Envisioned strategy for an early intervention in virus-suspected patients through non-invasive piezo- and pyro-electric-based wearable sensors

Sujoy Kumar Ghosh \(^a\) and Dipankar Mandal \(^a,b\)

The intervention of virus-infected persons mainly relies on diagnostic methods, and accordingly COVID-19 is not an exception. Thus, a major research direction in viral diagnosis is aligned towards biological- and biomedical-based approaches. In contrast, there is plenty of scope in wearable devices towards early intervention and continuous health status monitoring. Of particular interest, piezo- and pyro-electric wearable sensors can play a significant role by detecting physiological signals in the virus-affected patients. Remotely monitoring physiological signals, such as temperature, respiration, heart rate and other data is an added advantage, where the integration of artificial intelligence is possible. This can improve clinical decision-making paths. In this perspective, the most relevant piezo- and pyro-electric sensor-based wearable sensors for healthcare monitoring towards the early detection of virus-affected abnormalities are highlighted. To implement these types of sensors, the relevant fundamentals of piezo- and pyro-electricity are also discussed. Additionally, relevant materials and device structures are reviewed in order to understand the pros and cons, and thus further improvement can be applied according to the requirements. The envisioned strategy for the early detection of virus-suspected patients through non-invasive wearable sensors is becoming very important in non-contact-based physiological data collection, particularly in the context of the ongoing COVID-19 pandemic.

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1. Introduction

A common introduction regarding viral diagnosis and limitations is necessary to start discussing the ongoing COVID-19 crisis, which is an infectious respiratory illness caused by the novel coronavirus SARS-CoV-2. The World Health Organization (WHO) officially announced the COVID-19 outbreak as a pandemic due to its rapid human-to-human transmission. The basic signs and symptoms of COVID-19 may become noticeable between 2 to 14 days, i.e., the incubation period, after exposure. The common signs and symptoms include body temperature fluctuation during fever, cough, and shortness of breath or difficulty breathing. To date, no reliable vaccines are available in the market for this pandemic, and thus early rapid point-of-care diagnosis, isolation, surveillance and monitoring, and management are crucial. At this stage of emergency, the availability of a cost-effective and easily accessible health-care system is essential for frontline health workers, i.e., doctors, nurses and other medical staff, where clinicians can monitor the heartbeat, body temperature, coughing signal, etc. of COVID-19 patients round the clock without repeated direct physical contact with the patient, which simply prevent the further spread of the virus. To date, the research on COVID-19 mainly is focused on biological and biomedical research, which includes virus identification, gene sequencing, cogent protein structure analysis, development of diagnosis kits, vaccine trials and other medical interventions. However, before the invention of a vaccine, it is very important to reduce risks to clinicians and normal people exposed to the virus during treatment because clinicians are in direct contact with the COVID-19 confirmed and suspected patients. Also, many people are not going to hospitals because of the fear of being exposed to the virus.

Accordingly, in this scenario, wearable healthcare monitoring sensors with the help of artificial intelligence (AI) can play a significant role, through which the vital signs of COVID-19 suspected and confirmed patients including respiration rate, heart rate and body temperature can be monitored in real time.

1.1 How can wearable sensors help?

The common individuals are primarily divided into three categories.

(i) Not affected people,
(ii) Suspected cases,
(iii) Confirmed cases.

By continuous and real-time monitoring of physiological signals, initially individuals can be separated as suspected and non-suspected cases. Further monitoring of vital signs can help to separate the COVID-19 suspected cases from the common flu-infected people. The illness caused by the COVID-19 can be roughly divided into two stages. During the first stage, when symptoms are not acute, infected individuals can generally stay at home. For serious cases, this is followed by a second stage with worsening symptoms, in which some patients develop severe pneumonia. In these cases, patients must be admitted to a hospital as quickly as possible. The sooner they receive good medical treatment, the better the prognosis. During the first-stage, patients can engage in self-monitoring using personal healthcare systems primarily by acquiring their temperature, heart rate and respiration rate at home, and to call the health department or their family doctor in case of certain worsening conditions. Overall, wearable sensors can play a significant role in monitoring infected patients in the following manner:

I. The lightweight sensor attached to an individual’s body will record their vital signs including respiration rate, heart rate and temperature in real time and send the data through IoT to the clinician.

II. The data collected by the sensor will be analysed immediately by a computer algorithm. Subsequently, these parameters will be compared with standard values by a healthcare professional every 4–6 hours.

III. Finally, the data can be seen by a medical team, alerting them when a person’s health may be deteriorating. A schematic of this process is shown in Fig. 1.

1.2 Which problems can be solved by wearable sensors?

i. The continuous and real-time monitoring of health status by wearable sensors will not only allow doctors and nurses to respond quicker to the needs of patients and transfer them from the community to hospital when necessary, but also potentially reduce the exposure of healthcare staff to coronavirus.

ii. Reducing contact between individuals in quarantine and healthcare workers will also limit the use of personal protective equipment (PPE).

iii. Reduce the demand for COVID-19 testing kits.

iv. The physiological conditions of clinicians can also be directly recorded and possibly monitored day-by-day because generally, is not possible for the health status of doctors and nurses to be checked regularly by themselves or other medical staffs due to their continuous engagement with patients during this pandemic.

v. Quicker treatment is possible than standard care.

vi. Faster health check-up may reduce unnecessary queues.

2. Current approaches

Currently, very limited approaches have been observed in a few medical industries and academia to implement wearable sensors. For example, a sweat sensor was developed by Epicore Biosystems (Fig. 2a), a wearable patch was developed by Sensium for the early detection of patient deterioration (Fig. 2b), and thermal imaging wearables were developed by Rokit and used in China to detect COVID-19 (Fig. 2c). Furthermore, the Shanghai Public Health Clinical Centre (SPHCC) used the temperature sensor from the California-based connected health start-up VivaLNK to monitor COVID-19 patients (Fig. 2d). From the academic side, Imperial College London, in partnership with the National Health Service (NHS) organisation, implemented a sensor to remotely monitor people in quarantine at a special NHS facility near Heathrow airport, for example travellers from abroad or those wishing to travel to return home.
The main challenge associated with the current wearable sensors lies in their limited wearability, multi-functionality (i.e., individually and simultaneously detecting pressure and temperature), accuracy and low sensitivity. Pressure sensing can be achieved via several transduction mechanisms, such as, piezo-resistive, piezo-capacitive, piezo-electric and tribo-electric mechanisms. Among them, the piezo- and tribo-electric mechanisms are well suited due to their fast response time, high sensitivity and self-powered operation mode. However, despite the tremendous research progress, trboelectric sensors are not typically ideal, considering their long-term instability issues. Thus, more research is needed. On the other hand, piezoelectric sensors are ideally suited as pressure sensors since they show long-term stability. On the other hand, among the temperature transduction mechanisms, thermoelectricity and pyro-electricity show self-powered operation feasibility. However, the temperature sensing response time of thermoelectric sensors is limited by the thermal diffusivity of their material. In contrast to thermoelectric sensors, which depend on a spatial temperature gradient, pyroelectric sensors are highly desirable in the event where thermal gradients are difficult to access or the heat source temperature fluctuates similar to human body temperature variations. In this regard, piezo- and pyro-electric materials are ideal candidates for the development of temperature–pressure dual mode healthcare monitoring devices because in this case, with a single material, all biophysical signals can be detected, which is beneficial for the miniaturization of electronic systems. In this perspective, we will highlight a few novel approaches for piezo- and pyro-electric sensors that show feasibility for healthcare monitoring functionality.
3. Pyroelectric healthcare monitoring sensors

