Chapter

Innovative Wearable Sensors Based on Hybrid Materials for Real-Time Breath Monitoring

Mourad Roudjane and Younès Messaddeq

Abstract

This chapter will present the importance of innovative hybrid materials for the development of a new generation of wearable sensors and the high impact on improving patient's health care. Suitable conductive nanoparticles when embedded into a polymeric or glass host matrix enable the fabrication of flexible sensor capable to perform automatic monitoring of human vital signs. Breath is a key vital sign, and its continuous monitoring is very important including the detection of sleep apnea. Many research groups work to develop wearable devices capable to monitor continuously breathing activity in different conditions. The tendency of integrating wearable sensors into garment is becoming more popular. The main reason is because textile is surrounding us 7 days a week and 24 h a day, and it is easy to use by the wearer without interrupting their daily activities. Technologies based on contact/noncontact and textile sensors for breath detection are addressed in this chapter. New technology based on multi-material fiber antenna opens the door to future methods of noninvasive and flexible sensor network for real-time breath monitoring. This technology will be presented in all its aspects.

Keywords: wireless communication, wearable sensors, flexible antenna, innovative material, smart textile, breath monitoring

1. Introduction

Wearable sensors for vital signs monitoring are becoming key emerging technologies in different research fields such as medical science [1], sports and fitness [2], and military [3], to name a few. They present tremendous potential for providing a diagnosis of the subject's health status in their home, with the possibility to progress toward the concept of personalized medicine. The sensor is mostly made of conductive electrodes to detect the physiological signal, and it is an electronic device that seamlessly tracks and transfers all biometric data into an actionable base station with user interface for analysis and interpretation and data storage. Since two decades ago, tremendous efforts have been made in material sciences, radio frequency communications, and biomedical electronics research domains to transform these sensors as research and development laboratory tools to a commercial technology market. This transformation is driven by the urgent need to lower the high cost associated with health-care services in most countries, which continues to soar because of the increasing price of medical instruments and hospital care that would impose significant burdens on the socioeconomic structure of the countries [4]. Remote health-care noninvasive wearable sensors are ubiquitously alternative diagnostic tools for monitoring important physiological signs and activities of the patients in real time. New wearable devices with multi-functionalities are becoming more popular and are attracting more consumers. According to Idtechex's report on wearable sensors, the global wearable market is worth about 5 billion US dollars and is estimated to grow as much as 160 billions by 2028 [5]. Consequently, over the last few years, many companies developed and released to the market different wearable products in forms of smart watches, bracelets, skin patches, headbands, earphones, and fitness bands for unobtrusive, and when appropriate, continuous monitoring of some vital signs [6]. However, health data collected by these wearable devices suffer from inconsistencies and reliabilities that affect their usefulness and wellness for medical applications according to Erdmier, Hatcher, and Lee [7]. Wearable sensor for health monitoring systems can be integrated into textile fiber, clothes, and elastic bands or are directly attached to the human body [8]. The sensors are capable for measuring physiological parameters such as electrocardiogram (ECG), electromyogram (EMG), heart rate (HR), body temperature, electrodermal activity (EDA), blood pressure (BP), and breathing rate (BR) [9–11].

1.1 Remote health monitoring

The use of smartphones and tethered computers is now democratized worldwide and become a part of our daily life thanks to the recent advancement in mobile/computer technologies, which leads to a big change of every aspect of our lives starting from the way we work to social interactions. Remote health monitoring (or telemedicine) is no longer a science fiction as we used to see on our TV.

It starts to making initial inroads into health-care system. For example, a patient with chronic diseases, such as heart problems or diabetes, will continuously and simply monitor their health and send updates to their physician through the Internet. This can overcome the problem of infrequent clinical visits that can only provide a brief window into the physiological status of the patient. For a better acceptance of these technologies, wearable sensors' requirements must involve: comfort and ease to use, the ability to share the data with health-care professionals, very low energy consumption and long battery autonomy, and wireless communication with other devices [4, 12–17]. Remote health-care infrastructure is composed of wireless body area network, user interface smart digital assistant, and medical server for remote health-care monitoring system as illustrated in the general view architecture of **Figure 1**. In this architecture, a two-stage communication is used to transmit the health metrics recorded by the sensors to the remote health-care server. In the first stage, a short-range communication protocol is employed to transmit the measured data to a nearest gateway, such as smartphone, computer for advanced data processing, and display. The second stage consists of long range communication where the processed signal is transmitted to a server located in a health-care facility. The data can be transmitted over the Internet or cellular communication network such as general packet radio service (GPRS), 3G/4G, or Long-Term Evolution (LTE) services [18–20]. With the miniaturization of the electronic devices and powerful mobile computing capabilities, individuals are becoming capable of monitoring, tracking, and transmitting health biosignals continuously and in real time. These technologies will open a new era for humanity in particular when the access to medical centers and hospitals is more restricted such as in the case of a viral infection becoming a global pandemic.

