

TOPICAL REVIEW

Soft wearable sensors for monitoring symptoms of COVID-19 and other respiratory diseases: a review

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Soft wearable sensors for monitoring symptoms of COVID-19 and other respiratory diseases: a review

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Yuxuan Liu , Darpan Shukla, Holly Newman and Yong Zhu*

Department of Mechanical and Aerospace Engineering, North Carolina State University, Raleigh, NC 27695, United States of America

* Author to whom any correspondence should be addressed.

E-mail: yong_zhu@ncsu.edu

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Abstract

The COVID-19 pandemic has put extraordinary stress on medical systems and global society more broadly. The condition of infected patients may deteriorate rapidly due to overburdened hospital systems. This raises an urgent need for real-time and remote monitoring of physiological parameters to address the challenges associated with the COVID-19 pandemic. This review will present recent progress on soft wearable sensors that can potentially be used for monitoring respiratory diseases such as COVID-19. First, emerging monitoring devices and systems that can monitor key physiological parameters as suggested by the Centers for Disease Control and Prevention (e.g. body temperature, respiration rate, heart rate, oxygen saturation and body movement) are reviewed. Then, multimodal sensor systems consisting of two or more correlative sensors are presented. This review will conclude with challenges and future directions for wearable sensors for the diagnosis and therapy of respiratory diseases. While this review focuses on COVID-19, the sensing technologies reviewed can be applicable to other respiratory diseases such as H1N1 influenza.

1. Introduction

Coronavirus disease 2019 (COVID-19), a respiratory disease caused by SARS coronavirus 2 (SARS-CoV-2), has become a global pandemic. Because of the highly contagious and severe nature of the disease, an extraordinary burden has been placed on the global healthcare system. Overwhelmed healthcare systems are struggling not only with the timely screening and treatment of COVID-19 patients but also the clinical treatment of patients with other diseases. With almost all countries and areas around the world being heavily affected, the pandemic has caused enormous losses in health, life and the economy for global society.

Patients contracting COVID-19 usually exhibit respiratory symptoms [1]. Based on the severity of the symptoms, there are five case categories: asymptomatic, mild, moderate, severe and critical. Around 10% of all cases are severe enough to require intensive care, while most other cases are less severe [2, 3]. The Centers for Disease Control and Prevention (CDC) recommends that people who feel sick should stay at home in self-quarantine, as most people with COVID-19 have mild illness and can recover at home without medical care. However, this requires that the symptoms of isolating patients are carefully monitored by themselves or medical staff to prevent sudden deterioration of the illness, raising a high demand for continuous monitoring of the physiological parameters associated with typical symptoms of COVID-19. A wide range of symptoms can be detected in COVID-19 patients, including fever, cough, shortness of breath, difficulty breathing, fatigue and possible injury to the cardiac system, as suggested by the CDC [4, 5]. Relevant physiological parameters include body temperature, respiration rate and intensity, electrocardiograph (ECG), oxygen saturation and body movement (especially during coughing). Emerging wearable sensors for health monitoring could provide potential opportunities to detect and monitor these parameters rapidly and remotely, which offers the following advantages: (a) providing early warning of COVID-19 infection; (b) closely monitoring the clinical situation and preventing sudden worsening of the situation; (c) maintaining distance between medical staff and patients during screening and monitoring; (d) reducing the load of the

medical system with at-home monitoring; (e) enabling data-level analysis when integrated with artificial intelligence (AI).

A number of conventional sensors, including portable ones, have been reported for monitoring COVID-19 symptoms [1, 6–10]. While very promising [7], to the best of our knowledge no soft wearable sensors have yet been reported for this purpose. Wearable sensors would ideally require conformal contact with the surface of human body, which is crucial for a high signal-to-noise ratio, low motion artifacts and comfortable wear [11–14]. The human body exhibits a complex 3D curvilinear shape. In addition, human skin has a Young's modulus in the range of 140–600 kPa for the epidermis and 2–80 kPa for the dermis [15]. Due to the intrinsic elasticity and the wrinkling features of the skin, the local strain introduced to wearable items on the surface of the human body can reach as much as 100% during daily motion [16–18]. Soft wearable sensors must be flexible and even stretchable in order to match the 3D shape and deformation of human skin. In the past two decades, flexible and stretchable electronics have attracted significant interest [15, 19–44]. Many excellent review articles on soft wearable sensors based on flexible and stretchable electronics for health monitoring in general have emerged [13, 30, 33, 45–50], but not as yet for monitoring COVID-19 symptoms.

This review will focus on monitoring technologies using soft wearable sensors that are flexible and stretchable for monitoring typical COVID-19 symptoms. To begin, we review the monitoring of key physiological parameters including body temperature, respiration rate and intensity, oxygen saturation, ECG and body movement during coughing. For each parameter, its relevance to COVID-19 diagnosis and therapy, sensing mechanism(s) and representative sensors will be discussed. Considering that COVID-19 patients usually exhibit multiple symptoms, a section on correlative multimodal sensor platforms will be presented. Finally, challenges and future directions regarding soft wearable sensors for respiratory disease monitoring are discussed. While this review focuses on COVID-19, the sensing technologies reviewed can be applied to other respiratory diseases such as 2009 H1N1 influenza. Of note is that this review focuses on soft physical sensors for monitoring respiratory diseases. For soft electrochemical sensors, readers are referred to some recent excellent reviews [51–53].

2. Monitoring of physiological parameters

2.1. Body temperature

Body temperature is a key indicator of fever, a sign of bodily infection or inflammation. Fever is one of the most significant symptoms of COVID-19: 98% of infected patients are reported to have the early symptom of fever [54]. By monitoring the real-time body temperature, early diagnosis and observation of illness evolution can be realized. Body temperature is typically taken using infrared (IR) technologies, such as handheld IR thermometers and IR thermal imagers. While these methods are widely used in daily life, they provide a snapshot rather than continuous monitoring of body temperature. Wearable temperature sensors with good flexibility, fast response, adequate sensing range (25 °C–50 °C) and high sensitivity (down to ± 0.1 °C within the temperature range of 37 °C–39 °C and ± 0.2 °C outside the range) could address this issue [55, 56].

Temperature sensors include several major types: thermocouples, resistive temperature detectors (RTDs) and thermistors. Temperature can also be measured on the basis of the pyroelectric effect and thermoelectric effect [21, 55–57]. RTDs and thermistors are widely used in wearable sensors [58]. RTDs are made of metals and thermistors are made of ceramics or polymers. In general, thermistors can be categorized as negative temperature coefficient (NTC) type or positive temperature coefficient (PTC) type; sometimes RTDs are categorized as PTC type thermistors because they show a PTC. Wearable temperature sensors are typically made of metal- [59, 60], ceramic- [61], polymer- [62, 63] or carbon-based materials [64, 65]. For example, a Ni-based NTC thermistor, fabricated using laser direct writing (LDW), was used as an epidermal temperature sensor for monitoring the temperature of exhaled/inhaled breath with ultrahigh sensitivity [59]. The monolithic Ni–NiO–Ni structure consists of a NiO sensing channel, ~ 50 μm wide, formed by the laser reductive sintering technique shown in figure 1(a). The high sensitivity comes from the decreasing vacancy density in NiO formed by laser reductive sintering of Ni, which increases the activation energy required for electrical conduction. The ultralight and conformal sensor module can also be attached to different locations on the skin to monitor local body temperature (figure 1(b)). Graphene-based thermistors have been reported for wearable applications due to their excellent temperature sensitivity and mechanical robustness [57]. A graphene thermistor temperature sensor with a tunable thermal index was fabricated by embedding graphene in a polydimethylsiloxane (PDMS) matrix and using Ag nanowires (AgNWs) as the electrodes, as shown in figure 1(c) [65]. The sensor showed a stretchability of 50%. The sensitivity could be tailored by modifying the applied strain on the sensor; this was attributed to the strain-induced change in electrical contacts between adjacent crumpled graphene “balls”, which can lead to higher sensitivity under greater