3.1 Fundamentals of pyroelectricity

Pyroelectricity arises from the non-centrosymmetric crystal structure of materials, where a change in temperature ($T$) results in a change in polarization ($P$). Thus, the pyroelectric coefficient is described by,$^{11}$

$$\pi = \frac{\partial P}{\partial T}. \quad (1)$$

Additionally, a change in polarization also generates an output current, which is quantitatively described by

$$i_p = \pi A \frac{dT}{dt}. \quad (2)$$

where $A$ is the effective area. Evidently, the performance of pyroelectric devices is related with their area and temporal change in temperature, which forms the basis for modern thermal energy conversion devices. Since the pyroelectric current is generated from the change in surface charge density, it can also arise from temperature-induced strain in piezoelectric materials,$^{12}$ thermal dependence of the dielectric permittivity, and flexoelectric effects$^{13}$ from thermal gradients in all materials. To harvest thermal waste heat, pyroelectric devices mimic a thermodynamic heat engine. In this case, the polarization corresponds to the volume and the electric field is analogous to the pressure in the working fluid. Several thermodynamic cycles have been proposed for pyroelectric energy conversion. These cycles are represented by polarization ($P$)-electric field ($E$) cycles, where the thermal effect on polarization generated by an electric field is represented. Among them, the Brayton (i.e., two isoelectric ($2 \rightarrow 3, 4 \rightarrow 1$) and two adiabatic ($1 \rightarrow 2, 3 \rightarrow 4$) processes, Fig. 3a), Stirling (i.e., two isodisplacement ($2 \rightarrow 3, 4 \rightarrow 1$) and two isothermal ($1 \rightarrow 2, 3 \rightarrow 4$) processes, Fig. 3b), Carnot (two adiabatic ($2 \rightarrow 3, 4 \rightarrow 1$) and two isothermal ($1 \rightarrow 2, 3 \rightarrow 4$), Fig. 3c) cycles, and Ericsson (or Olsen) cycle (i.e., two isothermal and isoelectric processes, Fig. 3d) represent the most effective and employed thermodynamic cycles for pyroelectric energy conversion.$^{14}$ The energy conversion efficiency of this cycle is represented as,

$$\eta = \frac{\int_{T_L}^{T_H} E dP}{\int_{T_L}^{T_H} C(T) dT}, \quad (3)$$

where $T_H$ ($T_L$) is the temperature for the heat source (heat sink), $C(T)$ is the heat capacity and $\int E dP$ is the total electrical work done ($W_E$).$^{15}$
Additionally, the pyroelectric figure-of-merit can be represented as

\[
\text{FoM}_{\text{py}} = \frac{\pi^2 T}{C(T)\varepsilon_0 \varepsilon_r^3}. \tag{4}
\]

Thus, a higher energy conversion requires a simultaneous increment in \(\pi\) with a reduction in the dielectric permittivity \(\varepsilon_r\). Additionally, for pyroelectric devices, the maximum power can be represented as \(P_{\text{max}} = W_{\text{e}} f_{\text{max}} = \frac{W_{\text{e}}}{T}\), where \(f_{\text{max}}\) is the maximum cycling frequency.\(^{14}\) Since the thermal time-constant is higher than the electrical time constant, \(t_{\text{thermal}} = \frac{L^2}{\alpha}\), where \(L\) is the thickness and \(\alpha\) is the thermal diffusivity of the material. Thus, \(P_{\text{max}} = \frac{W_{\text{e}}/t_{\text{thermal}}}{1/L^2}\), which indicates that thin film devices have a higher power density than bulk devices.

This approach was recently demonstrated by relaxor ferroelectric thin films.\(^{17}\) In addition, a lower specific heat capacity is suitable for higher pyroelectric energy conversion according to eqn (4). However, according to eqn (2), it is evident that a higher \(\frac{dT}{dt}\) is required for higher pyroelectric current. Apparently, \(C(T)\) increases with an increase \(T\). Accordingly, researchers have used several strategies where large heat fluctuations are generated within materials even if the external thermal fluctuation is low. Thus, pyroelectric materials with a lower \(C(T)\) can harvest higher \(\frac{dT}{dt}\).

### 3.2 Wearable pyroelectric devices

#### 3.2.1 Electrode design

To enhance thermal fluctuations within the materials of devices, different electrode designs are necessary. For example, owing to their higher infrared light...
absorbing property, carbon nanotubes (CNTs) were used as an electrode for the fabrication of a higher performance PVDF-based pyroelectric generator (Fig. 4a(i)). The device was stimulated under a 1.45 W cm\(^{-2}\) near infrared (NIR) laser (808 nm) (Fig. 4a(ii)). CNTs showed the largest temperature fluctuations among other electrodes, such as graphene, aluminium (Al) and ITO (Fig. 4a(iii)), which enhanced the temperature-change rate, and finally caused an enhancement in the output current (\(\sim 9\) nA) (Fig. 4a(iv)). Besides CNTs, interconnected graphene nanoplatelets (GNPs) were also shown to be efficient screen-printed electrode (Fig. 4b(i)) with high electrical conductivity and high thermal radiation absorbance capability. Consequently, the electrode further enhanced the pyroelectric energy harvesting ability of PVDF (Fig. 4b(ii)). Generally, metal electrodes including Al and gold (Au) have high thermal and electrical conductivity, but low radiation absorbance (\(~0.1–0.3\)). Thus, illuminated thermal energy mostly reflects or transmits at the electrode surface, leading to a small change in temperature in the pyroelectric active material. The graphene ink electrode material absorbed most of the available radiation heat energy, thereby maximizing the rate of change in temperature, \(d\Delta T/dt\), which improved the pyroelectric energy harvesting device performance and effectiveness (Fig. 4b(iii)). For the graphene ink/PVDF/Al system, the closed circuit pyroelectric current was improved by 7.5 times, the open circuit voltage by 3.4 times, and the harvested energy by 25 times compared to the standard Al/PVDF/Al system electrode design (Fig. 4b(iv)). However, to use the higher electrical conductivity and thermal conductivity of the metal electrode as an additional benefit, another effective approach was demonstrated by partially covering the PVDF surface by micro patterned Al electrodes, which enhanced the heat transfer and achieved larger temperature fluctuations under an IR bulb (Fig. 4c(ii)).

Among several electrode coverage areas (100%, 88%, 70%, 53%, 45%, 28%, and 19%), the devices consisting of 45% and 53% coverage area showed about a 30% larger temperature change (Fig. 4c(iii)). This large temperature changes led to faster rates of change in temperature and a 400% higher voltage (as high as 60 V) and current output performance (Fig. 4c(iii)). Besides, an all-organic approach was demonstrated, where the photothermal effect of PEDOTs was used for pyroelectric energy harvesting by coating them on both sides of P(VDF-TrFE) (Fig. 4d(i)). The photothermally driven pyroelectric device was also further hybridized for harvesting solar light (Fig. 4d(ii)).