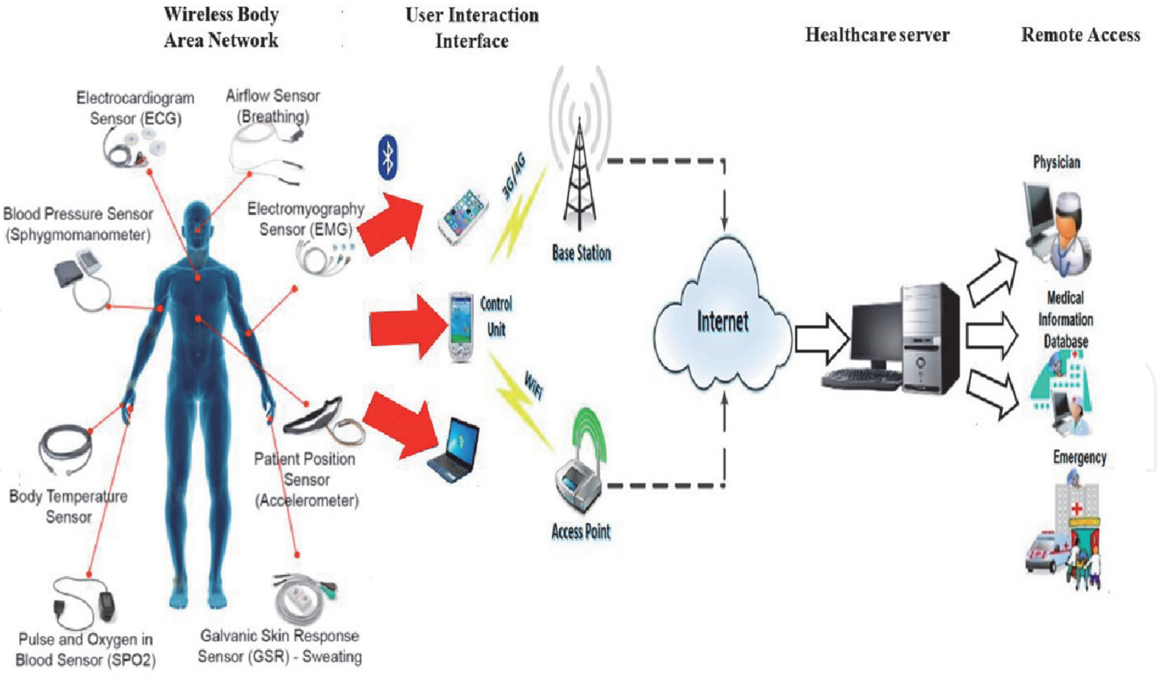

Figure 1.
General overview of the remote health monitoring system from. It is based on a wireless body area network, a user interaction interface, a healthcare server, and a remote access for medical professionals.

1.2 Human breathing mechanism

In general, breathing (or respiration) is an important physiological process of living organisms. It is defined by four parameters: inhalation time, exhalation time, breathing period, and breathing rate. For humans, this process results in air exchange between the lungs and external environment. It consists mostly of inhalation of oxygen through the nose or mouth into the lungs and flushing out carbon dioxide during exhalation. The entire process from the inhalation to exhalation is known as a breathing cycle. During breathing, inhalation is caused by the contraction of the human diaphragm which causes the intra-thoracic pressure to fall due to the enlargement of the thoracic cavity. The latter induces lung expansion due to inhalation. Once the gas exchange occur across the alveolar-capillary membrane [21], the exhalation of carbon dioxide allow the diaphragm and inter-costal muscles to relax. Consequently, the chest and abdomen return to the rest position.

1.3 Importance of breath monitoring with wearable sensors

Many evidences demonstrate the importance of breathing rate as a key parameter. It is an important indicator used to monitor the progression of illness. An abnormal BR has been shown to be an important predictor of serious events such as cardiac arrest and admission to an intensive care unit [22]. Besides, it is fundamental in the early detection of the risk of the occurrence of dangerous conditions such as sleep apnea [23], respiratory depression in post-surgical patients [24] and sudden infant death syndrome [25]. It was reported by Fieselmann et al. that a BR higher than 27 breaths/minute was the most important predictor of cardiac arrest in hospital wards [26]. Monitoring of breathing signal is also very important during anesthesia for evaluating sleep apnea disorders [27]. In addition, Tas et al have pointed out the importance of continuous monitoring of BR to study the effects of heroin administration with addicted person in order to avoid any heroin-induced respiratory depression during the treatment [28]. Although, BR is of paramount importance when assessing the health of a patient, it is still often measured clinically using a flow-meter embedded in a mouthpiece, or a mask rather than wearable and noninvasive devices with a continuous monitoring capability [29]. Therefore, technological development and clinical validation of new wearable sensors is more than urgent to feel the gap between the need of accurate and continuous measurement of breathing patterns and rate and the clinical practices [30]. This chapter covers the description of different classical techniques used for continuous monitoring of breath, and the new generation wearable sensor counterpart developed using different hybrid materials. The goal of this chapter is to make a clear and comprehensive review on the new generation of wearable sensors made of new flexible and biocompatible materials.

2. Wearable sensor classification

Wearable sensors can be classified into two categories: invasive and noninvasive. Invasive wearable sensors can be further classified as minimally invasive, such as subcutaneous electrodes to obtain the electromyography (EMG) signal, or as an implantable, such as a pacemaker. These kinds of sensors require a clinical intervention to place them inside the body. Noninvasive wearables may or may not be in physical contact with the body. These sensors are typically used in systems for continuous monitoring because they are ease to use and often do not require lots of assistance from a health-care professional. They could be in form of watches, bands or integrated into textile. Noncontact breath monitoring methods are becoming more popular since they have clear advantages over the contact methods. In the former, the patient comfort is taken into account, especially for long-term monitoring and improved accuracy as distress caused by a contact device may affect the breathing rate measurements.

3. Classic sensors: from bulk to flexible form

Classic sensor refers to the sensors often used by clinicians and health professionals to estimate the breathing frequency and pattern in the clinics or hospitals. In this section we will present these sensors in their original bulky and rigid form, and their innovative version based on plastic electronic technology. The latter are more soft, lightweight, and stretchable enabling applications that would be impossible to achieve using the bulky forms.

3.1 Respiratory airflow sensors

Airflow sensors are largely used in clinics and hospital for collecting the pattern trend of inhaled and exhaled air during breathing. The technology is based on measuring the amount of air inhaled and exalted using different mechanisms such as differential flowmeters for monitoring gases delivered by mechanical ventilators and recorded by commercial spirometers [31, 32], turbine flowmeters [33], and hot wire anemometers [31], similar to the differential flowmeters, and it consists of heated wires exchanging heat with the fluid flow. These sensors require often the use of mouthpiece which is uncomfortable for long-term use. In addition, they cannot be used for continuous monitoring of breath.

