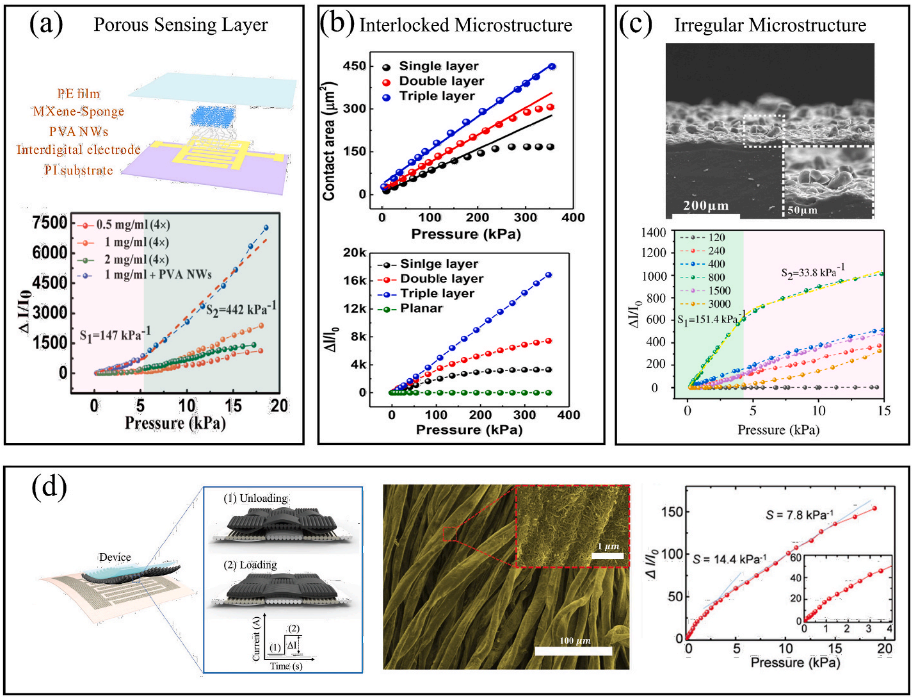

Fig. 6. The figure shows the performance of piezoresistive pressure sensors with different approaches. (a) The sensing performance of a piezoresistive sensor with porous sensing material separated by PVA insulating nano spacer from the interdigitated electrodes (Yue et al., 2018). (b) Performance of piezoresistive pressure sensor with multilayer interlocked micro dome sensing layer (Lee et al., 2018). (c) The sensing performance of piezoresistive pressure sensor with sandpaper templated sensing layer (Cheng et al., 2020). (d) Piezoresistive pressure sensor with conductive textile/IDE-based sensing mechanism (Liu et al., 2017).

et al., 2016). Researchers have explored alternative approaches to maintain the high-pressure sensitivity over a broad working range, such as using the Mayan pyramid structure (Jung et al., 2019), increasing the number of interlocked sensing layers (Lee et al., 2018), and employing hierarchical microstructures (Bae et al., 2016). Ko et al. reported a ferroelectric sensor with multiple layers of interlocked micro dome geometry and could achieve a linear sensitivity of 47.7 kPa^{-1} over an extended pressure range of 353 kPa (Fig. 6b) (Lee et al., 2018).

In addition to conventional techniques, nature-derived templates like tree leaves (Jian et al., 2017), sandpaper template (Cheng et al., 2020; Pang et al., 2018; Wang et al., 2016), colloidal self-assembled PS template (Zhang et al., 2017b), breath figure method (Wang et al., 2018), shark skin template (Jang et al., 2019), nylon mesh template (Wang et al., 2017a), banana leaf template (Nie et al., 2017), paper-based sensor (Pang et al., 2018; Zhan et al., 2017), and sand-blasting silicon technique (Kim et al., 2017), have proven successful in realizing high-performance piezoresistive pressure sensors. For example, Cheng et al. demonstrated a sandpaper-molded sensor fabrication technique using MXene as the sensing material, achieving a very high-pressure sensitivity of 151.4 kPa^{-1} below 5 kPa and maintaining an impressive sensitivity of 33.8 kPa^{-1} beyond 5 kPa (Fig. 6c) (Cheng et al., 2020).

Another promising non-conventional approach involves using textile or fabric-based materials to develop piezoresistive sensors (Ge et al., 2016; Li et al. 2019a, 2019b; Liu et al., 2017; Zhang et al., 2018). This method could potentially lead to the creation of smart textile for futuristic clothing applications. Wang et al. reported a textile-based piezoresistive sensor with a high sensitivity of 14.4 kPa^{-1} (Fig. 6d) (Liu et al., 2017). In their study, cotton was dip-coated with CNT to form conductive textile, and an interdigitated electrode was fabricated as a current collector for the sensor. Moreover, the sensor could maintain a linear sensitivity up to a high-pressure range of 20 kPa. The researcher successfully incorporated the sensor onto cloth, creating a smart wearable textile.