3.2.2 Materials design. Besides the electrode effect to improve the thermal absorbing property, material engineering is also very important to improve \(d\Delta T/dt\). Accordingly, an Er\(^{3+}\)-modified self-poled PVDF (Er–PVDF) film (Fig. 5a(i)) was demonstrated to exhibit enhanced pyroelectric property.\(^{22}\) Er\(^{3+}\), which is a good absorber of IR light (Fig. 5a(ii)), generated more heat within PVDF under same thermal oscillations, leading to enhanced thermal energy harvesting without any device structure engineering. Thus, under larger thermal fluctuations of \(\Delta T\sim 14\) K to 24 K, an increase in the output current of 12.5 nA to 15.5 nA was observed (Fig. 5a(iii)). Besides organic materials, inorganic materials have also been proven to be fruitful for pyroelectric energy harvesting.

For example, a thermally conductive AlN additive dispersed in thermally conductive networks in Pb\([\text{Mn}_{0.1}\text{Nb}_{0.6}\text{Sb}_{0.3}]_{2/3}\text{[Mn}_{0.2}\text{Sn}_{0.3}]_{1/3}\text{[Zr}_{0.5}\text{Ti}_{0.5}]_{1/3}\text{O}_{3}\) (lead magnesium niobate–lead antimony–manganese–lead zirconate titanate: PMN–PMS–PZT) ceramics improved the heat transfer and enhance their ferroelectric properties.\(^{23}\) The AlN filler efficiently transfers the diffused thermal wave energy to the thermally homogeneous PMN–PMS–PZT matrix due to the vibrations of the whole chain and phonon scattering (Fig. 5b). To enhance
pyroelectricity together with flexibility, the fabrication of organic/inorganic composites has been considered a good strategy. For example, benefiting from the higher pyroelectricity of PZT ceramic powder, a PVDF/PZT composite film showed a 30% enhancement in pyroelectricity (pyroelectric coefficient: $95 \ \mu C/m^2 K^{-1}$) in comparison to the PVDF film. In another approach, researchers observed that potassium–sodium niobate (KNN) powder was also capable of enhancing the pyroelectric co-efficient of a P(VDF–TrFE)/KNN composite-based flexible film up to $68 \ \mu C/m^2 K^{-1}$. A drastic improvement was overserved by incorporating BaTiO3 nanoparticles into PVDF. The triple-layer PVDF/BaTiO3 nanocomposite possessed a pyroelectric co-efficient as high as $268.49 \ \mu C/m^2 K^{-1}$. Furthermore, a strontium barium niobate, Sr$_{x}$Ba$_{y}$Nb$_{2}$O$_{6}$/polyurethane (PU), nanocomposite showed an extraordinary pyroelectric co-efficient of $380 \ \mu C/m^2 K^{-1}$. Additionally, a PVDF/lithium tantalate (LT) nanocomposite was also demonstrated to enhance the pyroelectricity of flexible PVDF.

However, despite the enormous progress in pyroelectric devices, the essential feature required for early intervention in the COVID-19 pandemic is that the devices should be wearable. Masks, which are primarily used for protection from the spread of the virus, is an ideal platform for wearability in the case of pyroelectric devices. The wearable feasibility of a pyroelectric device was demonstrated by attaching a commercially available Al-coated PVDF film to an N95 mask (Fig. 6a(i)). Thus, the mask itself works as a wearable breathing sensor by harvesting temperature oscillations generated during human respiration and exhalation (Fig. 6a(ii)). The dynamic process of breathing formed the time-dependent temperature fluctuations of around $12 \ ^\circ C$, which generated huge output responses (Fig. 6a(iii)). Additionally, the harvested output electrical signals were further used to operate commercial electronic gadgets such as a liquid crystal display (LCD) and array of light-emitting diodes (LEDs). In other perspectives, these types of approaches are very useful for monitoring breathing responses, in particular during the COVID-19 pandemic, such as by sensing the change in...
output responses, clinicians can easily monitor the change in the temperature gradient of the patients. Although a PVDF thin film could be used as a wearable pyroelectric device, nanofiber-based PVDF is required to overcome the limitation of the thermal sensitivity, flexibility and stretchability of PVDF thin films. Thus, to overcome this challenge, an all-fiber pyroelectric nanogenerator was demonstrated, which harvested as low as 2 K temperature variations. Furthermore, a highly aligned P(VDF-TrFE)-based pyroelectric energy harvester was also reported, which possesses a very high pyroelectric co-efficient of 68 μC m⁻² K⁻¹. In one step forward, an electrospun graphene oxide (GO)–PVDF composite nanofiber-based pyroelectric device was demonstrated as a highly efficient breathing sensor, which could detect very mild temperature fluctuations (≈4.5 K) during breathing at room temperature (Fig. 6b). Additionally, a commercially available PZT-based self-powered temperature sensor was demonstrated to harvest waste body heat, i.e., heat dissipation from the human body surface and from the process of respiration (Fig. 6c). The sensor could generate a significant output voltage without direct contact between the sensor and the human body.

Furthermore, for human body conformability, a stretchable pyroelectric device was developed using micropatterned P(VDF–TrFE) film which could detect very mild temperature fluctuations during breathing at room temperature (Fig. 6b).