The exhaled breath composition is complex, and it includes mainly a mixture of nitrogen, oxygen, CO_2, water vapor, and other components' traces such as volatile organic compounds (VOCs), acetone, carbon monoxide, ammonia, and nitric oxide

[34]. For instance, breath carbon monoxide test was used for neonatal jaundice diagnosis [35], breath ammonia was proposed for assessment of asthma and hemodialysis [36], and breath VOCs were used for the diagnosis of ovarian cancer [37]. A portable device for measuring human breath ammonia was developed based on a single use, disposable, inkjet printed ammonia sensor fabricated using polyaniline nanoparticles [38]. More recently, a novel miniaturized sensor based on organic semiconductor material was developed and successfully tested to monitor ammonia breath [39]. Flexible sensing platform based on the integration a self-healable polymer substrate with five kinds of functionalized gold nanoparticle films was used for sensing pressure variation as well as 11 kinds of VOCs [40]. A novel wearable electronic based on flexible printed multiwalled carbon nanotubes (MWCNTs)/polymer sensor array was designed as a compact armband to detect VOCs as well [41]. Despite the development of flexible electronics for breath sensing, many challenges need to be addressed for technology acceptance. For example, the biomarkers exhaled from breath may suffer from the interference from humidity or contamination from ambient air. In addition, most of the proposed sensors were tested in a controlled work place which is not the case in real-life conditions. Lastly, the proposed technologies was not tested for real-time breath monitoring. As consequence, the future wearable sensors shall address all these issues.

3.2 Respiratory inductance plethysmography sensors

This noninvasive method is based on tracking the chest and abdominal wall movements during breath to measure the changes in circumference during respiration which is directly related to the tidal volume [42]. The sensor consists of two bands placed around the abdomen at the level of the umbilicus and over the rib cage. The bands are made from a flexible conducting material used as strain gage, and the concept was first developed in 1967 and has since been established for monitoring patients in a clinics and hospitals [43]. The principle of the strain gage sensor is based on the change of the conductor resistance when it is subject to external force such as the case during the breathing process. Stretchable strain gage sensors have been developed based on various transduction mechanisms [44], such as capacitive [45] piezoelectric [46], and piezoresistive sensing [47]. A miniaturized strain sensor composed of a piezoresistive metal thin film set in a silicone elastomer substrate with a footprint smaller than that of a typical Band-Aid was developed [48]. Breathing rate and volume were measured by sticking one sensor on the rib cage and the other sensor on the abdomen.

More recently, nanomaterials have been shown to be promising as innovative strain sensors, and are based on nanoparticles building block based on CNT [49], graphene [50], and silver nanowires [51]. However, this technology has its limitation in monitoring patients throughout the day because the bands prone to slippage, which alter the breathing measurements.

3.3 SpO2 oximeter

SpO2 oximeter is another noninvasive method used to measure accurately both oxygen saturation in the blood and heart rate, and is widely used in hospitals and clinics to monitor patient at risk of hypoxia [52]. During breathing process, the inhaled oxygen binds to the hemoglobin in red blood cells, then it is transported throughout the body in arterial blood. The SpO2 pulsed sensor uses two light sources emitting typically at 660 (red) and 940 nm (infrared) to measure the percentage of hemoglobin in the blood that is saturated with oxygen using internal detector. This percentage is called blood saturation, or SpO2 [53]. The breathing rate is determined accurately using wavelet analysis technique of the plethysmogram as demonstrated by Leonard et al. [54]. The bulky wearable oximetry sensors used actually in clinics and hospitals are often placed on the finger, or wrist, but could be also on head, earphones, thigh, and ankle, and they have been widely commercialized [55].

The clinical oximeter versions are of portable size. However, a lot of efforts in material science and electronics were made to develop new versions at a wearable size able to monitor accurately the oxygenation. Recently, an electronic patch oximeter capable to perform the oxygenation sensing and communicate the data via wireless protocol was developed and successfully tested [56]. The electronic patch could be attached on different parts of the skin surface. Furthermore, an all organic optoelectronic oximeter sensor was designed with organic materials on flexible substrates to measure accurately the pulse oxygenation [57]. Ultraflexible organic photonic skin oximeter, and a miniaturized battery free flexible and wearable pulse oximeter were also reported recentely [58, 59]. The optoelectronic skins are extremely thin, lightweight and stretchable. Nevertheless, the patch-based technology is still at the research and development level and requires more improvement for long-term monitoring since patches may cause irritation for some people.

3.4 Electrocardiogram sensors

Breathing can be derived from an electrocardiogram (ECG) signal by measuring the electrical activity generated by the action potentials in heart muscle at each heartbeat [30]. The body-surface ECG is influenced by electrode motion with respect to the heart and by changes in thoracic electrical impedance of the lungs (air in and out), which is well-correlated with breath [60]. The breath is derived from the ECG fluctuations and this technique is called ECG-Derived Respiration (EDR) [61]. The ECG signals has for many years acquired using conventional wet silver-silver chloride (Ag/AgCl) electrodes which convert ionic current on the skin surface to electronic currents for amplification and signal processing [62]. The signal is recorded by measuring the voltage difference between two or more electrodes on the body surface over time [63, 64]. Although Ag/AgCl electrodes are cheap and disposable, they require the use of a conducting gel between the electrode and the skin. This may cause several problems related to the comfort of the users when used form long time, and signal corruption and base line drift caused by the change of the chemical properties of skin-electrode contact surface [64, 65].