In summary, various microstructures and innovative fabrication techniques have been explored to enhance the performance of piezoresistive pressure sensors. These advancements have resulted in improved sensitivity, faster response time, and better reproducibility. By understanding the complex relationships between microstructure size and spatial density, researchers have been able to optimize sensor performance. Additionally, unconventional approaches using nature-inspired templates and textile-based materials have opened new avenues for the development of high-performance piezoresistive pressure sensors with potential applications in wearable technology and smart textiles.

4.2.2.2. Response time and pressure resolution. Piezoresistive sensors based on contact resistance can achieve a fast response time. However, a lag exists between the input pressure and output signal due to the viscoelastic property of the polymer (Qiu et al., 2016). 3D nanostructured cellular network-based elastomers, such as graphene foam/aerogel, metallic and ceramic lattices, and carbon nanotubes aerogels, provide an excellent solution for achieving ultrafast response times under dynamic conditions (Qiu et al., 2016; Si et al., 2016). This exceptional dynamic response is attributed to the cellular nanostructure of these materials. In addition to cellulose nanostructures, a metal oxide framework-based porous piezoresistive sensor with high sensitivity was recently reported (Zhao et al., 2018). Due to the porosity of the sensing layer, the sensor achieved a fast response time of 60ms (Fig. 7a), enabling it to accurately monitor arterial pulse waveforms.

Sharp microstructures can reduce the viscoelastic behavior of the polymer and improve the response time. The reduced response time results from the effects of the micropatterned surface and the presence of air filling the space, significantly decreasing the viscoelastic drag on the surface. For example, Li et al. reported a pyramid microstructured PDMS/PPy-based pressure sensor with a sharp tip, achieving an ultrafast response time of 50 μs (Fig. 7b) (Li et al., 2018a). As previously mentioned, piezoresistive pressure sensors fabricated using functionalized nature-derived materials also exhibit excellent sensing performances. For instance, Fig. 7c shows the response time of a tissue paper-based piezoresistive pressure sensor functionalized by MXene to monitor physiological activity (Zheng et al., 2021).

Piezoresistive pressure sensors with high-pressure sensitivity can provide good pressure resolution. Recently, multilayered sensing layers with microdome features demonstrated remarkable linear sensing ranges and high-pressure sensitivity (Lee et al., 2018). Fig. 7d indicates