### Table 1 Summary of the performances of pyroelectric materials

<table>
<thead>
<tr>
<th>Material</th>
<th>Working conditions (ΔT, dT/dt)</th>
<th>Pyroelectric co-efficient (μC m⁻² K⁻¹)</th>
<th>Output current</th>
<th>Reference</th>
</tr>
</thead>
<tbody>
<tr>
<td>PVDF film</td>
<td>35 K</td>
<td>27</td>
<td>190 V, 11 μA</td>
<td>49</td>
</tr>
<tr>
<td>PVDF film</td>
<td>2.8 K, 0.014 K s⁻¹</td>
<td>33</td>
<td>60 V, 30 nA</td>
<td>20</td>
</tr>
<tr>
<td>PVDF film</td>
<td>25 K, 23 K s⁻¹</td>
<td>27.2</td>
<td>145 V, 120 nA</td>
<td>50</td>
</tr>
<tr>
<td>PVDF film</td>
<td>&gt;5 K</td>
<td>27.2</td>
<td>5.7 V, 109 nA</td>
<td>51</td>
</tr>
<tr>
<td>P(VDF-TrFE) film</td>
<td>13 K</td>
<td>43.9</td>
<td>13.65 V, 2.69 μA</td>
<td>52</td>
</tr>
<tr>
<td>P(VDF-TrFE) film</td>
<td>18.5 K, 100 K s⁻¹</td>
<td>2.48 V, 570 nA cm⁻²</td>
<td></td>
<td>34</td>
</tr>
<tr>
<td>Er-PVDF film</td>
<td>24 K, 2 K s⁻¹</td>
<td>33</td>
<td>15 nA</td>
<td>22</td>
</tr>
<tr>
<td>PVDF</td>
<td>12 K, 13 K s⁻¹</td>
<td>27</td>
<td>42 V, 2.5 μA</td>
<td>29</td>
</tr>
<tr>
<td>PVDF/PZT</td>
<td>30–50 °C</td>
<td>95</td>
<td></td>
<td>24</td>
</tr>
<tr>
<td>P(VDF-TrFE)/KNN</td>
<td>Not mentioned</td>
<td>68</td>
<td>Not mentioned</td>
<td>25</td>
</tr>
<tr>
<td>PVDF/BaTiO₃</td>
<td>130 °C</td>
<td>268.49</td>
<td></td>
<td>26</td>
</tr>
<tr>
<td>Sr₀.₃Bi₀.₇Nb₂O₆/polyurethane (PU)</td>
<td>300–335 K</td>
<td>380</td>
<td>Not mentioned</td>
<td>27</td>
</tr>
<tr>
<td>PVDF/lithium tantalate (LT)</td>
<td>Not mentioned</td>
<td>147</td>
<td>Not mentioned</td>
<td>28</td>
</tr>
<tr>
<td>PVDF nano-fiber</td>
<td>14 K, 1.5 K s⁻¹</td>
<td>0.062</td>
<td></td>
<td>30</td>
</tr>
<tr>
<td>PVDF nano-fiber</td>
<td>6 K, 10 K s⁻¹</td>
<td>20 nA</td>
<td></td>
<td>53</td>
</tr>
<tr>
<td>PVDF-GO nano-fibers</td>
<td>22 K, 2.12 K s⁻¹</td>
<td>0.027</td>
<td></td>
<td>32</td>
</tr>
<tr>
<td>PVDF-CH₃NH₂PbI₃, nanofibers</td>
<td>38 K, 2.26 K s⁻¹</td>
<td>0.044</td>
<td></td>
<td>54</td>
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<tr>
<td>Hydroxyapatite thin film</td>
<td>50 °C, 1 °C min⁻¹</td>
<td>12</td>
<td></td>
<td>36</td>
</tr>
<tr>
<td>Fluorapatite/gelatin</td>
<td>Not mentioned</td>
<td>0.05</td>
<td></td>
<td>35</td>
</tr>
<tr>
<td>Natural human skin</td>
<td>Not mentioned</td>
<td>0.021–0.027</td>
<td></td>
<td>37</td>
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<tr>
<td>Dentine and cementum</td>
<td>Not mentioned</td>
<td>0.025–0.0015</td>
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<td>38</td>
</tr>
<tr>
<td>Hoof tendon</td>
<td>−35 to 85 °C, 14 °C min⁻¹</td>
<td>0.004</td>
<td>Not mentioned</td>
<td>39</td>
</tr>
<tr>
<td>dabcoHReO₄ fibers</td>
<td>6 K, 0.2 K s⁻¹</td>
<td>8.5</td>
<td></td>
<td>40</td>
</tr>
<tr>
<td>ZnO nano-wires</td>
<td>30 K, 1.8 K s⁻¹</td>
<td>12–15</td>
<td></td>
<td>41</td>
</tr>
<tr>
<td>ZnO thin film</td>
<td>2 K min⁻¹</td>
<td>10</td>
<td></td>
<td>42</td>
</tr>
<tr>
<td>KNbO₃ nano-wires</td>
<td>39 K, 2 K s⁻¹</td>
<td>8</td>
<td>120 pA</td>
<td>43</td>
</tr>
<tr>
<td>PZT</td>
<td>45 K, 2.3 K s⁻¹</td>
<td>800</td>
<td></td>
<td>44</td>
</tr>
<tr>
<td>AlN thin film</td>
<td>Not mentioned</td>
<td>6–8</td>
<td></td>
<td>45</td>
</tr>
<tr>
<td>CdS</td>
<td>25 °C to −196 °C</td>
<td>4</td>
<td></td>
<td>46</td>
</tr>
<tr>
<td>CdSe</td>
<td>25 °C to −196 °C</td>
<td>3.5</td>
<td></td>
<td>46</td>
</tr>
<tr>
<td>BaTiO₃</td>
<td>12 K, 0.85 K s⁻¹</td>
<td>225–259</td>
<td></td>
<td>47</td>
</tr>
<tr>
<td>BiFeO₃</td>
<td>1.86 K, 0.21 K s⁻¹</td>
<td>Not mentioned</td>
<td></td>
<td>48</td>
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Fig. 7 Ultra-thin PZT-based self-powered healthcare monitoring sensor smartly operates LEDs and speaker modules using human vital signals (reproduced from ref. 57 with permission by the publisher).
Table 2  Summary of the nanofiber-based inorganic piezoelectric materials for a comparison of their energy harvesting performances

<table>
<thead>
<tr>
<th>Inorganic materials</th>
<th>Electrode</th>
<th>Voltage/current</th>
<th>Power</th>
<th>Reference</th>
</tr>
</thead>
<tbody>
<tr>
<td>PZT</td>
<td>Ag</td>
<td>6 V, 45 nA</td>
<td>1 μW cm$^{-2}$</td>
<td>58</td>
</tr>
<tr>
<td>ZnO</td>
<td>Au-coated textile</td>
<td>8 V, 2.5 μA</td>
<td>Not mentioned (NM)</td>
<td>59</td>
</tr>
<tr>
<td>ZnO</td>
<td>Au-coated textile</td>
<td>9.5 mV</td>
<td>NM</td>
<td>60</td>
</tr>
<tr>
<td>PZT/Ag/polymer</td>
<td>Ag polymer ink</td>
<td>19 V, 0.54 μA</td>
<td>1.6 W cm$^{-2}$</td>
<td>61</td>
</tr>
<tr>
<td>PZT</td>
<td>Carbon film</td>
<td>60 V, 500 nA</td>
<td>0.356 μW cm$^{-2}$</td>
<td>62</td>
</tr>
<tr>
<td>ZnO</td>
<td>Ag-coated fabric</td>
<td>4 V, 20 nA</td>
<td>NM</td>
<td>63</td>
</tr>
<tr>
<td>Yarn intersection ZnO/Pd-covered ZnO</td>
<td>Cu wires</td>
<td>3 mV, 17 pA</td>
<td>NM</td>
<td>64</td>
</tr>
<tr>
<td>BaTiO$_3$</td>
<td>Cu wire</td>
<td>1.9 V, 24 nA</td>
<td>NM</td>
<td>65</td>
</tr>
<tr>
<td>PZT ribbons</td>
<td>NM</td>
<td>30 pA</td>
<td>NM</td>
<td>66</td>
</tr>
</tbody>
</table>

Fig. 8  (a) (i) Ferroelectric polymer-based transducer for (ii) pulse wave monitoring under normal condition and (iii) under drug dose condition (reproduced from ref. 88 with permission by the publisher). (b) (i) P(VDF–TrFE)/BaTiO$_3$ composite micro-pillar array as (ii) healthcare monitoring sensor (reproduced from ref. 89 with permission by the publisher). (c) (i) P(VDF–TrFE) nanowire-based (ii) vital sign monitoring sensor (reproduced from ref. 90 with permission by the publisher).
PVDF/CH$_3$NH$_3$PbI$_3$ Ni
PVDF/BaTiO$_3$ Silver-coated nylon yarn 4 V 43.5
PVDF/PEDOT 1 V/0.15 mA Not mentioned 99
PVDF/PEDOT-coated PVDF nanothreads 14 V/29.8 mA 47
PVDF/PDMS Conductive PEDOT-coated PVDF nanothreads 16.2 mV Not mentioned 103
PVDF/PEDOT Conductive thread 1 V/20 mA Not mentioned 107

Table 3 Summary of the nanofiber-based organic piezoelectric materials for comparison of their energy harvesting performances

