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The role of optical fiber sensors in the new generation of healthcare devices: a review

Arnaldo Leal-Junior, * Jussara Silva, Leandro Macedo, Arthur Marchesi, Samilly Morau, Janine Valentino, Fabricyca Valentim and Magno Costa

This paper presents a review of optical sensor systems for wearable applications aiming at the new demands on healthcare motivated not only by the new paradigms in internet of things, but also in photonics development and artificial intelligence algorithms. In this context, the overview of musculoskeletal disorders and the role of wearable sensor systems in such applications are discussed. In addition, there is a comprehensive discussion of the components of wearable sensor systems with the novel developments and approaches in different wearable applications. Thus, the optical fiber sensor developments, approaches and applications are discussed for their use in smart textiles, biosensors and intrusive applications. Moreover, new developments on power supplies are discussed aiming at self-powered sensor systems. Therefore, this review paper can aid in the development of the new generation of wearable sensor systems in healthcare applications using optical fiber sensors and general optical based sensors, which can overcome or mitigate the shortcomings of conventional sensor technologies.

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1 Introduction

The internet of things (IoT) and sensor market have witnessed exponential growth in recent years.¹ IoT refers to the interconnected network of physical devices, vehicles, appliances, and even buildings, all embedded with sensors, software, and other technologies, allowing them to collect and exchange data.² This data-driven ecosystem has found applications across various sectors, including healthcare, agriculture, manufacturing, and smart cities, to name a few.³

Sensors play a pivotal role in the IoT scenario, serving as the eyes and ears of this interconnected world.⁴ These devices can capture a wide array of information, such as temperature, humidity, pressure, motion, and more, providing real-time data that can be analyzed and acted upon.⁵ The sensor market has expanded significantly to cater to the growing demands of IoT applications. Innovations in sensor technology, including miniaturization, increased sensitivity, and reduced power consumption, have enabled the proliferation of IoT devices and applications.⁶

As a vital application in IoT context, digital health applications are currently expanding, which comprise developments on novel technologies for healthcare, especially remote healthcare and smart sensors and actuators for monitoring health conditions and mitigating locomotor disorders.⁷ The population aging throughout the years, where

there is an exponential growth in the elderly population (over 65 years), motivates such growth in digital healthcare.⁸ The global demographic transition presents fresh challenges across various domains. From an economic standpoint, the growing elderly population heightens the strain on pension systems, particularly when coupled with a diminishing ratio of elderly individuals to those in the working-age bracket.⁹ Additionally, there is a distinct healthcare challenge associated with the aging population, characterized by age-related conditions such as immunosenescence, urological changes, and sensory impairments, encompassing issues like hearing loss, declining visual acuity, and degraded vestibular function.¹⁰

The technology of sensors, wireless communications, and data analysis has made significant advancements in recent years. As a result, the assessment of users' health is no longer limited solely to the clinical environment.¹¹ This enables the monitoring of various physical and physiological parameters of patients not only in clinics but also at home, where continuous monitoring of human activities is desirable for remote care, including diagnosis, emergency transportation, and monitoring the patient's health status.¹² Concurrently, there is a growing development of advanced materials resulting in 2D materials and heterostructures and metamaterials capable of responding to different optical, electromagnetic, chemical, thermal, and mechanical stimuli.¹³ In this context, intelligent photonic textiles have been proposed for user monitoring using a fully wearable and portable technology. From a certain perspective, these photonic textiles, when incorporated into clothing, can also

Federal University of Espírito Santo, Fernando Ferrari Avenue, Vitória, ES 29075-910, Brazil. E-mail: leal-junior.arnaldo@ieee.org



be considered multi-purpose devices (or structures) because they can simultaneously act as clothing and sensors.¹⁴ It is also important to mention that optical fibers are increasingly employed in the development of biosensors, through which various biomarkers, hormonal parameters, and physiological parameters, in general, can be measured. Thus, the use of artificial intelligence (AI) and cloud computing, combined with optical fiber sensing technology, can result in real-time, accurate, secure, and intelligent decision-making systems within the field of healthcare.¹⁵

Considering the wearable technologies for healthcare applications, there are a number of components to be analyzed and designed.¹⁶ To that extent, the role of optical fiber sensors in wearable devices as well as in the new generation of healthcare applications and human life improvements is not only limited to telecommunications and local passive optical networks, but also in sensor technologies and their integration into different structures of everyday life, including accessories,¹⁷ textiles¹⁸ and home devices.¹⁹ In this case, the components for a specific application generally involve the transduction element, the signal acquisition and conditioning circuits, the batteries and the system integration for the desired application,²⁰ where the latter can be intrusive applications or unobtrusive ones without the same restriction on biocompatibility. The contribution of this review is the overview and comparison of all components in an optical fiber sensor system, including not only the optical fiber sensing approach, but also the functionalization techniques. In addition, the review includes the discussion of the applications as well as the possibilities of textile integration and novel intrusive applications, including ingestible sensors. As another contribution, this review also includes the energy harvesting approaches with the possibility of achieving a novel self-powered optical fiber sensor system for biomedical applications.

This paper presents a review of the wearable sensor technologies focused on optical fiber sensors for healthcare and physiological parameter assessment. This review includes fundamental aspects on optical sensor technologies, including applications, integration into components, functionalization and energy harvesting. After this introduction in section 1, the remainder of this review paper is divided as follows. Section 2 presents an overview of the population aging and the main musculoskeletal disorders, which lead to the main demands and opportunities in wearable and healthcare sensor technologies. Then, section 3 presents optical fiber sensors with their main technologies, transduction mechanisms and implementation scenarios. Section 4 discusses the functionalization of sensor devices for the assessment of different physiological parameters. Batteries and new energy harvesting technologies are presented in section 5 as a possibility of the development of a new generation of self-powered sensor devices. Thereafter, sensor integration into textiles and components is discussed in section 6 with the latest advances in such areas. In another application scenario, biosensors (including intrusive

applications and ingestible sensors) are discussed in section 7. Finally, conclusions and future work suggestions are presented in section 8.

2 Overview of population aging and wearable sensor applications

Aging is a natural process involving changes characterized by physical and cognitive modifications resulting from the passage of time.²¹ The bodily changes occurring in this process lead to the deterioration of sensory (visual, vestibular, and somatosensory), musculoskeletal, and neuromuscular systems.²² These aspects can result in a decreased capacity to adapt to functional overloads, reflecting in postural control, which may predispose to fall occurrences.^{23,22}

The World Health Organization (WHO) states that the elderly population represents 20% of the global population, and the number of elderly individuals (above 65 years old) is projected to reach 1.5 billion by the end of 2050.²⁴ This fact is associated with the progressive advancement of preventive medicine, consequently leading to an extended life expectancy. However, along with increased life expectancy comes a rise in chronic-degenerative diseases, with their metabolic, physical, and emotional consequences, directly impacting the functionality, autonomy, and quality of life of many seniors.²⁵

Therefore, there is an urgent need for the development of public policies and technological resources that directly contribute to the improvement and maintenance of the health of this population segment.²⁶ Geriatric physical therapy is a specialized field that examines movement and functionality in elderly individuals. Through clinical assessments, it is capable of testing and quantifying different types of alterations stemming from kinematic functional disorders.²⁷

However, test protocols are currently carried out manually, relying on the evaluator's expertise. Moreover, the lack of quantitative data diminishes the precision of protocol outcomes.²⁸ As a result, there arises a need for sensory technologies capable of quantifying data and expressing more accurate and reliable results.

Considering that the elderly present greater physical and motor limitations, the risks of domestic accidents also increase. Approximately 35% of individuals (above 65 years old) experience one or more falls per year, which can lead to severe injuries and hospitalization with substantial costs to public health.²⁹ In this context, wearable sensor technology can aid in improving the overall quality of life for the elderly by remotely monitoring physiological signals, seeking to mitigate the consequences and sequelae of falls.^{30,31}

The primary advantage of wearable systems is their ability to collect data outside of laboratory and clinical environments.³² Hence, such systems are viable for analyzing fall detection and other physiological changes like blood pressure, heart rate, oxygen saturation, etc.^{31,33} In this regard, the IoT presents a promising solution to provide continuous,



objective, and holistic monitoring, alleviating the burden not only on human caregivers but also reducing the risk of alterations related to physiological parameters, thus aiding in clinical decision-making.³⁴

The increase in the elderly population and life expectancy, along with the consequent rise in the prevalence of chronic diseases, necessitates improvements in healthcare aimed at the well-being of this population.³⁵ Although there are many assessment protocols aiding in prevention, disease diagnosis, and rehabilitation, almost all of them are conducted manually and based on qualitative data, which diminishes result accuracy.²⁸

One of the leading causes of elderly hospitalization is falls. Falls are considered a serious public health issue, potentially resulting in severe injuries such as fractures, mobility changes, functionality alterations, and fear of falling again.³⁶ One validated approach in the literature to mitigate harm and prevent falls is the assessment of the elderly's fall risk using the timed up and go test experimental protocol.³⁷ Although a simple and widely used test by physiotherapists to assess mobility and fall risk in the elderly, this protocol is executed manually, potentially leading to low analysis precision and the loss of important kinematic parameters.³⁸

Thus, wearable devices capable of conducting quantitative analyses of health assessment protocols, with automatic, portable, remote data collection and temporary data storage, as well as data transmission, can contribute to better physiological monitoring of the elderly and, consequently, improve preventive health measures and aid in clinical diagnosis and rehabilitation. Fig. 1 shows an illustration of the role of wearable sensors in healthcare applications concerning the human being aging. These sensors enable the measurement of vital signs that can be used to automate healthcare protocols and provide data-driven decisions aiming to enhance the efficiency of medical treatments and the patient's recovery.

3 Flexible optical fiber sensor development

A flexible optical fiber sensor uses optical fibers as the sensing element and is designed to be flexible and bendable. These fibers can be easily shaped and molded to fit into tight spaces or conform to curved surfaces.³⁹ The use of these sensors is common in applications where traditional sensors are not suitable for use due to their rigidity or size limitations. Physical and chemical properties can be

monitored in harsh, underground, or subsea environments using flexible optical fiber sensors withstanding high temperatures, chemical corrosion, and electromagnetic interference.⁴⁰

In order to manufacture optical fibers, a variety of materials can be used, including silica glass, polymers, multimaterials (combinations of materials, such as silicon and silica, metals and silica, and polymers and chalcogenides, for example),⁴¹ and biological materials (spider silk⁴²). Polymers and silica glass are the most common materials.⁴³ There are distinct advantages and limitations to using silica glass fibers and POFs for the development of flexible optical fiber sensors. Due to their low attenuation, high transparency, and chemical inertness, silica glass fibers are ideal for high-performance optical sensing applications and sensor data transmission over long distances. The brittle nature of these materials, however, poses a challenge for applications that require intricate bending or shaping. On the contrary, POFs, such as those made from materials such as poly(methylmethylesiloxane) (PMMA), are highly flexible and biocompatible, making them suitable for applications requiring comfort and conformability, such as wearable health monitoring sensors. Furthermore, polymer fibers are easier to handle and fabricate than silica glass fibers, which facilitates the design of more versatile sensors. The attenuation of polymer fibers may be greater and their thermal stability may be lower than that of silica glass fibers, which may adversely affect their performance in certain sensing applications. Therefore, it is important to balance the trade-off between optical performance, mechanical flexibility, and ease of fabrication for flexible optical fiber sensors when choosing between silica glass and POFs.^{43–46}

A multimaterial approach to optical fiber development offers a pathway for improved performance, increased functionality, and customization. Multimaterial optical fibers can achieve improved transmission efficiency, expanded bandwidth, and reduced signal loss by combining different materials with complementary properties. With these fibers, multiple functionalities can be integrated into a single structure, enabling applications such as fiber amplification and nonlinear optical processes. Designers are able to tailor the properties of multimaterial fibers in order to meet specific application requirements, such as controlling the refractive index and managing dispersion. Multimaterials foster innovation in optical fiber technology, enabling new functionalities in optical communication, sensing, and high-energy lasers by pushing the frontiers of performance and enabling new material combinations.^{47,48}

3.1 Intensity variation sensors

The working principle of optical fiber sensors based on intensity variation consists of the measurement of the intensity of transmitted or reflected light. This is done when the optical fiber cable is subjected to a stimulus such as temperature,

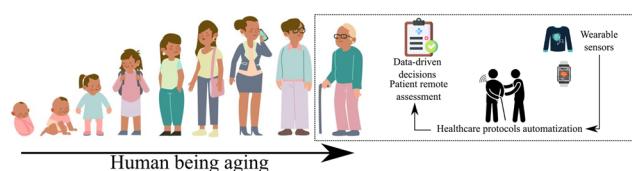


Fig. 1 The role of wearable sensors in healthcare applications concerning the human being aging illustration.



strain, or chemical reaction.⁴⁵ These external parameters modulate the analyzed output signal and correspondences can be established between the monitored parameter and the intensity of the transmitted or reflected light in sensing applications.⁴⁵ They can be used for different applications, including structural health monitoring,⁴⁹ biosensor development,⁵⁰ and healthcare applications.⁵¹ Rocha, Matilde, *et al.* have attached POFs to a handspring, and the sensor's force characterization was evaluated with a tensile tester. The sensor's sensitivity was characterized by force variations caused by decreasing the distance between the two handles of the handspring. At each step, the voltage (correlated to optical power intensity) was collected using an Arduino. The test was performed for the decrease of the distance between the two handles, and the voltage was collected at each step. The respective intensity of the force applied to the handspring handle was recorded, and the voltage was correlated to the optical power intensity variation.⁵² In another application, plantar pressure sensors were designed using POFs integrated into a commercial insole. The optical power intensity was characterized with respect to load application for monitoring plantar pressure distribution for ground reaction forces monitoring during the gait.⁵³

Intensity variation-based optical fibers can also be multiplexed for quasi-distributed sensing applications. The use of time-division multiplexing (TDM) of the sensors in ref. 54 is one possible approach, where the sensors are placed at different distances from the source and detector, so that, upon input, a single pulse of the appropriate duration produces a series of distinct pulses. As the sensors' outputs are interleaved in a time sequence, these pulses represent time samples of their outputs. The required duration of the input pulse is determined by the effective optical delay of the fiber connecting the sensor elements. With the repetitive pulsing of the system, each sensor can be addressed by time-selective gating of the detector output. There is also the possibility of side-coupling the light source in a POF (POF) sensor with a lateral section, which is reported in ref. 55. As a result of this approach, each lateral section is an intensity variation-based sensor, and each sensor has its own source of light. Each end of the fiber is equipped with two photodetectors. Light-emitting diodes (LEDs) are activated by microcontrollers, and signals are acquired when the LEDs are active. In order to avoid simultaneous activation of two or more light sources, the LED array is activated according to a predetermined sequence and frequency. Consequently, each LED is turned on one at a time. When the corresponding LED is active, the signal from each photodetector is acquired. This results in a matrix with $P \times D$ vectors, where D is the number of LEDs and P is the number of photodetectors. As a result, it is possible to decouple the effects of different parameters on each lateral section of the fiber. Experimental validation of the sensor has been conducted for three degrees of freedom applications involving simultaneous measurements of temperature, force, and angle. Fig. 2 shows a schematic representation of the working principle of

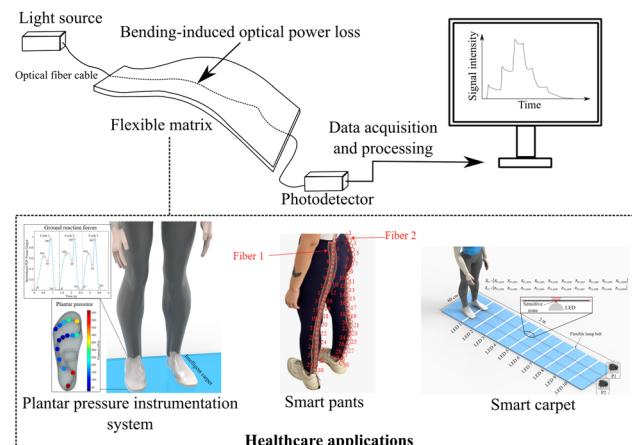


Fig. 2 Intensity variation-based optical fiber sensors' working principle schematic representation and health care application examples. Reproduced from ref. 19, 51 and 56 with permission from Optica Publishing Group, copyright 2023, MDPI AG, copyright 2019, MDPI, and MDPI AG, copyright 2019, respectively.

intensity variation-based optical fiber sensors. For healthcare applications, the optical fiber sensor is usually embedded into a flexible matrix responsible for protecting the cable and inducing optical loss when the sensing region is bent or stretched (or coupling LEDs in different regions along the optical fiber cable). These optical power losses modulate the output optical power intensity and they can be characterized according to the desired measurand. This technique can be employed in healthcare applications such as plantar pressure monitoring,⁵¹ smart pants for pose classification and movement intensity measurement,⁵⁶ and smart carpet for gait analysis.¹⁹