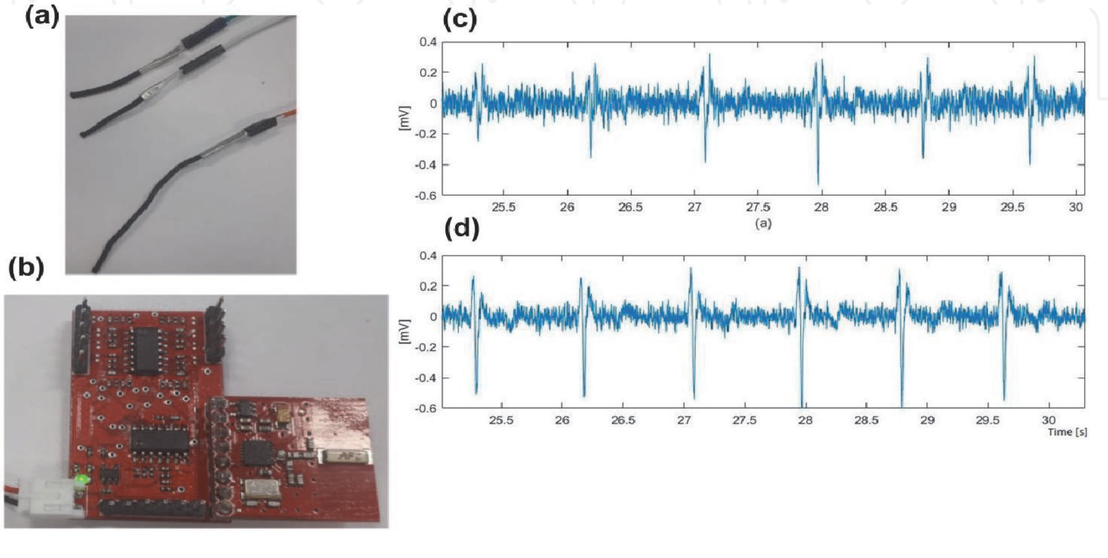

Figure 2.
Noninvasive ECG sensor made of flexible conductive fiber electrodes (a) connected to a wireless electronic system (b). ECG signals acquired with the conductive fiber (c) and with the Ag/AgCl electrodes (d) [68].

Flexibility of the electrodes is an important characteristic to achieve a better skin-electrode interface and to provide comfort to the users. Polymer dry electrode based on polydimethylsiloxane (PDMS) coated with metals such as Ti:Au [65], copper [66], and CNT [67] were previously developed. More recently, a biocompatible conductive polymer fiber doped with CNT was used to fabricate flexible and dry fiber electrodes for biopotential signals detection [68]. The ECG signal detection with these fiber electrodes was reported for the first time as shown in **Figure 2**, and their performance against moisture revealed no change in the quality of the detected signal in terms of signal-to-noise ratio. Nevertheless, the proposed system needs further development for medical application in particular the electrical sensing board.

Other contact-based methods were proposed such as humidity sensors, acoustic sensors, and air temperature sensors which are all reviewed in [30].

4. Imaging and microwave-based sensors

Beside the classical sensors, significant efforts were made by scientists to develop new noninvasive and contactless sensors capable to monitor breathing parameters. New approaches based on recording the chest and abdomen movements during breath using either camera imaging or microwave-based Doppler radar were proposed. For camera imaging, the working principal is based on evaluating posture changes and breathing rate of a subject often laying on a bed. For instance, infrared thermography based on wavelet decomposition [69], thermal imaging [70], camera-based systems [71], real-time vision-based methods [72], and a 3D vision tracking algorithm [73] were developed to measure the breathing rate in real time. For the microwave-based Doppler system, a radar source emitting pulsed or continuous microwave signal toward a subject, and the basic principle is that the Doppler radar signal reflected from the monitored subject is phase modulated by the subject's respiration and heartbeat. Continuous wave narrow-band radars [74], ultra-wide band (UWB) radars [75], and passive radar techniques [76] were proposed.

5. Textile as innovative sensors platform

Human-worn cloths since his birth, and they are considered as the first skin protection interface from the external environment. Clothing usually covers large parts of our body, and it was proposed for the first time in 1996 as the most appropriate platform to implement wearable systems for the unobtrusive monitoring of vital and bio-physiological signals [77]. Using conventional fabric manufacturing techniques (weaving, knitting, embroidery, and stitching), the combination of wireless communication sensing devices with functional yarn and fibers into a garment gives rise to the so-called smart textile (or electronic textile). A good smart textile is the one that has the capability of sensing accurately the external surrounding environment, while maintaining the same mechanical properties of a traditional garment. Based on the desired functionality, smart textile are divided into three categories: passive smart textile, which is considered as first generation model, has the minimum requirement for sensing an external stimulus, such as detecting the change of the external temperature; active smart textile is the second generation model with the capability to sens the surrounding environment such as temperature, light, odor, and reacts using various textile-based flexible and miniaturized actuators; finally, ultra (or very) smart textile is the third generation

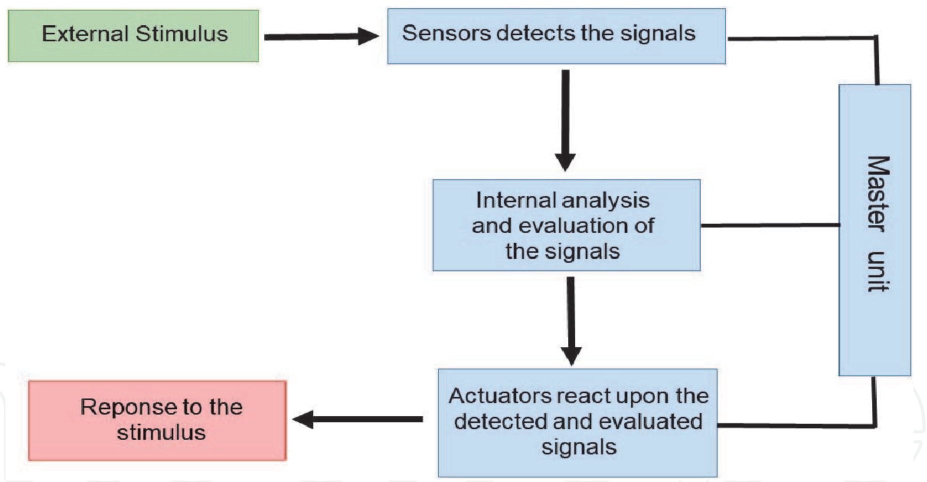

Figure 3.
Working principle of an active smart textile.

textile capable to sens, react, and adapt to the situation based on the learned experience from what it sensed and reacted to previously. Smart textile working principle is described in **Figure 3**.