<table>
<thead>
<tr>
<th>Electrospun fibers</th>
<th>Electrode</th>
<th>Voltage/current</th>
<th>Power</th>
<th>Ref.</th>
</tr>
</thead>
<tbody>
<tr>
<td>PVDF/graphene oxide</td>
<td>Cu-Ni plated fine knit polyester fabric</td>
<td>7 V</td>
<td>6.2 mW m$^{-2}$</td>
<td>32</td>
</tr>
<tr>
<td>PVDF</td>
<td>Conductive nanofiber membrane (PVP, PEDOT:PSS, ethyl alcohol, ethylene glycol and ionic liquid)</td>
<td>8 V/3.76 $\mu$A</td>
<td>Not mentioned</td>
<td>53</td>
</tr>
<tr>
<td>PVDF/CH$_3$NH$_3$PbI$_3$</td>
<td>Ni-Cu-coated fabrics</td>
<td>2 V/50 $\mu$A</td>
<td>0.8 mW m$^{-2}$</td>
<td>54</td>
</tr>
<tr>
<td>Pt/PVDF</td>
<td>Cu-Ni plated fabrics</td>
<td>30 V/1.38 mA</td>
<td>22 $\mu$W cm$^{-2}$</td>
<td>91</td>
</tr>
<tr>
<td>PVDF</td>
<td>Cu foil</td>
<td>76 mV/39 mA</td>
<td>577.6 $\mu$W cm$^{-2}$</td>
<td>92</td>
</tr>
<tr>
<td>Random PVDF</td>
<td>Al foil</td>
<td>2.21 V/4 $\mu$A</td>
<td>2.24 $\mu$W cm$^{-2}$</td>
<td>93</td>
</tr>
<tr>
<td>PVDF</td>
<td>PEDOT-coated PVDF nanofiber</td>
<td>48 V/6.0 $\mu$A</td>
<td>8.5 $\mu$W cm$^{-2}$</td>
<td>94</td>
</tr>
<tr>
<td>PVDF</td>
<td>Ag-coated nanothreads</td>
<td>3.4 V/4.4 $\mu$A</td>
<td>Not mentioned</td>
<td>95</td>
</tr>
<tr>
<td>PVDF</td>
<td>Ag conductive fabric</td>
<td>8.36 V/0.17 $\mu$A</td>
<td>77.69 $\mu$W cm$^{-2}$</td>
<td>96</td>
</tr>
<tr>
<td>PVDF</td>
<td>Ag-coated fiber</td>
<td>14 V/29.8 $\mu$A</td>
<td>5.10 $\mu$W cm$^{-2}$</td>
<td>97</td>
</tr>
<tr>
<td>PVDF-TrFE</td>
<td>Ag-coated nylon, CNT sheet</td>
<td>2.6 V/15 $\mu$A</td>
<td>1.53 $\mu$W cm$^{-2}$</td>
<td>98</td>
</tr>
<tr>
<td>PVDF</td>
<td>PEDOT</td>
<td>1 V/0.15 $\mu$A</td>
<td>Not mentioned</td>
<td>99</td>
</tr>
<tr>
<td>PVDF</td>
<td>Ni/Cu alloy</td>
<td>51 V/28.5 $\mu$A</td>
<td>10.5 $\mu$W cm$^{-2}$</td>
<td>100</td>
</tr>
<tr>
<td>PVDF</td>
<td>Ni/Cu alloy</td>
<td>42.5 V</td>
<td>125 $\mu$W cm$^{-2}$</td>
<td>101</td>
</tr>
<tr>
<td>BaTiO$_3$/PZT/CNT/PVDF</td>
<td>Carbon-filled polyethylene</td>
<td>6 V/4 $\mu$A</td>
<td>Not mentioned</td>
<td>102</td>
</tr>
<tr>
<td>PVDF-TrFE</td>
<td>Conductive thread</td>
<td>16.2 $\mu$V</td>
<td>Not mentioned</td>
<td>103</td>
</tr>
<tr>
<td>PVDF</td>
<td>Cu wire</td>
<td>2.3 V</td>
<td>1.05 $\mu$W cm$^{-2}$</td>
<td>104</td>
</tr>
<tr>
<td>PVDF</td>
<td>Ag coated nylon yarn</td>
<td>0.38 V/1.7 $\mu$A</td>
<td>14.81 $\mu$W cm$^{-2}$</td>
<td>105</td>
</tr>
<tr>
<td>PVDF/BaTiO$_3$</td>
<td>Graphene electrode of 3D micropatterned stretchable substrate</td>
<td>9.3 V, 189 $\mu$A</td>
<td>1.76 $\mu$W cm$^{-2}$</td>
<td>106</td>
</tr>
<tr>
<td>PVDF</td>
<td>PVDf-rGO</td>
<td>46 V, 18 $\mu$A</td>
<td>18.1 $\mu$W cm$^{-2}$</td>
<td>107</td>
</tr>
<tr>
<td>PVDF/BaTiO$_3$</td>
<td>Silver-coated nylon yarn</td>
<td>4 V</td>
<td>43.5 $\mu$W cm$^{-2}$</td>
<td>108</td>
</tr>
</tbody>
</table>
their ferroelectric hysteresis loop and remnant polarization ($P_r$). Since electrostriction is the origin of piezoelectricity, the electrostrictive strain ($\varepsilon$) hysteresis loop can be evaluated using the following equation: $\varepsilon = QP^2$ where the electrostriction coefficient can be evaluated using $d_{33} = 2Q_{r}P_{r}$. Conversely, using the $P_r$ value, the longitudinal piezoelectric strain constant, $d_{33}$, can be easily evaluated considering the macroscopic dimensional effect. In this case, $d_{33} = -\frac{P_r}{Y}$. Thus, $k^2 = \varepsilon + d_{33} = \varepsilon + \frac{d_{33}}{\varepsilon r_0}$, where $\varepsilon r_0$ is the dielectric permittivity, $\varepsilon_r$ is the measured dielectric constant and $\varepsilon_0$ is the vacuum permittivity ($\approx 8.854 \times 10^{-12}$). Here, $g_y$ is the voltage conversion coefficient, which is defined as $g_y = \frac{d_y}{\varepsilon r_0}$. Thus, the figure-of-merit (FoM$_p$) of a piezoelectric material is defined as FoM$_p = d_y g_y$ ($\text{Pa}^{-1}$).

### 4.2 Wearable healthcare monitoring pressure sensors

#### 4.2.1 Materials design.

To achieve skin-attachable and practical-to-use health-monitoring devices, power systems should be miniaturized, flexible, and sustainable. Accordingly, an ultrathin PZT film (thickness: $\approx 4.8 \mu m$) was developed using mechanical exfoliation followed by the inorganic-based laser lift-off technique. This film possessing excellent sensitivity ($\approx 0.018 \text{ kPa}^{-1}$) was used as a self-powered epidermal