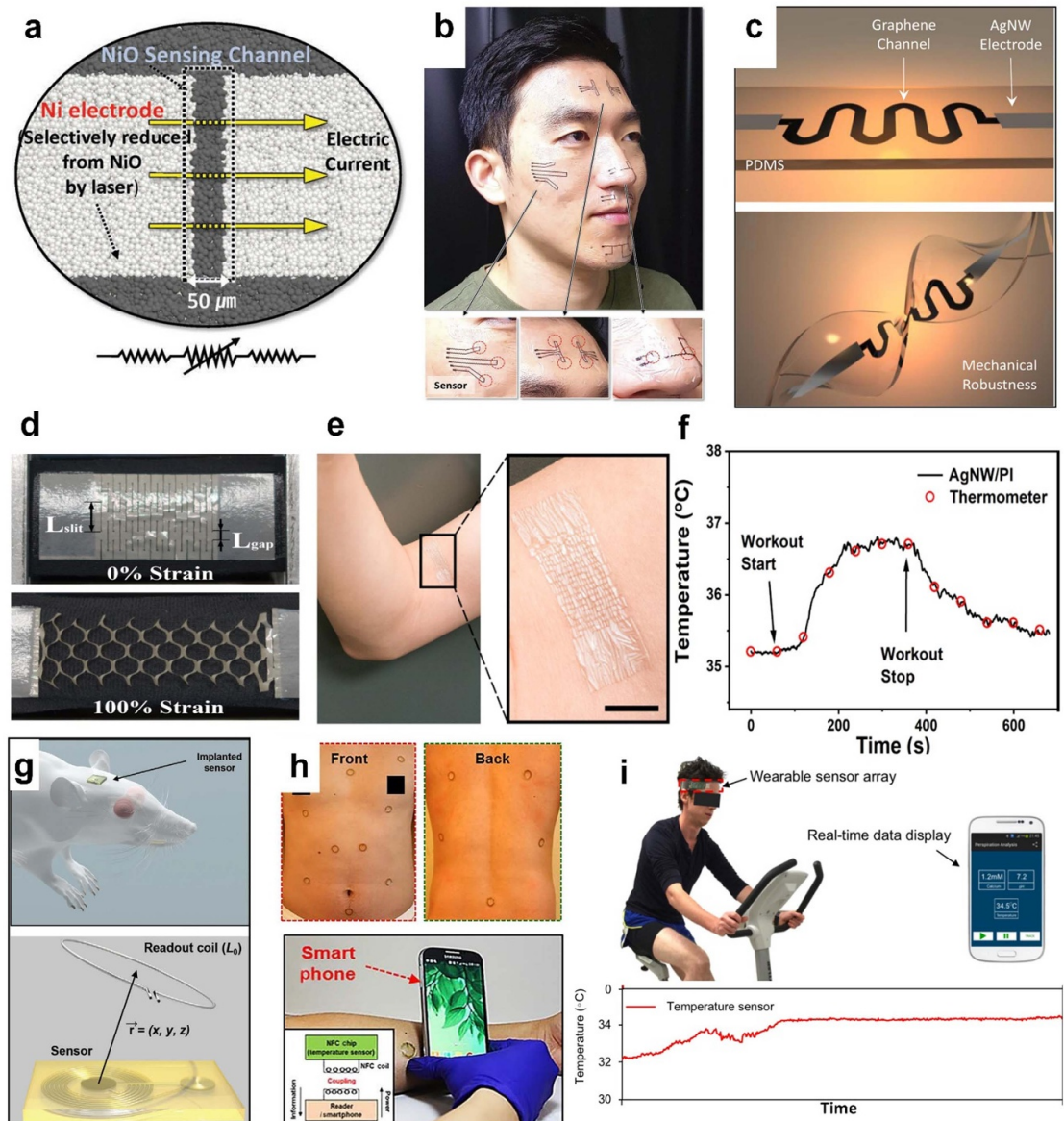


Figure 1. (a) Schematic of a Ni-based NTC thermistor with a NiO sensing channel formed by the laser reductive sintering technique. (b) Epidermal temperature sensors with a 25 μm thick PET substrate attached at various positions on the face. The sensing channels are labeled with red circles. (a), (b) Reproduced with permission from [59]. John Wiley & Sons. © 2019 WILEY-VCH Verlag GmbH & Co. KGaA, Weinheim. (c) Schematic diagrams of stretchable graphene thermistors with AgNW electrodes in relaxed (top panel) and twisted (bottom panel) states. Reprinted with permission from [65]. Copyright (2015) American Chemical Society. (d) Optical images of a AgNW/polyimide temperature sensor at tensile strains of 0% and 100%. (e) AgNW/polyimide temperature sensor attached to the skin near the biceps. Scale bar: 10 mm. (d), (e) Reprinted with permission from [60]. Copyright (2019) American Chemical Society. (f) Temperature recorded by the temperature sensor and IR thermometer during biceps workout (PI, polyimide). (g) Schematic illustration of the implantation of the temperature sensor in a rat. The inset shows a schematic illustration of the relative position of the sensor and the readout coil. [62] John Wiley & Sons. © 2020 WILEY-VCH Verlag GmbH & Co. KGaA, Weinheim. (h) Photographs of sensors at different locations on the front and back of the body (top panel) and short-range readout from a skin-mounted sensor using a smartphone with an inset to show the operational principles (bottom panel). From [67]. Reprinted with permission from AAAS. (i) Wearable sensing system worn on a subject's forehead during stationary cycling with a customized application to show monitoring data on a mobile phone (top panel) and the monitoring results of skin temperature (bottom panel). Reprinted with permission from [68]. Copyright (2016) American Chemical Society.

strain. Other carbon materials such as carbon nanotubes (CNTs) have been used in thermally responsive field-effect transistors (FETs). A stretchable temperature sensor array based on single-walled CNT FETs exhibited a high sensitivity of $6.5\% \text{ } ^\circ\text{C}^{-1}$ in the temperature range of $25 \text{ } ^\circ\text{C}$ – $45 \text{ } ^\circ\text{C}$. [66] These sensitive temperature sensors enabled by advanced materials are highly desirable for wearable temperature monitoring due to the small range of variance of human body temperature.

Common configurations of these wearable temperature sensors include epidermal patch [65, 66], band [69], belt [70] and watch type [71]. However, a major challenge with these wearable temperature sensors is to decouple the strain effect from the temperature effect. Several mechanical designs such as the island–bridge

structure, fractal patterns and kirigami structures may potentially be used to address the problem [72, 73]. Our group developed a strain-insensitive wearable temperature sensor by introducing the kirigami structure to a AgNW/polyimide composite thin film, as shown in figure 1(d) [60]. The device is highly stretchable and breathable as a result of the kirigami design. The temperature coefficient of resistance of the device can be tailored by adjusting the density of the AgNW network and the annealing temperature. The sensors show no deterioration of performance under tensile strains up to 100%. Finite element analysis confirms that the out-of-plane deformation under tensile strain, introduced by the kirigami structure is effective in reducing the local strain. This ultralight sensor was attached to human skin to monitor the change in skin temperature during workout, showing consistent results when compared with a commercial thermometer (figures 1(e) and (f)) and indicating an effective design to reduce artifacts from body movements. Apart from this structural strategy, stretchable material-enabled intrinsic stretchability is another strategy for making sensors highly stretchable. For instance, liquid metals have been used to accommodate the strain in temperature sensors as they can maintain the liquid phase even at room temperature and hence a stable conductive path under large strain [74, 75].

To remotely monitor the body temperature of COVID-19 patients, a wireless communication system is needed to transmit the collected data from the sensor to user interfaces such as smartphones, laptops or cloud platforms. By analyzing the data in real time, medical professionals can potentially make a diagnosis or treatment decision without in-person clinical observations. Near-field communication (NFC) and Bluetooth are two typical data transmission protocols widely used in wearable health monitoring systems. Wearable sensor modules integrated with a data acquisition system have attracted increasing attention. An LC resonance circuit-based bioresorbable temperature sensor with a NFC antenna was developed to monitor the local temperature change of a rat in a wireless and battery-free fashion, as shown in the top panel of figure 1(g) [62]. Temperature change can lead to change in the capacitance and hence the resonance frequency of the LC circuit, which can be measured by a reading circuit via electromagnetic coupling (bottom panel of figure 1(g)). Human body temperature mapping was achieved by attaching 65 temperature sensor modules on the body and collecting the data using NFC technology. The on-body distribution of the sensors is shown in the top panel of figure 1(h) [67]. The scheme of short-range readout from the skin-mounted sensors using a smartphone is shown in the lower panel of figure 1(h). With these flexible and miniaturized sensor modules, the mapping area can be set at any location on the human body. However, data transmission by NFC communication requires a very short distance (several centimeters) for connection, which is a drawback for system design and data quality. The Bluetooth-based data transmission approach has a longer applicable distance. Nyein *et al* integrated a flexible resistive temperature sensor with an electrochemical body fluid sensor to monitor body temperature during exercise [68]. As shown in figure 1(i), a flexible PCB with a Bluetooth module was used to transmit the signal to a user-friendly smartphone interface for data display. The time-series monitoring results can be used to analyze the health condition. Another similar platform was developed to continuously monitor the body temperature and other physiological parameters obtained from a microfluidic device [64]. The data acquisition system was designed with an integrated microcontroller and Bluetooth module to achieve remote and real-time signal reading and processing, which makes it promising for remote body temperature monitoring systems.

2.2. Respiration

Some of the most significant symptoms of COVID-19 are related to respiration, including shortness of breath, difficulty in breathing and possible cardiac damage caused by respiratory failure. Monitoring the respiratory rate is a simple yet effective way to evaluate the severity of respiratory disease. When the respiratory rate is greater than 30 breaths per minute the patient may face acute illness exacerbation, even into acute respiratory distress syndrome (ARDS). Real-time monitoring of respiratory rate with wearable sensors is important for identifying sudden deterioration. Several different mechanisms exist for measuring the respiratory rate, including detecting the difference between inhalation and exhalation, detecting the mechanical deformation of the human body associated with breathing and signal processing of other physiological parameters such as ECG and photoplethysmogram (PPG). The sensors are typically placed at several locations such as the nasal or oronasal area, chest and abdomen.

