Chapter

# Fiber Bragg Gratings as e-Health Enablers: An Overview for Gait Analysis Applications

*Maria de Fátima Domingues, Cátia Tavares, Tiago Leite, Nélia Alberto, Cátia Leitão, Carlos Marques, Ayman Radwan, Eduardo Rocon, Paulo Antunes and Paulo André*

## Abstract

Nowadays, the fast advances in sensing technologies and ubiquitous wireless networking are reflected in medical practice. It provides new healthcare advantages under the scope of e-Health applications, enhancing life quality of citizens. The increase of life expectancy of current population comes with its challenges and growing health risks, which include locomotive problems. Such impairments and its rehabilitation require a close monitoring and continuous evaluation, which add financial burdens on an already overloaded healthcare system. Analysis of body movements and gait pattern can help in the rehabilitation of such problems. These monitoring systems should be noninvasive and comfortable, in order to not jeopardize the mobility and the day-to-day activities of citizens. The use of fiber Bragg gratings (FBGs) as e-Health enablers has presented itself as a new topic to be investigated, exploiting the FBGs' advantages over its electronic counterparts. Although gait analysis has been widely assessed, the use of FBGs in biomechanics and rehabilitation is recent, with a wide field of applications. This chapter provides a review of the application of FBGs for gait analysis monitoring, namely its use in topics such as the monitoring of plantar pressure, angle, and torsion and its integration in rehabilitation exoskeletons and for prosthetic control.

**Keywords:** fiber Bragg gratings, e-Health enablers, gait analysis, plantar pressure, foot shear pressure, gait joint monitoring, instrumentation of prosthetic limbs

## 1. Fiber Bragg gratings: an introduction

Fiber Bragg gratings (FBGs) are sensing elements based on the longitudinal modulation of the refractive index of the optical fiber core. This type of device has all the advantages associated with optical fiber sensors, with the added feature of easily multiplexing several sensing points along one single fiber.

The production methodology of FBGs has evolved significantly since its initial approach. In the late 1970s, it was shown that optical fibers can be photosensitive, opening the door for FBGs production and its applications, both as sensing devices and in optical communications [1]. In 1981, Lam and Garside suggested that the formation of the FBGs was related to the interaction between UV light with defects in the doped silica core. Such findings lead to the later confirmation that the refractive index changes could be induced by doping the optical fibers core with germanium, given a new insight on the FBGs production [2, 3].

One decade has passed since new breakthroughs emerged regarding the FBGs production methodology. In 1989, Meltz et al. reported an FBG external inscription technique. The authors used a split 244 nm beam, which was later recombined in order to produce an interference pattern in the optical fiber core [4, 5]. With this technique, the authors were able to create a periodic and permanent change in the optical fiber core refractive index [5]. The reflected Bragg wavelength can be adjusted by changing the angle between the two split beams. In that way, the period of the interference pattern and the refractive index will change accordingly.

Alternatively, FBGs can be inscribed using phase masks, which are periodic patterns usually etched onto fused silica. In this technique, when the radiation from a UV laser is incident in the phase mask, the diffracted orders +1 and −1 are maximized, while the remaining ones are suppressed, creating an interferometric reflective pattern along the optical fiber core [6]. In **Figure 1**, the FBG inscription based on the phase mask technique as well as a representation of the FBG sensing mechanism is shown.

The FBG operational principle consists in monitoring the Bragg wavelength ($\lambda_{Bragg}$) shift reflected by the grating, as a function of the monitored parameter. The Bragg wavelength is dependent on the effective refractive index of the fiber core ($n_{eff}$) and the grating period ($\Lambda$) by the relation [4]:

$$\lambda_{Bragg} = 2 n_{eff} \Lambda \tag{1}$$

Therefore, the Bragg wavelength can be actuated by variations in the grating period or in optical fiber core effective refractive index. So, the Bragg wavelength dependence on strain and temperature can be translated by:

$$\Delta\lambda_{Bragg} = \lambda_{Bragg}(1 - \rho\alpha)\Delta\varepsilon + \lambda_{Bragg}(\alpha + \xi)\Delta T, \tag{2}$$

where the first term refers to the strain influence on the $\lambda_{Bragg}$ and the second describes the temperature effect. Hence, in Eq. (2), $\Delta\lambda_{Bragg}$ represents the shift of the Bragg wavelength, while $\rho$, $\alpha$, and $\xi$ are the photoelastic, thermal expansion,

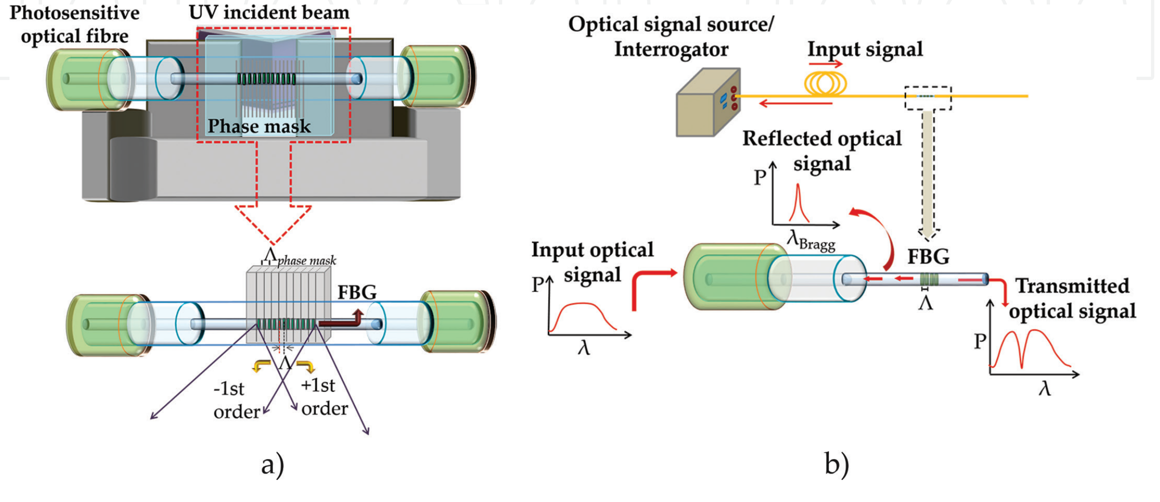


**Figure 1.**
*(a) Schematic representation of the setup typically used to inscribe FBG sensors in photosensitive optical fiber using the phase mask methodology; and (b) working principle of an FBG sensor.*

and thermo-optic coefficients of the fiber, respectively; $\Delta\varepsilon$ and $\Delta T$ corresponds to strain and temperature variations.

The FBG sensing mechanism comprises of a spectral broadband optical signal launched into the fiber, and an optical spectra analyzer to monitor the Bragg wavelength shifts. At the grating region, the Bragg wavelength component of the spectrum will be reflected, while in the transmitted optical signal that same Bragg wavelength component will be missing, as illustrated in **Figure 1b**.

Based in the described mechanisms, FBG sensors have a wide field of applications that range from their use for structural health monitoring, in oil and aeronautic industry and also as biomedical sensors and e-Health enablers, among others. Moreover, as the FBGs are elements with only few millimeters long, several gratings can be inscribed along the same optical fiber, allowing to multiplex a diverse network of sensing elements.

## 2. Gait analysis: relevance and impact in an e-Health scenario

Gait analysis research was given a pilot role in the nineteenth century, when the study of gait parameters started to be relevant in sports and medicine [7]. Regarding the medical point of view, from gait pattern analysis, a change in its normal parameters can reveal key information on patient's quality of life and/or in the evolution of different diseases. Gait disorders affect a large number of world population, since they are direct consequence of neurodegenerative diseases, such as spinal amyotrophic, multiple sclerosis, amyotrophic lateral sclerosis, neuromuscular diseases, cerebrovascular and cardiovascular pathologies, or even the physiological aging process [8–12]. Neurodegenerative diseases can be reflected in gait by showing a poor balance, a slower pace, shorter steps, lower free speed, and higher cadence [8–11].

The study of dynamic characteristics of human gait for clinical purposes has been widely reported lately. It aims to enhance the life's quality of patients suffering from gait disorders, and also, for their early detection, to enable early diagnosis and an adaptable treatment according to the evolution of the diseases or disorders [7, 13–16].

### 2.1 Gait analysis: gait cycle pattern

Gait analysis can be seen as the comprehensive study of the human locomotion, which as previously mentioned, has a major role in physical rehabilitation assessment, disorder diagnosis, surgical decision, and recovering follow up. Such study comprises the kinematic analysis (joint angles, angular velocities, and accelerations) and the kinetic analysis (ground reaction and joint forces) during the gait cycles [17, 18].

One gait cycle is the period of time between two consecutive contacts of the heel of the same foot with the floor. Generally, a cycle can be divided in two major phases: the stance phase, corresponding to the period in contact with the ground, which lasts for ~60% of the cycle; and the swing phase, corresponding to the period when there is no contact with the floor, and has a duration of ~40% of the total gait cycle [12, 19]. In **Figure 2**, the different phases are illustrated, along with events and periods that characterize a gait cycle.