![Fig. 9](image_url) (a) Applications of all-fiber nanogenerator towards healthcare monitoring such as vocal muscle vibrations during (a) speech, (b) coughing, and (c) swallowing and analysis of the responses of (d) speech signals and (e) coughing signals through STFT spectrograms and towards (f) wrist pulse detection (reproduced from ref. 91 with permission by the publisher).
piezoelectric sensor and nicely detected the radial/carotid artery pulse, respiratory activities, and trachea movements and utilized the acquired output voltage from human vital signals for the operation of LEDs and speaker modules (Fig. 7). This type of sensor is ideal for monitoring the health status of COVID-19 patient without any direct contact. However, since thin film-based inorganic sensors are limited by their stretchability, several inorganic nanowire-based piezoelectric materials such as ZnO, PZT, and BaTiO3 are also useful with textile-based electrode materials, leading to high performance energy harvesters and sensors. A list of the textile-based inorganic piezoelectric materials with their energy harvesting performance is presented in Table 2. Additionally, composite-based energy harvesters with inorganic materials as the main piezoelectric component and organic elastomers as a flexible and stretchable matrix have been enormously used in the energy harvesting research field. For example, a stretchable piezoelectric nanogenerator was realized by encapsulating ZnO nanowires in a parylene C polymer matrix. Similarly, several prototype flexible and stretchable piezoelectric energy harvesters were demonstrated using PDMS elastomer-based nanocomposites using various inorganic piezoelectric materials such as BaTiO3 nanoparticles and nanowires, core–shell structure (Ba,Ca)[Ti,Sn]O3/BaTiO3 (BCTS/BT) micro-crystals, PZT nanoparticles and nanowires, zinc stannate (ZnSnO3) nanocubes, KNbO3 nanorods, NaNbO3 nanowires, polybasic alkaline niobate, and (1−x)(Pb(Mg1/3Nb2/3)O3)−x(PbTiO3) (PMN–PT) nanowires. A super stretchable PMN–PT/MWCNT/PDMS composite-based piezoelectric energy harvester was demonstrated, which generated an outstanding energy harvesting performance (4 V and 500 nA) under strain as high as 200%. However, despite the successful application of inorganic/ceramic-based thin film and nanowire-constructed sensors...
towards healthcare monitoring, the implementation of regular life, lead (Pb)/semiconductor-based materials is still questionable and needs long-term observation before delivery to users. On the other hand, PVDF-based organic ferroelectric materials are promising candidates for self-powered integrated smart sensing systems because of their low modulus, intrinsic biocompatibility, flexibility, and conformability to almost any geometrical shape and size, and low-cost. Owing to their lower piezoelectricity, several strategies have been employed to improve the device performance. For example, the formation of microporous structures, composite structures, micro-/nanostructures, and nano-/micro fibers.\(^7^9\)

To enhance the piezoelectric property, the preparation of PVDF-based nanocomposites with the help of inorganic piezoelectric materials is an ideal design strategy. Accordingly, arollable energy harvester was proposed using a PVDF/potassium sodium niobate nanoparticle-based nanocomposite, which possessed a piezoelectric co-efficient of 53 pm V\(^{-1}\) and generated an output voltage of 18 V and current output of 2.6 \(\mu\text{A}\) under 50 N force.\(^8^0\) Similarly, several PVDF-based composite films were demonstrated, which showed very high energy harvesting performances, such as PVDF/K\(_{0.3}\)Na\(_{0.7}\)NbO\(_3\)–BaTiO\(_3\) (\(-160 \text{ V}\)),\(^8^1\) PVDF/Sm–PMN–PT nanowires with intercalation electrode (\(-320 \mu\text{A}\)),\(^8^2\) PVDF/PZT (\(-55 \text{ V}\)),\(^8^3\) PVDF/CH\(_3\)NH\(_3\)PbI\(_3\) (\(-1.8 \text{ V}, 37.5 \text{ nA under 2 kPa only}\)),\(^8^4\) PVDF/Ag@SiO\(_2\) (\(-53 \text{ V}\)),\(^8^5\) PVDF/3-D MOF (\(-143 \text{ pC N}^{-1}, 8.52 \text{ V kPa}^{-1}\)),\(^8^6\) and P(VDF–TrFE)/BaTiO\(_3\) (\(-45 \text{ V}\)).\(^8^7\)

In the direction of healthcare monitoring applications, the drug effect on the cardiovascular system was demonstrated using a flexible transducer fabricated from a ferroelectric polymer PVDF–TrFE–CFE-based thin film (Fig. 8a(i)).\(^8^8\) Through skin attachment of the transducer via a medical bandage, the wrist pulse and fingertip pulse wave were nicely recorded in real-time (Fig. 8a(ii)). Importantly, the effect of any drug dose to the cardiovascular system was also clearly identified through this type of pressure-sensing transducer. In this case, researchers administered glyceryl trinitrate (GTN) to male New Zealand white rabbits, 2.4–2.6 kg, to study the abnormality in their heart rate changes (Fig. 8a(iii)). GTN is an oral medicine to treat angina and heart failure, and consequently has been widely used in clinical practice and emergency medicine. Thus, the detection of the complicated changes in heartbeat signals and breathing rates indicates the capability of this type of sensor to evaluate the effect of cardiovascular drugs, which may play an important role in emergency medicine. Evidently, these types of pressure sensors are urgently needed in this pandemic situation of COVID-19. On the other hand, a highly sensitive (257 mV N\(^{-1}\)) and versatile platform for physiological signal monitoring was demonstrated by a nanoimprinting technology-based P(VDF–TrFE)/BaTiO\(_3\) composite micro-pillar array (Fig. 8b(i)).\(^8^9\) This sensor could detect and discriminate the breathing intensity such as normal breathing, deep breathing, laboured breathing and gasping (Fig. 8b(ii)). In a different approach, P(VDF–TrFE) nanowires were prepared using an anodized aluminium oxide (AAO) template (Fig. 8c(i)).\(9^0\) This nanowire-based device exhibited good sensitivity (458.2 mV N\(^{-1}\)) and could be used in human motion monitoring such as breathing responses and heartbeat pulse detection (Fig. 8c(ii)). However, for real-life applications, during the COVID-19 pandemic, light-weight and air-permeable textile-based sensors that can be used day and night are required. This functionality can only be fulfilled by electrospun piezoelectric nanofibers since they possess extraordinary sensitivity. The one-dimensionally (1D) confined nanofibers are \textit{in situ} poled and stretched due to their unique fabrication procedure. Generally, the dipoles of the PVDF chain are almost perpendicularly oriented to the fiber axis. Consequently, the highly aligned fibers show higher piezoelectricity than that of random fibers.\(^1^1\) Actually, during the fabrication process of aligned nanofibers, high stretching forces are exerted on the electrified solution jets
with additional mechanical stretching during collection of the nanofibers. Consequently, in situ poled-stretched nano-fibers with an ultra-high aspect ratio (i.e., several centimetres in length and micro/nano-size diameter) are produced. This dimensional reduction from film to a nano-fiber mat of the same thickness enhances the piezoelectric property and energy conversion efficiency of the nano-harvester owing to the space confinement effect. Additionally, the mixture of nano-fillers in PVDF nanofibers further enhances the dipolar orientations by in situ poling, stretching and interfacial interactions synergistically. 