5.1 Functionalized textile as electrodes

Smart textile is an important alternative which provides a more conformable and user friendly approach for breath monitoring. Textile-based dry electrode were reported to be as efficient and reliable as Ag/AgCl wet electrode [78]. They are made of conductive yarn that can be knitted in plain [79], honeycomb weave patterns [80], embroidered [81], or screen-printed directly onto fabric [82]. Conductive thread electrodes are often integrated into garment and connect to wireless electronic systems for data acquisition and transfer, which make their usage more easy and comfortable. The first smart textile platform based on knitted integrated sensors was proposed in 2005 [83]. The platform was able to acquire simultaneously, and in a natural environment, ECG, breathing activity, posture, temperature, and movement index signals. Later on, many researcher groups proposed smart shirt based on different type of conductive electrodes to measures ECG signal [80, 84], or textile piezoresistive sensor for detection of respiratory activity [85], such as the case for the European project Psyche a personal, cost-effective, multi-parametric monitoring system based on textile platforms and portable sensing devices [86]. However, these type of electrodes exhibit permanent performance degradation and significant reliability issues after repeated washing cycles, poor resistance to strain, and sweat oxidization problems in particular for silver-plated fibers [64, 65]. As a consequence, functionalized polymer-based sensors integrated into textile were proposed as an alternative solution. Weft-knitted strain sensor made from silver coated nylon was used to fabricate a belt, which can be worn around the chest or abdomen to monitor breathing rate [87]. Triboelectric nanogenerators (t-TENGs) was integrated into T-shirt garment by direct weaving of Cu-coated polyethylene terephthalate (Cu-PET) warp yarns and polyimide (PI)-coated Cu-PET (PI-Cu-PET) weft, to fabricate a chest strap for real-time breath monitoring [88]. Conductive silicone strap integrated into garment was also for breath monitoring [89].

5.2 Optical fibers for sensing

The development of smart textile requires further improvement of the functionalities of its fiber component. Recent advances in optical fiber technology did not only revolutionize the worlds of lasers and communications, but also the world of fashion and smart textile. In fact, when optical fibers were embedded into textile composites using either weaving or knitting manufacturing techniques [90], they give more functionalities to the garment such as light emitting for wearable luminous clothing [91], environmental conditions [92, 93], and health-care [94] monitoring. Using optical fiber, the breath monitoring principal is based on the changes of the optical properties of the light passing through the fiber during the expansion and contraction of the thorax and abdomen. These changes are caused by the increases of the fiber losses due to microbending [95], which enable the fiber to be used as a sensor. Flexible polymer optical fibers (POFs)-based sensor have been used previously for breath monitoring [96–98]. The POF-based sensors present some advantages such as the absence of electrical interference, in particular when used during magnetic resonance imaging [95], and good flexibility [99]. Moreover, highly flexible POFs was integrated into a carrier fabric that react to applied pressure to form a wearable sensing system [100]. Other types of optical fibers which have been implemented in the wearable sensor include: fiber Bragg grating (FBG) [101], macro-bending of single-mode fiber [102], and notched side-ablated POF on a fabric substrate [103]. With the FBG, a smart garment featuring 12 FBG sensors to monitor breath and heart rates was proposed and validated with gold standard instrument with both gender [101, 104]. However, this wavelength detection-based technology is very complex, and the fabrication technology of the sensors is very expansive, which jeopardizes the large scale manufacturing. Furthermore, the cable connection between the smart T-shirt and the optical spectrum interrogator reduces the mobility of the user and could create a discomfort when used for a long time.

6. New generation of intelligent textile

Wireless communication is one of the most critical requirement in the development of smart textiles as it eliminates all the mobility restrictions [105]. This communication is assured by an antenna, a key component for any wearable system. Embedding antennas into textile transform the garment into a user-network interface [106]. Several textile antenna designs were proposed for medical applications in the industrial, scientific and medical (ISM) and ultrahigh band, such as inverted-F antenna inkjet printed on fabric [107], flexible and stretchable patch square shape antenna on a 3-D printed substrate [108], and epidermal patch antenna [109]. With the rapid progress on the fabrication of conductive textile, silver yarn was used to create a 2.45 GHz spiral shape antenna integrated into textile and connected to an electronic system for heart rate monitoring, fall detection, and ambient temperature measurement [110]. Although the proposed wearable antennas were enabling radio frequency communication when connected to an electronic board, their weak performances and poor endurance to environmental conditions, such as moisture [111], require further improvements for future applications.

6.1 Development of new antenna

An efficient wearable antenna should fulfill important requirements such as thin thickness, robust, lightweight, and resistant to washing cycles. Moreover, it must be low cost for manufacturing [106]. Following these criteria, a next-generation fiber antennas that lend themselves to RF emission [112, 113] adaptable to existing civilian broadband mobile infrastructures was developed exclusively for real-time breath monitoring when integrated into textile. Four different antenna shapes were designed for a 2.45 GHz emission frequency: a leaky coaxial cable (LCX), a central fed-dipole, a loop, and a half-turn Archimedean spiral fiber. These antennas were fabricated using an inert hollow-core Polyimide-glass fiber of 362 μm diameter functionlized by plating the external and internal surfaces with thin films of conductive nanoparticles. For the LCX antenna, shown in **Figure 4(a)**, the external surface of the capillary was plated with 20 μm copper thin film using electrochemical deposition, and the internal surface was coated with 150 nm of silver layer using a redox chemical reaction [112]. Dipole, loop, and spiral fiber antennas were fabricated using the same hollow-core Polyimide-glass fiber and using the same silver deposition technique of the inner capillary as presented in **Figure 4(b)**. The schematic representations of these antennas are shown in **Figure 4(c)**–**(e)**.