The gait cycle can be further subdivided into six periods and eight functional events, five during the stance phase and three in the swing phase. Considering only one limb, the stance phase encompasses three different support periods. The first consists in a period of a double support, which is followed by single

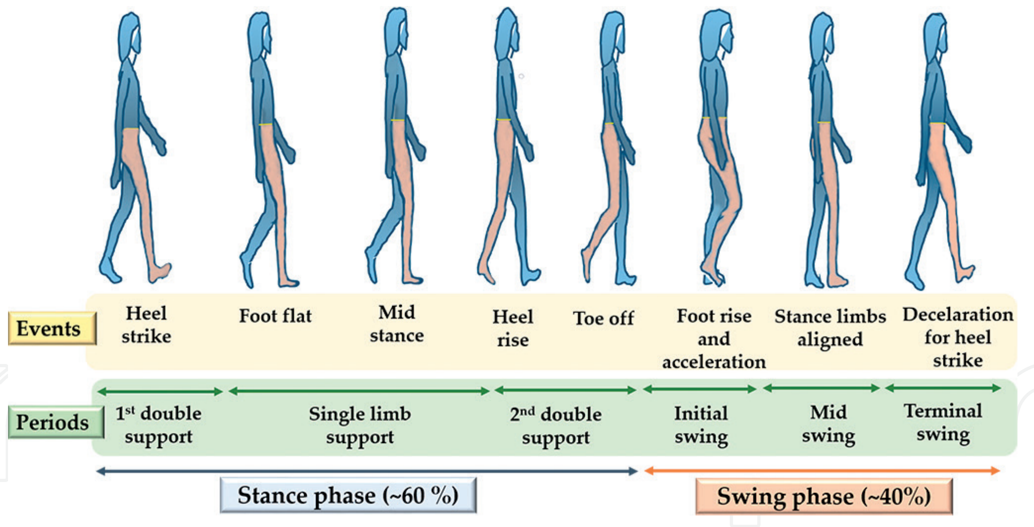


**Figure 2.**
*Representation of the stance and swing phases of a gait cycle.*

support and ends with the second double support period [18–20]. The double support period corresponds to the percentage of the cycle when both feet are simultaneously in contact with the floor and it describes the smooth transition between the left and the right single limbs support [18]. During the first double support, the heel strikes the floor (heel strike), marking the beginning of the gait cycle. The cycle evolves then toward the single period support, with the foot moving down toward the floor into a foot-flat position, where a stable support base is created for the rest of the body. Within the single support phase, the body is propelled over the foot, with the hip joint vertically aligned with ankle joint in the event characterized as the mid stance. From that point onward, the second double support phase starts, with the lower limb moving the body center of mass forward during the heel rise event, where the heel loses contact with the floor. The last contact of the foot with the floor is made by the big toe (hallux), at the toe off event, which also marks the end of the stance phase and the beginning of the swing phase [20].

During the swing phase, there is no contact between the plantar foot and the floor, and the limb continues its movement forward, which can be divided into three different periods: initial swing, mid swing, and terminal swing. In the initial swing, the lower limb vertical length should be reduced, for the foot to clear the floor and to accelerate forward by flexing the hip and knee, together with ankle dorsiflexion. The mid swing is characterized by the alignment of the accelerating limb with the stance limb. In this phase, the ankle and the hip joints are aligned. During the terminal swing, the limb undergoes a deceleration while it prepares for the contact with the floor, in the heel strike of the start of a new cycle [19–21]. As described, the swing phase is characterized by accelerations and decelerations of the lower limb, which require a more demanding muscular effort at the hip level [18].

## 2.2 Gait parameters

Gait analysis is a systematic procedure that allows the detection of negative deviations from normal gait pattern, as well as their causes. Based on such analysis, it is possible to quantify the parameters involved in the movement of the lower limbs and retrieve the mechanisms that rule the human body movement [22]. Based on the gait cycle pattern described earlier, there are several parameters that can be physically monitored in order to assess the patient's health: anthropometric, spatio-temporal, kinematic, kinetic, and dynamic electromyography (EMG), as shown in **Table 1** [22]. From such parameters, the ones that require a more specialized technology to be monitored outside the clinical environment, and therefore passible of being monitored in a gait e-Health architecture, are [7, 23, 24]:

- the stance and the swing phases duration for each foot;
- the walking velocity and gait cadence (number of steps per unit of time);
- The step length, width (distance between two equivalent points of both feet), and angles (direction of the foot during gait);
- the body posture (bending and symmetry) and the existence of tremors;
- the shear and the foot plantar pressure during the stance phase; and
- the direction and alignment of the limb segments with the ankle, knee, and hip joints.

The act of walking implies the movement of the whole body, and specifically, it requires a synchronized movement of each lower limb apart. Therefore, the gait pattern of an individual can be affected by a disorder in any segment of the body, like for instance, problems in the spinal cord or from a reduced knee flexion in patients with an anterior cruciate ligament reconstruction [25]. For that reason, the analysis of the gait cycle is a vital tool for the biomechanical mobility monitoring, as it can give crucial information not only about the lower limbs health condition, but also allows to infer details about other possible pathologies related to the dynamic movement of the body [26]. So, by monitoring the parameters previously listed, it is possible to assess the health conditions for the body parts involved in walking, namely the lower limbs and its joint. These parameters can be analyzed using objective and subjective techniques [7, 27, 28].

The subjective analysis is based on the observation of the patient while walking, and is generally performed in clinical environment under the supervision

| Parameters | Definition | Evaluation of: |
|---|---|---|
| Anthropometric | Related to the corporal dimensions of the human body. | Age, gender, height, weight, limb length, and body mass index. |
| Spatio-temporal | General gait parameters used for a simple objective gait evaluation, considering the time-distance characteristics. | Step and stride length, step width, cadence, velocity, stance and swing phases, and gait cycle events (for instance, heel strike). |
| Kinematic | Quantification of movements and geometric description of the lower limbs motion, without reference to forces. | Joint and segments angles, angular motion, acceleration, and segment trajectory. |
| Kinetic | Evaluation of the forces involved in the body locomotion. | Ground reaction forces, torque, and momentum. |
| EMG | Refers to the analysis of muscular activity, generally performed by using EMG surface electrodes. | Motor unit action parameters. |

**Table 1.**
*Parameters generally used for gait analysis (adapted from [22]).*

of a doctor or a therapist. For this analysis, the patient is asked to vary several gait-related parameters, while walking in a predetermined circuit [7]. This type of analysis is bit limited in the information that can be retrieved, nevertheless, this could be useful for an initial evaluation and posterior decision on which objective techniques should be used. In contrast to the subjective techniques, objective gait analysis is more of a quantitative evaluation of the parameters listed above. This type of analysis requires the use of different types of equipment and procedures to measure the gait parameters. These methodologies can be categorized according to the technology used, varying from the ones based on imaging, instrumented walking platforms or floor sensors, and wearable sensors [7, 24, 29].

For an e-Health architecture, the most suitable technology would be the one built using wearable sensors, which would be able to acquire the patient gait parameters, everywhere and under any conditions. Among those, FBGs can be considered as an objective technique for gait analysis (allows the quantification of parameters during gait analysis), which could be used in instrumented platforms or as wearable sensors [30]. Recently, the use of FBGs as wearable sensors for remote monitoring of patients has been reported [12, 31, 32].

### 2.3 Gait pattern monitoring: e-Health architecture

The Internet of Things (IoT) concept is the fusion between pervasive network connectivity and the computing capability expanded to sensing devices and objects, able to acquire and exchange data autonomously. In recent years, due to the potential gains brought to the citizens' quality of life, IoT is seen as a whole platform able to bridge people and objects by integrating the smart concept into people's life, namely, smart cities, homes, wearables, and mobility [33].

Within the vast field of applications provided by IoT, e-Health stands out as one of the most influential topics on life-quality of humans, as smart and connected healthcare services have been requested more enthusiastically. e-Health is gaining too much attention mainly due to the joint effect of the increase of insufficient and ineffective healthcare services, allied to the change in population demographics and the increasing demand of such change entails. The world's population aged over 60 years is expected to reach 2 billion by 2050 [34], which implies the rise of chronic diseases that may be translated on different degrees of mobility impairments, requiring a close monitoring and a patient-centered healthcare service, where the healthcare providers and patients are pervasively connected [12, 33]. Also aligned with such demands, the market for home medical devices is set to significantly grow from $27.8 billion in 2015 up to nearly $44.3 billion by 2020 [35]. The increase in available e-Health solutions is a remarkable step toward improving the healthcare services, along with the autonomy of debilitated or impaired citizens. Fundamentally, e-Health can be seen as the solution to help the elders and patients with chronic illness to live an active life, without compromising their mobility or daily routine [32].

e-Health systems use remote monitoring architectures composed of sensing devices responsible to collect patients' physiological information, analyze and store such data in the cloud. The information can afterward be wirelessly sent to the healthcare professionals for a decision/action. The continuous flow of information on the patient's condition improves the provided service at a lower cost, while simultaneously enhancing the life quality of patients, who need continuous attention [33].

The patients' physiological information can be collected by networked sensors, integrated in smart wearable systems, or placed within the patients living environment. Considering the specific case of gait analysis, the continuous, automatic, and remote monitoring of gait of impaired or under rehabilitation citizens allows the objective assessment for preventive and proactive supervision of the pathologies, as well as to closely assist the therapies in progress [36]. In this scenario, wearable sensing architecture allows not only the evaluation of the patients in the course of daily life activities, but also provides the feedback on the recovering/rehabilitation therapy to patients and medical staff through ubiquitous connectivity. Based on that feedback, new therapeutic instructions can be given remotely to the patient, maximizing the efficiency of the provided healthcare services [31, 32].

An e-Health architecture to monitor the gait pattern of a citizen/patient comprises three key elements: the monitoring system composed of a sensors' network (preferably wearable); a computer/analysis system to collect, analyze, and store the data; and finally, the wireless mobile gateway, responsible for data processing and wireless transmission to medical servers and decision centers [31, 32]. **Figure 3** schematizes the typical architecture involved in an e-Health scenario.

The first part of the considered architecture is responsible for the data sensing and consequently, for the information given to action centers. Therefore, it is crucial that the sensing network is as accurate and reliable as possible. The use of FBGs as IoT and e-Health enablers is becoming increasingly common, due to their sensing characteristics, when compared with the ones of their electronic counterparts, namely small size (in the order of micrometers), biocompatibility, multiplexing capability, immunity to electromagnetic interference, in addition to their high accuracy and sensitivity, even for applications in challenging environments [37–39]. Consequently, FBGs can be used as a reliable solution for the integration in e-Health architectures, as for monitoring sensing systems in biomechanics and physical rehabilitation. Some examples can be mentioned, covering the detection of bone strains, mapping of gait plantar and shear pressures, measuring of pressures in orthopedic joints and angles between the body segments, as have already been successfully reported [12, 30].

