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Review article

Triboelectric Nanogenerator-based smart biomedical sensors for healthcare

Shipra Solanki [a,b,1], Akhilesh Kumar Gupta [c,1], Udiptya Saha [a], Alexey V. Krasnoslobodtsev [c], Rajinder K. Gupta [b], Bansi D. Malhotra [a,d,*]

[a] Department of Biotechnology, Delhi Technological University, Shahbad Daulatpur, Delhi 110042, India
[b] Department of Applied Chemistry, Delhi Technological University, Shahbad Daulatpur, Delhi 110042, India
[c] Department of Physics, University of Nebraska at Omaha, Omaha, NE 68182, USA
[d] Biomedical Instrumentation Section, CSIR-National Physical Laboratory, Dr. KS Krishnan Road, New Delhi 110012, India

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ABSTRACT

Self-powered nanogenerators (NGs) use ambient mechanical energy to power devices, and NGs can replace batteries. The NGs converting mechanical energy into electrical energy utilizing the phenomenon of tribo-electrification and electrostatic induction are called Triboelectric Nanogenerators (TENGs). TENGs-based self-powered sensors have gained popularity as biomedical sensors due to their properties. The activities of elderly people with chronic diseases or younger ones doing exercise need continuous health monitoring. Remote healthcare monitoring also requires trustworthy and affordable solutions. For these reasons, self-powered biomedical devices are being developed to address the needs of patients, specifically in the elderly category.

This study reviews recent TENG-powered smart and wearable biomedical sensors. We use respiratory, tactile, cardiac, and sweat-based TENG sensors to monitor younger and older patients remotely and at home. Moreover, we also discuss different TENGs and *in vivo* sensors reported in recent years. Further, we discuss the drawbacks and limitations of the existing designs of TENGs, including poor performance, complex fabrication strategies with a large number of electronic elements, problems with miniaturization, etc. We strongly believe that developing smart and wearable self-powered devices capable of reading physiological data will determine the health monitoring at home for older people in the future.

Introduction

Medical diagnostics has recently evolved far beyond the assessment of external symptoms. Medical doctors generally depend on the results of different bio/chemical tests to diagnose and treat various diseases. The clinical diagnosis of a patient's condition may require continuous monitoring outdoors or even while resting at home [1,2]. The traditional hospital-centered diagnostic process and monitoring is shifting towards patient-centered monitoring, and extensive efforts are being made for early detection and well-being instead of diagnosis and treatment. The power requirement for the continuous working of such monitoring devices varies from a few microwatts to a few tens of milliwatts, and a reliable supply of energy to these devices in a regular manner is of prime importance. The use of batteries is the most viable option for such applications [3]. However, the use of batteries has not been found reliable, owing to their size and longevity limitations. Thus, extensive research is in progress to develop micro and nano technologies offering new possibilities for less power consumption devices. In addition, low-cost biomedical devices capable of independent, sustainable, and maintenance-free operation can be built using novel nanomaterials. The idea of automated self-powered devices utilizing environmental energy has revolutionized the field of biomedical monitoring [4–7].

The nanogenerators have been considered as potential alternatives to batteries providing stable, safe, and low-cost power supply for the functioning of various devices [8,9]. Many kinetic, mechanical, chemical, thermal, and electrical changes occur in a human body via different processes that can be used for energy harvesting [10–13]. The *in vivo* biological and mechanical energy produced by body movements such as breathing, heart beating, blood flow, muscle stretching, lung vibration, metabolic reactions, etc., can be utilized in different implantable biomedical devices with the help of many piezoelectric (PENGs) and triboelectric nanogenerators (TENGs) [14–18]. First introduced by Wang's group in 2006, nanogenerators have turned out to be the most promising devices for biomechanical energy harvesting and have

* Corresponding author.
E-mail address: bansi.malhotra@gmail.com (B.D. Malhotra).
[1] These authors contributed equally.

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enabled many systems to work without any external power supply [14,19]. Since its inception, the design of nanogenerators has evolved with materials varying from original zinc oxide nanowires to composites and polymers like polyvinylidene fluoride (PVDF) and even metals [20–23]. These recent improvements have increased power output, efficiency, utility, stability, and safety in PENGs and TENGs, making them applicable in diverse fields [24,25]. However, the major problem with PENGs is that their output is limited and insufficient to directly drive a medical device [26].

Certain emergency situations, such as an unexpected fall of an older person at home alone, might delay or even prevent seeking aid in time [27]. Falling can cause lacerations, fractures, and hematomas. This has stimulated the development of a novel fall detection system based on TENGs pressure sensing. The generated signals can be delivered to a platform such as a smartphone, to classify situations and data for validation [27]. Li et al. have demonstrated that TENG-based fall detection and sleep monitoring devices are possible [28]. The proposed devices can help build smart homes and hospitals to avoid fall-related injuries and quickly inform medical assistants if such emergency situations occur.

Current efforts in the field are geared towards the search for an efficient TENG for biomedical applications, including self-powered sensors, pacemakers, stimulators, etc. This review mainly focuses on the recent progress in smart biomedical sensors based on TENG. First, the basic fundamentals of TENG are discussed, followed by detailed specific results of some key applications in this field, including different materials used to fabricate wearable and implantable TENG-based sensors. The major limitations and prospects are also discussed along with the recent technology commercialization report. Finally, we conclude that TENG-based technology would be ideal for smart homes and smart hospitals designed specifically for older people, as shown in Fig. 1A.

Basic working principle of TENG

The triboelectric effect is a type of contact electrification wherein certain materials become electrically charged when separated from a different material with which they were in contact [29,30]. The different mechanisms of triboelectrification have been debated for many years, with contesting claims of its origin in electron transfer, ion transfer, or transfer of material species [31]. The TENG devices essentially contain one dielectric material that can develop a charge during contact electrification and a conducting material for the flow of current. Broadly the TENG devices can be fabricated on four operational models: vertical contact separation mode [30,32,33], lateral sliding mode [34,35], single electrode mode [36], and free-standing triboelectric layer mode [37]. These four operational modes are schematically shown in Fig. 1B. The detailed description of these different working modes can be found in literature (as referenced above). The major advantages and disadvantages of different operational modes of TENGs used in self-powered biosensing devices are listed in the Table provided in the

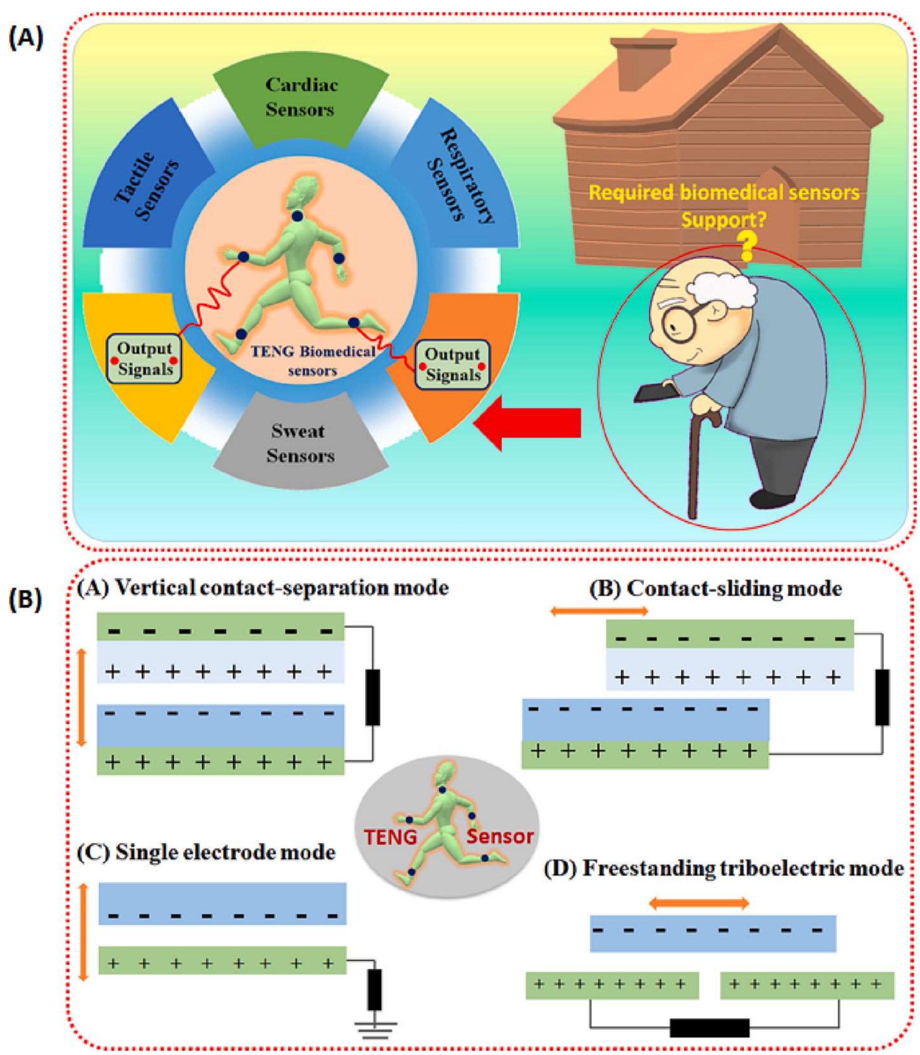

Fig. 1. (A) Perspective of TENG-based smart biomedical sensors toward the next-generation era of health monitoring (B) Different working modes of TENG devices.

supplementary file. Table 1 summarizes a few TENG based biosensing devices with their characteristics and respective operational modes.

Materials used in contact surfaces of TENG

The output of a TENG depends linearly on the density of the triboelectric charges governed by the materials used in fabricating the device. Several materials like polymers, metals, glass, paper, fabrics, and inorganic composites can be used for designing different TENG devices. Polymers like polydimethylsiloxane (PDMS), polyethylene terephthalate (PET), polytetrafluoroethylene (PTFE), and polyvinyl alcohol (PVA) have been extensively used due to flexibility, elasticity, and non-toxic nature.