The designed antennas were integrated into textile as shown in **Figure 4(f)** using different techniques. The LCX, dipole, and loop antennas were weaved using a computerized loom, while the spiral fiber antenna was sewed into a textile. The metal-glass-polymer fiber permits the antenna integration into a textile without compromising comfort or restricting movement of the user due to its flexibility.

The emissive properties in free space, and the gain of the designed antennas were investigated both experimentally and numerically, and it was found that the central frequencies of these antennas were around 2.45 GHz, with a gain of 1.76 dBi for LCX, 1.76 dBi for loop, 3.41 dBi for dipole, and 2.37 dBi for spiral fiber antenna [112–114]. For environmental endurance, the used composite metal-glass-polymer fiber shields efficiently the antenna from the external perturbation, and it becomes more sustainable to moisture effects when additional super-hydrophobic coating was added to the antenna external surface and to the surrounding textile [115]. Further studies on the resistance of the antennas against external deformations have shown that the spiral fiber was more sensitive in comparison to the rest of the antennas, with a central frequency shifts of about 360 MHz [113, 114].

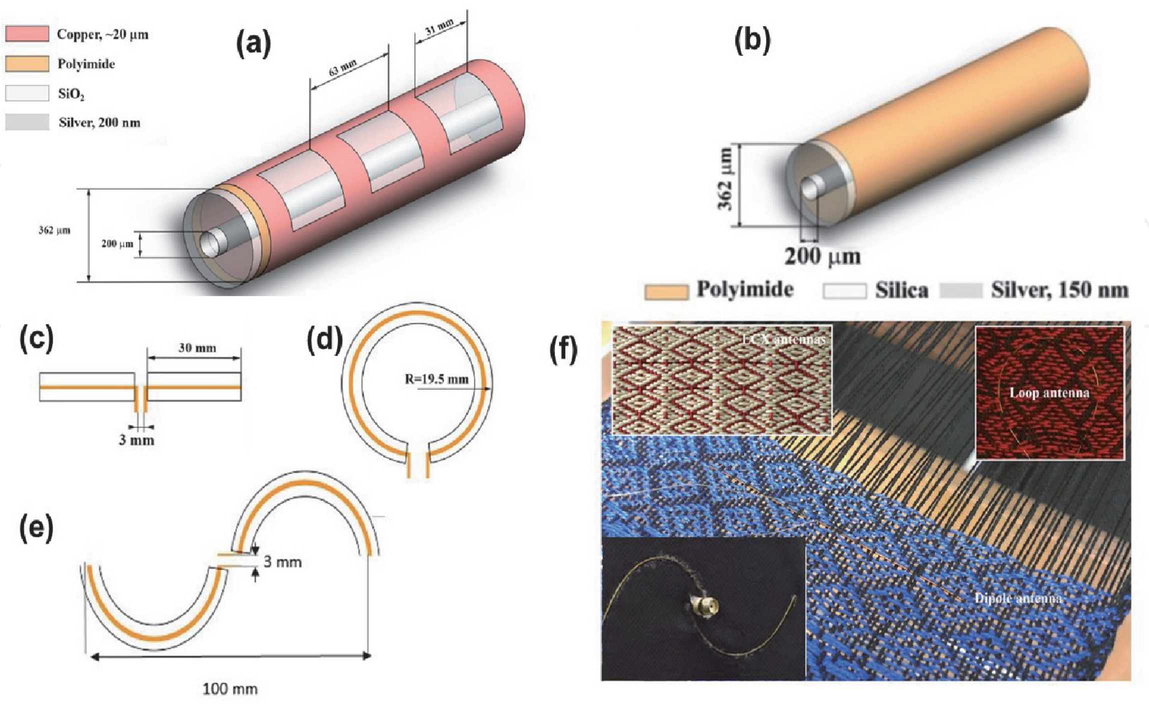

Figure 4.
(a) Structure of the LCX fiber antenna; (b) structure of the polyimide-coated hollow-core silica fiber. Geometry of (c) linear dipole, (d) loop, and (e) spiral fiber antenna fabricated from the conductive multi-material polyimide hollow-core silica fiber and designed for the 2.4 GHz communication network. (f) LCX, loop, and dipole antennas weaved into a cotton fabric (top).

6.2 Wireless communicating textile

The high sensitivity of the spiral shape antenna enables it to perform extra duty in addition to its classical role. Indeed, it has been demonstrated for the first time that when this antenna is integrated into a T-shirt on the chest position (see **Figure 5(a)**), access to breath monitoring was possible through the continuous measurements of its central frequency shift, which is induced by the movement of the chest during breath [114, 116]. As shown in **Figure 5(b)**, the movement of the chest causes deformation to the antenna shape, and the central frequency shift is continuously measured with a vector network analyzer (VNA) and plotted against time as presented in **Figure 5(d)**. A strong correlation between the frequency shifting and the breathing periods were observed [116].

Improvement were introduced to the T-shirt to make it more comfortable for daily use by developing a new smart textile featuring a breath sensor based on Bluetooth communication protocol as shown in **Figure 5(c)**. Breath monitoring was measured from the received signal strength indicator (RSSI) (see **Figure 5(e)**) emitted by the sensor, made of spiral antenna and a Bluetooth transmitter, and detected by a base station. The breathing patterns and rates were estimated after data processing of the RSSI signal using Fast Fourier Transformation (FFT) method [114].