Temperature and humidity are the two signals most widely used to differentiate inhalation and exhalation because the airflow inhaled and exhaled from ambient air and the human body have different temperature and humidity levels. By continuously detecting these two parameters, the respiration profile can be obtained. The most widely used configuration for these two types of respiratory sensor is to integrate the sensor modules and other components (e.g. the data acquisition module and power source) on a facemask, such as is already being used during the COVID-19 pandemic. This design has been widely used for years to regulate the gas flow from inhalation and exhalation for optimized sensing performance [76–78]. A

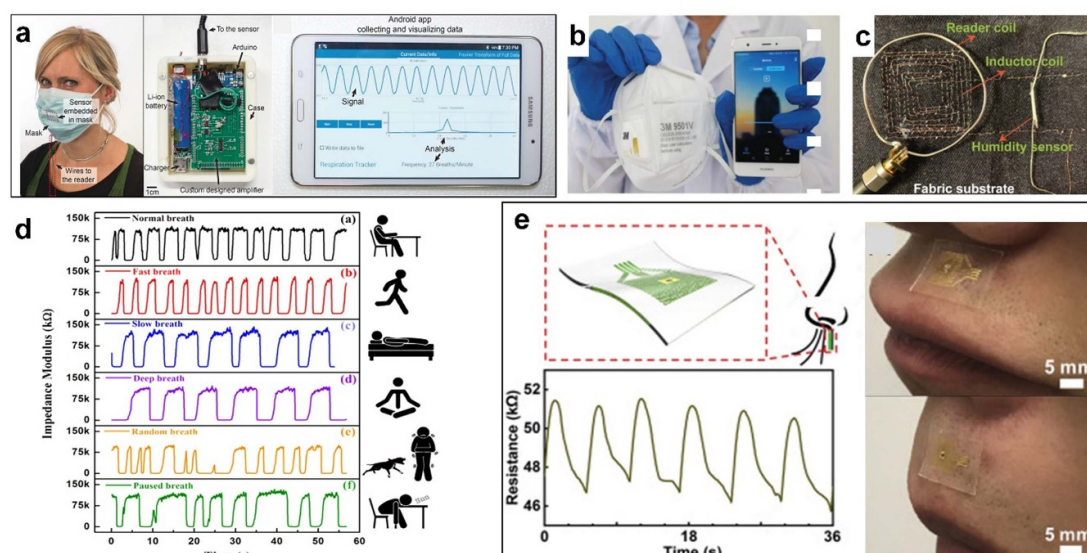


Figure 2. (a) A paper-based sensor integrated into a surgical mask (left panel), the data acquisition system (middle panel) and the Android app which can display the monitoring results (right panel). [79]. John Wiley & Sons. © 2016 WILEY-VCH Verlag GmbH & Co. KGaA, Weinheim. (b) Photograph of a breath monitoring system on a mask using a respiratory humidity sensor. [80] John Wiley & Sons. © 2019 WILEY-VCH Verlag GmbH & Co. KGaA, Weinheim. (c) Optical image of a breath monitoring system with a fiber inductor coil for LC wireless monitoring. [80] John Wiley & Sons. © 2019 WILEY-VCH Verlag GmbH & Co. KGaA, Weinheim. (d) Six respiration patterns recorded by a respiratory humidity sensor: normal breath, fast breath, slow breath, deep breath, random breath and paused breath. Reprinted with permission from [81]. Copyright (2019) American Chemical Society. (e) Schematic illustration of a stretchable respiratory sensor that can be attached to the upper lip to monitor respiration patterns (left panel) and optical images of the respiration sensor mounted on top of the upper lip with two motions, including pouting and compressing lips (right panel). Reprinted from [82], Copyright (2020), with permission from Elsevier.

paper-based humidity respiration sensor was integrated onto a surgical facemask with data acquisition electronics, as shown in figure 2(a) [79]. The respiration profile can be obtained and visualized in real time through a user-friendly Android app. An LC wireless human breath monitoring system has been reported, consisting of a humidity sensor made of yarns integrated into a 3M N95 mask (figure 2(b)) [80]. The respiration signals could be transmitted wirelessly by the resonant circuit shown in figure 2(c). Dai *et al* presented a humidity sensor made of a polymer sensing material on a ceramic substrate, which was used to distinguish different respiratory patterns including normal, fast, slow, deep, random and paused breath, as shown in figure 2(d). The sensor exhibited short response and recovery times (0.29 and 0.47 s, respectively) [81]. With a fast and accurate sensing performance, the respiratory signals during exercise were monitored and analyzed. However, these sensor systems are rigid and bulky due to the facemask configuration and nonstretchable materials used. A stretchable respiratory sensor attached to the upper lip (figure 2(e)) was developed to detect the temperature of airflow based on the thermal convection effect [82]. Different respiration patterns (i.e. panting, during fright, during meditation and when sitting and sleeping), the respiration rate during exercise and the respiration rate during sleep were monitored, demonstrating the ability to monitor respiration in real time with desirable signal quality.

Despite the promising advances in soft respiratory temperature and humidity sensors for real-time monitoring of respiratory symptoms, these sensors are prone to environmental interference. The humidity sensors based on capacitance change due to differences in water absorption in the dielectric materials may not function properly when the ambient temperature is below 0 °C, as the moisture in the air condenses on the sensor surface [79]. The temperature sensors used to detect the temperature difference between breath air flow and ambient air can only work when the ambient temperature is below that of the breath air flow. Detection of the change in chest contraction and expansion when breathing can address this challenge. Optical approaches have been used to detect the chest deformation associated with respiratory monitoring. But these optical sensor systems are usually complex with a large footprint, and so are not suitable for wearable applications. An alternative approach is to use wearable mechanical sensors such as strain sensors and pressure sensors to monitor the chest deformation associated with respiration [83–85]. Typically, strain and pressure signals for respiration monitoring can be obtained by placing the sensors on several locations on the body, including the umbilicus, upper abdomen, xiphoid process, upper thorax and diagonal. A comprehensive study of placing strain sensors at these locations showed that the accuracy of the detected

respiratory rate is similar at all these locations, but the upper thorax is the most comfortable location according to feedback from 30 subjects [86].

However, two issues still need to be solved when using these strain and pressure sensors: (a) the sensors need to be soft and stretchable to be conformal with the human body for accurate strain measurement and comfortable to wear; (b) the sensitivity of the sensors needs to be high to capture the respiration profile with high accuracy and resolution. A highly sensitive wearable strain sensor with good stretchability was used as a respiration sensor [87]. Copper was deposited onto a carbonized fabric and the sensor encapsulated by elastomer. The high gauge factor (~ 3557.6 and ~ 47.8 in the strain range from 0% to 48% and from 48% to 150%, respectively) endows the sensor with high sensitivity when monitoring respiration. The stretchability of the device is as high as 300%. With a convolutional neural network model, normal breath, tachypnea and tachypnea with cough can be distinguished by this sensor.

In addition to the two mechanisms discussed above, the respiration rate can also be obtained by extracting respiratory modulation from cardiac signals such as ECG and PPG. ECG and PPG both exhibit three respiratory modulations: baseline wander, amplitude modulation and frequency modulation. These respiratory modulations are related to the physiological processes caused by breathing and heartbeat. A number of algorithms have been developed to estimate breath rate by analyzing one or more of these modulations. In one such algorithm, breathing rate (BR) was extracted from PPG signals using the empirical mode decomposition (EMD) method, which is an adaptive time–frequency analysis approach [88]. Here, the signal is decomposed into intrinsic mode functions which are groups of frequency- and amplitude-modulated components of PPG. The component corresponding to the highest frequency provides information about the heart rate (HR), while the component corresponding to the lowest frequency can give an estimate of BR. A mean envelope can be calculated based on the first loop of EMD, which is further used for estimating BR. Furthermore, the dominant frequency peak is evaluated using the power spectral density of the respiratory element. This algorithm is relatively simple and less computationally expensive than other reported algorithms. It has been validated on real-time data acquired using wearable sensors with a mean absolute error of 0.0044 Hz (0.26 breaths per minute) [88]. For more details on development of the algorithm readers are referred to the review by Charlton *et al* [89].

2.3. Oxygen saturation monitoring

Oxygen saturation (SpO_2), referring to the percentage of hemoglobin that is saturated with oxygen, is an important physiological parameter for evaluating the overall health condition. It is typically measured using a pulse oximeter. The mechanism is to measure the absorption of light by oxygenated and nonoxygenated hemoglobin in blood at different wavelengths. A light source is used to emit light with different wavelengths and a photodetector converts the transmitted or reflected optical signals to electrical signals [90, 91]. The profile obtained by the pulse oximeter is called the PPG; the SpO_2 level and other physiological signals (e.g. pulse rate and respiration rate) can be extracted from the PPG.