3.2 Fiber Bragg gratings

Uniform fiber Bragg gratings (FBGs) are periodic perturbations of the refractive index along the optical fiber length. They are formed by exposing the fiber core to an intense light interference pattern. The resulting permanent grating acts as a narrowband reflector, reflecting a specific wavelength of light while transmitting all other wavelengths.⁵⁷ FBGs have found key applications in routing, filtering, control, and amplification of optical signals in high-capacity wavelength division multiplexing telecommunication networks.⁵⁸ They are also used in fiber optic sensors and quasi-distributed thermophysical measurements.⁵⁹ FBG sensors can be utilized in the design of flexible optical fiber sensors by encapsulating them with protective materials to ensure their functional integrity and survivability under harsh conditions or by engraving the Bragg gratings in POFs.^{60,61} These sensors can be used in structural health monitoring of civil infrastructure, such as bridges, tunnels, and buildings,⁶² mechanical deformation and temperature fluctuation monitoring in asphalt pavement,^{63–65} temperature and dynamic strain in electrical machine monitoring,⁶⁶ real-time monitoring of the curing process of resin-based carbon



fiber reinforced plastic and aluminum alloys,⁶⁷ and measurement of deformation accumulation and elastic recovery of materials in asphalt roads.⁶⁸ The reflected wavelength (known as Bragg wavelength) is proportional to the grating period (distance between subsequent gratings) and the effective refractive index, where these parameters change when the optical fiber is subjected to strain and/or temperature variations where a relationship between the Bragg wavelength and the monitored parameter is established in sensing applications.⁶⁹

In addition to uniform FBGs, there are non-uniform gratings such as long-period fiber Bragg gratings, chirped fiber Bragg gratings, phase-shifted fiber Bragg gratings, and tilted fiber Bragg gratings (TFBGs).⁷⁰ TFBGs are a type of fiber optic sensor that consists of a fiber Bragg grating with a certain angle between the grating plane and the fiber axis. This angle causes the grating to have a different structural geometry along the radial, azimuth, and axial orientations.⁷¹ This makes TFBGs a candidate for multifunctional fiber-optic components with distinct characteristics. In TFBGs, tilting the Bragg grating causes non-uniform spatial compression (or elongation) of the grating pitch. This is manifested as a decrease (or increase) in the tilt angle of the grating plane.⁷² Due to this non-uniform spatial compression, the spacing of the grating plane changes along the fiber cross-section, with the minimum occurring in the center. Due to the non-uniform tilt plane modulation and the stress-induced refractive index variation along the fiber cross-section, the phase matching condition for each pitch value is partially disturbed.⁷⁰

TFBGs have a significantly weaker mode coupling between the core and the cladding, and the cladding mode coupling changes significantly, allowing high-sensitivity bending response measurements to be conducted in the cladding mode.⁷³ The resonance wavelengths of TFBG core and cladding modes are sensitive to temperature, while their transmission power is temperature-independent. The transmission power of cladding modes, however, is bending-dependent, while the resonance wavelengths of both the core mode and cladding mode are bending-sensitive. As a result of these unique properties of TFBGs, it is possible to discriminate between bending and temperature effects simultaneously.⁷⁴ TFBGs can be used as sensors in various applications such as structural health monitoring,⁷⁵ biomedical sensing,⁷⁶ and chemical sensing.⁷⁷ Similar to uniform FBGs, TFBGs can be used as flexible optical fiber sensors encapsulating the TFBG sensors in flexible structures ensuring sensor integrity while the bending resistance is increased.^{78–80} In these applications, the higher sensitivity to bending compared to uniform FBGs is a key feature in high-sensitivity flexible optical fiber sensors based on TFBGs, and it can be employed for real-time monitoring of human breathing,⁸¹ hydrogen sensors,⁸² and biochemical sensors.⁸³ Fig. 3 shows the FBG working principle and some applications. In these sensors, part of a broadband incident light coupled into an optical fiber cable is reflected by the Bragg gratings. The reflected wavelength (called Bragg wavelength) is affected by variations in strain and temperature.⁸⁴ For sensing

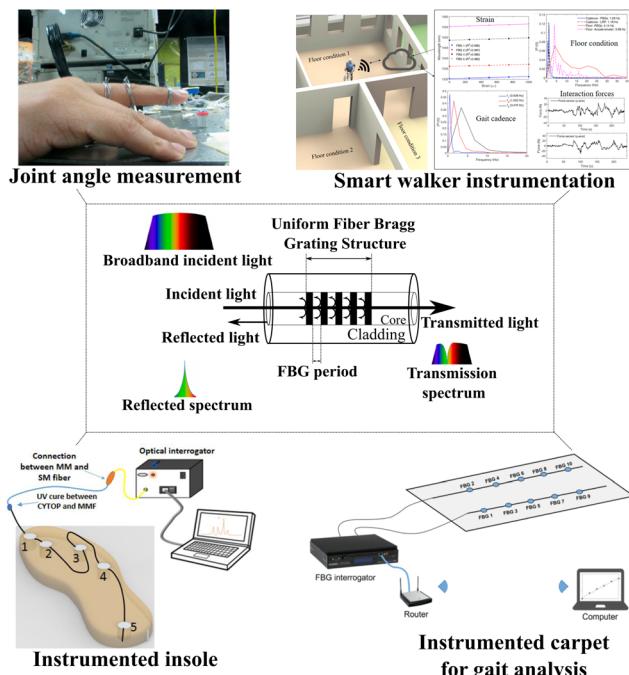


Fig. 3 FBG working principle and healthcare applications using FBG examples. Reproduced from ref. 85–88 with permission from IEEE, copyright 2023, MDPI AG, copyright 2014, MDPI AG, copyright 2017, and IEEE, copyright 2019, respectively.

purposes, FBG sensors can be embedded in different structures in healthcare applications for gait analysis,⁸⁵ joint angle measurement,⁸⁶ pressure plantar monitoring,⁸⁷ and smart walker instrumentation⁸⁸ as examples.

3.3 Interferometers

The most common interferometer techniques in optical fibers include the Michelson interferometer, Mach-Zehnder interferometer, Fabry-Perot interferometer, and Sagnac interferometer.⁸⁹ In the Michelson interferometer, an input light signal is split into two different paths by a splitter (coupled into two separate optical fibers, for example). These paths contain mirror elements, such as Bragg gratings. The reflected signals of the two paths are recombined, creating an interference pattern. This interference pattern is affected by external perturbations such as temperature and strain variation. It can be characterized according to the interest parameter and used as a sensor.^{90,91} The working principle of the Mach-Zehnder interferometer based on optical fibers consists of splitting a light signal into two different paths by means of splitters, where one arm of the interferometer serves as a reference and the other is utilized as the sensing element. The split signal is then recombined and an interference pattern is created. This interference pattern is affected by external perturbations applied to the sensor arm. It can be used for sensing applications through the characterization of the interference pattern with respect to the variation of the monitored external parameter.^{92,93}



In optical fibers, Fabry-Perot interferometers work by interfering with light waves that are reflected back and forth between two parallel mirrors (such as two Bragg gratings separated by cavities). In the Fabry-Perot cavity, light from a source is transmitted through an optical fiber, where it is reflected between the two mirrors. An optical power detector receives the reflected light through the fiber. During the reflection of light waves, constructive and destructive interference produces the interference pattern. As the cavity length can be altered by environmental factors, such as temperature or strain, the Fabry-Perot interferometer is suitable for use as a fiber sensor.^{94,95} Sagnac interferometers are based on the Sagnac effect, which is the phase shift of light propagating in a rotating reference frame. This interferometer splits a light beam from a source into two beams that travel in opposite directions around a closed loop, typically a fiber optic coil that maintains polarization. As the two beams recombine at a detector, the phase difference between them produces the interference pattern. The two beams have the same phase when the loop is stationary, but when the loop rotates, the phase of one beam is shifted relative to the other beam due to the Sagnac effect. As the phase shift is proportional to angular velocity, the Sagnac interferometer can be used as a rotation sensor.⁹⁶ Moreover, external perturbations imply a phase

shift of the transmitted optical signal and it can be used for monitoring external parameters in sensing applications.^{89,97,98} For all aforementioned interferometers, the optical fiber cables can be attached to flexible structures to develop flexible optical fiber sensors for temperature monitoring,⁹⁹ mechanical vibration sensors,¹⁰⁰ and breathing sensors,¹⁰¹ as examples. Fig. 4 shows a schematic representation of the optical circuit of Mach-Zehnder, Michelson, and Fabry-Pérot interferometers in optical fibers. Furthermore, by analyzing the modulation of the interference pattern for external perturbations, these techniques can be used in the monitoring of respiration,¹⁰² the arterial pulse wave sensor,¹⁰³ the monitoring of human physiological activity¹⁰⁴ and the detection of thrombus,¹⁰⁵ as examples (shown in Fig. 4).

3.4 Distributed optical fiber sensors

In optical fiber distributed sensing, the most common techniques are based on Rayleigh, Brillouin, and Raman scattering. They use different reflectometry and demodulation schemes, including optical time-domain and optical frequency-domain reflectometry.¹⁰⁶ Rayleigh scattering is based on the inhomogeneity of the fiber medium, causing localized variations of density which result in fluctuations of refractive index. It is an elastic scattering process, which means that scattered light has the same frequency as the incident light. The intensity of the backscattered light is proportional to the local refractive index of the fiber, which is affected by external perturbations such as temperature, strain, and pressure. By analyzing the backscattered light, it is possible to determine the location and magnitude of perturbations along the fiber.¹⁰⁷ Raman scattering occurs when light interacts with the vibrational modes of a material, resulting in a shift in the frequency of the scattered light. In optical fiber sensing, Raman scattering can be used to determine the location of an event by measuring the arrival time of the scattered light. Raman scattering is affected by temperature and strain perturbations. This results in a temperature and strain-dependent scattering cross-section and spontaneous Raman scattering that can be used for distributed sensing.¹⁰⁸ When light interacts with acoustic waves in a material, Brillouin scattering occurs, resulting in a shift in the frequency of the scattered light. In optical fiber sensing, Brillouin scattering can be used to determine the Brillouin frequency shift by measuring the central frequency of the Brillouin scattered light, which is linearly dependent on both strain and temperature. By measuring the scattered signal time, the external perturbation can be localized along the optical fiber length.¹⁰⁸

Optical-frequency domain reflectometry (OFDR) works by sending a probe signal into an optical fiber and measuring the backscattered light as a function of frequency. It is possible to obtain information about the optical fiber by analyzing the frequency-dependent backscattered light, such as the location and magnitude of any changes in the fiber's refractive index or scattering properties. In this manner, high

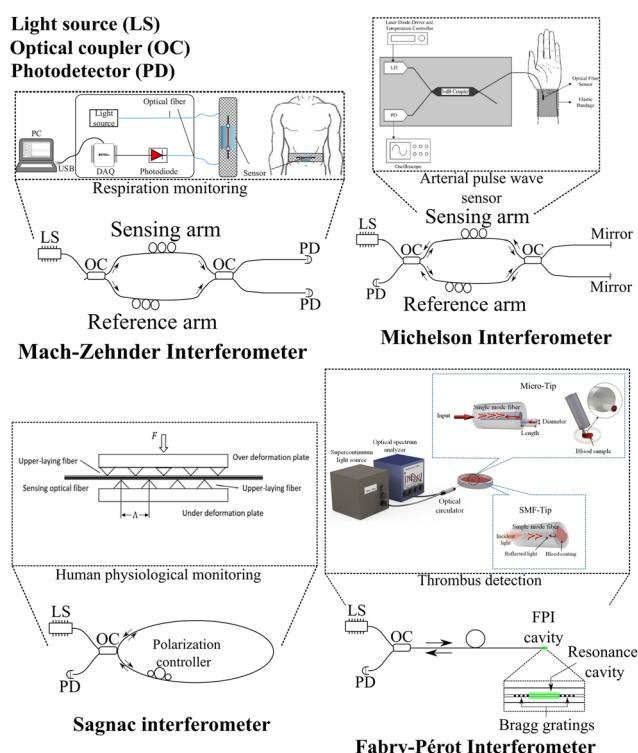


Fig. 4 Optical circuit representation of Mach-Zehnder, Michelson, Sagnac, and Fabry-Pérot interferometers in optical fibers and healthcare application examples using these techniques for sensing applications. Reproduced from ref. 102-105 with permission from Optica Publishing Group, copyright 2019, John Wiley and Sons, copyright 2010, IOP publishing group, copyright 2021, and MDPI AG, copyright 2023, respectively.



spatial resolution and sensitivity can be achieved with distributed sensing along the length of the fiber. OFDR has many applications, including monitoring structural health, measuring temperature and strain, and detecting and locating fiber faults.¹⁰⁹ The working principle of optical-time domain reflectometry (OTDR) for distributed optical fiber sensing is based on the observation of coherent Rayleigh backscattering from a sensing fiber in the time domain. The backscattering field from each scattering element interferes with the receiver when light from a coherent pulse reaches the scattering elements. This backscattering field results in a distributed coherent speckle pattern whose local phase and intensity depend on local disturbances when measured over the entire fiber length. As a result of analyzing backscattered light changes, it is possible to detect and locate disturbances along the length of the fiber.¹¹⁰ Transmission-reflection analysis (TRA) is another approach to distributed sensing with optical fibers. Using an unmodulated continuous wave light source, it measures the transmitted and backscattered optical powers directly. In distributed monitoring systems, this technique is used to track single perturbations. The TRA method is simpler and less expensive than other methods including OTDR and OFDR, which require modulated light sources and require fast electronics to achieve high resolution.^{111,112} Fig. 5 shows a schematic representation of the working principle of distributed sensors in optical fibers. In these sensors, part of incident light is scattered and this reflected signal can be used to measure external perturbations. They can be employed in different healthcare applications, such as upper limb mechanical perturbation detection,¹⁸ liver tissue temperature distribution measurement,¹¹³ and shape-sensing guiding system.¹¹⁴ Table 1 summarizes the key information about the optical fiber sensors employed in healthcare settings discussed in this chapter.

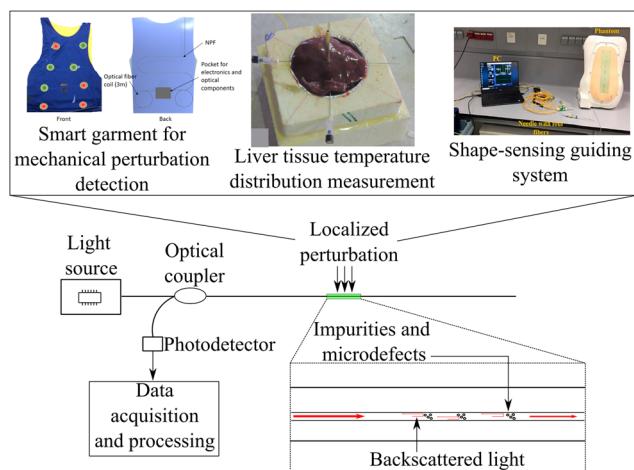


Fig. 5 Optical fiber distributed sensor working principle schematic representation and examples of healthcare applications. Reproduced from ref. 18, 113 and 114 with permission from IEEE, copyright 2020, IOP Publishing Group, copyright 2018, and Elsevier, copyright 2024, respectively.

4 Optical biosensor functionalization

Generally, as the next step after the definition of the transduction mechanism to be used (among the different ones in optical fiber sensors¹²¹), functionalization plays a key role on biosensor development.¹²² In this case, the functionalization methods and common approaches to achieve higher selectivity and sensitivity on the biosensors are discussed below.

Surface chemical functionalization is one of the most common strategies to modify the surface of materials used as a sensor. It is a simple and effective approach to altering the surface properties of a material to achieve a specific goal.¹²³ The success of a biosensor in detecting a specific analyte will depend on how effective its surface functionalization process is. This means that both the adhesion mechanism of a receptor on a substrate (transducer) and the selectivity provided by it are fundamental steps in the functionalization of a surface and in the construction of a sensory device.