Although the new wireless communication smart textile meets most of the requirements for a reliable and efficient portable system, its capabilities for an accurate monitoring of breath required further development. In fact, detecting breath using only one sensor placed on the middle of the chest is not enough because the breathing mechanism induces the displacement the chest and the abdomen, and the latter was not detected by the smart textile.

To enhance the performances of the smart textile, a new design featuring an array of six sensors was proposed [117] as shown in **Figure 6(a)**. The sensors were

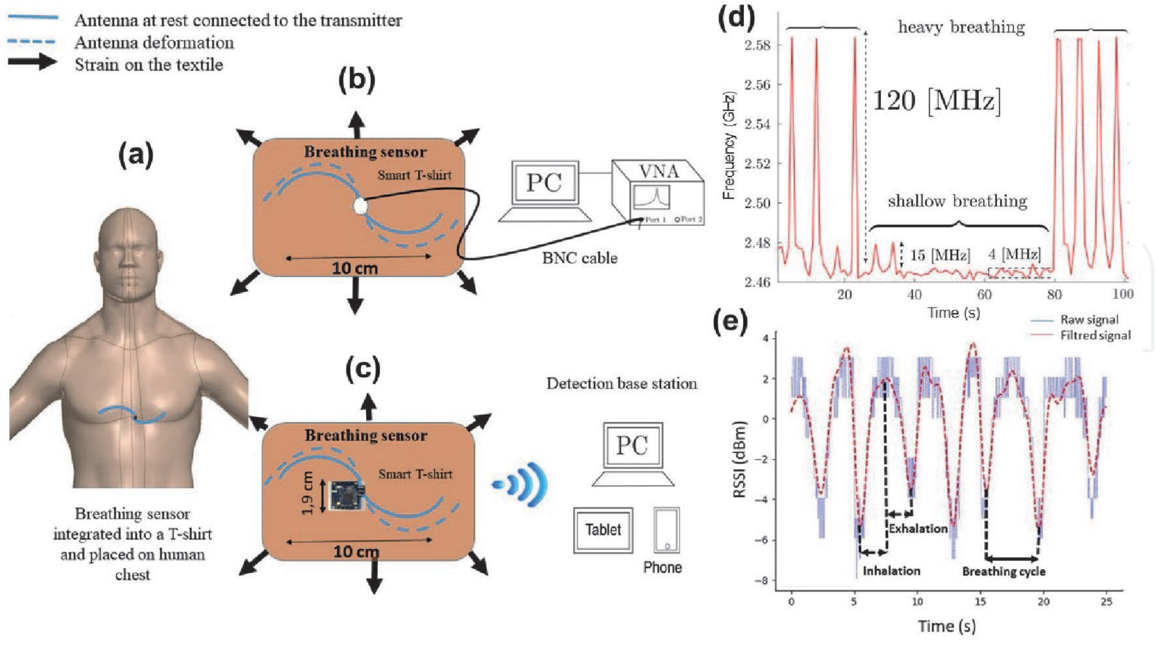

Figure 5.
Schematic representation of the working principle for breath detection: (a) breathing sensor is placed on the chest of human body; spiral antenna configuration change under the stretching load caused by the chest expansion during the breathing; (b) and the induced central frequency shift is measured using a VNA, (c) or through the RSSI signal detected wirelessly a base station. (d) Resonant frequency shift of the multi-material fiber spiral fiber antenna integrated into textile as a function of time during breathing pattern measurements. (e) Breathing signal from received signal strength indicator (RSSI) measurements (blue) and the processed signal (red).