When someone is infected by SARS-CoV-2, inflammation of the lungs impedes the transport of oxygen from the air into the bloodstream, leading to a decrease in the SpO_2 value. A healthy person has an SpO_2 level of 95%–100%, while COVID-19 patients may show an SpO_2 level as low as 90% in fatal cases [92, 93]. However, mild cases usually present SpO_2 values over 94%, which is relatively safe; patients can be monitored at home as recommended by the World Health Organization [94]. Continuous monitoring of SpO_2 is crucial to identify the severity of COVID-19 progression and determine the treatment strategy. Pulse oximeters are commercially available and are typically clipped on a person's finger, toe or ear lobe. However, the commercially available ones face several limitations including motion artifacts and high power consumption when used for long-term health monitoring. Soft pulse oximeters with low energy consumption have emerged in recent years with the aim of solving these problems.

A miniaturized thin pulse oximeter was developed that can be attached to fingernails, toenails and other locations on the human body, as shown in figures 3(a) and (b) [95]. By employing a soft and optically opaque encapsulation material, the device is flexible and mechanically compliant. NFC technology-based wireless charging for the sensor makes wireless monitoring possible. Hong *et al* developed an integrated noninvasive wearable smart band system with a built-in pulse oximeter to monitor the SpO_2 level before and after exercise [96]. The photodetector and light-emitting diode (LED) are placed on the backside of the smart band, as shown in figure 3(c). The SpO_2 level is estimated using the direct current (DC) and alternating current (AC) components of the PPG signal obtained by a red LED and an infra-red LED. After signal processing, peaks can be detected and the time difference between the peaks is used to calculate HR, as shown in figure 3(d). Bluetooth data transmission provides real-time and remote monitoring of multiple physiological parameters.

A customized soft pulse oximeter was developed by Abdollahi *et al* using advanced 3D printing. The soft pulse oximeter can be mounted on the fingertip or toe, as shown in figures 3(e) and (f) [97]. The specific user's finger or toe is scanned first and then a PDMS cuff for the finger or toe is printed. The toe type sensor

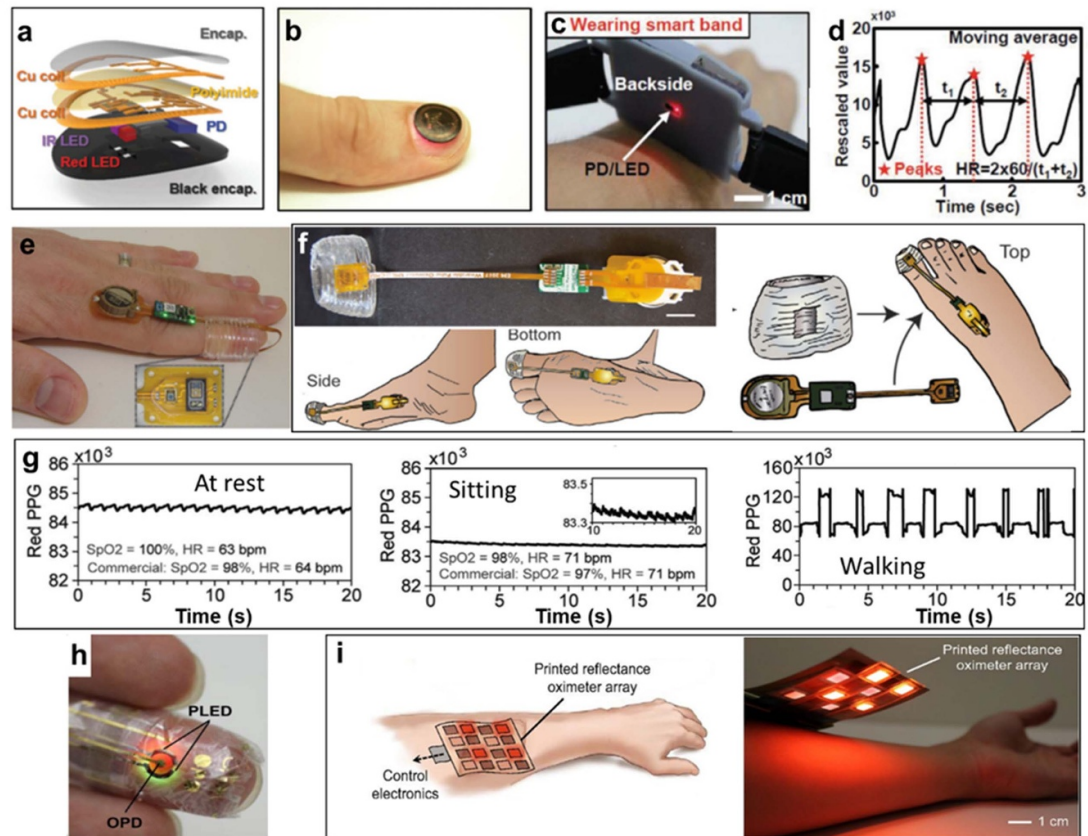


Figure 3. (a) Exploded-view schematic illustration of a thin flexible pulse oximeter with a millimeter-scale size. (b) Image of a thin flexible pulse oximeter mounted on a thumbnail during an operation. (a), (b) [95]. John Wiley & Sons. © 2016 WILEY-VCH Verlag GmbH & Co. KGaA, Weinheim. (c) A smart band with a built-in pulse oximeter to monitor SpO_2 . (d) The PPG signal after the signal conditioning process with labeled peaks (red stars) and the time difference between the peaks (t_1 and t_2). The equation for calculating HR is shown at the bottom. (c), (d) [96]. John Wiley & Sons. © 2018 WILEY-VCH Verlag GmbH & Co. KGaA, Weinheim. (e) Representative image of a patient-specific wearable pulse oximeter tested on the index finger. (f) Complete wearable toe pulse oximeter with the data acquisition and power modules on the side, bottom and top of the toe. Scale bar: 1 cm. (g) The pulse oximeter signal recording of a red PPG when the subject is at rest, sitting and walking. (h) Photograph of optoelectronic skin attached to a fingertip with a pulse oximeter. (i) Schematic of an organic light emitting diode and organic photodiode based oximeter array to map oxygenation of the reconstructed skin (left panel), and photo of the oximeter array on top of a person's forearm (right panel). (g)–(i) [97]. John Wiley & Sons. © 2020 WILEY-VCH Verlag GmbH & Co. KGaA, Weinheim.

was tested when the user was at rest, sitting and walking. The battery and Bluetooth module used to supply power and transmit signals are placed on top of the foot to avoid applying pressure to the pulse oximeter. The SpO_2 value and HR measured by the 3D printed pulse oximeter when at rest and sitting are comparable with those obtained using commercial devices, but it is not usable when walking due to the large artifact coming from body movement, as shown in figure 3(g). Fully soft pulse oximeters with soft components (e.g. light source and photodetector) may enhance the signal quality because there is no modulus mismatch between functional components and the substrate.

Flexible and stretchable organic light emitting diodes (OLEDs) and organic photodiodes (OPDs) have emerged in recent years, presenting a promising solution to the fabrication of fully soft pulse oximeters with enhanced comfort and signal quality. Yokota *et al* fabricated optoelectronic skin with a pulse oximeter and optical display modules enabled by OPDs and OLEDs, respectively [98]. Due to the ultrathin nature of the device and pre-stretching of the elastomer substrate, the whole device can endure a bending radius of less than $100\ \mu\text{m}$ and stretchability up to 200%, making it suitable for attachment to the fingertip as an epidermal sensor (figure 3(e)). By fabricating the OLED and OPD based oximeter into a reflectance array, 2D mapping of the SpO_2 level in the forearm under normal and ischemic conditions was obtained, as shown in figure 3(f) [91]. The device was fabricated by screen printing of functional materials onto a flexible plastic substrate. This study showed that the forehead provides the strongest pulse signal. These devices demonstrated the promising potential of flexible and stretchable pulse oximeters for continuous monitoring of physiological parameters (e.g. SpO_2 level and HR) for patients at home in the COVID-19 pandemic and other clinical situations.

2.4. ECG and HR monitoring

Electrophysiological signals reveal the electrical properties of biological cells and tissues. Amongst the various electrophysiological signals, the ECG reflects the electrical activity of the heart by detecting the small electrical changes that occur due to the depolarization and repolarization of the cardiac muscles during each cardiac cycle [99]. The ECG waveform constitutes three main components—the P wave that shows atrial depolarization, the QRS complex that depicts depolarization of the ventricles and the T wave that signifies repolarization of the ventricles [100]. Patients infected with COVID-19 have been reported to show abnormalities in the general ECG waveform such as fragmented QRS, QT elongation (>500 ms), ST–T change or T-wave inversion [101]. Fragmented QRS is associated with myocardial injury, the ST–T change leads to ischemia that caused by restriction of blood flow, while the QT elongation could result in an increasing risk of ventricular arrhythmia [102, 103].