In the following sections, the use of FBGs to monitor different body segments involved in gait will be explored.

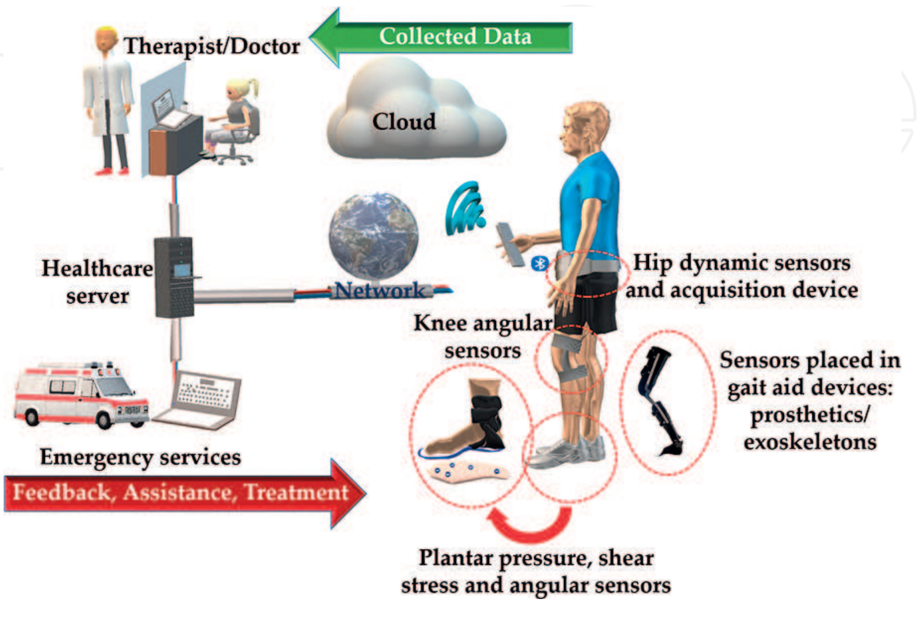


**Figure 3.**
*Possible architecture for a gait e-Health monitoring system.*

## 3. Plantar pressure and shear analysis

The assessment of plantar and shear pressures is of great importance for the gait health evaluation analysis, aiming to understand the effects induced in/by the body and to prevent the ulceration of the foot [40]. The pressure ulcers occur when tissue is compressed during prolonged periods of time, resulting in a wound that can infect and cause amputation, or in more severe cases, the patient's death [41]. An early identification of individuals at risk of foot ulceration (people with diabetes mellitus and peripheral neuropathy) is one of the primary means to reduce its incidence [40, 41]. Due to the poor load distribution, resulting from the reduced sensitivity of the foot, abnormally high plantar pressures occur in certain areas of the foot, and when that happens, it can lead to the growth of pressure sores in these locations [42]. The most affected areas are those with bony prominences, such as under the metatarsal bones, where the majority of plantar neuropathic ulcers occur [43]. Correct and continuous mapping of plantar pressure can prevent the occurrence of these pathologies, with the adoption of different walking habits or the use of correction equipment. On the other hand, in the case of the existence of ulcers, a redistribution of the forces imposed on the foot during walking aids healing and prevents further ulceration.

The force applied to the skin surface by a supporting structure has two components: the pressure acting normal to the surface and the shear stress acting in a tangential direction. Many authors have suggested that shear stresses have a pathogenic factor in the development of plantar ulcers [40–43]. This shear stress exists if there is sliding between two surfaces (foot and shoe), and it is closely related to friction [44]. Despite the importance of shear monitoring in assessing gait patterns, only normal pressure is widely reported. The lack of a validated and commercially available shear stress sensor is one of the main reasons why shear analysis is not as referenced as plantar pressure.

There are several solutions in the market for measuring plantar pressure, static in the form of fixed platforms, and wearable as shoes insoles. In regards of the importance of gait-related pathologies in the general population, and in the elder generation in particular, several works have been developed to improve the state-of-the-art. The literature reports the use of various technologies of plantar pressure and shear sensors, such as magneto-resistors, strain gauges, piezoelectric materials, capacitive sensors, and micro-strip antennas and coils.

As an alternative to these electronic devices, optical fiber-based sensors stand out due to their small diameter (hundreds of micrometers) and robustness, biocompatibility, high precision, electromagnetic insensitivity, as well as being electrically free at the point of measurement and owning the property of being able to multiplex several sensors in the same fiber, which allows to simultaneously monitor different parameters [45, 46].

Following such path, research studies have been carried out with FBG-based sensors used to measure not only plantar pressure but also shear parameters.

### 3.1 Plantar pressure sensors

The plantar pressure monitoring devices can be presented as fixed platforms or as insoles to be used directly in the footwear. Platform systems are typically constructed of several pressure sensors arranged in an array embedded in the floor or in a rigid platform. These systems can be used for static and dynamic studies, but are generally restricted to clinics and laboratories. In the case of the static tests, the patient stands still on the platform. On the other hand, for the dynamic tests, the platform is placed on the floor and the patient walks through it.

The application of these types of measuring systems has the advantage of being easy to use, since the platforms are flat and stationary. Nevertheless, these systems also present disadvantages, since they are influential to the patient's gait, once during the examination, he/she will have to tread specific areas of the platform surface [47].

Insole sensors can be incorporated into shoes, so that the measurements truly reflect the interface between the foot and the shoe. These systems, as they are flexible and portable, allow a greater accuracy of the acquired data, regarding the natural gait of the patient, and also greater variety of studies with different walking tasks, footwear, and even diverse floors/terrains [47]. However, insoles usually have a reduced number of sensors compared to platforms. The main requirements for the development of wearable/in-shoe sensors are: mobility, reduced number of cables, low power consumption, low cost, high acquisition frequency, proper sensitivity, noninvasiveness, and do not represent any danger to the user. Therefore, fiber optic sensors, due to their characteristics, have proven to be a reliable solution in this type of applications. Also, within the range of fiber optic sensors, the FBGs seem to be the best solution, since their multiplexing capability allows to have multiple sensors into a single fiber, reducing the number of cables needed in the insole. In this section, some recently developed work using optical fibers with Bragg gratings, as plantar pressure sensors in fixed platforms and in-shoe equipment, will be described.

The first work with FBG sensors incorporated in platforms for the measurement of plantar pressure was published in 2003, when Hao's team developed an insole shaped device with a silica optical fiber with five FBGs [48]. The insole was constituted of 10 layers of carbo-epoxy, among which the optical fiber was placed. The sensing FBG units were placed strategically at the main pressure points (heel and metatarsal areas). The device was tested in static tests to determine which areas have the greatest and lowest pressure at different user positions. The results showed that the sensors had an average sensitivity of 5.44 pm/N.

In 2014, Suresh et al. published a work comparing the use of FBGs and piezoelectric (PZT) sensors for gait monitoring at low and high speeds [49]. To manufacture the optical sensing platform, the FBGs were embedded between layers of a carbon composite material (CCM) in the form of an arc. After that, both types of sensors were placed on the underside of a commercial shoe (**Figure 4a**). For the dynamic test and to verify the behavior of both types of sensors, a healthy male walked on a treadmill wearing those shoes at various speeds. For the FBGs sensors, a mean pressure sensitivity of 1.3 pm/kPa was obtained. The study revealed that the FBG sensors have a better performance in the static moments and at lower speeds, while the piezoelectric sensors had greater performance for higher speeds.

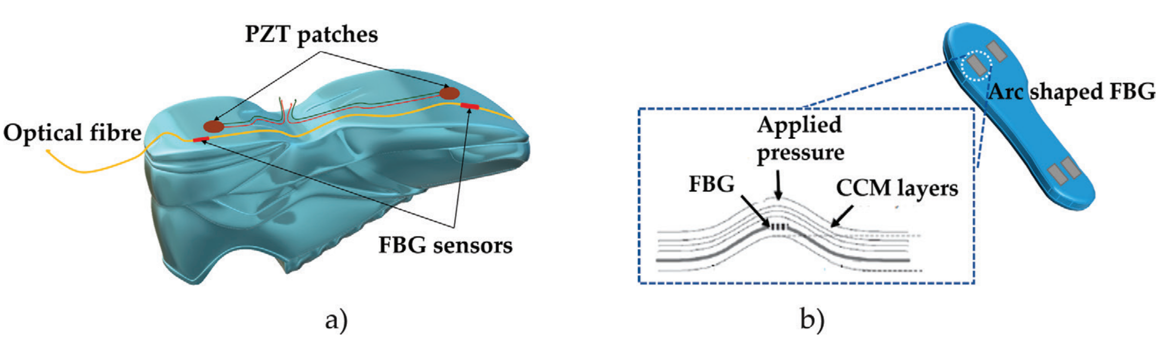


**Figure 4.**
*(a) Scheme of the shoe with the attached FBG and PZT sensors (adapted from [49]); and (b) schematic representation of an arc shaped FBG pressure sensor (left) and the insole sensing scheme (right) (adapted from [50]).*

Other approach was made by the same team, in which similar FBG cells were incorporated in an insole structure, as can be verified in the **Figure 4b**. The device was constituted by four arc-shaped cells, strategically placed in the forefoot and heel area. In this study, the plantar pressure was analyzed in both the fixed platform and the in-shoe systems. An average pressure sensitivity of 1.2 pm/kPa was obtained [50].

In 2016, Liang et al. proposed a sensing system based on six FBGs inscribed into a single fiber, which was embedded in silicone rubber [51]. The data registered by the optical sensors were compared to the ones collected through an i-Step P1000 digital pressure plate with 1024 barometric sensors. For the sensors' validation, 11 participants were tested, and according to the results, the viability of the optical sensor for this kind of measurement was demonstrated. Additionally, four different foot supporting types were successfully identified.