Aside from normal TENGs, the materials used for wearable devices should have properties like high flexibility, which allows them to be integrated into non-planar and curvilinear body systems, deformability to be attached to a human body, and self-healing capacity so that device performance can be restored after minor damage. Highly soft, breathable, comfortable, and sustainable wearable TENGs have been designed that have shown remarkable superiority in mechanical energy harvesting and self-powered sensing due to ease of fabrication and promising output performance [38,39]. Smart textile fabrics having lightweight and high energy-converting efficiency have been employed in wearable TENGs [40–43]. Bioresorbable materials, like silk and genetically modified spider silk, can be used in self-powered devices [44–46]. Wearable TENGs can have a fabric-based structure in which the whole fabric, like nylon, wool, or yarn, is used as a triboelectric layer or a fiber-based system with specially designed single fibers or a collection of fibers functioning as TENGs [47,48]. The most appropriate way to construct a wearable TENG is to use existing fabric with a proper design. The modification can be done to the entire fabric or a part of it [49–51]. Single thread-based TENGs have also been designed to fit one triboelectric layer into the other in coaxial tubular structures [52].

Implantable devices need biocompatible materials, robust for prolonged usage, completely non-toxic, and attachable to the tissues. Compared to synthetic polymers, natural bioresorbable polymers like cellulose, chitin, silk fibroin, rice paper, egg white, etc., have been found more suitable as they are easily processable, biodegradable and have the high film-forming ability [53]. Recently, there has been a lot of focus on research using nanomaterials like metal oxides, graphene, carbon nanotubes, metal dichalcogenides, and MXenes as TENG contact surfaces. Some nanomaterials can serve as negative tribo layers and conduct or insulating polymers; for example, polymers like PTFE are electrically insulating. A two-dimensional titanium carbide MXenes-based TENG was recently demonstrated to harvest energy from muscle movements with peak power output up to $\sim$ 0.5 – 0.65 mW [54].

It has been found that almost all materials exhibit triboelectricity. However, choosing the right pair of materials that can provide maximum output is presently a difficult task. Two types of approaches are followed to increase the surface charge density for a particular type of substrate to improve performance. One is to select the materials for the two contact surfaces in TENGs that have the maximum possible separation in the triboelectric series. The other is to increase the surface area by nanostructuring. Different nanostructuring techniques like

Table 1
Characteristics of different TENGs-based biomedical sensors.

S. No.	Type	Materials used	Size	Mode of operation	Function	Ref.
1.	Cardiac pacemaker	PTFE-Kapton	–	Contact-separation mode	Regulation of cardiac physiological activity	[74]
2.	Sleep sensor	ALPF-Cu	1.8 cm^2	Contact-separation mode	Monitoring of body movements during sleep	[104]
3.	Respiratory sensor	PTFE	$5.5 \times 2 \times 1$ cm^3	Contact-separation mode	Real time display of breathing patterns through air flow detection	[83]
4.	Sleep sensor	PDMS-Al	300×300 mm^2	Contact-separation mode	Breath and heartbeat monitoring	[117]
5.	Motion sensor	ABS-Cu	–	Contact-separation mode	Monitor motion of arm	[118,119]
6.	Caridovascular sensor	FEP-PET	1.5×3 cm^2	Contact-separation mode	Blood pressure monitoring	[78]
7.	Coronary heart disease sensor	Kapton-Cu	20×10 mm^2	Contact-separation mode	Pulse monitoring system	[75]
8.	Respiratory sensor	FEP-PDMS	33×33 mm^2	Contact-separation mode	Pulse and breath measurements	[120]
9.	Tactile sensor	Ionogel-PDMS	2×1.5 cm^2	Contact-separation mode	Monitoring touch and bending of fingers	[121]
10.	Touch sensor	P(VDF-TrFE) - PDMS	–	Contact-separation mode	Real-time monitoring of bodily motion	[122]
11.	Respiration sensor	Nylon-PTFE	5×5 cm^2	Contact-sliding mode	Monitoring of change in abdominal circumference	[123]
12.	Pulse sensor	PET	40×20 mm^2	Single-electrode mode	Pulse monitoring with wearable device	[124]
13.	Bladder sensor	PET-PDMS	1 cm	Contact-separation mode	To monitor the fullness of bladder	[125]
14.	Respiratory sensor	PTFE-Al	$3 \times 2 \times 0.1$ cm^3	Contact-separation mode	Accurate, continuous, and real-time monitoring of multiple physiological and pathological signs	[126]
15.	Eye motion sensor	Ecoflex and PEDOT:PSS	1.0–1.5 cm	Non-contact electrostatic induction	Eye motion sensor	[107]
16.	Drowning sensor	PA11 and PVDF	–	Contact-separation mode	Drowning sensor	[127]
17.	Respiratory sensor	PAN and PVDF	Mask	Contact-separation mode	Respiratory sensor	[128]
18.	Wear debris sensor in artificial joints	PE and iron	–	Contact-sliding mode	Wear debris sensor in artificial joints	[101]
19.	Tactile sensor	PDMS	63 mm $\times$ 47 mm $\times$ 350 μm	Contact-separation mode	Tactile sensor	[129]
20.	Sweat sensor	PTFE and copper	–		Sweat sensor	[96]

PTFE = polytetrafluoroethylene, ALPF = aluminum-plastic laminated film, PDMS = polydimethylsiloxane, ABS = acrylonitrile butadiene styrene, FEP = fluorinated ethylene propylene, PET = Polyethylene Terephthalate, PVDF = polyvinylidene fluoride, PA11 = polyamide 11, PAN = Polyacrylonitrile, PE = Polyethylene.

lithography, electrospinning, and surface texturing via plasma treatment can be utilized to increase the contact area of two tribo-layers. Although a triboelectric series arranges the materials according to their triboelectric nature, this approach lacks a standard of calibration. Thus, there is an urgent need to study these materials, their surface structure, patterning, functionalization, etc., for the optimal design of TENG-based biomedical sensors.

TENGs-based biomedical sensors

TENGs have been investigated for several biomedical applications, which fall into three basic categories: monitoring the desired biological activity, measuring a biological quantity, or eliciting some biological response [6,55]. The energy otherwise lost through walking, breathing, heartbeat, etc., can be harvested effectively by TENGs, giving rise to implantable and wearable electronic sensors that do not require an external supply of energy, such as batteries. This may not only obviate the need for surgeries to replace drained batteries, which can be expensive and painful, but it can also prevent potential toxicity associated with battery leakage. There are three approaches to achieving these applications: wholly implantable devices, non-implantable wearable devices, and a combination of the two (i.e., a wearable device that powers a separate implanted device). The wearable TENGs can be worn directly on the body, as an independent unit, or indirectly integrated into clothing, shoes, or other accessories. This section discusses different types of wearable and implantable self-powered sensors reported in the literature describing their basic structure and operation.

This kind of setup has been a favorite choice for continuous monitoring systems independent of external energy needs. For instance, these have been effectively demonstrated in the fabrication of large-scale textile-based TENG arrays capable of generating constant cues regarding sleep monitoring using a pressure-sensitive setup [56]. Sensors monitoring biomedical motions have also gathered much attention leading to their rapid development [57,58]. Flexibility and portability are essential characteristics for the successful functioning of wearable devices. Some of the exciting designs of wearable TENG systems have

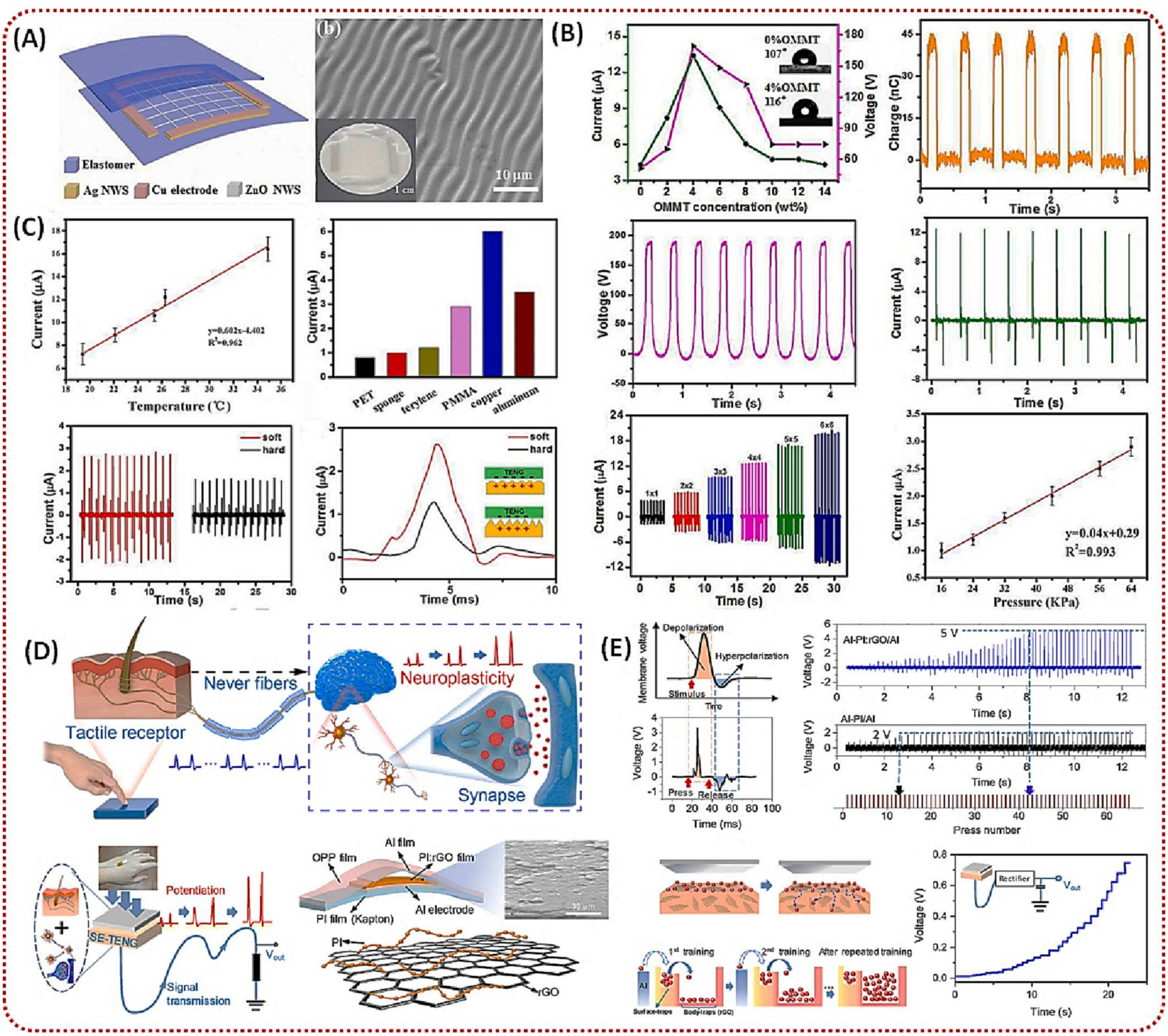

Fig. 2. Tactile Sensors (A) Schematic diagram of the PO-STENG and morphology of elastomer (B) Output of TENG (C) Short circuit current of the PO-STENG contacted with soft and hard materials, the shape of the current peak of the PO-STENG contacted with soft and hard materials [67] (D) An artificial tactile sensor in comparison with a tactile receptor and the related afferent nerve system (E) Electrical output performances and operation mechanism [68].