Continuous monitoring of ECG has become an important diagnostic tool for timely evaluation and treatment of irregular heart rhythms in COVID-19 patients [104]. Conventionally, ECG mapping is realized by attaching wet Ag/AgCl electrodes with conductive gels on the human skin using adhesives. However, these gelled electrodes can cause skin irritation after prolonged use. Additionally, the electrolytic gel dries over time, degrading the signal quality [20, 105]. Dry electrodes can avoid such limitations, showing promise for long-term continuous heart monitoring. Wearable ECG electrodes can utilize either fabric-based or silicone-based substrates to render the desired flexibility [106]. Optimizing the geometry and material choices of these electrodes can enable soft, skin-like wearable devices that could maintain conformity with human skin and possess low electrode–skin impedance [107].

Towards that end, Kim *et al* developed a hygroscopic auxetic sensor that provides stable monitoring of ECG over a long period (7 days) [108]. The hygroscopic characteristic of the electrodes facilitates absorption of sweat or water while the voids in the serpentine network provide pathways for air penetration and sweat evaporation (figure 4(a)). The auxetic structure has a negative Poisson's ratio and matches with deformation of the skin, making it possible to maintain conformal contact between the sensor and the skin during repeated movements, as shown in figures 4(b) and (c). No degradation in the quality of the ECG signals was observed during vigorous exercise, after salt-water treatment or after submersion in water. Our group has developed dry ECG electrodes using AgNWs that performed as well as the commercially available gel electrodes when the subject was resting and showed fewer motion artifacts [109]. To enhance the flexibility and stretchability, we printed dry ECG electrodes in stretchable shapes (e.g. Greek cross fractal patterns and serpentine patterns) using electrohydrodynamic printing of AgNWs, as shown in figure 4(d). The printed patterns showed stable performance under repeated bending and stretching cycles, making them suitable for wearable healthcare applications [110]. Gas permeability is another important consideration for long-term wear. Our group has reported two methods to achieve permeable and stretchable AgNW-based ECG sensors: (a) employing porous electrodes fabricated by the breath figure method [111], (b) integrating the sensors onto a textile sleeve [112]. The corresponding electrodes are shown in figures 4(e) and (f), respectively. Furthermore, our group developed a multimodal sensor chest patch that could detect high-quality ECG signals (in addition to skin hydration and body motions) [45]. The multimodal sensor patch featured several attributes such as self-adhesion, compact design and wireless data transmission to data logging devices. A rechargeable polymer battery was used to power the sensor for 37 h. Ferrari *et al* measured ECG and electromyogram (EMG) signals from an ultrathin (<1 μm), ultra-conformable and imperceptible single and multi-electrode array, which was fabricated by inkjet printing of poly(3,4-ethylenedioxythiophene) polystyrene sulfonate (PEDOT:PSS) onto a decal transfer paper [113]. The tattoo-like electrodes were used to test skin contact impedance for 48 h to showcase their adaptability for long-term use. Moreover, the electrodes could connect to a portable ECG monitor for an easy market-available personal monitoring device. Besides, the tattoo-based electrodes are preferable and allow hairs to grow through the electrodes without delaminating or distorting the skin–electrode contact.

Besides utilizing electrodes for mapping the bioelectrical signals of the human body, wearable pressure sensors based on piezoresistive or piezoelectric sensing are commonly employed for monitoring physiological variables such as heartbeat or pulse [116, 117]. In a piezoresistive sensor, application of an external pressure indirectly alters the density and contact of the conductive fillers in the sensor, which leads to a change in resistance. Piezoresistive sensors can either employ an active material or incorporate special geometries for achieving high sensitivity and wide pressure detection ranges [117]. Yang *et al* designed a bio-inspired, hierarchically microstructured, flexible piezoresistive sensor by sandwiching polyaniline/polyvinylidene fluoride nanofiber (HPPNF) film between microdome-shaped interlocking electrodes, as shown in figure 4(i) [118]. The microdome-structured electrodes in combination with HPPNF film helped to drastically increasing the contact area and provided an increased 3D deformation rate, giving the sensor a high sensitivity of 53 kPa^{-1} , a pressure detection range of 58.4–960 Pa, an excellent mechanical stability of over 50 000 cycles and a fast response time of 38 ms. The sensor could conformably adhere to the

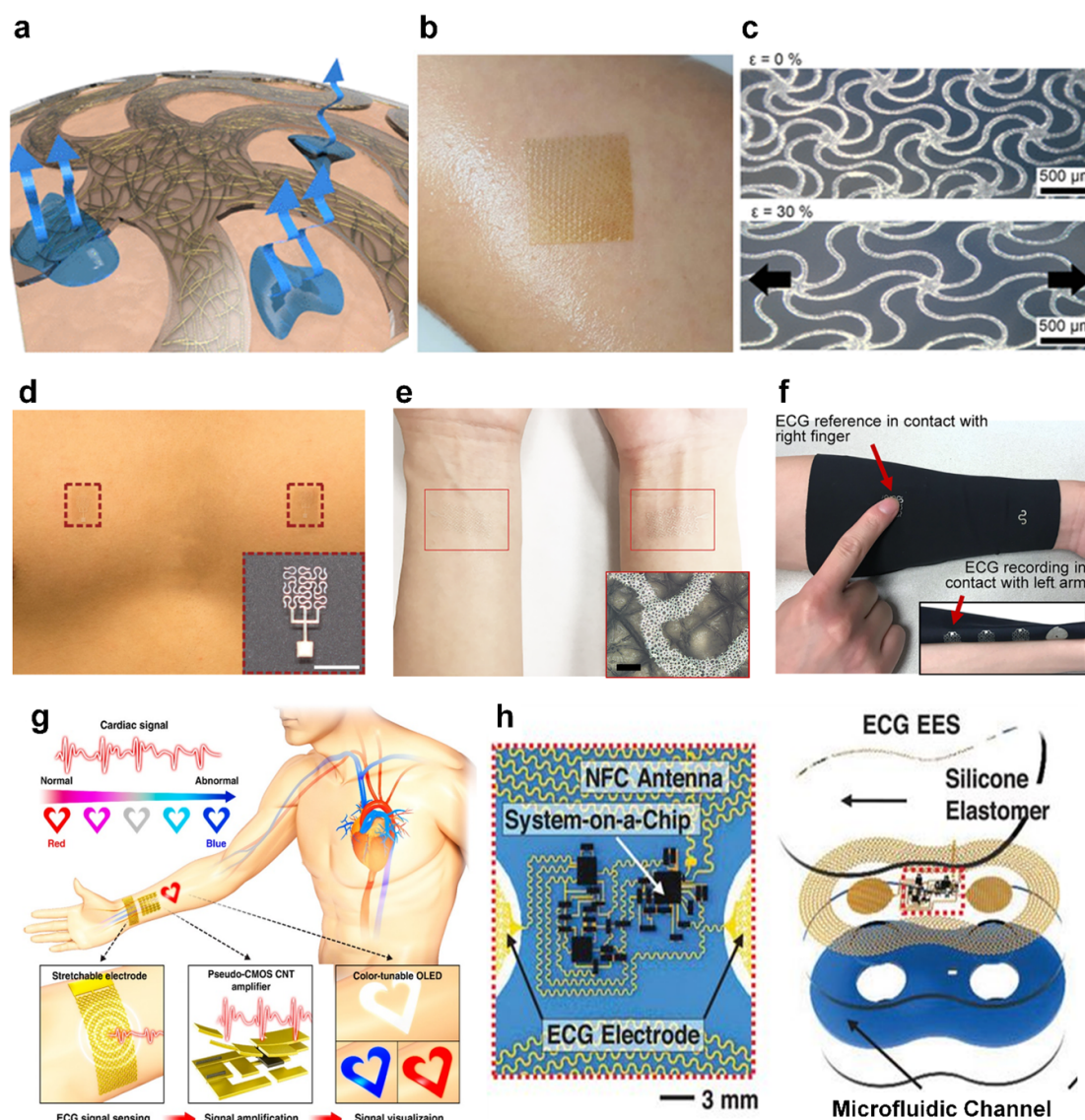


Figure 4. (a) Schematic diagram showing the breathability of a hygroscopic ECG electrode on skin. (b) Camera images of this ECG sensor being worn during exercise for 1 h. (c) Optical microscope images of an ECG electrode at relaxed and uniaxially stretched states. (a)–(c) Reprinted with permission from [108]. Copyright (2018) American Chemical Society. (d) AgNW dry ECG electrodes fabricated by electrohydrodynamic printing mounted on the chest. The inset shows a magnified image of an electrode. Scale bar: 5 mm. Reproduced from [110] with permission of The Royal Society of Chemistry. (e) Photograph of the porous ECG electrodes fabricated by the breath figure method based on AgNWs mounted on left and right forearms. The inset shows a magnified image of the porous electrode. Scale bar: 400 μm . Reprinted with permission from [111]. Copyright (2020) American Chemical Society. (f) Photograph of an ECG sensor integrated onto a textile sleeve. Reprinted with permission from [112]. Copyright (2019) American Chemical Society. (g) Schematic illustration of a real-time wearable cardiac monitoring system. Reprinted with permission from [114]. Copyright (2017) American Chemical Society. (h) Schematic illustration of an ECG epidermal electronic systems (EES) and an exploded-view schematic illustration. From [115]. Reprinted with permission from AAAS.