Briefly, the functionalization of a surface to act as a biosensor has important aspects, such as the orientation of the receptor, the inert area, and the total area. When united, these aspects result in fundamental parameters in the detection of an analyte. The first one is the sensitivity, which is defined as the variation of the sensory signal as a function of the variation of the analyte; and the second is selectivity, which is defined as the difference in response under a variety of different analytes.¹²⁴

Plasma-based biosensing can be performed either by means of a frequency associated with the surface plasmon resonance in the presence of an analyte (thin film-based biosensors), or by the intensity of the signal associated with the analyte (nanostructure-based biosensors).^{124,125} These plasmonic nanostructures can be obtained by different approaches, namely: top-down and bottom-up. Considered the most expensive and time-consuming, the top-down approach leads to the fabrication of periodic nanoarrays by more complex techniques, such as lithography, laser ablation, ion milling, and chemical etching. In other words, in the top-down approach, bulk materials are reduced to nanoscale materials by chemical, mechanical and physical methods. The bottom-up strategy is the most commonly used because it is a low-cost, large-scale approach. It results in the non-periodic or quasi-periodic arrangement of nanostructures, obtained from the synthesis of self-assembled plasmonic nanoparticles on transparent and reflective substrates. That is, the bottom-up approach atomic or molecular substances are assembled through physical processes or chemical reactions to produce nanoscale materials. Although this approach is less sensitive than biosensing, it can be applied in a scalable way to ensure greater use of the technique. Manufacturing techniques include physical or chemical vapor deposition and evaporation, molecular beam epitaxy, and bio/chemical processes (production of supramolecular complexes, self-assembled monolayers, and protein–polymer nanocomposites).^{126–128}

Table 1 Summary of key information about optical fiber sensors applied in healthcare settings

Application	Optical fiber material	Sensor sensitivity	Sensor technique	Ref.
Plantar pressure monitoring	Polymer	—	Intensity variation	17
Smart pants for pose classification	Polymer	Ranging from 1.43 mV mm ⁻¹ to 568.09 mV mm ⁻¹	Intensity variation	56
Smart carpet	Polymer	—	Intensity variation	19
Joint angle measurement	Polymer	0.0229 V Å ⁻¹	Intensity variation	115
Joint angle measurement	Silica	1.20 pm με ⁻¹	Bragg gratings	116
Smart walker instrumentation	Polymer	1.57 ± 0.15 pm με ⁻¹ for strain and 20.17 pm °C ⁻¹ for temperature	Bragg gratings	88
Instrumented insole	Polymer	From 7.71 pm kPa ⁻¹ to 8.51 pm kPa ⁻¹ for pressure and from 18.1 pm °C ⁻¹ to 18.9 pm °C ⁻¹ for temperature	Bragg gratings	87
Instrumented carpet for gait analysis	Polymer	—	Bragg gratings	117
Respiration monitoring	Silica	—	Mach-Zehnder interferometer	102
Arterial pulse wave measurement	Silica	—	Michelson interferometer	103
Rapid monitoring of human physiological activity	Polymer	—	Sagnac interferometer	118
Thrombus detection	Silica	7 nm μL ⁻¹ for the micro-tip fiber type and 8.7 nm μL ⁻¹ for the single-mode fiber type	Fabry-Pérot interferometer	105
Respiratory rate monitoring	Polymer	0.35 nm/%	Optical time domain reflectometry combined with Bragg gratings	119
Liver tissue temperature distribution measurement	Silica	10 pm °C ⁻¹	Optical backscatter reflectometry	113
Protein concentration quantifying	Silica	Ranging from 2.6 nm RIU ⁻¹ to 19.7 nm RIU ⁻¹	Optical backscatter reflectometry	120

Several materials with plasmonic properties can be used for biosensing, among which metals, oxides and nitrides can be mentioned. Noble metals such as palladium, gold and silver have excellent plasmonic properties that help to disperse resonance and absorb light in the infrared and visible regions, but have disadvantages such as high cost or ease of oxidation, like silver.^{124,129} Aluminum has low cost and process compatibility with silicon technologies and exhibits plasmonic resonance in the ultraviolet range. Despite being a promising metal compared to others, aluminum has high chemical instability and easily forms oxides, which decreases the effect of plasmonic resonance.¹³⁰⁻¹³² Copper has near-infrared plasmonic resonance, low cost and high plasmonic conductivity, but like aluminum, its surface can be easily oxidized and the plasmonic effect decreases.^{130,133,134} Semiconductor materials, such as metallic oxides, metallic chalcogenides, metallic nitrides, and silicon, also exhibit plasmonic properties that result in absorption, scattering and near field enhancement in nanostructures. Unlike metals that have a high concentration of carriers and natural resonant frequencies, the concentration of carriers in semiconductors is tunable by the doping procedure, allowing control of the resonant frequencies from visible to far-infrared.^{130,135}

SAMs can be obtained by various combinations of adsorbate–substrate pairs, such as normal alkanoic acids on aluminum oxide or silver, alkylsilanes on hydroxylated surfaces, disulfides on gold, alkanethiols on coinage metals such as platinum, gold, silver, and copper, and non-metals

such as gallium arsenide (GaAs), indium phosphide (InP), and indium tin oxide.¹³⁶ Among these, it is likely that SAMs based on alkanethiols on gold are one of the best options in biosensing because they show great advantages in device fabrication, such as flexibility and stability. This occurs because the bond between gold and sulfur (substrate-adsorbate) has a high affinity for each other and, therefore, it is difficult to desorb the organosulfur species (binding enthalpy is between 167 and 188 kJ mol⁻¹).¹³⁷

Detection based on bioreceptors has become an excellent detection alternative to more advanced analytical techniques, such as scanning electron microscopy,¹³⁸ transmission electronic microscopy,¹³⁹ Raman and infrared spectroscopy,¹⁴⁰ among others. Bioreceptors are key components in the detection of an analyte of biological (e.g. cortisol¹⁴¹ and cholesterol¹⁴²) or synthetic (e.g. micro- and nanoplastics¹⁴³) origin and play a fundamental role in the development of biosensors. These components are constituted by molecules or biological elements that undergo a conformational change or a biochemical reaction when they recognize a certain analyte. This phenomenon generates a type of signal (optical, electrochemical, thermal, acoustic, etc.) that can be transduced into an electrical signal and processed by electronics.^{127,144}

To achieve a highly effective biosensor that is able to accurately recognize its target, some requirements are necessary that are directly related to its bioreceptor. The selectivity of a biosensor depends specifically on the bioreceptor's compatibility with the analyte and ability to



distinguish it from other contaminating species that are mixed in the sample. Stability also plays a key role in biosensing when it requires the analyte to be continuously monitored. In this case, the stability of the biosensor against environmental weather conditions requires that the bioreceptor has affinity with the analyte and that it does not suffer degradation over time.¹²⁷

Since bioreceptors are primary components in the construction of biosensors (alongside transducers and electronics/display), they provide a broad category based on the principle of biorecognition.¹⁴⁵ The first of these is called catalytic biosensors, in which their receptors interact with the analyte and result in a new product of a biochemical reaction.¹⁴⁶ Receptors based on enzymes, cells, tissues and microorganisms are considered catalytic biosensors.¹⁴⁷ On the other hand, non-catalytic biosensors (commonly called affinity biosensors) are those in which the analyte irreversibly binds to the receptors, preventing any biochemical reaction.¹⁴⁶ Antibodies and aptamers (DNA and RNA) can be classified as non-catalytic biosensors.¹⁴⁸ Next, the main types of bioreceptors found in the literature are summarized and how their mechanism of biorecognition with the analyte occurs.

1. Enzymes: enzymes are globular proteins composed of amino acids and whose function is to catalyze biochemical reactions. They are elements widely used as bioreceptors due to their high capacity for biorecognition and excellent catalytic properties when selectively reacting with the substrate. Enzyme-based biosensors have a recognition mechanism whose detection efficiency depends on the catalytic activity of the enzyme and its binding capacity. The product of the biochemical reaction with the analyte can be detected directly or in the presence of an indicator.^{149–151} Approaches to improve the use of enzymes in biosensors can be done by genetic and chemical modification.¹⁴⁶ Genetic modification consists of facilitating the access of the substrate to the active site of the enzyme through fusion protein technology or site-directed mutagenesis.¹⁵² Chemical modification is used to change the properties of enzyme key residues or the overall surface¹⁵³ and approaches such as site-specific chemical modification, nonspecific modification of the enzyme surface, chemical cross-linking, and use of polymers can be cited.¹⁴⁶

A greater number of enzymes with specific biorecognition elements have been used for direct optical detection of disease biomarkers, such as glucose, cholesterol, urea, creatine, lactate (further information is described elsewhere¹⁵⁴). However, a direct-assay-type SPR biosensor has been constructed in recent years to detect chiral amino acids.¹⁵⁵ Albeit chiral recognition techniques are widely used, ultra-low concentration of amino acid isomers still remains a challenge for detection. The enzyme *Rasamsonia emersonii* D-amino acid oxidase was used as the receptor for capture of the specific D-amino acid enantiomer. The detection of the linear range and the signal amplitude has been improved when gold nanorods and graphene oxide were combined on gold film and used to fabricate a sensor assembled with the

enzyme. As a result, a limit of detection of 1.09×10^{-9} mM was achieved, providing a strong capability to determine the composition of the isomer mixture.

2. Cells: cell-based biosensors consist of genetically modified living cells and can be of both animal and microbial origin (bacteria, fungi, algae, protozoa and viruses). Due to their ability to survive in harsh environments, similar to bacteria and fungi, whole-cell biosensors are used to detect specific analytes under extreme conditions, such as low pH, high temperatures, presence of pollutants, etc. While many microbial cell-based biosensors utilize the natural ability of a microorganism to detect a specific analyte, animal cell-based biosensors are more complex due to their growth rate, morphological and structural characteristics, and nutritional requirements.^{127,156,157} Microorganism immobilization techniques can be described as physical methods, such as adsorption and entrapment, or chemical methods, like chemical binding and cross-linking.¹⁵⁸ The immobilization matrix of microorganisms also plays an important role in their immobilization and can be reviewed elsewhere.¹⁵⁹

Different microorganisms, such as bacteria, algae and yeasts, have been used in microbial biosensors in environmental monitoring applications and antimicrobial susceptibility testing. Tahirbegi *et al.* developed a fast pesticide detection method through a microfluidic device with integrated optical pH, oxygen and algal fluorescence for *in situ* analysis. The detection of pesticide concentration was made *via* metabolism/photosynthesis of *Chlamydomonas reinhardtii* algal cells in tap water. The device provides a fast quantification of pesticides in less than 10 min and detection in nanomolar concentrations.¹⁶⁰ However, in the case of animal cells, the growth rate, nutritional requirement and structural characteristics are very different from those of microbial cells. Therefore, this specific kind of cell-based biosensor needs proper design and suitable fabrication.¹⁶¹ For example, Caluori *et al.* placed cardiomyocytes on a microelectrode array to record the beating-force of diseased cardiomyocytes through atomic force microscopy.¹⁶²

3. Antibodies: antibody-based bioreceptors (or immunosensors) are extensively used as diagnostic tools due to their excellent specificity and affinity with their cognate antigen. These biosensors have antibodies as biorecognition components and use the antibody-antigen interaction as an analyte detection mechanism. Antibodies are proteins composed of two heavy and two light peptide chains, and have the shape of a Y, through which interactions occur. Typically, these interactions develop between antibody paratopes and antigen epitopes and allow multiple recognitions to take place according to the nature of the antibody.^{163–165} Technically, the immobilization of antibodies on substrates can occur using two different approaches, namely random and site-directed antibody immobilization. The first consists of a simpler method of random adsorption of antibodies onto a surface, while the second approach improves the availability of the antigen binding site.¹⁶⁶



Recently, Ucci *et al.* designed a lab-on-fiber optrode assisted by an oriented antibody immobilization strategy.¹⁶⁷ Lab-on-fiber technologies integrate functional materials onto optical fiber substrates at the micro and nanoscale to provide a multifunctional all-in-fiber device for many applications, ranging from optical processing, safety, and environmental monitoring.¹⁶⁸ The authors investigated the immobilization strategies to improve them through the use of hinge carbohydrates by involving homemade antibodies. First, an optimized protocol using microfluid surface plasmon resonance was created and then transferred to the lab-on-fiber platform to improve the biofunctionalization protocols. The capability of antigen recognition in ultra-low detection limits was achieved and the proposed oriented antibody immobilization strategy was able to detect Cripto-1 at a concentration of 0.05 nM, *i.e.*, a 10-fold enhancement compared to the random approach.

4. Aptamers: aptamers are synthetic nucleic acids (DNA and RNA sequences) formed by small single-stranded oligonucleotides that bind to their cognate target (proteins, peptides, amino acids, drugs, metal ions and whole cells) with high affinity and selectivity. The high affinity is due to its ability to be able to integrate into the structure of macromolecules or to incorporate small molecules into its own structure. The high stability of aptamers over a wide range of temperature and pH, the possibility of chemically synthesizing them and the ability of thermal refolding bring several advantages to biosensors based on aptamers.^{169–173} As approaches for the immobilization of aptamers on a gold surface, the techniques that have recently been used can be cited, such as immobilization *via* direct thiolation or thiolated short linkers, streptavidin/biotin interaction, as well as DNA nanostructures and reduced graphene oxide as immobilization platforms. More information can be found in more detail elsewhere.¹⁷⁴

In this regard, Ning *et al.* developed a J-shaped optical fiber localized surface plasmon resonance (LSPR) aptasensor able to detect pathogenic bacteria from actual water samples in one step for only 30 min without any sample pre-

treatment. For the first time an LSPR aptamer biosensor was designed for the rapid and ultrasensitive detection of *Helicobacter pylori* (*H. pylori*), reaching a detection limit as low as 45 CFU mL⁻¹ and a wide linear range from 1.0 × 10² CFU mL⁻¹ to 1.0 × 10⁸ CFU mL⁻¹. A further enhancement in LSPR signal response occurs when a spacer nucleic acid with a short stem-loop structure is adopted to control the aptamer density on gold nanoparticles on the surface of the J-shaped optical fiber probe.¹⁷⁵

In addition to the preparation of the device using functionalization mechanisms, the detection of physical and chemical quantities is performed following a transduction mechanism, some of them were discussed in section 4 for optical fiber sensor applications. It is also worth noting that the evolution of artificial intelligence associated with the IoT indicates that the use of sensors tends to increase, increasingly integrating with human beings.¹⁷⁶

Sensors can be classified into several categories.¹⁷⁷ Depending on the type of measurement performed and the detection signal identified, it is possible to classify different types of sensors, as shown in Table 2. Those based on recognition of biological systems are called biosensors. Receptor-based biosensors provide more efficient binding in terms of sensitivity and selectivity. In general, biosensors can be classified according to the biomolecule which can successfully identify the biological analyte, *i.e.*, anti-bodies, aptamers or enzymes (Fig. 6), main bioreceptors.¹⁷⁸

In summary, analytes contained in biological samples are bound to biological macromolecules and produce a biological signal that is passed to the transducer for further processing. Once the analytes are trapped on the surface of these bioreceptors, the interaction can cause changes in the form of light, heat, pH, mass change, charge change, *etc.*¹⁹⁴

This leads to another common classification of sensors that occurs by the way the detected signals are converted for analysis, which can be optical, thermal, electrical, electrochemical or piezoelectrical mechanisms.¹⁸³

In this context, the science of biosensors has become an area independent of modern analytical chemistry, due to the

Table 2 Classification of sensors based on signal detection and measurand

Signal detection	Signal conversion	Measurand	Ref.
Physical	PhotoElastic	Acoustic wave – sound pressure (dB)	179
	Piezoelectric	Physiological signals – pressure (kPa)	180
	Thermoelectric	Heat flux of human body – calorific power (W)	181
	Thermomagnetic	Temperature – radiation (Gy h ⁻¹)	182
	Magnetoelectric	Magnetic microbeads – magnetic field (kA m ⁻¹)	183
Chemical	Chemiluminescence (CL)	Propranolol – CL intensity (au)	184
	Chemiacoustic	Ammonia – S11 (dB)	185
	Chemoresistive	Breath acetone – resistance (au)	186
	Chemocapacitive	CWA and VX gas – capacitance change (%)	187
	Electrochemical	Nitrate – current (µA)	188
Biological	Photoelectric	Urea (enzyme) – transmission (%)	189
	Chemoresistive	miRNA-21(DNA) – resistance (au)	190
	Chemiluminescence (CL)	Bacteria (DNA) – CL intensity (au)	191
	Electrochemical	TCA and nitrite (Hb) – current (µA)	192
	Electrochemical	Cortisol (antibody) – current (µA)	193



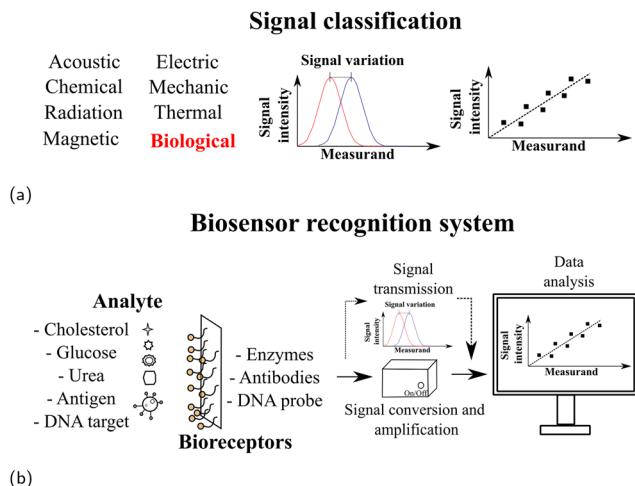


Fig. 6 (a) Classification of sensors based on the detection signal and (b) biosensor biorecognition system.

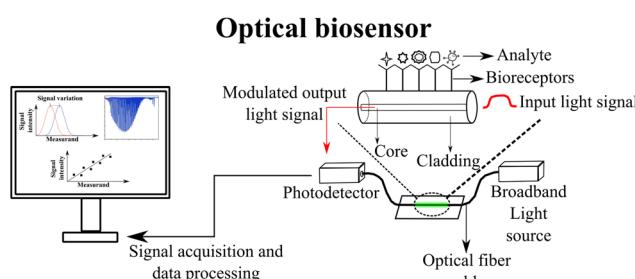


Fig. 7 Optical biosensor schematic representation.

combination of a biological component with a physical-chemical detector.¹⁹⁵ Optical biosensors combine the high specificity of the bioreceptor with the high sensitivity of optical transducers, *i.e.* the analyte/bioreceptor interaction causes changes in light parameters, specifically (Fig. 7).

Currently, the main research centers in the world's demand for biosensors with greater efficiency, precision and economy, seeking to meet the growing demand for clinical purposes related to the constant advancement of molecular biology and the need for diagnosis and disease prevention.