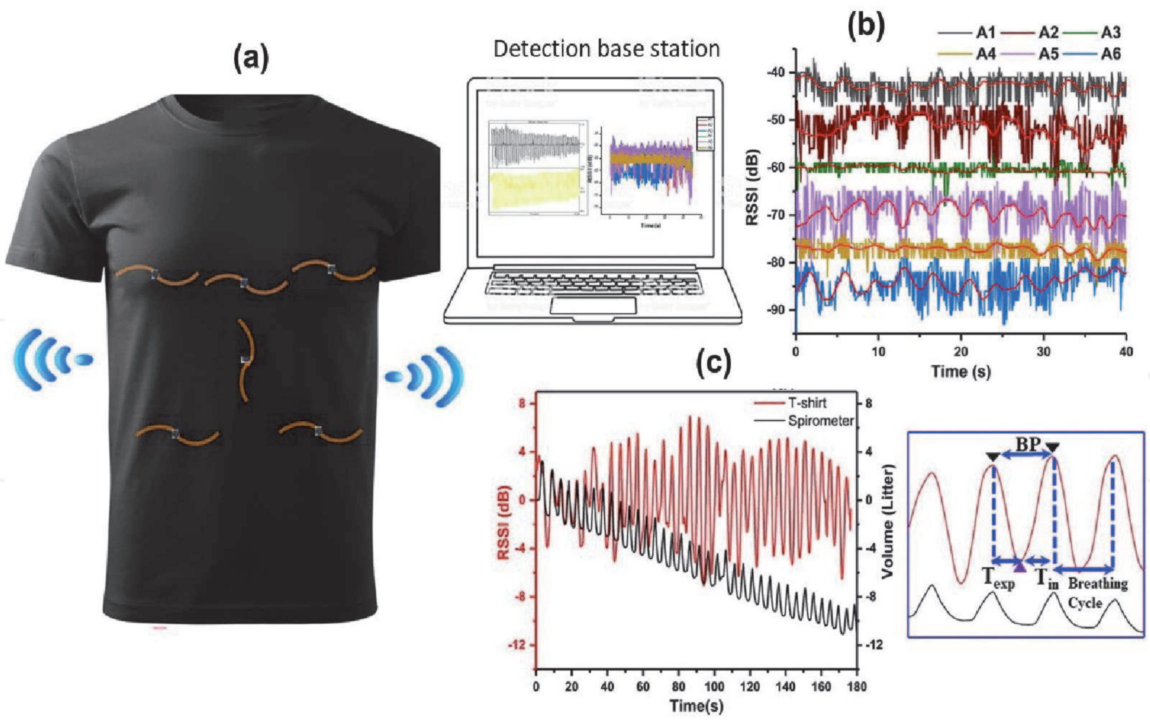

Figure 6.
(a) Breathing sensor array integrated into a stretchable T-shirt together with the portable base station for signal detection; (b) raw and smooth RSSI signals recorded with the smart T-shirt based on the sensor array Ai (i = 1–6); and (c) breathing pattern obtained from the received and filtered RSSI signal recorded using the wireless communication smart T-shirt (red), and the standard reference (black). The inset, the experimental measurements of the breathing parameters.

installed strategically on both left and right sides of the thorax and the abdomen to monitor breathing from both compartments (see **Figure 6(b)**). The emitted six RSSI signals are displayed and plotted in real-time on the detection base station. The breathing pattern were then extracted from the analysis of an RSSI trace based on two criteria: a large variation of the RSSI amplitude, and a clear RSSI signal oscillation. The validation of the new textile was performed by direct comparison with a flow meter used as a medical gold standard reference. A statistical analysis based on Bland-Altman analysis [118, 119] showed a good agreement between the breathing parameters measured with the smart T-shirt and those with a standard reference system. An example of breath-to-breath comparison is displayed in **Figure 6(c)**. These two synchronized signals present similar oscillation of the corresponding signals during the breathing process, and both systems record accurately 37 breathing cycles during the test. These experiments were performed with seven volunteers of different in both standing and seating postures. In addition to breath monitoring, the new textile was also capable to monitor successfully stimulated apnea by stopping breathing for a short period of time during breath detection [117].

7. Data processing and machine learning

Signal processing is a very crucial step to extract breathing patterns and parameters. This step is as important as the development of the wearable acquisition system, and the choice of the processing method can define the quality and accuracy of the output results. Signal processing could be performed either in real time using algorithms integrated into the processing unit of the wearable sensor, or performed off line by applying different transformations on the stored data. Data processing will depends on the sensing technology. For example, in the case of facial tracking sensor [73], image processing techniques were used to enhance the recorded thermal images and to remove unwanted noise, while for the wearable patch sensor network [120], real-time data processing and fusion algorithms integrated into the processing units were used. Most, if not all, processing methods are based on filters. Filters are often used in bio-physiological applications such as for the spectral contents of ECG signals [64], or for vital sign detection such as RSSI signals [114]. For breath signal, the spectral frequency range is well known, and signal filters can be applied to the raw data to minimize high-frequency noise while preserving the breathing envelope signal. A band-pass filter with cutoff frequencies of 0.05 and 1.9 Hz is applied to compensate for possible drifts and to reduce the noise level in the signals [64]. FFT algorithms, that converts the signal from the time domain to the frequency domain and vice versa, are then applied to extract the spectral features such as breathing rate.

Machine learning was proposed as another alternative method for breath monitoring. This method is based on pattern recognition and capabilities of computer machines to learn and execute a task. For instance, Convolutional Neural Network (CNN) method was used in identifying asynchronous breathing [121]. The results were very satisfactory (sensitivity of 98.5% and specificity of 89.4%) when trained with different amount of training data sets and different types of training data sets. Deep neural network (DNN) is another methods used for the firs time to improve phone recognition [122]. It has been demonstrated that when the DNN-based fine-grained algorithm is integrated into smartphone for acoustic recognition, and the breathing rate monitoring was achieved with the same professional-level accuracy [123]. The main advantages of the CNN and DNN methods are both robustness and precision if trained with sufficient training data, and become more attractive when combined with wearable sensors for health monitoring.

8. Conclusions and perspectives

In this chapter, we have covered two major aspects: the importance of breath monitoring as a prevention method for early detection of several diseases; and the most recent achievements in hybrid material science, enabling the development of new noninvasive and flexible wearable sensors for breath monitoring. We have reviewed the working principle of these sensors with the requirements needed for industrial and clinical acceptance. When the functionality of fibers is augmented through a combination of chemical processes, their integration into fabric give arise to a new platform for wearable sensors called smart textile. The examples presented in this chapter illustrate the recent applications based on smart textile for breath monitoring in which the fabric is enriched with new functionalities while maintaining the mechanical properties and the comfort of the user. Wireless communication using smart textile based on multi-material fiber antenna is the most recent achievement in technological development in which the capability for RF emission at 2.45 GHz and monitoring breath in real time is assured by new generation fiber antennas. The advantages of such a technology are its endurance to the environmental changes and the reliability of the measurements. At this stage, the potential use of smart textile technology has not yet fully explored. We believe that future research will be focused on developing highly flexible materials sensitive to electrical, chemical, and physical changes with high endurance to the environmental conditions, capable to integrate textile for health-care monitoring. At this level, it will be important to develop alternatives for the powering, communication, and signal processing of such systems, in particular if someone would like to deploy such technology in the northern regions. Improving the power efficiency generated through thermoelectric, piezoelectric, and photoelectric effects could pave the road for new powerless system. Machine learning methods have shown lot of success on predicting events with up to 96% of sensitivity. This method could be of an important use for data processing and extraction of biopotential and vital signs.

Author details

Mourad Roudjane[1*†] and Younès Messaddeq[1,2†]

1 Center for Optics, Photonics and Lasers (COPL), Department of Physics, Université Laval, Québec, Canada

2 Instituto de Química, UNESP, C.P, Araraquara, SP, Brazil

[*]Address all correspondence to: mourad.roudjane.1@ulaval.ca

[†]M.R wrote the chapter and Y.M revised and approved the text.

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