In 2017, Domingues et al. reported the development of two noninvasive solutions with FBGs in silica optical fiber incorporated in cork to monitor the body center of mass displacements and vertical ground reaction forces induced in the foot plantar surface during gait [12, 32]. One of the solutions, containing five FBGs, was developed to act as a fixed platform, and the other, with six FBG sensors, to be used as an instrumented insole to be adapted in a shoe as shown in **Figure 5**. Although the insole is made of five FBGs multiplexed in the same fiber, a clear isolation of each sensing point was also demonstrated, as seen in **Figure 6a**. Upon the calibration of the sensors located at point 1 (heel area), when the increasing load is applied in that point, only the FBG 1 shows a Bragg wavelength shift, proportional to the load applied (**Figure 6b**) [12].

The obtained results demonstrated the accuracy and reliability of the proposed systems to monitor and map the vertical active forces on the foot's plantar area during gait, **Figure 7**, with a sensitivity up to 11.06 pm/N.

The top graphic representation corresponds to the values independently registered by the five FBGs along time during two gait cycles, where the sum of the forces acquired by each FBG corresponds to the typical gait pattern [12]. In more detail, in the bottom graphic representation, it is possible to see which points of the insole are more actively pressed during the stance phase of the gait cycle. The dark blue representations in the foot, corresponds to the foot area that supports a higher load in the different stages of the stance phase [12, 32].

In the same year, a polymer optical fiber (POF) sensing system based on FBGs to measure foot plantar pressure was also described [52]. The plantar pressure signals were detected by five FBGs recorded in a cyclic transparent optical polymer (CYTOP) fiber, which was embedded in a cork platform in the form of an insole to monitor plantar pressure during gait. Initially, two studies were made with the insole as a fixed platform, one in which the user walked through the sensing

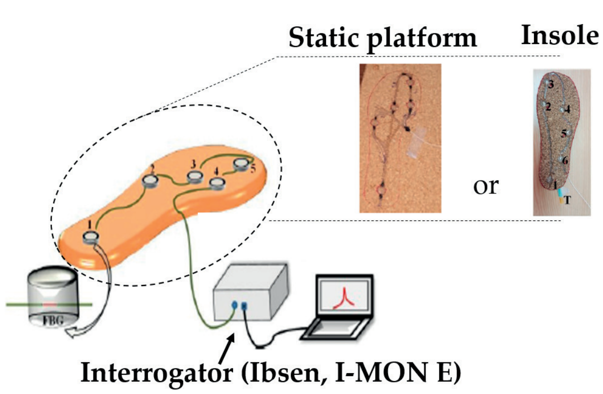


**Figure 5.**
*Schematic representation of the cork insole FBG monitoring system (adapted from [12]).*

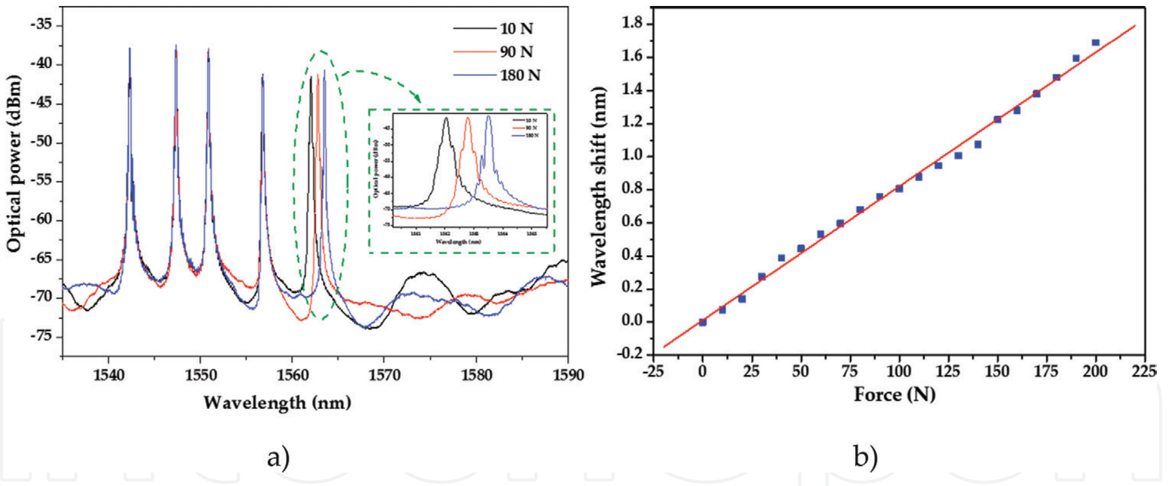

**Figure 6.**
*(a) Reflection spectra of the five FBGs multiplexed in the cork insole for three different load values; and (b) Bragg wavelength shift dependence on the load applied for FBG1 (adapted from [12]).*

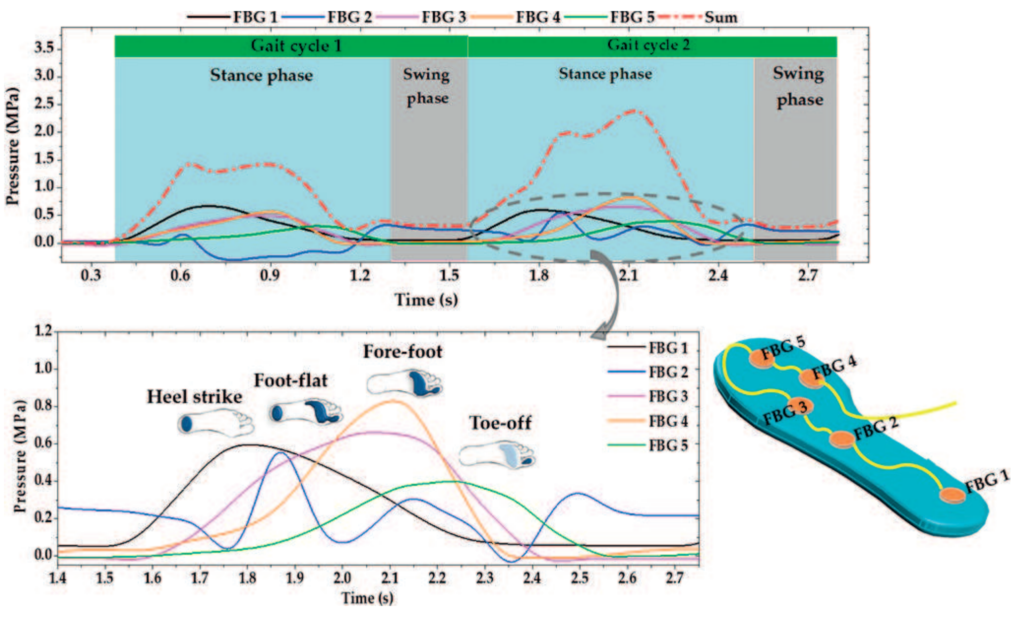

**Figure 7.**
*Representation of two complete gait cycles registered using a cork insole instrumented with five multiplexed FBGs (adapted from [12, 32]).*

structure and another where he stood in the platform just moving the body center of mass. The data obtained from this device in both tests showed good repeatability and a sensitivity twice as high as the solutions based on silica optical fiber. Additionally, a team of researchers from Shanghai presented a sensing platform based on FBGs using the fused deposition modeling (FDM) method for the construction of the structure [53]. This platform was composed of several cylindrical structures in polylactic acid (PLA) with the FBG inside them. This device was designed to be used as a fixed broad platform for plantar pressure monitoring, which demonstrated to have a reliable mechanical performance.

Finally, a noninvasive and efficient insole FBG-based architecture for monitoring plantar pressure was presented in [32]. This work stands out from the others, because the authors introduced a whole IoT solution with the insole sensors integrated with a wireless transceiver, exhibited high energy efficiency and secured data transmission, to ensure the mobility and privacy of user data. The presented data reflected the precision of the proposed system, with the sensors having sensitivities up to 7.8 pm/kPa.

### 3.2 Plantar pressure and shear sensors

FBGs have also great potential for measuring shear stresses in the shoe. Although there are reports of sensors developed for shear measurement [54, 55], measuring plantar and shear pressure simultaneously is more attractive and provides more insights about the wellbeing of the foot and the overall health of the person. Due to the advantages of fiber optics, and FBGs in particular, several researchers have been working on the design of FBG-based cells able to measure these two forces simultaneously. Although the main objective is the measurement of these parameters during gait, none of the studies refers the introduction of the developed sensing cells in insoles or platforms, presenting only the cells in its isolated form.

The first work published with simultaneous shear and vertical forces sensing with FBGs goes back to the year of 2000. The team of Koulaxouzidis developed a cell, using three optical fibers with an FBG each, embedded in a block of elastic material, as schemed in **Figure 8a** [56]. The developed sensor was able to measure the vertical stress, as well as the magnitude and direction of the shear stress on its top surface. The experimental results showed a good repeatability and a resolution near to 5 kPa in the measurement of both forces. Later, in 2013, Zhang et al. developed an identical sensor to the previous one. In this case, two POFs with one FBG each were used, one of them was placed horizontally (hPOF, hFBG), while the other was tilted (tPOF, tFBG). Both fibers were embedded in a soft polydimethylsiloxane (PDMS) matrix, as shown in **Figure 8b** [57]. The sensor had a 27 mm length and width, and a 22 mm height. In this work, the obtained pressure sensitivity was 0.8 pm/Pa in a full range of 2.4 kPa, and the shear stress sensitivity was 1.3 pm/Pa for a full range of 0.6 kPa.

In 2015, Chethana et al. developed an optical sensor ground reaction force measurement platform for gait and geriatrics studies [58]. The developed system consisted of eight FBGs to measure the respective soil reaction forces on the three axes (x, y, and z). Four of the FBGs were placed at the vertices of the measuring platform, monitoring the shear motions on the x and y-axes (two for the x-axis and two for the y-axis motions detection). The remaining four FBGs were placed one on each frame supporting leg to measure the plantar pressure exerted on these zones. According to the authors, the optical fiber sensors platform for ground reaction force measurements presented a zero cross-force sensitiveness in all three loading axes [58].