5 Energy harvesting for wearable sensor systems

In the development of actuators, sensors and active components, the power supply is always a significant element.¹⁹⁶ Considering the additional demands of size and power density of wearable devices, the demands of such systems are even larger and significant research in this field has been conducted in the last few years.¹⁹⁷ In this context, self-powered sensors and devices emerged in the last few years as a sustainable and affordable option for wearable device development.¹⁹⁸ For this reason, different technologies for energy harvesting are proposed and we discuss some of them in this section. It is important to mention that this is

not an exhaustive review in energy harvesting (interested readers are advised to check review papers focused on this topic^{199–202}). However, piezoelectric, triboelectric, and solar concentrators, as well as thermoelectric approaches are discussed in this section.

Certain crystals and ceramics have a unique way to generate energy. The phenomenon of the materials that are capable of generating an electrostatic charge when subjected to mechanical stress are known as piezoelectricity.²⁰³ The stress causes a polarization in the piezoelectric crystal, which leads to an opposite charge configuration, when the stress is released, it goes back to the initial state. The conservation of mechanical energy is satisfied generating electrical energy in an external circuit.²⁰⁴

The mechanical energy from movements and vibrations surrounds us in multiple ways and typically is wasted due to the lack of commercialization of the appropriate technology for its capture. This kind of technology has been studied since the end of the last century and brought recent progress that can be used into our daily life such as promising technology for medical applications and other areas where energy is limited or difficult to obtain.

An important piezoelectric energy harvesting method was proposed by ref. 205. The study focuses on the human gait as an energy harvesting model and how different loads impact the project. Also, the material selection for this kind of project is essential to obtain high quality results. Bimorphs and plectra were responsible for generating electrical energy from mechanical movements. The research concluded that the replacement of batteries with the renewable source of energy afforded by energy harvesters has reduced maintenance and consumption. Another application of piezoelectric technology is in the implantable field, such as an organ harvesting system.²⁰⁶ This research focuses on conformal piezoelectric energy harvesters (CPEHs) that utilize the natural contractile and relaxation movements of organs to generate electrical energy. The primary organs under analysis are the heart, lungs, and diaphragm, all of which exhibit a significant energy potential. This could enhance the applicability of such technology in biomedical monitors and wearable electronics.

The concept of tribology appears in many situations during our daily tasks. For example, wheels, trains and moving cars.²⁰⁷ Triboelectricity is based on the friction between different materials to generate electrostatic charges that can be harvested by an external circuit.²⁰⁴ Furthermore, it allows us to construct or integrate wearable power supply mechanisms. As another important approach for energy harvesting in wearable applications, the triboelectric nanogenerators (TENGs) represent an important recent technology that offers numerous advantages, including energy harvesting capabilities, biocompatibility and self-powering capabilities.²⁰⁸

Utilizing human biomechanical energy and TENG technology,²⁰⁷ a self-sustaining system that provides a continuous source of electricity for mobile electronics is



proposed. The project involves a multilayered attached-electrode contact-mode TENG and a power management circuit. This specific type of TENG was designed with a Kapton film and electrodes to harness human biomechanical energy.²⁰⁹ The act of walking or running by humans creates contact between the Kapton film and the electrodes, generating energy. The advantage of this technology is the ability to stay connected without the need to pause and recharge devices, including wearable technology, sensors and more.

Alopecia is a common topic of debate within the scientific community. This is primarily due to the fact that 50% of all men suffer from alopecia by the age of fifty,²¹⁰ prompting the exploration of new solutions for this issue. The study of triboelectric nanogenerators (TENGs) is proposed as a means of providing therapeutic electrical stimulation (ES). TENG-ES enhances hair growth cycles by promoting capillary blood flow, stimulating cellular proliferation and differentiation, and, more importantly, can serve as a self-sufficient treatment without the need for third-party equipment interventions.²⁰⁸

Moreover, a wearable electrical stimulation device (M-ESD) was designed by ref. 211. As is common in TENG projects, it harnesses body movements as a source of energy to enhance blood vessel growth factors and increase the number of hair follicles. The M-ESD consists of a TENG with two friction layers connected by a soft Ecoflex belt and a pair of interlaced electrodes. Such results indicate that the TENGs are not only used for energy harvesting proposes, but also can enhance the wearable and healthcare device capabilities by promoting therapeutic alternatives for different applications.

The solar energy harvesting technology involves using solar cells to convert sunlight energy into electrical energy *via* the photovoltaic effect.²¹² Solar cells are made of semiconductor junctions that have a characteristic energy level called a band gap. When incident photons have higher energy than this band gap, the solar cell generates an electrical current.²¹³ The generated power is then optimized by a conditioning circuit and stored in an energy storage device (ESD) or used to power an electronic load.²¹⁴ The input from the solar cell goes through a boost converter controlled by a Maximum Power Point Tracking (MPPT) algorithm. MPPT algorithms force the solar cells to work around their Maximum Power Point (MPP), which is normally at 80% of their open circuit voltage.²¹³ The performance of solar energy harvesting systems depends on many factors, such as the operating conditions of the load (sensor measurements and connectivity), time, location and position of the solar cells. It is also difficult to theoretically estimate the generated power when solar cells are included in a wearable device. Therefore, testing the performance of the solar energy harvesting system under different daily life scenarios is necessary.²¹⁴ Additionally, the cost of large-scale production and fabrication of energy harvesting wearable devices is a concern.

For example, they can be used to power physiological sensors that monitor vital signs such as the heart rate, blood pressure, and body temperature. This can be especially useful

for patients who need continuous monitoring, such as those with chronic conditions or those recovering from surgery.²¹⁵ Additionally, solar-powered wearables can be used to power wireless communication modules that transmit data from the sensors to healthcare providers, allowing for remote monitoring and telemedicine. This can improve patient outcomes and reduce healthcare costs.²¹²

In addition, wearable luminescent solar concentrators (LSCs) can be integrated with fiber solar cells to harvest additional photons and fully utilize the advantage of harvesting light from all directions.¹⁴³ LSCs are made from an amphiphilic polymer conetwork matrix and luminescent dyes, and they assist in enlarging the photon harvesting area, recycling lost photons, and utilizing the complete surface of the fiber dye-sensitized solar cells (FDSSCs). The results of the study showed a remarkable FDSSC power conversion efficiency enhancement of 84%, LSC concentration factor of 1.57, and device optical efficiency of 7.85%.¹⁴³ Additionally, wearable LSCs demonstrated a sustainable power conversion efficiency even after 1000 bending cycles, which demonstrates their wearability.

A thermoelectric generator (TEG) is a device that converts heat energy into electrical energy using the Seebeck effect.²⁰⁰ The Seebeck effect is a phenomenon where a temperature difference between two dissimilar conductors or semiconductors produces a voltage difference between them.²¹⁶ In a TEG, two different types of semiconductors (p-type and n-type) are connected in series, forming a thermocouple. When one side of the thermocouple is heated and the other side is cooled, a voltage difference is generated between the two sides, which can be used to power an electrical load. The efficiency of a TEG is determined by the thermoelectric properties of the semiconductors used, such as the Seebeck coefficient, electrical conductivity, and thermal conductivity.²⁰⁰ TEGs have a wide range of applications, including waste heat recovery, power generation in remote locations, and energy harvesting from body heat. Drawbacks of thermoelectric generators include their relatively low efficiency compared to other energy conversion technologies, their high cost, and their limited power output. Additionally, thermoelectric generators are often sensitive to temperature changes and may require complex thermal management systems to maintain optimal performance.²¹⁷

Applications can be found in various wearable electronic devices, such as health monitoring systems, performance monitoring systems, and wireless communication devices. These TEGs can harvest energy from the human body heat and convert it into electrical energy to power these devices, eliminating the need for external power sources or batteries. The TEGs can be attached to the skin or clothing, making them convenient and unobtrusive for the user. The high power density of these TEGs also makes them suitable for applications that require continuous and long-term monitoring. Additionally, they can be used for personalized thermoregulation.²¹⁷ Fig. 8 shows an illustration of harvesting energy applications: energy harvesting and storage



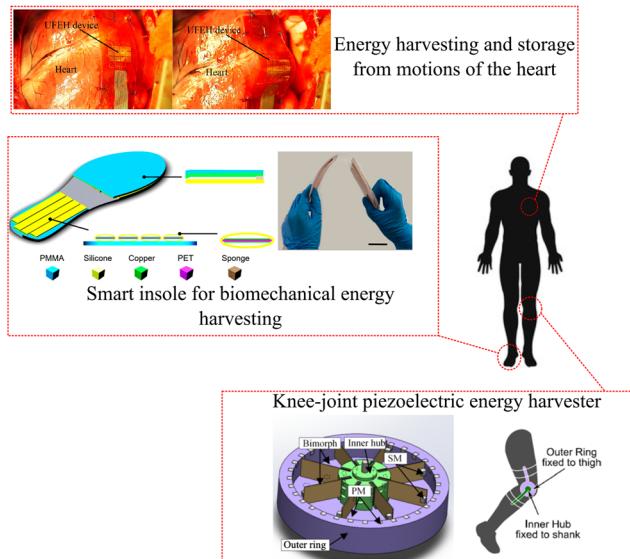


Fig. 8 Applications of energy harvesting using human physiological systems. Reproduced from ref. 218–220 with permission from IOP Publishing Group, copyright 2016, Springer Science and Business Media LLC, copyright 2015, and American Chemical Society (ACS), copyright 2020, respectively.

from motions of the heart,²¹⁸ smart insole for biomechanical energy harvesting,²¹⁹ and knee-joint piezoelectric energy harvester.²²⁰ These devices are capable of harvesting energy to develop self-powered sensor systems in healthcare settings. Table 3 summarizes some key features of the aforementioned energy harvesting systems.

6 Sensor integration into textiles and components

A number of challenges are involved in developing and implementing smart textiles for biomedical applications, including their dependability and washability, their high manufacturing costs, and their effectiveness after washing. Researchers around the globe are continuously working to

advance smart textile-based biomedical devices, but the development of wearable smart textile standards and testing procedures is now the focus of intense activity. Additionally, smart textiles have the potential to provide doctors with a complete picture of physiological responses, motivate people to take more active roles in their healthcare, assist in exercise, and record progress.^{221–226}

Structures that incorporate sensors, actuators, and other components have the capability of sensing, responding, and adapting to changes in their environment. A variety of materials can be incorporated to construct these structures, called smart structures. These materials include metals, polymers, and composites, and they can be rigid or flexible. In smart structures, sensors and other components can be used to monitor their conditions, detect changes in their environment, and respond to those changes. A smart structure can, for instance, be used to detect and diagnose equipment failures, monitor production lines, and track the movement of goods and materials within a manufacturing facility.²²⁷ Besides their use in aerospace and civil engineering,^{228,229} they can also be applied to biomedical engineering.^{230,231}

Due to their high flexibility and impact resistance, optical fiber sensors, specifically POFs, are well-suited to smart textiles and intrusive applications. Optical fibers based on silica can be more prone to breakage than polymer fibers, which may result in injury to users when there is a breakage due to glass punctures. Moreover, POFs can be embedded in soft structures, making them a suitable instrumentation option for wearable robots and biomedical applications.⁵¹ Compared to electronic sensors, they include compactness, lightness, flexibility, immunity to electromagnetic interference, chemical stability, and multiplexing capabilities.^{56,232} These features have pushed POF sensor technology into a wide range of smart textile applications.^{56,232,233}

Optical fibers can be embedded into textile fabrics using various textile fabrication techniques. Weaving, knitting, and non-weaving techniques are among them. In weaving, the

Table 3 Summary of energy harvesting systems key features

Application	Energy type	System voltage	Output power	Ref.
Human gait as an energy harvesting mode	Piezoelectricity	20 V (peaks observed during the gait cycle)	2.06 ± 0.3 mW	205
Organ harvesting system	Piezoelectricity	3.7 V	Time-averaged power density of 1.2 $\mu\text{W cm}^{-2}$	206
Energy harvesting from wheels, trains, and cars	Triboelectricity	—	1.044 mW (7.34 W m^{-3})	207
Energy harvesting in wearable applications using triboelectric nanogenerators	Triboelectricity	10–100 V	1–100 mW	208
Energy harvesting from human activities (e.g. running and walking)	Triboelectricity	—	—	209
Harnessing of body movements as a source of energy to enhance blood vessel growth factors and increase the number of hair follicles	Triboelectricity	6.2 V	5.488 to 70 μW	211
Feeding physiological sensors for vital sign monitoring	Solar energy	Ranging from 90 V to 2 kV	Ranging from 90 μW to 0.2 mW	215
Feeding wireless communication modules used in healthcare devices	Solar energy	1.7996 V	159.1 mW	212
Skin-attached device for energy harvesting	Thermoelectricity	Ranging from 1.27 to 3 V	—	217



most commonly used technique, warp and weft yarns are interlaced in a basic woven structure. In order to ensure effective transmission and sensing, optical fibers are commonly woven into fabrics in an unbent condition or with a limited bending angle through the weaving technique.²³⁴ External perturbations can modulate the optical output signal in sensing applications, as reported in ref. 235, where a POF is embedded into a textile for respiration frequency monitoring using a smartphone-based interrogation technique and remote access to information through a Web server. Leal-Junior, Arnaldo, *et al.* have proposed a smart textile in a similar approach, where POFs embedded into a textile were used to monitor the displacement caused by respiration, vibration due to heartbeat, and bending induced in the gait while a user is operating a smart walker²³⁶ and to assess the heart rate under dynamic movements.²³⁷ Najafi, Bijan, *et al.* have demonstrated smart socks embedded with highly flexible optical fibers where temperature and pressure were assessed for the management of biomechanical risk factors associated with diabetic foot amputation.²³⁸ Li, Tianliang, *et al.* have proposed a smart-textile-based gesture recognition application with additional monitoring of respiratory signals under different postures using fiber Bragg gratings embedded into textiles.²³⁹ Avellar, Leticia, *et al.* have proposed POF-based pants for a fully optical fiber-integrated smart textile for remotely monitoring the biomechanics of lower limbs.⁵⁶ These reported applications highlight the suitability of optical fiber sensors for the development of wearable devices and smart textiles due to the aforementioned advantages compared to electronic sensors.

Besides the aforementioned biomedical applications, optical fiber sensors can be embedded into structures for structural health monitoring, as reported in a review study conducted by Sasy Chan, Y. W., *et al.* where they provided

state-of-the-art optical fiber sensing technologies and their practical application in railway infrastructures²⁴⁰ and for real-time damage detection in smart composites proposed in ref. 241. Moreover, the recent advances in aerospace composite structure monitoring through optical fiber-based sensors are reported in ref. 242 and 243. Finally, it is worth mentioning that these applications demonstrate that optical fiber sensors are a flexible technology that has been explored in sensor integration for a wide variety of applications, from biomedical to industrial applications, enhancing the efficiency and safety of these applications by providing measurements that can be utilized for data-driven decisions. Fig. 9 shows some examples of textiles integrated with sensors for plantar pressure assessment,²⁴⁴ impact monitoring,²⁴⁵ and gesture recognition (Table 4).²⁴⁶

7 Sensors for vital sign monitoring

In healthcare settings, monitoring vital signs is crucial as it enables early detection of deterioration. This can be achieved by, for example, analyzing blood pressure, respiratory rate, temperature, and oxygen saturation. The use of sensors in vital sign monitoring enables healthcare providers to obtain accurate and real-time data regarding a patient's physiological parameters. When analyzing the data obtained by the sensors, clinical decisions can be made with respect to treatment plans, medication doses, and the need for a further evaluation.²⁴⁷ Furthermore, the interest in development of sensors to monitor vital signs is growing, with the World Health Organization predicting that 1 in 6 people will be over the age of 60 by 2030. The number of people aged 60 years and older is expected to increase from 1 billion in 2020 to 1.4 billion by 2050.²⁴⁸ Additionally, aging is accompanied by a natural decline in physical capabilities, including muscle strength, power, and endurance. Several factors contribute to this decline, including changes in muscle mass, neuromuscular function, and cardiovascular capacity.²⁴⁹ As a result of these factors, there is an increasing demand for sensor development for healthcare applications.

In addition to optical fiber sensors, various sensor modalities such as electromechanical sensors, thermistors, accelerometers, electrodermal activity sensors, acoustic sensors, and RFID tags find utility within healthcare contexts.²⁵⁰ Optical fiber sensors offer different advantages for healthcare applications, including high sensitivity allowing precise measurement of vital signs, miniaturization facilitating integration into wearable or implantable devices for continuous monitoring, immunity to electromagnetic interference ensuring compatibility with medical equipment, and multiplexing capability enabling simultaneous monitoring of diverse vital signs at different anatomical locations.²⁵¹ The employment of optical fiber sensors for monitoring vital signs is already well-consolidated.

A POF-based sensor was proposed by Leal-Junior *et al.* for simultaneous measurement of breath and heart rate under dynamic movement.²³⁷ As part of the employed setup, the

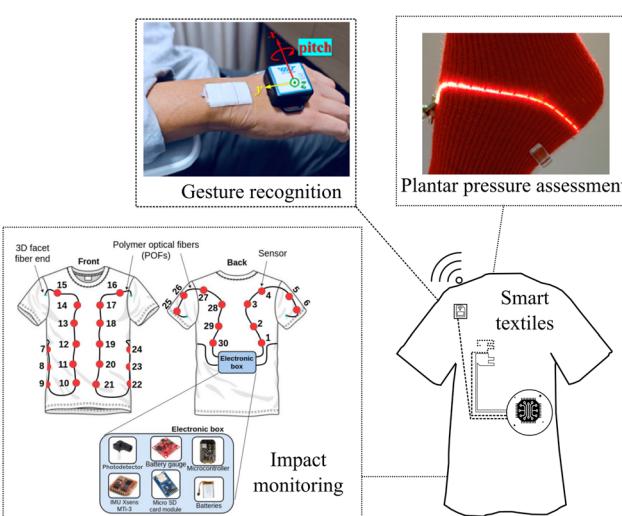


Fig. 9 Applications of textile integrated with sensors as examples. Reproduced from ref. 244–246 with permission from MDPI AG, copyright 2019, IEEE, copyright 2021, and Optica Publishing Group, copyright 2024, respectively.