4.2.2 Electrode design. The most challenging factor is the fabrication of electrodes for nano-fiber-based devices because nanofibers are highly air permeable. Conventionally, rigid metal plate electrodes are used to prepare the device because electrode deposition is not possible due to the high surface roughness of the nano-fiber mat, leading to the formation of a discontinuous film and the conducting paste can easily penetrate the nanofiber mat. However, the use of a metal electrode degrades the device performance over the time due to the large mismatch in mechanical properties (such as Young’s modulus and Poisson’s ratio) between the piezoelectric component and electrode materials. Accordingly, conducting textile-based electrodes are a suitable choice for device

Table 4 The relevant equations for pyroelectric and piezoelectric devices (l is the thickness, A is the area and t represents time)

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Pyroelectric</th>
<th>Piezoelectric</th>
</tr>
</thead>
<tbody>
<tr>
<td>Charge (Q)</td>
<td>$Q = \pi A \Delta T$</td>
<td>$Q = d_j A \Delta \sigma$</td>
</tr>
<tr>
<td>Short-circuit current ($I_{sc}$)</td>
<td>$I_{sc} = \pi A \frac{\Delta T}{\Delta t}$</td>
<td>$I_{sc} = d_j \frac{\Delta \sigma}{\Delta t}$</td>
</tr>
<tr>
<td>Open-circuit voltage ($V_{oc}$)</td>
<td>$V_{oc} = \frac{\pi A}{\varepsilon T}$</td>
<td>$V_{oc} = \frac{d_j}{\varepsilon T}$</td>
</tr>
<tr>
<td>Stored energy ($\frac{1}{2} Q V^2$)</td>
<td>$E = \frac{1}{2} \pi A l (\Delta T)^2$</td>
<td>$E = \frac{1}{2} \frac{d_j^2}{\varepsilon T} A l (\Delta \sigma)^2$</td>
</tr>
</tbody>
</table>

Fig. 12 Prawn shell (reproduced from ref. 114 with permission by the publisher)-based skin conformable device towards healthcare monitoring applications. (a) Digital photograph of the prawns and their shells (in the upper inset) with the molecular structure of chitin present in the shell (in the lower inset). (b) FE-SEM images of the demineralised prawn shell consisting of chitin nanofibers (in the left inset). (c) Prawn shell-based healthcare monitoring device. (d) Piezoelectric strain versus electric field hysteresis loop of the prawn shell. (e) Wearable prawn shell-based device attached to the throat (upper panel) and generated output voltage from the device upon repeated coughing (middle panel) with the short-term Fourier transform (STFT) of the output voltage for determining frequency of the coughing (lower panel). (f) (i) Photograph of the device attached to the human wrist (upper panel) for (ii) real-time monitoring of the heartbeat pulse consisting of (iii) typical pulse characteristics with (iv) STFT spectrums showing the range of frequency of the wrist pulse.
integration. A comparison is shown in Table 3, where it can be observed that metal electrodes are not a good choice as electrode materials, generating a lower electrical output in comparison to conducting fiber-based textile electrode materials, which are compatible with nanofiber-based organic piezoelectric materials. 

Fig. 13  (a) Stretchable graphene nanosheet electrode-based piezo-/pyro-electric hybrid device (reproduced from ref. 117 with permission by the publisher). (b) PVDF-based hybrid energy harvester (reproduced from ref. 118 with permission by the publisher).
Employing this design strategy, a wearable nano-tactile sensor was fabricated using Pt/PVDF-based highly aligned nanofibers as the piezoelectric component and an interlocked micro-fiber-based conducting textile based electrode.\(^{11}\) The device possessed a very high piezoelectric figure of merit \(\text{FoM}_p = 5 \times 10^{-11} \text{ Pa}^{-1}\) in comparison to pure PVDF nanofibers \(\text{FoM}_p = 9.7 \times 10^{-11} \text{ Pa}^{-1}\). Additionally, the sensor exhibited very high sensitivity \((\sim 600 \text{ mV N}^{-1})\), which enabled its use in healthcare monitoring. Of particular interest, this textile-based device served as an epidermal mechno-acoustic device, which could simultaneously capture the signals from articulator muscle groups and acoustic vibrations from the vocal cords (Fig. 9a). Thus, when the device was attached to the throat, it could sensitively monitor the muscle activity of the vocal cord during different physiological activities, such as speech (Fig. 9a), coughing (Fig. 9b), and swallowing (Fig. 9c). As is known, dry cough and tiredness are the most common symptoms of COVID-19, and thus these physiological changes can be accurately detected by this sensor, making it applicable for the early intervention of COVID-19. The output responses (in terms of voltage) generated from the sensor during different physical activities were further analysed by short-term Fourier transform (STFT)/direct Fourier transformation, which selectively indicated the vibrational frequencies of speech (Fig. 9d), and coughing (Fig. 9e). These types of accurate sensing functionality are essential for the real-time monitoring of health status related to COVID-19 suspected/confirmed patients. Also, this sensor accurately detected wrist pulse responses (Fig. 9f). Recent studies have shown that COVID-19 significantly affects the cardiovascular system, causing the heart to beat faster and harder to supply oxygen to major organs, which leads to heart failure and other cardiovascular diseases (CVD).\(^{109}\) Evidently, heartbeat monitoring simultaneously with the detection of body temperature fluctuations can possibly play a major role in the healthcare monitoring of suspected COVID-19 patients.

### 4.2.3 Biodegradable design.

In the current pandemic situation, disposable PPE, such as protective suits, facial masks, facial shields, gloves, gowns, and boot covers, are used in large quantities globally and mostly made of plastic materials. This will generate a large amount of waste materials. Therefore, healthcare monitoring sensors should be biodegradable, which will partially reduce the electronic waste.\(^{110}\) Accordingly, to fulfill this requirement, several wearable bio-inspired piezoelectric pressure sensors (i.e., bio-e-skin) have been designed.

A highly sensitive \((\sim 0.8 \text{ V kPa}^{-1})\) and flexible (Fig. 10a) structurally stable ferroelectric and piezoelectric \((d_{33} \sim 20 \text{ pC N}^{-1})\) fish gelatin nanofiber (GNF)-based sensor was demonstrated as a self-powered healthcare monitoring, ultra-lightweight \((ca. 0.6 \text{ g cm}^{-1})\) sensor, which successively monitored several biophysical signals.\(^{111}\) Notably, the read out pulse signals by this skin conformable sensor was as accurate as that measured by a bulky commercially available heartbeat measuring device (Fig. 10c). Using the gelatin nanofiber-based sensor, the radial artery pulse signals were measured under different conditions such as under rest, after exercise and after...
smoking. By analysing the artery pulse waveforms using parameters such as the time delay ($\Delta t_r$) between the detected leaks and augmentation index (AIr), the clinicians could easily identify whether the change was normal or abnormal. Therefore, this type of analysis is very fruitful and effective for early intervention in COVID-19-suspected patients since COVID-19 affects the cardiovascular system. Additionally, bio-e-skin was also demonstrated to be useful in robotic prosthesis (Fig. 10d).

Similarly, PLLA-based human skin interactive bio-e-skin exhibited excellent sensitivity (22 V N$^{-1}$) to harvest biomechanical motions and muscle movement of the oesophagus (the food pipe) and displayed distinct patterns, allowing the signals generated by the oesophagus during drinking and swallowing to be differentiated.\textsuperscript{112} Also, it could detect and discriminate the speech pattern of male and female voices. Furthermore, several bio-waste materials such as fish skin ($d_{33} \sim -3$ pC N$^{-1}$, sensitivity $\sim 27$ mV N$^{-1}$)\textsuperscript{113} (Fig. 11) and prawn shells ($d_{33} \sim -2$ pC N$^{-1}$)\textsuperscript{114} (Fig. 12) after the demineralization process served as piezoelectric materials and directly used as self-powered wearable healthcare monitoring devices, which were capable of sensing subtle pressure variations in radial artery and carotid artery blood pressure. Additionally, other biomaterials such as fish swim bladders ($d_{33} \sim 22$ pC N$^{-1}$)\textsuperscript{115} and fish scales ($d_{33} \sim -5$ pC N$^{-1}$, sensitivity $\sim 23.5$ $\mu$V Pa$^{-1}$)\textsuperscript{116} were also used as high performance biomechanical motion harvesters.