In 2018, Tavares et al. developed a cell with the same operating principle as previously reported, but using only one silica optical fiber with two FBGs placed individually in two adjacent cavities, one made of cork and another of polylactide acid (PLA), as shown in **Figure 9** [59]. For the cells' calibration, the used method was similar to the one described in Ref. [57], and the obtained values were

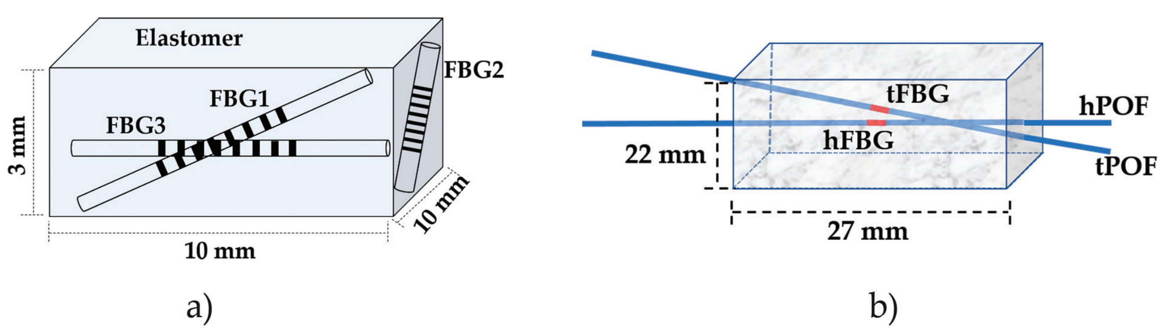


**Figure 8.**
*Schematic representation of the FBG-based sensor cell developed in (a) silica (adapted from [56]) and (b) POF (adapted from [57]) fibers.*

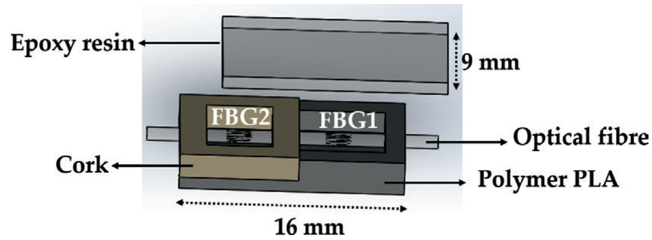


**Figure 9.**
*Scheme of the shear and pressure sensing cell with its different components and respective dimensions (adapted from [59]).*

compared with a 3-axial electronic force sensor. The results demonstrated that the developed device is a reliable solution for simultaneous measurement of shear and vertical forces. This solution has a great advantage over previous ones, since it only requires one optical fiber, which facilitates its incorporation into insoles. Therefore, several points, along the foot plantar surface, can be measured with a single optical fiber [12, 51, 52], with the advantage of being able to simultaneously diferentiate the two diferent forces (shear and pressure).

There are also studies using FBGs for vertical and shear forces measurements, but in which shear measurement is indirectly inferred from temperature variations. The authors argue that a rise in temperature in a certain area of the foot presupposes that there was friction between the surface of the foot and the shoe (shear force) [44, 60]. Najafi's team published a work in 2017 with the validation of a smart-textile based on fiber-optics with FBGs (SmartSox) for simultaneous measurement of temperature, pressure, and joint angles in patients with diabetic peripheral neuropathy (PND), where irregular temperature increase suggested the presence of shear forces [44]. In this study, FBG sensors were placed in socks that were successfully tested in a clinical setting by 33 individuals with PND to evaluate plantar pressure and temperature during normal gait velocity in a clinical setting.

## 4. Lower limb joints monitoring

The knee, hip, and ankle have a key role in gait, as it allows the body locomotion with muscles' minimum energy consumption and provides stability to walk in different terrain relief. During gait, the lower limb joints act together in order to provide the smoothest locomotion for the body. In **Figure 10**, the kinematics of the lower limbs in the different phases of gait are represented, namely the stance (a) and the swing phases (b) [14, 28].

At the beginning of the stance phase, in first double support and at the heel strike, the hip is flexed at 30°, the knee is extended and the ankle is at a neutral position. As the loading response approaches with the foot flat, the hip continues in a flexed mode as the knee starts flexing 5–10°, along with the ankle plantar flexing up to 20°, for the weight acceptance, shock absorption, and to propel the body forward. At the mid stance, the hip is extended, the knee flexed by 5–10° and the ankle is dorsi-flexed, with the purpose to move the body over the stationary foot. As the heel rises, with the ankle dorsi-flexing at 15°, the hip is extended at 15–30° and the knee is extended and then flexing. At the last point of the stance phase, in the toe off moment, the hip is flexing, the knee is also flexing at 30–40° and the ankle has a plantar flexing of 20–30°, in the preparation for the swing phase and the transfer of the load to the other limb [61].

At the initial swing, the hip continues flexing at 15°, the knee is flexing up to 65°, and the ankle is plantar flexed at 10° to clear the foot from the floor and advance the

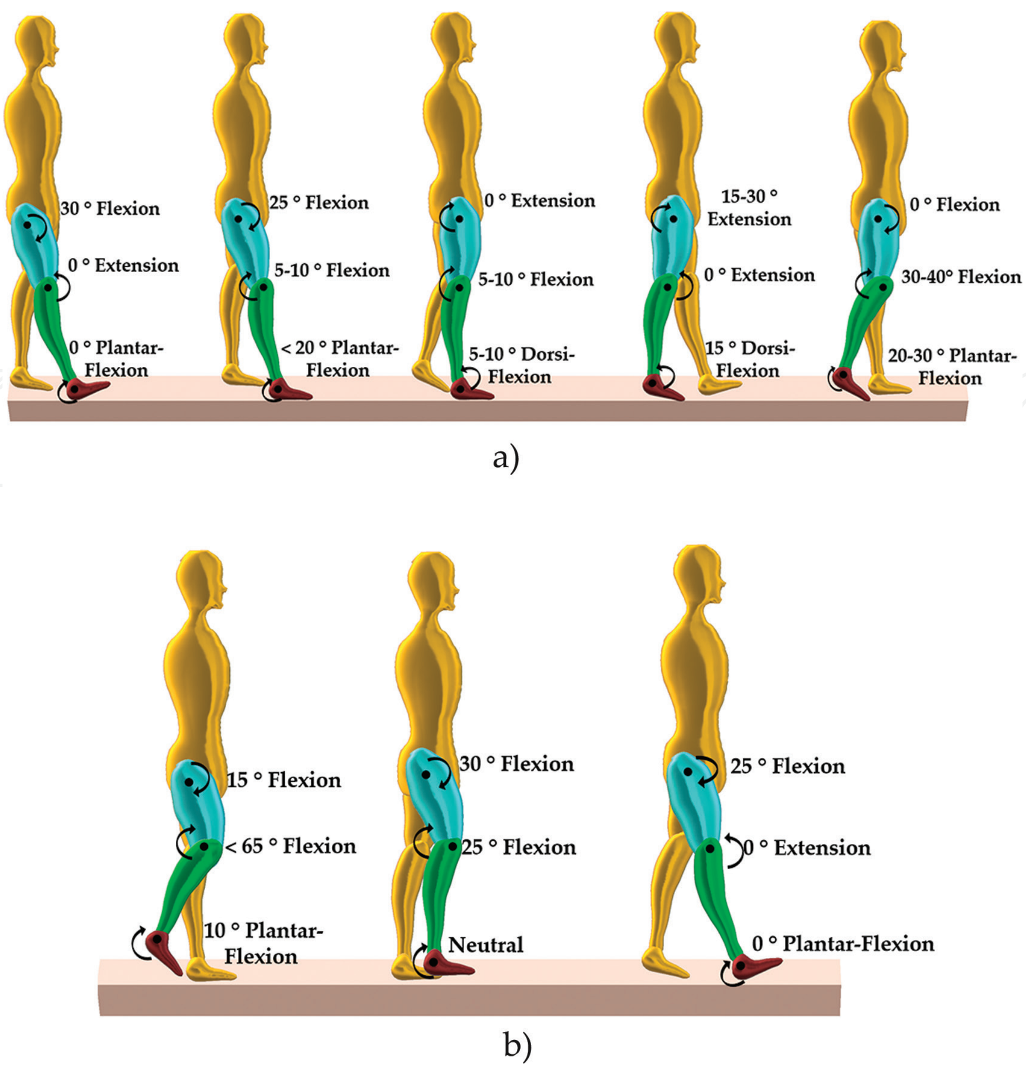


**Figure 10.**
*Schematic representation of the lower limbs kinematics involved in the: (a) stance and (b) swing phases.*

limb. At the mid swing, the hip is flexed at 30°, while the knee is flexing at 25°, and the ankle is at a neutral position. At the terminal swing, the hip is flexed at 25°, the knee is extended, and the ankle is at a 0° plantar flexion, to prepare the next heel contact at the beginning of the new stance phase [61].

The use of objective techniques to evaluate the health conditions of the knee can be a powerful tool for researchers and medical staff, providing relevant information about tendon-ligament strains and vibration, pressure, angular range of movements, and even temperature [62, 63].

There are numerous conventional techniques that can be used to monitor the joint conditions, such as stereo-optic, solid state, and piezo-resistive sensing methods, which employ accelerometers, magneto-resistive sensors, flexible goniometers, electromagnetic tracking systems, among others [24, 62, 64]. However, these techniques usually require complex and expensive electronics, which are susceptible to magnetic interferences and also cannot be used in humid/wet environments. Therefore, they do not represent an ideal solution for wearable sensing configurations, where the human transpiration may influence the sensors performance. So, the increasing research in the field of optical fiber sensors has also been focusing in the introduction of FBG technology in monitoring the lower limb joints during walking. Optical fiber sensors can be easily adapted to curved surfaces and various contours of the human body, especially the knee, a joint with complex anatomy [62, 63, 65–67].