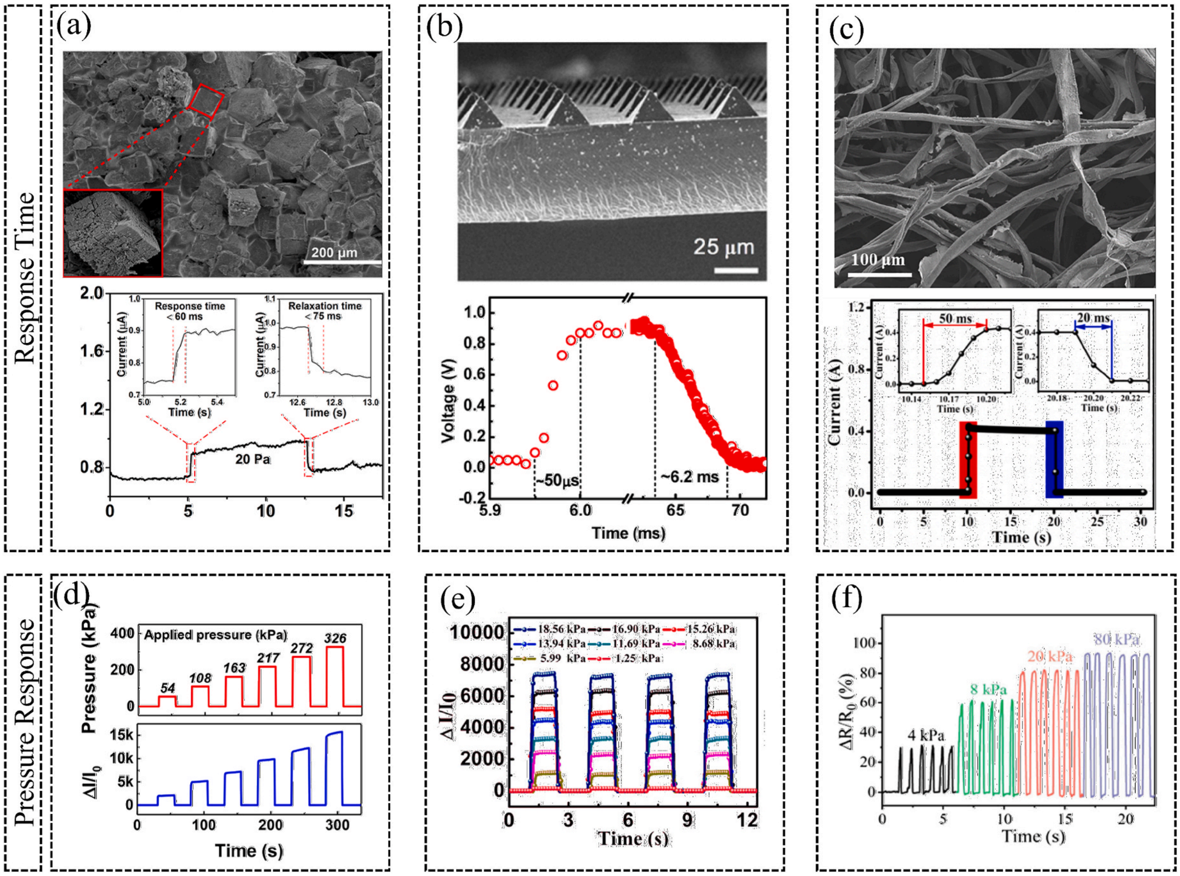

Fig. 7. Pressure response of piezoresistive pressure sensor with different fabrication methods. (a) A metal oxide framework-based piezoresistive pressure sensor shows a response time of 60ms under a 20Pa load (Zhao et al., 2018). (b) An ultrafast response time of 50 μs was achieved from a piezoresistive pressure sensor with a sharp micro pyramid structure and short channel length (Li et al., 2018a). (c) MXene functionalized cotton fabric-based piezoresistive pressure sensor shows a response time of 50ms (Zheng et al., 2021). (d) A linear pressure resolution was achieved with a multilayer interlocked sensing layer showing excellent pressure resolution up to 326 kPa (Lee et al., 2018). (e) Piezoresistive sensor with porous sensing layer and nano space-based approach shows a high-pressure resolution up to a broad pressure range (Yue et al., 2018). (f) Pressure resolution with polyester fabric functionalized by thiolated graphene shows moderate pressure resolution up to 80 kPa (Zhang et al., 2018).

that the sensor also offers good pressure resolution with a high signal level. This broad linear sensing range is advantageous for various applications, ranging from subtle pressure detection (arterial pulse waveform detection) to high-pressure sensing (tactile sensing). Due to the extensive linear sensing range, the sensor can effectively resolve pressure up to 328 kPa. Fig. 7e displays the pressure response of a piezoresistive sensor with a porous sensing layer functionalized by 2D materials (Yue et al., 2018). The sensor uses interdigitated electrodes with insulating nano-spacers, resulting in a significant relative change in current along with good pressure resolution. Lastly, Fig. 7f shows the pressure response of a polyester textile functionalized by a thiolated graphene pressure sensor (Zhang et al., 2018). Although the pressure sensor has two linear sensing ranges, the pressure response is not as high as that of the other two pressure sensors. However, the straightforward fabrication process may be appealing for low-cost sensor development for cardiovascular monitoring. Table 2 presents several reports on recent advancements in piezoresistive pressure sensors.

4.3. *Application of pressure sensor for pulse waveforms detection*

4.3.1. *Capacitive pressure sensors*

Linearity is a critical parameter for capacitive pressure sensors in detecting pulse waveforms. A linear response ensures that the sensor's output is proportional to the input pressure, providing accurate and consistent measurements across a range of pressure values. Non-linear sensors may result in inaccurate detection and analysis of the arterial pulse waveform.

Fig. 8a illustrates the detection of arterial pulse waveforms with a capacitive pressure sensor having a porous dielectric layer (Choi et al., 2020). The sensor effectively detected pulse waveforms at rest and during activity. Although capacitive pressure sensors with porous dielectric layers have low pressure sensitivity, resulting in poor SNR, sensitivity enhancement techniques can significantly improve performance. The pressure sensitivity of the capacitive pressure sensor in Fig. 8a was enhanced by adding MWCNTs below the percolation threshold, increasing the dielectric permittivity. Consequently, the sensor effectively detected weak pulse waveform signals with high signal output, and major peaks were easily discernible. The linearity of the sensor was also improved due to the increased dielectric permittivity, enabling accurate and reliable pulse waveform detection.