Table 4 Summary of key information about optical fiber sensor integration into textiles and components

Application	Structure	Sensor technique	Ref.
Respiration frequency monitoring using a smartphone-based interrogation technique	Smart garment	Intensity variation-based using POF	235
Cardiorespiratory frequency monitoring during smartwalker operating	Smart garment	Intensity variation-based using POF	236
Plantar pressure and temperature monitoring	Smart socks	Intensity variation-based using POF	239
Remote monitoring of the lower limb biomechanics for pose classification and motion quantifying	Smart pants	Intensity variation-based using POF	56
Real-time damage detection in smart composite	Composites	Fiber Bragg gratings and Fabry-Pérot interferometer	241
Aerospace composite structure health monitoring	Aerospace composites	Brillouin-based distributed sensing	243

sensor is positioned on the subject's chest in various regions to record both breathing and heartbeat-induced body vibrations. The sensor is lightweight and flexible, allowing for comfortable wear during a variety of activities. Furthermore, signal processing techniques were applied to the sensor data to eliminate the influence of body movement on the sensor response. To further reduce the effects of movement, the analysis is conducted in the frequency domain. As a result of this approach, the sensor was able to accurately measure both the breath rate and heart rate even during dynamic movements such as walking or other activities.

Haseda Y. *et al.* have proposed a polymer optical fiber Bragg grating sensor for blood pressure measurement.²⁵² The objective of this study was to measure pulse wave signals and calculate blood pressure using a setup comprising polymer optical fiber Bragg grating (POFBG) sensors. A POFBG sensor was placed on the brachial artery of the left elbow, while a silica-FBG sensor was located nearby for comparative purposes. Using reference blood pressure measurements obtained from an electrical sphygmomanometer, a calibration curve was constructed using pulse wave signals sampled at 1 kHz. In order to measure pressure changes, the FBG sensor system consists of a laser light source and an interrogator that detects shifts in the Bragg wavelength. To enhance the accuracy of blood pressure calculations, data processing techniques such as filtering and partial least squares regression (PLSR) were applied. As a whole, the setup and working principle of POFBG sensors demonstrated their potential for non-invasive and wearable vital sign monitoring with improved safety and reliability.

Leal-Junior *et al.* have proposed smart textiles for wearable sensing using optical fiber sensors.²⁵³ This study proposes a smart textile system that integrates multiplexed optical fiber sensors within a neoprene textile fabric. Between two layers of the fabric, sensors consisting of POFs and flexible light-emitting diodes (LEDs) are embedded into the textiles. In this setup, it is possible to simultaneously measure multiple parameters, such as temperature and interaction forces, with high sensitivity, linearity, repeatability, and resolution. Using controlled pressures and different temperatures to the sensors, the sensors are able to detect pressure interactions between the user and the environment, enabling the measurement of body temperature and the tracking of

activity levels. With low crosstalk between sensors, the system ensures accurate and independent measurements, making it a promising technology for wearable health monitoring.

Avellar L. *et al.* have proposed a fully portable smart textile integrated with optical fiber sensors.⁵⁶ This setup involves integrating POF sensors into a flexible textile for monitoring lower limb biomechanics and activity recognition. These sensors used multiplexed intensity variation to detect movements such as walking, squatting, and kicking. Each sensor was characterized by vertical displacements to enable machine learning algorithms to normalize the data. These data can be used for posture monitoring, which provides

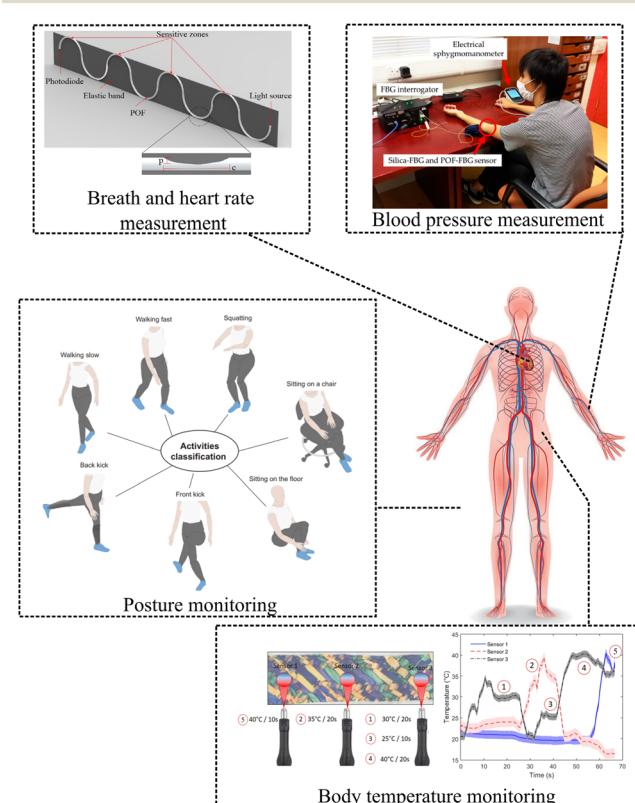


Fig. 10 Examples of vital sign monitoring applications using optical fiber sensors. Reproduced from ref. 56, 237, 252 and 253 with permission from Elsevier, copyright 2019, MDPI AG, copyright 2019, Springer Science and Business Media LLC, copyright 2020, and Optica Publishing Group, copyright 2023, respectively.



Table 5 Summary of vital sign monitoring using optical fiber sensors concerning the sensor type

Application	Sensor type	Ref.
Simultaneous measurement of breath and heart rate under dynamic movement	Intensity variation-based using POF	237
Blood pressure measurement	Polymer optical fiber Bragg gratings	252
Body temperature monitoring	Intensity variation-based using POF	253
Posture monitoring	Intensity variation-based using POF	56

valuable insights into daily activities and biomechanical patterns. Patients with mobility impairments or chronic conditions may benefit from the ability to monitor their movements in real time, which can contribute to early detection of movement abnormalities and improve their quality of life. Finally, Fig. 10 summarizes the aforementioned applications for monitoring vital signs using optical fiber sensors. Table 5 summarizes the vital sign monitoring using optical fiber sensors discussed in this chapter concerning the employed sensing technique.

8 Biosensors and intrusive applications

The need to continuously monitor the metabolites of the human body can provide additional understanding about the patient's diagnosis and improve the treatment needed for his recovery. Alternative monitoring is often done by devices located in specialized laboratories and requires skilled labor to carry out the measurements. Typically, fluids are sampled from patients at regular intervals and taken for analysis. This means that monitoring performed in this way has a high susceptibility to human errors and can mask the diagnosis. As an alternative to these typical procedures, biosensors can be used as invasive monitoring devices and provide continuous detection of chemical and biological species in medical applications.²⁵⁴

Briefly, there are three different types of biosensors that can invasively monitor patients: wearable (applied to the skin), ingestible (ingested into the gastrointestinal tract) and implantable (implanted into the body directly in contact with the skin).²⁰¹ Despite the advantages that invasive biosensing can provide in clinical analysis, it still has some problems that need to be overcome, such as its stability, biocompatibility, and calibration. In addition, no approach has been able to couple all the analytical requirements for biosensing, often presenting different performance on the benchtop with *in vivo* monitoring.²⁵⁵ However, new materials and manufacturing processes have emerged to minimize these deficiencies associated with biosensors and overcome the inherent challenges of invasive monitoring.

Regarding the manufacturing processes of these implantable and wearable biosensors, it is essential to consider the mechanisms for converting biological recognition into a detectable signal. This requires that the manufacturing process be seen as an important step in the development of a biosensor, not only considering the production costs with respect to the technique used, but also

the minimization of instrumental errors and the possibility of transitioning to non-invasive sensing.²⁵⁶ Invasive or minimally invasive biosensor fabrication techniques, such as lithography, screen printing technology and inkjet printing, can be cited as viable and low-cost approaches and will be described as follows.

1. Lithography is an excellent procedure that allows the components of a biosensor to be built systematically by superimposing layers. It is a technique with wide applications and that covers several possibilities of surface fabrication. Lithography is a technique that depends on the medium which is used to transfer the geometric pattern to the substrate and therefore is broadly classified in different ways, which can be cited as the most used photolithography, stereolithography, and electron beam lithography.^{256,257} Without a doubt, photolithography is one of its best-established and most widely used aspects in industry, mainly due to its large-scale production and low cost.²⁵⁸ In this process, UV light is used to transfer a geometric pattern established by a mask onto a substrate coated with a photoresist material. Despite its low resolution of approximately 1 μm , approximation and contact photolithography are still widely used throughout industry and modern research.²⁵⁹

2. The stereolithography process relies on photopolymerization or light-initiated polymerization and allows three-dimensional structures to be constructed. First, the digital models of the 3D figure are converted into a standard tessellation language file format so that they are virtually cut into sequential layers. This information is sent to the stereolithography apparatus which builds the 3D figure. The apparatus emits a high-energy laser that writes on the surface of the resin and traces the model of the 3D figure layer by layer by sequential polymerization. In other words, the construction of the 3D object is done by polymerizing two-dimensional cross sections.²⁶⁰ Its application in biosensing can be done by the manufacture of biosensors and their components or in the development of integrated devices. 3D printing technology in electrochemical biosensors, for example, has been widely used because it can provide customizable manufacturing of its electrodes, allowing full control over the shape and size of the desired electrode and printing on a specific matrix.²⁶¹

3. Electron beam lithography (EBL) is a process used to create nanostructures with a high standard of resolution and alignment accuracy. These nanostructured arrangements are formed by sweeping a focused electron beam over a surface coated with an electron resist film.



After immersing the device in an appropriate solution, exposed or unexposed resist will be completely removed from the surface.^{135,262} In the field of biosensors, silver nanodisks have already been designed by EBL on a sapphire substrate and used to detect heat-killed *Escherichia coli*.²⁶³ Nanoelectrode arrays were also obtained by EBL by creating nanoholes in thin polycarbonate films deposited on a macroelectrode.²⁶² In another case, EBL was used to fabricate nanostructures resembling filamentous virions on Si/SiO₂ substrate, as reported by Agrawal *et al.*¹³⁵

4. Screen printing technology (SPT) is a manufacturing process for microelectronic devices or components. The method is highly competitive among other means of manufacturing microelectrodes due to its low cost, easy operation and scalability. It consists of painting different types of substrates, such as ceramics, plastics, and paper, using specific paints or pastes that allow adaptability, design and modifications according to manufacturing needs.²⁶⁴ This ink or paste is squeezed through a frame or woven mesh and printed onto a matrix.²⁶⁵ SPT is widely used in the fabrication of electrode systems used in electrochemical biosensors. The working electrode, reference electrode and counter electrode can be printed on inert substrates and enable an electrochemical system for the manufacture of the biosensor.²⁶⁶ In addition, its miniaturized dimension allows only one drop of solution, *i.e.*, on a microliter scale, to be analyzed. The diffusion distance is also reduced (distance from the analyte to the surface of the receptor), decreasing the incubation time and providing a lighter measurement.²⁶⁷

5. Finally, inkjet printing technology consists of a printer containing an ink chamber and a channel that takes the ink to an injector nozzle, where the ink is digitally controlled. An inkjet printer is classified as a drop-on-demand (DOD) printer, which implies the ejection of ink only when necessary to avoid spending on necessary materials.²⁶⁸ The versatility of the technique, added to its low cost and manufacturing speed, allowed the technology to be used to print electronics²⁶⁹ and allowed manufacturing 2D²⁷⁰ and 3D structures.²⁷¹ The development of biosensors manufactured using inkjet printing began in the mid-1990s and since then, it has been a field extensively investigated by academic research and highly viable for industry. They fall into virus sensors, enzymatic biosensors and non-enzymatic biosensors, which may be reviewed in greater detail elsewhere.²⁶⁸

In summary, biosensor devices of high quality and precision require techniques that allow the fabrication of complex arrangements and that use multiple components. Lithography, screen printing technology and inkjet printing technology start to play a role as more functional techniques in the manufacture of these devices by enabling the formation of sensory components by simplified methods and offering versatility of operation. In addition, variations in techniques allow the miniaturization of biosensor devices to be manufactured and used in an invasive way, in which the

size of the device plays a fundamental role when implanted in the individual to monitor body fluids.

Ingestible biosensors are used as an alternative to implantable and wearable biosensors as they allow access to places within the human body that are impossible for the other two types. Thanks to advances in bioengineering, microelectronics and materials science, these biosensors could be manufactured in the form of capsules (and for this reason they are called ingestible biosensing capsules, IBCs) that are capable of monitoring multiple functions.²⁷² Its small capsule size (in the order of a few centimeters) and its tablet format contribute to the IBC being swallowed easily and allowing the monitoring of biomarkers of interest in the gastrointestinal tract to be carried out.²⁷³

Lu *et al.* have classified IBCs into two categories according to their locomotion, with passive IBCs being those that follow the peristaltic motion of the gastrointestinal tract and active IBCs being those that can be magnetically guided.²⁷² This was only possible thanks to the significant advances in the miniaturization of processors that made it possible to manufacture devices that are smaller, smarter, and more operationally efficient. Advances in battery technology also played an important role in the development of ingestible biosensors, allowing them to operate for long periods with a tiny device.²⁷⁴

The use of ingestible biosensors makes it possible to measure a wide range of biomarkers from the mucous membrane of the guts, including electrolytes, metabolites,²⁷³ hormones, enzymes, and gases,²⁷⁵ for example. The by-products of the microorganism ecosystem that inhabit the gastrointestinal tract, *i.e.*, the human gut microbiota, can also be monitored by these devices.^{274,276} A list of possible markers can be monitored in this way, such as the gastric juice electrolytes that control pH (hydrochloric acid, potassium chloride and sodium chloride),^{277,278} digestive enzymes (amylase, pepsin, trypsin, *etc.*),^{274,279} O₂-equivalent, H₂ and CO₂ gases, and microbial community,^{280,281} beyond imaging,²⁸² temperature monitoring,²⁸³ pressure,²⁸⁴ and drug delivery units.²⁸⁵

Regarding materials for manufacturing this kind of device, some essential components can be highlighted, such as structural materials and packaging, materials for electronic components, logic devices, communications and power sources, as reported by Bettinger, C.²⁸⁶ For encapsulation, some substances are considered generally recognized as safe (GRAS), such as gelatin, cellulose, zinc oxide, and metallic iron, among others. Polymers, ceramics, pigments, metals, and minerals are also classified as GRAS and can be used in encapsulation as well as device electronics.²⁸⁶ Galvanic couples, such as Zn|Cu, Mg|Cu,²⁸⁷ and melanin|ceramics²⁸⁸ are very energy efficient and can supply large voltages to the system when needed. On the other hand, energy collected from bodies of living subjects in chemical, thermal and mechanical ways is under development and could be used for self-powered electronics.²⁸⁹



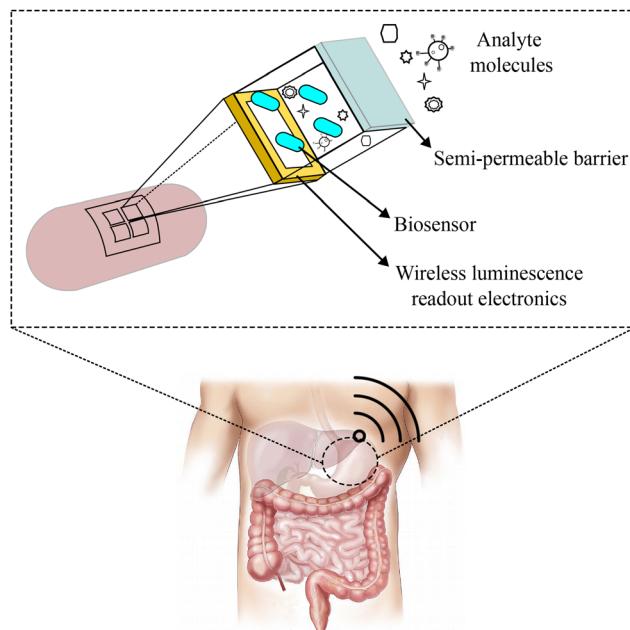


Fig. 11 Schematic representation of an ingestible biosensor.

In summary, the amount of information that an ingestible biosensor can provide about the gastrointestinal tract is enormous. Knowledge about the microbiome and how it impacts the other organs of the human body, the behavior of metabolites and hormones in the face of changes in the gastrointestinal environment and understanding about electrolytes in the intestine in response to the ingestion of food, medication and supplements are some of the questions that can be answered with the use of ingestible devices.²⁷⁴ Fig. 11 shows a schematic representation of an ingestible biosensor with its main parts highlighted. Table 6 summarizes some key information about the aforementioned biosensors used in intrusive applications.