The ideal technology to monitor limb joints should be able to measure curvature, being useful not only to monitor the motion of the lower limb segments, as well as to evaluate all the corporal posture [65]. The development of a smart garment, based on FBGs and flex sensing technologies, to monitor the body posture and lower and upper limbs' movements, was reported. An FBG-based sensing belt was produced by encapsulating FBG sensors inside a synthetic silica gel, as depicted in **Figure 11**, which was afterward attached to a garment for monitoring joints and body posture. The encapsulation of the FBG was made with an applied pre-stress, so the sensor is able to monitor both extension and compression deformations. The proposed FBG belt, fixed near the limb joints, is influenced by the body posture shifts, and the consequent sensor's Bragg wavelength shift was correlated with the angles at the limb joint [65]. Although the results presented by Abro et al. are only related to tests made at the upper limbs, the reasonable results obtained within the tests and exercises are a good indication of its potential application for the monitoring of the lower limbs motion.

### 4.1 Knee flexion-extension monitoring

From the lower limb three joints (ankle, knee, and hip), the knee is one of the body joints most prone to develop osteoarthritis [68]. Therefore, the supervision and monitoring of the motion of the knee are of crucial importance in the medical and physical rehabilitation field [37, 62, 67, 69, 70].

Rocha et al. suggest a wearable knee motion sensor, designed with a single FBG, embedded in a stretchable band of polyvinyl chloride (PVC) material and placed in the center of the knee joint, as schematized in **Figure 12a** [67]. The objective is to measure the knee movements from the straight leg to the maximum knee flexion and to obtain graphically the pattern of human gait, by monitoring flexion and extension, with the joint acting as a rotation axis, as represented in **Figure 12b**. The PVC band with the FBG was attached to an elastic ribbon (knee brace), by metallic pressure-buttons that ensure the stability of the sensing band while walking or running. In the reported work, the authors tested the proposed solution on a treadmill, under different types of run and speed, accompanied by video recorder [67]. The video was used to define the starting time of the stance and swing phases in order to correlate the data provided by the FBG sensor to the different phases of the walking routine [67, 71, 72].

When the leg is straight, the FBG sensor is in the resting position since there is neither flexion nor extension in the optical fiber. Once the bending movement of the knee starts, during walking, it results in an extension of the optical fiber, inducing a strain in the FBG sensor, positioned at the center of the knee joint. Consequently, a positive shift of the reflected Bragg wavelength is obtained. The reverse bending movement, from the maximum knee flexion point to straight leg, leading to a relaxation of the FBG, return to its initial Bragg wavelength value [67]. By monitoring the wavelength shift during these movements, the gait pattern of the patient could be characterized.

Although the researchers Rocha et al. show a clear characterization of the movement of the knee joint during the gait cycle, they also point out, as a drawback, the noise induced in the signal by vibration, considering that better results are achieved at lower speed, softening the influence of the elastic factor of the knee band [67].

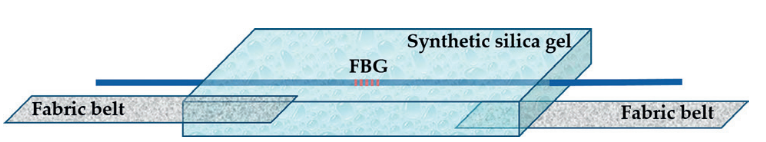


**Figure 11.**
*Schematic representation of the FBG belt proposed by Abro et al. (adapted from [65]).*

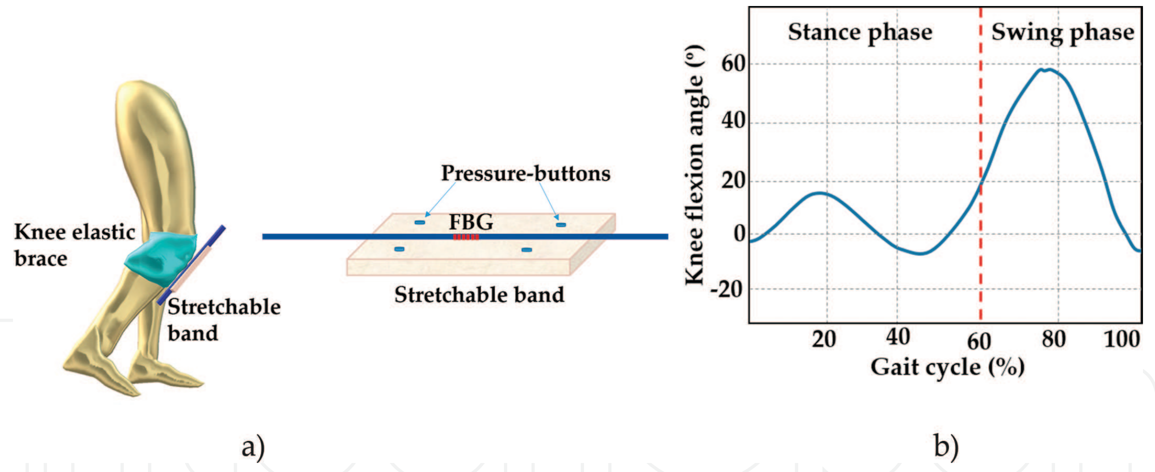


**Figure 12.**
*(a) Schematic representation of an FBG-based solution for knee movements monitoring (adapted from [67]); and (b) typical keen angle pattern during gait.*

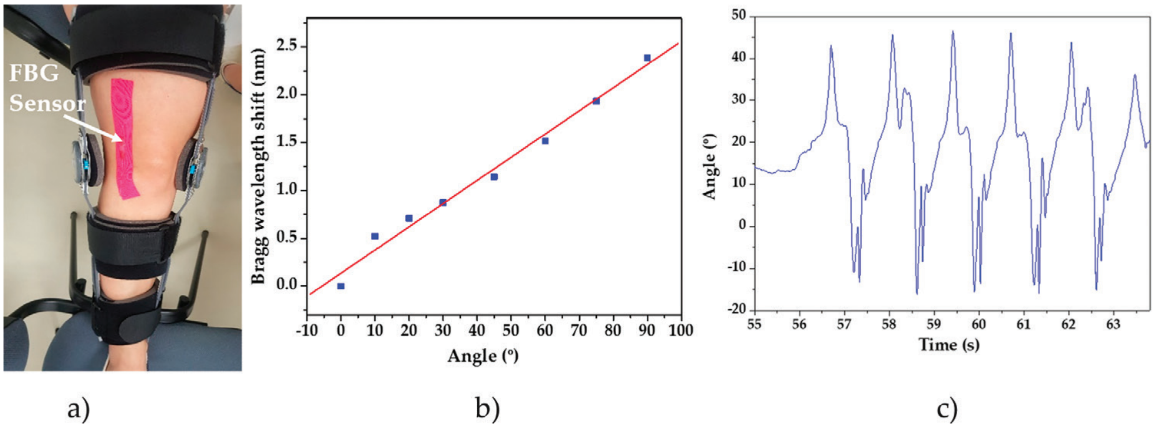


**Figure 13.**
*(a) Photograph of the kinesio tape with an embedded FBG for keen angle monitoring; (b) Bragg wavelength shift dependence with the keen flexion angle; and (c) knee flexion/extension angle during six gait cycles.*

Similar results can be achieved using kinetic tape (elastic adhesive tape) with an embedded FBG. The kinetic tape is attached to the lower limb, starting at the quadriceps area and ending at the beginning of the tibia, with the FBG placed just a few centimeters above the knee rotation axis, as shown in **Figure 13a**. Such configuration is a more stable solution, since the fiber containing the FBG is only actuated by the rotation of the knee, which stretches the kinetic tape inducing a strain and consequent positive wavelength shift in the FBG. During the calibration process, using an angle lock goniometer for angles ranging between 0 and 90°, a direct relation between the knee angle and the Bragg wavelength shift was found as displayed in **Figure 13b**. In **Figure 13c** is presented the flexion/extension angles, along time for six gait cycles, obtained with the solution represented in **Figure 13a**, and which as a similar behavior as reported by the authors in Ref. [67], but with a considerable reduced noise level.

### 4.2 Ankle flexion and dorsi-flexion monitoring

Umesh et al. proposed an FBG goniometer based on the deflection produced in an optical fiber by variation of the angle of the goniometer [73]. The purpose of the sensor is to measure the range of movement (ROM), which for the ankle joint the movement can be classified as ROM plantar-flexion and ROM dorsi-flexion. Plantar flexion is described as the rotation that increments the angle described between foot and the shin, and the dorsi-flexion is the rotation that results in a lower angle. The sensor is an assembly of two discs of 30 mm, overlapped by two discs with smaller diameter (5 mm). The two pairs of discs are circled by a rubber belt, to ensure synchronized rotation between them. The optical fiber with the FBG sensor is placed in a cantilever, connected to the upper belt. The rotation arm is linked to the side of the foot and its movement motivates the rotation of the correspondent disc. This rotation moves the cantilever and creates strain in the FBG, which can be rewritten in angle values, by proper calibration. The characterization of these two rotations has crucial importance in clinical diagnosis, helping the evaluation of the limitations of this joint. Furthermore, it is a noninvasive method of measurement with the advantages that optical fibers offer, and that can counteract to limitations of conventional electro-goniometers and video tracking systems as electromagnetic interference, size, and fragility [73].

### 4.3 Tendons and ligaments monitoring

Beyond their ability to measure the flexion, dorsiflexion, and extension of the joints involved in gait, FBGs can broaden their usage to applications related to the tendons and muscles monitoring. Although it may fall a bit out of the scope of e-Health, it is worth mentioning the application of FBGs to perform pressure mapping, and monitor strain and length of tendons and ligaments, when under load or locomotion. Ren et al. presented an FBG sensor embedded in a micro-shape memory alloy tube which is able to measure the displacement of the tendon [74]. To verify the performance of the sensor, the initial tests were made in the Achilles tendon and the results compared with the ones obtained simultaneously with a two-camera stereovision sensor. The fiber-based sensor was also applied to a cadaver knee tendon, in the medial and lateral collateral ligament, to record the deformation of the ligaments in simulated postures. The results proved that the FBG sensor has high sensitivity and low signal-to-noise ratio, without loss of accuracy. It is also easily implemented and minimally invasive to the biological tissues, projected to be applied *in-vivo*, after some improvements [74].