Fig. 8b presents a pulse waveform obtained from a capacitive pressure sensor with a porous micropyramid dielectric layer (Yang et al., 2019b). The micropatterned sensor's signal acquisition capability is several times better than the sensor with a porous dielectric layer. Micropatterns on the surface reduce the young's modulus at the surface, making it easier to deform under small pressure. Furthermore, pores within the micropatterned dielectric layer further decrease the young's modulus of the sensor, resulting in high pressure sensitivity. As a result, the sensor effectively detected arterial pulse waveforms with high SNR.

Finally, Fig. 8c shows wrist arterial waveform detection using a supercapacitive pressure sensor (Cho et al., 2017). High-pressure sensitivity can be achieved with supercapacitive pressure sensing due

Table 2
Recent advancements in piezoresistive pressure sensor.

Type	Sensing Material	Material Structure	Sensitivity (kPa^{-1})	Response Time (ms)	Working Range (kPa)	Ref.
Interdigital Electrodes	MXene	Sandpaper Molded Micropatterned Sensing Layer	151.4 (<4.7 kPa)	229	0–15	(Cheng et al., 2020)
	GNP	Silk Molded Micropatterned Sensing Layer	1875.3 (0–20 kPa)	1.3	0–40	(He et al., 2019)
	PEDOT: PSS	Sandpaper Molded Micropatterned PDMS	851 (0–3 kPa)	0.15	0–20	(Wang et al., 2016)
	MXene, PVA nanospacers	Polymer Foam	147 (0–5.37 kPa)	138	0–18.56	(Yue et al., 2018)
	Graphite-flakes modified non-woven fabrics	Non-woven fabric/Silver coated fabric electrodes	6.4 (0–800 kPa)	4	0–800	(Xu et al., 2020)
	Graphene	Hierarchical Microstructure	8.5 (0–12 kPa)	70	0–12	(Bae et al., 2016)
	Carbonized Paper	Rough Microstructured Carbonized Paper	5.67 (0–0.42 kPa)	80	0–20	(Chen et al., 2018)
Interlocked Sensing Layer	rGO, PVDF	Multilayer Microdome Structure	47.7 (0–400 kPa)	60	0–400	(Lee et al., 2018)
	Graphene	Fingerprint like patterned PDMS	110 (0–0.2 kPa)	80	0–75	(Xia et al., 2018)
	CNT	Microdome Structure	47.062 (0–1 kPa)	24	0–26	(Park et al., 2018)
Flat Electrodes	PEDOT: PSS	Biomimicking Shark Skin	62.56 (0–0.7 kPa)	Not Available	0–6	(Jang et al., 2019)
	Porous Carbon	Metal Oxide Framework Derived Porous Carbon	15.63 (0–2 kPa)	135	0–2	(Zhao et al., 2018)
	Carbonized Silk Nanofiber	Nanofibrous Silk	34.47 (0–0.4 kPa)	16.7	0–5	(Wang et al., 2017b)