been demonstrated in various areas, including biomedical monitoring [59,60], sleep monitoring [61], sound recording [33,62], and cuffless blood pressure measurement [63].

Tactile and motion sensors

Several electronic devices mimicking human skin with their application in artificial intelligence, robotic skin, and biomedical sensors were reported. These devices can be used to detect pressure, strain, roughness or temperature, or many stimuli simultaneously to function as real skin employed in automatic control, surveillance, remote operation, and security systems [64,65]. These sensors, also known as e-skin, are flexible, stretchable, and have the capacity for self-healing [66]. The human body is considered a conductor encapsulated in a dielectric layer. Human skin can function as an excellent positive triboelectric layer and can be used to power small electronic devices by exploiting energy harvesting using TENGs. TENGs sense motion or touch when the triboelectric layer comes in contact with a human body. They have high output efficiency and are the most explored types of TENGs as they have the potential to be used in touch pad technology for a flexible tactile sensor.

Total of nine movements of body muscles and joints can be effectively used for TENG development [69]. Most of these devices have the single-electrode (SE) based conformation, and their performance depends on the effective contact area of the device. Such devices are also called human body TENGs or H-TENGs, similar to a system where the body is connected to the ground through various loads or resistances [69]. An ultra-stretchable TENG was proposed using an organic compound and polydimethylsiloxane (PDMS) as triboelectric layers for monitoring temperature and pressure, as shown in Fig. 2A [67]. The device worked on SE mode of operation. It consisted of PDMS top layer with silver nanowires and zinc oxide nanowires grid along with copper tape, making a hybrid conducting electrode. The output of the device intensified 7.2 μA to 16.4 μA when the temperature increased from 19.4 °C to 34.9 °C, which is the average environmental temperature of daily human life, as shown in Fig. 2B and 2C. Another sensor that works on the motion of eyelids has been demonstrated to distinguish between voluntary and involuntary eye blinks by measuring *Orbicularis Oculi* muscle motion based on non-contact electrostatic induction [70]. A single-electrode self-powered intelligent tactile sensor with learning and memory performances can be constructed [68]. The key component is a well-designed negative triboelectric layer: the polyimide/reduced graphene oxide (PI/rGO) hybrid layer. The reduced graphene oxide (rGO) is embedded in the friction layer to act as electron-body traps to obtain memory performances. In a practical application, the afferent nerve systems are employed to sense external stimuli and to initiate action potentials. Contact electrification produces a triboelectric potential, which activates the synaptic system and transmits mechanical information to the end terminals, as shown in Fig. 2D. When the 1st press is applied to the device, TENG changes the mechanical pressure to electrical signals, and output increases to 0.004 V. After 42 continuous presses, the output voltage reaches a saturated state of 5 V. As the number of presses increases, the output voltage increases because electrostatic electrons migrate from the surface to the body. However, after enough training, the triboelectric electrons are captured in the PI: rGO layer reach the maximum, and the output voltage becomes saturated. Also, the output signals of the device are connected to a capacitor using a rectifier bridge, and the tactile sensor is used to charge capacitors (0.22 μF), as illustrated in Fig. 2E.

Nanostructured TENG-based devices were fabricated with a high open-circuit voltage of 128 V upon contact with the human skin. This value was significantly higher than that of a TENG without a nanostructure (51.6 V), which was attributed to the larger effective contact area of the nanostructured design. Furthermore, upon finger touch stimulation, a remarkable 266 $\mu W/cm^2$ power density was achieved with the nanostructured TENG. These results suggested that the skin-stimulated elastomer-based TENG could open new possibilities in the development of power sources for wearable sensors in the future [71]. Also, a hybrid self-powered sensor with a higher power density and sensibility was constructed by integrating a TENG and a PENG. The fabricated device and its power generation performance obtained 200 V with an open-circuit voltage, a short-circuit current ISC of 8 μA, and a power density of 0.35 mW cm^{-2}. The sensor was then successfully integrated into a glove to collect the electrical signal output generated by the gesture. The deep learning algorithms to recognize and control gestures in real-time were utilized in combination with tactile sensors. Such a combination could lead to applications in artificial intelligence, such as human–computer interaction, signal monitoring, and smart sensing [72].

The selection of materials is an important criterion that decides the performance of such devices. Other factors like the structure and integrity of the contact layers are also considered to be essential factors that influence the sensor's activity of the tactile sensors. Despite the immense achievements in the field of attachable TENGs, a significant issue needs to be considered. Reserachers generally ignore the comfort of the human body that needs to be taken care of while designing such wearable devices. Materials should be selected to avoid skin irritation and have a breathable structure. At the same time, the devices should integrate with the human body easily. The studies on the e-skin where this issue has been dealt with efficiently should help resolve this issue.

Cardiac sensors

Many TENGs have been fabricated to give complementary power to prolong the battery life of implantable devices and provide independent power supplies. TENGs have been used as cardiac sensors as well as implantable pacemakers [19,73]. The pacemakers can also derive energy from periodic cardiac movements or other body movements. One example is an implantable TENG (iTENG) designed by Ouyang et al., which utilizes the energy of the periodic cardiac motion and converts it to electrical energy stored in a capacitor to power a cardiac pacemaker, preventing its deterioration and increasing its operational life [74]. The functioning of the iTENGs is based on the contact-separation mode, consisting of PTFE as one of the triboelectric layers and sponge as a spacer. The assembly is packed in a flexible Teflon and PDMS layer, as shown in Fig. 3A. The *in vivo* performance was investigated by implanting the device in a porcine, and the energy harvested from each cardiac cycle was found to be 0.495 μJ. The same group fabricated a conventional self-powered pulse sensor and correlated its functioning with the analysis of diseased conditions like coronary heart disease, atrial septal defect, and atrial fibrillation [75]. A complete system was built to show the device's practical application in mobile diagnosis, which showed that it is possible to use the pulse waveform analysis of two TENGs simultaneously to differentiate the degree of arteriosclerosis. Ryu et al. demonstrated that the enclosed five-stacked i-TENGs could convert mechanical energy into electricity at 4.9 $\mu W/cm^3$ [76].

The same study also demonstrated a pre-clinical test device that successfully harvested energy using real-time output voltage and data monitored via Bluetooth. The device could charge a lithium-ion battery. Such a self-rechargeable cardiac pacemaker integrated with TENG power management system can certainly help with the continuous charging of a cardiac pacemaker, as shown in Fig. 3B. Furthermore, effective integration of a cardiac pacemaker with the i-TENG confirmed the self-rechargeable cardiac pacemaker system's ventricular pacing and sensing operation mode. This proof-of-concept technology could pave the way for new self-recharging implantable medical devices [76].

Zheng et al. reported a cardiac pacemaker utilizing energy from breathing and powering the pacemaker through the energy stored in a capacitor [26]. Lin et al. fabricated a low-cost, non-invasive, and self-powered wireless heart rate monitoring system driven by energy from the human walk [77]. It is integrated with a pulse sensor, a storage unit, and a TENG working on free-standing triboelectric layer mode. It