human epidermis to monitor clear wrist pulse signals along with other physiological signals, as shown in figure 4(j). In another example, arterial pulse was captured by a facile, low-cost, biodegradable all-paper-based piezoresistive pressure sensor [119]. The sensor consists of a dip-coated AgNW tissue paper as the sensing material, a nanocellulose paper (NCP) with printed silver electrodes as the bottom layer and a top NCP encapsulation layer. The porous and rough surface of the AgNW-coated tissue paper possesses conductive microfibers. On the application of pressure, these microfibers bridge the gap between the fingers of the interdigitated electrodes, leading to more conductive pathways and hence an increased current. The sensor was able to detect the pressure applied from the wrist artery, from which the pulse rate is calculated. The sensor has an ultralow power consumption ($\sim 10^{-8}$ W) and is mechanically robust showing only a moderate increase in resistance after 100 cycles of $\pm 180^\circ$ folding.

Piezoelectric sensors, which generate a voltage in response to an external pressure, have attracted interest due to their easy signal acquisition and self-powering capability when compared with piezoresistive sensors

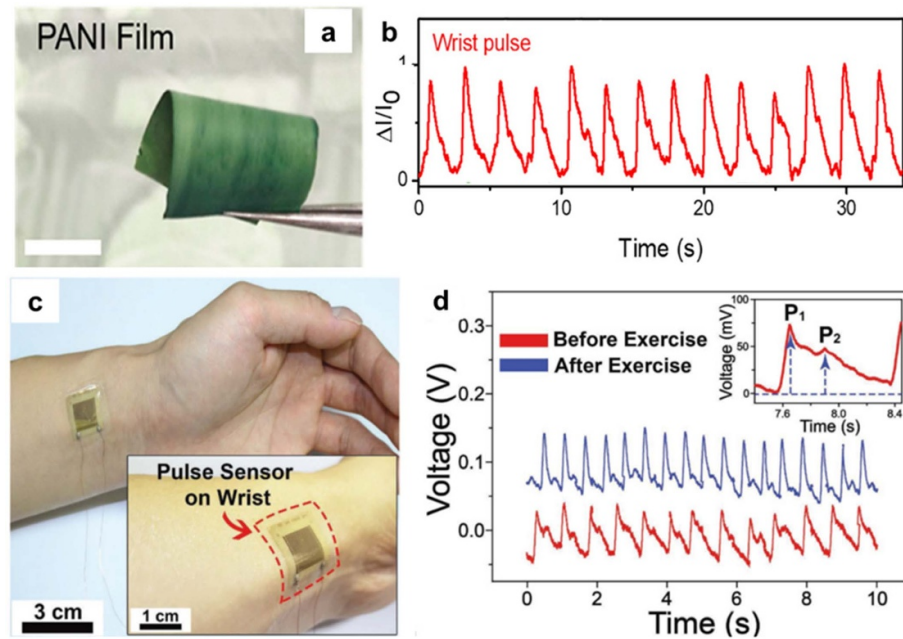


Figure 5. (a) Optical photograph of the hierarchical polyaniline nanofiber film. (b) Human wrist pulse signals measured by a sensor fabricated from polyaniline nanofiber film. (a), (b) Reprinted with permission from [118]. Copyright (2021) American Chemical Society. (c) Photograph of a PZT-based piezoelectric pulse sensor conformally attached to a human wrist. The inset shows the deformation of the sensor when a vessel moves. (d) Radial artery pulse signals detected by the PZT-based piezoelectric pulse sensor before and after exercise. (c), (d) [120] John Wiley & Sons. © 2017 WILEY-VCH Verlag GmbH & Co. KGaA, Weinheim.

[117]. To overcome the power consumption issues faced by conventional pulse sensors, Park *et al* fabricated a self-powdered, flexible, thin film $\text{Pb}[\text{Zr}_{x}\text{Ti}_{1-x}]\text{O}_3$ (PZT)-based piezoelectric sensor that could measure carotid and radial pulse signals from near-surface arteries, as shown in figures 5(c) and (d) [120]. The ultrathin nature of the epidermal sensor provides outstanding flexibility, which aids in conforming uniformly to the uneven topology of human skin. The real-time pulse signals detected from the sensor can be transferred wirelessly to a smartphone using an Arduino microcontroller unit and a Bluetooth transmitter. The captured signals exhibit characteristic peaks of the peripheral artery waveform, which could provide insights for biomedical use. Likewise, Shin *et al* developed a highly sensitive, wearable pressure sensor for real-time HR measurement using a ZnO nanoneedle and poly(vinylidene difluoride) (PVDF) hybrid film [121]. The hybrid film showed enhanced permittivity and a reduced elastic modulus compared with ZnO nanorods due to the large aspect ratio and hexagonal vertically grown pyramid structure of the ZnO nanoneedles. Moreover, the device incorporates reduced graphene oxide (rGO) electrodes and a screen-printed Bluetooth antenna, which could transmit the received pulse signals to a smartphone within 8 m without any distortion or time delay. This ZnO/PVDF film-based sensor exhibited a higher magnitude of electrical resistance change compared with a pristine PVDF film-based device and a lowest detectable pressure of 4 Pa. The sensor also displayed mechanical stability after 1000 cycles of repeated loading and unloading, making it suitable for day-to-day use. The sensor was able to detect the HR from the radial artery during normal and post-exercise conditions and showed similar oscillations compared with those detected by a wired pressure sensor.

A sensing system that can visually display the detected signal would facilitate the application of wearable sensors in healthcare. Koo *et al* utilized voltage-dependent color-tunable OLEDs to colorimetrically display the ECG signals measured from an ultrathin ($< 3 \mu\text{m}$) serpentine-shaped Au electrode, as shown in figure 4(g) [114], which could distinguish a normal signal from an abnormal one. The device comprises four CNT-based transistors, collectively called p-MOS inverters, which amplify the retrieved ECG signals, achieving a high signal-to-noise ratio. Apart from the display, wireless power transmission and communication modules can also be integrated to enable the remote monitoring of ECG signals. Alberto *et al* reported a quickly deployable passive 'electronic tattoo' sensor for electrophysiological monitoring [122]. The electronic patch is untethered, battery-free and ultrathin ($\sim 5 \mu\text{m}$), with a stretchable Ag-In-Ga coil for wireless power transmission ($\sim 300 \text{ mW}$ when transferred over the skin). The sensor patch exhibits a 'data-on-demand' feature, i.e. it receives the desired power and simultaneously communicates real-time ECG and body temperature data via Bluetooth when a scanning device (i.e. smartphone) is brought close to it.

COVID-19 has created numerous challenges in neonatal care for nurses and mothers [123]. Comfortable and continuous monitoring of the health of newborns is urgently needed. Rogers and co-workers developed a wireless, battery-free bimodal sensor system that can gently interface the soft neonatal skin due to its low modulus and ultrathin nature [115]. One sensor system consists of an ECG sensor that incorporates fractal geometries in the form of filamentary metal mesh microstructures, as shown in figure 4(h). A variety of signals can be extracted from the ECG waveform, including the QRS complex, HR, HR variability and pulse arrival time of a neonate. The other sensor system is placed on the base of the foot to measure PPGs. The recorded data from both systems could be continuously streamed to nearby data aggregators using a magnetic loop antenna tuned with NFC protocols.

2.5. Cough monitoring

Cough is a common symptom of COVID-19. According to a recent study, out of 21 COVID-19 patients 48% exhibited cough as a symptom [124]. COVID-19 spreads through respiratory droplets released when an infected person coughs or sneezes [125]. Detecting and monitoring cough frequency and intensity is significant for screening and diagnosing suspected cases in the early stages to prevent the spread of the disease in public areas (airports, schools, shopping malls, etc.). Conventional devices such as spirometers, pneumotachographs and peak flow meters are used to detect the cough peak flow, which is used for evaluating cough strength and the risk of cough dysfunction [126–128]. These devices require mouthpieces or masks, which collect the airflow during coughing. However, these devices make subjects uncomfortable and lead to unnatural behavior, and hence give unreliable detection results. Contactless cough strength detection methods based on cough sounds are being developed. By recording cough sounds with microphones and analyzing them using smartphones, cough frequency and intensity can be obtained and displayed through a user interface [129]. However, a few impediments such as bulkiness and an inability to monitor continuously still hinder the application of these devices for long-term monitoring of respiratory diseases such as COVID-19 [127]. In recent years, accessory-based [130], epidermal [131, 132] and textile-based [133] wearable devices with cough detecting and monitoring capability have been developed to enable continuous and unobtrusive monitoring of cough intensity and frequency.