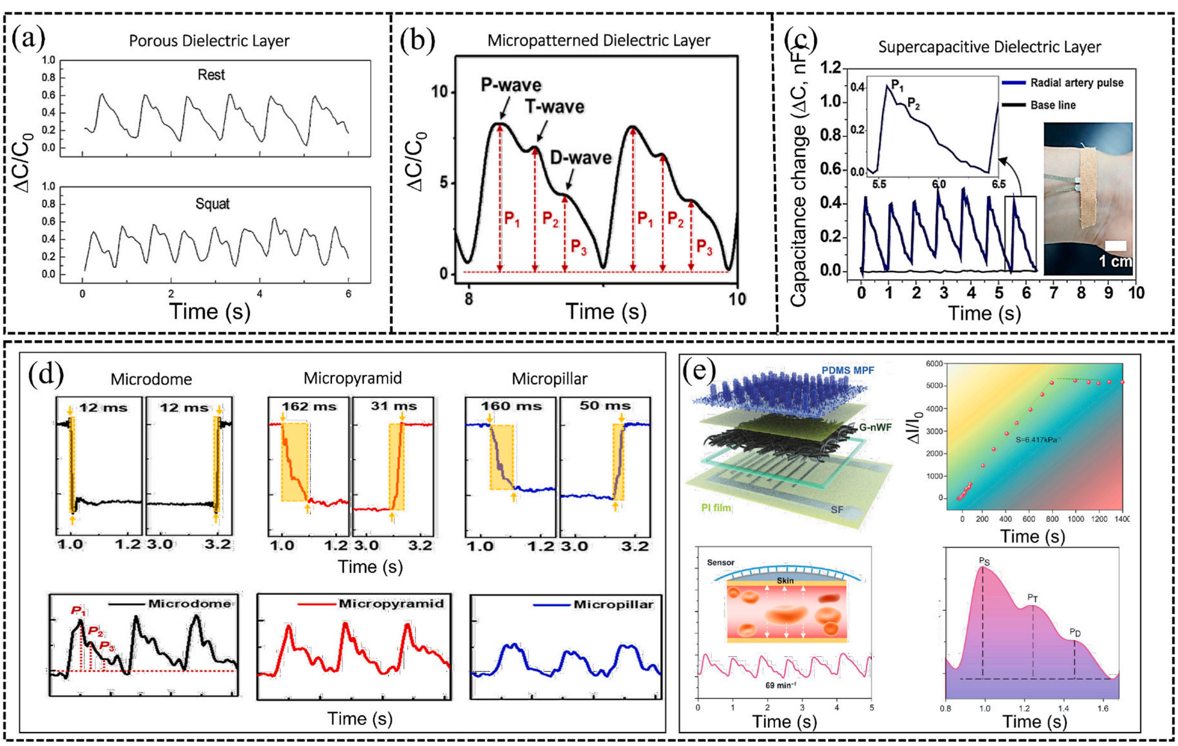

Fig. 8. Detection of wrist arterial pulse waveforms with capacitive and piezoresistive pressure sensors. (a) Wrist arterial pulse waveform using a capacitive pressure sensor with porous dielectric layer functionalized by MWCNTs (Choi et al., 2020). (b) The wrist arterial pulse waveform was acquired using a micropatterned dielectric layer with pores (Yang et al., 2019b). (c) A supercapacitive pressure sensor is employed to acquire wrist arterial pulse waveform (Cho et al., 2017). (d) Influence of different microstructure on the wrist arterial pulse waveform detection using piezoresistive pressure sensors (Park et al., 2018). (e) High quality pulse waveforms detection using conductive fabric based piezoresistive pressure sensor (Xu et al., 2020).

to the electric double layer formation. Consequently, the sensor effectively detected arterial pulse waveforms. The signal level is noticeably better than the noise level, and the pressure sensor obtained superior-quality continuous pulse waveform. The linearity of the supercapacitive pressure sensor is also improved due to the electric double layer formation, providing consistent and accurate measurements across a range of pressure values.

In summary, capacitive pressure sensors, with their high sensitivity, linearity, SNR, and broad pressure range, show great potential for detecting pulse waveforms. Different sensing mechanisms, such as porous dielectric layers, micropatterned surfaces, and supercapacitive pressure sensing, can be used to further enhance the sensor's performance in acquiring high-quality arterial pulse waveforms while maintaining linearity across various pressure values.

4.3.2. Piezoresistive pressure sensors

Fig. 8d and e shows arterial pulse waveform detection using piezoresistive pressure sensing. Fig. 8d presents the effects of various microstructures on the acquired pulse waveform (Park et al., 2018). According to the study, the micro dome feature achieved the most significant change in contact area, resulting in high-pressure sensitivity. The fast response time stems from the high reaction force due to a more significant change in contact area compared to other microstructures. The micro dome pressure sensor attained the fastest response time and high-pressure sensitivity, leading to exceptional quality pulse waveform detection. A micro pyramid sensor could also detect high-quality pulse waveforms, although its response time was slightly higher than the sensor with a micro dome structure.

Linearity is also a crucial characteristic for piezoresistive pressure sensor to detect arterial pulse waveforms accurately. A high linear sensing range is essential for obtaining a stable pulse waveform from the wrist artery and preventing waveform distortion due to non-linearity. In addition, a linear sensing range is necessary for maintaining sufficient sensitivity to detect pulse waveforms effectively under external pressure.

Xu et al. demonstrated a piezoresistive pressure sensor with a broad linear sensing range based on conductive fabric for pulse waveforms detection (Xu et al., 2020). The sensor achieved high sensitivity of 6.417 kPa^{-1} with a linear sensing range of 800 kPa, as shown in Fig. 8e. This wide sensing range resulted from the conductive non-woven fabric functionalized with graphene flakes and an interdigitated silver-coated fabric. Consequently, the sensor easily detected pulse waveforms while maintaining high stability and linearity. The high linearity and sensitivity also contributed to an improved signal-to-noise ratio, enabling more accurate detection of pulse waveforms.