9 Conclusions

This paper presented a review of wearable sensor applications for healthcare using optical fiber sensors. In this case, the overview of population aging and the demands that led to the development of new healthcare technologies are discussed. Then, the development of wearable and healthcare sensors is discussed considering all steps and components on the wearable optical fiber sensor fabrication, namely the optical fiber sensor approaches, fiber functionalization and novel technologies for power supply and self-powered sensor systems. In addition, the applications with textile integrated sensors as well as biosensors and intrusive applications are presented. Thus, this review presents an overview of wearable devices and indicates the possibility of combining the aforementioned technologies for the new generation of wearable devices, where the self-powered integrated optical systems for both unobtrusive and intrusive applications can be used for different healthcare devices and approaches.

It is possible to envisage the popularization of the optical fiber biosensing techniques using low cost platforms due to their advantages related to miniaturization, flexibility and multiplexing techniques, where multipurpose structures can be developed for vital sign monitoring. In the future, it may be possible to integrate the optical fiber biosensing platforms with the energy harvesting systems discussed in this paper to achieve a self-powered optical fiber sensing platform for biomedical applications. Another advantage of such systems is the possibility of using both in continuous monitoring and as need-based devices, since the optical fiber sensors can provide continuous monitoring of different parameters with low energy consumption. It is also worth mentioning that commercially available optical fiber sensors are already available with some startup companies beginning the commercialization of smart textiles and glucose sensors.

Table 6 Summary of key information about the discussed biosensors used in intrusive applications

Application	Locomotion	Sensor description	Ref.
Biomarkers for metabolites	Passive	The proposed sensor composition includes a battery-free ingestible biosensing system that integrates a self-powered glucose biofuel cell biosensor	273
Biomarkers for gases	Passive	This biosensor is an ingestible electronic capsule that incorporates electronic components for measuring H ₂ , CO ₂ , and O ₂ concentrations in the gut. It uses a combination of thermal conductivity and semiconducting sensors, with high accuracy for measuring gas concentrations	275
Monitoring of by-products of the microorganism ecosystem that inhabit the gastrointestinal tract	Passive	The capsule utilizes a combination of thermal conductivity and semiconducting sensors, with the ability to adjust sensor heating elements for selectivity and sensitivity to different gases	274, 276
Monitoring of microbial community	Passive	The proposed ingestible sensor is a miniaturized detector and transmitter packed into a capsule for tracking and quantifying bioactive compounds in the gut to assess microbiome effectiveness	280, 281
Drug delivery unity	Passive	Ingestible device designed for highly predictable intraluminal drug delivery in the gut. It features a drug reservoir, pH and temperature sensors, a stepper motor, and a transceiver for real-time wireless communication. The device can relay data to a computer, allowing for precise drug delivery control based on pH and temperature changes in the gut	285



Author contributions

A. L-J. conceptualization, funding acquisition, investigation, methodology, project administration, supervision, visualization, writing – original draft, writing – review & editing; J. S. conceptualization, investigation, methodology, visualization, writing – original draft; L. M. methodology, visualization, writing – original draft; A. M. methodology, visualization, writing – original draft; S. M. investigation, visualization, writing – original draft; J. V. conceptualization, investigation, methodology, visualization, writing – original draft; F. V. investigation, visualization, writing – original draft; M. C. conceptualization, investigation, methodology, visualization, writing – original draft, writing – review & editing.

Conflicts of interest

There are no conflicts to declare.

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Notes and references

- 1 S. Li, L. D. Xu and S. Zhao, *J. Ind. Inf. Integr.*, 2018, **10**, 1–9.
- 2 S. Sicari, A. Rizzardi and A. Coen-Porisini, *Comput. Netw.*, 2020, **179**, 107345.
- 3 A. Zanella, N. Bui, A. Castellani, L. Vangelista and M. Zorzi, *IEEE Internet Things J.*, 2014, **1**, 22–32.
- 4 G. A. Akpakuwu, B. J. Silva, G. P. Hancke and A. M. Abu-Mahfouz, *IEEE Access*, 2017, **6**, 3619–3647.
- 5 T. Islam, S. C. Mukhopadhyay and N. K. Suryadevara, *IEEE Sens. J.*, 2017, **17**, 577–584.
- 6 L. Romeo, A. Petitti, R. Marani and A. Milella, *Sensors*, 2020, **20**, 1–23.
- 7 A. Leal-Junior and C. Marques, *IEEE Instrum. Meas. Mag.*, 2021, **24**, 41–49.
- 8 P. P. Jayaraman, A. R. M. Forkan, A. Morshed, P. D. Haghghi and Y. B. Kang, *Wiley Interdiscip. Rev., Data Min. Knowl. Discov.*, 2020, **10**, 1–23.
- 9 A. Turner, *Philos. Trans. R. Soc., B*, 2009, **364**, 3009–3021.
- 10 E. Jaul and J. Barron, *Front. Public Health*, 2017, **5**, 1–7.
- 11 S. Majumder, T. Mondal and M. Deen, *Sensors*, 2017, **17**, 130.
- 12 L. P. Malasinghe, N. Ramzan and K. Dahal, *J. Ambient Intell. Humaniz. Comput.*, 2019, **10**, 57–76.
- 13 G. H. Lee, H. Moon, H. Kim, G. H. Lee, W. Kwon, S. Yoo, D. Myung, S. H. Yun, Z. Bao and S. K. Hahn, *Nat. Rev. Mater.*, 2020, **5**, 149–165.
- 14 Kenry, J. C. Yeo and C. T. Lim, *Microsyst. Nanoeng.*, 2016, **2**, 16043.
- 15 A. Leal-Junior, L. Avellar, V. Biazi, M. S. Soares, A. Frizera and C. Marques, *Opto-Electron. Adv.*, 2022, 210098.
- 16 S. Abdulmalek, A. Nasir, W. A. Jabbar, M. A. M. Almuhaya, A. K. Bairagi, M. A.-M. Khan and S.-h. Kee, *Healthcare*, 2022, **10**, 1993.
- 17 A. G. Leal-Junior, C. R. Díaz, C. Marques, M. J. Pontes and A. Frizera, *Opt. Laser Technol.*, 2019, **116**, 256–264.
- 18 A. G. Leal-Junior, D. Ribeiro, L. M. Avellar, M. Silveira, C. A. R. Diaz, A. Frizera-Neto, W. Blanc, E. Rocon and C. Marques, *IEEE Sens. J.*, 2021, **21**, 2995–3003.
- 19 L. Avellar, A. Leal-Junior, C. Diaz, C. Marques and A. Frizera, *Sensors*, 2019, **19**(14), 3156.
- 20 S. M. A. Iqbal, I. Mahgoub, E. Du, M. A. Leavitt and W. Asghar, *npj Flexible Electron.*, 2021, **5**, 9.
- 21 W. G. Macena, L. O. Hermano and T. C. Costa, *Revista Mosaicum*, 2018, **27**, 223–236.
- 22 M. Latash, M. Levin, J. Scholz and G. Schöner, *Medicina*, 2010, **46**, 382.
- 23 W. J. Chodzko-Zajko, D. N. Proctor, M. A. F. Singh, C. T. Minson, C. R. Nigg, G. J. Salem and J. S. Skinner, *Med. Sci. Sports Exercise*, 2009, **41**, 1510–1530.
- 24 S. Usmani, A. Saboor, M. Haris, M. A. Khan and H. Park, *Sensors*, 2021, **21**(15), 5134.
- 25 L. A. de Melo and K. C. de Lima, *Cien. Saude Colet.*, 2020, **25**, 3869–3877.
- 26 G. M. D. Miranda, A. da Cruz Gouveia Mendes and A. L. A. da Silva, *Rev. Bras. Geriatr. Gerontol.*, 2016, **19**, 507–519.
- 27 S. de Liz Sofiatti, M. M. de Oliveira, L. M. Gomes and K. V. S. Vieira, *Revista Brasileira Militar de Ciências*, 2021, **7**, 17.
- 28 R. A. Lourenço, V. G. Moreira, R. G. B. de Mello, I. de Souza Santos, S. M. Lin, A. L. F. Pinto, L. P. Lustosa, Y. A. de Oliveira Duarte, J. A. Ribeiro, C. C. Correia, H. N. Mansur, E. Ribeiro, R. R. D. Corte, E. Ferriolli, C. A. Uehara, A. Maeda, T. Petroni, T. S. Lima, S. F. Durão, I. Aprahamian, C. M. Avesani and W. J. Filho, *Geriatr. Gerontol. Aging.*, 2018, **12**, 121–135.
- 29 W. H. Organization, *WHO global report on falls*, 2007, pp. 1–47.
- 30 J. E. Lewis and M. B. Neider, *Ergon. Des.*, 2017, **25**, 4–10.
- 31 M. Wu and J. Luo, *Online Journal of Nursing Informatics Contributors*, 2019, Available at <https://www.himss.org/resources/wearable-technology-applications-healthcare-literature-review>.
- 32 N. Niknejad, W. B. Ismail, A. Mardani, H. Liao and I. Ghani, *Eng. Appl. Artif. Intell.*, 2020, **90**, 103529.
- 33 J. Li, Q. Ma, A. H. Chan and S. S. Man, *Appl. Ergon.*, 2019, **75**, 162–169.
- 34 T. G. Stavropoulos, A. Papastergiou, L. Mpaltadoros, S. Nikolopoulos and I. Kompatsiaris, *Sensors*, 2020, **20**, 2826.
- 35 A. C. Kogan, K. Wilber and L. Mosqueda, *J. Am. Geriatr. Soc.*, 2016, **64**, e1–e7.
- 36 K. Kulinski, C. DiCocco, S. Skowronski and P. Sprowls, *Front. Public Health*, 2017, **5**, 4.
- 37 D. Podsiadlo and S. Richardson, *J. Am. Geriatr. Soc.*, 1991, **39**, 142–148.
- 38 L. C. A. Andrade, G. L. dos Anjos Costa, L. G. B. Diogenes and P. H. R. Pimentel, *Res., Soc. Dev.*, 2021, **10**, e321101321615.
- 39 M. R. Kaysir, A. Stefani, R. Lwin and S. Fleming, *2017 Conference on Lasers and Electro-Optics Pacific Rim (CLEO-PR)*, 2017, pp. 1–2.



40 H.-E. Joe, H. Yun, S.-H. Jo, M. B. Jun and B.-K. Min, *Int. J. Precis. Eng. Manuf. - Green Technol.*, 2018, **5**, 173–191.

41 Z. Liu, Z. Zhang, H.-Y. Tam and X. Tao, *Photonics*, 2019, **6**, 48.

42 Z. Liu, M. Zhang, Y. Zhang, Y. Xu, Y. Zhang, X. Yang, J. Zhang, J. Yang and L. Yuan, *Sens. Actuators, A*, 2020, **313**, 112179.

43 Y. Koike and M. Asai, *NPG Asia Mater.*, 2009, **1**, 22–28.

44 M. Elsherif, A. E. Salih, M. G. Muñoz, F. Alam, B. AlQattan, D. S. Antonysamy, M. F. Zaki, A. K. Yetisen, S. Park, T. D. Wilkinson and H. Butt, *Adv. Photonics Res.*, 2022, **3**, 2100371.

45 K. Peters, *Polymer optical fiber sensors - A review*, 2011.

46 O. Strobel and J. Lubkoll, *2010 20th International Crimean Conference “Microwave Telecommunication Technology”*, 2010, pp. 16–20.

47 J. Ballato and P. Dragic, *Materials*, 2014, **7**, 4411–4430.

48 A. Leal-Junior, L. Macedo, A. Frizera and M. J. Pontes, *IEEE Photonics Technol. Lett.*, 2022, **34**, 611–614.

49 S. Liehr, P. Lenke, M. Wendt, K. Krebber, M. Seeger, E. Thiele, H. Metschies, B. Gebreselassie and J. C. Munich, *IEEE Sens. J.*, 2009, **9**, 1330–1338.

50 L. Macedo, R. W. M. Pires Junior, A. Frizera, M. J. Pontes and A. Leal-Junior, *Opt. Fiber Technol.*, 2022, **72**, 103001.

51 A. G. Leal-Junior, C. A. Diaz, L. M. Avellar, M. J. Pontes, C. Marques and A. Frizera, *Sensors*, 2019, **19**, 3156.

52 M. Rocha, P. Antunes, P. André, N. Alberto and M. F. Domingues, *2023 IEEE 7th Portuguese Meeting on Bioengineering (ENBENG)*, 2023, pp. 76–79.

53 A. G. Leal-Junior, A. Frizera, L. M. Avellar, C. Marques and M. J. Pontes, *IEEE Sens. J.*, 2018, **18**, 2362–2368.

54 A. D. Kersey, in *Distributed and Multiplexed Fiber Optic Sensors*, Wiley, 2011, pp. 277–314.

55 A. G. Leal-Junior, C. R. Díaz, C. Marques, M. J. Pontes and A. Frizera, *Opt. Laser Technol.*, 2019, **111**, 81–88.

56 L. Avellar, A. Frizera and A. Leal-Junior, *Biomed. Opt. Express*, 2023, **14**, 3689–3704.

57 K. Hill and G. Meltz, *J. Lightwave Technol.*, 1997, **15**, 1263–1276.

58 C. Giles, *J. Lightwave Technol.*, 1997, **15**, 1391–1404.

59 Y. Liu, A. Zhou, Q. Xia, Y. Zhao, H. Deng, S. Yang and L. Yuan, *IEEE Sens. J.*, 2019, **19**, 10728–10735.

60 X. Xiaoqiang, S. Ziqi, M. Yan and D. Yang, *2021 33rd Chinese Control and Decision Conference (CCDC)*, 2021, pp. 3150–3157.

61 A. Theodosiou and K. Kalli, *Opt. Fiber Technol.*, 2020, **54**, 102079.

62 Y. Qiu, Q. B. Wang, H. T. Zhao, J. A. Chen and Y. Y. Wang, *Shanghai Jiaotong Daxue Xuebao*, 2013, **18**, 129–139.

63 H. Zhou, Y. Wang, Z. Zhou and C. Ma, *Measurement*, 2023, **222**, 113565.

64 G. Kashaganova, A. Kozbakova, T. Kartbayev, G. Balbayev, K. Togzhanova, Z. Alimseitova and S. Orazaliyeva, *Electronics*, 2023, **12**(11), 2491.

65 A. Kerrouche, T. Najeh and P. Jaen-Sola, *Sensors*, 2021, **21**(11), 3639.

66 T. Li, J. Guo, Y. Tan and Z. Zhou, *IEEE Sens. J.*, 2020, **20**, 12074–12087.

67 J. Huang, J. Zeng, Y. Bai, Y. Wang, K. Wang, X. Wu, Z. Cheng and D. Liang, *Opt. Fiber Technol.*, 2020, **57**, 102216.

68 M. Liao, S. Liang, R. Luo and Y. Xiao, *Constr. Build. Mater.*, 2022, **342**, 128029.

69 H. Hisham, *Fiber bragg grating sensors: development and applications*, CRC Press, 2019.

70 J. Albert, L. Y. Shao and C. Caucheteur, *Tilted fiber Bragg grating sensors*, 2013.

71 X. Dong, H. Zhang, B. Liu and Y. Miao, *Tilted fiber bragg gratings: Principle and sensing applications*, 2011.

72 X. Chen, J. Xu, X. Zhang, T. Guo and B.-O. Guan, *IEEE Photonics Technol. Lett.*, 2017, **29**, 719–722.

73 L.-Y. Shao, L. Xiong, C. Chen, A. Laronche and J. Albert, *J. Lightwave Technol.*, 2010, **28**, 2681–2687.

74 P. Kisala, D. Harasim and J. Mroczka, *Opt. Express*, 2016, **24**, 29922–29929.

75 D. Kinet, P. Mégret, K. W. Goossen, L. Qiu, D. Heider and C. Caucheteur, *Sensors*, 2014, **14**, 7394–7419.

76 S. Korganbayev, M. De Landro, A. Wolf, D. Tosi and P. Saccamandi, *IEEE Sens. J.*, 2022, **22**, 15999–16007.

77 Y. Singh, S. K. Raghuwanshi, O. Prakash and P. K. Saini, *Opt. Quantum Electron.*, 2021, **53**, 664.

78 G. Allwood, G. Wild and S. Hinckley, *Electronics*, 2017, **6**(4), 92.

79 W. Bao, F. Chen, H. Lai, S. Liu and Y. Wang, *Sens. Actuators, B*, 2021, **349**, 130794.

80 A. Lopez Aldaba, A. González-Vila, M. Debliquy, M. Lopez-Amo, C. Caucheteur and D. Lahem, *Sens. Actuators, B*, 2018, **254**, 1087–1093.

81 C. Massaroni, M. Zaltieri, D. Lo Presti, A. Nicolò, D. Tosi and E. Schena, *IEEE Sens. J.*, 2021, **21**, 14069–14080.

82 C. Zhang, X. Chen, X. Liu, C. Shen, Z. Huang, Z. Wang, T. Lang, C. Zhao, Y. Zhang and Z. Liu, *Int. J. Hydrogen Energy*, 2022, **47**, 6415–6420.

83 B. Raju, R. Kumar, M. Senthilkumar, R. Sulaiman, N. Kama and S. Dhanalakshmi, *Sustain. Energy Technol. Assessments*, 2022, **52**, 102306.