The most common method for wearable cough detection is based on mechanical sensors such as pressure and strain sensors. By mounting a pressure or strain sensor on the throat, the throat-related movement caused by coughing can be detected [134]. For example, a highly stretchable, transparent silk fibroin-based capacitive pressure sensor has been used to detect coughing [135]. Conductive silver nanofibers are used as the electrodes for the capacitive sensor. By mounting the sensor at the throat, the variations in capacitance can be detected and used to profile the motion status of the throat, which can then be used to estimate the patient's cough intensity and duration, as shown in figures 6(a) and (b). A stretchable fiber nanogenerator has been designed to convert mechanical energy to electric energy [136]. The fiber has a core–sheath coaxial structure with an air gap between the core electrode and the sheath electrode. When the fiber is pressed, the air gaps decrease and the electrostatic effect is enhanced, leading to a more positive charged sheath electrode and current flow across the core and sheath layers. Based on this mechanism, the fiber generator can be used to detect human motion when placed at different locations on the human body (left panel in figure 6(c)). When placed on the throat, the sensor could monitor coughing and other subtle movements (e.g. swallowing, drinking and speaking) (right panel of figure 6(c)). Moreover, continuous monitoring can be achieved due to the self-powering feature of the nanogenerator.

As well as pressure sensors, strain sensors (e.g. piezoresistive and piezoelectric sensors [138, 139]) have also been used as soft cough monitoring sensors. A rGO woven fabric-based piezoresistive strain sensor showed a high sensitivity (gauge factor of 416 within 0%–40% strain and 3667 within 48%–57% strain) [133]. The sensor was mounted on the prominentia laryngea to detect the mechanical vibration caused by coughing or other physiological motions such as speaking, chewing and swallowing. Mao *et al* developed a piezoelectric sensor by depositing ZnO nanorods onto rGO-modified silk fabrics [137]. Based on the piezoelectric effect of the ZnO nanorods, electrical responses to the mechanical deformation of the sensor were recorded. These signals can be used to assess the severity of coughs when the sensor is attached to the throat, as shown in figures 6(d) and (e). A self-powered flexible hybrid piezoelectric sensor with pressure and strain detecting capabilities was developed using single-crystal $(1-x)\text{Pb}(\text{Mg},\text{Nb})\text{O}_{3-x}\text{PbTiO}_3$ (PMN-PT) [131]. The piezoelectric coefficient d_{33} of PMN-PT can be up to 2500 pC N^{-1} , 90 times higher than that of ZnO. The sensor is attached to the throat (figure 6(f)), where the voltage signal is detected due to vibration of the vocal cords when the person coughs (figure 6(g)).

Stethoscopes are a commonly used means to detect coughs. A wearable stethoscope, made of layered piezoelectric film over silicone rubber, placed in a vest can comfortably and easily record the sound of the lungs [140]. The silicone rubber reduces ambient noise and provides a physical interface with the skin as it closely matches the acoustic impedance of the body. This device is particularly helpful to physicians for

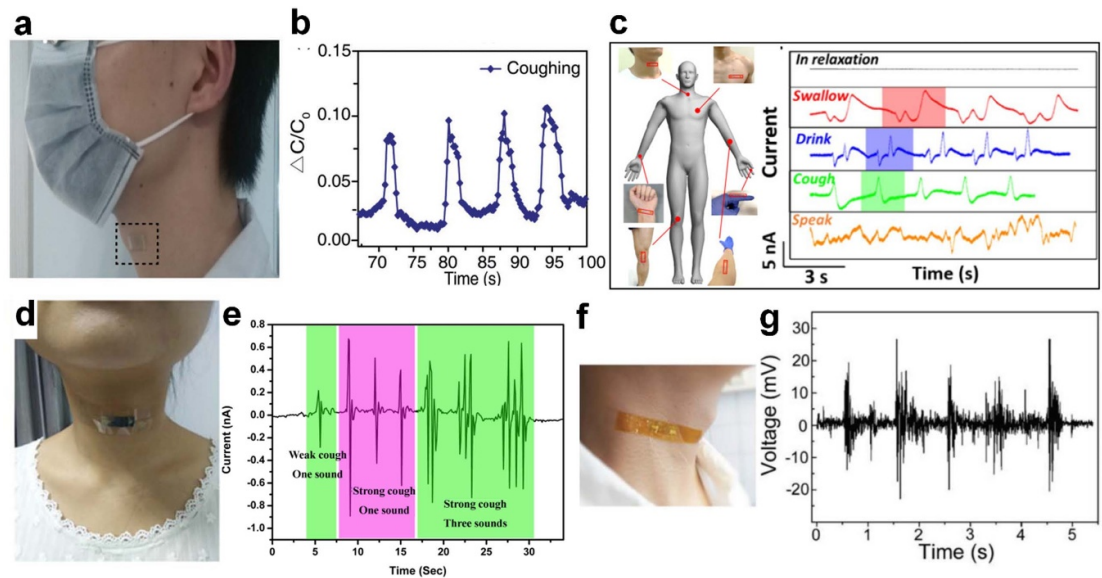


Figure 6. (a) Photograph of a silk-based pressure sensor attached to the throat to detect throat-related motion. (b) The coughing profile as monitored by the silk-based pressure sensor. (a), (b) [135]. John Wiley & Sons. © 2019 WILEY-VCH Verlag GmbH & Co. KGaA, Weinheim. (c) Schematic illustration of a fiber generator as wearable sensor attached to different positions on the human body to detect human motion (left panel), and the monitored throat-related activities including swallowing, drinking water, coughing and speaking. Reprinted from [136], Copyright (2017), with permission from Elsevier (d) A picture of a ZnO nanorod-based piezoelectric strain sensor attached to the throat to monitor coughing. (e) Real-time monitoring results of human coughing with different intensities. (d), (e) Reproduced from [137]. © IOP Publishing Ltd All rights reserved. (f) Picture of a $(1-x)\text{Pb}(\text{Mg},\text{Nb})\text{O}_{3-x}\text{PbTiO}_3$ (PMN-PT) ribbon-based piezoelectric strain sensor attached to the throat to monitor coughing. (g) The voltage signal detected by the PMN-PT ribbon-based piezoelectric strain sensor when the subject coughs. (f), (g) [131] John Wiley & Sons. © 2017 WILEY-VCH Verlag GmbH & Co. KGaA, Weinheim.

diagnosing COVID-19 patients as they can continuously and safely listen to the patient's lungs. To monitor mechano-acoustic signals, the Rogers group developed a device containing a flexible printed circuit board (FPCB), an accelerometer, a microcontroller and a wireless inductive charging circuit to charge a rechargeable 45 mAh lithium-ion polymer battery [141]. This device was mounted onto the suprasternal notch, a unique anatomical location, to measure HR, body energy expenditure and other parameters such as talking time and cadence.

3. Integrated multimodal sensor platforms

Patients with COVID-19 usually present multiple symptoms. Multimodal and correlated sensing of multiple physiological parameters (e.g. temperature, respiration rate, SpO_2 levels, electrophysiology, cough frequency and intensity) can lead to more precise diagnosis and treatment for COVID-19 patients. Kabiri Ameri *et al* developed a stretchable and transparent graphene-based multimodal sensor with sub-micrometer thickness [14]. The sensor is conformable to deformed skin with stable mechanical and electrical performance. The sensor patch was used to detect physiological signals including ECG, EMG, electroencephalogram (EEG), skin temperature and skin hydration. ECG and body temperature signal collection systems and profiles are shown in figures 7(a) and (b), respectively. This type of epidermal sensor is promising for unobtrusive continuous monitoring of physiological parameters. However, the interface between the soft sensor modules and the rigid silicon-based integrated circuits is challenging. To solve this problem, a wearable sensor patch with an ECG sensor and a temperature sensor was fabricated by printing functional materials on a flexible Kapton polyimide substrate that also hosts the silicon chips, as shown in figure 7(c) [142]. The fabrication procedure was modified to be compatible with the manufacturing of the FPCB. The sensor patch was used to monitor the ECG signal of a subject at rest and before, during and after exercise. The HR was extracted from the ECG signal. The thermistor was integrated with an analog to digital converter and a Bluetooth module for wireless monitoring of skin temperature.