5. Piezo-/pyro-electric hybrid sensor

Since the pyroelectric effect also generates thermal stress within a material under thermal fluctuations ($\Delta T$) and originates from the polarization change ($\Delta P$), all pyroelectric materials are also piezoelectric. Therefore, it can be written as $\Delta P = d_{ij} \sigma + \pi \Delta T$. Thus, it is very obvious that researchers have made several efforts to combine the piezo and pyroelectric effects in one

![Fig. 15](image-url) (a) BaTiO$_3$-based pyro–piezoelectric hybrid energy harvesting and (b) real-time temperature and pressure variations mapping (reproduced from ref. 119 with permission by the publisher).
device with one material. Table 4 presents the relevant equations for pyroelectric and piezoelectric devices.

Simultaneous monitoring of physiological signals with body temperature fluctuations is the ideal solution for early intervention in the virus/influenza-infected human body. Thus to realize a skin conformable piezo-/pyro-electric hybrid wearable electronic device, a highly stretchable (~30%) micro-patterned P(VDF-TrFE), as the piezo- and pyro-electric material, and patterned PDMS-carbon nanotube (CNT) composite with graphene nanosheets as the electrode material was constructed (Fig. 13a).187

Furthermore, a transparent single-structure tribo-/piezo-/pyro-electric hybrid sensor was developed, which was capable of sensing pressure and temperature simultaneously even under stretching conditions (Fig. 13b). The device could be conformably attached on different parts of the body for real-time monitoring of various human vital signs including breath, heartbeat pulse, breathing and swallowing, and thus capable of being used for the early intervention of COVID-19.188

With one step further, a PVDF-based textile hybrid energy harvester was fabricated, where a PVDF nanofiber membrane was sandwiched between a conductive nanofiber membrane and CNT electrodes (Fig. 14a).55 The conductive nanofiber membrane was prepared by mixing PVP, PEDOT:PSS, ethyl alcohol, ethylene glycol and ionic liquid. The flexibility of the nanogenerator was enhanced by using an electrospray thermoplastic polyurethane (TPU) nanofiber membrane as a substrate. This device demonstrated a high performance as a piezo-/pyro-electric energy harvesting device that can concurrently harvest thermal and mechanical energies.

In another approach, methylammonium lead iodide (CH$_3$NH$_3$PbI$_3$) mixed electrospray PVDF composite nanofiber-based flexible and wearable devices were prepared, which were capable of harvesting mechanical and thermal energies (Fig. 14b(i)).14 This device generated an output power of 0.8 mW m$^{-2}$ under human touch (Fig. 14b(ii)) and 0.2 mW m$^{-2}$ power output from non-contact infrared radiation. Notably, this device could sense thermal fluctuations of different arbitrary frequencies (Fig. 14b(iii)), and thus more applicable in real-life healthcare monitoring. In addition to organic wearable devices, several inorganic material-based wearable hybrid devices were demonstrated to be useful for simultaneous healthcare monitoring.

An innovative approach was demonstrated to detect temperature and pressure simultaneously using a ferroelectric BaTiO$_3$-based pyro–piezoelectric sensor system (Fig. 15a).189 A flexible 4 x 4 array sensor system was developed to sense real-time temperature and pressure variations induced by the finger. It was further used for voltage mapping of instantaneous pyro- and piezoelectric signals (Fig. 15b), where each unit of the 4 x 4 array sensor was capable of simultaneously detecting piezo and pyro-electric signals.

Additionally, micropatterned (1-x)Pb(Mg,Nb)O$_3$-xPbTiO$_3$ (PMN-PT) ribbons (fabrication process is shown in Fig. 16a) were also prepared, which possess excellent piezoelectric and pyroelectric properties.188 These ribbons were further utilized to build human activity energy harvesting and monitoring systems. The sensor was conformably attached on the surface of human skin, exhibited high sensitivity for the detection of human body motions and could detect acoustic sounds precisely during coughing (most common symptom of COVID-19) (Fig. 16b). The sensor was used for monitoring temperature-related activities caused for instance by warm water flow and even light illumination (Fig. 16c).

Furthermore, a scalable ceramic-polymer composite based on three-dimensional (3-D) interconnected piezoelectric PZT micro-foam (Fig. 17a) was introduced for the concurrent harvesting of mechanical and thermal energies.121 This 3-D composite was scalable on a large scale and could be stretched and bent by the fingers with ease without fracture, and highly conformable to different parts of the human body, e.g., shoulders and knees, suggesting its applicability in wearable devices (Fig. 17b). Also, it was shown that its piezoelectric energy harvesting performance was not degraded under long term (2000 cycles) heating–cooling cycles (25–35 °C) and also its pyroelectric energy harvesting performance was not degraded under long term (2000 cycles) compressive strain (~8%) (Fig. 17b). The coupled output under simultaneous mechanical and thermal fluctuations was significantly enhanced in comparison to the individual piezo- and pyro-electric output (Fig. 17b). Overall, piezo-/pyro-electric hybrid energy harvesters represent efficient devices to simultaneously monitor human physiological signals including human body temperature fluctuations.
6. Conclusion and future outlook

In conclusion, we presented a range of flexible and wearable pressure and temperature dual functional sensors, which can work simultaneously. These sensors are ideal candidates for application as healthcare monitoring sensors for early intervention of COVID-19. Since many companies are working on applying wearable sensors for the remote monitoring of COVID-19 patients, the use of these sensors may work well. Importantly, with one material having multi-sensing capability, several types of health status can be monitored. Also, these sensors are attachable to PPE, which clinicians and other health

Fig. 17 (a) PZT-based 3-D micro-foam with large-scale processability and conformability to different parts of the human body and (b) mechanical and thermal hybrid energy harvesting feasibility (reproduced from ref. 121 with permission by the publisher).

Fig. 18 Image showing envisioned smart personal protection equipment (PPE), which is typically used by clinicians in the treatment of COVID-19. The PPE consists of several wearable sensors/devices with wireless healthcare monitoring functionality, which is urgently needed in this current situation (picture courtesy ref. 122).
staff are required to wear. Thus, in hospitals and other COVID-19 prone areas, healthcare monitoring is possible day and night without any interruption. In this regard, the development of smart PPE consisting of wireless communication and integrated temperature-pressure hybrid sensors is urgently needed in order to monitor the physiological conditions of patients without compromising protective functions, as proposed in Fig. 18. In this direction, more research is needed to explore wearable piezo-/pyro-electric hybrid sensors and their applicability towards healthcare monitoring functions. Besides organic and inorganic materials, piezoelectric 2D materials are also ideal candidates due to their outstanding advantages in terms of ultrathin, transparency, flexibility, large surface-to-volume ratio, and stackable layers, which are beneficial for the development of lightweight and high-performance multifunctional applications. Additionally, triboelectric nanogenerators are also validated to be fruitful for the filtration and deactivation of SARS-CoV-2. Furthermore, flexible and rollable magnetoelectric materials also play a significant role in wearable and wireless sensors. Therefore, researchers from different sectors, such as, materials scientists, physicists, chemists, engineers, and other related research fields need to be involved more deeply. Since the COVID-19 pandemic is a global challenge, researchers around the globe need to work together to address the many challenges in order to implement wearable healthcare motoring sensors, which can prevent the further spread of COVID-19 and possibly end this crisis in near future.

Conflicts of interest
The authors declare no competing financial interest.

Acknowledgements
This work was financially supported by a grant from the Science and Engineering Research Board (EEQ/2018/001130), Government of India.

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