In summary, piezoresistive pressure sensors, with their high sensitivity, linearity, SNR, and broad pressure sensing range, offer promising capabilities for detecting pulse waveforms. Different microstructures, such as micro domes and micro pyramids, can further enhance the sensor's performance in acquiring high-quality arterial pulse waveforms. Moreover, the use of conductive fabrics functionalized with graphene flakes and interdigitated silver-coated fabric can enable broad linear sensing ranges and higher sensitivity, resulting in more accurate and stable pulse waveform detection.

5. Machine learning perspective

5.1. Data acquisition and pre-processing

Data acquisition and pre-processing are crucial for analyzing cardiovascular pulse waveforms with flexible capacitive and piezoresistive pressure sensors. Data acquisition involves collecting raw signals and converting them into digital values using an analog-to-digital converter (ADC). A high sampling rate is necessary to avoid aliasing artifacts and ensure accurate waveform representation. Pre-processing techniques include filtering, normalization, and segmentation to enhance signal quality and reliability.

Motion artifacts pose a challenge when acquiring pulse waveforms, especially with wearable sensors. Adaptive filtering and combining data from multiple sensors, such as accelerometers or gyroscopes, can help address this issue. Assessing data quality is essential before feature extraction and machine learning stages, using metrics like signal-to-noise ratio (SNR) and root mean square error (RMSE).

In conclusion, effective data acquisition, pre-processing, and data quality assessment are vital for accurate and reliable analysis of cardiovascular pulse waveforms. These steps pave the way for successful machine learning-based analysis and prediction in cardiovascular health monitoring and disease diagnosis.

5.2. Feature extraction and selection

Feature extraction and selection are essential steps in analyzing cardiovascular pulse waveforms using machine learning techniques. These steps involve extracting meaningful features from pre-processed data and selecting the most relevant features for further analysis.

Feature extraction transforms pre-processed data into a set of features that effectively represent pulse waveforms. These features capture temporal, morphological, and frequency-domain characteristics of the signal. Methods like wavelet analysis, principal component analysis (PCA), and independent component analysis (ICA) are commonly used for this purpose.

Feature selection aims to identify the most relevant and informative features to reduce dimensionality, improve computational efficiency, and prevent overfitting. Various methods include filter methods, wrapper methods, and embedded methods, which evaluate features based on individual relevance, training machine learning models, or integrating feature selection within the learning process.

In conclusion, feature extraction and selection play vital roles in cardiovascular pulse waveform analysis using machine learning techniques. These steps help effectively represent signals and identify the most relevant features for subsequent analysis, ultimately improving the performance of machine learning models in predicting and diagnosing cardiovascular diseases.

5.3. Machine learning algorithms for pulse waveforms analysis

Machine learning algorithms have shown great potential in enhancing the accuracy and efficiency of cardiovascular health monitoring and disease diagnosis using pulse waveforms analysis. Commonly used algorithms can be categorized as supervised, unsupervised, and deep learning algorithms.

Supervised learning algorithms, such as Support Vector Machines (SVM), Decision Trees, Random Forests, and k-Nearest Neighbors (k-NN), are widely used when labeled data is available. These algorithms learn from training data to make predictions or classifications on new, unseen data.

Unsupervised learning algorithms, like clustering algorithms, are employed when labeled data is scarce or unavailable. They identify underlying patterns or structures in the data without explicit supervision.

Deep learning algorithms, such as Convolutional Neural Networks (CNN) and Recurrent Neural Networks (RNN), use artificial neural networks with multiple layers to learn hierarchical representations of the data. CNNs are effective in learning spatial features from signals, while RNNs capture temporal dependencies in time-series data.

In conclusion, various machine learning algorithms have demonstrated their potential in pulse waveforms analysis, improving cardiovascular health monitoring and disease diagnosis. The choice of an appropriate algorithm depends on the problem, data, and desired outcome. Continuous advancements in machine learning and growing data availability are expected to further enhance pulse waveforms analysis capabilities in the future.

5.4. Applications

5.4.1. Cuffless blood pressure monitoring

Traditional blood pressure monitoring systems rely on cuff-based auscultatory and oscillometric techniques (Fig. 9a) (Choi et al., 2013). These methods involve occluding arterial blood flow using an inflatable cuff, which is unsuitable for long-term use, such as 24-h ambulatory

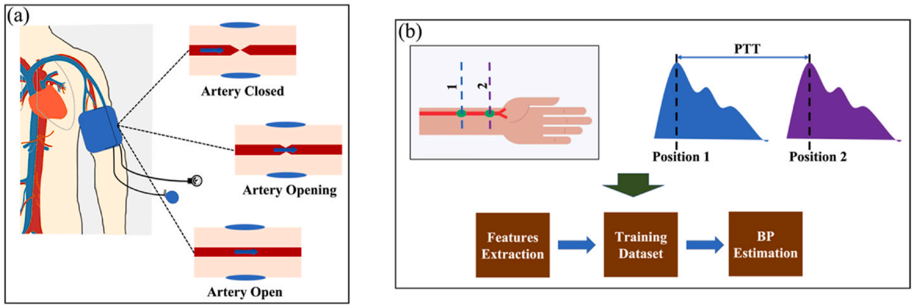

Fig. 9. (a) Cuff-based blood pressure measurement technique. (b) Cuffless blood pressure monitoring using machine learning based on pulse transit time.

blood pressure measurement (Rachim and Chung, 2019). As a result, researchers have sought alternative, non-invasive cuffless methods for blood pressure measurement.