## 5. Prosthetic and exoskeletons applications

For severe impaired citizens, it is common to adapt prosthetic lower limbs, in order to offer patients mobility. The interface pressure inside the prosthetic sockets is of major relevance, in order to avoid ulcerations in the patients and evaluate its suitability. Moreover, the application of robotics technology to improve the wellbeing of debilitated patients has been highly investigated in the past few years. In particular, exoskeletons can be wearable devices prone to be used to restore functional movements of amputees and persons with paralysis. Therefore, this section surveys the use of FBG sensors for the development and evaluation of prosthetic limbs, in addition to control and automation of exoskeletons.

### 5.1 Prosthetic limbs

The partial or total limb amputation is one of the oldest treatment options available in medicine. Unfortunately, the frequency of the lower limb amputation is growing worldwide. Traffic accidents, particularly motorcycle accidents, health problems (including diabetes, arterial hypertension, chronic renal failure, and hypercoagulability), and advanced age are the main causes. Additionally, this is a predominant incident in countries affected by landmines and other natural disasters, including, for instance, earthquakes. Due to the socioeconomic impact (with the consequent inability to work and socialize), the interference on the life quality, and other complications, such as hematoma, infections, necrosis, contractures, neuromas, and phantom pain; this is a relevant public health problem.

The key element of amputee rehabilitation is the engineering of solutions, appropriated for individuals to recover their physical capabilities. A prosthesis or artificial limb is a device, whose function is to substitute the limb that was lost, with cosmetic and functionality for the amputee. A lower limb prosthesis results from the assembly of several components, including socket, shank, ankle, and foot, as schematized in **Figure 14**.

The socket is the most relevant component of the artificial limb, since it constitutes the critical interface between the amputee's stump and the amputee. The design and fitting of the socket are also the most difficult processes, due to the particularity of each amputee's stump. When wearing the prostheses, the appropriated fit and comfort are critical factors that contribute to its successful use. Nevertheless, many amputees still complain about discomfort or pain, reporting a set of problems, including edema, pressure ulcers, dermatitis, and skin irritation, due to the use of the prostheses [76]. This is particularly related with the changes in the residual limb soft tissues (volume, shape, sensitivity, composition, among others), which vary during the day due to factors such as temperature, activity, and hydration.

As result, in the last years, several measurement systems have been proposed to assess the interface pressure between the residual limb and the prosthetic socket [77]. This includes electrical strain gauge [78], F-socket transducer arrays [79, 80], and finite element analysis [81–83]. The output from these systems has been used to improve the socket design. Nevertheless, despite the technological advances in the existing socket design and the measurement systems, available sockets still exhibit many weaknesses. For instance, apart from the high accuracy and sensitivity provided, the use of strain gauges requires modifying the sockets with openings for accommodation of the device. This procedure interferes in the socket shape, and consequently in the accuracy of the pressure measurements.

In the case of the F-sockets, these systems present flexibility, good sensitivity, and ease of use. Additionally, in contrast to strain gauges, no change in the


**Figure 14.**
*Typical transtibial prosthesis (adapted from [75]).*

socket shape is required, since F-sockets are quite thin, which can be placed *in-situ* between the residual limb and the prosthetic socket. Nevertheless, the nonlinearity, hysteresis, drift, and vulnerability to electromagnetic interferences are the main limitations. Additionally, the shear stresses are not accounted for, when this system is used.

The finite element analysis is a numerical modeling alternative, which, when applied to the residual limb, predicts the soft tissues load distributions and magnitudes. This information has been assisting the technicians during the socket design. Nonetheless, although some models already considered thresholds for tissue injury and adverse adaptation, and other researchers have included in the models parameters, such as comfort and pain threshold, several complaints are still reported from the use of the prostheses, due to the subjectivity, difficulty to evaluate these factors, and the inter- and intra-individual loading [83].

Consequently, new sensing methodologies with minimal limitations toward accurate measurements of the interface pressure within prosthetic sockets are essentially required. Thereby, the FBG technology was pointed out as a potential alternative to conventional methodologies [84]. In 2010, Kanellos et al. proposed a 2D optical FBG-based pressure sensor, predicting to be suitable for several biomedical applications, namely biomechanics, rehabilitation, and orthotics, including amputee sockets [85]. The device consists of FBGs embedded into a thin polymer layer of PDMS, with the minimum thickness of the sensing pad set to 2.5 mm. The sensor exhibited a maximum fractional pressure sensitivity of 12 $MPa^{-1}$, with a spatial resolution of 1× 1 $cm^2$, also revealing no hysteresis and real-time operation possibility. Due to the elasticity and ductility of the polymer, which match human skin behavior, the system becomes a flexible 2D pressure sensing surface. This configuration is appropriate to be attached or anchored to irregular shaped objects/bodies, allowing to translate more accurately all the phenomena that may occur in them. These properties meet the requirements of human machine interfaces, comprising amputee sockets, as initially predicted.

Apart from the medical requirements imposed on the FBG embedded-based sensor pads, which include real time acquisition, high sensitivity and resolution, and increased dynamic range, these systems also need to comply with a set of demands related to fabrication and packaging processes. These conditions result from the diversity of the patients to be treated, and also from their real life conditions. Thus, the influence of the fiber embedding depth (center and top position of pad cross-section), the thickness of the polymer sheet (2 and 3 mm), and the fiber type (hydrogenated SMF-28 and nonhydrogenated GF1B) were assessed in [86]. The results of this study reveal that the sensor pads rigidity and durability are enhanced, when the Bragg grating, inscribed into nonhydrogenated fiber, is embedded at the polymer center, with a thickness of 3 mm.

Results of the first investigation of the ability of FBGs to measure interface pressure between the stump of a trans-tibial amputee and a patellar tendon bearing (PTB) prosthetic sockets are presented in [75]. The patellar tendon (PT) bar was the key analyses' area since this supports the majority of the transtibial amputee's body weight, when the subject is using the PTB socket. In Ref. [75], the FBGs were first embedded into an epoxy material (NOA 61), aiming to acquire the required protection to withstand the high pressure values up to 230 kPa at the PT bar [78]. After that, this sensing pad was placed between two silicone polymeric sheets forming the pressure sensor, as schematized in **Figure 15**.

Since the initial contact of the PT with the sensor surface is mostly pressure concentrated, this behavior was imitated using a ball bearing, and positive wavelength shift of 3.8 nm was observed for a maximum load of 30 N, **Figure 16a**.

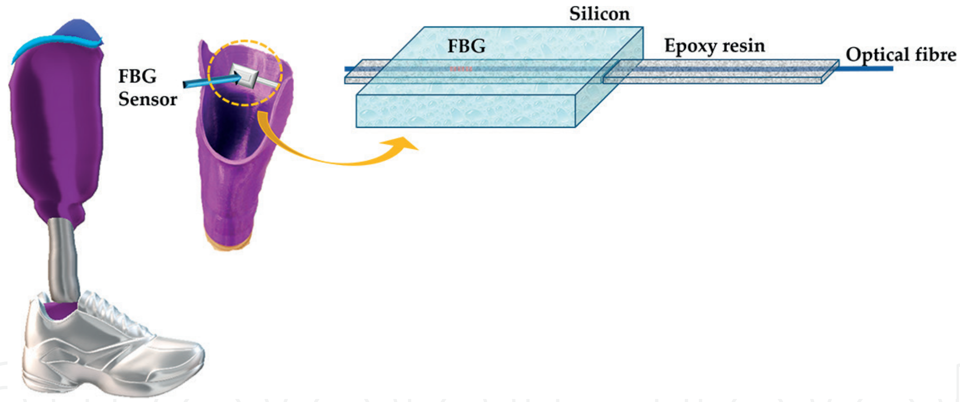


**Figure 15.**
*Schematic representation of an FBG-based system for monitor the interface pressure between the socket and the amputee (adapted from [75]).*

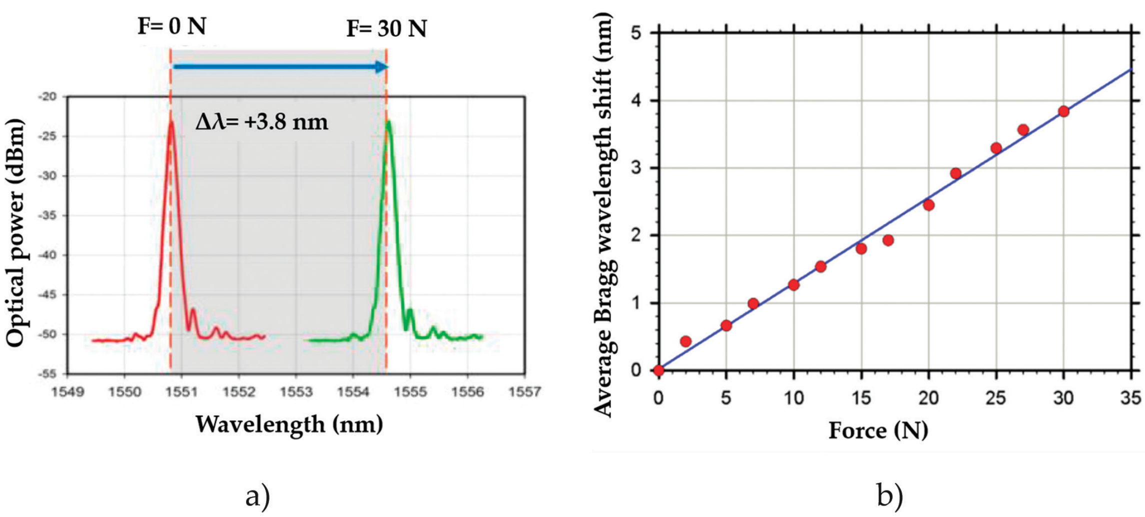


**Figure 16.**
*(a) Representation of the maximum Bragg wavelength shift for maximum load applied (adapted from [75]); and (b) average Bragg wavelength shift as function of the applied force (adapted from [75]).*

Thereafter, an experimental set up was designed to assess, *in-situ*, the sensor performance, while attached to the inner socket wall. Although there was no subject involved in these tests, consideration was taken to reproduce a real-life situation, as much as possible. The results obtained for the different load cycles reveal the suitability of the sensor to accomplish pressure measurements on the socket stump interface, especially in the PT region. From the calibration procedure, a proportional wavelength shift dependence with the load applied was found **Figure 16b**.