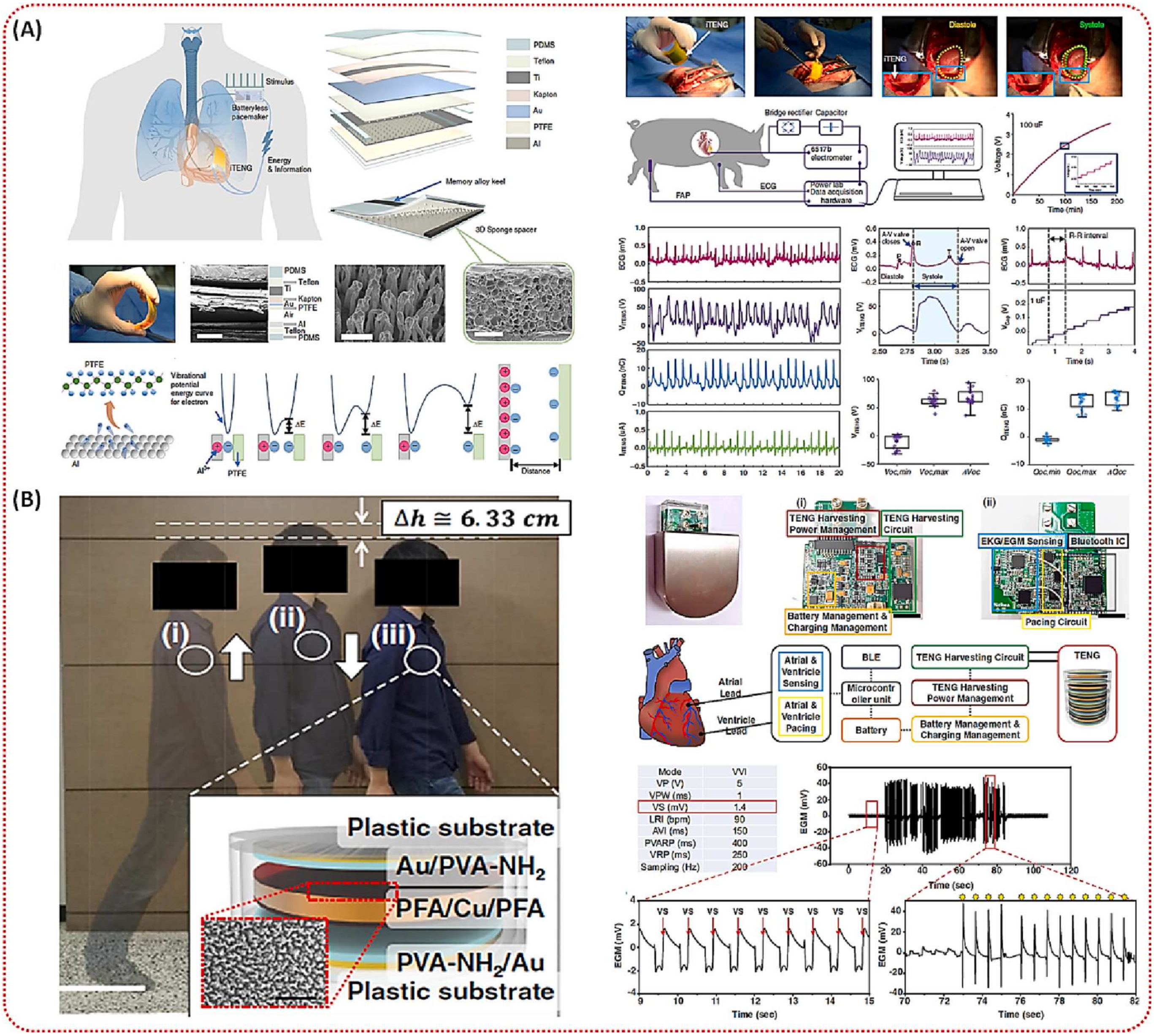

Fig. 3. (A) Illustration of symbiotic cardiac pacemaker system and their working mechanism; In vivo energy harvest and electrical characterization [74], (B) Overview of the *in vivo* triboelectric nanogenerator driven by inertia; Self-rechargeable cardiac pacemaker system [76].

consisted of copper-coated PTFE and thin copper film as free-standing triboelectric layers with an acrylic backbone. The working concept was based on the relative sliding of PTFE and copper layers during walking, and the energy stored in a capacitor was used to run the attached commercial sensor. A power of 2.28 mW was delivered by using this assembly. The authors called it a downy TENG, and the system was connected to Bluetooth for data transmission, which was finally displayed on a screen of a smartphone. The work presented a cost-effective route for daily healthcare monitoring. Chen et al. proposed a pressure sensor with a hierarchical structure suitable for monitoring pulse and blood pressure for diagnosing artery and heart conditions [78]. In this study, nanostructured PET and PTFE were used as triboelectric layers in the TENG, which was used to distinguish between wide ranges of pressure variations.

Though TENGs-based pulse or cardiac sensors deriving energy directly or indirectly from cardiac motion have emerged as a promising technology, some challenges still remain, including (1) minimal size; the size has to be very small to get integrated into the body for implantable pacemakers; (2) high energy density; the output current achieved till now has been relatively low; therefore some structural changes and innovative designs need to be introduced to fabricate commercially applicable devices; (3) efficient fixation with bio-tissue; new hybrid materials must be developed to attach devices to the body to avoid irritation and hurdle free operation; (4) long-term biocompatibility to safeguard the body as well as device preventing bio-fouling.

Respiratory sensors

The monitoring of human respiration provides a lot of information about a person's physiological and pathological conditions, which are useful for health assessment and the illness prognosis [17,79,80]. Periodic movement of the lungs can be utilized to fabricate TENGs functioning as efficient indicators of physiological health and help diagnose respiratory diseases. TENGs-based respiratory sensors, mostly harvesting mechanical energy due to the movement of lungs, have been developed for battery-free wearable electronics for intense healthcare monitoring in households and outside hospitals in the form of wearable textiles. The wearable textile sensors mostly utilize the mechanical energy of breathing for their operation and function as strain sensors [17,81].

One such sensor has been reported by Ning et al. using PTFE fibers and silver-coated nylon fibers. Both PTFE and Ag-coated nylon fibers were alternately wound around a stretchable substrate to form the helical fiber strain sensor (HFSS), as shown in Fig. 4A. The stretching and releasing of the substrate allowed contact electrification between the two layers and generation of a potential difference. The fabricated strain sensor was stitched into a chest strap and fastened to the chest to monitor the breathing and heartbeat signals (Fig. 4B). Both inhalation and exhalation processes generated continuous electric signals in HFSS [82]. The relationship between the volume of exhaled air over time and open circuit voltage was found to be directly proportional which allowed to develop the self-powered smart spirometer to quantify the airflow volume per expiration. Similarly, a screen-printed membrane-based TENG was fabricated using silver nanowires and PVDF to monitor breath.

The device was integrated into the mask, with open circuit voltage in the SE mode of TENG, where two layers came into contact due to the inward or and outward flow of air, facilitating the development of

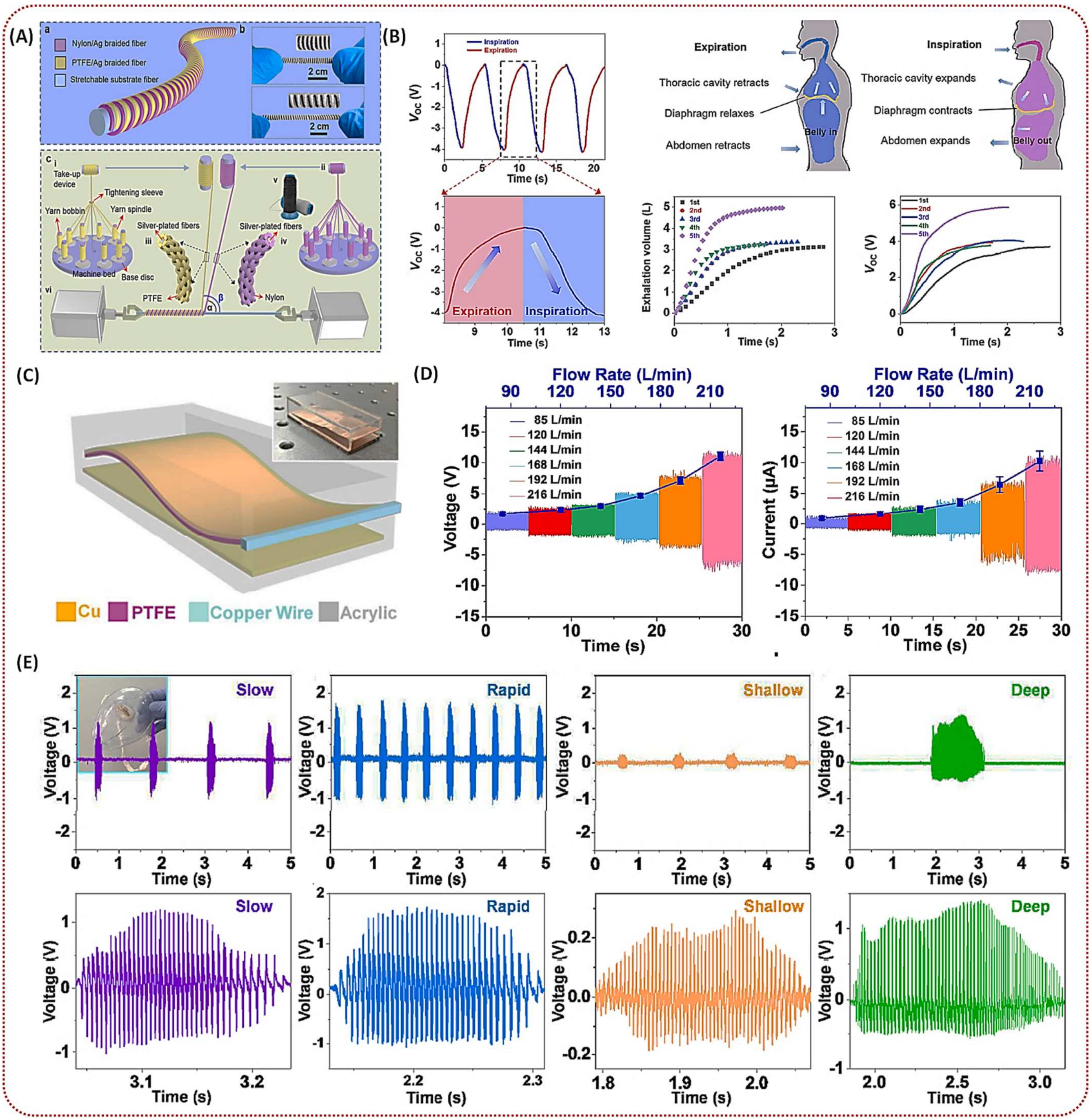

Fig. 4. (A) Fabrication process of the helical fiber strain sensor (HFSS) (B) expansion/contraction of the chest and abdomen as the human body breathes [82] (C) air flow-driven TENG (D) The output voltage and current signals recorded at different airflow speeds (E) The real-time respiratory signals of the TENG, recorded from four different human breathing behaviors [83].

charges at different tribo-polarities. It was demonstrated that such a design could measure respiration rate and analyze finger movements. The membrane had excellent properties like high skin-friendliness, outstanding electric output performance, and large-scale fabrication feasibility.