84 W. Du, X. Tao, H. Tam and C. Choy, *Compos. Struct.*, 1998, **42**, 217–229.

85 A. Leal-Junior, L. Avellar, W. Blanc, A. Frizera and C. Marques, *IEEE Internet Things J.*, 2024, **11**, 9587–9598.

86 C. Perez-Ramirez, D. Almanza-Ojeda, J. Guerrero-Tavares, F. Mendoza-Galindo, J. Estudillo-Ayala and M. Ibarra-Manzano, *Sensors*, 2014, **14**, 24483–24501.

87 D. Vilarinho, A. Theodosiou, C. Leitão, A. Leal-Junior, M. Domingues, K. Kalli, P. André, P. Antunes and C. Marques, *Sensors*, 2017, **17**, 2924.

88 A. G. Leal-Junior, A. Frizera, A. Theodosiou, C. Díaz, M. Jimenez, R. Min, M. J. Pontes, K. Kalli and C. Marques, *IEEE Sens. J.*, 2019, **19**, 9221–9228.

89 W. Li, Y. Yuan, J. Yang and L. Yuan, *Photonic Sens.*, 2021, **11**, 31–44.

90 X. Chen, S. He, D. Li, K. Wang, Y. Fan and S. Wu, *IEEE Sens. J.*, 2016, **16**, 349–354.

91 L. Liu, P. Lu, H. Liao, S. Wang, W. Yang, D. Liu and J. Zhang, *IEEE Sens. J.*, 2016, **16**, 3054–3058.



92 Z. Tian and S. S.-H. Yam, *IEEE Photonics Technol. Lett.*, 2009, **21**, 161–163.

93 X. Wang, D. Chen, H. Li, G. Feng and J. Yang, *IEEE Sens. J.*, 2017, **17**, 100–104.

94 D. Lyu, J. Peng, Q. Huang, W. Zheng, L. Xiong and M. Yang, *IEEE Sens. J.*, 2021, **21**, 57–61.

95 Y. Chen, Y. Zheng, H. Xiao, D. Liang, Y. Zhang, Y. Yu, C. Du and S. Ruan, *Sensors*, 2022, **22**(15), 5748.

96 B. Culshaw and A. Kersey, *J. Lightwave Technol.*, 2008, **26**(9), 1064–1078.

97 Y. Wang, H. Yuan, X. Liu, Q. Bai, H. Zhang, Y. Gao and B. Jin, *IEEE Access*, 2019, **7**, 85821–85837.

98 J. Zhao, Y. Zhao, L. Bai and Y. Zhang, *Sens. Actuators, A*, 2020, **314**, 112245.

99 M. Szczerska, *Chemosensors*, 2022, **10**(6), 228.

100 S. Wu, L. Wang, X. Chen and B. Zhou, *J. Lightwave Technol.*, 2018, **36**, 2216–2221.

101 X. Li, D. Liu, R. Kumar, W. P. Ng, Y. Q. Fu, J. Yuan, C. Yu, Y. Wu, G. Zhou, G. Farrell, Y. Semenova and Q. Wu, *Meas. Sci. Technol.*, 2017, **28**, 035105.

102 R. Wang, J. Zhao, Y. Sun, H. Yu, N. Zhou, H. Zhang and D. Jia, *Biomed. Opt. Express*, 2019, **11**, 316.

103 S. Eom, J. Park and J. Lee, *Microw. Opt. Technol. Lett.*, 2010, **52**, 1318–1321.

104 C. Ke, Y. Cai, T. Zhao and Z. Li, *IOP Conf. Ser. Earth Environ. Sci.*, 2021, **769**, 042039.

105 M. Ghasemi, J. Oh, S. Jeong, M. Lee, S. Bohlooli Darian, K. Oh and J. K. Kim, *Biosensors*, 2023, **13**, 817.

106 P. Lu, N. Lalam, M. Badar, B. Liu, B. T. Chorpeling, M. P. Buric and P. R. Ohodnicki, *Appl. Phys. Rev.*, 2019, **6**, 041302.

107 X. Bao and Y. Wang, *Advanced Devices Instrumentation*, 2021, **2021**, DOI: [10.34133/2021/8696571](https://doi.org/10.34133/2021/8696571).

108 S. Adachi, *2008 SICE Annual Conference*, 2008, pp. 329–333.

109 K. Yuksel, M. Wuilpart, V. Moeyaert and P. Megret, *2009 11th International Conference on Transparent Optical Networks*, 2009, pp. 1–5.

110 Y. Muanenda, *J. Sens.*, 2018, **2018**, 1–16.

111 M. Silveira, A. Frizera, A. Leal-Junior, D. Ribeiro, C. Marques, W. Blanc and C. A. Camilo, *Opt. Fiber Technol.*, 2020, **58**, 102303.

112 L. Avellar, A. Frizera, H. Rocha, M. Silveira, C. Díaz, W. Blanc, C. Marques and A. Leal-Junior, *Photonics Res.*, 2023, **11**, 364.

113 R. Correia, S. James, S.-W. Lee, S. P. Morgan and S. Korposh, *J. Opt.*, 2018, **20**, 073003.

114 Z. Katrenova, S. Alisherov, T. Abdol and C. Molardi, *Sens. Bio-Sens. Res.*, 2024, **43**, 100616.

115 A. G. Leal-Junior, A. Frizera-Neto, M. J. Pontes and T. R. Botelho, *Meas. Sci. Technol.*, 2017, **28**, 125103.

116 S. Umesh, S. Padma, T. Srinivas and S. Asokan, *IEEE Sens. J.*, 2018, **18**, 216–222.

117 A. Leal, L. Avellar, M. J. Pontes, C. A. Díaz, C. Marques and A. Frizera Neto, *Seventh European Workshop on Optical Fibre Sensors*, 2019.

118 X. Wang, H. Zhou, M. Chen, Y. He, Z. Zhang, J. Gan and Z. Yang, *Sci. China Inf. Sci.*, 2024, **67**, 1300–1310.

119 A. Grillet, D. Kinet, J. Witt, M. Schukar, K. Krebber, F. Pirotte and A. Depre, *IEEE Sens. J.*, 2008, **8**, 1215–1222.

120 D. Tosi, C. Molardi, M. Sypabekova and W. Blanc, *IEEE Sens. J.*, 2021, **21**, 12667–12678.

121 C. Broadway, R. Min, A. G. Leal-Junior, C. Marques and C. Caucheteur, *J. Lightwave Technol.*, 2019, **37**, 2605–2615.

122 M. S. F. Soares, *PhD Thesis*, University of Aveiro, 2021.

123 J. Wang, *Coatings for Biomedical Applications*, Woodhead Publishing, 2012, pp. 143–175.

124 M. Oliverio, S. Perotto, G. C. Messina, L. Lovato and F. D. Angelis, *ACS Appl. Mater. Interfaces*, 2017, **9**, 29394–29411.

125 R. T. Hill, *WIREs Nanomed. Nanobiotechnol.*, 2014, **7**, 152–168.

126 V. Nocerino, B. Miranda, C. Tramontano, G. Chianese, P. Dardano, I. Rea and L. D. Stefano, *Chemosensors*, 2022, **10**, 150.

127 V. Naresh and N. Lee, *Sensors*, 2021, **21**, 1109.

128 P. I. Dolez, *Nanoengineering*, Elsevier, 2015, pp. 3–40.

129 M. E. Hamza, M. A. Othman and M. A. Swillam, *Biology*, 2022, **11**, 621.

130 A. Negm, M. M. R. Howlader, I. Belyakov, M. Bakr, S. Ali, M. Irandejad and M. Yavuz, *Materials*, 2022, **15**, 7289.

131 M. J. McClain, A. E. Schlather, E. Ringe, N. S. King, L. Liu, A. Manjavacas, M. W. Knight, I. Kumar, K. H. Whitmire, H. O. Everitt, P. Nordlander and N. J. Halas, *Nano Lett.*, 2015, **15**, 2751–2755.

132 M. W. Knight, N. S. King, L. Liu, H. O. Everitt, P. Nordlander and N. J. Halas, *ACS Nano*, 2013, **8**, 834–840.

133 G. V. Naik, V. M. Shalaev and A. Boltasseva, *Adv. Mater.*, 2013, **25**, 3264–3294.

134 K. M. McPeak, S. V. Jayanti, S. J. P. Kress, S. Meyer, S. Iotti, A. Rossinelli and D. J. Norris, *ACS Photonics*, 2015, **2**, 326–333.

135 A. Agrawal, S. H. Cho, O. Zandi, S. Ghosh, R. W. Johns and D. J. Milliron, *Chem. Rev.*, 2018, **118**, 3121–3207.

136 C. Park, R. Colorado, A. Jamison and T. Lee, *Reference Module in Materials Science and Materials Engineering*, Elsevier, 2016.

137 G. M. Whitesides and P. E. Laibinis, *Langmuir*, 1990, **6**, 87–96.

138 T. Yu, F. Zhong, F. Zhang, C. Ying and G. Geng, *J. Phys.: Conf. Ser.*, 2021, **2002**, 012010.

139 P. Roingeard, P.-I. Raynal, S. Eymieux and E. Blanchard, *Rev. Med. Virol.*, 2018, **29**, e02019.

140 L. Mandrile, A. M. Giovannozzi, F. Pennecchi, A. Saverino, C. Lobascio and A. M. Rossi, *Anal. Methods*, 2015, **7**, 2813–2821.

141 C. Leitão, A. Leal-Junior, A. R. Almeida, S. O. Pereira, F. M. Costa, J. L. Pinto and C. Marques, *Biotechnol. Rep.*, 2021, **29**, e00587.

142 V. Narwal, R. Deswal, B. Batra, V. Kalra, R. Hooda, M. Sharma and J. Rana, *Steroids*, 2019, **143**, 6–17.

143 C.-J. Huang, G. V. Narasimha, Y.-C. Chen, J.-K. Chen and G.-C. Dong, *Biosensors*, 2021, **11**, 219.

144 Y. Tang, T. J. Hardy and J.-Y. Yoon, *Biosens. Bioelectron.*, 2023, **234**, 115361.



145 S. Shukla, P. Govender and A. Tiwari, *Advances in Biomembranes and Lipid Self-Assembly*, Elsevier, 2016, pp. 143–161.

146 H. H. Nguyen, S. H. Lee, U. J. Lee, C. D. Fermin and M. Kim, *Materials*, 2019, **12**, 121.

147 A. Hasan, M. Nurunnabi, M. Morshed, A. Paul, A. Polini, T. Kuila, M. A. Hariri, Y. Kyu Lee and A. A. Jaffa, *BioMed Res. Int.*, 2014, **2014**, 1–18.

148 E. C. Wilkison, K. L. Singampalli, J. Li, D. D. Dixit, X. Jiang, D. H. Gonzalez and P. B. Lillehoj, *Anal. Bioanal. Chem.*, 2023, **415**, 3983–4002.

149 D. W. G. Morrison, M. R. Dokmeci, U. Demirci and A. Khademhosseini, Clinical Applications of Micro- and Nanoscale Biosensors, in *Biomedical Nanostructures*, ed. K. E. Gonsalves, C. R. Halberstadt, C. T. Laurencin and L. S. Nair, 2007, DOI: [10.1002/9780470185834.ch17](https://doi.org/10.1002/9780470185834.ch17).

150 Y.-C. Zhu, L.-P. Mei, Y.-F. Ruan, N. Zhang, W.-W. Zhao, J.-J. Xu and H.-Y. Chen, *Advances in Enzyme Technology*, Elsevier, 2019, pp. 201–223.

151 D. J. Monk and D. R. Walt, *Anal. Bioanal. Chem.*, 2004, **379**, 931–945.

152 M. Campàs, B. Prieto-Simón and J.-L. Marty, *Semin. Cell Dev. Biol.*, 2009, **20**, 3–9.

153 R. C. Rodrigues, Á. Berenguer-Murcia and R. Fernandez-Lafuente, *Adv. Synth. Catal.*, 2011, **353**, 2216–2238.

154 K. Sadani, P. Nag, X. Thian and S. Mukherji, *Biosens. Bioelectron.: X*, 2022, **12**, 100278.

155 Z. Zhou, Z. Yang, L. Xia and H. Zhang, *Biosens. Bioelectron.*, 2022, **198**, 113836.

156 N. Gupta, V. Renugopalakrishnan, D. Liepmann, R. Paulmurugan and B. D. Malhotra, *Biosens. Bioelectron.*, 2019, **141**, 111435.

157 M. E. Inda, M. Mimee and T. K. Lu, *J. Allergy Clin. Immunol.*, 2019, **144**, 645–647.

158 L. Su, W. Jia, C. Hou and Y. Lei, *Biosens. Bioelectron.*, 2011, **26**, 1788–1799.

159 N. Lobsiger and W. J. Stark, *Anal. Sci.*, 2019, **35**, 839–847.

160 I. B. Tahirbegi, J. Ehgartner, P. Sulzer, S. Zieger, A. Kasjanow, M. Paradiso, M. Strobl, D. Bouwes and T. Mayr, *Biosens. Bioelectron.*, 2017, **88**, 188–195.

161 N. Gupta, V. Renugopalakrishnan, D. Liepmann, R. Paulmurugan and B. D. Malhotra, *Biosens. Bioelectron.*, 2019, **141**, 111435.

162 G. Caluori, J. Pribyl, M. Pesl, S. Jelinkova, V. Rotrekl, P. Skladal and R. Raiteri, *Biosens. Bioelectron.*, 2019, **124–125**, 129–135.

163 B. Byrne, E. Stack, N. Gilmartin and R. O'Kennedy, *Sensors*, 2009, **9**, 4407–4445.

164 H. W. Schroeder and L. Cavacini, *J. Allergy Clin. Immunol.*, 2010, **125**, S41–S52.

165 S. Sharma, H. Byrne and R. J. O'Kennedy, *Essays Biochem.*, 2016, **60**, 9–18.

166 A. Makaraviciute and A. Ramanaviciene, *Biosens. Bioelectron.*, 2013, **50**, 460–471.

167 S. Ucci, S. Spaziani, G. Quero, P. Vaiano, M. Principe, A. Micco, A. Sandomenico, M. Ruvo, M. Consales and A. Cusano, *Biosensors*, 2022, **12**(11), 1040.

168 M. Pisco and A. Cusano, *Sensors*, 2020, **20**(17), 4705.

169 E. M. McConnell, J. Nguyen and Y. Li, *Front. Chem.*, 2020, **8**, DOI: [10.3389/fchem.2020.00434](https://doi.org/10.3389/fchem.2020.00434).

170 S. Song, L. Wang, J. Li, C. Fan and J. Zhao, *TrAC, Trends Anal. Chem.*, 2008, **27**, 108–117.

171 S. Tombelli, M. Minunni and M. Mascini, *Biomol. Eng.*, 2007, **24**, 191–200.

172 T. Hermann and D. J. Patel, *Science*, 2000, **287**, 820–825.

173 J. Arlett, E. Myers and M. Roukes, *Nat. Nanotechnol.*, 2011, **6**, 203–215.

174 F. V. Oberhaus, D. Frense and D. Beckmann, *Biosensors*, 2020, **10**, 45.

175 W. Ning, S. Hu, C. Zhou, J. Luo, Y. Li, C. Zhang, Z. Luo and Y. Li, *Anal. Chim. Acta*, 2023, **1278**, 341733.

176 Y. Zheng, C. Liu, N. Y. G. Lai, Q. Wang, Q. Xia, X. Sun and S. Zhang, *Front. Bioeng. Biotechnol.*, 2023, **11**, DOI: [10.3389/fbioe.2023.1190211](https://doi.org/10.3389/fbioe.2023.1190211).

177 V. Naresh and N. Lee, *Sensors*, 2021, **21**(4), 1109.

178 L. Farzin, M. Shamsipur, L. Samandari and S. Sheibani, *Microchim. Acta*, 2018, **185**, 276.

179 C. Chen, R. Wang, X. Chen and S. Liu, *Optics Frontiers Online 2020: Distributed Optical Fiber Sensing Technology and Applications*, 2021, p. 116070O.

180 L. Xie, X. Zi, Q. Meng, Z. Liu and L. Xu, *Sensors*, 2019, **19**(7), 1656.

181 C. Jesús, F. Socorro, H. J. R. D. Rivera and M. R. D. Rivera, *J. Therm. Anal. Calorim.*, 2014, **116**, 151–155.

182 A. V. Tregubov, V. V. Prikhodko, A. S. Alekseyev, S. G. Novikov, G. V. Tertyshnikova and A. V. Zhukov, *IEEE Sens. Lett.*, 2022, **6**, 1–4.

183 J. Rife, M. Miller, P. Sheehan, C. Tamanaha, M. Tondra and L. Whitman, *Sens. Actuators, A*, 2003, **107**, 209–218.

184 Y. Qi and F.-R. Xiu, *J. Lumin.*, 2016, **171**, 238–245.

185 N. Wang, N. Zhang, T. Wang, F. Liu, X. Wang, X. Yan, C. Wang, X. Liu, P. Sun and G. Lu, *Sens. Actuators, B*, 2022, **350**, 130854.