Toward using these sensors *in-situ*, the performance of these sensing pads was broadly assessed concerning the sensitivity, durability, and hysteresis error [87]. Similar to the work of Kanellos et al., three production parameters were investigated, which are the FBG embedding depth (top, bottom, and neutral layers of the sensing pad), the sensing pad thickness (1, 2, and 3 mm), and the type/hardness of sensing pad materials [PDMS (harder) and silicone rubber (softer)]. The best sensor's performance (highest sensitivity and accuracy) was obtained for the FBG embedded in the neutral layer of PDMS and with the thicker sensing pads. An FBG array was produced with these conditions and used for interface pressure measurements within prosthetic sockets. Additionally, to further assess the performance of the proposed sensing pad, these were evaluated *in-situ*, in a traumatic transtibial amputee using a total surface bearing socket, with 6 mm silicone liner. The results were validated comparing the data obtained with the FBG technology to the pressure measurements acquired by the F-socket sensors. Although the data obtained for the 8 sub-regions of the amputees' residual limb follow the same tendency; higher pressure values were registered by the FBG sensors. The difference was attributed to the sensors' thickness, which is 3 mm in the case of the FBG sensing pads and 0.2 mm in the F-socket sensing mats.

Aiming to eliminate the previous limitations and provide a simpler and more practical sensing procedure, Al-Fakih et al. proposed an innovative customized FBG-instrumented silicone liner, which consists of two silicone layers with 12 FBGs embedded between them, with the gratings located in clinical interest points [88]. In this study, a custom gait simulation machine was built to test the performance of the sensing system during an amputee's simulation gait. The data were validated with the findings obtained using an F-socket. The FBG technology revealed sensitivity and accuracy similar to the ones obtained with the F-socket technology. Nevertheless, this new design can be used repeatedly in clinical and research setting, which is an important benefit compared to the F-socket mats that, due to drift and calibration issues, are usually discarded after each utilization.

Recently, the technologies used to assess the interface pressure between the residual limb and the prosthetic socket, and the challenges found concerning the development of new solutions of sockets for limb prostheses were reviewed in [89]. In this chapter, FBGs are pointed out as one of these technologies. Additionally, the study observes that due to the high risk of the damaging of fibers, their applications are still limited, with further studies still required to confirm their suitability in this field. Nevertheless, the shown advantages of this technology over other sensing methodologies, especially regarding drift and linearity, and the constant low satisfaction level of the amputees, are enough motivations to continue investing on this technology.

### 5.2 Rehabilitation exoskeletons

The application of robotics, in particular robotic exoskeleton systems, to improve the wellbeing of debilitated patients is already being adopted. This technology is being used in human power augmentation, and its application has become more prominent, as to provide alternative solutions for physically limited people support in their daily movements [90].

Exoskeletons are known to be wearable robots (robotic exoskeletal structures), with a strict physical and cognitive interaction with the human user, since, typically, it operates alongside human limbs. Although the scientific and technological research on the development and implementation of exoskeletons began in the early 60s, only recently, its application in rehabilitation and functional substitution of movements have been implemented in patients with motor disorders [91].

Robotic exoskeletons provide unique methods for rehabilitation, by promoting the patient engagement in its training, and retrieving better quantitative feedback and improved functional outcome for patients. In a future perspective, the development of more effective exoskeletons is insight, with solutions for a real-time biological synergy-based exoskeleton, which will allow disabled patients to regain normal mobility capabilities [92, 93].

The exoskeleton feedback is based on the information, which is retrieved from the embedded sensors in its structure. The current exoskeleton designs can have up to several dozens of sensors, to monitor variables, such as rotation, torque, tilt, pressure, position, velocity, neurological signals, among others. As the sensing systems integrated in the robot are the key devices for its proper performance, the research field on robotics already has a mature and overspread technology, offering good sensitivity, precise measurements, and competitive price, with sensing systems often based on solid-state sensing [64]. Nevertheless, this technology has also shown some drawbacks, due to its susceptibility to electromagnetic interferences resulting from the electric inertial motors. This interference could be enough to degrade the sensors signal, sending erroneous information to the control devices, leading the exoskeleton to perform erratically, and risking injuring the patient.

Rehabilitation robotics applications also require the analysis of the body motion, in order to close control loops around defined joints. Commercial optical systems, such as Vicon, are considered the standard in human motion analysis. Although Vicon provides accurate position information, it has some significant limitations, such as high costs and limited measure volume, since it has to be used in laboratories with fixed equipment, which prevents its use in rehabilitation robotics applications [94]. On the other hand, soft exoskeletons require even more imperceptible sensors, typically sensor heads with thicknesses below 0.5 mm, in which electronic devices present some drawbacks, including long term instability, inconsistency, excessive drift, and the restriction to a small sensing area requiring the use of more sensors to monitor larger areas [64]. As an alternative to these electronic and optical sensors, the optical fiber sensors offer a small and robust solution, able to acquire kinematic and kinetic measurements, enhancing the exoskeleton performance by adding further responsiveness, controllability, and flexible motion. Nevertheless, the use of FBG sensors in exoskeletons is not yet widely explored, with only a limited number of studies reported. Recently, Domingues et al. reported the instrumentation of an insole with FBG sensors for plantar pressure monitoring [12, 32]. The reported wearable device is able to be adapted to exoskeletons structures, and dynamically retrieve the gait pattern of the patient.

Although there is a shortage of studies regarding the adaptation of FBG sensing technologies to exoskeletons, for gait aid there are already some reports focusing on its application in robot fingers and glove-based devices [95–97]. Park et al. presented an FBG-based solution to monitor the force in exoskeleton fingers [95]. The authors embedded the optical fiber sensors in a finger-like plastic 3D mesh, inspired in the design of arthropod limbs, near the fingers base, for enhanced sensitivity. With the developed structure, it is possible to detect forces down to 0.02 N, with a resolution of ~0.15 N. The robot hand instrumented by Park et al. was able to be operated in a hybrid control scheme, with the fingers being capable to sense small forces, with the advantage of being able to have all the FBG sensors in one single fiber, due to FBGs multiplexing ability [95].

Jiang et al. also described the design and production of an instrumented robotic hand with three fingers that enable both pinch and power grips. The optical FBG sensors were embedded in both the rigid plastic and soft skin material that constitutes the hand bone structure. In the rigid plastic material, the authors included eight FBGs for force sensing, while in the soft skin, they integrated six FBGs strain sensors for tactile monitoring, providing information on the location of the contact points [96]. Although there are already some studies related to the upper limbs motion aid, some work is still needed concerning the application of FBG technology to exoskeletons for gait rehabilitation of patients, which demands a direct focus on the lower limbs synergy between the patient and the exoskeleton.

Key topics for further development of exoskeletons in rehabilitation scenarios include the need for robust human-robot multimodal cognitive interaction, safe and dependable physical interaction, true wearability and portability, and user aspects such as acceptance and usability [91]. It should be able to augment the ability and/or to treat skeletal parts, which are weak, ineffective, or injured due to a disease or a neurological condition. Therefore, the exoskeleton should be designed to work in parallel with human body and be actuated either passively and/or actively [98].

## 6. Conclusion

e-Health has been widely investigated in recent years, building on technological advances, especially in fields such as sensing and networking. Building on such gains, more innovations are expected to enhance the life quality of citizens, especially debilitated and elder ones. Gait analysis stands out as one promising solution, which can help in the rehabilitation of locomotive impairments, in addition to early diagnosis of other pathologies, such as ulcers in patients with diabetes. Various solutions have been proposed in the literature for close monitoring and analysis of gait. However, recently, FBGs have been pointed out as a promising alternative for a sensing technology to analyze gait movement, building on advantages such as small size, rigidness, low-cost, low power consumption, and minimally invasive. Due to its recent adoption and promising advantages, this chapter has provided a thorough review of research and design efforts of FBG-based sensors for gait analysis. The chapter initially explains the sensing principle underlying the FBG technology, after that the topic of gait analysis and the different phases of gait cycle are described, and then moves toward required e-Health monitoring solutions. Efforts toward the design of solutions to monitor plantar pressure and shear forces are discussed. Monitoring of plantar pressure, independently, is first presented, then simultaneous monitoring of plantar and shear forces is further elaborated. The chapter then moves toward monitoring of lower limb joints, which also play key roles in the gait analysis, since their wellbeing affects the gait cycle pattern. The use of optical fiber sensing in prosthetic and exoskeletons concludes the topics discussed in the chapter. This chapter represents a thorough review of research efforts in the design of optical fiber-based sensors in gait analysis, covering all related topics of monitoring plantar pressure, shear forces, knee and joints, and integration in prosthetic and exoskeletons.

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## Author details

Maria de Fátima Domingues[1*], Cátia Tavares[2], Tiago Leite[1,2], Nélia Alberto[1], Cátia Leitão[1,2], Carlos Marques[1,2], Ayman Radwan[1], Eduardo Rocon[3], Paulo Antunes[1,2] and Paulo André[4]

1 Instituto de Telecomunicações, Campus Universitário de Santiago, Aveiro, Portugal

2 Department of Physics & I3N, University of Aveiro, Campus Universitário de Santiago, Aveiro, Portugal

3 Centro de Automática y Robótica, CSIC-UPM, Madrid, Spain

4 Department of Electrical and Computer Engineering, Instituto de Telecomunicações, Instituto Superior Técnico, University of Lisbon, Lisbon, Portugal

*Address all correspondence to: fatima.domingues@ua.pt

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