Airflow-driven self-powered sensors have also been introduced, capable of harvesting energy from vibration [83]. The device consisted of a nanostructured *n*-PTFE film with copper film as a charge collector, as shown in Fig. 4C, exhibiting oscillatory behavior due to the variation of airflow. The device had a maximum output power of 1.3 mW at a load resistance of 15.1 MΩ. The output voltage variation with air flow rate showed alternating values at a given air flow rate. The performance of this device was good under highly humid conditions, suggesting that it could effectively operate under conditions of normal human respiration. The relationship between different breathing patterns and the electric signals produced by *n*-PTFE TENG was established by studying the accumulative transferred charge and volume of air transported during the exhalation process as a function of time, as shown in Fig. 4D and 4E). Both exhibited a very good match along the entire time domain showing that the volume of air exchanged during respiration can be measured by finding the total amount of transferred charge detected by the TENG. Upon installation in a regular mask, the device could provide a real-time display of different breathing patterns.

Respiratory TENGs could be used for the sensitive detection of gases like ammonia, acetone, and nitrogen dioxide [84,85]. The devices are preferably coated with a selective nanomaterial (Ce-doped ZnO-PANI) nanocomposite film in case of NH_3 sensor and NaOH-treated WO_3 layer in the case of NO_2 sensor) that acts as a sensing layer. Further, the output current is varied by the selective absorption or adsorption of the target gas molecules at the surface of the nanomaterials [83]. The wireless energy transmission systems have the added advantage of providing stable gas measuring conditions as the mechanical stimulus does not interfere with the sensing materials [79]. The most extensively studied TENGs-based sensors are respiratory sensors, and some of these have already been commercialized. In spite of these interesting developments, it is a long way to go in introducing devices for mass application due to several challenges. One of the biggest challenges is energy utilization and conversion efficiency, the available respiration energy of an average adult is 0.1 to 1 W, and energy conversion efficiency achieved till date is less than 1 %. The most likely reason is the loss of mechanical energy due to the design of the desired device. TENGs with mechanical properties comparable to tissues may have maximum mechanical displacement with respiration, resulting in the reduced mechanical loss.

Sweat sensors

Sweat is a bio-fluid that contains many chemicals like sodium, potassium, chlorine, calcium, glucose, uric acid, cortisol, and many biomarkers [86–88]. It has even been used to analyze different drugs, alcohol, and heavy metals [89–91]. Hence, it can be used for monitoring human health conditions in a non-invasive manner to optimize the training of athletes, physical exercise of rehabilitation patients, and the elderly. *In-situ* sweat sensors have immense potential to provide continuous health monitoring at molecular level. Wearable sweat sensors are a growing field of research as they can be used to collect large quantities of data for personalized health care. But they still face challenges like the supply and management of energy in the device, the data collection mechanism, analysis of the obtained results, etc. Thus, new sweat collection self-powered devices with integrated artificial intelligence have been designed and developed. Such devices have opened possibilities to facilitate personalized healthcare monitoring and aid in disease analysis.

The current trend towards the development of self-powered sweat sensors, is to use TENGs-based self-functional textiles for scavenging energy and a collection of samples. An output of 1.71 mW was collected from a poly (3,4-ethylene dioxythiophene) polystyrene sulfonate (PEDOT: PSS)-coated fabric with mild jumping at 2 Hz and load resistance of 59.7 MΩ using hybrid TENG and PENG mechanism [94]. A patch-based self-powered wearable sweat analysis system exploiting human motion for energy harvesting and analyzing Na^+ and K^+ in the sweat through wireless monitoring was reported by Gai et al. [95]. The energy was harvested using a hybrid nanogenerator module with a spring-coupled mechanical design consisting of PTFE and nylon layers as tribo-pairs. The biomarker (Na^+ and K^+) sensing was achieved using a 3D-printed polylactic acid (PLA). Similarly, a wearable self-powered sweat sensor using a commercial flexible circuit board with PTFE and copper as triboelectric layers was developed [93,96]. A replaceable microfluidic pH monitor and Na^+ sensor were integrated with the energy harvesting device, as shown in Fig. 5A. The engineered device displayed a high power output of ~ 416 mW m^{-2}. The TENG energy generated during exercise was stored in a capacitor, and the attached microfluidic electrochemical sweat sensor patch continuously performed biomarker monitoring. The device performance was not much affected even after many cycles during on-body human trials, as shown in Fig. 5B.

Self-powered and self-healing sweat sensor based on nanocellulose hydrogel has been recently reported [97]. The sensor was demonstrated to have self-healing efficiencies up to 95 % within 10 s, a stretchability of 1530 %, and a conductivity of 0.6 S m^{-1}. Certain designs allow for movement-monitoring sweat-resistant sensors. One such system uses two superhydrophobic outer layers to resist sweat absorbance and two self-cleaning tribo layers compromising of the application of resin and PDMS, as shown in Fig. 5C [93]. The electric signals generated during exercise on dry as well as sweaty TENG surfaces exhibit different characteristic patterns, as shown in Fig. 5D. It was found that the sweat-resistant TENG has a 2-fold increased output after sweat evaporation as compared to normal TENG because of sweat contamination effect in the normal TENG after heavy exercise.

Despite extensive research in this area, self-powered sweat sensors are far from being realized for commercial health monitoring devices. The major challenge is sweat sample collection as the sweat generated for different reasons (i.e., physical activity, thermal heating, by the body at rest or in stress) has different compositions. Mostly the sweat generated during exercise is monitored. But to achieve continuous monitoring, sweat collection and analysis under low metabolism is required. Moreover, sweat is a complicated fluid with more complexity than blood. Thus, its contamination should be avoided. In this context, closed collection systems have been found to be useful. The amount of sweat produced by the body at times may not be sufficient for analysis which may affect the continuous functioning of the device. A distributive multipoint sweat collection may prove helpful in such cases. Self-powered TENGs-based sweat sensors have shown potential as future technologies, but integrating the self-powered system with the sensor increases the complexity of the design. Efforts should be made to obtain and utilize advanced materials and integration methods to fabricate commercially successful products.

Other sensors

Various other types of biomedical sensors based on TENGs have been reported [98–102]. A bladder sensor combined with a micro-actuator capable of monitoring the fullness of the bladder was proposed using polyethylene terephthalate (PET) and polyvinyl chloride (PVC) as two triboelectric layers [103]. The voltage of the TENG sensor increased from 35.6 to 114 mV when the external force increased from 0 to 6.86 N. The group tested the integrated TENG sensor with the actuator using the benchtop setup *in vitro*. Song et al. reported a body movement sensing TENG with a sandwich structure having a floating cantilever whose movement monitored sleep conditions [104]. A neural sensor successfully monitoring different neural signals with varying amplitudes and latencies has been reported, which functioned as a neuro-modulator to

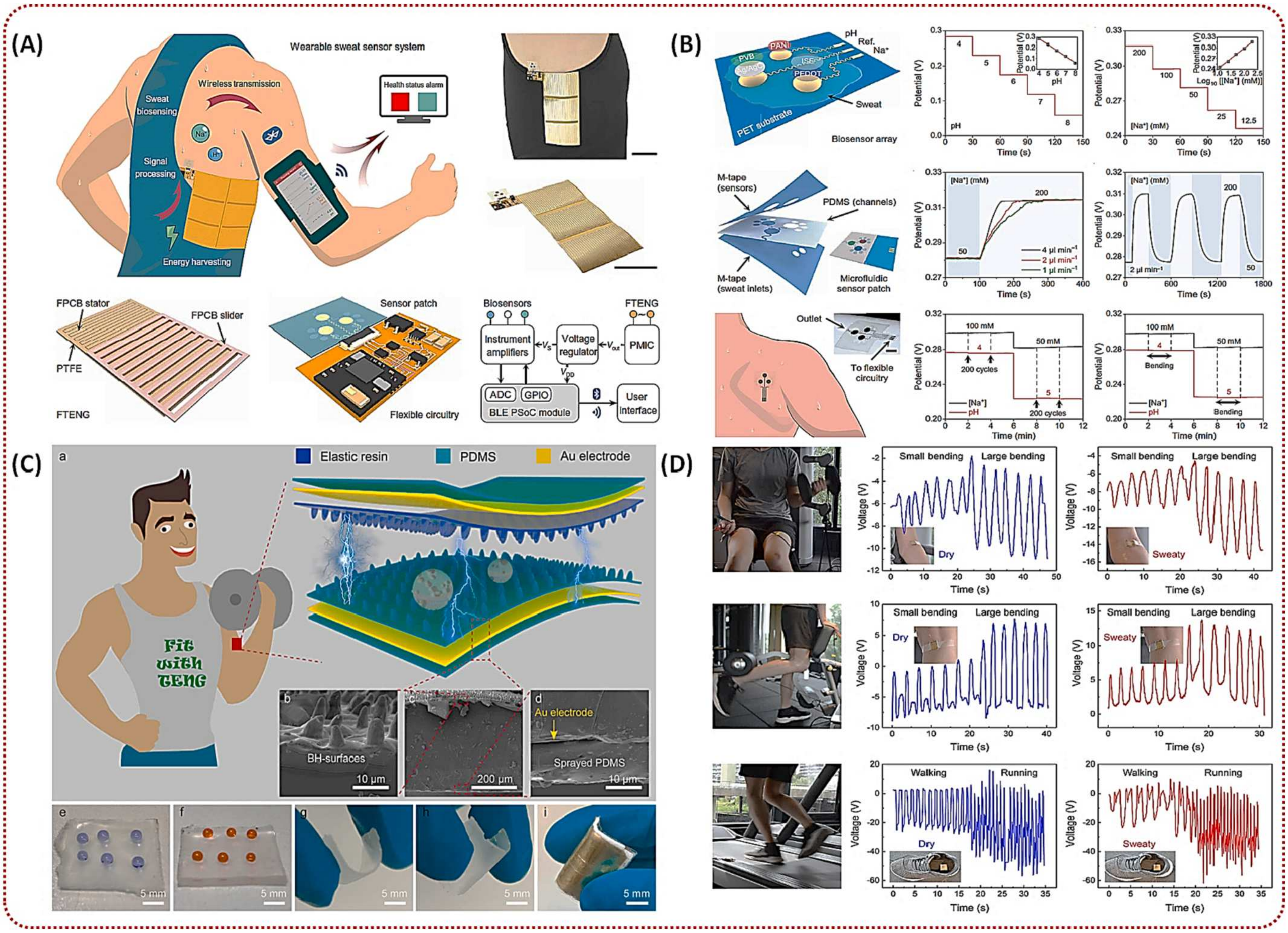

Fig. 5. (A) Battery-free wireless and non-invasive molecular monitoring (B) Demonstration of the microfluidic-based patch and their response [92] (C) Schematic structure of bioinspired sweat-resistant wearable triboelectric nanogenerator (BSRW-TENG) (D) Output response of different activities [93].

control muscle movement [105].