One promising method involves using pulse transit time (PTT) as a surrogate for blood pressure (Fig. 9b) (Griggs et al. ; Liu et al., 2019; Choi et al., 2013). *PTT* is the time required for a pulse to travel between two arterial sites as shown in Fig. 9b (Choi et al., 2013; Ding et al., 2017). It has a negative correlation with blood pressure, meaning as blood pressure increases, PTT decreases and vice versa (Landry et al., 2020; Lee et al., 2019b). Various models have been developed to estimate blood pressure based on PTT, requiring calibration against standard cuff-based systems and utilizing machine learning algorithms for training and testing (Parry et al.).

Several machine learning-based algorithms have been tested for blood pressure estimation, including Regularized linear regression, Feedforward Neural Network, Decision Tree Regression, Non-linear Auto-Regressive Models with Exogenous Input (NARX-NN), Long-Short Term Memory Neural Network (LSTM-NN), Random Forrest Algorithm, Adaboost Regression, Support Vector Machine, Multi Adaptive Regression Spline (MARS), and many more (Geerthy Thambiraj and Uma Gandhi and Umapathy Mangalanathan and 2020; Kachuee et al., 2017; Kyriacou, 2021; Landry et al., 2020; Senturk et al., 2020; Sharifi et al., 2019).

However, the need for frequent calibration is a significant drawback of PTT-based blood pressure estimation (Ding et al., 2017; Huynh et al., 2019; Kachuee et al., 2017; Sharifi et al., 2019). Alternative approaches involve extracting exclusive features from pulse waveforms or combining both PTT and pulse waveform features. Miao et al. developed a multisensory approach for feature extraction from pulse waveforms to provide a robust and accurate model (Miao et al., 2020). They extracted 35 features from pulse waveforms and trained the data using weakly supervised feature selection.

Recent studies for cuffless blood pressure estimation primarily focus on PPG-based measurement techniques. Flexible pressure sensors can be an alternative to PPG-based pulse waveform detection due to their ability to acquire more intricate features of pulse waveforms. A limited number of studies on cuffless blood pressure estimation using flexible pressure sensors and pressure sensor arrays have shown promising results (Luo et al., 2016; Rachim et al., 2021). More research is needed to fully understand the potential of flexible pressure sensors for estimating blood pressure in a cuffless manner.

5.4.2. *Cardiovascular disease prediction*

Machine learning-based artificial intelligence has the potential to revolutionize traditional Chinese medicine (TCM) diagnosis methods. TCMs, which has been practiced in China for thousands of years, involves practitioners diagnosing patients based on the subtlety of their pulse. However, this method requires years of practice, and diagnoses may differ among practitioners. Machine learning, in conjunction with pulse waveforms detection devices, can help classify different pulse waveforms more effectively and consistently.

Various studies have explored machine learning in the context of TCM diagnosis. Tang et al. validated a TCM pulse diagnostic model with an artificial neural network (ANN), achieving a diagnostic accuracy of 80% (Tang et al., 2012). Guo et al. analyzed and recognized TCM pulse based on Hilbert-Huang Transform (HHT) and Random Forrest algorithm in patients with coronary heart disease, with the best recognition accuracy of 90.21% (Guo et al., 2015). Amaratunga et al. discussed the application of machine learning in hypertension studies using four case studies, involving deep learning and support vector machines to predict hypertension development over ten years (Amaratunga et al., 2020). One case study used photoplethysmography (PPG) to assess hypertension, classifying hypertensive and normotensive cases with high sensitivity and positive predictive value.

In recent years, TCM-based pulse analyses have also been applied to study pregnant women's pulses using machine learning. Chen et al. reported a deep learning approach to classify pregnancy using pulse pressure waves, achieving an accuracy of 84% (Chen et al., 2020). By incorporating flexible pressure sensors capable of capturing more intricate pulse waveform features, the potential for diagnosing cardiovascular diseases (CVDs) can be further explored. The application of machine learnings to TCM-based pulse diagnosis shows promise for early detection of cardiovascular diseases, with the potential to offer more accurate and consistent diagnoses than traditional methods.