186 A. T. Güntner, I. C. Weber, S. Schon, S. E. Pratsinis and P. A. Gerber, *Sens. Actuators, B*, 2022, **367**, 132182.

187 S. G. Song, S. Ha, H. J. Cho, M. Lee, D. Jung, J. H. Han and C. Song, *ACS Appl. Nano Mater.*, 2019, **2**, 109–117.

188 R. Amali, H. Lim, I. Ibrahim, Z. Zainal and S. Ahmad, *J. Electroanal. Chem.*, 2022, **918**, 116440.

189 L. Cheng, W. Zheng, Y.-N. Zhang, X. Li and Y. Zhao, *IEEE Trans. Instrum. Meas.*, 2023, **72**, 1–7.

190 M. Govindasamy, C. R. Jian, C. F. Kuo, A. H. Hsieh, J. L. Sie and C. H. Huang, *Microchim. Acta*, 2022, **189**, 374.

191 Z. Gu, A. Fu, L. Ye, K. Kuerban, Y. Wang and Z. Cao, *ACS Sens.*, 2019, **4**, 2922–2929.

192 W. Weng, J. Liu, C. Yin, H. Xie, G. Luo, W. Sun and G. Li, *Int. J. Electrochem. Sci.*, 2019, **14**, 4309–4317.

193 O. Parlak, S. T. Keene, A. Marais, V. F. Curto and A. Salleo, *Sci. Adv.*, 2018, **4**, eaar2904.

194 S. Qazi and K. Raza, *Smart Biosensors in Medical Care*, Academic Press, 2020, pp. 65–85.

195 A. A. Karyakin, *Sensors for Environment, Health and Security*, Dordrecht, 2009, pp. 255–265.

196 C. Gonçalves, A. F. da Silva, J. Gomes and R. Simoes, *Inventions*, 2018, **3**, 1–13.



197 F. Narita and M. Fox, *Adv. Eng. Mater.*, 2018, **20**, 1–22.

198 S. Khalid, I. Raouf, A. Khan, N. Kim and H. S. Kim, *International Journal of Precision Engineering and Manufacturing-Green Technology*, 2019, **6**, 821–851.

199 Y. Zhang, J. Zhou, Y. Zhang, D. Zhang, K. T. Yong and J. Xiong, *Adv. Sci.*, 2022, **9**, 1–28.

200 S. Zhu, Z. Fan, B. Feng, R. Shi, Z. Jiang, Y. Peng, J. Gao, L. Miao and K. Koumoto, *Energies*, 2022, **15**, 1–27.

201 T. Lu, S. Ji, W. Jin, Q. Yang, Q. Luo and T.-L. Ren, *Sensors*, 2023, **23**, 2991.

202 S. Naval, A. Jain and D. Mallick, *Eng. Res. Express*, 2022, **4**, 045022.

203 A. Toprak and O. Tigli, *Appl. Phys. Rev.*, 2014, **1**, 031104.

204 L. Wang, Z. Fei, Y. Qi, C. Zhang, L. Zhao, Z. Jiang and R. Maeda, *ACS Appl. Energy Mater.*, 2022, **5**, 7091–7114.

205 M. Pozzi, M. S. H. Aung, M. Zhu, R. K. Jones and J. Y. Goulermas, *Smart Mater. Struct.*, 2012, **21**, 075023.

206 C. Dagdeviren, B. D. Yang, Y. Su, P. L. Tran, P. Joe, E. Anderson, J. Xia, V. Doraiswamy, B. Dehdashti, X. Feng, B. Lu, R. Poston, Z. Khalpey, R. Ghaffari, Y. Huang, M. J. Slepian and J. A. Rogers, *Proc. Natl. Acad. Sci. U. S. A.*, 2014, **111**, 1927–1932.

207 S. Niu, X. Wang, F. Yi, Y. S. Zhou and Z. L. Wang, *Nat. Commun.*, 2015, **6**, 8975.

208 G. Conta, A. Libanori, T. Tat, G. Chen and J. Chen, *Adv. Mater.*, 2021, **33**, 2007502.

209 S. Niu, S. Wang, L. Lin, Y. Liu, Y. S. Zhou, Y. Hu and Z. L. Wang, *Energy Environ. Sci.*, 2013, **6**, 3576–3583.

210 J. Qi and L. A. Garza, *Cold Spring Harb. Perspect. Med.*, 2014, **4**, a013615.

211 Z. Zhao, Y. Lu, Y. Mi, J. Meng, X. Wang, X. Cao and N. Wang, *Biosensors*, 2022, **12**, 1127.

212 A. Páez-montoro, M. García-valderas, E. Olías-ruíz and C. López-ongil, *Sensors*, 2022, **22**(10), 3950.

213 T. Zhang, Z. Wen, Y. Liu, Z. Zhang, Y. Xie and X. Sun, *iScience*, 2020, **23**, 101813.

214 C. Xu, Y. Song, M. Han and H. Zhang, *Microsyst. Nanoeng.*, 2021, **7**, 25.

215 A. Ali, H. Shaukat, S. Bibi, W. A. Altabey, M. Noori and S. A. Kouritem, *Energy Strat. Rev.*, 2023, **49**, 101124.

216 M. N. Hasan, M. Nafea, N. Nayan and M. S. Mohamed Ali, *Adv. Mater. Technol.*, 2022, **7**, 2101203.

217 A. J. Bandodkar, S. P. Lee, I. Huang, W. Li, S. Wang, C. J. Su, W. J. Jeang, T. Hang, S. Mehta, N. Nyberg, P. Gutruf, J. Choi, J. Koo, J. T. Reeder, R. Tseng, R. Ghaffari and J. A. Rogers, *Nat. Electron.*, 2020, **3**, 554–562.

218 B. Lu, Y. Chen, D. Ou, H. Chen, L. Diao, W. Zhang, J. Zheng, W. Ma, L. Sun and X. Feng, *Sci. Rep.*, 2015, **5**, 16065.

219 Z. Zhou, L. Weng, T. Tat, A. Libanori, Z. Lin, L. Ge, J. Yang and J. Chen, *ACS Nano*, 2020, **14**, 14126–14133.

220 Y. Kuang, Z. Yang and M. Zhu, *Smart Mater. Struct.*, 2016, **25**, 085029.

221 M. P. Sikka, S. Kumar and S. Garg, *2023 Joint International Conference on Digital Arts, Media and Technology with ECTI Northern Section Conference on Electrical, Electronics, Computer and Telecommunications Engineering (ECTI DAMT NCON)*, 2023, pp. 511–516.

222 M. Popescu and C. Ungureanu, *Materials*, 2023, **16**(11), 4075.

223 I. I. Shuvo and P. I. Dolez, *Functional and Technical Textiles*, Woodhead Publishing, 2023, pp. 333–372.

224 M. K. Kangazi and A. A. Merati, *Advances in Healthcare and Protective Textiles*, Woodhead Publishing, 2023, pp. 23–56.

225 J. Yin, S. Wang, A. Di Carlo, A. Chang, X. Wan, J. Xu, X. Xiao and J. Chen, *Med-X*, 2023, **1**, 3.

226 H. Li, R. Fan, B. Zou, J. Yan, Q. Shi and G. Guo, *J. Nanobiotechnol.*, 2023, **21**, 73.

227 A. Leal-Junior and A. Frizera-Neto, *Optical Fiber Sensors for the Next Generation of Rehabilitation Robotics*, Elsevier, 2022.

228 S. G. Farag, *2019 4th MEC International Conference on Big Data and Smart City (ICBDSC)*, 2019, pp. 1–5.

229 S. Javadinasab Hormozabad, M. Gutierrez Soto and H. Adeli, *Expert Syst.*, 2021, **38**(8), e12845.

230 E. Yarali, M. Baniasadi, A. Zolfagharian, M. Chavoshi, F. Arefi, M. Hossain, A. Bastola, M. Ansari, A. Foyouzat, A. Dabbagh, M. Ebrahimi, M. J. Mirzaali and M. Bodaghi, *Appl. Mater. Today*, 2022, **26**, 101306.

231 M. Ramezani and Z. Mohd Ripin, *J. Funct. Biomater.*, 2023, **14**(7), 347.

232 A. Leal-Junior, L. Avellar, A. Frizera and C. Marques, *Sci. Rep.*, 2020, **10**, 13867.

233 A. Issatayeva, A. Beisenova, D. Tosi and C. Molardi, *Sensors*, 2020, **20**, 1–16.

234 Z. Gong, Z. Xiang, X. OuYang, J. Zhang, N. Lau, J. Zhou and C. C. Chan, *Materials*, 2019, **12**(20), 3311.

235 L. G. Gomes, R. de Mello and A. Leal-Junior, *Opt. Fiber Technol.*, 2023, **78**, 103313.

236 A. G. Leal-Junior, C. R. Díaz, M. F. Jiménez, C. Leitão, C. Marques, M. J. Pontes and A. Frizera, *IEEE Sens. J.*, 2019, **19**, 567–574.

237 A. G. Leal-Junior, C. R. Díaz, C. Leitão, M. J. Pontes, C. Marques and A. Frizera, *Opt. Laser Technol.*, 2019, **109**, 429–436.

238 B. Najafi, H. Mohseni, G. S. Grewal, T. K. Talal, R. A. Menzies and D. G. Armstrong, *J. Diabetes Sci. Technol.*, 2017, **11**, 668–677.

239 T. Li, F. Qiao, P. Huang, Y. Su, L. Wang, X. Li, H. Li, Y. Tan and Z. Zhou, *IEEE Sens. J.*, 2022, **22**, 19336–19345.

240 Y. W. Sasy Chan, H.-P. Wang and P. Xiang, *Symmetry*, 2021, **13**(12), 2251.

241 R. de Oliveira, O. Frazão, J. Santos and A. Marques, *Comput. Struct.*, 2004, **82**, 1315–1321.

242 H. Rocha, C. Semprimoschnig and J. P. Nunes, *Eng. Struct.*, 2021, **237**, 112231.

243 S. Minakuchi and N. Takeda, *Photonic Sens.*, 2013, **3**, 345–354.

244 C. Guignier, B. Camillieri, M. Schmid, R. M. Rossi and M.-A. Bueno, *Sensors*, 2019, **19**, 3011.

245 L. Avellar, G. Delgado, C. Marques, A. Frizera, A. Leal-Junior and E. Rocon, *IEEE Sens. J.*, 2021, **21**, 20078–20085.

246 K. Xiao, Z. Wang, Y. Ye, C. Teng and R. Min, *Biomed. Opt. Express*, 2024, **15**, 1892.



247 W. Q. Mok, W. Wang and S. Y. Liaw, *Int. J. Nurs. Pract.*, 2015, **21**, 91–98.

248 W. H. Organization, Ageing and health — who.int, [https://www.who.int/news-room/fact-sheets/detail/ageing-and-health#:~:text=By%202030%2C%201%20in%206,will%20double%20\(2.1%20billion\)](https://www.who.int/news-room/fact-sheets/detail/ageing-and-health#:~:text=By%202030%2C%201%20in%206,will%20double%20(2.1%20billion)), 2022, [Accessed 14 mar 2024].

249 H. Suominen, *Eur. Rev. Aging Phys. Act.*, 2010, **8**, 37–42.

250 E. Batista, M. A. Moncusi, P. López-Aguilar, A. Martínez-Ballesté and A. Solanas, *Sensors*, 2021, **21**, 6886.

251 W. Lyu, S. Chen, F. Tan and C. Yu, *Photonics*, 2022, **9**, 50.

252 Y. Haseda, J. Bonefacino, H.-Y. Tam, S. Chino, S. Koyama and H. Ishizawa, *Sensors*, 2019, **19**, 5088.

253 A. Leal-Junior, L. Avellar, A. Frizera and C. Marques, *Sci. Rep.*, 2020, **10**, 13867.

254 E. P. Córcoles and M. G. Boutelle, *Biosensors and Invasive Monitoring in Clinical Applications*, Springer International Publishing, 2013.

255 G. Rong, S. R. Corrie and H. A. Clark, *ACS Sens.*, 2017, **2**, 327–338.

256 P. C. Pandey, G. Pandey and R. J. Narayan, *Front. Bioeng. Biotechnol.*, 2020, **8**, 894.

257 D. Ma, S. Chon, S. Cho, Y. Lee, M. Yoo, D. Kim, D. Y. Lee and J. K. Lim, *J. Electroanal. Chem.*, 2020, **876**, 114720.

258 J. Lietard, D. Ameur, M. J. Damha and M. M. Somoza, *Angew. Chem., Int. Ed.*, 2018, **57**, 15257–15261.

259 S. Fruncillo, X. Su, H. Liu and L. S. Wong, *ACS Sens.*, 2021, **6**, 2002–2024.

260 R. Raman and R. Bashir, *Essentials of 3D Biofabrication and Translation*, Elsevier, 2015, pp. 89–121.

261 G. Remaggi, A. Zaccarelli and L. Elviri, *Chemosensors*, 2022, **10**, 65.

262 F. Virgilio, M. Prasciolu, P. Ugo and M. Tormen, *Microelectron. Eng.*, 2013, **111**, 320–324.

263 N. A. Cinel, S. Butun and E. Özbay, *Opt. Express*, 2012, **20**, 2587.

264 S. Singh, J. Wang and S. Cinti, *ECS Sens. Plus*, 2022, **1**, 023401.

265 M. U. Ahmed, M. M. Hossain, M. Safavieh, Y. L. Wong, I. A. Rahman, M. Zourob and E. Tamiya, *Crit. Rev. Biotechnol.*, 2015, 1–11.

266 D. Albanese, M. D. Matteo and C. Alessio, *Computer Aided Chemical Engineering*, Elsevier, 2010, pp. 283–288.

267 F. Arduini, L. Micheli, D. Moscone, G. Palleschi, S. Piermarini, F. Ricci and G. Volpe, *TrAC, Trends Anal. Chem.*, 2016, **79**, 114–126.

268 A. Hussain, N. Abbas and A. Ali, *Chemosensors*, 2022, **10**, 103.

269 V. Beedasy and P. J. Smith, *Materials*, 2020, **13**, 704.

270 J. Li, M. C. Lemme and M. Östling, *ChemPhysChem*, 2014, **15**, 3427–3434.

271 K. Zub, S. Hoeppener and U. S. Schubert, *Adv. Mater.*, 2022, **34**, 2105015.

272 T. Lu, S. Ji, W. Jin, Q. Yang, Q. Luo and T.-L. Ren, *Sensors*, 2023, **23**(6), 2991.

273 E. De la Paz, N. H. Maganti, A. Trifonov, I. Jeerapan, K. Mahato, L. Yin, T. Sonsa-ard, N. Ma, W. Jung, R. Burns, A. Zarrinpar, J. Wang and P. P. Mercier, *Nat. Commun.*, 2022, **13**, 7405.

274 K. Kalantar-zadeh, N. Ha, J. Z. Ou and K. J. Berean, *ACS Sens.*, 2017, **2**, 468–483.

275 K. Kalantar-Zadeh, K. J. Berean, N. Ha, A. F. Chrimes, K. Xu, D. Grando, J. Z. Ou, N. Pillai, J. L. Campbell, R. Brkljača, K. M. Taylor, R. E. Burgell, C. K. Yao, S. A. Ward, C. S. Mc-Sweeney, J. G. Muir and P. R. Gibson, *Nat. Electron.*, 2018, **1**, 79–87.

276 K. Oliphant and E. Allen-Vercoe, *Microbiome*, 2019, **7**(1), 91.

277 J. P. Geibel and M. Physiology, *World J. Gastroenterol.*, 2005, **11**, 5259–5265.

278 C. Cheng, Y. Wu, X. Li, Z. An, Y. Lu, F. Zhang, B. Su and Q. Liu, *Sens. Actuators, B*, 2021, **349**, 130781.

279 W. Xu, H. Yang, W. Zeng, T. Houghton, X. Wang, R. Murthy, H. Kim, Y. Lin, M. Mignolet, H. Duan, H. Yu, M. Slepian and H. Jiang, *Adv. Mater. Technol.*, 2017, **2**, 1700181.

280 S. Mukherjee, S. Suleman, R. Pilloton, J. Narang and K. Rani, *Sensors*, 2022, **22**(11), 4228.

281 D. Smith and S. Jheeta, *Gastrointestinal Disorders*, 2020, **2**, 3–11.

282 S. N. Adler, Y. Metzger, C. Misrachi and R. Klar, *Gastrointest. Endosc.*, 2007, **65**, AB165.

283 P. L. Rajbhandary and G. Nallathambi, 2020 *42nd Annual International Conference of the IEEE Engineering in Medicine Biology Society (EMBC)*, 2020, pp. 4652–4655.

284 M. Brun, W. Michalek, B. Surjanhata and B. Kuo, *Gastroenterology*, 2011, **140**, S-865.

285 P. J. van der Schaar, F. Dijksman, J. Shimizu, C. Wanke and P. D. Siersema, *Gastroenterology*, 2011, **140**, S-766.

286 C. J. Bettinger, *Angew. Chem.*, 2018, **130**, 17190–17203.

287 P. Nadeau, D. El-Damak, D. Glettig, Y. L. Kong, S. Mo, C. Cleveland, L. Booth, N. Roxhed, R. Langer, A. P. Chandrasekaran and G. Traverso, *Nat. Biomed. Eng.*, 2017, **1**, 0022.

288 Y. J. Kim, W. Wu, S. E. Chun, J. F. Whitacre and C. J. Bettinger, *Proc. Natl. Acad. Sci. U. S. A.*, 2013, **110**, 20912–20917.

289 C. Dagdeviren, Z. Li and Z. L. Wang, *Annu. Rev. Biomed. Eng.*, 2017, **19**, 85–108.