A self-powered artificial mechanoreceptor module was demonstrated by Han et al. in which the TENG device acted as a tactile sensor with the current generated varying with pressure applied and serving as an input for a spiking neural network (SNN) as shown in Fig. 6A [106]. The TENG sensor could function at a low pressure of $\sim$ 3 kPa. The neuromorphic tactile system was integrated with a breath monitoring system with a wind and bending-type mechanoreceptor for monitoring human breath. Such mechanoreceptors might be beneficial for applications in tactile systems for robotics, prosthetics, and medical and healthcare applications. Other TENGs-based motion sensors that can sense small displacements, such as eye muscles, have been reported. An eye motion sensor comprised of a new TENG configuration named Non-Attached Electrode-Dielectric Triboelectric Sensor (NEDTS) was used to monitor voluntary and involuntary eye blinks [107]. The conductive electrodes of TENG were, however, not bonded to the dielectric materials by any coating or sputtering process, as shown in Fig. 6 B. The voltage was generated in a separate conductor by non-contact electrostatic induction via triboelectric interaction in TENG during motion. Since the movement of eye muscles triggered the output of the TENG and the displacement was small, the position of the attachment of sensor was found to be very important. The device was integrated into a human–machine interface (HMI) system to assist disabled people. The number of blinks was used for hands-free computer navigation and even drone control. The sensor also monitored the driver's fatigue and sleep condition.

A TENG-based sensor capable of monitoring small arterial pressure changes due to movement in different body parts like the throat was reported by Yang et al. [62]. It showed a vibration-induced electricity generation and was termed a bionic membrane sensor as it could be utilized to measure rapidly changing frequency over extensive frequency ranges from 0.1 to 3.2 kHz. The sensor patch was fabricated using layers of PET, PTFE, and ITO, which were used to distinguish dynamic patterns of the cardiovascular system, human voice monitoring, and arterial pulse wave monitoring, as shown in Fig. 6C.

Gait-based identity recognition is an emerging technology to provide high-quality healthcare services for elderly people. As the number of elders and empty nesters is increasing globally, their health monitoring is becoming a challenge worldwide because they have more probability of developing diseases like hypertension, hyperglycemia, asthma, stroke, cognitive decline, etc. Studies have shown that gait variability can be related to functional health and indicate chronic diseases like Parkinson's. Long-time gait monitoring and data acquisition help in the health assessment of the elderly. The gait monitoring devices based on TENG fabrics designed for a foot are considered very important because the activities of the foot are one of the major sources for harvesting kinetic energy from the human body. Insoles with energy harvesting and gait sensing capacities have proven to be very efficient as their demand for comfortability and flexibility is low [108]. The smart TENG insoles capable of real-time monitoring have been used for the detection of emergency falls and monitoring of rehabilitation, as shown in Fig. 6D. Anaya et al. used an integrated system of non-contact triboelectric

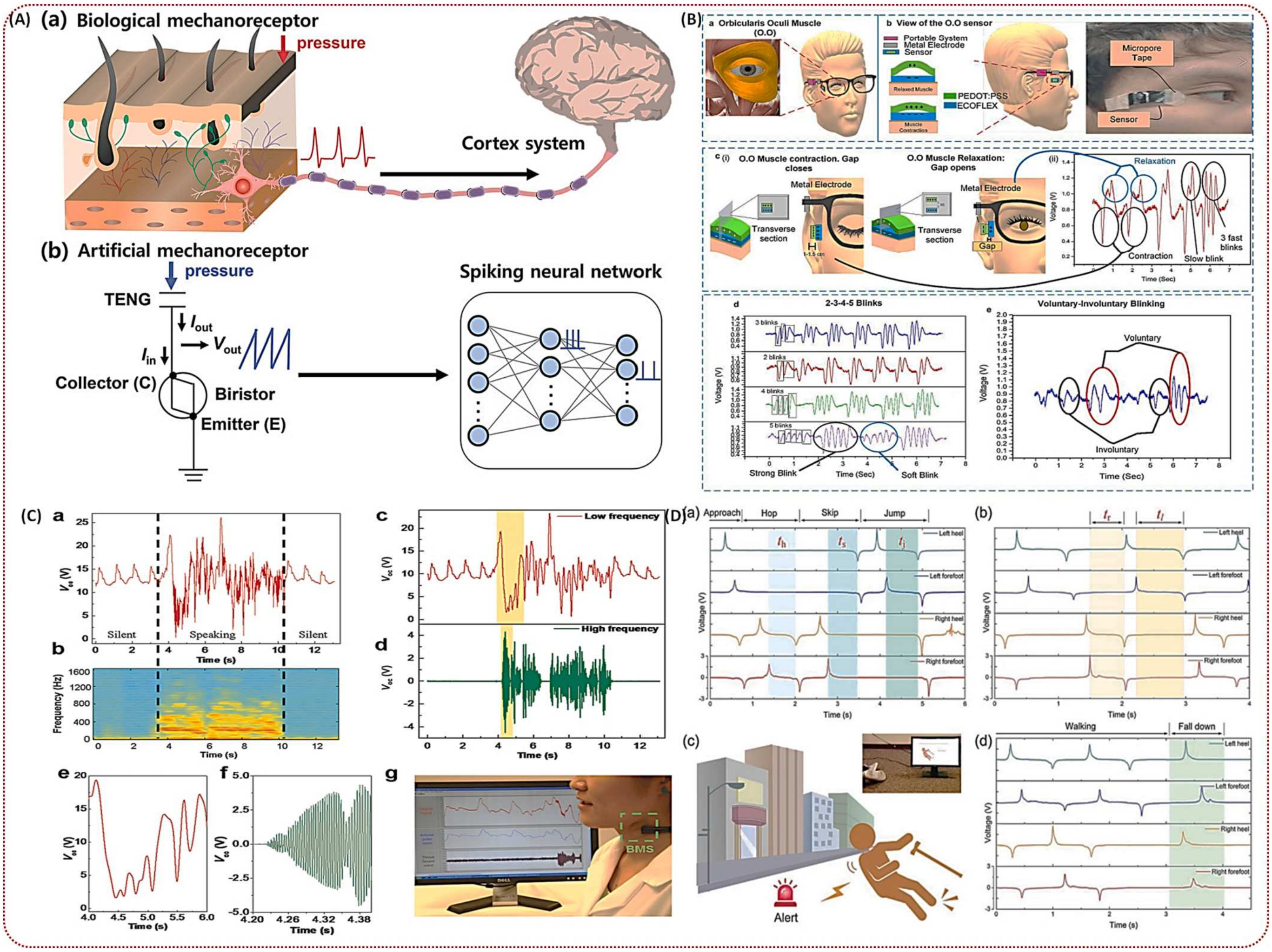

Fig. 6. (A) Biological and artificial tactile perception systems and their behavior concerning the SNN for neural processing [106] (B) Self-powered eye motion biomedical sensor and their response [107] (C) Voice recognition sensor and their responses [62] (D) Demonstrations of the smart insoles for warning of fall down [108].

systems for multiparameter monitoring. The system was found very useful for monitoring the activities of older adults using proximity and collision sensors [109]. Similarly, an artificial intelligence based TENG powered fall detection system was designed for the detection of accidents during walking in elderly [110].

With the innovative assembly of the TENGs at different locations on a body, there has been a lot of research on TENG-based sensors focusing on high power density, multiple functions, reduced biological damage, and improved chronic implantation. Biodegradable TENGs have been found suitable for wound-healing applications [111]. Peng et al. have conducted a study to investigate the practical sensor applications of these TENGs [112]. They evaluated the previous years' research publications and patents filed to correlate the experimental development with the commercial success of the sensors. It was concluded that stronger collaboration between the research community and industry resulted in more patents on the proposed devices. We suggest increasing industry-academia partnerships to convert research findings into commercial devices.Supplementary Materials.

Challenges in the applications of TENGs

He et al. monitored whole-body physiological signals towards next-generation technology. They demonstrated the TENG sensors' self-powered flexible, stretchable, and high-sensing mechanoluminescent features. The fabricated TENG sensor can detect various degrees of human physiology signals for recognition, as shown in Fig. 7A. Based on these electrical signals, we can figure out how the human body moves, which can help with medical rehabilitation for younger as well as older people [113].

In the last few decades, wearable electronics has significantly evolved due to the use of more flexible materials and increased portability by nature. Artificial intelligence (AI) techniques will eventually amplify wearable electronics' intelligence and provide more reliable and straightforward solutions to problems and resonating tasks for smart home applications, as shown in Fig. 7B. [114].

TENG technology has undergone rapid development in fundamental understanding and applications over the last decade. However, there are several hurdles in their application on a commercial scale. Though triboelectrification has been known for many years, more studies and experiments should be done to understand the origin of electrostatic charges at the material surface. A major limitation of TENG-driven self-powered electronic devices is due to irregular and random ambient environments contributing to the devices instability and erratic electrical output [115]. Several optimizations in the structure and design of the TENG devices have been done to overcome this drawback [116]. A possible solution could be to combine energy harvesting with storage so that excess energy produced during one cycle can be utilized later during the low-output phase. There is also a challenge in designing better

Fig. 7. Applications of TENGs Sensors towards next-generation era (A) Monitoring whole-body physiological signal [113] (B) The intelligent wearable electronics toward digital humans for smart home application [114].

materials. Since the output power is proportional to the square of the surface charge density and is related to the dielectric constant of the materials, there is thus a considerable need to optimize these properties for maximum performance [112]. A possible solution could be to develop a suitable surface micro/nanostructures to enhance surface area and create effective contacts. Durability, stability, and packaging are the other issues that need to be considered before making the TENGs-based devices applicable on a large scale. Since the operation of such devices is based on the physical contact of the materials, the materials used should be wear and tear-resistant for long-lasting performance. The triboelectric surfaces are affected by the ambient environmental conditions that can get contaminated easily, thus requiring efficient packaging materials for their protection. Furthermore, the TENGs fabricated so far have complicated and sophisticated procedures which are challenging to reproduce on a large scale, including the characteristics of different TENGs-based biomedical sensors summarized in Table 1.