A wireless, battery-free vital signs monitoring system that can collect ECG, PPG and skin temperature data through a soft and noninvasive interface with human skin is shown in figure 7(d) [115]. The configurations of the ECG and PPG epidermal electronics systems are shown in figures 7(e) and (f), respectively. HR, HR variability, respiratory rate, SpO_2 level, and body temperature can be extracted from the collected signals. A magnetic loop antenna was integrated into the platform to simultaneously transmit data

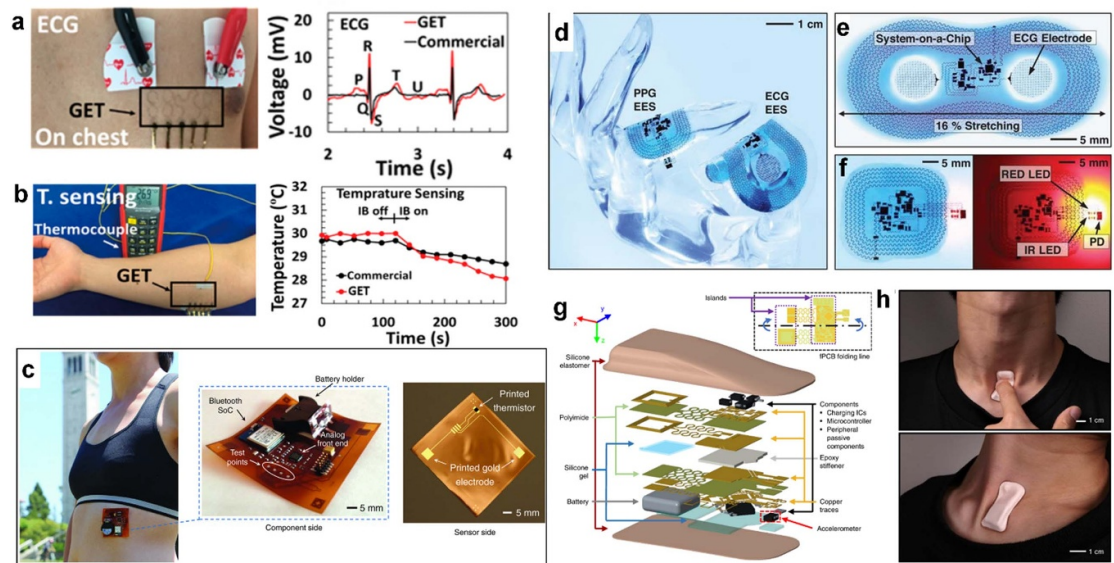


Figure 7. (a) The ECG and (b) body temperature sensors based on a graphene temporary tattoo multimodal sensor patch and the corresponding signals collected by the sensor patch, where GET represents graphene electronic tattoo. Reprinted with permission from [14]. Copyright (2017) American Chemical Society. (c) Photograph of the wearable sensor patch with printed sensors and hard silicon-based components on flexible Kapton polyimide substrate worn by the subject (left panel); the printed gold ECG electrodes and the thermistor before component assembly are shown in the right panel. [142] John Wiley & Sons. © 2016 WILEY-VCH Verlag GmbH & Co. KGaA, Weinheim. (d) Images of epidermal sensor patches for monitoring ECG and PPG, which are draped over a finger model (EES, epidermal electronic system). (e) Image of an ECG epidermal sensor patch stretched uniaxially in the horizontal direction by $\sim 16\%$. (f) PPG epidermal sensor patch operating in a lighted and a dark room (PD, photodiode). (d)–(f) from [115]. Reprinted with permission from AAAS. (g) Exploded schematic illustration of the active components, interconnect schemes and enclosure architecture of a stretchable, wireless and multimodal mechano-acoustic sensor system. (h) Images of the mechano-acoustic sensor system attached to the suprasternal notch during movements of the neck. Reprinted with permission from Springer Nature Customer Service Centre GmbH: Springer Nature, Nature Biomedical Engineering [141], Copyright © 2019, The Author(s), under exclusive licence to Springer Nature Limited.

wirelessly using the NFC protocols. The effective elastic modulus of the sensor patch is in the range of 200–300 kPa, which minimizes the stresses at the skin interface associated with body movement. The gentle adhesion (van der Waals force), enabled by the inherently thin, soft sensor patch, avoids possible skin injury when the sensor is removed after each use. A stretchable, wireless mechano-acoustic sensor system was designed and tested comprehensively with regard to hardware integration and data acquisition, as shown in figure 7(g) [141]. The sensor modules are encapsulated by silicone elastomer and mounted on a unique location on the suprasternal notch to enable multimodal sensing (figure 7(h)). Stretchability is enabled by the bridge–island strategy, in which two rectangular regions with rigid components (‘island’) are connected with each other by serpentine interconnects (‘bridge’). Simultaneous monitoring of ECG, EEG, electrooculogram, throat motions and skin temperature can be achieved. Time and frequency-domain analysis of these signals is enabled by advanced data algorithms with high fidelity. However, the motion artifacts for the cardiac signal and respiratory rate signal still need to be reduced when using this system as a practical continuous monitoring platform.

4. Summary and perspective

This review has presented a brief overview of soft wearable sensors with the potential for monitoring symptoms of COVID-19. In the past decade or so, the rapid development of flexible and stretchable electronics has transformed the field of wearable sensors. Wearable sensors can now be soft, thin, conformal to the skin and deformable with the skin with no mechanical mismatch (i.e. they have a similar Young’s modulus). A variety of symptoms of COVID-19 can be monitored by different types of soft wearable sensors, such as body temperature (temperature sensor), respiration (temperature and humidity sensors), oxygen saturation (pulse oximeter), ECG and HR (ECG electrode), and cough (pressure and strain sensors and stethoscope).

System-level integration, including a wireless communication module, power source and display in these sensors, has significantly enhanced their practical applicability in the healthcare field. By continuously monitoring physiological parameters, wearable sensors can play an important role in rapid screening and monitoring of COVID-19 patients. Wearable technologies can reduce the burden on the hospital system by

monitoring symptoms at improvised hospitals or at home. Integrated with AI and machine learning, the large amount of sensing data may facilitate timely and precise medical treatment decisions by medical professionals. In addition to COVID-19, the presented wearable technologies can help respond to other respiratory diseases (e.g. 2009 H1N1 influenza, asthma and chronic obstructive pulmonary disease) and other types of diseases in general.

While the concept of telehealth has drawn much attention in the past decades, wearing comfort, data accuracy and reliability, scalable manufacturing and long-term monitoring capability are still challenging, and these factors need to be addressed before wearable sensors can become an important part of the healthcare system. The following directions are emerging and may address the challenges mentioned above.

- (a) Advancement of flexible, stretchable, conformal, and permeable functional materials. A large number of flexible and stretchable materials, especially composite materials containing nanomaterials such as nanoparticles, nanowires, CNTs and 2D materials, have been developed recently [24, 33, 56, 143–151]. For epidermal (or on-skin) electronics, the materials should be conformal to the skin [30, 45, 152–156]. To achieve this, it is critical to reduce their thickness as bending stiffness is proportional to thickness cubed. For comfortable wear, the materials should be gas permeable. A variety of methods have been developed recently to fabricate permeable materials, [157–159] including a scalable breath figure method [111]. In addition, research on developing green wearable sensors using biodegradable materials is emerging [160–163].
- (b) Enablement of multimodal sensing that can monitor and correlate multiple symptoms [164–167]. One challenge is the cross-sensitivity of different sensors. One sensor may respond to multiple stimuli, making it difficult to identify and quantify the stimuli. Novel sensing mechanisms and/or designs with low cross-sensitivity are needed to decouple the stimuli [167–169]. Multimodal sensing can facilitate the application of AI in telehealth. AI-enabled data fusion can achieve comprehensive monitoring of disease features, leading to more accurate diagnosis and treatment decisions [170–172].
- (c) Development of low-cost, scalable fabrication methods. Both top-down and bottom-up approaches are used to fabricate soft wearable devices. Printed electronics on flexible substrates (e.g. plastics) are well established. This presents new challenges with regard to printing 1D and 2D nanomaterials on stretchable substrates, including surface properties of the substrate, ink formulation and post-processing [173]. As an example, methods such as screen printing, electrohydrodynamic printing and gravure printing of AgNWs have been reported [110, 174, 175]. Hybrid methods that combine bottom-up and top-down approaches have been demonstrated with promise [176, 177]. When possible, it is conducive to fabricate different types of sensors using the same material. For example, AgNWs have been used to fabricate strain sensors, hydration sensors and ECG and EMG electrodes on the same patch [45, 112]. Often, heterogeneous integration of different materials is necessary, where interface mismatch in terms of mechanical, thermal and electrical properties should be accounted for [21, 178].
- (d) Achievement of continuous, long-term monitoring. A major challenge for this is to maintain a positive power balance, which can be addressed by minimizing the power consumption of the sensors and/or maximizing the power supply (e.g. either batteries or energy harvesting) [179, 180]. New materials and sensing modalities that consume less power and low-power electronics have been emerging. For example, OLEDs and OPDs are being developed for the pulse oximeter system to decrease energy consumption [90]. In parallel, efforts have been devoted to energy harvesting from the body (e.g. in the form of heat and movement/strain using soft materials) [179], while enhancing energy storage capabilities using batteries and supercapacitors [181–185].

Data availability statement

No new data were created or analysed in this study.

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ORCID iDs

Yuxuan Liu  <https://orcid.org/0000-0001-8196-1054>

Yong Zhu  <https://orcid.org/0000-0002-3862-5757>

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