6. Summary and outlook

Recent advancements in biomedical applications allow the usage of flexible sensors for advanced cardiovascular monitoring. Although the wrist is the most suitable location for gathering arterial pulse waveform, there are still many challenges in achieving a standalone device for continuous monitoring of the pulse waveform using flexible sensors. For instance, Flexible pressure sensors must be located directly on the wrist artery to get a high-quality pulse waveform. Therefore, placing the sensor directly on the wrist can be challenging for people. Although flexible pressure sensor does not rely on the skin tonality issue, unlike PPG, obesity can hamper gathering high-quality pulse waveform. The curvilinear structure of the skin is also a challenge towards conformal contact of the sensor on the skin. However, recent technologies allow the thin design of the sensor that can alleviate the skin conformability issue. Another challenge for an adequate sensor adaptation for gathering cardiovascular biomarkers is the noise associated with body movement. Since getting the signal from the wrist artery entails proper positioning of the sensor all the time, the movement makes it difficult to achieve. However, using a sensor matrix instead of a single-pixel sensor can rectify the problem. To obtain a high-quality pulse waveform, the sensors should have high sensitivity and broad pressure sensing range with linearity and fast response time. Both capacitive and piezoresistive pressure sensors can achieve high-pressure sensitivity with a wide pressure sensing range and quick response time. Sensitivity is the most critical parameter for a flexible pressure sensor to monitor pulse waveform from the wrist artery. The sensor should maintain high-pressure sensitivity over a broad pressure range; otherwise, the effective pulse waveform detection will be hampered. Sensors having high-pressure sensitivity can effectively resolve little pressure.

The pressure sensitivity of both capacitive and piezoresistive sensors depends on the sensing materials, porosity inside the sensing materials, or micropatterns on the surface of the sensing material. Having porosity inside the sensing layer or micropatterns on the surface of the sensing layer decreases the young's modulus, which improves the pressure sensitivity of the sensor. A capacitive pressure sensor based on parallel plate electrodes can achieve high-pressure sensitivity by incorporating such porosity inside the dielectric layer or introducing micropatterns on the dielectric layer/electrodes. By tuning the dielectric properties of the sensing material, the sensitivity of the capacitive pressure sensor can be further improved. In addition, supercapacitive/ionotropic pressure sensing can be another perspective sensing technology for biomedical applications since the sensitivity of such sensors does not rely solely on the parallel plate electrode mechanism but instead on the electric double layer formation on the electrode/electrolyte interface. Due to the formation of an electric double layer, ultra-high areal capacitance can be achieved, which increases under pressure resulting in high-pressure sensitivity. This high-pressure sensitivity can also improve the SNR of the sensor, thus enabling high-quality pulse waveform detection. Micropatterns on the surface can also improve sensing performance for piezoresistive pressure sensors. Under pressure, the micropatterns change shape, leading to the sensing area increases leading to higher current flow. Therefore, by carefully controlling the micropatterns, the pressure sensitivity of the piezoresistive pressure sensor can be improved. The sensing performance of the piezoresistive pressure sensor depends significantly on the electrodes, unlike the capacitive pressure sensor. Besides using plane electrodes sandwiching the sensing layer, different packaging methods show high-performing sensors.

Although there are numerous reports on high-performance flexible pressure sensors, there are still some challenges in fabricating such sensors. One of the challenges is to ensure a broad and linear pressure sensing range. Most of the pressure sensors have a non-linear or multi-slope pressure sensing range. To obtain a universal cuffless blood pressure monitoring device based on a flexible pressure sensor, ensuring a reliable pressure sensor is necessary. Another challenge for sensor fabrication is the reproducibility, reliability, and facile production of the sensors. Complicated cleanroom fabrication of the sensors comes up with the increased cost of production, production lead time, and resources. Although there are reports on low-cost sensor fabrication using the natural template-based approach, ensuring uniformity, reproducibility, and repeatability of such fabrication techniques can still be challenging. The future goal for such solid-state sensors can be a non-invasive cuffless blood pressure measurement system. Since solid-state sensors such as capacitive and piezoresistive sensors can achieve high sensitivity, they have prospects of detecting the pulse waveform and, with suitable calibration, measuring the systolic and diastolic pressure. Since pulse detection requires high-pressure sensitivity, more study is essential for such sensor platforms and the feasibility of using such sensors for blood pressure measurement where commercial optical pulse monitoring system is more prevalent.

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.

Data availability

Data will be made available on request.

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