From the theoretical modeling perspective, progress has been made in understanding the triboelectrification. A few reports on new and less complicated TENG designs have recently been suggested. However, much remains to be done to investigate the behavior of such systems from the first principles and make them commercially viable.

Conclusion and future prospects

The discovery of TENGs is a milestone in green energy harvesting and powering biomedical devices. A maximum of 85 % energy conversion efficiency has been demonstrated till date. We have discussed different biomedical sensors that function using TENG technology. Other practical applications involving TENGs include water decontamination, controlled drug delivery, biodegradable electronics, voice recognition, and bio-monitoring. TENGs can be utilized to fabricate self-powered wearable and implantable devices for long-term functioning allowing even unattended monitoring of older patients continuously. The development and utilization of TENGs is expected to have a significant effect on the healthcare industry. Though the technology has evolved at a considerable pace from its first discovery, it is still far from its final stage of development. To fully utilize the benefits of TENGs, more work needs to be done on the miniaturization of devices, improvement in biocompatibility, stability, and durability of materials; simplification of designs; more efficient energy storage techniques, and low-cost device fabrication to make them suitable for bulk fabrication. It is expected that the next few years will soon witness the considerably improved efficiency of some TENGs for commercialization.

CRediT authorship contribution statement

Shipra Solanki: Conceptualization, Writing – original draft. **Akhilesh Kumar Gupta:** Conceptualization, Writing – original draft. **Udiptya Saha:** Writing – review & editing. **Alexey V. Krasnoslobodtsev:** Writing – review & editing. **Rajinder K. Gupta:** Writing – original draft, Writing – review & editing. **Bansi D. Malhotra:** Conceptualization, Writing – review & editing, Supervision.

Declaration of Competing Interest

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

Data availability

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

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Appendix A. Supplementary data

Supplementary data to this article can be found online at https://doi.org/10.1016/j.seta.2023.103233.

References

[1] Solanki S, Pandey CM, Gupta RK, Malhotra BD. Biotechnol J 2020;15(5):1900279.
[2] Kumar S, Pandey CM, Hatamie A, Simchi A, Willander M, Malhotra BD. Global Chall 2019;3(12):1900041.
[3] RamRakhyani AK, Mirabbasi S, Chiao M. IEEE Trans Biomed Circuits Syst 2010;5(1):48–63.
[4] Wang ZL. Faraday Discuss 2015;176:447–58.
[5] Zhang X-S, Han M-D, Wang R-X, Zhu F-Y, Li Z-H, Wang W, et al. Nano Lett 2013;13(3):1168–72.
[6] Zhang N, Tao C, Fan X, Chen J. J Mater Res 2017;32(9):1628–46.
[7] Gupta AK, Kumar RR, Ghosh A, Lin S-P. Mater Lett 2022;310:131541.
[8] M. Parvez Mahmud, N. Huda, S.H. Farjana, M. Asadnia, C. Lang, Adv. Energy Mater., 8 (2) (2018) 1701210.
[9] Hinchet R, Kim S-W. ACS Nano 2015;9(8):7742–5.
[10] Hannan MA, Mutashar S, Samad SA, Hussain A. Biomed Eng Online 2014;13(1):79.
[11] Sue C-Y, Tsai N-C. Appl Energy 2012;93:390–403.
[12] Romero E, Warrington R, Neuman M. Physiol Meas 2009;30(9):R35.
[13] Hsu C-H, Gupta AK, Purwidyantri A, Prabowo BA, Chen C-H, Chuang C-C, et al. Chemosensors 2022;10(3):94.
[14] Sun J, Yang A, Zhao C, Liu F, Li Z. Sci Bull 2019;64(18):1336–47.
[15] Liu Y, Wang L, Zhao L, Yao K, Xie Z, Zi Y, et al. Adv Electron Mater 2019;6(1):1901174.
[16] Shi L, Hu Y, He Y. Nano Energy 2020;71:104582.
[17] Su Y, Wang J, Wang B, Yang T, Yang B, Xie G, et al. ACS Nano 2020;14(5):6067–75.
[18] Wang ZL. ACS Nano 2013;7(11):9533–57.
[19] Liu Z, Li H, Shi B, Fan Y, Wang ZL, Li Z. Adv Funct Mater 2019;29(20):1808820.
[20] Li X, Sun M, Wei X, Shan C, Chen Q. Nanomaterials 2018;8(4):188.
[21] Lee M, Chen CY, Wang S, Cha SN, Park YJ, Kim JM, et al. Adv Mater 2012;24(13):1759–64.
[22] Choi D, Choi MY, Choi WM, Shin HJ, Park HK, Seo JS, et al. Adv Mater 2010;22(19):2187–92.
[23] Cha SN, Seo JS, Kim SM, Kim HJ, Park YJ, Kim SW, et al. Adv Mater 2010;22(42):4726–30.
[24] Hu W, Wei X, Zhu L, Yin D, Wei A, Bi X, et al. Nano Energy 2019;57:600–7.
[25] Vivekananthan V, Chandrasekhar A, Alluri NR, Purusothaman Y, Kim WJ, Kang C-N, et al. Mater Lett 2019;249:73–6.
[26] Zheng Q, Shi B, Fan F, Wang X, Yan L, Yuan W, et al. Adv Mater 2014;26(33):5851–6.
[27] Jeon S-B, Nho Y-H, Park S-J, Kim W-G, Tcho I-W, Kim D, et al. Nano Energy 2017;41:139–47.
[28] Li R, Wei X, Xu J, Chen J, Li B, Wu Z, et al. Micromachines 2021;12(4):352.
[29] Wang ZL, Wang AC. On the origin of contact-electrification. Mater Today 2019.
[30] Yang W, Chen J, Jing Q, Yang J, Wen X, Su Y, et al. Adv Funct Mater 2014;24(26):4090–6.
[31] Niu S, Wang ZL. Nano Energy 2015;14:161–92.
[32] Chen J, Zhu G, Yang W, Jing Q, Bai P, Yang Y, et al. Adv Mater 2013;25(42):6094–9.
[33] Fan X, Chen J, Yang J, Bai P, Li Z, Wang ZL. ACS Nano 2015;9(4):4236–43.
[34] Wen Z, Chen J, Yeh M-H, Guo H, Li Z, Fan X, et al. Nano Energy 2015;16:38–46.
[35] Jing Q, Zhu G, Bai P, Xie Y, Chen J, Han RPS, et al. ACS Nano 2014;8(4):3836–42.
[36] Yang Y, Zhang H, Chen J, Jing Q, Zhou YS, Wen X, et al. ACS Nano 2013;7(8):7342–51.
[37] Wang S, Xie Y, Niu S, Lin L, Wang ZL. Adv Mater 2014;26(18):2818–24.
[38] Zhao Z, Yan C, Liu Z, Fu X, Peng LM, Hu Y, et al. Adv Mater 2016;28(46):10267–74.
[39] Gupta AK, Hsu C-H, Lai S-N, Lai C-S. ECS J Solid State Sci Technol 2020;9(11):115019.
[40] Liu J, Gu L, Cui N, Xu Q, Qin Y, Yang R. Res 2019;2019:1091632.
[41] Kwak SS, Yoon HJ, Kim SW. Adv Funct Mater 2019;29(2):1804533.
[42] Dong K, Wang Y-C, Deng J, Dai Y, Zhang SL, Zou H, et al. ACS Nano 2017;11(9):9490–9.
[43] Zhang N, Huang F, Zhao S, Lv X, Zhou Y, Xiang S, et al. Matter 2020;2(5):1260–9.
[44] Zhang Y, Zhou Z, Sun L, Liu Z, Xia X, Tao TH. Adv Mater 2018;30(50):1805722.
[45] Kuzma M, Gerhard E, Shan D, Yang J. Advances in Bioresorbable Electronics and Uses in Biomedical Sensing. in: Interfacing Bioelectronics and Biomedical Sensing, Springer; 2020. p. 29–72.
[46] Wen D-L, Liu X, Deng H-T, Sun D-H, Qian H-Y, Brugger J, et al. Nano Energy 2019;66:104123.
[47] Paosangthong W, Torah R, Beeby S. Nano Energy 2019;55:401–23.
[48] Wang W, Yu A, Zhai J, Wang ZL. Adv Fiber Mater 2021;3(6):394–412.

[49] Hu Y, Zheng Z. Nano Energy 2019;56:16–24.
[50] Hatamie A, Angizi S, Kumar S, Pandey CM, Simchi A, Willander M, et al. J Electrochem Soc 2020;167(3):037546.
[51] Chen G, Li Y, Bick M, Chen J. Chem Rev 2020;120(8):3668–720.
[52] Yang Y, Xie L, Wen Z, Chen C, Chen X, Wei A, et al. Mter Interfaces 2018;10(49): 42356–62.
[53] Jiang W, Li H, Liu Z, Li Z, Tian J, Shi B, et al. Adv Mater 2018;30(32):1801895.
[54] Dong Y, Mallineni SSK, Maleski K, Behlow H, Mochalin VN, Rao AM, et al. Nano Energy 2018;44:103–10.
[55] Li Z, Zheng Q, Wang ZL, Li Z. Research 2020;2020:8710686.
[56] Lin Z, Yang J, Li X, Wu Y, Wei W, Liu J, et al. Adv Funct Mater 2017;28(1): 1704112.
[57] Yi F, Lin L, Niu S, Yang PK, Wang Z, Chen J, et al. Adv Funct Mater 2015;25(24): 3688–96.
[58] Yang T, Pan H, Tian G, Zhang B, Xiong D, Gao Y, et al. Nano Energy 2020;72: 104706.
[59] Yang W, Chen J, Wen X, Jing Q, Yang J, Su Y, et al. ACS appl mater interfaces 2014;6(10):7479–84.
[60] Yan C, Deng W, Jin L, Yang T, Wang Z, Chu X, et al. ACS appl mater interfaces 2018;10(48):41070–5.
[61] Meng K, Zhao S, Zhou Y, Wu Y, Zhang S, He Q, et al. Matter 2020;2(4):896–907.
[62] Yang J, Chen J, Su Y, Jing Q, Li Z, Yi F, et al. Adv Mater 2015;27(8):1316–26.
[63] Meng K, Chen J, Li X, Wu Y, Fan W, Zhou Z, et al. Adv Funct Mater 2019;29(5): 1806388.
[64] X. Wang, J. Liang, Y. Xiao, Y. Wu, Y. Deng, X. Wang, M. Zhang, Journal of Physics: Conference Series, IOP Publishing, 2018, pp. 012009.
[65] Parida K, Kumar V, Jiangxin W, Bhavanasi V, Bendi R, Lee PS. Adv Mater 2017;29 (37):1702181.
[66] Ding W, Wang AC, Wu C, Guo H, Wang ZL. Adv Mater Technol 2019;4(1): 1800487.
[67] Cheng Y, Wu D, Hao S, Jie Y, Cao X, Wang N, et al. Nano Energy 2019;64:103907.
[68] Wu C, Kim TW, Park JH, Koo B, Sung S, Shao J, et al. ACS Nano 2019;14(2): 1390–8.
[69] Zhang R, Hummelgård M, Örtegren J, Olsen M, Andersson H, Olin H. Nano Energy 2019;57:279–92.
[70] Anaya DV, He T, Lee C, Yuce MR. Nano Energy 2020:104675.
[71] Xu R, Luo F, Zhu Z, Li M, Chen B, Appl ACS. Electron Mater 2022;4(8):4051–60.
[72] Zhu Y, Sun F, Jia C, Huang C, Wang K, Li Y, et al. Sustainability 2022;14(17): 10875.
[73] Mathew AA, Chandrasekhar A, Vivekanandan S. Nano Energy 2021;80:105566.
[74] Ouyang H, Liu Z, Li N, Shi B, Zou Y, Xie F, et al. Nat commun 2019;10(1):1–10.
[75] Ouyang H, Tian J, Sun G, Zou Y, Liu Z, Li H, et al. Adv Mater 2017;29(40): 1703456.
[76] Ryu H, Park H-M, Kim M-K, Kim B, Myoung HS, Kim TY, et al. Nat commun 2021; 12(1):1–9.
[77] Lin Z, Chen J, Li X, Zhou Z, Meng K, Wei W, et al. ACS Nano 2017;11(9):8830–7.
[78] Chen S, Wu N, Lin S, Duan J, Xu Z, Pan Y, et al. Nano Energy 2020:104460.
[79] Su Y, Yang T, Zhao X, Cai Z, Chen G, Yao M, et al. Nano Energy 2020;74:104941.
[80] Su Y, Yao M, Xie G, Pan H, Yuan H, Yang M, et al. Appl Phy Lett 2019;115(7): 073504.
[81] Guan X, Xu B, Wu M, Jing T, Yang Y, Gao Y. Nano Energy 2021;80:105549.
[82] Ning C, Cheng R, Jiang Y, Sheng F, Yi J, Shen S, et al. ACS Nano 2022;16(2): 2811–21.
[83] Wang M, Zhang J, Tang Y, Li J, Zhang B, Liang E, et al. ACS Nano 2018;12(6): 6156–62.
[84] Wang S, Tai H, Liu B, Duan Z, Yuan Z, Pan H, et al. Nano Energy 2019;58:312–21.
[85] Su Y, Wang J, Wang B, Yang T, Yang B, Xie G, et al. ACS Nano 2020.
[86] Brasier N, Eckstein J. Digital Biomarkers 2019;3(3):155–65.
[87] Choi J, Bandodkar AJ, Reeder JT, Ray TR, Turnquist A, Kim SB, et al. ACS Sensors 2019;4(2):379–88.
[88] Bariya M, Nyein HYY, Javey A. Nat Electron 2018;1(3):160–71.
[89] Tai LC, Gao W, Chao M, Bariya M, Ngo QP, Shahpar Z, et al. Adv Mater 2018;30 (23):1707442.
[90] Brothers MC, DeBrosse M, Grigsby CC, Naik RR, Hussain SM, Heikenfeld J, et al. Acc chem res 2019;52(2):297–306.
[91] Mohan AV, Rajendran V, Mishra RK, Jayaraman M. TrAC, Trends Anal Chem 2020;131:116024.
[92] Li W, Lu L, Kottapalli AGP, Pei Y. Nano Energy 2022:107018.
[93] Song Y, Min J, Yu Y, Wang H, Yang Y, Zhang H, et al. Sci adv 2020;6(40): eaay9842.
[94] Zhu M, Shi Q, He T, Yi Z, Ma Y, Yang B, et al. ACS Nano 2019;13(2):1940–52.
[95] Gai Y, Wang E, Liu M, Xie L, Bai Y, Yang Y, et al. Small Methods 2022:2200653.
[96] Y. Song, J. Min, Y. Yu, H. Wang, Y. Yang, H. Zhang, W. Gao, Sci. adv, 6 (40) eaay9842.
[97] Qin Y, Mo J, Liu Y, Zhang S, Wang J, Fu Q, et al. Adv Funct Mater 2020:2201846.
[98] Jao Y-T, Yang P-K, Chiu C-M, Lin Y-J, Chen S-W, Choi D, et al. Nano Energy 2020; 50:513–20.
[99] Wu Y, Li Y, Zou Y, Rao W, Gai Y, Xue J, et al. Nano Energy 2022;92:106715.
[100] Yang L, Liu C, Yuan W, Meng C, Dutta A, Chen X, et al. Nano Energy 2022: 107807.
[101] Liu Y, Zhao W, Liu G, Bu T, Xia Y, Xu S, et al. Nano Energy 2021;85:105967.
[102] Zhou Z, Padgett S, Cai Z, Conta G, Wu Y, He Q, et al. Biosens Bioelectron 2020; 155:112064.
[103] Arab Hassani F, Mogan RP, Gammad GG, Wang H, Yen S-C, Thakor NV, et al. ACS Nano 2018;12(4):3487–501.
[104] Song W, Gan B, Jiang T, Zhang Y, Yu A, Yuan H, et al. ACS Nano 2016;10(8): 8097–103.
[105] Lee S, Wang H, Shi Q, Dhakar L, Wang J, Thakor NV, et al. Nano Energy 2017;33: 1–11.
[106] Han J-K, Tcho I-W, Jeon S-B, Yu J-M, Kim W-G, Choi Y-K. Adv Sci 2022;9(9): 2105076.
[107] Anaya DV, He T, Lee C, Yuce MR. Nano Energy 2020;72:104675.
[108] Lin Z, Wu Z, Zhang B, Wang YC, Guo H, Liu G, et al. Adv Mater Technol 2022;4 (2):1800360.
[109] Anaya DV, Zhan K, Tao L, Lee C, Yuce MR, Alan T. Nano Energy 2021;90:106486.
[110] Wang S, Gao J, Lu F, Wang F, You Z, Huang M, et al. Nano Energy 2023;108: 108230.
[111] Li Z, Feng H, Zheng Q, Li H, Zhao C, Ouyang H, et al. Nano Energy 2018;54: 390–9.
[112] Peng H, Fang X, Ranaei S, Wen Z, Porter AL. Nano Energy 2017;35:358–69.
[113] He M, Du W, Feng Y, Li S, Wang W, Zhang X, et al. Nano Energy 2021;86:106058.
[114] Zhang Z, He T, Zhu M, Sun Z, Shi Q, Zhu J, et al. npj Flexible Electron 2020;4(1): 1–12.
[115] Zhang C, Zhou L, Cheng P, Yin X, Liu D, Li X, et al. Appl Mater Today 2020;18: 100496.
[116] Niu S, Liu Y, Zhou YS, Wang S, Lin L, Wang ZL. IEEE Trans Electron Devices 2014; 62(2):641–7.
[117] Ding X, Cao H, Zhang X, Li M, Liu Y. Sensors 2018;18(6):1713.
[118] Maharjan P, Toyabur R, Park J. Nano Energy 2018;46:383–95.
[119] Bhatta T, Maharjan P, Salauddin M, Rahman MT, Rana SS, Park JY. Adv Funct Mater 2020;30(36):2003276.
[120] Liu Z, Zhao Z, Zeng X, Fu X, Hu Y. Nano Energy 2019;59:295–301.
[121] Zhao G, Zhang Y, Shi N, Liu Z, Zhang X, Wu M, et al. Nano Energy 2019;59: 302–10.
[122] Ha M, Lim S, Cho S, Lee Y, Na S, Baig C, et al. ACS Nano 2018;12(4):3964–74.
[123] Zhang Z, Zhang J, Zhang H, Wang H, Hu Z, Xuan W, et al. Nanoscale Res Lett 2019;14(1):354.
[124] Cui X, Zhang C, Liu W, Zhang Y, Zhang J, Li X, et al. Sens Actuator A Phys 2018; 280:326–31.
[125] Hassani FA, Mogan RP, Gammad GG, Wang H, Yen SC, Thakor NV, et al. ACS Nano 2018;12(4):3487–501.
[126] Ma Y, Zheng Q, Liu Y, Shi B, Xue X, Ji W, et al. Nano Lett 2016;16(10):6042–51.
[127] Feng M, Wu Y, Feng Y, Dong Y, Liu Y, Peng J, et al. Nano Energy 2021;93:106835.
[128] H. He, J. Guo, B. Illas, A. Gaczy, B.z. Istak, V. Hliva, D. Tarak, J.G. Kovacs, I. Harmati, K. Molnar, Nano Energy, 89 (2021) 106418.
[129] J. He, Z. Xie, K. Yao, D. Li, Y. Liu, Z. Gao, W. Lu, L. Chang, X. Yu, Nano Energy, 81 (2021) 105590.